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Brian C. J. Moore

Cochlear Hearing Loss

physiological, psychological and technical issues

Second Edition

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BRIAN C. J. MOORE
Department of Experimental Psychology,
University of Cambridge,
Downing Street, Cambridge CB2 3EB,
England

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Contents

Preface

The is a second edition of *Cochlear Hearing Loss*, which was first published in 1998. The book is intended to impart an understanding of the changes in perception that take place when a person has a cochlear hearing loss. I have tried to interrelate physiological data and perceptual data. The book aims to present both data and concepts in an integrated way. The main goal is to convey an understanding of the perceptual changes associated with cochlear hearing loss, of the difficulties faced by the hearing-impaired person and of the possibilities and limitations of current hearing aids. The reader should come away with an impression not only of what happens but also of why it happens. I have attempted to write at a level that is accessible and comprehensible to students in audiology and in speech and hearing science. I hope, therefore, that the book will be considered suitable as a textbook for such students.

The book assumes some prior knowledge about sound. For example, it assumes familiarity with the decibel and the concept of the spectrum of a sound. However, an extensive glossary is provided to give brief explanations of technical terms used in the book. Also the concept of linear and nonlinear systems is explained in some detail, as this concept is crucial for understanding this book, but it is often poorly explained in textbooks on hearing.

Appreciation of the perceptual effects of cochlear hearing loss can be helped by listening to simulations of these effects. A compact disc containing simulations of different aspects of cochlear damage (Moore, 1997) can be obtained by writing to the author at: Department of Experimental Psychology, University of Cambridge, Downing Street, Cambridge CB2 3EB, England. The cost of the disc is 12 pounds sterling or 20 US dollars. Payment (a cheque payable to B.C.J. Moore) should be enclosed with the order. The disc also illustrates the effectiveness of various types of amplification and compression in compensating for the effects of cochlear hearing loss.

This second edition has been extensively updated and revised. In particular, new material has been added on: dead regions in the cochlea, pitch perception, loudness perception, the precedence effect, informational masking, speech perception and hearing aids.

I would like to thank the colleagues who commented on preliminary drafts of parts of the first or second editions of this book. They include José Alcántara, Thomas Baer, Deniz Başkent, Peter Blamey, Brent Edwards, Brian Glasberg, Hedwig Gockel, Ervin Hafter, Robert Peters, Marina Rose, Aleksander Sek, Michael Stone, Deborah Vickers, Joyce Vliegen and Magdalena Wojtczak. I would also like to thank Ian Cannell, Brian Glasberg, Aleksander Sek and Michael Stone for considerable assistance in producing figures. The second edition was written while I was on sabbatical leave at the Department of Psychology, University of California at Berkeley. I thank Ervin Hafter for inviting me to come and for the use of his office, and Lee Hafter for many splendid dinners.

Brian C.J. Moore
March, 2007

1 Physiological Aspects of Cochlear Hearing Loss

I INTRODUCTION

Hearing loss caused by damage to the cochlea is probably the most common form of hearing loss in the developed countries. Its most obvious symptom, and the one that is almost always assessed in the clinic, is an elevation of the threshold for detecting sounds. However, it is also accompanied by a variety of other changes in the way that sound is perceived. Even if sounds are amplified (e.g. by a hearing aid) so that they are well above the threshold for detection, the perception of those sounds is usually abnormal; the person with cochlear hearing loss often reports that the sounds are unclear and distorted, and that it is hard to hear comfortably over a wide range of sound levels. A common complaint is difficulty in understanding speech, especially when background sounds or reverberation are present. One of the main aims of this book is to explain why these problems occur and why current hearing aids are of limited benefit in compensating for the problems.

The book assumes that the reader has a basic knowledge of physics and acoustics, for example an understanding of what is meant by terms such as *sinusoid*, *spectrum*, *frequency component* and the *decibel*. The reader who is not familiar with these terms should consult a textbook, such as *An Introduction to the Psychology of Hearing* (Moore, 2003) or *Signals and Systems for Speech and Hearing* (Rosen and Howell, 1991). For the reader who knows these things, but needs a reminder, many of the key terms are defined briefly in the Glossary. Most of the terms that appear in italics in the text are defined in the Glossary. One concept that may not be familiar is that of a *linear* system. This topic is of importance, since the normal peripheral auditory system shows significant *nonlinearities*, whereas the system becomes more linear when cochlear damage occurs. Hence, this chapter starts with a description of the properties of linear and nonlinear systems. It then goes on to consider the physiology and the function of the normal and damaged cochlea.

II LINEAR AND NONLINEAR SYSTEMS

The auditory system is often thought of as a series of stages, the output of a given stage forming the input to the next. Each stage can be considered as a device or system, with an input and an output. For a system to be linear, certain relationships between the input and output must hold true. The following two conditions must be satisfied:

1. If the input to the system is changed in magnitude by a factor *k*, then the output should also change in magnitude by a factor *k*, but be otherwise unaltered. This condition is called *homogeneity*. For example, if the input is doubled, then the output is doubled, but without any change in the form of the output. Thus, a plot of the output as a function of the input would be a straight line passing through the origin (zero input gives zero output) – hence the term *linear system*. Such a plot is called an *input-output function*. An example of such a function is given in panel (a) of Figure 1.1.
2. The output of the system in response to a number of independent inputs presented simultaneously should be equal to the sum of the outputs that would have been obtained if each input were presented alone. For example, if the response

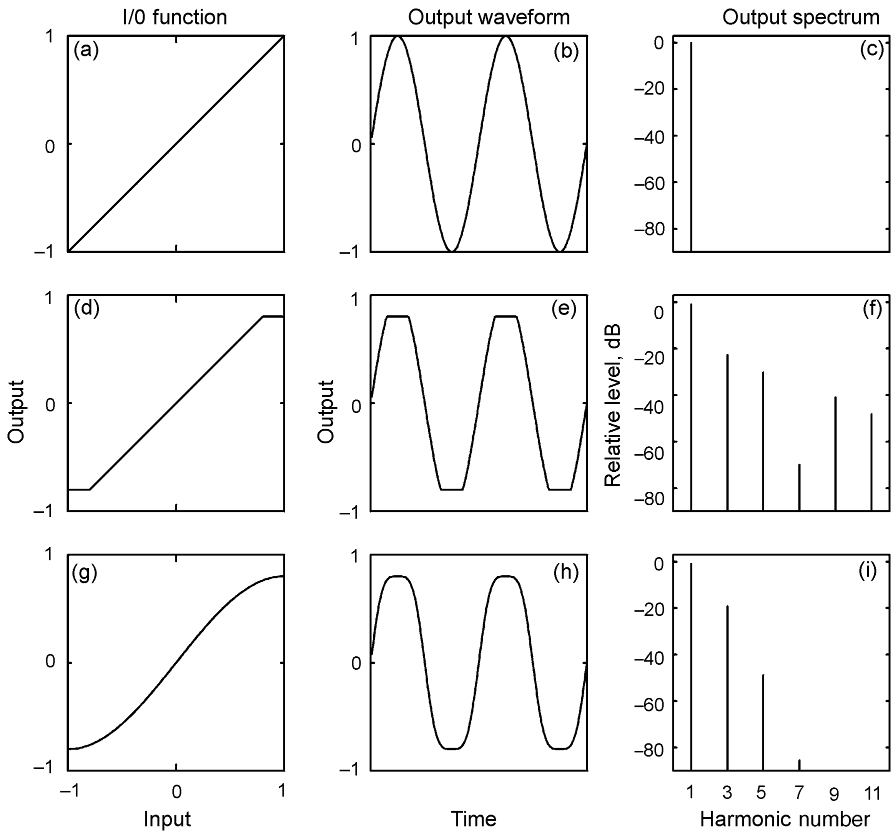


Figure 1.1. The left column shows input-output functions for a linear system (a), a nonlinear system with 'hard' peak clipping (d) and a nonlinear system with more progressive 'saturation' (g). The middle column shows outputs from these systems in response to a sinusoidal input. The third column shows spectra of the outputs.

to input A is X and the response to input B is Y, then the response to A and B together is simply X + Y. This condition is known as *superposition*.

When describing a system as linear, it is usually assumed that the system is *time-invariant*. This means that the input-output function does not change over time. For example, if the input is I and the output is O, the relationship between the input and the output would be:

$$O = cI, \tag{1.1}$$

where c is a constant that does not vary with time.

When a sinusoid is used as an input to a linear system, the output is a sinusoid of the same frequency. This is illustrated in panel (b) of Figure 1.1. The *amplitude* and *phase* of the output may, however, be different from those of the input. Assume that the input is a single sinusoid whose *waveform* as a function of time, I(t), can be described by:

$$\mathrm{I}(t) = A\sin(2\pi ft), \tag{1.2}$$

where A is the peak amplitude of the input and f is the frequency in Hz (cycles per second). The output as a function of time, O(t), could then be represented by:

$$\mathrm{O}(t) = G \times A\sin(2\pi ft + \phi), \tag{1.3}$$

where G is a constant representing the amplitude ratio between the input and output, and ϕ is a constant representing the phase shift between the input and output. G is sometimes referred to as the *gain* of the system.

Since the output of a linear system in response to a sinusoidal input is itself sinusoidal, the spectrum of the output, by definition, consists of a single frequency component. This is illustrated in panel (c) of Figure 1.1. More generally, the output of a linear system never contains frequency components that were not present in the input signal. The response of a linear system may, however, vary with the frequency of the input sinusoid. This is equivalent to saying that the constants G and ϕ in Equation 1.3 can vary with the input frequency.

In practice, many devices or systems are linear as long as the input is not too large. Excessive inputs may cause the system to become nonlinear; more details are given later on. Such a system is usually called linear, even though it can become nonlinear under extreme conditions. As an example, consider a loudspeaker. The input is a voltage and the output is a movement of the cone of the loudspeaker, which can produce audible sound waves. For the types of inputs that are typically used for a loudspeaker, the response is approximately linear; the conditions of homogeneity and superposition are obeyed. If the input voltage varies in a sinusoidal manner, the movement of the cone is almost sinusoidal. However, if the frequency of the input is changed, holding the magnitude of the input constant, the magnitude of the movement of the cone may vary. In this case, we would say that the loudspeaker does not have a 'flat' frequency response; G may vary with frequency. Similarly, ϕ may vary with frequency. Other examples of systems that usually operate in a nearly linear way are microphones, amplifiers and the output transducers used in hearing aids (often called receivers).

When waveforms other than sinusoids are applied as the input to a linear system, the output waveform will often differ from that of the input. For example, if the input to a linear system is a square wave, the output is not necessarily a square wave. This is one reason for the popularity of sinusoids in auditory research; sinusoids are the only waveforms which are always 'preserved' by a linear system. If a system is linear, then it is relatively easy to predict its output for any arbitrary complex input. As a first step, the output is measured as a function of frequency for a sinusoidal input. Essentially, the values of G and ϕ are determined as a function of the input frequency. To predict the output for a complex input, a *Fourier analysis* of the input is performed. This gives a description of the input in terms of the amplitudes and phases of its sinusoidal components. The output for each of the sinusoidal components comprising the input can then be calculated. Finally, using the principle of superposition, the output in response to the whole complex can be calculated as the sum of the outputs in response to its individual sinusoidal components. This is a powerful method, and it gives another reason for using sinusoids as stimuli.

As mentioned earlier, many linear systems become nonlinear if the input is made large enough. An example is shown in panel (d) of Figure 1.1. The input-output function is linear over a large part of its range, but it flattens out for large positive or negative values of the input. This is sometimes called *saturation* or peak clipping and can occur in systems such as transistor amplifiers and condenser microphones. In the example shown, the clipping is symmetrical, in that it occurs at the same absolute value for positive and negative values of the input. When a sinusoid is used as input to such a system and the peak amplitude of the sinusoid, A, is sufficiently large, the output is no longer sinusoidal. This is illustrated in panel (e) of Figure 1.1. The output is periodic, with the same *period* as the input sinusoid, but the waveform is distorted. In this case, the output contains frequency components that are not present in the input. This is illustrated in panel (f) of Figure 1.1. In general, the output of a nonlinear system in response to a single sinusoid at the input contains one or more components (sinusoids) with frequencies that are integer multiples of the frequency of the input. These components are referred to as *harmonics*, and the nonlinear system is said to introduce *harmonic distortion*. For example, if the input was a sinusoid with a frequency of 500 Hz, the output might still contain a component with this frequency, but components with other frequencies might be present too, for example 1000, 1500, 2000 . . . Hz.

Another example of a nonlinear input-output function is shown in panel (g) of Figure 1.1. In this case, the function does not show 'hard' clipping, but the slope becomes more shallow when the absolute value of the input or output exceeds a certain value. This type of input-output function can occur in valve (tube) amplifiers, moving coil microphones and loudspeakers, and it can also occur in the auditory system. The output waveform, shown in panel (h) of Figure 1.1, is less distorted than when hard clipping occurs, and the output spectrum, shown in panel (i), reveals less harmonic distortion.

If the input to a nonlinear system consists of two sinusoids, then the output may contain components with frequencies corresponding to the sum and difference of the two input frequencies, and their harmonics, as well as the original sinusoidal components that were present in the input. These extra components are said to result from *intermodulation distortion*. For example, if the input contains two sinusoids with frequencies f_1 and f_2, the output may contain components with frequencies $f_1 - f_2, f_1 + f_2$, $2f_1 - f_2$, $2f_2 - f_1$ and so on. These components are referred to as intermodulation distortion products, and, in the case of the auditory system, they are also called *combination tones*.

When a system is nonlinear, the response to complex inputs cannot generally be predicted from the responses to the sinusoidal components comprising the inputs. Thus, the characteristics of the system must be investigated using both sinusoidal and complex inputs.

Often, the input and output magnitudes of a system are plotted on logarithmic axes (the decibel scale is an example). In that case, the input and output magnitudes are not specified as instantaneous values (e.g. as the instantaneous voltage in the case of an electrical signal). Generally, instantaneous magnitudes can have both positive and negative values, but it is not possible to take the logarithm of a negative number. Instead, the input and output are averaged over a certain time, and the magnitude is expressed as a quantity which cannot have a negative value. Typically, both the input and output are specified in terms of their *power* (related to the *mean-square value*) or their *root-mean-square value*. Sometimes, the peak amplitude may be used. For a linear system, the condition of homogeneity still applies to such measures. For example, if the input power is doubled, the output power is also doubled. When plotted on 'log-log' axes, the input-output function of a linear system is a straight line with a slope of unity. To see why this is the case, we take the logarithm of both sides of Equation 1.1 ($O = cI$). This gives:

$$\log(O) = \log(cI) = \log(c) + \log(I) \tag{1.4}$$

The value of $\log(c)$ is itself a constant. Therefore, since $\log(O)$ is simply equal to $\log(I)$ plus a constant, the slope of the line relating $\log(O)$ to $\log(I)$ must be unity. In a nonlinear system, the slope of the input-output function on logarithmic axes differs from unity. Say, for example, that the output is proportional to the square of the input:

$$O = cI^2 \tag{1.5}$$

Taking logarithms of both sides gives:

$$\log(O) = \log(cI^2) = \log(c) + 2\log(I) \tag{1.6}$$

In this case, the slope of the input-output function on logarithmic axes is two. When the slope of the input-output function is greater than one, the *nonlinearity* is referred to as *expansive*. If the output were proportional to the square-root of the input ($O = cI^{0.5}$), the slope of the function would be 0.5. When the slope of the input-output function is less than one, the nonlinearity is referred to as *compressive*. Examples of input-output functions plotted on log-log axes will be presented later in this chapter, in connection with the response of the basilar membrane (BM) within the cochlea.

III STRUCTURE AND FUNCTION OF THE OUTER AND MIDDLE EAR

Figure 1.2 shows the structure of the peripheral part of the human auditory system. It is composed of three parts, the outer, middle and inner ear. The outer ear includes the pinna and the auditory canal, or meatus. The pinna and meatus together create a broad *resonance* which enhances sound levels at the eardrum, relative to those obtained in the absence of the listener's head, over the frequency range from about 1.5 to 5 kHz.

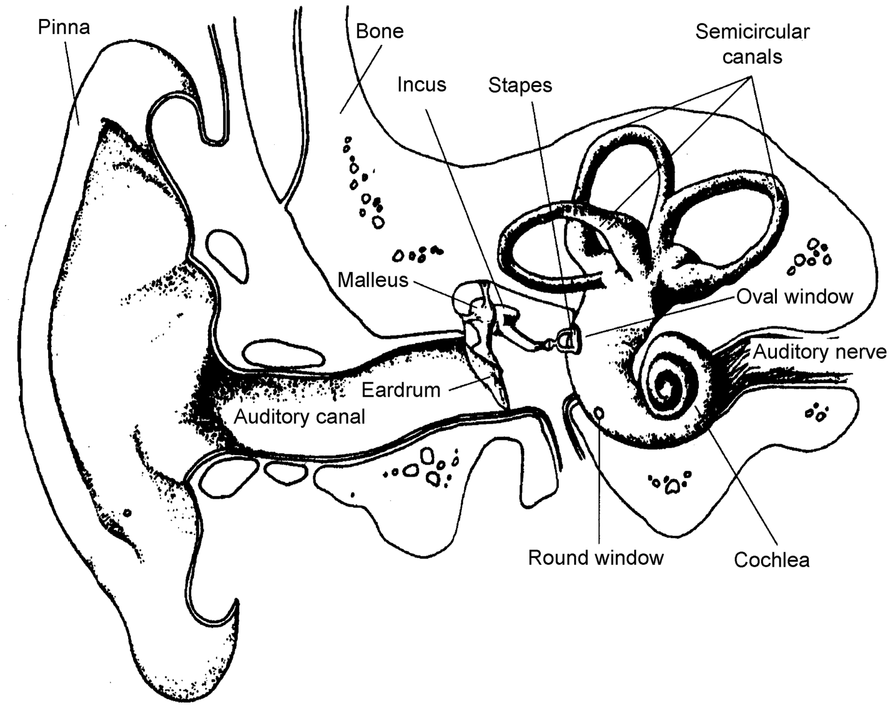


Figure 1.2. Illustration of the structure of the peripheral auditory system showing the outer, middle and inner ear. Redrawn from Lindsay and Norman (1972).

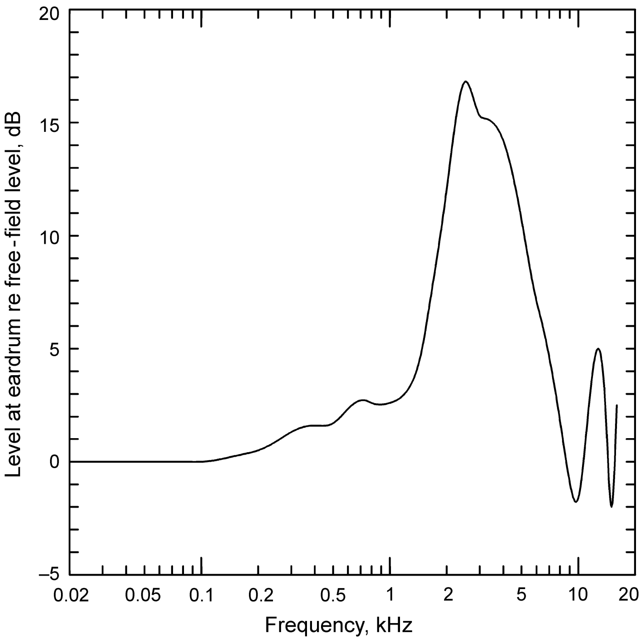


Figure 1.3. The difference between the sound level measured at the eardrum (for a sound coming from the frontal direction) and the sound level in the free field at the point corresponding to the centre of the listener's head. Data from Shaw (1974).

This is illustrated in Figure 1.3. The maximum boost is typically about 12–15 dB in the region around 2.5 kHz. When a hearing aid is fitted so as to block the meatus, this broad resonance is lost for a behind-the-ear aid, reduced in magnitude for a concha aid, and shifted to higher frequencies for a completely-in-the-canal aid. This has to be taken into account when fitting hearing aids and especially when adjusting the gain as a function of frequency.

At medium and high frequencies, the sound reaching the eardrum is significantly modified by the pinna, head and upper torso. Specifically, when the sound contains a broad range of frequencies, the pinna introduces a complex pattern of peaks and notches into the spectrum. This pattern varies systematically with the direction of the sound source relative to the head, and the spectral patterns thus provide important information about the location of sound sources (see Chapter 7 for more details).

Sound travels down the meatus and causes the eardrum, or tympanic membrane, to vibrate. The eardrum forms the outer boundary of the middle ear. These vibrations are transmitted through the middle ear by three small bones, the ossicles, to a membrane-covered opening in the bony wall of the spiral-shaped structure of the inner ear – the cochlea. This opening is called the oval window and it forms the outer boundary of the middle ear. The three bones are called the malleus, incus and stapes (popularly known as the hammer, anvil and stirrup), the stapes being the lightest and smallest of these and the one which actually makes contact with the oval window.

The major function of the middle ear is to ensure the efficient transfer of sound energy from the air to the fluids in the cochlea. If the sound were to impinge directly onto the oval window, most of it would simply be reflected back, rather than entering the cochlea. This happens because the resistance of the oval window to movement is very different from that of air. This is described as a difference in acoustical impedance. The middle ear acts as an impedance-matching device or transformer that improves sound transmission and reduces the amount of reflected sound. This is accomplished mainly by the 27 : 1 ratio of effective areas of the eardrum and the oval window, and to a small extent by the lever action of the ossicles. Transmission of sound energy through the middle ear is most efficient at middle frequencies (500–5000 Hz), which are the ones most important for speech perception; see Chapter 8. This is illustrated in Figure 1.4, adapted from Glasberg and Moore (2006), which shows an estimate of the relative effectiveness of transmission through the middle ear as a function of frequency.

The ossicles have minute muscles attached to them which contract when we are exposed to intense sounds. This contraction, known as the *middle ear reflex* or *acoustic reflex*, is probably mediated by neural centres in the brain stem (Liberman and Guinan, 1998). The reflex can be triggered by sound of any frequency, but it reduces the transmission of sound through the middle ear only at low frequencies (below about 1000 Hz). It may help to prevent damage to the delicate structures inside the cochlea. However, the activation of the reflex is too slow to provide any protection against impulsive sounds, such as gunshots or hammer blows.

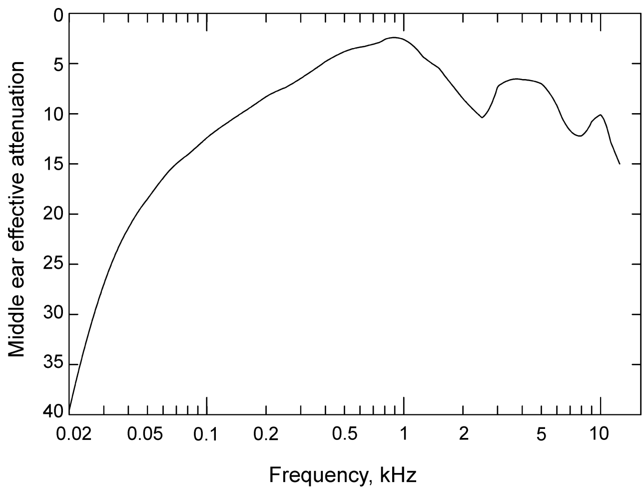


Figure 1.4. The transfer function of the middle ear, plotted as relative response versus frequency. The response was arbitrarily labelled as 2.6 dB at 1 kHz. The estimate comes from Glasberg and Moore (2006).

For moderate sound levels (below about 90 dB SPL), the outer and middle ear behave essentially as linear systems; they do not introduce significant harmonic or intermodulation distortion. However, at high sound levels both the tympanic membrane and the ossicles may vibrate in a nonlinear manner, and the acoustic reflex also introduces nonlinearity. These nonlinearities may result in audible harmonic and intermodulation distortion.

IV STRUCTURE AND FUNCTION OF THE NORMAL COCHLEA

IV.1 THE COCHLEA, THE BASILAR MEMBRANE AND THE ORGAN OF CORTI

The inner ear is also known as the cochlea. It is shaped like the spiral shell of a snail. However, the spiral shape does not appear to have any functional significance, and the cochlea is often described as if the spiral had been ‘unwound’. The cochlea is filled with almost incompressible fluids, and it has bony rigid walls. It is divided along its length by two membranes, Reissner’s membrane and the BM (see Figure 1.5).

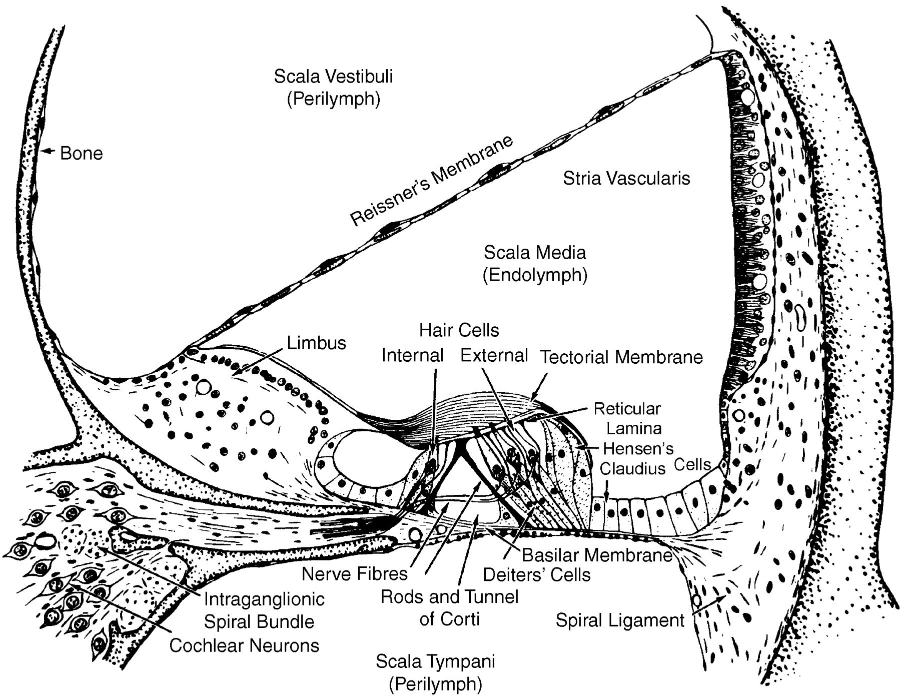


Figure 1.5. Cross-section of the cochlea, showing the BM, Reissner’s membrane and the organ of Corti. Redrawn from Davis (1962).

The start of the spiral, where the oval window is situated, is known as the *base*; the other end, the inner tip, is known as the *apex*. It is also common to talk about the basal end and the apical end. At the apex there is a small opening (the helicotrema) between the BM and the walls of the cochlea which connects the two outer chambers of the cochlea, the *scala vestibuli* and the *scala tympani*. Inward movement of the oval window results in a corresponding outward movement in a membrane covering a second opening in the cochlea – the round window. Such movements result in pressure differences between one side of the BM and the other (i.e. the pressure is applied in a direction perpendicular to the BM), and this results in movement of the BM (see below for details). The helicotrema eliminates any pressure differences between the scala vestibuli and the scala tympani at very low frequencies. This prevents the BM from moving significantly in response to movements of the oval window caused by jaw movements or by slow changes in air pressure (such as occur when changing altitude). The helicotrema also reduces movement of the BM in response to low-frequency sounds.

On the side of the cochlea closest to the outer wall (the right-hand side in Figure 1.5), there is a structure called the stria vascularis. This plays a strong role in the metabolism of the cochlea and in creating the voltages (electrical potentials) that are essential for the normal operation of the cochlea. The stria vascularis is sometimes colloquially described as the 'battery' of the cochlea.

A third membrane, called the *tectorial membrane*, lies above the BM, and also runs along the length of the cochlea. Between the BM and the tectorial membrane are hair cells, which form part of a structure called the *organ of Corti* (see Figures 1.5 and 1.6). They are called hair cells because they appear to have tufts of

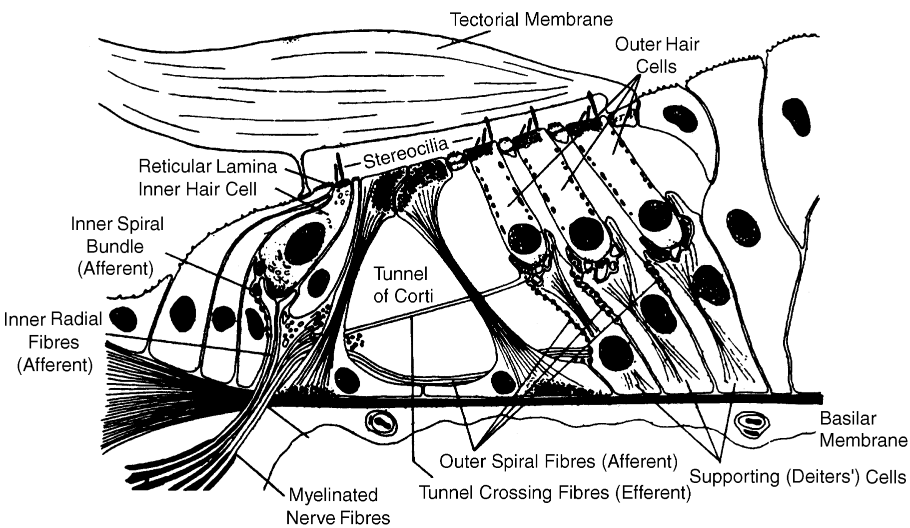


Figure 1.6. Cross section of the organ of Corti as it appears in the basal turn of the cochlea.

hairs, called *stereocilia*, at their apexes. The hair cells are divided into two groups by an arch known as the tunnel of Corti. Those on the side of the arch closest to the outside of the spiral shape are known as *outer hair cells* (OHCs), and they are arranged in three rows in cats and up to five rows in humans, although the rows are often somewhat irregular in humans (Wright *et al.*, 1987). The hair cells on the other side of the arch form a single row, and are known as *inner hair cells* (IHCs). The stereocilia on each OHC form a V- or W-shaped pattern, and they are arranged in rows (usually about three) that are graded in height, the tallest stereocilia lying on the outside of the V or W. The stereocilia on each IHC are also arranged in rows graded in height, but the arrangement is more like a straight line or a broad arc. In humans, there are about 12 000 OHCs (per ear), each with about 140 stereocilia protruding from it, while there are about 3500 IHCs, each with about 40 stereocilia.

The tectorial membrane, which has a gelatinous structure, lies above the hair cells. It appears that the stereocilia of the OHCs actually make contact with the tectorial membrane, but this may not be true for the IHCs. The tectorial membrane appears to be effectively hinged at one side (the left in Figure 1.6). When the BM moves up and down, a shearing motion is created; the tectorial membrane moves sideways (in the left–right direction in Figure 1.6) relative to the tops of the hair cells. As a result, the stereocilia at the tops of the hair cells are moved sideways. The movement occurs via direct contact in the case of the OHCs, but in the case of the IHCs it may be produced by the viscous drag of fluid streaming between the upper part of the organ of Corti and the tectorial membrane. The movement of the stereocilia of the IHCs leads to a flow of electrical current through the IHCs, which in turn leads to the generation of action potentials (nerve *spikes*) in the neurones of the auditory nerve. Thus, the IHCs act to transduce mechanical movements into neural activity.

The IHCs and OHCs have very different functions. The great majority of afferent neurones, which carry information from the cochlea to higher levels of the auditory system, connect to IHCs; each IHC is contacted by about 20 neurones (Spoendlin, 1970). Thus, most information about sounds is conveyed via the IHCs. The main role of the OHCs is actively to influence the mechanics of the cochlea. The OHCs have a motor function, changing their length, shape and stiffness in response to electrical stimulation (Ashmore, 1987; Yates, 1995), and they can therefore influence the response of the BM to sound. The OHCs are often described as being a key element in an *active mechanism* within the cochlea. The function of this active mechanism is described in more detail below.

The action of the OHCs is partly under the control of higher centres of the auditory system. There are about 1800 efferent nerve fibres that carry information from the auditory system to the cochlea, most of them originating in the superior olivary complex of the brain stem. Many of these efferent fibres make contact with the OHCs, and can affect their activity (Liberman and Guinan, 1998). Thus, even the earliest stages in the analysis of auditory signals are partly under the control of higher centres.

IV.2 TUNING ON THE BASILAR MEMBRANE

When the oval window is set in motion by a sound, a pressure difference occurs between the upper and lower surface of the BM. The pressure wave travels almost instantaneously through the incompressible fluids of the cochlea. Consequently, the pressure difference is applied essentially simultaneously along the whole length of the BM. This causes a pattern of motion to develop on the BM. The pattern does not depend on which end of the cochlea is stimulated. Sounds which reach the cochlea via the bones of the head rather than through the air do not produce atypical responses.

The response of the BM to stimulation with a sinusoid takes the form of a travelling wave which moves along the BM from the base towards the apex. The amplitude of the wave increases at first and then decreases rather abruptly. The basic form of the wave is illustrated in Figure 1.7, which shows schematically the instantaneous displacement of the BM for four successive instants in time, in response to a low-frequency sinusoid. The four successive peaks in the wave are labelled 1, 2, 3 and 4. This figure also shows the line joining the amplitude peaks, which is called the *envelope*. The envelope shows a peak at a particular position on the BM.

The response of the BM to sounds of different frequencies is strongly affected by its mechanical properties, which vary progressively from base to apex. At the base the BM is relatively narrow and stiff. This causes the base to respond best to high frequencies. At the apex the BM is wider and much less stiff, which causes the apex to respond best to low frequencies. Each point on the BM is tuned; it responds best (with greatest displacement) to a certain frequency, called the *characteristic frequency* (CF), or best frequency, and responds progressively less as the frequency

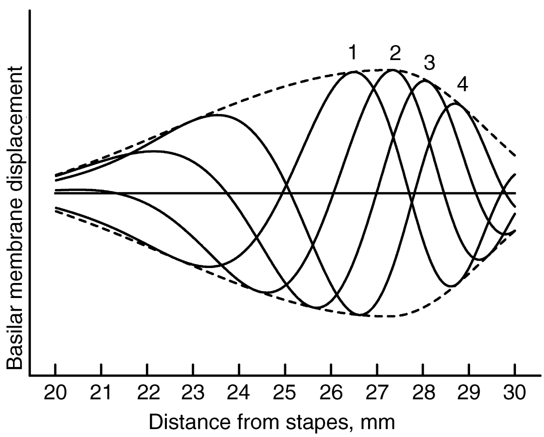


Figure 1.7. The solid lines show the instantaneous displacement of the BM at four successive instants in time (labelled 1–4), derived from a cochlear model. The pattern moves from left to right, building up gradually with distance, and decaying rapidly beyond the point of maximal displacement. The dashed line represents the envelope traced out by the amplitude peaks in the waveform.

is moved away from the CF. It is now believed that the tuning of the BM arises from two mechanisms. One is referred to as the *passive mechanism*. This depends on the mechanical properties of the BM and surrounding structures, and it operates in a roughly linear way. The other is the *active mechanism*. This depends on the operation of the OHCs, and it operates in a nonlinear way. The active mechanism depends on the cochlea being in good physiological condition, and it is easily damaged. The travelling wave shown in Figure 1.7 is typical of what is observed when only the passive mechanism is operating.

Figure 1.8 shows the envelopes of the patterns of vibration for several different low-frequency sinusoids (data from von Békésy, 1960). Sounds of different frequencies produce maximum displacement at different places along the BM, that is, there is a frequency-to-place transformation. If two or more sinusoids with different frequencies are presented simultaneously, each produces maximum displacement at its appropriate place on the BM. In effect, the cochlea behaves like a frequency analyser, although with less than perfect resolution. The resolution is often described in terms of the sharpness of tuning. This refers to the ‘narrowness’ of the response patterns on the BM. In the case of responses to a single tone, as shown in Figure 1.7, it refers to the spread of the response along the BM; sharp tuning would be associated with a narrow spread.

Most of the pioneering work on patterns of vibration along the BM was done by von Békésy (1960). The vibration patterns found by von Békésy were rather broad; for example, the pattern for a 400-Hz sinusoid extended along almost the whole length of the BM (see Figure 1.8). However, these patterns probably reflect only the passive mechanism. The active mechanism would not have been functioning in von Békésy’s experiments, for two reasons. Firstly, he had to use very high sound levels – about 140 dB SPL; such high levels are known to damage the active mechanism. Secondly, he used cadaver ears, and the active mechanism ceases to function after death.

Recent work measuring BM responses to sound differs from that of von Békésy in several ways. Firstly, living animals have been used. Great care has been taken to keep the animals in good physiological condition during the measurements and to minimize trauma caused by the necessary surgery. Secondly, the techniques themselves are designed to be minimally invasive. Finally, rather than measuring the response of several different points on the BM to a single frequency, measurements have usually been made of the responses of a single point to sinusoids of differing frequency. In this case, the sharpness of tuning is often measured by adjusting the *level* at each frequency to produce a fixed response on the BM. If a point on the BM is sharply tuned, then the sound level has to be increased rapidly as the frequency is moved away from the CF. If the tuning is broad, then the sound level has to be increased only gradually as the frequency is moved away from the CF. The results show that the sharpness of tuning of the BM depends critically on the physiological condition of the animal; the better the condition, the sharper is the tuning (Khanna and Leonard, 1982; Sellick, Patuzzi and Johnstone, 1982; Leonard and Khanna, 1984; Robles, Ruggero and Rich, 1986; Ruggero, 1992; Robles and Ruggero, 2001).

The health of the cochlea is often monitored by placing an electrode in or near the auditory nerve, and measuring the combined responses of the neurones to tone bursts

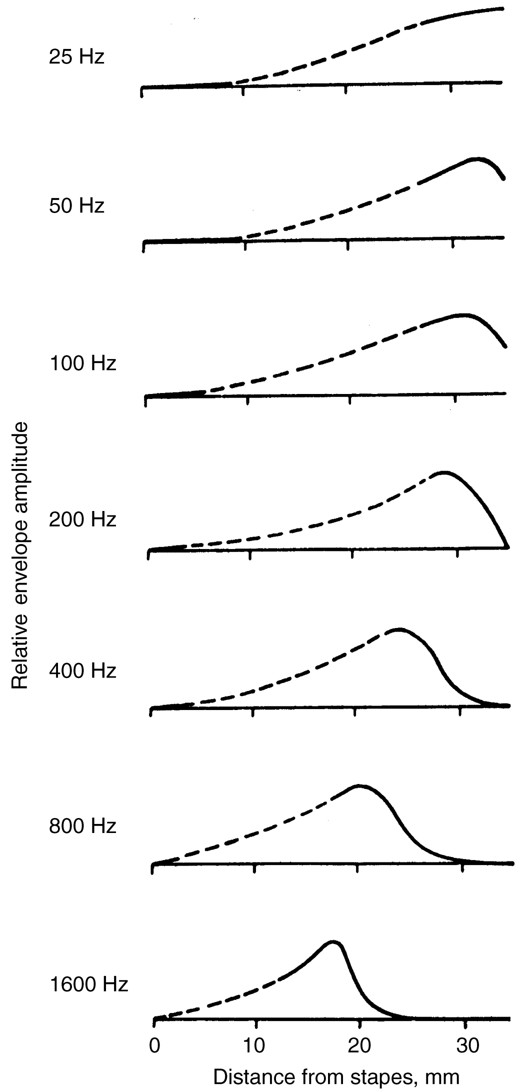


Figure 1.8. Envelopes of patterns of vibration on the BM for a number of low-frequency sounds. Solid lines indicate the results of actual measurements, while the dashed lines are von Békésy's extrapolations. Redrawn from von Békésy (1960).

or clicks; this response is known as the *compound action potential* (AP), or CAP. The lowest sound level at which an AP can be detected is called the AP threshold. Usually, the BM is sharply tuned when the AP threshold is low, indicating that the cochlea is in good physiological condition and the active mechanism is functioning.

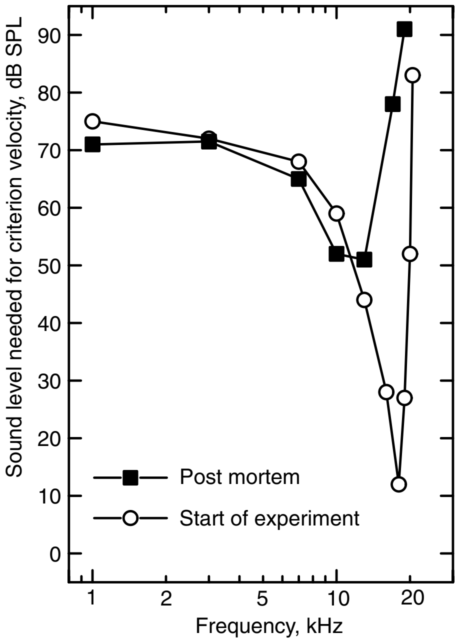


Figure 1.9. Tuning curves measured at a single point on the BM. Each curve shows the input sound level required to produce a constant velocity on the BM, plotted as a function of stimulus frequency. The curve marked by open circles was obtained at the start of the experiment when the animal was in good physiological condition. The curve marked by filled squares was obtained after the death of the animal. Data from Sellick, Patuzzi and Johnstone (1982).

An example is given in Figure 1.9, which shows the input sound level (in dB SPL) required to produce a constant velocity of motion at a particular point on the BM, as a function of stimulus frequency (Sellick, Patuzzi and Johnstone, 1982). This is sometimes called a *constant velocity tuning curve*. It is not yet clear whether the effective stimulus to the IHCs is BM vibration amplitude (equivalent to displacement) or BM velocity. However, for a given frequency, velocity is directly proportional to displacement: the greater the amplitude, the faster the movement. At the start of the experiment, when AP thresholds were low, a very sharp tuning curve was obtained (open circles). This curve reflects the contribution of both the passive and the active mechanisms. As the condition of the animal deteriorated, the active mechanism ceased to function. The tuning became broader, and the sound level required to produce the criterion response increased markedly around the tip. The broad tuning curve recorded after death (filled squares) reflects the tuning produced by the passive mechanism alone.

The frequency at the tip of the tuning curve (the CF – where the sound level was lowest) shifted downwards when the condition of the animal deteriorated. This occurred because, at least for high CFs, the active mechanism gives maximum gain for a frequency that is somewhat above the best frequency determined by the passive mechanism. In a healthy ear, the CF is determined mainly by the active mechanism. When the active mechanism ceases to function, the sharply tuned tip of the tuning curve is lost and the CF shifts to a lower frequency determined by the passive mechanism.

Even in a healthy ear, the balance between the active and passive mechanisms can change with sound level. If the input level of a sinusoid is held constant, and its frequency is varied, then the response of the BM (velocity or amplitude) at a specific point shows a peak for a specific frequency. However, the frequency which gives the maximum response often varies with the input sound level (Ruggero *et al.*, 1997). For high CFs, this frequency decreases with increasing sound level, as the relative contribution of the active mechanism decreases. This implies that, for a fixed input frequency, the place on the BM showing the maximum response shifts towards the base with increasing sound level. Usually, the CF is specified for a low input sound level.

In summary, in a normal healthy ear each point along the BM is sharply tuned, responding with high sensitivity to a limited range of frequencies and requiring higher and higher sound intensities to produce a response as the frequency is moved outside that range. The sharp tuning and high sensitivity reflect the active process mediated by the OHCs.

IV.3 THE NONLINEARITY OF INPUT-OUTPUT FUNCTIONS ON THE BASILAR MEMBRANE

In a normal healthy ear, the response of the BM is nonlinear; when the input magnitude is increased, the magnitude of the response does not grow directly in proportion to the magnitude of the input (Rhode, 1971; Rhode and Robles, 1974; Sellick, Patuzzi and Johnstone, 1982; Robles, Ruggero and Rich, 1986; Ruggero, 1992; Ruggero *et al.*, 1997; Robles and Ruggero, 2001). This is illustrated in Figure 1.10, which shows input-output functions of the BM for a place with a CF of 8 kHz (from Robles, Ruggero and Rich, 1986). A series of curves is shown; each curve represents a particular stimulating frequency, indicated by a number (in kHz) close to the curve. The output (velocity of vibration) is plotted on a logarithmic scale as a function of the input sound level (in dB SPL – also a logarithmic scale). If the responses were linear, the functions would be parallel to the dashed line. Two functions are shown for a CF tone (8 kHz), one (at higher levels) obtained about one hour after the other. The slight shift between the two was probably caused by a deterioration in the condition of the animal. An 'idealized' function for a CF tone, with the ordinate scaled in dB units (i.e. as $20\log_{10}$(velocity)) is shown in Figure 1.11.

While the function for the CF tone is almost linear for very low input sound levels (below 20–30 dB) and approaches linearity at high input sound levels (above 90 dB), the function has a very shallow slope at mid-range levels. This indicates a *compressive nonlinearity*: a large range of input sound levels is compressed into a smaller range of responses on the BM. The form of this function can be explained in the following way. At low and medium sound levels the active mechanism amplifies the response on the BM. The amplification may be 50 dB or more (Robles and Ruggero, 2001).

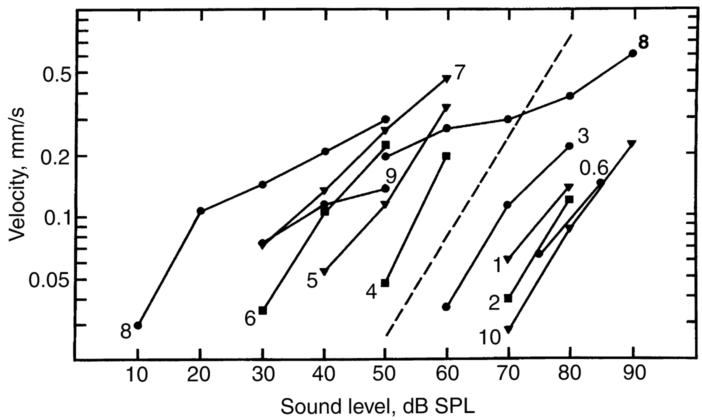


Figure 1.10. Input-output functions for a place on the BM with CF = 8 kHz. The stimulating frequency, in kHz, is indicated by a number close to each curve. The dashed line indicates the slope that would be obtained if the responses were linear (velocity directly proportional to sound pressure). Redrawn from Robles, Ruggero and Rich (1986).

At very low sound levels, below 20–30 dB, the amplification is roughly constant and is at its maximal value. As the sound level increases, the amplification progressively reduces. Thus, the response grows more slowly than it would in a linear system. When the sound level is sufficiently high, around 90 dB SPL, the active mechanism is unable to contribute any amplification, and the response becomes linear (although some researchers have reported a shallow growth of response even at very high sound levels). Hence, at high levels, the 'passive' response becomes dominant.

The nonlinearity mainly occurs when the stimulating frequency is close to the CF of the point on the BM whose response is being measured. For stimuli with

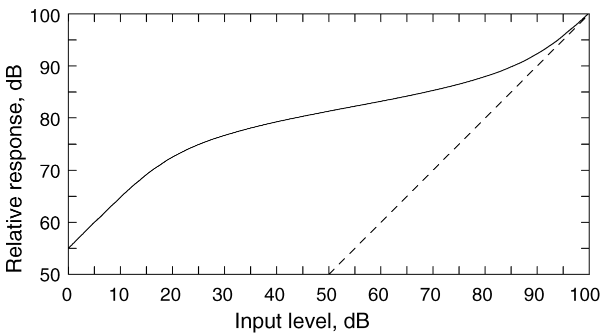


Figure 1.11. Schematic input-output function of the BM for a sinusoid at CF. A decibel scale is used for both axes. The ordinate is scaled (arbitrarily) so that an input of 100 dB gives an output of 100 dB. The dashed line shows the slope that would be obtained if the response were linear.

frequencies well away from the CF, the responses are more linear. Hence, the curves for frequencies of 7 and 9 kHz (close to CF) show shallow slopes, while the curves for frequencies below 7 kHz and above 9 kHz show steeper (linear) slopes. Effectively, the compression occurs only around the peak of the response pattern on the BM. As a result, the peak in the distribution of vibration along the BM flattens out at high sound levels, which partly accounts for the broad tuning observed by von Békésy (1960).

Given that the BM response at moderate sound levels is highly nonlinear, one might expect that, in response to a single sinusoid, harmonics would be generated and the waveform on the BM would be significantly distorted. In fact, this does not seem to happen (Cooper and Rhode, 1992; Ruggero *et al.*, 1997). The reason why is not fully understood. Perhaps the active mechanism involves feedback onto the BM in such a way that potential harmonic distortion is filtered out by the passive mechanism. However, in response to inputs containing more than one sinusoidal component, significant distortion can occur. This is described later on in this chapter.

IV.4 TWO-TONE SUPPRESSION

Experiments using two tones have revealed another aspect of nonlinearity on the BM, namely *two-tone suppression*. The effect is analogous to an effect that was first discovered from measurements of the responses of single neurones in the auditory nerve (see below for details). The response to a tone (called the 'probe' tone) with frequency close to the CF of the place on the BM being studied can be reduced by a second tone with a higher or lower frequency, especially when the second tone is higher in level than the probe tone (Rhode, 1977; Patuzzi, Sellick and Johnstone, 1984; Ruggero, Robles and Rich, 1992). The effect is illustrated in Figure 1.12 for a probe tone at 8.6 kHz (close to the CF) and a suppressor tone at 10.6 kHz. Ruggero, Robles and Rich (1992) provided a detailed comparison of the properties of mechanical two-tone suppression on the BM and two-tone suppression measured in the auditory nerve. They concluded that all of the properties match qualitatively (and mostly quantitatively) and that two-tone suppression in the auditory nerve probably originates from two-tone suppression on the BM.

IV.5 COMBINATION TONE GENERATION

Another aspect of BM nonlinearity is the generation of distortion products in response to two or more sinusoidal inputs. These products are often called *combination tones*. When two sinusoids are presented simultaneously, and their frequency separation is not too great, their response patterns overlap on the BM. It appears that, at the point of overlap, distortion products are generated that behave like additional sinusoidal tones. For example, if the two primary tones presented to the ear have frequencies f_1 and f_2 ($f_2 > f_1$), the distortion products have frequencies such as $2f_1 - f_2$ and $f_2 - f_1$. The distortion products appear to propagate along the BM to the locations tuned to their own frequencies (Robles, Ruggero and Rich, 1991). For example, the combination

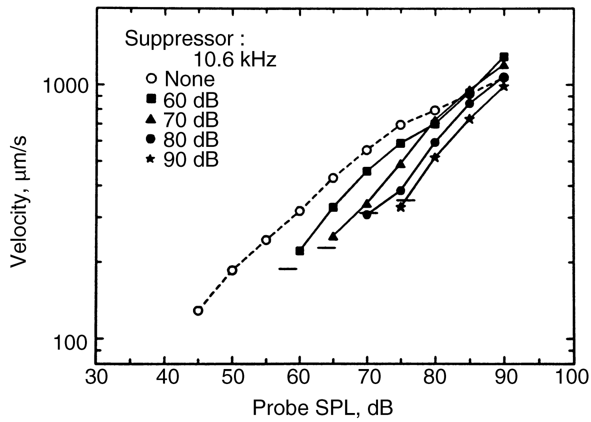


Figure 1.12. An example of two-tone suppression for a place on the BM with CF = 8.6 kHz. The dashed curve with open circles shows an input-output function for an 8.6-kHz tone alone, referred to as the probe. The solid curves show input-output functions when a suppressor tone was added to the probe. The suppressor was presented at each of several overall levels, from 60 to 90 dB SPL, as indicated by the key in the figure. The solid curves are truncated at the level where the suppressor led to an increase in response rather than a decrease. Redrawn from Ruggero, Robles and Rich (1992).

tone with frequency $2f_1 - f_2$ produces a local maximum on the BM at the place tuned to $2f_1 - f_2$.

Human listeners can sometimes hear these additional tones. The one with frequency $2f_1 - f_2$ is especially easy to hear. It is audible even for relatively low levels of the primary tones when f_2 is about 1.2 times f_1 (Smoorenburg, 1972a, 1972b). For example, if the two primary tones have frequencies 1000 and 1200 Hz, then a tone is heard with frequency 800 Hz. The combination tone with frequency $2f_1 + f_2$ is much harder to hear. This may be the case because this tone has a higher frequency than the two primary tones. Although $2f_1 + f_2$ is probably generated at the point where the response patterns of f_1 and f_2 overlap, it does not propagate to the location on the BM tuned to $2f_1 + f_2$; this would involve propagation in the 'wrong' direction, from the apex towards the base.

IV.6 RESPONSES OF THE BASILAR MEMBRANE TO COMPLEX SOUNDS

Consider the response of the BM to two sinusoids, of different frequencies, presented simultaneously. Assume that the sinusoids are equal in level, so that two-tone suppression is small. The kind of pattern of vibration that occurs depends on the frequency separation of the two sinusoids. If this is very large, then the two sinusoids produce two, effectively separate, patterns of vibration on the BM. Each produces a maximum at the place on the BM which would have been excited most had that component been presented alone. Thus, the response of the BM to a low-frequency sinusoid of moderate *intensity* is essentially unaffected by a high-frequency sinusoid, and vice versa. In this case, the BM behaves like a frequency analyser, breaking down the complex sound into its sinusoidal components. Correspondingly, when we listen to two sinusoids with widely spaced frequencies, we hear two separate tones, with two different pitches. When the two sinusoids are relatively close in frequency, however, the patterns of vibration on the BM interact, so that some points on the BM respond to both of the sinusoids. At those points, the displacement of the BM as a function of time is not sinusoidal but is a complex waveform resulting from the interference of the two sinusoids. When the two sinusoids are sufficiently close in frequency, there is no longer a separate maximum in the pattern of vibration for each of the component sinusoids; instead there is a single, broader, maximum. Thus, the BM has failed to separate (resolve) the individual frequency components. Correspondingly, when two sinusoids are very closely spaced in frequency, we cannot hear two separate tones, each with its own *pitch*; rather, we hear a single sound corresponding to the mixture. This is described more fully in Chapter 3, Section VIII.

Consider now the more complex case of the pattern of responses on the BM to a periodic complex tone, such as a voiced vowel or an instrument playing a note. Such a tone typically contains many harmonics of a common *fundamental frequency*. For example, the note A4 would have a fundamental component with a frequency of 440 Hz, and higher harmonics with frequencies of 880, 1320, 1760 ... Hz. The harmonics are equally spaced on a linear frequency scale. However, the mapping of CF to distance along the BM roughly follows a *logarithmic* scale. For example, sinusoids with frequencies of 400, 800, 1600 and 3200 Hz would produce peaks that were roughly equally spaced along the BM (see Figure 1.8 above). When a harmonic complex tone is presented to the ear, the lower harmonics each give rise to a separate peak on the BM, while the higher harmonics give responses that overlap, so that there are not distinct peaks corresponding to the individual harmonics. A perceptual consequence of this is that individual low harmonics can often be 'heard out' as separate tones, while higher harmonics cannot be individually heard; this is described more fully in Chapter 3, Section VII. These factors play a crucial role in the perception of complex tones, as is explained in Chapter 6, Section II.

IV.7 OTOACOUSTIC EMISSIONS

Evidence supporting the idea that there are active biological processes influencing cochlear mechanics has come from a remarkable phenomenon first reported by Kemp (1978), although predicted by Gold (1948). If a low-level click is applied to the ear, then it is possible to detect sound being reflected from the ear, using a microphone sealed into the ear canal. The early part of this reflected sound appears to come from the middle ear, but some sound can be detected for delays from 5 to 60 ms following the instant of click presentation. These delays are far too long to be attributed to the middle ear, and they almost certainly result from activity in the cochlea itself. The reflected sounds are known as *evoked otoacoustic emissions*. They have also been called *Kemp echoes* and *cochlear echoes*.

Although the input click in Kemp's experiment contained energy over a wide range of frequencies, only certain frequencies were present in the reflected sound. Kemp suggested that the reflections are generated at points on the BM, or in the IHC/OHC transduction mechanism, where there is a gradient or discontinuity in the mechanical or electrical properties. The response is nonlinear, in that the reflected sound does not have an intensity in direct proportion to the input intensity. In fact, the relative level of the reflection is greatest at low sound levels; the emission grows about 3 dB for each 10 dB increase in input level. This nonlinear behaviour can be used to distinguish the response arising from the cochlea from the linear middle ear response. Sometimes the amount of energy reflected from the cochlea at a given frequency may exceed that which was present in the input sound (Burns, Keefe and Ling, 1998). Indeed, many ears emit sounds in the absence of any input, and these can be detected in the ear canal (Zurek, 1981). Such sounds are called *spontaneous otoacoustic emissions*, and their existence indicates that there is a source of energy within the cochlea which is capable of generating sounds. Kemp (2002) and others have suggested that the emissions are a by-product of the active mechanism.

Cochlear emissions can be very stable in a given individual, both in waveform and frequency content, but each ear gives its own characteristic response. Responses tend to be strongest between 500 and 2500 Hz, probably because transmission from the cochlea back through the middle ear is most efficient in this range, as described earlier. Cochlear emissions can be measured for brief tone bursts as well as clicks, and it is even possible to detect a reflected component in response to continuous stimulation with a pure tone.

When the ear is stimulated with two tones, an emission may be detected at the frequency of one or more combination tones, particularly $2f_1 - f_2$. Such emissions are called *distortion-product otoacoustic emissions*. This confirms that the combination tone is present as a mechanical disturbance in the cochlea, as a travelling wave on the BM.

Sometimes the transient stimulation used to evoke a cochlear echo induces a sustained oscillation at a particular frequency, and the subject may report hearing this oscillation as a tonal sensation. The phenomenon of hearing sound in the absence of external stimulation is known as *tinnitus*. It appears that tinnitus may arise from abnormal activity at several different points in the auditory system, but in a few cases it corresponds to mechanical activity in the cochlea.

In summary, several types of otoacoustic emissions can be identified, including evoked emissions, spontaneous emissions and distortion-product emissions. While the exact mechanism by which otoacoustic emissions are generated is not understood, there is agreement that it is connected with the active mechanism in the cochlea.

V NEURAL RESPONSES IN THE NORMAL AUDITORY NERVE

Most studies of activity in the auditory nerve have used electrodes with very fine tips, known as microelectrodes. These record the nerve impulses, or spikes, in single auditory nerve fibres (often called single units). The main findings, summarized below, seem to hold for most mammals.

V.1 SPONTANEOUS FIRING RATES AND THRESHOLDS

Most neurones show a certain baseline firing rate, called the *spontaneous rate*, in the absence of any external stimulus. Liberman (1978) presented evidence that auditory nerve fibres could be classified into three groups on the basis of their spontaneous rates. About 61 % of fibres have high spontaneous rates (18–250 spikes per second), 23 % have medium rates (0.5–18 spikes per second) and 16 % have low spontaneous rates (less than 0.5 spikes per second). The spontaneous rates are correlated with the position and size of the synapses of the neurones on the IHCs. High spontaneous rates are associated with large synapses, primarily located on the side of the IHCs facing the OHCs. Low spontaneous rates are associated with smaller synapses on the opposite side of the IHCs. The spontaneous rates are also correlated with the thresholds of the neurones. The threshold is the lowest sound level at which a change in response of the neurone can be measured. High spontaneous rates tend to be associated with low thresholds and vice versa. The most sensitive neurones may have thresholds close to 0 dB SPL, whereas the least sensitive neurones may have thresholds of 80 dB SPL or more.

V.2 TUNING CURVES AND ISO-RATE CONTOURS

The tuning of a single nerve fibre is often illustrated by plotting the fibre's threshold as a function of frequency. This curve is known as the *tuning curve* or frequency-threshold curve (FTC). The stimuli are usually tone bursts, rather than continuous tones. This avoids the effects of long-term adaptation (a decrease in response over time that can occur with continuous stimulation), and also makes it easier to distinguish spontaneous from evoked neural activity. The frequency at which the threshold of the fibre is lowest is called the characteristic frequency (CF) (the same term is used to describe the frequency to which a given place on the BM is most sensitive). Some typical tuning curves are presented in Figure 1.13. On the logarithmic frequency scale used, the tuning curves are usually steeper on the high-frequency side than on the low-frequency side. It is generally assumed that the tuning seen in each single auditory nerve fibre occurs because that fibre responds to activity in a single IHC at a particular point on the BM. Iso-velocity tuning curves on the BM are similar in shape to neural FTCs (Khanna and Leonard, 1982; Sellick, Patuzzi and Johnstone, 1982; Robles, Ruggero and Rich, 1986; Ruggero *et al.*, 1997).

The CFs of single neurones are distributed in an orderly manner in the auditory nerve. Fibres with high CFs are found in the periphery of the nerve bundle, and there is an orderly decrease in CF towards the centre of the nerve bundle (Kiang *et al.*, 1965). This kind of arrangement is known as *tonotopic organization* and it indicates that the place representation of frequency along the BM is preserved as a place representation in the auditory nerve.

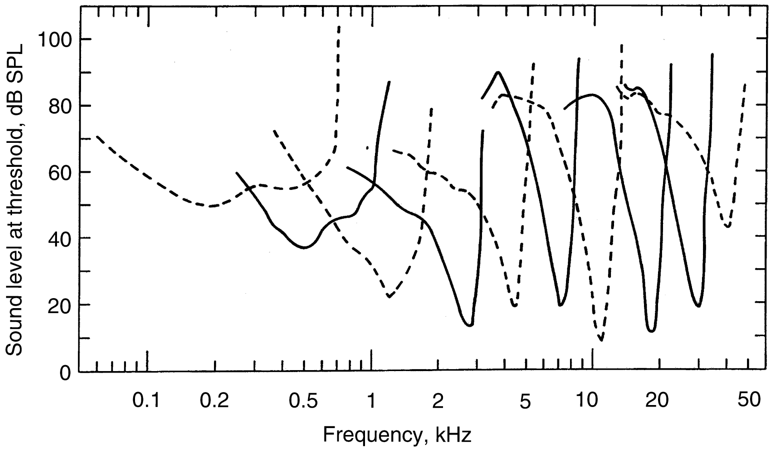


Figure 1.13. A sample of tuning curves (also called frequency-threshold curves) obtained from single neurones in the auditory nerve of anaesthetized cats. Each curve shows results for one neurone. The sound level required for threshold is plotted as a function of the stimulus frequency (logarithmic scale). Redrawn from Palmer (1987).

In order to provide a description of the characteristics of single fibres at levels above threshold, iso-rate contours can be plotted. To determine an iso-rate contour, the intensity of sinusoidal stimulation required to produce a predetermined firing rate in the neurone is plotted as a function of frequency. The resulting curves are generally similar in shape to tuning curves, although they sometimes broaden at high sound levels. Also, for high CFs, the frequency at the tip (the lowest point on the curve) may decrease slightly with increases in the predetermined firing rate. This reflects the change in BM tuning with level described earlier.

V.3 RATE-VERSUS-LEVEL FUNCTIONS

Figure 1.14 shows schematically how the rate of discharge for three auditory nerve fibres changes as a function of stimulus level. The curves are called *rate-versus-level functions*. In each case, the stimulus was a sinusoid at the CF of the neurone. Consider first the curve labelled (a). This curve is typical of what is observed for neurones with high spontaneous firing rates. Above a certain sound level the neurone no longer responds to increases in sound level with an increase in firing rate; the neurone is said to be saturated. The range of sound levels between threshold and the level at which saturation occurs is called the *dynamic range*. For neurones with high spontaneous rates, this range is often quite small, about 15–30 dB. Curve (b) is typical of what is observed for neurones with medium spontaneous rates. The threshold is slightly higher than for (a) and the dynamic range is slightly wider. Curve (c) is typical of what is observed for neurones with low spontaneous rates. The threshold is higher than for (b). The firing rate at first increases fairly rapidly with the increasing sound level, but

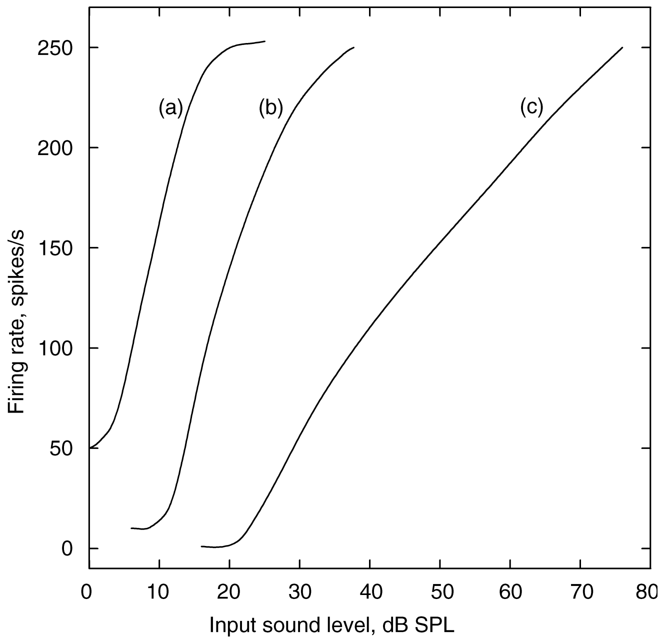


Figure 1.14. Schematic examples of how the discharge rates of single auditory neurones vary as a function of stimulus level. The curves are called *rate-versus-level functions*. In each case, the stimulus was a sinusoid at the CF of the neurone. Curves (a), (b) and (c) are typical of what is observed for neurones with high, medium and low spontaneous firing rates, respectively.

then the rate of increase slows down. The firing rate continues to increase gradually with increasing sound level over a wide range of levels. This has been called sloping saturation (Sachs and Abbas, 1974; Winter, Robertson and Yates, 1990).

The shapes of rate-versus-level functions can be understood in terms of two functions (Yates, 1990; Patuzzi, 1992). This is illustrated in Figure 1.15. The first function is the input-output function of the BM, illustrated schematically in the top-right panel. The second is the function relating the spike rate in a specific neurone to the magnitude of the BM response. This second function is similar in form for different neurones, showing saturation when the BM amplitude is a certain factor above the value required for threshold, but it varies in the magnitude required for threshold. Three such functions are illustrated schematically in the top-left panel of Figure 1.15. The rate-versus-level functions corresponding to these three functions are shown in the bottom-right panel.

The variation across neurones depends mainly on the type of synapse, as discussed earlier. Neurones with low thresholds have large sensitive synapses. They start to respond at very low sound levels, where the input-output function on the BM is nearly linear. As the sound level increases, the BM displacement increases

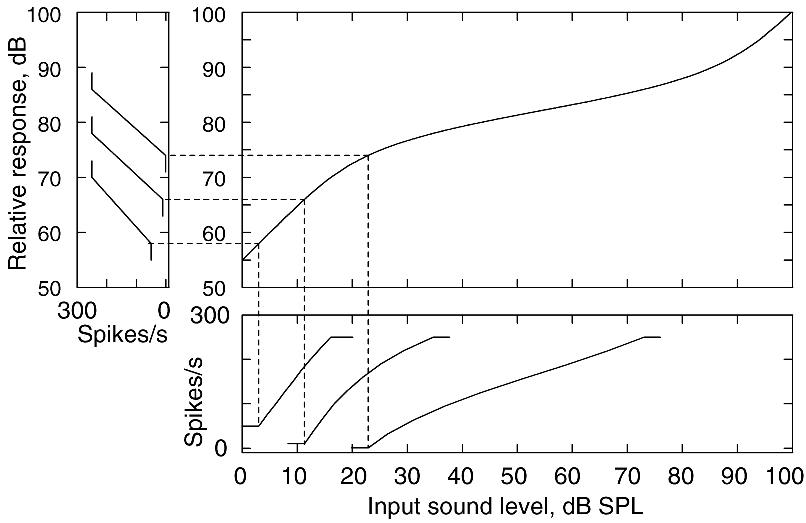


Figure 1.15. Schematic illustration of how the shapes of rate-versus-level functions can be accounted for in terms of the BM input-output function (top-right panel), and the functions relating neural firing rate (APs/sec) to amplitude on vibration on the BM (top-left panel). Three such functions are shown, corresponding to synapses with different sensitivities. The resulting three rate-versus-level functions are shown in the bottom panel. Adapted from Patuzzi (1992).

in a nearly linear manner, and the neurone saturates relatively early, giving a small dynamic range, as shown by the left-most curve in the lower panel. Neurones with higher thresholds have less sensitive synapses. They respond over the range of sound levels where the BM input-output function shows a strong compressive nonlinearity. Hence, a large increase in sound level is needed to increase the BM displacement to the point where the neurone saturates, and the neurone has a wide dynamic range, as shown by the right-most curve in the lower panel.

V.4 TWO-TONE SUPPRESSION

Auditory neurones show an effect that is exactly analogous to the two-tone suppression on the BM that was described earlier. Indeed, the neural effect was discovered long before the BM effect. The tone-driven activity of a single fibre in response to one tone can be suppressed by the presence of a second tone. This was originally called two-tone inhibition (Sachs and Kiang, 1968), although the term *two-tone suppression* is now generally preferred, since the effect does not appear to involve neural inhibition. Typically the phenomenon is investigated by presenting a tone at, or close to, the CF of a neurone. A second tone is then presented, its frequency and intensity are varied and the effects of this on the response of the neurone are noted. When the frequency and

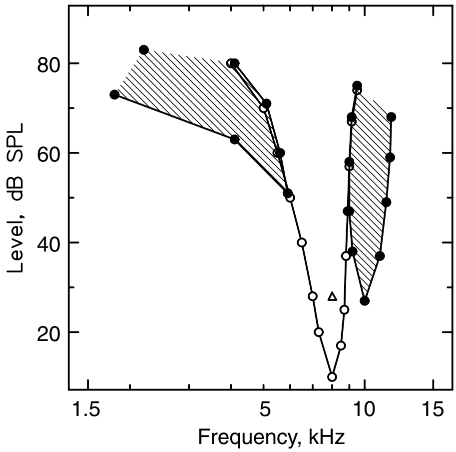


Figure 1.16. Neurophysiological data from Arthur, Pfeiffer and Suga (1971). The open circles show the tuning curve (threshold versus frequency) of a single neurone with a CF at 8 kHz. The neurone was stimulated with a tone at CF, and just above threshold (indicated by the open triangle). A second tone was then added and its frequency and intensity were varied. Any tone within the shaded areas bounded by the solid circles reduced the response to the tone at CF by 20 % or more. These are the suppression areas.

intensity of the second tone fall within the excitatory area bounded by the tuning curve, this usually produces an increase in firing rate. However, when they fall just outside that area, the response to the first tone is reduced or suppressed. The suppression is greatest when the suppressor falls in one of two frequency regions on either side of the excitatory response area, as illustrated in Figure 1.16. The suppression begins very quickly when the suppressor is turned on, and ceases very quickly when it is turned off (Arthur, Pfeiffer and Suga, 1971). This is consistent with the likely origin of the suppression as a mechanical effect on the BM.

V.5 PHASE LOCKING

In response to a sinusoid with a frequency below about 5 kHz, the nerve firings tend to be *phase locked*, or synchronized, to the evoked waveform on the BM. A given nerve fibre does not necessarily fire on every cycle of the stimulus but, when firings do occur, they occur at roughly the same phase of the waveform each time. Thus, the time intervals between firings are (approximately) integer multiples of the period of the stimulating waveform. For example, a 500-Hz tone has a period of 2 ms; the waveform repeats regularly every 2 ms. The intervals between nerve firings in response to a 500-Hz tone are approximately 2, or 4, or 6, or 8 ms, and so on. Neurones do not fire in a completely regular manner, so that there are not exactly 500, or 250 or 125 spikes/s. However, information about the period of the stimulating

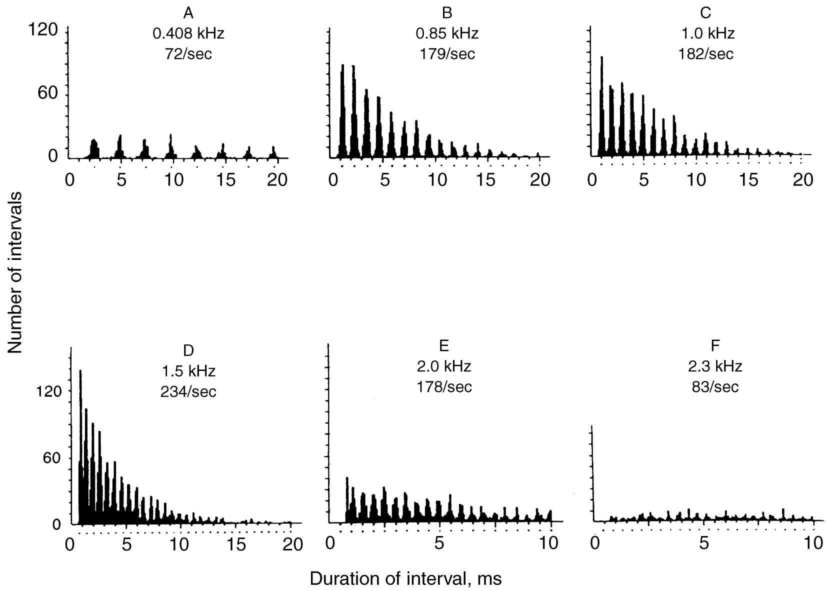


Figure 1.17. Interspike interval histograms for a single auditory neurone (in the squirrel monkey) with a CF of 1.6 kHz. The frequency of the sinusoidal input and the mean response rate in spikes per second are indicated above each histogram. All tones had a level 80 dB SPL. Notice that the time scales in E and F differ from those in A to D. Redrawn from Rose *et al.* (1968).

waveform is carried unambiguously in the temporal pattern of firing of a single neurone.

One way to demonstrate phase locking in a single auditory nerve fibre is to plot a histogram of the time intervals between successive nerve firings. Several such *interspike interval histograms* for a neurone with a CF of 1.6 kHz are shown in Figure 1.17. For each of the different stimulating frequencies (from 0.408 to 2.3 kHz in this case), the intervals between nerve spikes lie predominantly at integer multiples of the period of the stimulating tone. These intervals are indicated by dots below each abscissa. Thus, although the neurone does not fire on every cycle of the stimulus, the distribution of time intervals between nerve firings depends closely on the frequency of the stimulating waveform.

Phase locking does not occur over the whole range of audible frequencies. In most mammals it becomes progressively less precise for stimulus frequencies above 1 kHz, and it disappears completely at about 4–5 kHz (Rose *et al.*, 1968), although the exact upper limit varies somewhat across species (Palmer and Russell, 1986). Phase locking improves in precision with increasing sound level at low levels, and then stays roughly constant in precision over a very wide range of sound levels.

Information from phase locking contributes to the ability to localize sounds in space, and it probably also plays a role in the perception of pitch and in the perception of speech when background sounds are present. These aspects of auditory perception are discussed more fully in Chapters 5, 7 and 8.

VI TYPES OF HEARING LOSS

A *conductive hearing loss* is caused by a reduced efficiency of sound transmission through the outer and/or middle ear. This may be caused by wax (cerumen) in the ear canal, damage to the eardrum produced by infection or trauma, damage to or stiffening of the ossicles in the middle ear, or fluid in the middle ear caused by infection. It results in an attenuation of sound reaching the cochlea, so that sounds appear quieter than normal. The amount of loss may vary with frequency, so sounds may appear to have a somewhat different tonal quality from normal. However, these are the main perceptual consequences of a conductive loss; unlike cochlear hearing loss, it does not generally result in marked distortions or abnormalities in other aspects of sound perception. Conductive hearing loss can often be treated by drugs (to cure infections) or surgery.

Cochlear hearing loss involves damage to the structures inside the cochlea. It can arise in many ways, for example by exposure to intense sounds or to ototoxic chemicals (such as certain antibiotics, drugs used to treat high blood pressure or solvents), by infection, by metabolic disturbances, by some forms of allergies, by autoimmune disorders and as a result of genetic factors. These agents can produce a variety of types of damage to the cochlea, and, to complicate matters further, the damage may extend beyond the cochlea. For example, an infection may produce damage at several sites, such as the auditory nerve and higher centres in the auditory pathway. When both cochlear and neural structures are involved, the more general term *sensorineural hearing loss* is used.

Finally, hearing loss can occur through damage to structures or neural systems occurring at a level in the auditory system beyond the cochlea, for example in the auditory nerve or the auditory cortex. Such types of hearing loss are given the general name *retrocochlear hearing loss*. A relatively common cause of retrocochlear loss is the growth of a benign tumour (often called an acoustic neuroma or vestibular schwannoma) which presses on the auditory nerve.

Although sensorineural hearing loss can involve structures other than the cochlea, it is common for the most serious damage to occur within the cochlea. This is probably true for the majority of cases of *presbyacusis*, the hearing loss that is associated with ageing. Furthermore, it is common for the damage to be largely confined to certain specific structures within the cochlea. This book is concerned with cases where the hearing loss arises primarily from damage to the cochlea. Henceforth, when the phrase ‘hearing impairment’ is used, it should be taken to imply hearing impairment caused by cochlear damage. However, it should be borne in mind that, while it is relatively easy to produce ‘pure’ cochlear damage in animal models, such ‘pure’ damage may be relatively rare in hearing-impaired humans.

In the last twenty years there has been a considerable increase in understanding of the physiology and function of the normal cochlea. Along with this has come an improved understanding of the changes in function that are associated with cochlear hearing loss. The rest of this chapter reviews the structure and function of the impaired cochlea.

VII PHYSIOLOGY OF THE DAMAGED COCHLEA

There is strong evidence that the functioning of the normal cochlea depends upon the operation of an active mechanism that is linked to the integrity of the OHCs. This mechanism may involve feedback of energy onto the BM, via the OHCs, and it plays an important role in producing the high sensitivity of the BM to weak sounds and the sharp tuning on the BM. The normal BM has a strongly nonlinear response, which results in compressive input-output functions, two-tone suppression and combination-tone generation.

Cochlear hearing loss often involves damage to the OHCs and IHCs; the stereocilia may be distorted or destroyed, or entire hair cells may die. The OHCs are generally more vulnerable to damage than the IHCs. Some examples of OHC damage are shown in Figure 1.18. When OHCs are damaged, the active mechanism tends to be reduced in effectiveness or lost altogether. The function of the OHCs can also be adversely affected by malfunctioning of the stria vascularis (Schmiedt, 1996). As a result, several changes occur: the sensitivity to weak sounds is reduced; so sounds need to be more intense to produce a given magnitude of response on the BM, the tuning curves on the BM become much more broadly tuned and all of the frequency-selective nonlinear effects weaken or disappear altogether.

VII.1 BASILAR MEMBRANE RESPONSES

There have been many studies showing that the responses of the BM are highly physiologically vulnerable. One example has already been given; see Figure 1.9. Generally, the tuning on the BM becomes less sharp, and the sensitivity around the tip is reduced, when the cochlea is damaged. In the great majority of studies, the changes in BM responses have been associated with some form of damage to the OHCs, either directly or via some form of metabolic disturbance.

The effects of cochlear damage on the input-output functions of the BM of a chinchilla are illustrated in Figure 1.19 (Ruggero and Rich, 1991). The solid curve with black squares, labelled ‘Before’, shows the input-output function obtained when the cochlea was in good condition; the stimulus was a sinusoid with frequency corresponding to the CF of 9000 Hz. The curve shows a compressive nonlinearity for input sound levels between about 30 and 90 dB SPL. In contrast, the response to a sinusoid with a frequency of 1000 Hz, well below the CF, is steeper and is almost linear (solid curve with open circles).

To manipulate the functioning of the cochlea, the animal was injected with furosemide (also known as frusemide), a diuretic that is known to disrupt hair cell potentials, and hence disrupt the function of the hair cells. The dashed curves in

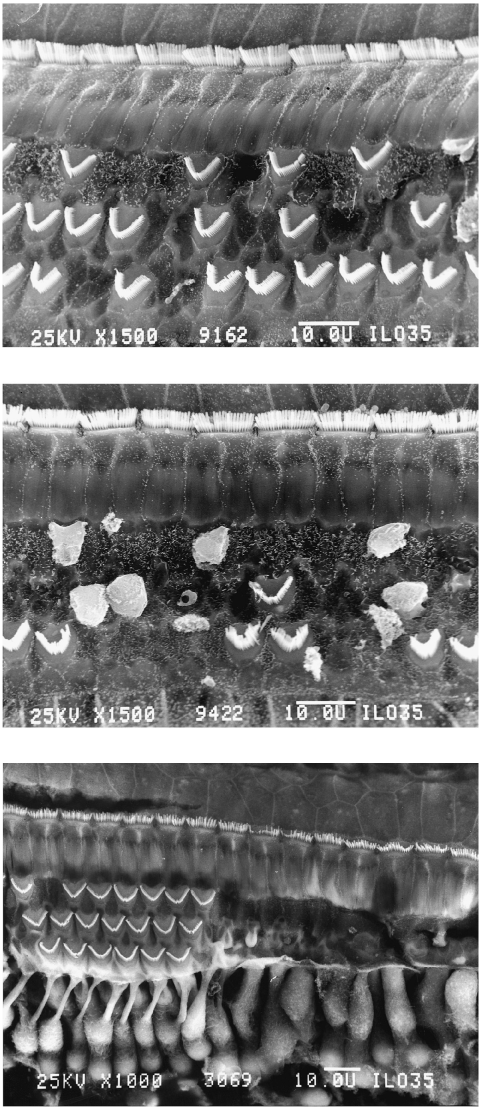


Figure 1.18. Examples of patterns of damage to the OHCs. For the top two panels, the damage was caused by the administration of aminoglycosides. For the bottom panel, the damage was of unknown origin. In the upper panel, the stereocilia of some OHCs are missing. In the middle panel, the stereocilia of the OHCs are mostly either missing or grossly deformed. In the bottom panel, the stereocilia of the OHCs are completely missing over a certain region. The bottom panel also shows expansion of the supporting cells to replace lost hair cells. The electron micrographs were supplied by Dr Andrew Forge of the Institute of Laryngology and Otology, University College London Medical School.

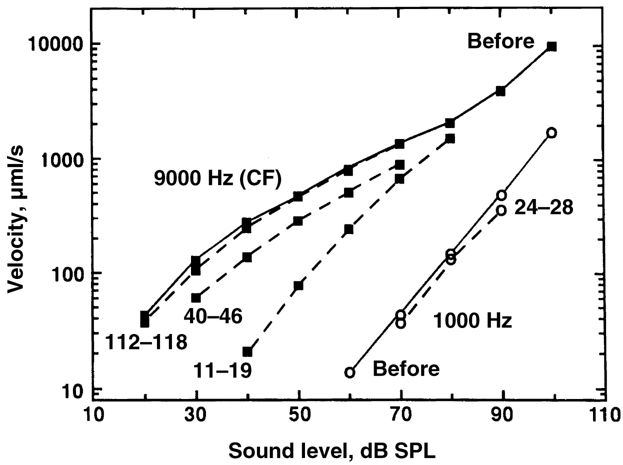


Figure 1.19. Input-output functions on the BM immediately preceding (solid lines) and following (broken lines) an intravenous furosemide injection. See text for details. Redrawn from Ruggero and Rich (1991).

Figure 1.19 were obtained at various times after injection of the drug; the time is indicated by a range in minutes next to each curve. Shortly after the injection (11–19 minutes), the input-output function for the CF tone was markedly altered. The biggest alteration was at low sound levels. To produce a given response on the BM (say, 40 µm/s), the input level had to be increased by about 25 dB relative to the level measured before the injection. However, the response to a CF tone at a high level (80 dB SPL) was almost normal. This is consistent with the idea that the contribution of the active mechanism reduces progressively as the sound level is increased above about 40 dB SPL. After a sufficiently long time (112–118 minutes), the input-output function returned to normal. In this case, the cochlear damage was reversible. Larger doses of the drug, or treatment with other drugs, can result in permanent cochlear damage.

Note that the injection of the drug did not change the input-output function for the 1000-Hz tone (see the curve with open symbols labelled 24–28). This is consistent with the idea that the active mechanism mainly influences responses around the peak of the response pattern evoked by a sinusoidal tone. Responses to tones with frequencies well away from the CF are linear, and remain so when the active mechanism is damaged.

VII.2 NEURAL RESPONSES

Some of the first evidence for a physiologically vulnerable active mechanism came from studies of the responses of single neurones in the auditory nerve. Robertson and Manley (1974) showed that the normal, sharp tuning seen in auditory neurones could be altered by reducing the oxygen supply to the animal. The tuning curves became less sharp, and at the same time the sensitivity around the tip decreased. These changes were similar to those found subsequently in BM responses with similar manipulations (see Figure 1.9 above). The changes in BM tuning and sensitivity found by Robertson and Manley were reversible. Similar effects were reported by Evans (1975), who also found that a reversible degradation in tuning could be produced by the ototoxic agents cyanide and furosemide. Evans and Harrison (1976) used the drug kanamycin to produce selective damage to the OHCs. They found that the threshold and tuning properties of auditory nerve fibres were dependent on the integrity of the OHCs.

VII.3 STRUCTURE–FUNCTION CORRELATION

Liberman and his colleagues used noise exposure and ototoxic drugs, separately or in combination, to produce a variety of types of damage to the hair cells in the cochlea. They then measured neural responses and compared them with structural changes in the cochlea (for a review, see Liberman, Dodds and Learson, 1986). After studying the properties of a given single neurone, the neurone was injected with horseradish peroxidase. This labelled the neurone so that it could be traced to the IHC with which it synapsed in the organ of Corti. In this way, the neural response properties could be directly compared with the structural changes in the hair cells and the immediately surrounding structures in the organ of Corti.

Figure 1.20 shows the situation when there is partial destruction of the OHCs, with intact IHCs. This pattern of damage is typical of that associated with moderate doses of ototoxic drugs, but is less typical of noise exposure. The left part of the figure schematically illustrates the pattern of structural damage. The view is looking down onto the tops of the hair cells; each stereocilium appears as a small dot. The three

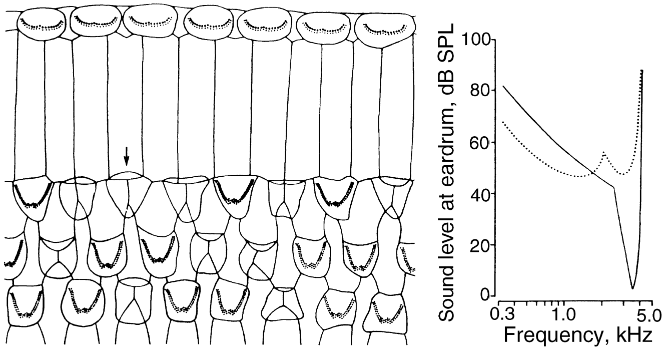


Figure 1.20. The left part shows a schematic diagram of an organ of Corti with sub-total loss of OHCs and intact IHCs. The right part shows a normal neural tuning curve (solid) and an abnormal tuning curve (dotted) associated with this kind of damage. Adapted from Liberman, Dodds and Learson (1986).

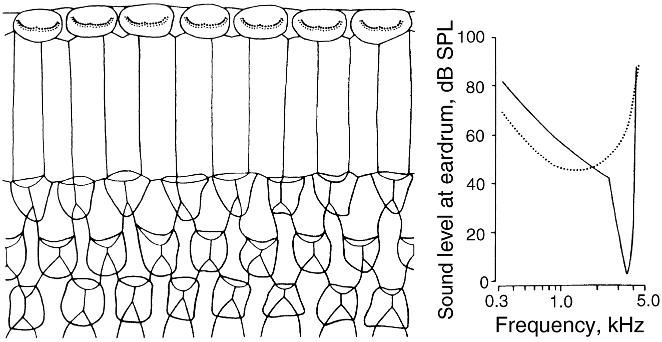


Figure 1.21. As Figure 1.20, but for the situation with total loss of OHCs with intact IHCs.

rows of OHCs appear at the bottom. In the figure, some OHCs are missing or lack stereocilia (see arrow).

The right-hand part of Figure 1.20 shows a typical normal tuning curve (solid curve) and a tuning curve associated with this type of damage (dotted curve). The normal tuning curve shows a sharply tuned 'tip' and a broadly tuned 'tail'. The abnormal tuning curve also appears to have two sections, but the tip is elevated, and the tail is hypersensitive (i.e. thresholds in the region of the tail are lower than normal). The elevated tip may reflect the operation of the active mechanism at reduced effectiveness. The pattern of results suggests that the active mechanism is tuned to a frequency slightly above the resonance frequency of the passive mechanism at that place. This is consistent with the observation that, in a normal ear and at high CFs, the peak response on the BM shifts towards the base of the cochlea with increasing sound level (see Section IV.2).

