

Ciaran Simms
Denis Wood

Solid Mechanics
and its Applications

Pedestrian and Cyclist Impact

A Biomechanical Perspective

 Springer

Pedestrian and Cyclist Impact

SOLID MECHANICS AND ITS APPLICATIONS

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Aims and Scope of the Series

The fundamental questions arising in mechanics are: *Why?*, *How?*, and *How much?* The aim of this series is to provide lucid accounts written by authoritative researchers giving vision and insight in answering these questions on the subject of mechanics as it relates to solids.

The scope of the series covers the entire spectrum of solid mechanics. Thus it includes the foundation of mechanics; variational formulations; computational mechanics; statics, kinematics and dynamics of rigid and elastic bodies; vibrations of solids and structures; dynamical systems and chaos; the theories of elasticity, plasticity and viscoelasticity; composite materials; rods, beams, shells and membranes; structural control and stability; soils, rocks and geomechanics; fracture; tribology; experimental mechanics; biomechanics and machine design.

The median level of presentation is the first year graduate student. Some texts are monographs defining the current state of the field; others are accessible to final year undergraduates; but essentially the emphasis is on readability and clarity.

Ciaran Simms • Denis Wood

Pedestrian and Cyclist Impact

A Biomechanical Perspective

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To Lynn

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Foreword

In the world today there are roughly 800 million cars, trucks and buses. There are also perhaps 500 million motorcycles and other motorised two-wheeled vehicles. No one really knows how many bicycles there are but, since 1990 the annual production of bicycles has been around 100 million, so it is likely that there are more bicycles in the world than all other road vehicles put together. Currently of road traffic deaths globally, three quarters of those deaths are to vulnerable road users, mainly pedestrians and cyclists. More important however are the seriously injured survivors. Although data are fragmentary in many parts of the world, for every death there are at least three casualties with serious, disabling injuries; brain or neurological damage or disruption of a major joint in the lower limbs. Financially and socially, the costs of the survivors represent some 80% of the costs of traffic injuries to society.

Hence the importance of this book. Ciaran Simms and Denis Wood have produced a book which is relevant to public health policy makers, transportation planners and particularly vehicle designers and the legislators who influence vehicle exterior design. By giving a biomechanical perspective to the subject of pedestrian and bicyclist impacts, they emphasize the crucial importance of exterior design of all road vehicles. Historically, vehicle crashworthiness has been skewed by activities in the United States where vehicle occupants have been the sole concern because of the highly motorised, car based conditions in North America. Modern car design now provides the occupants with airbags, seat belts and deforming structures which minimise the crash forces by providing large ride-down distances in severe impacts. It seems only right that the much more numerous pedestrians and cyclists in the world should get a few centimetres of useful ride-down distance and the appropriate exterior shape to reduce the severity of the injuries which they receive.

The authors provide an excellent historical perspective of the topic, illustrating its complexity and imperfectly documented epidemiology. They then explain in detail the current tools which are available through experimental testing, improved information on impact injury mechanics, injury criteria and modelling. This leads to the conclusion that although our knowledge is still imperfect what we do know can and should be applied. The benefits of such an approach on a global scale could be enormous.

By bringing together past research and current knowledge the authors have produced an important book which should be in the library of every biomechanical engineer, public health specialist, every car, truck and bus designer and every legislator who is concerned with traffic safety.

Murray Mackay, FREng.
President – IRCOB

Isle of Man, April 2009

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Chapter 1

Introduction

In 2004 the World Health Organisation predicted that road traffic injuries will become the third leading contributor to the global burden of disease by 2020, unless appropriate countermeasures are taken [1]. The proportion of road fatalities who are pedestrians and cyclists varies in different countries, but overall the protection of pedestrians is the most important road traffic safety priority [2]. This is best achieved using a combination of road engineering, vehicle design, legislation/enforcement and accident avoidance technology. The separation of pedestrians and cyclists from fast-moving motorised vehicles is clearly an ideal approach to preventing fatalities, and there are now also pre-crash sensing methods combined with brake-assist technology aimed at preventing the occurrence of pedestrian and cyclist accidents [3, 4]. However, these approaches cannot prevent all accidents, and vehicle/pedestrian and vehicle/cyclist collisions will remain a real and frequent problem in most countries for the foreseeable future.

This book is aimed at understanding the physical processes which occur when pedestrians and cyclists are struck by motorised vehicles. We provide a clear overview of the importance of pedestrian and cyclist impacts and the principal goals are to show how pedestrian and cyclist pre-impact movements and vehicle design influence subsequent injury outcome. This involves recourse to several academic disciplines: epidemiology, mechanics and anatomy/physiology. Therefore, this book presents pedestrian and cyclist impact from a biomechanical perspective.

We have had a basic understanding of the relationship between vehicle design and pedestrian injuries since the 1960s [5], but the safety of pedestrians and cyclists was not a serious consideration in vehicle design until the 1980s. This was encouraged by the popular but mistaken belief that little could be done to protect pedestrians in the event of a vehicle impact [6, 7]. This is best illustrated by a statement in a review paper in the *Journal of Accident Analysis and Prevention* in 1972 which concluded that “it would appear that pedestrians and vehicles are just not compatible” [6]. This common perception permitted manufacturers to continue to neglect an area of vehicle design not governed by legislation and not considered to provide added value to the vehicle. However, times have changed, and the advent of New Car Assessment

Programs has meant that there is now substantial public appetite in many countries for the regulation of vehicle design for pedestrian safety.

This book provides a comprehensive treatment of the biomechanics of pedestrian and cyclist impacts. It should therefore be of value to new and established researchers alike. We give a detailed treatment of the mechanics of pedestrian impact, as well as a review of the accident databases and the relevant injury criteria used to assess pedestrian and cyclist injuries. A further focus is the effect on injury outcome of the variability of initial position and speed of pedestrians and cyclists relative to the striking vehicle and the influence of vehicle design.

This is the first structured book on the mechanics of pedestrian and cyclist impact presented from a biomechanical perspective. It is intended to become a 'one stop' source for understanding the mechanics of pedestrian and cyclist impacts, and the book features a comprehensive treatment of the impact equations. Therefore, this book will serve as a primer and teaching tool for new researchers but also as a reference text for established researchers. The book highlights important improvements in pedestrian safety in recent decades and provides the necessary theoretical foundations for the reader to fully understand these advances.

This book is principally designed to be read sequentially, with later chapters building on fundamental foundations presented in earlier chapters. However, considerable effort has also been made to bring coherence to individual chapters. The book is structured as follows:

- Chapter 2 summarises pedestrian and cyclist injuries using the available accident databases.
- Chapter 3 provides an analysis of the movements of pedestrian and cyclists when struck by the front of vehicles.
- Chapter 4 presents the relationship between pedestrian projection distance and vehicle impact speed which is used for accident reconstruction and for biomechanics research.
- Chapter 5 is a summary of the injury mechanisms and injury prediction criteria that are used in the assessment of vehicle design.
- Chapter 6 describes the existing standardised tests to assess pedestrian safety in production vehicles.
- Chapter 7 provides mathematical formulations for impact modelling.
- Chapter 8 presents the state-of-the-art of existing simulation tools for modelling pedestrian and cyclist impact.
- Chapter 9 considers the relationship between vehicle related injuries and ground related injuries for pedestrians and cyclists.
- Chapter 10 provides a detailed analysis of the influence of vehicle design on pedestrian and cyclist injuries.
- Chapter 11 presents conclusions and perspectives for future research.

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Chapter 2

Pedestrian and Cyclist Injuries

Introduction

This chapter presents a general epidemiology of pedestrian and cyclist accidents. We begin with an overview of the scale and worldwide variability of both pedestrian and cyclist injuries. However, there is far more information available for pedestrian than for cyclist accidents and it can be difficult to distinguish between them, as they are sometimes jointly categorised in accident databases as Vulnerable Road Users [1]. In other cases, cyclist accidents are excluded from accident databases altogether [2] and even where they are included, underregistration is significant.¹ However, the general pattern of injury severity for pedestrians and cyclists is similar [4] and, due to their higher significance and the increased data available, the focus of this chapter is on pedestrian injuries. Pedestrian accident databases are the principal source of information, but it will be seen that those available mostly contain accidents from a small number of industrialised countries. Therefore, their findings must be interpreted cautiously when assessing pedestrian and cyclist injury levels in less industrialised countries. Nevertheless, the existing accident databases provide the best available overview of pedestrian and cyclist injuries.

Global View of Pedestrian and Cyclist Fatality and Injury Rates

The World Health Organisation has estimated that worldwide 1.2 million people are killed in road traffic crashes and that up to 50 million are injured each year [5]. The proportion of road accident fatalities who are pedestrians and cyclists varies substantially throughout the world, with large divisions evident between low and high income countries, as shown in Table 2.1. For example, in the US, France and Sweden vehicle occupants represent more than 60% of all road fatalities and the proportion of pedestrians and cyclists is 12–13% and 2–9% respectively. However,

¹ For example, only 15% of cycling accidents involving bodily injury are recorded in Belgium [3].

Table 2.1 Proportion of overall road traffic fatalities who are pedestrians and cyclists in different countries, presented in increasing order of pedestrian risk.

Country	Percentage of road accident fatalities who are pedestrians	Percentage of road accident fatalities who are cyclists
The Netherlands	10% [11]	22% [11]
China (Beijing only)	10–15% [12]	ca. 30% [13]
Malaysia	10–15% [12]	NA
Thailand	10–15% [12]	NA
France	12% [14]	3.4% [15]
USA	13% [11]	2% [11]
Germany	13% [14]	NA
Italy	17% [16]	5.3% [15]
Sweden	13% [14, 17]	8.6% [15]
Australia	18% [11]	4% [11]
Ireland	20% [18]	2.5% [18]
UK	21% [19]	4.6% [15]
Portugal	23% [14]	3.8% [15]
Chile	24% [12]	NA
Japan	30% [16]	10% [20]
South Africa	40% [21]	3% [22]
India (Delhi only)	42% [11]	14% [11]
Peru	45% [12]	NA
Ethiopia	51% [12]	NA
Kuwait	55% [6]	NA
Ivory Coast	75% [6]	NA

this relationship is largely inverted in many African, Asian and South American countries. In the Netherlands, pedestrians account for 10% and cyclists for 22% of fatalities. In contrast, pedestrians account for over 50% of road accident fatalities in Kuwait and Ethiopia and 75% in the Ivory Coast [6], and pedal and motorcyclists combined rank first for road accident fatalities in India and Surinam [6]. An explanation of these variations lies in the cultural as well as socio-economic differences between these regions. However, some reports are conflicting as the availability of detailed accident data in lower income countries is generally poor, and the majority of pedestrian fatalities occur in countries which do not routinely collect detailed road accident statistics. Therefore, this information shortfall in countries where it is most needed must be addressed if worldwide pedestrian and cyclist fatalities are to be significantly reduced.

It has been predicted that the number of road accident fatalities will rise to 2 million by 2020, and that 85% of these collisions will occur in low-income countries. Furthermore, these projections indicate that 75% of road accident fatalities will be pedestrians, cyclists and motorcyclists [7].

Apart from regional variations, the urban/rural divide is also important for distinguishing the relative importance of pedestrian fatalities. The majority of pedestrian accidents occur in built-up areas [8], and in London pedestrians account for half of

road fatalities [9]. In Kenya, pedestrians represent 70% of all urban fatalities, but just over 30% of rural traffic fatalities [10].

Main Pedestrian and Cyclist Injury Database Sources

In many countries, the police record basic information following an accident, such as time of day, scene description, vehicle type, presence of tyre/blood/glass traces on the road, etc. Similarly, hospitals usually record the principal injuries. However, detailed accident databases linking police and hospital records and predicting the range of impact speeds and vehicle/pedestrian/cyclist configurations at impact require significant resources. Therefore, pedestrian accident databases are either general and contain a large number of cases, or in-depth and contain a small number of detailed accident reconstructions.

Some highly motorised countries routinely maintain pedestrian accident databases,² and these mostly show consistent findings: about 80% of pedestrians were standing up and moving across the road when struck from the side by the fronts of passenger cars [23, 24]. About 60% of pedestrians made no avoidance manoeuvre such as jumping, accelerating, turning away or stopping [25]. The distribution of impact directions for cyclists is more complex, but lateral impact on the cyclist lower limb by the vehicle front is most common [26], accounting for about 60% of accident configurations [27–30]. A breakdown of the relative importance of the different impact directions for pedestrian accidents in the APROSYS database and for Japanese serious cyclist injuries and fatalities is given in Figures 2.1a&b respectively. While Figure 2.1a shows that 89% of pedestrians are struck on either the left or the right side, Figure 2.1b shows that for cyclists only 66% of fatalities involve side impact, and front/rear impacts are relatively more important for cyclists than for pedestrians.

Older databases such as the US Pedestrian Injury Causation Study (PICS) contain only accidents from the 1970s, and vehicle design and pedestrian injury profiles have changed considerably since then [25]. In recognition of this, the Pedestrian Safety Working Group of the International Harmonised Research Activities (IHRA)³ body have collated recent pedestrian accidents from Australia, Europe, Japan and the USA to a common format [2], see Table 2.2. The principal IHRA findings are presented here: the dataset consists of 1605 pedestrian accidents which resulted in 9463 injuries, of which 6158 were AIS1⁴ injuries and 3305 were AIS2-6 injuries. The IHRA database does not include pedestrian injuries from developing countries or any cyclist accidents, but it is the most comprehensive and up-to-date pedestrian accident database available.

² Eg German In Depth Accident Study (GIDAS), US Pedestrian Crash Data Study (PCDS), APROSYS European In Depth Pedestrian Database, Australia Transport Safety Bureau Fatal File.

³ IHRA are part of the United Nations Economic Commission for Europe (UNECE).

⁴ AIS Abbreviated Injury Scale, categorising injuries from minor (AIS1) to unsurvivable (AIS6) [33].

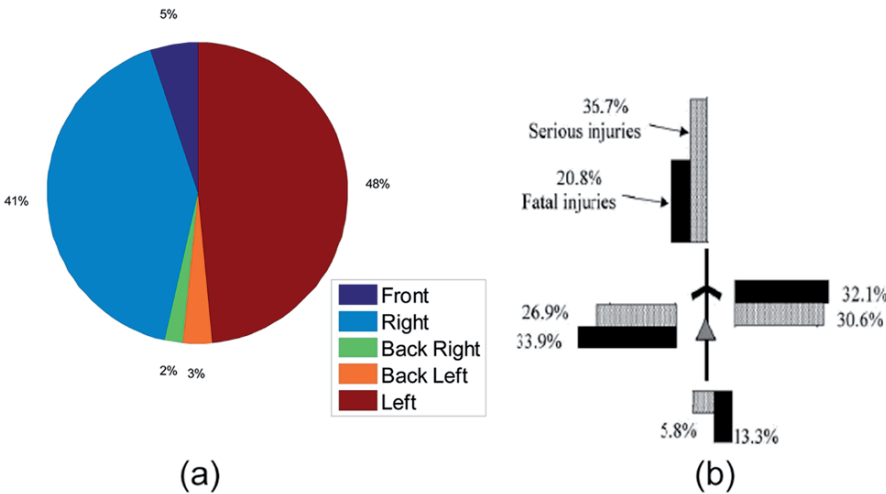


Fig. 2.1 Relative contribution of impact directions for (a) pedestrians in the APROSYS database (impact angles not available) adapted from [31] and (b) Japanese cyclist serious injuries and fatalities, adapted from [32].

Table 2.2 IHRA pedestrian accident data sources [2].

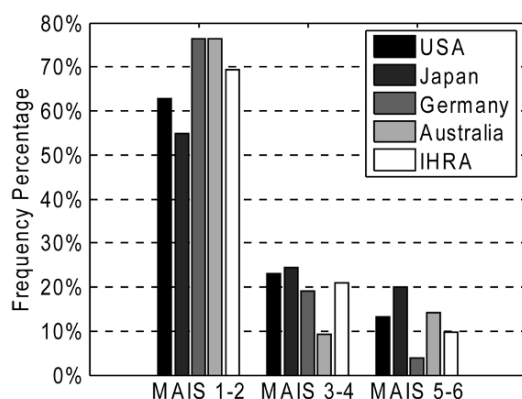
Country	Pedestrian accident database source
Japan	240 urban pedestrian accident cases collected by the Japanese Accident Research Institute (JARI) between 1987 and 1988 and in-depth case study data of pedestrian accidents conducted by the Japanese Institute for Traffic Accident Research and Data Analysis (ITARDA) between 1994 and 1998.
Germany	783 urban and rural cases collected between 1985 and 1998 by the German In-Depth Accident Study (GIDAS). Accidents in which pedestrians were overrun or impact speed could not be established were excluded.
United States	521 urban cases between 1994 and 1999 were collected in the Pedestrian Crash Data Study (PCDS). The first point of contact between the vehicle and the pedestrian had to be forward of the top of the A-pillar.
Australia	80 urban accidents between 1999 and 2000 were collected. These included 64 with passenger cars, SUVs or 1 box type vehicles where the pedestrian was standing, walking or running away. The main point of contact with the pedestrian on the vehicles was forward of the top of the A-pillar.

Distribution of Pedestrian Injuries

The IHRA data shows that a wide variety of pedestrian body regions are injured. However, it is important to distinguish between the severity of different injuries and the frequency with which these occur. In the following presentation of the IHRA data, AIS1 injuries are not included since the large number of these minor injuries considerably skews the dataset. The overall distribution of AIS2-6 pedestrian injur-

Table 2.3 The main IHRA pedestrian body regions injured and their frequency [2].

Body region	% AIS2-6 injuries
Head	31.4
Face	4.2
Neck	1.4
Chest	10.3
Abdomen	5.4
Pelvis	6.3
Arms	8.2
Legs	32.6

**Fig. 2.2** Distribution of IHRA pedestrian injury severities, adapted from [2].

ies to both adults and children categorised by body region is shown in Table 2.3. Figure 2.2 shows that, in all four source countries, approximately 70% of injuries are Maximum AIS [MAIS] 1–2, 20% are MAIS3–4 and 10% are MAIS5–6.

The head (excluding the face) and leg regions each account for about one third of AIS2–6 injuries, while the thorax region (pelvis, abdomen and chest) accounts for over 20% of AIS2–6 injuries, see Table 2.3. The remainder is principally accounted for by injuries to the face, neck and arms. These results compare well with German in-depth accident data from 1973 to 2000, where it was reported that the head and face accounted for nearly 30% of serious injuries to both adults and children, while thoracic injury accounted for 26% of adult and 19% of child injuries [34]. An Australian study of fatal pedestrian accidents between 1997 and 1999 found that 50% of pedestrian deaths were from AIS4+ injuries to the head/chest or a combination of these. However, 17% of fatal injuries were due to trauma of the chest, spine or abdomen [34].

Analysis of UK hospital records showed that the most common location of leg skeletal injury in pedestrian fatalities was the tibia (30%), fibula (25%), pelvis (21%) and femur (14%). Fibula fractures occur in 78% of cases where a tibia fracture is sustained, but when tibia fracture occurs, the risk of injuries to the ankle and knee

is reduced. The risk of femur and pelvis injuries is also reduced by a tibia fracture, especially in combination with a fibula fracture [35].

The distribution of IHRA pedestrian AIS2-6 injuries for both adults and children as a function of body region and striking vehicle component is shown in Tables 2.4 and 2.5. The tables contain a lot of information, and parts (a) and (b) have been constructed to present the same data in a slightly different manner. For example, row 2 in Table 2.4a shows that for adult AIS2-6 injuries caused by the front bumper, 3% are to the head and 71% are to the lower leg. Table 2.4a column 2 also shows that 28% of all adult AIS2-6 injuries are to the head. By comparison, column 2 in Table 2.4b shows that for adult AIS2-6 injuries to the head, 21% of cases are caused by the bonnet/wing top and 24% are caused by the windscreen frame/A-pillars.

The bumper, front panel and bonnet leading edge are the main sources of leg injury for both adult and child pedestrians. For mid-body regions, the bonnet leading edge and bonnet top are the leading sources of injuries for both adults and children. The top surface of the bonnet is the main cause of head injuries to children, but surprisingly, a substantial number of child head injuries are attributed to the windscreen. For adults, the windscreen glass is the main source of pedestrian head injury, followed by the windscreen frame/A-pillar and top surface of the bonnet and wing [2].

Distribution of Cyclist Injuries

The IHRA data does not include cyclist injuries, but Japanese accident data between 1995 and 1998 showed that 72% of cyclist fatalities resulted from head injuries, compared to 64% for pedestrians. For serious injuries, the leg accounted for 43% of cyclist injuries compared to 49% for pedestrians, and head injuries accounted for 21% of cyclist and 22% of pedestrian serious injuries [32]. Swedish accident data from 1989–1990 indicated that 31% of cyclist injuries were to the head and face, and 25% to the foot, knee, wrist and leg [37]. The breakdown of cyclist injuries in the German In Depth Accident Study (GIDAS) [38] and the European APROSYS in-depth accident database [23] are given in Tables 2.6 and 2.7.

There is some variability evident: the GIDAS data shows cyclist head injuries in 42% of cases, compared to 28% of cases for the APROSYS data. The GIDAS data also shows substantially higher knee injury risk for cases where cyclists were struck from the side by cars compared to head-on impact [38]. The GIDAS data shows that 61.4% of cyclists struck by the fronts of cars suffered osseous and/or ligament injuries to the knee, but tibia plateau fractures were much less common in cyclists than pedestrians [38].

However, both the APROSYS and GIDAS data indicate that, as for pedestrians, cyclist injuries are spread throughout the body and the upper and lower extremities and the head are more frequently injured than the midbody regions. Analysis of 136 bicycle accidents in Germany in the 1970s concluded that impact with the road surface was responsible for 31% of injuries, while the bumper, bonnet leading edge

Table 2.4a IHRA adult pedestrian AIS2-6 injuries by body region and vehicle contact [2]: numbers and percentage of body region injuries by each vehicle component.

	Head	Face	Neck	Chest	Abdomen	Pelvis	Arms	Legs					Un-known	Total
								Over-all	femur	knee	Lower leg	foot		
Front Bumper	20 (3%)	2 (0%)	- (0%)	2 (0%)	3 (0%)	3 (0%)	3 (0%)	16 (3%)	29 (5%)	69 (11%)	429 (71%)	29 (5%)		605 (100%)
Bonnet/wing top	140 (31%)	9 (2%)	1 (0%)	122 (27%)	39 (9%)	35 (8%)	73 (16%)	21 (5%)	3 (1%)	1 (0%)	1 (0%)	2 (0%)	1 (0%)	448 (100%)
Bonnet/wing leading edge	7 (2%)	2 (1%)	1 (0%)	36 (11%)	65 (20%)	80 (24%)	28 (9%)	46 (14%)	33 (10%)	5 (2%)	24 (7%)	1 (0%)	-	328 (100%)
Wind-screen glass	303 (70%)	52 (12%)	11 (3%)	28 (6%)	3 (1%)	10 (2%)	22 (5%)	1 (0%)	-	-	1 (0%)	1 (0%)	-	432 (100%)
Wind-screen frame/A-pillars	159 (56%)	28 (10%)	5 (2%)	34 (12%)	7 (2%)	14 (5%)	29 (10%)	5 (2%)	1 (0%)	-	-	-	2 (1%)	284 (100%)
Front Panel	-	1 (1%)	-	8 (8%)	13 (14%)	6 (6%)	5 (5%)	9 (9%)	9 (9%)	10 (10%)	32 (33%)	3 (3%)	-	96 (100%)
Others	33 (21%)	7 (5%)	-	29 (19%)	9 (6%)	12 (8%)	11 (7%)	6 (4%)	4 (3%)	5 (3%)	26 (17%)	13 (8%)	-	155 (100%)
Total	662 (28%)	101 (4%)	18 (1%)	259 (11%)	139 (6%)	160 (7%)	171 (4%)	104 (4%)	79 (3%)	90 (4%)	513 (22%)	49 (2%)	3 (0%)	2348 (100%)

Table 2.4b IHRA adult pedestrian AIS2-6 injuries by body region and vehicle contact [2]: numbers and percentage of vehicle component association with each body region injury.

	Head	Face	Neck	Chest	Abdomen	Pelvis	Arms	Legs					Un-known	Total
								Over-all	femur	knee	Lower leg	foot		
Front Bumper	20 (3%)	2 (2%)	-	2 (1%)	3 (2%)	3 (2%)	3 (2%)	16 (15%)	29 (37%)	69 (77%)	429 (84%)	29 (59%)		605 (26%)
Bonnet/wing top	140 (21%)	9 (9%)	1 (6%)	122 (47%)	39 (28%)	35 (22%)	73 (43%)	21 (20%)	3 (4%)	1 (1%)	1 (0%)	2 (4%)	1 (33%)	448 (19%)
Bonnet/wing leading edge	7 (1%)	2 (2%)	1 (6%)	36 (14%)	65 (47%)	80 (50%)	28 (16%)	46 (44%)	33 (42%)	5 (6%)	24 (5%)	1 (2%)	-	328 (14%)
Wind-screen glass	303 (46%)	52 (52%)	11 (61%)	28 (11%)	3 (2%)	10 (6%)	22 (13%)	1 (1%)	-	-	1 (0%)	1 (2%)	-	432 (18%)
Wind screen frame/A-pillars	159 (24%)	28 (28%)	5 (28%)	34 (13%)	7 (5%)	14 (9%)	29 (17%)	5 (5%)	1 (1%)	-	-	-	2 (66%)	284 (12%)
Front Panel	-	1 (1%)	-	8 (3%)	13 (9%)	6 (4%)	5 (3%)	9 (9%)	9 (11%)	10 (11%)	32 (6%)	3 (6%)	-	96 (4%)
Others	33 (5%)	7 (7%)	-	29 (11%)	9 (6%)	12 (8%)	11 (6%)	6 (6%)	4 (5%)	5 (6%)	26 (5%)	13 (27%)	-	155 (7%)
Total	662 (100%)	101 (100%)	18 (100%)	259 (100%)	139 (100%)	160 (100%)	171 (100%)	104 (100%)	79 (100%)	90 (100%)	513 (100%)	49 (100%)	3 (100%)	2348 (100%)

Table 2.5a IHRA child pedestrian AIS2-6 injuries by body region and vehicle contact [2]; number and percentage of body region injuries by each vehicle component.

	Head	Face	Neck	Chest	Abdomen	Pelvis	Arms	Legs					Unknown	Total
								Over-all	femur	knee	Lower leg	foot		
Front Bumper	4 (3%)	-	-	1 (1%)	2 (2%)	-	3 (3%)	3 (3%)	30 (30%)	7 (7%)	47 (47%)	2 (2%)	1 (1%)	100 (100%)
Bonnet/wing top	83 (61%)	6 (4%)	1 (1%)	17 (13%)	5 (4%)	8 (6%)	13 (10%)	2 (1%)	-	-	-	-	-	135 (100%)
Bonnet/wing leading edge	8 (13%)	-	3 (5%)	7 (11%)	13 (21%)	5 (8%)	7 (11%)	4 (7%)	7 (11%)	1 (2%)	6 (10%)	-	-	61 (100%)
Wind-screen glass	41 (75%)	4 (7%)	1 (2%)	2 (4%)	2 (4%)	2 (4%)	1 (2%)	1 (2%)	-	-	-	-	1 (2%)	55 (100%)
Wind screen frame/A-pillars	9 (75%)	-	-	1 (8%)	-	-	2 (17%)	-	-	-	-	-	-	12 (100%)
Front Panel	5 (29%)	-	-	1 (6%)	-	1 (6%)	1 (6%)	-	5 (29%)	1 (6%)	3 (185)	-	-	17 (100%)
Others	12 (19%)	-	1 (2%)	9 (15%)	3 (5%)	1 (2%)	4 (6%)	9 (15%)	5 (8%)	-	13 (21%)	5 (8%)	-	62 (100%)
Total	162 (37%)	10 (2%)	6 (1%)	38 (9%)	25 (6%)	17 (4%)	31 (7%)	19 (4%)	47 (11%)	9 (2%)	69 (16%)	7 (2%)	2 (0%)	442 (100%)

Table 2.5b IHRA child pedestrian AIS2-6 injuries by body region and vehicle contact [2]: number and percentage of vehicle component association with each body region injury.

	Head	Face	Neck	Chest	Abdomen	Pelvis	Arms	Legs					Un-known	Total
								Over-all	femur	knee	Lower leg	foot		
Front Bumper	4 (2%)	-	-	1 (6%)	2 (8%)	-	3 (10%)	3 (16%)	30 (64%)	7 (78%)	47 (68%)	2 (2%)	1 (50%)	100 (23%)
Bonnet/wing top	83 (51%)	6 (60%)	1 (17%)	17 (45%)	5 (20%)	8 (47%)	13 (42%)	2 (11%)	-	-	-	-	-	135 (31%)
Bonnet/wing leading edge	8 (5%)	-	3 (50%)	7 (18%)	13 (52%)	5 (29%)	7 (23%)	4 (21%)	7 (15%)	1 (11%)	6 (9%)	-	-	61 (14%)
Wind-screen glass	41 (25%)	4 (40%)	1 (17%)	2 (5%)	2 (8%)	2 (12%)	1 (3%)	1 (5%)	-	-	-	-	1 (50%)	55 (12%)
Wind screen frame/A-pillars	9 (6%)	-	-	1 (3%)	-	-	2 (6%)	-	-	-	-	-	-	12 (3%)
Front Panel	5 (3%)	-	-	1 (3%)	-	1 (6%)	1 (3%)	-	5 (11%)	1 (11%)	3 [36]	-	-	17 (4%)
Others	12 (7%)	-	1 (17%)	9 (24%)	3 (12%)	1 (6%)	4 (13%)	9 (47%)	5 (11%)	-	13 (19%)	5 (8%)	-	62 (14%)
Total	162 (100%)	10 (100%)	6 (100%)	38 (100%)	25 (100%)	17 (100%)	31 (100%)	19 (100%)	47 (100%)	9 (100%)	69 (100%)	7 (100%)	2 (100%)	442 (100%)

Table 2.6 Injury frequency of cyclist body regions for all injury severities in German (GIDAS) accidents 1995–2003 [38].

Head	42%
Neck	6%
Thorax	21%
Upper extremity	44%
Abdomen	5%
Pelvis	11%
Thigh	7%
Knee	25%
Lower leg	15%
Foot	13%

Table 2.7 Frequency of involvement of cyclist body regions for all injury severities in the APROSYS database ($N = 57$) [23].

Head	28%
Thorax	14%
Upper extremity	23%
Spine	2%
Abdomen	0%
Lower extremity	45%

and windscreen accounted for 8, 10 and 14% respectively [39]. The more recent APROSYS data shows that many injuries cannot easily be attributed to a specific vehicle region [23]. However, in cases where the source could be ascertained, about 10% of injuries were from head impact with the windscreen and 2% from the A-pillar, about 9% of injuries were from thigh impact with the bonnet, 14% were from knee impact with the bonnet leading edge and 11% were lower leg injuries from the bumper. Approximately 42% of cyclist injuries were attributed to impact with the ground. However, there were only 53 cyclists included in this accident database, and further data is required. Similarly, Japanese accident data showed that the bonnet leading edge caused femur, tibia and fibula fractures in cyclists, whereas for pedestrians injury to these body regions resulted from contact with the bumper and front apron [32], but the sample size was fairly small (< 169).

The Japanese accident data shows a fivefold higher fatality risk for pedestrians than cyclists [32], see Figure 2.3. However, more data from other sources is needed to corroborate these findings as the implication of this much lower cyclist fatality risk compared to pedestrians combined with the significantly higher proportion of cyclist fatalities (22%) compared to pedestrians (11%) in the Netherlands (Table 2.1) is that a very large number of cyclist accidents must occur in that country. Overall, cyclists are involved in fewer accidents leading to serious injury or death than pedestrians, but they are involved in more accidents resulting in minor injury [20].

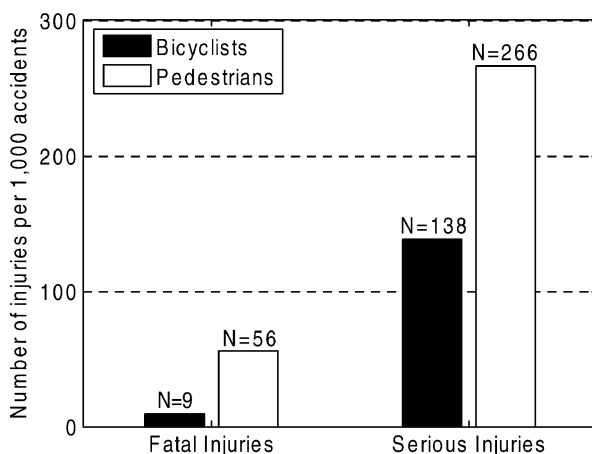


Fig. 2.3 Relative risk of fatality and serious injury for pedestrians and cyclists from Japanese real accident data, adapted from [32].

Injury Risk as a Function of Age and Sex

In the highly motorised countries represented by the IHRA dataset, child pedestrians are disproportionately represented: 31% of all pedestrian accidents involved children aged 15 and younger, although this group accounted for only 18% of the overall population in those countries [40]. The increased risk for children has previously been attributed to the height of children relative to the vehicle front [8]. For pedestrian accidents in a predominantly urban environment in Seattle, fatality rates were lowest in the age range 22–34, while children and young adults had a slightly higher fatality rate, and those over 65 had a much higher fatality rate [41]. Males accounted for 59% of the fatalities in these cases. Analysis of cycling accidents in Sweden between 1989 and 1990 showed that children and adults under the age of 20 are disproportionately represented in cycling accidents, and 55% of cases involved males [37]. A database of 136 accidents involving cyclists on the regional roads of the Brussels region between 1998 and 2000 showed that male cyclists in the 20–30 age bracket are at most risk, while the number of women cyclists sustaining injuries was low [3], see Figure 2.4. In South Africa, 57% of pedestrian fatalities were in the 22–44 year old age group, and 76% of pedestrian fatalities were males [21].

The severity of pedestrian and cyclist injuries is also age dependent [42, 43]. Younger people have a greater likelihood of low severity injuries, while older people have a greater likelihood of high severity injuries [2, 8]. However, the IHRA data in Figure 2.5 also shows that children are injured at slightly lower impact velocities than adults in most cases. Other researchers have reported that pedestrians over 60 are more than four times as likely to die than younger people [42] and while 10% of pedestrians under 30 with a pelvic injury died, this was increased to 40% for those over 75 [44]. Similarly, Rogers concluded that the fatality risk of cyclists over the

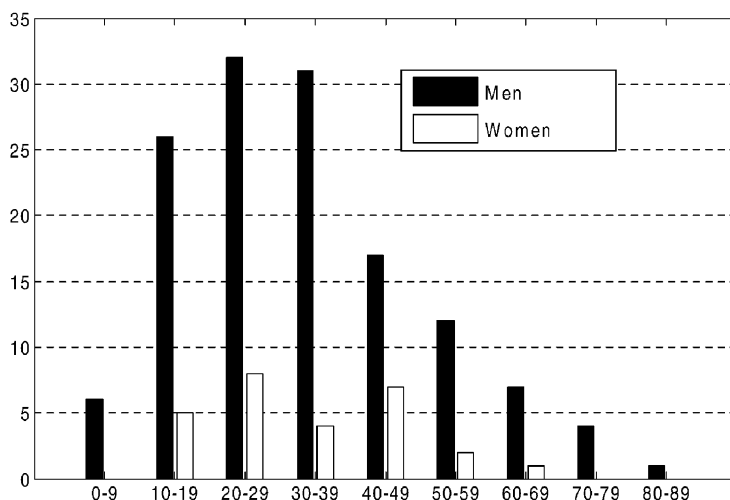


Fig. 2.4 Age and gender of cyclists sustaining injuries in the Brussels region between 1998 and 2000, adapted from [3].

age of 65 was 5.8 times higher than for cyclists under the age of 14 [43], and in Sweden the most severe bicycle related injuries were with people over 65 [37].

Overall, these findings reflect the different behaviour of pedestrians compared to cyclists and the differing socio-economic realities in the respective catchment areas where data collection was performed, and underline the need for all governments to engage in detailed accident data collection, rather than relying on research data from other jurisdictions.

The Distribution of Vehicle Impact Speeds

The strong correlation between impact speed and pedestrian injury severity is well-established [34, 45–48]. Figure 2.6 shows pedestrian fatality risk data from the World Health Organisation [49] and from Anderson et al. [50] after Walz et al. [46]. The data shows a sharp increase in pedestrian fatality risk for impact speeds in the region of 40 km/h. The cumulative frequency of vehicle impact speeds for pedestrian accidents for the four constituent countries in the IHRA study is given in Figure 2.7, and similar trends were recently found from On The Spot analysis of accidents in the UK [51]. The trends for Japan, Germany and Australia are similar, but for the USA a larger proportion of accidents occurred at lower speeds. In all four countries about 50% of pedestrian accidents occur at impact speeds below 25 km/h and about 90% of pedestrian injury cases occur at impact speeds below 50 km/h. Figure 2.7 shows that pedestrian injury accidents predominantly occur at speeds be-

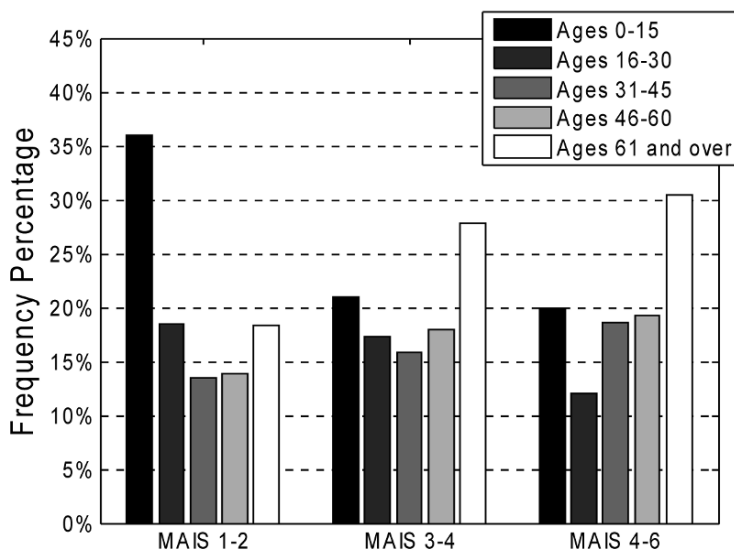


Fig. 2.5 IHRA pedestrian injuries as a function of age and injury severity, adapted from [2].

low 50 km/h. About 25% of pedestrian accidents occur at vehicle impact speeds in excess of 40 km/h, and Figure 2.6 indicates that the majority of these are fatal.

Figure 2.8a shows the cumulative frequency of pedestrian fatalities [48], indicating that 20% of pedestrian fatalities occur for vehicle impact speeds below about 33 km/h. Combining the pedestrian fatality risk as a function of vehicle impact speed (Figure 2.6) with the probability distribution of pedestrian fatalities derived from the cumulative frequency data in Figure 2.8a, the relative incidence of vehicle impact speeds in pedestrian accidents can be derived, see Figure 2.8b. This shows that a large majority of pedestrian accidents occur at impact speeds below 40 km/h, and therefore vehicle design should be focussed on injury reduction measures at these relatively low impact speeds.

In the IHRA pedestrian data lower MAIS injuries occurred at lower impact velocities for all four constituent countries and the average impact velocity for children aged 0-14 years is about 28 km/h, while for over 14 year olds the average impact velocity is about 32 km/h. For lower extremity trauma, the relationship between vehicle impact speed and pedestrian injuries is more complex. In an analysis of PCDS data, Matsui [52] found that at impact velocities of 20–30 km/h the main pedestrian lower extremity injuries are knee ligament damage, while at impact velocities of 40 km/h bony fractures predominate.

Walking speeds of 1.2–1.7 m/s and 0.9–1.2 m/s have been reported for younger and older pedestrians respectively [53, 54]. Although faster than pedestrians, cyclist travel speeds at impact are also low: 60% of vehicle-cyclist accidents in Japan occurred at cyclist speeds of 5 km/h or less, and 90% at 10 km/h or less [32], which partly explains the similarities in injury patterns between the two categories. How-

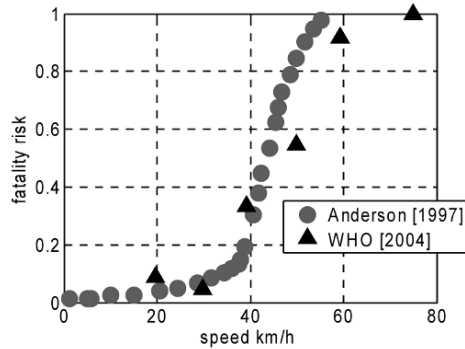


Fig. 2.6 Probability of pedestrian fatality as a function of vehicle impact speed (km/h), adapted from [46, 49, 50].

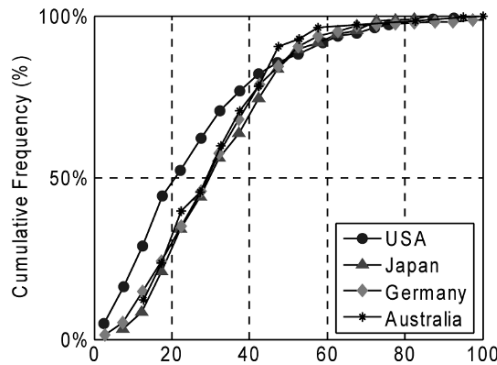


Fig. 2.7 Cumulative frequency of vehicle impact speed for pedestrian injury accidents (km/h), adapted from [2].

ever, vehicle impact speeds for cycling accidents are higher than for pedestrian accidents: German accident data showed that 50% of cars struck cyclists below 35 km/h and 76% below 50 km/h, see Figure 2.9. The equivalent pedestrian data (Figure 2.5) shows that 50% of pedestrian accidents occur at impact speeds below 25 km/h. Vehicle cyclist impact speeds greater than 70 km/h are rare, but injury severity increases with vehicle impact speed in a similar manner as for pedestrians: 95% of cyclist injuries were minor (MAIS1 or 2) for vehicle impact speeds below 30 km/h. However, this is reduced to 67% for impact speeds between 31–50 km/h and 10% above 70 km/h impact speed [30].

Analysis of pedestrian and cyclist accidents in Japan [20] led to cumulative fatality and serious injury risk curves as a function of vehicle driver's perceived speed at impact, see Figure 2.10. The data shows that the 50th percentile impact speed for serious injuries is higher for cyclists than for pedestrians. However, the 50th percentile impact speed for fatalities is higher for pedestrians than cyclists. This is

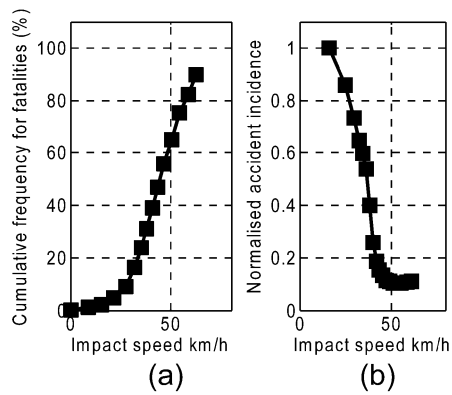


Fig. 2.8 (a) Cumulative frequency of pedestrian fatalities versus vehicle impact speed, adapted from [48] and (b) normalised fatality incidence versus vehicle impact speed (km/h) for pedestrian accidents, adapted/derived from [46, 48].

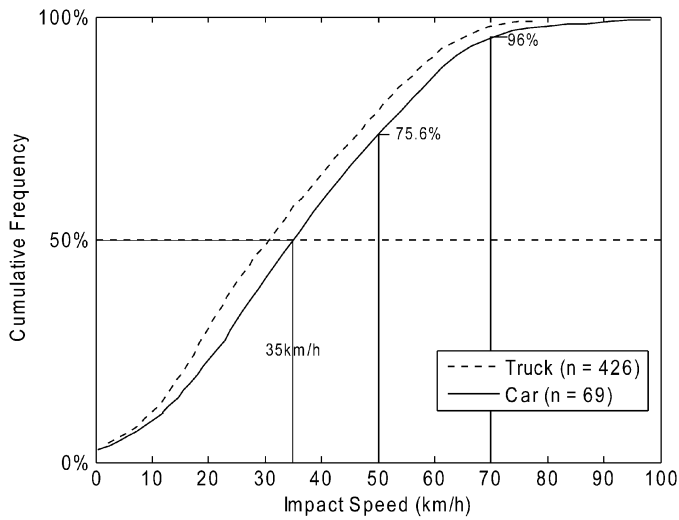


Fig. 2.9 Cumulative impact speed for cyclists struck in 614 accidents between 1973 and 1986, adapted from [30].

quite surprising, as we have already seen that the same research group concluded that the fatality risk for pedestrians is five times higher than for cyclists (see Figure 2.2). Clearly more data from other jurisdictions is required to understand these conflicting results.