Figure 1.21 shows the situation when there is a total loss of OHCs with intact IHCs. This pattern of damage is most easily produced with large doses of ototoxic drugs. The bowl-shaped abnormal tuning curve completely lacks the sharp tip because the active mechanism is completely destroyed. The broad tuning of the curve probably depends largely on the passive mechanism.

Figure 1.22 shows the situation where there is severe damage to the stereocilia of the IHCs and the stereocilia of the first row of OHCs (the row closest to the IHCs). The OHC damage is sufficient to eliminate completely the sharply tuned tip of the tuning curve, suggesting that the active mechanism is particularly dependent on the first row of OHCs. In addition, the whole curve is shifted upwards, that is sensitivity is much less than normal. This overall loss of sensitivity (compare the tuning curves in Figures 1.20 and 1.21) can probably be attributed to the IHC damage. According to Liberman *et al.* (1986), significant damage to the IHC stereocilia is always associated with reduced sensitivity on the tail of the tuning curve.

Liberman *et al.* did not find any cases of pure IHC damage, without OHC damage. It appears that the OHCs are generally more vulnerable to damage than the IHCs, and so damage to the IHCs is nearly always associated with damage to the OHCs.

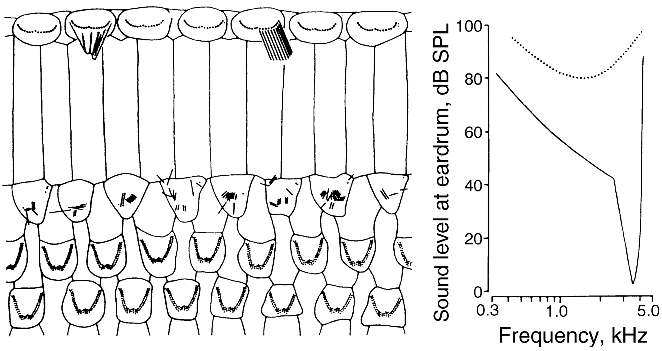


Figure 1.22. As Figure 1.20, but for the situation with severe damage to both OHC and IHC stereocilia. Most of the damage to the OHC stereocilia is confined to the first row of hair cells.

However, cases approximating pure IHC damage were sometimes found. Figure 1.23 shows an example of a neurone contacting an IHC missing the entire tall row of stereocilia (see arrow) in a region with only modest disarray of OHC stereocilia. The tuning curve shows a tip that is almost normal in shape, but the tip and tail are both shifted upwards by about 40 dB. It appears that the active mechanism was operating, but the transduction mechanism had greatly reduced sensitivity.

These findings can be summarized as follows. The OHCs are responsible for the sharp tips of the tuning curves. When the OHCs are damaged, the sharp tip becomes elevated, or may disappear altogether. This can cause a threshold elevation around the tip of the tuning curve of 40–50 dB. The condition of the OHCs determines the ratio of tip to tail thresholds. Damage to the IHCs causes an overall loss of sensitivity. This is apparent in the tail of the tuning curve, whether or not the OHCs are damaged. Pure OHC damage either leaves the tail unaffected or causes hypersensitivity (lower thresholds) in the tail. When both OHCs and IHCs are damaged, thresholds are usually

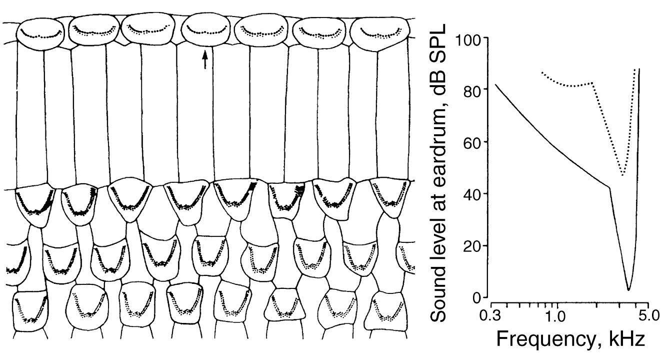


Figure 1.23. As Figure 1.20, but for a situation with moderate damage to IHC stereocilia and minimal damage to OHC stereocilia.

greatly elevated, by 80 dB or more, and the tuning curve is broad, without any sign of a sharp tip.

Liberman *et al.* found that they could 'account for' most of the threshold shift in the neurones by evaluating the presence or absence of OHCs and IHCs and the condition of their stereocilia. Furthermore, behavioural thresholds after noise exposure have been found to be correlated with patterns of damage in hair cell stereocilia (Slepecky *et al.*, 1982). It seems likely that, for many types of acquired cochlear hearing loss, the primary cause of the loss is loss of function of the OHCs and/or IHCs. For moderate losses, where thresholds are elevated by less than 50 dB, it may often be the case that the main cause of the loss is damage to the OHCs, with consequent impairment of the active mechanism. In this case, the loss of absolute sensitivity (elevation of threshold) occurs mainly because of reduced responses on the BM to low-level sounds. For more severe losses, it is likely that both OHCs and IHCs are damaged. When the IHCs are damaged, a greater response of the BM is needed to produce a 'threshold' amount of neural activity. In extreme cases, the IHCs may be completely non-functional over a certain region of the BM, in which case there is no transduction of BM-vibration in that region and no responses of neurones innervating that region. This is referred to as a *dead region* and is discussed in more detail in Chapters 2 and 3.

VII.4 OTOACOUSTIC EMISSIONS

Evoked otoacoustic emissions are reduced in magnitude by cochlear hearing loss. Human ears with hearing losses exceeding 40–60 dB (see Chapter 2, Section II.4 for the definition of hearing loss) usually show no detectable emissions (Gorga *et al.*, 1997). The emissions appear to be particularly associated with OHC function. The emissions are abolished in ears which have been exposed to intense sounds or to drugs which adversely affect the operation of the cochlea. In the former case, the emissions may return after a period of recovery. This suggests that the emissions are linked to the active mechanism. The measurement of cochlear emissions provides a sensitive way of monitoring the physiological state of the cochlea, and it is now being commonly applied in clinical situations, especially for assessing cochlear function in the very young (Kemp, 2002).

VII.5 PHASE LOCKING

The effect of cochlear damage on phase locking is not entirely clear. Harrison and Evans (1979) used the ototoxic drug kanamycin to produce hair cell damage (mainly to OHCs) in the guinea pig, and found that phase locking was not affected. However, Woolf, Ryan and Bone (1981) carried out a similar experiment using the chinchilla and found that phase locking was adversely affected by damage to the OHCs. For neurones with CFs corresponding to frequencies where the behavioural thresholds were elevated by 40 dB or more compared to normal, phase locking was significantly reduced. There was a reduction in the highest frequency at which phase locking could be observed, and the precision of phase locking over the range 0.4–3 kHz was reduced.

The reason for the discrepancy between studies is unclear. It could be related to a species difference. Another possibility is that the data of Harrison and Evans are atypical. Woolf *et al.* pointed out that the upper limit of phase locking reported by Harrison and Evans (5–7 kHz) was higher than usually observed in the guinea pig. The reason why phase locking should deteriorate as a consequence of cochlear damage is also unclear. Woolf *et al.* suggested that it could be connected with poorer mechanical coupling between the tallest stereocilia of the OHCs and the tectorial membrane. Whatever the reason, it seems clear that phase locking can sometimes be affected by cochlear damage, and this may have important perceptual consequences.

Cochlear damage can certainly affect phase locking to complex sounds, such as speech. Vowel sounds contain peaks in their spectral envelope at certain frequencies called *formants*; these correspond to resonances in the vocal tract (see Chapter 8 and Figure 8.3). Each formant is actually defined by several harmonics, whose amplitudes exceed those of adjacent harmonics. The formants are numbered, the lowest in frequency being called F1, the next F2 and so on. The frequencies of the formants, and especially the first two formants, are believed to be important for determining the identities of vowel sounds. At low levels in a normal auditory system, each neurone shows phase locking to a single harmonic or a small group of harmonics whose frequencies lie close to the CF of the neurone. Hence, the temporal response patterns vary markedly across neurones with different CFs. However, at higher levels, the temporal response patterns show a ‘capture’ phenomenon, in which the first two formant frequencies dominate the responses; neurones with CFs that are somewhat removed from a formant frequency may nevertheless show strong phase locking to that formant (Miller *et al.*, 1997). Most of the neurones with mid-range CFs show phase locking either to F1 or to F2. This may partly depend upon a suppression effect, related to two tone suppression, whereby the relatively strong harmonics close to the formant frequencies suppress the responses to weaker harmonics at adjacent frequencies. Whatever the mechanism, the temporal information coded in the phase locking may be used by the auditory system to determine the formant frequencies.

Miller *et al.* (1997) studied the effect on this phenomenon of cochlear damage caused by exposure to intense sound. After the acoustic trauma, capture by the second formant (which fell in the region of threshold elevation) was not observed; neurones with CFs adjacent to F2 did not show clear phase locking to F2, but showed more complex response patterns. The phase locking to formant frequencies observed in the normal auditory nerve may play an important role in the coding of the formant frequencies in the auditory system. If so, the reduced phase locking associated with cochlear damage might contribute to problems in understanding speech.

VIII CONCLUSIONS

The functioning of the normal cochlea is strongly dependent on an active mechanism that is physiologically vulnerable. This mechanism depends upon the integrity of the OHCs, and particularly their stereocilia. The active mechanism is responsible for the high sensitivity and sharp tuning of the BM. It is also responsible for a variety of nonlinear effects that can be observed in BM responses and neural responses. These effects include: the nonlinear input-output functions on the BM, the reduction in sharpness of tuning with increasing sound level and combination tone generation. Finally, the active mechanism is probably responsible for the generation of evoked and spontaneous otoacoustic emissions. The active mechanism strongly influences responses on the BM at low and medium sound levels, but its contribution progressively reduces as the sound level increases.

The OHCs are easily damaged by noise exposure, ototoxic chemicals, infection and metabolic disturbances. When they are damaged, the active mechanism is reduced in effectiveness or destroyed completely. This has several important consequences:

1. Sensitivity is reduced, so that the tips of tuning curves are elevated by up to 40–50 dB.
2. The sharpness of tuning on the BM is greatly reduced. The tip of the tuning curve may be elevated or may disappear altogether, leaving only the broad tuning of the passive BM.
3. Nonlinear effects such as compressive input-output functions on the BM, two-tone suppression and combination tone generation are reduced or disappear altogether.
4. Evoked and spontaneous otoacoustic emissions are reduced or disappear, at least in the frequency range corresponding to the damaged place.

The IHCs are the transducers of the cochlea converting the mechanical vibrations on the BM into neural activity. They are less susceptible to damage than the OHCs. When they are damaged, sensitivity is reduced. Damage primarily to the IHCs, with intact OHCs, is rare. When it does occur, sharp tuning may be preserved, but the whole tuning curve is elevated (less sensitive). More commonly, damage occurs both to the OHCs and the IHCs. In this case, the whole tuning curve is elevated, with a greater elevation around the tip than around the tail.

Damage to the hair cells sometimes, but not always, results in a reduction in phase locking; the precision with which neural impulses are synchronized to the cochlear-filtered stimulating waveform may be reduced. The reason why this occurs is unclear, but it may have important perceptual consequences.

2 Absolute Thresholds

I INTRODUCTION

The *absolute threshold* of a sound is the minimum detectable level of that sound in the absence of any other external sounds. The most obvious result of cochlear damage is reduced sensitivity to weak sounds; absolute thresholds are higher than normal. When measuring absolute thresholds, it is important to define the way in which the physical intensity of the threshold stimulus is measured. This chapter first describes the three common methods for specifying the threshold intensity. It then considers the origins of the loss of sensitivity associated with cochlear damage.

II MEASURES OF ABSOLUTE THRESHOLD

II.1 MINIMUM AUDIBLE PRESSURE (MAP)

The first method requires the measurement of the sound pressure at some point close to the entrance of the ear canal or inside the ear canal, using a small 'probe' microphone. Such probe microphones are now often incorporated in 'real-ear' measurement systems that are used in fitting hearing aids (Mueller, Hawkins and Northern, 1992). Ideally, the measurement of sound level is made very close to the eardrum. In all cases, it is necessary to specify the exact position of the probe, since small changes in position can markedly affect the results at high frequencies. The threshold so determined is called the *minimum audible pressure* (MAP). The sounds are usually, but not always, delivered by headphone. Usually, the absolute threshold is measured separately for each ear (called *monaural* listening).

II.2 MINIMUM AUDIBLE FIELD (MAF)

A second method uses sounds delivered by loudspeaker, preferably in a room whose walls are highly sound-absorbing. When reflections of sound from the walls, floor and ceiling are negligible, the listening conditions are described as *free field*. The measurement of sound level is made after the listener is removed from the sound field, at the point which had been occupied by the centre of the listener's head. The threshold determined in this way is called the *minimum audible field* (MAF). Usually, the sound is assumed to be coming from directly in front of the listener. Considerable errors can occur using this method if the reflections from the walls, floor and ceiling are not negligible, especially when sinusoids are used as stimuli. The errors can be reduced by the use of signals covering a wider frequency range, such as bands of noise or frequency modulated tones, which are also called warble tones. Usually, the absolute threshold is measured when listening with both ears (called binaural listening), but one ear can be plugged if it is desired to measure the threshold for each ear separately.

II.3 COMPARISON OF MAP AND MAF

Figure 2.1 shows estimates of the MAP, with sound level measured close to the eardrum (Killion, 1978), and estimates of the MAF published in an ISO standard (ISO 389-7, 2005). Note that the MAP estimates are for monaural listening and the MAF estimates are for binaural listening. On average, thresholds are about 1–2 dB lower when two ears are used as opposed to one. Both curves represent the average data from many young listeners with normal hearing. It should be noted, however, that individual subjects may have thresholds as much as 20 dB above or below the mean at a specific frequency and still be considered as 'normal'.

The 'audibility curves' for the MAP and MAF are somewhat differently shaped, since the physical presence of the head, the pinna and the meatus influences the sound field. The MAP curve shows only minor peaks and dips (±5 dB) for frequencies between about 0.2 and 13 kHz, whereas the MAF curve shows a distinct dip around

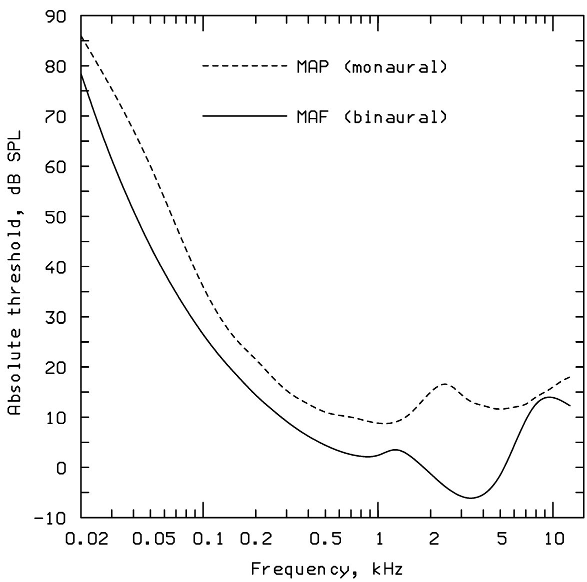


Figure 2.1. The absolute threshold expressed as the minimum audible pressure (MAP) and the minimum audible field (MAF) plotted as a function of frequency for young, normally hearing subjects.

3–4 kHz and a peak around 8–9 kHz. The difference derives mainly from a broad *resonance* produced by the meatus and pinna (see Figure 1.3). As a result, for a fixed sound source, the sound level measured at the eardrum is higher than the sound level measured in the free field, for sounds with frequencies close to 3 kHz.

Another factor which contributes to the shapes of the audibility curves is the transmission characteristic of the middle ear. Transmission is most efficient for mid-range frequencies and drops off markedly for very low and very high frequencies (see Figure 1.4). This can partly account for the rapid rise in threshold at low and high frequencies.

To a first approximation, the cochlea in normally hearing humans is equally sensitive to all frequencies from about 0.5 up to 15 kHz; variations in absolute threshold with frequency over this range probably reflect primarily the transmission characteristics of the outer and middle ear (Moore, Glasberg and Baer, 1997; Puria, Rosowski and Peake, 1997). Below 0.5 kHz, the sensitivity of the cochlea appears to decrease. This may reflect reduced gain from the active mechanism at low frequencies (Yates, 1995), internal noise produced by blood flow (Zwicker and Fastl, 1999) or an effect of the helicotrema (Cheatham and Dallos, 2001) (recall that this is a small opening which connects the two outer chambers of the cochlea, the *scala vestibuli* and the *scala tympani*, at the apical end of the cochlea). The helicotrema reduces the pressure difference applied across the basilar membrane (BM) for very low-frequency sounds.

The highest audible frequency varies considerably with age. Young children can often hear sinusoids with frequencies as high as 20 kHz, but for most adults threshold rises rapidly above about 15 kHz (Ashihara *et al.*, 2006). The loss of sensitivity with increasing age (*presbyacusis*) is much greater at high frequencies than at low, and the variability between people is also greater at high frequencies.

II.4 THE AUDIOGRAM

A third method of specifying absolute thresholds is commonly used in audiology; thresholds are specified relative to the average threshold at each frequency for young, healthy listeners with ‘normal’ hearing. In this case, the sound level is usually specified relative to standardized values produced by a specific earphone in a specific coupler. A coupler is a device which contains a cavity or series of cavities and a microphone for measuring the sound produced by the earphone. The preferred earphone varies from one country to another. For example, the Telephonics TDH 39, 49 or 50 is often used in the United States, while the Beyer DT48 is used in Germany. Thresholds specified in this way have units dB HL (hearing level) in Europe or dB HTL (hearing threshold level) in the United States. For example, a threshold of 40 dB HL at 1 kHz would mean that the person had a threshold which was 40 dB higher than ‘normal’ at that frequency. In psychoacoustic work, thresholds are normally plotted with threshold increasing upwards, as in Figure 2.1. However, in audiology, threshold elevations are shown as hearing losses, plotted downwards. The average ‘normal’ threshold is represented as a horizontal line at the top of the plot, and the degree of hearing loss is indicated by how much the threshold falls below this line. This type of plot is called an *audiogram*. Figure 2.2 compares an audiogram for a hypothetical hearing-

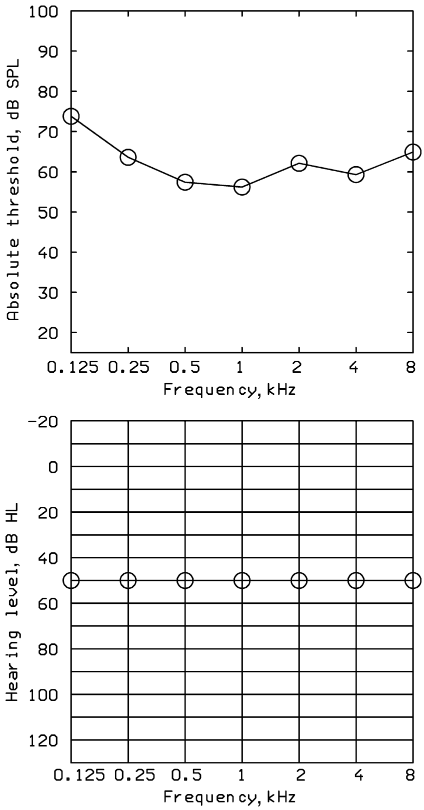


Figure 2.2. Comparison of a clinical audiogram for a 50-dB hearing loss at all frequencies (bottom), and the absolute threshold curve for the same hearing loss plotted in terms of the MAP (top).

impaired person with a ‘flat’ hearing loss with a plot of the same thresholds expressed as MAP values. Notice that, although the audiogram is flat, the corresponding MAP curve is not flat. Note also that thresholds in dB HL can be negative. For example, a threshold of −10 dB simply means that the individual is 10 dB more sensitive than the average.

The *sensation level* (SL) describes the level of a sound relative to the absolute threshold of the subject for that sound. For example, if the absolute threshold is 23 dB SPL, and the sound is presented at 33 dB SPL, then its level is 10 dB SL.

III DESCRIPTIONS OF THE SEVERITY OF HEARING LOSS

The measurement of absolute thresholds provides the easiest way of detecting and measuring damage to the auditory system. Often, the hearing loss varies with frequency. However, it is useful to have a way of describing the overall severity of a hearing loss. A common way of quantifying hearing loss is in terms of the absolute threshold for sinusoids, expressed in dB HL, and averaged over the frequencies 500, 1000 and 2000 Hz; this is often known as the *pure-tone average* (PTA) hearing loss. Goodman (1965) proposed the following classification:

PTA −10–26 dB	Normal limits
PTA 27–40 dB	Mild hearing loss
PTA 41–55 dB	Moderate hearing loss
PTA 56–70 dB	Moderately severe hearing loss
PTA 71–90 dB	Severe hearing loss
PTA over 90 dB	Profound hearing loss

A similar classification is still used, although nowadays the boundary between normal hearing and mild hearing loss is usually put at a PTA of 16 dB. Also, the frequencies included in the calculation of the PTA are sometimes extended to higher frequencies. Generally, the categories are indicative of the severity of the problems that are likely to be experienced in everyday life; greater losses are associated with greater problems. However, individuals with similar absolute thresholds may vary considerably in this respect.

IV CAUSES OF HEARING LOSS DUE TO COCHLEAR DAMAGE

Elevation of absolute threshold due to cochlear damage can occur in two main ways. First, loss of function of the *outer hair cells* (OHCs) impairs the active mechanism, resulting in reduced BM vibration for a given low sound level. Hence, the sound level must be larger than normal to give a just-detectable amount of vibration. It is likely that hearing loss caused solely by loss of the active mechanism cannot be greater than about 50 dB at low frequencies and 60 dB at high frequencies. Secondly, loss of function of *inner hair cells* (IHCs) can result in reduced efficiency of transduction, and so the amount of BM vibration needed to reach threshold is larger than normal. The functioning of both OHCs and IHCs can be affected by a variety of factors other than physical damage, including impaired operation of the stria vascularis (Schmiedt, 1996).

In extreme cases, the IHCs at certain places along the BM may be completely non-functioning. In such cases, vibration at those places is not detected by the neurones directly innervating them. Say, for example, the IHCs at the basal end of the cochlea are non-functioning. Neurones innervating the basal end, which would normally have high *characteristic frequencies* (CFs), will not respond. However, if a high-frequency sinusoid is presented, it may be detected if it produces sufficient BM vibration at a more apical region; this corresponds to downward spread of excitation. In other words, a high-frequency sound may be detected via neurones that are tuned to lower frequencies. Similarly, if there are no functioning IHCs at an apical region of the cochlea, a low-frequency sound may be detected via neurones that are tuned to higher frequencies. Because of this possibility, the 'true' hearing loss at a given frequency may be greater than suggested by the audiometric threshold at that frequency.

A region of the BM over which there are no functioning IHCs is referred to as a *dead region* (Moore and Glasberg, 1997). Moore (2004a) has suggested that a dead region can be defined as 'a region where the inner hair cells or/and neurones are functioning so poorly that a tone producing maximum vibration in that region is detected at a different place in the cochlea, using *off-frequency listening*.' Often the extent of a dead region is described in terms of the CFs of the functioning IHCs or neurones at the edge of the dead region (Moore, 2001). Say, for example, that the IHCs are non-functioning over a region of the BM having CFs in the range 4–10 kHz. The neurones at the boundaries of the dead region would have CFs of 4 and 10 kHz. This is described as a dead region extending from 4 to 10 kHz. Dead regions are not necessarily easily identified from the pure-tone audiogram. For example, when there is a low-frequency dead region, low frequencies are detected using the responses of neurones with CFs above the dead region (Thornton and Abbas, 1980; Florentine and Houtsma, 1983); see Chapter 3, Section IX. However, audiometric thresholds at low frequencies may not be markedly higher than audiometric thresholds at higher frequencies, because the neurones with high CFs may have *tuning curves* with shallow, low-frequency 'tails' (see Figure 1.22). Psychoacoustic procedures for determining the extent of dead regions are described in Chapter 3, Section IX.

V PERCEPTUAL CONSEQUENCES OF ELEVATED ABSOLUTE THRESHOLDS

The loss of sensitivity to weak sounds is one of the main causes of the problems experienced by people with cochlear hearing loss. Weak sounds, and especially the weaker sounds in speech, like 'p', 't' and 'k', simply are not heard. The effect of this on speech perception is discussed in Chapter 8, Section IV.

As mentioned earlier, individuals with similar absolute thresholds may vary considerably in terms of the difficulty that they experience in everyday life. These individual differences are partly explicable in terms of differing patterns of OHC and IHC loss. The most severe situation is when there are one or more extensive dead regions. Tones with frequencies corresponding to a dead region are often described as sounding 'noise-like' or 'distorted' (Huss and Moore, 2005a). When complex sounds such as speech are presented, only limited information can be extracted from the frequencies in the speech corresponding to the dead region (Vickers, Moore and Baer, 2001; Baer, Moore and Kluk, 2002). This is discussed in Chapter 8, Section IV.

3 Masking, Frequency Selectivity and Basilar Membrane Nonlinearity

I INTRODUCTION

Although loss of sensitivity is of major importance in creating hearing difficulty in everyday life, it is not the only factor involved. Cochlear damage produces a variety of changes in the perception of sounds that are well above the threshold for detection. This chapter considers the reduced frequency selectivity associated with cochlear hearing loss. *Frequency selectivity* refers to the ability of the auditory system to separate or resolve (to a limited extent) the spectral *components* in a complex sound. For example, if two tuning forks, each tuned to a different frequency, are struck simultaneously, two different tones can usually be heard, one corresponding to each frequency. This ability is also known as *frequency analysis* and *frequency resolution*; these terms will be used interchangeably in this book.

Frequency selectivity probably depends largely on the filtering that takes place in the cochlea (see Chapter 1, Section IV.2). The sinusoidal components of a complex sound each lead to a peak in the vibration pattern at a specific place on the BM (*basilar membrane*), and the components are coded independently in the auditory nerve, provided that their frequency separation is sufficiently large. Damage to the cochlea, and particularly to the *outer hair cells* (OHCs), leads to reduced sharpness of tuning on the BM and in single neurones of the auditory nerve. Hence, it is to be expected that frequency selectivity is poorer than normal in people with cochlear hearing loss.

This chapter starts by describing how frequency selectivity can be measured in normally hearing subjects, using *masking* experiments. Typically, the subject is required to detect a signal such as a sinusoid in the presence of a background (masking) sound. The results are used to provide a description of frequency selectivity in the normal auditory system. The description is based on two complementary concepts. One is called the *auditory filter* shape. This characterizes the tuning or selectivity available at a particular centre frequency. It can be thought of as analogous to measures of tuning at a single point on the BM (e.g. the constant velocity *tuning curve* described in Chapter 1; see Figure 1.9) or in single neurones of the auditory nerve (e.g. the *frequency-threshold curve* described in Chapter 1; see Figure 1.13). The second concept is that of the *excitation pattern*. This represents the distribution of activity produced by a given sound at different places (corresponding to different *characteristic frequencies*, or CFs) in the auditory system. It can be considered as analogous to measures of the magnitude of response at different points along the BM; see, for example, Figure 1.8.

The chapter also describes *non-simultaneous masking*, when a brief signal is presented either before or after a masker. Non-simultaneous masking can be used as a tool for measuring the effects of nonlinearities on the BM, such as compressive *input-output functions* and *two-tone suppression* (see Chapter 1, Sections IV.3 and IV.4).

The remainder of the chapter is concerned with the measurement of frequency selectivity in people with cochlear hearing loss. It describes how frequency selectivity and nonlinearities are reduced by cochlear hearing loss and discusses the perceptual consequences of these effects.

II THE MEASUREMENT OF FREQUENCY SELECTIVITY USING MASKING

II.1 INTRODUCTION

Sometimes, sounds we want to hear are made inaudible by other sounds. This is called masking. For example, it may be impossible to hear the doorbell ringing when music is being played at high volume on a stereo system. Masking may be defined as the process by which the threshold for detecting one sound (the signal) is raised in the presence of another sound (the masker). Masking reflects the limits of frequency selectivity; if a signal with a given frequency is masked by a masker with a different frequency, then the auditory system has failed to resolve the signal and the masker. Hence, by measuring the conditions necessary for one sound just to mask another, it is possible to characterize the frequency selectivity of the auditory system.

It should be noted that the tuning on the BM has often been measured using sinusoids whose frequency and intensity have been systematically varied. At any one time, only a single sinusoid is present. In contrast, measurements of masking usually involve the presentation of more than one sinusoid at the same time. This may be important in interpreting the results, as will be described later.

II.2 THE POWER-SPECTRUM MODEL

To explain the mechanism underlying auditory masking, Fletcher (1940) suggested that the peripheral auditory system behaves as if it contained an array of bandpass filters, with overlapping *passbands* (the passband of a given filter refers to the range of frequencies passed by the filter). These filters are now called the *auditory filters*. Each filter can be thought of as corresponding to a particular place on the BM. Fletcher and subsequent researchers (Patterson and Moore, 1986) proposed a model of masking known as the *power-spectrum model*, which is based on the following assumptions:

1. The peripheral auditory system contains an array of linear overlapping bandpass filters.

2. When trying to detect a signal in a noise background, the listener makes use of just one filter with a centre frequency close to that of the signal. Usually, it is assumed that the filter used is the one that has the highest signal-to-masker ratio at its output; this filter has the 'best' representation of the signal.
3. Only the components in the noise which pass through the filter have any effect in masking the signal.
4. The threshold for detecting the signal is determined by the amount of noise passing through the auditory filter. Specifically, threshold is assumed to correspond to a certain signal-to-noise ratio at the output of the filter. The stimuli are represented by their long-term power spectra, that is by the average power plotted as a function of frequency. The relative phases of the components and the short-term fluctuations in the masker are ignored.

None of these assumptions is strictly correct: the filters are not linear, but are level dependent (Moore and Glasberg, 1987), as would be expected from the properties of the filtering that takes place on the BM. Listeners can combine information from more than one filter to enhance signal detection, especially when the signal contains components with frequencies spread over a wide range (Spiegel, 1981; Buus *et al.*, 1986); noise falling outside the passband of the auditory filter centred at the signal frequency can affect the detection of that signal (Hall, Haggard and Fernandes, 1984); and short-term fluctuations in the masker can play a strong role (Patterson and Henning, 1977; Kohlrausch, 1988; Moore, 1988).

These violations of the assumptions do not mean that the basic concept of the auditory filter is wrong. Indeed, the concept is widely accepted and will be used extensively in the rest of this book. Nevertheless, it should be remembered that simplifying assumptions are often made in attempts to characterize and model the auditory filter.

II.3 ESTIMATING THE SHAPE OF A FILTER

The relative response of a filter as a function of the input frequency is often called the *filter shape*. For a physical device, such as an electronic filter, the filter shape can be measured in several ways, two of which are:

1. A sinusoid is applied to the input of the filter, keeping its magnitude constant, and the output magnitude is measured as a function of the input frequency. A plot of the output magnitude as a function of frequency then gives the filter shape.
2. A sinusoid is applied to the input of the filter, its frequency is systematically varied and its input magnitude adjusted so as to hold the output magnitude constant. A method similar to this is used to determine tuning curves on the BM or in the auditory nerve (see Chapter 1, Sections IV.2 and V.2).

If a filter is linear, the shapes resulting from these two methods are simply related: if the two curves are plotted on a *decibel* scale, then one is simply an inverted version of the other. For example, if the filter is a bandpass filter, then method (1) will give a curve with a peak at a certain frequency (the centre frequency of the filter), while method (2) will give a minimum at that same frequency. For a linear filter, it does not matter which method is used: the same end result can be obtained by either method.

The determination of the auditory filter shape at a particular centre frequency is complicated by several factors. First, the auditory filter is *not* linear; its shape depends on sound level. Hence, different results may be obtained depending on overall level and depending on whether it is the input to the filter or the output from the filter that is held constant. In practice, it may be difficult to hold either the input or the output constant in a masking experiment, although one can attempt to approximate these conditions. Secondly, in humans, the input to and output from the filter cannot be measured directly. Hence, the filter shape has to be derived from experimental data on masking, using certain assumptions.

Most methods for estimating the shape of the auditory filter at a given centre frequency are based on the assumptions of the power-spectrum model of masking. The signal frequency is held constant and the *spectrum* of the masker is manipulated in some way. By analogy with methods (1) and (2) above, two basic methods can be used:

1. The masker level is held constant and the signal level is varied to determine the threshold for detection of the signal. This is analogous to holding the input level to a filter constant and measuring its output; the signal threshold is assumed to be proportional to the masker level at the output of the filter.
2. The signal level is held constant and the masker level is varied to determine the value required just to mask the signal. This is analogous to holding the output of the filter constant and varying its input to achieve a fixed output.

Both of these approaches will be described in the following sections.

III ESTIMATING FREQUENCY SELECTIVITY FROM MASKING EXPERIMENTS

III.1 PSYCHOPHYSICAL TUNING CURVES

The measurement of *psychophysical tuning curves* (PTCs) involves a procedure which is analogous in many ways to physiological methods for determination of a tuning curve on the BM or a neural tuning curve (Chistovich, 1957; Small, 1959) (see Chapter 1, Sections IV.2 and V.2). The signal is fixed in level, usually at a very low level, say 10 dB *sensation level* (SL). The masker can be either a sinusoid or a narrow band of noise, although a noise is preferable, for the reasons given below.

For each of several masker centre frequencies, the level of the masker needed just to mask the signal is determined. Because the signal is at a low level, it is assumed that it will produce activity primarily at the output of one auditory filter. It is assumed further that, at threshold, the masker produces a constant output from that filter in order to mask the fixed signal. If these assumptions are valid, then the PTC indicates the masker level required to produce a fixed output from the auditory filter as a function of frequency; this is why the procedure is analogous to the determination of a tuning curve on the BM or a neural tuning curve. Both physiological procedures involve adjusting the input level for each frequency to produce a constant response. In the case of the BM, the response is a constant amplitude or velocity of vibration on the BM. In the case of a neural tuning curve, the response is a fixed number of neural spikes per second.

When measuring a PTC, the preferred masker is a band of noise rather than a sinusoid, as the use of a noise reduces the influence of *beats* (fluctuations in amplitude) caused by the interaction of the signal and the masker (Kluk and Moore, 2004). When a sinusoid is used as a masker, beats can provide a cue to indicate the presence of the signal when the masker frequency is close to the signal frequency, but not when the masker frequency is equal to the signal frequency. This can lead to a PTC with a very sharp tip (Kluk and Moore, 2004), which may be much sharper than the tip of the auditory filter.

Consider the two example PTCs shown in Figure 3.1; the data are taken from Moore and Glasberg (1986b). The circles show results from the normal ear of a subject with

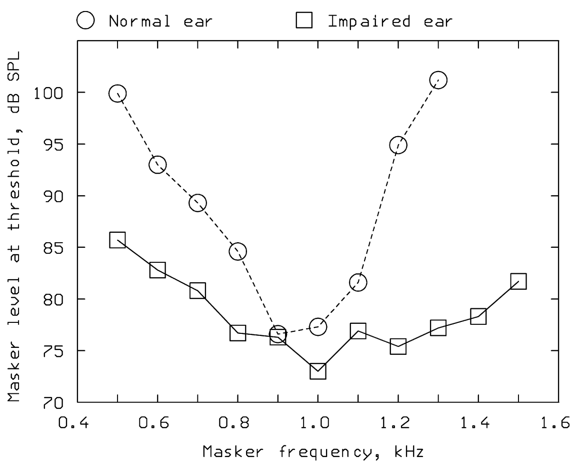


Figure 3.1. PTCs determined in simultaneous masking for the normal ear (circles and dashed line) and the impaired ear (squares and continuous line) of a subject with a unilateral cochlear hearing loss. The signal frequency was 1 kHz. The variable masker was a narrowband noise. A fixed notched noise was gated with the variable masker, to restrict off-frequency listening. The signal was presented at a level 10 dB above its masked threshold in the notched noise alone.

a unilateral cochlear hearing loss, while the squares show results from the impaired ear of the same subject. The PTC is clearly broader for the impaired ear.

One complication in interpreting these results is that the auditory filter giving the highest signal-to-masker ratio is not necessarily centred at the signal frequency. The process of detecting the signal through a filter which is not centred at the signal frequency is called *off-frequency listening*. In this context, it is the centre frequency of the filter that is 'off frequency'. There is good evidence that off-frequency listening can influence PTCs. When the masker frequency is above the signal frequency, the highest signal-to-masker ratio occurs for a filter centred below the signal frequency. This happens because the filter has a rounded tip, but steeply sloping 'skirts'. For a filter centred just below the signal frequency, the response to the signal is only slightly smaller than would be the case for a filter centred at the signal frequency. However, the response to the masker is markedly reduced by using the off-frequency filter. Similarly, when the masker frequency is below the signal frequency, the highest signal-to-masker ratio occurs for a filter centred above the signal frequency. Assuming that human listeners always select the 'best' filter, then, in both these cases, the masker level required for threshold is higher than would be the case if off-frequency listening did not occur. When the masker frequency equals the signal frequency, the signal-to-masker ratio is similar for all auditory filters that are excited, and off-frequency listening is not advantageous. The overall effect is that the PTC has a sharper tip than would be obtained if only one auditory filter were involved (Johnson-Davies and Patterson, 1979; O'Loughlin and Moore, 1981a, 1981b).

One way to limit off-frequency listening is to add to the main masker (the sinusoid or narrowband noise) a fixed broadband noise with a reasonably narrow spectral notch centred at the signal frequency (O'Loughlin and Moore, 1981a; Moore, Glasberg and Roberts, 1984; Patterson and Moore, 1986). The signal is usually presented at a level about 10 dB above its threshold when the notched noise alone is presented. For example, if the masked threshold of the signal in the notched noise is 15 dB SPL, then the signal would be presented at a level of 25 dB SPL. Schematic spectra of stimuli like this are shown in Figure 3.2. The notched noise makes it disadvantageous to use an auditory filter whose centre frequency is shifted much from the signal frequency; for such a shifted filter, more of the noise would pass through the filter. The notched-noise level is kept constant throughout the experiment, and the PTC is determined as before by finding the level of the main masker necessary to mask the signal, as a function of masker frequency. The effect of using such a noise, in addition to the variable masker, is illustrated in Figure 3.3. The main effect is a broadening of the tip of the PTC, while the slopes of the skirts are relatively unaffected.

The PTCs shown in Figure 3.1 were gathered in the presence of a notched noise of this type. For both the normal and impaired ears, the signal was at a level about 10 dB above masked threshold in the notched noise alone, and the notched-noise level was the same in the two ears. As well as reducing off-frequency listening, the noise had the advantage that it led to equal levels of the tips of the PTCs in the normal and impaired ears. Thus, the comparison of frequency selectivity between the two ears was not confounded by differences in overall level.

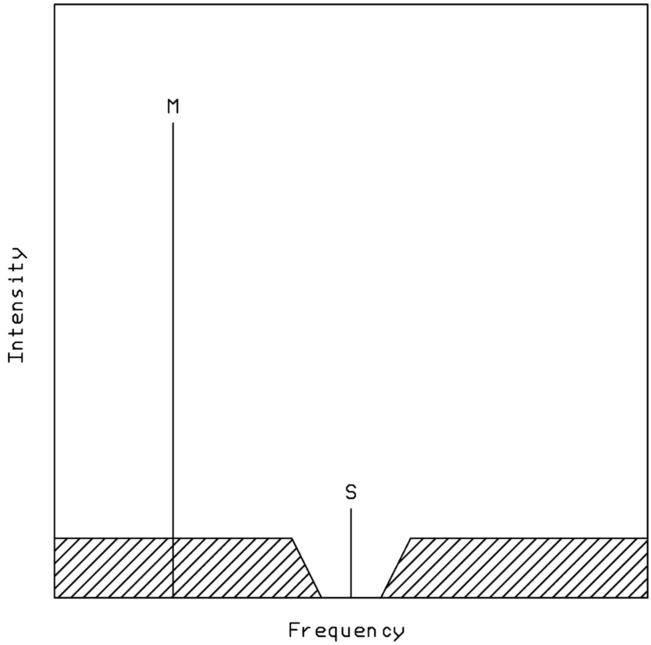


Figure 3.2. Schematic spectra of stimuli used to determine a PTC under conditions where off-frequency listening is restricted. The low-level notched noise is shown as the shaded regions. The signal is indicated by S and the masker by M.

III.2 THE NOTCHED-NOISE METHOD

To avoid strong violations of the assumptions of the power-spectrum model, it is necessary to use a masker that limits the extent to which off-frequency listening is useful. Also, the masker should limit the range of filter centre frequencies over which the signal-to-masker ratio is sufficiently high to be useful; this promotes the use of just one auditory filter. This can be achieved using a noise masker with a spectral notch around the signal frequency. For such a masker, the highest signal-to-masker ratio occurs for a filter which is centred reasonably close to the signal frequency, and performance is not improved (or is improved very little) by combining information over filters covering a range of centre frequencies (Patterson, 1976; Patterson and Moore, 1986; Moore, Glasberg and Simpson, 1992). The filter shape can then be estimated by measuring signal threshold as a function of the width of the notch. Usually, the noise level is kept constant and the signal level is varied to determine threshold, although some measurements have been performed with the signal level fixed and the masker level varied (Glasberg, Moore and Nimmo-Smith, 1984; Rosen, Baker and Kramer, 1992; Rosen, Baker and Darling, 1998).

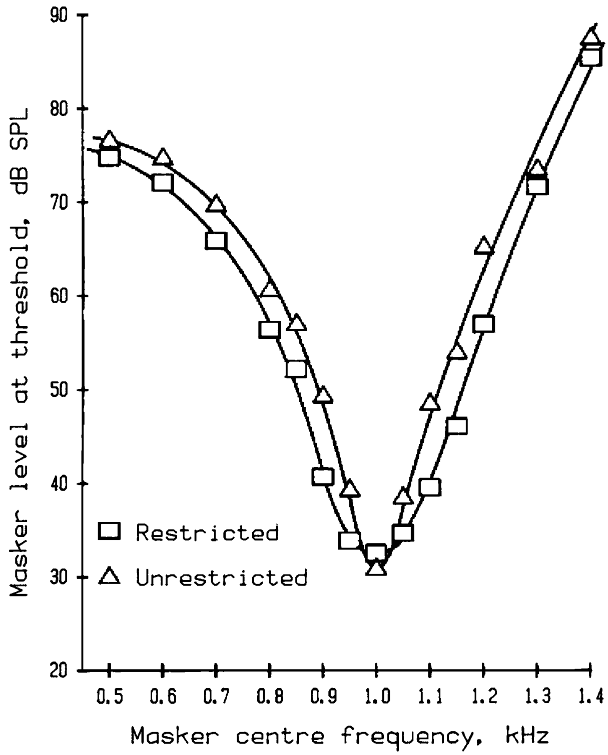


Figure 3.3. Comparison of PTCs where off-frequency listening is not restricted (triangles) and where it is restricted using a low-level notched noise centred at the signal frequency (squares). Data from Moore, Glasberg and Roberts, 1984a.

For moderate noise levels, the auditory filter is almost symmetrical on a linear frequency scale (Patterson, 1974, 1976; Patterson and Nimmo-Smith, 1980; Moore and Glasberg, 1987). Hence, the auditory filter shape can be estimated using a fixed-level notched-noise masker with the notch placed symmetrically about the signal frequency. The method is illustrated in Figure 3.4. The signal (indicated by the bold vertical line) is fixed in frequency, and the masker is a noise with a bandstop or notch centred at the signal frequency. The deviation of each edge of the noise from the centre frequency is denoted by Δf. The width of the notch is varied, and the threshold of the signal is determined as a function of the notch width. Since the notch is symmetrically placed around the signal frequency, the method cannot reveal asymmetries in the auditory filter, and the analysis assumes that the filter is symmetric on a linear frequency scale. This assumption appears not unreasonable, at least for the top part of the filter and at moderate sound levels, since PTCs are quite symmetric around the tips. For a signal symmetrically placed in a notched noise, the highest

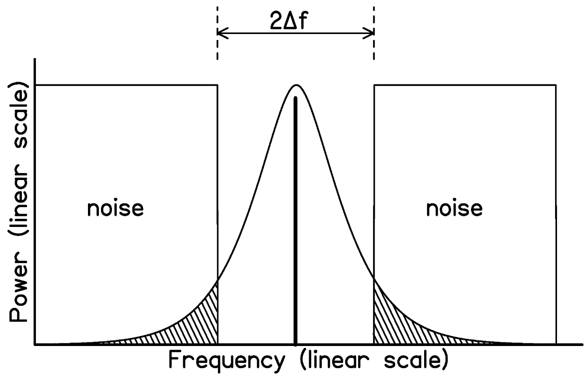


Figure 3.4. Schematic illustration of the notched-noise technique used by Patterson (1976) to determine the shape of the auditory filter. The threshold of the sinusoidal signal is measured as a function of the width of a spectral notch in the noise masker (overall width = 2Δf). The amount of noise passing through the auditory filter centred at the signal frequency is proportional to the shaded areas.

signal-to-masker ratio at the output of the auditory filter is achieved with a filter centred at the signal frequency, as illustrated in Figure 3.4. Using a filter not centred at the signal frequency reduces the amount of noise passing through the filter from one of the noise bands, but this is more than offset by the increase in noise from the other band.

As the width of the spectral notch is increased, less and less noise passes through the auditory filter. Thus, the threshold of the signal drops. Figure 3.4 is plotted with linear *power* (not decibels) on the ordinate. With such an ordinate, the total noise power passing through the auditory filter is proportional to the area under the filter in the frequency range covered by the noise. This is shown as the shaded areas in Figure 3.4. Assuming that threshold corresponds to a constant signal-to-masker ratio at the output of the filter, the change in signal threshold with notch width indicates how the area under the filter varies with Δf. The area under a function between certain limits is obtained by integrating the value of the function over those limits. Hence, by differentiating the function relating threshold (in linear power units) to Δf, the relative response of the filter at that value of Δf is obtained. In other words, the relative response of the filter (in linear power units) for a given deviation, Δf, from the centre frequency is equal to the slope of the function relating signal threshold to notch width at that value of Δf. If the threshold decreases rapidly with increasing notch width, that indicates a sharply tuned filter. If the threshold decreases slowly with increasing notch width, that indicates a broadly tuned filter. An example of an auditory filter shape obtained with this method is shown in Figure 3.5. Note that, although the derivation is based on the use of linear power units, the relative response of the filter is usually plotted on a decibel scale, as is done in Figure 3.5. The response of the filter at its tip was arbitrarily called 0 dB; in other words, it was assumed that

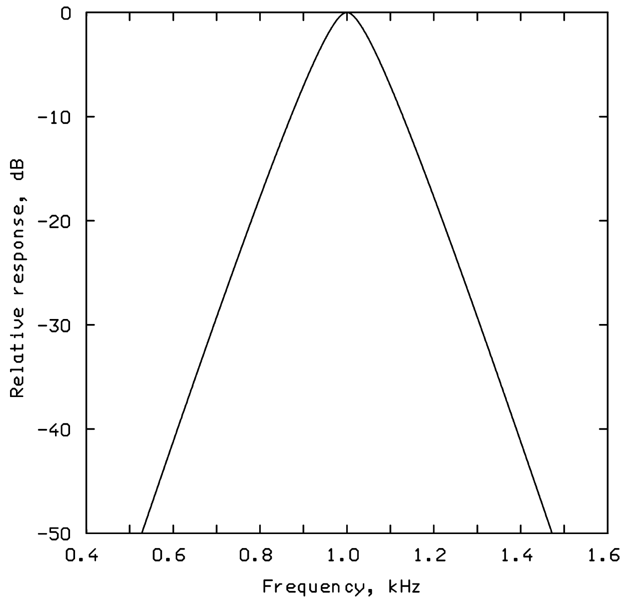


Figure 3.5. An example of an auditory filter shape obtained using the notched-noise method. The filter was centred at 1 kHz. The response is plotted relative to the response at the tip, which was arbitrarily called 0 dB.

for a signal at the centre frequency of the filter, the output magnitude was equal to the input magnitude.

When the auditory filter is asymmetric, as it is when high masker levels are used (see later), the filter shape can still be measured using a notched-noise masker if some reasonable assumptions are made, and if the range of measurements is extended to include conditions where the notch is placed asymmetrically about the signal frequency. The reader is referred elsewhere for details (Patterson and Nimmo-Smith, 1980; Glasberg and Moore, 1990).

One limitation of the notched-noise method occurs when the auditory filter is markedly asymmetric, as it is in some hearing-impaired people. In such cases, the method does not define the sharper side of the filter very well.

IV CHARACTERISTICS OF THE AUDITORY FILTER IN NORMAL HEARING

IV.1 VARIATION WITH CENTRE FREQUENCY

A bandpass filter is often characterized by its *bandwidth*, which is a measure of the effective range of frequencies passed by the filter. The bandwidth of a filter is often defined as the difference between the two frequencies at which the response has fallen by a factor of two in power (i.e. by 3 dB) relative to the peak response; this is known as the *half-power bandwidth* or −3-dB bandwidth. For example, if a filter has its peak response at 1000 Hz, and the response is 3 dB below the peak response at 900 Hz and at 1120 Hz, then the −3-dB bandwidth is 220 Hz. An alternative measure is the *equivalent rectangular bandwidth* (ERB). The ERB of a given filter is equal to the bandwidth of a perfect rectangular filter (a filter whose shape has a flat top and vertical edges) which has a transmission in its passband equal to the maximum transmission of the specified filter and transmits the same power of *white noise* as the specified filter. In other words, if we take a rectangular filter and scale it to have the same maximum height and area as our filter, then the bandwidth of the rectangular filter is the ERB of our filter. The equalization of height and area is done with the filter characteristic plotted in linear power coordinates. For auditory filters, the ERB is usually about 11 % larger than the −3-dB bandwidth. The ERB is related to what is sometimes referred to as the *critical bandwidth* (Fletcher, 1940; Zwicker, 1961; Scharf, 1970).

Figure 3.6 shows the ERBs of the auditory filters estimated in various studies (Moore and Glasberg, 1983c; Glasberg and Moore, 1990; Moore, 2003) using notched noise at moderate levels. The solid line in Figure 3.6 provides a good fit to the ERB values over the whole frequency range tested. It is described by the following equation:

$$\mathrm{ERB_N} = 24.7(0.00437F + 1), \tag{3.1}$$

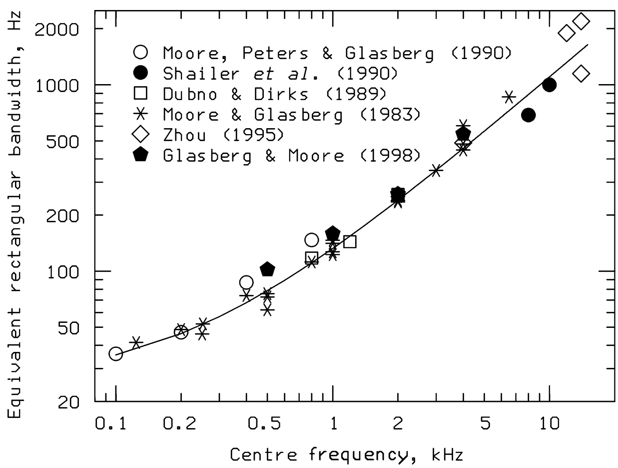


Figure 3.6. Estimates of the auditory filter bandwidth from a variety of experiments, plotted as a function of centre frequency. The solid line represents the equation (3.1) suggested by Glasberg and Moore (1990). Two points are shown for a centre frequency of 14 kHz, since filter bandwidths were estimated for two different noise spectrum levels, 40 and 50 dB; the ERB was greater at the higher level.

where F is the centre frequency in Hertz. The suffix N is used to denote the fact that this is the mean value of the ERB for normally hearing listeners. This equation is a modification of one originally suggested by Greenwood (1961). He based it on the assumption that each $\mathrm{ERB_N}$ corresponds to a constant distance along the BM. Although the constants in Equation 3.1 differ from those given by Greenwood, the form of the equation is the same as his. Each $\mathrm{ERB_N}$ corresponds to a distance of about 0.89 mm on the BM.

Auditory filter shapes for young normally hearing listeners vary relatively little across listeners; the standard deviation of the ERB across listeners at a given centre frequency is typically about 10 % of its mean value (Moore, 1987; Moore, Peters and Glasberg, 1990; Wright, 1996). However, the variability tends to increase at very low frequencies (Moore, Peters and Glasberg, 1990) and at very high frequencies (Patterson *et al*., 1982; Shailer *et al*., 1990).

Sometimes it is useful to plot psychoacoustical data on a frequency scale related to $\mathrm{ERB_N}$, called the *$\mathrm{ERB_N}$-number scale*. Essentially, $\mathrm{ERB_N}$ is used as the unit of frequency. For example, the value of $\mathrm{ERB_N}$ for a centre frequency of 1000 Hz is about 130 Hz, so an increase in frequency from 935 to 1065 Hz represents a step of one $\mathrm{ERB_N}$. A formula relating $\mathrm{ERB_N}$-number to frequency can be derived from Equation 3.1 (Glasberg and Moore, 1990):

$$\mathrm{ERB_N}\text{-number} = 21.4\log_{10}(0.00437F + 1), \tag{3.2}$$

where F is frequency in Hertz. The $\mathrm{ERB_N}$-number scale is comparable to a scale of distance along the BM, starting at the apical end. A frequency of 50 Hz produces peak activity near the apex; the corresponding $\mathrm{ERB_N}$-number is 1.83. A frequency of 15 000 Hz produces peak activity near the base; the corresponding $\mathrm{ERB_N}$-number is 38.9. A frequency of 1635 Hz produces peak activity about halfway along the BM; the corresponding $\mathrm{ERB_N}$-number is 19.5.

IV.2 VARIATION WITH LEVEL

If the auditory filter were linear, then its shape would not vary with the level of the stimuli used to measure it. However, this is not the case, as would be expected from measurements of the filtering on the BM. Moore and Glasberg (1987) presented a summary of measurements of the auditory filter shape using notched-noise maskers with notches placed both symmetrically and asymmetrically about the signal frequency. They concluded that the lower skirt of the filter becomes less sharp with increasing level, while the upper skirt becomes slightly steeper. Glasberg and Moore (1990) re-analysed the data from the studies summarized in that paper, and they also examined the data presented in Moore, Peters and Glasberg (1990) and Shailer *et al.* (1990). The re-analysis led to the following conclusions:

1. The auditory filter for a centre frequency of 1000 Hz is roughly symmetric on a linear frequency scale when the level of the noise (in a notched-noise exper-

iment) is approximately 51 dB/ERB$_N$. This corresponds to a noise spectrum level of about 30 dB (spectrum level is the level of a sound in decibels measured in a 1-Hz wide band). The auditory filters at other centre frequencies are approximately symmetric when the effective input levels to the filters are equivalent to the level of 51 dB/ERB$_N$ at 1 kHz (after making allowance for changes in relative level produced by passage of the sound through the outer and middle ear).

2. The low-frequency side of the auditory filter becomes less sharp with increasing level.
3. Changes in slope of the high-frequency side of the filter with level are less consistent. At medium centre frequencies (1–4 kHz) there is a trend for the slope to increase with increasing level, but at low centre frequencies there is no clear trend with level, and the filters at high centre frequencies show a slight decrease in slope with increasing level.

The statements above are based on the assumption that, although the auditory filter is not linear, it may be considered as approximately linear at any given noise level. Furthermore, the sharpness of the filter is assumed to depend on the level of the input to the filter, not the level of the output (Moore and Glasberg, 1987). This may not be strictly accurate (Rosen, Baker and Kramer, 1992; Glasberg, Moore and Stone, 1999; Baker and Rosen, 2006). Nevertheless, it appears that the lower side of the auditory filter broadens with increasing level whatever assumptions are made in analysing the data (Unoki *et al.*, 2006).

Figure 3.7 illustrates how the shape of the auditory filter varies with input level for a centre frequency of 1 kHz. In this figure, the output of the filter is plotted in decibels as a function of frequency, with the level of the input as parameter. At the centre frequency, the output level in dB was assumed to be equal to the input level in dB. In other words, the gain (the output level minus the input level) at the centre frequency was assumed to be 0 dB. The gain for frequencies well below the centre frequency is negative (the output level is lower than the input level), but the gain at these frequencies *increases* with increasing level. However, it would be more in accord with the functioning of the cochlea to assume that the gain is fixed for frequencies well below the centre frequency, and that the gain at the centre frequency decreases with increasing level (Ruggero *et al.*, 1997). Recall that the input-output function of the BM is linear for frequencies well below CF, but is compressive for frequencies close to CF (see Figure 1.10).

Figure 3.8 shows auditory filters with the same shapes as in Figure 3.7. However, in Figure 3.8 the shapes are plotted in terms of the relative filter gain, and the gain is assumed to be invariant with level for low frequencies; this is consistent with linear BM input-output functions for frequencies well below CF. When plotted in this way, the gain at the centre frequency is seen to decrease with increasing input level, consistent with a progressively reducing gain from the *active mechanism* with increasing level (see Chapter 1, Section IV.3). The dashed regions of the curves

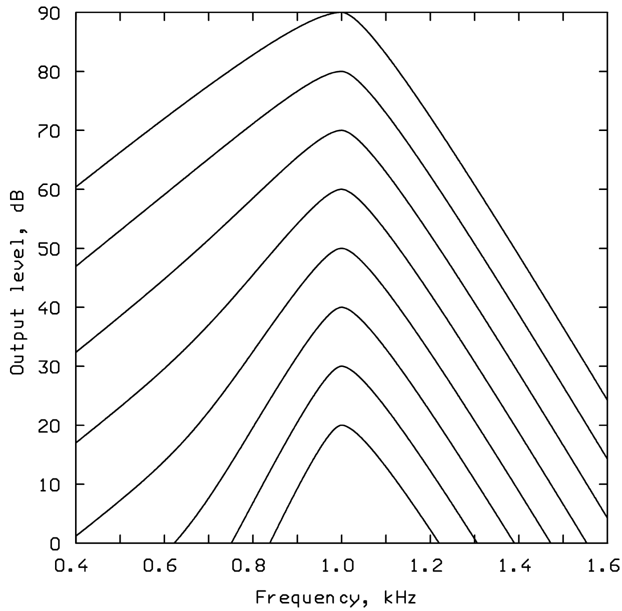


Figure 3.7. The shape of the auditory filter centred at 1 kHz, plotted for input sound levels ranging from 20 to 90 dB SPL/ERB$_N$ in 10-dB steps. The output of the filter is plotted as a function of frequency, assuming that the gain at the centre frequency is 0 dB. On the low-frequency side, the filter becomes progressively less sharply tuned with increasing sound level. At moderate sound levels, the filter is approximately symmetric on the linear frequency scale used.

are based on extrapolations. For example, when the input level is 20 dB, the filter response cannot be determined experimentally for regions more than 20 dB down from the tip.

A discrepancy with BM responses is observed for frequencies well above the centre frequency; BM input-output functions become nearly linear for input frequencies well above CF, whereas the filter gains shown in Figure 3.8 decrease with increasing level, indicating a *compressive nonlinearity*. It is possible that the discrepancy reflects errors of measurement. As mentioned earlier, the notched-noise method does not give a precise estimate of the slope of the steeper side of the auditory filter when the filter is markedly asymmetric. This is a particular problem at high sound levels, where the lower branch becomes very shallow. Thus, at high levels, there may well be significant errors in the estimates of the sharpness of the high-frequency side of the filter. Also, BM responses to tones well above CF are very small in amplitude, making it hard to measure the responses accurately. BM responses at these frequencies often show a 'plateau', a range of frequencies where the response hardly varies with frequency (Ruggero *et al.*, 1997). Such a plateau is not evident in neural tuning curves, so the effect in BM measurements may reflect an artefact of some kind.

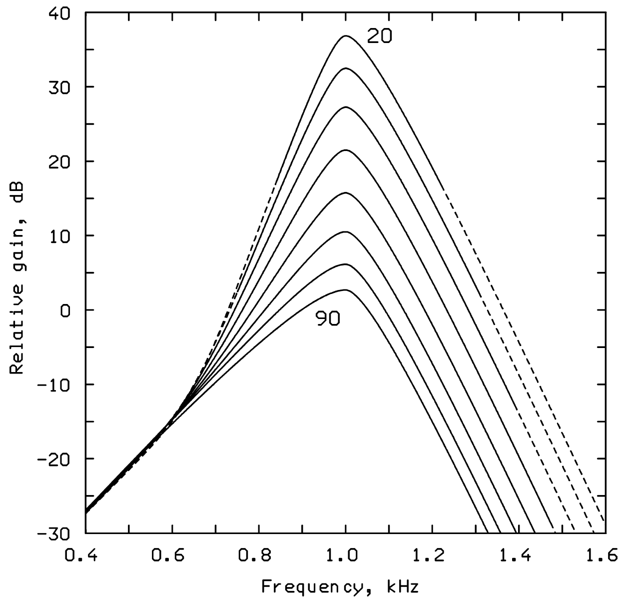


Figure 3.8. Auditory filters for the same input levels as in Figure 3.7. Here, the shapes are plotted in terms of the filter gain, and the gain is assumed to be invariant with level for low frequencies; this is consistent with linear BM input-output functions for frequencies well below CF. When plotted in this way, the gain at the centre frequency is seen to decrease with increasing input level, consistent with a progressively reducing gain from the active mechanism with increasing level. Regions of the curves which are dashed are based on extrapolation; the experimental data do not allow these regions of the filter responses to be determined.

IV.3 SUMMARY

In summary, the properties of auditory filters measured in masking experiments using notched noise are broadly consistent with the filtering measured on the BM and in the auditory nerve, in terms of variation with centre frequency and level. The variation of auditory filter shape with level can be interpreted as indicating a fixed gain on the low-frequency tail of the auditory filter and a gain that decreases with increasing level around the centre frequency. As a result, the filter broadens with increasing level.

V MASKING PATTERNS AND EXCITATION PATTERNS

V.1 MASKING PATTERNS

In the masking experiments described so far, the frequency of the signal was held constant, while the masker frequency was varied. These experiments are most appropriate for estimating the shape of the auditory filter at a given centre frequency.

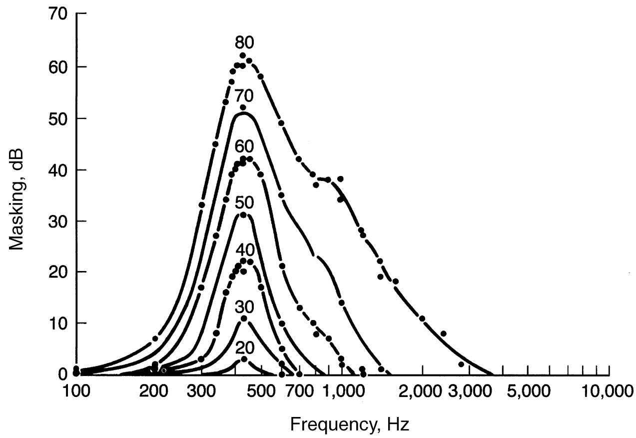


Figure 3.9. Masking patterns (masked audiograms) for a narrow band of noise centred at 410 Hz. Each curve shows the elevation in threshold of a sinusoidal signal as a function of signal frequency. The overall noise level for each curve is indicated in the figure. Data from Egan and Hake (1950).

However, in many experiments on masking, the signal frequency has been varied while the masker frequency was held constant. Typically, the masker level has been held constant and the signal level required for threshold has been measured as a function of the signal frequency (Wegel and Lane, 1924). The function relating masked threshold to the signal frequency is known as a *masking pattern*, or sometimes as a *masked audiogram*.

A typical set of results is shown in Figure 3.9; the data are taken from Egan and Hake (1950). In this figure, the signal threshold is expressed as the amount of masking, which is the amount by which the masked threshold exceeds the absolute threshold when both are expressed in decibels. Notice that on the high-frequency sides of the patterns the slopes become shallower at high levels. Thus, if the level of a low-frequency masker is increased by, say, 10 dB, the masked threshold of a high-frequency signal is elevated by more than 10 dB; the amount of masking grows nonlinearly (in an *expansive* way) on the high-frequency side. This has been called the upward spread of masking.

Masking patterns do not reflect the use of a single auditory filter. Rather, for each signal frequency the listener uses a filter centred close to the signal frequency. Thus, the auditory filter is shifted as the signal frequency is altered. One way of interpreting the masking pattern is as a crude indicator of the *excitation pattern* of the masker. The excitation pattern of a sound is a representation of the activity or excitation evoked by that sound as a function of CF or 'place' in the auditory system (Zwicker, 1970). It can be thought of as a kind of internal representation of the spectrum of the sound. In the case of a masking pattern, one might assume that the signal is detected when the excitation it produces is some constant proportion of the excitation produced by the masker in the frequency region of the signal; 'frequency region' here refers to places on the BM, or neurones, that have CFs close to the signal frequency. Thus, the threshold of the signal as a function of frequency is proportional to the masker's excitation level. The masking pattern should be parallel to the excitation pattern of the masker, but shifted vertically by a small amount. In practice, the situation is not so straightforward, since the shape of the masking pattern is influenced by factors such as the detection of beats (Moore, Alcántara and Dau, 1998), off-frequency listening and the detection of combination tones produced by the interaction of the signal and the masker (Greenwood, 1971).

V.2 RELATIONSHIP OF THE AUDITORY FILTER TO THE EXCITATION PATTERN

Moore and Glasberg (1983c) have described a method for deriving the shapes of excitation patterns using the concept of the auditory filter. They suggested that the excitation pattern of a given sound can be thought of as the output of the auditory filters as a function of their centre frequency. This idea is illustrated in Figure 3.10. The upper portion of the figure shows schematic auditory filter shapes for five centre frequencies. For each filter, the gain is plotted relative to the gain at the centre frequency, which was assumed to be 0 dB (as in Figure 3.5). A moderate input level was assumed; so each filter is symmetrical on the linear frequency scale used, but the bandwidths of the filters increase with increasing centre frequency. The dashed line represents a 1-kHz sinusoidal signal whose excitation pattern is to be derived. The lower panel shows the relative response of each filter in response to the 1-kHz signal, plotted as a function of the centre frequency of each filter; this is the desired excitation pattern.

To see how this pattern is derived, consider the output from the filter with the lowest centre frequency. This has a relative response to the 1-kHz tone of about −40 dB, as indicated by point 'a' in the upper panel. In the lower panel, this gives rise to the point 'a' on the excitation pattern; the point has an ordinate value of −40 dB and is positioned on the abscissa at a frequency corresponding to the centre frequency of the lowest filter illustrated. The relative responses of the other filters are indicated, in order of increasing centre frequency, by points 'b' to 'e', and each leads to a corresponding point on the excitation pattern. The complete excitation pattern was actually derived by calculating the filter outputs for filters spaced at 10-Hz intervals.

In deriving the excitation pattern, excitation levels were expressed relative to the level at the tip of the pattern, which was arbitrarily labelled as 0 dB. To calculate the excitation pattern for a 1-kHz tone with a level of, say, 60 dB, the level at the tip would be labelled as 60 dB, and all other excitation levels would correspondingly be increased by 60 dB. However, for the calculation to be valid, the filter shapes would have to be appropriate for an input level of 60 dB.

Note that, although the auditory filters were assumed to be symmetric on a linear frequency scale, the derived excitation pattern is asymmetric. This happens because the bandwidth of the auditory filter increases with increasing centre frequency.

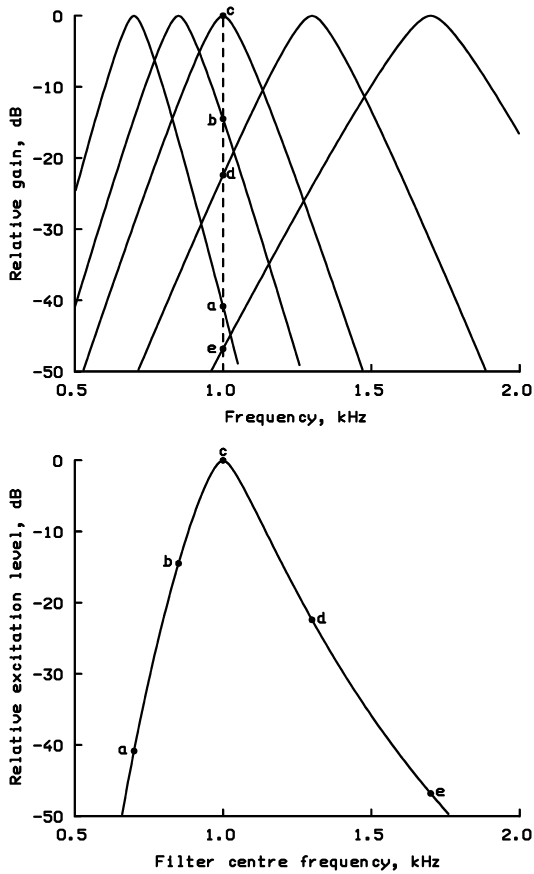


Figure 3.10. An illustration of how the excitation pattern of a 1-kHz sinusoid (dashed line) can be derived by calculating the outputs of the auditory filters as a function of their centre frequency. The top panel shows five auditory filters, centred at different frequencies, and the bottom half shows the calculated excitation pattern.

V.3 CHANGES IN EXCITATION PATTERNS WITH LEVEL

In the normal auditory system, the low-frequency side of the auditory filter becomes less steep with increasing sound level (Figures 3.7 and 3.8). Correspondingly, the excitation patterns evoked by narrowband sounds (such as sinusoids) become less steep on the high-frequency side. This is illustrated in Figure 3.11, which shows excitation patterns for a 1-kHz sinusoid at levels ranging from 20 to 90 dB, calculated in a manner similar to that described above but this time plotted on a logarithmic frequency scale. The excitation patterns have the same general form as the masking patterns shown in Figure 3.9, and show the same nonlinear expansive growth on the high-frequency side.

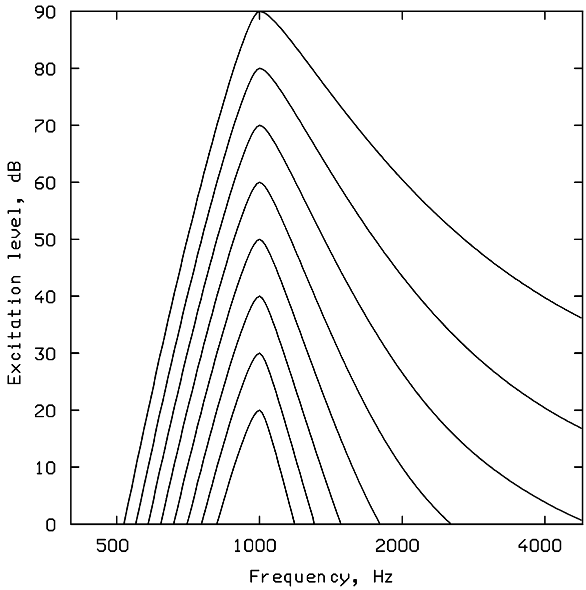


Figure 3.11. Excitation patterns calculated according to the procedure described in the text for 1-kHz sinusoids ranging in level from 20 to 90 dB SPL in 10-dB steps. The frequency scale is logarithmic.

In calculating the excitation patterns in this figure, it was assumed that the auditory filters had a gain of 0 dB at their tips for all input levels, as in Figure 3.7. However, as argued earlier, the gain at the tip probably decreases with increasing input level, as in Figure 3.8. Hence, the excitation patterns in Figure 3.11 are misleading: they do not directly reflect the magnitude of BM responses. On the BM, the magnitude of response at the tip of the excitation pattern grows in a compressive manner with increasing sound level, while the magnitude of response on the high-frequency side grows linearly. This effect is not directly revealed in simultaneous masking experiments, because the masked threshold depends mainly on the signal-to-masker ratio at the output of the auditory filter. Excitation patterns calculated as described here will be described as psychoacoustical excitation patterns, to distinguish them from excitation patterns on the BM.

V.4 POSSIBLE EFFECTS OF SUPPRESSION

As pointed out earlier, physiological measures of the tuning of the BM have usually been obtained using sinusoids whose frequency and intensity have been systematically varied. At any one time, only a single sinusoid is present. In contrast, the psychophysical measures of masking described above involved the presentation of more than one sinusoid at the same time. If the filtering on the BM were linear, and if the assumptions of the power-spectrum model of masking were valid, this should not matter; estimates of the auditory filter shape derived using complex stimuli should correspond well to the frequency-response shape measured with single sinusoids. However, the filtering on the BM is *not* linear, as described in Chapter 1.

In the power-spectrum model of masking, the masking of a signal by a masker with a different centre frequency is assumed to be caused by the spread of the excitation produced by the masker to the 'place' corresponding to the signal frequency (Wegel and Lane, 1924; Fletcher, 1940; Zwicker, 1970; Zwicker and Fastl, 1999). The masking of the signal is assumed to occur because auditory filters tuned close to the signal frequency are excited by the masker. However, an alternative possibility is that the signal is suppressed by the masker (Wightman, McGee and Kramer, 1977; Weber, 1983; Pickles, 1984; Delgutte, 1988, 1990; Moore and Vickers, 1997; Yasin and Plack, 2005). A masker may suppress the response to a signal that is remote in frequency without itself producing excitatory activity at the signal place (Hind *et al.*, 1967; Sachs and Kiang, 1968), a phenomenon described in Chapter 1 as *two-tone suppression*.

At present, the relative importance of the spread of excitation and suppression remains somewhat controversial. Overall, the results suggest that simultaneous masking might result from a combination of suppression and spread of excitation (Delgutte, 1990; Beveridge and Carlyon, 1996; Moore and Vickers, 1997). The spread of excitation probably dominates when the masker contains frequency components reasonably close to the signal frequency, while suppression plays a greater role when the masker frequency is remote from the signal frequency, and especially when it is well below the signal frequency.

As pointed out by Delgutte (1990, 1996), the evidence indicating that strong suppression can occur for masker frequencies well below the signal frequency has important implications for the interpretation of masking patterns, such as those shown in Figure 3.7. The signal threshold may seriously overestimate the excitation produced by the masker at the place corresponding to the signal frequency when the masker is much lower in frequency than the signal. Put another way, the high-frequency sides of excitation patterns may be more restricted, and have steeper slopes, than indicated by masking patterns.

VI NON-SIMULTANEOUS MASKING

VI.1 BASIC PROPERTIES OF NON-SIMULTANEOUS MASKING

The phrase simultaneous masking is used to describe situations where the masker is present for the whole time that the signal occurs. Masking can also occur when a brief signal is presented just before or after the masker; this is called non-simultaneous masking. This type of masking is introduced here since it can be used as a tool to measure the effects of suppression on the internal representation of sounds.

Two basic types of non-simultaneous masking can be distinguished: (1) *backward masking*, in which the signal precedes the masker (also known as pre-stimulatory masking) and (2) *forward masking*, in which the signal follows the masker (also known as post-stimulatory masking). The emphasis here will be on forward masking, since it is generally the greater of the two types of masking, at least in well-trained subjects, and since it is better understood than backward masking. Forward masking might occur via several different mechanisms. One is a reduction in sensitivity of recently stimulated neurones; the neurones stimulated by the masker may be 'fatigued', and this reduces the response to a signal that comes just after the masker (Meddis and O'Mard, 2005). A second possible mechanism is based on a persistence in the pattern of neural activity evoked by the masker at some level above the auditory nerve. The response to the masker may take some time to decay when the masker is turned off (Plomp, 1964b). If the signal is presented during this decay time, the persisting response to the masker may mask the response to the signal. A third mechanism is central inhibition evoked by the masker, which persists for some time after the masker is turned off (Brosch and Schreiner, 1997).

A fourth mechanism may sometimes play a role. The response of the BM to the masker takes a certain time to decay, and for small intervals between the signal and the masker this may result in forward masking (Duifhuis, 1973). This mechanism is probably only important for low frequencies (for which the decay time on the BM is longer) and when the masker-signal interval is less than a few milliseconds.

Figure 3.12 shows the results of an experiment on forward masking using broadband white noise as the masker (Moore and Glasberg, 1983a). The noise level is plotted in terms of the level within the ERB_N of the auditory filter centred at the signal frequency. The signal was a brief burst of a 4-kHz sinusoid, and the delay, *D*, of the signal after the end of the masker could have one of three values. In the figure, *D* is specified as the time from the end of the masker to the end of the signal. The signal threshold was measured for several different levels of the masker. The main properties of forward masking are as follows:

1. Forward masking is greater the nearer in time to the masker that the signal occurs. This is illustrated in the left panel of Figure 3.12. When the delay *D* of the signal is plotted on a logarithmic scale, the data fall roughly on a straight line. In other words, the amount of forward masking, in dB, is a linear function of log(*D*).
2. The rate of recovery from forward masking is greater for higher masker levels. Thus, regardless of the initial amount of forward masking, the masking decays to zero after 100–200 ms.

3. A given increment in masker level does not produce an equal increment in amount of forward masking. For example, if the masker level is increased by 10 dB, the masked threshold may only increase by 3 dB. This contrasts with simultaneous masking, where, for broadband noise maskers, threshold usually corresponds to a constant signal-to-masker ratio (Hawkins and Stevens, 1950). This effect can be quantified by plotting the signal threshold as a function of masker level. The resulting function is called a *growth-of-masking* function. Several such functions are shown in the right panel of Figure 3.12. In simultaneous masking, such functions would have slopes close to one (for a broadband noise masker); this slope is indicated by the dashed line. In forward masking, the slopes are less than one, and the slopes decrease as the value of *D* increases.
4. The amount of forward masking increases with increasing masker duration for durations of up to at least 20 ms. The results for greater masker durations vary somewhat across studies. Some studies show an effect of masker duration for durations of up to 200 ms (Kidd and Feth, 1982), while others show little effect for durations of beyond 50 ms (Fastl, 1976).

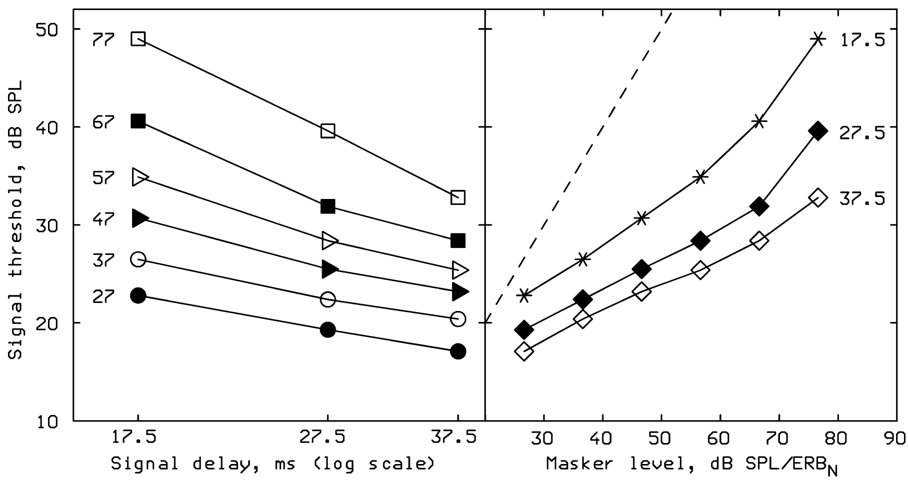


Figure 3.12. The left panel shows the threshold of a brief 4-kHz signal, plotted as a function of the time delay of the signal after the end of the noise masker. Each curve shows results for a different noise level; the noise level is specified in terms of the level within a 1-ERB$_N$-wide band centred at the signal frequency. The right panel shows the same thresholds plotted as a function of masker level. Each curve shows results for a different signal delay time (17.5, 27.5 or 37.5 ms). The dashed line has a slope of 1, indicating linear growth of masking. Note that the slopes of the growth-of-masking functions decrease with increasing signal delay. The data are taken from Moore and Glasberg (1983a).

VI.2 EVIDENCE FOR SUPPRESSION FROM NON-SIMULTANEOUS MASKING

Chapter 1 described how the response to a tone of a given frequency can sometimes be suppressed by a tone with a different frequency, a phenomenon known as two-tone suppression. For other complex signals, similar phenomena occur and are given the general name lateral suppression, or *suppression*. This can be characterized in the following way: strong activity at a given CF can suppress weaker activity at adjacent CFs.

I consider next the perceptual consequences of suppression, and how its effects can be measured. It was suggested above that suppression may have some influence on simultaneous masking, when the masker frequency is well below the signal frequency. However, Houtgast (1972) has argued that simultaneous masking is not an appropriate tool for detecting the effects of suppression. In simultaneous masking, the masking stimulus and the signal are processed simultaneously in the same auditory filter. He argued that any suppression applied to that filter will affect the neural activity caused by both the signal and the masker. In other words, the signal-to-masker ratio in a given frequency region (corresponding to a given place on the BM) will be unaffected by suppression, and thus the threshold of the signal will remain unaltered.

This argument may not be correct in all cases. Suppression might change the signal-to-masker ratio at the output of a given auditory filter if the signal and masker are widely separated in frequency. However, Houtgast's argument certainly seems reasonable when applied to cases where the signal and masker have coincident or near-coincident frequency components. Say, for example, that the masker consists of two frequency components, at 500 and 2000 Hz, while the signal is a sinusoid at 2000 Hz. If the signal and masker are presented simultaneously, then the 500 Hz component in the masker may produce suppression at the place on the BM tuned to 2000 Hz, but that suppression will be applied equally to the 2000-Hz component of the masker and to the 2000-Hz signal, without changing their ratio. If the masking of the signal is determined mainly by the 2000-Hz component of the masker, then the suppression produced by the 500-Hz component will have no measurable effects.

Houtgast suggested that the effects of suppression could be measured by presenting the masker and the signal successively, for example by using forward masking. It is assumed that the masked threshold of the signal is monotonically related to the excitation evoked by the masker in neurones tuned close to the signal frequency. Thus, increases in masker-evoked excitation will lead to increases in the signal threshold, while decreases in masker-evoked excitation (produced, for example, by suppression) will lead to decreases in signal threshold. It is also assumed that any suppression effects occurring within the masker (e.g. suppression of one frequency component in the masker by another frequency component within the masker) cease as soon as the masker is turned off.

Following the pioneering work of Houtgast (1972, 1973, 1974), many researchers have reported that there are systematic differences between the results obtained using simultaneous and non-simultaneous masking techniques. An extensive review is provided by Moore and O'Loughlin (1986). One major difference is that non-simultaneous masking reveals effects that can be directly attributed to suppression. A good demonstration of this involves a psychoacoustical analogue of two-tone suppression. Houtgast (1973, 1974) measured the threshold for detecting a 1-kHz signal following a 1-kHz non-simultaneous masker. He then added a second tone to the masker and measured the threshold again. He found that sometimes the addition of this second tone produced a reduction in the threshold, an effect that has been called two-tone unmasking. He attributed this to suppression of the 1-kHz component in the masker by the second component. If the 1-kHz component is suppressed, this reduces the excitation evoked by the masker in the frequency region around 1 kHz, producing a drop in the threshold for detecting the signal. Houtgast mapped out the combinations of frequency and intensity over which the 'suppressor' produced a reduction in signal threshold exceeding 3 dB. He found two regions, one above 1 kHz and one below, as illustrated in the right panel of Figure 3.13. The regions found were similar to the suppression areas which can be observed in single neurones of the auditory nerve, as illustrated in the left panel of Figure 3.13 (these data were also shown in Chapter 1). Similar results have been obtained by Shannon (1976).

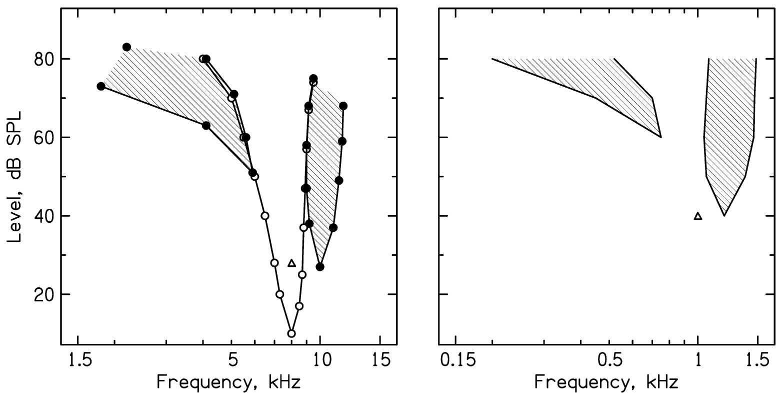


Figure 3.13. The left panel shows neurophysiological data from Arthur, Pfeiffer and Suga (1971). The open circles show the tuning curve (threshold versus frequency) of a single neurone with a CF at 8 kHz. The neurone was stimulated with a tone at CF, and just above threshold (indicated by the open triangle). A second tone was then added and its frequency and intensity varied. Any tone within the shaded areas bounded by the solid circles reduced the response to the tone at CF by 20 % or more. These are the suppression areas. The right panel shows the psychoacoustical data of Houtgast (1974) obtained using non-simultaneous masking. The signal was a 1-kHz sinusoid. The masker contained two components. One was a 1-kHz sinusoid which was fixed in level at 40 dB (indicated by the open triangle). The second component (the 'suppressor') was a sinusoid which was varied both in frequency and in level. The shaded areas indicate combinations of frequency and level where the second tone reduced the threshold by 3 dB or more.

Under some circumstances, the reduction in threshold (unmasking) produced by adding one or more extra components to a masker can be partly explained in terms of additional cues provided by the added components, rather than in terms of suppression. Specifically, in forward masking the added components may reduce 'confusion' of the signal with the masker by indicating exactly when the masker ends and the signal begins (Moore, 1980, 1981; Moore and Glasberg, 1982, 1985; Neff, 1985; Moore and O'Loughlin, 1986). This may have led some researchers to overestimate the magnitude of suppression as indicated in non-simultaneous masking experiments. However, not all unmasking can be explained in this way.

VI.3 THE ENHANCEMENT OF FREQUENCY SELECTIVITY REVEALED IN NON-SIMULTANEOUS MASKING

A second major difference between simultaneous and non-simultaneous masking is that the frequency selectivity revealed in non-simultaneous masking is greater than that revealed in simultaneous masking. For example, PTCs determined in forward masking are typically sharper than those obtained in simultaneous masking (Moore, 1978), as illustrated in Figure 3.14. A low-level notched-noise masker was gated with

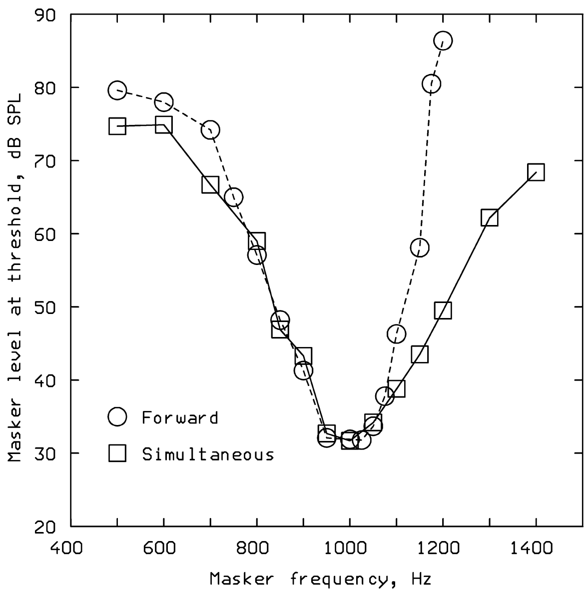


Figure 3.14. Comparison of PTCs determined in simultaneous masking (squares) and forward masking (circles). A low-level notched noise was gated with the masker to provide a consistent detection cue in forward masking and to restrict off-frequency listening. Data from Moore, Glasberg and Roberts (1984).

the main narrowband noise masker to restrict off-frequency listening and to provide a consistent detection cue in forward masking. The difference between the two curves is particularly marked on the high-frequency side.

Two different explanations have been proposed for the sharper tuning found in forward masking. According to Houtgast (1974), the sharper tuning arises because the internal representation of the masker (its excitation pattern) is sharpened by a suppression process, with the greatest sharpening occurring on the low-frequency side. In simultaneous masking, the effects of suppression are not seen, since any reduction of the masker activity in the frequency region of the signal is accompanied by a similar reduction in signal-evoked activity. In other words, the signal-to-masker ratio in the frequency region of the signal is unaffected by the suppression. In forward masking, on the other hand, the suppression does not affect the signal. For maskers with frequencies above that of the signal, the effect of suppression is to sharpen the excitation pattern of the masker, resulting in an increase of the masker level required to mask the signal. Thus, the suppression is revealed as an increase in the slopes of the PTC.

An alternative explanation is that described earlier; in simultaneous masking, the signal may be suppressed by the masker, and this increases the effectiveness of the masker for masker frequencies well above and below the signal frequency. In non-simultaneous masking, the masker does not suppress the signal, and so the masker is less effective (Delgutte, 1988, 1990). Hence, when determining a PTC, the masker level has to be increased on the skirts of the PTC.

Whichever of these two explanations is correct, the same inference can be drawn: the tuning curve measured in non-simultaneous masking gives a more accurate indication of the tuning that occurs for single sinusoids. Thus, the PTC measured in non-simultaneous masking is likely to be closely related to BM tuning curves and neural tuning curves, while the PTC measured in simultaneous masking is likely to be broader than those curves.

VI.4 RELATION BETWEEN THE GROWTH OF FORWARD MASKING AND THE BASILAR MEMBRANE INPUT-OUTPUT FUNCTION

As described earlier, growth-of-masking functions in forward masking usually have slopes less than one; when the masker level is increased by X dB, the signal level at threshold usually increases by less than X dB. Oxenham and Moore (1995) have suggested that the shallow slopes of the growth-of-masking functions can be explained, at least qualitatively, in terms of the compressive input-output function of the BM. Such an input-output function is shown schematically in Figure 3.15. It has a shallow slope for medium input levels, and a steeper slope for very low input levels. Assume that, for a given time delay of the signal relative to the masker, the response evoked by the signal at threshold is directly proportional to the response evoked by the masker. Assume, as an example, that a masker with a level of 40 dB produces a signal threshold of 10 dB. Consider now what happens when the masker level is

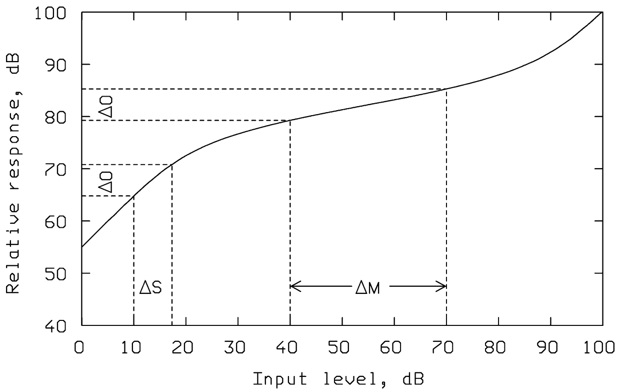


Figure 3.15. The curve shows a schematic input-output function on the BM. When the masker is increased in level by ΔM, this produces an increase in response of ΔO. To restore signal threshold, the response to the signal also has to be increased by ΔO. This requires an increase in signal level, ΔS, which is markedly smaller than ΔM.

increased by 30 dB. The increase in masker level, denoted by ΔM in Figure 3.15, produces a relatively small increase in response or output, ΔO. The small increase occurs because the masker level falls in a range where the input-output function of the BM is highly compressive (i.e. it has a shallow slope). To restore the signal to threshold, the signal has to be increased in level so that the response to it increases by ΔO. However, this requires a relatively small increase in signal level, ΔS, as the signal level falls in the range where the input-output function is relatively steep. Thus, the growth-of-masking function has a shallow slope.

According to this explanation, the shallow slope arises from the fact that the signal level is lower than the masker level; so the masker is subject to more compression than the signal. The difference in compression applied to the masker and signal increases with increasing difference in level between the masker and signal. Hence, the slope of the growth-of-masking function should also decrease with increasing difference in level between the masker and signal. This can account for the progressive decrease in the slopes of the growth-of-masking functions with increasing time delay between the signal and masker (see Figure 3.12); longer time delays are associated with greater differences in level between the signal and masker.

A prediction of this explanation is that the growth-of-masking function for a given signal time delay should increase in slope if the signal level is high enough to fall in the highly compressive region of the input-output function. Such an effect can be seen in the growth-of-masking function for the shortest delay time in Figure 3.12; the function steepens for the highest signal level. The effect is also apparent, but less clearly so, for the longer time delays of the signal.

For the data shown in Figure 3.12, the signal was relatively long; its overall duration was 20 ms. For such a signal, the signal level at threshold is generally below the masker level. It is this difference in level that gives rise to the shallow slopes of the growth-of-masking functions. However, if the signal is made very brief, and it is presented immediately after the end of the masker, the signal level at threshold may be close to the masker level. Under these conditions, the growth-of-masking function is predicted to have a slope of one; each 10-dB change in level should lead to a 10-dB change in signal threshold.

Oxenham and Plack (1997) performed just such an experiment. They measured forward masking for a 6-kHz sinusoidal masker and a signal of the same frequency. They used a very brief signal and a very short time delay between the masker and signal. The level of the signal at threshold was found to be approximately equal to the masker level, and the growth-of-masking function had a slope of one; each 10-dB increase in masker level was accompanied by a 10-dB increase in signal level. This is illustrated in Figure 3.16 (filled symbols) and is consistent with the explanation offered above.

In another condition, they used a masker frequency well *below* the signal frequency. The masker frequency was 3 kHz instead of 6 kHz. In this condition, the growth-of-masking function had a slope much *greater* than one; a 10-dB increase in masker level

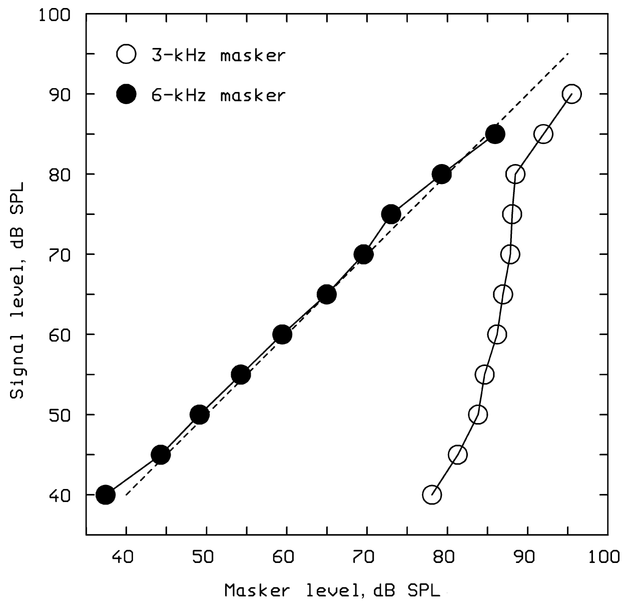


Figure 3.16. Data from Oxenham and Plack (1997) for normally hearing subjects. Thresholds for detecting a 6-kHz signal following a 3-kHz or 6-kHz masker are shown. The signal level was fixed, and the masker level was varied to determine the threshold. Symbols represent the mean thresholds of three normally hearing subjects. The dashed line has a slope of one; linear growth of masking would give data parallel to this line.

was accompanied by a 40-dB increase in signal level, as shown by the open symbols in Figure 3.16. This can be explained in the following way: the signal threshold depends on the response evoked by the masker at the CF corresponding to the signal frequency. The growth of response on the BM for sinusoids with frequencies well below the CF is linear. Thus, the signal is subject to compression while the masker is not (essentially the opposite of the situation illustrated in Figure 3.15). This gives rise to the steep growth-of-masking function.

The ratio of the slopes of the growth-of-masking functions for the 6-kHz masker and the 3-kHz masker reflects the difference in slope of the BM input-output function for a tone at the CF (6 kHz) and a tone well below CF (3 kHz). The slopes actually differ by about a factor of four. Assuming that the response to the 3-kHz tone is linear (at the place with CF = 6 kHz), this implies that the input-output function for the 6-kHz tone has a slope of about 0.25 for mid-range sound levels. This is consistent with direct estimates of the slopes of BM input-output functions, as described in Chapter 1, Section IV.3.

VII THE AUDIBILITY OF PARTIALS IN COMPLEX TONES

A *complex tone* is a sound that contains many sinusoidal components and that evokes a subjective sensation of *pitch*. The sounds produced by musical instruments are nearly always complex tones. When the *waveform* of a sound repeats regularly as a function of time, the sound is said to be *periodic*. For such sounds, the sinusoidal components have frequencies that are integer multiples of a common *fundamental frequency* (F0). The components are referred to as *harmonics*. The harmonics are numbered, the first harmonic being the fundamental component. For example, the note A4 played on an oboe would have an F0 (first harmonic) of 440 Hz, a second harmonic of 880 Hz, a third harmonic of 1320 Hz and so on. Some sounds contain frequency components whose frequencies are not integer multiples of a common fundamental. However, they may still evoke a pitch. Examples are the tones produced by bells and gongs. Such sounds are called inharmonic complex tones. The term *partial* is used to refer to any frequency component in a complex tone, whether it is a harmonic or not.

To a limited extent, human listeners are able to hear pitches corresponding to the individual partials in complex tones. This is not the usual way that complex tones are perceived; rather, it is common to hear a single pitch corresponding to the F0; see Chapter 6, Section VII.1. However, it is possible to ‘hear out’ some individual partials if attention is directed in an appropriate way.

Plomp (1964a) and Plomp and Mimpen (1968) investigated the limits of the ability to hear out partials in complex tones. They used tones with twelve sinusoidal components. The listener had a three-position switch that they could use to hear either the complex tone or one of two comparison sinusoidal tones. One of the comparison tones had the same frequency as a partial in the complex; the other lay halfway between that frequency and the frequency of the adjacent higher or lower partial. The listener had to judge which of these two comparison tones was a component

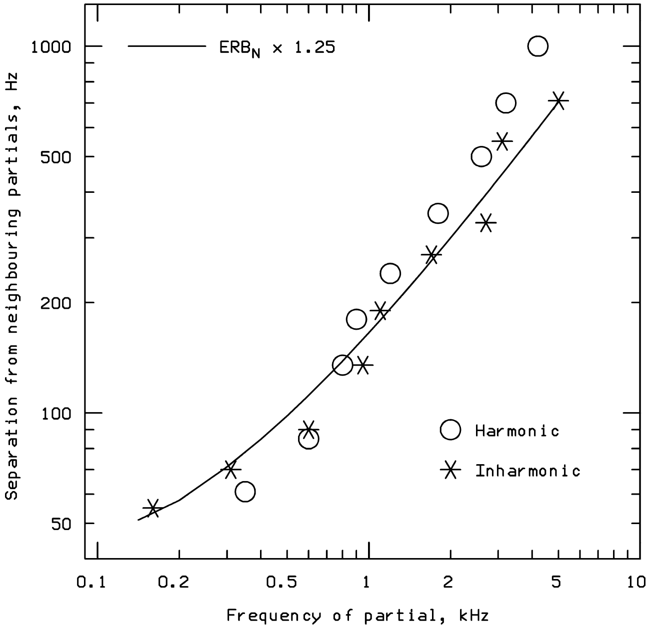


Figure 3.17. Data from Plomp (1964a) and Plomp and Mimpen (1968) on the audibility of partials in complex tones. The symbols show the frequency separation of a partial from neighbouring partials necessary for that partial to be heard out with 75 % accuracy, plotted as a function of the frequency of the partial. The solid curve shows the value of ERB$_N$ multiplied by 1.25, plotted as a function of frequency.

of the complex. Plomp and Mimpen used two types of complex tone: a harmonic complex tone containing harmonics 1 to 12, where the frequencies of the components were integer multiples of that of the fundamental, and an inharmonic complex tone, where the frequencies of the components were mistuned from simple frequency ratios. Figure 3.17 shows their results, plotted in terms of the frequency separation between a given partial and neighbouring partials necessary for that partial to be heard out with 75 % accuracy.

Plomp and Mimpen's data are consistent with the hypothesis that a partial can just be heard out from neighbouring partials when it is separated from those partials by 1.25 times the value of ERB$_N$ at the frequency of the partial. The solid line in Figure 3.17 shows the value of ERB$_N$ multiplied by 1.25. This line fits the data rather well. The data of Moore and Ohgushi (1993) on the audibility of partials in inharmonic complex tones are also consistent with the idea that partials can be heard out with about 75 % accuracy when the partials are separated by about 1.25 ERB$_N$. Separations greater than 2 ERB$_N$ are sufficient to give near-perfect scores, while separations less than 1 ERB$_N$ lead to very low scores. However, partials with frequencies above about

3000 Hz become difficult to hear out even when they are separated by more than 2 ERB_N (Moore *et al.*, 2006b). This suggests that the ability to hear out partials depends not only on frequency selectivity but also on *phase locking* (which declines at high frequencies). Partials that can be heard out are often described as 'resolved', while partials that cannot be heard out are described as 'unresolved'.

For harmonic complex tones, the lower harmonics are separated by more than 1.25 ERB_N. For example, for the note A4 mentioned above, the separation between the second and third harmonics (880 and 1320 Hz) corresponds to about 3 ERB_N. However, the higher harmonics are separated by less than 1.25 ERB_N. For example, the separation between the ninth and tenth harmonics (3960 and 4400 Hz) corresponds to only 0.9 ERB_N. Generally, about the first to the fifth harmonics in a complex tone can be heard out reasonably well. Harmonics above the eighth are very difficult to hear out. Between the fifth and eighth harmonics there is a transition region, where the harmonics can be heard out to some extent; they are partially resolved. These results have important implications for theories of the pitch perception of complex tones, as described in Chapter 6, Section VII.2.

Unpublished studies conducted in my laboratory indicate that people with cochlear hearing loss have great difficulty in hearing out partials from complex sounds, as would be expected from the reduced frequency selectivity commonly associated with cochlear hearing loss (which is described in more detail below). This is discussed further in Chapter 6, Sections IX.1 and IX.2.

VIII EFFECTS OF COCHLEAR DAMAGE ON FREQUENCY SELECTIVITY IN SIMULTANEOUS MASKING

VIII.1 COMPLICATING FACTORS

Comparisons of frequency selectivity in normally hearing and hearing-impaired people are complicated by several factors. One factor is the sound level of the stimuli used. As described above, the auditory filters of normally hearing subjects broaden on the low-frequency side with increasing level. This effect probably depends on the active mechanism in the cochlea. At low sound levels, this mechanism probably strongly enhances the tuning on the low-frequency side of the auditory filter. However, as the sound level increases, the contribution of the active mechanism is progressively reduced and the low-frequency side of the auditory filter broadens.

The active mechanism is often damaged or completely non-functioning in ears with cochlear damage, as described in Chapter 1, Sections VII.1 and VII.2. Hence, changes in frequency selectivity with level are absent or much less pronounced (Moore, Laurence and Wright, 1985; Stelmachowicz *et al.*, 1987; Nelson and Schroder, 1997). As a result, the differences between normally hearing and hearing-impaired subjects tend to decrease at high sound levels.

A second complicating factor is off-frequency listening. Some measures of frequency selectivity, especially PTCs, can be strongly influenced by off-frequency listening. More importantly, the role of off-frequency listening may vary markedly depending on the sensation level (SL) of the stimuli and the frequency selectivity of the subject. For example, if PTCs are measured for a normally hearing subject using a signal at 70 dB SPL, there is a strong potential for off-frequency listening in some conditions. On the other hand, if a hearing-impaired subject with an absolute threshold of 60 dB SPL is tested with a 70 dB SPL signal, the potential for off-frequency listening is much reduced. In practice, the extent to which off-frequency listening affects masked thresholds for hearing-impaired people depends on the way that absolute thresholds vary with frequency. Although these problems can be reduced by gating a notched noise with the variable masker, as described earlier, this has not usually been done in studies with hearing-impaired subjects. Hence, in what follows, studies of PTCs will be described only briefly, while measures controlling off-frequency listening will be described in more detail.

VIII.2 PSYCHOPHYSICAL TUNING CURVES

There have been many studies comparing PTCs in normal-hearing subjects and subjects with cochlear hearing loss (Leshowitz, Linstrom and Zurek, 1975; Hoekstra and Ritsma, 1977; Zwicker and Schorn, 1978; Bonding, 1979b; Florentine *et al.*, 1980; Tyler, Wood and Fernandes, 1982; Carney and Nelson, 1983; Festen and Plomp, 1983; Stelmachowicz *et al.*, 1985; Nelson, 1991; Kluk and Moore, 2004, 2005). Although the studies differ in detail, their results usually show that PTCs are broader than normal in the hearing-impaired subjects; an example was given in Figure 3.1. However, it is difficult to quantify the differences from normal, owing to the problems discussed above. Most studies have found that the sharpness of tuning of PTCs decreases with increasing absolute threshold, although the correlation between threshold and sharpness of tuning varies markedly across studies. No systematic differences in PTCs have been reported between cochlear losses of different origin, such as noise-induced, Ménière's disease, ageing and hereditary losses.

Sometimes, PTCs have been found whose tips are shifted well away from the signal frequency (Thornton and Abbas, 1980; Florentine and Houtsma, 1983; Turner, Burns and Nelson, 1983). This can happen when the signal frequency falls in a dead region. The most-studied situation of this type is when there is a low-frequency dead region, that is when there is severe damage to the inner hair cells (IHCs) at the apical end of the cochlea (see Chapter 2, Section IV). The detection of low-frequency tones is then mediated by neurones with high CFs. If the signal to be detected has a frequency corresponding to the dead region, the tip of the tuning curve lies well above the signal frequency. In other words, a masker centred well above the signal frequency is more effective than a masker centred close to the signal frequency. This happens because the higher-frequency masker lies closer to the CFs of the neurones mediating detection of the signal.

An example of a PTC obtained from a subject with a presumed mid-frequency dead region is shown in Figure 3.18 (Moore and Alcántara, 2001). The absolute thresholds, specified as the sound pressure level at the eardrum, are shown by crosses.

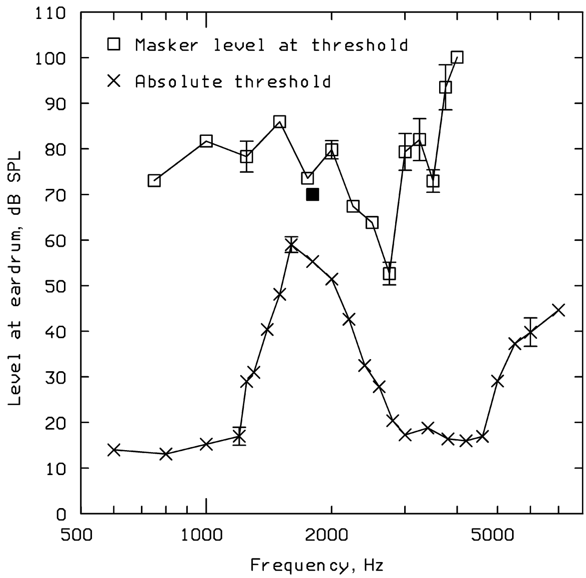


Figure 3.18. Data obtained from a subject with a presumed mid-frequency dead region. The crosses indicate the absolute thresholds. The open squares show a PTC obtained using a sinusoidal signal and a narrowband noise masker. The signal had a frequency of 1800 Hz. It was presented at 15 dB SL (solid square).

Absolute thresholds for frequencies from 600 to 1200 Hz and from 3000 to 5000 Hz are nearly normal, but there is a distinct threshold elevation between 1300 and 2800 Hz. Probably, absolute thresholds for frequencies in the range 1300–2800 Hz were determined by excitation spreading to lower CFs (for frequencies from 1300 to 1600 Hz) or higher CFs (for frequencies from 1800 to 2800 Hz). For determination of the PTC (open squares), the signal had a frequency of 1800 Hz and it was presented at a level 15 dB above the absolute threshold, as indicated by the filled square. The tip of the PTC falls well above the signal frequency. In fact, the tip falls at 3000 Hz, which is the frequency at which absolute threshold has returned to a nearly normal value.

PTCs with shifted tips have also been found for people with high-frequency dead regions. In these cases, the tips of the PTCs are shifted downwards (Moore *et al.*, 2000; Kluk and Moore, 2005). An example is shown in Figure 3.19.

PTCs have sometimes been found that are W-shaped rather than V-shaped (Hoekstra and Ritsma, 1977). Two factors seem to contribute to producing these W-shaped PTCs. The first is the detection of beats caused by the interaction of the signal and the masker (Kluk and Moore, 2005). Beats are easy to detect when the frequency of the signal is close to that of the masker, but do not occur when the signal and masker frequencies are equal. This leads to a local minimum in the PTC

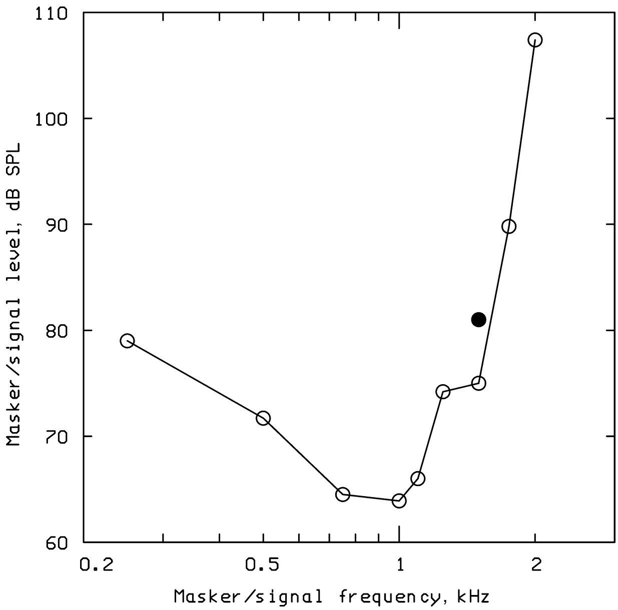


Figure 3.19. A PTC obtained from a subject with a high-frequency dead region. The signal frequency was 1.5 kHz.

at the signal frequency. The PTC may have additional minima at masker frequencies for which the beats become too rapid to provide a useful cue, or, for subjects with a dead region, at the CF of the place where the signal is detected, that is at the boundary of the dead region. The influence of beats as a cue can be reduced by increasing the bandwidth of the masker. This is illustrated in Figure 3.20. Note that the PTC has two tips for the sinusoidal masker but has only a single broad minimum for the masker with a bandwidth of 320 Hz.

A second way in which W-shaped PTCs can arise is via the interaction of the signal and masker to give a combination tone (or combination band of noise, if the masker is a band of noise). The combination tone involved is the simple difference tone, with frequency equal to the difference between the masker frequency and the signal frequency, and it can be more easily audible than the signal when the listener has good low-frequency hearing and poor high-frequency hearing. The difference tone cannot be heard when the signal frequency equals the masker frequency (because the difference tone would then have a frequency of 0 Hz), but it can be heard when the signal frequency is close to the masker frequency. This gives rise to a local minimum in the PTC at the signal frequency, again with a second minimum at the CF of the place where the signal is detected. The difference tone can be masked using a lowpass noise (Kluk and Moore, 2005), and when this is done the minimum in the PTC at the signal frequency is eliminated, leaving only a single minimum. This is illustrated in Figure 3.21.

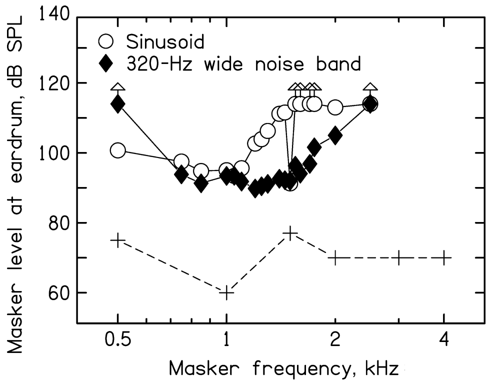


Figure 3.20. Illustration of the influence of the bandwidth of the masker on PTCs for a hearing-impaired subject, using a signal frequency of 1500 Hz. The + symbols indicate the absolute thresholds. The open circles show the PTC obtained using a sinusoidal masker. It has two tips, including a very sharp one at the signal frequency. The filled diamonds show the PTC obtained using a 320-Hz wide band of noise as a masker. This PTC shows only a single broad minimum, which is shifted slightly below the signal frequency. Up-pointing arrows indicate that the required masker level exceeded the maximum possible value. Data from Kluk and Moore (2005).

VIII.3 AUDITORY FILTER SHAPES MEASURED WITH NOTCHED NOISE

Auditory filter shapes of subjects with cochlear impairments have been estimated in several studies using notched-noise maskers (Tyler *et al.*, 1984; Glasberg and Moore, 1986; Dubno and Dirks, 1989; Laroche *et al.*, 1992; Peters and Moore, 1992; Stone, Glasberg and Moore, 1992; Leek and Summers, 1993; Sommers and Humes, 1993; Leeuw and Dreschler, 1994). The results generally agree in showing that auditory filters are broader than normal in hearing-impaired subjects and that, on average, the degree of broadening increases with increasing hearing loss.

Consider, as an example, the study of Glasberg and Moore (1986). They measured auditory filter shapes in subjects with unilateral and bilateral cochlear impairments; in the former case, differences between the normal and impaired ears cannot be attributed to extraneous factors such as age or ability to concentrate. The same noise spectrum level (50 dB, corresponding to an overall level of 79 dB SPL at 1 kHz) was used for the normal and impaired ears. Results for six subjects at a centre frequency of 1 kHz are shown in Figure 3.22. The filter shapes for the normal ears (upper panel) are similar to one another, and all show a high degree of frequency selectivity. The filter shapes for the impaired ears (lower panel) vary greatly in shape across subjects, but they are all broader than those for the normal ears on at least one side. The most

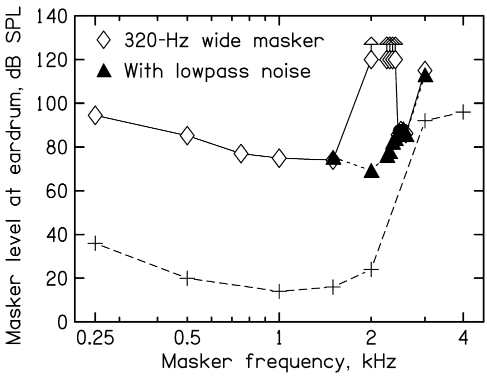


Figure 3.21. Illustration of the influence of combination-tone detection on PTCs for a subject with a high-frequency dead region. The + symbols indicate the absolute thresholds. The signal frequency was 2.5 kHz. The open diamonds show the PTC obtained using a 320-Hz wide noise masker. The filled triangles show the PTC obtained when a lowpass noise was added to the 320-Hz wide masker. Up-pointing arrows indicate that the required masker level exceeded the maximum possible value. Data from Kluk and Moore (2005).

common feature is a marked broadening on the low-frequency side. This indicates a marked susceptibility to masking by low-frequency sounds, such as car noise and air-conditioning noise.

Figure 3.23 summarizes values of the ERB of the auditory filter obtained from subjects with cochlear hearing loss in my laboratory and in the laboratory of Robert Peters. The summary was prepared by Michael Stone. The upper panel shows ERB values plotted relative to the value for young normally hearing subjects at moderate sound levels, as given by Equation 3.1. The lower panel shows ERB values plotted relative to the value for young normal subjects tested at the same sound pressure level as the impaired subjects, assuming that the ERB for normal subjects varies with level as described by Glasberg and Moore (1990). In both cases, the ERB values are plotted as a function of the absolute threshold (dB HL) at the test frequency. There is a clear trend for the ERB values to increase with increasing absolute threshold. The increase is less in the lower panel, since the auditory filters for normal subjects broaden with increasing level of the test stimuli. However, the trend is still quite clear. There is also considerable scatter in the data, indicating that the ERB of the auditory filter cannot be predicted reliably from the absolute threshold.

The scatter in the data shown in Figure 3.23 may reflect different ways in which cochlear damage can occur, as described in Chapter 1, Section VII.3 (Liberman and Dodds, 1984; Liberman, Dodds and Learson, 1986). In one common case, the damage is primarily to the OHCs and the IHCs are intact. This results in reduced effectiveness of the active mechanism, but a relatively normal mechanism for transducing BM movement into neural responses. In this case, the degree of auditory-filter broadening

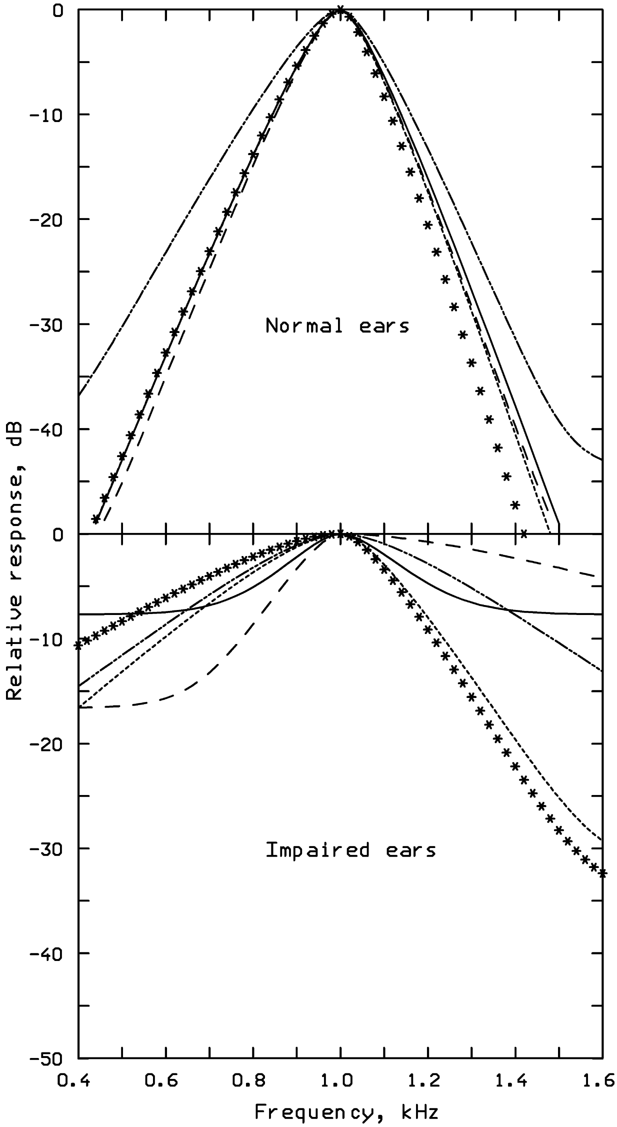


Figure 3.22. Auditory filter shapes at a centre frequency of 1 kHz for the normal (top) and impaired (bottom) ears of six subjects with unilateral cochlear impairments.

is correlated with the loss of sensitivity (elevated absolute threshold), and the broadening of the auditory filter can occur on both sides or mainly on the low-frequency side.

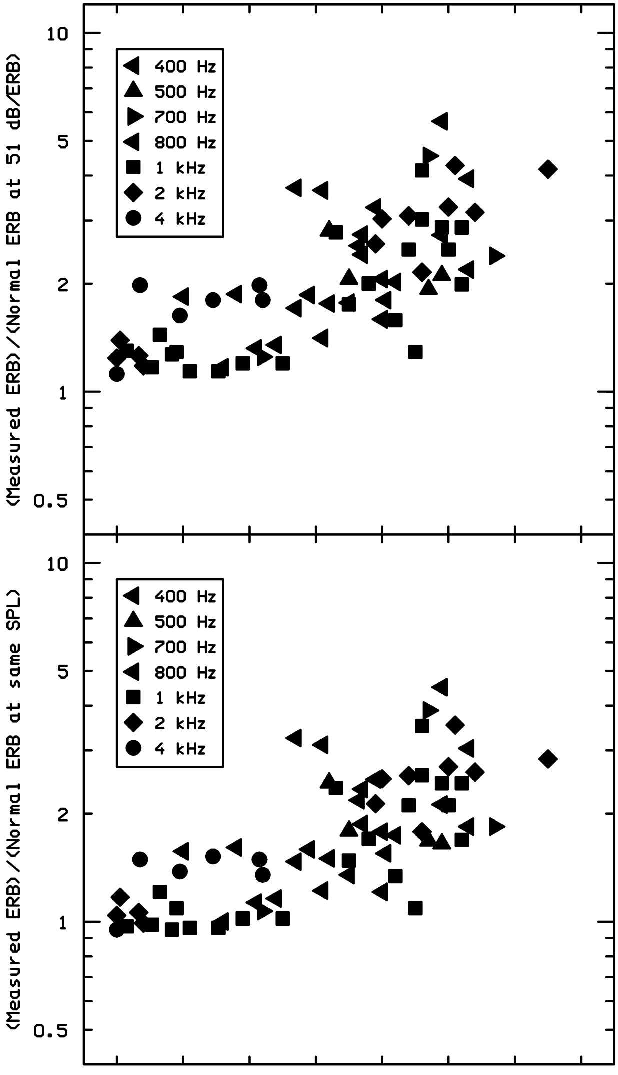


Figure 3.23. Values of the ERB of the auditory filter for subjects with cochlear hearing loss, plotted as a function of absolute threshold (dB HL) at the test frequency. The ERB values are plotted either relative to the values given by Equation 3.1, or relative to the ERBs for young normally hearing subjects tested at the same sound level.

In a second case, both the OHCs and IHCs are damaged. Both the active mechanism and the transducer mechanism are adversely affected. This results in a considerable elevation of absolute threshold and the filter is normally broadened on both sides. In a final (rare) case, the damage is mainly to the IHCs, with minimal damage to the OHCs. The transducer mechanism is severely affected in this case, but the active mechanism still functions, albeit with reduced effectiveness because of the high sound level needed to evoke a neural response. In this case, the absolute threshold is elevated, but the sharpness of tuning is only slightly worse than normal.

In summary, all methods of estimating frequency selectivity suggest that, on average, it decreases with increasing hearing loss above about 30 dB. However, for a given elevation in absolute threshold, there can be considerable scatter both in the degree of broadening and in the asymmetry of the auditory filter. This large variability may arise from different types of hair cell dysfunction within the cochlea.

IX THE USE OF MASKING TO DIAGNOSE DEAD REGIONS

IX.1 THE THRESHOLD-EQUALIZING NOISE (TEN) TEST

Moore *et al.* (2000) have described a simple method for diagnosing dead regions in the clinic. The method involves measuring the threshold for detecting a sinusoidal tone presented in a special background noise called threshold-equalizing noise (TEN), and so the method is called the TEN test. The noise is spectrally shaped so that the masked threshold, specified in dB SPL, is approximately the same for all frequencies in the range 250–10 000 Hz for people with normal hearing. The noise level is specified in terms of the level in a 1-ERB_N-wide band centred at 1000 Hz (i.e. the level in the frequency range 935–1065 Hz). For normally hearing subjects, the masked threshold is approximately equal to the TEN level/ERB_N when the threshold is measured using a method similar to that used for determining the audiogram in clinical audiology.

For people with hearing loss, but without any dead region, the masked threshold in the noise is usually slightly (2–5 dB) higher than normal. However, when the signal frequency falls in a dead region, the signal is only detected when it produces sufficient vibration at a remote region in the cochlea where there are surviving IHCs and neurones. The amount of vibration at this remote region is less than in the dead region, and so the noise is very effective in masking it. Thus, the signal threshold is expected to be markedly higher than normal. This is illustrated in Figure 3.24 for a hypothetical case of a person with a dead region extending from 1.07 kHz upwards (indicated by the shaded areas). It is assumed that there is some hearing loss below 1.07 kHz, and the amount of BM vibration required for detection of a signal is indicated by the horizontal dashed line at 40 dB. The signal to be detected is a 1.5 kHz tone. In quiet (upper panel) the signal has to be presented at a level of about 67 dB SPL for the BM vibration evoked by the signal to be detected; detection occurs at the place tuned just below 1.07 kHz. The lower panel illustrates the situation when TEN with a level of 70 dB SPL/ERB_N is presented. This produces a roughly constant

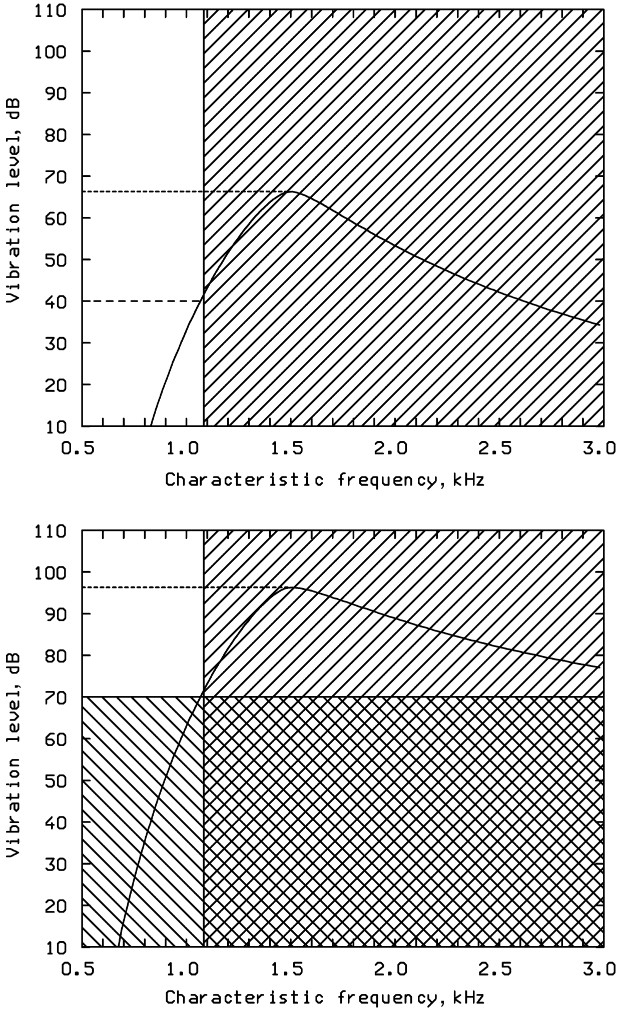


Figure 3.24. Illustration of the rationale behind the TEN test for diagnosis of dead regions, for a hypothetical person with a high-frequency dead region starting at 1.07 kHz. Envelopes of BM vibration patterns are illustrated for a 1.5-kHz tone at the absolute threshold (top panel) and at masked threshold in TEN with a level of 70 dB SPL/ERB$_N$ (bottom panel). All vibration levels are expressed relative to the vibration level produced by a 1-kHz tone at 0 dB SPL.

level of BM vibration at all CFs of about 70 dB. For the signal to be detected, the signal level has to be increased so that its vibration level just reaches that produced by the TEN at the place where the signal is detected. This requires a signal level of about 97 dB. Thus, the threshold for detecting the signal in the TEN is much higher than normal.

To validate the test, Moore *et al*. (2000) measured thresholds in the TEN and PTCs in the same subjects. PTCs with shifted tips (indicating that the signal frequency fell in a dead region) were usually associated with thresholds in the TEN that were 10 dB or more higher than normal. When the tips of the PTCs were not shifted, thresholds in the TEN were usually within 10 dB of the normal value (equal to the level/ERB_N of the TEN). This led Moore *et al.* to suggest the following criteria for diagnosing a dead region at the signal frequency:

1. The masked threshold should be 10 dB or more above the level/ERB_N of the TEN.
2. The TEN should produce at least 10 dB of masking, that is the masked threshold in the TEN should be at least 10 dB above the absolute threshold.

The second criterion was introduced since it is important that the masked threshold is actually determined by the TEN, rather than by the elevated absolute threshold. If the TEN does not produce at least 10 dB of masking, then the result must be regarded as inconclusive and the test should be repeated (if possible) using a higher level of the TEN. The TEN test was recorded onto compact disc (CD), with the intention that the CD is replayed via a two-channel audiometer in order to administer the test.

IX.2 THE TEN(HL) TEST

There were several problems with the version of the TEN test described above. The first was that levels were specified in dB SPL, which meant that absolute thresholds had to be measured twice, once in dB HL (as is done in routine clinical practice) and once in dB SPL. This increased the time needed to run the test. A second problem was that the levels of the tone and noise indicated on the audiometer were different from the levels actually obtained. A third problem was that it was sometimes difficult to get the TEN level high enough, either because the TEN became too loud or because of distortion in the earphone or audiometer. To solve these problems, Moore, Glasberg and Stone (2004) introduced a new version of the TEN test, called the TEN(HL) test, using a noise that is spectrally shaped so that, when used with normally hearing subjects, it gives equal masked threshold in dB HL when used with TDH 39, 49 or 50 earphones. Because all levels are specified in dB HL, absolute thresholds only need to be measured once. Furthermore, with the TEN(HL) test, both tone and noise levels are equal to the levels indicated on the audiometer dial. The noise for the TEN(HL) test has a restricted bandwidth compared with the noise used for the earlier TEN test (TEN(HL)-test tone frequencies are from 500 to 4000 Hz). This reduces the *loudness* of the TEN(HL) for a given level per ERB_N. Also, the TEN(HL) was designed as a 'low-noise noise', having minimal amplitude fluctuations (Pumplin, 1985), which allows higher levels/ERB_N without distortion. The criteria for diagnosing a dead region using the TEN(HL) are the same as described earlier for the first version of the TEN test, except that all levels are specified in dB HL instead of dB SPL.

IX.3 PREVALENCE OF DEAD REGIONS ASSESSED USING THE TEN(HL) TEST

The TEN(HL) test has been used in several studies to assess the incidence of dead regions in people with cochlear hearing loss. In one relatively large-scale study (Aazh and Moore, 2007a), the TEN(HL) test was applied using a test frequency of 4000 Hz only, for 98 ears with absolute thresholds between 60 and 85 dB HL at 4000 Hz. Thirty-six ears met the criteria for a dead region. There was no statistically significant difference in the slope of the audiogram or pure-tone average (PTA) between ears with and without dead regions, although an earlier study had suggested that a steeply sloping audiogram might be associated with the presence of a dead region (Preminger, Carpenter and Ziegler, 2005). The prevalence of dead regions exceeded 50 % for hearing losses greater than 70 dB. However, the presence/absence of a dead region could not be predicted reliably from the audiogram and some ears with hearing losses of 80 dB or more did not have any dead region.

In a second large-scale study (Vinay and Moore, 2007a), the TEN(HL) test was used with 308 subjects (556 ears), having hearing losses ranging from mild to severe. All possible test tone frequencies (from 500 to 4000 Hz) were used. Results were presented only for frequencies and ears for which the TEN(HL) level could be made high enough to produce at least 10 dB of masking. Classifying by subject, 177 (57.4 %) were found to have a dead region in one or both ears for at least one frequency. Fifty-four females (54.5 %) and 123 males (58.8 %) had dead regions in one or both ears. Classifying by ear, 256 (46 %) were diagnosed as having a dead region at one frequency or more: 233 ears (41.9 %) had only a high-frequency dead region, 13 ears (2.3 %) had only a low-frequency dead region and 10 ears (1.8 %) had a dead region at both low and high frequencies, with a surviving middle-frequency region. It was not possible reliably to predict the presence/absence of a dead region from the audiogram. However, for each test frequency, 59 % or more of ears had a dead region when the absolute threshold was above 70 dB HL. A very steep slope of the audiogram was found to be suggestive of a high-frequency dead region but did not provide a reliable diagnostic method.

X EFFECTS OF COCHLEAR DAMAGE ON FORWARD MASKING AND SUPPRESSION

Several studies have compared results from simultaneous and forward masking in an attempt to measure suppression effects in hearing-impaired subjects. Usually, the signal level has been fixed, and the masker has been varied to determine a masked threshold. Essentially, it is assumed that threshold is reached when the masker produces a fixed output from the auditory filter centred at the signal frequency.

Wightman, McGee and Kramer (1977) compared PTCs in simultaneous and forward masking for subjects with high-frequency hearing losses. They found that, when both the signal and the masker were in regions of normal sensitivity, the PTCs obtained in forward masking were sharper than those found in simultaneous masking. However, when the signal was in the frequency region of reduced sensitivity, or just below it, the differences were much reduced. They also attempted to measure suppression more directly, using the two-tone unmasking paradigm described earlier. Unmasking was observed when the ‘suppressor’ tone fell in frequency regions of normal sensitivity, but not when it fell in the region of the hearing loss. Wightman *et al.* concluded that the suppression mechanism was rendered ineffective by the hearing loss. They took their results to support the hypothesis that, in normally hearing subjects, suppression is responsible for the differences between PTCs in simultaneous and forward masking. Penner (1980b) and Mills and Schmeidt (1983) studied two-tone unmasking in subjects with noise-induced hearing loss and obtained similar results to those of Wightman *et al.*

Moore and Glasberg (1986b) measured PTCs in simultaneous and forward masking separately for each ear of subjects with unilateral moderate cochlear losses. A notched-noise was gated with the narrowband variable masker to restrict off-frequency listening, to provide a consistent detection cue in forward masking and to equate the levels of the PTCs around their tips for the normal and impaired ears. For the normal ears, they found, as expected, that the PTCs were usually sharper in forward masking than in simultaneous masking. For the impaired ears, the PTCs were broader, and the differences between simultaneous and forward masking were reduced or absent. An example of their results is given in Figure 3.25. These results are consistent with the idea that suppression is reduced or absent in people with moderate cochlear hearing loss.

In another experiment using five subjects with unilateral cochlear hearing loss, Moore and Glasberg (1986b) used notched noise as a simultaneous and a forward masker. The signal level was fixed and the masker level was varied to determine threshold for each notch width used. For the normal ears, the rate of change of masker level at threshold with notch width was greater in forward masking than in simultaneous masking, indicating a higher effective degree of frequency selectivity in forward masking. For the impaired ears, the differences between simultaneous and forward masking were smaller, although they were completely absent for only one subject. These results suggest that the suppression mechanism was reduced in effectiveness in the impaired ears but was not completely inoperative, except in one case.

In summary, the results of the various experiments are in agreement in showing that:

1. Two-tone unmasking in forward masking is small or absent when the stimuli fall in a frequency region where there is cochlear hearing loss.
2. Differences in frequency selectivity between simultaneous and forward masking are much smaller than normal in hearing-impaired subjects.

Taken together, these findings strongly support the idea that suppression is reduced or absent in people with cochlear hearing loss.

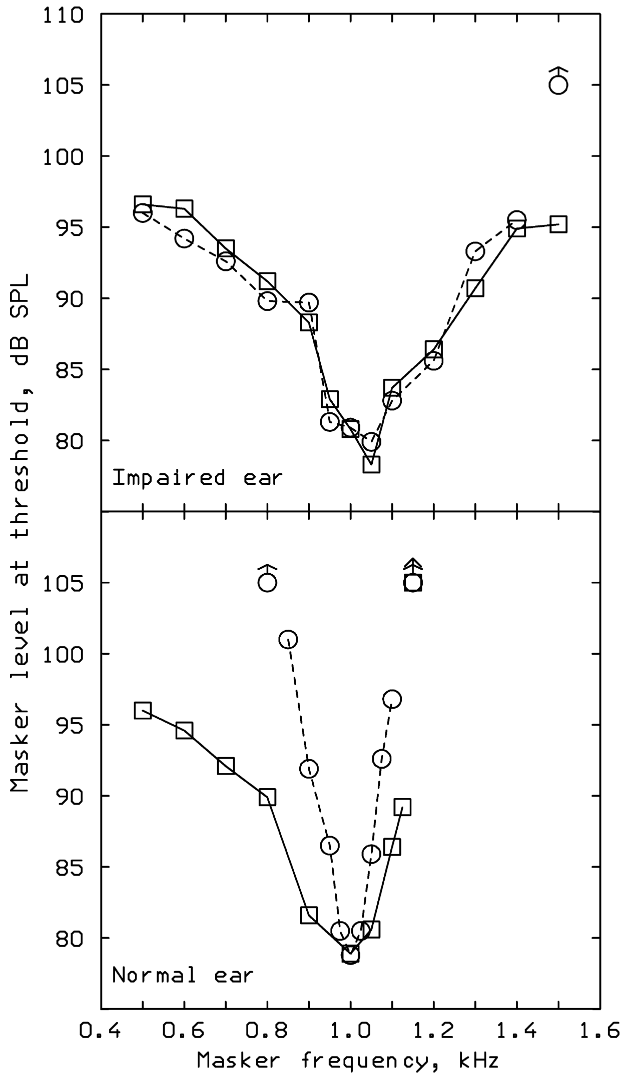


Figure 3.25. PTCs determined for the normal (bottom) and impaired (top) ears of a subject with a unilateral cochlear impairment, for simultaneous masking (squares) and forward masking (circles). Up-pointing arrows indicate that the masker level required for threshold was greater than 105 dB SPL. Data from Moore and Glasberg (1986b).

XI EFFECTS OF COCHLEAR HEARING LOSS ON BM INPUT-OUTPUT FUNCTIONS

It was argued above that the slopes of growth-of-masking functions in forward masking for normally hearing subjects depend strongly on the compression that occurs on the BM. The shallow slopes that are typically observed (see Figure 3.12) are a consequence of the fact that the masker typically falls in a range of levels where the BM input-output function is highly compressive (has a shallow slope), while the signal typically falls in a range of levels where the function is more linear (the slope is steeper). If the compression on the BM is lost as a consequence of cochlear hearing loss, then the growth-of-masking functions in forward masking should have slopes (in dB per dB) close to unity, except when the signal is very close to its absolute threshold. Furthermore, the slope should remain close to unity, regardless of the relative frequencies of the masker and signal, as all frequencies should be processed linearly. Empirical data have confirmed these predictions (Oxenham and Moore, 1995, 1997; Oxenham and Plack, 1997). This is illustrated in Figure 3.26, which shows individual data from three subjects with moderate cochlear hearing loss in the same conditions as those used for the normally hearing subjects in Figure 3.16. In contrast to Figure 3.16, the data for all three hearing-impaired subjects in Figure 3.26 show linear growth-of-masking functions for both the 6-kHz and the 3-kHz masker. This is consistent with the view that cochlear damage results in a loss of BM compression.

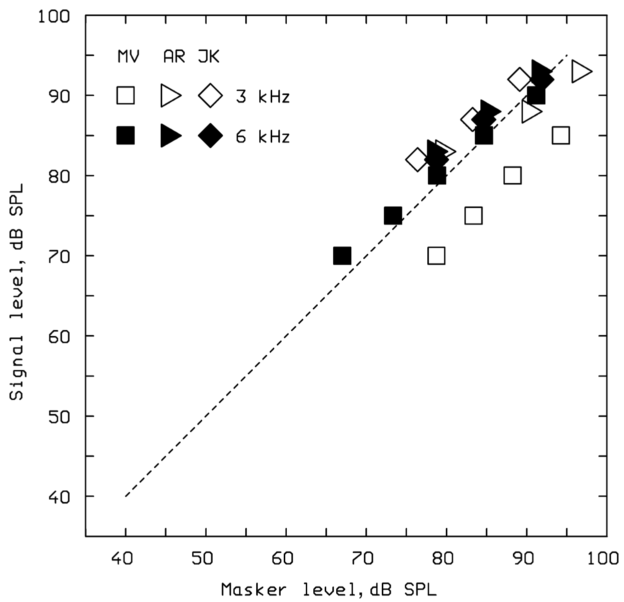


Figure 3.26. Data from Oxenham and Plack (1997) for three subjects with cochlear hearing loss. Individual data from subjects MV, AR and JK are plotted. Thresholds for detecting a 6-kHz signal following a 3-kHz or 6-kHz masker are shown. The signal level was fixed, and the masker level was varied to determine the threshold. The dashed line has a slope of one; linear growth of masking would give data parallel to this line.

XII PERCEPTUAL CONSEQUENCES OF REDUCED FREQUENCY SELECTIVITY, DEAD REGIONS, LOSS OF SUPPRESSION AND STEEPER BM INPUT-OUTPUT FUNCTIONS

Frequency selectivity plays an important role in many aspects of auditory perception. Hence, when frequency selectivity is reduced in people with cochlear hearing loss, these aspects of perception can be expected to alter. Some of these aspects – specifically, loudness perception, frequency discrimination, pitch perception and speech perception – are discussed in more detail in later chapters of this book. The steeper BM input-output functions produced by cochlear damage also affect many aspects of auditory perception (Moore and Oxenham, 1998); again, several of these are discussed later in this book. The next section considers some of the aspects of perception that are directly affected by reduced frequency selectivity.

XII.1 SUSCEPTIBILITY TO MASKING

From the studies discussed earlier, it is obvious that masking effects are often more pronounced in hearing-impaired than in normally hearing subjects. However, the size of the difference depends greatly on the spectral characteristics of the signal and masker. When the signal and masker overlap spectrally (e.g. when the signal is a sinusoid and the masker is a broadband noise without distinct spectral peaks), masked thresholds are usually only slightly higher for hearing-impaired than for normal listeners, except when the signal frequency falls in a dead region. However, when the signal and masker differ in spectrum, masking may be considerably greater for the hearing impaired. There are two situations where this might apply. One is when the average spectrum of a masking sound differs from that of a signal. For example, the signal may be a high-frequency warning siren, and the masker may be primarily low-frequency noise from air-conditioning or machinery. The second is when the signal and the masker differ in their short-term spectra. For example, the signal might be a vowel sound from an attended talker and the masker might be a different vowel sound from an interfering talker. A hearing-impaired person with reduced frequency selectivity will be less able than normal to take advantage of the spectral differences between the two vowels. This issue is examined in more detail in Chapter 8, Section VI.2.

XII.2 TIMBRE PERCEPTION

Timbre is often defined as that attribute of auditory sensation in terms of which a listener can judge that two sounds having the same loudness and pitch are dissimilar. Differences in timbre allow us to distinguish between the same note played on, say, the piano or the flute. Timbre depends on both spectral and temporal aspects of sounds. Changing either the long-term spectral shape of a sound or its temporal *envelope* may lead to a change in perceived timbre (see Handel, 1995, for a review).

The aspects of timbre perception that are affected by spectral shape almost certainly depend on the frequency selectivity of the ear. The spectral shape of a sound is represented in the excitation pattern evoked by the sound, and the timbre is related to the shape of the excitation pattern (Plomp, 1976). In an ear with cochlear damage, frequency selectivity is usually reduced. Hence, the excitation pattern contains less detail about the spectrum than would be the case for a normal ear. This leads to a reduced ability to distinguish sounds on the basis of their spectral shape (Summers and Leek, 1994). Also, the internal representation of spectral shape may be influenced by suppression. Suppression in a normal ear may enhance the contrast between the peaks and dips in the excitation pattern evoked by a complex sound (Tyler and Lindblom, 1982; Moore and Glasberg, 1983b). This may make it easier to pick out spectral features, such as *formants* in vowel sounds. The loss of suppression associated with cochlear damage would thus lead to a greater difficulty in picking out such features.

For many sounds, such as different musical instruments (e.g. xylophone versus harpsichord), the differences in spectral shape are so large that even very poor frequency selectivity allows their discrimination. However, it may be more difficult for a person with reduced frequency selectivity to distinguish sounds that differ in spectrum in more subtle ways. There has been little study of the ability of hearing-impaired people to distinguish musical instruments. However, different steady-state vowels are primarily distinguished by their spectral shapes, and studies of vowel perception suggest that reduced frequency selectivity can, indeed, lead to difficulties in the discrimination and identification of vowels. These studies are reviewed in Chapter 8, Section VI.2.

XII.3 PERCEPTUAL CONSEQUENCES OF DEAD REGIONS

People with high-frequency dead regions also usually have severe or profound high-frequency loss. Without amplification, the higher frequencies in sounds will not be audible, making it hard to hear speech sounds like ‘s’ and ‘sh’. With sufficient amplification, sounds whose frequencies fall within the dead region can sometimes be detected, although in practice it is difficult to restore audibility when the hearing loss is greater than about 90 dB. Hence, a simple lack of audibility may make speech perception more difficult. The effects of dead regions on speech perception are discussed in more detail in Chapter 8, Section IV.

Sounds whose frequency components fall in a dead region are detected at the ‘wrong’ place in the cochlea. For example, if a person has a high-frequency dead region extending from 1500 Hz upwards, a 2000-Hz tone will be detected, if amplified sufficiently, via the place tuned to 1500 Hz. Tones whose frequencies fall well within a high-frequency or low-frequency dead region are usually described as sounding noise-like or highly distorted (Huss and Moore, 2005a). However, such descriptions do not seem to provide a reliable method for diagnosing dead regions, since people without dead regions sometimes describe tones as sounding distorted or noise-like (Huss and Moore, 2005a). The effects of dead regions on pitch perception are discussed in Chapter 6, Section V.

4 Loudness Perception and Intensity Resolution

I INTRODUCTION

Loudness corresponds to the subjective impression of the magnitude of a sound. The formal definition of loudness is: that attribute of auditory sensation in terms of which sounds can be ordered on a scale extending from quiet to loud (ANSI, 1994). Because loudness is subjective, it is very difficult to measure in a quantitative way. Estimates of loudness can be strongly affected by bias and context effects of various kinds (Gabriel, Kollmeier and Mellert, 1997; Laming, 1997). For example, a sound with a moderate level (say 60 dB SPL) may appear quieter when presented just after a high-level sound (say 100 dB SPL) than when presented before the high-level sound. This chapter starts by describing some of the ways in which loudness perception has been studied in normally hearing people. It then goes on to consider changes in loudness perception associated with cochlear hearing loss.

II LOUDNESS PERCEPTION FOR NORMALLY HEARING PEOPLE

II.1 EQUAL-LOUDNESS CONTOURS AND LOUDNESS LEVEL

It is often useful to be able to compare the loudness of sounds with that of a standard, reference sound. The most common reference sound is a 1000-Hz sinusoidal tone, presented in a *free field* (i.e. with no reflections from the walls, floor or ceiling) to both ears and coming from directly in front of the listener. The *loudness level* of a sound is defined as the level of a 1000-Hz tone that is equal in loudness to the sound. The unit of loudness level is the *phon*. Thus, the loudness level of any sound in phons is the level (in dB SPL) of the 1000-Hz tone to which it sounds equal in loudness. For example, if a sound appears to be as loud as a 1000-Hz tone with a level of 45 dB SPL, then the sound has a loudness level of 45 phons. To determine the loudness level of a given sound, the subject is asked to adjust the level of a 1000-Hz tone until it appears to have the same loudness as that sound. The 1000-Hz tone and the test sound are presented alternately rather than simultaneously.

In a variation of this procedure, the 1000-Hz tone is fixed in level, and the test sound is adjusted to give a loudness match. If this is repeated for various different frequencies of a sinusoidal test sound, an *equal-loudness contour* is generated (Fletcher and Munson, 1933). For example, if the 1000-Hz tone is fixed in level at 40 dB SPL,

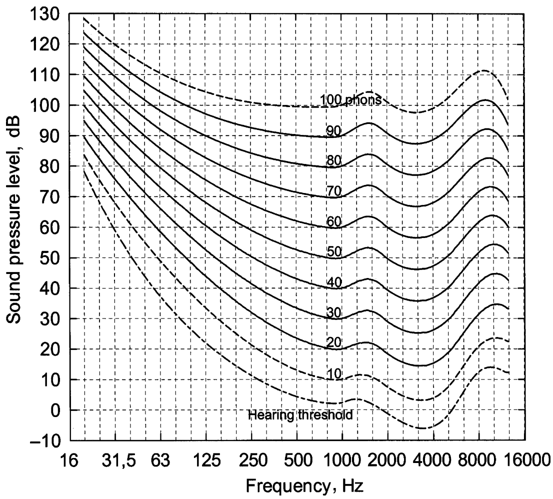


Figure 4.1. Equal-loudness contours for various loudness levels. The lowest curve is the absolute threshold (MAF) curve. The curves are based on a recent standard (ISO 226, 2003).

then the 40-phon equal-loudness contour is generated. The equal-loudness contours published in a recent standard (ISO 226, 2003) are shown in Figure 4.1. The figure includes the *absolute threshold* (MAF) curve. The listening conditions were similar to those for determining the MAF curve, namely that the sound came from a frontal direction in a free field. The equal-loudness contours are of similar shape to the MAF curve, but tend to become flatter at high loudness levels.

Note that the subjective loudness of a sound is not directly proportional to its loudness level in phons. For example, a sound with a loudness level of 80 phons appears much more than twice as loud as a sound with a loudness level of 40 phons.

II.2 THE SCALING OF LOUDNESS

Several methods have been developed that attempt to measure 'directly' the relationship between the physical magnitude of sound and perceived loudness (Stevens, 1957). In one, called *magnitude estimation*, sounds with various different levels are presented, and the subject is asked to assign a number to each one according to its perceived loudness. In a second method, called *magnitude production*, the subject is asked to adjust the level of a sound until it has a loudness corresponding to a specified number.

On the basis of results from these two methods, Stevens suggests that loudness, L, is a *power function* of physical intensity, I:

$$L = kI^{0.3}, \tag{4.1}$$

where k is a constant, depending on the subject and the units used. In other words, the loudness of a given sound is proportional to its intensity raised to the power 0.3. Note that this implies that loudness is *not* linearly related to intensity; rather, it is a compressive function of intensity. An approximation to Equation 4.1 is that the loudness doubles when the intensity is increased by a factor of 10; the latter corresponds to a 10-dB increase in level. In practice, Equation 4.1 only holds for sound levels above about 40 dB SPL. For lower levels than this, the loudness changes with intensity more rapidly than predicted by Equation 4.1 (Hellman, 1976).

The unit of loudness is the *sone*. One sone is defined arbitrarily as the loudness of a 1000-Hz sinusoidal tone at 40 dB SPL, presented binaurally from a frontal direction in a free field. Figure 4.2 shows the relationship between loudness level in phons and loudness in sones. For a 1000-Hz tone, presented under the conditions specified above, the loudness level is equal to the physical level in dB SPL. Figure 4.2 is based on the predictions of a loudness model (Moore, Glasberg and Baer, 1997; ANSI, 2005), but it is consistent with empirical data obtained using scaling methods (Hellman and Zwislocki, 1961). Since the loudness in sones is plotted on a logarithmic scale, and the decibel scale is itself logarithmic, the curve shown in Figure 4.2 approximates a straight line for loudness levels above 40 phon, the range for which Equation 4.1 holds.

Another method of relating loudness sensations to physical intensity involves categorical judgements of loudness. This method has been used extensively to assess loudness perception in hearing-impaired people. The subject is presented with a test

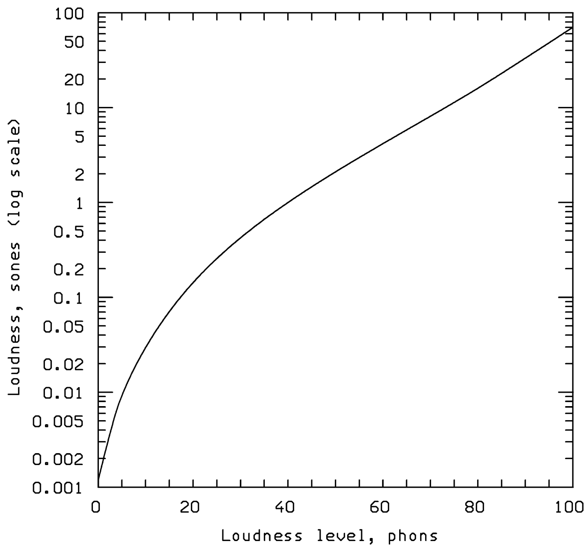


Figure 4.2. Loudness in sones (log scale), plotted as a function of loudness level in phons. The curve is based on a loudness model (Moore, Glasberg and Baer, 1997).

sound and is asked to judge its loudness by using one of several verbal categories (Pascoe, 1978; Hellbrück and Moser, 1985; Kiessling, Steffens and Wagner, 1993; Cox *et al*., 1997). An example is the Loudness Growth in 1/2-Octave Bands (LGOB) test described by Allen, Hall and Jeng (1990) and adapted by Pluvinage (1989) and by Moore *et al*. (1992). The stimuli are half-octave wide bands of noise centred at 500, 1000, 2000 and 4000 Hz. In an initial phase of the test, each centre frequency is tested separately. On each trial, the noise band is presented as a series of three bursts, and the subject is required to indicate the loudness of the bursts by pressing one of seven buttons labelled 'cannot hear', 'very soft', 'soft', 'comfortable', 'loud', 'very loud' and 'too loud'. The sound level is varied randomly from trial to trial over the range from 30 to 110 dB SPL.

In the second phase, all stimuli consistently eliciting responses of 'cannot hear' or 'too loud' are eliminated, and a series of trials is presented in which both the centre frequency and level of the noise bands are randomized. The results of the second phase are used to construct functions relating perceived loudness to level at each centre frequency. Sometimes, the verbal categories are transformed to a numerical scale, for example ranging from 0 to 50.

Estimates of loudness obtained by numerical ratings or categorical ratings can be strongly influenced by factors such as the range of stimuli used and the instructions given (Heller, 1991; Hellbrück, 1993; Hohmann, 1993; Elberling, 1999). Often, listeners distribute their responses across the range of available categories whatever range of stimulus levels is used (Garner, 1954). For example, if the stimuli range in level from 30 to 90 dB SPL, then (for normally hearing listeners) the 30-dB stimulus might be judged as 'very soft' and the 90-dB stimulus might be judged as 'too loud'. If the stimuli range in level from 20 dB to 110 dB SPL, then the 20-dB stimulus might be judged as 'very soft' and the 110-dB stimulus might be judged as 'too loud'. However, there is no easy way of determining what is the 'correct' range to use. Hence, caution is needed in interpreting results obtained using these methods, especially when comparing across studies using different methods and different stimulus ranges.

II.3 THE DETECTION OF INTENSITY CHANGES

The normal auditory system is remarkable both in terms of its absolute sensitivity and in terms of the range of sound intensities to which it can respond. The most intense sound that can be heard without immediately damaging the ears has a level about 120 dB above that of the faintest sound that can be detected. This corresponds to a ratio of intensities of 1 000 000 000 000 : 1. Over this whole range, relatively small changes in sound intensity can be detected.

The smallest detectable change in intensity has been measured for many different types of stimuli by a variety of methods. The three main methods are:

1. Modulation detection. The stimulus is *amplitude modulated* (i.e. made to vary in amplitude) at a slow regular rate and the listener is required to detect the modulation. Usually, the modulation is sinusoidal.

2. Increment detection. A background stimulus is presented, and the subject is required to detect an increment in the level of the background. The background may either be presented continuously or be turned on sometime before and off after the incremented portion.
3. Intensity discrimination of gated or pulsed stimuli. Two (or more) separate pulses of sound are presented successively, one being more intense than the other(s), and the subject is required to indicate which pulse was the most intense.

In all of these tasks, the subjective impression of the listener is that a change in loudness is being detected. For example, in method 1, the modulation is heard as a fluctuation in loudness. In method 2, the increment is heard as a brief increase in loudness of the background, or sometimes as an extra sound superimposed on the background (although, when the increment is small, the listener may hear a change but not know whether it was an increase or decrease in loudness). In method 3, the most intense pulse appears louder than the other(s). Although there are some minor discrepancies in the experimental results for the different types of method, the general trend is similar. For wideband noise, or for bandpass filtered noise, the smallest detectable intensity change is approximately a constant fraction of the intensity of the stimulus. If I is the intensity of a noise band and ΔI is the smallest detectable change in intensity (both in linear units), then $\Delta I/I$ is roughly constant. This is an example of *Weber's Law*, which states that the smallest detectable change in a stimulus is proportional to the magnitude of that stimulus. The value of $\Delta I/I$ is called the *Weber fraction*. If the smallest detectable change in level is expressed in decibels, that is as $\Delta L = 10\log_{10}[(I + \Delta I)/I]$, then this, too, is constant. For wideband noise, ΔL has a value of about 0.5 to 1 dB. This holds from about 20 to 100 dB SL (Miller, 1947). The value of ΔL increases for sounds which are close to the absolute threshold.

For sinusoids, Weber's Law does not hold (Riesz, 1928; Harris, 1963; Viemeister, 1972; Jesteadt, Wier and Green, 1977b). Instead, it is found that $\Delta I/I$ decreases somewhat with increasing I. This has been called the 'near miss' to Weber's Law. The data of Riesz for modulation detection show a value of ΔL of 1.5 dB at 20 dB SL, 0.7 dB at 40 dB SL, and 0.3 dB at 80 dB SL (all at 1000 Hz). The change in ΔL with level is somewhat less for pulsed tones (method 3 above) than for modulated tones (method 1 above).

III EFFECTS OF COCHLEAR HEARING LOSS ON LOUDNESS PERCEPTION

Most people with cochlear hearing loss show a phenomenon called *loudness recruitment*, also called *recruitment* (Fowler, 1936; Steinberg and Gardner, 1937). This may be described as follows. The absolute threshold is higher than normal. When the level of a sound is more than 4–10 dB above the elevated absolute threshold, the rate of growth of loudness level with increasing sound level is greater than normal (Moore, 2004b). When the level is sufficiently high, usually around 90–100 dB SPL, the loudness reaches its 'normal' value; the sound appears as loud to the person with impaired hearing as it would to a normally hearing person. With further increases in sound level above 90–100 dB SPL, the loudness grows in an almost normal manner.

A complementary way of describing this effect is in terms of *dynamic range*. This refers to the range of sound levels between the absolute threshold and the level at which sounds become uncomfortably loud. For example, in a person with normal hearing, the threshold for detecting a mid-frequency sound might be around 0 dB SPL, while the sound might become uncomfortably loud at a level of around 100 dB SPL. The dynamic range would then be 100 dB. Typically, in people with cochlear hearing loss, the absolute threshold is elevated, but the level at which sounds become uncomfortably loud is about the same as normal (except in cases of severe or profound loss). Hence, the dynamic range is reduced compared to normal. Sometimes, especially in cases of severe hearing loss, the loudness does not reach its 'normal' value at high sound levels but remains somewhat below that which would occur in normal hearing. This is called under-recruitment or partial recruitment. The opposite case also sometimes occurs, where high-level sounds appear louder to a hearing-impaired person than to a normally hearing person. This is called over-recruitment or hyperacusis. However, the most common pattern is that of complete recruitment, where for sounds at high level the loudness is about the same for normally hearing and hearing-impaired people.

Loudness recruitment is most easily demonstrated in people who have a unilateral cochlear hearing impairment. It is then possible to obtain loudness matches between the two ears. Usually, a tone is presented alternately to the two ears. The level is fixed in one ear, and the level is adjusted in the other ear until the loudness is judged to be the same at the two ears. The clinical version of this method is known as the alternating binaural loudness balance (ABLB) test. Figure 4.3 shows an example of measurements of this type obtained by Moore and Glasberg (1997). Results were obtained at five different centre frequencies, using sinusoidal tone bursts as stimuli. The exact experimental method is described in Moore, Wojtczak and Vickers (1996). The level of the tone in the normal ear that matches the loudness of the tone in the impaired ear is plotted as a function of the level in the impaired ear. At each frequency tested there is clear evidence of loudness recruitment; the slopes of the curves (in dB/dB) are greater than unity. For the higher frequencies tested, the level required for equal loudness in the normal ear approaches the level in the impaired ear at high levels, that is complete loudness recruitment occurs. For the lowest frequency, only partial recruitment occurs, at least up to the highest level tested. The solid lines are predictions of a model that is described below.

Loudness recruitment can also be demonstrated in people with bilateral cochlear hearing loss. One method of doing this involves the use of the categorical loudness scaling procedure described earlier. Some examples of growth-of-loudness curves

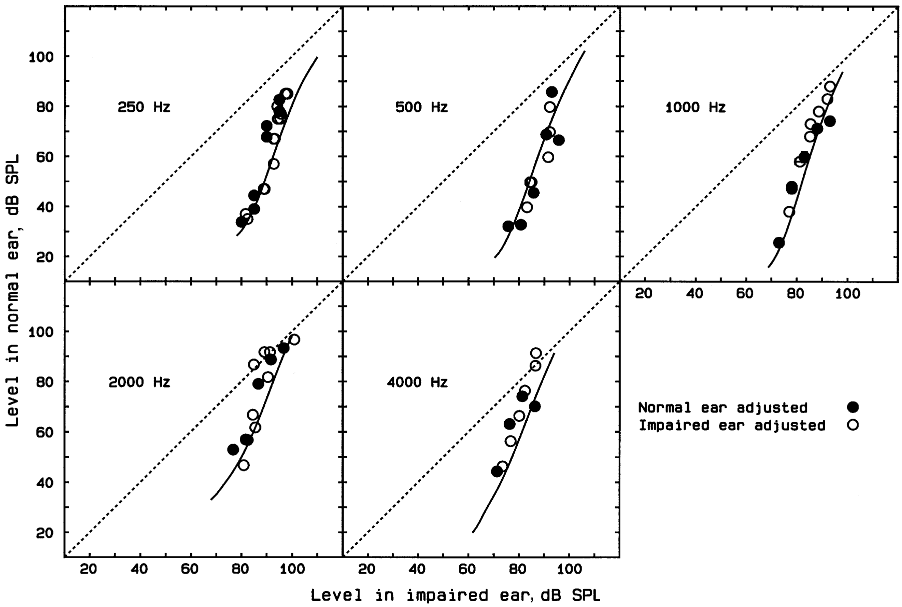


Figure 4.3. Loudness-matching functions for a subject with a unilateral cochlear hearing loss. Each panel shows results for a different signal frequency. The curves are predictions of the model described in the text. The diagonal dotted lines indicate equal levels in the two ears.

obtained using the LGOB method (Allen, Hall and Jeng, 1990) are given in Figure 4.4 (data from Moore *et al.*, 1992d). The numbers 1–5 correspond to the five categories of loudness from 'very soft' to 'very loud'. In each panel, the dashed line indicates typical results for subjects with normal hearing. The four panels at the top show results for a subject with marked loudness recruitment; all of the curves have steep slopes, and at high levels the loudness reaches or even exceeds that which would occur in a normal ear. The bottom four panels show results for a subject with little or no loudness recruitment. The curves are shifted down vertically relative to those for normally hearing subjects, but the slopes are almost normal. This pattern of results is common in cases of conductive hearing loss, but rare in cases of cochlear damage. Most subjects with cochlear hearing loss give results between these two extremes, a typical example being shown in the middle four panels.

A third way of measuring loudness recruitment is to use magnitude estimation or magnitude production, as described earlier for normally hearing subjects (Stevens, 1957). These techniques give a similar pattern of results to the categorical loudness scaling procedures; almost all subjects with cochlear damage show some degree of loudness recruitment, as indicated by the fact that the rate of growth of loudness with increasing sound level is greater than normal (Hellman and Meiselman, 1990, 1993).

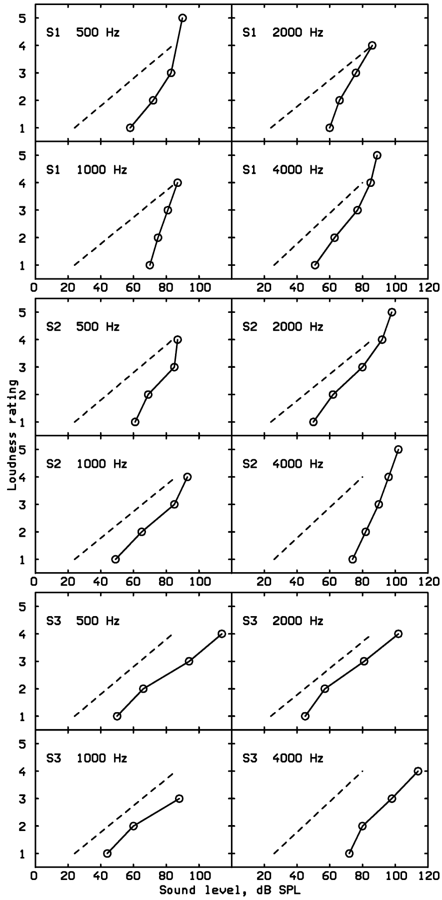


Figure 4.4. Results of three subjects for a categorical loudness scaling procedure (circles). The dashed lines show typical results for normally hearing subjects. The centre frequency of the 1/2-octave-wide band of noise used is indicated in each panel. A loudness rating of 1 corresponds to 'very soft' and a rating of 5 corresponds to 'very loud'.

On average, the steepness of loudness growth curves increases with increasing absolute threshold at the test frequency (Miskolczy-Fodor, 1960; Glasberg and Moore, 1989; Hellman and Meiselman, 1990, 1993; Kiessling, Steffens and Wagner, 1993). This is consistent with the idea that threshold elevation and loudness recruitment are both linked to the loss of the active mechanism in the cochlea. However, there can be considerable individual differences in the rate of loudness growth for a given degree of hearing loss; examples are given later in this chapter (see Figure 4.10). These individual differences probably arise largely from differing relative amounts of OHC (outer hair cell) and IHC (inner hair cell) damage, as will be explained later in this chapter.

When the absolute threshold is high, the dynamic range can be very small indeed. For example, for subject S2 in Figure 4.4, a 75-dB SPL tone at 4000 Hz was judged 'very soft', while a 100-dB SPL tone was judged 'very loud', indicating a dynamic range of only about 25 dB. In some cases, the dynamic range can be as small as 10 dB.

The measures of loudness recruitment described above were all obtained with bursts of sound whose amplitude envelope was constant during their presentation time. However, loudness recruitment also influences the perception of sounds whose amplitude fluctuates from moment to moment. Examples of such sounds include speech and music. For speech, the most prominent fluctuations occur at rates from about 0.5 Hz up to 32 Hz (Steeneken and Houtgast, 1980). For sounds that are amplitude modulated at rates up to 32 Hz, recruitment results in a magnification of the perceived amount of fluctuation (Moore, Wojtczak and Vickers, 1996); the sound appears to 'wobble' more than it would for a normally hearing person. The magnification of the perceived fluctuation is roughly independent of the modulation rate over the range 4–32 Hz. The fact that the magnification does not decrease with increasing modulation frequency is consistent with the idea that recruitment results mainly from the loss of fast-acting compression on the basilar membrane (BM). This idea is elaborated later in this chapter.

IV A MODEL OF NORMAL LOUDNESS PERCEPTION

To understand how loudness recruitment occurs, it is helpful to consider models of loudness perception. Most models of loudness are based on concepts proposed by Fletcher and Munson (1937) and by Zwicker and Scharf (Zwicker, 1958; Zwicker and Scharf, 1965). Figure 4.5 shows a block diagram of a model of loudness perception developed by Moore, Glasberg and Baer (1997), but based on these earlier models. This model forms the basis for an ANSI standard (ANSI, 2005). The first two stages are fixed filters to account for the transmission of sound through the outer and middle ear; see Chapter 1, Figures 1.3 and 1.4.

The next stage is the calculation of a psychoacoustical *excitation pattern* for the sound under consideration; the concept of the excitation pattern was described in Chapter 3, Sections V.2 and V.3. The excitation patterns are calculated from *auditory filter* shapes. Panel A of Figure 4.6 shows excitation patterns for a 1000-Hz sinusoidal

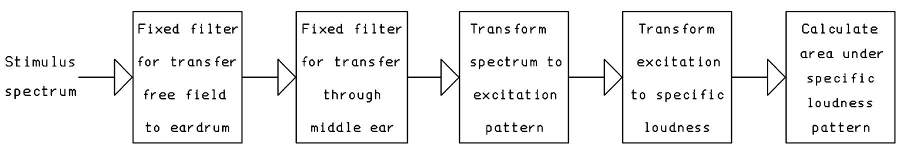


Figure 4.5. A block diagram of the loudness model developed by Moore, Glasberg and Baer (1997).

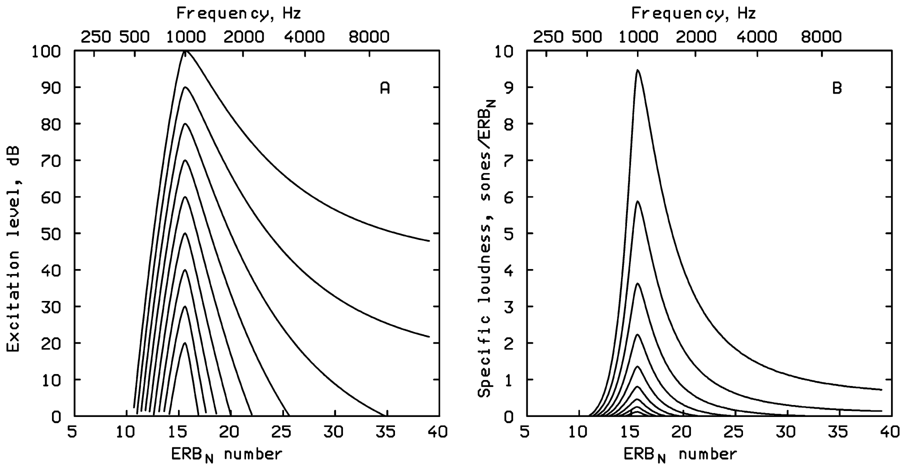


Figure 4.6. Panel A shows psychoacoustical excitation patterns for a 1000-Hz sinusoid with a level ranging from 20 to 100 dB SPL in 10-dB steps. The patterns are plotted on an ERB$_N$-number scale. The corresponding frequency is shown at the top. Panel B shows the corresponding specific loudness patterns, calculated using the model of Moore, Glasberg and Baer (1997).

tone with a level ranging from 20 to 100 dB SPL in 10-dB steps. The frequency scale of the excitation pattern has been translated to the *ERB$_N$-number scale* described in Chapter 3, Section IV.1; each 1-ERB$_N$ step corresponds roughly to a constant distance of about 0.89 mm along the BM. The corresponding frequency is shown at the top. Note that these excitation patterns were calculated assuming that the gain of each auditory filter was 0 dB at its centre frequency. Calculated in this way, the excitation patterns resemble *masking patterns*. Effectively, the patterns in Figure 4.6 were calculated ignoring the effects of BM compression. On the BM, the gain of each place decreases with increasing level for an input at the *characteristic frequency* (CF), while the gain is level-independent for frequencies well below the CF.

The effects of BM compression are taken into account in the next stage, which is a transformation from excitation level to *specific loudness*, *N*. The specific loudness is a kind of loudness density, the loudness per ERB$_N$. It represents the loudness that would be evoked by the excitation within a 0.89-mm range on the BM if it were possible to present that excitation alone (without any excitation at adjacent regions on the BM). The relationship between excitation and *N* is specified with the excitation expressed in linear power units, *E*, rather than excitation level in dB, L_E. The relationship between *E* and L_E is:

$$L_E = 10\log_{10}(E/E_0), \tag{4.2}$$

where E_0 is the peak excitation produced by a 1000-Hz tone at 0 dB SPL (free-field presentation, frontal incidence). For most values of *E*, and for centre frequencies above

500 Hz, the specific loudness is related to the excitation at a given centre frequency by the following equation:

$$N = C[(E + A)^{\alpha} - A^{\alpha}], \tag{4.3}$$

where C, A and α are constants. The value of A is equal to two times the peak excitation produced by a sinusoidal signal at absolute threshold; this latter quantity is denoted by E_{THRQ}. When E_{THRQ} is expressed in decibels ($10\log_{10}E_{THRQ}$), it is denoted L_{THRQ}. Equation 4.3 applies over the range $E_{THRQ} < E < 10^{10}$.

The value of α is equal to 0.2, so Equation 4.3 defines a compressive relationship between excitation and specific loudness. For example, when E is much greater than the threshold value, if E is increased by a factor of 10, N increases by a factor of only 1.58. However, the function becomes steeper when E approaches E_{THRQ}. When E is less than E_{THRQ}, the specific loudness is small, but not zero, and a different (steeper) function is used to characterize the relationship between N and E; see Moore, Glasberg and Baer (1997) for details. A different function is also used when the value of E is very high.

The compressive nonlinearity in the model can be thought of as representing the overall effects of the transformation from the physical stimulus to neural activity. At least two nonlinearities contribute to this overall effect: the compressive nonlinearity of the BM *input-output function*, and the nonlinear transformation from BM velocity or amplitude to neural activity (Yates, 1990). The model is unrealistic in the sense that the filtering on the BM and the nonlinear input-output function on the BM are not two separate processes; it is almost certainly incorrect to represent them as two sequential stages in the model. Nevertheless, the end result in the model is probably similar to what would be obtained in a more realistic model of BM responses, such as proposed by Giguère and Woodland (1994), Lopez-Poveda and Meddis (2001) or Zhang *et al.* (2001).

The gain applied by the active mechanism in the cochlea appears to decrease somewhat at low frequencies (Yates, 1995; Robles and Ruggero, 2001). This results in an increase in the sound level required to produce a given response on the BM and an increase in the slope of the input-output function of the BM. In the model, these effects are simulated by increasing the value of E_{THRQ} and increasing the value of α. To implement the change in E_{THRQ}, a term, G, is introduced, which represents the *low-level* gain of the active mechanism at a specific frequency, *relative* to the gain at 500 Hz and above (which is assumed to be constant). G can be thought of as a gain 'offset'. The value of G is always less than or equal to 1 in linear units. Its value in decibels ($10\log_{10}G$) is denoted G_{dB}; this is always 0 or a negative number. It is assumed that the product of G and E_{THRQ} is constant. For example, if E_{THRQ} is a factor of 10 higher than the value at 500 Hz and above, then G is equal to 0.1 (corresponding to $G_{dB} = -10$ dB). The equation relating specific loudness to excitation then becomes:

$$N' = C[(GE + A)^{\alpha} - A^{\alpha}] \tag{4.4}$$

The values of the constants A and α both increase as G becomes smaller, with decreasing centre frequency below 500 Hz.

If the specific loudness is plotted as a function of $\mathrm{ERB_N}$-number, the resulting pattern is called a *specific loudness pattern*. Specific loudness patterns for a 1000-Hz sinusoidal tone with a level ranging from 20 to 100 dB SPL are shown in panel B of Figure 4.6. The overall loudness of a given sound, in sones, is assumed to be equal to the sum of the loudness across all one-$\mathrm{ERB_N}$-wide bands. This is equal to the total area under the specific loudness pattern plotted as N versus $\mathrm{ERB_N}$. Thus, the loudness of any sound, including a single sinusoid, is assumed to depend partly on the spread of excitation along the BM.

V A MODEL OF LOUDNESS PERCEPTION APPLIED TO COCHLEAR HEARING LOSS

V.1 INTRODUCTION

The perception of loudness may be affected by at least four changes that occur with cochlear hearing loss:

1. Elevation in absolute threshold, which may be caused by loss of function of OHCs or IHCs, neural degeneration or a combination of all of these (Schuknecht, 1993). Reduced functioning of the stria vascularis may also be involved (Schmiedt, 1996), but it is assumed that this can be modelled indirectly as reduced functioning of the OHCs and IHCs.
2. A reduction in or loss of the compressive nonlinearity in the input-output function of the BM, which is mainly associated with OHC dysfunction; see Chapter 1, Section VII.1.
3. Loss of *frequency selectivity*, which results in broader excitation patterns, which is again associated mainly with loss of OHC function; see Chapter 1, Section VII.1 and Chapter 3, Section VIII.
4. Complete loss of IHCs or functional neurones at certain places within the cochlea (*dead regions*); see Chapter 1, Section IV, Chapter 2, Section IV and Chapter 3, Sections VIII and IX.

Individual differences in loudness-growth functions may reflect differences in any or all of these factors. Moore and Glasberg (1997) considered the possible influence of each of these factors and how they could be incorporated into a loudness model. A revised version of this model was developed (Moore and Glasberg, 2004) to take into account recent data showing that, for sound levels very close to absolute threshold, the rate of growth of loudness with increasing level is similar for normal and impaired ears (Buus and Florentine, 2002; Moore, 2004b). The revised model of Moore and Glasberg (2004) is described in outline below; some details of the model are omitted for simplicity.

V.2 ELEVATION OF ABSOLUTE THRESHOLD

As described in Chapter 2, Section IV, elevation of absolute threshold due to cochlear damage can occur in two main ways. First, loss of function of the OHCs can result in reduced BM vibration for a given low sound level. Hence, the sound level has to be increased to give a just-detectable amount of vibration. Secondly, loss of function of IHCs and/or neurones can result in reduced efficiency of transduction, so the amount of BM vibration needed to reach threshold is higher than normal. In principle, it is possible to partition the overall hearing loss at a given frequency into a component due to OHC dysfunction and a component due to IHC (and neural) dysfunction:

$$\mathrm{HL}_{\mathrm{OHC}} + \mathrm{HL}_{\mathrm{IHC}} = \mathrm{HL}_{\mathrm{TOTAL}} \tag{4.5}$$

For example, if the total hearing loss at a given frequency is 60 dB, 40 dB of that loss might be due to OHC dysfunction and 20 dB to IHC dysfunction. Note that the proportion of a hearing loss that is attributed to OHC or IHC dysfunction is not the same as the proportion of OHCs and IHCs that are damaged or lost. In the above example, 40 dB of the hearing loss was attributed to OHC dysfunction and 20 dB to IHC dysfunction, but this does *not* imply that damage to the OHCs was twice as great as damage to the IHCs.