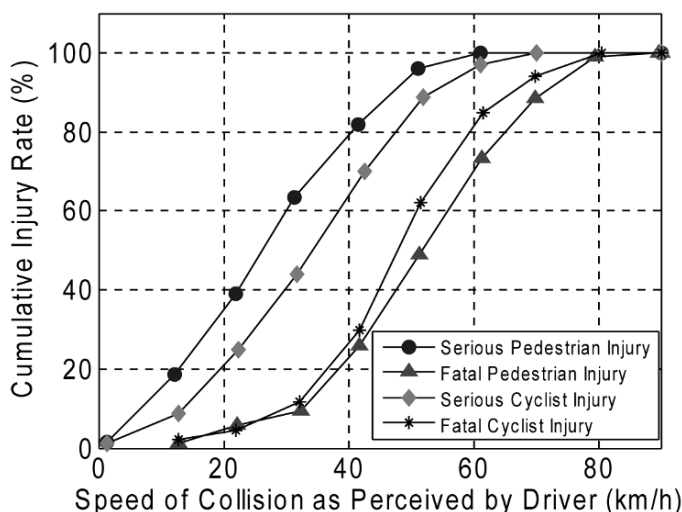


Fig. 2.10 Pedestrian and cyclist cumulative fatality and serious injury rate as a function of driver's perceived collision speed, adapted from [20].

Injuries from Vehicle and from Ground Contact

Most vehicle pedestrian collisions terminate following pedestrian contact with the ground, and the latter frequently leads to significant injuries, with about 8% of serious or fatal pedestrian head injuries attributed to ground contact [55]. Cyclist injuries from contact following a vehicle impact have also been reported [56, 57]. Tables 2.8 and 2.9 show a comparison between vehicle and ground related injuries from the IHRA dataset for adults and children respectively. Contacts with the ground account for 20–25% of AIS2–6 head, face and arm injuries, but only a low proportion of midbody or lower extremity injuries result from contact with the ground. The APROSYS data indicates that ground contact is especially important for child pedestrians and for cyclists, though only seven cyclist cases were included in that database [23], and this trend for child pedestrians is also evident in the IHRA data (Tables 2.8 and 2.9). Due to their obvious importance, ground contact pedestrian injuries are covered in detail in Chapter 9.

Injury Risk as a Function of Vehicle Size and Type

Several studies have observed that larger passenger cars result in more serious pedestrian injuries and higher fatality rates than smaller cars [58–60]. However, there is an even stronger relationship between vehicle type (cars, trucks, SUVs, etc.) and pedestrian injury/fatality risk. This has become increasingly important in recent

Table 2.8 IHRA adult pedestrian injuries as a function of body region and vehicle and ground contact [2].

	Head	Face	Neck	Chest	Abdomen	Pelvis	Arms	Legs					Unknown	Total
								Overall	Femur	Knee	Lower leg	Foot		
Vehicle	662	101	18	259	139	160	171	104	79	90	513	49	3	2348
Ground	125	18	2	21	2	8	32	6	4	3	5	14	1	241
Ratio of vehicle/ground injuries	5.3	5.6	6	12.3	69.5	20	5.3	17.3	19.8	30	102.6	3.5	3	9.7

Table 2.9 IHRA child pedestrian injuries as a function of body region and vehicle and ground contact [2].

	Head	Face	Neck	Chest	Abdomen	Pelvis	Arms	Legs					Unknown	Total
								Overall	Femur	Knee	Lower leg	Foot		
Vehicle	162	10	6	38	25	17	31	19	47	9	69	7	2	442
Ground	46	4	1	1	10	1	63	–	–	–	–	–	–	–
Ratio of vehicle/ground injuries	3.5	2.5	–	38	–	17	3.1	–	–	–	–	–	2	7.0

years because the distribution of vehicle types now includes a substantial number of Light Trucks and Vans (LTVs), including Sport Utility Vehicles (SUVs). In 2004 SUVs represented 15% of new vehicle registrations in Europe [61], while in the US almost 50% of new vehicles were either light trucks or SUVs [62]. Lefler and Gabler [62] used a combination of the Pedestrian Crash Data Study (PCDS) data, the General Estimates Data and the Fatal Accident Reporting System data from the US to show that 11.5% of pedestrians struck by large SUVs are killed, compared with 4.5% for pedestrians struck by cars. When the data was subdivided into three impact velocity ranges [0–20 km/h], [21–40 km/h] and [41–60 km/h], LTVs were found to be more likely to cause AIS3+ injury than cars in all three speed ranges.

Roudsari et al. [63] used the PCDS to analyse 3146 injuries among 386 pedestrians. There was no significant difference in mean impact speed between LTVs and passenger cars involved in pedestrian impacts. There were 159 adults with head injuries, of which 46 were struck by LTVs. They reported that the likelihood of adult head injuries was slightly higher for LTV (54%) than car crashes (46%) ($p = 0.16$). In addition, 39% of head injuries from LTV impacts were attributed to the secondary ground impact, compared to only 7% for cars, see chapter 9. However, the likelihood of thorax injuries was considerably higher for LTV crashes (37%) than for cars (20%) ($p = 0.001$). Similarly, the likelihood of abdomen injuries was considerably higher for LTV crashes (33%) than for cars (18%) ($p = 0.003$). In contrast, Roudsari et al. [64] found no difference between LTVs and passenger cars in the

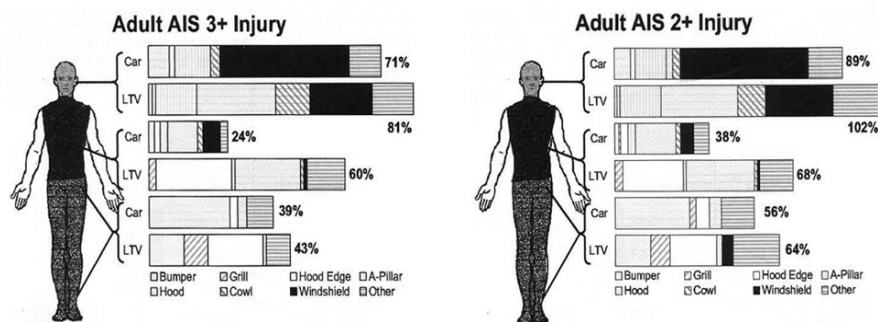


Fig. 2.11 Distribution of pedestrian injuries as a function of vehicle type, adapted from [67].

percentage of lower extremity injuries, while Ballesteros et al. [65] found that LTVs resulted in a higher percentage of injuries to the lower extremities above the knee.

Longhitano et al. [66, 67] also used the PCDS to analyse the influence of vehicle type on pedestrian injury distribution, see Figures 2.11 and 2.12. Ground impact injuries were not included. They found AIS3+ head injuries in 71% of cases for car impacts compared to 81% of cases for LTVs. AIS3+ injuries of the mid-body regions were found in car impacts in only 25% of cases compared to 60% of cases for LTVs. They concluded that the head was the most frequently injured body region for both vehicle categories, but the next most important body region is the lower extremities for impacts with cars, whereas for impacts with LTVs it is the torso. In less motorised countries, the most severe pedestrian injuries (AIS4-5) occur in accidents which involve either buses and trucks [68]. Analysis of STAT19 and the Heavy Vehicle Crash Injury Study found that pedestrians involved in collisions with the driver's side of trucks usually suffer serious or fatal injuries, and there is the additional danger of being run-over by the truck or projected into the path of oncoming traffic. Pedestrians struck by the vehicle middle region often fall to the ground, and can be trapped and dragged under the vehicle. Pedestrians struck on the passenger side are often not seen by the driver, and there are a relatively large number of collisions in this region [69, 70].

For vehicle/cyclist collisions, injuries caused by the bicycle itself are low severity and infrequent [39], and therefore it is unlikely that bicycle design has a significant influence on cyclist injuries. However, similar to pedestrians, more pronounced bonnet leading edges result in more severe cyclist injuries compared to vehicles with smoother bonnet leading edges. However, the proportion of cyclist injuries from the windscreen is higher in cases with lower bonnet leading edges [39] and cyclist collisions with trucks generally result in higher injury severity than collisions with passenger cars [57]. Similarly, analysis of 169 Japanese cyclist accidents with vehicle impact velocities of 30–50 km/h and cyclists aged thirteen or older showed that, like pedestrians, cyclists suffer fewer fatalities in collisions with passenger cars compared to minivans or SUVs [32]. These researchers also reported the cumulative fatality frequency for pedestrians and cyclists as a function of impact speed for both

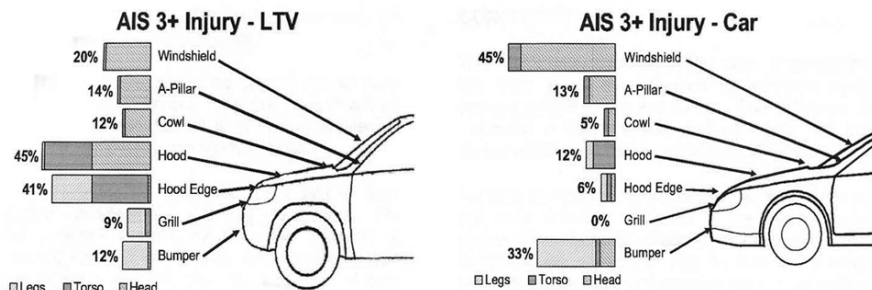


Fig. 2.12 sources of pedestrian AIS three plus injuries from Light Trucks and Vans (LTVs) and cars, adapted from [67].

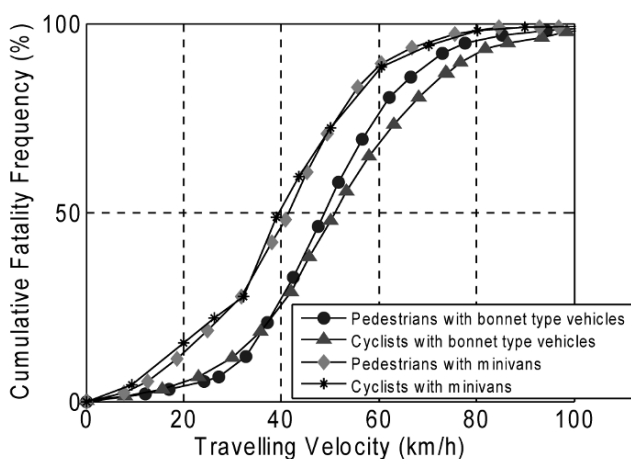


Fig. 2.13 Cumulative fatality frequency for pedestrians and cyclists as a function of impact speed for both bonnet type vehicles and minivans for Japanese accident data, adapted from [32].

bonnet type vehicles and minivans, see Figure 2.13. This is discussed in detail for both pedestrians and cyclists in Chapter 10.

Injuries and Disabilities

For non-fatal injuries, the resulting disability level is important. In particular, knee joint injuries are rarely life-threatening but frequently lead to long-term disabilities, and therefore bear a high societal cost [71–73]. This is because closed long bone fractures usually heal satisfactorily but knee ligament injuries may not [74]. However, high energy open fractures of the long bones also have severe consequences [75].

There is some disagreement in the literature about the current incidence of pedestrian knee injuries. In a study of 857 fatal pedestrian accidents, Teresinski and Madro [76] found 80% of cases resulted in injuries to the knee ligaments and/or femoral/tibial epiphyses. However, Otte et al. [38] reported that 16.6% of injured pedestrians in Germany from 1995–2003 suffered knee trauma, but only 1.2% of these were bony or ligamental injuries. Of these, tibial plateau fractures and medial collateral ligament lesions were the most frequent injuries. Similarly, 20.9% of cyclists suffered knee injuries, but only 0.5% of these were bony or ligamental injuries [38]. An analysis of 80 pedestrian accidents in Australia found that 2% suffered from ruptured knee ligaments, 21% a fractured tibia or fibula and there were no femur fractures [77]. Clearly, more data is required to resolve these conflicting findings.

Pedestrian Injury Trends over Time

In most highly motorised countries, the number of pedestrian fatalities has been reducing in the last few decades. This is partly due to increases in pedestrian safety, but also due to a reduction in walking rates. In the United States, pedestrian fatalities reduced by 13% from 5489 to 4784 between 1994 and 2006 [78]. Similarly, pedestrian fatalities in Europe reduced from about 13,000 per year to about 6000 per year between 1980 and 2000 [79]. However, a report estimating global road fatalities found a rapid growth in road crashes observed in India and China: from 1987–1995, overall road deaths increased by 40% in China, of which 10–15% were pedestrians [12]. It has been projected that by 2020 75% of road accident fatalities will be pedestrians, cyclists and motorcyclists [7].

Analysis of detailed accident databases within highly motorised countries has yielded more specific trends. In Japan, pedestrian femur injuries decreased from 17 to 4% and knee injuries from 10 to 1% from 1978–1998. During the same period, chest injuries increased from 3–11% and lower leg injuries from 19–36% [80]. A review of pedestrian and cyclist accident cases collected by GIDAS in Germany showed that head, neck and abdomen injury frequencies have remained largely unchanged, but there has been a reduction in thorax, upper extremity, thigh, knee, lower leg, foot and pelvis injury frequency for both pedestrians and cyclists, see Tables 2.10 and 2.11 [38].

Concluding Remarks

The global scale of pedestrian and cyclist fatalities and injuries has been presented, and major variations between different countries have been noted. It is clear that pedestrian and cyclist injury and fatalities are by far the most important road accident consideration in low-income countries. Results from accident databases

Table 2.10 GIDAS pedestrian accidents changes over time: all injuries [38].

Body region	Accident year 1985–1993	Accident year 1995–2003	Reduction in percentage injured
Head	56%	53%	3%
Neck	4%	5%	–1%
Thorax	22%	20%	2%
Upper extremity	44%	37%	7%
Abdomen	9%	7%	2%
Pelvis	16%	13%	3%
Thigh	16%	9%	7%
Knee	31%	22%	9%
Lower leg	26%	18%	8%
Foot	18%	14%	4%
Total	1463	1117	–

Table 2.11 GIDAS cyclist accidents changes over time: all injuries [38].

Body region	Accident year 1985–1993	Accident year 1995–2003	Reduction in percentage injured
Head	52%	42%	10%
Neck	5%	6%	–1%
Thorax	21%	21%	0%
Upper extremity	48%	44%	4%
Abdomen	6%	5%	1%
Pelvis	12%	11%	1%
Thigh	12%	7%	5%
Knee	32%	25%	7%
Lower leg	21%	15%	6%
Foot	17%	13%	4%
Total	2304	2018	–

from highly motorised countries give a detailed view of pedestrian injury distributions. They show that vehicle impact speed, vehicle type and pedestrian age are all strongly correlated with pedestrian injury and fatality risk. Furthermore, it is generally possible to attribute pedestrian and cyclist injuries to contact with specific vehicle parts and to distinguish between vehicle and ground contact as the source of injury. It is this data analysis which is directing the legislative and vehicle design efforts to reduce the severity of pedestrian injuries. While the number of pedestrian fatalities is reducing in many highly motorised countries, this is not the case for countries such as China and India where the level of motorisation is increasing rapidly. The relative lack of data for cyclist accidents and the frequently conflicting data on pedestrian injuries means that further widespread accident data collection is required.

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Chapter 3

Pedestrian and Cyclist Impact Kinematics

Introduction

Pedestrian and cyclist motion during impact determines which body regions are injured and the sources of these injuries. Furthermore, both pedestrians and cyclists are projected substantial distances during impact and the total projection distance is frequently used for accident reconstruction purposes (Chapter 4). However fundamentally, the movement of pedestrians and cyclists during and after vehicle impact is key to understanding the resulting injury mechanisms and for informing improved vehicle design (Chapter 10).

The kinematics of pedestrian and cyclist impacts are characterised by a large degree of variability. This is compounded by the fact that 79% of pedestrians and almost all cyclists were reported to be in motion at impact [1, 2]. However, there is generally insufficient evidence available following an accident to determine these complex movements, and therefore high speed videos of staged tests using crash dummies or cadavers struck by vehicle fronts are often used to record the kinematics of these impacts.

This chapter provides an overview of pedestrian and cyclist motion during and after vehicle impact. Descriptions of the most frequent impact configurations are given and film evidence from real accidents is presented where possible. Due to their superior clarity, images from staged tests are used to complement the real accident film evidence. The kinematics of pedestrian impact are far better understood than those of cyclist impacts, but there are many similarities in their motion patterns [3–5]. Nonetheless, the bent knee riding posture, high location of the pelvis and the higher transverse velocity of cyclists can result in a kinematic response that differs between cyclists and pedestrians [2]. However, the interaction between the cyclist and the bicycle following vehicle impact is minimal [6], the influence of bicycle mass is slight [7] and cyclist kinematics are unaffected by bicycle deformation [8], so the biggest differences between pedestrian and cyclist motion arise from the respective initial positions at impact.

Sources for Studying Pedestrian and Cyclist Movement

Due to their transient nature, the kinematics of pedestrian and cyclist impacts have mainly been studied using high speed video recordings of staged tests performed with crash test dummies or cadavers. These tests yield valuable information on the general kinematics and permit correlation between pedestrian/cyclist motion and vehicle damage and injury locations on the body. The subsequent development of computational models for impact analysis has yielded predictions of body motion which largely concur with the results of staged tests, see Chapter 8. More recently, a number of real vehicle pedestrian accidents have been recorded by security cameras and amateur photographers and made freely available. The image clarity from these recordings is much poorer than for staged tests, but they are of great interest and are presented in this chapter since they relate to actual collisions and therefore improve our fundamental understanding of the impact process. These tests also permit assessment of the validity of staged tests and computational predictions of pedestrian and cyclist motion.

Classification of Pedestrian and Cyclist Impact Configurations

The most important factors determining pedestrian and cyclist kinematics are the shape and speed of the vehicle and the height, stance and speed of the body relative to the vehicle at the instant of impact. There appears to be no film evidence of cycling accidents, but film evidence of pedestrian accidents shows that significant variations in the impact configurations and the resulting kinematics frequently occur. For practical reasons, this chapter focuses on the most common impact sequences, but individual accident cases are unlikely to be fully described by the descriptions in this chapter. This is confirmed by staged tests showing (for example) large differences in body rotation in the impact phase depending on leg configuration at the instant of vehicle contact [9].

We have seen in Chapter 2 that about 80% of pedestrians were standing up and moving across the road when struck from the side by the fronts of passenger cars [10, 11], and about 60% of pedestrians made no avoidance maneuver prior to impact such as jumping, accelerating, turning away or stopping [12]. Therefore the most important pedestrian accident scenario is a side-struck pedestrian moving at walking pace across the vehicle line of travel. However, this configuration may not be representative of children, as one analysis found that nearly half of child pedestrians struck by vehicles were running when struck [13].

For cyclists, a survey of 138 accidents in the Brussels region showed that 89 cases (64%) occurred at crossroads and 49 (36%) occurred away from crossroads [14]. Analysis of US, German, Dutch and Japanese accident data showed that cyclists, like pedestrians, are most frequently hit from the side by the fronts of vehicles [4, 15, 16], though the proportion of front/rear struck cyclists is higher than for pedestrians, see Chapter 2. Due to the elongated shape of bicycles, almost 40% of nominally

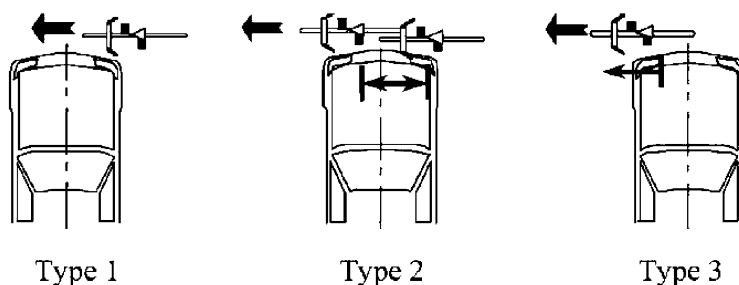


Fig. 3.1 Categories of side struck cyclist impact showing cases where the bicycle is struck but the cyclist does not suffer a direct hit, adapted from [2].

side impact cases are oblique impacts, though this is more frequent for adults than children [5]. Furthermore, there are a considerable number of cases where either the front or the rear side of the bicycle is struck, but the cyclist is not directly impacted [15, 16]. In an analysis of Japanese accident data from 1995–1998, three categories were established: (1) front of the bicycle up to (but not including) the saddle struck by vehicle, (2) saddle within two thirds of the front end of the vehicle front measured from the struck side, and (3) saddle in final third of vehicle front, see Figure 3.1 [2]. These categories are significant for determining subsequent head impact on the vehicle: in category 1, cyclist head impact occurred in only 6 out of 16 cases (38%), while in category 2 head impact occurred in 34 out of 49 (69%) of cases. No head impacts were recorded in category 3. There are therefore many more cyclist accident cases which do not involve a head impact on the vehicle front compared to pedestrian accident cases, and this may be largely responsible for the five-fold difference in fatality risk for pedestrians compared to cyclist reported by the same researchers [2], see Chapter 2. Another difference is that cyclist's legs are generally not weight-bearing and this may reduce leg injuries.

These analyses show that the most common impact configuration for cyclists and pedestrians is impact from the side by the front of a moving vehicle. Due to the similarities of cyclist with pedestrian kinematics, and the far greater detail available for the latter, we now largely focus on the description of pedestrian motion in vehicle impacts, but descriptions of cyclist impact kinematics are also given where available.

An early description of pedestrian kinematics was given by Stuerz and Suren [17]. However, the seminal work on pedestrian motion was by Ravani et al. [18], who analysed 460 pedestrian accidents in California. They showed that pedestrian motion is determined by the relative geometric configuration and speed of the vehicle and pedestrian at impact and that injury patterns are dependent on the subsequent trajectory type. They reported that impacts by the vehicle front are the most common (67%), sidewipes occur about 18% of the time and pedestrian impact by the rear of vehicles is rare (2%). The remainder were front corner impacts.¹ Con-

¹ Contrary to popular opinion, pedestrians are run over in less than 2% of all cases [19].

Table 3.1 Relative frequency of impact configurations for vehicle fronts striking pedestrians [18].

Wrap projection	Forward projection	Somersault	Roof vault	Fender vault
45%	35%	13%	5%	2%

sidering only the frontal collision cases, 80% of accidents could be classified into one of five basic trajectory categories: wrap projection, forward projection, somersault, roof vault and fender vault, and the relative frequencies are given in Table 3.1. These categories are named to describe the respective motion of the pedestrian in each case and have been widely adopted in the literature. However, somersault and roof vault are special cases of wrap projection, while fender vault is an incomplete wrap projection, so that there are three fundamental categories: sideswipe, wrap projection and forward projection. The pedestrian kinematics in each category are distinct and will be described in detail. However, although these classifications are useful, the inherent variability of vehicle pedestrian impacts means that some cases do not adequately fit a single categorisation.

Pedestrian Sideswipe Collisions

Sideswipes occur when contact between a pedestrian and the side of a moving vehicle results in a twisting moment which rotates the pedestrian about his/her longitudinal axis. The precise kinematics depends on the height and speed of the pedestrian relative to the vehicle. The linear impulse is mostly determined by the pre-impact speed of the pedestrian. The rotational impulse depends on the vehicle speed at impact, the pedestrian half-width and the effective retardation between the pedestrian and the side of the vehicle. For an adult pedestrian contacting the mid-region of a vehicle where the height is substantially above the pedestrian centre of gravity, the pedestrian generally falls to the ground quite close to the point of impact, though projected somewhat in the direction of vehicle travel. This is illustrated by Figure 3.2, which shows a staged impact between a VW Golf at 40 km/h and a child dummy moving transversely at 13 km/h [20]. The orientation at the instant of ground impact can be inferred from Figure 3.2e, showing early head impact with the ground. There is some restitution which projects the pedestrian from the vehicle in the transverse direction.

In sideswipe cases where a pedestrian is struck by the front wing of a car, the movement is more complex. This is shown in a low velocity collision recorded by a security camera in Helsinki, see Figure 3.3 [21]. A female pedestrian walked into the side of a passenger car in the region of the front wheel arch. The first contact was between her knees and the wheel arch. This decelerated her legs while her upper body momentum resulted in rotation over the side of the bonnet. However, this was counteracted by the large rotation about her longitudinal axis induced by

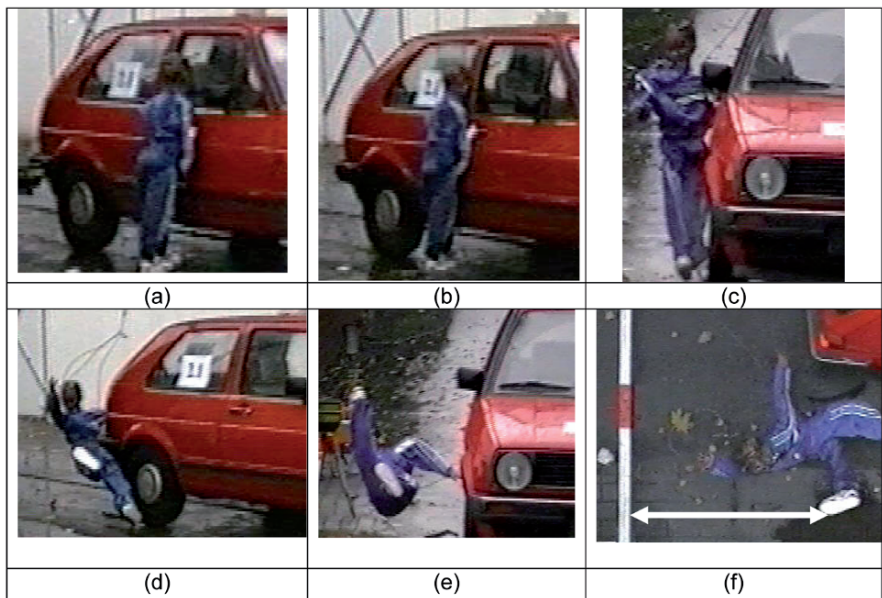


Fig. 3.2 Staged test between a VW Golf traveling at 40 km/h and a child dummy traveling transversely at 13 km/h, arrow in (f) shows projection distance, adapted from [20].

the frictional torque, as highlighted by the arrows indicating the direction in which she is facing. The pedestrian’s motion in the direction of vehicle travel was small and there was a significant voluntary response as she did not wrap over the bonnet or fall to the ground.

In higher velocity side-swipes where the pedestrian is struck by the vehicle wing, the shoulders/head may contact the A-pillar or windscreen, depending on the height and speed of the pedestrian. This can be a severe impact and results in a substantially increased projection distance in the direction of vehicle travel, as shown in a staged impact between a VW Golf at 44 km/h and an adult dummy moving transversely at 14 km/h [20], see Figure 3.4. The pedestrian dummy wraps over the front wing of the vehicle.

Wrap Projection

Pedestrian or cyclist wrap projection occurs when the principal vehicle impact force is applied below the centre of gravity. This is the most common sequence for pedestrians struck by vehicle fronts (Table 3.1) and can occur regardless of the orientation of the pedestrian at impact. A real-life example of this can be seen in an accident involving a sports car and an onlooker at a race track, see Figure 3.5 [22]. The bumper and bonnet leading edge contacts on the legs/thigh are below the pedestrian’s centre



Fig. 3.3 Real-world low velocity pedestrian sideswipe collision in Helsinki: arrows show the direction the woman is facing [21].

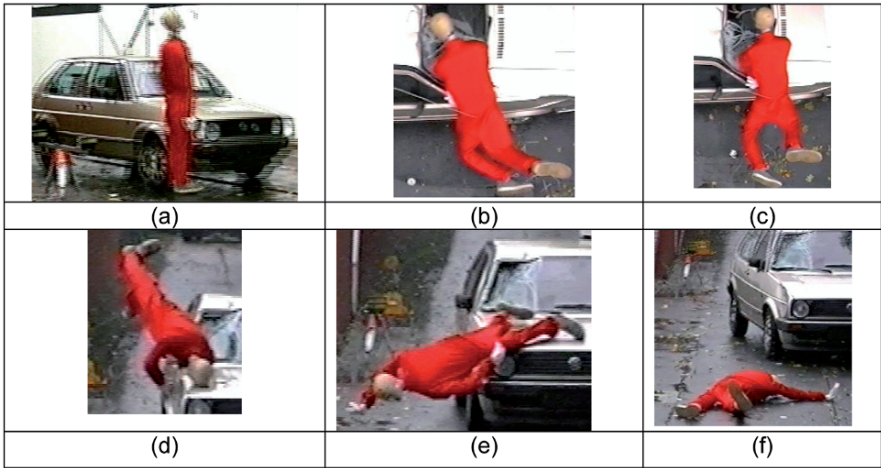


Fig. 3.4 Staged test between a VW Golf traveling at 40 km/h and an adult dummy traveling transversely at 13 km/h, adapted from [20].

of gravity and the net effective torque results in ‘wrapping’ of the body over the bonnet. The rate of rotation depends on the degree of eccentricity of the contact and the impact speed, see Chapter 7. In very low speed impacts the pedestrian may be able to prevent head impact by active bracing using the arms. For moderate to

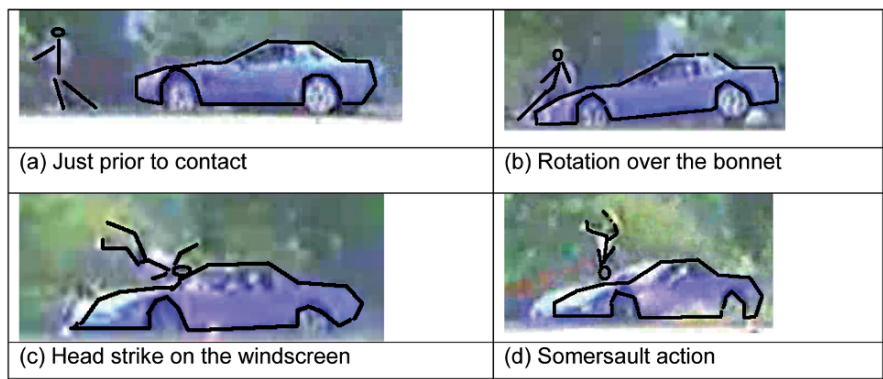


Fig. 3.5 Real accident pedestrian wrap projection involving a sports car striking an adult male pedestrian, adapted from [22].

high severity collisions, primary impact is followed by a further contact where the head and/or arms/shoulders are struck by the bonnet/windscreen area on the vehicle, and the pedestrian may then continue to rotate towards the roof, see Figure 3.5. If the vehicle is not braking sufficiently, the pedestrian may pass over the roof of the vehicle and land behind it, and this is known as roof vault. If braking is sufficient, the pedestrian may rotate completely on or above the bonnet prior to separation from the vehicle, and this is called somersault.² When the pedestrian has a substantial pre-impact transverse velocity across the vehicle, transverse momentum of the upper body may cause the head to miss the bonnet/windscreen, and instead the pedestrian vaults over the side of the vehicle. These are fender vault cases, and occur especially when initial contact is close to the far corner of the vehicle or is at the near corner and the pedestrian is projected back away from the vehicle. The projection distance in the direction of vehicle travel is then substantially reduced as there is only partial engagement with the vehicle.

Wrap projection is the most common but also the most complex impact scenario and the typical sequence will be described in greater detail with the aid of staged collisions using crash dummies and cadavers. Dummy tests are more repeatable, but cadaver tests are more biofidelic.

When the front of a passenger car strikes a side-facing adult pedestrian, the first contact is between the bumper and the knee region, see Figures 3.6 and 3.7, which show 39–40 km/h staged pedestrian impacts with cadavers [23, 24]. The following timings relate to these staged impacts. Initially the struck side leg is accelerated in the direction of vehicle travel. Due to inertia effects, the movements of the non-contacted body regions lag behind those of the struck region, facilitated by a combination of articulation in the joints, soft tissue deformation and bending of the long bones. The lower leg and foot on the struck side fold underneath the bumper and the non-struck leg remains still for at least 30 ms after initial vehicle contact.

² Although not shown here, this is what happened in the case involving the sports car in Figure 3.5.

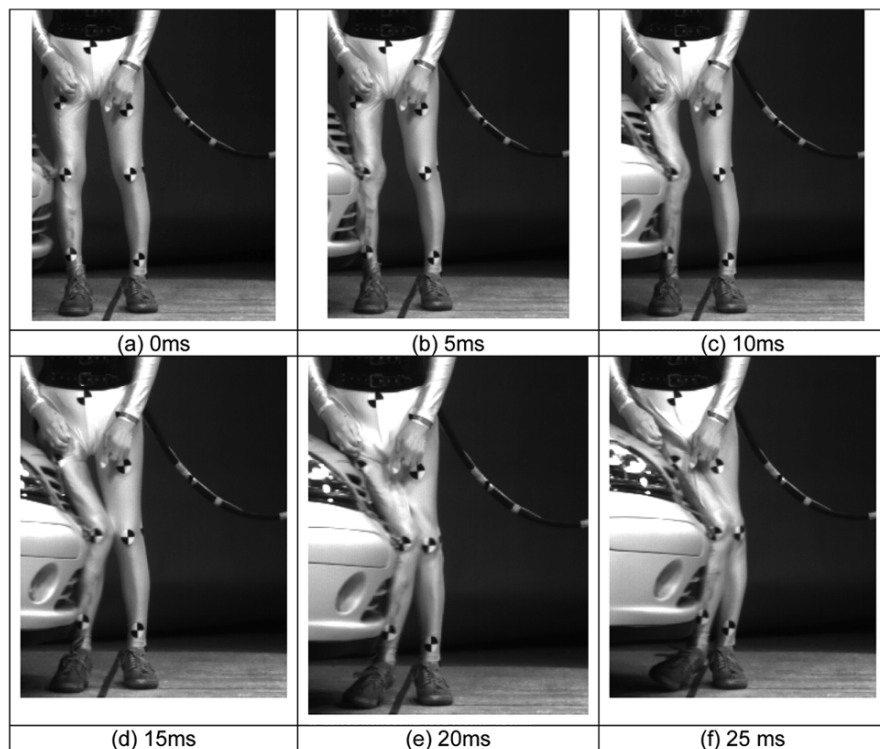


Fig. 3.6 Evolution of Leg kinematics for side facing pedestrian cadaver struck by a passenger car at 39 km/h, adapted from [23].

Similarly, the pelvis and the whole upper body remain almost horizontal for 40 ms, see Figure 3.7.

By about 20 ms, there is contact between the bonnet leading edge and the pedestrian upper leg/pelvis on the struck side, the severity of which depends on the vehicle shape. Newer cars generally have much smoother front profiles than older cars and the bonnet leading edge is sometimes difficult to distinguish, see Chapter 10. Rotation of the upper body onto the bonnet then proceeds so that 120ms after primary impact the torso has rotated sufficiently for the arm and shoulder on the struck side to contact the bonnet, see Figures 3.7 and 3.8. The motion of the head initially lags behind the torso as its inertia causes lateral bending in the neck. This can be clearly seen at 90 ms in Figure 3.8. However, the resulting lateral bending moment in the neck causes the head to ‘catch up’ with the torso rotation and by about 120 ms the head contacts the vehicle in the region of the bonnet/windscreen/grille.³ In these

³ The exact contact location is highly significant for the injuries and will be discussed further in Chapter 10.

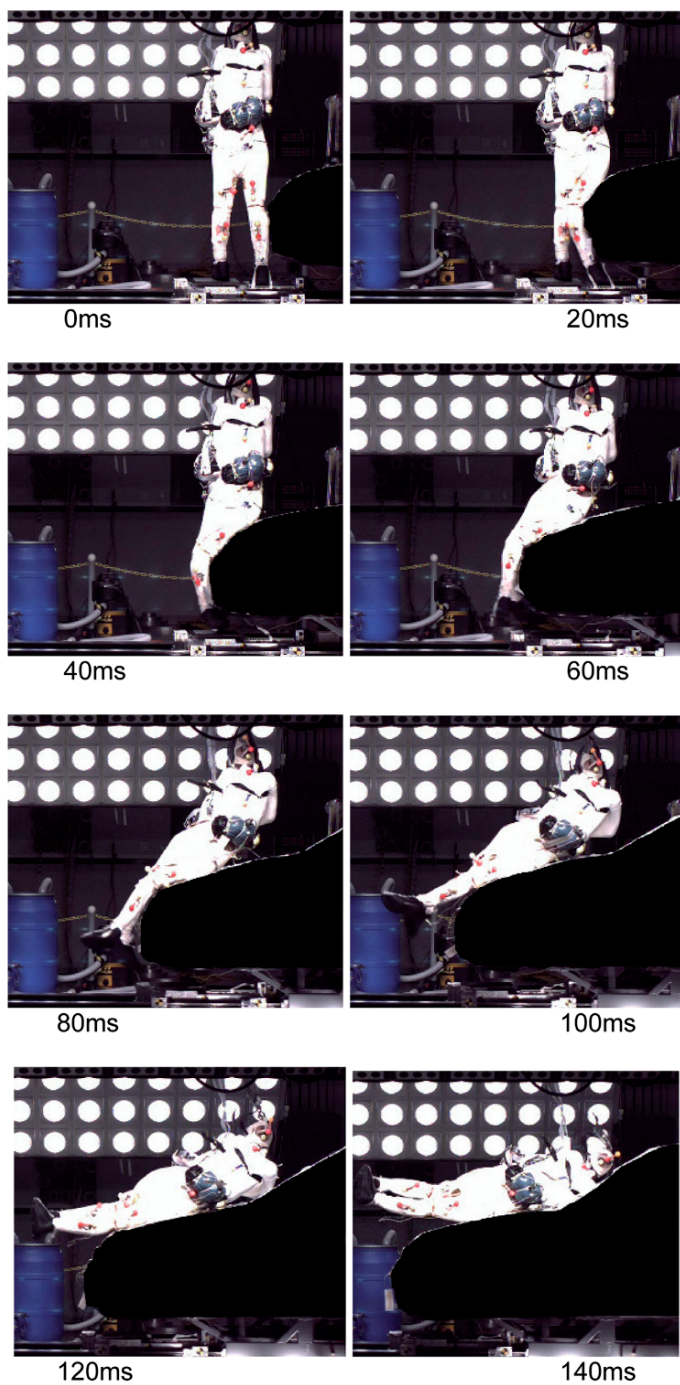


Fig. 3.7 Pedestrian wrap projection for cadaver tests at 40 km/h, adapted from [24].

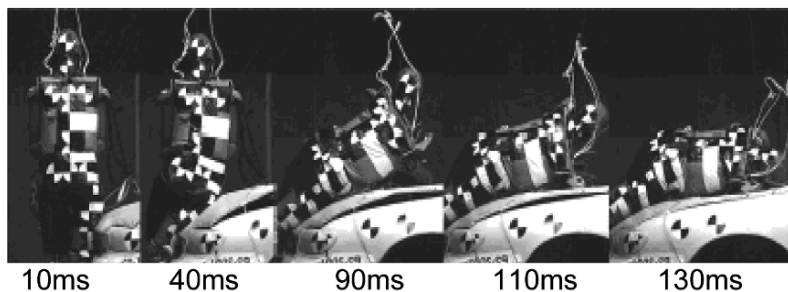


Fig. 3.8 Evolution of wrap trajectory of Polar pedestrian dummy onto bonnet of passenger car, adapted from [25].

tests, this whipping motion in the neck results in an increased head impact velocity for the pedestrian.

We have seen that approximately 80% of pedestrians were moving when struck and 85% were hit laterally (Chapter 2), implying transverse motion of the pedestrian relative to the vehicle in most cases. In wrap projection there is therefore a transverse offset between the location on the vehicle of the leg and the shoulder/head contacts. This offset depends mostly on the pedestrian pre-impact velocity, size, orientation and stance and the vehicle speed and the height of bumper. Transverse motion of the struck leg is typically arrested by friction, but the pedestrian's body then pivots about the bonnet leading edge in a three-dimensional manner which results in a head contact on the bonnet/windscreen which is both transversely as well as longitudinally offset from the initial leg contact location. Since there are substantial stiffness variations between the scuttle, windscreen and A-pillar for most cars [26], minor changes in head impact position can result in large variations in head injuries, see Chapter 10. In some cases the transverse offset results in a fender vault, especially where the initial contact of the moving pedestrian is close to the far corner of the vehicle. The head/shoulder contact on the bonnet/windscreen is then generally replaced by a head strike on the ground close to the side of the vehicle.

Wrap projection impacts for cyclists are similar: bumper impact with the cyclist's leg is followed by impact of the bonnet leading edge with the upper leg and the upper body then wraps around the bonnet [3, 4]. This is clearly shown in recent staged cyclist impact tests using cadavers struck by a passenger vehicle at an oblique angle [27], see Figure 3.9 compared to Figures 3.5–3.8.

Pedestrian/Cyclist Head Contact in Wrap Projections

The length measured along the vehicle front profile from the ground to the pedestrian/cyclist head impact location is the Wrap Around Distance WAD [28], which provides a useful measure of which vehicle front-end components contact the head

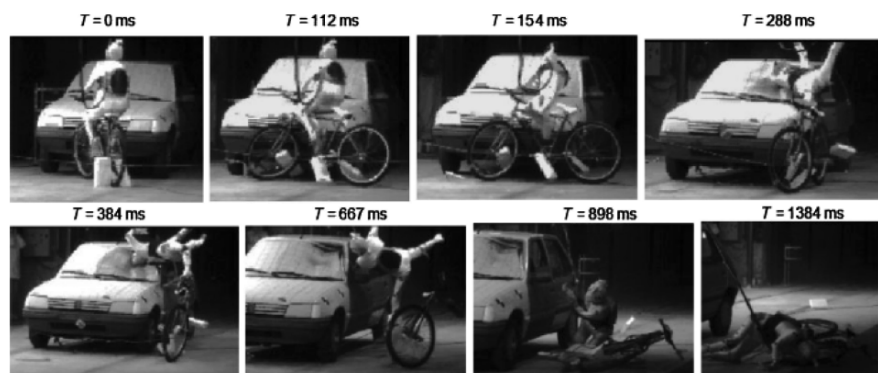


Fig. 3.9 Oblique cadaver vehicle cyclist impact leading to wrap trajectory and fender vault, adapted from [27].

in pedestrian/cyclist impacts, see Chapter 10. The height of the head from the ground for pedestrians and cyclists is nearly the same [2], but for moderate and high speed collisions, the WAD is greater than the overall standing height [29]. Figure 3.10 shows pedestrian and cyclist head impact locations on the windscreen/bonnet of bonnet-type vehicles in Japanese accidents from 1995–1998.

The position of head impact on the vehicle depends largely on the pedestrian/cyclist height and transverse velocity and the vehicle impact speed. If the body is hit from the side, the shoulder frequently contacts the top surface of the bonnet or the windscreen first, tending to reduce the severity of the head impact. In contrast, if a vehicle strikes the body from the front or the rear, head shielding does not occur [29]. Accident analysis has shown that 47% of tall pedestrians (1.7–1.9 m) struck by a car suffered a head/vehicle impact compared to only 31% of short pedestrians (1.5–1.69 m). Similarly, taller pedestrians are twice as likely to have a head impact on the windscreen, while shorter ones are 50% more likely to have a head impact on the bonnet [30]. Similarly to pedestrians, child cyclist head impacts occur most frequently on the bonnet, while the heads of adult cyclists more often impact the rear of the bonnet or the windscreen [5]. In general, a low bonnet leading edge height results in the pedestrian/cyclist being carried further onto the car and closer to the windshield [29].

The pedestrian centre of mass is considerably lower than that of a cyclist [4, 31], and it was initially predicted that this leads to more severe impacts for cyclists than pedestrians [4], but fundamental mechanics (Chapter 7) and recent fatality risk data for pedestrians and cyclists (Chapter 2) contradict this. Instead, it has been proposed that increased sliding of the cyclist's body on the bonnet occurs due to the higher initial pelvis height in cyclists compared to pedestrians, and this could account for their larger WAD [2, 31]. This is confirmed by accident data which showed that, for cyclists struck by bonnet type vehicles, head impact occurs with the relatively compliant windscreen or roof rather than the stiffer bonnet or windscreen frame region, partially explaining their lower injury risk [2].

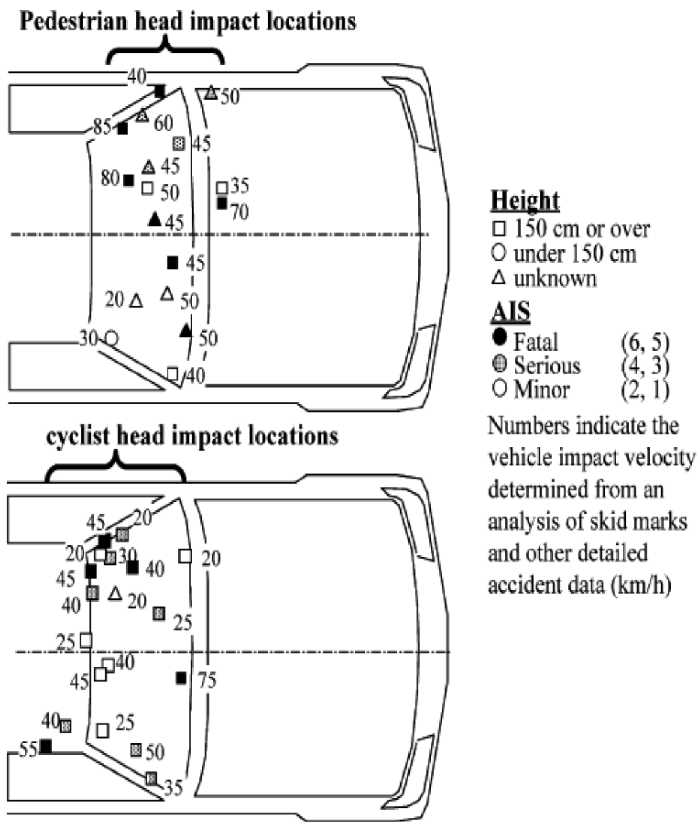


Fig. 3.10 Pedestrian and cyclist head impact locations for impacts with bonnet type vehicles, adapted from [2].

The higher cyclist centre of gravity may also partially explain the reported lower head impact velocity and impact angle [32], and hence the lower injury risk. Cadaver tests of cyclists hit by passenger cars show that the thigh and pelvis slide over the bonnet after impact, and then the upper body hits the bonnet while rotating about the pelvis. It was observed in these tests that the cyclist’s head and shoulders maintained an almost horizontal orientation up to 100 milliseconds after impact [32].

Multibody simulations of a cyclist travelling at 15 km/h perpendicular to a vehicle impacting at 30 km/h showed a 67 cm lateral shift of the head impact location on the vehicle, but the longitudinal head impact location was hardly affected [7]. The lateral direction shift is approximately equal to the transverse velocity multiplied by the time taken to reach head contact. The influence of bicycle speed on the velocity of head impact with the vehicle front is unclear as conflicting findings have been reported [15, 33].

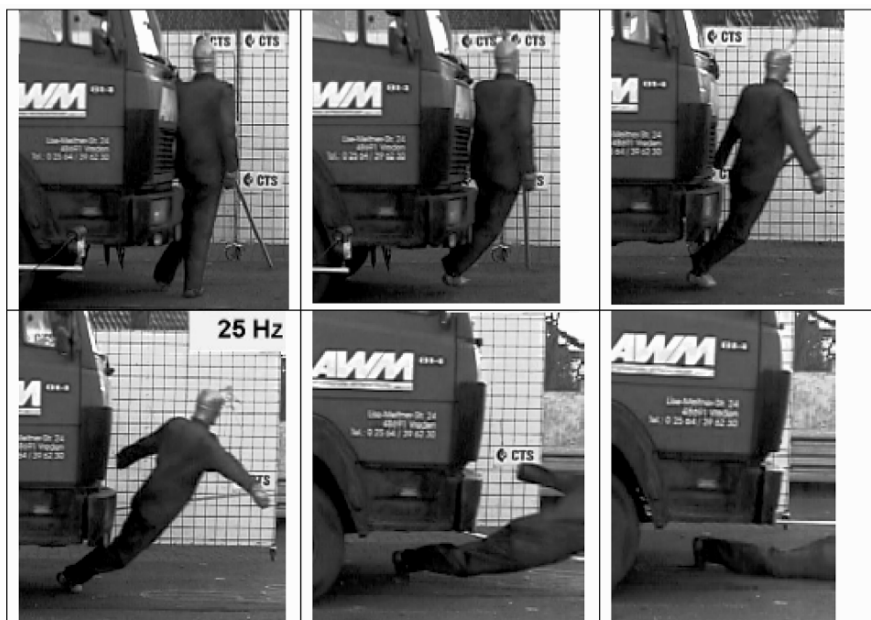


Fig. 3.11 Pedestrian forward projection kinematics: vehicle impact speed of 16.9 km/h against a stationary dummy [34].

Forward Projection

Forward projection occurs when the pedestrian/cyclist centre of gravity is within the primary impact zone. This typically occurs where an adult is struck by a minivan or a truck, a small adult is struck by a large SUV or a child is struck by a passenger car, van, truck or an SUV. The kinematics of pedestrian forward projection are more straightforward than for wrap projection. The location of first contact depends on the size of the struck body relative to the vehicle front, but subsequent contacts follow in very quick succession since the pedestrian/cyclist is struck by an almost flat and vertical surface, see Figure 3.11 [34]. As a result, the mid-body region is rapidly accelerated in the direction of vehicle travel with very little rotation so that the mid-body region and the vehicle attain an almost common post impact velocity. If the lower extremities and the head are not directly contacted (for example, when an SUV strikes an adult pedestrian), the body folds around the front shape of the vehicle, often leading to head impact on the bonnet (for adults) or bonnet leading edge (for children), see Figure 3.12.

Forward projection impacts occur provided the height of the vehicle's front leading edge is above the pedestrian/cyclist's centre of gravity height. In such cases, the impact will be effectively concentric and the mid-body region and the vehicle front attain a common post impact velocity. When the bonnet leading edge height

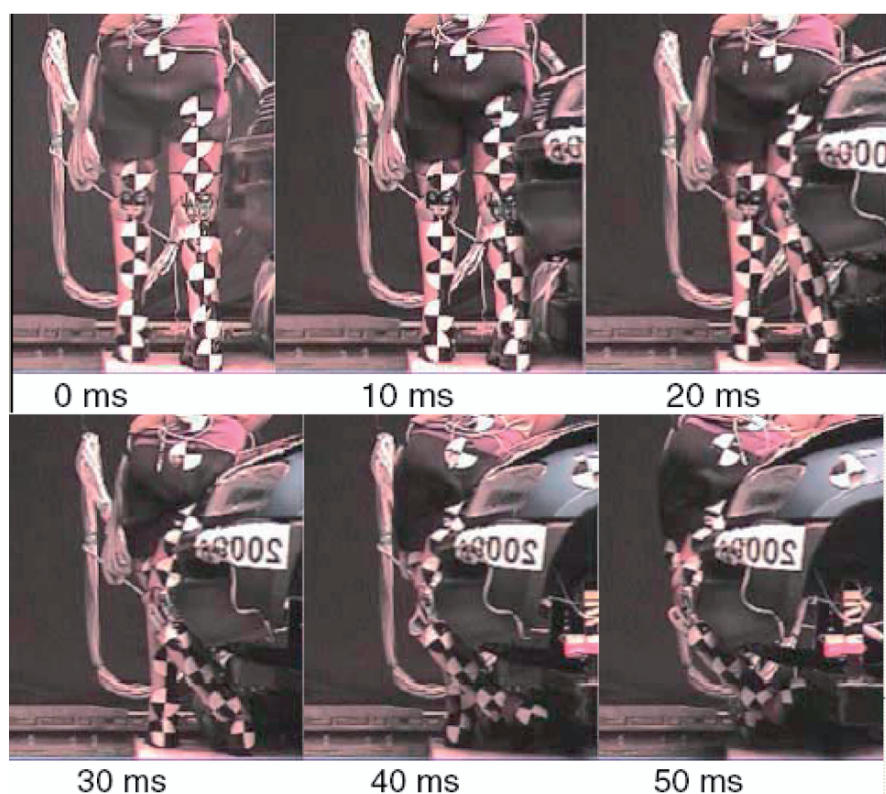


Fig. 3.12 Adult Polar pedestrian dummy struck by an SUV at 40 km/h: forward projection with substantial wrap of the upper body over the bonnet, adapted from [35].

is just below the body centre of gravity height, the resulting kinematics appear as a combination of wrap and forward projection, see Figure 3.12.

Figure 3.13 shows pedestrian and cyclist head impact locations on the wind-screen/bonnet of minivan-type vehicles in Japanese accidents between 1995 and 1998. Due to the flat fronted nature of these vehicles, there is little facility for sliding over the front profile, and WADs are similar to standing heights and no clear differences between pedestrians and cyclists are evident. The head impact speed for child pedestrians is usually lower than the pre-impact vehicle speed and head contact usually occurs with the bonnet top [13].

Multibody simulations have been used to predict that for head-on collisions between cars and cyclists, knee contact with the vehicle front-end causes the upper body to pivot upwards and slide over the bonnet [36]. However, in forward projection cases the bonnet leading edge is at or above the cyclist's hips, and then the cyclist rotates about the bonnet edge and does not slide over the top of the bonnet, see Figure 3.14.

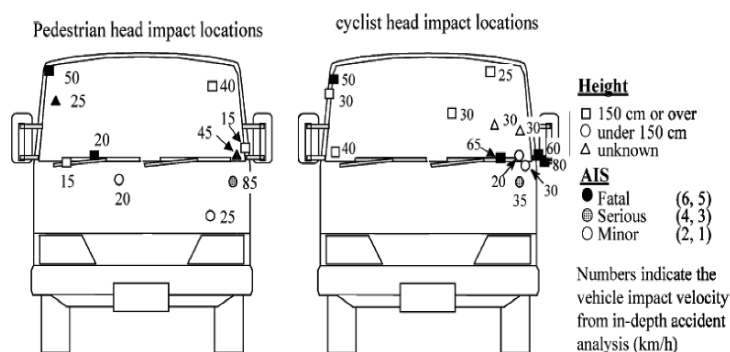


Fig. 3.13 Pedestrian and cyclist head impact locations for impacts with minivan type vehicles, adapted from [2].

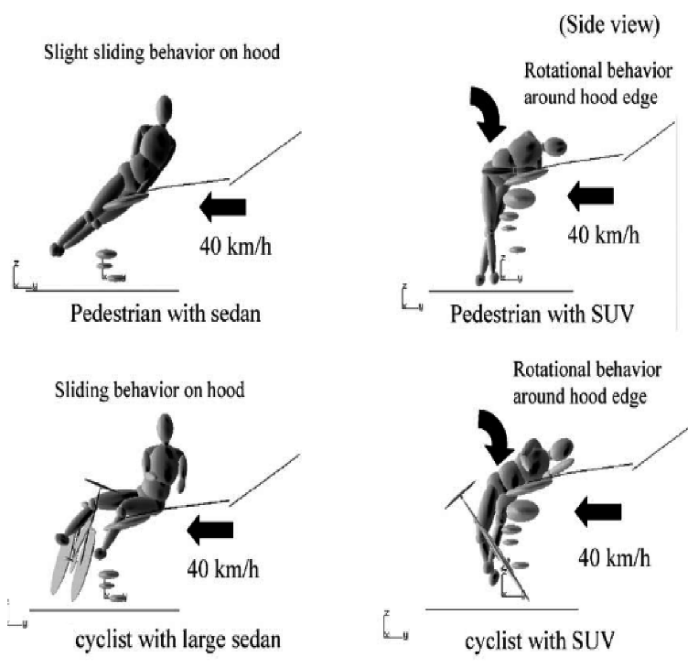


Fig. 3.14 Differences between pedestrian and cyclist kinematics when struck by cars and SUVs, adapted from [2].

Post Head Impact Kinematics for Forward and Wrap Projection Cases

Due to the focus on head injuries, pedestrian and cyclist movement after head impact with the vehicle is not as well documented as the pre head impact motion. Fig-

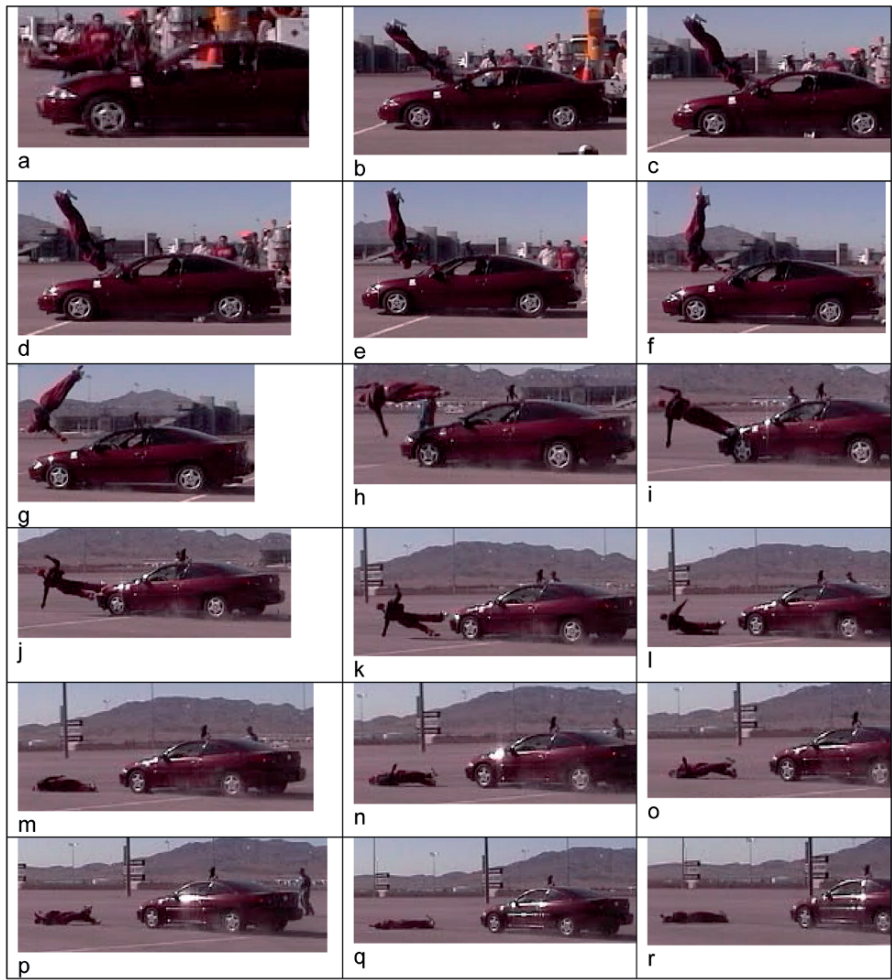


Fig. 3.15 Post head impact kinematics in staged test, adapted from [37].

ure 3.15 shows the results of a staged wrap projection impact of a stationary adult pedestrian dummy at 58 km/h [37]. Only the post head impact movement is shown. Evidence from this and other staged tests shows little restitution in the head impact [38], which therefore attains a common linear velocity with the bonnet/windscreen impact location immediately after head impact (Figures 3.15b&c). The body then pivots about the head, either clockwise in higher severity collisions (lifting the legs towards the roof) or anticlockwise in lower severity collisions (dropping the body back towards the bonnet due to gravity). The kinematics depend mostly on the vehicle impact speed, and the impact angle and angular velocity of the body at head impact, see Chapter 7.

Separation of the pedestrian/cyclist from the vehicle following head impact in wrap projections marks the beginning of the airborne trajectory (Figures 3.15e–k), and the occurrence of this depends substantially on vehicle braking. In cases where the body does not drop back onto the bonnet, separation will occur if the vehicle is braking (as seen in Figure 3.15d). However, in the subsequent somersault type movement of the body further contacts with the bonnet are possible if vehicle braking is insufficient. In lower severity cases where the body drops back onto the bonnet following head impact, separation will occur if braking is sufficient to overcome the friction between the bonnet and the body, or if the body slides/rolls/bounces off the side of the bonnet. Otherwise, the pedestrian/cyclist may be carried on the bonnet of the vehicle for an indeterminate distance. In the airborne period, the body's centre of gravity undergoes approximately parabolic motion, though this is difficult to ascertain visually from staged test evidence. Some flailing of the limbs can be observed, see Figures 3.15e–j.

For forward projection, following separation from the vehicle (mostly due to braking), the body falls over and contacts the ground due to gravitational effects and friction at the feet. In cases where the vehicle does not brake sufficiently following impact, the vehicle may run over the pedestrian.

The contact with the ground is highly variable for both wrap and forward projection pedestrian impacts, and prediction of ground contact kinematics is very difficult. However, the rotational movement of the body during the airborne portion in most wrap projection cases has two common effects on the subsequent contact with the ground. The first is that due to centrifugal forces the pedestrian/cyclist's body is usually extended rather than flexed at ground contact, and the second is that the velocity of impact with the ground is different for the various body regions. Figures 3.15k–q show one possible ground contact sequence: the feet contact the ground first, and the rolling motion result in first hip and then head contact following in quick succession. However, due to the variability, in other cases the head contacts the ground first [39], and this is then followed by the hip and feet contacts.

The vertical ground contact in the staged test in Figure 3.15 shows very little restitution, but there is substantial sliding of the dummy to rest. The degree of sliding versus rolling depends on the effective coefficient of retardation and the shape and orientation of the body when the ground is contacted, see Chapter 9.

In addition to the effect of braking on vehicle/pedestrian separation following head impact, pre-impact braking also reduces the height of the vehicle front due to compliance in the suspension system. This increases the angular velocity of the pedestrian post primary impact (see Chapter 7) and therefore reduces the time to head contact. However, the vehicle retardation due to braking reduces the forward movement of the bonnet/windscreen and also reduces the linear velocity of the pedestrian centre of gravity, which increases the time to head contact, so that a degree of cancellation occurs.

There is no evidence to date of differences between pedestrian and cyclist ground contact mechanisms, and this should be a focus of future research.

Concluding Remarks

Film evidence of pedestrian and cyclist kinematics from real accidents and from staged tests has been presented. In most cases pedestrian kinematics can be classified as either wrap or forward projection, depending on the height of the principal impact forces relative to the body centre of gravity, and body stance and vehicle speed at impact. The principles of wrap and forward projection apply equally to cyclists as to pedestrians. However, despite these well known classifications, there is substantial variability in the impact movements for both pedestrians and cyclists, especially in the later phases of the impact where contact with the ground occurs. The motion patterns described in this chapter are key to understanding the causes of pedestrian and cyclist injuries (Chapter 2) and for the development of vehicle front structures for pedestrian and cyclist protection (Chapter 10).

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Chapter 4

The Relationship between Vehicle Impact Speed and Pedestrian and Cyclist Projection Distance

Introduction

Many factors affect pedestrian and cyclist injury risk in a vehicle collision, but the most important determinant is the vehicle impact speed. This has been observed empirically (see Chapter 2) and is also evident from a theoretical basis, see Chapter 7. Vehicle impact speed therefore has implications for legislators in designing speed limits for urban areas, for automotive engineers to reduce vehicle aggressivity and for biomechanics research into injury causation. It also has legal implications in determining culpability following a collision [1] and driver statements of vehicle speed are therefore unreliable and other independent methods to estimate vehicle impact speed have been proposed.

The methods frequently used to estimate vehicle speed include the use of witness statements, tyre skid marks on the road, impact locations/damage on the vehicle and pedestrian projection distance. However, witness statements are subjective, tyre skid marks are now less common due to ABS braking and impact locations/damage on the vehicle depend on the collision configuration and are unreliable predictors of vehicle speed [2]. An alternative method is therefore necessary, and the strong correlation between vehicle impact speed and the distance that impacted pedestrians and cyclists are projected in an impact has been known for over 30 years (e.g. [3]). Since then, a variety of methods have been proposed to estimate vehicle impact speed based on measurement of the pedestrian projection distance.¹ This chapter presents accident investigation data relating to pedestrian and cyclist projection distances and a comparison is made with staged tests involving pedestrian/cyclist dummies and cadavers. A theoretical framework for modelling pedestrian impact and projection is then proposed, and a review of models in the literature is given. There are many such models, but emphasis is placed on those predicated on representing the mechanics of the collision event rather than on the many regression models based on staged tests or accident data.

¹ In accident reconstruction vehicle impact speed is estimated from pedestrian projection distance. However, in reality the vehicle speed is the independent variable.

Stages of Pedestrian and Cyclist Projection

Chapter 3 showed that most pedestrian impacts can be categorised according to the relative geometries of the vehicle and pedestrian into wrap projection, forward projection or fender vault cases. The latter represents an ‘incomplete’ impact between the vehicle and the pedestrian, and vehicle speed reconstruction should generally not be attempted from pedestrian projection distance in these cases. Many cyclist impacts can also be categorised as wrap projection or forward projection, but due to the interaction of bicycles with the striking vehicle there are also many cases where full engagement of the vehicle with the cyclist does not occur, see Chapter 3. In these cases, vehicle speed reconstruction should not generally be attempted from cyclist projection distance.

For wrap and forward projection cases (which together represent about 80% of pedestrian accidents, see Chapter 3), the video still-shots of vehicle pedestrian collisions presented in Chapter 3 show that there are three principal stages in the impact process which apply to both wrap and forward projection:

1. The vehicle-pedestrian impact in which momentum is transferred from the vehicle to the pedestrian. This stage terminates when the pedestrian and vehicle separate.
2. Depending on the collision configuration, there follows either
 - a. a period of flight terminated by first contact with the ground for wrap projection; or
 - b. a fall-over stage terminated by contact of the pedestrian midbody region with the ground for forward projection.
3. Slide/roll/bounce to rest following first contact with the ground [4, 5].

The total projection distance is the combined distance travelled by the pedestrian in these three stages. Where there is a substantial pre-impact velocity of the pedestrian perpendicular to the direction of vehicle travel, the projection distance has components parallel and perpendicular to the vehicle travel direction. However, both fundamental mechanics and empirical evidence [6] indicate that these are substantially uncoupled and, assuming that full engagement with the vehicle front has occurred, the pedestrian projection distance parallel to the vehicle travel direction can still be used to estimate vehicle impact speed.

The same principles should apply for cyclists, but the addition of the bicycle makes the impact process more complex and there is much less real accident or staged data available for cyclist impacts compared to pedestrian impacts. Due to the generally increased cyclist speed relative to pedestrians (see Chapter 3), interactive effects between cyclist speed and the level of impact engagement with the vehicle are possible. This was predicted using multibody modelling of vehicle/cyclist impacts: early work showed projection distances for cyclists struck from the side increase if the bicycle’s initial speed is increased from 0 to 20 km/h [7], though a more recent study has reported the opposite effect [8], and further research is required. Chapter 3 showed that cyclist impact from the side by the front of vehicles is

the most frequent collision configuration, but there are many cases where either the front or the rear side of the bicycle is struck, but the cyclist is not directly engaged in the impact. Multibody modeling of side struck cyclists where full engagement has occurred showed that the kinematics of cyclists and pedestrians in wrap and forward projection impacts are similar [9–12], see Chapter 3. In particular, it has been noted that interaction between the cyclist and the bicycle is negligible following vehicle impact, so that the principal differences arise from the initial position of the cyclist compared to a pedestrian [9, 12]. These mainly result in head impact further back along the bonnet/windscreen compared to pedestrians (see Chapters 3 and 10) and in these cases, cyclist and pedestrian projection distances may be expected to be similar. Analysis of pedestrian and cyclist accident data has confirmed this [13].

For cyclist accidents, both the cyclist and the bicycle are impacted, and the projection distance of bicycles has been reported to be greater than for cyclists [14], probably due to the higher coefficient of restitution for the vehicle/bicycle and lower friction for bicycle/ground impacts compared to the vehicle/cyclist and cyclist/ground impacts. Both cyclist and bicycle projection distances could therefore be used to calculate the vehicle impact speed, but staged tests have shown greater variations in bicycle projection distances compared to projection distances of dummies or cadavers used as surrogate cyclists [7, 15], and therefore the cyclist projection distance is generally used for bicycle accident reconstruction purposes.

Post Impact Separation from the Vehicle

Separation of the pedestrian or cyclist from the vehicle following the impact stage is complex and may be difficult to ascertain. The video stills in Chapter 3 showed that, following head contact with the bonnet or windscreen in wrap projection cases, the body of the pedestrian may drop back down onto the bonnet or continue to rotate away from the bonnet. This depends on the angular velocity of the body following head impact, which is mainly a function of the vehicle and pedestrian geometry and the impact speed, see Chapter 7. Separation may occur at high speeds due to rotation of the pedestrian over the roof of the vehicle. In cases where the pedestrian's body drops back onto the bonnet due to gravity following head impact, separation will also occur if vehicle braking is sufficient to overcome friction between the pedestrian and the bonnet/windscreen. Similar principles apply to the separation of a cyclist from the striking vehicle, but the higher pre-impact transverse velocity may result in earlier separation, though further research is required to clarify this. In cases where separation of the pedestrian or cyclist from the vehicle does not occur immediately following the impact stage, the relationship between vehicle impact speed and projection distance is indeterminate.

Effective Coefficient of Retardation in the Ground Contact

The Coulomb friction coefficient μ is the ratio of the friction force to the normal force between two contacting surfaces. However, pedestrian contact with the road surface following vehicle impact is complex due to the combination of rigid body sliding and rolling, intermittent contact and body deformation. Therefore the effective coefficient of retardation rather than the coefficient of friction is a more appropriate term and Wood and Simms [16] analysed accident reconstructions and staged tests and found that the coefficient of retardation is approximately normally distributed² with a reported range of $\mu = 0.39$ to 0.87 . There is no data relating to the coefficient of retardation for cyclists, but it is unlikely to be significantly different than for pedestrians.

Accident Data

There are many sources of accident data relating vehicle impact speed to pedestrian and cyclist projection distance [6, 17–24]. Vehicle impact speed in accident reconstructions can sometimes be determined from tyre skid-marks or video analysis and related to the measured pedestrian projection distance, and these are referred to as accident reconstruction cases. Other data sources are the numerous staged collisions with pedestrian dummies and cadavers which have been performed since the 1960s (e.g., [4, 16, 21, 25–27]). Attempts to use a model to understand the biomechanical relationship between vehicle impact speed and pedestrian projection distance must involve a comparison with reconstructed accidents or with staged tests, and therefore a summary of the accident data is presented here.

Figure 4.1 shows pedestrian wrap projection data from accident reconstructions [6, 17–24]. Significant scatter is evident, probably reflecting some differences in data collection methods, but also showing the variability of vehicle pedestrian impacts, even for cases categorised as wrap projections. However, despite this variability, Figure 4.1 shows a clear pattern in the data and the reported scatter is reasonably consistent for all of the datasets except for that of Otte [22]. Otte reported on 355 pedestrian accidents between 1985 and 2001, including some cases with mean deceleration levels³ post separation of over 3 g which are much higher than those in the other datasets. Figure 4.2 shows the Otte wrap projection data [22] separately to all of the other accident reconstruction wrap projection data [6, 17–21, 23, 24], which have been now grouped as a single data set, as well as four recorded cases of pedestrian fender vault [6]. The maximum theoretical impact velocity for a given projection distance based on Searle's particle projection model [28] assuming a horizontal launch and a plastic impact, see equation (4.3), a maximum coefficient of friction of $\mu = 0.87$ [16], a maximum pedestrian mass of 117 kg [29] and a minimum

² Mean = 0.56, standard deviation 0.1.

³ Mean deceleration ($\bar{a}$) based on the equation of uniform acceleration: $\bar{a} = v^2/2s$.

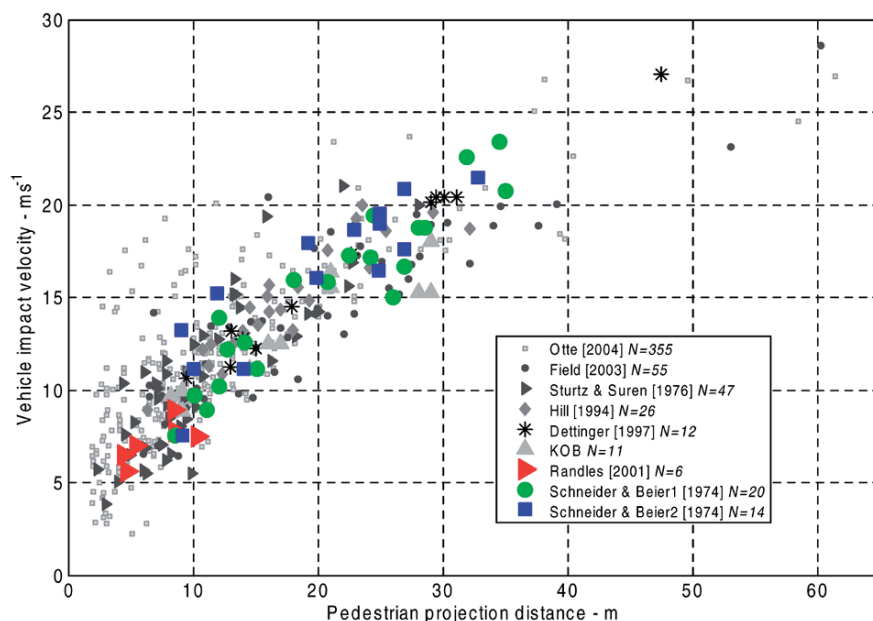


Fig. 4.1 Relationship between vehicle impact speed and pedestrian projection distance for wrap projection cases from accident reconstructions [6, 17–24].

vehicle mass of 597 kg [29]) is also shown. Comparing this theoretical curve with the wrap projection accident data and the four fender vault accident cases indicates that some of the Otte [22] cases were fender vault cases which were misclassified as wrap projections. This illustrates the practical difficulties associated with classification of the kinematics for some vehicle pedestrian accidents when limited post accident information is available. For comparison with forward projection cases and with model predictions, only the accident reconstruction dataset of [6, 17–21, 23, 24] are therefore used, and the Otte 2004 [22] data are not included.

Figure 4.3 shows the real-world accident reconstruction data for pedestrian forward projection cases [2, 6, 18, 19, 24]. There are much fewer cases compared to wrap projection, and many involve children, because most adult pedestrian impacts by passenger cars are wrap projection cases. There is no clear distinction between adult and child pedestrian projection distances in forward projection, as expected by consideration of equation (4.3).

A comparison of the wrap and forward projection accident data is shown in Figure 4.4. The relationship between vehicle impact speed and projection distance is broadly similar in both cases, but for a given vehicle impact speed, the pedestrian projection distance is slightly higher for forward compared to wrap projection cases. This results from a higher momentum transfer in forward projection cases, and also leads to more severe injuries (see Chapter 10), but further research is required to investigate if the body orientation at the instant of ground contact is influenced in

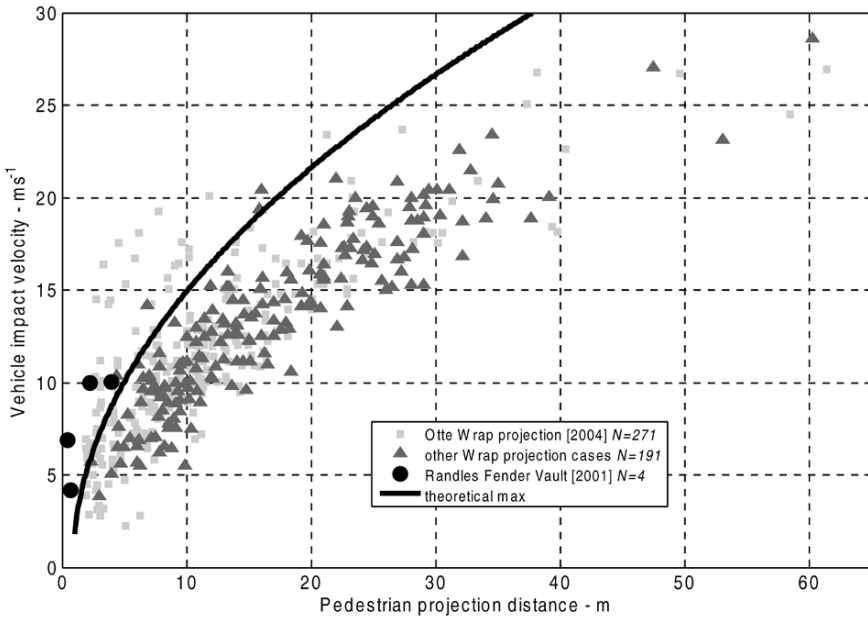


Fig. 4.2 Relationship between vehicle impact speed and pedestrian projection distance for Otte [22] and other [6, 17-21, 23, 24] wrap projection cases, as well as four reported fender vault cases [6] and a theoretical maximum impact velocity curve.

a predictable manner by vehicle shape, as this may affect the slide to rest distance following initial ground impact, see Chapter 9. However, a general implication is that pedestrian projection distance is influenced by vehicle shape.

Otte [22] reported cyclist projection distances from 349 bicycle accidents between 1985 and 2001, and it is assumed that these were wrap projection cases as the vehicles involved were described as cars, see Figure 4.5. The Otte [22] pedestrian wrap projection cases from the same paper and the previously outlined theoretical maximum velocity curve are also shown. Figure 4.5 indicates that some of the cyclist cases were not full engagement wrap projections – similar to the pedestrian data from the same paper – but there appears to be no other real accident data for comparison purposes. It is also clear that the relationship between vehicle impact speed and projection distance is very similar for pedestrians and cyclists, as had previously been reported [14]. However, the influence of higher bonnet leading edge height on projection distance is greater for cyclists than for pedestrians (25% higher for impact speeds of 30 km/h [7]), probably due to the higher pelvis height of the former.

The clear functional relationships between impact speed and projection distance in forward and wrap projection evident in Figures 4.1–4.5 for pedestrians and cyclists are only observed in cases where separation of the vehicle and pedestrian occurred following the impact process. As stated previously, in cases where the ped-

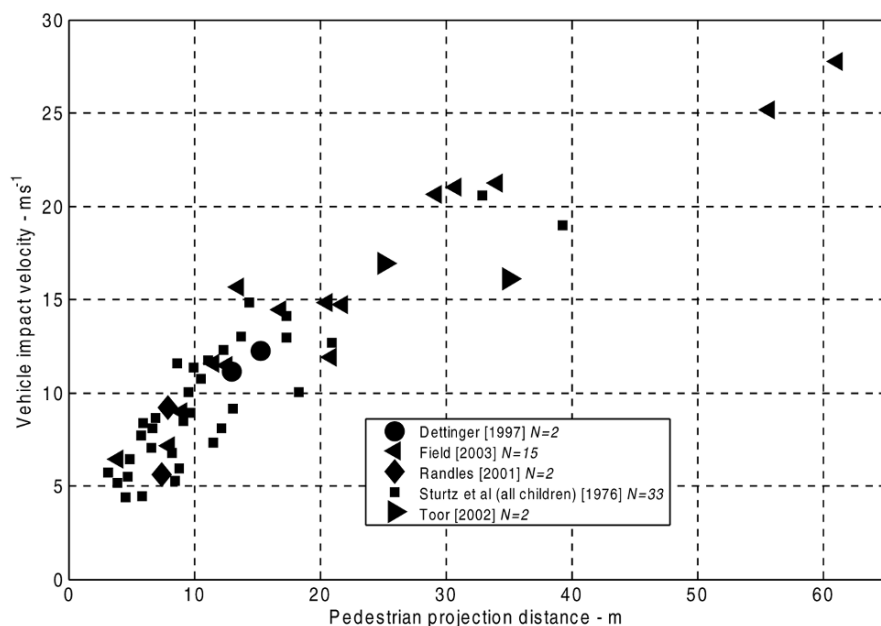


Fig. 4.3 Relationship between vehicle impact speed and pedestrian projection distance for real-world forward projection cases [2, 6, 18, 19, 24].

estrian or cyclist is carried on the vehicle, the projection distance is indeterminate, as shown by analysis of film evidence of a pedestrian accident [6].

Staged Tests

Several researchers have published projection distance data for staged pedestrian and cyclist impact tests (e.g. [4, 15, 27, 30]), but this data will not be considered in detail here except to complement and compare to the available accident data. Figure 4.4 showed limited data for pedestrian accident wrap projection cases for projection distances less than 5 m. To address this, Fugger et al performed dummy impact tests in the speed range 8–43 km/h [4]. Figure 4.6 shows their data together with the pedestrian accident reconstruction wrap projection data from [6, 17–21, 23, 24]. In the 1980s, the KOB pedestrian accident research project compared accident reconstruction cases to staged dummy and cadaver tests, see Figure 4.7 [21].

For cyclists, Huibers and Janssen [7] performed staged side impact tests using cadavers and dummies as surrogate cyclists, and Haight and Eubanks [27] performed staged front-rear impacts using a crash dummy as a surrogate cyclist, see Figure 4.8. The available cyclist accident projection data from Otte [22] is also shown, together with the previously defined theoretical maximum impact velocity curve. It has been

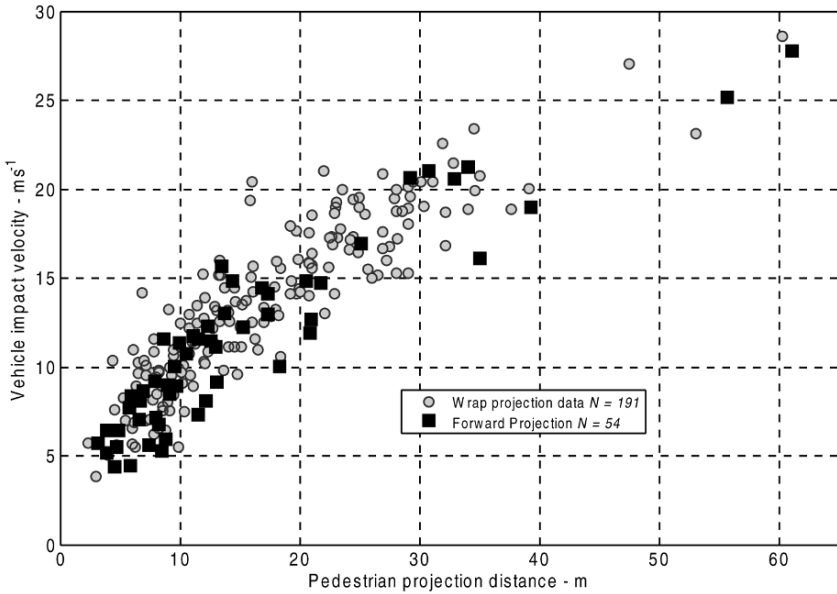


Fig. 4.4 Comparison between wrap projection [6, 17-21, 23, 24] and forward projection [2, 6, 18, 19, 24] cases from accident reconstructions.

reported that impact direction has very little influence on cyclist projection distance [7], but Figure 4.8 does not support this conclusion, as the front/rear dummy tests reported by [27] show significantly higher projection distances for a given vehicle impact speed, and further research is required to clarify this. Staged cyclist impact tests using child and adult dummies struck by passenger cars showed that the projection distance for a child dummy is approximately 30% higher than for an adult dummy in a 30 km/h impact [7]. Since child impact cases with passenger cars are usually forward projections, and adult impacts are generally wrap projection cases, this corroborates the finding for pedestrians that forward projections result in slightly larger projection distances than wrap projection cases, see Figure 4.4.

Comparison between Accident Reconstructions and Staged Tests

Accident reconstructions have the advantage that they relate to real-world events, but there can be significant error in the estimation of vehicle speed, in the measurement of pedestrian projection distance [31] and in the categorisation of individual collision cases, as seen with the Otte data [22]. By comparison, in staged tests using crash dummies the vehicle impact speed and projection distance can be accurately

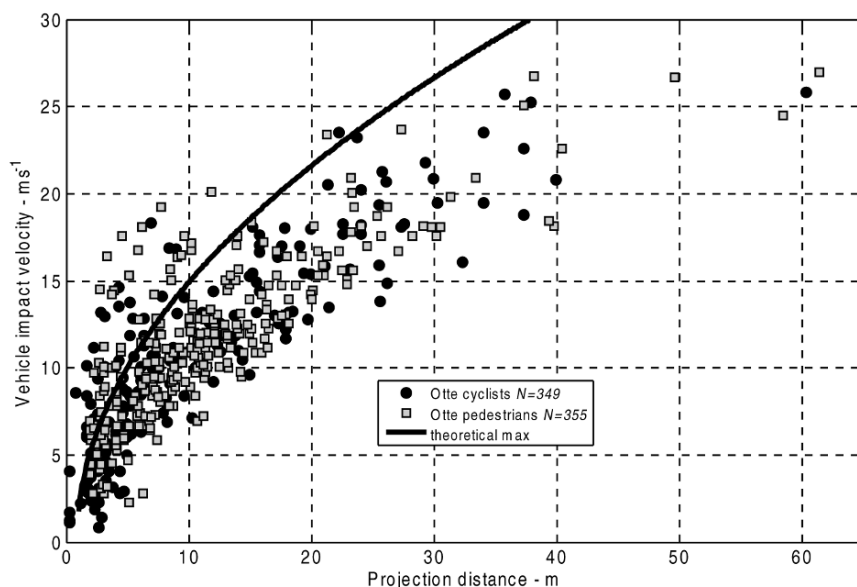


Fig. 4.5 Wrap projection cyclist projection distances as a function of vehicle impact speed for accidents reported by Otte [22]. The Otte pedestrian wrap projection cases and the theoretical maximum vehicle impact velocity curve from a horizontal slide model are also presented.

measured, but there are inevitable problems with the biofidelity of crash dummies⁴ and with the range of human anthropometrics and vehicle shapes represented by crash dummies and test vehicles respectively, see Chapter 8. Furthermore, only a limited range of collision configurations can be tested. The KOB pedestrian data in Figure 4.7 and the cyclist data in Figure 4.8 show that staged tests have less variability than accident cases, and the pedestrian dummy tests contain the least variability. Real accidents contain a much greater range of initial impact configurations and the staged cyclist tests generally do not contain any of the partial engagement cases discussed previously for real accidents. Staged tests using cadavers clearly have improved biofidelity compared to dummy tests, but the subjects tend to be limited to older people who may not represent the shape, mass and stiffness of the wider pedestrian population. In addition, the same issues of limited representation of the range of actual impact configurations apply and the sample size of the available test data is small. Finally, pedestrians are nearly always moving when struck ([33] and Chapter 3), and this has generally not been reproduced in published staged tests due to technical difficulties.

The accident reconstruction data shows that despite the strong correlation there is also considerable variability in vehicle impact speed for a given pedestrian projection distance. Consideration of the impact process shows that there are many con-

⁴ A computational study found that the use of a crash dummy with incorrect mass distribution can introduce significant errors in pedestrian projection distance [32].

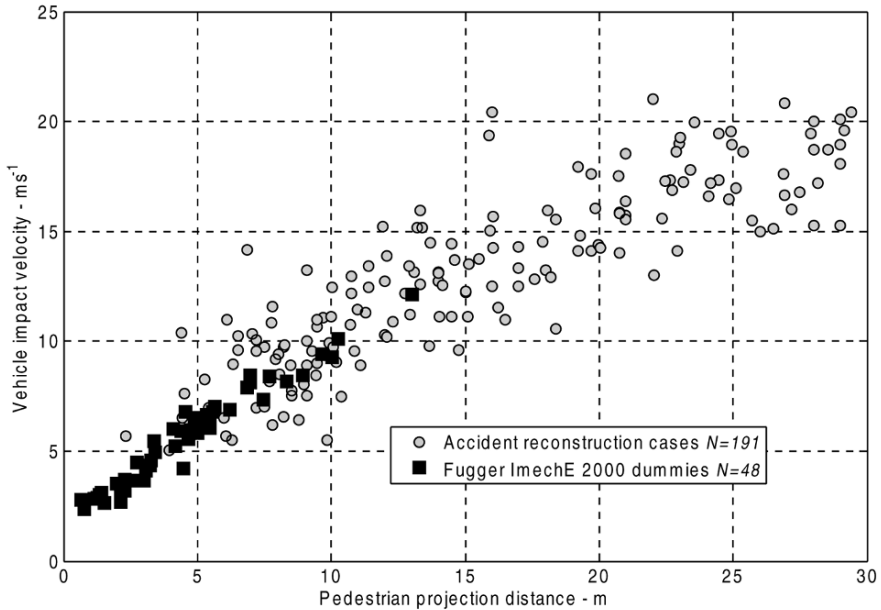


Fig. 4.6 Comparison between pedestrian wrap projection data from real accidents [6, 17–21, 23, 24] and from low speed staged dummy impacts [4].

founding factors that contribute towards this variability, including vehicle geometry, pedestrian anthropometrics, impact duration, degree of vehicle braking, effects of suspension dynamics, continued pedestrian/vehicle interaction after secondary impact in a wrap projection, pedestrian pre-impact transverse velocity,⁵ foot-ground contact during the fall-over phase in forward projections, and the effects of slide, roll and bounce on effective retardation following contact with the ground.