The value of $\mathrm{HL}_{\mathrm{TOTAL}}$ at a given frequency can be measured directly (it simply corresponds to the audiometric threshold in dB HL). In the model, it is specified at the standard audiometric frequencies of 125, 250, 500, 1000, 2000, 4000, 6000 and 8000 Hz. Linear interpolation is used to calculate values at intermediate frequencies. However, the values of $\mathrm{HL}_{\mathrm{OHC}}$ and $\mathrm{HL}_{\mathrm{IHC}}$ can only be estimated indirectly. Of course, once $\mathrm{HL}_{\mathrm{OHC}}$ is estimated, then $\mathrm{HL}_{\mathrm{IHC}}$ can also be estimated using Equation 4.5. The model incorporates default values for $\mathrm{HL}_{\mathrm{OHC}}$, which are calculated from the overall hearing loss at each audiometric frequency.

It is described below how the values of $\mathrm{HL}_{\mathrm{OHC}}$ and $\mathrm{HL}_{\mathrm{IHC}}$ are used within the model to produce the overall effect of elevating the absolute threshold by $\mathrm{HL}_{\mathrm{TOTAL}}$.

V.3 REDUCED COMPRESSIVE NONLINEARITY

It is assumed that the reduction of the compressive nonlinearity on the BM produced by hearing loss is related to the value of $\mathrm{HL}_{\mathrm{OHC}}$. The value of $\mathrm{HL}_{\mathrm{OHC}}$ is used to modify the gain offset, G. To simulate hearing loss due to loss of OHC function, G_{dB} is decreased (made a bigger negative number). This is associated with increases in the value of the exponent α. The changes in G_{dB} and α lead to a steepening of the function relating specific loudness to excitation level.

Figure 4.7 shows the transformation from psychoacoustical excitation level to specific loudness for several values of the peak excitation level at absolute threshold. The number next to each curve shows the value of L_{ETHRQ} *relative* to the value for normal hearing for frequencies above 500 Hz. For the left-most curve, the actual value of L_{ETHRQ} was 3.6 dB, which corresponds to the peak excitation level in quiet at the

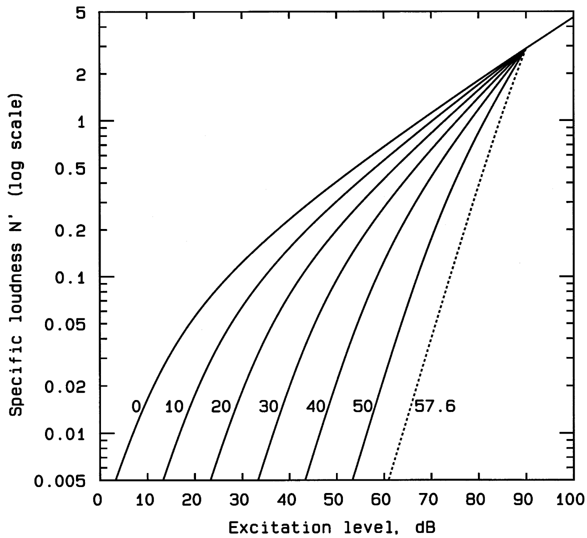


Figure 4.7. The transformation from psychoacoustical excitation level to specific loudness for several values of the peak excitation level at absolute threshold. The number next to each curve shows the value of L_{ETHRQ} *relative* to the value for normal hearing for frequencies above 500 Hz.

absolute threshold for monaural listening for frequencies above 500 Hz. For normal hearing, the relative value of the threshold excitation level never exceeds about 23 dB; this is the value for a 50-Hz tone. The curves for higher values of the relative threshold excitation level are applicable only to impaired hearing.

For frequencies above 500 Hz, the maximum gain of the cochlear amplifier is assumed to be 57.6 dB. To simulate impaired hearing, G_{dB} for frequencies above 500 Hz is made negative (relative to the 'normal' value of 0 dB) by up to 57.6 dB, by an amount equal to HL_{OHC}. The value of HL_{OHC} is not allowed to be greater than 57.6 dB. For frequencies below 500 Hz, G_{dB} has a negative value even for normal hearing. The value for normal hearing is denoted $G_{dBnorm}(f)$, where (f) indicates that the value is a function of frequency, f. To simulate impaired hearing, the gain offset G_{dB} is made a bigger negative number by an amount equal to HL_{OHC}, but with a limit of 57.6 + $G_{dBnorm}(f)$ dB. For example, if G_{dBnorm} at a specific frequency is -11 dB, the maximum possible value of HL_{OHC} is 46.6 dB, and the maximum reduction in G_{dB} is 46.6 dB.

V.4 REDUCED INNER HAIR CELL/NEURAL FUNCTION

In some earlier models (Moore and Glasberg, 1997; Launer, Hohmann and Kollmeier, 1997), the effect of loss of IHC and/or neural function was modelled by a simple attenuation of the excitation level at the frequency in question. However, it seems plausible that damaged IHCs may respond very poorly to weak inputs, but respond in a more nearly normal way when the input signal is well above the threshold value. To accommodate this possibility, two cases are considered. When the excitation level, L_E, exceeds L_{ETHRQ} by an amount less than or equal to $\mathrm{HL_{IHC}}$, that is when:

$$L_E - L_{ETHRQ} \le \mathrm{HL_{IHC}}, \tag{4.6}$$

then the excitation level is reduced by an amount equal to $\mathrm{HL_{IHC}}$. When the excitation level exceeds L_{ETHRQ} by an amount greater than $\mathrm{HL_{IHC}}$, that is when:

$$L_E - L_{ETHRQ} > \mathrm{HL_{IHC}}, \tag{4.7}$$

then the excitation level is reduced by:

$$\mathrm{HL_{IHC}^2}/(L_E - L_{ETHRQ}) \tag{4.8}$$

This has the effect of progressively reducing the attenuation applied to the excitation level as L_E increases, the largest amount of attenuation being equal to $\mathrm{HL_{IHC}}$. For example, if L_{ETHRQ} is 50 dB and $\mathrm{HL_{IHC}}$ is 10 dB, then the attenuation applied to the excitation level is reduced from 10 to 2 dB as L_E is increased from 60 to 100 dB. Note that the value of $\mathrm{HL_{IHC}}$ has no effect on the values of G or α.

V.5 REDUCED FREQUENCY SELECTIVITY

Frequency selectivity is usually reduced in cases of cochlear hearing loss; see Chapter 3, Section VIII. For a sinusoidal stimulus, this leads to an excitation pattern which is broader in an impaired ear than in a normal ear. Moore and Glasberg (2004) developed a series of equations that describe empirically how the sharpness of the auditory filters varies with sound level and with hearing loss. The degree of broadening with hearing loss was assumed to depend specifically on $\mathrm{HL_{OHC}}$ rather than on $\mathrm{HL_{TOTAL}}$. Figure 4.8 shows psychoacoustical excitation patterns calculated from filter shapes based on these equations, for flat hearing losses of 40 and 60 dB. For the purpose of calculating these curves, the values of $\mathrm{HL_{OHC}}$ were assumed to be 36 and 54 dB respectively, that is the value of $\mathrm{HL_{OHC}}$ was assumed to be 90 % of $\mathrm{HL_{TOTAL}}$. The patterns were calculated for 1-kHz tones with levels from 20 to 100 dB SPL and they are plotted on an $\mathrm{ERB_N}$-number scale, with the corresponding frequency shown at the top. Notice that the patterns are broader for the larger hearing loss, but the change in the patterns with level is smaller for the larger hearing loss. The dashed lines indicate the peak excitation level that would be produced by a tone at absolute threshold. Excitation falling below these lines would be inaudible. The curves below the dashed lines have been plotted only to show more clearly the changes in shape with level.

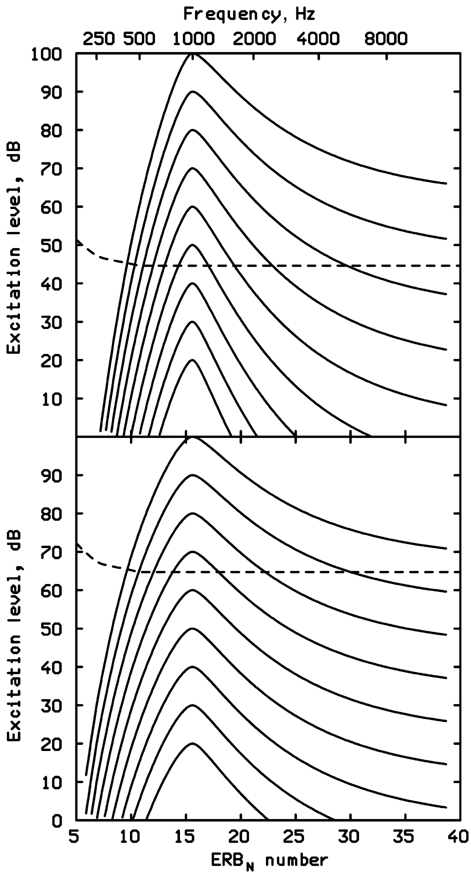


Figure 4.8. Psychoacoustical excitation patterns calculated for hypothetical hearing-impaired people with flat losses of 40 dB (upper panel) or 60 dB (lower panel). The value of HL_{OHC} was assumed to be 36 dB for the former and 54 dB for the latter. Patterns are shown for tones with levels ranging from 20 to 100 dB SPL. The dashed lines indicate the peak excitation level that would be produced by a tone at absolute threshold. Excitation falling below these lines would be inaudible.

V.6 COMPLETE LOSS OF FUNCTIONING IHCs OR NEURONES (DEAD REGIONS)

As described in Chapters 1, 2 and 3, sometimes a person with cochlear hearing loss may have a region of the cochlea where there are no functioning IHCs and/or neurones. The effects of a dead region are included in the loudness model by setting the excitation to a very low value (effectively zero) over the frequency range corresponding to the region assumed to be dead. This has the effect that the specific loudness evoked from that region is zero.

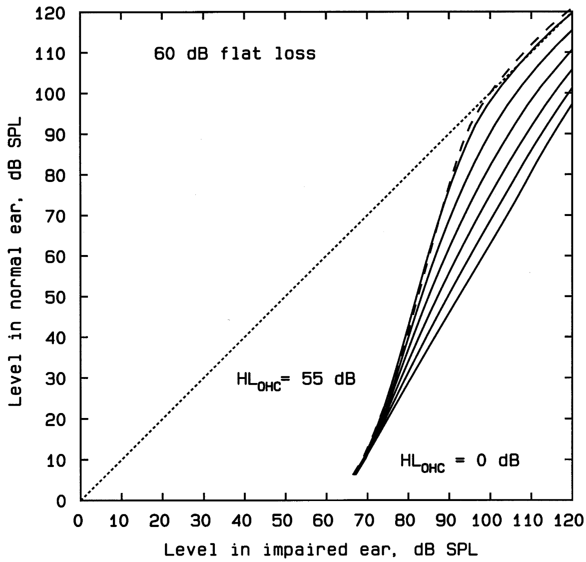


Figure 4.9. The effect on predicted loudness-matching functions of changing the value of the parameter HL_{OHC} for a hypothetical subject with a flat hearing loss of 60 dB in one ear and normal hearing in the other ear. The figure illustrates the effect of varying the parameter HL_{OHC}; values of HL_{OHC} were 0, 10, 20, 30, 40, 50 and 55 dB (dashed line). The diagonal dotted line indicates equal levels in the two ears.

V.7 USING THE MODEL TO ACCOUNT FOR LOUDNESS RECRUITMENT

The model can be used to evaluate the relative importance of the factors described above. Indeed, it is possible within the model to include or remove each factor; for example, frequency selectivity can be set to 'normal' values instead of having the values appropriate for the hearing loss. Analyses of this type indicate that *the main cause of loudness recruitment is the steeper input-output function on the BM*. Reduced frequency selectivity (greater spread of excitation) also plays some role, as suggested by Kiang, Moxon and Levine (1970) and by Evans (1975), but that role is not very great. The minor role of greater-than-normal spread of excitation in producing loudness recruitment is also suggested by experimental studies in which noise was used to mask the excitation at CFs remote from the signal frequency (Hellman, 1978; Moore *et al.*, 1985; Hellman and Meiselman, 1986; Zeng and Turner, 1991); the noise had little effect on the rate of loudness growth measured for hearing-impaired subjects.

The solid lines in Figure 4.3, which was presented earlier in this chapter, show predictions of the model. This subject was diagnosed as having a dead region at low frequencies, and to model her data it was assumed that the dead region extended from 0 to 280 Hz. Thus, for a signal at 250 Hz, the growth of loudness with increasing level in the impaired ear was predicted to depend mainly on the upward spread of excitation. In that case, the broadening of the auditory filters plays a significant role. Note that the predicted growth of loudness would be steeper if no dead region were assumed. Generally, the model fits the data rather well.

Figure 4.9 illustrates the effect of varying the parameter HL_{OHC} for a hypothetical subject with a hearing loss of 60 dB at all frequencies in one ear, with the other ear having completely normal audiometric thresholds. The figure shows the predicted sound levels required to match the loudness of a 1-kHz sinusoidal tone between the two ears. To generate these predictions, the loudness was calculated as a function of level for each ear separately, and the loudness functions were used to calculate the levels giving equal loudness in the two ears. The value of HL_{OHC} was varied in 10-dB steps from 0 to 50 dB (solid curves); the dashed line shows predictions for HL_{OHC} = 55 dB. The functions have an initial portion of shallow slope for very low sensation levels (SLs), a steeper mid-level portion and then a shallower high-level portion. This is consistent with empirical data (Miskolczy-Fodor, 1960; Hellman and Zwislocki, 1964; Hellman, 1997; Moore, 2004b).

Empirically measured loudness growth functions can vary markedly in slope across subjects, even for subjects with similar absolute thresholds (Hellman and Meiselman, 1990, 1993; Kiessling, Steffens and Wagner, 1993; Hellman, 1997; Moore and Glasberg, 1997). The model can account for the range of empirically measured loudness growth functions by allowing HL_{OHC} to vary across subjects.

In summary, the loudness model developed by Moore and Glasberg (2004) can account for the rapid growth of loudness typically associated with cochlear hearing loss. By varying the parameter HL_{OHC}, and taking into account dead regions, the model can also account for individual variability in loudness growth functions. According to the model, the main cause of loudness recruitment is reduced compressive nonlinearity of the input-output function on the BM, although reduced frequency selectivity (broader excitation patterns) also plays some role.

VI EFFECTS OF BANDWIDTH ON LOUDNESS

VI.1 NORMAL HEARING

For normally hearing subjects, if the bandwidth of a sound, such as a noise, is varied keeping the *overall* intensity fixed, the loudness remains constant as long as the bandwidth is less than a certain value, called the critical bandwidth (CB) for loudness. If the bandwidth is increased beyond the CB, the loudness increases (Zwicker, Flottorp and Stevens, 1957). In other words, for a fixed overall intensity, a sound appears louder when its spectrum covers a wide frequency range than when its spectrum covers a narrow frequency range. The reason for this effect can be understood by considering how specific loudness patterns change with bandwidth. With increasing bandwidth up to the CB, the specific loudness patterns become lower at their tips, but broader; the decrease in area around the tip is almost exactly cancelled by the increase on the skirts, so that the total area remains almost constant (Moore and Glasberg, 1986d; Moore, Glasberg and Baer, 1997). When the bandwidth is increased beyond the CB, the increase on the skirts is greater than the decrease around the tip, and the total area, and hence the predicted loudness, increases. Since the increase depends on the summation of specific loudness at different CFs, the increase in loudness is often described as *loudness summation*.

VI.2 IMPAIRED HEARING

Several studies have shown that loudness summation is reduced in people with cochlear hearing loss; the increase in loudness with increasing bandwidth is less than occurs in normally hearing people (Scharf and Hellman, 1966; Bonding, 1979a; Florentine and Zwicker, 1979; Bonding and Elberling, 1980). Often, the experiments have involved comparisons of loudness between narrowband noises and broadband

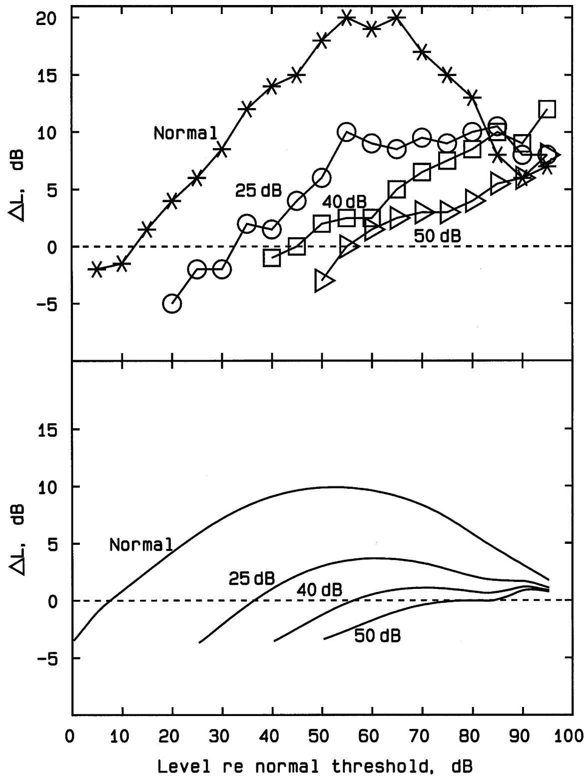


Figure 4.10. The top panel plots data of Bonding and Elberling (1980), showing differences in overall level, ΔL, between a 230-Hz-wide noise and a 1600-Hz-wide noise required for equal loudness. Both noises were geometrically centred at 1000 Hz. Results are shown for normally hearing subjects (asterisks), and subjects with hearing losses of 25 dB (circles), 40 dB (squares) and 50 dB (triangles). The bottom panel shows predictions of the model assuming that $HL_{OHC} = 0.8\ HL_{TOTAL}$.

noises. Typically, one sound, the reference sound, is fixed in level, and the other sound, the comparison sound, is adjusted in level to achieve equal loudness. The two sounds are alternated. For normally hearing subjects, the level of the sound with the greater bandwidth is usually lower than that of the sound with the narrower bandwidth, at the point of equal loudness. The extent of this level difference, ΔL, varies with overall level, being greatest at moderate sound levels (Zwicker, Flottorp and Stevens, 1957; Zwicker and Scharf, 1965; Bonding, 1979a; Bonding and Elberling, 1980; Zwicker and Fastl, 1999). For hearing-impaired subjects, the value of ΔL is usually markedly smaller than normal.

The top panel of Figure 4.10 shows an example of results from a study of Bonding and Elberling (1980). They tested thirty subjects with sensorineural hearing losses, assumed to be primarily cochlear in origin. Subjects were selected to have reasonably flat losses of 25 ($N = 12$), 40 ($N = 10$) or 50 ($N = 8$) dB. Twenty-five normally hearing subjects were also tested. The reference signal was always a 230-Hz-wide band of noise geometrically centred at 1000 Hz. The comparison stimulus was a 1600-Hz-wide band of noise geometrically centred at 1000 Hz. The figure shows the mean value of ΔL, the difference in overall level between the comparison and reference sounds at the point of equal loudness, plotted as a function of the level of the reference sound relative to the average absolute threshold for the normally hearing subjects at 500, 1000 and 2000 Hz.

The data for the normally hearing subjects show that ΔL increases with increasing level up to about 50–60 dB, and then declines. The maximum value of ΔL is larger by about a factor of two than is typically observed in the literature (Zwicker, Flottorp and Stevens, 1957). The data for the hearing-impaired subjects generally show less loudness summation (smaller values of ΔL), and the values of ΔL decrease with increasing hearing loss.

The lower panel of Figure 4.10 shows the predictions of the model assuming that HL_{OHC} was 0.8 HL_{TOTAL}. With this value, the general form of the predictions was close to that of the data, although the predicted form was similar when HL_{OHC} was assumed to be 0.9 HL_{TOTAL}. The predicted values of ΔL are generally somewhat smaller than the observed values, especially for the normally hearing subjects; the model predictions are more in line with data in the literature (Zwicker, Flottorp and Stevens, 1957; Zwicker and Scharf, 1965). Overall, the main features of the data are predicted reasonably well by the model, especially when the inherent variability in this type of experiment is taken into account.

At low SLs, the values of ΔL are slightly negative for both normally hearing and hearing-impaired subjects. This effect is correctly predicted by the model. It can be described by saying that when the bandwidth of a low-level noise is increased, keeping the overall level constant, the loudness decreases. It occurs for the following reason. At low SLs, specific loudness decreases rapidly with decreasing excitation level (see Figure 4.7). As the bandwidth is increased, keeping overall intensity constant, the excitation spreads over a wider range of CFs, but the excitation in the central part of the excitation pattern decreases. The decrease in the central part of the specific loudness pattern more than offsets the increase at the edges, and so overall loudness decreases.

In summary, loudness summation is typically reduced in people with cochlear hearing loss. In contrast to what happens with normally hearing people, the loudness of a sound of fixed intensity does not usually increase markedly with increasing bandwidth beyond the CB. It is likely that the reduced loudness summation observed in people with cochlear hearing loss depends both on loss of the compressive nonlinearity on the BM (which is the main cause of loudness recruitment) and on reduced frequency selectivity.

VII EFFECTS OF COCHLEAR HEARING LOSS ON INTENSITY RESOLUTION

As described earlier, the ability to detect changes in intensity, or to compare the intensity of two separate sounds, is usually assumed to be based on the loudness sensations evoked by the sounds. In people with cochlear hearing loss, a given change in intensity usually results in a larger-than-normal change in loudness. Hence, it might be expected that intensity discrimination would be better than normal. However, this expectation is based on the assumption that the just-detectable change in loudness is unaffected by the cochlear damage, and this assumption may not be valid.

Following early evidence that loudness recruitment was specific to cochlear dysfunction (Dix, Hallpike and Hood, 1948), there were many studies exploring the possibility that better-than-normal intensity discrimination could be used as an indirect measure of loudness recruitment, and hence as an indicator of cochlear hearing loss (Lüscher and Zwislocki, 1949). These studies led to the Short Increment Sensitivity Index (SISI) test (Jerger, Shedd and Harford, 1959; Jerger, 1962), reviewed by Buus, Florentine and Redden (1982a, 1982b). This test measures the ability to detect brief (200-ms) 1-dB increments in level of a continuous tone presented at 20 dB SL. Typically, cochlear hearing loss leads to an *improvement* in the ability to detect the changes in level.

However, intensity resolution is not always better than normal in people with cochlear hearing loss. The results depend on the level of testing and on the pattern of hearing loss of the subject. Typically, people with cochlear hearing loss show better intensity discrimination than normal when the comparison is made at equal, low, SLs, for example 10 or 20 dB SL. However, intensity discrimination is usually about the same as normal (and is sometimes worse than normal) when the comparison is made at equal, high, sound pressure levels. This is true for increment detection (Buus, Florentine and Redden, 1982a, 1982b), modulation detection (Glasberg and Moore, 1989; Turner, Zwislocki and Filion, 1989) and pulsed-tones intensity discrimination (Glasberg and Moore, 1989; Turner, Zwislocki and Filion, 1989; Schroder, Viemeister and Nelson, 1994).

It is not entirely clear why loudness recruitment only leads to better-than-normal intensity discrimination at low SLs. It is possible that the variability in the loudness sensation increases with increases in the slope of the loudness growth function. A steeper loudness growth function in an ear with recruitment leads to a larger change in average loudness for a given change in intensity, but this does not necessarily lead to improved intensity discrimination, because the variability of the loudness sensation increases by about the same factor (Zwislocki and Jordan, 1986).

VIII PERCEPTUAL CONSEQUENCES OF ALTERED LOUDNESS PERCEPTION

VIII.1 CONSEQUENCES OF LOUDNESS RECRUITMENT AND REDUCED DYNAMIC RANGE

The most prominent change in loudness perception associated with cochlear hearing loss is loudness recruitment and reduced dynamic range. For sounds with inherent amplitude fluctuations, such as speech or music, this results in an exaggeration of the perceived dynamic qualities. The sound appears to fluctuate more in loudness than it would for a normally hearing person. When listening to speech, the loudness differences between consonants and vowels may be greater than normal. When listening to music, the *forte* passages may be perceived at almost normal loudness, but the *piano* passages may be inaudible. Simulations of these effects can be found on a compact disc available from the author (Moore, 1997). One might regard the normal auditory system as containing a built-in fast-acting *automatic gain control* (*AGC*) system; this system is lacking or reduced in effectiveness in people with cochlear hearing loss. The perception of exaggerated loudness changes can play a role in several other aspects of auditory perception. The role in temporal processing is reviewed in Chapter 5, Section IV.3, and the role in speech perception is reviewed in Chapter 8, Section VIII.1.

The reduction in dynamic range has practical implications for the design and use of hearing aids. When hearing aids were first developed, they acted primarily as linear amplifiers, except that they had some form of output limiting or clipping to prevent the user from being exposed to excessively loud sounds. When the volume control was set so that the more intense sounds were at a comfortable loudness, then some weak sounds were inaudible owing to the reduced dynamic range. If the volume control was set to make the weak sounds audible, then intense sounds became unpleasantly loud (or sounded distorted because of limiting/clipping). In practice, the amount of amplification selected for use in everyday life was often rather small (Leijon, 1990), meaning that many weak environmental and speech sounds were inaudible.

These problems have been alleviated by the use of hearing aids incorporating *automatic volume control* or *compression*; such aids are described in more detail in Chapter 9.

VIII.2 PERCEPTUAL CONSEQUENCES OF REDUCED LOUDNESS SUMMATION

As described earlier, a narrowband sound and a broadband sound with the same intensity usually differ considerably in loudness for normally hearing people (the broadband sound being louder), but differ less or not at all in loudness for people with cochlear hearing loss. This means that the relative loudness of complex sounds may differ for normally hearing and hearing-impaired people; sound A might be judged louder than sound B by a normally hearing person but might be judged as less loud than sound B by a hearing-impaired person. This complicates the fitting of hearing aids, especially hearing aids with multi-band compression (see Chapter 9 for details of such aids).

VIII.3 PERCEPTUAL CONSEQUENCES OF ALTERED INTENSITY DISCRIMINATION

As reviewed above, intensity discrimination by people with cochlear hearing loss can be better than normal when the comparison is made at equal SL. However, intensity discrimination is not better than normal and can be worse than normal when the comparison is made at equal SPL. Even when intensity discrimination is worse than normal, this does not appear to lead to marked problems, since it is rare in everyday life for critical information to be carried by small changes in intensity. Although intensity contrasts can convey information in speech, the contrasts involve rather large changes in intensity, changes that are well above the threshold of detection for both normally hearing and hearing-impaired people.

Potential problems can arise when hearing aids incorporating fast-acting compression are used; details are given in Chapter 9, Section III.4. Such hearing aids reduce differences in level between sounds. If a large amount of compression is used, then information-bearing intensity contrasts may be reduced to the point where they are difficult for the hearing-impaired person to discriminate. This can lead to impaired speech perception (Plomp, 1994; Noordhoek and Drullman, 1997). However, moderate amounts of compression do not seem to have such deleterious effects, and even have beneficial effects by restoring the audibility of weak sounds. These issues are discussed further in Chapter 9.

5 Temporal Resolution and Temporal Integration

I INTRODUCTION

All sounds are characterized by pressure variations over time. However, if the intensity of a sound remains constant over time, and if its frequency content is also constant over time, the sound is said to be steady; it is heard as an unchanging sound. An example is a sustained tone produced by a musical instrument such as the oboe. A sound such as a broadband white noise (which contains equal energy at all frequencies) is also heard as a steady sound, even though its waveform is not regular but fluctuates rapidly from moment to moment. In the case of the noise, the fluctuations are mostly too rapid to be heard distinctly. *Temporal resolution* refers to the ability to detect changes over time. Often, it involves the detection of changes in the envelope of a sound, for example the detection of a brief gap in a sound or the detection of amplitude modulation (AM) of a sound. *Forward* and *backward masking* may also be regarded as situations involving temporal resolution (Moore *et al.*, 1988). If a brief signal is presented close in time to a masker, then temporal resolution may be insufficient to separate the signal and masker, and masking will occur. One way to conceptualize this process is to assume that the representation of sounds at higher levels in the auditory system takes some time to build up and decay. For example, one might explain forward masking by assuming that, when the masker is turned off, the internal representation of the masker takes some time to decay. If a brief signal is presented during this decay time, then forward masking may occur.

It is important to distinguish between temporal resolution (or acuity) and *temporal integration* (or summation). The latter refers to the ability of the auditory system to combine information over time to enhance the detection or discrimination of stimuli; generally, the longer a sound is, the better it is detected and discriminated. Temporal integration is described in the last part of this chapter.

Many studies have shown that temporal resolution can be adversely affected by cochlear hearing loss. To understand why temporal resolution is affected, it is helpful to use a model of temporal processing in the normal auditory system and to consider how the different stages of the model may be altered by cochlear pathology. That is the approach taken in this chapter.

In characterizing temporal resolution in the auditory system, it is important to take account of the filtering that takes place in the peripheral auditory system. The different frequency components in complex sounds are partly resolved on the basilar membrane (BM). One can regard the auditory system as having frequency-selective 'channels', each channel being responsive to a limited range of frequencies. Temporal resolution depends on two main processes: analysis of the time pattern occurring within each frequency channel and comparison of the time patterns across channels. This chapter concentrates on within-channel processes, since there have been few studies of across-channel processing in hearing-impaired subjects.

II MODELLING WITHIN-CHANNEL TEMPORAL RESOLUTION IN NORMAL HEARING

An example of a model of temporal resolution is illustrated in Figure 5.1. Each stage of the model is discussed below.

II.1 BANDPASS FILTERING

There is an initial stage of bandpass filtering, reflecting the action of the *auditory filters*. For simplicity, only one filter is shown; in reality, there is an array of parallel channels, each like that shown in the figure. When a brief signal is passed through a bandpass filter, the filter responds over a longer duration than that of the input signal. Generally, the narrower the filter, the more the output is stretched in time relative to the input. This is illustrated in Figure 5.2, which shows the response of a simulated auditory filter to a brief impulse. The filter was centred at 1000 Hz. The narrowest filter had a bandwidth of 130 Hz (which corresponds to the value of ERB_N for a centre frequency of 1000 Hz) and the other filters had bandwidths two, four and eight times that value. The response, called the *impulse response*, clearly shortens in time as the filter becomes broader. The auditory filters have bandwidths that decrease progressively with decreasing centre frequency (Glasberg and Moore, 1990); see Chapter 3, Section IV.1. One might expect, therefore, that the auditory filters would play some role in limiting temporal resolution, this effect being greater at low centre frequencies. The evidence relating to this expectation will be presented later.

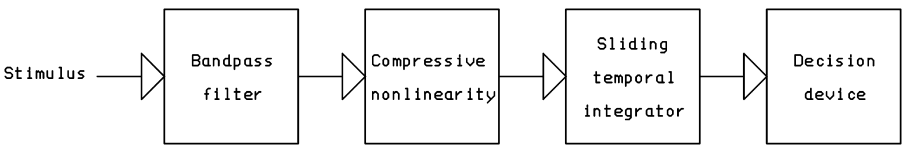


Figure 5.1. A block diagram showing the stages of a model of temporal resolution.

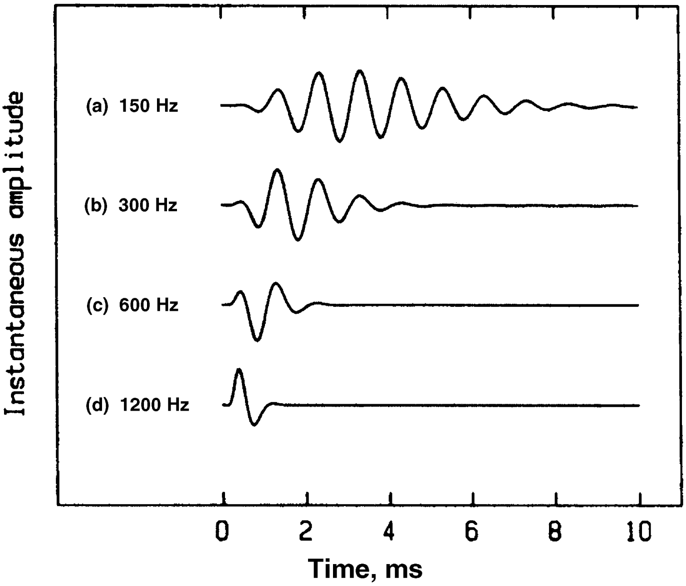


Figure 5.2. The response to a brief impulse of a simulated auditory filter, with a bandwidth of 150 Hz, and filters with bandwidths two times, four times or eight times greater than that value. All filters are centred at 1 kHz. The responses are not drawn to scale. The peak amplitude of the response actually increases as the filter bandwidth increases.

II.2 THE NONLINEARITY

Each filter is followed by a nonlinear device. This nonlinear device is meant to reflect the operation of several processes that occur in the peripheral auditory system. For example, nerve spikes tend to occur at a specific phase of the stimulating waveform on the BM. An effect resembling this can be achieved by a nonlinear process called half-wave rectification; in this process, only the parts of the waveform with a specific polarity (e.g. the positive parts) are passed, while the parts of the waveform with opposite polarity are set to zero. Another significant nonlinearity is the compressive *input-output function* of the BM; see Chapter 1, Section IV.3 and Figure 1.10. In recent models of temporal resolution, the nonlinear device includes these two processes, that is rectification and a compressive nonlinearity, resembling the compressive input-output function on the BM (Oxenham and Moore, 1994, 1997; Moore, Peters and Glasberg 1996). As noted in Chapters 3 and 4, it is unrealistic to treat the filtering on the BM and the compressive nonlinearity as separate stages, but this probably does not seriously undermine the usefulness of the model. A method for determining the characteristics of the nonlinearity is described later in this chapter.

II.3 THE SLIDING TEMPORAL INTEGRATOR

The output of the nonlinear device is fed to a 'smoothing' device, which can be implemented either as a lowpass filter (Viemeister, 1979) or a sliding temporal integrator (Moore *et al.*, 1988; Plack and Moore, 1990). Usually, the smoothing device is thought of as occurring after the auditory nerve; it is assumed to reflect a relatively central process. The device determines a weighted average of the output of the compressive nonlinearity over a certain time interval, or 'window'. The weighting function is often modelled as a pair of back-to-back exponential functions, as illustrated in Figure 5.3. This weighting function is sometimes called the 'shape' of the temporal window. Most weight is given to the output of the nonlinear device at times close to the temporal centre of the window, and progressively less weight is given to the output at times farther from the centre. The window itself is assumed to slide in time, so that the output of the temporal integrator is a weighted running average of the input. This has the effect of smoothing rapid fluctuations while preserving slower ones. When a sound is turned on abruptly, the output of the temporal integrator takes some time to build up. Similarly, when a sound is turned off, the output of the integrator takes some time to decay.

It is often assumed that backward and forward masking depend on the process of build up and decay. For example, if a brief signal is rapidly followed by a masker (backward masking), the response to the signal may still be building up when the

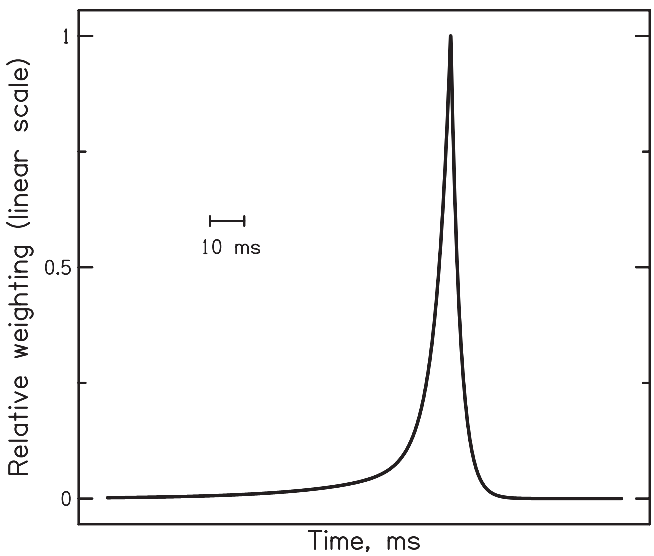


Figure 5.3. The 'shape' of the temporal window. This is a weighting function applied to the output of the nonlinear device. It performs a weighted running average of the output of the nonlinear device. The shape is plotted with a linear ordinate as a function of time.

masker occurs. If the masker is sufficiently intense, then its internal effects may 'swamp' those of the signal. Similarly, if a brief signal follows soon after a masker (forward masking), the decaying response to the masker may swamp the response to the signal.

The operation of the sliding temporal integrator is illustrated in Figure 5.4. The panels on the left-hand side show several different signals applied to the input of

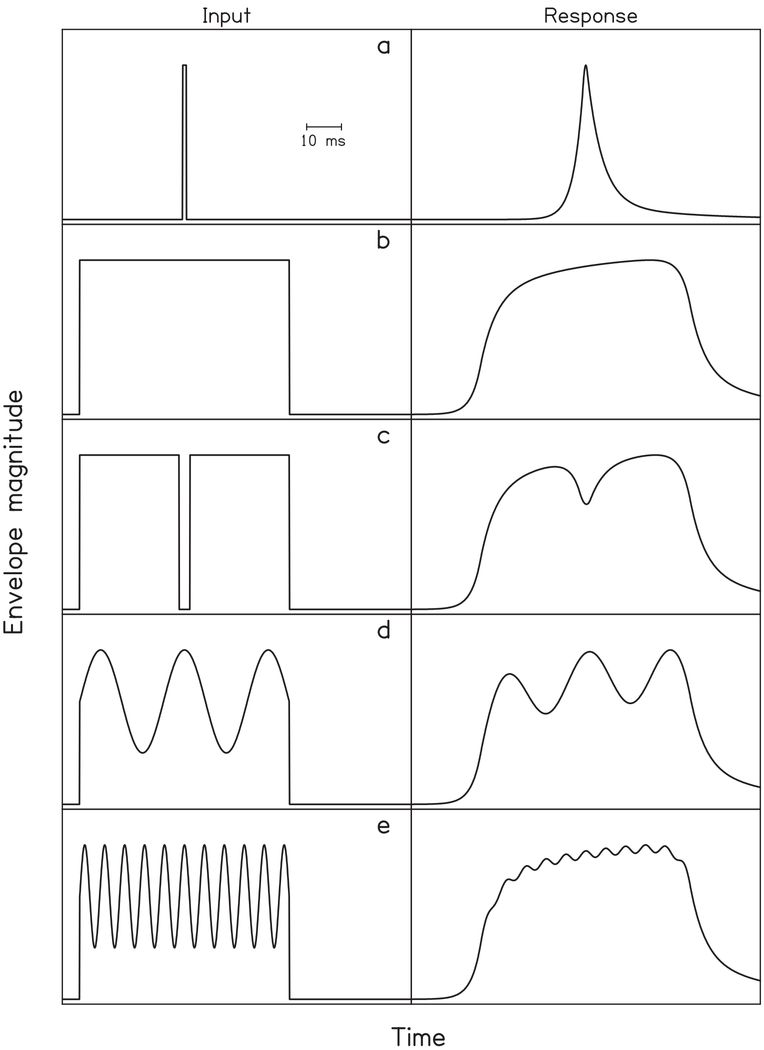


Figure 5.4. Examples of the influence of the sliding temporal integrator on the envelopes of sounds. The panels on the left show inputs to the sliding temporal integrator. The panels on the right show the corresponding outputs.

the integrator. These signals can be thought of as corresponding to the envelopes of audio signals, not the waveforms. On the right are shown the outputs of the sliding temporal integrator. In response to a brief pulse (panel a), the output builds up and then decays. The build up is more rapid than the decay because the window shape is asymmetric in time. In fact, the impulse response shown in the right-hand part of panel a is simply the window shape played backwards in time. The asymmetry assumed for the window shape makes it possible to account for an asymmetry between forward and backward masking; when the time interval between the masker and the signal is increased, backward masking decreases more rapidly than forward masking (Oxenham and Moore, 1994).

In response to an input with a rectangular envelope (panel b), the output builds up, stays at a nearly steady value for some time and then decays. In response to an input with a temporal gap (panel c), the output shows a build up, a steady part, a 'dip', another steady part and then a decay. The dip corresponds to the gap, but it is like a partially filled-in representation of the gap. In response to an input with a slow sinusoidal fluctuation, such as might be produced by an amplitude-modulated tone (panel d), the output also shows a sinusoidal modulation; slow fluctuations are preserved at the output of the sliding temporal integrator. In response to an input with a fast sinusoidal fluctuation (panel e), the output shows a much reduced amount of fluctuation; fast fluctuations are attenuated at the output of the sliding temporal integrator.

II.4 THE DECISION DEVICE

The output of the sliding temporal integrator is fed to a decision device. The decision device may use different 'rules' depending on the task required. For example, if the task is to detect a brief temporal gap in a signal, the decision device might look for a dip in the output of the temporal integrator (Moore, 2003). If the task is to detect AM of a sound, the device might assess the amount of modulation at the output of the sliding temporal integrator (Viemeister, 1979).

II.5 CHARACTERIZING THE NONLINEAR DEVICE AND THE SLIDING TEMPORAL INTEGRATOR

In the model, the characteristics of the auditory filters are based on auditory filter shapes derived from masking experiments, as described in Chapter 3, Section IV. However, it is also necessary to define the characteristics of the nonlinearity and the sliding temporal integrator. An approach to this problem was described by Oxenham and Moore (1994). They performed an experiment in which they used a noise masker and a brief 6-kHz signal. In one set of conditions, the signal was presented after the masker (forward masking). In another set, it was presented before the masker (backward masking). In a third set of conditions, the signal was presented between two bursts of the masker; this involves a combination of forward and backward masking. As described earlier, forward and backward masking can be accounted for in terms of the build up and decay processes at the output of the sliding temporal integrator.

An interesting effect is observed in cases when a forward and backward masker are combined. It might be thought that if two different maskers are equally effective (i.e. each produces the same amount of masking) the combination of the two maskers would result in a doubling of the signal energy required for threshold (Green and Swets, 1974); this corresponds to an increase in the signal level at threshold of 3 dB. In fact, the signal threshold often increases by more than this. The amount by which signal threshold exceeds the prediction is referred to as excess masking. Combining two equally effective non-simultaneous maskers (one forward and one backward) *consistently* results in excess masking, usually of 7–12 dB at moderate sound levels (Wilson and Carhart, 1971; Cokely and Humes, 1993).

This excess masking can be explained if it is assumed that each stimulus (the forward masker, signal and backward masker) is subjected to a compressive nonlinearity before the effects of the stimuli are combined in a *linear* temporal integrator (Penner and Shiffrin, 1980; Penner, 1980a), as assumed in the model of temporal resolution shown in Figure 5.1. To understand how the compressive nonlinearity accounts for excess masking, consider the following. Imagine that two equally effective non-simultaneous maskers (one forward and one backward) are presented together. At the output of the temporal integrator, the decay of response to the forward masker is summed with the build up of response to the backward masker. It is assumed that, at the time when the brief signal produces its own maximum response at the output of the temporal integrator, the effects of the forward and backward masker are equal (as they are equally effective maskers). The integrator itself is a linear device, and so the internal effect evoked by the two maskers is simply double the effect evoked by either alone. Thus, in order to reach threshold, the level of the signal has to be increased relative to the level required for a single masker. In fact, to reach the signal threshold, the *internal* effect of the signal must also be doubled. This requires more than a 3-dB increase in signal threshold because the signal itself is independently compressed.

Oxenham and Moore (1994) showed that their results could be used to separate the effects of the temporal integrator and the compressive nonlinearity prior to the integrator. A good fit to their forward and backward masking data was obtained when the stimulus intensity at the output of the simulated auditory filter was raised to a power between 0.25 and 0.35. If, for example, the intensity is raised to the power 0.3, then a 10-fold increase in power (corresponding to 10 dB) would be needed to double the internal effect of the signal. Thus, for two equally effective maskers, one forward and one backward, excess masking of 7 dB is predicted.

Oxenham and Moore (1994) also used their data to derive the weighting characteristic, or shape, of the temporal window (see the schematic illustration in Figure 5.3), following an approach proposed by Moore *et al.* (1988). However, the derivation is rather complex, and is beyond the scope of this book. The reader is referred to the original publications for details. In fact, the weighting function shown in Figure 5.3 corresponds to that derived by Oxenham and Moore.

In the next section, data on temporal resolution in normally hearing people are presented and interpreted in terms of the model. Then, data on temporal resolution in hearing-impaired people are described and evaluated.

III TEMPORAL RESOLUTION IN NORMAL HEARING

III.1 THE EFFECT OF CENTRE FREQUENCY ON GAP DETECTION

As mentioned above, if the auditory filter plays a role in limiting temporal resolution, one would expect temporal resolution to improve with increasing frequency. Several researchers have measured thresholds for detecting a gap in narrowband sounds (tones or bands of noise) as a function of centre frequency; the duration of the gap is adjusted to find the point where it is just detectable. When a narrowband sound is turned off and on abruptly to introduce a temporal gap, energy ‘splatter’ occurs outside the nominal frequency range of the sound; this may be heard as a click or a thud. People can use the click or thud as a cue for detecting the gap. However, it is usually argued that this reflects the use of spectral information rather than temporal information; it depends upon frequency selectivity rather than upon temporal resolution per se. Two methods have been used to prevent the splatter being detected. In one, the narrowband sound is presented in a background sound, usually a broadband noise, designed to mask the splatter. In the other, the gap is introduced by turning the sound off and on again gradually; this reduces spectral splatter, but it prevents the use of very brief gaps.

The pattern of results found for the detection of gaps in bands of noise depends on the bandwidth of the noise, and on how that bandwidth varies with centre frequency. If the *relative* bandwidth is held constant (i.e. the bandwidth is a constant proportion of the centre frequency), then gap thresholds decrease monotonically with increasing centre frequency (Fitzgibbons and Wightman, 1982; Fitzgibbons, 1983; Shailer and Moore, 1983). This is the effect that would be expected if the auditory filters played a role in limiting gap detection. However, if the absolute bandwidth is held constant, gap thresholds do not vary markedly with centre frequency (Shailer and Moore, 1985; de Filippo and Snell, 1986; Eddins, Hall and Grose, 1992). On the other hand, if the bandwidth is varied keeping either the centre frequency or the upper spectral edge fixed, gap thresholds increase with decreasing noise bandwidth (Shailer and Moore, 1983, 1985; Eddins, Hall and Grose, 1992; Glasberg and Moore, 1992; Snell, Ison and Frisina, 1994; Eddins and Green, 1995).

To understand this pattern of results, it is necessary to take account of the fact that noise bands fluctuate randomly in amplitude from moment to moment. The rapidity of these fluctuations increases with increasing bandwidth. The slow fluctuations are easy to hear as changes in loudness from moment to moment. More rapid fluctuations are heard as a kind of ‘roughness’. Gap thresholds for noise bands are probably partly limited by the inherent fluctuations in the noise (Shailer and Moore, 1983, 1985; Green, 1985; Eddins and Green, 1995). Randomly occurring dips in the noise are ‘confused’ with the gap to be detected. The confusion is maximal for dips comparable in duration to the gap. In practice, this means that noise with a narrow bandwidth, and hence slow fluctuations, creates the greatest confusion and gives the largest gap thresholds. The data are consistent with this view.

In summary, gap thresholds measured using bands of noise do not provide clear evidence of an improvement in temporal resolution with centre frequency. However, the gap thresholds are influenced by the inherent fluctuations in the noise. The slow fluctuations associated with small bandwidths lead to large gap thresholds.

Shailer and Moore (1987) studied the ability of subjects to detect a temporal gap in a sinusoid, which has no inherent fluctuations. To mask spectral splatter associated with the introduction of the gap, the sinusoid was presented in a continuous noise with a spectral notch at the frequency of the sinusoid. In one of their conditions, called preserved phase, the sinusoid was turned off at a positive-going zero crossing (i.e. as the waveform was about to change from negative to positive values) and it started (at the end of the gap) at the phase it would have had if it had continued without interruption. Thus, it was as if the gap had been 'cut out' from a continuous sinusoid.

Shailer and Moore (1987) found that the gap threshold was roughly constant at about 5 ms for centre frequencies of 400, 1000 and 2000 Hz. Moore, Peters and Glasberg (1993) measured gap thresholds for centre frequencies of 100, 200, 400, 800, 1000 and 2000 Hz, using a condition similar to the preserved-phase condition of Shailer and Moore. The gap thresholds were almost constant, at 6–8 ms, over the frequency range 400–2000 Hz, but increased somewhat at 200 Hz, and increased markedly, to about 18 ms, at 100 Hz. Individual variability also increased markedly at 100 Hz.

Overall, the results of experiments using narrowband stimuli indicate that temporal resolution does not vary markedly with frequency, except at very low frequencies (200 Hz and below). This suggests in turn that the smoothing produced by the auditory filters does not play a major role, except perhaps at very low frequencies.

III.2 TEMPORAL MODULATION TRANSFER FUNCTIONS

Gap detection experiments give a single number – the gap threshold – to describe temporal resolution. A more general approach is to measure the threshold for detecting changes in the amplitude of a sound as a function of the rapidity of the changes. In the simplest case, white noise is sinusoidally amplitude modulated, and the amount of modulation required to detect the modulation is determined as a function of modulation rate. The function relating the threshold amount of modulation to modulation rate is known as a *temporal modulation transfer function* (TMTF). Modulation of white noise does not change its long-term magnitude spectrum. An example of the results is shown in Figure 5.5, adapted from Bacon and Viemeister (1985). For low modulation rates, performance is limited by the amplitude resolution of the ear, rather than by temporal resolution. Thus, the threshold is almost independent of modulation rate for rates up to about 30–50 Hz. As the rate increases beyond 50 Hz, temporal resolution starts to have an effect; the threshold increases, and for rates above about 1000 Hz the modulation is difficult to detect at all. Thus, sensitivity to AM decreases

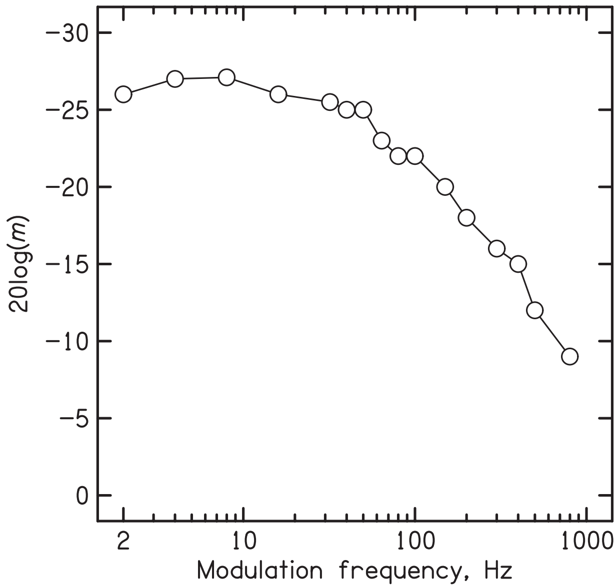


Figure 5.5. A TMTF. A broadband white noise was sinusoidally amplitude modulated, and the threshold amount of modulation required for detection was plotted as a function of modulation rate. The amount of modulation is specified as 20log(m), where m is the modulation index (m can vary between 0 and 1; a value of 1 corresponds to 100 % modulation). The higher the sensitivity to modulation, the more negative is 20log(m). Adapted from Bacon and Viemeister (1985).

progressively as the rate of modulation increases. The shapes of TMTFs do not vary much with overall sound level, but the ability to detect the modulation does worsen at low sound levels.

III.3 THE RATE OF RECOVERY FROM FORWARD MASKING

As described in Chapter 3, the threshold for detecting a signal in forward masking decreases progressively with increasing time delay of the signal relative to the end of the masker (see Figure 3.12). As described earlier, the rate of recovery from forward masking is often considered as a measure of temporal resolution; the more rapidly the threshold drops as the time interval between the masker and signal increases, the better is temporal resolution. However, the rate of decrease of threshold (often described as the decay of masking) depends on the masker level; the decay is more rapid for higher masker levels. This effect can be understood by considering the form of the input-output function on the BM, as illustrated schematically in Figure 1.11 (Oxenham and Moore, 1997). To understand this, consider an example based on the following assumptions:

1. The threshold for detecting a brief signal is measured for two different times following the end of the masker, t_1 and t_2, where $t_1 < t_2$.
2. The internal effect of the masker, E_M, as reflected in the output of the sliding temporal integrator, decays by a factor X over this time; this decay is determined by the shape of the temporal window, as illustrated in Figure 5.3.
3. X is independent of the overall masker level. This is equivalent to assuming that the temporal integrator is linear.
4. The threshold for detecting the brief signal is reached when the peak internal effect evoked by the signal, E_S, is a constant proportion of the internal effect of the masker at the times t_1 and t_2.

To take a specific example, assume that $X = 0.1$; the internal effect of the masker at time t_1 is 10 times the effect at t_2. In order to reach threshold at the two different delay times, the internal effect of the signal, E_S, has to be a factor of 10 greater at time t_1 than at time t_2. Consider now how much the signal intensity has to be changed in order to give a change in its internal effect by a factor of 10. If the masker is intense, then the signal thresholds will be relatively high; the signal levels at threshold might fall in the range 40–80 dB SPL, where the input-output function on the BM is highly compressive. Assuming a typical amount of mid-level compression, the signal intensity would have to change by a factor of about 1000 in order to change its internal effect by a factor of 10. This corresponds to a change in signal level of about 30 dB. Thus, when the masker is sufficiently intense to give signal thresholds in the range 40–80 dB, the signal threshold would decay by 30 dB when its delay time following the masker was increased from t_1 to t_2.

Consider now the situation where the masker is at a lower level, and the signal threshold is correspondingly lower. Say, for example, that thresholds fall in the range 10–30 dB, where the input-output function on the BM approaches linearity. In this case, the intensity of the signal needs to be changed by a factor only a little greater than 10 to change its internal effect by a factor of 10. In other words, the signal threshold would decay by only a little more than 10 dB when its delay time following the masker was increased from t_1 to t_2. Clearly, the decay of masking is much less in this case than in the case where the signal levels were higher.

In practice, the input-output function on the BM becomes progressively more compressive as the input level is increased over the range 15 to about 50 dB. For the masker levels and signal durations typically used in forward masking experiments, signal thresholds fall within this range. Hence, a progressive change in the rate of decay of forward masking would be expected with increasing masker level; that is exactly what is observed (see Figure 3.12). Notice, however, that the rate of decay of forward masking is predicted to depend on the signal level at short delay times, not the masker level. Of course, higher masker levels usually imply higher signal levels. But a higher signal level at threshold could also be achieved by using a shorter signal, for example.

In summary, the rate of decay of forward masking is greater for higher masker levels. This can be understood by assuming that the sliding temporal integrator is linear, but the nonlinearity prior to the temporal integrator is more compressive at medium levels than at low levels; less compression leads to a slower rate of decay of forward masking.

IV TEMPORAL RESOLUTION IN PEOPLE WITH COCHLEAR DAMAGE

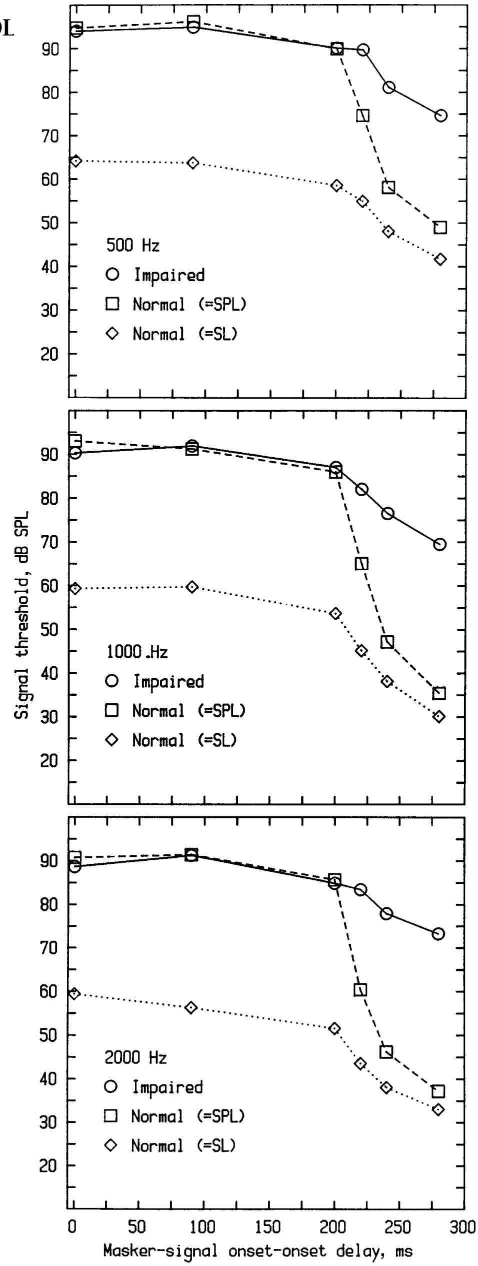
Some measures of temporal resolution in subjects with cochlear damage appear to show reduced temporal resolution, while others do not. Several factors can affect the results, and not all of these are directly connected with temporal processing itself. This section considers several of these factors.

IV.1 THE INFLUENCE OF SOUND LEVEL ON GAP DETECTION AND THE RATE OF DECAY OF FORWARD MASKING

One important factor influencing measures of temporal resolution is the sound level used. Many measures of temporal resolution show that performance in normally hearing subjects worsens at low sensation levels (SLs) (Plomp, 1964b; Shailer and Moore, 1983; Buus and Florentine, 1985; Fitzgibbons and Gordon-Salant, 1987; Peters, Moore and Glasberg, 1995). This is not unique to temporal resolution; performance on many tasks worsens at low SLs, presumably because less neural information is available, or because of the greater effects of internal noise at low SLs. It is not generally possible to test hearing-impaired subjects at high SLs, because they have *loudness recruitment*; sounds with levels of 90–100 dB SPL appear as loud as they would to a normally hearing listener, as described in Chapter 4, Section III. On some measures of temporal resolution, such as the detection of gaps in bands of noise, hearing-impaired subjects appear markedly worse than normal subjects when tested at the same SPLs, but only slightly worse at equal SLs (Fitzgibbons and Wightman, 1982; Tyler *et al.*, 1982; Glasberg, Moore and Bacon, 1987; Nelson and Thomas, 1997). For example, Nelson and Thomas (1997) measured gap detection thresholds for normally hearing and hearing-impaired listeners using a 650-Hz-wide noise centred at about 2850 Hz. At equal SLs, most hearing-impaired listeners showed near-normal performance,

→

Figure 5.6. The threshold for the detection of a brief sinusoidal signal presented at various times during and following a noise masker. The temporal position of the signal is plotted relative to the onset of the masker; times greater than 200 ms correspond to forward masking. Mean results are shown for the normal and impaired ears of five subjects with unilateral cochlear damage. Each panel shows results for a different signal frequency. The masker was presented at a single high level to the impaired ear (84 dB SPL). The masker was presented to the normal ear either at the same SPL or the same SL.



especially when the SL was relatively low. However, at equal SPLs, and also when the level was adjusted to give comfortable loudness, the hearing-impaired listeners had markedly larger gap thresholds than normal.

As mentioned earlier, the rate of decay of forward masking is often considered as a measure of temporal resolution; the more rapidly the threshold drops as the time interval between the masker and signal increases, the better is temporal resolution. The effect of overall level for the case of forward masking is illustrated in Figure 5.6, adapted from Glasberg, Moore and Bacon (1987). They tested subjects with cochlear hearing loss in one ear only. The brief sinusoidal signal was presented at various times during and following a noise masker. The masker was presented at a fixed high level to the impaired ear of each subject. In the normal ear, it was presented both at the same SPL and at the same SL as for the impaired ear.

It is clear from the figure that the rate of decay of forward masking is much more rapid for the normal ear than for the impaired ear when the comparison is made at equal SPL. However, at equal SL, the difference is much reduced. This effect of level is not unexpected, since the rate of decay of forward masking in normal ears depends strongly on sound level, as described in Chapter 3, Sections IV.2 and V.2 and earlier in this chapter. The slow rate of decay of forward masking found for normally hearing people at low sound levels was explained above in terms of the reduced compressive nonlinearity of the BM input-output function at low levels. Similarly, the slow rate of decay of forward masking that is commonly found in people with cochlear hearing loss probably depends partly on the loss of compressive nonlinearity on the BM. A prediction derived from this explanation is that the rate of decay of forward masking in people with cochlear hearing loss should vary less with level than is the case for normally hearing subjects. Indeed, if the input-output function on the BM is completely linear, then the rate of decay of forward masking should not vary at all with level. Experimental results are consistent with this prediction. In subjects with moderate-to-severe cochlear hearing loss, the rate of decay of forward masking does appear to be almost invariant with level (Oxenham and Moore, 1995, 1997).

In summary, two common measures of temporal resolution – gap detection and the rate of decay of forward masking – both show deleterious effects of cochlear hearing loss when the normal and impaired ears are compared at equal SPLs. At equal SLs, the discrepancy between normal and impaired hearing is less. Unfortunately, people with cochlear hearing loss usually listen at low SLs, since loudness recruitment makes it impossible to present sounds at high SLs without discomfort occurring. Hence, in practice, cochlear hearing loss leads to poorer temporal resolution than normal.

IV.2 THE INFLUENCE OF AUDIBLE BANDWIDTH ON TEMPORAL MODULATION TRANSFER FUNCTIONS AND GAP DETECTION

Another important consideration is the bandwidth available to the listeners. This can be clearly seen by consideration of studies measuring the TMTF. Several studies measuring TMTFs for broadband noise carriers showed that hearing-impaired listeners were generally less sensitive to high rates of modulation than normal listeners (Formby, 1982; Lamore, Verweij and Brocaar, 1984; Bacon and Viemeister, 1985). However, this may have been largely a consequence of the fact that high frequencies were inaudible to the impaired listeners (Bacon and Viemeister, 1985); most of the subjects used had greater hearing losses at high frequencies than at low, as is typical in cases of noise-induced hearing loss. When the broadband noise is lowpass filtered, as a crude simulation of the effects of threshold elevation at high frequencies, normally hearing subjects also show a reduced ability to detect modulation at high rates (Bacon and Viemeister, 1985).

Bacon and Gleitman (1992) measured TMTFs for broadband noise using subjects with relatively flat hearing losses. They found that at equal (high) SPLs performance was similar for hearing-impaired and normally hearing subjects. At equal (low) SLs, the hearing-impaired subjects tended to perform better than the normally hearing subjects. Moore, Shailer and Schooneveldt (1992) controlled for the effects of listening bandwidth by measuring TMTFs for an octave-wide noise band centred at 2 kHz, using subjects with unilateral and bilateral cochlear hearing loss. Over the frequency range covered by the noise, the subjects had reasonably constant thresholds as a function of frequency, both in their normal and their impaired ears. This ensured that there were no differences between subjects or ears in terms of the range of audible frequencies in the noise. To ensure that subjects were not making use of information from frequencies outside the nominal passband of the noise, the modulated carrier was presented in an unmodulated broadband noise background. The results for the subjects with unilateral impairments are shown in Figure 5.7. It can be seen that performance is similar for the normal and impaired ears, both at equal SPL and equal SL, although there is a slight trend for the impaired ears to perform better at equal SL.

Studies of gap detection also show clear effects of the audible frequency range of the stimuli. Thresholds for detecting gaps in broadband noise become progressively larger as the audible frequency range of the stimuli is reduced by increasing high-frequency hearing loss (Florentine and Buus, 1984; Buus and Florentine, 1985; Salvi and Arehole, 1985).

To summarize the results so far, people with cochlear hearing loss often show reduced temporal resolution as a result of the low SL of the stimuli and/or the reduced audible bandwidth of the stimuli. When these factors are controlled for, hearing-impaired subjects often perform as well as, or even better than, normal.

IV.3 THE INFLUENCE OF CHANGES IN THE COMPRESSIVE NONLINEARITY

For certain types of sounds, the temporal resolution of subjects with cochlear hearing loss seems to be worse than normal even when the stimuli are well above threshold and when all of the components of the stimuli fall within the audible range. This happens mainly for stimuli that contain slow random fluctuations in amplitude, such as narrow bands of noise. For such stimuli, subjects with cochlear damage often show larger gap detection thresholds than normal (Fitzgibbons and Wightman, 1982; Florentine and Buus, 1984; Buus and Florentine, 1985; Glasberg, Moore and Bacon, 1987). However, gap detection is not usually worse than normal when the stimuli are sinusoids, which

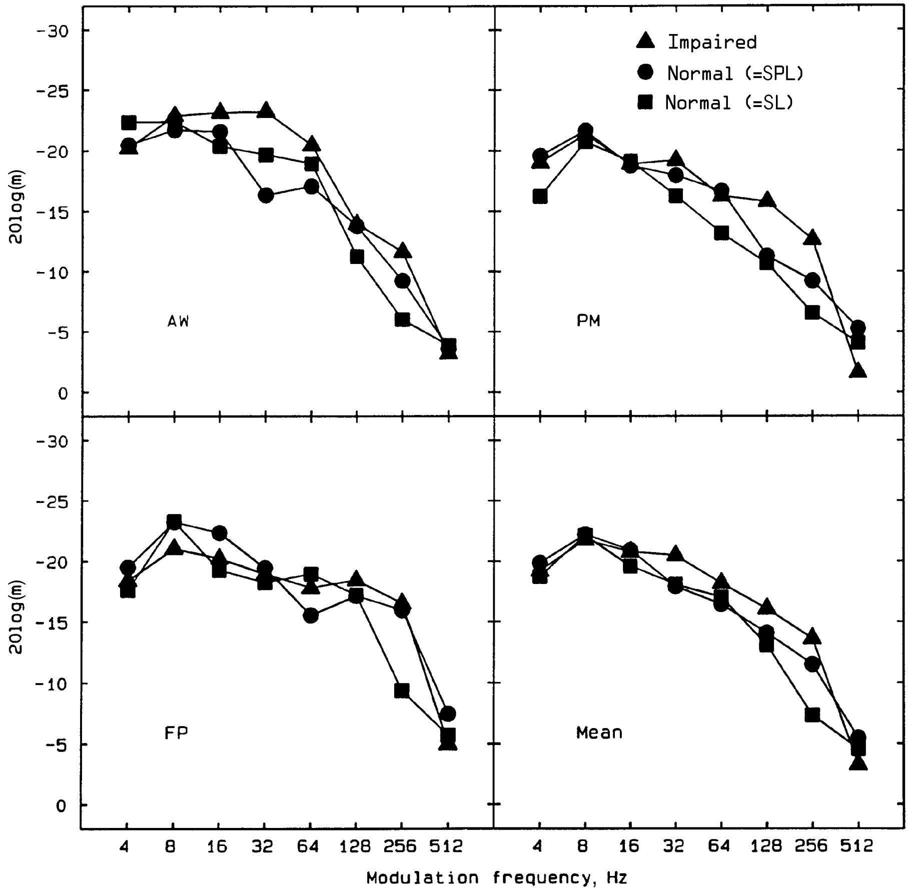


Figure 5.7. TMTFs obtained using a bandpass noise carrier for the normal and impaired ears of three subjects with unilateral cochlear hearing loss. Mean results for the three subjects are shown in the bottom right panel. Data from Moore, Shailer and Schooneveldt (1992).

do not have inherent amplitude fluctuations (Moore and Glasberg, 1988b; Moore *et al.*, 1989). Glasberg, Moore and Bacon (1987) and Moore and Glasberg (1988b) suggested that the poor gap detection for narrowband noise stimuli might be a consequence of loudness recruitment, the abnormally rapid growth of loudness with increasing intensity that occurs commonly in cases of cochlear hearing loss; see Chapter 4, Section III. For a person with recruitment, the inherent amplitude fluctuations in a narrowband noise would result in larger-than-normal loudness fluctuations from moment to moment (Moore, Wojtczak and Vickers, 1996), so inherent dips in the noise might be more confusable with the gap to be detected.

This idea can also be expressed in terms of the model of temporal resolution. It seems likely that loudness recruitment is caused primarily by a reduction in the compressive nonlinearity found in the normal cochlea; see Chapter 4, Section V.3. When cochlear damage occurs, the cochlea behaves in a more linear way, and the input-output function of the BM becomes less compressive, having a slope closer to unity (on log-log coordinates); see Figure 1.19.

To assess the idea that steeper input-output functions lead to impaired gap detection for stimuli with fluctuating envelopes, Glasberg and Moore (1992) processed the envelopes of narrow bands of noise so as to modify the envelope fluctuations. The envelope was processed by raising it to a power, N. If N is greater than unity, this has the effect of magnifying fluctuations in the envelope, thus simulating the effects of recruitment; higher powers correspond to greater degrees of simulated recruitment. If N is less than unity, fluctuations in the envelope are reduced. This represents a type of signal processing that might be used to compensate for recruitment; it resembles the operation of a fast-acting compressor or automatic gain control (AGC) system; see Chapter 9.

The values of N used were 0.5, 0.66, 1, 1.5 and 2. For $N = 1$, the stimuli were the same as unprocessed noise. A value of N of 2 simulates the type of recruitment typically found in cases of moderate to severe cochlear damage, where, for example, a 50-dB range of stimulus levels gives the same range of loudness as a 100-dB range of stimulus levels in a normal ear. Several different bandwidths of the noise were used.

Some examples of the envelopes of unprocessed and processed stimuli are shown in Figure 5.8. The envelopes are plotted on a logarithmic (dB) scale, as this seems more relevant to loudness perception than a linear amplitude scale. The bottom panel shows the envelope of a 'normal' noise band ($N = 1$) with a bandwidth of 10 Hz. The top panel shows the effect of squaring the envelope ($N = 2$), while the middle panel shows the result of raising the envelope to the power 0.5. The envelope fluctuations are obviously greatest in the top panel and smallest in the middle panel.

To prevent the detection of spectral splatter associated with the gap or with the envelope processing, the stimuli were presented in a continuous background noise. The spectrum of the noise was chosen so that it would be as effective as possible in masking the splatter while minimizing its overall loudness.

Figure 5.9 shows an example of results obtained using a subject with unilateral hearing loss of cochlear origin. The stimuli were presented at 85 dB SPL, a level which was well above the absolute threshold for both the normal and impaired ears (although the SL was lower in the impaired ear). The results for the normal ear were very similar to those of three normally hearing subjects who were also tested. Gap thresholds increased significantly with decreasing noise bandwidth, as expected.

For all noise bandwidths, gap thresholds increased as N increased. This effect was particularly marked for the smaller noise bandwidths. There was a significant interaction between bandwidth and N, reflecting the fact that changes in gap threshold with N were greater for small bandwidths. This supports the idea that fluctuations in the noise adversely affect gap detection; greater fluctuations lead to worse performance,

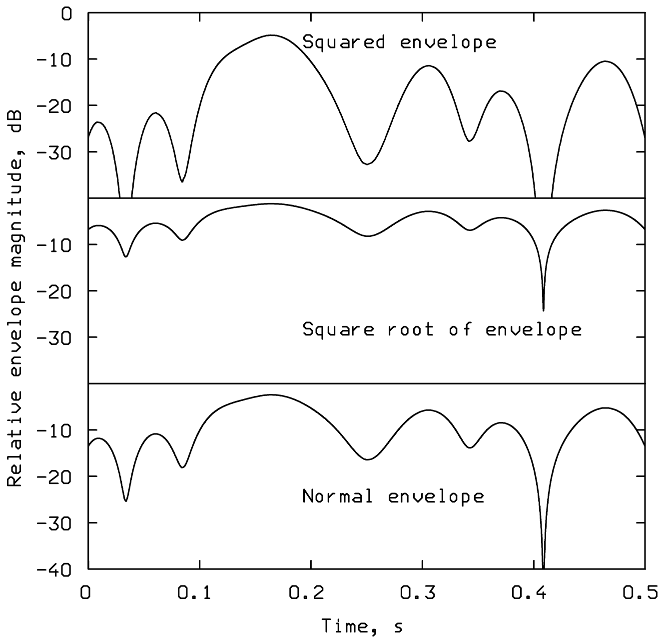


Figure 5.8. Examples of the envelopes of noise bands with $N = 1$ (unprocessed, bottom panel), $N = 0.5$ (middle panel) and $N = 2$ (top panel). The noise bandwidth was 10 Hz. The envelope magnitudes are plotted on a decibel scale.

especially when the fluctuations are slow.

Gap thresholds were larger for the impaired than for the normal ear. The overall geometric mean gap threshold was 12.8 ms for the normal ear and 27.2 ms for the impaired ear. Performance for the normal ear with $N = 2$ was roughly similar to performance for the impaired ear with unprocessed noise bands ($N = 1$); geometric mean gap thresholds were 26.9 ms for the former and 26.5 ms for the latter. Thus, the simulation of recruitment in the normal ear was sufficient to produce impaired gap detection, comparable to that actually found in the impaired ear.

For both normally hearing and hearing-impaired subjects, the effects of changing N decreased with increasing noise bandwidth. One reason for this is that slow fluctuations can be followed by the auditory system, whereas rapid fluctuations are smoothed to some extent by the central temporal integration process described earlier. The rapid fluctuations of the wider noise bands are smoothed in this way, thus reducing their influence on gap detection.

The results suggest that, for most subjects with cochlear hearing loss, recruitment, or, equivalently, a reduction in the compressive nonlinearity on the BM, may provide a sufficient explanation for increased gap thresholds. Thus, it is not usually necessary to assume any abnormality in temporal processing occurring after the cochlea. However,

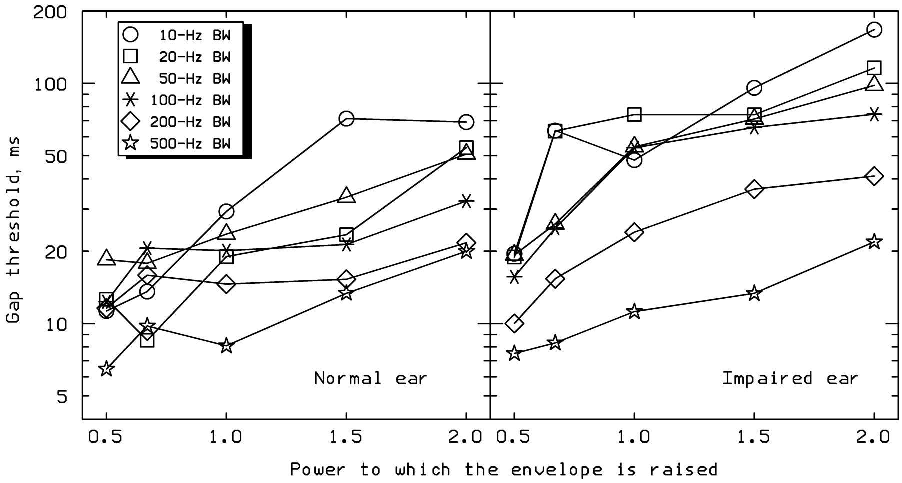


Figure 5.9. Thresholds for detecting a gap in a noise band whose envelope had been processed to enhance or reduce fluctuations. Gap thresholds are plotted as a function of the power to which the envelope was raised, with noise bandwidth as parameter; the higher the power, the greater the fluctuations. Results are shown for each ear of a subject with unilateral cochlear hearing loss. Data from Glasberg and Moore (1992).

a few subjects show impairments in temporal resolution even using non-fluctuating stimuli (Jesteadt *et al.*, 1976; Moore and Glasberg, 1988b; Moore *et al.*, 1989; Plack and Moore, 1991) or noise stimuli with a fairly large bandwidth (Nelson and Thomas, 1997), especially subjects with large hearing losses. It is possible that the subjects showing this impaired resolution had damage to both outer hair cells (affecting the active process and the compressive nonlinearity) and inner hair cells (reducing the amount of information conveyed in the auditory nerve), or that they had a retrocochlear component to their hearing loss.

For deterministic stimuli that have no inherent random fluctuations, hearing-impaired subjects can actually perform a little better than normally hearing subjects when tested at equal SLs. This applies, for example, to the detection of gaps in sinusoids (Moore and Glasberg, 1988b; Moore *et al.*, 1989).

V TEMPORAL INTEGRATION AT THRESHOLD

V.1 TEMPORAL INTEGRATION IN NORMALLY HEARING PEOPLE

It has been known for many years that the absolute threshold for detecting a sound depends upon the duration of the sound (Exner, 1876). For durations up to a few hundred milliseconds, the intensity required for threshold decreases as the duration increases. For durations exceeding about 500 ms, the sound intensity at threshold is roughly independent of duration. Many workers have investigated the relation between threshold and duration for tone pulses, over a wide range of frequencies and durations. The early work of Hughes (1946) and Garner and Miller (1947) indicated that, over a reasonable range of durations, the ear appears to integrate the intensity of the stimulus over time in the detection of short-duration tone bursts. This is often called temporal integration. In other words, the threshold corresponds to a constant *energy* rather than a constant intensity. For durations up to about 500 ms, the following formula is approximately true:

$$I \times t = \text{constant}, \tag{5.1}$$

where I is the threshold intensity for a tone pulse of duration t.

Thresholds as a function of duration are often plotted on dB versus log-duration coordinates. When plotted in this way, energy integration is indicated by the data falling on a straight line with a slope of −3 dB per doubling of duration; each time the duration is doubled, the intensity at threshold is halved, corresponding to a 3-dB decrease in level. Although the average data for a group of subjects typically give a slope close to this value, the slopes for individual subjects can differ significantly from −3 dB per doubling. This suggests that it would be unwise to ascribe too much significance to the average slope. It seems very unlikely that the auditory system would actually integrate stimulus energy; it is almost certainly neural activity which is integrated (Zwislocki, 1960; Penner, 1972). It may also be the case that the auditory system does not actually perform an operation analogous to integration. Rather, it may be that the threshold intensity decreases with increasing duration partly because a longer stimulus provides more chances to detect the stimulus through repeated sampling. This idea is sometimes called multiple looks (Viemeister and Wakefield, 1991).

V.2 TEMPORAL INTEGRATION IN PEOPLE WITH COCHLEAR HEARING LOSS

For people with cochlear hearing loss, the change in threshold intensity with signal duration is often smaller than it is for normally hearing people. If the thresholds are plotted on dB versus log-duration coordinates, the slopes are usually much less in absolute value than the typical value of −3 dB/doubling found for normally hearing people. This is often described as reduced temporal integration (Gengel and Watson, 1971; Pedersen and Elberling, 1973; Elliott, 1975; Chung, 1981; Hall and Fernandes, 1983; Carlyon, Buus and Florentine, 1990). There is a trend for higher absolute thresholds to be associated with flatter slopes. In other words, the greater the hearing loss, the more reduced is the temporal integration.

V.3 EXPLANATIONS FOR REDUCED TEMPORAL INTEGRATION IN PEOPLE WITH COCHLEAR HEARING LOSS

A number of explanations have been advanced to account for reduced temporal integration in people with cochlear hearing loss (Florentine, Fastl and Buus, 1988). One is that it results from a reduction or complete loss of the compressive nonlinearity on the BM. This leads to steeper input-output functions on the BM and to steeper rate-versus-level functions in the auditory nerve; see Chapter 1, Sections VII.1 and VII.2. According to the models of temporal integration proposed by Zwislocki (1960) and by Penner (1972), this will automatically lead to reduced temporal integration. Figure 5.10 illustrates schematically two rate-versus-level functions: the left-hand curve shows a typical function for a low-threshold neurone in a normal auditory system; the right-hand curve shows a typical function for a neurone in an auditory system with OHC damage. The curve is shifted to the right, reflecting a loss of sensitivity, and is steeper, reflecting the loss of the compressive nonlinearity on the BM. It is assumed that there is some residual compression on the normal BM

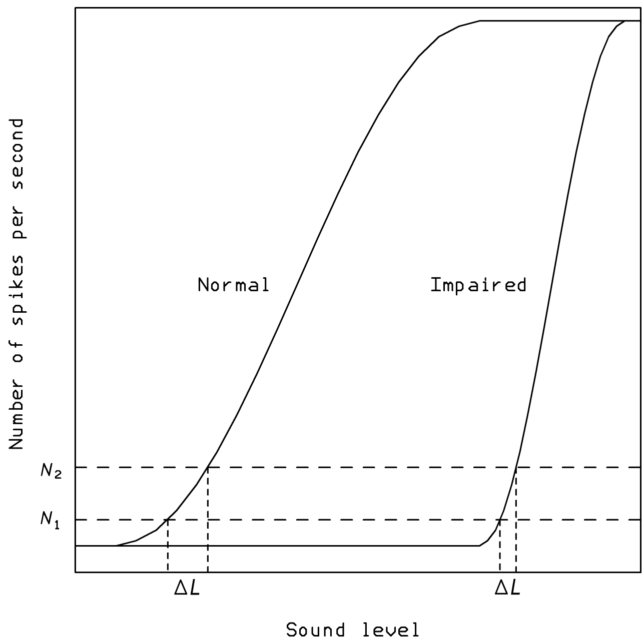


Figure 5.10. Schematic illustration of rate-versus-level functions in single neurones of the auditory nerve for a normal ear (left curve) and an impaired ear (right curve). The horizontal dashed lines indicate the mean firing rate needed for threshold for a long-duration sound (lower line, rate N_1) and a short-duration sound (upper line, rate N_2) on the assumption that threshold corresponds to a fixed total number of spikes.

at levels close to threshold; although the input-output function steepens at low levels (as illustrated in Figure 1.10), it does not become completely linear.

Consider now how steeper rate-versus-level functions can lead to reduced temporal integration. Assume that, to a first approximation, the threshold for detecting a sound requires a fixed number of neural spikes to be evoked by that sound. Assume also, for the sake of simplicity, that the neurones involved in detection at absolute threshold are relatively homogeneous in terms of their rate-versus-level functions. The lower dashed horizontal line in Figure 5.10 indicates the number of neural spikes per second, N_1, needed to achieve absolute threshold for a long-duration sound; in practice, of course, the absolute threshold depends upon the activity of many neurones, but if they are all similar, then the argument can be illustrated by considering just one. Notice that N_1 is the spike rate; the total number of spikes evoked by a sound of duration D would be $N_1 \times D$. If the duration of the sound is decreased by a factor R, then the level has to be increased to restore the total number of spikes evoked by the sound. Assume that the higher spike rate needed for the shorter-duration sound is N_2, where $N_2 = R \times N_1$. For example, if the duration is halved, the spike rate has to be increased by a factor of two to achieve the same total spike count. The increase in level, ΔL, needed to achieve this increased rate is greater for the normal than for the impaired ear, because the rate-versus-level function is steeper for the impaired ear, and this explains the reduced temporal integration.

Although this explanation is plausible, there are some difficulties associated with it. First, it is based on the assumption that there is some residual compression on the normal BM at levels close to threshold, which may not be the case (Plack and Skeels, 2007). Secondly, rate-versus-level functions are not always steeper than normal in cases of hearing loss. For example, cats with hearing loss caused by exposure to intense sounds do not show steeper-than-normal rate-versus-level functions (Heinz, Issa and Young, 2005).

VI TEMPORAL INTEGRATION AT SUPRATHRESHOLD LEVELS

A phenomenon similar to temporal integration at threshold is found for the perception of the loudness of sounds that are above the detection threshold. For normally hearing subjects, if the intensity of a sound is held constant and the duration is varied, the loudness increases with duration for durations up to 100–200 ms. If a sound of a particular duration is chosen as a reference sound, and a sound of shorter duration is adjusted so that it appears equally loud to the reference sound, then the shorter duration sound is adjusted to a higher intensity. As a very rough summary of the data, it may be stated that, for durations up to about 80 ms, constant energy leads to constant loudness; each time the duration is halved, the intensity has to be doubled to maintain constant loudness. This is equivalent to saying that the level must be increased by 3 dB for each halving of duration. However, considerable variability occurs across studies. For reviews, see Zwislocki (1969), Scharf (1978) and Buus, Florentine and Poulsen (1997).

The amount of temporal integration for loudness is sometimes quantified by estimating the difference in level between equally loud short and long tones. For example, if the difference in level required for equal loudness of a 100-ms tone and a 10-ms tone is 10 dB, it may be stated that the amount of temporal integration is 10 dB. Florentine, Buus and Poulsen (1996) and Buus, Florentine and Poulsen (1997) found that, for normally hearing listeners, the amount of temporal integration for sinusoidal tone bursts varied with sound level, being greatest for medium sound levels. They explained this in terms of the function relating loudness to sound level, pointing out that this function was shallower for medium sound levels than for very low or very high sound levels; see Chapter 4, Figure 4.2. The explanation is similar, in principle, to that given in the previous section to explain differences in temporal integration at threshold between normally hearing and hearing-impaired people.

To understand how temporal integration for loudness might be related to the slopes of loudness-growth functions, consider the following example. Assume that the duration and level of a reference sound are fixed, and the duration of a test sound is decreased, keeping the level fixed, until the loudness is half of the loudness of the reference sound. Assume also that the duration at which the loudness is halved is similar for all values of the reference level. To measure the amount of temporal integration, the level of the shorter sound has to be increased by an amount sufficient to double its loudness. If the function relating loudness to level is relatively steep, as it is at very low and very high sound levels, then only a small increase in sound level, say 6 dB, will be needed to double the loudness. If the function relating loudness to level is relatively shallow, as it is at medium sound levels, then a larger increase in sound level, say 12 dB, will be needed to double the loudness. This can explain why temporal integration is greater at medium levels than at very low or very high levels.

As described in Chapter 4, the functions relating loudness to sound level are typically steeper for people with cochlear hearing loss than for normally hearing people. According to the arguments given above, this should lead to reduced temporal integration for loudness. One study of the effect of duration on loudness for hearing-impaired people (Pedersen and Poulsen, 1973) did not show such an effect. However, Pedersen and Poulsen only tested people with hearing impairment caused by *presbyacusis*, and these people had mild hearing losses (30–40 dB) at the test frequency of 1000 Hz.

In a more recent study, Buus, Florentine and Poulsen (1999) obtained loudness matches between 5- and 200-ms tones as a function of level for fifteen listeners with cochlear hearing loss and for seven age-matched listeners with normal hearing. For the normal listeners, the amount of temporal integration, defined as the level difference between equally loud short and long tones, varied non-monotonically with level and was largest at moderate levels, as found previously (Florentine, Buus and Poulsen, 1996; Buus, Florentine and Poulsen, 1997). The maximum amount of temporal integration varied across subjects from about 15 dB to about 30 dB. The results varied widely across hearing-impaired listeners, but for most the amount of temporal integration increased as the level of the 200-ms tone was increased up to about 100 dB SPL, and the maximal amount of temporal integration was usually less than 20 dB. The impaired listeners' amounts of temporal integration at high SPLs were often larger than normal, although temporal integration was reduced near threshold. When evaluated at equal SLs, the amount of temporal integration well above threshold usually was at the low end of the normal range, except for listeners whose audiometric thresholds were near-normal at low frequencies and increased rapidly at high frequencies (who may have had high-frequency *dead regions*); these listeners showed larger-than-normal amounts of temporal integration for frequencies just inside the region of near-normal hearing.

Buus, Florentine and Poulsen (1999) argued that their results were consistent with what would be expected from the slopes of loudness-growth functions and from a hypothesis called the equal-loudness-ratio hypothesis. This states that the loudness ratio between long and brief tones of equal SPL is the same at all SPLs. They suggested that the larger-than-normal amounts of temporal integration found for subjects with steeply sloping hearing loss could be explained by the fact that the loudness of a tone with frequency just inside the region of near-normal hearing grows less rapidly with increasing level than for a normal ear (Hellman, 1994).

VII PERCEPTUAL CONSEQUENCES OF ABNORMAL TEMPORAL PROCESSING IN PEOPLE WITH COCHLEAR HEARING LOSS

VII.1 CONSEQUENCES OF ABNORMAL TEMPORAL RESOLUTION

It has been argued above that the sliding temporal integrator in the model of temporal resolution is probably normal in most people with cochlear hearing loss. However, the nonlinearity preceding the integrator is less compressive in impaired ears than in normal ears. For stimuli with inherent slow amplitude fluctuations (such as narrow bands of noise), this can lead to poorer temporal resolution, since the inherent fluctuations become more confusable with the temporal feature to be detected. However, for deterministic stimuli (such as sinusoids) or for broadband noise stimuli, performance is similar for normal and impaired ears, when the comparison is made at equal SLs. Unfortunately, most sounds in everyday life are characterized by unpredictable fluctuations in amplitude from moment to moment. For such sounds, people with cochlear damage will have more difficulty than normal in following the temporal structure of the sounds. In addition, temporal resolution may be poor because the sounds are at low SLs and/or because the audible bandwidth of the stimuli is restricted. All of these factors can lead to problems in understanding speech and in discriminating and identifying music and environmental sounds. For example, it may be difficult for a person with cochlear hearing loss to detect a weak consonant sound following soon after a relatively intense vowel sound.