Regression Models Relating Impact Speed to Pedestrian Projection Distance

The strong correlations in Figure 4.4 indicate that regression techniques can be used to define a mathematical relationship with which to reconstruct vehicle impact speed from pedestrian projection distance, and this has been done since the 1960s [26]. In particular, several researchers have used the relationship between projection velocity

⁵ The complexity of vehicle pedestrian interaction is illustrated by the finding that pedestrians moving towards the centre of the vehicle at impact are normally thrown in the direction opposite to their original direction of travel. In contrast, pedestrians moving away from the centre of the vehicle at impact continue in that direction [23].

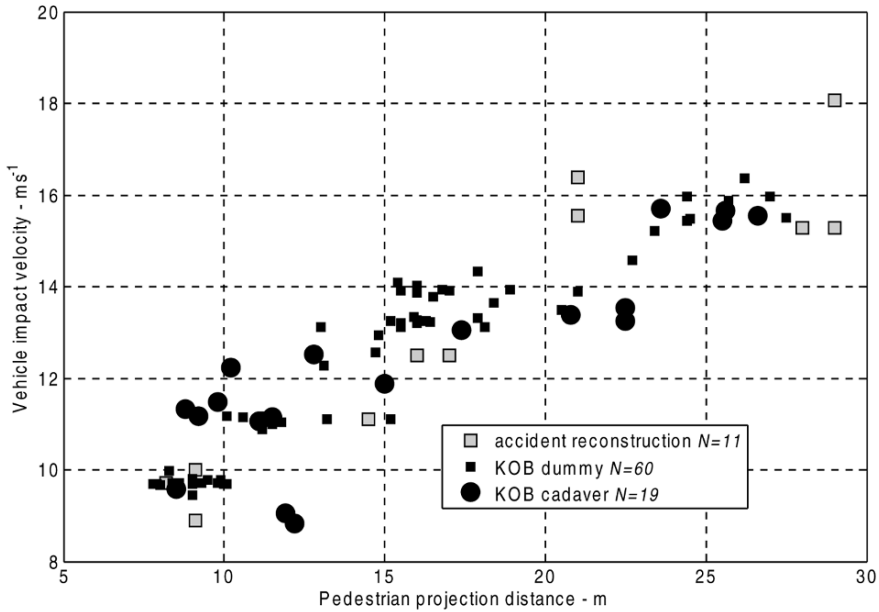


Fig. 4.7 Comparison between KOB [21] real accident data staged cadaver and dummy tests.

and square root of projection distance inherent in Searle's particle launch model⁶ to regress impact velocity versus the square root of pedestrian projection distance for both wrap and forward projections, see Figures 4.9 and 4.10 [33, 34]. This simple approach yields a good fit over a range of projection distances for which sufficient accident reconstruction data is available. However the accident data is sparse below about 5 m projection distance for both wrap and forward projections, and a comparison with the dummy impact tests performed by Fugger et al. [4, 30] shows that the regression curve overestimates impact velocity at low projection distances for both wrap and forward projections, see Figures 4.9 and 4.10.

A further problem with the regression approach arises when the variability in the data is considered. Evans and Smith [33] used accident data from Hill [20] and Dettinger [17, 18] to yield a regression model for wrap projection cases with 95%ile confidence limits which clearly do not bound 95% of accidents when other researcher's data are considered, see Figure 4.11. This highlights a pitfall of the empirical approach, which is that predictions are only valid for the dataset used to perform the regression. Since Evans and Smith [2, 33] used a subset of the accident datasets which have less variability, their resulting confidence limits are too tight. Toor et al. [2] partially addressed this by using additional data to yield their regression model, and their 90% confidence limits predictions are therefore more representative, see Figure 4.11.

⁶ See equation (4.3) to follow.

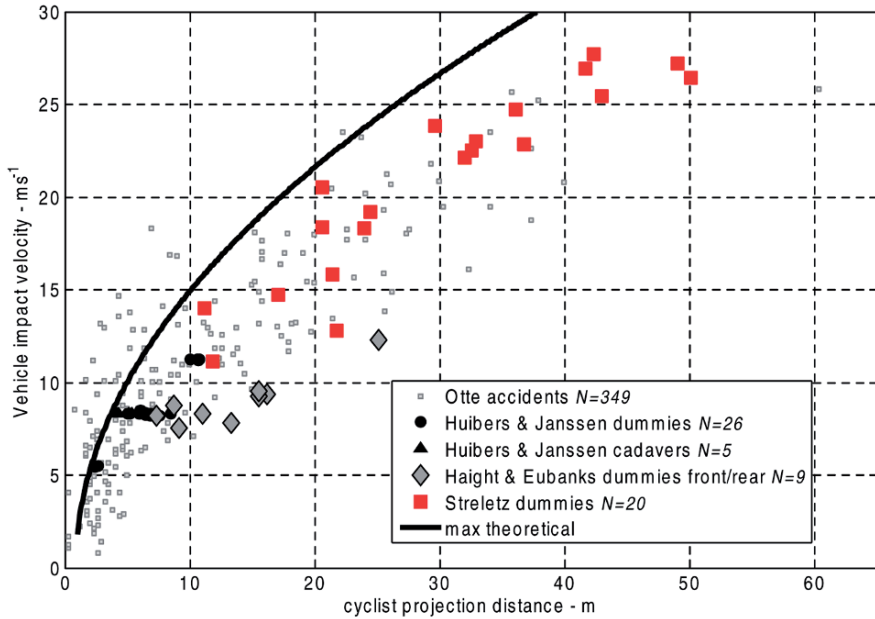


Fig. 4.8 Cyclist projection distances: real accidents from Otte [22], cadaver and dummy tests from Huibers and Janssen and Streletz [7, 15] and dummy front-rear tests from Haight and Eubanks [27]. The theoretical maximum vehicle impact velocity curve from a horizontal slide model is also presented.

However, from a biomechanical perspective the real drawback of this empirical approach is that it does not contribute towards our understanding of the mechanics of the impact process. For forward projection, the pedestrian always falls to the ground due to friction acting on the feet, and therefore the projection distance will be at least the height of the pedestrian centre of gravity above the ground, even for very low velocity impact. This was clearly shown in dummy impact tests performed by Fugger et al. [30], but regressions of the accident reconstruction data do not reflect this, due to the lack of very low speed data. Therefore, regression models not only provide no physical insight, but also yield physically incorrect predictions at low impact speeds for forward projection.

Physics Based Models Relating Impact Speed to Pedestrian Projection Distance

A better modelling approach is to consider the mechanics of the collision process. Examination of film evidence in Chapter 3 and the pedestrian accident reconstruction data in Figure 4.4 indicates that there are well-defined impact processes govern-

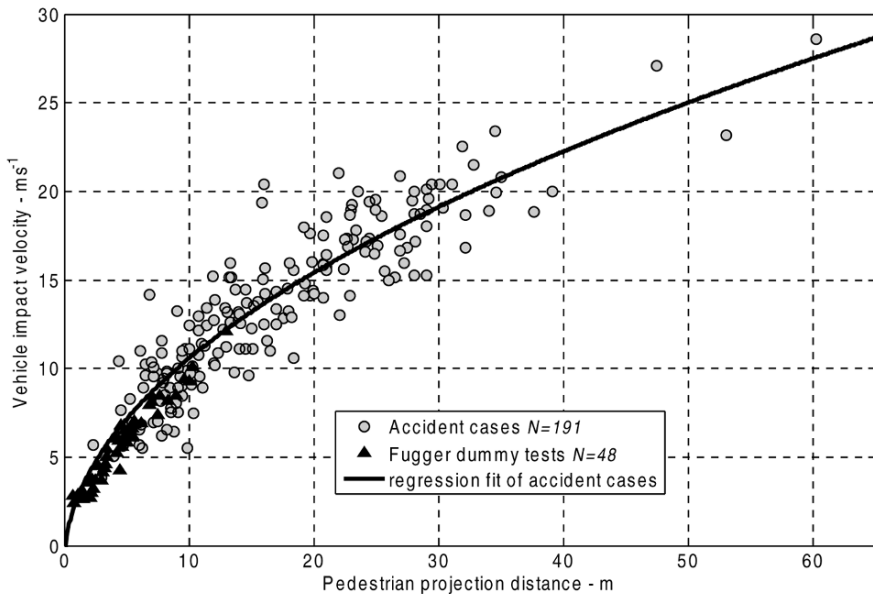


Fig. 4.9 Linear regressions between vehicle impact speed and square root of pedestrian projection distance for wrap projection cases compared to accident reconstruction data from [6, 17–21, 23, 24] and staged dummy data from [4].

ing the total projection distance – albeit different for wrap and forward projection. Furthermore, the variability in the accident reconstruction data also suggests that deterministic modelling is inherently inappropriate. Therefore, physics based models which include the principal sources of uncertainty and which can successfully predict both the average and the range of impact speeds for different pedestrian projection distances are required. Two such models will be presented here for wrap and forward projection respectively which capture the impact process and also provide a means of vehicle impact speed reconstruction from pedestrian projection distance. These models will be compared to others presented in the literature, but as a precursor, the particle projection model presented by Searle is presented first, since it provides insight to the projection process.

Theoretical Considerations: The Particle Projection Model

A major simplification of the projection process is achieved by considering the pedestrian as a particle with initial velocity components $\dot{x}(t_{1+})$ and $\dot{z}(t_{1+})$ and an initial

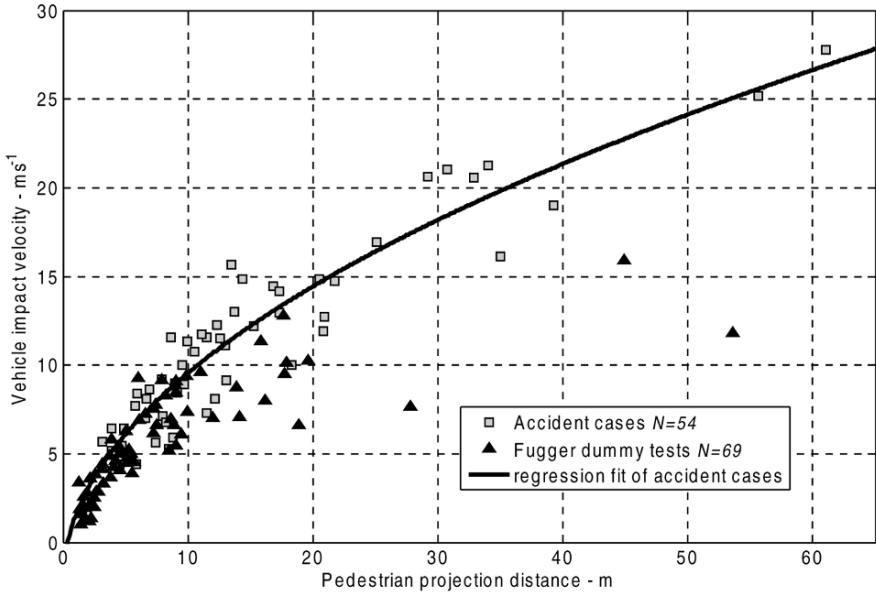


Fig. 4.10 Linear regressions between vehicle impact speed and square root of pedestrian projection distance for forward projection cases compared to accident reconstruction data from [2, 6, 18, 19, 24] and staged dummy data from [30].

height p above the road surface as proposed by Searle [28].⁷ The intermittent reaction force from the ground to the particle is R and the retardation force is μR .

The equations of motion for the particle are

$$\begin{aligned} m\ddot{x} &= -\mu R, \\ m\ddot{z} &= R - mg. \end{aligned} \quad (4.1)$$

Searle solved these equations to yield:

$$[\dot{x}(t_{1+}) + \mu\dot{z}(t_{1+})]^2 = 2\mu g(S - \mu p). \quad (4.2)$$

The total projection distance S in equation (4.2) depends on the relationship between the retardation coefficient μ and the vertical launch velocity $\dot{z}(t_{1+})$. Searle's model accounts for restitution effects in the vertical pedestrian/ground contact and he used an elegant analysis to show that intermittent contact between the pedestrian and the ground has no effect on the effective retardation and hence on the overall projection distance. However, application of Searle's equation requires knowledge of the horizontal and vertical launch velocities and therefore his equation cannot gener-

⁷ Using the notation from Chapter 7: $\dot{x}(t_{1+})$ refers to the x direction component of velocity of the pedestrian cg just after head contact with the vehicle.

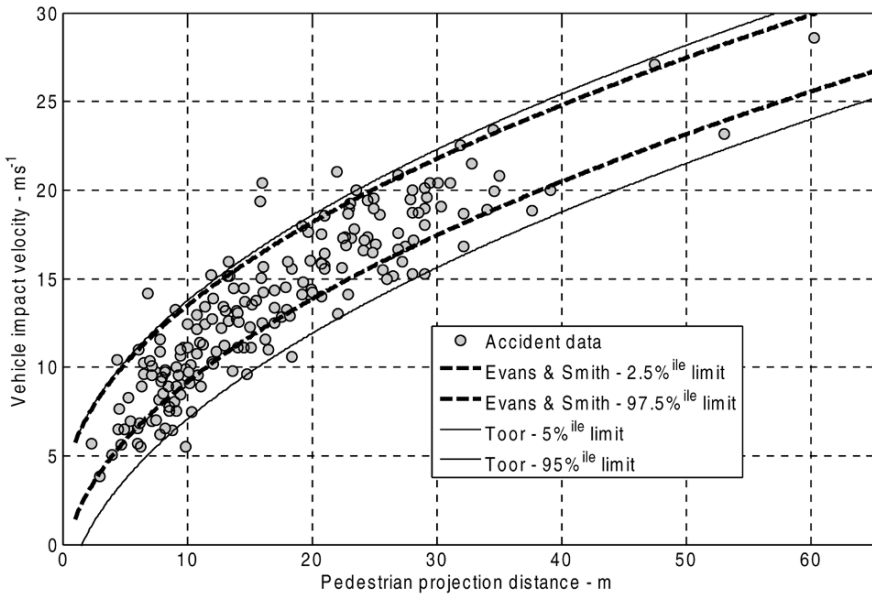


Fig. 4.11 Predicted limits from empirical models from Evans and Smith [33] and Toor et al. [2] for wrap projection compared to accident data from [6, 17–21, 23, 24].

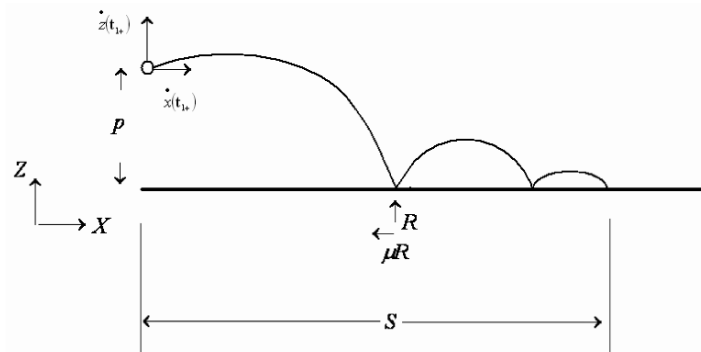


Fig. 4.12 Pedestrian particle projection model proposed by Searle [28].

ally be directly applied. Moreover, the equation relates projection distance to launch velocity rather than vehicle impact speed. In order to relate directly to the vehicle pre-impact speed ($\dot{X}(t_{0-})$), the impact process must be considered. Considering the simplest case of a concentric impact (i.e., $h = 0$ in equation (7.12) and hence the vertical launch velocity is zero), then assuming zero restitution the conservation of momentum yields:

$$\dot{X}(t_{0-}) = \sqrt{2\mu g(S - \mu p)} \left[\frac{M + m}{M} \right]. \quad (4.3)$$

Equation (4.3) is a major simplification because it assumes a concentric impact on a ‘particle’ pedestrian, but it does show that the projection distance is approximately proportional to the square of the collision speed, is only weakly dependent on variations in vehicle/pedestrian mass and is inversely proportional to the coefficient of retardation. Assuming limiting values ($\mu_{\max} = 0.87$, $m_{\max} = 117$ kg and $M_{\min} = 597$ kg [16, 29]) the maximum theoretical vehicle impact speed for a given projection distance presented in Figures 4.2, 4.5 and 4.8 can be found. However, the rotation of the pedestrian onto the vehicle front and the subsequent flight phase in wrap projection are not accounted for using this approach. Similarly, the impact phase and eccentricity of the friction force at the feet with respect to the body centre of gravity in forward projection are not included.

A more representative approach is achieved by observing the three principal projection stages for both wrap and forward projections [5, 35, 36] and physically based models should explicitly represent the projection distances associated with these stages. Separate models for wrap and forward projection are presented here.

Wrap Projection

For wrap projection, the total distance traveled (S_{total}) results from impact, flight and slide/roll/bounce to rest such that

$$S_{\text{total}} = S_{\text{impact}} + S_{\text{flight}} + S_{\text{slide/roll/bounce}}. \quad (4.4)$$

The impact theory in Chapter 7 shows that a single segment representation of the pedestrian can adequately predict whole-body kinematics in wrap projections. Therefore, using the post-head-impact horizontal and vertical velocity of the pedestrian centre of gravity predicted by a single segment model as launch velocities for the particle flight model proposed by Searle [28] yields a hybrid model for wrap projection which accounts for the impact stage, flight through the air and the subsequent bouncing and skidding to rest [29]. For a given set of input parameters, this model is deterministic and therefore predicts a unique result. However, given the statistical distribution of the input variables for this model (vehicle and pedestrian mass, degree of braking, coefficient of retardation, bonnet droop due to braking, vehicle shape), the application of Monte Carlo methods yields a distribution of vehicle impact speeds for a given projection distance, see Figure 4.13 [5].

The single segment hybrid model scatterplot predictions encompass the wrap projection accident data very well. Statistical analysis shows that the standard deviations predicted by the model are similar to those of the accident reconstruction data (see Figure 4.14), though it can be seen that at very low projection distances the standard deviations for the accident data are greater than the model predictions. This underlines the problem that for wrap projection cases the uncertainty due to

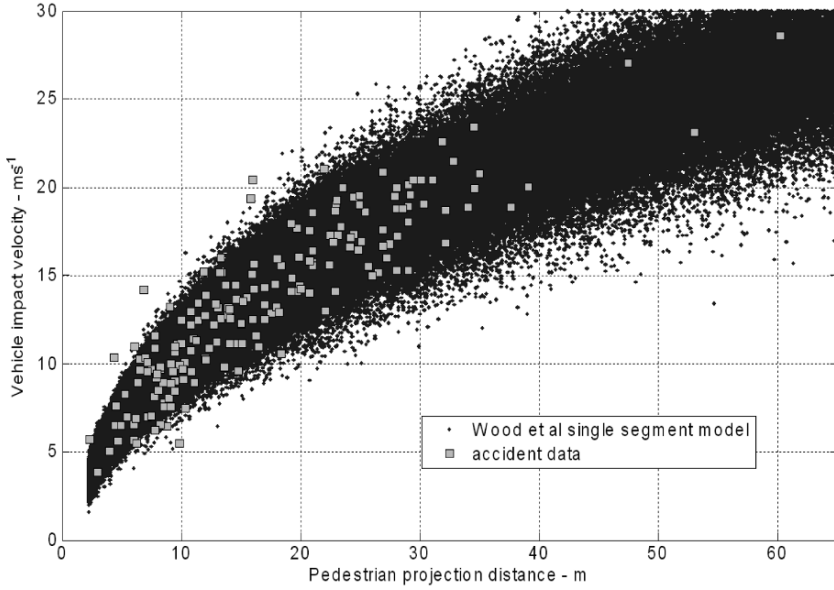


Fig. 4.13 Comparison of single segment model scatterplot predictions [5] with pedestrian wrap projection accident data from [6, 17–21, 23, 24].

the complexity of the vehicle pedestrian interaction introduces relatively high errors at low impact speeds. At high speeds, the strong relationship between impact speed and projection distance inherent in the slide model predominates.

Pedestrian Forward Projection

For forward projection, the total distance traveled (S_{total}) results from impact, fall-over and slide/roll/bounce to rest (see Figure 4.15) such that

$$S_{\text{total}} = S_{\text{impact}} + S_{\text{fall-over}} + S_{\text{slide/roll/bounce}}. \quad (4.5)$$

Considering momentum conservation, restitution (e) and the impact duration (t_{impact}), the projection distance (S_{impact}) during primary contact is

$$S_{\text{impact}} = \frac{M}{M + m} [1 + e] \dot{X}(t_0-) t_{\text{impact}}. \quad (4.6)$$

Assuming no further interaction with the vehicle and that the pedestrian has remained upright with the feet in contact with the ground, then a fall-over period begins due to the combination of gravitational effects and friction acting at the feet. Representing the falling pedestrian as a rigid segment with centre of gravity height

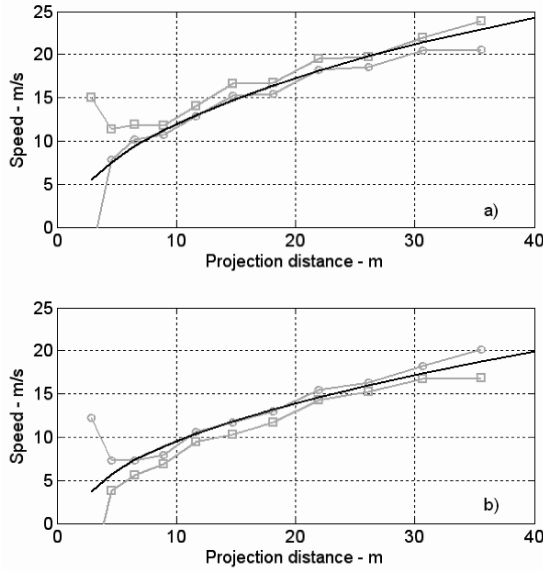


Fig. 4.14 Comparison of (a) +1 and (b) -1 standard deviation predictions (black) with the 95% confidence ranges for the real world standard deviations (grey), adapted from [37].

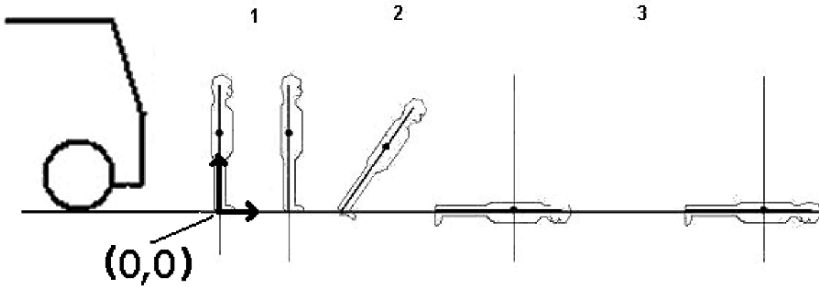


Fig. 4.15 Single segment representation of pedestrian falling over in forward projection, adapted from [38].

h and radius of gyration k , ground reaction force R and ground friction force μR , the planar equations of motion of the pedestrian segment are

$$\begin{aligned} m\ddot{x} &= -\mu R, \\ m\ddot{y} &= mg - R, \\ mk^2\ddot{\theta} &= hR \sin(\theta) + \mu Rh \cos(\theta). \end{aligned} \quad (4.7)$$

During the fall-over phase, the model assumes that the feet remain mostly in contact with the ground [30] and this constraint is expressed mathematically by

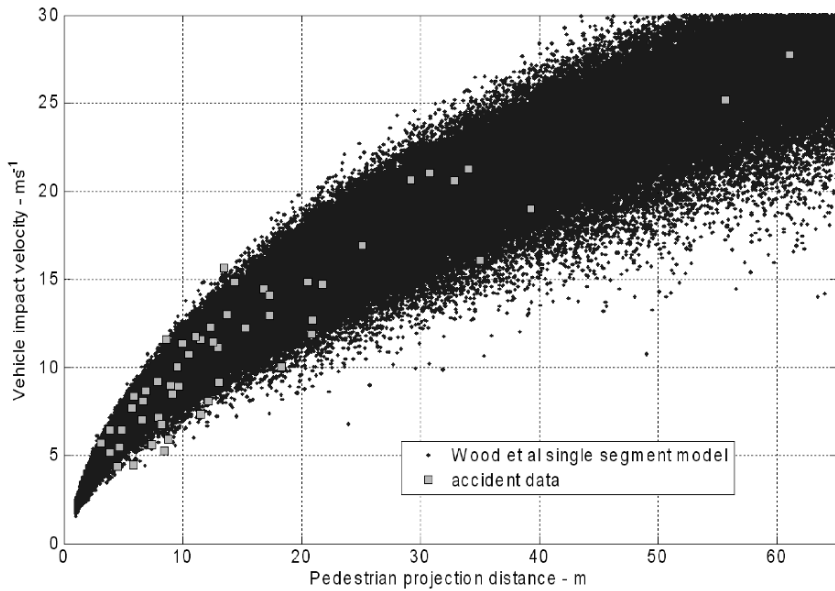


Fig. 4.16 Scatterplot predictions for single segment pedestrian forward projection model [5] compared to accident data from [2, 6, 18, 19, 24], adapted from [5].

$$\ddot{y} = h(\dot{\theta})^2 \cos(\theta) + h\ddot{\theta} \sin(\theta). \quad (4.8)$$

Although it has been seen in Chapter 3 that the inertia of the feet can cause them to wrap underneath the bumper, a similar modelling approach for motorcycle projection found that the vertical ground reaction force (R) and hence the horizontal friction force (μR) reduces rapidly with increasing fall-over angle [39]. Equation (4.8) couples the vertical and rotational motion of the pedestrian segment and these differential equations can be numerically integrated until the fall-over angle has reached ninety degrees, and the horizontal distance travelled during this phase ($S_{\text{fall-over}}$) is recorded. At the end of the fall-over phase, the segment cg strikes the ground, resulting in a loss of horizontal momentum, after which the equation of uniform acceleration is used to find the distance travelled in sliding to rest ($S_{\text{slide/roll/bounce}}$).

Using statistical distributions of vehicle and pedestrian mass, coefficient of retardation and restitution, Monte Carlo techniques were again used to produce a scatterplot of pedestrian projection distances for a range of vehicle impact speeds, see Figure 4.16 [5]. The model predictions encompass the real-world data very well. There is insufficient data to perform statistical analysis, but it can be seen that the model predicts the tendency for a minimum (nonzero) projection distance even at impact speeds close to zero, presumably due to pedestrian fall-over even for trivial impacts.

Table 4.1 Tables for prediction of vehicle collision velocity for pedestrian forward and wrap projection in cases where only the projection distance and impact type (wrap or forward projection) are known, adapted from [1].

Confidence levels for velocity prediction	Forward projection						Wrap projection			
	$(V_{col} \text{ (m/s)} = A\sqrt{S(m)} + B)$						$(V_{col} \text{ (m/s)} = C \times [S(m) - S_o]^D)$			
	Adults			Children						
	A	B	r	A	B	r	So	C	D	r
Probable lower	3.5	-1.7	0.99	3.4	-1.6	0.99	1.8	3.7	0.47	0.99
Probable upper	3.9	-1.5	0.99	3.8	-1.4	0.99	1.4	4.3	0.46	0.99
Normal lower	3.0	-1.6	0.99	2.9	-1.5	0.99	1.9	2.9	0.49	0.99
Normal upper	4.3	-1.4	0.99	4.1	-1.2	0.99	1.2	4.8	0.46	0.99
Overall lower	2.5	-1.3	0.99	2.4	-1.2	0.99	1.9	2.3	0.51	0.99
Overall upper	4.6	-1.2	0.99	4.4	-1.0	0.99	1.2	5.3	0.46	0.99

Confidence Limits for Vehicle Impact Speed Prediction

Using these two Monte Carlo based wrap and forward projection models, regression analysis was used to derive confidence limits for vehicle impact speed prediction based on pedestrian projection distances [1]. Three sets of limits were proposed: the ‘probable’ range corresponded to the 50%ile range between the lower and upper 25%ile limits. This is appropriate for general injury and civil law cases when the confidence required of the speed estimate corresponds to ‘the balance of probability’. The second range proposed was the 95%ile range from the lower to the upper 2.5%ile limit. This is the ‘Normal’ confidence limit and has application in in-depth biomechanics research. The final range proposed was the 99.8%ile range from the lower to the upper 0.1%ile limit. This is the ‘Overall’ confidence limit where the likelihood of the collision speed being less than or more than the confidence range is each 1 in 1000. The lower limit corresponds to the stringent requirement of ‘beyond reasonable doubt’ required in criminal cases. Reconstruction tables for these limits are in frequent use for accident investigation purposes, see Table 4.1.

Other Models

Han and Brach also developed a model for pedestrian impact, flight and slide to rest. Their model included the effect of road gradient and they also used a Monte Carlo approach to account for uncertainty [36]. However, they used an arbitrary function to represent pedestrian travel distance during primary impact instead of explicit modelling of the vehicle/pedestrian impact interaction. They presented mean curves based on regressions of their Monte Carlo scatterplot data, and therefore the ability of their models to predict uncertainty is unknown. Detailed multibody modelling of pedestrian projection distance by Bhalla et al. using Madymo [32] showed that projection distance was highly dependent on the input parameters to the model (pedestrian ori-

entation, anthropometry, vehicle geometry, etc.). They also concluded that there is uncertainty inherent in vehicle speed estimation using projection distance formulae, as is evident from the accident data in Figure 4.4. Moreover, despite the detail included in their models, they did not show a better correlation with accident data than the particle based [28] or regression models [40] they used for comparison. More recently, detailed reconstruction of two pedestrian accident cases [41] was used to evaluate models categorised as either particle or regression based (e.g. [42]), single segment based [43] or based on multibody modelling [41]. For the first of the two accidents considered, all methods provided a solution close to the speed estimated by in-depth investigation. In the second accident there was no evidence of vehicle braking and therefore pedestrian separation from the vehicle was uncertain and reconstruction was only possible with multibody modelling techniques using Madymo and/or PC Crash software. Good agreement with the prediction from in-depth investigation was achieved, but the authors cautioned that many of the input parameters necessary for these models have default values in commercial software that have a very significant influence on results. Therefore, although these software packages are user-friendly, significant expertise is required to avoid meaningless predictions [41]. Other researchers have shown that Madymo modelling can be used to reproduce vehicle impact speed in pedestrian accidents if sufficient information regarding the pre-impact conditions is available [44], but it is unclear whether the method used to assess the accuracy of the modelling was valid. Very recently, genetic algorithms have been used to optimise multibody pedestrian impact simulations to determine pre-impact pedestrian posture and vehicle impact speed [45]. Although this method was applied to simulated accident data, it is clear that this method has the potential to substantially reduce the uncertainty in pedestrian accident reconstruction.

Some of the collision mechanisms remain poorly understood. For example, the extent of continued vehicle pedestrian interaction following head contact in wrap projections for both pedestrians and cyclists is not known, and the effects of pedestrian/cyclist speed and stance on projection distance are also poorly understood. The latter are important as they determine the effective launch angle in wrap projection cases and could be a contributory factor to additional head injuries. Similarly, pedestrian ground contact mechanisms and the effective retardation have not been adequately investigated.

It has been seen that detailed multibody modelling can be used to analyse vehicle/pedestrian and pedestrian/ground interactions. Coupled with increases in computing power, it would therefore be appropriate to combine these models with the Monte Carlo technique. The influence of each of the many sources of uncertainty which have been discussed in this chapter could then be evaluated in manner similar to the optimisation work of [45]. Regression of the predictions from this modelling could be used to provide formulae for accident reconstruction engineers depending on the level of information available following a collision.

Concluding Remarks

Data collected from accident reconstructions and from staged tests show that there are clear relationships between vehicle impact speed and pedestrian and cyclist projection distances for both wrap and forward projection pedestrian and cyclist impacts. However, there is substantial uncertainty associated with these relationships, due to variations in vehicle shape, retardation coefficient, suspension dynamics, relative vehicle/pedestrian or cyclist mass, vehicle braking level and pre-impact orientation and motion of the pedestrian or cyclist. Staged tests remove much of the measurement error associated with accident reconstructions, but fail to capture the range of impact configurations which occur in real-world collisions.

Regression analysis using staged test or accident data can be used to estimate vehicle impact speed from a measured pedestrian or cyclist projection distance. However, biomechanical models which explicitly address the three stages of impact, flight or fall-over and slide to rest, and the uncertainty associated with each of these are shown to adequately model the pedestrian projection process in a vehicle collision. Detailed multibody modelling or even finite element models have the potential to accurately reconstruct pedestrian and cyclist accidents. However, far more input data is required for these models. In the absence of this information, statistical techniques such as Monte Carlo modelling, combined with detailed multibody or finite element representations of the impact have the best potential for providing accident reconstruction data for pedestrian and cyclist impacts.

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Chapter 5

Injury Mechanisms and Injury Criteria

Introduction

Traumatic injuries occur when the mechanical tolerance limits of body structures are exceeded. Injuries are generally believed to result from excess strain induced by direct or indirect (i.e. inertial) loading and the most important advance in impact biomechanics has been the development of injury criteria relating the probability of trauma to mechanical parameters which can be measured using instrumented crash test dummies or cadavers, or evaluated in a numerical model. Without injury criteria, the severity of trauma in a staged test or an accident reconstruction cannot be evaluated.

Injury criteria are mostly estimated from experimental tests on cadavers or animals, but tolerance thresholds are sometimes derived from human volunteer tests. There are many problems associated with the development of injury criteria including sample size, age of cadaver subjects, scaling from animal tests to humans, inherent variability in biological samples and differences between cadavers and living people. Furthermore, limitations in sensor technology and in dummy biofidelity have resulted in injury criteria that correlate injuries with structural parameters (forces and accelerations etc) rather than tissue level loading or deformations (stress or strain). These factors should be considered when assessing the validity of a proposed injury criterion.

Chapter 2 showed that injuries to the head and lower extremities dominate for pedestrians and cyclists, but injuries to the midbody regions are also significant for serious injuries and fatalities. In this chapter, injury tolerances and criteria for the head, neck, thorax, abdomen, pelvis, thighs, knees and legs in common use in all impact biomechanics research are presented. The developments leading to these criteria are well summarised in [1] and in several review papers [2–4]. These injury tolerances and criteria are a necessary foundation for the proposed vehicle safety standards (Chapter 6) and in evaluating vehicle design for pedestrian and cyclist safety (Chapter 10). This chapter assumes a basic knowledge of anatomy, for which the reader is referred to standard texts (e.g. [5]).

For each of the body regions considered, there are substantial variations in injury tolerance, not only as a function of age and sex, but also arising from the state of health and other inherent biological variability. For example, the bones of children less than 10 years old are more flexible than the bones of adults [6] and head injuries in children are generally less severe than in adults in similar freefall situations [7]. Furthermore, the mechanical properties of bone vary with age [8] and the presence of osteoporosis dramatically reduces failure load in bone [3]. Therefore injury criteria should only be applied to individual accident cases with great caution, and predicting the probability of injury rather than a simple occurrence/non-occurrence of injury is generally more realistic. Many statistical approaches have been adopted for this purpose, but the logistic regression method [9] has been shown to yield better results than the classical maximum likelihood method [10].

Head Injuries

Head injuries in pedestrians and cyclists include fractures to the bones of the face and skull, and damage to the external soft tissue and to the brain. The close anatomical proximity of these structures means they are often injured together: an impact fracturing the skull will generally also cause external soft tissue damage, and may also lead to brain injuries. However, the mechanisms leading to different kinds of head injury vary substantially. We focus on skull fracture and brain injuries as these are most important for pedestrians/cyclists.

Under direct impact the skull may deform and fracture, and skull fractures are characterised as open or closed depending on whether the dura mater (outermost meningeal layer, see Figure 5.1) is damaged. Skull fractures nearly always occur due to a direct blow to the head, and fracture may occur at the site of the impact location, but also away from it. Drop tests using un-embalmed intact human cadaver heads have shown failure loads ranging from 4.5–14.1 kN, deflections at fracture of 3.4–16.6 mm and failure energies of 14.1–68.5 J [11].

The brain is an almost incompressible gel-like material (high bulk modulus but low shear stiffness) which is surrounded by the meningeal layers which are contained in the skull, see Figure 5.1. Brain injuries can be focal or diffuse. The major mechanisms of brain injury are pressure and shear stress due to pressure gradients or relative motion of the brain with respect to the skull [2]. A direct impact of the skull results in linear pressure gradients in the brain which can cause contusions at the site of impact and/or opposite it (coup and contre-coup injuries). Using finite element modelling and live animal tests, a pressure threshold for minor brain injuries of 173 kPa has been predicted [12].

Angular acceleration of the head results in shearing of the brain tissue, which can lead to Diffuse Axonal Injuries (DAI) spread throughout the brain. These typically manifest several hours after impact but arise directly from impact and not from secondary events such as bleeding or increased intracranial pressure. This has been verified experimentally, where the spatial distribution of shear strain in a mechanical

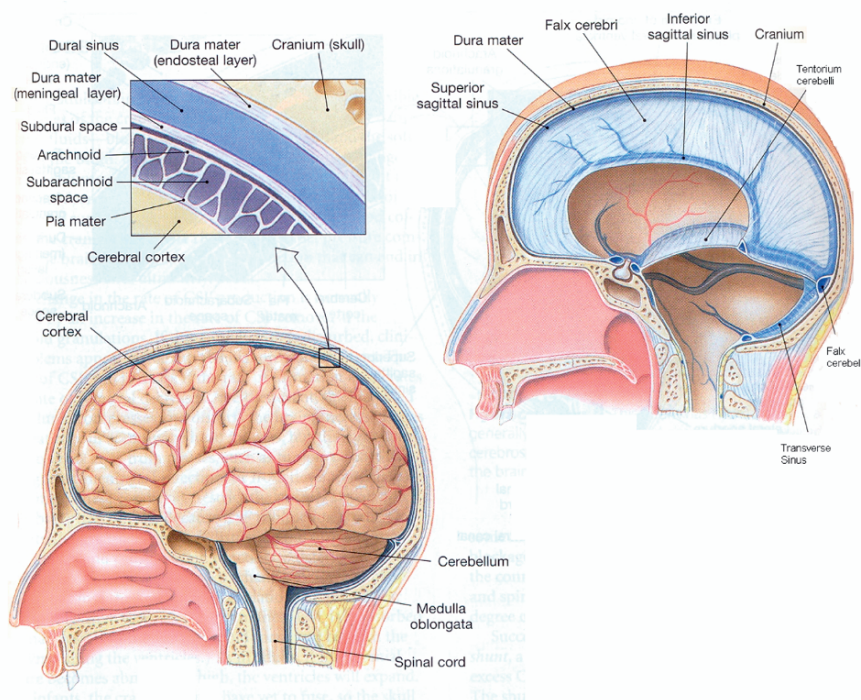


Fig. 5.1 Basic head anatomy showing brain, dura layers and skull: adapted from [5].

model subjected to angular acceleration was seen to match DAI induced in an animal model [13]. Angular acceleration also causes shear loading of the bridging veins connecting across the meningeal layers, which can lead to subdural haematoma [14].

Head Injury Criteria

The foregoing descriptions show that the term ‘head injury’ covers many potential traumatic events and no single mechanical parameter can predict the occurrence of them all. However, in spite of this, there is still only one head injury predictor which is in widespread use, and this is the Head Injury Criterion (HIC), which is based solely on the time history of the linear acceleration of the head centre of gravity. The HIC was developed from the Wayne State Tolerance Curve (WSTC) [15] which showed that the linear acceleration required for skull fracture is inversely related to impact duration. Since skull fracture is correlated with moderate concussion, the WSTC was proposed as a predictor for head injury. The data derives from (1) 1–6 ms duration linear impacts of cadaver foreheads, (2) 6–10 ms duration impacts in which cadaver and animal brain pressure responses were compared and (3) longer

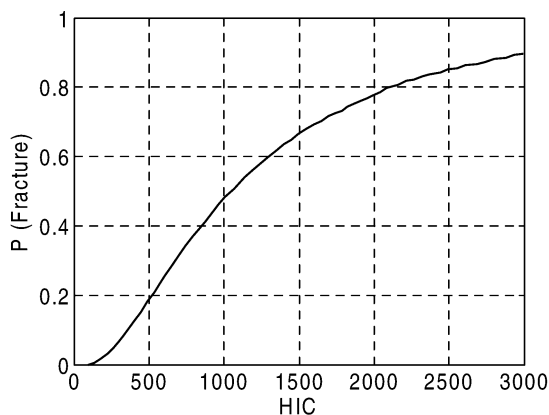


Fig. 5.2 Relationship between HIC and skull fracture probability [18], adapted from [1].

pulse duration non-injurious human volunteer tests. The HIC is iteratively calculated from the resultant head centre of gravity acceleration time history in g^1 units:

$$\text{HIC} = \max[t_2 - t_1] \left(\left[\frac{1}{[t_2 - t_1]} \right] \int_{t_1}^{t_2} [a] dt \right)^{2.5} \quad \text{where } (t_2 - t_1) \leq 36 \text{ ms.} \quad (5.1)$$

The HIC equation derives from power regression analysis of the WSTC [16, 17], and subsequent analysis [18] led to the proposal of a log-normal probability curve relating HIC score to skull fracture (see Figure 5.2), which shows that a HIC score of 1000 has an approximately 50% probability of a skull fracture. The HIC is very sensitive to high frequency noise in the head acceleration measurement, and the 3 ms criterion, also based on the WSTC, was therefore developed. The 3 ms threshold is the maximum acceleration sustained for at least 3 milliseconds, and a threshold of 80 g has been proposed for skull fracture [19].

There have been no direct demonstrations of functional brain damage to accompany the skull fractures in the Wayne State tests [1], and the HIC has been frequently criticised. However, in cadaver tests in which subdural and subarachnoid haematomas were produced, significant correlations were observed between AIS injury severity² and the corresponding HIC and 3 ms scores [20]. Similarly, from controlled experiments with cadavers compared to pedestrian head injuries, an abrupt transition from moderate to severe injury was found for HIC scores of 1100–1400, and a 50–60% probability of sustaining an injury of at least AIS3 has been predicted for a HIC of 1000 [21]. Furthermore, reconstruction of pedestrian accidents using an adult headform impactor in which the damage profile on the vehicle was matched to the accident case showed a strong correlation between HIC score and death [22].

¹ Acceleration due to gravity (9.81 ms^{-2}).

² See Chapter 2 for AIS definition

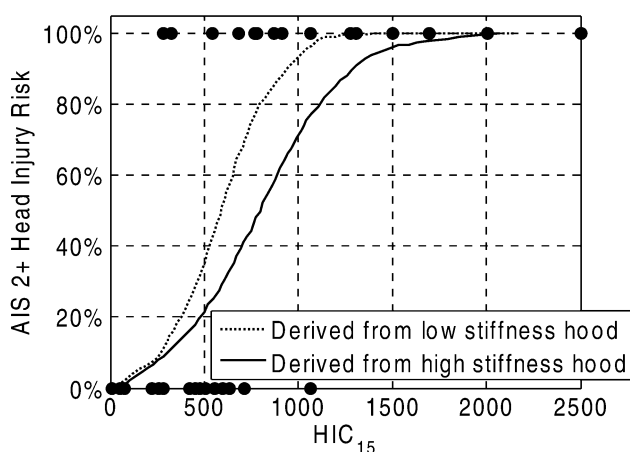


Fig. 5.3 Relationship between child pedestrian AIS2+ head injury risk and HIC score predicted from multibody model reconstructions of real accidents, adapted from [23].

Similarly, reconstruction of child pedestrian head impacts using a multibody model to predict the relationship between the HIC score and AIS2+ head injuries also yielded a strong correlation, see Figure 5.3 [23]. The figure shows that assuming a lower stiffness for the bonnet (hood) results in higher predicted injury risks for a given HIC score.

Nonetheless, a major flaw in the HIC is that it only accounts for linear acceleration. This is appropriate for skull fracture predictions, but ignoring the influence of rotational acceleration makes the HIC theoretically unsuitable for shear strain induced brain injury prediction. In recognition of this, Newman [24] combined both translational and rotational head acceleration in a predictor called the Generalised Acceleration Model for Brain Injury Threshold (GAMBIT) and in the Maximum Power Index model (HIP) [25]. The HIP correlated at least as well as the HIC with skull fractures in reconstructions of accidents involving professional American footballers, motorcyclists and pedestrians [26]. Other researchers have proposed brain tolerance thresholds to rotational velocity and/or acceleration: Figure 5.4 shows results of animal and physical modelling linking brain shear strain to head angular velocity and angular acceleration [27]. Further work with an animal model showed that 25% elongation in giant squid axons resulted in structural failure [28]. However so far there is no generally accepted criterion for brain injury induced by rotational head motion.

More recently the focus has been on using finite element models to reproduce brain injuries observed in animal models and in accident reconstructions. This approach has facilitated evaluation of stress and strain based injury hypotheses. For example, Takhounts et al. [29] have developed a simulated injury monitor (SIMon) and

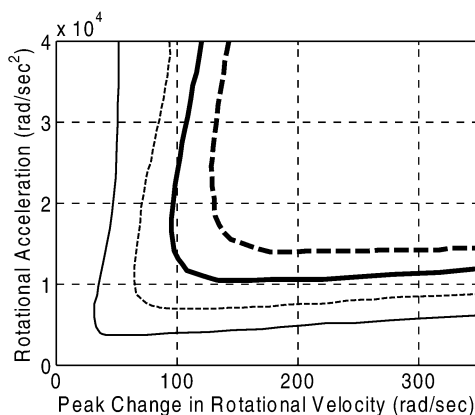


Fig. 5.4 Relationship between brain shear strain and changes in head angular acceleration and angular velocity. Critical strains: 0.05 (solid line), 0.1 (dashed line), 0.15 (heavy solid line) and 0.2 (heavy dashed line), adapted from [27].

proposed a Cumulative Strain Damage Measure³ (CSDM) to predict diffuse axonal injuries, a dilatation damage measure to predict contusions and a relative motion damage measure to predict subdural haematoma. Similarly, brain shear stresses and strain rates predicted using finite element modelling have agreed with the location and severity of axonal injuries described in medical reports [30].

For years validation of finite element head models for impact was limited as the only experimental data available was intracranial pressure from cadaver impact tests, and there is no unique relation between pressure and shear stress in transient loading of the brain [31]. However, more recent tests have recorded brain deformation during impact using high-speed biplanar X-ray imaging of cadaver heads in which neutral density targets (NDTs) were implanted in the brain. Relative skull/brain movements of up to 5 mm were recorded [32] and the trajectories of these NDTs have since been used for finite element model validations, see Chapter 8. Suitably validated finite element models can provide a means to predict pressure and strain distributions in the brain which can be compared with scaled animal tolerance values or with real-world injuries following an accident reconstruction simulation. For example, reconstruction of well-documented American football head trauma cases in which the rigid body skull motion was defined as input to finite element simulations led to the prediction of a 50% risk of moderate and severe neurological lesions for peak Von Mises brain stresses of 18 and 38 kPa respectively, and a local strain energy of 2.2 J for a 50th percent risk of skull fracture [33]. This led to an injury criterion based on peak brain von-Mises stress occurring in a volume of at least 10 contiguous finite elements in the head model (about 3 cm³ of

³ A critical CSDM level of 15% tensile strain was proposed following scaling of animal injury data [29].

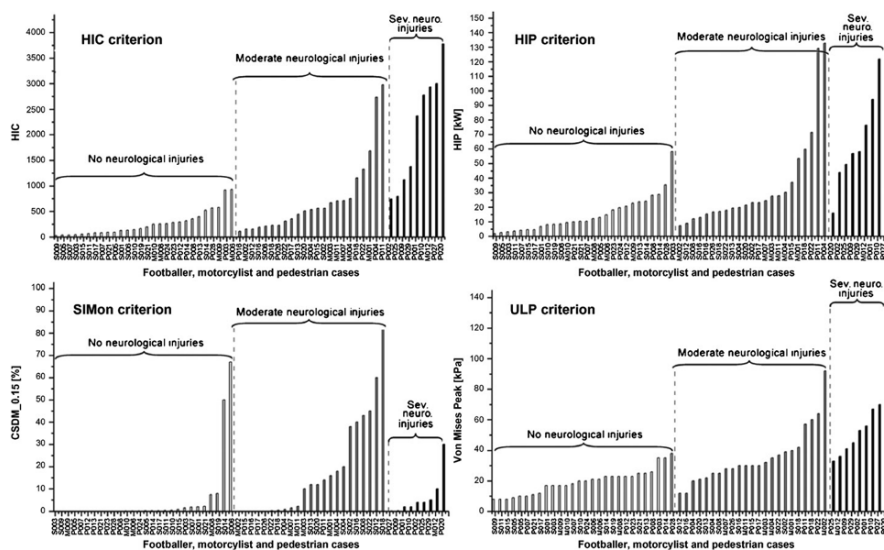


Fig. 5.5 Evaluation of HIC, HIP and ULP and SIMon head injury criteria using finite element model reconstruction of American football, motorcycle and pedestrian accidents, adapted from [26].

brain volume), known as the ULP head injury criterion.⁴ Very recently, the same researchers used their finite element model to reconstruct 61 accident cases consisting of 22 football, 12 motorcycle and 27 pedestrian accidents and compare the capability of the HIC, HIP, SIMon and the ULP head injury criteria [26] of predicting moderate and severe neurological injuries, as well as subdural haematoma and skull fractures. Results showed a good correlation of the HIC, HIP and ULP criteria with both moderate and severe neurological injuries, but poorer predictions for the SIMon CSDM model, see Figure 5.5. These results were then used in a logistic regression to derive a relationship between brain Von Mises stress and injury probability, see Figure 5.6. Given the variability of the accident data, all of the assessed injury criteria yielded good predictions of skull fracture, but a poorer ability to predict subdural haematoma. Despite its simplicity, it is striking that the HIC provided a good prediction of both skull fracture and brain injuries in the 61 accident cases considered.

The advantage of this accident reconstruction approach is that injury predictions are not deduced from cadavers or animal experiments, but from real accidents, about half of which were pedestrian accidents. However, a variation of up to 20% for the head impact velocity was estimated for the pedestrian accident reconstruction cases and, due to the strong dependence of impact severity on impact speed (see Chapter 7), there are limitations to this approach.

⁴ ULP – University of Louis Pasteur, Strasbourg.

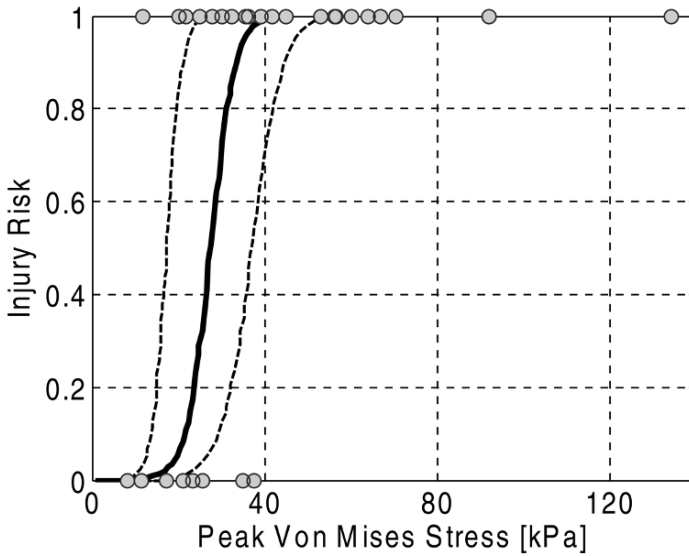


Fig. 5.6 Probability of moderate neurological injuries as a function of peak Von Mises stress [34] derived from accident reconstructions predicted by the ULP finite element head model, adapted from [26].

Finite element modelling has also been used to assess the influence of other variables on injury outcome. For example, intracranial stress has been found to depend on the geometry of the head, so that lower HIC thresholds should be used when analysing data from dummies larger than the 50th percentile male [35]. This is in broad agreement with the finding that children are less likely to suffer brain injuries for similar severity impacts [7]. The influence of impact direction also had a substantial effect on the predicted intracranial response [36]. When evaluating the global kinematic injury measures for rotational pulses, the change in angular velocity corresponded best with the intracranial strains found in the finite element model. In contrast, for translational impulses the HIC and HIP showed the best correlation for the predicted strain levels [36].

Spinal Injuries

The spine consists of lumbar, thoracic and cervical sections (Figure 5.7), but due to the slender, flexible and exposed nature of the cervical spine which comprises the neck, this region is the most vulnerable in pedestrian and cyclist impacts. Injuries to the neck often result from direct impact to the head, but also occur via inertial loading from the head. However, the latter are whiplash-type injuries which are not common in pedestrian/cyclist impact. When pedestrians and cyclists are struck

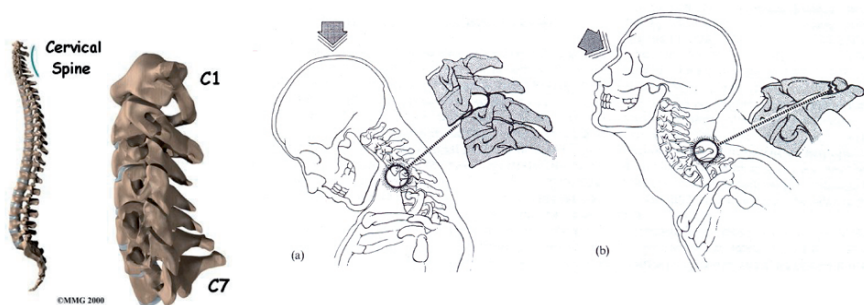


Fig. 5.7 (a) The whole spine [37] and (b) cervical spine injuries from head impact, adapted from [38].

by vehicles, compressive loading of the neck occurs frequently due to head contact on the bonnet/windscreen, and this may occur in a compression-flexion or a compression-extension mode, see Figure 5.7.

In attempts to measure threshold forces and moments for neck injuries, Mertz et al. [39] used the Hybrid III 50th percentile male dummy to simulate cases where serious neck injuries were recorded in American football players. They reported a peak compression tolerance of 4000N. Nyquist et al. [40] also used the Hybrid III 50th percentile male dummy to reconstruct collisions and reported tension and shear limits of 3300 N and 3000 N respectively. Mertz and Patrick [41] published tolerance limits for flexion and extension bending moments from volunteers and cadaver subjects. Ligamentous damage was found in a small female cadaver subject at an extension moment of 48 Nm, and this was scaled to 57 Nm as a threshold for a 50th percentile male. No injuries were produced during flexion testing. Yoganandan et al. [42] tested cadaveric specimens in axial tension under quasistatic and dynamic loading conditions and reported mean specimen failure loads of 1555 and 3373 N for isolated and intact specimens respectively. Following the finding that neck tensile failure loads are reduced when extension bending moments are superimposed [43, 44], the N_{ij} neck injury criterion [45, 46] was proposed:

$$N_{ij} = \frac{F_z}{F_{\text{threshold}}} + \frac{M_y}{M_{\text{threshold}}} \quad (5.2)$$

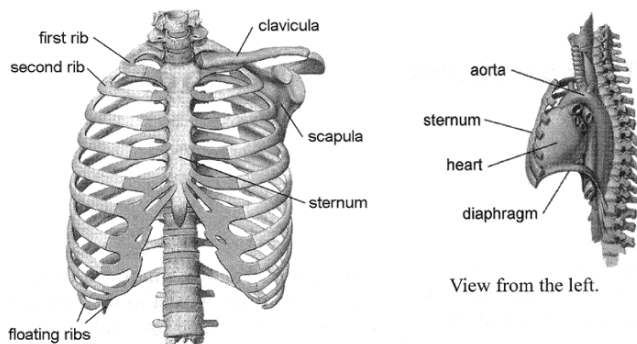
and Table 5.1 shows the current N_{ij} thresholds for the different loading modes adopted in FMVSS 208⁵ [47] for evaluation using the Hybrid III dummy family. The standard stipulates that the N_{ij} scores must not exceed 1.0 in any of the four loading mechanisms, i.e. compression/extension, compression/flexion, tension/extension or tension/flexion.

There is far less data available on the tolerance of the thoracic and lumbar spine regions to injury, and most of the work available has focused on functional motion

⁵ FMVSS208 United States Federal Motor Vehicle Safety Standard: Occupant Crash Protection.

Table 5.1 N_{ij} thresholds values currently in use in FMVSS 208 [47].

Dummy	Axial load (N)	Flexion (Nm)	Extension (Nm)
50th %ile male	6806 tension 6160 compression	310	135
5th %ile female	4287 tension 3880 compression	155	67
3 year old child	2120	68	27
6 year old child	2800	93	37

**Fig. 5.8** The thorax, adapted from [1].

units (ie, a vertebral pair with intra-vertebral disc and ligaments intact) to provide input data for numerical models. However, anterior shear tests to failure on the lumbar spine have shown static and dynamic soft tissue failures of 1290 and 1770 N respectively [48].

Thorax Injuries

The thorax consists of the rib cage with internal organs (Figure 5.8). Lateral injuries to the thorax are most common in vehicle pedestrian/cyclist collisions, resulting from compression of the rib cage and sternum, and viscous or inertial loading of the soft tissues of the internal organs.

The Thoracic Trauma Index (TTI) was proposed to assess blunt trauma injuries to the chest [49]. The TTI is based on acceleration, but also accounts for age and mass of the subject:

$$TTI = 1.4 \times \text{age} + \frac{1}{2}(\text{RIB}_{\text{acc}} + T12_{\text{acc}}) \frac{M}{M_{\text{ref}}}, \quad (5.3)$$

where RIB_{acc} is the maximum acceleration in g of ribs 4 and 8 on the struck side, $T12_{\text{acc}}$ is the maximal lateral acceleration in g of the lower thoracic spine, ‘age’

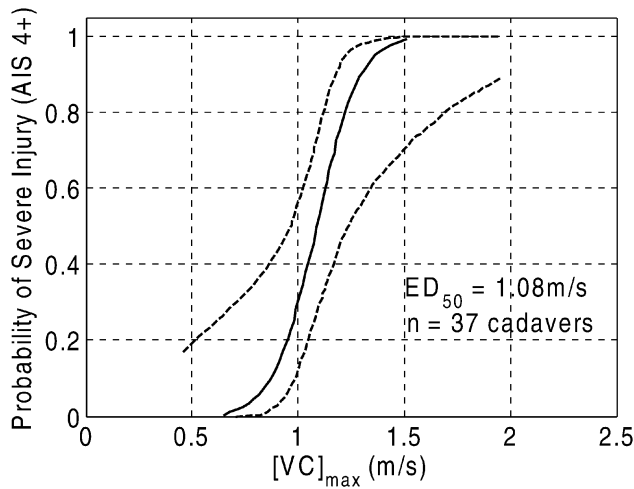


Fig. 5.9 VC correlation with injury, adapted from [52].

and M are the subject's age in years and mass in kg respectively, and M_{ref} is the reference mass (75 kg for a 50th percentile human male). TTI threshold values of 85g and 60g have been proposed for adults and children respectively [50].

The Viscous Criterion (VC) proposed by Lau and Viano [51] is predicated on the finding that the product of normalised compression and compression rate is correlated with thorax injuries:

$$\text{VC} = \frac{dD(t)}{dt} \times \frac{D(t)}{D(0)}, \quad (5.4)$$

where $D(t)$ is the time-varying torso compression and $D(0)$ is the initial torso thickness. Statistical analysis of thorax impacts to cadavers has shown that the maximum VC score is highly correlated with the risk of severe injury, see Figure 5.9. A maximum chest VC tolerance for adults of 1 m/s has been recommended for a 25% probability of serious injury [52]. A summary of thorax tolerance thresholds is given in Table 5.2.

For children, reconstruction of child pedestrian accidents using a child thorax dummy model validated against scaled cadaver data yielded the following threshold for a 20% fatality probability: 25% chest deflection, a VC score of 0.376 m/s and average rib/spine acceleration of 60 g [56]. However, considerable scatter was reported in the accident reconstruction data.

Table 5.2 Thorax injury thresholds to lateral impact: adapted from [1].

Tolerance level	Injury level	Reference
Force		
7.4 kN	AIS0	[53]
10.2 kN	AIS3	[53]
5.5 kN	25% probability of AIS ≥ 4	[54]
Acceleration		
T8-Y 45.2 g	25% probability of AIS ≥ 4	[54]
T12-Y 31.6 g	25% probability of AIS ≥ 4	[54]
60 g	25% probability of AIS ≥ 4	[55]
Compression to whole thorax 38.4%	25% probability of AIS ≥ 4	[54]
VC _{max} to whole thorax		
1.0 m/s	50% probability of AIS ≥ 3	[54]
1.47 m/s	25% probability of AIS ≥ 4	[54]

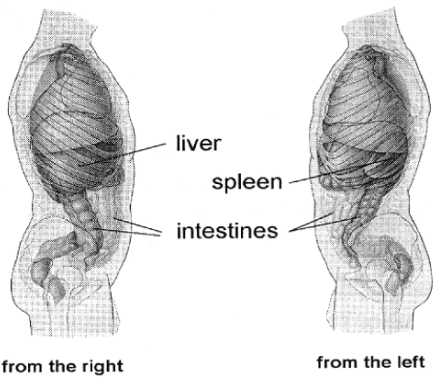


Fig. 5.10 The abdomen, adapted from [1].

Abdominal Injuries

The abdomen lies between the thorax and the pelvis and contains the mostly hollow intestines and the more solid liver, spleen, and pancreas (Figure 5.10). Abdomen impact can be life-threatening, though the presence of the floating ribs provides some protection for the upper abdomen in side impacts. Acceleration based injury criteria for the abdomen are unreliable because of its inherent deformability. Human cadaver studies indicate a maximum abdomen force tolerance of 4.4 kN [57].

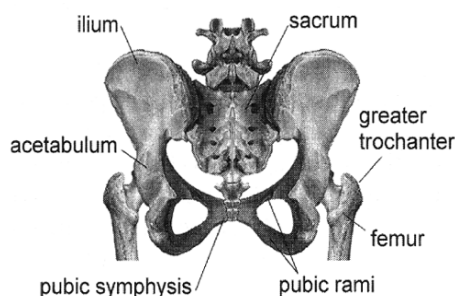


Fig. 5.11 The pelvis, adapted from [1].

Pelvis Injuries

The pelvis is a bony girdle which connects the spine to the lower extremities (Figure 5.11). The digestive organs are located within the pelvic ring and nerves and large blood vessels pass through it. The pelvis serves as an attachment point for muscles and pelvic fracture is debilitating and often associated with substantial bleeding and sensory and motor dysfunction.

Lateral pelvis impact is common when pedestrians and cyclists are struck by car fronts, especially if the bonnet leading edge height is similar to the pelvis height above the ground, see Chapter 10. However, pelvic fractures are a substantial factor in pedestrian morbidity and mortality [58], because the severity of these injuries is associated with a high degree of destructive energy transmitted to the body as a whole, as manifested by the number and severity of associated injuries [58]. Early analysis of pedestrian accidents showed that no pelvic fractures occurred at impact speeds below 20 km/h [59]. A study of more serious pedestrian accidents in the UK found that the pelvis appears to be more susceptible to damage by direct contact when it is loaded laterally, causing fracture of the iliac crest or the hip joint socket [60].

Large variations for pelvis fracture tolerance are reported between males and females: 4 kN for 5th percentile females and 10kN for 50th percentile males [61]. The US standard FMVSS 214 stipulates a peak acceleration tolerance of 130 g for the pelvis in a side impact. Furthermore, fracture tolerances vary depending on load path: for a single load path through the greater trochanter tolerances can be as much as halved [62] compared to a load path including the iliac crest. In side impact tests on cadavers the reported peak acceleration for pelvic fracture was 73 g [63].

Lower Extremity Injuries

Lower extremity injuries are frequent when pedestrians and cyclists are struck by vehicles, because this is usually the first body region struck, and a large number

of different injuries can occur [64]. From the bumper impact, the leg soft tissue and long bones are exposed to forces at the impact area, which can result in soft tissue injury and bone fracture. The knee joint is subjected to a shear force and bending moment, resulting in compressive loads on the nearest tibia and femoral condyles, and tensile forces in the knee ligaments and the joint capsule. During bumper contact the inertia of the foot causes the leg to rotate about its longitudinal axis, resulting in shearing of the soft tissue and torque is also transmitted to the knee [65]. These lower extremity injuries are rarely fatal, but although long bone injuries usually heal satisfactorily, ligament injuries in the knee may not [66], and cadaver tests have shown that direct impact at the joints produce more severe injuries than long bone impacts [67]. Furthermore, the extent of soft tissue injury associated with lower extremity bone fractures has a decisive influence on fracture healing and functional restoration. With closed fractures, the most long-term impairments result from primary soft tissue damage [65, 68]. Vascular injuries associated with leg fracture are also important, as 60% of all open tibia fractures involving vascular injuries ultimately resulted in amputation [69].

Development of injury criteria for lower extremity structures has been the focus of considerable research, but the influence of damage to one body structure on loading of adjacent structures makes this a complex task. For example, high-speed films of cadaver leg impacts have shown that long bone fractures occurred in the first few milliseconds after impact, and no soft tissue injuries of the knee joint were found in these cases [65]. The most frequent lower extremity injuries for pedestrians and cyclists are fractures of the tibia and femur (and less often, also the fibula) and soft tissue damage in the knee, especially the medial collateral ligament in side impact cases, see Chapter 2. Joint injury generally involves disruption to the ligaments and damage to the menisci and cartilage, but can also involve bony fractures in the region of the joint [3].

The Long Bones: Femur, Tibia and Fibula

Cortical bone is 45% stronger in tension than in compression [70] and the long bones of the lower extremities therefore usually fail from compressive loading due to inertial bending in impact. Although failure tests on long bones have been performed since the 19th century [71], much of the recent work on lower extremity injuries was done at the University of Virginia (UVA). Kerrigan et al. [72] reviewed failure bending moments for the isolated femur and the tibia bones, as well as adding results from their own work, see Tables 5.3 and 5.4. They found average dynamic failure moments for the femur and tibia of 412 and 310 Nm respectively. Tables 5.3 and 5.4 show the age and sex dependencies of fracture strength for the femur and tibia.

Axial compression due to self weight reduces the applied bending moment required for failure of the leg and the dynamic bending strength is 69% higher than for quasi static loading, with fractures showing a higher degree of comminution.

Table 5.3 Femur bending strength for males (M) and females (F), adapted from [72].

Study	Sample size	Failure moment	Notes
Weber 1859 in Messerer 1880	9 (4 M, 5 F)	233 Nm (M) 182 (F)	Quasi-static
Messerer 1880	12 (6 M, 6 F) Age 20–82	310 Nm (M) 180 Nm (F)	Quasi-static
Mather 1968	145 (91 M, 54 F)	318 Nm (M) 202 Nm (F)	
Motoshima 1960 in Yamada 1970	35 total (M + F)	234 Nm (20–39 yrs) 213 Nm (40–49 yrs) 203 Nm (50–59 yrs) 201 Nm (60–69 yrs) 184 Nm (70–89 yrs)	Quasi-static Females 5/6 as strong as males Bending strength isotropic
Martens et al. (1986)	33 (26 M, 7 F)	373 Nm (mid shaft fx)	Dynamic (<200 ms) 4 pt tests
Kerrigan et al. (2003)	8	Mean 412 Nm, SD 114 Nm	Dynamic

Table 5.4 Tibia bending strength for males (M) and females (F), adapted from [72].

Study	Sample size	Failure moment	Notes
Weber 1859 in Messerer 1880	9 (4 M, 5 F)	165 Nm (M) 125 (F)	Quasi-static
Messerer 1880	12 (6 M, 6 F) Age 20–82	207 Nm (M) 124 Nm (F)	Quasi-static
Motoshima 1960 In Yamada 1970	35 total (M + F)	108 Nm (20–39 yrs) 180 Nm (40–49 yrs) 175 Nm (50–59 yrs) 172 Nm (60–69 yrs) 164 Nm (70–89 yrs) 182 Nm (mean)	Quasi-static Females 5/6 as strong as males Bending strength isotropic
Schreiber et al. 1997	10 (ave age 51) 12 (ave age 68) 9 (ave age 72)	241 Nm (SD 49) 408 Nm (SD 115) 311 Nm (Sd 96)	Tested whole leg, not just tibia Quasi-static (2 M, 8 F) Dynamic (6 M, 6 F) Dynamic with axial load (4M, 5F)
Kerrigan et al. 2003	9	Mean 310 Nm, SD 60 Nm	Dynamic

However, in dynamic tests the peak strain was circa 60% lower and the energy to fracture was therefore similar for static and dynamic cases [73]. The Schreiber data in table 5 results from testing whole legs, which shows that the fibula makes the leg substantially stronger in bending.

The UVA research group subsequently performed additional impact tests on isolated leg and thigh specimens, and used Weibull survival models to generate fracture

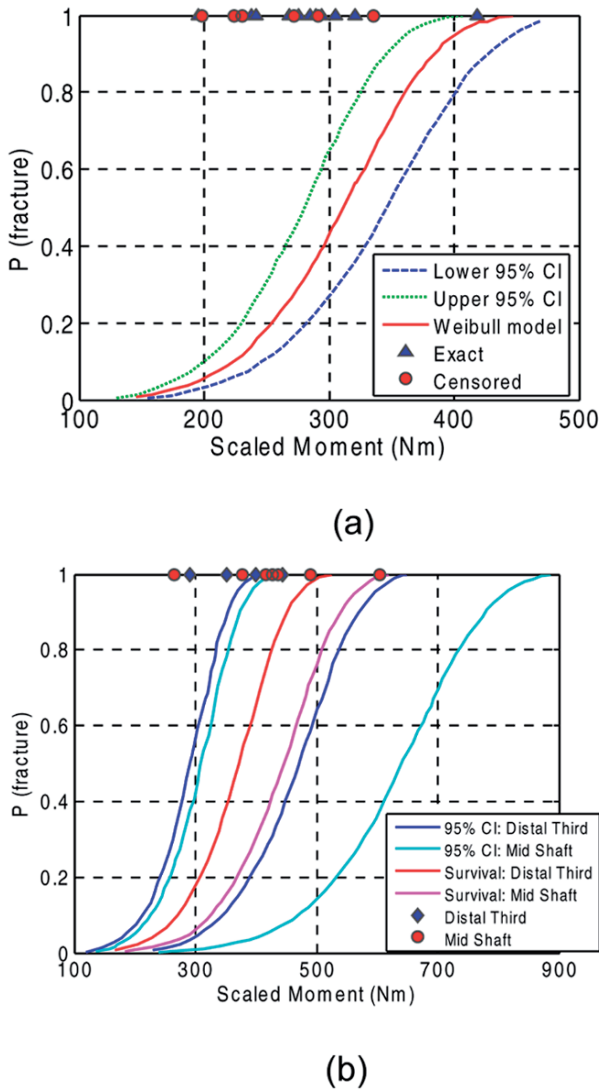


Fig. 5.12 Fracture probability as a function of bending moment in latero-medial bending of (a) the leg and (b) the thigh: raw and censored data and Weibull survival model with 95% confidence limits, adapted from [74].

probability curves as a function of bending moment in the leg and thigh, see Figure 5.12 [74].

In an attempt to replicate pedestrian impact loading conditions more faithfully, 13 full cadaver impact tests were performed [75], see Figure 5.13a, and these were

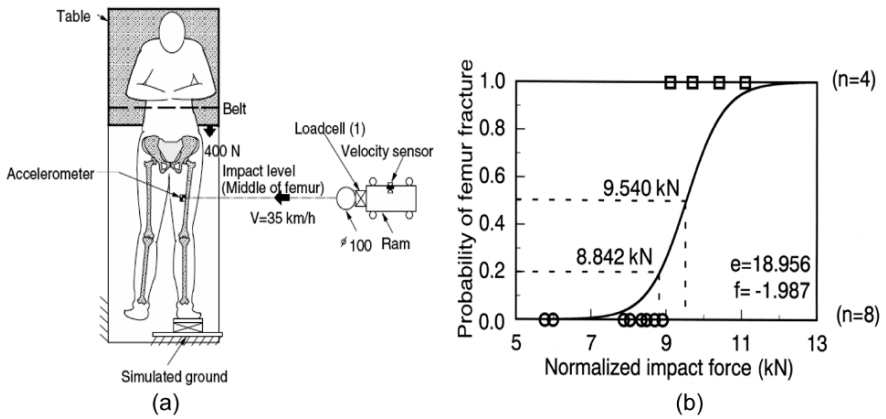


Fig. 5.13 (a) Femur impact tests performed on cadavers, (b) resulting fracture probability curve is a function of impact force, adapted from [75].

Table 5.5 TI threshold values from [73].

	Small female	Mid-size male	Large male
M_{crit}	115 Nm	225 Nm	307 Nm
F_{crit}	22.9 kN	35.9 kN	44.2 kN

used to generate a probability of fracture risk for the femur as a function of impact force, see Figure 5.13b.

The Tibia Compression Force Criterion (TCFC) adopted in UNECE R94⁶ limits the maximal axial load in the leg to 8 kN, but Schreiber et al. [73] found compressive failures in whole leg specimens at 6.2 and 7.4 kN. The corresponding Femur Force Criterion (FFC) limits the axial load in the femur to 7.6 kN for loads sustained for longer than 10 ms. The Tibia Index (TI) [76] combines the leg axial load and the bending moment in a manner similar to the N_{ij} criterion:

$$TI = \frac{M}{M_{crit}} + \frac{F}{F_{crit}} \tag{5.5}$$

where the critical bending moment (M_{crit}) and axial load (F_{crit}) are given in Table 5.5. A TI of 1 or greater exceeds the proposed injury threshold.

⁶ UNECE R94 standard: Uniform provisions concerning the approval of vehicles with regard to the protection of the occupants in the event of a frontal collision.

Table 5.6 Failure levels in knee bending and shear tests from [72].

Impactor displacement (mm)	Bending moment (Nm)	Shear force (N)
11.3	142	—
10.3	142	—
8.3	130	—
39.6	—	1667
15.7	—	563

Table 5.7 Cadaver knee injuries in lateral bending and shear loading, adapted from [72].

Specimen	Knee Joint Injuries							
	Ligaments				Meniscus		Bones	
	ACL	PCL	MCL	LCL	Med.	Lat.	Femur	Tibia
124R-Bend		■ ₂	■ _{1,3}				● ₁	
167L-Bend	■ ₁		■ ₃			▲ ₁		
135L-Bend			■ ₁					
167R-Shear	■ ₂						● ₂	
169R-Shear	■ ₄				▲ ₂	▲ ₁	● ₂	
Ligament Injuries: ■ ₁ Partial Tear, ■ ₂ Partial Avulsion from Tibial Surface, ■ ₃ Complete femoral avulsion, ■ ₄ Complete tibial avulsion Meniscus Injuries: ▲ ₁ Tear, ▲ ₂ Avulsion of anterior and posterior attachments Epiphysis Fractures: ● ₁ Fracture of condyle, ● ₂ Osteochondral fracture								

The Knee

The complexity of the knee joint means that failure loads of individual structures are difficult to measure, and lumped shear force and bending moment tolerance of the knee joint are often reported instead, since these are the predominant knee loading mechanisms for laterally struck pedestrians and cyclists. Kerrigan et al. [72] tested three knee joints to failure in pure valgus bending and two in lateral shear, see Table 5.6 [72]. Table 5.7 shows that, in the five cases tested, the medial cruciate ligament is injured most in lateral bending, whereas the anterior cruciate ligament and the femur and lateral meniscus are at high risk in both bending and shear. The mean failure bending moment in these tests was 137 Nm, much lower than 284 Nm reported by Kajzer et al. [77] for ligament avulsion. However, Kerrigan et al. [72] cast doubt on the experimental procedure for the Kajzer et al tests, which indicate the unlikely outcome that the bending strength of the femur and tibia/fibula complex is less than for lateral bending in the knee. In real accidents, pure shear or bending are unlikely, and a combination of both occurs.

The development of detailed finite element models of the knee (e.g. [78]) has allowed prediction of mechanical loading of specific structures such as the medial collateral ligament. To evaluate the likelihood of injuries, tolerance levels of the medial collateral ligament for 50th percentile males as a function of valgus knee bending moment using univariate Weibull survival models have been proposed [79]:

$$\begin{aligned}
 P(\text{injury})_{\text{first moment peak}} &= 1 - e^{4.338 \ln M - 20.4589}, \\
 P(\text{injury})_{\text{max moment}} &= 1 - e^{5.772 \ln M - 28.624}.
 \end{aligned}
 \tag{5.6}$$

Finally, a combination of cadaver impact tests and finite element modelling was used to derive a tolerance to lateral bending in the knee joint of 16° rotation and 13 mm shearing displacement, above which ligament injury is predicted [80].

The Ankle and Upper Extremities

The ankle joint connects the foot to the leg and, due to its position, direct pedestrian and cyclist ankle injuries are relatively infrequent. However, in one report ankle joint injuries were reported in 36% of fatal pedestrian accidents [81]. The most likely cause of ankle injuries is from indirect loading sustained when force is applied laterally or medially to the ankle by the tibia or fibula causing the joint to be severely adducted or abducted [60]. The arms and shoulders of pedestrian and cyclists are frequently injured in wrap projection cases when the upper body strikes the bonnet or windscreen, see Chapter 2. However, these are generally much less severe than the attendant head injuries and injury criteria for these body regions are not presented here.

Concluding Remarks

This chapter has presented the established injury criteria for the body regions most frequently injured in vehicle pedestrian and cyclist accidents. There is great variability in injury tolerance among different people, and predicting probabilities of injury is a more meaningful approach than simply predicting the occurrence or non-occurrence of injury based on mechanical parameters. In future, the results of more detailed biomechanical testing will help in the formulation of injury criteria which are more specific to age and sex and the state of health of a person, and which correlate injury to tissue level strains. In the meantime, the available models for injury prediction are crude, and prediction of injuries for individual cases cannot be done with any confidence. More reliable injury criteria are therefore required for future evaluation of injury reduction measures for pedestrians and cyclists.

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Chapter 6

Vehicle Design Standards for Pedestrian and Cyclist Safety

Introduction

By 1950, 60% of American households owned a car, yet vehicle design for occupant protection was poorly developed, and remained unregulated in the US until 1968¹ [1]. Regulation resulted in dramatic improvements in occupant safety [2]. Similarly, although researchers have had a basic understanding of the relationship between vehicle design and pedestrian injuries since the 1960s [3], the safety of vulnerable road users such as pedestrians and cyclists was not a serious consideration in vehicle design until the 1980s. This has been fuelled by the belief that little could be done to protect pedestrians in the event of a vehicle impact [4, 5] but also by vehicle manufacturers' reluctance to develop an area not governed by legislation and not considered to provide sufficient added value to the vehicle. Safety research for pedestrians and cyclists was therefore slow and poorly funded in comparison to vehicle occupants and scientific consensus on the requirements of vehicle design for pedestrian protection has still not been achieved. In consequence, existing standards are subject to updates and legal implementation is evolving and remains limited. However, there is now substantial public appetite in many countries for the regulation of vehicle design for pedestrian safety, as evidenced by the introduction of pedestrian safety testing by consumer driven safety organisations such as the New Car Assessment Programs [6] operational in Europe, Japan and Australia.²

This chapter provides a summary of the development of crash safety standards aimed at improving passenger vehicle front-end design for pedestrian and cyclist impacts. The principal focus has specifically been on improving pedestrian safety when struck by passenger vehicles but, following early research into cyclist impact mechanics [7], it has been generally assumed that standards governing vehicle design for pedestrian safety will also improve safety for cyclists [8–10]. However,

¹ FMVSS 208 introduced in 1968 by NHTSA (frontal impact legislation).

² For example, the EuroNCAP founded in 1997 publishes safety reports on new cars, and awards 'star ratings' based on vehicle performance in a variety of crash tests including impacts with pedestrians (<http://www.euroncap.com>).

it has been shown that the impact velocity and impact angle for cyclist's heads on the bonnet are lower than for pedestrians when struck by the fronts of passenger vehicles and test conditions should reflect these differences [11]. Similarly, accident data show that test methods for the fronts of buses, coaches and goods vehicles would be beneficial [12], especially in less motorised countries.

A significant difference in head injury risk between pedestrians and cyclists may exist if the cyclist is wearing a helmet, and a number of safety standards for cycling helmets are aimed at maintaining head linear acceleration at sublethal levels in drop tests (e.g. [13–15]). However, these have primarily evolved to protect cyclist's heads in the event of a fall to the ground: for example, the Snell Foundation test involves a drop height of 2.2 m, which is roughly 25% greater than an adult head height above the ground. The impact velocity for this test is 6.6 m/s (24 km/h), which is significantly lower than the mean reported vehicle/cyclist impact speed of ca. 35 km/h, see Chapter 2. The effectiveness of cycling helmets in protecting against head injuries resulting from vehicle impacts is the subject of significant debate (e.g. [16–19]). Nonetheless, bicycle helmets which pass the Snell test [15] have a significant potential to reduce the severity of cyclist head injuries in the event of a collision with a vehicle.

Bodies Developing Pedestrian Safety Standards

The main bodies which have developed pedestrian safety standards are Working Groups of the European Enhanced Vehicle Safety Committee (EEVC), the International Organization for Standardisation (ISO) and the International Harmonised Research Activities/United Nations Economic Commission for Europe (UNECE).³ The EEVC and ISO Pedestrian Safety Working Groups were created in 1987 to determine test methods and acceptance levels for assessing protection afforded to pedestrians by the fronts of cars in an accident. Recognising that the ISO and EEVC tests were formulated based on accident data collected in the 1970s and 1980s, the IHRA program was developed to analyze more recent pedestrian accidents and develop harmonised test procedures to reflect the current accident environment in the member countries. The IHRA accident data from Australia, Europe, Japan and the USA showed that the head and leg each account for almost one third of pedestrian injuries, with the bonnet top and windscreen glass being the leading causes of head injury for children and adults respectively, see Chapter 2. Therefore test procedures for child and adult head impact and the adult leg in the vicinity of the knee were developed by the IHRA [20]. The IHRA has now largely superseded the work of the ISO and, given the global nature of the production vehicle market, the IHRA has been working on a draft Global Technical Regulation (GTR) for pedestrian protec-

³ The United States National Highway Traffic Safety Administration (NHTSA) was independently active in this field until the 1990s, and now channels its input through the IHRA.

tion since 2002 [8]. This draft is largely based on the work of EEVC, ISO, IHRA and the Japanese Ministry for Land, Infrastructure and Transport [21].

Types of Test Proposed

There is substantial common membership of the EEVC, ISO and the IHRA pedestrian safety groupings and the types of tests proposed by these bodies are all very similar. All pedestrian safety standards are aimed at protecting pedestrians struck from the side by a vehicle at speeds up to 40 km/h, since this was judged to be the upper limit at which pedestrian protection could reasonably be provided and this speed includes over 80% of pedestrian accident cases [22–25], see Chapter 2. In all cases pedestrian subsystem impactor tests were developed in favour of pedestrian dummy tests, see Figure 6.1. This is quite different to the full scale dummy tests stipulated in frontal impact vehicle occupant protection standards (e.g. FMVSS 208 [26]) and several arguments have been proposed to justify this, mostly relating to test repeatability and pedestrian dummy biofidelity. Due to the substantially greater whole body motion of struck pedestrians compared to vehicle occupants, the validation requirements of a pedestrian dummy (even at the limited level of meeting kinematic corridors defined by cadaver tests) are difficult to achieve. Early pedestrian dummies such as the Hybrid II and the Rotationally Symmetric Pedestrian Dummy had poor biofidelity, see Chapter 8 [27–30]. Furthermore, since the typical pedestrian impact involves a chain of contacts of which the head is usually the last but most important, the use of a whole-body crash dummy to assess head impacts can result in substantial variability resulting from nominally identical test conditions. This is clearly not desirable for legislative tests. Finally, due to the range of human anthropometrics, a family of representative pedestrian crash dummies would be required and this was considered to be prohibitively expensive by the EEVC [31]. However, the choice of leg and headform impactors in lieu of a whole-body dummy makes it almost impossible to determine the net change in pedestrian or cyclist risk resulting from a specific vehicle design alteration [32].

The principal goal of a subsystem impactor test is to reproduce the force interaction between the corresponding vehicle and pedestrian body parts, and provide an assessment of the resulting injury likelihood by comparison with accident or cadaver data. A general assessment of the validity of an impactor can be made by comparing the vehicle damage from an impactor test with the corresponding damage in a staged whole-body dummy test or an accident reconstruction [33], but this does not guarantee reproduction of the impact force time history. In consequence the link between impactor dynamics and specific injury likelihood remains the subject of research.

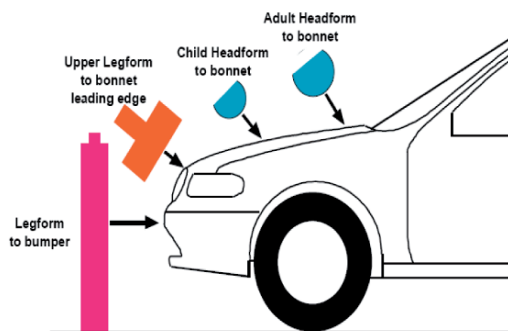


Fig. 6.1 Pedestrian subsystem tests, adapted from [34].

Subsystem Tests: Legform Impactor to Bumper

Accident data [24, 25] and staged tests with cadavers [35] have shown that the knee and the tibia are the most frequently injured body regions following contact with the bumper of passenger cars (see Table 2.4). Furthermore, these injuries occur early in the collision, before significant torso acceleration has commenced and therefore an isolated legform has been chosen for evaluating the bumpers of passenger cars, see Figure 6.1. Several legform impactors have been developed: an early prototype developed by INRETS in France [36] for the EEVC test is no longer used and the current EEVC legform was developed by the UK Transport Research Laboratory (TRL) [37, 38]. It consists of two foam-covered rigid segments representing the upper and lower leg of a 50th percentile adult male connected by a simulated knee joint. An alternative flexible legform impactor with a more biofidelic knee joint has been developed by the Japanese Automobile Manufacturers Association (JAMA) in collaboration with the Japanese Automobile Research Institute (JARI) [39]. Both legform impactors are designed to be propelled in free flight at 40 km/h against the vehicle bumper (see Figure 6.2), and the knee lateral bending angle and shear displacement and upper tibia lateral acceleration are compared to tolerance thresholds derived from biomechanical testing, see Table 6.1 for the EEVC and GTR proposed thresholds [40]. The ISO have also produced a specification for a legform impactor [41], but no physical prototype has been developed. For the IHRA draft Global Technical Regulation (GTR) [8], a higher tolerance of 19° for lateral knee bending is proposed [42] and the choice of the rigid TRL or the flexible JAMA-JARI legform remains under discussion. The flexible legform is more biofidelic [43] but its reliability has not been proven [21].

Dummy tests have shown that when the upper body mass is excluded, the knee bending moment in bumper impacts is decreased [46]. Nonetheless, the EEVC has considered the isolated legform impactor test representative of a whole-body dummy test if the bumper impact occurs at or below the knee joint level (i.e. 500 mm or less above the ground) [31]. For vehicles with high bumpers, the bumper impacts

Table 6.1 EEVC legform biomechanical thresholds for 40 km/h free flight impact to vehicle bumper.