VII.2 CONSEQUENCES OF REDUCED TEMPORAL INTEGRATION

One consequence of reduced temporal integration is that the hearing loss, as measured by the change in absolute threshold relative to 'normal' values, varies according to the duration of the test sounds; the hearing loss is not as great for short sounds as it is for long sounds. Consider, as an example, two sounds with durations of 400 and 10 ms. For a normally hearing person, the level required for detection of these two sounds might be, for example, 4 and 20 dB SPL, respectively; the shorter sound has to be about 16 dB higher in level to reach the absolute threshold. For a person with moderate cochlear hearing loss, the threshold for detecting the longer sound might be 54 dB SPL, that is 50 dB higher than normal. However, the threshold for detecting the shorter sound might be 60 dB SPL, which is only 40 dB higher than normal. Thus, the 'loss' relative to normal hearing is 10 dB less for the shorter sound than for the longer sound.

It is not obvious what is the 'correct' duration at which to measure absolute thresholds. Clinically, audiometric thresholds are usually measured using tones lasting several hundred milliseconds, and many prescriptive formulae for fitting hearing aids (see Chapter 9) are based on such measurements. However, it would seem just as valid to measure thresholds for brief sounds and to specify hearing loss in that way. Many hearing-impaired people have difficulty in detecting weak consonants, and those consonants are often of short duration, for example p, t or k.

As described above, temporal integration for loudness can differ for normally hearing and for hearing-impaired people, and the amount of temporal integration can vary markedly from one impaired listener to another. This can complicate the design of hearing aids that attempt to restore loudness to 'normal' (Kollmeier and Hohmann, 1995; Launer and Moore, 2003); see Chapter 9 for further discussion of such hearing aids. However, Buus, Florentine and Poulsen (1999) argued that, for most hearing-impaired listeners, the amount of *gain* necessary to restore normal loudness is independent of the duration of the input signal.

6 Pitch Perception and Frequency Discrimination

I INTRODUCTION

Pitch is a subjective attribute of sound defined in terms of what is *heard*. It is related to the physical repetition rate of the waveform of a sound; for a pure tone (a sinusoid), this corresponds to the frequency, and for a periodic complex tone to the fundamental frequency (F0). Increasing the repetition rate gives a sensation of increasing pitch. Pitch is defined formally as 'that attribute of auditory sensation in terms of which sounds can be ordered on a scale extending from low to high' (ANSI, 1994). Variations in pitch give rise to a sense of melody, are associated with the intonation of voices and provide cues as to whether an utterance is a question or a statement. Since pitch is a subjective attribute, it cannot be measured directly. Often, the pitch of a sound is assessed by adjusting the frequency of a sinusoid until the pitch of the sinusoid matches the pitch of the sound in question. The frequency of the sinusoid then gives a measure of the pitch of the sound. Sometimes, a periodic complex sound, such as a pulse train, is used as a matching stimulus. In this case, the repetition rate of the pulse train gives a measure of pitch.

The ability to detect changes in frequency over time is called frequency discrimination. Usually, the changes in frequency are heard as changes in pitch. It is important to distinguish between frequency selectivity and frequency discrimination. The former refers to the ability to resolve the frequency components of a complex sound, as described in Chapter 3, Section VII. If a complex tone with many harmonics is presented, a given harmonic can only be 'heard out' from the complex tone if it is separated from neighbouring harmonics by about 1.25 $\mathrm{ERB_N}$ (Plomp, 1964a; Moore and Ohgushi, 1993; Moore, 2003); see Chapter 3, Sections IV.1 and VII for a description of the $\mathrm{ERB_N}$-number scale and for a description of the ability to hear out harmonics. For example, for a complex tone with an F0 of 150 Hz, the sixth harmonic (900 Hz) is separated from the neighbouring harmonics (750 and 1050 Hz) by about 1.25 $\mathrm{ERB_N}$, and it would just be possible to 'hear it out' as a separate tone. Frequency discrimination often involves much smaller frequency differences. For example, a trained listener can just discriminate a 1000-Hz sinusoid from a 1003-Hz sinusoid when the two sinusoids are presented successively with a brief silent interval between them.

II THEORIES OF PITCH PERCEPTION

There are two traditional theories of pitch perception. One, the 'place' theory, is based on the fact that different frequencies (or frequency components in a complex sound) excite different places along the basilar membrane (BM), and hence excite neurones with different characteristic frequencies (CFs). The place theory assumes that the pitch of a sound is related to the excitation pattern produced by that sound; for a pure tone, the pitch is generally assumed to correspond to the position of maximum excitation.

An alternative theory, called the temporal theory, is based on the assumption that the pitch of a sound is related to the time pattern of the neural impulses evoked by that sound. These impulses tend to occur at a particular phase of the waveform on the BM, a phenomenon called phase locking; see Chapter 1, Section V.5. The intervals between successive neural impulses approximate integer multiples of the period of the waveform, and these intervals are assumed to determine the perceived pitch. The temporal theory cannot be applicable at very high frequencies, since phase locking becomes very weak for frequencies above about 5 kHz, although the upper limit varies across species (Palmer and Russell, 1986). However, the tones produced by most musical instruments, the human voice and most everyday sound sources have F0s well below 5 kHz.

Many researchers believe that the perception of pitch involves both place mechanisms and temporal mechanisms. However, one mechanism may be dominant for a specific task or aspect of pitch perception, and the relative role of the two mechanisms almost certainly varies with centre frequency.

The place and temporal theories were originally proposed to account for the perception of the pitch of pure tones. Later, they were extended to account for the perception of complex tones. This chapter follows a similar sequence. It first presents experimental data on the perception of pure tones and describes the theoretical implications of those data. The effect of cochlear hearing loss on the pitch perception of pure tones is then described. Next, the chapter presents data on the perception of complex tones, which are much more common in everyday life. Finally, the effect of cochlear hearing loss on the pitch perception of complex tones is described.

III THE PERCEPTION OF THE PITCH OF PURE TONES BY NORMALLY HEARING PEOPLE

III.1 THE FREQUENCY DISCRIMINATION OF PURE TONES

The smallest detectable change in frequency is called the frequency *difference limen*. There have been two common ways of measuring frequency discrimination. One measure involves the discrimination of two successive steady tones with slightly different frequencies. On each trial, the tones are presented in a random order and the listener is required to indicate whether the first or second tone is higher in frequency.

The frequency difference between the two tones is adjusted until the listener achieves a criterion percentage correct, for example 75 %. This measure will be called the DLF (*difference limen for frequency*). A second measure uses tones which are frequency modulated. In such tones, the frequency moves up and down in a regular periodic manner about the mean (carrier) frequency. The number of times per second that the frequency goes up and down is called the modulation rate. Typically, the modulation rate is rather low (between 2 and 20 Hz), and the changes in frequency are heard as fluctuations in pitch – a kind of 'warble'. To determine a threshold for detecting frequency modulation, two tones are presented successively; one is modulated in frequency and the other has a steady frequency. The order of the tones in each trial is random. The listener is required to indicate whether the first or the second tone is modulated. The amount of modulation (also called the modulation depth) required to achieve a criterion response (e.g. 75 % correct) is determined. This measure will be called the FMDL (frequency modulation detection limen).

An example of results obtained with the two methods is given in Figure 6.1 (data from Sek and Moore, 1995). Expressed in Hz, both DLFs and FMDLs are smallest at low frequencies, and increase monotonically with increasing frequency. Expressed as a proportion of centre frequency, as in Figure 6.1, DLFs are smallest for middle frequencies, and are larger for very high and very low frequencies. FMDLs vary less

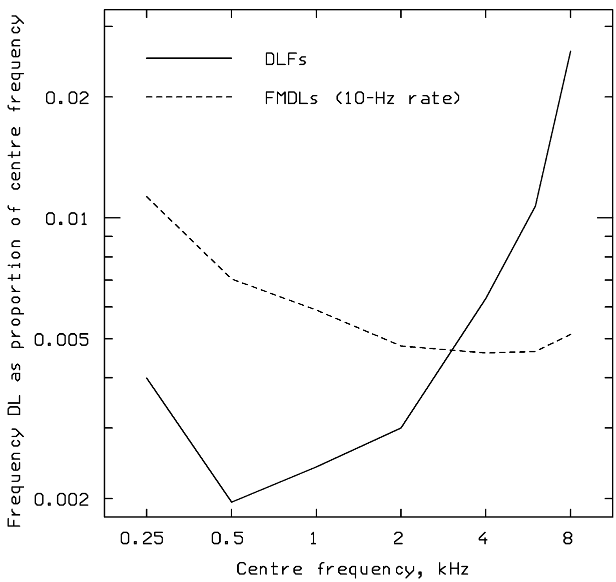


Figure 6.1. Thresholds for detecting differences in frequency between steady pulsed tones (DLFs) and for detecting frequency modulation (FMDLs), plotted as a percentage of the centre frequency and plotted against centre frequency. The modulation rate for the FMDLs was 10 Hz. The data are taken from Sek and Moore (1995).

with frequency than DLFs. Both DLFs and FMDLs tend to get somewhat smaller as the sound level increases (Wier, Jesteadt and Green, 1977; Nelson, Stanton and Freyman, 1983).

Place models of frequency discrimination (Henning, 1967; Siebert, 1970; Zwicker, 1970) predict that frequency discrimination should be related to frequency selectivity; both should depend on the sharpness of tuning on the BM. Zwicker (1970) has attempted to account for frequency discrimination in terms of changes in the excitation pattern evoked by the sound when the frequency is altered. Zwicker inferred the shapes of the excitation patterns from masking patterns such as those shown in Figure 3.9 (see Chapter 3 for details). In his original formulation of the model, Zwicker intended it to apply only to FMDLs; others (e.g. Freyman and Nelson, 1986) have tried to apply the model to account for DLFs.

The model is illustrated in Figure 6.2. The figure shows two excitation patterns, corresponding to two tones with slightly different frequencies. A change in frequency results in a sideways shift of the excitation pattern. The change is assumed to be detectable whenever the excitation level at some point on the excitation pattern changes by more than a certain threshold value. Zwicker suggested that this value was

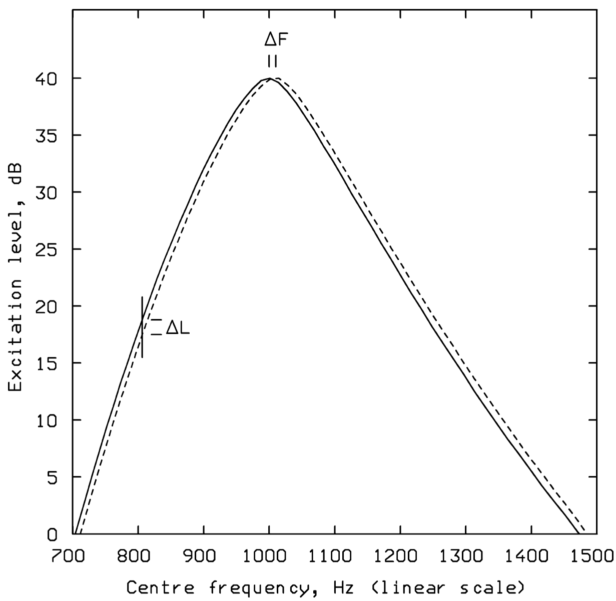


Figure 6.2. Schematic illustration of an excitation-pattern model for frequency discrimination. Excitation patterns are shown for two sinusoidal tones differing slightly in frequency; the two tones have frequencies of 1000 and 1010 Hz. It is assumed that the difference in frequency, ΔF, can be detected if the excitation level changes anywhere by more than a criterion amount. The biggest change in excitation level is on the low-frequency side. The change is indicated by ΔL.

about 1 dB. The change in excitation level is greatest on the steeply sloping low-frequency side of the excitation pattern. Thus, in this model, the detection of a change in frequency is functionally equivalent to the detection of a change in level on the low-frequency side of the excitation pattern. The steepness of the low-frequency side is roughly constant when the frequency scale is expressed in units of ERB_N (see Chapter 3, Section IV.1), rather than in terms of linear frequency. The slope is about 18 dB/ERB_N. To achieve a change in excitation level of 1 dB, the frequency has to be changed by ERB_N/18. Thus, Zwicker's model predicts that the frequency DL at any given frequency should be about one-eighteenth (= 0.056) of the value of ERB_N at that frequency. FMDLs do conform fairly well to this prediction of the model, especially when the modulation rate is fairly high (10 Hz or above), as illustrated in Figure 6.3. The dashed line in this figure shows the ratio FMDL/ERB_N, plotted as a function of centre frequency. The modulation rate was 10 Hz. The ratio is roughly constant, and its value is about 0.05, close to the value predicted by the model. However, DLFs vary more with frequency than predicted by the model (Moore, 1974; Moore and Glasberg, 1986d, 1989; Sek and Moore, 1995). This is illustrated by the solid line in Figure 6.3, which shows the ratio DLF/ERB_N. The ratio varies markedly with centre frequency. The DLFs for frequencies of 2 kHz and below are smaller than predicted by Zwicker's model, while those for frequencies of 6 and 8 kHz are larger than predicted.

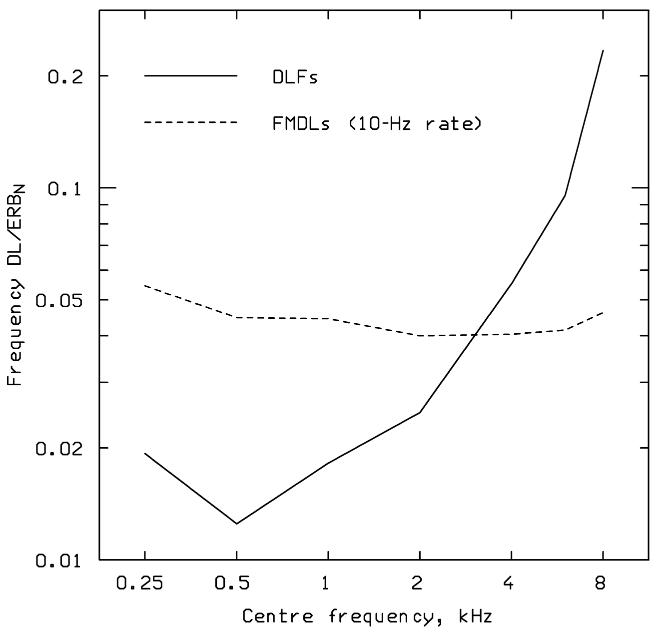


Figure 6.3. DLFs and FMDLs expressed relative to the value of ERB_N at each centre frequency, and plotted as a function of centre frequency. The modulation rate for the FMDLs was 10 Hz. The data are taken from Sek and Moore (1995).

The results for the FMDLs are consistent with the place model, but the results for the DLFs are not. The small DLFs at low frequencies probably reflect the use of temporal information from phase locking. Phase locking becomes less precise at frequencies above 1 kHz, and it is largely lost above 5 kHz. This can account for the marked increase in the DLFs at high frequencies (Goldstein and Srulovicz, 1977).

The ratio FMDL/ERB_N is not constant across centre frequency when the modulation rate is very low (around 2 Hz), but increases with increasing centre frequency (Moore and Sek, 1995, 1996; Sek and Moore, 1995). For low centre frequencies, FMDLs are smaller for a 2-Hz modulation rate than for a 10-Hz rate, while for high centre frequencies (above 4 kHz) the reverse is true. For very low modulation rates, frequency modulation may be detected by virtue of the changes in phase locking to the carrier that occur over time. In other words, the frequency is determined over short intervals of time, using phase-locking information, and changes in the estimated frequency over time indicate the presence of frequency modulation. Moore and Sek (1996) suggested that the mechanism for decoding the phase-locking information was 'sluggish'; it had to sample the sound for a certain time in order to estimate its frequency. Hence, it could not follow rapid changes in frequency and played little role for high modulation rates. Consistent with this, Gockel, Moore and Carlyon (2001) showed that, for sinusoidal carriers that were frequency modulated with an asymmetric function, the overall pitch was determined more by parts of the sound where the frequency changed slowly than by parts where the frequency changed rapidly.

In summary, measures of frequency discrimination are consistent with the idea that DLFs, and FMDLs for very low modulation rates, are determined by temporal information (phase locking) for frequencies up to about 4–5 kHz. The precision of phase locking decreases with increasing frequency above 1–2 kHz, and it is largely absent above about 5 kHz. This can explain why DLFs increase markedly at high frequencies. FMDLs for medium to high modulation rates may be determined by a place mechanism, that is by the detection of changes in the excitation pattern. This mechanism may also account for DLFs and for FMDLs for low modulation rates, when the centre frequency is above about 5 kHz.

III.2 THE PERCEPTION OF MUSICAL INTERVALS

Two tones which are separated in frequency by an interval of one *octave* (i.e. one has twice the frequency of the other) sound similar. They are judged to have the same name on the musical scale (e.g. C3 and C4). This has led several theorists to suggest that there are at least two dimensions to musical pitch. One aspect is related monotonically to frequency (for a pure tone) and is known as tone height. The other is related to pitch class (i.e. the name of the note) and is called tone chroma (Bachem, 1950). For example, two sinusoids with frequencies of 220 and 440 Hz would have the same tone chroma (they would both be called A on the musical scale), but, as they are separated by an octave, they would have different tone heights.

If subjects are presented with a pure tone of a given frequency, f_1, and are asked to adjust the frequency, f_2, of a second tone so that it appears to be an octave higher in pitch, they generally adjust f_2 to be roughly twice f_1. However, when f_1 lies above 2.5 kHz, so that f_2 would lie above 5 kHz, octave matches become very erratic (Ward, 1954). It appears that the musical interval of an octave is only clearly perceived when both tones are below 5 kHz.

Other aspects of the perception of pitch also change above 5 kHz. A sequence of pure tones with frequencies above 5 kHz does not produce a clear sense of melody (Attneave and Olson, 1971). It is possible to hear that the pitch changes when the frequency is changed, but the musical intervals are not heard clearly. Also, subjects with absolute pitch (the ability to assign names to notes without reference to other notes) are very poor at naming notes with frequencies above 4–5 kHz (Ohgushi and Hatoh, 1991).

These results are consistent with the idea that the pitch of pure tones is determined by different mechanisms above and below 5 kHz, specifically, by a temporal mechanism at low frequencies and a place mechanism at high frequencies. It appears that the perceptual dimension of tone height persists over the whole audible frequency range, but tone chroma only occurs in the frequency range below 5 kHz. Musical intervals are only clearly perceived when the frequencies of the tones lie in the range where temporal information is available.

III.3 THE EFFECT OF LEVEL ON PITCH

The pitch of a pure tone is primarily determined by its frequency. However, sound level also plays a small role. On average, the pitch of tones below about 2 kHz decreases with increasing level, while the pitch of tones above about 4 kHz increases with increasing sound level. The early data of Stevens (1935) showed rather large effects of sound level on pitch, but more recent data generally show much smaller effects (Verschuure and van Meeteren, 1975). For tones between 1 and 2 kHz, changes in pitch with level are generally less than 1 %. For tones of lower and higher frequencies, the changes can be larger (up to 5 %). There are also considerable individual differences both in the size of the pitch shifts with level and in the direction of the shifts (Terhardt, 1974a).

It has sometimes been argued that pitch shifts with level are inconsistent with the temporal theory of pitch; neural interspike intervals are hardly affected by changes in sound level over a wide range. However, changes in pitch with level could be explained by the place theory, if shifts in level were accompanied by shifts in the position of maximum excitation on the BM. On closer examination, these arguments turn out to be rather weak. Although the temporal theory assumes that pitch depends on the temporal pattern of nerve spikes, it also assumes that the temporal information has to be 'decoded' at some level in the auditory system. In other words, the time intervals between neural spikes have to be measured. It is quite possible that the mechanism which does this is affected by which neurones are active and by the spike rates in those neurones; these in turn depend on sound level.

The argument favouring the place mechanism is also weak. Chapter 1 described how the peak in the pattern of excitation evoked by a high-frequency pure tone shifts towards the base of the cochlea with increasing sound level. The base is tuned to higher frequencies; so the basalward shift should correspond to hearing an increase in pitch. At high sound levels, the basalward shift corresponds to a shift in frequency of one-half octave or more. Thus, the place theory predicts that the pitch of high-frequency pure tones should increase with increasing sound level, and the shift should correspond to half an octave or more at high sound levels. Although the observed shift is in the predicted direction, it is always much less than half an octave. If pitch is determined by a place mechanism, then the auditory system must have some way of compensating for changes in excitation patterns with level.

At present, there is no generally accepted explanation for the shifts in pitch with level. Given this, it seems that the existence of these pitch shifts cannot be used to draw any strong conclusions about theories of pitch. In any case, as already mentioned, the pitch shifts are rather small. In some people with cochlear hearing impairment, shifts in pitch with level can be much larger. Examples are described below.

IV FREQUENCY DISCRIMINATION OF PURE TONES BY PEOPLE WITH COCHLEAR HEARING LOSS

People with cochlear hearing loss usually have auditory filters that are broader than normal; see Chapter 3. Hence, the excitation pattern evoked by a sinusoid is also broader than normal. According to the place theory, this should lead to impaired frequency discrimination. According to the temporal theory, there should not necessarily be a relationship between frequency selectivity and frequency discrimination. However, frequency discrimination could be adversely affected by the reduced precision of phase locking that can occur in cases of cochlear damage; see Chapter 1, Section VII.5. Also, it may be the case that cochlear hearing loss somehow affects the ability to 'decode' the phase-locking information.

IV.1 DIFFERENCE LIMENS FOR FREQUENCY (DLFs)

DLFs for people with cochlear damage have been measured in several studies (Gengel, 1973; Tyler, Wood and Fernandes, 1983; Hall and Wood, 1984; Freyman and Nelson, 1986, 1987, 1991; Moore and Glasberg, 1986c; Moore and Peters, 1992; Simon and Yund, 1993). The results have generally shown that frequency discrimination is adversely affected by cochlear hearing loss. However, there is considerable variability across individuals, and the size of the DLF has not been found to be strongly correlated with the absolute threshold at the test frequency. Simon and Yund (1993) measured

DLFs separately for each ear of subjects with bilateral cochlear damage and found that DLFs could be markedly different for the two ears at frequencies where absolute thresholds were the same. They also found that DLFs could be the same for the two ears when absolute thresholds were different.

Tyler, Wood and Fernandes (1983) compared DLFs and frequency selectivity measured using psychophysical tuning curves (PTCs) (see Chapter 3, Section 3.1). They found a low correlation between the two. They concluded that frequency discrimination was not closely related to frequency selectivity, suggesting that place models were not adequate to explain the data. Moore and Peters (1992) measured DLFs for four groups of subjects: young normally hearing, young hearing impaired, elderly with near-normal hearing and elderly hearing impaired. The auditory filter shapes of the subjects had been estimated in earlier experiments using the notched-noise method (see Chapter 3, Section 3.2), for centre frequencies of 100, 200, 400 and 800 Hz. The DLFs for both impaired groups were higher than for the young normal group at all centre frequencies (50–4000 Hz). The DLFs for the elderly group with near-normal hearing were intermediate. The DLFs at a given centre frequency were generally only weakly correlated with the sharpness of the auditory filter at that centre frequency, and some subjects with broad filters at low frequencies had near-normal DLFs at low frequencies. These results suggest a partial dissociation of frequency selectivity and frequency discrimination of pure tones.

Overall, the results of these experiments do not provide strong support for place models of frequency discrimination. This is consistent with the conclusions reached earlier in this chapter, that DLFs for normally hearing people are determined mainly by temporal mechanisms for frequencies of up to about 5 kHz. An alternative way of accounting for the fact that cochlear hearing loss results in larger-than-normal DLFs is in terms of loss of neural synchrony (phase locking) in the auditory nerve; see Chapter 1, Section VII.5. Goldstein and Srulovicz (1977) described a model for frequency discrimination based on the use of information from the interspike intervals in the auditory nerve. This model was able to account for the way that DLFs depend on frequency and duration for normally hearing subjects. Wakefield and Nelson (1985) showed that a simple extension to this model, taking into account the fact that phase locking gets slightly more precise as sound level increases, allowed the model to predict the effects of level on DLFs. They also applied the model to DLFs measured as a function of level in subjects with high-frequency hearing loss, presumably resulting from cochlear damage. They were able to predict the results of the hearing-impaired subjects by assuming that neural synchrony was reduced in neurones with CFs corresponding to the region of hearing loss. Of course, this does not prove that loss of synchrony is the cause of the larger DLFs, but it does demonstrate that loss of synchrony is a plausible candidate.

Another possibility is that the central mechanisms involved in the analysis of phase-locking information make use of differences in the preferred time of firing of neurones with different CFs; these time differences arise from the propagation time of the travelling wave on the BM (Loeb, White and Merzenich,1983; Shamma, 1985).

The propagation time along the BM can be affected by cochlear damage (Ruggero, 1994; Ruggero *et al.*, 1996), and this could disrupt the processing of the temporal information by central mechanisms.

IV.2 FREQUENCY MODULATION DETECTION LIMENS (FMDLs)

Zurek and Formby (1981) measured FMDLs for ten subjects with sensorineural hearing loss (assumed to be mainly of cochlear origin) using a 3-Hz modulation rate and frequencies between 125 and 4000 Hz. Subjects were tested at a sensation level (SL) of 25 dB, a level above which performance was found (in pilot studies) to be roughly independent of level. The FMDLs tended to increase with increasing hearing loss at a given frequency. The worsening of performance with increasing hearing loss was greater at low frequencies than at high frequencies.

Zurek and Formby suggested two possible explanations for the greater effect at low frequencies. The first is based on the assumption that two mechanisms are involved in coding frequency, a temporal mechanism at low frequencies and a place mechanism at high frequencies. The temporal mechanism may be more disrupted by hearing loss than the place mechanism. An alternative possibility is that absolute thresholds at low frequencies do not provide an accurate indicator of the extent of cochlear damage, since these thresholds may be mediated by neurones with CFs above the test frequency. For example, if the inner hair cells (IHCs) at a region of the BM tuned to low frequencies are damaged, then low frequencies may be detected via the 'tails' of the tuning curves of neurones with higher CFs. In extreme cases, there may be a dead region at low frequencies.

Moore and Glasberg (1986c) measured both FMDLs and thresholds for detecting amplitude modulation (AM) (see Chapter 4, Section II.3), using a 4-Hz modulation rate. They used subjects with moderate unilateral and bilateral cochlear impairments. Stimuli were presented at a fixed level of 80 dB SPL, which was at least 10 dB above the absolute threshold. The FMDLs were larger for the impaired than for the normal ears, by an average factor of 3.8 for a frequency of 500 Hz and 1.5 for a frequency of 2000 Hz, although the average hearing loss was similar for these two frequencies. The greater effect at low frequencies is consistent with the results of Zurek and Formby, described above. The amplitude-modulation detection thresholds were not very different for the normal and impaired ears. These thresholds provide an estimate of the smallest detectable change in excitation level. Moore and Glasberg also used the notched-noise method (see Chapter 3, Section 3.2) to estimate the slopes of the auditory filters at each test frequency. The slopes, together with the amplitude-modulation detection thresholds, were used to predict the FMDLs on the basis of Zwicker's excitation-pattern model. The obtained FMDLs were reasonably close to the predicted values. In other words, the results were consistent with the excitation-pattern model.

Grant (1987) measured FMDLs for three normally hearing subjects and three subjects with profound hearing losses. The sinusoidal carrier was modulated in frequency by a triangle function three times per second. Stimuli were presented at 30 dB SL for the normal subjects and at a ‘comfortable listening level’ (110–135 dB SPL) for the impaired subjects. For all carrier frequencies (100–1000 Hz), FMDLs were larger, by an average factor of 9.5, for the hearing-impaired than for the normally hearing subjects. Grant also measured FMDLs when the stimuli were simultaneously amplitude modulated by a noise that was lowpass filtered at 3 Hz. The slow random amplitude fluctuations produced by this AM would be expected to impair the use of cues for frequency modulation detection based on changes in excitation level. Consistent with the predictions of the excitation-pattern model, the random AM led to increased FMDLs. Interestingly, the increase was much greater for the hearing-impaired than for the normally hearing subjects. When the random AM was present, thresholds for the hearing-impaired subjects were about 16 times those for the normally hearing subjects.

As described earlier, it is likely that, for low modulation rates, normally hearing subjects can extract information about frequency modulation both from changes in excitation level and from phase locking (Moore and Sek, 1995; Sek and Moore, 1995). The random AM disrupts the use of changes in excitation level, but does not markedly affect the use of phase-locking cues. The profoundly hearing-impaired subjects of Grant appear to have been relying mainly or exclusively on changes in excitation level. Hence, the random AM had severe adverse effects on the FMDLs.

Lacher-Fougère and Demany (1998) measured FMDLs for a 500-Hz carrier, using modulation rates of 2 and 10 Hz. They used five normally hearing subjects and seven subjects with cochlear hearing loss ranging from 30 to 75 dB at 500 Hz. Stimuli were presented at a ‘comfortable’ loudness level. The subjects with losses up to 45 dB had thresholds that were about a factor of two larger than for the normally hearing subjects. The subjects with larger losses had thresholds that were as much as ten times larger than normal. The effect of the hearing loss was similar for the two modulation rates. Lacher-Fougère and Demany suggested that cochlear hearing loss disrupts excitation-pattern (place) cues and phase-locking cues to a roughly equal extent.

Moore and Skrodzka (2002) measured FMDLs for three young subjects with normal hearing and four elderly subjects with cochlear hearing loss. Carrier frequencies were 0.25, 0.5, 1, 2, 4 and 6 kHz, and modulation rates were 2, 5, 10 and 20 Hz. FMDLs were measured both in the absence of AM, and with AM of a fixed depth (a peak-to-valley ratio of 6 dB) added in both intervals of a forced-choice trial. The added AM was intended to disrupt cues based on FM-induced AM in the excitation pattern. The results averaged across subjects are shown in Figure 6.4. Generally, the hearing-impaired subjects (filled symbols) performed markedly more poorly than the normally hearing subjects (open symbols). For the normally hearing subjects, the disruptive effect of the AM (triangles versus circles) tended to increase with increasing modulation rate, for carrier frequencies below 6 kHz, as found previously by Moore and Sek (1996). For the hearing-impaired subjects, the disruptive effective of the AM was generally larger than for the normally hearing subjects, and the magnitude of the disruption did not consistently increase with increasing modulation rate. For the 2-Hz modulation rate, the FMDL for the hearing-impaired subjects, averaged across the

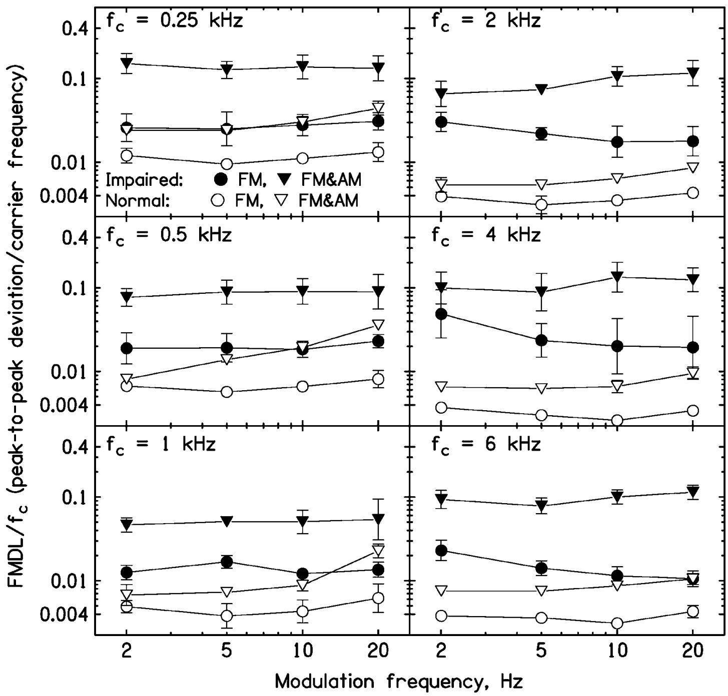


Figure 6.4. Data from Moore and Skrodzka (2002), showing FMDLs plotted as a function of modulation frequency. Each panel shows results for one carrier frequency. Mean results are shown for normally hearing subjects (open symbols) and hearing-impaired subjects (filled symbols). FMDLs are shown without added AM (circles) and with added AM (triangles). Error bars indicate ± one standard deviation across subjects. They are omitted when they would span a range less than 0.1 log units (corresponding to a ratio of 1.26).

four lowest carrier frequencies, was a factor of 2.5 larger when AM was present than when it was absent. In contrast, the corresponding ratio for the normally hearing subjects was only 1.45. It was argued earlier that the relatively small disruptive effect of AM at low modulation rates for normally hearing subjects reflects the use of temporal information (Moore and Sek, 1996). The larger effect found for the hearing-impaired subjects suggests that they were not using temporal information effectively. Rather, the FMDLs were probably based largely on excitation-pattern cues (FM-induced AM in the excitation pattern), and these cues were strongly disrupted by the added AM. Overall, the results suggest that cochlear hearing impairment adversely affects both temporal and excitation pattern mechanisms of FM detection.

In conclusion, FMDLs for hearing-impaired people are generally larger than normal. The larger thresholds may reflect both the broadening of the excitation pattern (reduced frequency selectivity) and a reduced ability to use cues based on phase locking.

V THE PERCEPTION OF PURE-TONE PITCH FOR FREQUENCIES FALLING IN A DEAD REGION

In some people with cochlear damage restricted to low-frequency regions of the cochlea, it appears that there are no functioning IHCs with CFs corresponding to the frequency region of the loss. In other words, there may be a low-frequency dead region. In such cases, the detection of low-frequency tones is mediated by neurones with high CFs. One way of demonstrating this is by the measurement of PTCs; see Chapter 3, Figures 3.18–3.20. If the signal to be detected has a low frequency, falling within the dead region, the tip of the tuning curve lies well above the signal frequency. In other words, a masker centred well above the signal in frequency is more effective than a masker centred close to the signal frequency (Thornton and Abbas, 1980; Florentine and Houtsma, 1983; Turner, Burns and Nelson, 1983). For example, Florentine and Houtsma (1983) studied a subject with a moderate to severe unilateral low-frequency loss of cochlear origin. For a 1-kHz signal, the tip of the PTC fell between 2.2 and 2.85 kHz, depending on the exact level of the signal.

The perception of pitch in such subjects is of considerable theoretical interest. When there is a low-frequency dead region, a low-frequency pure tone cannot produce maximum neural excitation at the CF corresponding to its frequency, since there are no neural responses at that CF. The peak in the neural excitation pattern must occur at CFs higher than the frequency of the signal. If the place theory is correct, this should lead to marked upward shifts in the pitch of the tone. In fact, this does not usually happen.

Florentine and Houtsma (1983) obtained pitch matches between the two ears of their unilaterally impaired subject. They presented the stimuli at levels just above absolute threshold, to minimize the spread of excitation along the BM. Pitch shifts between the two ears were small. Turner, Burns and Nelson (1983) studied six subjects with low-frequency cochlear losses. Three of their subjects showed PTCs with tips close to the signal frequency; they presumably had functioning IHCs with CFs close to the signal frequency. The other three subjects showed PTCs with tips well above the signal frequency; they presumably had low-frequency dead regions. Pitch perception was studied either by pitch matching between the two ears (for subjects with unilateral losses) or by octave matching (for subjects with bilateral losses, but with some musical ability). The subjects whose PTCs had tips above the signal frequency gave results similar to those of the subjects whose PTCs had tips close to the signal frequency; no distinct pitch anomalies were observed.

Huss and Moore (2005b) studied pitch perception in subjects with a variety of configurations of dead regions, diagnosed using the *threshold-equalizing noise* (TEN) test (Chapter 3, Section IX.3) and PTCs. Two tasks were used, a pitch-matching task and an octave-matching task. For the pitch-matching task, subjects were asked to adjust the frequency of a pure tone so that its pitch matched that of another pure tone with fixed frequency. The two tones were presented alternately. Matches were made across ears, to obtain a measure of *diplacusis*, and within one ear, to estimate the reliability of matching. For the octave-matching task, subjects were asked to adjust a tone of variable frequency so that it sounded one octave higher or lower than a fixed reference tone. Only a few subjects were able to perform this task reliably. The level for each frequency was chosen using a loudness model (Moore and Glasberg, 1997) so as to give a fixed calculated loudness.

Pitch matches within one ear were often erratic for tones falling more than half an octave into a low-frequency or high-frequency dead region, indicating that such tones usually do not evoke a clear pitch sensation. This is consistent with the large DLFs that are found for tones whose frequencies fall in a dead region (McDermott *et al.*, 1998; Thai-Van *et al.*, 2003; Kluk and Moore, 2006). Pitch matches across the ears of subjects with asymmetric hearing loss and octave matches within ears indicated that tones falling within a dead region are often perceived with a pitch different (usually higher) than ‘normal’. This was true for tones falling in both low-frequency and high-frequency dead regions. For tones with frequencies below 5 kHz, the pitches were usually not consistent with what would be expected from the use of either place information or temporal information alone. However, for tones with frequencies above 5 kHz, the pitches were roughly consistent with what would be expected from the place where the tones were detected.

One subject with highly asymmetric hearing loss made matches of a 0.5-kHz pure tone falling in a dead region in the worse ear with an AM tone in the other (better) ear. The carrier frequency of the AM tone was adjusted to about 3.8 kHz and the modulation frequency was adjusted to about 0.52 kHz. The matched carrier frequency corresponded roughly with the place where the tone in the worse ear was detected, while the matched modulation frequency corresponded roughly with the frequency of the tone in the worse ear, suggesting that some temporal information was preserved.

Taken together, the results of studies of pitch perception using people with dead regions indicate the following:

1. Pitch matches (of a tone with itself, within one ear) are often erratic, and frequency discrimination is poor, for tones with frequencies falling in a dead region. This indicates that such tones do not evoke a clear pitch sensation.
2. Pitch matches across the ears of subjects with asymmetric hearing loss, and octave matches within ears, indicate that tones falling within a dead region sometimes are perceived with a near-‘normal’ pitch and sometimes are perceived with a pitch distinctly different from ‘normal’.
3. The shifted pitches found for some subjects indicate that the pitch of low-frequency tones is not represented solely by a temporal code. Possibly, there needs to be a correspondence between place and temporal information for a ‘normal’ pitch to be perceived (Evans, 1978; Loeb, White and Merzenich,1983; Srulovicz and Goldstein, 1983). Alternatively, as noted earlier, temporal information may be ‘decoded’ by a network of coincidence detectors whose operation depends on the phase response at different points along the BM (Loeb, White and Merzenich,1983; Shamma and Klein, 2000). An alteration of this phase response by cochlear hearing loss (Ruggero *et al.*, 1996) may prevent the effective use of temporal information.

Finally, there have been several studies showing that DLFs may show a local 'enhancement' (be smaller than for adjacent frequencies) for tone frequencies falling just outside a dead region (McDermott *et al.*, 1998; Thai-Van *et al.*, 2003; Kluk and Moore, 2006). It has been suggested that this enhancement reflects the effects of cortical plasticity, whereby neurones that were previously tuned to frequencies within a dead region all become tuned to frequencies just below the edge frequency of the dead region (McDermott *et al.*, 1998; Irvine and Wright, 2005).

VI PITCH ANOMALIES IN THE PERCEPTION OF PURE TONES

Cochlear hearing loss at low or high frequencies can lead to changes in perceived pitch even when no dead region is present. For people with unilateral cochlear hearing loss, or asymmetrical hearing losses, the same tone presented alternately to the two ears may be perceived as having different pitches in the two ears. This effect is given the name diplacusis. Sometimes, different pitches are perceived even when the hearing loss is the same in the two ears. The magnitude of the shift can be measured by getting the subject to adjust the frequency of the tone in one ear until its pitch matches that of the tone in the other ear.

According to the place theory, cochlear damage might result in pitch shifts for two reasons. The first applies when the amount of hearing loss varies with frequency and especially when the amount of IHC damage varies with CF. When the IHCs are damaged, transduction efficiency is reduced, and so a given amount of BM vibration leads to less neural activity than when the IHCs are intact. When IHC damage varies with CF, the peak in the *neural* excitation pattern evoked by a tone will shift away from a region of greater IHC loss. Hence, the perceived pitch is predicted to shift away from that region. Early studies of diplacusis (de Mare, 1948; Webster and Schubert, 1954) were generally consistent with this prediction, showing that when a sinusoidal tone is presented in a frequency region of hearing loss the pitch shifts towards a frequency region where there is less hearing loss. For example, in a person with a high-frequency hearing loss, the pitch was reported to be shifted downwards. However, there are clearly cases where the pitch does not shift as predicted.

An alternative way in which pitch shifts might occur is by shifts in the position of the peak excitation on the BM; such shifts can occur even for a flat hearing loss. Chapter 1 (Sections VII.2 and VII.3) described work showing that the tips of tuning curves on the BM and of neural tuning curves were often shifted towards lower frequencies in cases of moderate to severe cochlear damage. This means that the maximum excitation at a given place is produced by a *lower* frequency than normal. Hence, for a given frequency, the peak of the BM response in an impaired cochlea would be shifted towards the base, that is towards places normally responding to higher frequencies. This leads to the prediction that the perceived pitch should be shifted upwards. Several studies have found that this is usually the case. For example, Gaeth and Norris (1965) and Schoeny and Carhart (1971) reported that pitch shifts were generally upwards regardless of the configuration of loss. However, it is also clear that individual differences can be substantial, and subjects with similar patterns of hearing loss (absolute thresholds as a function of frequency) can show quite different pitch shifts.

Burns and Turner (1986) measured changes in pitch as a function of intensity, by obtaining pitch matches between a tone presented at a fixed level (midway, in dB, between the absolute threshold and 100 dB SPL) and a tone of variable level. The tones were presented alternately to the same ear. As described earlier, normally hearing subjects usually show small shifts in pitch with intensity in this type of task; the shifts are rarely greater than about 3 %. The hearing-impaired subjects of Burns and Turner often showed abnormally large pitch-intensity effects, with shifts of up to 10 %. A common pattern was an abnormally large negative pitch shift with increasing level for low-frequency tones.

Burns and Turner (1986) obtained several other measures from their subjects, including PTCs in forward masking, DLFs, measures of diplacusis and octave judgements. There was a tendency for increased DLFs and increased pitch-matching variability in frequency regions where the PTCs were broader than normal. The exaggerated pitch-level effects occurred both in frequency regions where PTCs were broader than normal and (sometimes) in regions where both absolute thresholds and PTCs were normal. The results of the diplacusis measurements and octave matches indicated that the large pitch-intensity effects were mainly a consequence of large increases in pitch at low levels; the pitch returned to more 'normal' values at higher levels.

As pointed out by Burns and Turner, these results are difficult to explain by the place theory. There is no evidence to suggest that peaks in BM responses or in neural excitation patterns of ears with cochlear damage are shifted at low levels but return to 'normal' positions at high levels. Also, even in subjects with similar configurations of hearing loss, the pitch shifts and changes in pitch with level can vary markedly. Furthermore, as pointed out earlier, low-frequency pure tones can evoke low pitches even in people who appear to have no IHCs or neurones tuned to low frequencies.

The results are also problematic for the temporal theory. There is no obvious reason why systematic shifts in pitch should occur as a result of cochlear damage or of changes in level. Unfortunately, there seem to be no physiological data concerning the effects of level on phase locking in ears with cochlear damage. As pointed out earlier, it is possible that the central mechanisms involved in the analysis of phase-locking information make use of the propagation time of the travelling wave on the BM (Loeb, White and Merzenich,1983; Shamma, 1985). This time can be affected by cochlear damage, and this could disrupt the processing of the temporal information by central mechanisms.

In summary, the perceived pitch of pure tones can be affected by cochlear hearing loss, and changes in pitch with level can be markedly greater than normal. Large individual differences occur, even between subjects with similar absolute thresholds. The mechanisms underlying these effects remain unclear.

VII THE PITCH PERCEPTION OF COMPLEX TONES BY NORMALLY HEARING PEOPLE

VII.1 THE PHENOMENON OF THE MISSING FUNDAMENTAL

The pitch of complex tones does not, in general, correspond to the position of maximum excitation on the BM. Consider, as an example, a sound consisting of short impulses (clicks) occurring 200 times per second. This sound contains harmonics with frequencies at integer multiples of 200 Hz (200, 400, 600, 800 ... Hz). The harmonic at 200 Hz is the fundamental component and its frequency is called the fundamental frequency and is denoted F0. The sound has a low pitch, which is very close to the pitch of its fundamental component, and a sharp *timbre* (tone quality); it sounds very 'buzzy'. However, if the sound is filtered so as to remove the fundamental component, the pitch does not alter; the only result is a slight change in timbre. This is called the phenomenon of the missing fundamental (Ohm, 1843; Schouten, 1940). Indeed, all except a small group of mid-frequency harmonics can be eliminated, and the low pitch remains the same, although the timbre becomes markedly different.

Schouten (1940, 1970) called the low pitch associated with a group of high harmonics the *residue pitch*. Several other names have been used to describe residue pitch, including *virtual pitch*, periodicity pitch and low pitch. The term residue pitch will be used here. Schouten pointed out that it is possible to hear the change produced by removing the fundamental component and then reintroducing it. Indeed, when the fundamental component is present, it is possible to 'hear it out' as a separate sound. The pitch of that component is almost the same as the pitch of the whole sound. However, the presence or absence of the fundamental component does not markedly affect the pitch of the whole sound.

The perception of a residue pitch does not require activity at the point on the BM which would respond maximally to the fundamental component. Licklider (1956) showed that the low pitch of the residue could be heard when low-frequency noise was present that would mask any component at F0. Even when the fundamental component of a complex tone is present, the pitch of the tone is usually determined by harmonics other than the fundamental. Thus, the perception of residue pitch should not be regarded as unusual. Rather, residue pitches are normally heard when listening to complex tones.

The phenomenon of the missing fundamental is not consistent with a simple place model of pitch based on the idea that pitch is determined by the position of the peak excitation on the BM. However, more elaborate place models have been proposed, and these are discussed below.

VII.2 DISCRIMINATION OF THE REPETITION RATE OF COMPLEX TONES

When the repetition rate of a complex tone changes, all of the components change in frequency by the same ratio, and a change in residue pitch is heard. The threshold for detecting a change in repetition rate will be denoted the F0DL. The F0DL for a complex tone is usually smaller (better) than the DLF for a sinusoid at F0 (Flanagan and Saslow, 1958), and it can be smaller than the DLF for any of the individual sinusoidal components in the complex tone (Moore, Glasberg and Shailer, 1984). This indicates that information from the different harmonics is combined or integrated in the determination of residue pitch. This can lead to very fine discrimination; F0DLs are about 0.2 % for F0s in the range 100–400 Hz, provided that low harmonics are present (e.g. the third, fourth and fifth). F0DLs are larger (poorer) for complex tones that contain only high harmonics (above about the seventh) (Hoekstra and Ritsma, 1977; Moore, Glasberg and Shailer, 1984; Houtsma and Smurzynski, 1990; Carlyon and Shackleton, 1994; Plack and Carlyon, 1995; Carlyon, 1997; Bernstein and Oxenham, 2003; Moore *et al.*, 2006).

VIII THEORIES OF PITCH PERCEPTION FOR COMPLEX TONES

VIII.1 THE REPRESENTATION OF A COMPLEX TONE IN THE PERIPHERAL AUDITORY SYSTEM

To understand theories of pitch perception for complex tones, it is helpful to consider how complex tones are represented in the peripheral auditory system. A simulation of the response of the BM to a complex tone is illustrated in Figure 6.5. In this example, the complex tone is a periodic pulse train, containing many equal-amplitude harmonics. The lower harmonics are partly resolved on the BM, and give rise to distinct peaks in the pattern of activity along the BM. At a place tuned to the frequency of a low harmonic, the waveform on the BM is approximately a sinusoid at the harmonic frequency. For example, at the place with a CF of 400 Hz the waveform is a 400-Hz sinusoid. In contrast, the higher harmonics are not resolved and do not give rise to distinct peaks on the BM. The waveforms at places on the BM responding to higher harmonics are complex, but they all have a repetition rate equal to F0.

There are two main ways in which the residue pitch of a complex sound might be extracted. First, it might be derived from the frequencies of the lower harmonics that are resolved on the BM. The frequencies of the harmonics might be determined either by place mechanisms (e.g. from the positions of local maxima on the BM) or by temporal mechanisms (from the interspike intervals in neurones with CFs close to the frequencies of individual harmonics). For example, for the complex tone whose analysis is illustrated in Figure 6.5, the second harmonic, with a frequency of 400 Hz, would give rise to a local maximum at the place on the BM tuned to 400 Hz. The interspike intervals in neurones innervating that place would reflect the frequency of that harmonic; the intervals would cluster around integer multiples of 2.5 ms. Both of these forms of information may allow the auditory system to determine that there is a harmonic at 400 Hz.

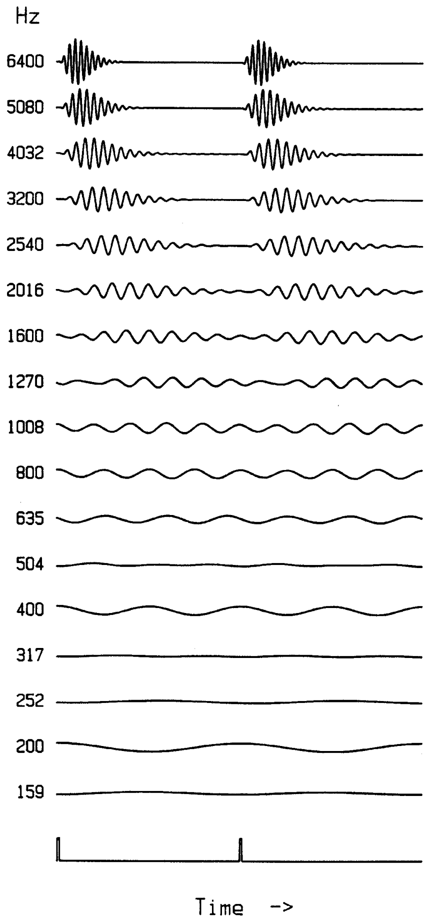


Figure 6.5. A simulation of the responses on the BM to periodic impulses of rate 200 pulses per second. Each number on the left represents the frequency which would maximally excite a given point on the BM. The waveform which would be observed at that point, as a function of time, is plotted opposite that number.

The auditory system may contain a pattern recogniser which determines the residue pitch of the complex sound from the frequencies of the resolved components (Goldstein, 1973; Terhardt, 1974b). In essence, the pattern recogniser tries to find the harmonic series giving the best match to the resolved frequency components; the F0 of this harmonic series determines the perceived pitch. Say, for example, that the first stage establishes frequencies of 800, 1000 and 1200 Hz to be present. The F0 whose harmonics would match these frequencies is 200 Hz. The perceived pitch corresponds to this inferred F0 of 200 Hz. Note that the inferred F0 is always the highest possible value that fits the frequencies determined in stage one. For example, an F0 of 100 Hz would also have harmonics at 800, 1000 and 1200 Hz, but a pitch corresponding to 100 Hz is *not* perceived. It is as if the pattern recogniser assumes that the harmonics are successive harmonics, such as the fourth, fifth and sixth, rather than non-successive harmonics like the eighth, tenth and twelfth.

The pitch of a complex tone may also be extracted from the higher unresolved harmonics. As shown in Figure 6.5, the waveforms at places on the BM responding to higher harmonics are complex, but they all have a repetition rate equal to F0. For the neurones with CFs corresponding to the higher harmonics, nerve impulses tend to be evoked by the biggest peaks in the waveform, that is by the waveform peaks close to envelope maxima. Hence, the nerve impulses are separated by times corresponding to the period of the sound. For example, in Figure 6.5 the input has a repetition rate of 200 periods per second; so the period is 5 ms. The time intervals between nerve spike would cluster around integer multiples of 5 ms, that is 5, 10, 15, 20 … ms. The pitch may be determined from these time intervals. In this example, the time intervals are integer multiples of 5 ms; so the pitch corresponds to 200 Hz.

Experimental evidence suggests that pitch can be extracted *both* from the lower harmonics and from the higher harmonics. Usually, the lower harmonics give a clearer residue pitch, and are more important in determining residue pitch, than the upper harmonics (Plomp, 1967; Ritsma, 1967; Moore, Glasberg and Peters, 1985). Also, F0DLs are smaller for tones containing only low harmonics than for tones containing only high harmonics (Hoekstra and Ritsma, 1977; Moore, Glasberg and Shailer, 1984; Houtsma and Smurzynski, 1990; Carlyon and Shackleton, 1994; Plack and Carlyon, 1995; Carlyon, 1997; Bernstein and Oxenham, 2003; Moore *et al.*, 2006). However, a residue pitch can be heard when only high unresolved harmonics are present (Ritsma, 1962, 1963; Moore, 1973; Houtsma and Smurzynski, 1990; Kaernbach and Bering, 2001). Although this pitch is not as clear as when lower harmonics are present, it is clear enough to allow the recognition of musical intervals and of simple melodies (Moore and Rosen, 1979; Houtsma and Smurzynski, 1990).

VIII.2 SPECTRO-TEMPORAL PITCH THEORIES

Several researchers have proposed theories in which both place (spectral) and temporal mechanisms play a role; these are referred to as spectro-temporal theories. The theories assume that information from both low harmonics and high harmonics contributes to the determination of pitch. The initial place/spectral analysis in the cochlea is followed by an analysis of the time pattern of the neural spikes evoked at each CF (Moore, 1982, 2003; Srulovicz and Goldstein, 1983; Patterson, 1987b; Meddis and Hewitt, 1991; Meddis and O'Mard, 1997). The model proposed by Moore is illustrated in Figure 6.6. The sound is passed through an array of bandpass filters, each corresponding to a specific place on the BM. The time pattern of the neural impulses at each CF is determined by the waveform at the corresponding point on the BM. The interspike intervals at each CF are determined. Then, a device compares the time intervals present at different CFs and searches for common time intervals. The device may also integrate information over time. In general, the time interval which is found

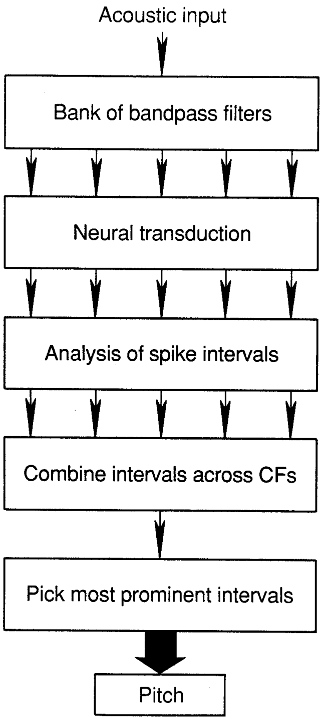


Figure 6.6. A schematic diagram of the spectro-temporal model of pitch perception for complex tones described by Moore (2003).

most often corresponds to the period of the fundamental component. The perceived pitch corresponds to the reciprocal of this interval. For example, if the most prominent time interval is 5 ms, the perceived pitch corresponds to a frequency of 200 Hz.

As described above, complex tones with low, resolvable harmonics usually give rise to clear pitches. F0DLs for such tones are small. Tones containing only high, unresolvable harmonics give less clear pitches, and F0DLs are larger, although F0DLs can be small for a tone whose lowest harmonic is around the eighth, even though all harmonics in the tone are probably unresolved. The effects of harmonic number probably arise because the temporal information conveyed by the low harmonics is less ambiguous than the temporal information conveyed by the high harmonics (Moore, 2003; Moore *et al.*, 2006).

Consider two complex tones, one containing three low harmonics, say 800, 1000 and 1200 Hz, and the other containing three high harmonics, say 1800, 2000 and 2200 Hz. For the first tone, the components are largely resolved on the BM. The neurones with CFs close to 800 Hz respond as if the input were an 800-Hz sinusoid. The time intervals between successive nerve impulses are multiples of the period of that tone, that is 1.25, 2.5, 3.75, 5.0 ... ms. Similarly, in neurones with CFs close to

1000 Hz the intervals between successive nerve spikes are multiples of 1 ms, that is 1, 2, 3, 4, 5 ... ms, and in neurones with CFs close to 1200 Hz the intervals are 0.833, 1.67, 2.5, 3.33, 4.17, 5.0 ... ms. The only interval that is in common across CFs is 5 ms, and this unambiguously defines a pitch corresponding to 200 Hz (the missing fundamental).

Consider now the response to the second complex tone, with three high harmonics. These harmonics are not resolved. They give rise to maximum activity at a place on the BM with CF close to 2000 Hz. Neurones with CFs around 2000 Hz are driven by a complex waveform. The temporal structure of the response is correspondingly complex. Each peak in the temporal fine structure of the waveform is capable of evoking a spike, so many different time intervals occur between successive spikes. The interval corresponding to F0, 5 ms, is present, but other intervals, such as 4.0, 4.5, 5.5 and 6.0 ms, also occur (Evans, 1978; Javel, 1980). Hence, the pitch is somewhat ambiguous. Increasing the number of harmonics leads to activity across a greater range of places on the BM. The pattern of interspike intervals is slightly different for each place, and the only interval that is in common across CFs is the one corresponding to the repetition period of the sound. Thus, the pitch becomes less ambiguous as the number of harmonics increases.

In summary, spectro-temporal theories can account for most existing data on the pitch perception of complex tones. The theories assume that both place analysis and temporal analysis are important, and that information about pitch can be extracted both from low harmonics and from high harmonics.

VIII.3 THE RELATIVE IMPORTANCE OF ENVELOPE AND TEMPORAL FINE STRUCTURE

Evidence that the time pattern of the waveform evoked on the BM by the higher harmonics plays a role comes from studies of the effect of changing the relative phase of the components in a complex tone. Changes in phase can markedly alter the waveform of a tone. This is illustrated in Figure 6.7, which shows the effect of changing the relative phase of one component in a complex tone containing just three harmonics, the ninth, tenth and eleventh. The left panels show waveforms of the individual sinusoidal components, and the right panels show the waveforms of the complex tones produced by adding together the sinusoidal components. For the case illustrated in the top half of the figure, all three components start in 'cosine' phase; this means that the components have their maximum amplitudes at the start of the waveform. Correspondingly, a peak in the envelope of the complex tone occurs at the start of the waveform, and a new peak in the envelope occurs for every ten oscillations in the temporal fine structure. At the point in time marked by the vertical dashed line, a peak in the waveform of the centre component (the tenth harmonic) coincides with minima in the waveforms of the two other harmonics. This gives a minimum in the envelope of the complex tone. Thus, the waveform of the complex tone has an envelope with distinct peaks and dips and is described as amplitude modulated. It might also be described as a 'peaky' waveform.

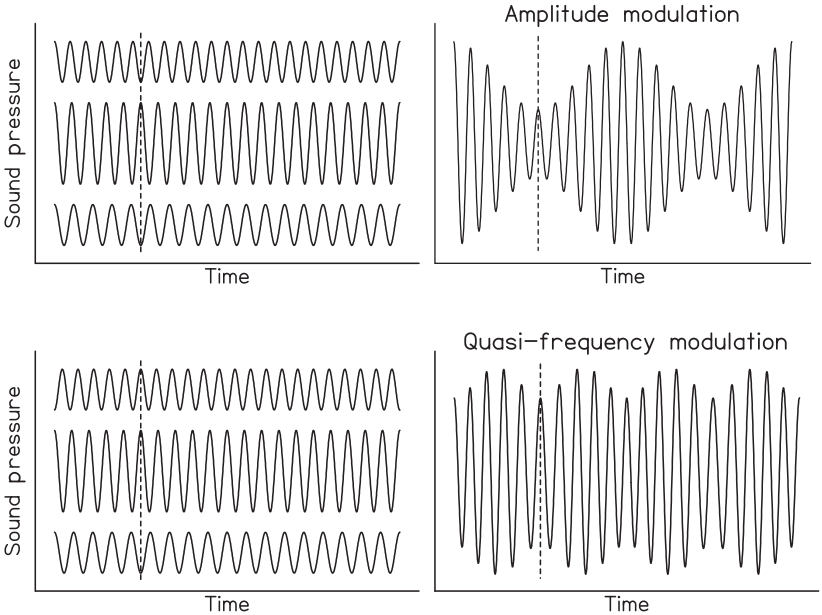


Figure 6.7. The effect of changing the relative phase of one component in a complex tone containing just three harmonics, the ninth, tenth and eleventh. The left panels show waveforms of the individual sinusoidal components, and the right panels show the waveforms of the complex tones produced by adding together the sinusoidal components. For one phase (upper panels), the waveform has an envelope with distinct peaks and dips; this waveform is sometimes called amplitude modulated. For the other phase (lower panels), the envelope is much flatter; this waveform is sometimes called quasi-frequency modulated.

For the case illustrated in the bottom half of the figure, the phase of the highest component is shifted by 180 °; that component starts with a minimum rather than a maximum in its waveform. As a result, the amplitude of the complex tone at the start of the waveform is not as high as for the case when all components started in cosine phase. At the point in time marked by the vertical dashed line, a peak in the waveform of the centre component (the tenth harmonic) coincides with a minimum in the waveform of the ninth harmonic and a peak in the waveform of the eleventh harmonic. The minimum in the envelope of the complex tone is less deep than the minimum when all components started in cosine phase. Thus, the envelope is much 'flatter', and the envelope actually shows two maxima for each period of the waveform. This waveform is sometimes called quasi-frequency modulated, as the time between peaks in the temporal fine structure fluctuates slightly; however, this is not easily visible in the figure.

If the harmonics have low harmonic numbers (say the second, third and fourth), they will be resolved on the BM. In this case, the relative phase of the harmonics is of little importance as the envelope on the BM does not change when the relative phases of the components are altered. However, if the harmonics have high harmonic numbers (as in Figure 6.7), then changes in the relative phase of the harmonics can result in changes in the envelope of the waveform on the BM. If this waveform has a peaky envelope, the repetition period will be clearly represented in the intervals between neural impulses. However, if the waveform has a flatter envelope, the repetition period will be less clearly represented.

For tones containing only high harmonics, phase can affect both the pitch value and the clarity of pitch (Moore, 1977; Patterson, 1987a; Shackleton and Carlyon, 1994). Moore *et al.* (2006) measured F0DLs using stimuli similar to those illustrated in Figure 6.7. To prevent subjects basing their judgements on the frequencies of individual harmonics, or on overall shifts in the excitation pattern, the number of the lowest component, *N*, was randomly varied by ±1 from one stimulus to the next. Their mean results for normally hearing subjects are shown in Figure 6.8. The F0DLs were not affected by the phase of the components when *N* was 7 or less. However, when *N* was 8 or more, the waveform with the peaky envelope led to smaller F0DLs than the waveform with the flatter envelope. F0DLs worsened progressively as *N* was increased from 8 to 12, and then flattened off.

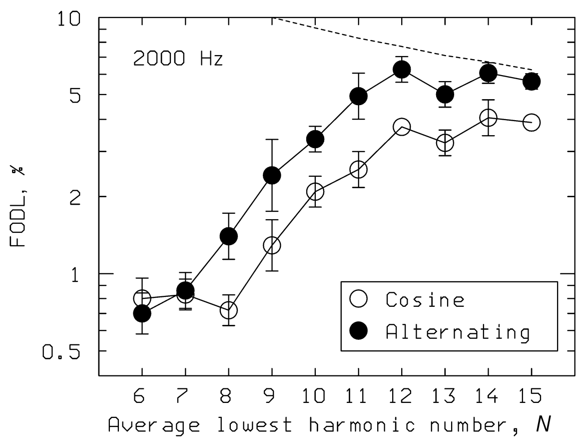


Figure 6.8. Mean results of Moore *et al.* (2006) for normally hearing subjects, showing F0DLs plotted as a function of the number, *N*, of the lowest harmonic in three-component complex tones. Open circles show F0DLs for tones with components added in cosine phase (giving a 'peaky' waveform) and filled circles show F0DLs for tones with components added in alternating phase (giving a 'flatter' waveform). The dashed line indicates the smallest F0DLs that could be achieved by making use of shifts in the excitation pattern of the stimulus.

To explain these results, Moore *et al.* (2006) proposed that, when N fell in the range 8–12, the harmonics were not resolved (which is why the F0DLs were influenced by the relative phase of the components), but temporal fine structure information could be used to achieve reasonably small F0DLs. The F0DLs were smaller for the cosine-phase than for the alternating-phase tones because the former contained just a few high-amplitude fine structure peaks per period, leading to a clear indication of the time interval corresponding to the F0. The worsening in performance with increasing N from 8 to 12 was argued to result from a progressive loss of the ability to use temporal fine structure information. For N above 12, it was argued that only envelope information could be used to extract the pitch, and not the temporal fine structure, as proposed earlier by Moore and Moore (2003).

When the harmonics have very high frequencies relative to F0, the interspike intervals may be too closely spaced or too ambiguous for their detailed timing to be used in determining the time interval corresponding to F0. Say, for example, that harmonics at 1900, 2000 and 2100 Hz are presented. The interval corresponding to F0, 10 ms, would be present, but other intervals very close to 10 ms would also be present, such as 9.5 and 10.5 ms. Under these conditions, the ability to use the information in the temporal fine structure appears to be lost, and the pitch can only be extracted from the bursts of neural firing that occur approximately every 10 ms, but are scattered around the interval of 10 ms. This is equivalent to using information derived from the envelope of the waveform on the BM, rather than its temporal fine structure, and it leads to poorer F0 discrimination (and a less clear pitch) than when temporal fine structure information can be used.

IX PITCH PERCEPTION OF COMPLEX TONES BY PEOPLE WITH COCHLEAR HEARING LOSS

IX.1 THEORETICAL CONSIDERATIONS

As described in Chapter 3, cochlear hearing loss is usually associated with reduced frequency selectivity; the auditory filters are broader than normal. This will make it more difficult to resolve the harmonics of a complex tone, especially when the harmonics are of moderate harmonic number. For example, for an F0 of 200 Hz, the fourth and fifth harmonics would be quite well resolved in a normal auditory system but would be poorly resolved in an ear where the auditory filters were, say, three times broader than normal. This is illustrated in Figure 6.9, which shows excitation patterns for a complex tone composed of the fourth to tenth harmonics of a 200-Hz fundamental for a normal auditory system (solid curve) and for an impaired auditory system (dashed curve) in which the auditory filters are three times broader than normal. The former shows distinct ripples corresponding to the lower harmonics, whereas the latter does not.

In the normal auditory system, complex tones with low harmonics give rise to clear pitches, while tones containing only high harmonics give less clear pitches.

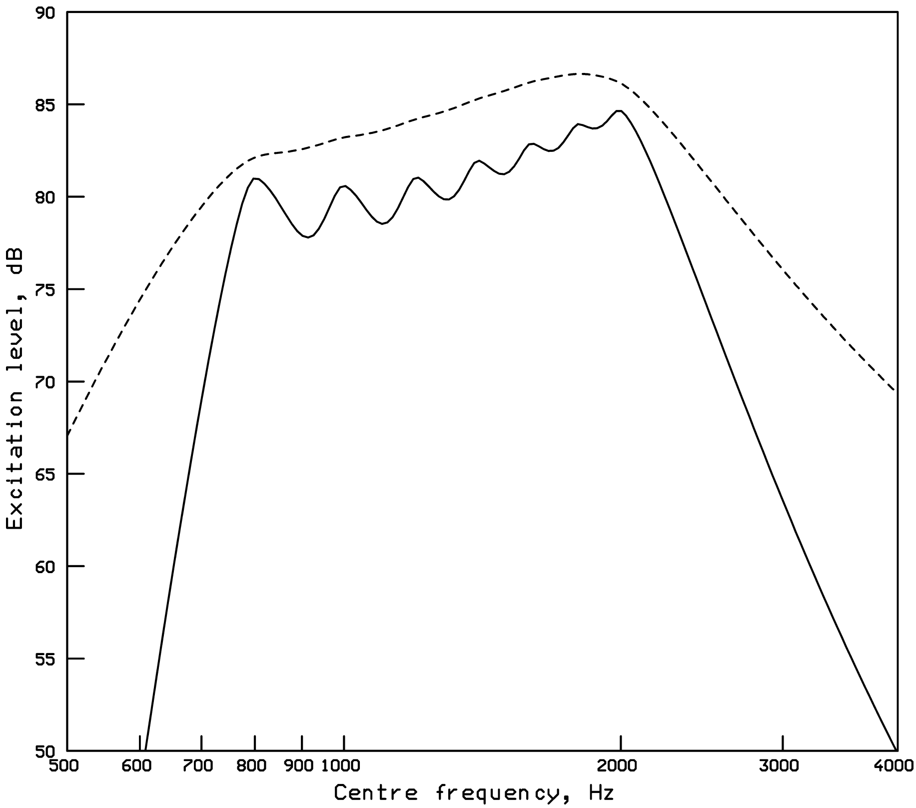


Figure 6.9. Psychoacoustical excitation patterns for a harmonic complex tone containing harmonics 4–10 of a 200-Hz fundamental. Each harmonic has a level of 80 dB SPL. Excitation patterns are shown for a normal ear (solid curve) and for an impaired ear (dashed curve) in which the auditory filters are three times as broad as normal.

As described earlier, the difference between the two types of tones probably arises because the temporal information conveyed by the low harmonics is less ambiguous than the temporal information conveyed by the high harmonics. Since cochlear hearing loss is associated with poorer resolution of harmonics, one might expect that this would lead to less clear pitches and poorer discrimination of pitch than normal. However, for complex tones with many harmonics, this effect should be small; spectro-temporal theories assume that information can be combined across different frequency regions to resolve ambiguities.

Spectro-temporal theories also lead to the prediction that the perception of pitch and the discrimination of repetition rate by subjects with broader-than-normal auditory filters might be more affected by the relative phases of the components than is the case for normally hearing subjects. For subjects with broad auditory filters, even the lower harmonics would interact at the outputs of the auditory filters, and higher harmonics would interact more than normal, giving a potential for strong phase effects.

In addition to producing reduced frequency selectivity, it is possible that cochlear damage leads to abnormalities in phase locking (neural synchrony) or in the ability to make use of phase-locking information. This could lead to less clear pitches and poorer discrimination of pitch than normal.

IX.2 EXPERIMENTAL STUDIES

The pitch discrimination of complex tones by hearing-impaired people has been the subject of several studies (Hoekstra and Ritsma, 1977; Rosen, 1987; Moore and Glasberg, 1988c, 1990; Moore and Peters, 1992; Arehart, 1994; Moore and Moore, 2003; Bernstein and Oxenham, 2006; Moore, Glasberg and Hopkins, 2006). Most studies have measured F0DLs. These studies have revealed the following:

1. There was considerable individual variability, both in overall performance and in the effects of harmonic content.
2. For some subjects, when F0 was low, F0DLs for complex tones containing only low harmonics (e.g. 1–5) were markedly higher than for complex tones containing only higher harmonics (e.g. 6–12), suggesting that pitch was conveyed largely by the higher, unresolved harmonics.
3. For some subjects, F0DLs were larger for complex tones with lower harmonics (1–12) than for tones without lower harmonics (4–12 and 6–12) for F0s up to 200 Hz. In other words, adding lower harmonics made performance *worse*. This may happen because, when auditory filters are broader than normal, adding lower harmonics can create more complex waveforms at the outputs of the auditory filters. For example, there may be more than one peak in the envelope of the sound during each period, and this can make temporal analysis more difficult (Rosen, 1986; Rosen and Fourcin, 1986).
4. The F0DLs were mostly only weakly correlated with measures of frequency selectivity. There was a slight trend for large F0DLs to be associated with poor frequency selectivity, but the relationship was not a close one. Some subjects with very poor frequency selectivity had reasonably small F0DLs.
5. There can be significant effects of component phase. In several studies, F0DLs have been measured with the components of the harmonic complexes added in one of two phase relationships, all cosine phase or alternating cosine and sine phase. The former results in a waveform with prominent peaks and low amplitudes between the peaks (as in the upper-right panel of Figure 6.7). The latter results in a waveform with a much flatter envelope (as in the lower-right panel of Figure 6.7). The F0DLs tended to be larger for complexes with components added in alternating sine/cosine phase than for complexes with components added in cosine phase. However, the opposite effect was sometimes found. The direction of the phase effect varied in an unpredictable way across

subjects and across type of harmonic complex. Phase effects can sometimes be stronger for hearing-impaired than for normally hearing subjects, although this is not always the case.

Hopkins and Moore (2007) studied how cochlear hearing loss affects sensitivity to changes in temporal fine structure in complex tones. To do this, they measured the ability to discriminate a harmonic complex tone, with F0 = 100, 200 or 400 Hz, from a similar tone in which all components had been shifted up by the same amount in Hertz, ΔF. For example, for an F0 of 100 Hz, the harmonic tone might contain components at 900, 1000, 1100, 1200 and 1300 Hz, while the shifted tone might contain components at 925, 1025, 1125, 1225 and 1325 Hz; the value of ΔF in this example is 25 Hz. People with normal hearing perceive the shifted tone as having a higher pitch than the harmonic tone (de Boer, 1956; Moore and Moore, 2003). The envelope repetition rate of the two sounds is the same (100 Hz), so the difference in pitch is assumed to occur because of a difference in the temporal fine structure of the two sounds (Schouten, Ritsma and Cardozo, 1962; Moore and Moore, 2003). To reduce cues relating to differences in the excitation patterns of the two tones, Hopkins and Moore (2007) used tones containing many components, and the tones were passed though a fixed bandpass filter. I consider here only one of their conditions, for which the filter was centred on the eleventh harmonic. To prevent components outside the passband of the filter from being audible, a background noise was added. In the presence of this noise, the differences in excitation patterns between the harmonic and frequency-shifted tones were very small.

The normally hearing subjects tested by Hopkins and Moore (2007) were able to perform this task well, presumably reflecting the ability to discriminate changes in the temporal fine structure of the harmonic and frequency-shifted tones. However, subjects with moderate cochlear hearing loss generally performed very poorly. The results suggest that moderate cochlear hearing loss results in a reduced ability, or no ability, to discriminate harmonic from frequency-shifted tones based on temporal fine structure.

Overall, these results suggest that people with cochlear hearing loss have reduced abilities to perceive the pitch, and to discriminate the F0, of complex tones. This appears to happen for at least two reasons: (1) the lower harmonics are less well resolved than normal, which makes it harder to determine the frequencies of individual low harmonics; (2) the ability to make use of temporal fine structure information (based on phase locking) is reduced.

X PERCEPTUAL CONSEQUENCES OF ALTERED FREQUENCY DISCRIMINATION AND PITCH PERCEPTION

X.1 EFFECTS ON SPEECH PERCEPTION

The perception of pitch plays an important role in the ability to understand speech. In all languages, the pitch patterns of speech indicate which are the most important words in an utterance; they distinguish a question from a statement and they indicate the structure of sentences in terms of phrases. In 'tone' languages, such as Mandarin Chinese, Zulu and Thai, pitch can affect word meanings. Pitch also conveys non-linguistic information about the gender, age and emotional state of the speaker. Supplementing lip reading (speech reading) with an auditory signal containing information only about the F0 of the voice can result in a substantial improvement in the ability to understand speech (Risberg, 1974; Rosen, Fourcin and Moore, 1981; Grant *et al*., 1985). The use of a signal that conveys information about the presence or absence of voicing (i.e. about whether a periodic complex sound is present or not) gives less improvement than when pitch is signalled in addition (Rosen, Fourcin and Moore, 1981). It seems likely that a reduced ability to discriminate pitch changes, as occurs in people with cochlear hearing loss, would reduce the ability to use pitch information in this way.

For complex tones, people with cochlear hearing loss are often more affected by the relative phases of the components than are normally hearing people. When a listener is reasonably close to a person speaking, and when the room has sound-absorbing surfaces, the waveforms reaching the listener's ears when a voiced sound is produced will typically have one major peak per period. These peaky waveforms may evoke a distinct pitch sensation. On the other hand, when the listener is some distance from the speaker, and when the room is reverberant, the phases of the components become essentially random (Plomp and Steeneken, 1973), with the result that the waveforms are less peaky. In this case, the evoked pitch may be less clear. The ability of hearing-impaired listeners to extract pitch information in everyday situations may be overestimated by studies using headphones or conducted in rooms with sound-absorbing walls.

For normally hearing listeners, several studies have shown that when two people are talking at once it is easier to 'hear out' the speech of individual talkers when their voices have different F0s (Brokx and Nooteboom, 1982). This effect probably arises in two main ways (Culling and Darwin, 1993). First, when the F0s of two voices differ, the lower resolved harmonics of the voices have different frequencies and excite different places on the BM. This allows the brain to separate the harmonics of the two voices and to attribute to one voice only those components whose frequencies form a harmonic series. This mechanism would be adversely affected by cochlear hearing loss, since reduced frequency selectivity would lead to poorer resolution of the harmonics. Secondly, the higher harmonics would give rise to complex waveforms on the BM, and these waveforms would differ in repetition rate for the two voices. The brain may be able to use the differences in repetition rate to enhance separation of the two voices. Both of these mechanisms might depend on the two voices having different short-term spectra. At any one time, the peaks in the spectrum of one voice would usually fall at different frequencies from the peaks in the spectrum of the other voice. Hence, one voice would dominate the BM vibration patterns at some places, while the other voice would dominate at other places. The local temporal patterns could be used to determine the spectral characteristics of each voice. This mechanism would also be impaired by cochlear hearing loss, for two reasons. First, reduced frequency selectivity would tend to result in more regions on the BM responding to the harmonics of both voices, rather than being dominated by a single voice. Secondly, abnormalities in temporal coding might lead to less effective representations of the F0s of the two voices. Consistent with this expectation, Rossi-Katz and Arehart (2005) showed that cochlear hearing loss markedly reduced the ability to use differences in F0 between two vowels to improve their identification when the differences in F0 were present only for the higher harmonics of the vowels. However, differences in F0 between the lower harmonics of the two vowels did lead to improved vowel identification.

X.2 EFFECTS ON MUSIC PERCEPTION

The existence of pitch anomalies (diplacusis and exaggerated pitch-intensity effects) may affect the enjoyment of music. Changes in pitch with intensity would obviously be very disturbing, especially when listening to a live performance, where the range of sound levels can be very large. There have been few, if any, studies of diplacusis for complex sounds, but it is likely to occur to some extent. One person studied in our laboratory was a professor of music who had a unilateral cochlear hearing loss. He reported that he typically heard different pitches in his normal and impaired ears, and that musical intervals in his impaired ear sounded distorted. Other subjects have reported that some musical notes do not produce a distinct pitch and that they get no pleasure from listening to music.

7 Spatial Hearing and Advantages of Binaural Hearing

I INTRODUCTION

Two ears are definitely better than one, for several reasons. First, differences in the intensity and time of arrival of sounds at the two ears provide cues that are used to localize sound sources. Secondly, when a desired signal and a background noise come from different locations, comparison of the stimuli reaching the two ears improves the ability to detect and discriminate the signal in the noise. Thirdly, when trying to hear a sound such as speech in the presence of background noise, the speech-to-noise ratio may be much higher at one ear than at the other ear. For example, if the speech comes from the left and the noise from the right, the speech-to-noise ratio will be higher at the left ear than at the right. Under these circumstances, people are able to make use of the ear receiving the higher speech-to-noise ratio. Finally, even when the signals reaching the two ears are identical, the ability to discriminate or identify the signals is often slightly better than when the signals are delivered to one ear only. These advantages of having two ears can be reduced by cochlear hearing loss, but this does not always happen. This chapter reviews several aspects of binaural and spatial hearing, comparing results for normally hearing and hearing-impaired people.

The chapter starts with a review of the ability to localize sounds in space, that is to determine the direction that sounds are coming from. It is useful to define some common terms used in studies of sound localization. The word *binaural* refers to situations where sound reaches both ears. When the stimulus arriving at the two ears is identical, this is referred to as *diotic*. When the sound is different at the two ears, this is called *dichotic*. When earphones are worn, the sound sometimes appears to come from within the head. Judgements of the position within the head (e.g. towards the left ear, towards the right ear or in the middle of the head) are referred to as lateralization judgements.

The directions of sound sources in space are usually defined relative to the head. For this purpose, three planes are defined, as illustrated in Figure 7.1 (Blauert, 1997). The *horizontal plane* passes through the upper margins of the entrances to the ear canals and the lower margins of the eye sockets. The *frontal plane* lies at right angles to the horizontal plane and intersects the upper margins of the entrances to the ear canals. The *median plane* lies at right angles to the horizontal and frontal planes; points in the median plane are equally distant from the two ears. The point of intersection of all three planes lies roughly at the centre of the head; it defines the origin for a coordinate system for specifying the angles of sounds relative to the head. The direction of a sound can be specified by its *azimuth* and its *elevation*. All sounds lying in the median plane have 0° azimuth. All sounds lying in the horizontal plane have 0° elevation. A

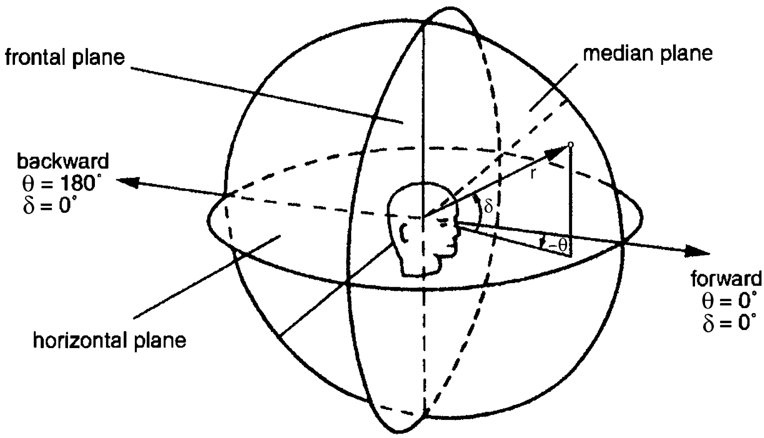


Figure 7.1. Illustration of the coordinate system used to define the position of sounds relative to the head. The azimuth is given by the angle θ (positive for leftwards directions and negative for rightwards directions) and the elevation by the angle δ (positive for upwards directions and negative for downwards directions). The distance is given by r. Adapted from Blauert (1997).

sound with 0 ° azimuth and 0 ° elevation lies directly in front of the head. A sound with 90 ° azimuth and 0 ° elevation lies directly opposite the left ear. A sound with 180 ° azimuth and 0 ° elevation lies directly behind the head. A sound with 0 ° azimuth and 90 ° elevation lies directly above the head, while a sound with 0 ° azimuth and 270 ° elevation lies directly below the head. Generally, the azimuth is the angle produced by projection onto the horizontal plane (θ in Figure 7.1), while the elevation is the angle produced by projection onto the median plane (δ in Figure 7.1).

There are two aspects to performance in the localization of sounds. The first aspect is concerned with how well the perceived direction of a sound source corresponds to its actual direction. For normally hearing people, the perceived direction generally corresponds reasonably well with the actual direction, although, for sinusoidal stimuli, errors sometimes occur in judging whether a sound is coming from in front or behind, or from above or below the horizontal (Stevens and Newman, 1936); this point is discussed in more detail later on. The second aspect of localization is concerned with how well subjects can detect a small shift in the position of a sound source. This aspect measures the resolution of the auditory system. When resolution is studied using stimuli presented via loudspeakers, the smallest detectable change in angular position, relative to the subject, is referred to as the *minimum audible angle* (MAA).

II THE LOCALIZATION OF SINUSOIDS

II.1 CUES FOR LOCALIZATION

Consider a sinusoidal sound source located to the left side of the head. The sound has to travel further to reach the right ear than to reach the left ear, so the sound is delayed in time in the right ear relative the left. In addition, the sound reaching the right ear is less intense than that reaching the left ear, because the head casts a kind of 'shadow' (see below for details). There are thus two physical differences in the sound at the two ears, the interaural time difference (ITD) and the interaural level difference (ILD), which could provide cues as to the location of the sound source. The magnitudes of these cues and the extent to which they are actually used by the auditory system are described below. Notice that the sounds at the two ears are normally fused and heard as a single sound image. For example, if an ILD is present, we do not hear a loud sound towards one side and a softer sound towards the other side; rather, a single sound image is heard towards the side receiving the more intense sound. The physical cues of ITD and ILD are used by the brain to create a percept at a specific subjective location.

The cues of ITD and ILD are not equally useful at all frequencies. Consider first the ILD, measured in dB. Low-frequency sounds have wavelengths which are long compared with the size of the head, and as a result they bend very well around the head. Thus, little or no 'shadow' is cast by the head. On the other hand, at high frequencies, where the wavelength is short compared with the dimensions of the head, a 'shadow', almost like that produced by an obstacle in a beam of light, occurs. ILDs are negligible at low frequencies (except when the sound source is very close to the head) but may be 20 dB or more at high frequencies. This is illustrated in Figure 7.2.

ITDs range from 0 (for a sound straight ahead) to about 650 μs for a sound at 90 ° azimuth (directly opposite one ear). This is illustrated in Figure 7.3. The ear at which

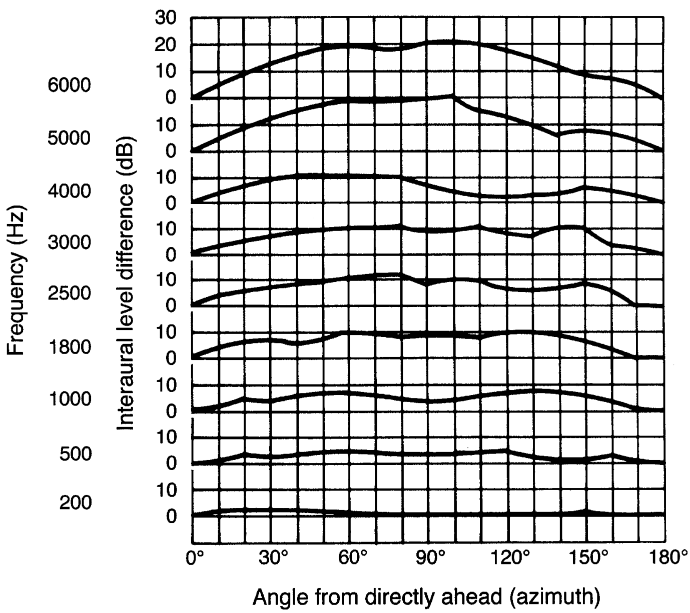


Figure 7.2. ILDs for sinusoidal stimuli plotted as a function of azimuth; each curve is for a different frequency. Adapted from Feddersen *et al.* (1957).

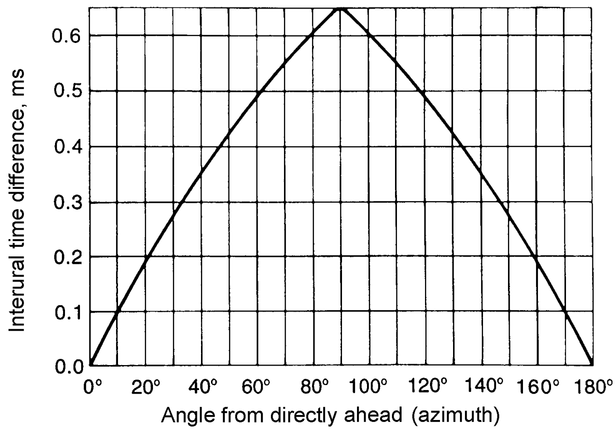


Figure 7.3. ITDs plotted as a function of azimuth. Adapted from Feddersen *et al.* (1957).

the sound arrives earlier is often called the leading ear; the other ear is called the lagging ear. For a given azimuth, the ITD is roughly independent of the frequency of the sound source. However, the usefulness of ITDs as a cue for localization does vary with frequency, as will be explained below. For a sinusoidal tone, a time difference is equivalent to a phase difference between the two ears. For example, if a 100-Hz tone is delayed at one ear by 100 μs, this is equivalent to a phase shift of one-tenth of a cycle. For low-frequency tones, the phase difference provides effective information about the location of the sound. However, for higher-frequency sounds, the phase difference provides an ambiguous cue.

Consider a sinusoid with a frequency of 769 Hz. This has a period of 1.3 ms. If the sinusoid is delayed by 650 μs at one ear relative to the other ear, then the sinusoid has a phase difference of 180 ° between the two ears. In other words, a peak in the waveform at the left ear coincides with a minimum in the right ear, and vice versa. In this situation, the auditory system has no way of knowing whether the leading ear was the left or the right; the sound at the right ear might be 650 μs ahead of that at the left ear or 650 μs behind. For a high-frequency sinusoid, say around 10 kHz, there may be many cycles of phase difference between the two ears. The auditory system has no way of determining which cycle in the left ear corresponds to a given cycle in the right ear. Thus, the phase difference at the two ears provides unambiguous information for frequencies below 750 Hz, but the information becomes increasingly ambiguous at higher frequencies.