	Knee bending angle	Knee shear displacement	Tibia head acceleration
EEVC	15°	6 mm	200 g
GTR	19°	6 mm	250 g

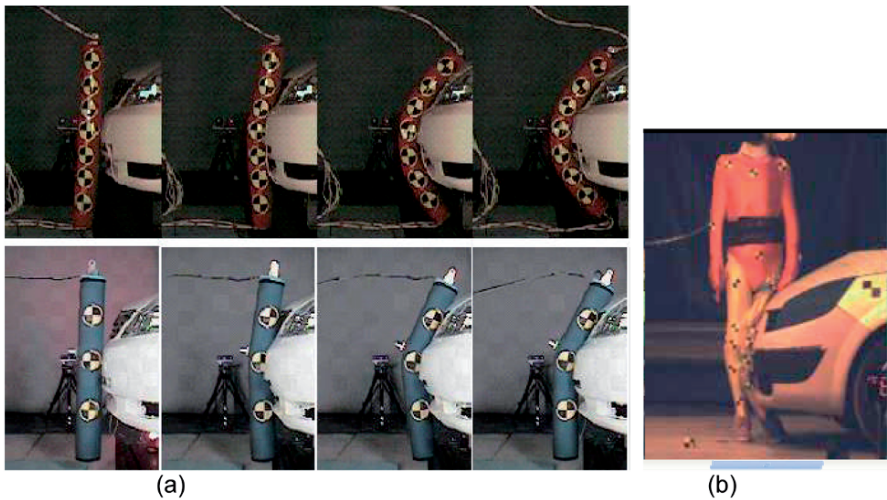


Fig. 6.2 (a) Comparison of TRL and JAMA-JARI legforms in contact with a vehicle bumper showing increased flexibility of the JAMA-JARI prototype, adapted from [44]. (b) cadaver tests, adapted from [45].

the femur part of the legform impactor and the absence of the hip joint constraint and upper body mass can lead to rigid body rotation of the legform impactor without lateral bending of the knee [47, 48]. Furthermore, there is no accelerometer in the femur component to assess the risk of fracture in this region. To account for this, the EEVC and GTR have allowed for an optional alternative horizontal upper legform to bumper test⁴ in cases where the bumper is located high and close to the bonnet leading edge [31]. However, the acceptance level for the upper legform test (see next section) in this configuration is demanding (5 kN force and 300 Nm bending moment) and the EEVC have used this to discourage a design trend towards higher bumpers [9]. The IHRA is now considering leg protection for high bumper vehicles and is researching an improved test procedure for the upper legform to high bumper vehicle test [42].

A comparison between impacts with the EEVC legform impactor and the Polar whole body pedestrian dummy (see Chapter 8) showed that in the first 15ms the knee bending angle time-history was similar in both cases, but between 20 and 30 ms the Polar dummy knee angle continued to increase while the legform impactor

⁴ The mass for the upper legform impactor in these cases is 9.5 kg, see next section.

knee angle did not [47]. A recent comparison between the EEVC legform impactor test and staged tests using cadavers struck by a Peugeot 206 and a Renault Twingo showed that the critical bending moment proposed by the EEVC for the legform impactor is appropriate, but that the shear displacement threshold is not [49].

Upper Legform to Bonnet Leading Edge

The height of the trochanter (top of the femur bone) for 5th–95th percentile males varies from 890–1040 mm and from 800–900 mm for females [50]. In comparison, measurements of popular cars published in 1990 showed bonnet leading edge heights ranging between 688–811 mm [51], and therefore the bonnet leading edge contact with an adult pedestrian is mostly in the region of the femur [52], see Chapter 10. In consequence, the EEVC upper legform impactor developed by TRL represents a (shortened) segment of an adult femur [31], see Figure 6.3. The design is intended to partially replicate the in-vivo situation where the flesh covered femur is constrained at the distal and proximal ends by the knee and hip joints and is free to bend in the middle. There are load cells at the upper and lower supports of the upper legform, and strain gauges measure the peak bending moment in the middle of the impactor. The upper legform test is configured such that the centre line of the impactor coincides with the bonnet leading edge reference line identified on the vehicle [40], see Figure 6.3.

Staged tests using an Ogle 50th percentile male dummy struck by instrumented car fronts [53] and multibody pedestrian models [54] showed that the effective mass, impact velocity and impact angle for the upper leg depend in a predictable manner on the shape of the vehicle front. For the staged tests, the bumper top-edge height from the ground was varied from 350–550 mm, the bonnet leading edge height varied from 600–850 mm and bumper leads of 50–250 mm were tested. The energy of deformation of the bumper-to-leg contact was then calculated from bumper load cells and accelerometer derived displacement of the leg relative to the vehicle. This resulted in input parameters for the EEVC upper legform impactor test where the vehicle's bonnet leading edge height and bumper lead are measured and look-up tables are used to define impactor velocity, impact angle and impact energy, see Figures 6.4 and 6.5. The impact mass is then determined from the impact velocity and energy. However, Japanese accident data showed that bumper lead was not significant for bonnet leading edge injuries [55] and for low bonnet leading edge heights the directions of the upper leg displacement and bonnet leading edge force vectors are not collinear, resulting in overestimation of the impact energy and the impact force/bending moment [56]. The upper legform impactor is a guided test, which does not allow rotation and sliding over the bonnet leading edge, a phenomenon which has been observed in the whole body models [56, 57]. In a comparison with the response of the Polar pedestrian dummy, the upper legform impactor tests were found to overestimate the severity of the pelvis impact with the vehicle front, es-

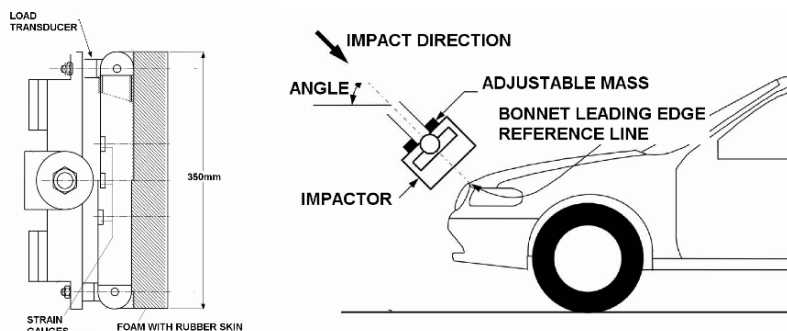


Fig. 6.3 The EEVC upper legform impactor and its application, adapted from [40].

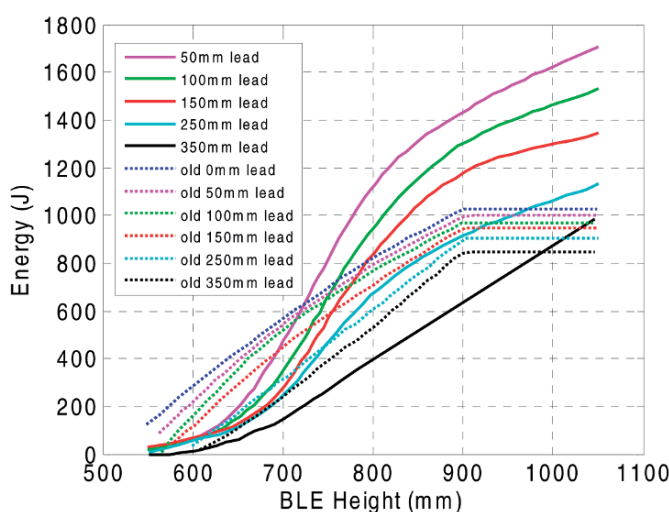


Fig. 6.4 Updated look-up tables for EEVC upper legform impact energy, adapted from [40].

pecially for high fronted sport utility vehicles [47]. To address these problems, an updated impact energy lookup table has been proposed by the EEVC, see Figure 6.4.

The biomechanical tolerance thresholds proposed for the impactor were intended to predict fractures of the femur and pelvis. The limits were 4 kN for the sum of the two load cells and a midshaft bending moment of 220 Nm, determined by replicating vehicle damage from real-world accidents using the upper legform impactor test and then correlating the impactor force and bending moment measurements with actual injuries [33]. However, very large scatter was evident in this data, see Table 6.2. This is probably due to the large variability in individual impact tolerance, but the chosen threshold values provide a poor means of distinguishing injury likelihood, even for the data from which these thresholds are derived, see also Chapter 5.

Table 6.2 Bonnet leading edge impactor forces and bending moments for accident reconstruction cases using EEVC upper legform impactor compared to reported injury level [33].

Injury level	Peak sum of impactor forces	Peak Bending moment
AIS1 or less	3.8–7.1 kN	217–271 Nm
AIS2	4.9–7.6 kN	236–403 Nm
AIS3	3.6–4.8 kN	167–246 Nm

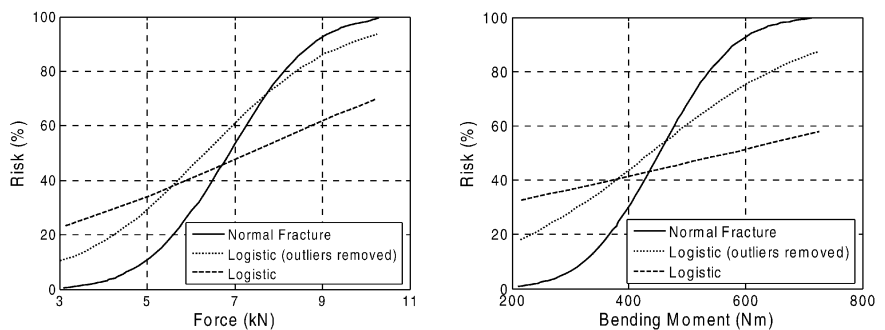


Fig. 6.5 Updated EEVC upper legform impactor tolerances, adapted from [40].

TRL and JARI accident reconstructions using an updated version of the upper legform impactor were subsequently combined to create a database of 39 cases [58], and cumulative normal distributions and logistic regressions resulted in femur/pelvis injury risk curves as a function of impactor force and bending moment, see Figure 6.5. Using this data and a recommended injury risk level of 20%, the EEVC threshold values were updated to 5 kN for the sum of impactor forces and a 300Nm bending moment limit [40].

Cadaver studies have shown that femur and pelvis injuries for impact energies below 200 J are insignificant and therefore no EEVC bonnet leading edge test is proposed for vehicles where the upper legform impact energy is less than 200 J. A recent comparison between the EEVC upper legform impactor test and staged tests using cadavers and multibody simulations showed that the upper leg impact angle and velocity in the EEVC specification are reasonably representative of real accidents for both a Peugeot 206 and a Renault Twingo [49]. However, the high incidence of spinal, thoracic and abdominal injuries in a sample of 80 pedestrian accidents highlights the types of serious injuries that are not addressed by the current subsystem testing protocols [59].

There is currently no ISO or IHRA upper legform test, partly due to difficulties encountered with this test by the EEVC [9], where the streamlined shape of modern passenger cars has resulted in a substantial reduction in bonnet leading edge injuries [9]. However, the IHRA plan further research in this area [60] as data from the USA shows a large number of mid-body injuries have been found due to the prevalence of light trucks and vans, and a means to evaluate these is important [42].

Headform to Bonnet Top: Adult and Child

The body region most frequently suffering serious injury in pedestrian accidents is the head [61], see Chapter 2. The location of head impact on the vehicle is a function of pedestrian stature, pre-impact stance and vehicle shape and speed (see Chapter 3), but in general head impact for shorter pedestrians occurs towards the front of the bonnet/wings, while for adults the head contacts towards the rear of the bonnet/wings or the windscreen/A-pillar (Chapter 3). Therefore, the EEVC and the ISO developed a child headform to assess child head injury risk from contact with the forward sections of the bonnet and an adult headform test to assess adult pedestrian injury risk from contact with the rear part of the bonnet, wings and scuttle. The EEVC headform tests are limited to the front of the vehicle up to, and including, the base of the windscreen, because these were the only areas that were thought feasible to improve [21]. The EEVC Working Group 17 considered that the headform impactor test is suitable for modelling cyclist injuries [62]. However, Maki et al have stated that cyclist head impact on the bonnet of passenger cars is rare, and have therefore questioned the validity of the EEVC test for cyclists [63].

Both the EEVC and ISO headforms are spherical and featureless to increase test repeatability and both have the same construction: a semi-rigid sphere, covered in a deformable rubber or foam ‘skin’, see Figure 6.6. The impactors are designed to be propelled in free flight at 40 km/h at a predetermined angle against various portions of the bonnet top. The IHRA headform specifications are the same as for the ISO [20].

The EEVC adult headform has shown good correlation of the resultant acceleration time history by comparison with cadaver tests [64]. In cadaver tests and computer simulations with a vehicle impact speed of 40 km/h, head impact velocities to the bonnet were 43–50 km/h, with the direction of impact⁵ approximately 65° to the horizontal [64]. Computer simulations also showed the head impact angle for adults to be circa 65° and 40–55° for children. However, the EEVC have recognised that head impact velocity varies substantially as a function of vehicle shape [31, 47] and the simplification of a fixed impact velocity for the headform test eliminates the influence of overall vehicle shape on head impact conditions, see Chapter 10. A recent comparison between the EEVC headform impactor test and cadaver tests with a Peugeot 206 and a Renault Twingo showed that the cadaver head impact velocity and the impact angle are generally lower than the EEVC specification [49].

The ISO/IHRA have proposed slightly different impactor specifications and impact conditions than the EEVC, see Table 6.3. The differences in the impactor masses are small, but there are substantial differences in the vehicle regions covered and in the prescribed impact conditions. For the EEVC tests, the child headform test area is bounded by Wrap Around Distances (WAD – see Chapter 3) of 1000–1500 mm, while the adult headform test area is bounded by a WAD of 1500–2100 mm, unless the latter extends beyond the windscreen, in which case the lower

⁵ The direction of the head impact velocity vector is found from the ratio of the vertical horizontal velocity of the head at the instant of impact with the bonnet.

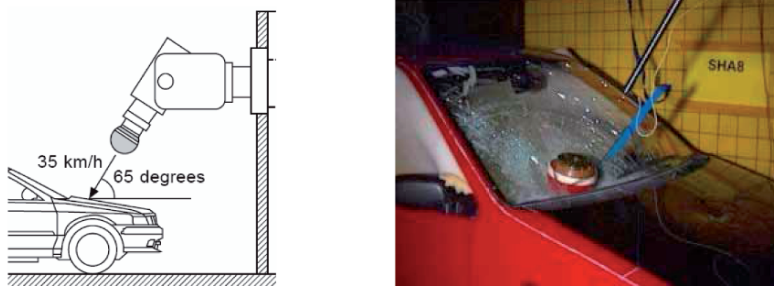


Fig. 6.6 (a) Configuration of headform to bonnet top test: adapted from [65], (b) application of headform impactor on vehicle windscreen, adapted from [66].

Table 6.3 EEVC, ISO and IHRA child headform impactor specifications, from [68].

Child headform	EEVC	ISO	IHRA
Mass	2.5 kg	3.5 kg	3.5 kg
Impact speed	40 km/h	Up to 40 km/h	Depends on vehicle shape
Wrap around distance	1000–1500 mm	1000–1500 mm	900–1400 mm
HIC/HPC Tolerance	HIC = 1000	HIC = 1000	HIC = 1000
Impact angle	50°		

windscreen frame forms the rear boundary of the test area. The EEVC test does not include the windscreen and only includes the windscreen frame if it lies within the specified WAD, even though the EEVC have accepted that most adult head impact locations are outside the zone considered by their test method [40].

The acceptance level for all headforms is a Head Injury Criterion⁶ (HIC) score of 1000, see Chapter 5. All proposed tests only account for the linear acceleration of the headform via the HIC, and do not include rotational acceleration effects which are known to be significant for brain trauma such as diffuse axonal injuries, see Chapter 5. However, a HIC value of 1000 has previously been found to be a good indicator of serious head injury for real pedestrian cases involving adults and children [67] and it was concluded that there was insufficient data available to propose a separate robust HIC acceptance level for child protection [31].

The IHRA have focused on head injuries resulting from impact with the vehicle front structure including the windscreen and A-pillars, and have developed head impact test conditions which depend on the vehicle shape. The IHRA investigated front shapes of passenger cars and categorised these into three groups: sedan and light and sport type vehicles, Sport Utility Vehicle [69] and one box [70] (see Figure 6.7) and these were used in computer simulations to determine the influence of vehicle front shape on head impact velocity, impact angle, WAD and effective head mass. The simulations showed the average effective head mass was close to actual

⁶ In some cases, the term Head Protection Criterion is used in favour of head injury criteria, but its means of calculation is identical.

Table 6.4 EEVC, ISO and IHRA adult headform impactor specifications, from [68].

Adult headform	EEVC	ISO	IHRA
Mass	4.8 kg	4.5 kg	4.5 kg
Impact speed	40 km/h	Up to 40 km/h	Depends on vehicle shape
Wrap around distance	1500–2100 mm	1500–2100 mm	1700–2400 mm
HIC/HPC Tolerance	HIC = 1000	HIC = 1000	HIC = 1000
Impact angle	65°		

head mass for cases of bonnet contacts, but only about 80% of actual head mass for cases of windscreen contact. However, for ease of testing the IHRA decided to specify only the average effective head mass for the adult and child headforms respectively. The headform impact velocity, angle and location are prescribed using look-up tables based on vehicle shape.

For the GTR, child and adult headform tests at 35 km/h are defined. The threshold HIC score for the bonnet or windscreen is 1000 and the windscreen frame and A-pillars are excluded from the test zone [42]. The reason for this exclusion is that it was considered that the A-pillar stiffness requirements for vehicle rollover protection preclude reduction of HIC scores for pedestrian head impact against the A-pillars. For the windscreen, the existing low HIC scores for mid-windscreen contacts mean that the advantages of including the windscreen in the test would be minimal [42], see Chapter 10.

The GTR Working Group reviewed the IHRA recommendation to take into account the shape of the front of the vehicle. However, they concluded that the EEVC tests, which do not account for the influence of vehicle shape, are in fact more severe and decided to follow the same approach, stating that the IHRA recommendations would be difficult to put in place in the context of a regulatory and certification approach [42].

Implementation into Legislation

The EEVC pedestrian impactor tests have been included in a European Directive [10] which is decomposed into two stages: phase 1 is in place since 2005 and specifies the EEVC pedestrian impactor tests. Phase 2 is due to be implemented in 2010, and tougher requirements as well as inclusion of brake-assist technology are envisaged, but this is currently under review [71]. Vehicles intended for off-road usage often have additional frontal protection structures ('bullbars') to protect the vehicle front-end from damage in minor collisions. These pose a significant injury risk to pedestrians and other vulnerable road users, see chapter 10, and a requirement to pass the EEVC impactor tests has been implemented in Europe [72] and Australia [73]. In Japan, adult and child headform impact tests based on the IHRA pedestrian safety working group recommendations are required [74]: the headform

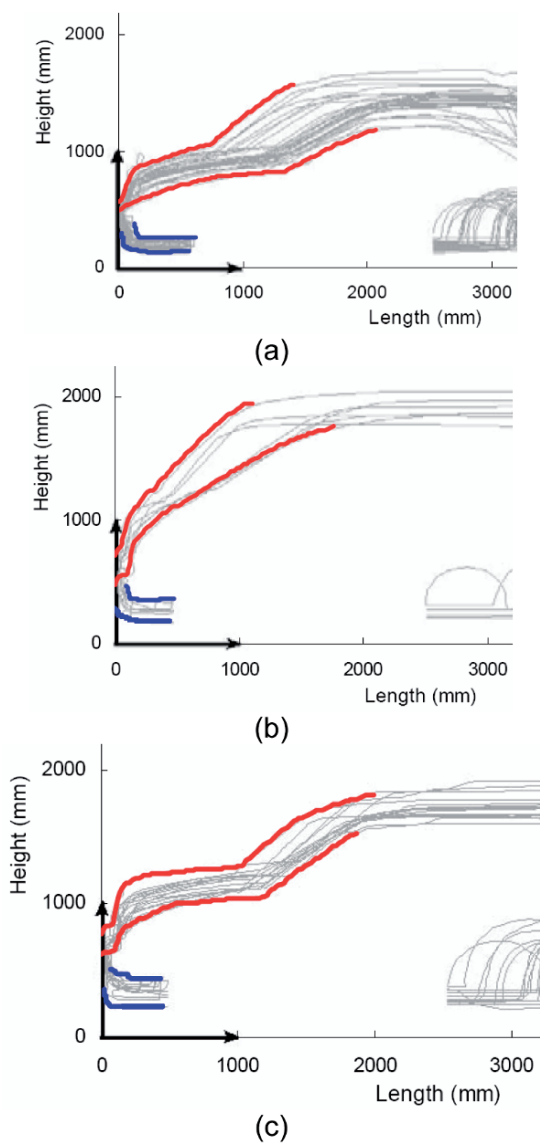


Fig. 6.7 The IHRA vehicle shape categories (a) sedan, light vehicle and sports types, (b) SUV and (c) 1-box vehicles, adapted from [70].

impact speed is 32 km/h and the impact angle is determined based on the IHRA vehicle categories (sedan, SUV or 1-box) [20]. There is no legislation for pedestrian safety in the United States, but legislation on pedestrian safety is evolving and subject to updates in many countries.

Concluding Remarks

The EEVC, ISO and the IHRA have all proposed subsystem impactor tests instead of whole body dummy tests to represent the critical impacts between vehicle fronts and pedestrian body regions. The proposed tests include a legform to bumper test, an upper legform to bumper or bonnet leading edge test, and adult and child headform impact tests with a range of vehicle regions. Not all standards include all of these tests, but despite some differences in the specifications, there is substantial similarity in the tests proposed by the different standards bodies.

It has generally been hoped that provision of pedestrian safety by improved vehicle front-end design will also improve cyclist safety, but more research is required to define cyclist-specific impactor test conditions and similarly, pedestrian and cyclist test methods for the fronts of buses, coaches and goods vehicles are also required.

The chief limitations of the sub-system tests are that this approach makes it very difficult to determine the overall change in pedestrian risk resulting from a specific vehicle design alteration, and there are some crucial midbody regions that are not covered by the current tests. Nonetheless, these tests have been adopted in the EuroNCAP consumer safety ratings and have been implemented legislatively in a number of jurisdictions including the EU, Japan and Australia.

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Chapter 7

Mathematical Formulations for Impact Modelling

Introduction

A wide variety of mathematical models have been applied to the study of pedestrian and cyclist impact. Early researchers formulated one-dimensional particle models of the human body to analyze the relationship between impact speed and pedestrian projection distance (e.g. [1], see Chapter 4). Today, finite element models of the human body with many thousands of degrees of freedom¹ are used to predict pedestrian injuries [2–4]. However, despite the level of detail in these models, they do not automatically yield meaningful injury predictions (see Chapter 8), and their inherent complexity can mask fundamental insights which a simpler model may emphasise.

This raises the important question of the appropriate complexity when formulating a model to analyze a problem. Clearly, a particle model of the human body cannot be used to predict detailed body kinematics, let alone body deformations and injuries, but it is less obvious whether a detailed multi-body or even finite element model is more appropriate to study pedestrian or cyclist projection distances than a simplified particle model. A guiding principle should be that if two models have been validated for a given purpose and yield the same prediction of a response, then the simpler model is better for that application, because additional detail which does not improve a model's predictions serves only to reduce clarity.² Furthermore, as models become more complex they usually require more detailed knowledge of the input parameters, and this is frequently a limiting factor.

A further problem is the validity of a model in a particular application. This is especially relevant for detailed models because the range of potential applications is much greater. For example, application of a multi-body pedestrian model as a cyclist has been found to cause problems.³ In contrast, this risk is greatly reduced for a

¹ This is the number of independent variables required to fully define a system.

² “Make everything as simple as possible, but not simpler” – Albert Einstein.

³ In the Madymo 6.2 Pedestrian model the hip, knee and ankle joint torque-angle relationships implemented to facilitate standing in the pedestrian application result in large internal moments when the model is placed in a seated position on a bicycle.

particle model because its limited domain of applicability is more readily apparent, and there is therefore little risk of using a particle model to predict injuries. However, the value of appropriately validated detailed models is undeniable as these can really help in the evaluation of pedestrian risk when struck by different vehicle shapes (see Chapter 10). Engineering judgement is therefore required to choose a model with an appropriate level of complexity.

To illustrate this compromise between model complexity and clarity of interpretation, this chapter explicitly develops the key features of the impact process for pedestrians and cyclists. The approach taken is to develop impact models in an increasing order of complexity, and compare the model predictions at each stage with the available experimental data. This allows the reader to observe fundamental insights from the basic models while also providing the background to more complex multi-body and finite element models that are now routinely used in research and industrial settings.

The mathematical principles required for modelling pedestrian and cyclist impact are of course identical. Therefore, for brevity the following developments are specifically applied to pedestrian impact modelling. However, the same theoretical approaches can be applied to cyclists, though the addition of the bicycle does increase the modelling complexity. It is hoped that this approach will encourage new researchers to use application of the basic laws of impulse and momentum to ensure that the predictions of their detailed models lie within bounds that can be predicted using basic mechanics. However, the exponential rise in complexity that accompanies additional modelling detail should convince most readers of the benefits of user-friendly proprietary software when assembling complex models.

Notation

Comparison of pedestrian impact models with increasing degrees of complexity requires a consistent approach to notation, and the notation used is introduced here. A standing pedestrian (mass m) and a vehicle (mass M) just prior to impact are shown in Figure 7.1. The global origin and the pedestrian coordinate system are (arbitrarily) chosen as the centre of gravity of the pedestrian at the instant of first contact with the vehicle (A). The vehicle coordinate system origin is at the centre of gravity (cg) of the vehicle (J). The positive X direction is aligned with the vehicle travel direction, and the positive Z direction is vertically upwards. The Y direction is then defined according to the right-hand coordinate system convention. The unit vectors in the X , Y and Z directions are $\vec{i}$, $\vec{j}$ and $\vec{k}$ respectively, such that $\vec{i} \times \vec{j} = \vec{k}$, where the arrow subscript denotes a vector quantity and $\times$ is the vector cross product.

The three time varying linear position degrees of freedom of the pedestrian cg are $x(t)$, $y(t)$ and $z(t)$. The corresponding linear position degrees of freedom of the vehicle cg are $X(t)$, $Y(t)$ and $Z(t)$. The angular velocities of the pedestrian and vehicle about the three principal axes are given by $\omega_x(t)$, $\omega_y(t)$ and $\omega_z(t)$, and

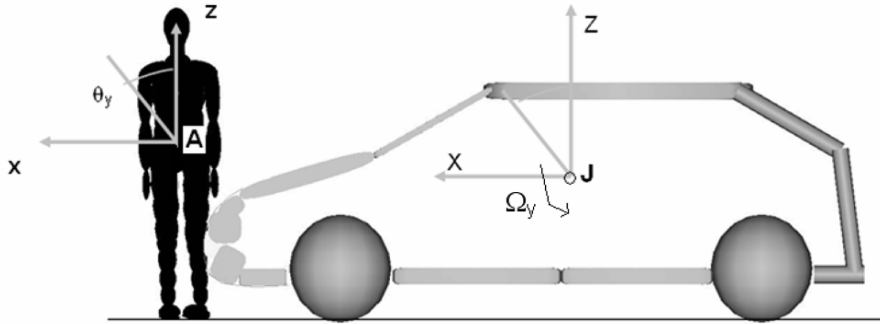


Fig. 7.1 Typical configuration of a side struck pedestrian and vehicle at the instant of first contact. The global origin is chosen to be at the pedestrian centre of gravity just before impact at A and the vehicle centre of gravity is at J.

$\Omega_x(t)$, $\Omega_y(t)$ and $\Omega_z(t)$ respectively. The dot superscript denotes the time derivative, so that the linear velocities and accelerations of the pedestrian cg are $\dot{x}(t)$, $\dot{y}(t)$ and $\dot{z}(t)$, and $\ddot{x}(t)$, $\ddot{y}(t)$ and $\ddot{z}(t)$ respectively. The same convention applies to the pedestrian angular velocity and the vehicle cg velocity and acceleration. However, for generalised three-dimensional rotations, it is not possible to directly relate angular velocity components to rotations about the principal axes since finite rotations cannot be described by vector operations. In this chapter, planar motion in the X – Z plane is assumed, and then the angular velocity of the pedestrian about the Y axis $\omega_y(t)$ is the time derivative of the angular orientation $\theta_y(t)$, see Figure 7.1.

A more detailed view of the pedestrian and the vehicle front is shown in Figure 7.2. The vertical and horizontal distances from the pedestrian's cg to the location of the vehicle impact on the body are h and b respectively. The distances from the cg to the feet and head are p and q respectively, such that the overall pedestrian length is $L = p + q$. If the pedestrian is stationary at impact and the vehicle is travelling at a velocity V , then the initial conditions are: $\dot{x}(0) = 0$, $\dot{X}(0) = V$.

Timing

It is helpful to define temporal markers which are common to nearly all pedestrian impacts. In most cases, the initial impact involves the bumper and bonnet leading edge area striking the leg and thigh region in quick succession, see Chapter 3. Although the bumper typically strikes the legs about 10–15 ms before the bonnet leading edge makes contact with the body (see Chapter 3), in this chapter a single impact representing both the bumper and bonnet leading edge contacts is assumed as a fist approximation. This occurs when F on the vehicle makes contact with the leg at B (see Figure 7.2), and is referred to as the primary impact, designated as time zero (t_0). The position of the pedestrian cg at this time is therefore $x(t_0)$, $y(t_0)$ and $z(t_0)$.

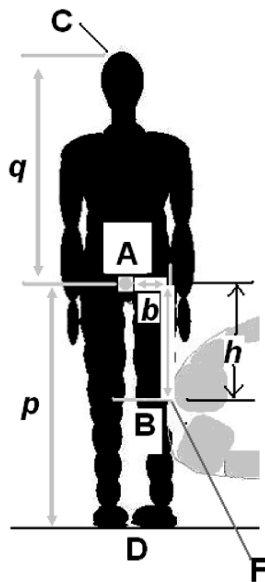


Fig. 7.2 Simplified geometric details of the pedestrian and vehicle front-end.

In the case of a low fronted vehicle striking a pedestrian, this is followed by rotation of the body over the front of the vehicle and further (generally more minor) impacts occur until the head makes contact with the bonnet or windshield, see Chapter 3. This is referred to as the secondary impact and designated as time (t_1), such that the position of the pedestrian cg at this time will be $x(t_1)$, $y(t_1)$ and $z(t_1)$. Following this, there is the possibility of further contacts with the vehicle front, depending mostly on braking level, but an airborne period usually follows, terminated by the first impact with the ground which is referred to as time t_2 . The collision is completely over following a period of slide, roll and bounce of the pedestrian to rest at time (t_{end}), see Chapter 3.

The plus and minus symbols are used to facilitate reference to an instant just before and just after an impact: therefore $\dot{x}(t_{0-})$ and $\dot{x}(t_{0+})$ refer to the X component velocity of the pedestrian cg just before and just after primary contact with the vehicle respectively.

Impulse and Momentum

The basis of all mechanics is Newton's Second Law:

$$\sum F = m \frac{dv}{dt}. \quad (7.1)$$

When there are no externally applied loads, $\sum F = 0$ and linear momentum (mv) is conserved. Therefore, assuming no interaction with the road during primary impact between a vehicle and a pedestrian, we can write down the conservation of horizontal momentum during primary impact:

$$M \dot{X}(t_{0-}) + m \dot{x}(t_{0-}) = M \dot{X}(t_{0+}) + m \dot{x}(t_{0+}). \quad (7.2)$$

It is common to define the velocity change during an impact using the delta symbol (Δv), such that for the pedestrian:

$$\Delta v = [\dot{x}(t_{0+}) - \dot{x}(t_{0-})] \quad (7.3)$$

and for the vehicle:

$$\Delta V = [\dot{X}(t_{0+}) - \dot{X}(t_{0-})]. \quad (7.4)$$

Assuming an at-rest pedestrian is struck by a vehicle in a concentric and plastic impact (no rotation or restitution) such that there is a common post impact velocity, equations (7.2) and (7.3) yield

$$\Delta v = \frac{M}{m + M} V. \quad (7.5)$$

For an average adult male pedestrian of mass 77 kg and a vehicle of mass 1200 kg [5], the velocity change of a stationary pedestrian centrally struck by a vehicle is therefore about 95% of the vehicle impact speed. During contact with the vehicle, the momentum of the pedestrian changes according to the impulse of the contact force (F_{contact}). Mathematically, this is found from integrating Newton's Second Law with respect to time:

$$m[\dot{x}(t_{0+}) - \dot{x}(t_{0-})] = \int_{t_{0-}}^{t_{0+}} F_{\text{contact}} dt. \quad (7.6)$$

Much of the complexity of pedestrian impacts relates to the force interaction described by equation (7.6), the articulation of the body and the fact that the impact is normally eccentric, so that the resulting motion is both linear and rotational.

In general, the integral in equation (7.6) cannot be explicitly evaluated in a crash test without detailed instrumentation. However, by noting that total energy is conserved and assuming that kinetic energy lost in the impact is converted to work done in deforming the contacting structures, we can derive a relation between impact speed, deformation and mean acceleration in a plastic concentric impact where the final common velocity is V_{final} . The loss in kinetic energy in the impact is

$$\Delta E_K = \frac{1}{2} [M V^2 - (m + M) V_{\text{final}}^2]. \quad (7.7)$$

But

$$V_{\text{final}} = \frac{M}{m + M} V \quad (7.8)$$

and therefore the loss in kinetic energy simplifies to

$$\Delta E_K = \frac{1}{2} V^2 \left[\frac{mM}{m+M} \right]. \quad (7.9)$$

Work is done deforming both the pedestrian and the vehicle to a combined crush of $d_{\max}$ at the contact interface. The energy balance equation is:

$$\frac{1}{2} V^2 \left[\frac{mM}{m+M} \right] = \int_0^{d_{\max}} F \, dd_{\text{crush}}. \quad (7.10)$$

Using mean acceleration ($\bar{a}$) as a crude measure of collision severity [6], and simplifying the integral in equation (7.10) using Newton's Second Law yields the following relationship between crush and velocity change and acceleration of the pedestrian:

$$\bar{a} = \left[\frac{1}{2} \frac{V^2}{d_{\max}} \right] \left[\frac{M}{m+M} \right]. \quad (7.11)$$

Equation (7.11) shows that the severity of a plastic, concentric impact increases with the square of the impact velocity for a fixed crush depth, and is inversely proportional to the deformation for a fixed impact velocity.

Single Segment Formulation Using Momentum Considerations

Considering the pedestrian as a rigid segment, and applying the conservation of linear and angular momentum during the primary impact and assuming that

1. the primary impact has no moment about the vehicle cg,
2. the vertical impulse is negligible, and
3. the impact is plastic,

the following system of equations results:

$$\begin{bmatrix} m & M & 0 \\ 0 & -Mh & mk^2 \\ 1 & -1 & h \end{bmatrix} \begin{Bmatrix} \dot{x}(t_{0+}) \\ \dot{X}(t_{0+}) \\ \dot{\theta}_y(t_{0+}) \end{Bmatrix} = \begin{Bmatrix} M\dot{x}(t_{0-}) = MV \\ M\dot{X}(t_{0-})h \\ 0 \end{Bmatrix}, \quad (7.12)$$

where k is the radius of gyration of the pedestrian, h is the vertical distance between the pedestrian cg and the assumed primary impact location on the vehicle (see Figure 7.2). This system of equations expresses the post primary impact kinematic variables ($\dot{x}(t_{0+})$, $\dot{X}(t_{0+})$ and $\dot{\theta}_y(t_{0+})$) in terms of the impact geometry (h) and inertia parameters (m , M and k) and the pre-impact vehicle speed (V).

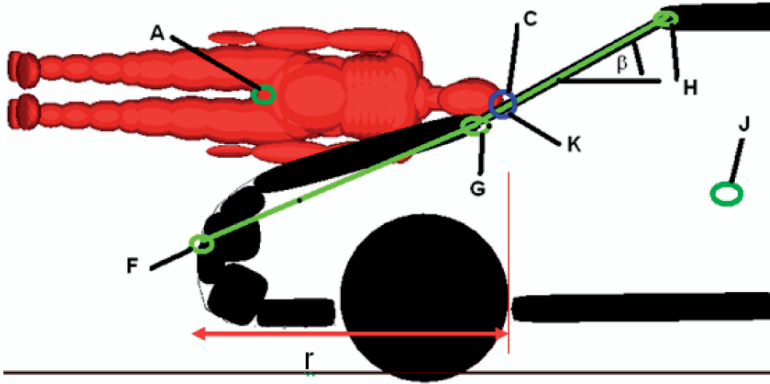


Fig. 7.3 Head impact configuration and vehicle shape parameters for a single segment momentum model.

Post Primary Impact Kinematics

In the time interval between primary and secondary impact, and assuming no further contacts, gravity is the only force acting on the pedestrian. Therefore, the pedestrian cg horizontal velocity and angular velocity remain constant, while the vertical velocity changes over time under the action of gravity.

Head Contact Time

At time t_1 the pedestrian's head (point C) strikes the bonnet or windscreen (point K), a horizontal distance r from the initial vehicle impact location at F, see Figure 7.3. The timing of head contact on the bonnet/windscreen depends on the geometry, inertia and initial velocities of the pedestrian and the vehicle. A simple search method can be implemented by considering the bonnet/windscreen as a line of slope $\tan(\beta)$. Head contact occurs when the X and Z coordinates of the head intersect with the bonnet/windscreen line such that

$$\begin{aligned} & \dot{z}(t_{0+})t_1 + q \cos \theta_y(t_1) - \frac{1}{2}gt_1^2 \\ &= \tan(\beta)[\dot{x}(t_{0+})t_1 + q \sin(t_1)] + G_z \\ &+ \tan(\beta)\left[-\left(\dot{X}(t_{0+})t_1 - \frac{1}{2}\mu gt_1^2\right) + b + r\right], \end{aligned} \quad (7.13)$$

where μ is the vehicle braking friction coefficient, and G_z is the vertical coordinate of point G at the base of the windscreen. This equation can be solved approximately (e.g. using the method of bisections) to yield the head contact time.

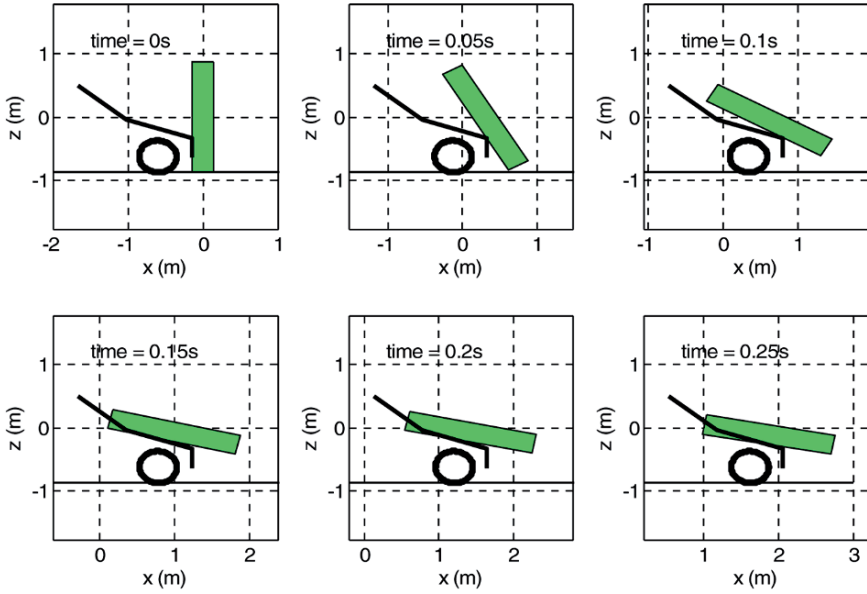


Fig. 7.4 Predictions of pedestrian and vehicle motion using single segment momentum based method: $h = 0.33$ m, $V = 10$ m/s.

Post Head Impact Kinematics

The conservation of linear and angular momentum can again be used to determine the post head impact velocities. However, in this impact a significant momentum exchange occurs in both the X and Z directions due to the oblique angle of the windscreen. Assuming again a plastic impact and that the impact has no moment about the vehicle cg yields:

$$\begin{bmatrix} m & 0 & 0 & M & 0 \\ 0 & m & 0 & 0 & M \\ 1 & 0 & q \cos \theta(t_1) & -1 & 0 \\ 0 & 1 & -q \sin \theta(t_1) & 0 & -1 \\ 0 & 0 & mk^2 & Mq \cos \theta(t_1) & -Mq \sin \theta(t_1) \end{bmatrix} \begin{Bmatrix} \dot{x}(t_{1+}) \\ \dot{z}(t_{1+}) \\ \dot{\theta}(t_{1+}) \\ \dot{X}(t_{1+}) \\ \dot{Z}(t_{1+}) \end{Bmatrix} \\
 = \begin{bmatrix} m\dot{x}(t_{1-}) + M\dot{X}(t_{1-}) \\ m\dot{z}(t_{1-}) \\ 0 \\ 0 \\ K_1 \end{bmatrix}, \quad (7.14)$$

where

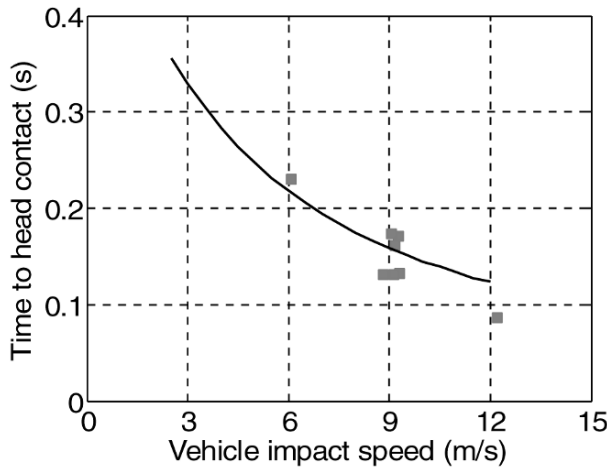


Fig. 7.5 Time to head contact as a function of vehicle impact speed, cadaver test cases [7] (grey squares) and single segment momentum model predictions (black line).

$$K_1 = mk^2\dot{\theta}(t_{1-}) - Mq \sin \theta(t_1)\dot{Z}(t_{1-}) + Mq \cos \theta(t_1)\dot{X}(t_{1-}).$$

This system of equations expresses the post head impact kinematic variables in terms of the impact geometry and inertia parameters and the pre head impact kinematics. Implementation of equations (7.12)–(7.14) describing the single segment pedestrian model using this momentum approach yields piece-wise continuous linear and angular velocity time histories, so the kinematics of pedestrian motion following vehicle impact can be predicted, see Figure 7.4. This approach assumes the contacts occur over infinitesimally short times.

The ability of this model to predict actual pedestrian impact kinematics can be assessed by comparison with staged tests. Severy et al. [7] performed dummy impact tests with a passenger car and Figure 7.5 compares the time taken to reach head contact as a function of vehicle impact speed for the cadaver tests and for the single segment momentum model. The model provides a very good fit to the experimental trend.

Cesari et al. [8] measured cadaver head impact velocity on vehicles as a function of passenger car impact speed. Figure 7.6a compares these cadaver test results with the single segment momentum model predictions, again showing the ability of the model to predict the experimental trends, though the assumption of rigidity in the model does result in some over-prediction of the head impact speed. The same approach has also been successfully used by Wood to predict the wraparound ratio for head impact locations on a vehicle, see Figure 7.6b [9].