The idea that sound localization is based on ITDs at low frequencies and ILDs at high frequencies has been called the *duplex theory* and dates back to Lord Rayleigh (1907). While it appears to hold for sinusoids, it is not strictly accurate for complex sounds, as will be explained later.

II.2 PERFORMANCE OF NORMALLY HEARING PEOPLE IN LOCALIZATION AND LATERALIZATION

Studies of localization using sinusoids have usually used tone bursts with gradual onsets and offsets to minimize cues related to the interaural timing of the envelope; envelope cues are discussed in more detail later in this chapter. Figure 7.4 shows the MAA for sinusoidal signals, plotted as a function of frequency (Mills, 1958, 1972). All stimuli were presented in the horizontal plane, that is with an elevation of 0 °. Each curve shows results for a different reference azimuth; the task was to detect a shift in azimuth from that direction. The MAA was smallest for a reference azimuth of 0 °, that is for sounds coming from directly in front of the subject. A shift as small as 1 ° could be detected for frequencies below 1000 Hz. Performance worsened around 1500 Hz. This is consistent with the duplex theory; at 1500 Hz phase differences between the two ears are ambiguous cues for localization, and ILDs are small. Performance worsened markedly when the reference direction was moved away from 0 ° azimuth. Indeed, for reference azimuths of 60 and 75 °, the MAA was so large that it could not be determined when the frequency was around 1500 Hz.

When resolution is studied using earphones, it is possible to study the effectiveness of ILDs alone or ITDs alone. The discriminability of interaural differences is usually measured in a two-alternative forced-choice task. Two successive binaural stimuli are presented and they differ in ILD or ITD. For example, one stimulus might be

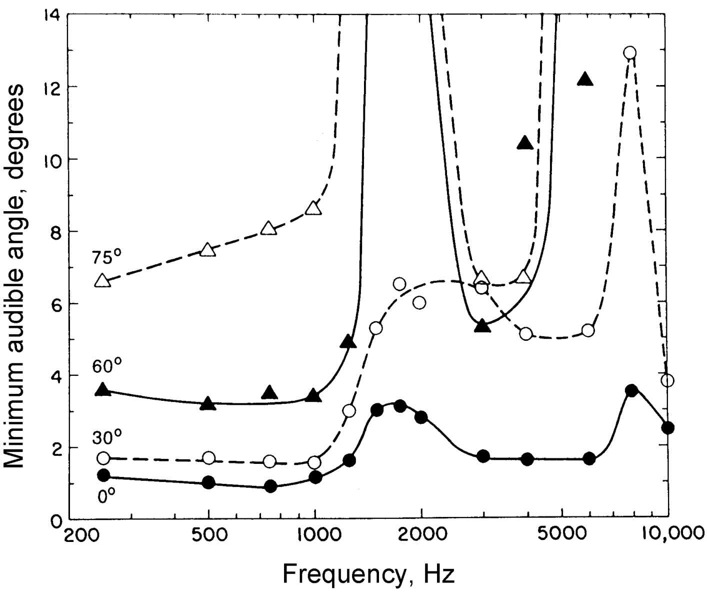


Figure 7.4. The MAA for sinusoidal signals, plotted as a function of frequency; each curve shows results for a different reference direction. Data from Mills (1958, 1972).

identical at the two ears, while the other stimulus might have an ITD of ΔT. The task of the subject is to say whether the first stimulus was to the left or the right of the second stimulus. In other words, the task is to identify a shift in the perceived location associated with a change in ILD or ITD.

Thresholds for detecting changes in ITD are smallest when the reference ITD is zero, corresponding to a sound heard in the centre of the head (Yost, 1974). In this case, a change in ITD of about 10 μs can be detected for a frequency of 900 Hz. Such a change would be produced by moving a sound source through an angle of about 1 ° relative to the subject; this corresponds well with the MAA value described above. The smallest detectable ITD increases somewhat at lower frequencies, although the threshold is roughly constant at about 3 ° when it is expressed as the smallest detectable change in relative phase at the two ears. Above 900 Hz, the threshold ITD increases markedly, and above 1500 Hz changes in ITD are essentially undetectable. The threshold ITD increases for all frequencies when the reference ITD is greater than zero, that is when the reference sound is heard towards one side of the head.

Thresholds for detecting changes in ILD are also smallest when the reference ILD is zero, corresponding to an identical sound at the two ears. In this case, changes in ILD of about 1 dB can be detected across a wide range of frequencies, although performance worsens slightly for frequencies around 1000 Hz (Mills, 1960; Yost and Dye, 1988). When the reference ILD is 15 dB, as would occur for a high-frequency sound well to one side of the head, the threshold change in ILD varies between 1.3 dB (at low and high frequencies) and 2.5 dB (around 1000 Hz).

In summary, the resolution of the binaural system, which determines the ability to detect changes in the position of a sound source, is best for sounds that come from directly ahead (0 ° azimuth). Resolution at low frequencies is based on the use of ITDs. Changes in ITD cannot be detected for frequencies above 1500 Hz. Changes in ILD can be detected over the whole audible frequency range, but, in practice, ILDs sufficiently large to provide useful cues to sound location occur primarily at high frequencies.

II.3 PERFORMANCE OF HEARING-IMPAIRED PEOPLE IN LOCALIZATION AND LATERALIZATION

There have been few studies of localization and lateralization of sinusoids by hearing-impaired subjects; most studies have used complex stimuli (see below). Nordlund (1964) measured free-field localization of sinusoids with frequencies 500, 2000 and 4000 Hz. He reported both the mean error (a measure of systematic errors or biases) and the standard deviation (a measure of resolution). For subjects with cochlear losses in both ears, only one out of thirteen showed any abnormal results (based on a 99.9 % confidence interval). For subjects with unilateral or highly asymmetric cochlear losses, seven out of nine showed at least one abnormal result. However, even for these subjects, the majority of results were within the normal range.

Häusler, Colburn and Marr (1983) reported limited measurements on subjects with unilateral cochlear losses (mainly Ménière's syndrome). Stimuli were presented via earphones. Thresholds for detecting changes in ITD were sometimes very large, but few specific details were given.

Hall, Tyler and Fernandes (1984) measured thresholds for detecting a change in ITD of a 500-Hz sinusoid at a level of 70 dB SPL. They tested six normally hearing subjects and ten subjects with mild-to-moderate symmetrical cochlear hearing losses. The ITD thresholds were markedly larger for the impaired subjects (mean 176 μs) than for the normally hearing subjects (mean 65 μs). The ITD thresholds tended to increase with increasing absolute threshold at 500 Hz, and three subjects with absolute thresholds less than 40 dB SPL had near-normal ITD thresholds. The poor performance of the subjects with higher absolute thresholds could have been partly due to the low sensation level (SL) of the stimuli, although it might also indicate a reduced ability to use phase-locking information.

Smoski and Trahiotis (1986) measured thresholds for detecting changes in ITD using 500-Hz sinusoids delivered by earphones. They used four subjects with mainly high-frequency cochlear losses caused by noise exposure or viral infections. At equal sound pressure level (80 dB SPL), the hearing-impaired subjects had higher ITD thresholds than the two normally hearing subjects tested. However, at an equal SL of 25 dB, ITD thresholds did not differ greatly for the normal and hearing-impaired subjects.

In summary, the limited data available on the localization and lateralization of sinusoids suggest that bilateral symmetric cochlear damage does not lead to marked abnormalities, except when the stimuli are at low SLs. Asymmetrical or unilateral damage more often leads to abnormalities. In most studies, abnormalities in localization were not closely related to audiometric thresholds, although sometimes absolute thresholds and ITD thresholds have been found to be correlated. Sometimes abnormal localization has been found in frequency regions where absolute thresholds are normal.

III THE LOCALIZATION OF COMPLEX SOUNDS

III.1 THE ROLE OF TRANSIENTS AND ACROSS-FREQUENCY COMPARISONS

All sounds which occur in nature have onsets and offsets, and many also change their intensity or their spectral structure as a function of time. ITDs of these transients provide cues for localization that are not subject to the phase ambiguities which occur for steady sinusoidal tones. Also, for sounds that cover a reasonable frequency range, phase ambiguities can be resolved by comparisons across frequency; the ITD that is common across all frequencies must be the 'true' ITD (Grantham, 1995; Stern and Trahiotis, 1995).

III.2 PERFORMANCE OF NORMALLY HEARING PEOPLE

Klump and Eady (1956) measured thresholds for discriminating changes in ITD using stimuli delivered via headphones. They compared three types of stimuli: band-limited noise (containing frequencies in the range 150–1700 Hz); 1000-Hz pure tones with gradual rise and fall times; and clicks of duration 1 ms. For a reference ITD of 0 °, the threshold ITDs were 9, 11 and 28 μs, respectively. Thus, the greatest acuity occurred for the noise stimulus. The single click gave rise to the poorest performance. However, threshold ITDs for click trains can be as small as 10 μs (Hafter and Dye, 1983) when the click rate is reasonably low (200 clicks per second or less). For bursts of noise (lowpass filtered at 5000 Hz), the ability to detect changes in ITD improves with duration of the bursts for durations up to about 700 ms, when the threshold change in ITD reaches an asymptotic value of about 6 μs (Tobias and Zerlin, 1959).

Yost, Wightman and Green (1971) investigated what frequencies were most important for the discrimination of changes in ITD of click trains. They did this by filtering the click trains and by adding filtered noise to mask the energy in certain frequency regions. They found that discrimination deteriorated for clicks which were highpass filtered so that only energy above 1500 Hz was present. However, discrimination was largely unaffected by lowpass filtering. Masking with a lowpass noise produced a marked disruption, while a highpass noise had little effect. Thus, discrimination of lateral position on the basis of time delays between the two ears depends largely on the low-frequency content of the clicks, although somewhat poorer discrimination is possible with only high-frequency components.

Henning (1974) investigated the lateralization of high-frequency sinusoids which were amplitude modulated; for an example of the waveform of an amplitude-modulated sound, see the upper-right panel of Figure 6.7. He found that the detectability of changes in ITD in the envelope of a 3900-Hz carrier modulated at a frequency of 300 Hz was about as good as the detectability of changes in ITD of a 300-Hz sinusoid. However, there were considerable differences among individual subjects; the threshold ITDs had values of 20, 50 and 65 μs for three different subjects (all stimuli had 250-ms durations and 50-ms rise–fall times). Henning found that the time delay of the envelope rather than the time delay of the fine structure (associated with the carrier frequency) within the envelope determines the lateralization. The signals could be lateralized on the basis of time delays in the envelope, even when the carrier frequencies were different in the two ears. Thus, it seems that for complex signals containing only high-frequency components, listeners extract the envelopes of the signals and compare the relative timing of the envelopes at the two ears. However, lateralization performance is best when the carrier frequencies are identical.

Overall, these results lead to some revision of the basic principles of the duplex theory. Complex sounds containing only high frequencies (above 1500 Hz) can be localized on the basis of ITDs, but this is done by comparing the timing of the envelope at the two ears.

III.3 PERFORMANCE OF PEOPLE WITH COCHLEAR HEARING LOSS

A survey of studies of localization and lateralization in hearing-impaired people, for studies up to 1981, has been given by Durlach, Thompson and Colburn (1981). A wide variety of stimuli were used in the studies, but the majority used either wideband noise or filtered noise. Durlach *et al*. concluded that many of the studies were hard to interpret because they did not distinguish between systematic errors in localization and poor resolution. Nevertheless, there was a clear trend for poor localization and lateralization to occur in people with unilateral or asymmetrical cochlear damage. Subjects with symmetrical cochlear losses often showed near-normal performance, especially when tested at reasonably high sound levels.

Häusler, Colburn and Marr (1983) studied sound localization and lateralization in hearing-impaired subjects using broadband noise as a stimulus. They measured MAAs for stimuli presented in free-field at various azimuths and ITD and ILD thresholds for stimuli presented via earphones. Stimuli were presented at levels ranging from 85 to 100 dB SPL. Some subjects with bilateral sensorineural losses (presumably cochlear in origin) showed MAAs within the normal range, both for stimuli at 0° azimuth, and for stimuli to one side. These subjects nearly always had good speech-discrimination scores. Other subjects with bilateral losses showed somewhat larger MAAs than normal, especially for stimuli located to the side. These subjects usually had poor speech-discrimination scores. Subjects with unilateral or asymmetric losses (mainly Ménière's syndrome) tended to have larger MAAs than normal, especially for sounds towards the side of the impaired ear, but a few subjects showed performance within the normal range.

Subjects with bilateral sensorineural losses usually showed thresholds for the detection of changes in ITD and ILD that were within the normal range, although a few showed larger ITDs than normal. Subjects with unilateral or asymmetric sensorineural losses (Ménière's) showed threshold ITDs that were normal or slightly larger than normal. However, threshold ILDs were mostly larger than normal.

Hawkins and Wightman (1980) measured thresholds for detecting changes in ITD and ILD for narrow bands of noise centred at 500 and 4000 Hz, using eight subjects with moderate cochlear losses, six of whom had symmetrical losses. The remaining two had unilateral losses caused by Ménière's syndrome. The stimuli were presented at 85 dB SPL and were delivered by earphones. For most subjects, ITD thresholds were much higher than normal at 4000 Hz (greater than 250 μs, as compared to 40–80 μs for normally hearing subjects). At 500 Hz, ITD thresholds were more variable, with two subjects showing performance within the normal range (12–29 μs) and the remainder showing thresholds in the range 33 μs to unmeasurable. The ILD thresholds were often close to normal, except for the subjects with unilateral losses. Abnormally large ITD thresholds were sometimes found when both ILDs and absolute thresholds were normal.

Smoski and Trahiotis (1986) measured thresholds for detecting a change in ITD using narrowband noises centred at 500 and 4000 Hz and an amplitude-modulated 4000-Hz sinusoidal carrier. Stimuli were delivered by earphones. They used four subjects with mainly high-frequency cochlear losses caused by noise exposure or viral infections. At equal sound pressure level (80 dB SPL), the hearing-impaired subjects had higher ITD thresholds than the two normally hearing subjects tested. The difference was small for the 500-Hz stimulus, but was large for the 4000-Hz stimuli. For example, for the 4000-Hz amplitude-modulated stimulus, the threshold ITD averaged 25 μs for the normally hearing subjects and about 250 μs for the hearing-impaired subjects. However, at an equal SL of 25 dB, ITD thresholds were much larger (200–600 μs), but did not differ greatly for the normal and hearing-impaired subjects.

Kinkel, Kollmeier and Holube (1991) measured thresholds for detecting changes in ITD and ILD of narrow bands of noise presented via earphones at centre frequencies of 500 and 4000 Hz. They used 15 normally hearing subjects and 49 subjects with hearing loss presumed to be mainly of cochlear origin. Stimuli were presented at 75 dB SPL or at the 'most comfortable level', whichever was the higher. The ITD thresholds were, on average, much larger for the hearing-impaired than for the normally hearing subjects, both at 500 Hz (210 μs versus 38 μs) and at 4000 Hz (530 μs versus 81 μs), although some hearing-impaired subjects had ITDs within the normal range. The average ILD thresholds were also larger for the hearing-impaired than for the normally hearing subjects, both at 500 Hz (4.7 dB versus 2.6 dB) and at 4000 Hz (5.1 dB versus 2.2 dB). However, many hearing-impaired subjects had ILD thresholds within the normal range.

Gabriel, Kollmeier and Mellert (1992) measured thresholds for detecting changes in ITD and ILD for narrowband noise stimuli presented via earphones at about 30 dB SL. The centre frequency of the noise was varied from 250 to 4000 Hz in one-octave steps. They used three subjects with symmetrical cochlear losses. Both ITDs and ILDs were generally larger than normal, although results varied markedly across subjects. One subject with a flat loss showed essentially no sensitivity to changes in ITD, except at 500 Hz, where the threshold was 100 μs. That subject also had unusually large ILD thresholds (3–8 dB, as compared to about 1 dB for normally hearing subjects). Even for two subjects with similar audiograms and similar causes of hearing loss (noise induced), performance differed markedly both in the values of the thresholds and in their pattern across frequency.

It is clear from these studies that binaural performance can vary markedly across subjects. Subjects with unilateral or asymmetric losses tend to show larger-than-normal thresholds for detecting changes in ITD and ILD. Subjects with symmetrical losses sometimes show normal or near-normal performance for broadband noise stimuli. However, they often show impaired performance for narrowband stimuli. It is possible, as pointed out by Colburn and Trahiotis (1992), that good performance for a restricted frequency range may be enough to ensure good performance for broadband stimuli.

Lacher-Fougère and Demany (2005) measured thresholds for detecting a change in interaural phase difference (IPD) (relative to a reference stimulus with an IPD of 0 °) of sinusoidally amplitude-modulated pure tones using seven normally hearing listeners and nine listeners with bilaterally symmetric hearing losses of cochlear origin. The IPDs were imposed either on the carrier signal alone and not on the envelope, or vice versa. The carrier frequency was 250, 500 or 1000 Hz, the modulation frequency was 20 or 50 Hz and the sound level was fixed at 75 dB SPL. Thresholds were typically higher for the hearing-impaired than for the normally hearing listeners for both types of IPD change, but the deficit was markedly larger for carrier IPDs than for envelope IPDs. In other words, the hearing-impaired subjects had particular difficulty in discriminating the stimuli based on their temporal fine structure.

III.4 REASONS FOR LARGE INTERAURAL TIME DIFFERENCE AND INTERAURAL LEVEL DIFFERENCE THRESHOLDS IN PEOPLE WITH COCHLEAR HEARING LOSS

The poor discrimination of ITDs, when it occurs, may be the result of several factors:

1. It may be partly caused by the relatively low SL of the stimuli. Hearing-impaired people are usually tested at low SLs; higher SLs are not possible because of their loudness recruitment. ITD discrimination in normally hearing subjects worsens markedly when the sound level is below about 20 dB SL (Häusler, Colburn and Marr, 1983), and so poor performance would also be expected in hearing-impaired people when tested at low SLs.
2. It may result from abnormalities in the propagation time of the travelling wave along the BM (basilar membrane) or in the phase of neural spike initiation, and from differences in travel time or phase of spike initiation between the two ears (Ruggero and Rich, 1987; Ruggero, Rich and Recio, 1993). As described above, discrimination of ITDs tends to be best for low-frequency sounds. The propagation time of the travelling wave to its point of maximum amplitude on the BM is probably in the range 5–9 ms for low-frequency sounds in a normal ear (Ruggero and Rich, 1987). Cochlear damage can result in marked changes in this propagation time, which may amount to as much as 1 ms (Ruggero, 1994). However, the direction of the change depends upon the exact nature of the underlying pathology (Ruggero, 1994). It seems likely that cochlear damage will often result in differences between the two ears in propagation time along the BM, especially for asymmetric losses, and this will have an effect similar to introducing a time delay between the two ears. The difference in travel time may be large compared to the range of ITDs that occurs naturally (up to 650 μs), and this would adversely affect the discrimination of ITDs. It may also strongly affect the perceived location of sound sources.
3. The people tested were often hearing-aid users. Hearing aids can introduce time delays which are a significant proportion of the ITDs that occur naturally. Extensive use of a monaurally fitted hearing aid could result in some form of adaptation to the abnormal ITD. This may impair discrimination of the ITD when the aid is not worn.
4. Poor discrimination of the ITD may be the result of abnormalities in phase locking, or in the ability to make use of the patterns of phase locking (loss of sensitivity to temporal fine structure; see Chapter 1, Section VII.5 and Chapter 6, Sections IV and IX).

Abnormalities in ILD discrimination may also have multiple causes:

1. As for ITD discrimination, they may result from the relatively low SL of the stimuli.
2. They may result from abnormal *input-output functions* on the BM and from differences in input-output functions between the two ears. Put in another way, differences in loudness between the two ears, or in the rate of growth of loudness with increasing sound level, may impair the discrimination of ILDs. In this context, it is noteworthy that ILD discrimination in normally hearing subjects can be markedly impaired by putting an earplug in one ear (Häusler, Colburn and Marr, 1983).

IV THE CONE OF CONFUSION, HEAD MOVEMENTS AND PINNA CUES

IV.1 THE CONE OF CONFUSION

If the influence of the pinnae is ignored, then the head may be regarded as a pair of holes separated by a spherical obstacle. If the head is kept stationary, then a given ITD will not be sufficient to define uniquely the position of the sound source in space. For example, if the sound at the left ear leads the sound at the right ear by 50 μs, this indicates that the sound lies to the left, but the sound could be in front of or behind the head, and above or below the horizontal plane. In fact, there is a *cone of confusion* such that any sound source on the surface of this cone would give rise to the same ITD

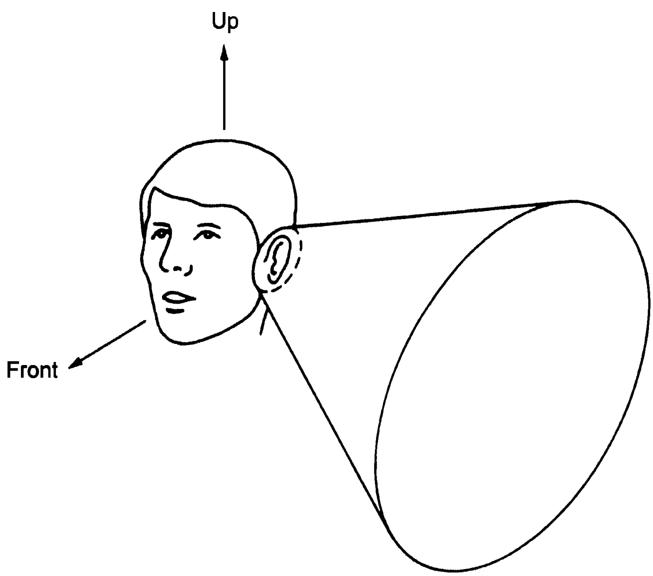


Figure 7.5. Illustration of the cone of confusion for a specific ITD. All points on the surface of the cone give rise to that same ITD.

(see Mills, 1972, and Figure 7.5). A similar ambiguity arises for ILDs. One special case applies when both the ITD and ILD are zero, that is the sound is the same at the two ears. In this case, the sound could be located anywhere in the median plane; the cone 'expands' to become a plane.

IV.2 THE ROLE OF HEAD MOVEMENTS

Ambiguities related to the cone of confusion may be resolved by head movements. Suppose, for example, that it is desired to resolve an ambiguity about the location of a sound source in the vertical direction, that is about its elevation. If the head is rotated about a vertical axis by, say, 20 ° and this results in a 20 ° shift in the apparent lateral position of the auditory image in relation to the head, then the sound source must be located in the horizontal plane. If the rotation of the head is accompanied by no change in the auditory image, then the sound must be located either directly above or directly below the head. Human listeners do seem to be able to use head movements to resolve ambiguities in this way (Wallach, 1940; Perrett and Noble, 1997). Intermediate shifts in the location of the auditory image lead to intermediate vertical height judgements (Wallach, 1940).

IV.3 INFORMATION PROVIDED BY THE PINNAE

Even when the head is fixed, or when the duration of a sound is too short to allow useful information to be gained from head movements, ambiguities associated with the cone of confusion do not seem to pose a problem in everyday situations. For example, a brief click can usually be localized accurately, without any confusion about whether it comes from in front of or behind the head. The ambiguities are resolved by information provided by the pinnae. Some of the sound reaching the ear enters the meatus directly, but some enters the meatus after reflection from one or more of the folds in the pinna. There are also some reflections from the shoulders and the upper part of the torso. When the direct sound and the reflected sound are added together, this results in a change in the spectrum of the sound reaching the eardrum. Say, for example, that the sound reflected from the pinna has a time delay of 75 μs (this is the time taken for a sound to travel a distance of about 2.5 cm), and that the sound source is a white noise, which contains a wide range of frequencies. For a frequency of 6667 Hz (period = 150 μs), the reflected sound is delayed by one-half of a period relative to the leading sound. Therefore, the direct and reflected sounds cancel, giving a dip in the spectrum at 6667 Hz. For a frequency of 13 333 Hz (period = 75 μs), the reflected sound is delayed by one whole period, and so the direct and reflected sounds add, giving a peak in the spectrum. Because there are usually several reflections from the pinna, the spectrum of the sound reaching the eardrum is complex, containing multiple peaks and dips. The spectral pattern varies systematically with the direction of the sound source (Wightman and Kistler, 1989; Blauert, 1997).

The spectral changes produced by the pinna can be used to judge the location of a sound source. Since it is the spectral *patterning* of the sound which is important, the information provided by the pinna is most effective when the sound has spectral energy over a wide frequency range. High frequencies, above 6 kHz, are especially important, since it is only at high frequencies that the wavelength of sound is sufficiently short for it to interact strongly with the pinna.

IV.4 LOCALIZATION USING PINNA CUES BY NORMALLY HEARING AND HEARING-IMPAIRED PEOPLE

Häusler, Colburn and Marr (1983) measured MAAs in the vertical direction for sounds coming from straight ahead, using broadband noise as a stimulus; in other words, subjects had to detect shifts in elevation for sounds presented in the median plane. Stimuli were presented at levels ranging from 85 to 100 dB SPL. One ear was plugged; so presumably the only cue available was the spectral patterning provided by the pinna and head. Subjects with normal hearing had MAAs in the range 1–10 °. Subjects with bilateral cochlear hearing loss fell into two groups. One group had MAAs within the normal range and also had good speech discrimination. The other group had MAAs that were so large that they could not be measured, and had poor speech discrimination. For subjects with unilateral losses due to Ménière's syndrome, some of the impaired ears also showed MAAs within the normal range, while others showed slightly larger-than-normal MAAs. MAAs were within the normal range for the normally hearing ear.

Noble, Byrne and Lepage (1994) required subjects to identify which loudspeaker from an array was emitting a broadband pink noise. They tested six normally hearing subjects and 87 subjects with sensorineural hearing loss. Of these, 66 subjects had a symmetrical loss that was assumed to be cochlear in origin. Generally, the hearing-impaired subjects did not perform as well as the normally hearing subjects. Differences in localization accuracy in different regions of auditory space were related to different configurations of hearing loss. The ability to judge the elevation of sounds (i.e. to judge location in the up–down dimension) depended on high-frequency sensitivity; high absolute thresholds at high frequencies were correlated with poor localization. Front–rear discrimination was correlated with absolute thresholds in the range 4–6 kHz, high thresholds being associated with poor discrimination.

In summary, hearing-impaired subjects can differ markedly in their ability to make use of the high-frequency spectral cues provided by the pinna. Some subjects appear to be completely unable to use these cues, while others use them almost normally. The poor performance, when it occurs, is probably due to two factors:

1. The high frequencies that carry pinna cues may be inaudible, or at very low SLs.
2. The peaks and dips in spectrum introduced by the pinna may be not be resolved because of reduced frequency selectivity (Jin *et al.*, 2002).

V GENERAL CONCLUSIONS ON SOUND LOCALIZATION

The auditory system is capable of using several physical cues to determine the location of a sound source. Time and intensity differences at the two ears, changes in the spectral composition of sounds due to head-shadow and pinna effects, and changes in all of these cues produced by movements of the head or sound source can all influence the perceived direction of a sound source. In laboratory studies, usually just one or two of these cues are isolated. In this way, it has been shown that sometimes a single cue may be sufficient for the accurate localization of a sound source. For real sound sources, such as speech or music, all of the cues described above may be available simultaneously. In this situation, the multiplicity of cues makes the location of the sound sources more definite and more accurate.

Hearing-impaired people show considerable individual differences in their ability to localize sounds. Often, in studies measuring the horizontal MAA for broadband sounds at high levels, performance is nearly normal, especially in cases of symmetrical losses. In this situation, several cues are available – ITDs, ILDs and pinna cues – and an ability to use any one of these, or any combination, is sufficient to give good performance. In studies using earphones, where usually only one cue is present at a time, the performance of hearing-impaired subjects is often poorer than normal, especially when narrowband stimuli are used.

VI THE PRECEDENCE EFFECT

VI.1 THE PRECEDENCE EFFECT FOR NORMAL HEARING

In many listening environments, the sound from a given source reaches the ears via a number of different paths. Some of the sound arrives by a direct path, but much of it only reaches the ears after one or more reflections from nearby surfaces (echoes). The echoes arrive after the direct sound. One might expect that the echoes would be heard as separate events and would disturb sound localization, but this does not usually happen. Instead, two or more brief sounds that reach the ears in close succession are heard as a single sound if the interval between them is sufficiently short. The interval over which fusion takes place is about 5 ms for a single click followed by a single echo but is longer when multiple echoes are present and may be as long as 50 ms for complex sounds, such as speech or music. This effect has been called *echo suppression* (Blauert, 1997).

If two or more successive sounds are heard as fused, the location of the total sound is determined largely by the location of the first sound. This is known as the *precedence effect* (Wallach, Newman and Rosenzweig, 1949), although it has also been called the *Haas effect*, after Haas (1951), the law of the first wavefront (Blauert, 1997) and *onset dominance* (Litovsky *et al.*, 1999). Since the first sound is the sound coming directly from the source, this means that localization is generally accurate despite the presence of echoes. The precedence effect is reflected in the finding that the ability to detect shifts in the location of the lagging sound (the echo) is reduced for a short time following the onset of the leading sound (Zurek, 1980; Perrott, Marlborough and Merrill, 1989).

VI.2 THE PRECEDENCE EFFECT FOR IMPAIRED HEARING

There have been relatively few studies of the precedence effect for hearing-impaired listeners. Cranford *et al.* (1993) assessed the precedence effect using clicks presented via two loudspeakers located at azimuths of -45 and $+45\,^{\circ}$. The clicks were either synchronous in the two loudspeakers (which would normally be heard as a sound coming from the front at $0\,^{\circ}$ azimuth) or the click in one loudspeaker was presented slightly before the click in the other, with a delay ranging up to 8 ms. When the precedence effect operates, the sound is perceived as coming from the leading loudspeaker when the delay is 1 ms or more (Blauert, 1997). Subjects were required to indicate whether the sound came from in front, from the left or from the right. Responses were counted as correct when the sound was reported as coming from the direction of the leading click (or from straight ahead when there was no delay). Otherwise, they were counted as incorrect. Four groups of subjects were tested, young and elderly, with and without cochlear hearing loss. The clicks were presented at an SL of 40 dB. The results indicated that both age and hearing loss had detrimental effects on performance, and performance was worst overall for the elderly hearing-impaired group.

Goverts, Houtgast and van Beek (2002) measured the precedence effect using lowpass-filtered 5-ms noise bursts, presented over headphones. Each noise burst was divided into leading and lagging parts, one lasting x ms (where x is less than 5 ms) and one lasting $5 - x$ ms. The two parts were given opposite ITDs of $+200$ and -200 μs. For example, the leading part might arrive earlier at the left ear than at the right ear, while the lagging part might arrive earlier at the right ear than at the left ear. The strength of the precedence effect was investigated by requiring subjects to judge whether the sound came from the left or the right. In one set of conditions, the leading and lagging parts had equal durations, and performance was measured as a function of the signal level or the level of a background filtered white noise. In another set of conditions, the relative duration of the leading and lagging parts was varied, using presentation at about 40 dB SL in quiet and at about 6 dB above the masked threshold in filtered white noise. Relative to normally hearing subjects, the six hearing-impaired subjects tested showed a decreased precedence effect, although there was substantial individual variability. The reduced performance for the hearing-impaired subjects could not be explained in terms of reduced audibility. The mean performance of the hearing-impaired subjects at 40 dB SL in quiet was similar to that of normally hearing subjects when tested in masking noise at a signal-to-noise ratio of 0 dB.

Roberts, Koehnke and Besing (2003) measured a different aspect of the precedence effect, namely the degree of perceptual fusion of the leading and lagging sounds, for subjects with normal and impaired hearing (the latter being older than the former). They used 4-ms bursts of noise as the leading and lagging sounds, and measured the lag-burst threshold (LBT), defined as the time delay between the onsets of the leading and lagging bursts at which the sounds were no longer perceived as fused. Stimuli were presented over headphones, but were processed to simulate listening in different types of rooms. A single stimulus (composed of a leading and lagging part) was presented on each trial, and subjects were asked to indicate whether it sounded more like 'one click' or more like 'two clicks'. LBTs were measured in simulated reverberant and anechoic environments in quiet and noise, at SLs of 10, 20, 30, 40 and 50 dB.

For the normally hearing subjects, mean LBTs were about 3–7 ms in anechoic conditions, regardless of the presence or absence of noise or of SL, and were 28–35 ms in reverberant conditions, almost independent of the presence or absence of noise or of SL. For the hearing-impaired subjects, mean thresholds in quiet anechoic conditions decreased from about 15 to 6 ms as the SL increased from 10 to 50 dB, and thresholds in noise in anechoic conditions decreased from about 10 to 8 ms as the SL increased from 10 to 30 dB. Thresholds in quiet reverberant conditions increased with increasing SL from about 38 to 47 ms. Thresholds in noisy reverberant conditions were about 32–34 ms, regardless of SL. Some, but not all, of the hearing-impaired subjects had LBTs that were markedly higher than normal. At first sight, one might think that higher LBTs indicate greater fusion of the leading and lagging sounds, which might be advantageous for localization. However, Roberts, Koehnke and Besing (2003) pointed to previous work with normally hearing subjects indicating that late-arriving echoes have more influence on perceived location than earlier-arriving echoes (Stellmack, Dye and Guzman, 1999). They argued, therefore, that fusion of late-arriving echoes with the leading sound might be disadvantageous.

In summary, the results suggest that the precedence effect operates in a somewhat abnormal way in hearing-impaired subjects. This may lead to difficulties in localizing sounds in reverberant rooms, especially when background noise is present. It should be noted that the precedence effect depends strongly on binaural processing, that is comparison of the signals reaching the two ears. Unilateral hearing loss can lead to a breakdown of both echo suppression and the precedence effect. Thus, room echoes become much more noticeable and the localization of sound in rooms becomes more difficult.

VII BINAURAL MASKING LEVEL DIFFERENCES (MLDs)

VII.1 MLDs FOR NORMALLY HEARING PEOPLE

The masked threshold of a signal can sometimes be markedly lower when listening with two ears than when listening with one (Hirsh, 1948). Consider the situation shown in Figure 7.6a. White noise from the same noise generator is fed to both ears via stereo headphones. Sinusoidal signals, also from the same signal generator, are fed separately to each ear and mixed with the noise. Thus, the stimuli at the two ears are identical. Assume that the level of the tone is adjusted until it is just masked by the noise, that is it is at its masked threshold, and let its level at this point be L_0 dB. Assume now that the signal is inverted in phase (equivalent to a phase shift of 180° or π radians) in one ear only (see Figure 7.6b). The result is that the signal becomes audible again. The tone can be adjusted to a new level, L_π, so that it is

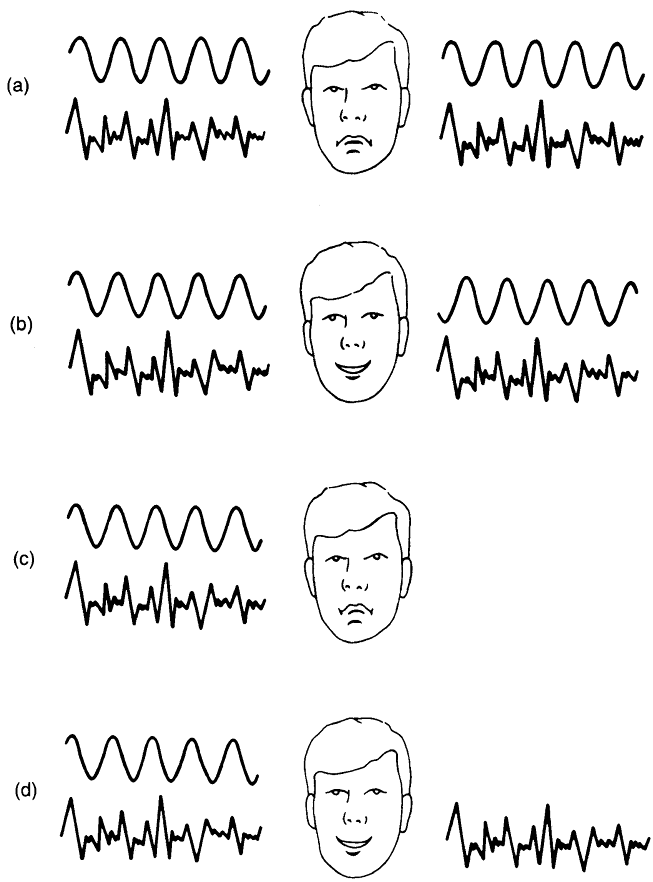


Figure 7.6. Illustration of two situations in which binaural MLDs occur. In conditions (a) and (c) detectability is poor, while in conditions (b) and (d), where the interaural relations of the signal and masker are different, detectability is good (hence the smiling faces).

once again at its masked threshold. The difference between the two levels, $L_0 - L_\pi$ (dB), is known as a *masking level difference* (MLD), and its value may be as large as 15 dB at low frequencies (around 500 Hz), decreasing to 2–3 dB for frequencies above 1500 Hz. Thus, simply by inverting the signal waveform at one ear the signal becomes considerably more detectable.

An example which is surprising at first sight is given in Figure 7.6c. The noise and signal are fed to one ear only, and the signal is adjusted to be at its masked threshold. Now the noise alone is added at the other ear; the tone becomes audible once again (Figure 7.6d). Further, the tone disappears when it, too, is added to the second ear, making the sounds at the two ears the same. Notice that it is important that the same noise is added to the non-signal ear; the noises at the two ears must be correlated or derived from the same noise generator. Release from masking is not obtained when an independent noise (derived from a second noise generator) is added to the non-signal ear.

The phenomenon of the MLD is not limited to pure tones. Similar effects have been observed for complex tones, clicks and speech sounds. Whenever the phase or level differences of the signal at the two ears are not the same as those of the masker, the ability to detect and identify the signal is improved relative to the case where the signal and masker have the same phase and level relationships at the two ears. Such differences only occur in real situations when the signal and masker are located in different positions in space. Thus, detection and discrimination of signals, including speech, are better when the signal and masker are not coincident in space than when they are coincident (e.g. when they come from the same loudspeaker).

To describe the conditions in experiments measuring MLDs, it is customary to use the symbols N (for noise) and S (for signal), each being followed by a suffix denoting relative phase at the two ears. A phase inversion is equivalent to a phase shift of 180 $^{\circ}$ or π radians. Thus, N_0S_π refers to the condition where the noise is in phase at the two ears and the signal is inverted in phase. N_u means that the noise is uncorrelated at the two ears. The suffix $_m$ indicates monaural presentation, that is presentation to one ear only. Table 7.1 gives the magnitude of the MLD for a variety of combinations of signal and noise. Four conditions for which there is no binaural advantage, N_0S_0, N_mS_m, N_uS_m and $N_\pi S_\pi$, all give about the same 'reference' threshold. The MLDs for the conditions shown in the table are obtained by expressing thresholds relative to this reference threshold.

In addition to improving the detectability of tones, conditions which produce MLDs also favour other aspects of the ability to discriminate signals. For example, when speech signals are presented against noisy backgrounds, speech intelligibility is better under conditions when the speech and noise do not have the same interaural phase (Licklider, 1948). However, the advantage for intelligibility is not as large as the advantage for detection. For single words in broadband white noise, the speech-to-noise ratio required for 50 % intelligibility is typically about 6 dB lower for N_0S_π

Table 7.1. Values of the MLD for various interaural phase relationships of the signal and masker. These results are typical for broadband maskers and low-frequency signals.

Interaural condition	MLD in dB
N_uS_π	3
N_uS_0	4
$N_\pi S_m$	6
N_0S_m	9
$N_\pi S_0$	13
N_0S_π	15

than for N_0S_0, whereas the threshold for detecting the words is about 13 dB lower (Levitt and Rabiner, 1967).

VII.2 MECHANISMS UNDERLYING MLDs

Models of the mechanisms underlying MLDs are complex, and it is beyond the scope of this book to describe them fully. Briefly, most recent models assume that firing patterns are compared for neurones with corresponding characteristic frequencies in the two ears. It is assumed that, at each characteristic frequency (CF), there is an array of delay lines which can delay the neural spikes from one ear relative to those from the other; each delay line has a characteristic time delay. These are followed by coincidence detectors, which count the numbers of spikes arriving synchronously from the two ears. If a sound is delayed by a time τ at the left ear relative to the right, then a delay of τ in the neural spikes from the right ear will cause the spikes from the left and right ears to be synchronous at the point of binaural interaction. Thus, the interaural delay of the signal is coded in terms of which delay line gives the highest response in the coincidence detectors.

This type of model was originally proposed to account for the ability to use ITDs in localizing sounds (Jeffress, 1948), but it has since been extended to account for MLDs and other aspects of binaural processing. It is useful to think of the outputs of the coincidence detectors as providing a kind of two-dimensional display; one of the dimensions is CF and the other is the interaural delay. The response to any stimulus is a pattern of activity in this display. When a signal and a masker have the same interaural delay time, τ, then they produce activity at overlapping points in the pattern; most activity lies around a line of equal delay (τ) versus CF. When a signal and masker have different ITDs, the pattern is more complex. The addition of the signal to the masker may cause activity to appear at points in the pattern where there was little activity for the masker alone. This could enhance the detection of the signal, giving an MLD. For further details, the reader is referred to Colburn (1996) and Stern and Trahiotis (1995).

VII.3 MLDs FOR PEOPLE WITH COCHLEAR HEARING LOSS

A survey of studies of the MLD using hearing-impaired subjects was presented by Durlach, Thompson and Colburn (1981). While there was considerable variability in the results across studies, generally it was found that cochlear hearing loss leads to reduced MLDs, even when the hearing loss is reasonably symmetrical. For example, Quaranta and Cervellera (1974) reported abnormally small MLDs in 86 % of cases. However, the relative sizes of the MLDs in different conditions (see Table 7.1) were often similar to normal. In some cases with asymmetric hearing losses, the pattern of results can be very different from normal. For example, in the study of Schoeny and Carhart (1971), thresholds for the condition N_0S_m were lower (better) when the signal was delivered to the better-hearing ear than when it was delivered to the ear with poorer hearing, even though the N_mS_m thresholds were similar for the two ears.

Several more recent studies have confirmed that MLDs are typically smaller than normal in subjects with cochlear hearing loss (Hall, Tyler and Fernandes, 1984; Jerger, Brown and Smith, 1984; Staffel *et al.*, 1990; Kinkel, Kollmeier and Holube, 1991). These studies have shown a trend for smaller MLDs in subjects with higher absolute thresholds, although the association is not generally strong, and subjects with similar absolute thresholds can have very different MLDs. MLDs also tend to be smaller the more asymmetrical is the loss (Jerger, Brown and Smith, 1984).

VII.4 POSSIBLE REASONS FOR SMALLER MLDs IN PEOPLE WITH COCHLEAR DAMAGE

In normally hearing subjects, MLDs increase with increasing sound level; so the reduced MLDs in hearing-impaired subjects could be partly a consequence of the low SL of the stimuli; the loudness recruitment associated with cochlear damage prevents the use of high SLs. However, this cannot be a complete explanation, since reduced MLDs for subjects with cochlear hearing loss occur when the stimuli are at SLs where normally hearing subjects show large MLDs (Quaranta and Cervellera, 1974). Also, reduced MLDs are found even when the relative levels of the stimuli at the two ears are adjusted so that an in-phase stimulus produces an image centred in the head (Quaranta and Cervellera, 1974).

Hall, Tyler and Fernandes (1984) suggested that the small MLDs found for hearing-impaired subjects might be partly a consequence of poor coding of temporal fine structure. They tested six normally hearing subjects and ten subjects with mild-to-moderate cochlear hearing losses. They found a significant correlation between the MLD at 500 Hz and ITD thresholds for 500-Hz sinusoids, supporting their suggestion. Kinkel, Kollmeier and Holube (1991) also found a correlation between the MLD at 500 Hz and the ITD threshold at 500 Hz. However, the correlation might have been due to the fact that both the MLD and the ITD threshold were correlated with age and with the absolute threshold at 500 Hz (Kinkel and Kollmeier, 1992).

Staffel *et al.* (1990) suggested that the small MLDs found for hearing-impaired subjects might be partly a consequence of asymmetry of the auditory filters at the two ears at a given centre frequency. If the filters have different amplitude and phase responses at the two ears, because of differences in the underlying pathology (e.g. different patterns of outer hair cell loss), this would lead to differences in the filter outputs in response to noise. Thus, the correlation of the noise would be reduced, which would in turn reduce the MLD; recall that MLDs do not occur when the noise is uncorrelated at the two ears. To test this hypothesis, they measured N_0S_0 and N_0S_π thresholds as a function of masker bandwidth, using a 500-Hz signal and noise centred at 500 Hz. Six normally hearing subjects and eighteen subjects with cochlear hearing loss were tested; ten of the latter had distinctly asymmetric hearing losses. Staffel *et al*. reasoned that interaural asymmetry of the auditory filters would have little effect for very small masker bandwidths but would have an increasing effect as bandwidth increased. However, the results did not support this hypothesis. The MLDs of the hearing-impaired subjects were smaller than normal both for a masker with a very small bandwidth (25 Hz) and for maskers with much wider bandwidths (400 and 800 Hz). Thus, while auditory-filter asymmetry may play a small role in reducing MLDs in hearing-impaired subjects, it does not seem to be a major factor.

In summary, the most likely causes of reduced MLDs found for people with cochlear damage are the reduced SL of the stimuli and reduced sensitivity to temporal fine structure.

VIII HEAD-SHADOW EFFECTS

VIII.1 BENEFITS OF HEAD SHADOW FOR NORMALLY HEARING PEOPLE

Most studies of the MLD have used conditions where performance cannot be improved simply by attending to one ear only. Usually, this means that the sounds were presented over earphones. Hence, these studies have been specifically concerned with the benefit of binaural processing, where information from the two ears is combined and compared. However, when listening for a signal in background noise in everyday listening situations (i.e. not using earphones), it is often the case that the signal-to-noise ratio is much better at one ear than at the other. The differences in signal-to-noise ratio occur as a result of the head-shadow effects illustrated in Figure 7.2. For example, if a speech signal comes from the left and an interfering noise comes from the right, the speech-to-noise ratio will be higher at the left ear than at the right. The improved signal-to-noise ratio occurs mainly at high frequencies (above about 2 kHz). An advantage of having two ears is that the listener can effectively ‘select’ the ear giving the higher signal-to-masker ratio.

Bronkhorst and Plomp (1988) studied the effects of ITDs and head shadow on the intelligibility of speech in noise, under binaural listening conditions. The sounds were recorded using a realistic model of a human head and torso (KEMAR; see Burkhard and Sachs, 1975). The noise had the same long-term average spectrum as the speech, and the speech and noise were recorded separately. The speech was presented from directly in front of KEMAR (0° azimuth), while the noise was presented at seven azimuths ranging from 0° (frontal) to 180°. For the recordings at 0° azimuth, ITDs and ILDs were essentially zero at all frequencies. The recordings of the noise were digitally processed to derive two types of signals. One was obtained by taking each of the recordings made at a non-zero azimuth and shaping the magnitude spectrum for each ear to match that obtained for a 0° azimuth. This preserved the ITD cues but removed the ILD cues. The other was obtained by taking the recording made at 0° azimuth and shaping the spectrum for each ear to match that for one of the non-zero azimuths. This meant that no ITD cues were present, but ILD cues appropriate for the non-zero azimuths were introduced.

The processed stimuli were presented via earphones. The speech-to-noise ratio required for 50 % of sentences to be understood (the *speech reception threshold*, SRT) was determined for each noise azimuth. The decrease in SRT, relative to the case where both speech and noise came from 0 ° azimuth, is called the *binaural intelligibility level difference* (BILD). The BILD due to ITDs was between 3.9 and 5.1 dB (for noise azimuths between 30 and 150 °). This BILD may be considered as a form of MLD resulting from comparison of the signals at the two ears, that is it reflects binaural processing. The BILD due to ILDs was 3.5–7.8 dB. This BILD probably has two components. One, like the BILD resulting from the ITDs alone, depends on binaural processing. The other depends on the fact that the signal-to-noise ratio at one ear is improved when the noise and speech are spatially separated. For example, for speech at 0 ° azimuth, when the noise is moved to the left, say to 45 ° azimuth, the signal-to-noise ratio improves at the right ear and worsens at the left ear. Performance can be improved by giving greater weight to the ear with the higher speech-to-noise ratio. In additional experiments, Bronkhorst and Plomp (1988) used the stimuli with ILDs alone, but they turned off the stimulus to one ear. Turning off the stimulus to the ear receiving the poorer signal-to-noise ratio produced little degradation in performance. They concluded that the advantage gained from ILDs mainly depends on the ear receiving the higher speech-to-noise ratio. However, this advantage decreases when the noise in the other ear is fairly loud.

When the unprocessed stimuli were used (i.e. when both ITDs and ILDs were present), the improvements were larger than for ITDs or ILDs alone, ranging from 5.8 to 10.1 dB. However, the overall effect was not as great as would be obtained if the BILDs gained from ITDs and ILDs were simply added.

In summary, spatial separation of speech and background noise can lead to a BILD of up to 10 dB. A large part of this effect is due to the fact that the speech-to-noise ratio is improved at one ear by head-shadow effects. A smaller part is due to binaural processing of ITDs.

VIII.2 BENEFITS OF HEAD SHADOW FOR HEARING-IMPAIRED PEOPLE

Bronkhorst and Plomp (1989) carried out similar experiments to those described above using seventeen subjects with symmetrical hearing losses and seventeen subjects with asymmetrical losses (differences in threshold between the two ears averaged over 500, 1000 and 2000 Hz, ranging from 5 to 31 dB). Most subjects were diagnosed as having mild to moderate cochlear hearing losses. The noise level was adjusted for each subject so as to be as far as possible above absolute threshold without being uncomfortably loud. When the speech and noise both came from 0 ° azimuth, the SRTs were, on average, 2.5 dB higher than found for normally hearing subjects. Both groups of hearing-impaired subjects showed 2.6–5.1 dB less binaural gain than normal when the noise azimuth was changed to 90 °. In other words, in this condition, SRTs were 5.1–7.6 dB higher than normal, a considerable difference.

The BILDs due to ILDs alone ranged from 0 dB to normal values of 7 dB or more. The size of the BILDs depended on the high-frequency loss in the ear receiving the higher speech-to-noise ratio; greater high-frequency losses were associated with a reduced advantage. This makes sense, since head-shadow effects are greatest at high frequencies, and if those high frequencies are inaudible, little advantage can be gained. The average BILD due to ITDs alone was nearly normal (4.2 dB as compared to 4.7 dB for normally hearing subjects) for subjects with symmetrical hearing losses. However, subjects with asymmetrical losses showed smaller BILDs, averaging 2.5 dB. When ITDs were introduced in stimuli already containing ILDs, the gain was 2–2.5 dB for both groups, comparable to what was obtained for normally hearing subjects.

In summary, subjects with cochlear hearing loss are generally less able than normal to take advantage of spatial separation of speech and interfering noise. When tested under conditions where speech and noise are spatially separated, they perform more poorly, relative to normal, than when the speech and noise come from the same position in space. The disadvantage appears to arise mainly from the inaudibility of high frequencies in the ear at which the speech-to-noise ratio is highest.

IX RELEASE FROM INFORMATIONAL MASKING

The types of simultaneous masking described in Chapter 3, Sections II and III can be explained largely in terms of processes occurring in the cochlea and auditory nerve. Masking of this type is sometimes called (inappropriately) *energetic masking*, and it occurs when the response of the auditory nerve to the masker-plus-signal is very similar to the response to the masker alone. When a masking sound is highly similar in some way to the signal, and/or when the properties of the masker vary in an unpredictable way from one stimulus to the next, there may be much more masking than would be expected from energetic masking alone. This 'extra' masking is called *informational masking*. It is assumed that informational masking occurs because the signal is confused with the masker, or because attention is directed to an inappropriate aspect of the sound (Carhart, Tillman and Greetis, 1969; Moore, 1980; Neff and Green, 1987; Freyman *et al.*, 1999).

When the task of a listener is to identify the speech of one talker in the background of another talker with similar characteristics, informational masking may play a strong role (Freyman *et al.*, 1999). Under these conditions, the amount of informational masking is greatly influenced by whether or not the target and background are perceived to come from the same location in space. This is illustrated by an experiment of Freyman *et al.* (1999). Normally hearing listeners were asked to identify nonsense syllables spoken by a female talker in the presence of a background of either *speech-shaped noise* or a second female talker. In one pair of conditions, the target and masker were both presented from two loudspeakers, one located directly in front of the listener (0 ° azimuth) and one located 60 ° to the right. The target speech from the loudspeaker at 0 ° was presented slightly earlier in time than the target speech from the loudspeaker at 60 °, which made the target speech appear to come from in front, owing to the precedence effect, as discussed in Section IV.1. In one condition (called here the coincident condition), the background sound was presented in the same way, so that it too was heard as coming from in front. In a second condition (called here the separated condition), the background sound was presented slightly earlier in time from the loudspeaker at 60 ° than from the loudspeaker at 0 °, which made the background appear to come from the right. The long-term average spectra of the target and background at the two ears were essentially the same for the two conditions. For the noise background, the percentage of key words identified was only 5–10 % better for the separated than for the coincident condition. However, for the female-talker background, the corresponding difference was 15–30 %. The greater effect for the speech masker was probably due to a release from informational masking caused by the perceived spatial separation of the target speech and the background.

I am not aware of any studies comparable to that of Freyman *et al.* (1999) but for hearing-impaired listeners. However, Arbogast, Mason and Kidd (2005) have studied the effect of spatial separation on speech recognition for hearing-impaired subjects using speech that was processed using a 'tone vocoder' so as to preserve mainly cues related to the temporal envelope of the speech in different frequency bands. The speech was filtered into fifteen 1/3-octave-wide bands, the envelope of each band was extracted, and the envelope for a given band was used to amplitude modulate a sinusoid with frequency corresponding to the centre of that band. The target sentences were generated by combining eight randomly chosen modulated sinusoids out of the fifteen. One of the maskers used, called the different band sentence (DBS) masker, was generated in a similar way to the target speech (but using a different sentence), by combining six randomly chosen modulated sinusoids, excluding the ones already used for the target speech. This was intended to reduce energetic masking, leading to a dominance of informational masking. The target sentences were presented from a loudspeaker directly in front of the listener (0 ° azimuth). The masker was presented at either 0 ° azimuth (coincident condition) or 90 ° azimuth (separated condition). The 'threshold' for each condition was determined as the target-to-masker ratio required for 51 % correct performance. The amount of spatial release from masking was then defined as the difference between thresholds for the coincident and separated conditions. They tested ten hearing-impaired subjects with mild-to-moderate hearing loss and ten age-matched normally hearing subjects. The masker level was set to approximately 40 dB SL whenever possible, but was lower than this for some of the normally hearing subjects and all of the hearing-impaired subjects because of comfort or equipment limitations.

The spatial release from masking for the DBS masker was, on average, about 15 dB for the normally hearing subjects and about 10 dB for the hearing-impaired subjects. The smaller release for the latter may have been due to the lower average SL of the stimuli or because of relatively more energetic masking, owing to the reduced frequency selectivity of the hearing-impaired subjects (see Chapter 3, Section VII). However, overall, the results suggest that the hearing-impaired subjects were able to take advantage of the spatial separation of the target and masker to reduce informational masking.

X DIOTIC ADVANTAGES

In many tasks, detection and discrimination performance is better when listening with two ears than with one, even when the same stimulus is delivered to the two ears. This is given the general name diotic summation. For example, the absolute threshold is about 1–2 dB lower for diotic than for monaural stimulation (Shaw, Newman and Hirsh, 1947), and thresholds for the detection of changes in frequency or intensity are also lower for diotic stimulation (Jesteadt, Wier and Green, 1977a).

Of perhaps more interest is the fact that the intelligibility of speech in noise is usually greater for diotic than for monaural stimulation. For normally hearing subjects, the diotic advantage in intelligibility for key words in simple sentences is about 9 % when baseline performance in the monaural condition is 70 % (Davis, Haggard and Bell, 1990), and about 20 % when monaural performance is 50 % (Davis and Haggard, 1982). Plomp and Mimpen (1979) reported that the SRT for sentences presented in noise with the same long-term average spectrum as the speech was 1.4 dB lower for diotic presentation than for monaural presentation. This corresponds to an improvement in intelligibility in difficult listening situations (when about 50 % of sentences are reported correctly) of about 20 %.

People with cochlear hearing impairment also show better speech intelligibility for diotic stimulation than for monaural stimulation. Laurence, Moore and Glasberg (1983) measured SRTs in noise of eight subjects with moderate to severe cochlear losses under three conditions: unaided, with linear hearing aids and with two-channel compression hearing aids (see Section III of Chapter 9 for more details of hearing aids with compression). For each condition, they compared three cases: listening with both ears (either unaided or binaurally aided), listening with the left ear only and listening with the right ear only. In the latter two cases, the unused ear was either plugged and muffed or fitted with a hearing aid that was turned off. For all three conditions, there was a significant advantage of binaural versus monaural listening, even when the speech and noise came from the same loudspeaker directly in front of the subject (this gives essentially diotic stimulation, at least for unaided listening). The advantage of using two ears was similar for unaided and aided conditions. On average, for the case where the speech and noise came from directly in front, the SRT was about 2 dB lower when listening binaurally than when listening with whichever single ear gave the lowest SRT. For the speech material they used, this is equivalent to an improvement in intelligibility of about 22 %. A similar average diotic advantage for subjects with cochlear hearing loss was reported by Moore *et al.* (1992).

In summary, performance on many tasks, including the understanding of speech in noise, is better with two ears than with one, even when the stimuli at the two ears are essentially the same. This diotic advantage occurs for both normally hearing people and people with cochlear damage. If anything, it may be larger for hearing-impaired people.

XI PERCEPTUAL CONSEQUENCES OF ABNORMAL BINAURAL AND SPATIAL HEARING IN PEOPLE WITH COCHLEAR DAMAGE

The most obvious consequence of abnormal binaural and spatial hearing is difficulty in localizing sound sources. Some people with cochlear hearing loss have essentially no ability to use spectral cues provided by the pinna. This may happen either because the cues are inaudible or because the patterns of spectral peaks and dips cannot be resolved. When pinna cues are lacking, this creates difficulty in deciding whether a sound comes from in front or behind, and from above or below. It should be noted that pinna cues are often not conveyed adequately by hearing aids. This happens for two reasons. First, the hearing aid's microphone may be in a position, such as behind the ear, where the pinna cues are not picked up. Secondly, most hearing aids do not amplify sounds in the frequency range above 6 kHz, where pinna cues are most effective. Hence, hearing-aid users often suffer from difficulty in resolving the cone of confusion.

Some people with cochlear damage have higher-than-normal thresholds for detecting changes in ITD and ILD. This applies especially to people with asymmetrical losses, and to sounds with narrow bandwidths. This means that the precision with which the azimuth of a sound source can be determined is reduced, making it more difficult to orient to a sound and to decide what physical object gave rise to that sound.

When listening in conditions of reverberation, cochlear hearing loss is sometimes but not always associated with a reduction in the precedence effect, which leads to increased difficulty in localizing sounds.

Many studies comparing the intelligibility of speech in noise for normally hearing and hearing-impaired subjects have used speech and noise coming from the same loudspeaker. Such studies underestimate the difficulty experienced by hearing-impaired people in everyday life, as they fail to take into account the reduced ability of people with cochlear damage to take advantage of spatial separation of a desired sound and one or more interfering sounds. This reduced ability shows up both in studies of the MLD using earphones and in studies using stimuli presented in a free field. The disadvantage in free field depends partly on reduced binaural processing abilities and partly on the reduced audibility of high-frequency sounds at the ear receiving the higher signal-to-background ratio. For normally hearing subjects, spatial separation of speech and background noise can reduce the SRT by up to 10 dB. For people with moderate cochlear damage, the advantage is 3–5 dB less. This is equivalent to a substantial loss in intelligibility.

8 Speech Perception

I INTRODUCTION

People with cochlear hearing loss frequently complain of difficulty with speech communication. The extent and nature of the difficulty depends partly on the severity of the hearing loss. People with mild or moderate losses can usually understand speech reasonably well when they are in a quiet room with only one person talking. However, they have difficulty when more than one person is talking at once, or when background noise or reverberation are present. People with severe or profound losses usually have difficulty even when listening to a single talker in a quiet room, and they generally have severe problems when background noise is present. Hence, their ability to understand speech relies heavily on lip reading and on the use of context.

There has been considerable controversy in the literature about the reasons for these difficulties in understanding speech. Some researchers have suggested that the difficulties arise primarily from reduced audibility; absolute thresholds are higher than normal, so the amount by which speech is above threshold, and the proportion of the speech spectrum which is above threshold, are both less than for normal listeners (Humes, Dirks and Kincaid, 1987; Zurek and Delhorne, 1987; Lee and Humes, 1993). In other words, it is argued that the difficulties occur mainly because part of the speech cannot be heard at all. Other researchers (Plomp, 1978, 1986; Dreschler and Plomp, 1980, 1985; Glasberg and Moore, 1989) have argued that the difficulty in understanding speech arises at least partly from a reduced ability to discriminate sounds which are well above the absolute threshold. Many of the problems of discrimination have been reviewed in earlier chapters. According to this point of view, even if speech is amplified so that it is audible, the hearing-impaired person will still have problems in understanding speech.

The evidence reviewed below indicates that, for mild losses, audibility is the single most important factor. However, for severe to profound losses, poor discrimination of suprathreshold (audible) stimuli is also of major importance.

II THE MAGNITUDE OF THE NOISE PROBLEM

The issue considered next is: how much worse than normal are hearing-impaired people in their ability to understand speech in noise? This is often quantified by estimating the speech-to-noise ratio required to achieve a certain degree of intelligibility, such as 50 % correct. This ratio is called the *speech reception threshold* (SRT) and is usually expressed in dB. The higher the SRT, the poorer the performance. For many of the common speech materials used, and especially for sentence lists (Bench and Bamford, 1979; Plomp and Mimpen, 1979; Nilsson, Soli and Sullivan, 1994), the percentage correct varies quite rapidly with changes in speech-to-noise ratio. For example, if the speech-to-noise ratio is set to a value giving about 50 % correct, increasing the ratio by 1 dB typically gives an increase in the percentage correct of 7–19 %. Correspondingly, even small differences in SRT between normally hearing and hearing-impaired listeners indicate substantial differences in the ability to understand speech in noise.

Plomp (1994) reviewed several studies which measured the SRT for sentences presented in a continuous background noise. The SRT was defined as the speech-to-noise ratio in dB required for 50 % of sentences to be identified completely correctly. The noise had the same long-term average spectrum as the speech; such noise is referred to as *speech-shaped noise*. For high noise levels, people with cochlear hearing loss had higher SRTs than normally hearing people. The increase in SRT varied from about 2.5 dB, for people with mild hearing losses caused by noise exposure or associated with ageing, to about 7 dB, for people with moderate to severe losses caused by Ménière's syndrome or by unknown pathologies. An elevation in SRT of 2.5 dB is sufficient to create a substantial loss of intelligibility in difficult listening situations.

The elevation in SRT can be much greater when a fluctuating background noise or a single competing talker is used instead of a steady noise. Normally hearing subjects are able to take advantage of temporal and spectral 'dips' in the interfering sound to achieve a much lower SRT than when steady background noise is used (Duquesnoy, 1983; Festen and Plomp, 1990; Hygge *et al.*, 1992; Baer and Moore, 1994; Moore, Glasberg and Vickers, 1995; Peters, Moore and Baer, 1998; Füllgrabe, Berthommier and Lorenzi, 2006). For them, the SRT when the background is a single talker is 7–18 dB lower than when the background is speech-shaped noise. People with cochlear hearing loss are less able than normally hearing people to take advantage of the temporal and spectral dips. For the former, SRTs are not greatly different for a steady noise background and a single talker background (Duquesnoy, 1983; Festen and Plomp, 1990; Hygge *et al.*, 1992; Peters, Moore and Baer, 1998; Lorenzi *et al.*, 2006a). Hence, when the background is a single talker, the SRT is 12 dB or more higher for people with cochlear damage than for normally hearing people. This represents a very large deficit.

Finally, as described in Chapter 7, people with cochlear hearing loss are less able than normally hearing people to take advantage of spatial separation of the target speech and the interfering sound(s). This can lead to a further elevation in SRT, relative to that found for normally hearing people, of 5–7 dB.

In summary, in some listening situations common in everyday life, such as trying to listen to one person when another person is talking, people with cochlear hearing loss may require speech-to-background ratios 16 dB or more higher than normal (Duquesnoy, 1983). This represents a very substantial problem. However, the majority of laboratory experiments show a less severe problem, as they have used as a background sound steady speech-shaped noise coming from the same direction as the target speech (or presented to the same ear via earphones).

III THE ROLE OF AUDIBILITY

III.1 THE ARTICULATION INDEX (AI) AND SPEECH INTELLIGIBILITY INDEX (SII)

There is no doubt that audibility is crucial for speech intelligibility; if part of the speech spectrum is below the absolute threshold or is masked by background sound, then information is lost, and intelligibility will suffer to some extent. The *Articulation Index* (AI) provides a way of quantifying the role of audibility (French and Steinberg, 1947; Fletcher, 1952; Kryter, 1962; Pavlovic, 1984). In recent work, the term *speech intelligibility index* (SII) has been used instead of AI, but the underlying concepts are similar (ANSI, 1997). The AI or SII are based on the assumption that speech intelligibility is uniquely related to a quantity that, for a normally hearing person, can be calculated from the long-term average spectra of the speech and background sound reaching the ear of the listener. The frequency range from about 200 to 9000 Hz, which is the range that contributes most to intelligibility, is divided into a number of bands. It is assumed that each band makes a certain contribution to speech intelligibility. That contribution is determined by the audibility of the speech in that band and by the relative importance of that band for intelligibility. The overall intelligibility is assumed to be related to a simple sum of the contributions from each band.

For a method using n bands, the SII is defined by the equation:

$$SII = \sum_{i=1}^{n} I_i A_i \tag{8.1}$$

The values of A_i represent the audibility of the different bands. For example, A_9 indicates the proportion of the speech spectrum that is audible in the ninth band. The values of A_i can vary between 0 and 1. The values of A_i depend on the spectra of the speech and the background, and on the level in each band relative to the absolute threshold. The values of I_i represent the relative importance of each band. The values of I_i add up to one. The values of I_i vary depending on the nature of the speech material used, for example whether it is composed of single words or sentences (Pavlovic, 1987). For each band, the value of A_i is multiplied by the value of I_i. This gives an estimate of the contribution of the ith band to intelligibility. The products are then summed across bands, as indicated by the Σ symbol.

It is assumed that the speech in each band covers a 30-dB range of levels when measured using 125-ms samples. The speech peaks, specified as the level exceeded 1 % of the time, have levels about 15 dB above the level corresponding to the root-mean-square (rms) value of the speech. The speech minima have levels about 15 dB below this mean level. A_i is the proportion of the 30-dB dynamic range in the ith band that is above both the absolute threshold and the masked threshold imposed by the background sound. For example, if no noise is present and the rms level of the speech in the ninth band is N dB above the absolute threshold for that band, then the value of A_9 is $(N + 15)/30$. If $N = 10$, then $A_9 = 0.833$. If N is greater than 15 dB, then the speech in that band is fully audible, and the value of A_9 is set to 1.0. If N is less than -15 dB, then the speech is completely inaudible in that band and A_9 is set to 0.

The SII gives a measure of the proportion of the speech spectrum that is audible, but a measure where the relative weighting of different frequencies reflects the relative importance of those frequencies for speech intelligibility. An SII value of 1.0 indicates that *all* of the speech spectrum is audible. For a hearing-impaired person, a hearing aid that resulted in an SII of 1.0 would be completely effective in restoring the audibility of speech. This would not ensure that the speech was intelligible. Indeed, sometimes the application of sufficient amplification to maximize the SII results in lower intelligibility of speech than when less amplification is used (Rankovic, 1991). However, achieving an SII of 1.0 would at least mean that intelligibility was not limited by part of the speech spectrum being below the absolute threshold. If a hearing aid gave an SII value less than 1.0, then intelligibility *might* be improved by increasing the amplification over the frequency range where part of the speech spectrum was inaudible.

To predict actual intelligibility scores, the empirical relationship between the value of the SII and intelligibility must first be determined for the specific speech material used and with a representative group of normally hearing subjects. This gives the intelligibility-articulation transfer function (Fletcher, 1952). Provided the subjects in a given experiment have a similar amount of training to the subjects used to determine the transfer function, the transfer function can then be used to predict intelligibility from the SII. For clearly articulated sentence materials, an SII of 0.7 gives near-perfect (98 %) intelligibility and an SII of 0.5 still gives scores above 90 % correct. For nonsense syllables, an SII of 0.7 gives about 90 % correct, and an SII of 0.5 gives about 70 % correct (ANSI, 1997).

III.2 USE OF THE AI OR SII TO PREDICT SPEECH INTELLIGIBILITY FOR THE HEARING IMPAIRED

Several researchers have examined the question of whether the AI or SII can be used to predict speech intelligibility for hearing-impaired listeners. Obviously, in this case, the absolute thresholds of the individual listeners must be used in calculating the values of A_i. If the AI or SII is successful in this respect, without any modification, this would imply that audibility is the main factor limiting intelligibility. While a few researchers have reported accurate predictions using the unmodified AI (Aniansson, 1974; Lee and Humes, 1993), most studies have shown that speech intelligibility is worse than would be predicted by the AI (Fletcher, 1952; Dugal, Braida and Durlach, 1978; Pavlovic, 1984; Pavlovic, Studebaker and Sherbecoe, 1986; Smoorenburg, 1992), especially for listeners with moderate or severe losses. The predictions are often quite accurate for those with mild losses.

An example of the results of this type of experiment is given in Figure 8.1; the data are from Pavlovic (1984). The speech materials were word lists presented under various conditions of filtering (broadband, lowpass and highpass) either with or without background white noise at a speech-to-noise ratio of 10 dB. Sixteen subjects with

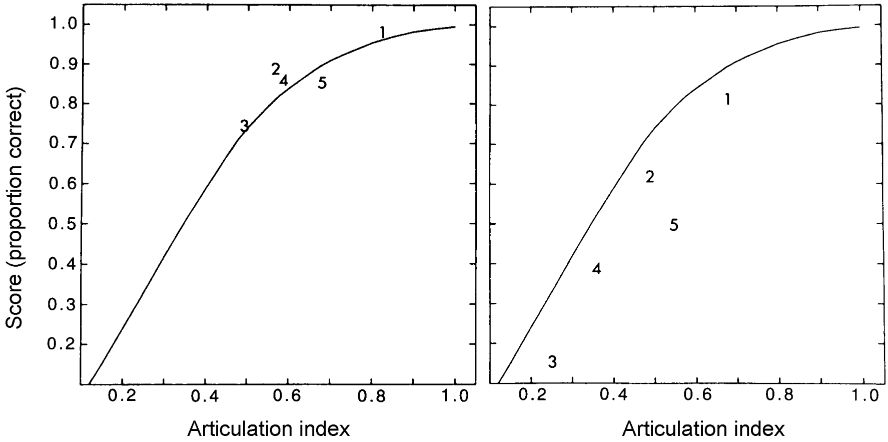


Figure 8.1. Results of Pavlovic (1984) comparing speech-recognition scores of hearing-impaired subjects with predictions based on the AI. Each number represents the mean score across subjects for a specific condition of filtering/background noise. For subjects with mild losses, the predictions are accurate (left panel); for subjects with more severe losses, the obtained scores fall below the predicted values (right panel).

noise-induced hearing loss were used. For the eight subjects with the mildest losses (thresholds better than 50 dB HL at 4000 Hz), the mean scores across subjects for the eight different conditions, indicated by the numbers in the left panel of the figure, are close to the predictions based on the AI (solid curve). For the eight subjects with more severe losses (thresholds 55 dB HL or worse at 4000 Hz), the mean scores, shown in the right panel, fall consistently below the predicted values.

Overall, the results from studies using the AI or SII suggest that, while audibility is of major importance, it is not the only factor involved, at least for people with moderate to severe cochlear losses. Suprathreshold discrimination abilities also need to be taken into account. Additionally, it should be noted that neither the AI nor the SII gives accurate predictions of speech intelligibility under conditions where the background noise is fluctuating (e.g. when there is a single background talker) or when reverberation is present, although a modification of the SII measure to deal with such situations has been proposed (Rhebergen and Versfeld, 2005).

III.3 THE INTELLIGIBILITY OF SPEECH IN NOISE AT HIGH OVERALL LEVELS

Another way of evaluating the importance of audibility is to measure the SRT in noise as a function of overall noise level. If the performance of hearing-impaired subjects is limited by part of the speech spectrum being below absolute threshold, then the SRT, expressed as a speech-to-noise ratio, should decrease progressively with increasing noise level; as the noise level is increased, the speech level has to be increased also, and so more and more of the speech spectrum should lie above absolute threshold. Furthermore, the SRT for hearing-impaired subjects should approach that for normally hearing subjects at sufficiently high noise levels.

One study, using 20 elderly subjects with mild hearing losses (mean threshold about 20 dB HL at medium frequencies, increasing to 40 dB HL at 4000 Hz), reported results consistent with these predictions (Lee and Humes, 1993). However, the results of most other studies have not shown the predicted pattern: the SRT for hearing-impaired subjects remains approximately constant for noise levels above a certain value, and the SRT at high noise levels remains greater for hearing-impaired than for normally hearing subjects (Plomp, 1978, 1986, 1994; Smoorenburg, 1992).

There is, however, a difficulty in interpreting these results. For normally hearing people, the intelligibility of speech in noise decreases somewhat at high overall levels (Fletcher, 1953; Studebaker *et al.*, 1999). This may happen partly because the auditory filters become broader with increasing level, as described in Chapter 3, Section IV.2. The SRTs for hearing-impaired subjects tested at high noise levels have often been compared with SRTs measured at lower noise levels using normally hearing subjects. The discrepancy between the results for the hearing-impaired and the normally hearing subjects would be less if the normally hearing subjects were tested at higher levels. However, the effect of level is probably not sufficient to explain all of the difference in SRTs between normal and hearing-impaired listeners. Overall, these results lead to the same conclusions as those derived from studies using the AI and the SII. For people with moderate to severe losses, factors other than audibility contribute to their relatively poor ability to understand speech in noise.

III.4 COMPARISON OF DETECTION AND RECOGNITION FOR SPEECH IN NOISE

Turner *et al.* (1992) measured psychometric functions (percentage correct as a function of signal level) for the detection and recognition of stop consonants (p, t, k, b, d, g) followed by the vowel ‘a’ (as in car), presented in white noise. They used two normally hearing subjects and three subjects with ‘flat’ moderate cochlear loss. They found that *detection* of the consonants occurred at the same speech-to-noise ratio for the normal and hearing-impaired subjects. However, for a given level of *recognition*, the hearing-impaired subjects required higher speech-to-noise ratios. They concluded that ‘the poorer-than-normal speech recognition in noise exhibited by some hearing-impaired listeners is not due to a deficit in detecting the speech signal in noise. Instead, their poorer speech-recognition performance in noise is due to their inability to efficiently utilize audible speech cues.’ Some caution is needed, however, in drawing strong conclusions from these results. Detection can be based on audibility of a very limited part of the spectrum of the speech, but recognition of speech requires information to be audible over a wider frequency range.

III.5 THE INTELLIGIBILITY OF SPEECH IN QUIET AT HIGH OVERALL LEVELS

As described earlier, hearing-impaired people with mild-to-moderate cochlear hearing losses do not generally have difficulty in understanding connected discourse in a quiet non-reverberant room. However, they may have some difficulty for isolated nonsense syllables. Subjects with severe losses can have considerable difficulty with speech in quiet. Turner and Robb (1987) tried to determine whether this difficulty could be explained in terms of audibility. They studied the identification of synthetic consonant–vowel syllables, composed of one of the six stop consonants (p, t, k, b, d, g) followed by the vowel 'a' (as in car). A model of filtering in the peripheral auditory system was used to estimate the portion of the speech spectrum that was above the threshold of audibility for a given presentation level. Several presentation levels were used for each subject. They tested four normally hearing subjects and five subjects with moderate-to-severe hearing losses.

For the normally hearing subjects, recognition improved monotonically as the audible portion of the stimulus increased, and performance was perfect when most of the spectrum was above threshold. For four out of the five hearing-impaired subjects, recognition scores were consistently lower than for the normally hearing subjects at a given level of audibility. For these subjects, scores remained below 100 %, even in cases where audibility was close to 100 %. For the remaining subject, the hearing loss at low frequencies was too severe for even 50 % audibility to be achieved. Overall, these results strongly suggest that reduced audibility is not sufficient to explain the relatively poor consonant recognition of the hearing-impaired subjects. It is noteworthy, however, that even presentation levels of 100 dB SPL were not sufficient to provide 100 % audibility for subjects with severe losses (although appropriate frequency-dependent amplification could ensure 100 % audibility at overall levels below 100 dB SPL).

III.6 SIMULATION OF HEARING LOSS BY SELECTIVE FILTERING (FREQUENCY-DEPENDENT ATTENUATION)

Yet another approach to studying the effects of audibility is to subject the stimuli to frequency-dependent attenuation (filtering) so as to imitate the effect of a hearing loss on audibility. Say, for example, that a hearing-impaired subject has absolute thresholds of 30, 40, 45, 50, 70 and 70 dB HL at 250, 500, 1000, 2000, 4000 and 6000 Hz, respectively. The filter would be designed to attenuate by 30 dB at 250 Hz, 40 dB at 500 Hz, 45 dB at 1000 Hz, and so on. The filtered stimuli are presented to normally hearing subjects. The filtering has the effect of matching the audiogram for the impaired ear (listening to unfiltered stimuli) and for the normal ears (listening to filtered stimuli). If audibility is the main cause of difficulty in speech understanding experienced by the hearing-impaired subject, then performance should be similar for the hearing-impaired subject listening to unfiltered speech and normally hearing subjects listening to filtered speech.

Only a few studies have used this approach. Fabry and van Tasell (1986) used six subjects with unilateral cochlear loss, filtering the stimuli to match the audiograms of the normal and impaired ears. Several of the subjects had absolute thresholds that were normal or near-normal at low frequencies, but increased markedly at high frequencies. They measured the identification of consonant–vowel nonsense syllables in quiet. For two subjects, scores were similar for unprocessed stimuli presented to the impaired ear and filtered stimuli presented to the normal ear. Patterns of errors were also similar for the two ears. For three subjects, performance was better for the impaired than for the normal ears, by an average of about 8 %. This may have happened because of the unfamiliarity of the filtered stimuli presented to the normal ears. Another possibility is that, for speech in quiet, loudness recruitment can actually have a small beneficial effect by increasing the loudness of sounds presented at low sensation levels (SLs) (Gatehouse and Haggard, 1987). For the remaining subject, the score was 11 % worse for the impaired than for the normal ear.

Unfortunately, there do not seem to have been any studies of this type examining the intelligibility of speech in noise. Thus, it is not known whether selective filtering to match the audibility of the speech in noise for normal and hearing-impaired listeners would produce equal performance for the two groups.

III.7 SIMULATION OF HEARING LOSS BY MASKING

Another approach uses masking noise with normal listeners to simulate the effects of the elevated thresholds associated with hearing impairment. The noise is spectrally shaped so that the masked audiograms of the normal listeners match the unmasked audiograms of the hearing-impaired listeners. This technique has been used by a number of different investigators (Fabry and van Tasell, 1986; Humes, Dirks and Kincaid, 1987; Zurek and Delhorne, 1987; Humes and Roberts, 1990; Dubno and Schaefer, 1992; Bacon, Opie and Montoya, 1998). In some studies, speech intelligibility performance for the noise-masked normal ears has been reasonably close to that of the hearing-impaired subjects whose absolute thresholds were simulated. However, this was not the case for a study performed by Needleman and Crandell (1995). They measured SRTs for sentences in noise using ten subjects with mild-to-moderate sensorineural hearing loss. Two normally hearing subjects were matched with each hearing-impaired subject, using spectrally shaped noise. The noise-masked normal subjects obtained significantly lower (better) SRTs than the hearing-impaired subjects. This suggests that factors other than reduced audibility were responsible for the relatively high SRTs of the hearing-impaired subjects.

A study by Bacon, Opie and Montoya (1998) also revealed the presence of suprathreshold deficits. They compared performance for hearing-impaired subjects and normally hearing subjects listening in a background of noise to simulate hearing impairment. SRTs were measured in steady and modulated background noises. For six out of eleven hearing-impaired listeners, the difference in SRTs for the steady and modulated backgrounds was less than for the noise-masked normal-hearing group. In other words, several of the hearing-impaired subjects showed a reduced ability to take advantage of dips in the background sound, even though audibility for the two groups was equated.

Two problems with the technique of noise masking should be noted. First, the technique is limited to the simulation of mild-to-moderate hearing losses; the simulation of severe losses would require unacceptably loud noise. Secondly, the technique does not simulate the loss of audibility alone. The noise also produces an effect resembling loudness recruitment (see Chapter 4, Section III). Indeed, some researchers have suggested that noise may be regarded as simulating the combined effects of threshold elevation and loudness recruitment (Humes and Roberts, 1990). However, the nature of the recruitment produced by background noise is probably different from that produced by cochlear hearing loss (Phillips, 1987). More generally, the background noise may affect not only detection thresholds but also the perception of suprathreshold sounds. This idea is supported by the observation that using masking noise to match the audiograms of normal and impaired listeners has greater effects on speech intelligibility than using frequency-dependent attenuation (Fabry and van Tasell, 1986).

III.8 CONCLUSIONS ON THE ROLE OF AUDIBILITY

Taken together, the results reviewed above strongly suggest that one or more factors other than audibility contribute to the difficulties of speech perception experienced by those with moderate or greater cochlear losses. This is especially true in situations where the stimuli are presented at high levels and/or in background noise, and when the background is fluctuating. In other words, the difficulties arise partly from abnormalities in the perception of sounds that are above the threshold of audibility. For those with mild losses, audibility is probably the dominant factor.

IV INFLUENCE OF DEAD REGIONS ON SPEECH PERCEPTION

There have been several reports suggesting that people with moderate-to-severe hearing loss at high frequencies often do not benefit from amplification of high frequencies, or even perform more poorly when high frequencies are amplified (Moore, Laurence and Wright, 1985; Murray and Byrne, 1986; Ching, Dillon and Byrne, 1998; Hogan and Turner, 1998; Turner and Cummings, 1999; Amos and Humes, 2001). It was suspected that at least some of the subjects tested had high-frequency *dead regions*, but no test for the presence of dead regions was conducted in those studies.

In two recent studies, subjects were tested for the presence of dead regions using both psychophysical tuning curves (PTCs) and the *threshold-equalizing noise* (TEN) test (Vickers, Moore and Baer, 2001; Baer, Moore and Kluk, 2002). All subjects had high-frequency hearing loss, but some had high-frequency dead regions and some did not. The edge frequency of a dead region is denoted f_e. Generally, the subjects with dead regions had more severe high-frequency hearing losses than those without dead regions. The speech stimuli were vowel–consonant–vowel (VCV) nonsense syllables. Prior to presentation via earphones, the stimuli were subjected to the frequency-dependent amplification prescribed by the Cambridge formula (Moore and Glasberg, 1998); see Chapter 9, Section II.2 for more details of this. The goal of the amplification was to restore audibility as far as possible, while avoiding excessive loudness. Subjects were tested using broadband speech (upper frequency limit 7500 Hz) and speech that was lowpass filtered with various cutoff frequencies.

In one study (Vickers, Moore and Baer, 2001), all stimuli were presented in quiet. In another study (Baer, Moore and Kluk, 2002), the stimuli were presented in steady speech-shaped noise, using a speech-to-noise ratio chosen to give a moderate reduction in performance relative to that measured in quiet. The general pattern of results was similar for the two studies. For subjects without dead regions, performance improved progressively with increasing cutoff frequency. This indicates that these subjects were able to make use of high-frequency information. For subjects with dead regions, scores tended to increase with increasing cutoff frequency up to $1.5f_e$ to $2f_e$. For higher cutoff frequencies, scores remained roughly constant or decreased slightly (not significantly). These results indicate that the subjects with dead regions extracted little or no information from frequencies that were well inside the dead region.

Vestergaard (2003) presented data broadly consistent with the findings summarized above. He tested experienced users of hearing aids, using the TEN test to diagnose dead regions. He measured speech intelligibility using lowpass filtered word lists. Subjects listened through their own hearing aids, all of which provided substantial amplification at frequencies where there was hearing loss. For subjects with no dead region or high-frequency dead regions starting at 3 kHz or above, performance tended to improve progressively with increasing cutoff frequency. For subjects with high-frequency dead regions starting in the range 0.75–1.5 kHz, the pattern was less consistent, but the mean scores showed no clear change with increasing cutoff frequency above 1 kHz.

Mackersie, Crocker and Davis (2004) compared speech intelligibility for two groups of subjects, with and without high-frequency dead regions, diagnosed using the TEN test. The audiometric thresholds were matched for the two groups. Subjects were tested while wearing a hearing aid which provided substantial gain in frequency regions of hearing loss. Speech identification scores were obtained for unfiltered stimuli and stimuli that were lowpass filtered at f_e, $1.41f_e$ and $2f_e$. Filter settings for the ears without suspected dead regions were the same as settings for the threshold-matched counterparts. In quiet and in low levels of noise (signal-to-noise ratios close to 15 dB), scores were significantly higher for the wideband (unfiltered) stimuli than for the filtered stimuli, and performance was similar for the ears with and without dead regions. In high levels of noise (signal-to-noise ratios close to 0 dB), mean scores were highest for the wideband speech for the ears without dead regions, but performance reached an asymptote at a lower cutoff frequency for the ears with dead regions. In other words, when substantial noise was present, the subjects with dead regions did not benefit from amplification of frequency components well inside the dead region.

Overall, these results support the idea that subjects with extensive high-frequency dead regions do not make as effective use of audible speech information at high frequencies as subjects without such dead regions.

Recently, Vinay and Moore (2007b) measured the recognition of highpass filtered nonsense syllables as a function of filter cutoff frequency for hearing-impaired subjects with *low*-frequency hearing loss. Sixteen of the subjects had low-frequency (apical) cochlear dead regions and twelve did not. The diagnosis of any dead region was made using the TEN(HL) test (Moore, Glasberg and Stone, 2004), and PTCs were used to define the value of f_e more precisely. Stimuli were amplified differently for each ear, using the Cambridge formula (see Chapter 9, Section II.2). The mean results for subjects without dead regions are shown in the upper panel of Figure 8.2. Scores were high (about 78 %) for low cutoff frequencies, remained approximately constant for cutoff frequencies up to 862 Hz and then worsened with increasing cutoff frequency. For subjects with low-frequency dead regions, performance was typically

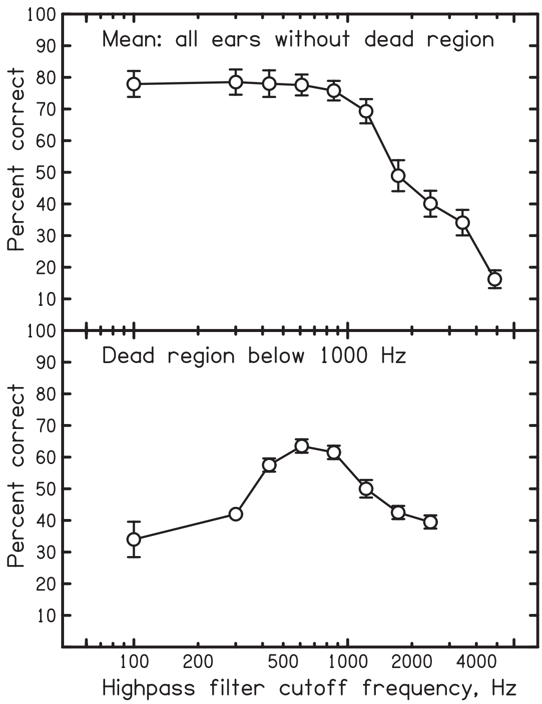


Figure 8.2. Results of Vinay and Moore (2007b) showing the intelligibility of nonsense syllables measured as a function of the cutoff frequency of a highpass filter for subjects with low-frequency hearing loss. The upper panel shows mean results for subjects without any dead region. The lower panel shows an example of the results for a subject with a low-frequency dead region with f_e = 1000 Hz.

poor for the lowest cutoff frequency (100 Hz), improved as the cutoff frequency was increased to about $0.57f_e$ and worsened with further increases. The lower panel in Figure 8.2 shows an example of the results for a subject with f_e equal to 1000 Hz. These results indicate that people with low-frequency dead regions are able to make effective use of frequency components that fall in the range $0.57f_e$–f_e, but that frequency components below $0.57f_e$ have deleterious effects.

Overall, the ability to understand speech is usually poor for subjects with extensive dead regions (whether at low frequencies or high frequencies), even for speech in quiet.

V CORRELATION BETWEEN PSYCHOACOUSTIC ABILITIES AND SPEECH PERCEPTION

Assuming that the speech perception difficulties of people with moderate or greater cochlear losses are partly caused by abnormalities in the perception of sounds that are above the threshold for detection, the question naturally follows: 'What psychoacoustical factors have the most influence on speech intelligibility?' One method for studying this problem is to examine statistical relationships between measures of these abilities in hearing-impaired subjects. Typically, subjects have been given a battery of psychoacoustical and speech-perception tests. Ideally, a large number of subjects should be tested (for statistical validity), the subjects should be given practice on each test until their performance is stable and sufficient data should be gathered to give a reliable and accurate score on each test. Unfortunately, this would take an enormous amount of time and, in practice, compromises are inevitable. Either a large number of subjects can be tested, but with little time for practice on each test or for the execution of each test, or fewer subjects can be tested, with each given more training and more time per test.

Several researchers have conducted correlational studies (Dreschler and Plomp, 1980, 1985; Patterson *et al.*, 1982; Tyler *et al.*, 1982; Festen and Plomp, 1983; Horst, 1987; Glasberg and Moore, 1989; van Rooij and Plomp, 1990; Lutman, 1991). Most studies have measured SRTs in noise and have assessed whether the SRTs can be 'accounted for' (in a statistical sense) by performance on various psychoacoustic tests. If the results of a given psychoacoustic test are highly correlated with the results of a measure of speech perception, then one might argue that the variations in performance on the speech-perception test can be 'accounted for' by the variations on the psychoacoustic test. On the other hand, if performance on the two tests has a very low correlation, then some factor(s) other than that specific psychoacoustical measure must account for variability in the results of the speech-perception test. Of course, a correlation between two measures does not necessarily imply that one causes the other.

The correlational studies indicate that a substantial part of the variability in SRTs is not explained by absolute thresholds. Most studies also agree that suprathreshold abilities such as frequency selectivity (Horst, 1987) or temporal resolution (Tyler *et al.*, 1982) can account for a significant proportion of the variance in the SRTs.

However, the effects of these variables are difficult to separate from the effects of absolute threshold.

Glasberg and Moore (1989) tested nine subjects with unilateral cochlear hearing loss; performance using the normal ear acted as a control for performance using the impaired ear, and the normal ear was compared with the impaired ear both at equal SL and at equal SPL. One possible drawback of using such subjects is that the impaired ear may become 'lazy' or 'neglected' in some sense, giving poorer discrimination abilities than might be expected from the pure-tone audiogram (Hood, 1984; Moore *et al.*, 1997). In addition, the pathologies producing unilateral loss may not be typical of hearing losses in general. Hence, Glasberg and Moore (1989) also tested six subjects with bilateral losses of cochlear origin. They measured SRTs in quiet and in two levels (60 and 75 dB SPL) of a noise with the same long-term average spectrum as the speech. They also measured performance on a variety of psychoacoustic tests, which were generally conducted at three centre frequencies: 0.5, 1.0 and 2.0 kHz. Most subjects had absolute thresholds for their impaired ears in the range 40–60 dB HL at these frequencies, but a few had thresholds as low as 30 dB HL or as high as 80 dB HL at one of the test frequencies.

The SRTs were higher for the impaired than for the normal ears, both in quiet and in noise. The elevation in quiet could be largely accounted for by the higher absolute thresholds for the impaired ears. Taking the results for all ears together, the SRT in quiet was highly correlated with the mean absolute threshold at 0.5, 1 and 2 kHz ($r = 0.96$, $p<0.001$). For the impaired ears only, the corresponding correlation was $r = 0.82$ ($p < 0.001$).

The level of the 75-dB noise was sufficient to raise the SRTs well above those measured in quiet, and, at this noise level, the SRTs were higher for the impaired ears than for the normal ears. The mean SRT in the 75-dB noise, expressed as a speech-to-noise ratio, was -0.9 dB for the normal ears. For the impaired ears of the subjects with unilateral impairments, the corresponding mean SRT was 7.1 dB. Thus, the impaired ears required, on average, an 8 dB higher speech-to-noise ratio to achieve 50 % intelligibility. For the subjects with bilateral impairments, the mean SRT was 1.3 dB, which is 2.2 dB higher than normal. Taking the results for all ears together, the SRTs were correlated with the mean absolute threshold at 0.5, 1 and 2 kHz ($r = 0.74$, $p < 0.001$). For the impaired ears only, the correlation reduced to $r = 0.56$ ($p < 0.01$). These correlations are lower than those found between the SRTs in quiet and the absolute thresholds. This indicates that, for noise levels sufficient to raise the SRT well above that measured in quiet, a significant proportion of the variance in the SRTs was not accounted for by variations in absolute threshold.

While the SRTs in the 75-dB noise were correlated with the mean absolute thresholds at 0.5, 1 and 2 kHz, they were more highly correlated with three measures of suprathreshold discrimination. These measures were the frequency discrimination of pure tones, the frequency discrimination of complex tones and the threshold for detecting temporal gaps in bands of noise. The higher correlations were found both for the results of all ears and for the impaired ears only. This is consistent with the hypothesis that, in high levels of background noise, the ability to understand speech is determined more by suprathreshold discrimination abilities than by absolute sensitivity.

Ching, Dillon and Byrne (1997) attempted to determine the importance of psychoacoustical factors for speech recognition after the effects of audibility had been taken into account. To do this, they presented speech in quiet over a wide range of levels and under various conditions of filtering. For each level and condition, the SII was calculated and the number of key words in sentences that were correctly identified was measured. The results for eight normally hearing subjects were used to determine the transfer function relating the speech scores to the values of the SII. Twenty-two hearing-impaired subjects were tested. The speech scores for these subjects were expressed as *deviations* from the values predicted from the SII. These deviations represent the extent to which speech scores are better or worse than expected for a given amount of audibility of the speech. The deviations of the speech scores from the predicted values at high SLs were significantly correlated with a measure of frequency selectivity at 2 kHz (obtained using a notched-noise masker) and with a measure of temporal resolution (Zwicker, 1980).

Overall, the results of studies using the correlational approach have been somewhat inconclusive. They have provided evidence that the intelligibility of speech in noise for subjects with moderate or greater cochlear losses is influenced by suprathreshold discrimination abilities, but they have not provided a clear indication of the relative importance of the various psychoacoustic factors that might be involved.

VI ASSESSING THE EFFECTS OF FREQUENCY SELECTIVITY ON VOWEL AND CONSONANT PERCEPTION

Some researchers have attempted to relate measures of frequency selectivity in specific frequency regions to the intelligibility of speech sounds for which important information is carried in those frequency regions. This more focused approach can yield useful insights into the importance of frequency selectivity for speech perception.

VI.1 CONSONANT PERCEPTION

Preminger and Wiley (1985) measured PTCs for a 500-Hz signal and a 4000-Hz signal in subjects with cochlear hearing loss of various configurations. Losses were classified as high-frequency, flat or low-frequency; there were two subjects in each group. The test stimuli were consonant–vowel syllables which were categorized into three groups on the basis of the predominant spectral energy in the consonant parts of the sounds; the three groups were low frequency (e.g. w, b, m, l), high frequency (e.g. t, d, k, z, s) and diffuse (e.g. v, f). The subjects with high-frequency loss had broadened PTCs at 4000 Hz, but normal PTCs at 500 Hz. These subjects achieved higher performance for low-frequency consonants than for high-frequency consonants. The subjects with flat hearing losses showed almost no frequency selectivity at 4000 Hz (the PTCs were 'flat') and they both performed poorly at identifying high-frequency consonants. One subject also had a very broad PTC at 500 Hz, and that subject performed poorly at identifying low-frequency consonants. The other subject showed some tuning at 500 Hz and achieved better identification of low-frequency consonants. For the subjects with low-frequency loss, the relation between PTC tuning and consonant identification was not so clear.

Thibodeau and van Tasell (1987) estimated frequency selectivity at 2000 Hz by measuring the percentage-correct detection of a 2000-Hz sinusoid as a function of the width of a spectral notch in a noise, the notch being centred at 2000 Hz. Two normally hearing subjects and seven subjects with moderate flat sensorineural (probably cochlear) losses were used. They also measured the ability to discriminate two synthetic stop-vowel syllables, di and gi, which were identical except for spectral transitions (glides in the *formant* frequencies; see below for a description of formants) in the range 1800–2500 Hz. These spectral transitions provided the information necessary to distinguish the two consonants. Discrimination was measured both in broadband noise and in noise with a spectral notch around 2000 Hz. There was a significant correlation between scores on the two tasks; subjects with poorer frequency selectivity were also poorer at discriminating the syllables.

These results indicate that effects of reduced frequency selectivity on consonant discrimination can be observed when the information in the speech is restricted to a certain frequency range. Poor frequency selectivity is associated with poor consonant discrimination. However, natural speech contains information over a wide frequency range, making it much harder to show specific effects of reduced frequency selectivity.

VI.2 VOWEL PERCEPTION

Vowel sounds are characterized by peaks in their spectra at certain frequencies. These peaks correspond to *resonance*s in the vocal tract. They are called formants, and are numbered, the lowest in frequency being called the first formant (F1), the next the second formant (F2) and so on. The pattern of frequencies at which the formants occur is thought to play an important role in determining the perceived vowel identity. The first three formants are the most important ones. The *excitation pattern* evoked by a vowel can be regarded as an internal representation of the spectral shape of the vowel. This is illustrated in Figure 8.3. The top panel shows the spectrum of a synthetic vowel 'i' (as in 'bead') on a *linear* frequency scale, while the middle panel shows the same spectrum on an *ERB_N-number scale* (see Chapter 3, Section IV.1). The bottom panel shows the excitation pattern for the vowel plotted on an ERB_N-number scale. This pattern was calculated as described in Chapter 3, Section IV.2 for a normal auditory system.

Several aspects of the excitation pattern are noteworthy. First, the lowest three peaks in the excitation pattern do not correspond to formant frequencies but rather to individual lower harmonics; these harmonics are resolved in the normal peripheral auditory system and can be heard out as separate tones under certain conditions, as described in Chapter 3, Section VII. Hence, the centre frequency of the first formant is

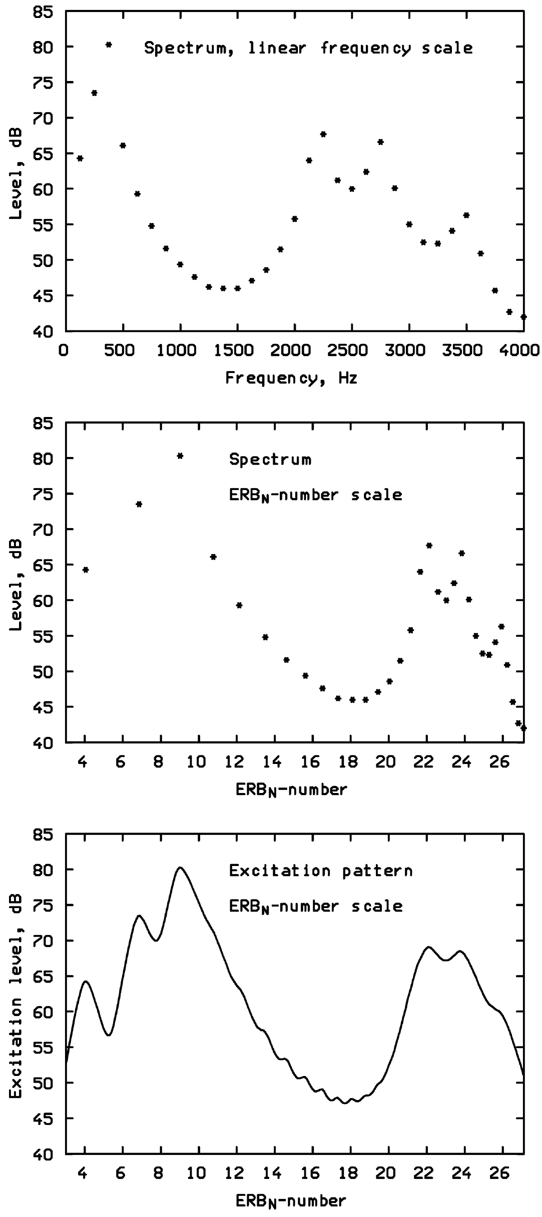


Figure 8.3. The top panel shows the spectrum of the synthetic vowel 'i' as in 'bead', plotted on a linear frequency scale. The middle panel shows the spectrum plotted on an ERB_N-number scale. The bottom panel shows the excitation pattern for that vowel, calculated for a normal ear, and plotted on an ERB_N-number scale.

not directly represented in the excitation pattern; it must be inferred from the relative levels of the peaks corresponding to the individual lower harmonics.

A second noteworthy aspect of the excitation pattern is that the second, third and fourth formants, which are clearly separately visible in the original spectrum, are not well resolved. Rather, the second and third formants form a single prominence in the excitation pattern, with only a small dip to indicate that two formants are present. The fourth formant appears as a small 'shoulder' on the upper side of this prominence.

Figure 8.4 shows excitation patterns for the same vowel, but calculated for impaired auditory systems in which the auditory filters were assumed to have equivalent rectangular bandwidths (ERBs) twice as great as normal (top panel) or four times as great as normal (bottom panel); these degrees of broadening are typical of moderate and severe cochlear hearing losses, respectively. It is obvious that spectral details are less well represented in these excitation patterns; the peaks and dips corresponding to the lower harmonics are less clear, and the second, third and fourth formants are represented by only a single broad prominence. Indeed, in the lower panel, there is only a small, 2–3 dB, peak to indicate the presence of the second, third and fourth formants.

Given these considerations, one might expect that vowel identification would be less accurate in hearing-impaired subjects than in normally hearing subjects, and that vowel identification would worsen progressively with decreasing frequency selectivity. There are at least some data to support this idea. Van Tasell, Fabry and Thibodeau (1987) measured confusions among seven synthetic steady state vowels for ten normally hearing subjects and three subjects with cochlear hearing loss. Confusions were greater for the hearing-impaired than for the normal subjects. They also used some of the vowels as forward maskers; the threshold of a brief sinusoidal signal was measured at each harmonic frequency. These vowel masking patterns (VMPs) can be used to infer the internal representation of the vowels, including the effects of *suppression* (see Chapter 3, Sections VI.2 and VI.3). The VMPs of the impaired subjects showed poorer preservation of the vowels' formant structures than did the VMPs of the normal subjects. Vowel identification accuracy appeared to be related mainly to the positions of peaks in the VMPs rather than to the levels of peaks or between-peak characteristics.

Turner and Henn (1989) also compared vowel identification with measures of frequency selectivity for two normally hearing subjects and three subjects with cochlear losses. For each of several frequencies of a sinusoidal signal, they measured the signal threshold in forward masking for a sinusoidal masker of variable frequency but fixed level of 95 dB SPL. The masking results were used to calculate the excitation patterns evoked by the vowels. They suggested that, the more similar the excitation patterns of two vowels, the more often they would be confused. This suggestion received some support when the measure of similarity was based on the entire shape of the calculated excitation patterns. However, it received stronger support when they used a measure that emphasized the similarities of spectral peak locations.

Although these experiments support the idea that reduced frequency selectivity can adversely affect vowel identification, everyday experience and studies using natural

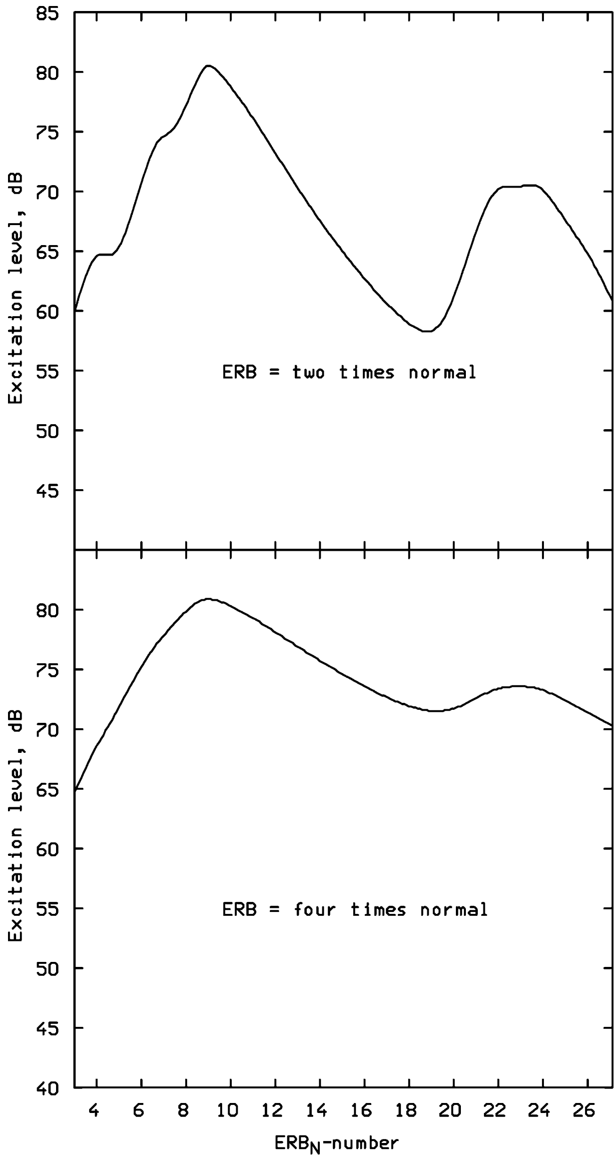


Figure 8.4. The excitation pattern for the same vowel as in Figure 8.3 but calculated for an impaired ear with auditory filters two times broader than normal (top panel) or four times broader than normal (bottom panel).

speech indicate that vowel identification by subjects with moderate cochlear hearing loss is often rather good. This may happen for at least two reasons. First, the spectral differences between vowels are often very large, so that frequency selectivity has to be grossly impaired to prevent the differences being detected. Secondly, naturally produced vowels contain temporal cues (such as duration), as well as spectral cues, and these cues may be used to compensate for the effects of reduced frequency selectivity.

VII INFLUENCE OF LOSS OF SENSITIVITY TO TEMPORAL FINE STRUCTURE

In Chapters 6 and 7, evidence was presented that people with cochlear hearing loss often have a reduced sensitivity to the temporal fine structure of sounds. Here, the possible influence of this on speech perception is considered. When a complex broadband sound such as speech is analysed in the cochlea, the result is a series of bandpass-filtered signals, each corresponding to one position on the basilar membrane (BM). Each of these signals contains two forms of information: fluctuations in the envelope (the relatively slow variations in amplitude over time) and fluctuations in the temporal fine structure (the rapid oscillations with rate close to the centre frequency of the band). This is illustrated in Figure 8.5. The role of envelope and temporal fine structure cues in speech perception has been studied by filtering the speech into several contiguous frequency bands (each like one of the bands shown in Figure 8.5) and processing the signal in each band so as to preserve only envelope or temporal

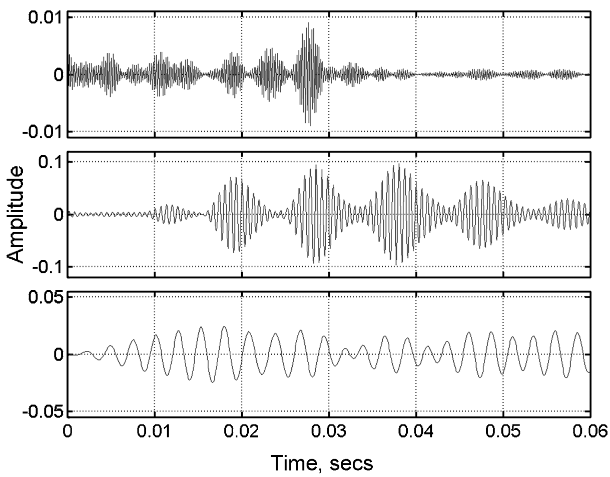


Figure 8.5. Outputs of 1-ERB$_N$-wide bandpass filters centred at 369, 1499 and 4803 Hz (from bottom to top) in response to a segment of a speech signal. For each centre frequency, the signal can be considered as a slowly varying envelope imposed on a more rapid temporal fine structure.

fine structure cues. Envelope cues alone can be presented by deriving the envelope in each band and using the envelope to amplitude modulate a noise band or sinusoid centred at the same frequency as the band from which the envelope was derived (Dudley, 1939). With a moderate number of bands (4–16), envelope cues alone can lead to a high intelligibility for speech in quiet both for normally hearing people (Shannon *et al.*, 1995; Loizou, Dorman and Tu, 1999) and for hearing-impaired people (Turner, Souza and Forget, 1995; Souza and Boike, 2006; Baskent, 2006). However, for normally hearing listeners, the intelligibility of speech based on envelope cues alone is very poor when a fluctuating background sound such as a single talker or an amplitude-modulated noise is present (Nelson *et al.*, 2003; Qin and Oxenham, 2003; Stone and Moore, 2003a); presumably, the same would apply for hearing-impaired listeners. This suggests that envelope cues alone do not allow effective listening in the dips of a background sound.

To effectively make use of the dips in a fluctuating background sound, the auditory system must have some way of determining whether the signal that is present during the dips is dominated by the target speech or by the background sound. The normal auditory system may do this by using information derived from neural phase locking to the temporal fine structure; changes in phase locking of auditory nerve discharges when a dip occurs indicate that the target speech is present in the dip. The difficulties experienced by hearing-impaired people when trying to listen in the dips may reflect a loss of ability to extract or use information from the temporal fine structure.

Lorenzi *et al.* (2006b) studied the role of temporal fine structure cues in speech perception by processing speech so as to remove envelope cues as far as possible while preserving temporal fine structure cues. They did this by filtering the signal into sixteen contiguous frequency bands, extracting the envelope in each band and dividing the signal in each band by the envelope (Smith, Delgutte and Oxenham, 2002). The resulting signal in each band had a constant envelope amplitude, but a time-varying temporal fine structure. The band signals were then recombined. For comparison, they also measured performance with unprocessed speech and speech processed to preserve envelope cues while removing temporal fine structure cues (using the processing described earlier).

The ability to identify nonsense syllables presented in quiet was measured for three groups of subjects: normally hearing, and young and elderly with mild-to-moderate flat hearing loss. For the intact speech, the normally hearing group achieved perfect scores and both groups of hearing-impaired subjects performed nearly as well as the normally hearing group. After moderate training, all groups achieved high scores (about 90 % correct) for the speech with envelope cues. After more extensive training, the normally hearing group also achieved about 90 % correct for the speech with mainly temporal fine structure cues. However, both hearing-impaired groups performed much more poorly, with most subjects scoring close to the level that would be expected from the use of cues such as overall duration. These results indicate that moderate hearing loss causes a dramatic deterioration in the ability to use temporal fine structure cues for speech perception.

In a second experiment, Lorenzi *et al.* (2006b) tested only the subjects from the young hearing-impaired group. The stimuli were intact (unprocessed) nonsense syllables presented in a steady background noise and in a background noise that was sinusoidally amplitude modulated at an 8-Hz rate with a 100 % depth. The speech-to-noise ratio was fixed individually at the level yielding about 50 % correct identification for the speech in steady noise. The difference in scores for the two types of noise provides a measure of the ability to listen in the dips of the modulated background, called masking release (Füllgrabe, Berthommier and Lorenzi, 2006). The amount of masking release was found to be highly correlated ($r = 0.83$) with scores obtained in the first experiment using speech processed to preserve mainly temporal fine structure cues while removing temporal envelope cues. In other words, subjects who performed relatively well when listening to speech in quiet containing mainly temporal fine structure cues showed a relatively large masking release when listening to intact speech. However, the amount of masking release for these subjects was markedly smaller than is typically found for normally hearing subjects under similar conditions (Gustafsson and Arlinger, 1994; Füllgrabe, Berthommier and Lorenzi, 2006). Overall, the pattern of the results supports the hypothesis that listening in the dips depends on the use of temporal fine structure information, and that the greatly reduced ability of hearing-impaired subjects to listen in the dips is partly a result of the loss of ability to use temporal fine structure cues.

It remains unclear whether the reduced ability of hearing-impaired subjects to make use of temporal fine structure reflects reduced precision of phase locking or a reduced ability to extract information from the patterns of phase locking. For a complex signal like speech, the broadened auditory filters of hearing-impaired subjects result in temporal fine structure at a given centre frequency that is more complex than normal, and that varies more rapidly than normal. This in itself may make it more difficult to use the temporal fine structure information, since, as described in Chapter 6 (Section III.1), the mechanism that 'decodes' the temporal fine structure information appears to be 'sluggish' and cannot track rapid changes in frequency (Moore and Sek, 1996).

VIII THE USE OF SIMULATIONS TO ASSESS THE IMPORTANCE OF PSYCHOACOUSTIC FACTORS IN SPEECH PERCEPTION

Another approach to assessing the importance of suprathreshold discrimination abilities for speech perception is to simulate the effect of one specific aspect of hearing impairment by processing the stimuli in a way that mimics the effect of this aspect. The processed stimuli are then used in tests with normally hearing subjects. Provided the simulation is accurate, this makes it possible to study the effect of that aspect in isolation. This general strategy has been used to study the importance of three psychoacoustical factors: loudness recruitment combined with threshold elevation, frequency selectivity and temporal resolution.

VIII.1 SIMULATIONS OF LOUDNESS RECRUITMENT COMBINED WITH THRESHOLD ELEVATION

The effects of loudness recruitment combined with threshold elevation can be simulated by splitting the signal into a number of frequency bands and expanding the range of levels in each band before recombining the bands. Effectively, this simulation does the opposite of what happens in a normal cochlea, where stimuli are filtered along the BM, and fast-acting compression is applied at each characteristic frequency (see Chapter 1, Section IV.3). The goal of the simulation is to process the level of the stimulus in each band, on a moment-by-moment basis, so as to create loudness sensations in a normal ear that would resemble those produced in an impaired ear with recruitment. Say, for example, it is desired to simulate a hearing loss where the absolute threshold for a centre frequency of 1 kHz is 50 dB SPL, and 'normal' loudness is reached at 100 dB SPL. If the momentary level in the frequency band around 1 kHz is 50 dB SPL, then the signal in that band is attenuated by about 50 dB, so that its momentary level is close to the absolute threshold for normal hearing, that is roughly 0 dB SPL. If the momentary level is 100 dB or higher, no attenuation is applied; the signal level is unchanged. If the momentary level is at an intermediate value, say 75 dB SPL, then it is attenuated by an intermediate amount, 25 dB in this example.

Villchur (1974) used a three-band system to process speech so as to simulate the effects of recruitment associated with severe hearing loss. The stimulus in each band was processed using a fast-acting expander. Subjects with unilateral hearing impairments judged processed stimuli presented to their normal ear to sound 'similar' or 'very similar' to unprocessed stimuli presented to their impaired ear. The intelligibility of the processed speech was not measured in formal tests. However, Villchur concluded that recruitment is a sufficient cause for loss of intelligibility in cases of severe hearing loss.

In a later study (Villchur, 1977), he used a 16-channel system, with computer-controlled attenuators to achieve expansion of the range of levels in each band. A severe sloping hearing loss was simulated. The intelligibility of speech in quiet at a level of 94 dB was adversely affected by the processing, both for isolated words and for sentences. The intelligibility of words and sentences in white noise was also adversely affected by the processing, the percentage correct decreasing from about 69 to 50 %. An even greater effect was found for speech in speech-shaped noise.

Duchnowski and Zurek (Duchnowski, 1989; Duchnowski and Zurek, 1995) used digital signal processing to implement a 14-band system. In one experiment, the processing was adjusted to simulate the hearing losses of the subjects tested by Zurek and Delhorne (1987); these subjects had mild-to-moderate hearing losses. The test stimuli were consonant–vowel (CV) syllables, presented in quiet and in various levels of speech-shaped noise. Generally, the pattern of results obtained using the simulation with normally hearing subjects matched the pattern obtained for the impaired subjects of Zurek and Delhorne. Thus, for these subjects, the threshold elevation and associated loudness recruitment appear sufficient to account for their difficulties in understanding speech.

In a second experiment, the hearing loss of subjects with more severe losses was simulated. The stimuli were amplified either by a constant amount across frequency (a 'flat' frequency-gain characteristic) or by an amount that increased with frequency (high-frequency emphasis). When the flat frequency-gain characteristic was used, the pattern of results from the impaired subjects closely matched the results from the normal subjects listening to the processed stimuli. However, when high-frequency emphasis was employed, the impaired subjects generally performed more poorly than their normal counterparts listening to processed stimuli. This suggests that some factor other than threshold elevation and loudness recruitment contributed to the speech-perception problems of the hearing-impaired subjects.

Moore and Glasberg (1993) used digital signal processing to implement a 13-band system. They simulated a flat moderate hearing loss (condition R2), a flat severe hearing loss (condition R3) and a loss that increased with frequency, being mild at low frequencies and severe at high frequencies (condition RX). They also included a control condition (R1) in which stimuli were processed through the simulation, but no expansion of the range of levels was applied in any frequency band.

Moore and Glasberg also assessed whether there are deleterious effects of recruitment after the hearing loss has been corrected as far as possible by linear amplification, as would be used in a conventional hearing aid. To do this, they ran a set of conditions in which the stimuli in conditions R2, R3 and RX were subjected to the frequency-dependent gain recommended by the NAL revised procedure (Byrne and Dillon, 1986; see Chapter 9, Section II.2 for more details). This gain is appropriate for speech presented at a moderate conversational level of about 65 dB SPL. This gave three more conditions, R2+, R3+ and RX+.

For speech in quiet, the simulation produced, as expected, a reduction in the ability to understand low-level speech. The level of speech required for 50 % intelligibility was 16 dB SPL in condition R1, 64 dB in condition R2, 79 dB in condition R3 and 53 dB in condition RX. However, speech at sufficiently high levels was highly intelligible in all conditions. Also, linear amplification according to the NAL prescription reduced the 50 % intelligibility points to less than 50 dB SPL for all conditions, and gave high intelligibility for speech at normal conversational levels. Thus, linear amplification was rather effective in improving the intelligibility of speech in quiet, although it did not allow speech to be both intelligible and comfortable over a wide range of sound levels; speech with input levels above about 70 dB SPL was judged to be unpleasantly loud in the conditions involving linear amplification.

For speech presented at a fixed input level of 65 dB SPL, against a background of a single competing talker, simulation of hearing loss produced substantial decrements in performance. This is illustrated in Figure 8.6 for the case where the background speech had a level of 71 dB SPL. Performance was almost perfect for condition R1, but not for the conditions simulating threshold elevation and loudness recruitment. The speech-to-background ratios in conditions R2 and RX had to be 11–13 dB higher than in condition R1 to achieve similar levels of performance. Linear amplification according to the NAL prescription improved performance markedly for the conditions simulating flat losses, but was less effective for the condition simulating a sloping

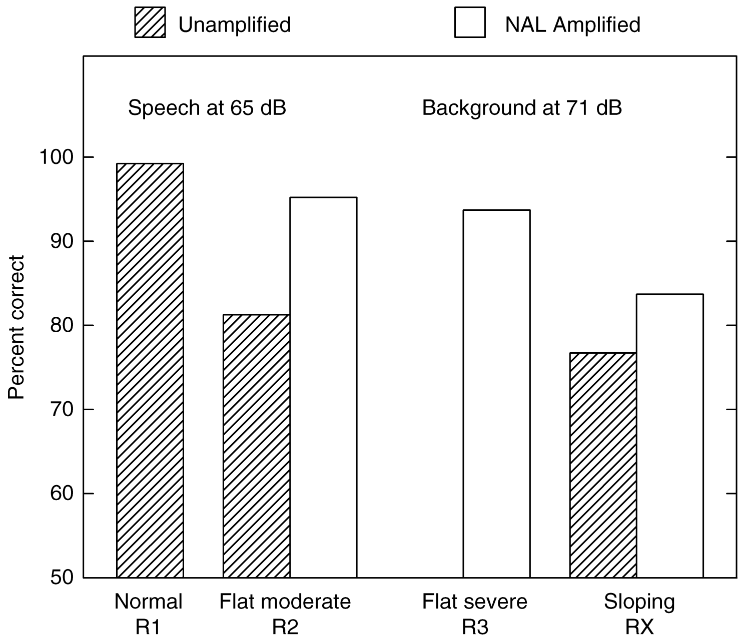


Figure 8.6. Comparison of the percentage correct scores for the different conditions of Moore and Glasberg (1993), for a speech-to-background ratio of −6 dB. No score is shown for condition R3, since the target speech was inaudible for that condition.

loss. Performance in this condition remained well below that in the control condition, even after linear amplification.

In a second study, Moore, Glasberg and Vickers (1995) used similar processing conditions (R1, R2, R3 and RX), but the speech was presented in a background of speech-shaped noise instead of a single competing talker. The input level of the speech was fixed at 65 dB SPL, while the level of the background noise varied from 65 to 74 dB SPL. The speech in condition R3 was inaudible. For conditions R2 and RX, the speech-to-noise ratios had to be up to 6 dB higher than in the control condition (R1, unprocessed stimuli) to achieve similar levels of performance. When linear amplification according to the NAL prescription was applied before the simulation, performance improved markedly for conditions R2 and RX, and did not differ significantly from that for R1. For condition R3, performance with simulated NAL amplification remained below that for condition R1; the decrement in performance was equivalent to about a 1-dB change in speech-to-noise ratio.

The results obtained using speech-shaped noise differ in two ways from the results obtained using a single talker as the interfering sound: first, the deleterious effects of the simulations were generally smaller with speech-shaped noise; secondly, linear amplification according to the NAL formula restored performance to normal for two of the three conditions tested (R2+ and RX+) using speech-shaped noise, whereas it did not when a single competing talker was used. In the remaining condition (R3+), the degradation in performance was significant, but was equivalent to a change of only 1 dB in speech-to-noise ratio.

The differences between the two sets of results can be understood in the following way. Normally hearing people can take advantage of spectral and temporal dips in a background of a single competing talker. Hence, the speech-to-background ratio required to achieve a given level of performance is markedly lower than when the background is speech-shaped noise. However, 'dip listening' requires the ability to hear over a wide range of sound levels (a wide dynamic range). Consider, as an example, a situation where the target speech has a level 10 dB lower than that of the interfering speech and where the overall level is such that the interfering speech is loud, but not uncomfortably so. The overall loudness is determined mainly by the levels of peaks in the interfering speech. In any given frequency band, the target speech may contain useful information at a level 40 dB below the peak level of the interfering speech (since for one voice the range of levels is about 30 dB – see Section III.1 on the AI). Such information may be audible during brief pauses in the interfering speech. Ideally, then, brief sounds should be audible and discriminable when their levels are 40 dB below the highest comfortable level.

Loudness recruitment, either real or simulated, reduces the range of levels over which sounds are both audible and comfortable. If the level of the speech is set so that the more intense parts of the speech are comfortably loud, the weaker parts may be inaudible. Hence, people with recruitment cannot exploit dip listening as effectively as normally hearing people (Peters, Moore and Baer, 1998). When a background of speech-shaped noise is used, dip listening is of much less importance, since the noise does not contain dips of sufficient magnitude or duration. Hence, speech intelligibility depends more on the higher-level portions of the speech, and these are less affected by reduced dynamic range. Furthermore, linear amplification, which ensures that the higher level portions of speech are clearly audible, is effective in compensating for the simulated recruitment, except when the hearing loss is severe (condition R3).

Moore *et al.* (1997) presented simulations of hearing impairment to the normal ears of subjects with moderate to severe unilateral cochlear hearing loss. The intelligibility of speech in quiet and in background sounds was compared with that obtained for the impaired ears using unprocessed stimuli. The results of loudness matches between the two ears were used to tailor a simulation of threshold elevation combined with loudness recruitment individually for each subject. The subjects reported that the simulation produced sounds with appropriate loudness and dynamics, but the processed speech in the normal ear appeared markedly more clear than the unprocessed speech in the impaired ear. Performance for the impaired ears in identifying speech in quiet and in background sounds was markedly worse than for the normal ears using the simulation of threshold elevation and loudness recruitment. Moore *et al.* suggested that the relatively poor results for the impaired ears might be caused partly by a form of 'neglect' which is specific to subjects with unilateral or asymmetric loss (Hood, 1984). This idea was supported by results obtained using bilaterally hearing-impaired subjects, with similar amounts of hearing loss to the unilaterally hearing-impaired subjects. The speech identification scores of the bilaterally impaired subjects were markedly better than for the impaired ears of the unilaterally hearing-impaired subjects. However, the scores for identifying speech in background noise were still somewhat worse than those for the normal ears listening to the simulation of threshold elevation and loudness recruitment.

In one respect, the results of the simulations differ from those obtained using subjects who have actual cochlear hearing loss. When threshold elevation and loudness recruitment are simulated using normally hearing subjects, performance is markedly better when the interfering sound is a single talker than when it is speech-shaped noise. However, as described earlier, people with cochlear hearing loss often show little or no difference in intelligibility for these two backgrounds (Duquesnoy and Plomp, 1983; Festen and Plomp, 1990; Moore, Glasberg and Stone, 1991; Peters, Moore and Baer, 1998). This suggests that some factor other than recruitment contributes to the reduced ability of hearing-impaired people to ‘listen in the dips’. One possible factor is reduced frequency selectivity (Baer and Moore, 1994). Another possible factor is reduced sensitivity to temporal fine structure, the effects of which were discussed earlier in this chapter.

In summary, the results of the simulations indicate that the reduced dynamic range associated with loudness recruitment can lead to difficulty in understanding speech in background sounds. When the background sound is a steady noise, the difficulty can be effectively compensated by linear amplification of the type commonly used in hearing aids, except when the loss is severe. When the background sound is fluctuating, as for a single talker, the deleterious effects of the reduced dynamic range are greater and are not fully compensated by linear amplification.

VIII.2 SIMULATIONS OF REDUCED FREQUENCY SELECTIVITY

To simulate reduced frequency selectivity, the spectra of stimuli are ‘smeared’ or ‘smoothed’ on a moment-by-moment basis so that the excitation pattern produced in a normal ear resembles (albeit crudely in many of the simulations) the pattern that would be produced in an impaired ear using unprocessed signals. Several different techniques have been used to perform the spectral smearing. Early studies used analogue signal processing (Villchur, 1977; Summers and Al-Dabbagh, 1982; Summers, 1991). In more recent studies, digital signal processing techniques have been used. An illustration of this type of processing is shown in Figure 8.7. A short segment of the signal is taken, and its spectrum is calculated using a mathematical technique called the *Fast Fourier Transform* (FFT). The resulting spectrum is then smeared or blurred so as to decrease the contrast between peaks and valleys in the spectrum. The modified spectrum is transformed back into a temporal waveform using an *Inverse Fast Fourier Transform* (IFFT). This is repeated for a series of overlapping segments, and the resulting processed segments are added together. Hence, this is referred to as the overlap-add method (Allen, 1977). The method has been used in several studies (Celmer and Bienvenue, 1987; Howard-Jones and Summers, 1992; ter Keurs, Festen and Plomp, 1992,

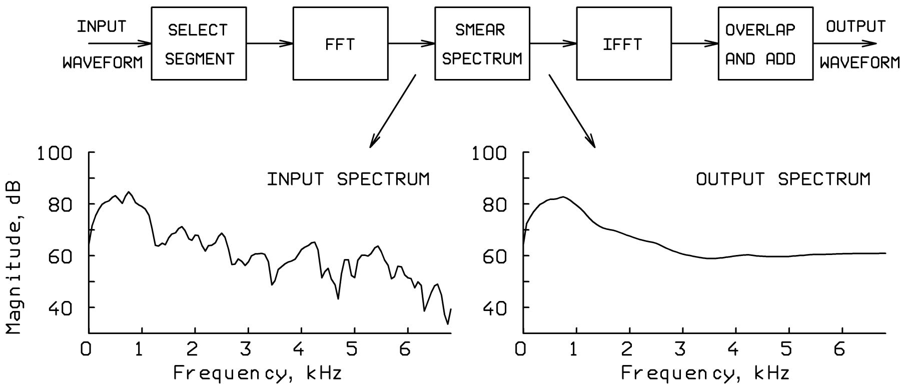


Figure 8.7. The upper part shows the sequence of operations used to perform spectral smearing. The lower part shows examples of short-term spectra at the input and output of the smearing process.

1993; Baer and Moore, 1993, 1994). Examples of sounds processed in this way are on a compact disc which can be obtained from the author (Moore, 1997).

It should be noted that such spectral smearing does not simulate all of the effects of reduced frequency selectivity. Specifically, the time patterns at the outputs of the auditory filters are affected by reduced frequency selectivity in a way that is not reproduced by the simulations. Instead, the simulations alter the time patterns of the stimuli in a complex way that is a by-product of the specific processing used. Essentially, the simulations may be regarded as mimicking the consequences of reduced frequency selectivity in terms of place coding in the auditory system, but not in terms of temporal coding.

Ter Keurs, Festen and Plomp (1992) smeared the spectrum of each segment in such a way as to simulate the effect of broadened auditory filters whose bandwidth was a constant proportion of the centre frequency. They smeared the spectrum of speech and noise separately and then added the speech and noise together. The SRT for speech in noise increased once the filter bandwidth used in the simulation was increased beyond about 23 % of the centre frequency (roughly double the normal bandwidth of the auditory filter). Vowel identification was affected more than consonant identification. In a second study (ter Keurs, Festen and Plomp, 1993), they compared the effects of the smearing using either speech-shaped noise or a single talker as the background sound. For both types of background, SRTs increased when the smearing bandwidth was increased. For unsmeared speech, SRTs were 5–7 dB lower when the background was a single talker than when it was speech-shaped noise. This difference decreased as the smearing bandwidth was increased. Hence, the effect of the spectral smearing on SRTs was greater for the speech masker than for the noise masker.

Baer and Moore (1993) measured the intelligibility of speech in quiet and in speech-shaped noise. Normally hearing subjects listened to sentence material that had been processed to simulate varying degrees of loss of frequency selectivity. When speech in noise was used, the speech was mixed with the noise prior to processing. The procedure for simulating impaired frequency selectivity used a realistic form of spectral smearing, based on measured characteristics of auditory filters in hearing-impaired people. It was based on a procedure that was previously validated using non-speech signals (Moore, Glasberg and Simpson, 1992). The stimuli were processed so that the excitation pattern produced in a normal ear, calculated over a short period of time, would resemble that found with unprocessed stimuli in an impaired ear.

Several different types of smearing were used, simulating specific degrees of broadening and asymmetry of the auditory filter. Sentences in quiet and in speech-shaped noise were smeared and presented to normally hearing listeners in intelligibility tests. Some of the results are illustrated in Figure 8.8. The intelligibility of speech in quiet (the three left-most bars in Figure 8.8) was hardly affected by spectral smearing, even for smearing that simulated auditory filters six times broader than normal. However, this probably reflects a ceiling effect, as scores were close to perfect in all conditions. The intelligibility of speech in noise was adversely affected by the smearing, especially for large degrees of smearing and at a low speech-to-noise ratio (−3 dB; see the three right-most bars in Figure 8.8). Simulation of asymmetrical broadening of

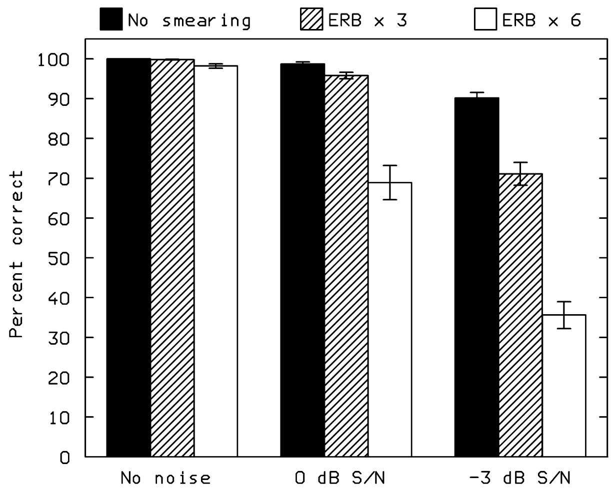


Figure 8.8. Results of Baer and Moore (1993) showing the percentage of key words correct for three amounts of spectral smearing and three different amounts of background noise. The three amounts of smearing were: no smearing (solid bars), simulation of auditory filters with bandwidths three times greater than normal (diagonally shaded bars) and simulation of auditory filters with bandwidths six times greater than normal (open bars). Error bars indicate ± one standard error.

the lower side of the auditory filter had a greater effect than simulation of asymmetrical broadening of the upper side, suggesting that upward spread of masking may be particularly important.

In a second study, Baer and Moore (1994) used a single competing talker as the background, instead of speech-shaped noise. The results were similar in form to those found using speech-shaped noise. Specifically, performance worsened with increasing smearing, and the worsening was greater at the more adverse speech-to-background ratio. The results agreed with those of ter Keurs, Festen and Plomp (1993) in showing that the deleterious effects of spectral smearing were greater for a speech masker than for a noise masker. Hence, the difference in masking produced by speech and noise maskers was less for spectrally smeared than for unprocessed stimuli. This is consistent with the finding noted earlier, that for normally hearing subjects SRTs are lower when the background is speech than when it is noise, while for hearing-impaired subjects the difference is smaller or absent.

In summary, the results of experiments on spectral smearing suggest that reduced frequency selectivity does contribute significantly to the difficulties experienced by people with cochlear hearing loss in understanding speech in the presence of background sounds.

VIII.3 SIMULATION OF THE COMBINED EFFECTS OF THRESHOLD ELEVATION, RECRUITMENT AND REDUCED FREQUENCY SELECTIVITY

Nejime and Moore (1997) simulated the effect of loudness recruitment and threshold elevation together with reduced frequency selectivity. To implement the combined simulation, they first applied the simulation of reduced frequency selectivity (as used by Baer and Moore, 1993) and then applied the simulation of threshold elevation combined with loudness recruitment (as used by Moore and Glasberg, 1993). Examples of sounds processed using the combined simulation can be found on a compact disc which can be obtained from the author (Moore, 1997). In their first experiment, four conditions were simulated: a moderate flat loss with auditory filters broadened by a factor of three (B3R2), a moderate-to-severe sloping loss with auditory filters broadened by a factor of three (B3RX) and those conditions with linear amplification applied prior to the simulation processing (B3R2+, B3RX+). The amplification in these two latter conditions was similar to what might be used in a well-fitted linear hearing aid. For conditions B3R2 and B3RX, performance was markedly worse than for a control condition (normal hearing, condition R1) tested in a previous study. For conditions B3R2+ and B3RX+, linear amplification improved performance considerably. However, performance remained below that for condition R1 by between 5 and 19 %. Thus, linear amplification did not fully compensate for the deleterious effects of the simulation.

In the second experiment of Nejime and Moore (1997), the broadening of the auditory filters was made more realistic by making it a function of the absolute threshold at the centre frequency of the auditory filter; greater hearing loss was associated with a broader filter. Three different hearing losses were simulated: a moderate-to-severe sloping loss with variable broadening of the auditory filters (BXRX), the same moderate-to-severe sloping loss with linear amplification (BXRX+) and the same broadening of the auditory filters but without the simulation of loudness recruitment and threshold elevation (BX). For condition BXRX, performance was markedly worse than in condition R1, while performance in condition BX was somewhat worse than for condition R1. For condition BXRX+, linear amplification according to the NAL procedure improved performance to a large extent but it remained worse than for condition R1.

Comparison of the results with previous results using simulations only of recruitment and threshold elevation showed significantly poorer performance for the combined simulation. This was true both without and with linear amplification. The results suggest that the reduced frequency selectivity simulated by spectral smearing is a suprathreshold factor that is relatively unaffected by amplification and which combines roughly additively with the effects of loudness recruitment and threshold elevation.

VIII.4 SIMULATION OF REDUCED TEMPORAL RESOLUTION

In Chapter 5, Section II, temporal resolution was described in terms of a four-stage model consisting of an array of bandpass filters (the auditory filters) each followed by a compressive nonlinearity, a sliding temporal integrator and a decision device. It was argued that cochlear hearing loss can affect the first two stages, the filters and the nonlinearity, but that it usually does not affect the last two. Cochlear hearing loss results in a broadening of the auditory filters and a reduction in the compressive nonlinearity. The end result is that temporal resolution is reduced only for certain specific stimuli, namely those with slowly fluctuating envelopes. In some situations, temporal resolution may be poor simply because of the low SL of the stimuli for people with cochlear hearing loss.

The simulation of loudness recruitment described above effectively includes these effects; specifically, it models the effects of reduced SLs and reduced compressive nonlinearity. Indeed, as described in Chapter 5, Section IV.3, the simulation of loudness recruitment in this way results in poorer detection of temporal gaps in narrow bands of noise. One might question, therefore, whether it is appropriate to go any further in simulating the effects of abnormalities in temporal processing. However, some researchers have taken the approach of simulating the effects of reduced temporal resolution ‘directly’ by temporal smearing analogous to the spectral smearing used to simulate reduced frequency selectivity. Such work can be valuable because it provides a way of estimating the relative importance for speech intelligibility of temporal modulations at different rates.

Drullman, Festen and Plomp (1994) carried out a study of the effects of temporal envelope smearing on speech intelligibility. The study was not designed specifically as a simulation of reduced temporal resolution, but in effect the processing was equivalent to such a simulation. They split the input signal into a number of frequency bands covering the range 100–6400 Hz. The bands had widths of 0.25, 0.5 or 1 octave. The narrowest bandwidth corresponds roughly with the ERBs of normal auditory filters, while the wider bandwidths correspond roughly with the auditory filter bandwidths that might occur in people with moderate-to-severe cochlear losses. They extracted the envelope of the signal in each band and processed the envelope so as to smooth out rapid fluctuations. The envelope for each band was lowpass filtered with various cutoff frequencies. As the cutoff frequency was reduced, the envelope became more and more smeared in the time domain, simulating loss of temporal resolution. For all filtering bandwidths (0.25, 0.5 or 1 octave), the SRT of sentences in speech-shaped noise was almost unaffected by lowpass filtering the envelopes with cutoff frequencies of 64, 32 or 16 Hz; the SRT was elevated by only about one decibel relative to that for a control condition with no envelope filtering. However, lowpass filtering with cutoff frequencies of 8 or 4 Hz produced mean elevations in SRT of 2.4 and 5.6 dB relative to the control condition. Thus, removing very fast envelope fluctuations had little effect, but removing slow fluctuations led to impaired intelligibility. These results indicate that fluctuations of the envelopes of the filtered bands at rates of 16 Hz and above are relatively unimportant for speech intelligibility. Since most people with cochlear hearing loss can easily detect modulation at rates up to 16 Hz (e.g. see Figure 5.7), this suggests that reduced temporal resolution is not a factor limiting speech intelligibility for most people with cochlear hearing loss.

Hou and Pavlovic (1994) carried out a study that was aimed at directly simulating the effects of reduced temporal resolution in the auditory system. They first filtered the speech into 23 bands using simulated auditory filters with bandwidths corresponding to those found in normal ears. They then smeared the temporal envelope in each band using a sliding temporal integrator similar to that described in Chapter 5, Section II.3 for the model of temporal resolution (Moore *et al.*, 1988; Plack and Moore, 1990). When the temporal integrator performed less smearing than the temporal integrator in the normal auditory system (see Chapter 5, Section II.3), the smearing had no effect on the intelligibility of nonsense syllables in quiet. When the temporal smearing was greater, small but significant decreases in intelligibility occurred. However, even when the smearing was about three times as large as that occurring in a normal auditory system, the decrease in intelligibility was only 5–10 %.

Hou and Pavlovic's results indicate that temporal smearing can produce reduced intelligibility of nonsense syllables in quiet. They argued that reduced temporal resolution in the impaired auditory system was probably as detrimental to speech recognition as reduced frequency selectivity. However, it remains unclear whether temporal smearing is an appropriate way to model reduced temporal resolution in hearing-impaired subjects. The temporal smearing used by Hou and Pavlovic would lead to reduced temporal resolution for all types of stimuli, whereas, as described in Chapter 5, Section IV, reduced temporal resolution in people with cochlear hearing loss generally occurs only for stimuli at low SLs or with slowly fluctuating envelopes.

In any case, the shape of the temporal integrator measured for hearing-impaired subjects is not usually markedly different from normal (Plack and Moore, 1991).

IX CONCLUSIONS

People with mild or moderate cochlear losses can usually understand speech reasonably well when they are in a quiet room with only one person talking. However, they have difficulty when more than one person is talking at once, or when background noise or reverberation are present. People with severe or profound losses usually have difficulty even when listening to a single talker in a quiet room, and they generally have severe problems when background noise is present. Such difficulties may be especially pronounced when an extensive dead region is present. The elevation of the SRT in noise, relative to the value for normally hearing subjects, is greater when the background sound is fluctuating (e.g. a single talker) than when it is a steady noise. People with cochlear hearing loss are less able than normally hearing people to take advantage of temporal and spectral dips in fluctuating background sounds. Also, people with cochlear hearing loss are less able than normally hearing people to exploit spatial separation between the target and interfering sounds.

The speech-perception difficulties of people with mild cochlear hearing losses can be accounted for primarily in terms of audibility, at least for speech in quiet. With suitable linear amplification, such people can usually understand speech almost as well as normally hearing people. However, for people with moderate-to-severe cochlear hearing losses, suprathreshold discrimination abilities also need to be taken into account. Although audibility is still of primary importance, it does not fully account for the difficulties experienced by such people.

It has proved difficult to determine the relative importance of different psychoacoustic factors that might affect speech intelligibility. However, evidence from a range of studies, including correlational studies, studies of the use of specific speech cues, and simulations, all indicate that reduced frequency selectivity plays a significant role. There is also evidence, mainly from simulations, that loudness recruitment is a significant factor. However, the most important aspect of loudness recruitment appears to be reduced dynamic range, rather than distorted loudness relationships. Finally, a reduced ability to make use of temporal fine structure cues appears to contribute to difficulty in 'listening in the dips' of a background sound.

9 Hearing Aids

I INTRODUCTION

Cochlear hearing loss can rarely be 'cured', although drugs can be of benefit in some cases; for a review of drug treatments, see Dobie (1997). Hearing aids are the primary method for alleviating problems associated with cochlear hearing loss. However, people with cochlear hearing loss frequently complain that their hearing aids are of limited benefit. They find that the aids are sometimes useful in helping them to hear weak sounds, but that the aids do not help very much, if at all, when background noise is present. It seems clear that hearing aids are not like eyeglasses. A well-fitted pair of eyeglasses can give good vision if the problem is caused by poor focus of the image on the retina, as is often the case. Hearing aids do not restore hearing to 'normal'. They can partially compensate for the loss of sensitivity produced by cochlear damage, but they do not compensate for the suprathreshold changes in perception caused, for example, by reduced frequency selectivity or reduced ability to use temporal fine structure cues. This chapter reviews the capabilities and limitations of current hearing aids, and considers some ways in which hearing aids may be improved in the future. It also discusses methods of fitting hearing aids to suit the individual.

II LINEAR AMPLIFICATION

II.1 THE DIFFICULTY OF RESTORING AUDIBILITY USING LINEAR AIDS

The primary goal of hearing aids is to restore audibility via frequency-selective amplification. Up to about 1990, most hearing aids operated essentially as linear amplifiers; for low and moderate input levels, they applied gain that was independent of the input level. For example, such an aid might amplify by 20 dB at 1 kHz, and that same gain would be applied regardless of the input level. A 1-kHz sinusoid with an input level of 40 dB SPL would be amplified to a level of 60 dB SPL, while an input level of 80 dB SPL would be amplified to 100 dB SPL. In practice, most linear hearing aids incorporate some means of limiting the maximum output, and, if the limiter comes into operation, the aid behaves in a nonlinear manner, for example significant *harmonic* and *intermodulation distortion* may be produced (see Chapter 1, Section II). However, for convenience, aids which are normally used over the linear part of their operating range will be referred to here as linear.

It became apparent very soon after hearing aids first came into use that it was not practical to use linear amplification to compensate fully for the loss of audibility caused by cochlear hearing loss (Steinberg and Gardner, 1937). The major factor preventing this was loudness recruitment and the associated reduced dynamic range, as described in Chapter 4, Section III. Say, for example, a person had a cochlear hearing loss of 60 dB at all frequencies. The highest comfortable level (HCL) for such a person would typically be around 90–100 dB HL. If the person had a hearing aid that fully compensated for the loss of audibility, the aid would apply a gain of 60 dB at all frequencies. However, that would mean that any sound with a level above about 40 dB HL would be amplified to a level exceeding the HCL. To avoid uncomfortable loudness, the gain selected by users of linear hearing aids is often small, and may be insufficient to ensure audibility of speech at normal conversational levels over a wide frequency range (Leijon, 1990). Simulations of the problems associated with linear amplification can be found on a compact disc which is available from the author (Moore, 1997).

Another problem with linear hearing aids is that users often find it necessary to adjust the volume control to deal with different listening situations. The overall level of speech and other sounds can vary considerably from one situation to another (Pearsons, Bennett and Fidell, 1976; Killion, 1997), and people with cochlear hearing loss do not have sufficient dynamic range to deal with this. However, adjustment of the volume control can be difficult, especially for elderly people, who may have limited manual dexterity, and placing a hand near the aid changes the acoustical conditions and may induce feedback (see below for further discussion of feedback). This situation has been improved by the introduction of hearing aids with remote control of volume, but even with remote control, the need to adjust the volume control frequently is irksome.

II.2 PRESCRIPTIVE FITTING RULES FOR LINEAR HEARING AIDS

Even when linear hearing aids include output limiting, it has been found to be impractical to compensate fully for loss of audibility. Rather, various prescriptive rules have been developed that can be used to derive an appropriate frequency-gain characteristic from the pattern of hearing loss as a function of frequency (Lybarger, 1978; McCandless and Lyregard, 1983; Byrne and Dillon, 1986; Moore and Glasberg, 1998). These rules prescribe the amount of gain that is required as a function of frequency. Sometimes, suprathreshold measures such as the uncomfortable loudness level (ULL) or most comfortable level (MCL) are also taken into account. Such rules always prescribe less gain than the hearing loss at a given frequency. Typically, the rules prescribe a gain that is between one-third and one-half of the hearing loss. For example, if the hearing loss is 60 dB at 1 kHz, then the prescribed gain is between 20 and 30 dB.

The goal of most of these rules is to select a frequency-gain characteristic that will amplify speech so as to improve intelligibility, while avoiding making speech uncomfortably loud. One popular rule is the NAL(R) formula devised at the National Acoustic Laboratories of Australia (Byrne and Dillon, 1986). The rule works solely on the basis of the pure-tone audiogram. It was derived by dividing speech into a number of contiguous 1/3-octave bands or one-octave bands covering the range 250–6000 Hz and working out what gain would be required in each band to amplify the speech in that band to the MCL, assuming that the speech was at a normal conversational level (typically 65–70 dB SPL overall when the talker and listener are separated by about one metre). It should be noted that the user of a hearing aid is usually able to adjust the overall gain via the volume control on the aid. Hence, the main point of prescriptive rules, including NAL(R), is to select an appropriate 'shape' for the frequency-gain characteristic. However, the NAL(R) formula prescribes a gain that is intended to correspond to the average gain actually used. The goal of the NAL(R) procedure seems very reasonable. Making each frequency band in speech equally loud can ensure that no single band contributes excessively to loudness. Thus, the overall amplification can be increased, increasing the proportion of the speech spectrum that is audible while maintaining a comfortable loudness.

To check whether the NAL(R) procedure does actually meet its goal, Moore and Glasberg (1998) used a model of loudness similar to that described in Chapter 4, Section V. They calculated *specific loudness patterns* evoked by a sound with a spectrum corresponding to the long-term average spectrum of speech (Byrne *et al.*, 1994), assuming an overall speech level of 65 dB SPL. The specific loudness patterns were calculated for several hypothetical hearing losses, varying in severity and type (e.g. flat losses, losses increasing towards high frequencies and so on). Specific loudness patterns were calculated both for the unamplified speech spectrum and for speech spectra which had been subjected to frequency-selective amplification according to various rules, including NAL(R). In principle, the NAL(R) formula should result in a specific loudness pattern that is flat (constant specific loudness) over the frequency range that is most important for speech, namely 500–5000 Hz. However, Moore and Glasberg found that the specific loudness patterns were not flat; they usually showed a mid-frequency peak, with the specific loudness decreasing at higher frequencies. Moore and Glasberg proposed a new formula, the Cambridge formula, which did lead to specific loudness patterns that were approximately flat, according to the loudness model. This formula calls for slightly less mid-frequency gain and slightly more high-frequency gain than the NAL(R) formula.

Although the various prescriptive rules do lead to different frequency-gain characteristics for the same audiogram, it has proved surprisingly difficult to demonstrate differences between the rules in terms of resulting speech-recognition scores; for a review, see Humes (1991). The intelligibility of speech in noise does not seem to be greatly affected by the shape of the frequency-gain characteristic, provided the amplification is sufficient to place most of the speech spectrum above absolute threshold. For example, van Buuren, Festen and Plomp (1995) measured speech reception thresholds (SRTs) for speech in speech-shaped noise using a wide range of frequency-gain characteristics. These characteristics were chosen so that the average speech spectrum, after amplification, fell in the range from 5 dB above the absolute threshold to just below UCL. The SRTs were similar for most of the frequency-gain characteristics. However, performance was somewhat worse (SRTs were higher) when the amplified speech spectrum had high levels at low frequencies.

At first sight, these results appear to suggest that the frequency-gain characteristic is unimportant. However, in the study by van Buuren, Festen and Plomp, the speech was presented over a very restricted range of sound levels. A given frequency-gain characteristic might give satisfactory results over a small range of levels, but might be much less successful for speech covering a wide range of levels, as occurs in everyday life. Say, for example, that a frequency-gain characteristic with low-frequency emphasis was used. The low-frequencies in the speech might be presented at a level just below the ULL, while the high frequencies might be only 5 dB above absolute threshold. If the overall level was increased, the low frequencies would become uncomfortably loud, while if the level was decreased by more than 5 dB, some of the high frequencies would become inaudible. In practice, it would be better to use a frequency-gain characteristic that placed speech with a moderate overall level (65–70 dB SPL) more towards the middle of the listener's dynamic range at all frequencies. A formula such as the Cambridge formula does something close to that. If speech is presented at an equal comfortable level in all frequency bands, then increasing or decreasing the level of the speech by a *moderate* amount will not lead to the speech becoming uncomfortably loud or partly inaudible.

In any case, it should be realized that the goal of amplifying all bands of speech to the MCL could only be realized in practice for one specific speaker talking at one specific level. The long-term average spectrum of speech can vary markedly from one speaker to another. Also, the average shape of the speech spectrum for a given talker varies with vocal effort; a higher vocal effort leads to relatively more high-frequency emphasis (ANSI, 1997). Thus, at best, a prescriptive rule can give a frequency-gain characteristic that will be appropriate for 'typical' listening situations encountered by a 'typical' user. In practice, it may be necessary to do some 'fine tuning' of a hearing aid around the frequency-gain characteristic recommended by a given rule (Moore and Glasberg, 1998).

III COMPRESSION AMPLIFICATION

It was suggested many years ago that the problems associated with linear amplification, as described above, could be alleviated by the use of an automatic gain control (AGC) system (Steinberg and Gardner, 1937). With AGC, it is possible to amplify weak sounds more than stronger ones, with the result that the wide dynamic range of the input signal is compressed into a smaller dynamic range at the output. Hence, AGC systems are also called *compressors*. Although this idea sounds simple, in practice there are many ways of implementing AGC, and there is still no clear consensus as to the 'best' method, if there is such a thing. To understand why things are not so simple, it is helpful first to consider some of the basic characteristics of AGC systems.

III.1 BASIC CHARACTERISTICS OF AUTOMATIC GAIN CONTROL SYSTEMS

An AGC amplifier is an amplifier whose gain is determined by a control signal. The gain is defined as the output voltage divided by the input voltage, or, if both are expressed in decibels, as the output level minus the input level. For example, if the output voltage is twice the input voltage, the gain is 2 (in linear terms) or 6 dB (since $20\log_{10}(2) = 6$). The control signal is derived either from the input to the amplifier or from its output. The gain is reduced as the input level is increased. For inputs below a certain level, most AGC amplifiers act as linear amplifiers. Over the range where the amplifier is linear, the output is directly proportional to the input. If the output level in decibels is plotted as a function of the input level in decibels, the result is a straight line with a slope of one; see Chapter 1, Section II. Once the input level exceeds a certain value, the gain is reduced and the slope of the line becomes less than one. This is illustrated by the *input-output function* in Figure 9.1. The *compression threshold* is defined as the input level at which the gain is reduced by 2 dB, relative to the gain applied in the region of linear amplification (ANSI, 2003). For example, if the gain were 25 dB for input levels well below the compression threshold, the compression

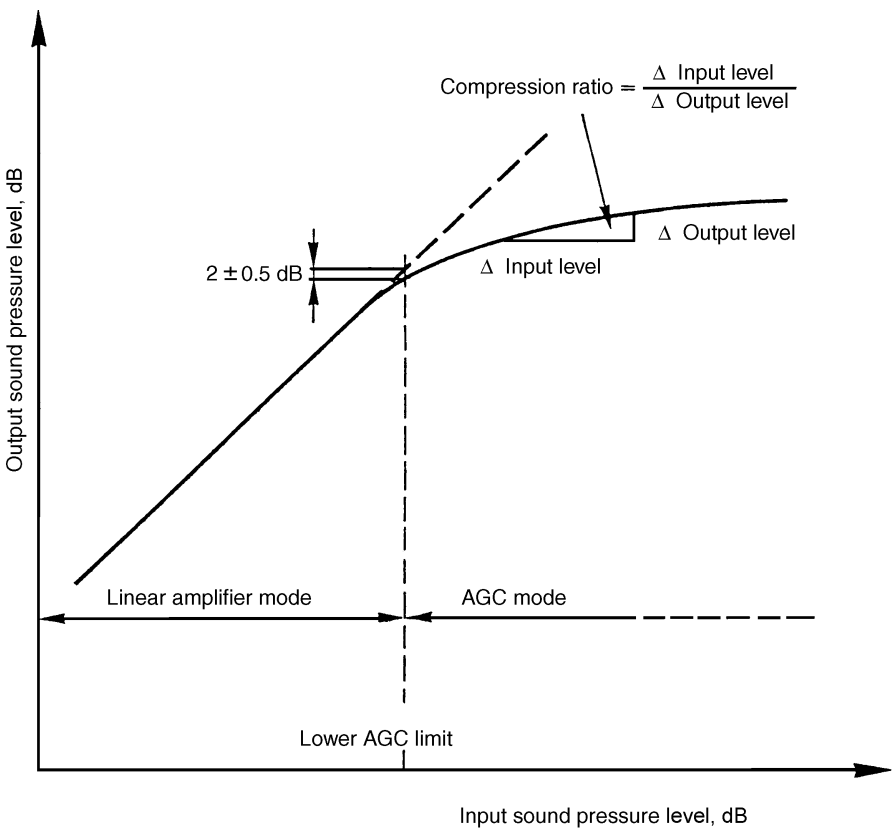


Figure 9.1. An input-output function for an AGC system. The output level in decibels is plotted as a function of the input level in decibels.

threshold would be the input level at which the gain was reduced to 23 dB. One reason for having a compression threshold, with linear amplification for lower levels, is that it is impractical to continue to increase the gain indefinitely as the input level decreases. A second reason is that the use of high gain for very low-level inputs can make microphone noise or low-level environmental noise sound intrusive. Indeed, for very low-level inputs, some hearing aids reduce the gain to prevent such noises from being audible; this is called expansion, as opposed to compression.

The 'amount' of compression is specified by the *compression ratio*, which is the change in input level (in decibels) required to achieve a 1-dB change in output level (for an input level exceeding the compression threshold); the compression ratio is equal to the reciprocal of the slope of the input-output function in the range where the compression is applied. For example, a compression ratio of three means that the output grows by 1 dB for each 3-dB increase in input level. The compression ratio is measured using a steady signal such as a continuous sinusoid. When the level of the sinusoid is changed, the gain is allowed to stabilize before a measurement is taken. The compression ratio measured in this way will be referred to as CR_{static}.

When the input level is high, the gain of an AGC amplifier, expressed in decibels, may become negative, that is the signal is attenuated rather than being amplified. Consider the example shown in Figure 9.2. For input levels below 40 dB SPL, the

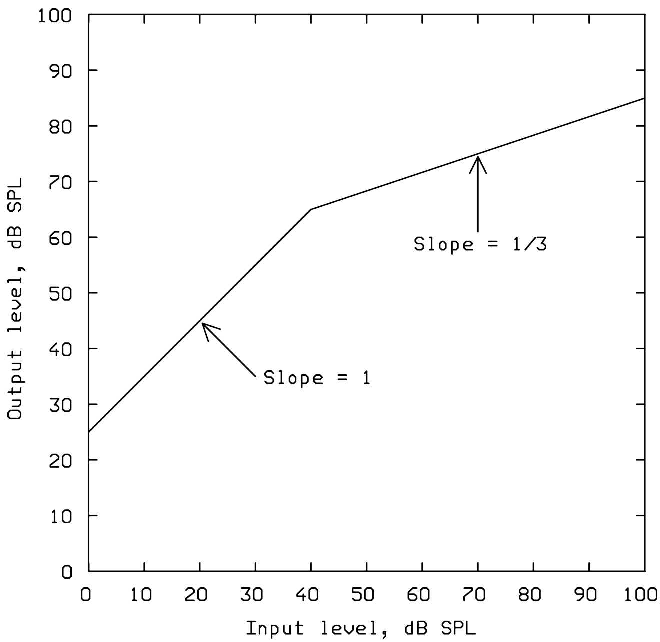


Figure 9.2. A schematic input-output function for an AGC system with a compression threshold of 43 dB SPL, and a compression ratio of 3. Notice that for input levels above about 78 dB SPL the output level is lower than the input level, that is the system acts as an attenuator.

AGC amplifier applies a gain of 25 dB. The gain starts to decrease for input levels above 40 dB SPL. For an input level of 43 dB SPL, the gain is reduced to 23 dB. Therefore, the AGC amplifier has a compression threshold of 43 dB SPL. The value of CR_{static} is 3. When the input level is increased from 40 to 100 dB SPL (a 60-dB range), the output level increases from 65 to 85 dB SPL (a 20-dB range). Thus, for an input level of 100 dB SPL, the signal is attenuated by 15 dB. This is not necessarily a bad thing. Many people, both normally hearing and hearing-impaired, find that sounds with levels of 100 dB SPL and above are unpleasantly loud. Reducing the sound level can make the loudness more acceptable, without impairing the ability to detect or discriminate between sounds.

AGC amplifiers vary in how quickly they react to changes in input sound level. Typically, the speed of response is measured by using as an input a sound whose level changes abruptly between two values, normally 55 dB SPL and 90 dB SPL. When the sound level abruptly increases, the gain decreases, but this takes time to occur. Hence, the output of the amplifier shows an initial spike or 'overshoot', followed by a decline to a steady value. This is illustrated in Figure 9.3. The time taken for the output to get within 3 dB of its steady value is called the *attack time*, and is labelled t_a in Figure 9.3. When the sound level abruptly decreases, the gain increases, but again this takes time to occur. Hence, the output of the amplifier shows an initial 'dip',

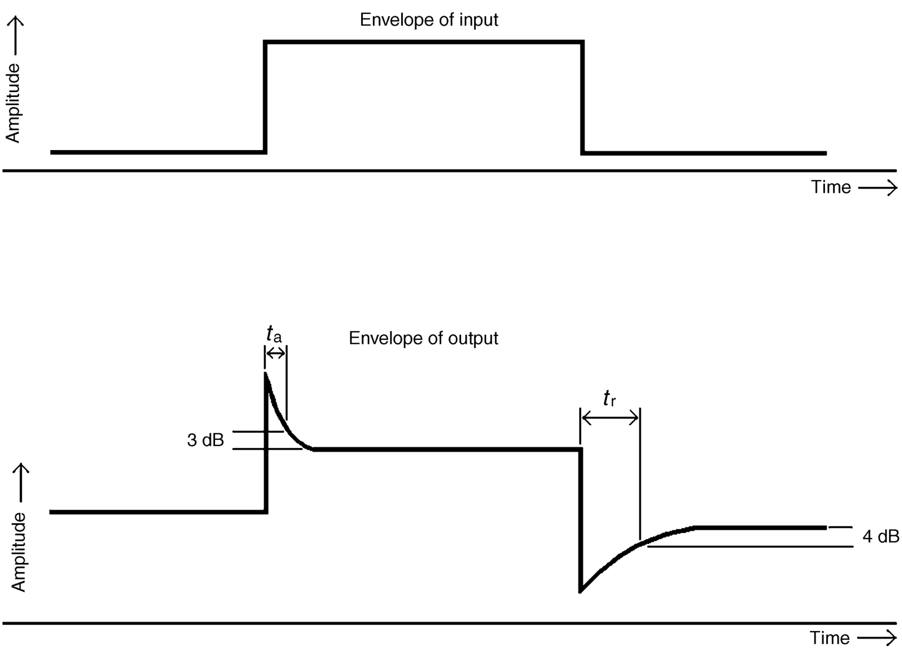


Figure 9.3. Illustration of the temporal response of an AGC system. The envelope of the input signal is shown at the top. The envelope of the output of the system is shown at the bottom.

followed by an increase to a steady value. The time taken for the output to increase to within 4 dB of its steady value is called the *recovery time* or *release time*, and is labelled t_r in Figure 9.3.

The gain-control signal in an AGC system resembles a smoothed version of the envelope of the input signal. The higher the value of the smoothed envelope, the lower is the gain. Also, the longer the attack and recovery times, the greater is the amount of smoothing. As a result of this smoothing, the gain changes do not follow rapid fluctuations in the envelope of the signal; only the slower fluctuations are followed. Thus, the amount of compression applied to a signal like speech is often less than would be expected from CR_{static}, especially when the compression is slow acting.

One way to quantify this effect is to use as input to the AGC system a signal which is sinusoidally amplitude modulated, with modulation frequency f_m (Braida *et al.*, 1982; Stone and Moore, 1992, 2003a). It is convenient to specify the modulation depth of the input signal, D_{in}, as the difference in level between a peak and a valley of the modulated waveform. For example, D_{in} might be 10 dB. The modulation depth of the signal at the output of the system, D_{out}, is often less than D_{in}. The effective compression ratio (Stone and Moore, 1992), CR_{eff}, can be defined as:

$$CR_{eff} = D_{in}/D_{out} \tag{9.1}$$

For example, if D_{in} is 10 dB and D_{out} is 5 dB, the value of CR_{eff} is 2. For a given speed of compression, the value of CR_{eff} decreases with increasing f_m and approaches 1 when f_m is high. In other words, no compression is applied when f_m is high. The value of CR_{eff} only approaches the nominal compression ratio, CR_{static}, for very low modulation rates.

A measure that may be more meaningful than CR_{eff} is the fractional reduction of modulation, f_r (Stone and Moore, 2003a), which is defined as the change in modulation depth produced by the AGC system divided by the original modulation depth. This measure indicates the relative amount of modulation removed by the compressor, and is given by:

$$f_r = \frac{CR_{eff} - 1}{CR_{eff}} \tag{9.2}$$

For example, if CR_{eff} is equal to 2 for a given value of f_m, f_r is equal to 0.5, meaning that the AGC system removed 50 % of the modulation at that value of f_m. If CR_{eff} is equal to 4, f_r is equal to 0.75, meaning that 75 % of the modulation is removed. For the amplitude modulation rates that dominate the envelope of speech, roughly 2 to 32 Hz (Plomp, 1983), a slow-acting compression system (with long attack and release times) typically removes little of the modulation (certainly much less than would be expected from CR_{static}), while a fast-acting system (with short attack and release times) typically removes a substantial amount of the modulation for low modulation rates (up to about 10 Hz), but not for higher rates (Stone and Moore, 2003a; Moore, Stone and Alcántara, 2001).

III.2 VARIETIES OF AUTOMATIC GAIN CONTROL SYSTEMS

AGC systems have been designed in many different forms, mostly on the basis of different rationales or design goals. For reviews, see Moore (1990), Hickson (1994), Dillon (1996) and Kates (2005). This section briefly describes some of the different types.

Some systems are intended to adjust the gain automatically for different listening situations. Essentially, they are intended to relieve the user of the need to adjust the volume control to deal with these situations. Usually, such systems change their gain slowly with changes in sound level; this is achieved by making either t_r, or t_r and t_a, relatively long (usually t_r is a few hundred milliseconds). These systems are often referred to as *automatic volume control* (AVC). AVC systems are described in more detail later on.

AGC is often used primarily to limit the maximum output of hearing aids, to prevent discomfort and/or circuit overload at high input sound levels. The value of CR_{static} is usually very high (10 or more) and the compression threshold is also high (typically around 85 dB SPL). Such systems are known as *compression limiters*. Compression limiters usually have a small value of the attack time, t_a (1–10 ms), so as to respond rapidly to sudden increases in sound level. The recovery time, t_r, is also usually fairly small (20–100 ms). Compression limiters are quite widely used in hearing aids, even though they can adversely affect sound quality for high-level inputs.

An alternative type of compressor, with lower compression ratios and lower compression thresholds, has been used in hearing aids in attempts to make the hearing-impaired person's perception of loudness more like that of a normal listener. These are sometimes called wide dynamic range compressors (WDRC), as the compression operates over a wide range of input sound levels. Typically, the compression threshold is about 45 dB SPL, and CR_{static} has values up to about 3. One aim of such systems is to ensure that the weaker consonant sounds of speech, such as p, t and k, are audible without the more intense sounds (e.g. vowels) being uncomfortably loud. To achieve this goal, the compressors usually have short time constants. The attack times are typically 2–10 ms and the recovery times are typically 20–150 ms. Systems with time constants in this range are often referred to as fast-acting compressors or syllabic compressors, since the gain changes over times comparable to the durations of individual syllables in speech.

III.3 RATIONALES FOR THE USE OF MULTI-BAND COMPRESSION (AND NOISE REDUCTION)

Several authors (Villchur, 1973; Mangold and Leijon, 1979; Laurence, Moore and Glasberg, 1983) have proposed that syllabic compression should be applied separately in two or more frequency bands. The idea is that the signal picked up by the microphone is filtered to split it into several frequency bands, and each band has its own independent fast-acting AGC amplifier. There are at least two reasons why this might be beneficial. First, the amount of hearing loss often varies markedly with frequency; typically, hearing loss is greater at high frequencies than at low frequencies and the dynamic range decreases with increasing frequency. Hence, the amount of compression should vary with frequency, and this requires that compression be applied independently in more than one band. A second reason is that relatively weak high-frequency components in speech (e.g. those associated with k, p, t), which can be important for intelligibility, are often accompanied by, or follow rapidly after, relatively intense low-frequency components, often associated with vowels. This requires more gain at high frequencies than at low. However, other high frequency sounds (e.g. 's' sounds in speech, or the sound of cutlery on plates) can be relatively intense. To prevent these from becoming too loud, the gain at high frequencies must be adjusted rapidly from moment to moment. The use of fast-acting AGC in two or more separate bands can help to make weak high-frequency components audible, even in the presence of intense low-frequencies, while the more intense high-frequency sounds do not become uncomfortably loud. More generally, even when a person has a flat hearing loss, it makes sense to control the gain 'locally' in frequency regions, rather than have the gain at all frequencies be determined by whatever frequency region happens to contain the highest energy at a given time. Examples of the effectiveness of a fast-acting two-channel compression system can be found on a compact disc which is available from the author (Moore, 1997).

One problem that arises with the use of fast compression is that, unless the compression is *very* fast-acting, brief low-level portions of a speech signal may not be made audible. However, increasing the speed of an AGC system can lead to distortion of various types (Souza, 2002), and may have deleterious effects on speech intelligibility in difficult listening conditions (Stone and Moore, 2003a, 2004). Hence, the attack and release times of syllabic compressors are usually chosen as a compromise; they are short enough to give reasonable audibility of most parts of the speech signal, but not so short that they introduce a large amount of distortion.

III.4 RESEARCH ON THE EFFECTIVENESS OF MULTI-BAND SYLLABIC COMPRESSION

This section is mainly concerned with research on the effectiveness of fast-acting (syllabic) multi-band compression. In principle, such compression could restore the loudness perception of a hearing-impaired person roughly to 'normal'; it could be thought of as replacing the fast-acting compression that occurs in a normal cochlea. The compression in the cochlea is lost or reduced when the outer hair cells (OHCs) are damaged, as described in earlier chapters. However, the dynamic aspects of loudness perception are not restored to normal, since, as described earlier, even fast-acting compressors do not affect amplitude modulation depth for the higher modulation rates that occur in sounds such as speech.

Research on the effectiveness of multi-band compression has given somewhat conflicting results. The conflict arises partly from differences in the way that the compression systems have been implemented and partly from differences in methods of evaluation. Individual differences between the subjects used may also have played a role.

Comprehensive reviews of results using multi-channel compression have been provided by Hickson (1994), Dillon (1996) and Souza (2002). Some general trends can be discerned from the results:

1. For speech in quiet, benefits of compression have often been found, in a variety of systems, when the speech materials used have covered a wide range of levels, as occurs in everyday life (Villchur, 1973; Lippmann, Braida and Durlach, 1981; Laurence, Moore and Glasberg, 1983; Moore, Laurence and Wright, 1985; Moore *et al.*, 1992; Moore and Glasberg, 1986a, 1988a; Stone *et al.*, 1999). When the speech material has been presented at one reasonably high level, and when the speech material has been carefully equalized in level during the recording process (as was the case in many studies), compression does not show benefits over linear amplification (Lippmann, Braida and Durlach, 1981).
2. For speech in steady background noise, benefits of fast-acting compression have sometimes (but not always) been found for systems with a small number of bands (Villchur, 1973; Laurence, Moore and Glasberg, 1983; Moore, Laurence and Wright, 1985; Moore *et al.*, 1992; Moore and Glasberg, 1986a, 1988a; Ringdahl *et al.*, 1990). Benefits have not usually been found for systems with a large number of bands (Lippmann, Braida and Durlach, 1981; Bustamante and Braida, 1987), although Yund and Buckles (1995a, 1995b) found some benefit of increasing the number of bands up to eight. For speech presented in a fluctuating background sound with spectral and temporal dips, increasing the number of bands up to eight leads to a small improvement in speech intelligibility (Moore, Peters and Stone, 1999).
3. The extent of the benefit of compression for the understanding of speech in noise depends upon how the frequency-gain characteristic was chosen for the control condition using linear amplification. Under laboratory conditions using speech in noise with carefully controlled levels and fixed spectra, it may be possible to use linear amplification to make the speech audible and comfortable over a wide frequency range. Under these conditions, there may be little benefit of compression. However, if the linear condition is set up so that speech and/or environmental sounds with reasonably high input levels are not amplified to unpleasantly loud levels, then lower overall gains must be used. Under these conditions, benefits of compression may become apparent because the compression allows more amplification for weaker sounds while not amplifying intense sounds excessively.

In summary, clear benefits of multi-band compression have been demonstrated under conditions which are representative of those occurring in everyday life. The benefits have also become clear from consumer acceptance; sales of hearing aids with multi-band compression have increased markedly over the last 20 years, and most hearing aids now have some form of multi-band compression.

III.5 METHODS FOR INITIAL FITTING OF HEARING AIDS WITH MULTI-BAND COMPRESSION

Most researchers and clinicians agree that the fitting of multi-band compression hearing aids should be a two-stage process. In the first stage, an initial fitting is made, usually based on the audiometric thresholds of the patient, and sometimes combined with data on loudness discomfort levels. Fitting methods based on loudness scaling were popular at one time (Pluvinage, 1989), but such methods have largely fallen out of favour, perhaps because of the difficulty of obtaining consistent and reproducible results using loudness scaling (Elberling, 1999). Examples of methods for initial fitting based on the audiogram are NAL-NL1 (Byrne *et al.*, 2001), DSL (Cornelisse, Seewald and Jamieson, 1995; Scollie *et al.*, 2005), CAMEQ (Moore, Glasberg and Stone, 1999) and CAMREST (Moore, 2000). The second stage of fitting involves some form of fine tuning to suit individual needs and preferences.

Of the above methods, all but DSL are based on a loudness model similar to that described in Chapter 4, Section V. For a person with cochlear hearing loss, the specific loudness pattern evoked by a sound depends both on the physical spectrum of the sound reaching the ear (including the effects of any amplification applied to the sound) and on the pattern of hearing loss for that ear. The methods are based on the idea of applying amplification so as to achieve one or more goals in terms of the shape of the specific loudness pattern and the overall loudness. For all methods, the loudness model was used to determine the frequency- and level-dependent gains that would meet the goals as accurately as possible. This was done for a large variety of different audiometric configurations, and the results were used to derive rules relating the gains to the degree of hearing loss at each frequency.

The NAL-NL1 and CAMEQ methods have similar goals. Both aim to give a 'flat' specific loudness pattern over the frequency range 500–5000 Hz for a sound with the long-term average spectrum of speech, and also to give similar overall loudness to what would be experienced by a person with normal hearing. The range 500–5000 Hz covers the most important frequencies for speech perception (ANSI, 1997). For NAL-NL1, the aim is to achieve these goals over a wide range of sound levels, whereas for CAMEQ the aim is to achieve the goals for levels from 65 to 85 dB SPL. For levels below 65 dB SPL, the CAMEQ method aims to optimize the audibility of the speech, which sometimes results in greater loudness than normal at low levels.

The average speech spectrum for low-to-medium levels has more power at low frequencies than at high frequencies (ANSI, 1997). Hence, for a normal ear, the specific loudness pattern evoked by speech at 65 dB SPL has its highest values at low frequencies, around 500 Hz (see the solid line in Figure 9.4). With the gains recommended by CAMEQ, the specific loudness pattern is flatter than normal for low-to-medium input levels (dashed line in Figure 9.4); so the *timbre* is expected to be somewhat 'sharper' or 'brighter' than normal. Speech at 85 dB SPL (shouted

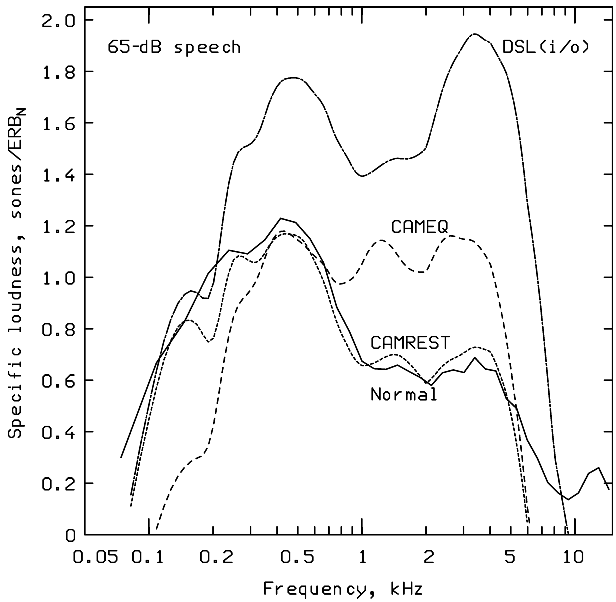


Figure 9.4. The solid line shows the specific loudness pattern evoked in a normal ear by a sound with the same long-term average spectrum as speech at 65 dB SPL. The dashed line shows the pattern evoked in an impaired ear with a typical moderate sloping hearing loss, when the 65-dB speech spectrum is subjected to amplification according to the CAMEQ procedure. The short-dashed line shows the corresponding pattern for the CAMREST procedure, while the dash-dotted line shows the pattern for the DSL(i/o) procedure.

speech) has a spectral shape with relatively more high-frequency energy (ANSI, 1997). As a result, the specific loudness pattern for a normal ear is almost flat for mid-range frequencies (solid line in Figure 9.5). Hence, for 85-dB SPL speech, there is little difference between the specific loudness patterns evoked in a normal ear by unamplified speech and an impaired ear with CAMEQ-prescribed amplification (dashed line in Figure 9.5).