These results show that application of the conservation of momentum to a single segment model of the pedestrian is sufficient to give a good representation of the overall kinematics of pedestrian motion following a vehicle impact. Specific kin-

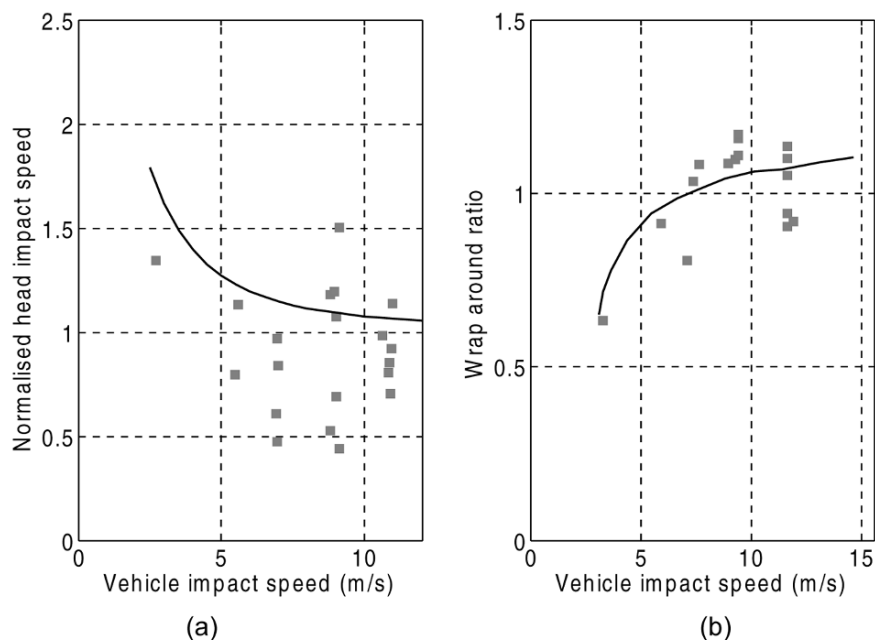


Fig. 7.6 (a) Head impact velocity as a function of vehicle impact speed: Cadaver test cases [8] (grey squares) and single segment momentum model (black line); (b) Wood single segment momentum model prediction of wrap around ratio to head contact [9] compared to cadaver tests [8].

ematic features such as the time to head contact, head impact velocity and wrap around distance to head impact location can be successfully predicted. It is noteworthy that this is achieved without explicit modelling of the time varying contact forces, the deformation of the contacting surfaces and the articulation of the pedestrian body segments. It can therefore be concluded that pedestrian kinematics are predominantly governed by inertia and geometry.

Pedestrian Formulation Using an Ordinary Differential Equation (ODE) Approach

We have seen that pedestrian kinematics can be quite well represented using the conservation of momentum applied to a single segment formulation. This approach predicts pedestrian motion without any knowledge of the contact interaction forces, and this is a major advantage in cases where the effective contact stiffness is not known. However, this approach cannot be directly used to predict pedestrian injuries, since injuries are related to forces and accelerations rather than velocity.

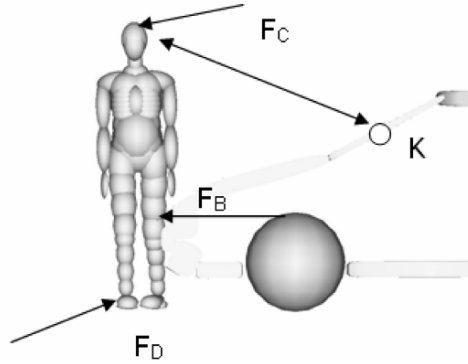


Fig. 7.7 Forces on the pedestrian for implementation of a single segment ODE approach.

Dynamic rigid body interaction between the vehicle and the pedestrian can be modeled by explicit integration of the equations of motion for the system. Due to the discontinuous and complex nature of contact interactions, numerical integration techniques are used in favour of an analytical integration approach. This approach yields the time varying contact forces, and can directly incorporate the stiffness of different contact segments and is therefore a major step towards prediction of pedestrian injuries. It does however require some *a priori* knowledge of the contact behaviour between the pedestrian and the vehicle.

The planar equations of motion of the pedestrian segment can be written down by consideration of the forces and moments which act on it. It is assumed that

1. Point F on the vehicle strikes point B on the pedestrian, and after this point C on the pedestrian strikes point K on the windshield.
2. There is a ground reaction acting on the feet at point D.

The contact force vectors ($\vec{F}_B$, $\vec{F}_C$ and $\vec{F}_D$) are non-zero only when the respective contacting surfaces intersect. The simplest contact force model is a linear spring, where the contact penetration is analogous to spring deflection. At each time-increment in the integration procedure, the penetrations (pen_B , pen_C and pen_D) are evaluated and the contact force is the product of the penetration depth and the specified contact stiffness. An equal and opposite force is applied to each half of the contact pair for the duration of the next time-step. In this way the impulse of each contact is applied. The equations of motion can be expressed in matrix format as follows:

$$\begin{bmatrix} m & 0 & 0 & 0 \\ 0 & m & 0 & 0 \\ 0 & 0 & mk^2 & 0 \\ 0 & 0 & 0 & M \end{bmatrix} \begin{Bmatrix} \ddot{x} \\ \ddot{z} \\ \ddot{\theta} \\ \ddot{X} \end{Bmatrix} = \begin{bmatrix} F_{Bx} + F_{Cx} + F_{Dx} \\ F_{Bz} + F_{Cz} + F_{Dz} - mg \\ \vec{r}_{BA} \times \vec{F}_B + \vec{r}_{CA} \times \vec{F}_C + \vec{r}_{DA} \times \vec{F}_D \\ -\mu Mg - F_{Bx} - F_{Cx} \end{bmatrix}, \quad (7.15)$$

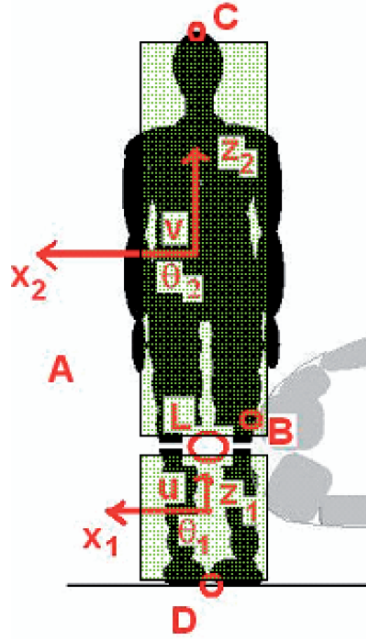


Fig. 7.8 Two segment rigid body pedestrian with hinge at L (the knee): lower and upper body segments with mass centres at u and v respectively.

where

$$\vec{r}_{BA} = [-Lt \cos(\phi - \theta) \ 0 \ -Lt \sin(\phi - \theta)],$$

$$\vec{r}_{CA} = [q \sin \theta \ 0 \ q \cos \theta],$$

$$\vec{r}_{DA} = [-p \sin \theta \ 0 \ -p \cos \theta],$$

$$\vec{F}_B = [k_B \text{pen}_B \ 0 \ 0],$$

$$\vec{F}_C = [k_C \text{pen}_C \sin \beta \ 0 \ k_C \text{pen}_C \cos \beta],$$

$$\vec{F}_D = [k_D \text{pen}_D \ 0 \ 0],$$

$$\phi = \tan^{-1}(h/b),$$

$$Lt = \sqrt{h^2 + b^2}.$$

Equation (7.15) can be solved simultaneously at each timestep for the accelerations and a time marching numerical integration scheme (such as Euler) can be used to predict positions and velocities.

A comparison between kinematic predictions using this ODE approach and the previous momentum based method is given in Figure 7.9a. For the momentum based

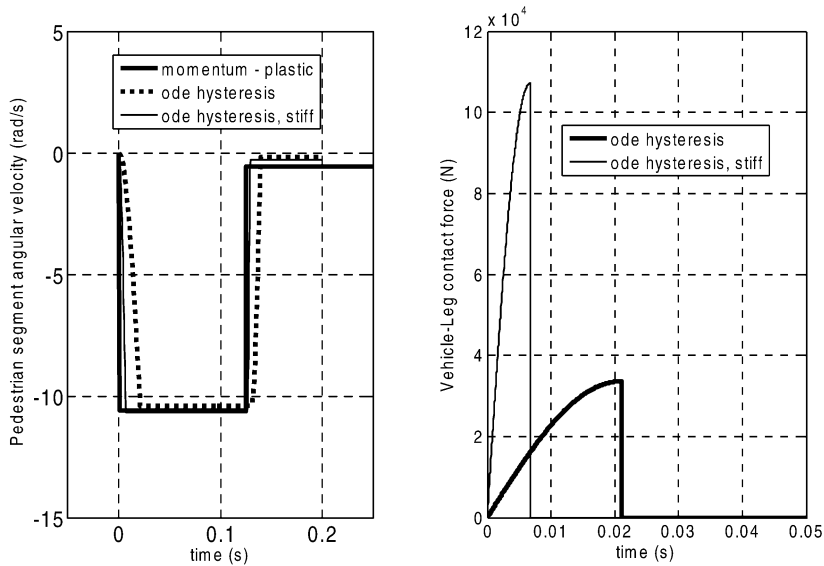


Fig. 7.9 (a) Pedestrian single segment angular velocity predictions from momentum considerations showing step changes in angular velocity compared to the ODE approach; (b) time history of leg contact force from ODE method with hysteresis showing the effect of a stiffer contact.

model, each contact results in a step change in angular velocity, whereas for the ODE model the angular impulse occurs over a finite period of time. A special case of the ODE model is when the contact stiffness is very high and there is no restitution. The effect of this hysteresis is that the resulting kinematics are almost identical to the zero-restitution momentum model.

The ODE approach of the pedestrian with time varying contact forces allows prediction of the principal contact forces and whole-body kinematics as function of contact stiffness, see Figure 7.9b. The area under the curves (impulse) is similar for a stiffer or a more compliant contact, but the peak force is much higher for the stiffer contact. This ODE approach forms the basis of more sophisticated commercial multi-body codes such as Madymo [10].

Rigid Body ODE Approach with a Hinge Segment

The articulating joints of the body mean that the rigid segment representation of a pedestrian has limited accuracy. The simplest articulation is to allow a hinged joint at L , for example at the knee (see Figure 7.8), such that rotation of the upper body relative to the lower body in the $X-Z$ plane is possible. The pedestrian now consists of a lower leg segment (mass m_1 , radius of gyration k_1) and an upper body segment (m_2 , k_2), and a rotational spring with stiffness k_θ at L can be used to represent the

lateral angular stiffness of the knees. The two segment approach with other joint locations (e.g. the neck) may be more appropriate in other applications.

The second derivative of the constraint provided by the hinge joint at L is

$$\begin{aligned}\ddot{x}_2 &= \ddot{x}_1 + \frac{L_1}{2}\ddot{\theta}_1 \cos \theta_1 - \frac{L_1}{2}\dot{\theta}_1^2 \sin \theta_1 + \frac{L_2}{2}\ddot{\theta}_2 \cos \theta_2 - \frac{L_2}{2}\dot{\theta}_2^2 \sin \theta_2, \\ \ddot{z}_2 &= \ddot{z}_1 + \frac{L_1}{2}\ddot{\theta}_1 \sin \theta_1 - \frac{L_1}{2}\dot{\theta}_1^2 \cos \theta_1 - \frac{L_2}{2}\ddot{\theta}_2 \sin \theta_2 - \frac{L_2}{2}\dot{\theta}_2^2 \cos \theta_2. \quad (7.16)\end{aligned}$$

Assuming impact occurs on the upper body segment, the corresponding equations of motion of the two body segments and the constraint provided by the joint can be expressed in matrix format as

$$\begin{bmatrix} m_1 & 0 & 0 & 0 & 0 & 0 & 0 & -1 & 0 \\ 0 & m_1 & 0 & 0 & 0 & 0 & 0 & 0 & -1 \\ 0 & 0 & m_1 k_1^2 & 0 & 0 & 0 & 0 & \frac{L_1}{2} \cos \theta_1 & \frac{L_1}{2} \sin \theta_1 \\ 0 & 0 & 0 & m_2 & 0 & 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 0 & m_2 & 0 & 0 & 0 & 1 \\ 0 & 0 & 0 & 0 & 0 & m_2 k_2^2 & 0 & -\frac{L_2}{2} \cos \theta_2 & \frac{L_2}{2} \sin \theta_2 \\ 0 & 0 & 0 & 0 & 0 & 0 & M & 0 & 0 \\ 1 & 0 & \frac{L_1}{2} \cos \theta_1 & -1 & 0 & \frac{L_2}{2} \cos \theta_2 & 0 & 0 & 0 \\ 0 & 1 & -\frac{L_1}{2} \sin \theta_1 & 0 & -1 & -\frac{L_2}{2} \sin \theta_2 & 0 & 0 & 0 \end{bmatrix} \begin{Bmatrix} \ddot{x}_1 \\ \ddot{z}_1 \\ \ddot{\theta}_1 \\ \ddot{x}_2 \\ \ddot{z}_2 \\ \ddot{\theta}_2 \\ \ddot{X} \\ F_{Lx} \\ F_{Lz} \end{Bmatrix} = \begin{bmatrix} F_{Dx} \\ F_{Dz} - m_1 g \\ \vec{r}_{DU} \times \vec{F}_D + k_\theta(\theta_2 - \theta_1) \\ F_{Bx} + F_{Cx} \\ F_{Bz} + F_{Ca} - m_2 g \\ \vec{r}_{BV} \times \vec{F}_B \times \vec{r}_{CV} \times \vec{F}_C + k_\theta(\theta_1 - \theta_2) \\ -F_{Bx} - F_{Cx} - \mu M g \\ \frac{L_1}{2}\dot{\theta}_1^2 \sin \theta_1 + \frac{L_2}{2}\dot{\theta}_2^2 \sin \theta_2 \\ \frac{L_1}{2}\dot{\theta}_1^2 \cos \theta_1 + \frac{L_2}{2}\dot{\theta}_2^2 \cos \theta_2 \end{bmatrix},$$

where

$$\begin{aligned}\vec{r}_{DU} &= \begin{bmatrix} -\frac{L_1}{2} \sin \theta_1 & 0 & -\frac{L_1}{2} \cos \theta_1 \end{bmatrix}, \\ \vec{r}_{LU} &= \begin{bmatrix} \frac{L_1}{2} \sin \theta_1 & 0 & \frac{L_1}{2} \cos \theta_1 \end{bmatrix}, \\ \vec{r}_{LV} &= \begin{bmatrix} -\frac{L_2}{2} \sin \theta_2 & 0 & -\frac{L_2}{2} \cos \theta_2 \end{bmatrix},\end{aligned}$$

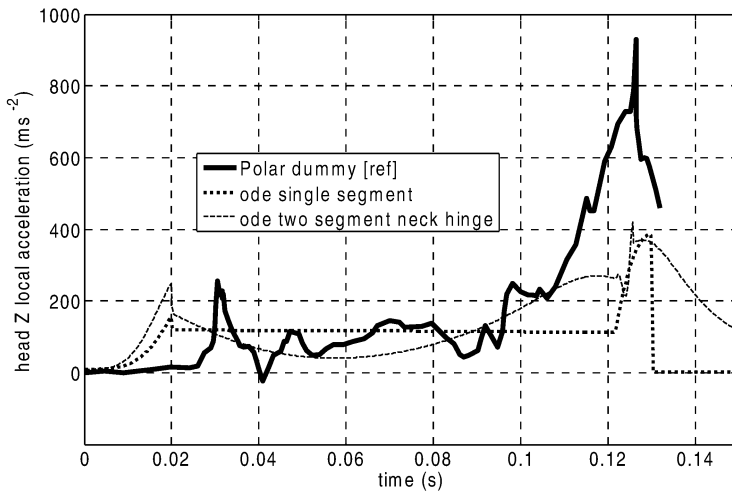


Fig. 7.10 Head component acceleration for Polar dummy impact test [11] and comparisons with one and two segment ODE models.

$$\begin{aligned}
 \vec{r}_{CV} &= \begin{bmatrix} \frac{L_2}{2} \sin \theta_2 & 0 & \frac{L_2}{2} \cos \theta_2 \end{bmatrix}, \\
 \vec{r}_{BV} &= [-Lt_1 \cos(\phi - \theta) & 0 & -Lt_1 \sin(\phi - \theta)], \\
 \vec{F}_B &= [k_B \text{ pen}_B & 0 &], \\
 \vec{F}_C &= [k_C \text{ pen}_C \sin \beta & 0 & k_C \text{ pen}_C \cos \beta], \\
 \vec{F}_D &= [= k_B \text{ pen}_B & 0 & 0], \\
 \phi &= \tan^{-1} \left(\frac{(h + e)}{b} \right), \\
 e &= q - \frac{L_2}{2}, \\
 Lt_1 &= \sqrt{(h + e)^2 + b^2}.
 \end{aligned} \tag{7.17}$$

This system of equations looks complex but can be solved as before using numerical integration. A rotational spring at A with very high values of k_θ will ensure that the response of the two segment model approaches that of the single segment ODE model.

The ability of these simplified ODE models to predict head accelerations can be evaluated by comparison with staged tests. Okamoto et al. [11] published video images and acceleration time histories for the Polar pedestrian dummy impacted by a passenger car. A comparison of the experimental results with the corresponding predictions of the single and two segment ODE pedestrian model are given in Figure 7.10, using a pivot at the neck in this case. Figure 7.11 compares the whole

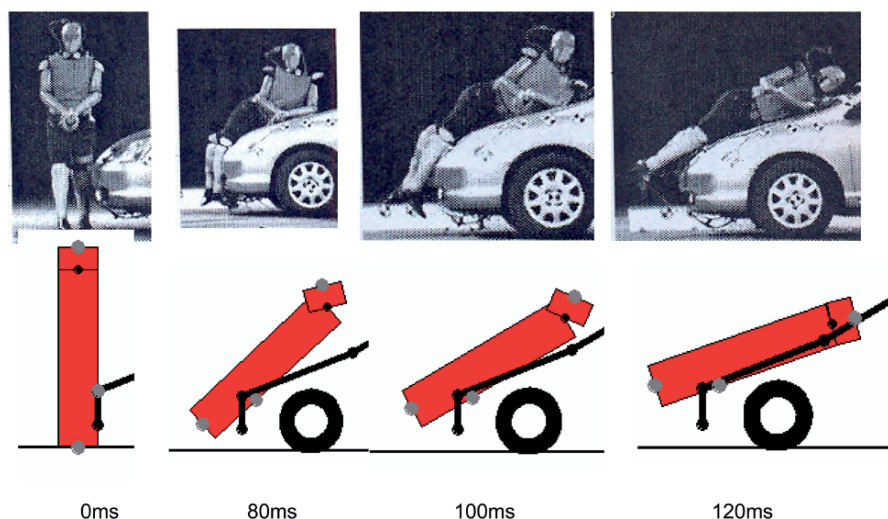


Fig. 7.11 Graphical comparison of Okamoto Polar dummy impact test [11] with ODE two segment prediction.

body kinematics of the staged test with the two segment ODE prediction. Despite the simplicity of the models, good correspondence with the experimental results is evident. In particular, it can be seen that the addition of the pivot at the neck results in substantial improvements in the ability of the model to reproduce head motion in experimental tests. A two segment momentum model with the pivot at the knee has also been used to analyze the effects of bumper height on pedestrian kinematics [12] and with a pivot at the neck to model the range of pedestrian head impact velocities on vehicles [9].

Three-Dimensional Effects

In reality, detailed understanding of vehicle pedestrian impact requires a three-dimensional approach, because there is frequently a transverse motion of the pedestrian relative to the vehicle, and because the leg stance at the instant of impact results in rotation of the pedestrian about his/her longitudinal axis, see Chapter 3. There is a significant increase in mathematical complexity when three-dimensional dynamics and kinematics are considered, mostly because of the nature of three-dimensional rotations (see for example [13]). Commercial codes such as Madymo are three-dimensional, but for the sake of brevity, implementation in 3-D is not included here.

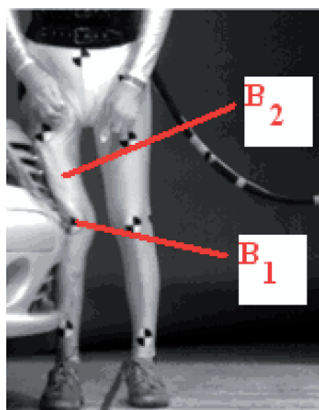


Fig. 7.12 The disadvantages of associated with a rigid body approach become clear when localised loading/deformation is considered, adapted from [14].

Problems with a Rigid Body Approach

The rigid body approach cannot be used to calculate deformations. This is a significant limitation, because local effects around the site of an impact determine the extent of injury. Consider two points B_1 and B_2 on the front surface of the leg, separated by a short distance, see Figure 7.12. Now consider the bumper of a vehicle striking the leg at B_1 without contacting B_2 . It is clear that there would be soft tissue damage at B_1 and probably underlying bone damage as well (see Chapter 2). However, it is unlikely that there would be any soft tissue damage at B_2 as the bumper does not make contact with the body in this region. However, a multi-body representation of this interaction would show only very small differences in the dynamic response between B_1 and B_2 : since both points belong to the same rigid body, kinematic differences are solely due to the rotation of the segment. The amplitude of this is found from the cross product of the position difference vector with the angular velocity vector, and the position difference vector is small (recall B_1 is close to B_2).

A much more realistic representation of impact is possible when local deformations are allowed. This means that although rigid body models are ideal for predicting whole-body dynamics, they are not capable of predicting deformations which are known to be significant in injury causation, see Chapter 5. The most popular method of analysing the deformation of continua is the finite element method. A rigorous derivation of the finite element equations is beyond the scope of this book, and is covered adequately in many textbooks (e.g. [15]). However, in this book, we do briefly introduce the method and demonstrate its application to the mechanics of pedestrian impact.

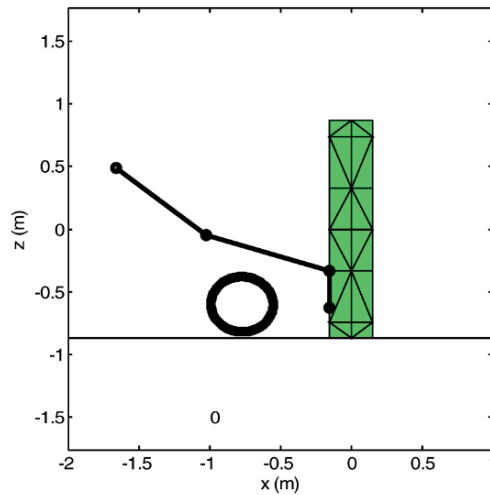


Fig. 7.13 Single segment finite element representation of the pedestrian using triangular elements.

A Finite Element Approach to Pedestrian Impact

In the finite element method, the rigid segment representation of the human body is subdivided into a series of simple geometric shapes, see Figure 7.13. The elements are connected at nodes, and (in this case) the nodes are vertices of triangles.

The governing differential equation is solved simultaneously at each node, and values of the state variables (or functions thereof such as strain) required at locations not represented by a node are found by interpolation using the known nodal values. In this case, the planar kinematics of each node are found by numerical integration of the structural dynamics equation (essentially Newton's Second Law) applied simultaneously at each node

$$m\ddot{x} = F_{\text{ext}} - F_{\text{int}} + F_{\text{body}}, \quad (7.18)$$

where F_{ext} are externally applied forces, F_{int} are the internal forces and F_{body} are the gravity forces. For simplicity, the vehicle is still represented as a rigid body constrained to move only in the X direction. Since each node of the finite element pedestrian has 2 degrees of freedom (planar motion of a particle), there will be $2n$ simultaneous equations for the pedestrian to be solved at each time increment in the numerical integration procedure:

$$\begin{bmatrix} M_{11} & M_{12} & M_{13} & \dots & M_{1n} \\ M_{21} & & & & \\ \vdots & & M & & \\ & & & & \\ M_{n1} & & & & M_{nn} \end{bmatrix} \begin{bmatrix} \ddot{x}_1 \\ \ddot{z}_1 \\ \ddot{x}_2 \\ \dots \\ \ddot{x}_n \\ \ddot{z}_n \end{bmatrix} = \begin{bmatrix} F_{\text{ext}_x1} \\ F_{\text{ext}_z1} \\ F_{\text{ext}_x2} \\ \dots \\ F_{\text{ext}_xn} \\ F_{\text{ext}_zn} \end{bmatrix} - \begin{bmatrix} F_{\text{int}_x1} \\ F_{\text{int}_z1} \\ F_{\text{int}_x2} \\ \dots \\ F_{\text{int}_xn} \\ F_{\text{int}_zn} \end{bmatrix} + \begin{bmatrix} F_{\text{body}_x1} \\ F_{\text{body}_z1} \\ F_{\text{body}_x2} \\ \dots \\ F_{\text{body}_xn} \\ F_{\text{body}_zn} \end{bmatrix}. \quad (7.19)$$

In addition, the X direction equation of motion of the car must be integrated. It will be the same as before:

$$M\ddot{X} = \mu Mg - F_{Bx} - F_{Cx}, \quad (7.20)$$

where the horizontal components of the contact force on the car from the leg (F_B) and the head (F_C) are equal and opposite the forces acting on the pedestrian. The nodal deflections can be used to calculate strains which can be used with injury criteria to yield detailed injury predictions.

In addition to the geometric, inertial and contact characteristic information required for the ODE based model, a finite element formulation requires knowledge of the material properties for all deformable elements. This is a major challenge, since human tissues are mostly nonlinear, anisotropic, viscoelastic and undergo large deformations [16]. Therefore determination of the material properties of human tissues is an important scientific pursuit in itself. The implication of this is that the potential benefits of the finite element pedestrian formulation over a rigid body formulation may be largely nullified if insufficient data relating to the mechanical properties of the tissues are available.

To illustrate the finite element method, simplified linear elastic material properties ($E = 1e9$ MPa) were implemented, and the kinematics predicted by the FE formulation are shown in Figure 7.14. The colours of each element are proportional to the effective (von Mises) stress predicted by the model. In addition, the time histories of the pedestrian segment angular velocity are shown together with the corresponding predictions from the rigid segment ODE model. Since the stiffness of the finite element structure is sufficiently high to suppress large deformations, the resulting motion is almost identical to the rigid body formulation, see Figure 7.4.

The developments in this chapter show that there is a hierarchy in which the finite element formulation can predict the ODE response when the Young's modulus is very high (Figure 7.14), and the ODE model can predict the momentum response when the contact stiffness is very high (Figure 7.9). Therefore, the finite element

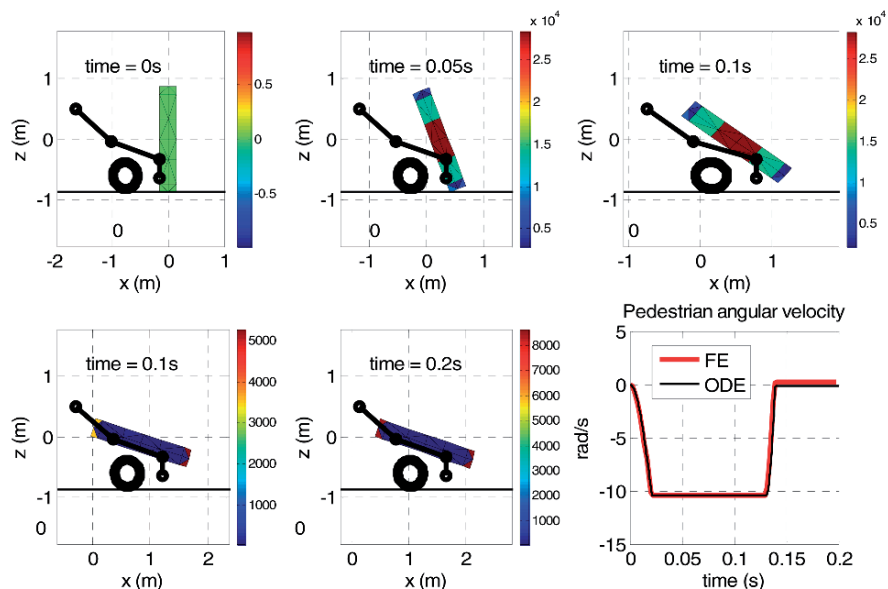


Fig. 7.14 Pedestrian kinematics and effective stress predictions from single segment finite element formulation ($E = 1e9$ MPa).

formulation can clearly be used to predict real-world events such as time to head contact, head impact velocity and contact forces.

However, it is a considerable challenge to assess the ability of a finite element pedestrian model to predict those real-world events which cannot also be predicted by the momentum or rigid body models. Furthermore, variations in material properties greatly alter the stress predictions but may have very little influence on the resulting kinematic predictions, and therefore an FE model of pedestrian impact which is only kinematically validated must not be used for stress predictions. Therefore, since they are unvalidated, the Von Mises stress predictions in Figure 7.14 have very little value. Finite element models predict element strains from nodal displacements, and element stresses from the constitutive laws. In an experimental procedure, the measurement of transient stress or strain is very challenging, and therefore although a finite element pedestrian model can predict stress/strain in the head at say the instant of contact with the windscreen, there is currently no experimental vehicle pedestrian impact data available to which such predictions can be compared. The predictions of finite element models depend strongly on geometry and material properties (viscoelasticity and damping, etc.), but also on element size and shape and boundary conditions. This means that attempts to ‘validate’ a finite element model using kinematics alone are incomplete. It has been shown in this chapter that the kinematics are predominantly a function of the inertial and geometric characteristics, and substantial changes in material properties have only minor influences on overall body kinematics. In contrast, the predicted stress and strain results will

of course be strongly influenced by alterations in material properties. In practice, researchers have attempted to validate their finite element models on a component basis, and these efforts will be treated in further detail in Chapter 8.

Concluding Remarks

This chapter has reviewed the common physical frameworks for analysing the impact interaction between pedestrians and vehicles. The equations of linear and angular impulse and momentum are shown to provide a good prediction of the kinematics of pedestrian motion, but step changes in segment velocities are inherent in this approach. Numerical integration of the equations of motion of rigid/hinged pedestrian representations result in the additional predictions of time-varying contact forces and body segment accelerations which also correspond well to experimental data. Finally, finite element analysis yields additional predictions of localised stress and strain, which are clearly crucial to injury outcome. However, although finite element models can predict body kinematics adequately, the validity of the corresponding stress/strain predictions is difficult to assess experimentally. In particular, it has been shown that the use of kinematic parameters to validate finite element pedestrian models is incomplete. These fundamental considerations provide a framework for interpreting the analyses presented in subsequent chapters.

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Chapter 8

Models for Simulating Impact

Introduction

This chapter provides an overview of the physical and computational models frequently used in the simulation of pedestrian and cyclist accidents. The main strengths and weaknesses of the different approaches are presented and the extent to which each method has been validated using experimental data is assessed. Finally, the state-of-the-art for these methods is presented and a summary of what each can yield is provided. These models form the basis for assessing the influence of vehicle design on pedestrian and cyclist injuries which will be presented in Chapter 10.

Pedestrian Physical Dummy Models

Crash test dummies have long been a successful research and design tool for assessing injury risk to vehicle occupants. However, although the use of a crash test dummy for pedestrian and cyclist research initially appears promising, several difficulties have been found with this approach [1]. One practical problem is the risk of damage to dummy components from very hard contacts. More fundamentally however, as discussed in Chapter 6, the whole body motion of struck pedestrians and cyclists is substantially greater than normally occurs for vehicle occupants, making it more difficult to reproduce motion within kinematic corridors from cadaver tests. Unsurprisingly therefore, the trajectories of early simplified pedestrian dummy prototypes such as the Hybrid II and the Rotationally Symmetric Pedestrian Dummy did not compare well with cadaver tests [2–5]. This is particularly problematic for the analysis of head impact, which is usually the last but most important of the body impacts on the vehicle. Poor overall kinematic biofidelity can result in substantially unrealistic head impact conditions. As a result, the legislative tests presented in Chapter 6 use subsystem components to represent pedestrian body regions in favour of using a whole body dummy.

Nonetheless, in the late 1990s Honda began the development of the Polar dummy [6, 7], which today appears to be the only whole-body physical pedestrian dummy in active use for research purposes, see Figure 8.1. The Polar dummy was designed to be representative of a 50th percentile adult American male and is a modification of the Thor frontal crash test dummy [8]. The principal alterations from the original Thor design were an increase in the contour of the thorax jacket, a reduction of the stiffness of the spine, an altered knee to allow for some lateral bending and truncation of the arms to reduce test variability. Following initial validation tests, the foam thickness used for the flesh was reduced and the stiffness of the ‘skin’ covering and joints was altered to improve the kinematics when compared to cadaver trajectories [9].

A comparison of the Polar II dummy trajectory for lateral impact from a recent small passenger car at 40 km/h showed favourable results, see Figure 8.1 [10]. Further comparison with the cadaver kinematic response corridors also showed a good correspondence, see Figure 8.2. The Pedestrian Dummy Task Group of the Society of Automotive Engineers evaluated the repeatability of the Polar II pedestrian dummy and found very consistent results for the trajectories of the head, the first thoracic vertebra (T1) and the pelvis, as well as the head resultant acceleration. Subsequent modification of the knee joint to increase the lateral bending stiffness resulted in significant improvements in the moment angle response of the isolated dummy knee in valgus bending [11].

The cadaver trajectories in the principal direction of impact are longer than the Polar dummy trajectory, and the time of head strike in the cadaver tests is later than in the dummy tests [10]. These results were at least partly attributed to increased cadaver sliding up the bonnet due to different clothing and significantly different damage patterns measured on the bonnet leading edge for cadavers compared to the dummy tests. These differences result in discrepancies in the head impact velocities for the dummy compared to cadavers, see Figure 8.3. Furthermore, significant difficulties were found in precise positioning of the cadavers due to their compliance and also due to soft tissue creep. Recognising that very small changes in initial conditions significantly influence overall kinematics [12], this poses a significant technical difficulty in attempting precise comparisons between dummy and cadaver kinematics. For example, orientation of the dummy about the vertical axis between 0 and 100 degrees resulted in a 40% change in head impact speed, and a 300% change in HIC score [13]. Furthermore, lack of muscle tension in the cadavers was identified as a potential cause of poor correlation of body segment trajectories.

Mathematical Models

Mathematical models of vehicle pedestrian impacts have been formulated since the 1970s. There are two main approaches which are currently followed: multibody models in which body segments are assumed to be rigid, and finite element models in which body segments are assumed to deform according to the constitutive proper-



Fig. 8.1 Comparison Polar II pedestrian dummy (left) and cadaver (right) kinematics a 40 km/h lateral impact with a recent small passenger car, adapted from [10].

ties of the component materials. In some cases, these two approaches are combined to take advantage of the speed of multibody computations as well as the detail avail-

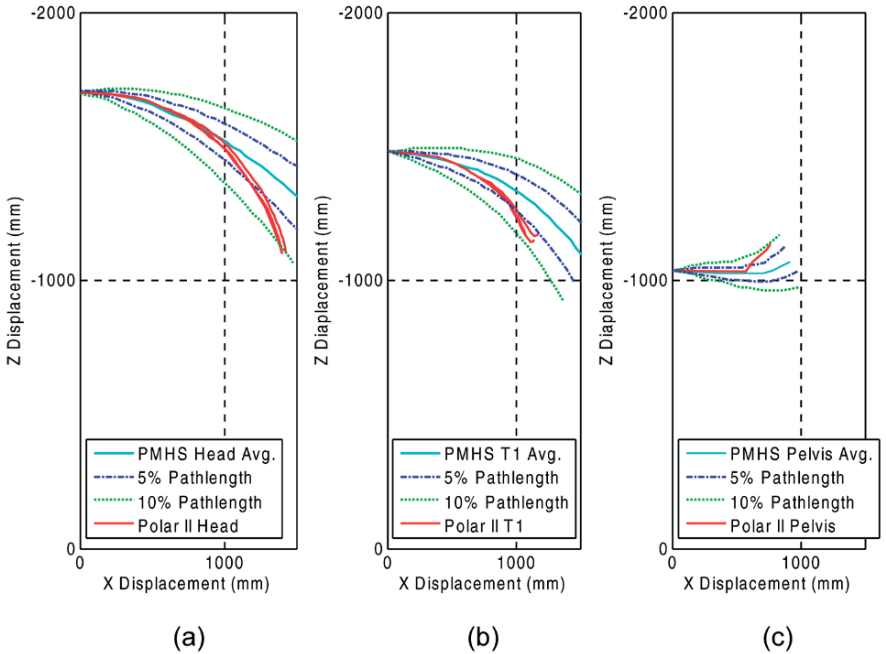


Fig. 8.2 Polar dummy trajectory comparisons to cadaver corridors [12] from lateral impact of a recent passenger car at 40 km/h with cadavers for (a) the head, (b) T1 and (c) the pelvis, adapted from [10].

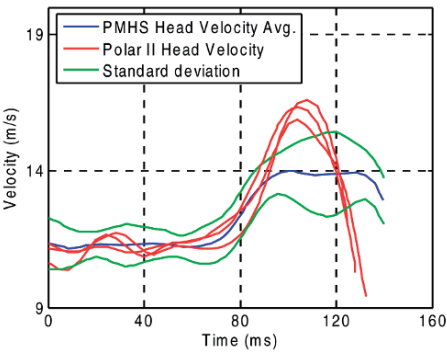


Fig. 8.3 Polar dummy head velocity comparison to cadaver corridors [12] from lateral impact of a recent passenger car at 40 km/h with cadavers, adapted from [10].

able in finite element models. Finite element models are clearly an attractive means to derive injury parameters, but their complexity has three disadvantages. The first is that significant computational effort is required so that even with modern computer power long simulation times are required. The second is that these models require a large number of input parameters which are frequently not very well known.

Thirdly, the complexity of these models can mask fundamental relationships immediately evident with simpler models. This problem is compounded by the large amount of output data created by detailed models, especially when interpreted by less experienced users. A more detailed treatment of the physics underpinning these methods and their respective advantages and pitfalls is given in Chapter 7.

Multibody Models

Multibody models are based on the assumption that segments of a system can be considered as rigid and that kinematic joints constrain the relative motion between segments. Lumped parameter joint stiffnesses can be applied to approximate the bio-mechanical constraints imposed by ligaments and muscles, and penetration search algorithms can detect contact between surfaces attached to different rigid bodies. These kinds of models have been applied to the analysis of pedestrian impacts since the mid-1970s, e.g. [14, 15]. Since then, their complexity has increased substantially with many more segments used to represent the articulations of the human body, generalisation from a 2D to a 3D representation, increased biofidelity of the inertial, stiffness and contact characteristics and more and higher quality experimental data (cadaver kinematic corridors, etc.) are now available for validation purposes.

The development of multibody crash simulation codes from first principles is time-consuming and the popular use of crash simulations in both industry and research has been greatly facilitated by commercial software. The most important multibody crash simulation software today is Madymo¹ developed by the Dutch research Corporation TNO. A particular strength of Madymo is the availability of state-of-the-art crash simulation models, including a family of pedestrian surrogates. A brief description of the Madymo pedestrian model is given here, but other models also developed using the Madymo software [16, 17] or using other commercial packages [18–20] follow a very similar formulation.

The first published Madymo pedestrian model [21] was planar and consisted of only two segments. The current model is three-dimensional with 52 rigid bodies representing the skeletal system and 64 ellipsoids representing the outer surface of the body based on anthropometry of an average Western European male derived using the Ramsis software [22]. The inertia properties are derived from Schneider et al. [23] and the structure of the kinematic joints connecting the rigid body segments is shown in Figure 8.4 [24]. To represent a range of anthropometries, a three-year-old and a six-year-old child as well as a 5th percentile female, a 50th percentile male and a 95th percentile male pedestrian model are available, see Figure 8.4b. The child model anthropometry is based on the Q child dummy [25], whereas the adult anthropometries are based on Ramsis [22].

Lumped parameter joint stiffnesses are based on a combination of volunteer range-of-motion data, cadaver impact tests and simulation results from a Madymo

¹ Madymo: MAtheMatical DYnamic MOdel TNO.

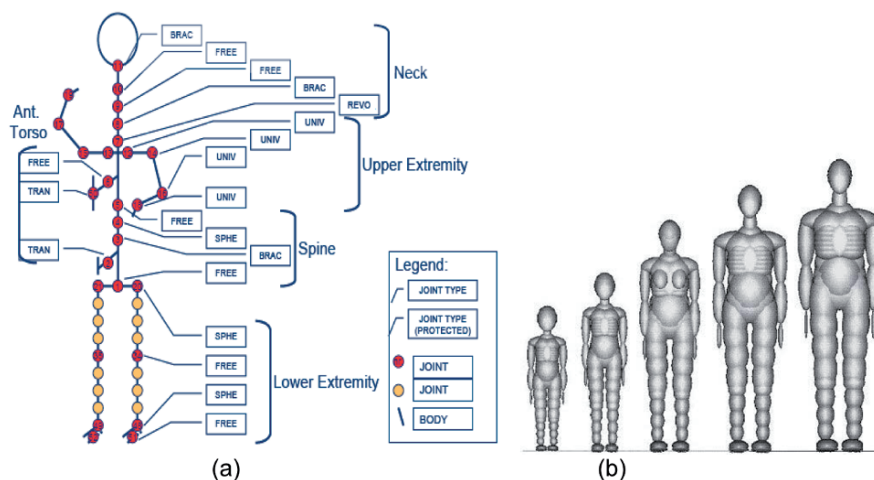


Fig. 8.4 (a) Structure of Madymo multibody model: BRAC = rigid; UNIV = universal; SPHE = spherical; TRAN = translational; REVO = revolute. (b) Madymo pedestrian model family, adapted from [24].

vehicle occupant model [26–35]. Angular stiffness functions for leg bending were derived from quasi-static bending tests on cadaver specimens [36] and fracture of the femur and tibia is facilitated via frangible joints.

The contact characteristics are a very important component of a pedestrian impact simulation. The stiffness of the contact determines the peak force, while the hysteresis loop determines the degree of energy absorption and is critical for accurate kinematic predictions. In Madymo the contact stiffness is defined via user supplied loading/unloading functions, hysteresis paths and damping. In the Madymo pedestrian model these are based on published literature, but they have been optimised to improve results for a variety of impactor test configurations [37], see Figure 8.5.

Contact stiffness depends on the properties of both halves of a contact pair. In cases where one contacting surface is much stiffer than the other, the effective stiffness is similar to that of the more compliant surface. The loading and unloading functions and the degree of hysteresis can be estimated from acceleration measurement of rigid free flight impactor tests if the impactor travels perpendicular to the contact surface and no rotation occurs, see Figure 8.6. These tests need to be performed in several places on the vehicle in order to gain a stiffness map for accurate vehicle contact simulation [39]. Recently, stiffness corridors for multibody model contacts have been measured for the European car fleet based on 425 EuroNCAP tests [40].

The kinematic validity of the Madymo pedestrian model can be assessed by comparison with cadaver tests, Figure 8.7. The trajectories of the head, pelvis, knee and foot show that there is a very close correspondence between the simulation and

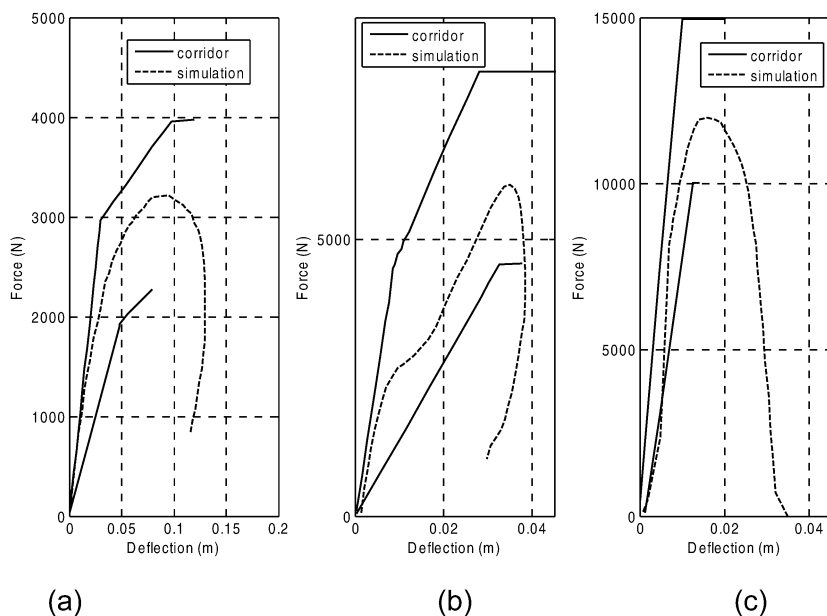


Fig. 8.5 Madymo pedestrian model simulations for contact stiffness optimisation: impactor mass 23.4 kg (a) thorax impact at 6.5 m/s, (b) abdomen impact at 9.4 m/s and (c) pelvis impact at 9.8 m/s [38], adapted from [37].