The NAL-NL1 method often prescribes little or no compression at high frequencies, even for a person with a hearing loss that is greatest at high frequencies. This is a side effect of the aim of providing a similar overall loudness to normal for speech over a wide range of sound levels. The minimal prescribed compression at high frequencies seems undesirable, since much research, as reviewed earlier, suggests that compression at high frequencies can increase the range of sound levels over which speech is both comfortable and intelligible for people with high-frequency hearing loss. Also, most hearing-impaired people prefer more compression at high frequencies than at low frequencies (Keidser *et al.*, 2007). CAMEQ prescribes more compression at high frequencies than NAL-NL1.

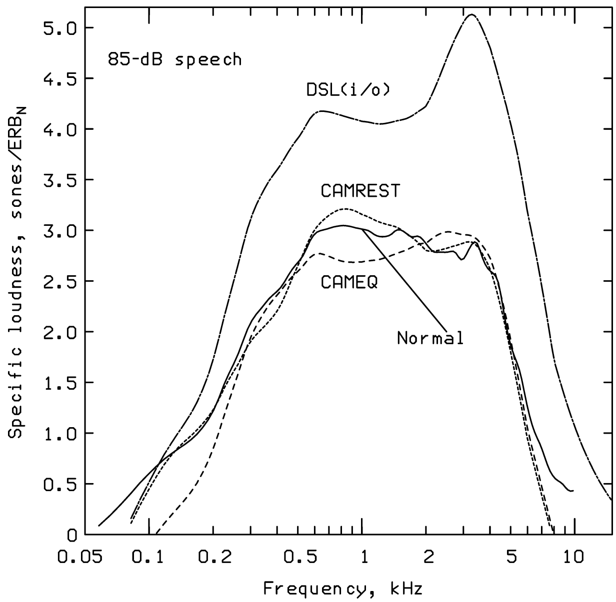


Figure 9.5. As Figure 9.4, but for speech at 85 dB SPL (and with the spectral characteristics of shouted speech).

A second difference between the CAMEQ and NAL-NL1 methods is that NAL-NL1 prescribes less high-frequency gain than required to produce a flat specific loudness pattern when the high-frequency hearing loss is severe or profound. This is based on evidence that the benefit from high-frequency amplification is reduced for such people, an effect which has been called reduced efficiency (Hogan and Turner, 1998) or desensitization (Ching, Dillon and Byrne, 1998). However, such desensitization may mainly occur for people with high-frequency dead regions; see Chapter 8, Section IV. People with severe hearing loss but without dead regions do appear to benefit from high-frequency amplification (Vickers, Moore and Baer, 2001; Baer, Moore and Kluk, 2002). Hence, it seems likely that NAL-NL1 prescribes too little gain at high frequencies for people with severe hearing loss but without dead regions. See below for further discussion of this point.

The other method that was developed using a loudness model is called CAMREST (Moore, 2000). This aims to restore the specific loudness pattern to normal for speech-like stimuli over a wide range of sound levels. In other words, it aims to make both the overall loudness and the timbre of speech similar to normal. The short-dashed lines in Figures 9.4 and 9.5 illustrate the specific loudness patterns evoked by speech at 65 and 85 dB SPL in an impaired ear with CAMREST-prescribed amplification. The patterns are very close to those for the normal ear for mid-range frequencies. Note that, for 85-dB speech, the gains prescribed by CAMREST are similar to those prescribed by CAMEQ.

The remaining procedure for initial fitting, DSL(i/o) (Cornelisse, Seewald and Jamieson, 1995), has the goal of fitting 'the acoustic region corresponding to the "extended" normal auditory dynamic range into the hearing-impaired individual's residual auditory dynamic range' (Cornelisse, Seewald and Jamieson, 1995). The method is based on 'complete' compensation for recruitment, that is restoration of the dynamic range to normal and complete restoration of the audibility of speech sounds. The procedure is widely used for fitting hearing aids to children (Scollie *et al.*, 2000). DSL(i/o) generally prescribes higher gains than NAL-NL1, CAMEQ or CAMREST, and the effect of this can be seen in the specific loudness patterns (dash-dotted lines) in Figures 9.4 and 9.5. According to the loudness model, the DSL(i/o) procedure leads to overall loudness that is greater than normal, with particularly high specific loudness of around 3–4 kHz.

It should be emphasized that none of the procedures takes into account the effect of the speed of the AGC system used. Unless the AGC system operates extremely fast (which is rarely, if ever, the case in practice), the dynamic aspects of loudness will not be restored to normal or near normal. As described earlier, slow modulations in speech will be compressed more than fast modulations.

The CAMREST, CAMEQ and DSL(i/o) methods have been compared in three studies (Moore, Alcántara and Marriage, 2001; Alcántara, Moore and Marriage, 2004; Marriage, Moore and Alcántara, 2004), all of which made use of the Danalogic 163D digital behind-the-ear (BTE) hearing aid, which incorporates fast-acting compression acting independently in fourteen overlapping channels grouped into six adjustable frequency regions. The NAL-NL1 method was not included in these studies as it is only usable with aids having a maximum of four channels. All three studies had the same basic design. Each subject was fitted with one or two aids programmed according to each of the three initial fitting methods. The computer software implementing the methods was used to produce target insertion gains for two input levels (55 and 80 dB SPL) for each of six centre frequencies. These gains are referred to as G55 and G80, respectively. These target gains were then programmed into the aids using the manufacturer's fitting software. All fittings were verified by measuring real-ear gains, using a probe microphone system. Where necessary, the nominal gains in the manufacturer's software were adjusted so as to achieve gains within ±3 dB of the target values. The order of use of each procedure was counterbalanced across subjects.

Directly following the initial fitting, checks were made that the loudness and quality of speech and music stimuli were acceptable. The speech stimuli were recorded samples of male and female talkers speaking with normal and raised effort and presented at levels of approximately 60 and 80 dB SPL, respectively. The music was a piece of jazz with piano, double bass and drums, presented at about 75 dB SPL. The goal was to perform the minimum fine tuning necessary to make the aids acceptable to the subjects. Changes to the fitting were made only if the subject reported that some sounds were uncomfortably loud or too quiet, or that some sounds had an unnatural quality.

Subjects were then asked to wear the aids for one week, after which they returned to the laboratory, when further adjustments were made if requested, based on experiences with the aids during that week. In a few cases, subjects returned after a shorter interval because the fit was not satisfactory. Again, the changes made were the smallest necessary to achieve a satisfactory fit. The difference between the adjusted gains and the *initial* target gains (i.e. those recommended by the procedure) was taken as the main measure of the adequacy of the initial fit for each procedure.

In the first study (Moore, Alcántara and Marriage, 2001), ten subjects with moderate sensorineural bilateral hearing loss were tested. All were experienced hearing-aid users (at least two years of previous aid use, some with compression aids and some with linear aids with output limiting). All subjects were fitted bilaterally for the study. Figure 9.6 shows the mean gain adjustments; these are the differences between the target gains recommended by each procedure and the gains resulting after fine tuning. The smaller these adjustments, the more appropriate is the initial fit. The gain adjustments for both CAMEQ and CAMREST were small at all frequencies for both input levels. This indicates that, on average, these two methods produced initial fittings that did not systematically deviate from preferred fittings. The mean gain adjustments for DSL(i/o) were negative, especially at high frequencies, and more so at the higher input level. This indicates that, on average, the gains prescribed by DSL(i/o) at high frequencies were greater than the preferred gains (although a small part of this effect can be attributed to the need to reduce high-frequency gains to reduce feedback with DSL(i/o) in a few cases).

The second study (Alcántara, Moore and Marriage, 2004) was similar to the first study, except that the hearing aids were fitted unilaterally, in the ear of choice. The pattern of results, shown in Figure 9.7, was very similar to that obtained in the first study, which used bilateral fitting. Since loudness is greater when listening with two ears than when listening with one ear (Moore and Glasberg, 2007), one might expect that patients fitted unilaterally with the CAMEQ or CAMREST procedures would prefer more gain than patients fitted bilaterally, especially for low input sound levels. The results do not reveal any evidence for such an effect. This may indicate that overall loudness per se is not critical in determining initial satisfaction with a hearing-aid fitting. It is possible that satisfaction with gains for low-level inputs is determined mainly by the audibility of weak speech sounds. Absolute thresholds for detecting sounds are only slightly affected by the use of two ears versus one ear. Differences in loudness produced by unilateral versus bilateral aiding should be smaller for high input sound levels than for low input sound levels, as the loudness in the unaided ear usually approaches 'normal' values at high sound levels, as a result of loudness recruitment. Thus, it is not surprising that, for the 85-dB input level, the gain changes needed to achieve satisfactory fittings were similar for the bilateral and the unilateral fittings.

In the third study (Marriage, Moore and Alcántara, 2004), twenty subjects with no previous experience of hearing aids were tested. Ten of these were bilaterally aided; the other ten were unilaterally aided. An initial analysis showed that the gain adjustments did not depend on whether subjects were fitted with one or two aids. This was consistent with the results found for experienced users. Hence, data were collapsed across subjects fitted unilaterally and bilaterally. Figure 9.8 shows the mean gain

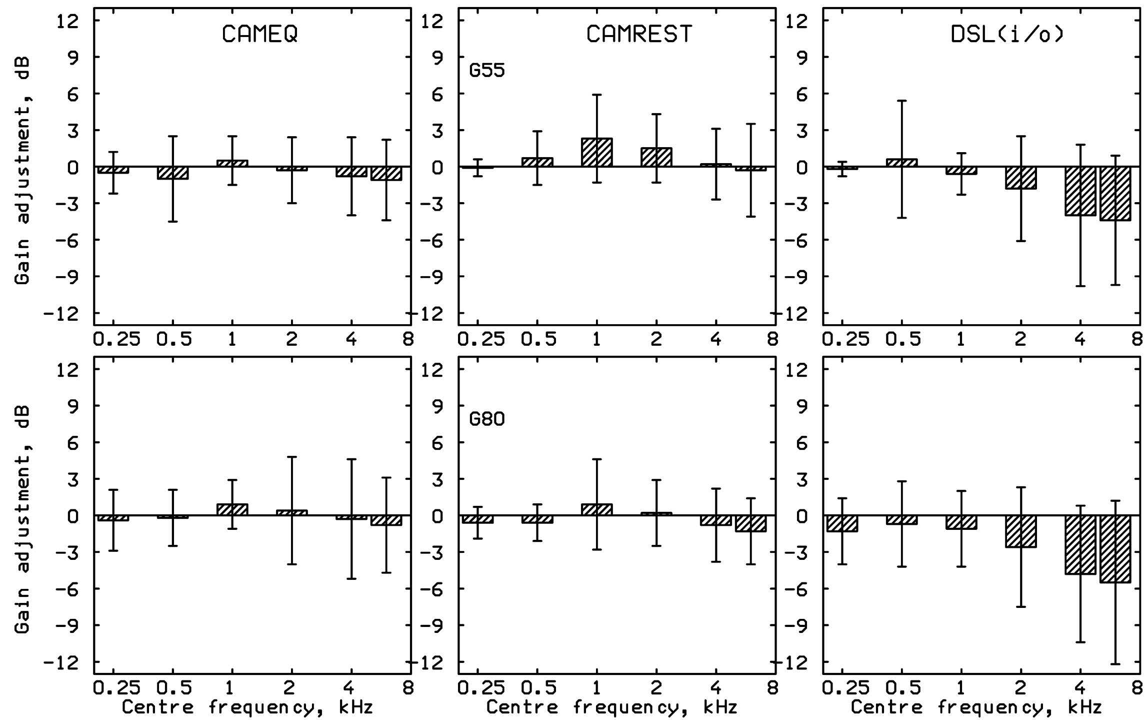


Figure 9.6. The mean gain adjustment for experienced subjects fitted bilaterally. These are deviations between the target gains recommended by each procedure and the gains resulting after minimal fine tuning. The smaller these adjustments, the more appropriate is the initial fit. Error bars show ± one standard deviation across subjects.

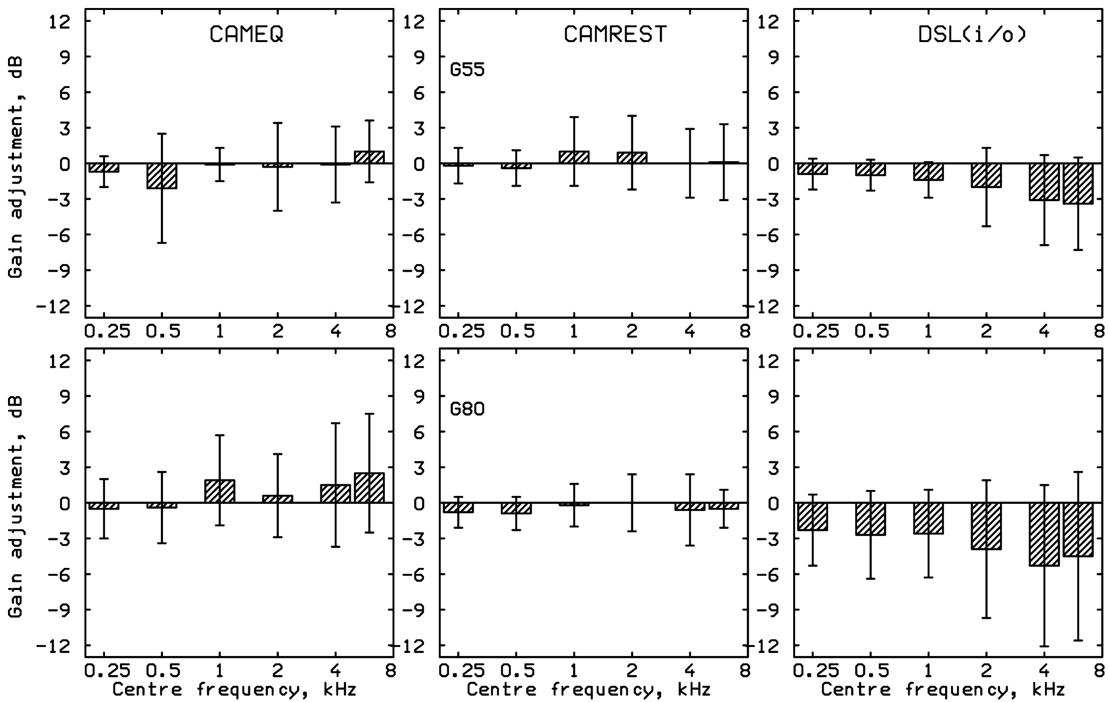


Figure 9.7. As Figure 9.6, but for experienced users fitted unilaterally.

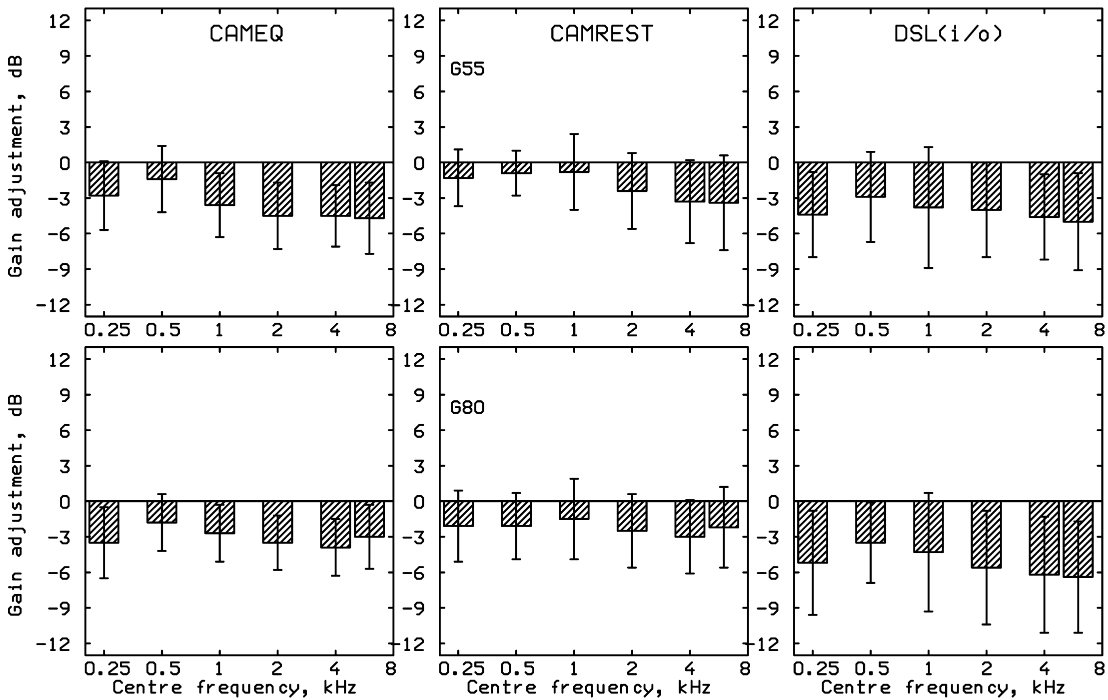


Figure 9.8. As Figure 9.6, but for inexperienced users fitted unilaterally or bilaterally.

adjustments. On average, gain reductions were required for all three procedures, and the reductions were especially large for DSL(i/o). The reductions for CAMEQ and CAMREST contrast with the results for the experienced groups, shown earlier.

Overall, these results suggest that, on average, the gains prescribed by CAMEQ and CAMREST are comparable to the gains preferred by experienced hearing-aid users. However, some individuals prefer more gain than prescribed and some prefer less gain. Inexperienced users prefer less gain than prescribed by CAMEQ or CAMREST, but the amount of gain adjustment required to achieve satisfactory fits is relatively small, typically being less than 3 dB and rarely exceeding 5 dB.

A study of Smeds (2004) supports the use of lower gains than prescribed by CAMEQ for inexperienced users. She compared two methods for fitting fast-acting multi-channel compression, using the Danalogic 163D hearing aid. One method, called NormLoudn, was based on an early version of CAMEQ. The other method, called LessLoudn, was based on the manufacturer's fitting method, and gave less than 'normal' overall loudness. Twenty-one first-time hearing-aid users were fitted with each method. Preference in everyday life was evaluated using interview, questionnaire and diary. Laboratory tests included paired-comparison judgements of preference and loudness and a speech-recognition test. Nineteen out of the twenty-one subjects stated that they preferred LessLoudn in everyday life. Also, the questionnaire and the diary showed a clear preference for LessLoudn in all types of listening situations. Paired comparisons of preference in the laboratory showed that LessLoudn was preferred to NormLoudn in all situations tested, except for soft speech in low-level noise, where there was no significant preference for one method over the other. Speech-recognition scores were similar for the two fittings. Overall, the results suggested that new users preferred gains that were 3–5 dB lower than prescribed by CAMEQ.

III.6 METHODS FOR FINE TUNING HEARING AIDS WITH MULTI-BAND COMPRESSION

It is generally accepted that even the best initial fitting methods do not give satisfactory fittings for all clients. Adjustments are often necessary to suit individual needs and preferences. However, there is no standard method for making such adjustments. Adaptive methods have been described, based on judgements of the loudness and tone quality of speech and music signals presented over a range of input levels, but it has been found that the fittings obtained with such methods have relatively poor reproducibility (Moore *et al.*, 1999, 2005).

One approach to fine tuning is the concept of the 'trainable' hearing aid (Elberling and Hansen, 1999; Zakis, 2003; Dillon *et al.*, 2006). The basic idea is that the user adjusts the gain and/or frequency response of the hearing aid, preferably via a remote control, to achieve satisfactory loudness and sound quality in each of several listening situations. Over time, the aid learns to recognize both each specific situation and the preferred setting for that situation, and the aid then adjusts itself automatically to achieve these settings. To my knowledge, such a system is not yet available in commercial hearing aids.

III.7 SLOW-ACTING AUTOMATIC GAIN CONTROL SYSTEMS

Although benefits of fast-acting multi-band compression have been demonstrated, it is not clear that such compression is the best way to compensate for the effects of loudness recruitment. There is certainly some evidence that the use of high compression ratios (greater than 2 to 3) in such systems can have deleterious effects on speech intelligibility in quiet and in noise (Plomp, 1994; Crain and Yund, 1995; Hohmann and Kollmeier, 1995; Verschuure and Dreschler, 1996; Souza, 2002). For people with severe to profound hearing loss, it may be necessary to use high compression ratios to make sure that speech is both audible and comfortable over the wide range of sound levels encountered in everyday life. However, such high compression ratios are not optimal for intelligibility at moderate input levels when fast compression is used.

As described earlier, it has often been found that, for speech presented at a single level, an optimally adjusted linear amplifier gives speech intelligibility at least as good as, and sometimes better than, that provided by multi-band compression amplifiers. Therefore, an alternative approach to the problem of compensating for reduced dynamic range is to use slow-acting AVC to deliver speech at a single level, or over a small range of levels, regardless of the input level. However, conventional AVC systems involve a compromise between conflicting requirements. For speech with a fixed average level, the level may fluctuate markedly from moment to moment, and may drop to very low values during brief pauses in the speech. If the gain of an aid changes significantly during the speech itself, or during the pauses in the speech, then 'breathing' or 'pumping' noises may be heard, which are objectionable to the user. These effects are particularly marked when moderate levels of background noise are present, as is often the case in everyday situations. The noise appears to come and go. It 'rushes up' during pauses in the speech, and becomes quieter when the speech starts again. In addition to these objectionable effects, the temporal envelope of the speech may be distorted by the changes in gain (Stone and Moore, 2007). To avoid these problems, the gain should change relatively slowly as a function of time, that is t_a and t_r should be long.

On the other hand, it is important to protect the user from sudden intense sounds, such as a door slamming or a cup being dropped. This requires the gain to change much more rapidly. This problem is usually dealt with by having an AVC system with a fast attack time (t_a in the range 1–10 ms), and a longer recovery time (t_r in the range 300–2000 ms). The fast attack time provides protection against sudden increases in sound level; the gain drops very quickly when the input sound level suddenly increases. However, such a system has the disadvantage that the gain of the aid drops to a low value immediately after an intense transient; the aid effectively goes 'dead' for a while. A further problem is that a recovery time of a few hundred milliseconds is not sufficiently long to prevent 'breathing' and 'pumping' sounds from being heard.

An AVC system developed in my laboratory provides a better solution to these problems (Moore and Glasberg, 1988a; Moore, Glasberg and Stone, 1991; Stone *et al.*, 1999). This system, referred to as dual-front-end AGC, is illustrated in Figure 9.9. The signal picked up by the microphone (1) is fed to an AGC amplifier (2) whose gain

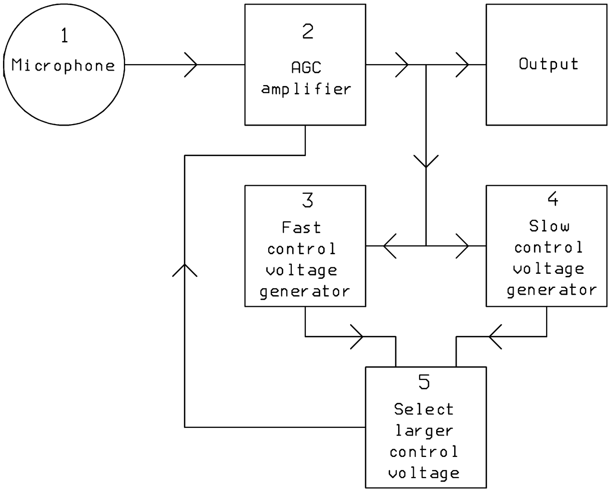


Figure 9.9. A block diagram of the dual-front-end AGC system.

is determined by a control voltage. There are two control voltage generators (3 and 4). One of the control voltage generators (4) has a lower compression threshold than the other one, and it changes only slowly in response to changes in sound level. Normally, the operation of the AGC amplifier is determined by this slow-acting control voltage. The effect is that the gain changes slowly as the listening situation changes, in a way which is almost unnoticed by the user and which avoids audible 'breathing' noises. The fast-acting control voltage generator (3) has a compression threshold about 8 dB higher than that of the slower control voltage generator. Normally, it has little effect on the operation of the AGC amplifier. However, if an intense sound suddenly occurs, the fast-acting control voltage increases rapidly. When it exceeds the control voltage generated by (4), it is selected by circuit (5) to control the gain of the amplifier. The gain is rapidly reduced, thus protecting the user from over-amplification of the intense sound. However, the fast-acting control voltage decreases rapidly following the cessation of the intense sound, and control returns to the slow-acting control voltage. If the intense sound is brief, it does not significantly change the value of the slow-acting control voltage. Thus, following an intense transient, the gain returns to the original value determined by the overall level of the speech. In this way, intense transients are selectively attenuated, but the overall gain for the speech is held almost constant, except for a very brief period following the transient. Note that, although the transient is attenuated, it remains clearly audible, and can be identified, for example as a door slamming or a cup being dropped. A system similar to this has been incorporated in several commercial hearing aids and in cochlear implants. In some cases, dual time constant systems have been implemented independently in several frequency bands.

An alternative approach is Adaptive Dynamic Range Optimization (ADRO; Blamey, 2005). ADRO processing is used in both cochlear implants (see below) and hearing aids to control the level of sounds independently in a large number (usually 32 or 64) of narrow frequency bands. The distribution of levels is estimated in each band and expressed as percentiles. For example, the 90th percentile is the level that is exceeded 10 % of the time and the 30th percentile is the level that is exceeded 70 % of the time. These percentiles provide measures of the dynamic range of the input signal that are compared with the dynamic range of the listener's hearing. The latter is estimated for each centre frequency (each band) using measures of threshold (T) and loudness (with interpolation to determine values for centre frequencies between the standard audiometric frequencies). There are three target levels for each centre frequency: the maximum output level (denoted M), the comfort level (denoted C) and the audibility level (denoted A), which may be defined as C – 20 dB or T, whichever is the greater. The gain is usually slowly adapted for each centre frequency so as to satisfy a series of 'rules'. The rules are prioritized, and 'fuzzy logic' is used to decide which rule should 'win' when the results of the different rules are incompatible. Typical examples of the rules used in hearing aids are:

1. The output level of each band should always be below the M level. Fast output compression limiting is used to achieve this, when necessary.
2. The 90th percentile should be below the C level. The gain is reduced slowly if this rule is not satisfied, to keep sounds comfortable.
3. The 30th percentile should be above the A level. The gain is increased slowly if this rule is not satisfied, to keep sounds audible.
4. There is a limit to the gain applied in each frequency band to avoid over-amplification of low-level background noise.

The maximum rate of decrease of gain is about 9 dB per second, while the maximum rate of increase of gain is about 3 dB per second. This difference allows the gain to be decreased reasonably quickly in response to increases in sound level, but avoids increases in gain during brief pauses in speech, thus avoiding the 'pumping' noises described earlier. When all four rules are satisfied, the gain remains unchanged and the ADRO acts in a similar way to an optimized linear amplifier. Evaluations of ADRO have shown that it leads to high listening comfort and good speech intelligibility over a wide range of speech levels and for a wide range of types of hearing loss (Blamey, 2005).

III.8 COMPARISONS OF SLOW-ACTING AND FAST-ACTING SYSTEMS

Studies comparing slow-acting and fast-acting AGC systems have given mixed results. Stone *et al*. (1999) compared the slow-acting dual-front-end AGC system with fast-acting four-channel compression. Both algorithms (along with two others) were implemented in a wearable digital hearing aid. Eight subjects with moderate-to-severe cochlear hearing loss were tested in a counter-balanced design. The characteristics of the compression were adjusted to suit the individual subjects. Subjects had at least two weeks' experience with each system in everyday life before evaluation using the APHAB questionnaire (Cox and Alexander, 1995) and using measures of speech intelligibility in quiet and noise. The APHAB scores did not indicate clear differences between the two systems. Scores for the words in quiet were high for both systems at both 50 and 80 dB SPL. The speech-to-noise ratios required for 50 % intelligibility were low (indicating good performance) and similar for the two systems.

Hansen (2002) examined the influence of release times from 40 to 4000 ms on both subjectively assessed speech intelligibility and sound quality, using a simulated 15-channel hearing aid. Various everyday life sounds were recorded binaurally through hearing aid microphones placed in BTE hearing-aid cases. The different compression settings were assessed using these sounds in a paired-comparison procedure. Both normally hearing and hearing-impaired subjects were tested. Gains were adjusted so that 'normal' speech received the insertion gain recommended by the NAL-R formula (Byrne and Dillon, 1986). In the first experiment, CR_{static} was fixed at 2.1 : 1, the compression threshold was set to a very low value and only the time constants were varied. All subjects showed a significant preference for the longest release time (4000 ms) over the two shorter release times (400 ms and 40 ms), for both quality and speech intelligibility. In the second experiment, various combinations of release time, CR_{static} and compression threshold were used. The hearing-impaired subjects showed a significant preference for the hearing-aid setting with a long release time (4 s) and a low compression threshold (20 dB SPL), for both quality and speech intelligibility. With a short release time (40 ms), a lower compression threshold (20 dB SPL) was preferred over a higher threshold (50 dB SPL). For the normally hearing subjects, the same settings were preferred for speech intelligibility, but no significant differences were found for sound quality.

Gatehouse, Naylor and Elberling (2006a) compared two-channel fast-acting compression, two-channel slow-acting AVC and a linear system, all implemented in the same hearing aid. Fifty hearing-impaired subjects were tested, each of whom experienced each system for a 10-week period. Outcome measures included speech intelligibility under diverse conditions and self-reported disability. Overall, each compression system was superior to the linear system in terms of listening comfort, satisfaction, self-reported intelligibility and measured speech intelligibility. The slow-acting system was given better ratings for listening comfort than the fast-acting system, but for reported and measured speech intelligibility the fast-acting system gave better results. There was no overall difference in satisfaction between the slow-acting and fast-acting systems. However, some individual subjects did show a distinct preference for one compression system over the other. Gatehouse, Naylor and Elberling (2006b) suggested that these preferences were related to the 'auditory ecology' of the subjects, for example the range of sound levels that they encountered in everyday life, and how often they were exposed to rapid variations in sound level.

III.9 GENERAL CONCLUSIONS ABOUT COMPRESSION

Compression can be beneficial in two ways. First, it can allow speech in quiet to be understood over a wide range of sound levels without adjustment of the volume control and without the speech ever becoming uncomfortably loud. This can be achieved either by using a slow-acting AVC system, such as the dual-front-end AGC or ADRO, or by using multi-channel syllabic compression. Many people find it unnecessary to use the volume control with hearing aids incorporating compression. Indeed, some compression aids do not have volume controls. Secondly, for speech at low to medium levels, compression can improve the intelligibility of speech in background noise, presumably by making speech cues more audible.

There is little evidence to support the idea that the benefits of compression accrue from restoring the perception of loudness to ‘normal’. Indeed, if fast-acting compression is set up to do this, hearing-aid users often complain that everything sounds too ‘noisy’. In practice, the aids must be set up so as to ‘undercompensate’ for the loudness recruitment. Compression sufficient to restore loudness perception to ‘normal’ sometimes has deleterious effects on speech intelligibility and is sometimes not liked by users (Moore, Lynch and Stone, 1992; Plomp, 1994). It seems likely that the benefits of compression arise mainly from the fact that compression increases audibility while avoiding discomfort from loud sounds.

IV SOME GENERAL PROBLEMS WITH HEARING AIDS

This section briefly discusses some problems that are inherent to most hearing aids.

IV.1 INADEQUATE GAIN AT HIGH FREQUENCIES

A common problem is inadequate gain at high frequencies relative to low or medium frequencies. For example, the gains achieved at high frequencies using the manufacturer’s fitting software may be less than the gains entered into the software (Hawkins and Cook, 2003; Aazh and Moore, 2007b). This is particularly the case when fitting people with hearing losses that increase rapidly at high frequencies. It is now relatively common (and is recommended practice) to check the real-ear gain of hearing aids using a probe-tube microphone whose tip is placed inside the ear canal, close to the eardrum (Mueller, Hawkins and Northern, 1992). This can reveal ‘errors’ in fitting. The high-frequency gain can sometimes be increased electronically, using frequency-response shaping circuitry in the hearing aid. However, often it is necessary to resort to changes in the ‘plumbing’, for example by changing the tubing leading from a BTE aid to the earmould, or by the use of acoustic resistors (dampers) in the tubing. However, even after such modifications, the obtained frequency-gain characteristic may differ substantially from the one recommended by a prescriptive rule. Inadequate high-frequency gain is common when BTE aids are used, although the gain at high frequencies (and the effective frequency range) can be improved markedly by placing the receiver (the miniature loudspeaker that generates the sound) in the ear canal.

Sometimes it is impossible to achieve adequate gain at high frequencies because of problems with acoustic feedback, which is discussed in the next section, although this problem has been reduced with improvements in methods for reducing feedback. Indeed, the microphones and receivers (miniature loudspeakers) used in hearing aids often have a response which is deliberately restricted at high frequencies, to prevent acoustic feedback. Thus, many hearing aids produce little or no amplification above 5 to 6 kHz. This is not necessary or desirable from an auditory point of view. Speech intelligibility in quiet and in noise can be improved for people with mild or moderate sensorineural hearing loss by providing amplification for frequencies up to at least 7 kHz (Skinner and Miller, 1983; Vickers, Moore and Baer, 2001; Baer, Moore and Kluk, 2002; Hornsby and Ricketts, 2006). Also, increasing the frequency range may lead to a more effective use of cues to localization provided by the pinnae (see Chapter 7, Sections IV.3 and IV.4). Finally, provided that the distortion of the hearing aid is kept low, a wide frequency range does not lead to a harsh or tinny sound quality (Killion, 1993).

An exception occurs for people with dead regions at high frequencies. For such people, high-frequency tones that fall well within the dead region are often reported to sound highly distorted or noise-like (Huss and Moore, 2005a). In such cases, it appears that speech intelligibility is better if amplification is provided for frequencies up to about $1.7f_e$ (where f_e is the edge frequency of the dead region), but not for higher frequencies, as described in Chapter 8, Section IV.

IV.2 ACOUSTIC FEEDBACK

Feedback occurs when the sound generated by a hearing aid leaks back to the microphone of the aid. The sound is then picked up by the microphone and amplified further. This sets up a self-sustaining oscillation. It is usually heard as a ‘whistling’ sound. Feedback is usually unpleasant to listen to, and it may be accompanied by considerable distortion. In practice, feedback often limits the amount of gain that can be used in a hearing aid, especially at medium to high frequencies. Sometimes, the user of an aid does not hear feedback (e.g. when it occurs at a high frequency where the audiometric loss may be severe), but becomes embarrassed about wearing the aid because they are aware that others can hear the feedback.

Feedback can be reduced in a number of ways. One way is to improve the seal of the aid or earmould to the ear canal so that less sound leaks from the receiver to the microphone. This can be done by making a better-fitting aid or mould, by smearing the ear canal or earmould with a substance like petroleum jelly before inserting the aid or by using a soft foam tip fitted over the earmould or hearing aid. For a BTE aid, increasing the thickness of the tubing leading from the aid to the earmould may reduce feedback. Another method is to increase the distance between the microphone and the receiver, for example by placing the receiver in the ear canal. A third way of reducing feedback is by smoothing the frequency response of the aid. Feedback usually occurs at one specific frequency where there is a peak in the frequency-gain characteristic, or in the response of the transmission path from the ear canal to the microphone, or both. If the frequency-gain characteristic is smoothed, then the overall gain can often be increased before feedback starts again. The frequency response can be smoothed either by electronic filtering or by modifications to the earmould and tubing, for example by the use of acoustic resistors.

Many modern hearing aids achieve feedback reduction in a very different way, using digital signal processing. Often, the aid estimates the characteristics of the feedback path from the receiver to the microphone and then constructs a digital filter which is used partially to cancel the feedback signal. In some aids, the digital filter adapts over time; so it can allow for changes in the feedback path caused, for example, by slight changes in the position of the earmould or by bringing a telephone to the ear. Modern systems of this type typically allow an increase in high-frequency gain of about 10–18 dB before feedback occurs (Freed and Soli, 2006).

Feedback-reduction systems have allowed greater gain to be achieved with vented earmoulds, or even with 'open' fittings, where the ear canal is essentially left unoccluded. However, adaptive digital feedback systems can sometimes lead to problems when the sounds being picked up by the hearing-aid microphone contain strong tonal components. This can happen, for example, when listening to music. The tonal components may be 'interpreted' by the aid as feedback, and the attempt to cancel the components can lead to considerable distortion. Also, it is sometimes possible to hear ringing sounds or tones *after* an external tonal sound has been turned off.

IV.3 PEAKINESS OF FREQUENCY RESPONSE

The frequency response of a hearing aid on a real ear often shows two or three distinct peaks caused by resonances in the acoustical delivery system. This is especially true for BTE aids. A single broad peak around 3 kHz is desirable, since this mimics the normal response of the outer ear; see Chapter 1, Figure 1.3. However, additional peaks are often present. It is possible to reduce these peaks, smoothing the overall frequency response, by suitable modifications to the tubing and/or by the use of acoustic resistors (Libby, 1981; Killion, 1982). It is now widely accepted that the frequency response should be smoothed in this way as far as possible. While one study showed no significant differences in judged clarity and pleasantness between hearing aids with unsmoothed and smoothed responses (Cox and Gilmore, 1986), most other studies have shown that hearing aids with smoothed frequency responses are preferred (Libby, 1981; Mertz, 1982; van Buuren, Festen and Houtgast, 1996). For example, van Buuren, Festen and Houtgast (1996) found that large peaks in the frequency response adversely affected intelligibility (more for impaired than for normal listeners) and subjective sound quality. Smooth responses gave the best results for intelligibility and pleasantness.

In any case, there are at least three benefits of smoothing the frequency response other than effects of sound quality:

1. It can reduce acoustic feedback, as described above.
2. It can reduce the distortion (including temporal distortion produced by rapid phase changes) that often occurs at frequencies around peaks in the response.
3. It can allow a greater proportion of the speech spectrum to be above threshold before the ULL is reached.

In spite of these benefits, in clinical practice measures to smooth the frequency response are often not taken. This happens either because of ignorance or laziness on the part of the hearing-aid dispenser, or from the mistaken belief of the dispenser that there is no benefit in smoothing the response. It is to be hoped that this state of affairs will improve in the future, especially with the increasing use of real-ear probe tube measurement systems.

IV.4 THE OCCLUSION EFFECT

People who wear hearing aids often complain that their own voice sounds unnatural, being excessively loud and 'hollow-sounding'. The effect occurs when an earmould is worn that blocks the entrance to the ear canal, and it is called the *occlusion effect*. The sound of the person's own voice is transmitted through the bones of the head into the ear canal (meatus). The transmission occurs mainly through the flexible (cartilaginous) outer part of the ear canal. When the ear canal is blocked, the sound cannot escape, and the resulting sound pressure at the eardrum is markedly higher than would occur for an unblocked ear. Using a probe tube microphone, Killion, Wilber and Gudmundsen (1988) showed that sound levels exceeding 90 dB SPL could occur for speech sounds such as 'a' as in car. An illustration of this effect is available on a compact disc available from the author (Moore, 1997).

One way of alleviating the occlusion effect is to vent the earmould, by drilling a hole through it. Also, it is becoming increasingly popular to use 'open' fittings, in which the ear canal is hardly occluded; often the sound is fed into the ear canal via a small inconspicuous tube with no earmould at all. This largely eliminates the occlusion effect. Acoustic feedback can be avoided in such fittings by the use of digital feedback reduction, as described above, provided that a large amount of amplification is not required. With such open fittings, it is not usually possible to apply a significant amount of gain at low frequencies (Kates, 2005).

An alternative way of reducing the occlusion effect is based on observations by Zwislocki (1953). He showed that the transmission of bone-conducted sound into the ear canal was much less when an earplug was inserted deeply into the ear canal than when it was inserted only into the flexible outer part. The inner part of the ear canal is composed of rigid bone, which accounts for the reduced transmission. Zwislocki's observations have been confirmed by Schroeter and Poesselt (1986) and by Killion, Wilber and Gudmundsen (1988), who showed that the sound level in the ear canal produced by a person's own voice could be reduced by as much as 20 dB at low frequencies by using a deeply seated earplug or earmould, as compared to a shallow one. This effect is also illustrated on the compact disc referred to above (Moore, 1997). This finding can be used to reduce the occlusion effect in three ways:

1. By using an aid that is fitted deeply inside the ear canal (all-in-the-ear devices).
2. By using an earmould that extends deeply into the ear canal.
3. By using a receiver in a soft casing that is fitted deeply in the ear canal.

IV.5 TIME DELAYS

Digital hearing aids can produce significant delays of the audio signal, ranging from about 3 ms up to 10 ms. These delays can have a number of disturbing effect. First, the user of a hearing aid hears their own voice both via the bones of the head (bone conduction) and, after a delay, via the hearing aid. This can affect the perception of their own voice, making it sound 'coloured' or as if there were an echo (Stone and Moore, 1999, 2005). The delay can also have effects on speech production, causing a slowing of speech rate (Stone and Moore, 2002, 2003b). However, both of these effects are small when the delay is less than 20 ms, so these effects are not of great importance given the delays that typically occur in hearing aids.

A more serious problem occurs when the delay varies across frequency. This may happen because of the signal processing in the aid (Kates, 2005) or because of the use of a large vent or an open fitting. In the latter case, the low frequencies 'leak' directly into the ear canal and are not delayed, while the high frequencies are amplified by the aid and are heard after a delay. Stone and Moore (2003b) showed that across-frequency delays of more than 9 ms could produce disturbing effects during speech production and could adversely affect sound quality. Delays of this magnitude do occur in some digital hearing aids.

A final disturbing effect may occur when the delay differs across the two ears. Large interaural delays can occur when an aid is fitted unilaterally, and this may disrupt sound localization based on ITD cues (see Chapter 7, Section II.1). Consistent with this expectation, the ability to determine the azimuth of sounds is poorer when one hearing aid is worn than when two aids are worn (Kobler and Rosenhall, 2002), although this effect may partly be a result of the disruption of ILD cues.

Also, users of a unilaterally fitted hearing aid show biases in their localization judgements (Simon, 2005). Even when hearing aids are fitted bilaterally, the delay produced by the aids may differ across the ears. This may initially disrupt sound localization, but listeners appear to acclimatize to the difference in ITD, so that, after a few weeks of experience, localization is not adversely affected (Drennan *et al.*, 2005).

V METHODS FOR IMPROVING THE SPEECH-TO-NOISE RATIO

V.1 MULTI-CHANNEL NOISE REDUCTION

Consider a hypothetical situation where a hearing-impaired person is trying to understand speech in the presence of an intense noise whose energy is concentrated in a narrow frequency range (a rather rare circumstance in everyday life). The speech-to-noise ratio can be improved by filtering the input signal into a number of bands and reducing the gain for any band that is dominated by noise (Ono, Kanzaki and Mizoi, 1983; van Dijkhuizen, Festen and Plomp, 1991; Rankovic, Freyman and Zurek, 1992). Several commercial hearing aids incorporate a form of noise reduction based on this rationale. These aids have methods for determining whether a particular frequency band is dominated by speech or by noise. Usually, these methods are based on the amount or rate of amplitude modulation in that frequency band; speech is a highly modulated sound with modulation concentrated at low rates (Plomp, 1983), whereas background noises tend to be less modulated and to have higher modulation rates. If a band is dominated by noise, then the gain in that band is reduced. Such systems have not been shown to improve the intelligibility of speech, although they can sometimes improve listening comfort (Alcántara *et al.*, 2003; Ricketts and Hornsby, 2005).

V.2 DIRECTIONAL MICROPHONES

The intelligibility of speech in background noise can be improved by increasing the signal-to-noise ratio. As described in Chapter 8, Section II, under difficult listening conditions, each 1-dB improvement in speech-to-noise ratio typically gives an increase of intelligibility of 7 to 19 %. One of the few ways of achieving a true improvement in speech-to-noise ratio is to use a directional microphone. Some hearing aids have omni-directional microphones that respond equally to sounds from all directions. In contrast, a directional microphone is more sensitive to sounds coming from certain directions than to sounds coming from other directions. In the case of a hearing-aid microphone, it is assumed that the user will normally face towards the desired sound source, and so the microphone is designed to be more sensitive to sounds coming from the front than to sounds coming from the sides or back. In practice, the directional pattern is altered by placement of the microphone close to the head and body, and the direction of maximum sensitivity is displaced somewhat away from straight ahead. For example, if the microphone is placed above the left ear, the direction of maximum sensitivity is usually at an azimuth of about 20–45 $^\circ$, that is 20–45 $^\circ$ to the left of straight ahead.

The *directivity* of a microphone is a measure of the sensitivity to sounds coming from the direction of maximum response relative to the sensitivity to sounds from other directions. In principle, a higher directivity gives a greater improvement in speech-to-noise ratio whenever the noise comes from one or more directions other than the desired direction. However, the beneficial effects of directional microphones are reduced in reverberant rooms, since, in such rooms, some of the interfering sound comes from the frontal direction.

Simple directional microphones have been used in hearing aids for many years. The directivity of early systems was not high, but directivity has improved in recent years. Many modern systems are adaptive; the pattern of directivity is continually adjusted so as to minimize the most prominent interfering source. In some aids, this adaptive operation is performed independently in several different frequency bands. As a result, in a situation where the target speech comes from in front (0 $^\circ$ azimuth) and one or more interfering sounds come from the side or the rear, the intelligibility of the target speech can be considerably improved compared to what is obtained with omni-directional microphones or unaided listening (Ricketts and Henry, 2002; Wouters, Berghe and Maj, 2002; Bentler, Palmer and Mueller, 2006; Blamey, Fiket and Steele, 2006). However, a possible disadvantage of adaptive directional microphones operating independently at the two ears is that they can disrupt cues to sound localization when hearing aids are fitted bilaterally (Van den Bogaert *et al*., 2006).

Directional microphones are most commonly used with BTE hearing aids. However, a disadvantage of such aids is that they do not allow effective use of sound localization cues provided by the pinna; see Chapter 7, Section IV.3. Hence, the benefit of the directional microphone comes at the cost of a loss of some localization cues.

V.3 BINAURAL PROCESSING ALGORITHMS

Kollmeier and his co-workers have described several methods of processing sounds that are based on the use of binaural cues (interaural level differences, ILDs, and interaural time differences, ITDs) to enhance speech-to-background ratios (Kollmeier, Peissig and Hohmann, 1993; Kollmeier and Koch, 1994; Wittkop, Hohmann and Kollmeier, 1996). It is assumed that the user of the system wears microphones mounted in or close to their ear canals and that audio signals can be transmitted from one hearing aid to the other. The simplest of their processing methods works in the following way. The sound is split into a large number of frequency bands. The ITD and ILD within each band are determined on a moment-by-moment basis. If the ITD and ILD are small within a given band, then the signal within that band probably came from directly in front of the head (although it could in fact come from any direction in the median plane; see Chapter 7, Section IV.1). In that case, the signal in that band is passed unaltered. If the ITD and/or ILD are large within a given band, that indicates that the signal in that band is dominated by sound coming from a direction that is off to one side. In this case, the signal in that band is attenuated. In practice, the amount of attenuation is related to the magnitudes of the ITDs and ILDs, and the attenuation is made to vary smoothly over time and across frequency bands. The overall effect of the processing is that sounds from the frontal direction are preserved, while sounds from other directions are attenuated.

Evaluations of this system (Kollmeier, Peissig and Hohmann, 1993) showed that it could give significant improvements in the intelligibility of speech in a ‘cocktail party’ situation (with several interfering speakers at various angles), provided that there was no reverberation; the improvements were roughly equivalent to those produced by a 5-dB change in speech-to-background ratio. However, the performance of the algorithm worsened when reverberation was present.

Several more complex schemes have been developed and evaluated, with promising results (Kollmeier, Peissig and Hohmann, 1993; Kollmeier and Koch, 1994; Wittkop, Hohmann and Kollmeier, 1996). However, the schemes are computationally intensive, and they introduce time delays in the signal that may be unacceptable; see Section IV.5. Further evaluations are necessary to assess how well such schemes may work in everyday situations.

VI TRANSPOSITION AIDS FOR SEVERE AND PROFOUND HEARING LOSS

People with severe or profound hearing loss at high frequencies often get little or no benefit from conventional hearing aids, possibly because many of them have dead regions at high frequencies (Moore *et al.*, 2000; Moore, 2001). One possible way of providing benefit to such people is by the use of frequency transposition, whereby inaudible high-frequency components in the input signal are moved downwards in frequency (transposed) so as to make them audible. Hearing aids based on transposition have been available for several years, but have met with only limited success (Johansson, 1966; Velmans and Marcuson, 1983; McDermott *et al.*, 1999; McDermott and Dean, 2000; McDermott and Knight, 2001).

There may be several reasons for the limited success. First, the signal processing used to perform the transposition may result in artefacts and distortion of various kinds. Secondly, the transposed frequency components sound unnatural, and it may take a long time for the user to become accustomed to the sound of the transposition and to learn to make effective use of the transposed information. Thirdly, in some forms of processing, the transposed components are superimposed on lower-frequency non-transposed components, which may lead to masking. Finally, transposition devices have usually been fitted without any definite knowledge of whether or not the patient had a dead region, or of the extent of any dead region.

Recently, Robinson, Baer and Moore (2007) evaluated a transposition method using subjects with high-frequency dead regions, the edge frequency of which, f_e, was diagnosed using the threshold-equalizing noise (TEN) test and determined more precisely using *psychophysical tuning curves*, as described in Chapter 3, Sections VIII.2 and IX.1. Subjects were chosen with values of f_e close to 1000 Hz. The processing was adapted to the value of f_e for each subject. Frequency components from $2f_e$ to $2.7f_e$ were transposed to the range f_e to $1.7f_e$. Frequency components below $1.7f_e$ were amplified, but unaffected by transposition. Transposition only occurred if the ratio of high-frequency to low-frequency energy was sufficiently high; this is called conditional transposition. As a result, transposition mainly occurred when fricative sounds, like ‘s’ and ‘f’, were present. The conditional transposition also prevented high-frequency background noise of moderate level from being transposed.

The transposition processing was compared with a control condition using linear amplification for frequencies up to $1.7f_e$, and attenuation above that. It was assumed that this would lead to the best intelligibility that could be achieved with linear amplification (Vickers, Moore and Baer, 2001; Baer, Moore and Kluk, 2002). All tests were laboratory based. In one test, consonant discrimination was assessed using vowel–consonant–vowel (VCV) stimuli. Two subjects out of seven performed significantly better with transposition than in the control condition, and none performed significantly worse. The perception of affricates (such as the ‘ch’ in church) was consistently improved. In a second test, the detection of ‘s’ or ‘z’ at the end of words was assessed using word pairs such as ‘cat’ and ‘cats’; on each trial a single word was presented and subjects had to decide which of the two words had been presented. Averaged across subjects, the detection of ‘s’ or ‘z’ was significantly improved, with five out of seven individual subjects improving significantly. This processing is currently being evaluated using wearable digital hearing aids.

Another promising form of transposition involves ‘nonlinear frequency compression’ (Simpson, Hersbach and McDermott, 2005, 2006). The signal is divided into two frequency ranges, with a dividing frequency f_c. Frequency components below f_c are amplified but are otherwise left unaltered. Frequency components above f_c are amplified and shifted downwards in frequency, with the amount of shift increasing as the frequency increases above f_c. Simpson, Hersbach and McDermott (2005) compared an experimental device incorporating this form of transposition with conventional hearing aids, using experienced hearing-aid users with moderate-to-severe sensorineural hearing loss and sloping audiograms. The recognition of monosyllabic words was assessed. Eight of the seventeen subjects showed a significant improvement with transposition, while only one showed a significant score decrease; the others showed no significant change. However, less good results were obtained for listeners with steeply sloping audiograms (Simpson, Hersbach and McDermott, 2006).

In summary, while commercial hearing aids incorporating transposition have met with only limited success, new methods for performing transposition are being developed, and have given promising results for some types of hearing-impaired people. These new methods may appear in commercial hearing aids in the near future.

VII COCHLEAR IMPLANTS

Cochlear implants represent a form of treatment for severe to total deafness arising from damage to the cochlea. Usually, such deafness is associated with loss of function of the OHCs and IHCs, but the auditory nerve survives to some extent. The treatment involves the implantation of electrodes so as to stimulate the surviving neurones electrically. This results in sensations of sound.

It is beyond the scope of this book to give more than a brief overview of this topic. For reviews, see Clark, Tong and Patrick (1990), Tyler (1993), Parkins (1997) and Zeng (2004). Briefly, the systems currently in use have two major parts. One part is external to the patient. It consists of a microphone to pick up the sound and a digital processor that analyses the sound and derives from it signals appropriate for delivery to the individual electrodes. These signals are transmitted through the skin to the implanted part of the system. This consists of a receiver of the transmitted signals and a device which separates the signals intended for the individual electrodes and delivers each of these signals to its appropriate electrode. The electrodes are usually implanted within the cochlea, by insertion through the round window.

Most cochlear implants currently in use are multi-channel devices, using several electrodes. These attempt to exploit the tonotopic or 'place' organization of the cochlea; see Chapter 1, Section IV.2. The electrodes are distributed as far as possible along the length of the cochlea and each is intended to stimulate neurones with a limited range of characteristic frequencies (CFs) (here, CF is used to refer to the frequency that would best excite a given neurone if the cochlea were not damaged). In practice, it is difficult to insert the electrode array very deeply into the cochlea, so the electrodes mainly lie close to neurones with medium and high CFs. Also, the electrical current from a given electrode or electrode pair tends to spread out within the cochlea, so the stimulation is not confined to neurones with a small range of CFs.

In a cochlear implant, information is transmitted in two basic ways: in terms of which electrode is active or of the distribution of current across electrodes ('place' coding) and in terms of the amplitude and time pattern of stimulation on individual electrodes ('temporal' coding); for a review, see Moore (2003). Changes in the place of stimulation generally give rise to changes in perceived timbre; stimulation at the basal end gives a 'sharp' timbre, while stimulation towards the apical end gives a more mellow or bass-like timbre. Sometimes, the percepts associated with stimulation of different places are described in terms of differences in pitch, rather than differences in timbre (McDermott and McKay, 1997; McKay, McDermott and Carlyon, 2000). Changes in the rate of stimulation on a single electrode can give rise to changes in perceived pitch, if the rate is below a few hundred hertz (Pijl and Schwarz, 1995; Pijl, 1997); for a review, see Moore and Carlyon (2005). Also, if an electrode is stimulated with brief electrical pulses at a high rate (say 1000 pulses per second), amplitude modulation of the pulses can lead to the percept of a pitch related to the modulation rate (McKay, McDermott and Clark, 1994; McDermott and McKay, 1997).

Many different coding strategies have been used in cochlear implants. Most devices filter the signal into about 20 frequency channels; the output from each channel is used to derive a signal which stimulates a specific electrode. In some devices, the stimulating current may be divided between two electrodes so as to create a 'virtual', intermediate, electrode (Donaldson, Kreft and Litvak, 2005). In one coding strategy, called compressed analogue (CA), the input waveform is first compressed and then filtered, and the resulting analogue waveforms are applied directly to the electrodes (Eddington, 1983). In another strategy, the envelopes of the outputs of the channels are used to control the amplitudes of brief electrical pulses that are delivered at a constant high rate (typically about 800 pulses per second). The pulses are interleaved on the different electrodes, hence the name continuous interleaved sampling (CIS) (Wilson *et al.*, 1991). The interleaving is done to reduce interactions of electrical fields between electrodes so as to improve the specificity of place coding (Parkins, 1997). This type of processing is incorporated in several commercial devices. In another processing strategy, called SPEAK, the signal is again filtered into a number of channels. About every 4 ms, between one and ten (typically six) channels with the largest outputs are selected and corresponding electrodes are stimulated by current pulses whose amplitudes are determined by the respective channel outputs.

Results obtained using cochlear implants vary widely across patients even for a single type of implant. However, it is remarkable that, for each of the different types of processing schemes described above, some patients have achieved good levels of speech understanding without lip-reading. Almost all patients find that the quality of their lives has been improved by the implants. However, most patients have great difficulty when background sounds are present, perhaps because of poor spectral resolution (Fu and Nogaki, 2005) or because information about the temporal fine structure of sounds is not conveyed effectively by cochlear implants, except at very low frequencies (Moore and Carlyon, 2005).

When cochlear implants were first introduced, they were restricted to adults with total or near-total hearing loss. However, the results have been sufficiently good that their use has been extended to adults and children (McCormick, Archbold and Sheppard, 1994) with significant residual hearing. Also, increasing numbers of people are being fitted with both a hearing aid and a cochlear implant, either in the same ear or in opposite ears. The implant can be used to provide information about medium and high frequencies, and the hearing aid can provide information about low frequencies. The combination often works markedly better than either device alone (Tyler *et al.*, 2002; Ching, Incerti and Hill, 2004; Turner *et al.*, 2004; Ching *et al.*, 2005).

VIII CONCLUDING REMARKS

Hearing aids have improved considerably in recent years. Hearing aids are available with low distortion and with smooth wideband frequency responses, although it is still the case that high-frequency gain and bandwidth are often inadequate. Most hearing aids have a high degree of flexibility in shaping their frequency-gain characteristic. In principle, it is usually possible to tailor the frequency response of a hearing aid to suit an individual patient. Hearing aids are also available that offer reasonably effective compensation for loudness recruitment. In practice, however, many hearing-impaired people are still being fitted with hearing aids that have an irregular frequency response with an inappropriate shape, and whose compression characteristics are not adjusted appropriately.

The occlusion effect, the perception that the user's own voice is too loud and unnatural in quality, can be largely eliminated by the use of open fittings, and such fittings are becoming increasingly popular. However, with open fittings it is not possible to apply much gain at low frequencies, so such fittings are not suitable for people who have low-frequency hearing loss. An alternative method of reducing the occlusion effect is to use a receiver fitted deeply into the ear canal.

Even the best possible current hearing aids do not restore hearing to normal; especially when listening to speech in noise or competing speech, hearing-impaired people listening through hearing aids perform less well than normally hearing people. This situation has been improved to some extent through the use of directional microphones, although these are most often used with BTE hearing aids, and these aids do not allow the effective use of sound-localization cues provided by the pinnae. People with severe or profound hearing losses are now often given a cochlear implant, or a cochlear implant in combination with a hearing aid; the combination usually works better than either device alone.

Glossary

This glossary defines most of the technical terms which appear in the text. Sometimes the definitions are specific to the context of the book and do not apply to everyday usage of the terms. The glossary also defines some terms not used in the text but that may be found in the literature on hearing. Terms in italics within definitions are themselves entries in the glossary.

Absolute threshold	The minimum detectable level of a sound in the absence of any other external sounds. The manner of presentation of the sound and the method of determining detectability must be specified.
Acoustic reflex	A contraction of muscles in the middle ear which reduces the transmission of sound through the middle ear, mainly for frequencies below 1000 Hz.
Active mechanism	A mechanism within the cochlea which amplifies the response of the *basilar membrane* to weak sounds, enhances the sharpness of *tuning* and introduces *nonlinearity*. It depends on the function of the *outer hair cells*, and is easily damaged.
AGC	See *Automatic gain control*.
Amplitude	The instantaneous amplitude of an oscillating quantity (e.g. sound pressure or voltage) is its value at any instant, while the peak amplitude is the maximum value that the quantity attains. Sometimes the word 'peak' is omitted when the meaning is clear from the context.
Amplitude modulation (AM)	The process whereby the amplitude of a carrier is made to change as a function of time.
Apex	The inner tip of the spiral-shaped cochlea; the opposite end from the *base*.
Articulation Index (AI)	A measure of the proportion of the speech spectrum that is audible for a given listening situation and a given listener. Each spectral region is given a weighting according to its contribution to intelligibility. The AI varies between 0 and 1.
Attack time	The time taken for the output of an *automatic gain control* circuit to get within 3 dB of its steady value when the input level is abruptly increased (usually by 25 dB).

Audiogram	A graph showing the *absolute threshold* for *pure tones* as a function of *frequency*. It is usually plotted as hearing loss (deviation from the average threshold for young normally hearing people) in *decibels* as a function of frequency, with increasing loss plotted in the downward direction.
Auditory filter	One of an array of bandpass *filters* that are assumed to exist in the peripheral auditory system. The characteristics of the filters are often estimated in *masking* experiments.
Aural harmonic	A *frequency component* generated by *nonlinearity* in the auditory system in response to a sinusoidal input. Its frequency is an integer multiple of the frequency of the input.
Automatic gain control (AGC) system	A system whose *gain* changes over time so as to reduce the range of levels at the output relative to the range at the input; the gain decreases as the input level increases.
Automatic volume control (AVC) system	A form of *automatic gain control* in which the *gain* changes slowly with time. The effect can be similar to adjusting a volume control by hand.
AVC	See *Automatic volume control*.
Azimuth	The angle of a sound relative to the centre of a listener's head, projected onto the *horizontal plane*. It is represented by θ in Figure 7.1.
Backward masking	A form of *masking* in which the signal is presented before the masker.
Bandwidth	A term used to refer to a range of *frequencies*. The bandwidth of a bandpass *filter* is often defined as the difference between the two frequencies at which the response of the filter has fallen by 3 dB (i.e. to half power). It can also be defined in terms of the *equivalent rectangular bandwidth*.
Base	The end of the cochlea closest to the oval window and stapes.
Basilar membrane (BM)	A membrane inside the cochlea which vibrates in response to sound and whose vibrations lead to activity in the auditory pathways (see Chapter 1).
Beats	Periodic fluctuations in peak *amplitude* which occur when two *sinusoids* with slightly different *frequencies* are superimposed.
Bel	A unit for expressing the ratio of two powers. The number of bels is the logarithm to the base 10 of the power ratio.
Best frequency	See *Characteristic frequency*.

BF	See *Characteristic frequency*.
BILD	See *Binaural intelligibility level difference*.
Binaural	A situation involving listening with two ears.
Binaural intelligibility level difference (BILD)	The decrease of the *speech reception threshold* (SRT) in a background sound when the speech is spatially separated from the background, relative to the SRT when both speech and background come from the same position in space.
Binaural masking level difference (BMLD or MLD)	This is a measure of the improvement in detectability of a signal which can occur under binaural listening conditions. It is the difference in threshold of the signal (in decibels) for the case where the signal and masker have the same phase and level relationships at the two ears and the case where the interaural phase and/or level relationships of the signal and masker are different.
CB	See *Critical bandwidth*
Characteristic frequency (CF), Best frequency (BF)	The frequency at which the threshold of a given single neurone is lowest, that is the frequency at which it is most sensitive. CF is also used to describe the frequency to which a given place on the *basilar membrane* is most sensitive.
Cochlear echoes	See *Evoked otoacoustic emissions*.
Cochlear hearing loss	A hearing loss produced by damage to the structures inside the cochlea.
Combination tone	A *tone* perceived as a component of a complex stimulus which is not present in the sensations produced by the constituent components of the complex when they are presented alone. Also used to describe a physical *frequency component* present in the output of a nonlinear system, but absent from the input, when the input consists of two or more *sinusoids*.
Complex tone	A *tone* composed of a number of *sinusoids* of different *frequencies*.
Component	One of the *sinusoids* composing a complex sound. Also called a *frequency component*.
Compound action potential (CAP)	A gross measure of the response of the auditory nerve to clicks or short tone bursts.
Compression	A process which causes the output of a system to cover a smaller *dynamic range* than the input of the system. In such a system, the *gain* decreases as the input level increases.
Compression limiter	A type of *automatic gain control* circuit with a high *compression threshold* and short *attack* and *recovery*

	times. It is used to prevent over-amplification of intense sounds.
Compression ratio	A property of an *automatic gain control* circuit. It is the change in input level (in decibels) required to achieve a 1-dB change in output level (once the *compression threshold* is exceeded).
Compression threshold	The input level at which the gain of an *automatic gain control* circuit is reduced by 2 dB, relative to the gain applied in the region of linear amplification.
Compressive nonlinearity	A *nonlinearity* in which the output grows by a smaller factor than the input when the input is increased. For example, if the input is doubled, the output increases by less than a factor of two.
Compressor	A type of *automatic gain control* circuit in which the *gain* changes fairly rapidly when the input level changes. The gain decreases as the input level increases, so the range of levels at the input is compressed to a smaller range at the output.
Conductive hearing loss	A hearing loss produced by reduced transmission of sound through the outer and/or middle ear.
Cone of confusion	A surface defining all points in space that lead to a given interaural time difference or a given interaural level difference.
Constant velocity tuning curve	See *Tuning curve*.
Critical bandwidth (CB)	A measure of the 'effective *bandwidth*' of the *auditory filter*. It is often defined empirically by measuring some aspect of perception as a function of the bandwidth of the stimuli, and trying to determine a 'breakpoint' in the results. However, such breakpoints are rarely clear. In this book, the term *equivalent rectangular bandwidth* is used in preference to critical bandwidth.
Cycle	That portion of a periodic function that occurs in one period.
Dead region	A region in the cochlea where the *inner hair cells* and/or neurones are functioning so poorly that a *tone* which produces peak *basilar-membrane* vibration in that region is detected via an adjacent region where the *inner hair cells* and/or neurones are functioning more effectively.
Decibel	One-tenth of a *bel*, abbreviated dB. The number of dB is equal to ten times the logarithm (base 10) of the ratio of two *intensities*, or 20 times the logarithm of the ratio of two *amplitudes* or pressures.

Dichotic	A situation in which the sounds reaching the two ears are not the same.
Difference limen (DL)	Also called the just-noticeable difference (JND) or the differential threshold. The smallest detectable change in a stimulus. The method of determining detectability must be specified.
Difference limen for frequency (DLF)	The smallest detectable difference in *frequency* between two successive *pure tones*.
Diotic	A situation in which the sounds reaching the two ears are the same.
Diplacusis	Binaural diplacusis describes the case when a *tone* of fixed *frequency* evokes different *pitches* in the left and right ears.
Directional microphone	A microphone whose sensitivity to sounds varies with the direction of the sounds.
Directivity	A measure of the sensitivity of a microphone to sounds coming from the direction giving maximum response, relative to the sensitivity to sounds from other directions.
Distortion-product otoacoustic emissions	One or more *frequency components* emitted from the ear in response to two or more primary *tones*. The distortion products are not present as frequency components in the input signal.
Duplex theory	A theory proposed by Lord Rayleigh which assumes that sound localization depends on the use of interaural time differences at low *frequencies* and interaural level differences at high frequencies.
Dynamic range	The range of sound levels over which a system operates in a specified way. For example, for a single neurone in the auditory nerve it is the range of sound levels between the threshold and the level at which *saturation* occurs. In the case of auditory perception at a given *frequency*, it refers to the range of sound levels between the threshold for detection and the level at which the sound becomes uncomfortably loud.
Echo suppression	The phenomenon that when a sound is rapidly followed by some echoes (reflections of the sound from nearby surfaces), the echoes are perceptually fused with the leading sound and are not heard as separate events.
Elevation	The angle of a sound relative to the centre of a listener's head, projected onto the *median plane*. It is represented by δ in Figure 7.1.
Energetic masking	A form of *masking* which can be explained in terms of processes occurring in the cochlea and auditory nerve.

	Energetic masking occurs when the response of the auditory nerve to the masker-plus-signal is very similar to the response to the masker alone. It can be contrasted with *informational masking*.
Envelope	The envelope of any function is the smooth curve passing through the peaks of the function.
Equal-loudness contour	A curve plotted as a function of *frequency* showing the *sound pressure level* required to produce a given *loudness level*.
Equivalent rectangular bandwidth (ERB)	The ERB of a *filter* is the *bandwidth* of a rectangular filter which has the same peak transmission as that filter and which passes the same total power for a *white noise* input. The ERB of the *auditory filter* is often used as a measure of the *critical bandwidth*.
ERB	See *Equivalent rectangular bandwidth*.
ERB_N	The average *equivalent rectangular bandwidth* of the *auditory filter* as determined using young normally hearing subjects tested at moderate sound levels.
ERB_N-number scale	A scale in which the *frequency* axis has been converted into units based on the values of ERB_N. Each one-ERB_N step corresponds to a distance of about 0.89 mm on the *basilar membrane*.
Evoked otoacoustic emissions	Sounds emitted from the ear in response to sounds applied to the ear. The emissions are thought to be generated by the *active mechanism* in the cochlea.
Excitation pattern	A term used to describe the distribution of activity produced by a given sound at different places (corresponding to different *characteristic frequencies*) in the auditory system. Sometimes the term is used to describe the effective level of excitation (in *decibels*) at each CF. Psychoacoustically, the excitation pattern of a sound can be defined as the output of the *auditory filters* as a function of centre frequency.
Expansive nonlinearity	A *nonlinearity* in which the output grows by a greater factor than the input when the input is increased. For example, if the input is doubled, the output increases by more than a factor of two.
Expansion	The opposite of *compression*. It is sometimes used in hearing aids for low-level inputs; the gain is reduced to prevent microphone noise or low-level environmental noise from being audible.
F0	See *Fundamental frequency*.
F0DL	The smallest detectable difference in repetition rate between two successive *complex tones*.

Fast Fourier Transform (FFT)	A mathematical technique for calculating the *amplitudes* and *phases* of the *frequency components* in a sample of a sound of finite duration.
FFT	See *Fast Fourier Transform*.
Filter	A device with an input and an output whose response varies with *frequency*. For example, a bandpass filter shows a relatively large response to *sinusoids* whose frequencies fall within its *passband*, and shows a smaller response to frequencies outside the passband.
Filter shape	The relative response of a *filter* plotted as a function of the input *frequency*.
Frequency modulation detection limen (FMDL)	The smallest amount of *frequency modulation* needed to distinguish a frequency-modulated *tone* from a steady tone.
Formant	A *resonance* in the vocal tract which is usually manifested as a peak in the spectral *envelope* of a speech sound.
Forward masking	A form of *masking* in which the signal is presented after the masker.
Fourier analysis	A mathematical technique for determining the *amplitude*s and *phase*s of the sinusoidal frequency *components* in a complex sound.
Free field	A field or system of waves free from the effects of boundaries.
Frequency	For a *sinusoid*, the frequency is the number of *period*s occurring in one second. The unit is cycles per second, or hertz (Hz). For a complex periodic sound the term ‘repetition rate’ is used to describe the number of periods per second (pps).
Frequency analysis	See *Frequency selectivity*.
Frequency component	See *Component*.
Frequency modulation (FM)	The process whereby the *frequency* of a carrier is made to change as a function of time.
Frequency resolution	See *Frequency selectivity*.
Frequency selectivity	The ability of the auditory system to separate or resolve (to a limited extent) the spectral components in a complex sound.
Frequency-threshold curve	See *Tuning curve*.
Frontal plane	The plane lying at right angles to the *horizontal plane* and intersecting the upper margins of the entrances to the ear canals. See Figure 7.1.
Fundamental frequency (F0)	The fundamental frequency of a periodic sound is the *frequency* of that sinusoidal component of the sound that has the same *period* as the periodic sound.

Gain	The gain of a system (e.g. a *filter* or an amplifier) is the magnitude of the output divided by the magnitude of the input. For example, if the output is ten times the input, the gain is said to be 10. The gain is often expressed in *decibels*, in which case the gain is equal to the output level minus the input level.
Growth-of-masking function	A function relating the masked threshold of a signal to the level of the masker.
Haas effect	See *Precedence effect*.
Half-power bandwidth	For a bandpass *filter*, the difference between the two *frequencies* at which the response has fallen by a factor of two in power (i.e. by 3 dB) relative to the peak response.
Harmonic	A harmonic is a component of a *complex tone* whose *frequency* is an integer multiple of the *fundamental frequency* of the complex.
Harmonic distortion	Sinusoidal components present at the output of a nonlinear system when a single *sinusoid* is present at the input. The components have frequencies that are integer multiples of the *frequency* of the input.
Hearing level (HL)	The *absolute threshold* for a sound for a given listener expressed in *decibels* relative to the average absolute threshold for the same sound for young listeners with no known hearing defect.
HL	See *Hearing level*.
Homogeneity	A property of a *linear* system whereby if the input magnitude is changed by a certain factor the output magnitude changes by that same factor.
Horizontal plane	The plane passing through the upper margins of the entrances to the ear canals and the lower margins of the eye sockets. See Figure 7.1.
Hz	See *Frequency*.
IFFT	See *Inverse Fast Fourier Transform*.
Impulse response	The response of a system to a brief impulse. For a *linear* system, the characteristics of the system are completely defined by the impulse response.
Informational masking	A type of *masking* that occurs when a masking sound is highly similar in some way to the signal, and/or when the properties of the masker vary in an unpredictable way from one stimulus to the next. The amount of masking under these conditions is greater than would be expected from *energetic masking* alone.
Inner hair cells (IHCs)	Specialized cells lying between the *basilar membrane* and the *tectorial membrane*. They are transducers

	that convert mechanical vibration into electrical neural responses (action potentials or *spikes*) in the auditory nerve.
Input-output function	A plot of the magnitude of the output of a system as a function of the magnitude of the input.
Intensity	The sound *power* transmitted through a given area in a sound field. Units such as watts per square metre are used. The term is also used as a generic name for any quantity relating to amount of sound, such as power or energy, although this is not technically correct.
Intermodulation distortion	When the input to a system consists of two *sinusoids* and the output contains *frequency components* other than the two sinusoids, the extra components are said to result from intermodulation distortion. This only happens in a *nonlinear* system.
Interspike interval histogram	A histogram of the time intervals between successive nerve *spikes*. The peaks in the histogram indicate the time intervals that occur most often.
Inverse Fast Fourier Transform (IFFT)	A mathematical technique for calculating the *waveform* of a sound from the *amplitudes* and *phases* of its *frequency components*.
Kemp echoes	See *Evoked otoacoustic emissions*
Level	The level of a sound is specified in *decibels* in relation to some reference level. See *Sensation level* and *Sound pressure level*.
Limiter	See *Compression limiter*.
Linear	A linear system is a system which satisfies the conditions of *superposition and homogeneity* (see Chapter 1).
Loudness	The subjective impression of the magnitude of a sound. It is defined as the attribute of an auditory sensation in terms of which sounds may be ordered on a scale extending from quiet to loud.
Loudness level	The loudness level of a sound, in *phons*, is the sound pressure level in *decibels* of a *sinusoid* of *frequency* 1 kHz which is judged by the listener to be equivalent in *loudness*, when both sounds are presented binaurally from a frontal direction in a *free field*.
Loudness recruitment	See *Recruitment*.
Loudness summation	An expression reflecting the idea that the *loudness* of sounds depends on summing the *specific loudness* occurring in different *critical bandwidths* or *equivalent rectangular bandwidths*. It is sometimes used to describe the observation that, for a fixed overall

	intensity, a sound appears louder when its spectrum covers a wide *frequency* range than when its *spectrum* covers a narrow frequency range.
MAA	See *Minimum audible angle*.
Magnitude estimation	A method used to investigate the relationship between the physical magnitude of a stimulus (e.g. the *intensity* of a sound) and its subjective magnitude (e.g. its *loudness*). A series of stimuli with differing magnitudes is presented, and the subject is asked to assign a number to each stimulus, reflecting its subjective magnitude.
Magnitude production	A method used to investigate the relationship between the physical magnitude of a stimulus (e.g. the *intensity* of a sound) and its subjective magnitude (e.g. its *loudness*). The subject is asked to adjust the magnitude of a stimulus so that the subjective impression of its magnitude corresponds to a number given by the experimenter. This is repeated for a series of numbers.
Masked audiogram	See *Masking pattern*.
Masking	The amount (or the process) by which the threshold of audibility for one sound is raised by the presence of another (masking) sound.
Masking level difference (MLD)	See *Binaural masking level difference*.
Masking pattern, masked audiogram	A graph of the detection threshold of a sinusoidal signal, plotted as a function of the signal *frequency*, determined in the presence of a fixed *masking* sound. Sometimes, the thresholds are expressed as an amount of masking (in *decibels*).
Mean-square value	A quantity related to the *power* of a signal. It is obtained by squaring the instantaneous value of the signal (e.g. deviation of sound pressure from the mean atmospheric pressure) and taking the average of the squared values over time.
Median plane	The plane containing all points that are equally distant from the two ears. See Figure 7.1.
Middle ear reflex	A contraction of muscles in the middle ear which reduces the transmission of sound through the middle ear, mainly at frequencies below 1000 Hz.
Minimum audible angle (MAA)	The smallest detectable change in angular position of a sound source, relative to the subject's head.
Minimum audible field (MAF)	The minimum audible level (*absolute threshold*) of a sinusoidal signal as a function of *frequency* for a signal presented in *free field* from a frontal direction. The signal level is measured at the position corresponding

	to the centre of the listener's head, after the listener has been removed from the sound field.
Modulation	A change in a particular dimension of a stimulus. For example, a *sinusoid* may be modulated in *frequency* or in *amplitude*.
Modulation rate	For periodic *modulation*, the number of *periods* per second of the modulation (in either *amplitude* or *frequency*). Sometimes also called the modulation frequency.
Monaural	A situation in which sounds are presented to one ear only.
Noise	Any unwanted sound. *White noise* is a sound whose *power* per unit *bandwidth* is constant, on average, over the range of audible *frequencies*. See also, *Pink noise*.
Nonlinearity	A property of a system that is not *linear*, that is, does not obey the conditions of *superposition and homogeneity*.
Non-simultaneous masking	A form of *masking* in which the signal either proceeds the masker (*backward masking*) or follows the masker (*forward masking*).
Occlusion effect	An effect occurring when the ear canal is blocked. The person's own voice sounds unnaturally loud and 'boomy'.
Octave	The interval between two *tones* when their *frequencies* are in the ratio 2 : 1.
Off-frequency listening	The process of detecting a signal through an *auditory filter* which is not centred at the signal frequency.
Omni-directional microphone	A microphone equally sensitive to sounds from all directions.
Onset dominance	See *Precedence effect*.
Organ of Corti	A complex structure lying between the *basilar membrane* and the *tectorial membrane*. It contains the *inner* and *outer hair cells*.
Outer hair cells (OHCs)	Specialized cells lying between the *basilar membrane* and the *tectorial membrane*. They play an important role in the *active mechanism* of the cochlea.
Partial	Any sinusoidal *frequency component* in a *complex tone*. It may or may not be a *harmonic*.
Passband	The range of *frequencies* passed by a bandpass *filter*. Usually, it is specified as the range over which the response of the filter falls within certain limits.
Passive mechanism	The mechanism that gives rise to *tuning* on the *basilar membrane* in the absence of the *active mechanism*. It is revealed when the active mechanism ceases to

	function (e.g. after death). The passive mechanism depends on the basic mechanical properties of the basilar membrane and surrounding structures.
Period	The smallest time interval over which a *periodic waveform* repeats itself. For a *sinusoid*, it corresponds to the reciprocal of the *frequency*.
Periodic waveform	A *waveform* that repeats itself regularly as a function of time.
Phase	The phase of a *periodic waveform* is the fractional part of a *period* through which the *waveform* has advanced, measured from some arbitrary point in time. It may be expressed in degrees (°) or radians; $180^\circ = \pi$ radians.
Phase locking	The tendency for nerve firings (*spikes*) to occur at a particular *phase* of the stimulating *waveform* on the *basilar membrane*.
Phon	The unit of *loudness level*.
Pink noise	This is a noise whose *spectrum level* decreases by 3 dB for each doubling of *frequency*.
Pitch	That attribute of auditory sensation in terms of which sounds can be ordered on a scale extending from low to high. Variations in pitch give a sense of melody.
Power	A measure of energy per unit of time. It is difficult to measure the total power generated by a sound source, and it is more common to specify the magnitudes of sounds in terms of their *intensity*, which is the sound power transmitted through a unit area in a sound field.
Power function	A mathematical expression in which the magnitude of a quantity y is proportional to the magnitude of a quantity x raised to a power α: $y = \mathrm{C} \times x^{\alpha}$, where C is a constant. The *loudness* of a *sinusoid* in *sones* is often assumed to be a *power* function of *intensity*, with $\alpha = 0.3$, for sound levels above about 40 dB SPL.
Precedence effect	A description of the finding that, in the presence of echoes, the subjective location of a sound is determined primarily by the physical location of the leading part of the sound (usually this is the direct sound that reaches the ears before the echoes).
Presbyacusis (also called presbycusis)	The hearing loss that is associated with ageing.
Psychometric function	A plot of performance (e.g. percentage correct) in a detection or discrimination task as a function

of the size of the stimulus to be detected or discriminated.

Psychophysical tuning curve (PTC) A curve showing the level of a narrowband masker needed to mask a fixed sinusoidal signal, plotted as a function of masker *frequency*.

Pure tone A sound wave whose instantaneous pressure variation as a function of time is a sinusoidal function. Also called a simple *tone*.

Pure-tone average The hearing loss in dB HL averaged over specified *frequencies*, usually 500, 1000 and 2000 Hz, but sometimes including higher frequencies.

Rate-versus-level function A plot of the firing rate of a single neurone as a function of the sound level applied to the ear.

Recovery time The time taken for the output of an *automatic gain control* circuit to get within 4 dB of its steady value when the input level is abruptly decreased (usually by 25 dB).

Recruitment This refers to a more rapid than usual growth of *loudness* with increase in stimulus level, which occurs in people with cochlear hearing loss.

Release time See *Recovery time*.

Residue pitch Also known as *virtual pitch*, low pitch and periodicity pitch. The low pitch heard when a group of *partials* is perceived as a coherent whole. For a harmonic *complex tone*, the residue pitch is usually close to the pitch of the fundamental component, but that component does not have to be present for a residue pitch to be heard.

Resonance An enhancement of the *intensity* of a sound that occurs when its *frequency* equals or is close to the natural frequency of vibration of an acoustic system or air-filled cavity. The word is also used to describe the process by which the enhancement occurs.

Retrocochlear hearing loss A hearing loss caused by damage to the auditory nerve and/or higher levels of the auditory pathway.

Rms See *Root-mean-square value*.

Root-mean-square (rms) value A quantity obtained by squaring the instantaneous value of a *waveform* (e.g. deviation of sound pressure from the mean atmospheric pressure), taking the average of the squared values over time and then taking the square root of the average.

Saturation	The phenomenon whereby at medium to high sound levels a neurone no longer changes its rate of firing in response to increases in sound level.
Scala tympani	One of the two outer chambers of the cochlea. It lies on the opposite side of the *basilar membrane* from the *organ of Corti*.
Scala vestibuli	One of the two outer chambers of the cochlea. It lies on the same side of the *basilar membrane* as the *organ of Corti*.
Sensation level (SL)	The level of a sound in *decibels* relative to the threshold level for that sound for the individual listener.
Sensorineural hearing loss	A general term used to describe hearing loss caused by damage to the cochlea and/or the auditory nerve and higher levels of the auditory pathway.
Simple tone	See *Pure tone*.
Simultaneous masking	A form of *masking* in which the masker occurs over the whole time for which the signal is presented.
Sinusoid, Sine wave, Sinusoidal vibration	A *waveform* whose variation as a function of time is a sine function. This is the function relating the sine of an angle to the size of the angle.
Sone	A unit of *loudness*. A 1-kHz *sinusoid* presented binaurally in *free field* from a frontal direction is defined as having a loudness of one sone when its level is 40 dB SPL. The loudness in sones roughly doubles for each 10-dB increase in sound level above 40 dB SPL.
Sound pressure level (SPL)	This is the level of a sound in *decibels* relative to an internationally defined reference level. The latter corresponds to an *intensity* of 10^{-12} W/m^2, which is equivalent to a sound pressure of 20 μPa.
Specific loudness	A term used in *loudness* models to denote the loudness evoked by a sound in each *critical bandwidth* or each *ERB*$_N$.
Specific loudness pattern	A plot of specific *loudness* as a function of *frequency*, usually with frequency expressed on a *critical-bandwidth* scale or *ERB*$_N$*-number scale*.
Spectrogram	A display showing how the short-term *spectrum* of a sound changes over time. The abscissa is time, the ordinate is *frequency* and the amount of energy is indicated by the lightness or darkness of shading. Spectrograms are often used in the analysis of speech sounds.
Spectrum	The spectrum of a sound wave is the distribution in *frequency* of the magnitudes (and sometimes the phases)

	of the components of the wave. It can be represented by plotting *power, intensity, amplitude* or *level* as a function of frequency.
Spectrum level	This is the level of a sound in *decibels* measured in a 1-Hz wide band. It is often used to characterize sounds with continuous spectra such as *noises*. A *white noise* has a long-term average spectrum level that is independent of *frequency*. A *pink noise* has a spectrum level that decreases by 3 dB for each doubling of frequency.
Speech intelligibility index (SII)	See *Articulation index*.
Speech reception threshold (SRT)	A measure of the level of speech required to achieve a given degree of intelligibility (e.g. 50 % correct). When measured in the presence of background *noise*, the SRT is often expressed as the speech-to-background ratio, in *decibels*.
Speech-shaped noise	*Noise* whose *spectrum* matches the long-term average spectrum of speech.
Spike	A single nerve impulse or action potential.
SPL	See *Sound pressure level*.
Spontaneous otoacoustic emissions	Sounds emitted from the ear in the absence of any external sound source. They are thought to be generated by the *active mechanism* in the cochlea.
Spontaneous rate	The average number of action potentials per second produced by a neurone in the absence of an input signal.
SRT	See *Speech reception threshold*.
Stereocilia	Small hair-like structures that project from the tops of the *inner* and *outer hair cells*.
Superposition	A property of a *linear* system whereby the response to two or more inputs presented simultaneously (e.g. two *sinusoids* with different *frequencies*) is the sum of the responses to the inputs presented individually.
Suppression	The process whereby excitation or neural activity at one *characteristic frequency* is reduced by the presence of excitation or neural activity at adjacent characteristic frequencies.
Tectorial membrane	A gelatinous structure lying above the *stereocilia* of the *inner* and *outer hair cells*.
Temporal integration	The phenomenon whereby performance on a specific task improves as the duration of the stimulus is increased. It is often used to describe the fact that the intensity of a sound at the *absolute threshold* for detection decreases with increasing duration.

Temporal modulation transfer function (TMTF)	The *modulation* depth required for detection of sinusoidal *amplitude modulation* of a carrier, plotted as a function of modulation *frequency*.
Temporal resolution	The ability to detect changes in the time pattern of a sound, usually changes in the *envelope*, under conditions where spectral cues are not available.
Threshold-equalizing noise (TEN)	A *noise* that is spectrally shaped such that, when used as a masker of a sinusoidal signal, the masked threshold is independent of the signal *frequency*, over a specified frequency range.
Timbre	That attribute of auditory sensation in terms of which a listener can judge that two sounds similarly presented and having the same *loudness* and *pitch* are dissimilar. Put more simply, it relates to the quality of a sound.
Time invariant	Description of a system or mechanism whose properties do not vary over time.
Tinnitus	The perception of a sound in the absence of any external sound applied to the ear.
TMTF	See *Temporal modulation transfer function*.
Tone	A sound wave capable of evoking an auditory sensation having *pitch*.
Tonotopic organization	A property of the auditory system that *characteristic frequency* (CF) is represented in an orderly spatial arrangement. For example, the CF of the *basilar membrane* varies monotonically with position along the membrane. In the auditory nerve, CF varies smoothly with the position of neurones within the nerve.
Tuning	The property of a system that it responds best to a limited range of sinusoidal *frequencies*.
Tuning curve	For the *basilar membrane*, this is a graph of the sound level required to produce a fixed response at a specific point (constant velocity or constant *amplitude*), plotted as a function of *frequency*. For a single nerve fibre it is a graph of the lowest sound level at which the fibre will respond, plotted as a function of frequency. This is also called a *frequency-threshold curve* (FTC). See also *Psychophysical tuning curve*.
Two-tone suppression	A phenomenon observed on the *basilar membrane* and in single neurones of the auditory nerve whereby the response to a tone close to the *characteristic frequency* is reduced by a second tone at a higher or lower frequency.
Virtual pitch	See *Residue pitch*.

Waveform	A term used to describe the form or shape of a wave. It may be represented graphically by plotting instantaneous *amplitude* or pressure as a function of time.
Weber fraction	The smallest detectable change in a stimulus, ΔS, divided by the magnitude of that stimulus, S.
Weber's Law	A 'law' stating that the smallest detectable change in a stimulus, ΔS, is proportional to the magnitude of that stimulus, S. In other words, ΔS/S = constant. The 'law' holds reasonably accurately for the intensity discrimination of *white noise*. For *sinusoids*, ΔS/S decreases somewhat with increasing sound level, which is called the 'near miss' to Weber's law.
White noise	A *noise* with a *spectrum level* that does not vary as a function of *frequency*.

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