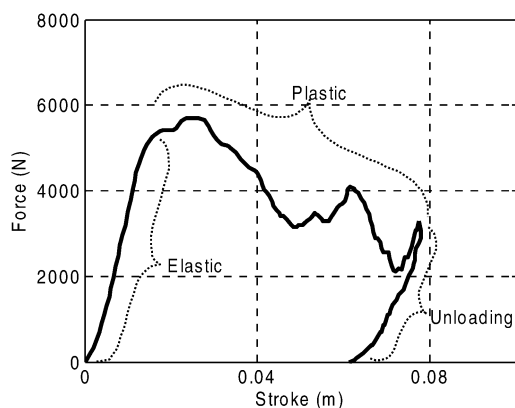


Fig. 8.6 Typical dynamic loading and unloading properties, adapted from [24].

the staged tests, see Figure 8.8. In a reconstruction of 18 cadaver tests using the Madymo pedestrian model, the predicted head impact location was quantified (time and magnitude of the peak) and found to be over 90% accurate [37]. A different multibody pedestrian model also formulated in Madymo showed good prediction

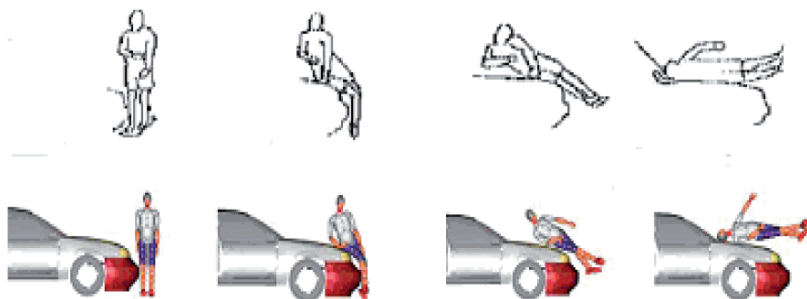


Fig. 8.7 Kinematic comparison of Madymo multibody pedestrian model with cadaver tests, adapted from [43].

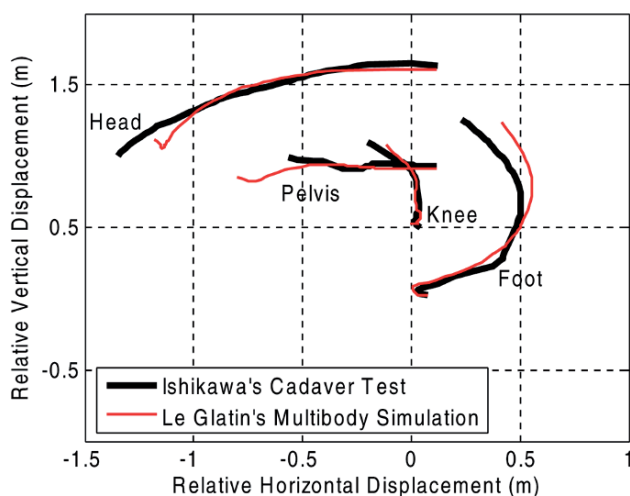


Fig. 8.8 Trajectories of the head, pelvis knee and foot for the Madymo pedestrian model compared to a cadaver test, adapted from [43].

of bumper force time-history and head resultant acceleration by comparison with cadaver tests [16], see Figure 8.9.

Other researchers have used real world accident reconstructions to assess model validity. However, the degree of uncertainty associated with pedestrian accident reconstruction [41] means that their use for pedestrian model validation also has limitations. For example, when the Madymo pedestrian model was used to reconstruct real-world accidents, the main parameter affecting the HIC value was the pedestrian orientation relative to the vehicle at impact [42], and yet this information can rarely be established with confidence following an accident.

Multibody modeling has also shown that the height of the bonnet leading edge, and the bumper height and lead and the length of the bonnet determine the gross

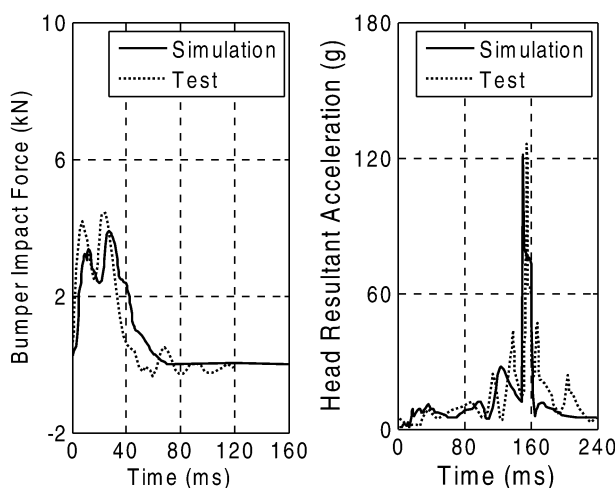


Fig. 8.9 Time history comparison of bumper and head resultant acceleration for a Madymo pedestrian model with cadaver tests, adapted from [16].

motion of the pedestrian. In contrast, the deformation properties of the vehicle front exhibit only a secondary effect on the motion pattern itself [44, 45]. The results of multibody modeling have been frequently used to assess the influence of vehicle design on pedestrian injury, see Chapter 10. Multibody modeling of cyclist impacts using Madymo also showed good ability of the model to reproduce cadaver kinematics for a 50 km/h impact of a Peugeot 205 against a side-facing cadaver [46], see Figure 8.10.

Finite Element Models

In the finite element method the assumption of rigid body motion is relaxed and the domain of interest is discretised into simple geometric shapes or ‘elements’ which can deform under load according to an assigned constitutive relationship between stress and strain known as the material law, see Chapter 7. For impact analysis, the explicit integration method is commonly used, where the displacement of the nodes (or vertices) of each element are found by numerical integration of the nodal accelerations at each time increment which are found from Newton’s Second Law. This method allows the prediction of stresses and deformations in the body resulting from a given set of boundary conditions. This permits not only simulation of long bone bending, but also the idealised kinematic joints generally used in multibody models can be replaced by deforming contact surfaces constrained by ligament and muscle elements which have a fully three-dimensional representation. This is clearly

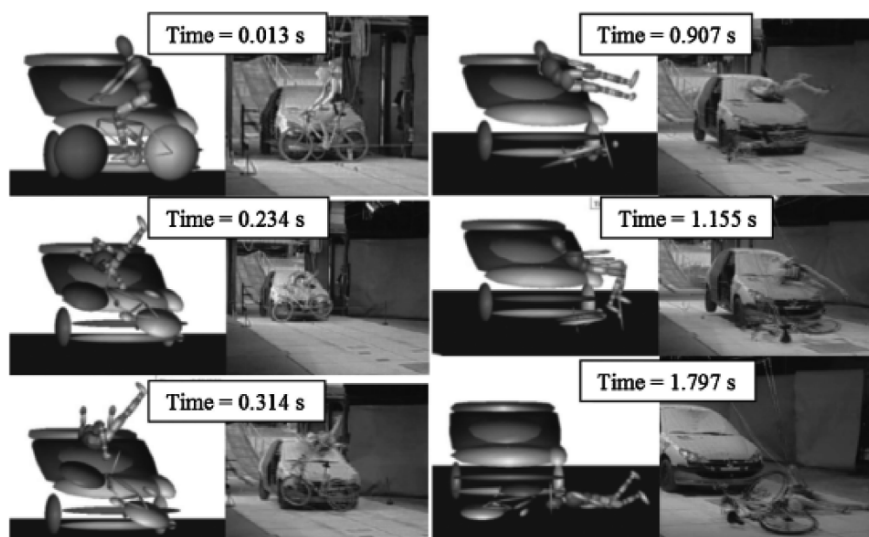


Fig. 8.10 Multibody response of a cyclist impact using Madymo compared to cadaver tests for a 50 km/h impact of a Peugeot 205 against a side-facing cadaver [46].

a very powerful technique, and development of finite element models has become the principal focus of pedestrian impact research in recent years.

The ‘accuracy’ of a finite element model for pedestrian or cyclist impact depends fundamentally on the geometric description of the pedestrian and the vehicle, application of suitable material laws and implementation of contact algorithms which faithfully reproduce the interactions between contacting segments. The issues relating to contact interactions are in general similar to those previously discussed for multibody models. With regard to model geometry, dramatic improvements in medical imaging techniques and initiatives like the Visible Human Project [47] have resulted in the availability of high definition cross-sectional images of the human body. As a result recent finite element models reconstruct the principal skeletal tissues of the body with great geometrical accuracy. In contrast, the development of material laws to represent the constitutive behaviour of biological tissue has been much more difficult. The mechanical behaviour of most biological tissues is nonlinear elastic, anisotropic and viscoelastic² as well as being age, sex and species dependent [48, 49]. Soft tissues like muscles and ligaments are capable of large deformations and present particular challenges for finite element representation. Furthermore, there are significant differences between the tensile and compressive properties for many biological tissues and, together with the need for damage algorithms, these complexities mean there is a real risk of incorrect predictions of tissue stress/strain resulting from simplifications such as the frequent assumption of linear elastic isotropy for the behaviour of human soft tissues. Nonetheless, significant advances are being

² Strain rate dependent behaviour as well as hysteresis and relaxation effects.

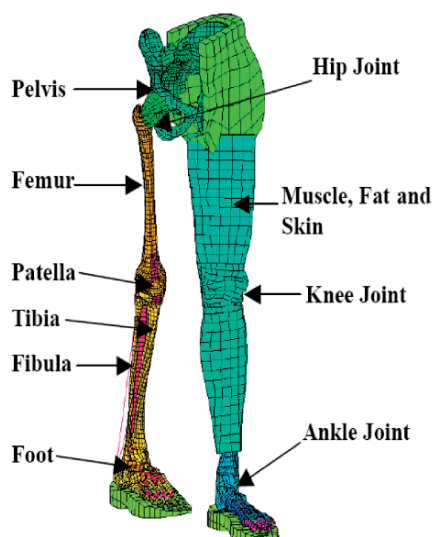


Fig. 8.11 The Toyota THUMS leg model, adapted from [56].

made in this active research area,³ and the material laws applied to soft tissues in recent finite element pedestrian models are generally far more suitable than those used only ten years ago.

Finite element models for application in pedestrian (and more recently in cyclist) impact analysis have been developed since the mid 1990s [50–54]. These early formulations mostly featured simplified geometries for the long bones, one-dimensional force models for ligaments, idealised kinematic joint formulations, linear elastic material properties and limited attempts were made at model validation [55]. Today the three main finite element models developed for pedestrian impact research are the Total HUMAN Model for Safety (THUMS) developed by Toyota [56] (Figure 8.11), the lower limb finite element model developed by Honda [57] (Figure 8.12) and the Lower Limb Model for Safety (LLMS) developed at INRETS [58]. There are also a number of finite element head models [59, 60] that have been used in the reconstruction of pedestrian accidents for the development of head injury criteria that include brain deformations, see Chapter 5.

The skeletal structures in recent finite element pedestrian models are generally modelled using shell elements for cortical bone and solid elements for cancellous bone and bone yield stress is included. Ligaments and tendons are modelled using elements ranging from one dimensional force models to solid brick elements. Muscles and fat are mostly modelled using solid elements. The compressive properties of soft tissues influence the accuracy of bone fracture predictions due to their energy absorption capabilities and quasilinear and non-linear viscoelastic material

³ Elsevier has recently started a new journal dedicated to this field: *Journal of the Mechanical Behaviour of Biomedical Materials*.

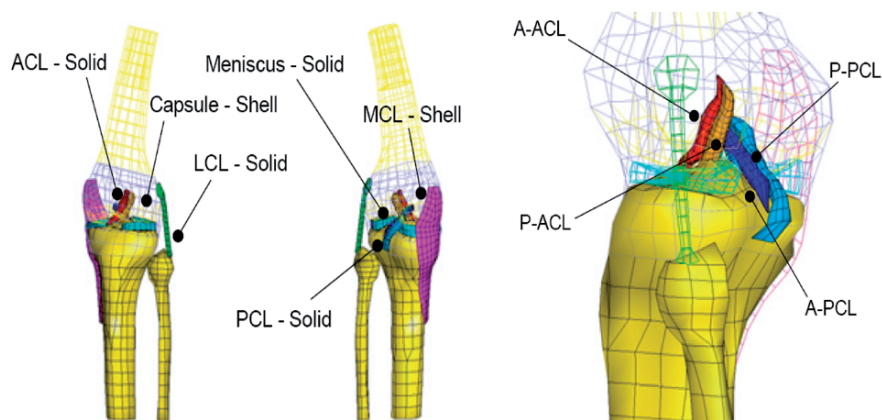


Fig. 8.12 Details of the knee joint in Honda FE pedestrian model, adapted from [57].

models have been developed for knee ligament, flesh and fat [57, 61]. Other researchers have used an elasto-plastic model for the knee ligaments, with strain-rate dependent characteristics including ultimate strength and material constants derived from tests on rabbit ligaments at various strain rates [62, 63]. However, the biphasic viscoelastic nature of articular cartilage remains greatly simplified by the use of a linear elastic material law [64], and the bones of the feet have been modelled as rigid. Similarly, the upper body of the Honda pedestrian model consists of rigid body sections [55], and the muscle representation in existing models generally do not attempt to reproduce the known non-linear, anisotropic viscoelastic behaviour of muscle tissue in compression [48, 65].

The ‘validation’ of finite element models for pedestrian impact presents significant difficulties. Considering the hierarchy of modelling approaches from conservation of momentum to multibody dynamics to finite element models, the complexity of finite element models is mainly justified by the ability of this method to provide stress and strain information which cannot be provided by either of the two less complex approaches, see Chapter 7. Therefore, reproducing whole body kinematics or even rigid body dynamic quantities such as force/moment time histories or linear accelerations are necessary but not sufficient conditions for validation of a finite element model. Rather, to validate a finite element model as a tool which can tell us more about pedestrian or cyclist impact than a multibody model can, the finite element model should be able to predict some experimental stress or strain measurements. To avoid the risks associated with extrapolation, the experimental tests used for model validation should be performed at the upper and lower limits of the impact severities of interest, thereby allowing the ‘validated’ model to be used for interpolation purposes at intermediate severity levels. Furthermore, due to the large number of degrees of freedom in finite element models, the input parameters (material constants etc) should not be based on the same experimental data used to validate the model. It will be seen that even the most sophisticated finite element pedestrian

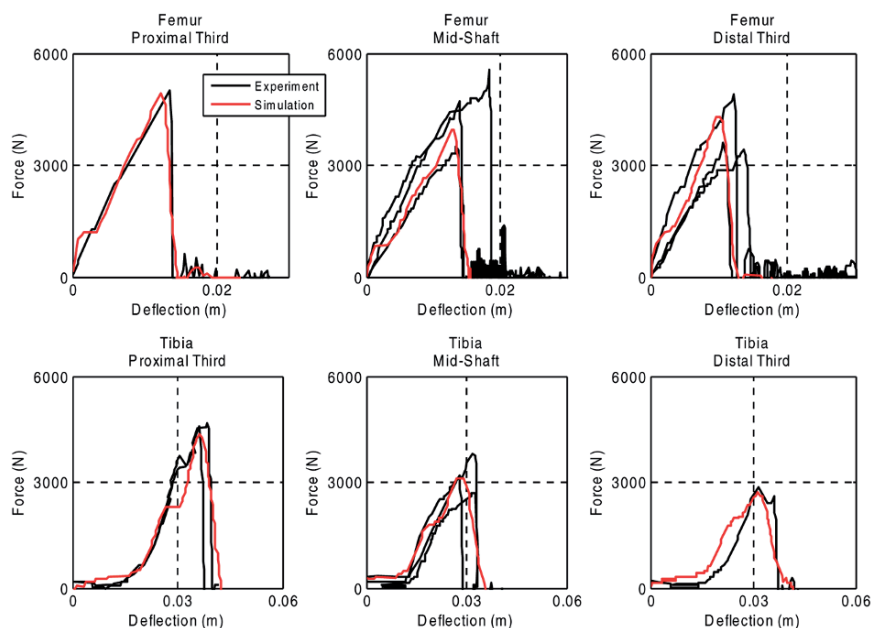


Fig. 8.13 Honda pedestrian model predictions of 3 point dynamic bending tests on the long bones of the lower extremity, adapted from [57].

models in the literature struggle with this definition of validation, due to the lack of appropriate experimental data. Therefore, pedestrian and cyclist experiments recording tissue strains in human subjects are urgently needed for further validation of finite element models [55].

Commencing with component level validation, good correspondence has been achieved for finite element predictions of isolated long bone force-deflection behaviour compared with experimental tests performed at loading rates of 1.5 m/s, see Figure 8.13. Similarly, predicted isolated knee ligament force-deflection behaviour compared quite well to experimental tests on isolated knee ligament tissues [57], see Figure 8.14.

Partial validation of the material laws used for the knee ligaments in the LLMS pedestrian model [64] has been achieved by comparison of the isolated knee ligament model with cyclic tests performed on canine medial collateral ligaments [66]. However, uniaxial tests are generally insufficient to characterise three-dimensional behaviour, and more complete validation would involve correlation of predicted tissue strain in the knee with experimental measurements in cadavers.

The detailed mass distribution of finite element models and the facility for long bone as well as soft tissue deformation mean that the whole body kinematics can be very well captured using finite element models, see Figure 8.15.

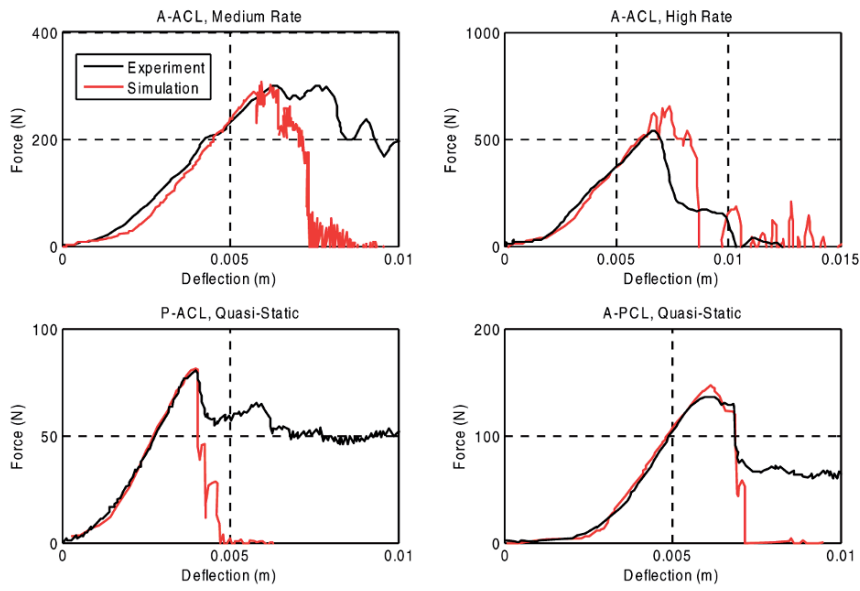


Fig. 8.14 Honda pedestrian model predictions of ligament force-deflection behaviour compared to experimental tests, adapted from [57].

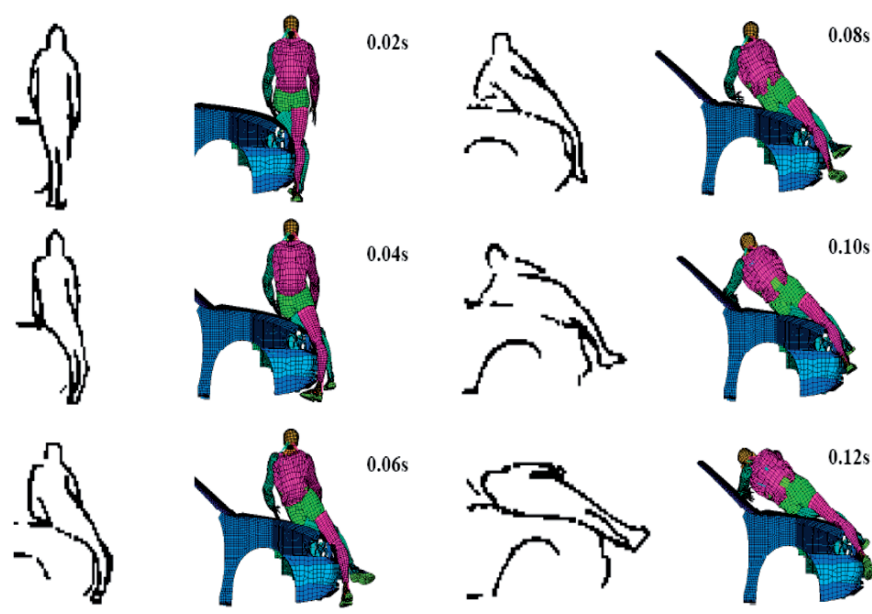


Fig. 8.15 Comparison of the THUMS FE model [56] with cadaver impact tests, adapted from [67].

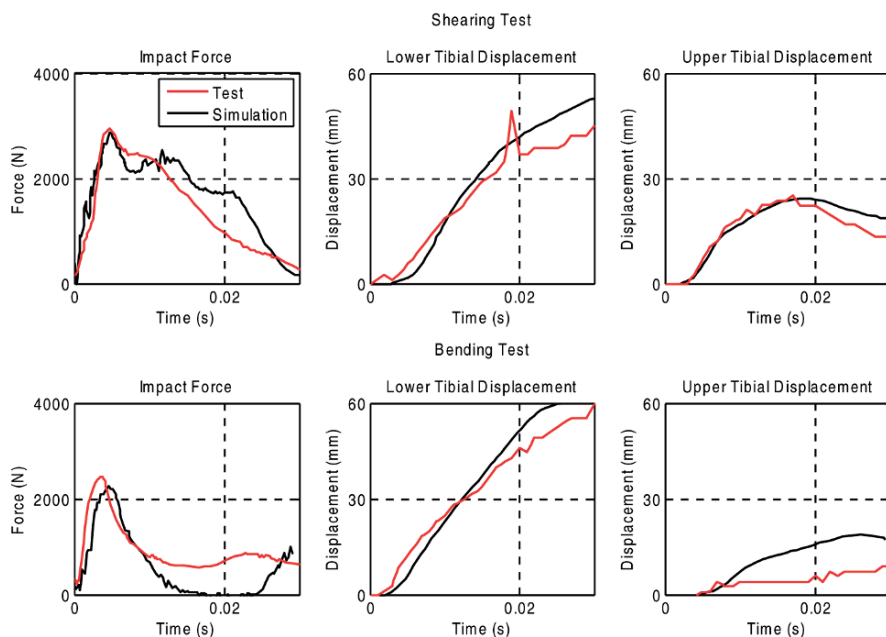


Fig. 8.16 Comparison of the THUMS FE model [56] with cadaver impact tests for shear and bending of the knee joint, adapted from [29, 68].

A comparison of the THUMS model with experimental knee shear and bending tests [29, 68] performed showed very good results for impact force and lower tibia displacement, but poorer correspondence for upper tibia displacement in the bending test, see Figure 8.16.

Mixed results were found in a comparison between the THUMS model predictions and cadaver tests where the cortical strain in the femoral shaft and the superior ramus was recorded using strain gauges [69], see Figure 8.17. However, using a fracture threshold of $2500\mu\epsilon$ [70] the simulations correctly predicted no femoral shaft fracture in all cases. For the superior ramus, fracture occurred in three of the five cases, and although this could not be reliably predicted with the model,⁴ the predicted strain magnitudes are similar to those measured. It appears that these are the only available data comparing finite element tissue strains with experimentally measured tissue strain in cadavers for pedestrian impact.

⁴ Bone quality was considered to be responsible for the experimental variation.

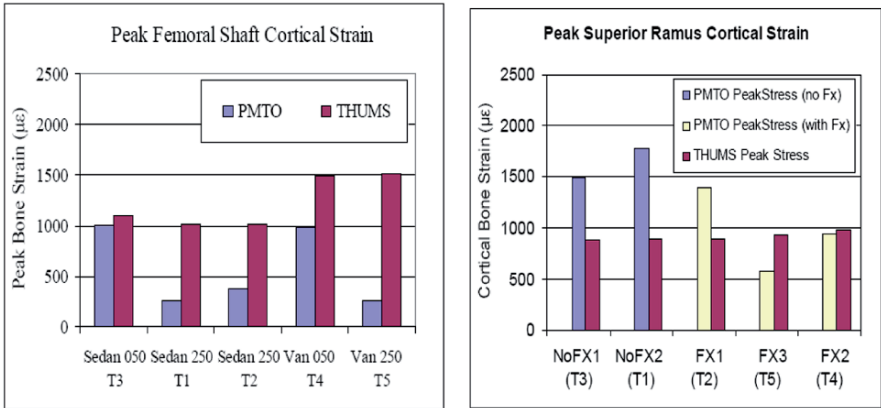


Fig. 8.17 Comparison of cortical bone strain in the femoral shaft and the superior ramus from cadaver tests with THUMS model predictions [69].

Application of Finite Element Pedestrian and Cyclist Models

Despite the problems with validation, finite element modeling using the LLMS has been used to simulate impact tests on isolated cadaver legs. Comparison of the model predictions with observed knee injuries led to the proposal of 16° of lateral bending angle and a shearing displacement of 15 mm as soft tissue injury tolerance levels for the knee [71]. Finite element modeling of active muscle forces also showed that knee ligament strains and knee shearing displacement and bending angles reduce when the leg muscles are active, thereby reducing knee ligament injury risk [72, 73]. It has also been predicted that the strain in the medial collateral ligament for a 50th percentile male struck by a passenger car depends on the lower bumper stiffness, while for the 5th percentile female and six-year-old child the collateral ligament strain depends on the upper bumper stiffness [74]. Finite element modeling of cyclist impacts at 32 km/h has also been performed [54, 75], see Figure 8.18. Comparison of global trajectories and accelerations were good, and the model was used to predict strain levels in the knee ligaments for a variety of input conditions. However, tissue level strain validation data was not available and the accuracy of the finite element predictions is therefore unknown.

Concluding Remarks

Given the importance of legislation as a driving force in effecting vehicle design changes (see Chapter 6), and the overwhelming preference of legislators to use physical rather than virtual tests in the assessment of production vehicles, the development of a suitable pedestrian dummy has clear attractions. The focus to date has

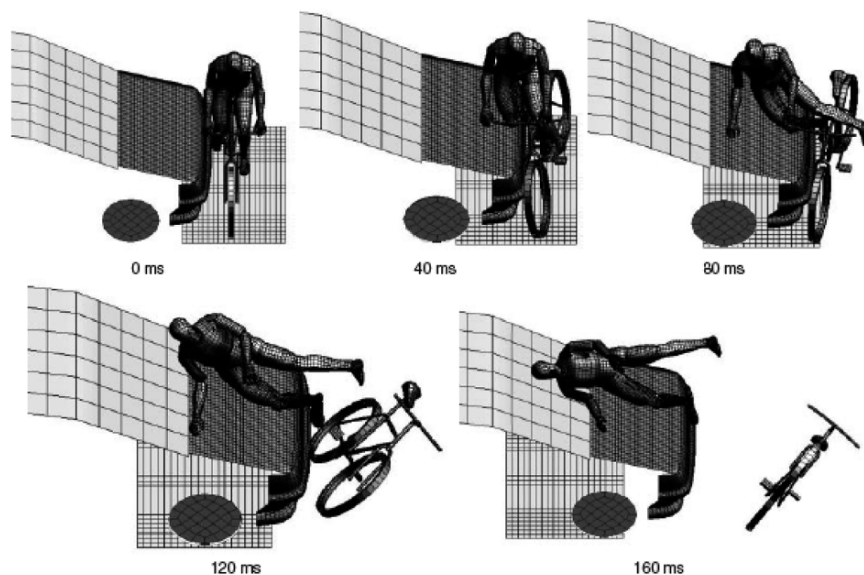


Fig. 8.18 Finite element modeling of cyclist impacts using LLMS FE leg model and Hybrid III whole body model [54].

been on improving the ability of the Polar pedestrian dummy to reproduce cadaver impact trajectories and head velocity time-histories. While a good dummy trajectory is clearly a prerequisite, validation of parameters such as head acceleration and knee shear force/bending moment used in injury criteria evaluation is a necessary next step. Considering the simplifications required, physical dummy testing cannot predict tissue stresses/strains, and therefore injury predictions using a dummy are limited to rigid body parameters. Nonetheless, a pedestrian dummy which reproduces not only the movement but also the principal dynamics of pedestrian and cyclist impacts would be a powerful tool to assess the influence of speed, shape and stiffness on pedestrian injuries as well as the effectiveness of proposed injury countermeasures. Such a dummy could also be used in the validation of multibody computer simulations and in the refinement of subsystem component tests [12]. To a large extent, the Polar dummy fulfills these requirements.

Multibody pedestrian and cyclist simulation models have the potential to provide similar information as a pedestrian dummy. The assumption of rigidity rules out predictions of tissue stress/strain, but the well established relationships between rigid body parameters and injury (see Chapter 5) mean that a multibody model subjected to the same validation process as a pedestrian dummy (ie reproducing both kinematic and dynamic quantities measured using cadavers) can also be used to establish the influence of speed, shape and stiffness on pedestrian and cyclist injuries. Furthermore, multibody models are inherently suitable for parametric tests which can be impractical to perform experimentally. Therefore, a validated multibody ped-

estrian model can be used to establish the effect of changes in vehicle shape or stiffness or the influence of variations in bone strength related to age and sex on overall injury predictions. Furthermore, considering the broad range of human anthropometrics, scaling techniques have been used to propose multibody models representing a range of statures [37]. However, validation of these scaled models requires additional experimental data.

Finite element models of pedestrian and cyclist impact have the potential to predict the influence of vehicle speed, shape and stiffness on tissue stress/strain, but so far validation of the tissue level behaviour of finite element models is very limited. Nonetheless, the work of Snedeker et al in this respect is promising [69], and reconstruction by Marjoux et al [76] of over 60 real-world accidents using finite element head models validated at the tissue strain level has led to the proposal of new strain based brain injury criteria which are a major advance on the Head Injury Criterion [77], see Chapter 5.

Bose et al. have proposed knee ligament failure thresholds by simulating impact tests on isolated knee joints using the LLMS [78]. By reproducing the global force/moment responses and by assessing injuries during necropsy, partial avulsion of the medial collateral ligament at 15° lateral knee bending and 105 Nm bending moment were predicted. Furthermore, by using estimated strain failure thresholds for the collateral (28%) and cruciate (22%) ligaments, failure of both the medial collateral and posterior cruciate ligaments were predicted, though necropsy results revealed only medial collateral ligament failure. The model predicted a non-homogeneous strain distribution in the posterior cruciate ligament, and the authors argued the importance of considering local strain levels for predicting tissue failure. However, although these predictions are plausible and advance our knowledge of knee injuries, there remains a lack of validation of the tissue strains for the whole knee joint and the development of the required experimental procedure will be a technically challenging but valuable process.

Ultimately, finite element models can be used as a design and assessment tool for pedestrian and cyclist safety and aspects such as the interaction between bone fracture and ligament failure have been shown using the Honda finite element pedestrian model [55], where a failure of one tissue substantially reduced the likelihood of failure in the other. The parametric approach to assessing design variables described for multibody models can clearly also be applied to finite element models, though the computational effort required is much greater. Similarly, scaling techniques to represent different pedestrian statures can be performed [79, 98]. However, significant additional experimental data is required for validation since scaling to represent different ages necessitates the provision of age dependent material laws for the biological tissues, and these are poorly defined for soft tissues.

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Chapter 9

Ground Contact Injuries

Introduction

Chapter 3 showed that most vehicle pedestrian and cyclist collisions terminate following pedestrian or cyclist contact with the ground, and the latter frequently leads to significant injuries, with about 8% of serious or fatal pedestrian head injuries attributed to ground contact [1]. While there have been no direct studies of cyclist ground contact kinematics, and cyclists are not included in the IHRA dataset, some smaller in-depth studies have studied cyclist injuries from ground contact: the European in-depth pedestrian database (APROSYS) has indicated that ground contact is especially important for cyclists, though only seven cyclist cases were included [2]. In the absence of direct studies of differences between pedestrian and cyclist ground contact kinematics, this chapter focuses on pedestrian ground contact.

The process of ground contact for both pedestrians and cyclists involves a combination of slide, roll and bounce to rest, see Figure 9.1, and frequently leads to further significant injuries [3, 4]. The overall injury severity is largely determined by the head impact [4, 5] and, although pelvis fractures from ground contact have been reported [6], thorax and abdomen injuries are much less likely from ground impact than from vehicle impact [7].

The IHRA review of pedestrian accidents presented in Chapter 2 showed that contacts with the ground account for 20–25% of head, face and arm injuries, but only a low proportion of midbody or lower extremity injuries [9]. A review of the Pedestrian Crash Data Study (PCDS) from the US found that 19% of injuries to pedestrians from all vehicle types resulted from contact with the ground [7]. Finally, detailed On The Spot analysis (OTS) of real accidents in Britain found over 30% of both non-minor and fatal injuries were attributed to ground contact, see Figure 9.2 [10].

Pedestrian/ground contact injuries are therefore an important consideration, and in this chapter we address the relative severity of vehicle versus ground impacts, the high degree of variability associated with the ground contact and the influence of



Fig. 9.1 Post head impact kinematics in a staged test showing slide, roll and bounce in ground contact, adapted from [8].

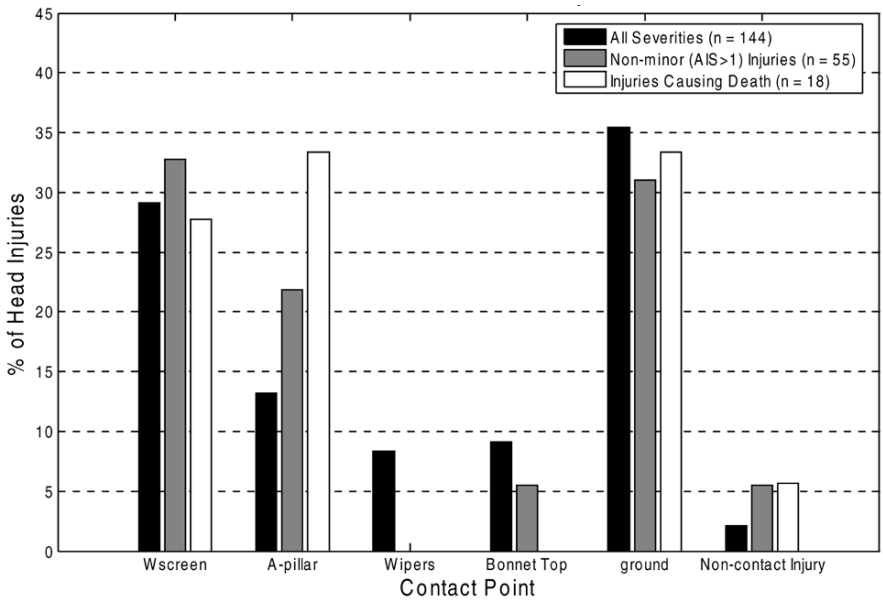


Fig. 9.2 Principal causes of pedestrian head injuries in UK on-the-spot accidents, adapted from [10].

vehicle impact speed and vehicle design on subsequent pedestrian ground contact injuries. Finally, some suggestions for reducing ground contact injuries are presented.

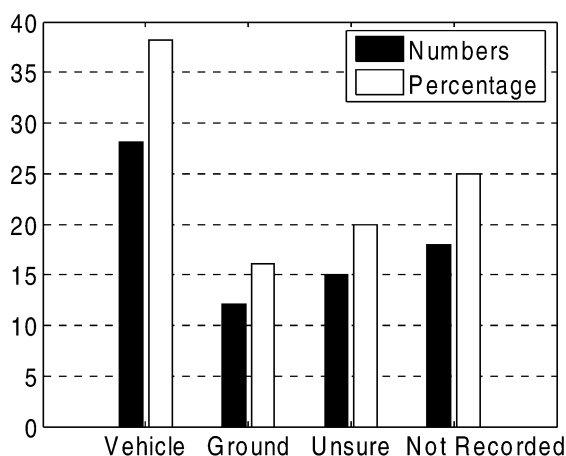


Fig. 9.3 APROSYS accident data showing high proportion of cases where source of injury (vehicle or ground) could not be ascertained, adapted from [2].

Relative Severity of Ground versus Vehicle Impact

Assessing the severity of injuries from ground contact relative to those from vehicle contact helps to define the practical limits for pedestrian protection that can be achieved by improved vehicle design. However, this is a difficult task and it is often even difficult to establish whether injury was caused by vehicle or by ground contact [2, 4]. For example, the APROSYS database showed that in 45% of cases the source of injury was either not recorded or could not be ascertained, see Figure 9.3 [2].

Nonetheless, the general severity of ground contact injuries is consistently reported as lower than for vehicle contact injuries [1, 2, 4, 9, 11–16]. In the Pedestrian Crash Data Study (PCDS), the ground was responsible for 523 out of 2180 injuries [15, 17]. However, 93% of injuries from ground impact were at minor AIS levels, and serious injuries were more likely to be caused by the vehicle [15]. Similarly, analysis of German accident data showed that, while 65% of pedestrians are injured by road impact, primary (or vehicle) impact caused twice as many injuries as secondary (or ground) impact [4]. Further analysis showed that 38% of pedestrians suffered higher injury severity from primary impact, 50% suffered the same injury severity from primary and secondary impact and 12% of pedestrians suffered higher injury severity due to the secondary impact [4]. Table 9.1 shows that the percentage of AIS2+ injuries to each body region from the primary impact is somewhat higher than from the secondary impact. For example 41% of all head injuries from secondary impact were AIS2+, compared to 50% of head injuries due to primary impact [4].

French accident data showed about 8% of serious or fatal head injuries result from ground contact compared to vehicle contact [1]. An in-depth study of 77 cases

Table 9.1 Percentage of AIS2+ injuries to various body regions which occur in primary and secondary impact, adapted from [4].

	Primary impact	Secondary impact
Head	50%	41%
Chest	39%	27%
Abdomen	40%	24%
Pelvis	32%	21%
Legs	38%	30%

Table 9.2 APROSYS real accident data showing the relative importance of ground and vehicle related pedestrian injuries, adapted from [2].

Pedestrians			
Injury severity	Primary impact (%)	Secondary impact (%)	Ratio Primary/Secondary
All	68	16	4.3
AIS1	68	24	2.8
AIS2+	79	9	8.8
Serious All	62	22	2.8
Fatal All	80	9	8.9
Child All	45	21	2.1
Elderly All	78	12	6.5
Elderly AIS1	34	11	3.1
Elderly AIS2+	79	3	26.3
Cyclists			
All	48	42	1.1

in Australia found that the road was the most common cause of head injuries, but it was not the cause of any AIS3+ head injuries [11]. Similarly, the APROSYS database showed that 36% of pedestrians received a head injury from secondary impact compared to 43% from primary impact, and a greater proportion of the latter were AIS2+ [2]. The ratio of primary to secondary injury percentages in Table 9.2 shows the overall dominance of primary injuries over secondary injuries in the APROSYS dataset. However, a case-by-case analysis showed that the severity of the secondary impact is often the same as for the primary impact, and the primary impact is particularly important for elderly people, while secondary impact appears more significant for children [2].

Variability of Ground Contact

Some of the accident findings presented above are directly conflicting, and the evidence is confusing and can only be understood by considering the complexity of

pedestrian ground contact and the influence of vehicle shape and speed on pedestrian motion prior to contact with the ground. However, before assessing vehicle shape and speed, it should be recognised that the complex motion of the human body due to its deformability and the articulated motion in the joints as well as the relatively long time lapse between first vehicle contact and ground contact means that there is considerable variability associated with pedestrian ground contacts [17–19]. Multibody analysis using the Madymo pedestrian model showed that the HIC₃₆ score and the 3 ms acceleration criterion for the head, pelvis and chest from ground contact is far less predictable than for a vehicle impact [19]. In fact, the timing and peak magnitude of the ground contact force appeared to be almost random with respect to the initial conditions. This variability can occur even where the vehicle impact speed is constant [18]. Dummy tests showed that, when the initial body contact with the ground occurs with the buttocks or the back, the severity of the head impact with the ground is substantially decreased [12, 18]. Similarly, when the head is the first body part hit by the ground, the HIC scores are over 500, but otherwise they are low [12]. Therefore great caution is needed when assessing ground related pedestrian injuries in specific accident cases.

Vehicle Impact Speed Effects

Early European studies of pedestrian accidents showed that the severity of the vehicle and ground impacts was speed dependent. At higher impact speeds, head injuries were primarily due to vehicle rather than road impact, whereas at low speed the ground impact is more severe [4, 17, 20, 21]. Figure 9.4 shows that below 7 m/s impact speed the injury severity from the ground impact is higher than from the vehicle impact. However, this relationship is reversed at higher speeds [20, 22].

A more recent Australian study of accident data found that two thirds of the serious injuries and half the Harm¹ for low speed crashes occurred from contact with the front of the car, while the ground was involved in 11% of severe injuries, but 25% of the Harm. The ground was associated with 23% of all Harm in low speed crashes and about 10% of all Harm in high-speed impact crashes [23].

There is some evidence that the proportion of direct head impacts on the ground is higher at low impact speeds [12]. Accident analysis showed that at low speeds the head mostly hits the ground first, while increasing the speed results in much more whole body rotation [4]. This is corroborated by dummy tests [5] at 20 km/h, which showed that often the head strikes the road first, while at 20–30 km/h often the legs strike first. Other dummy tests were used to conclude that, unlike the horizontal component of head ground impact speed, the vertical component is largely insensitive to vehicle impact speed [12]. However, the relative importance of these impact components remains unknown, and requires further work.

¹ Harm: total societal cost.

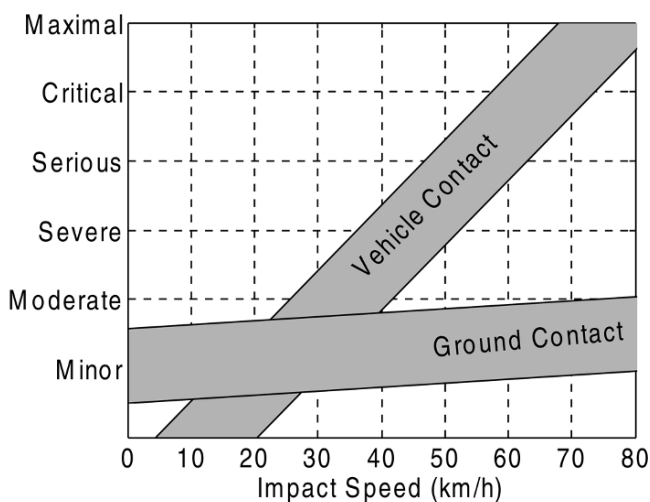


Fig. 9.4 Relative influence of vehicle versus ground injuries as a function of vehicle impact speed, adapted from [20].

Staged tests using adult and child sized pedestrian dummies struck by passenger cars showed a clear correlation between vehicle impact speed and maximum resultant head acceleration from vehicle impact. However, this relationship does not hold for ground impact, see Figure 9.5 [24]. For child dummy impacts, head/ground impact is more severe than the corresponding head/vehicle impact in most of the tests, despite the severe head/vehicle impact at collision speeds above 32 km/h [12]. The impact speed of the head against the ground is generally higher for the child than for the adult [12].

Similarly, multibody modelling of vehicle and ground impacts using the Madymo pedestrian model struck by a passenger car showed that the head 3 ms score for ground contact is always less than for vehicle contact at a given speed and pedestrian pre-impact stance [19]. This trend increases at higher impact speeds as the vehicle impact severity is primarily a function of vehicle impact velocity, while the ground impact severity appears to be primarily a function of the vertical drop height of the pedestrian [19]. The peak load in the head/ground contact for impact speeds ranging from 5–20 m/s is far higher than the vehicle/head contact, but the associated head velocity change is much lower, see Figure 9.6. This is due to the high stiffness of the ground contact compared to the vehicle contact. This results in high accelerations of short duration for the ground contact, yielding a higher HIC score, but a lower 3 ms head acceleration score. In most cases the impact energy for the pedestrian ground impact is substantially lower than for the vehicle pedestrian contact and overall, the relative severity of the ground compared to the vehicle contact reduces with vehicle impact speed, though there are exceptions to this when the first point of contact with the ground is with the head. The location on the head of the head/ground impact is highly variable, and clear patterns could not be determined.

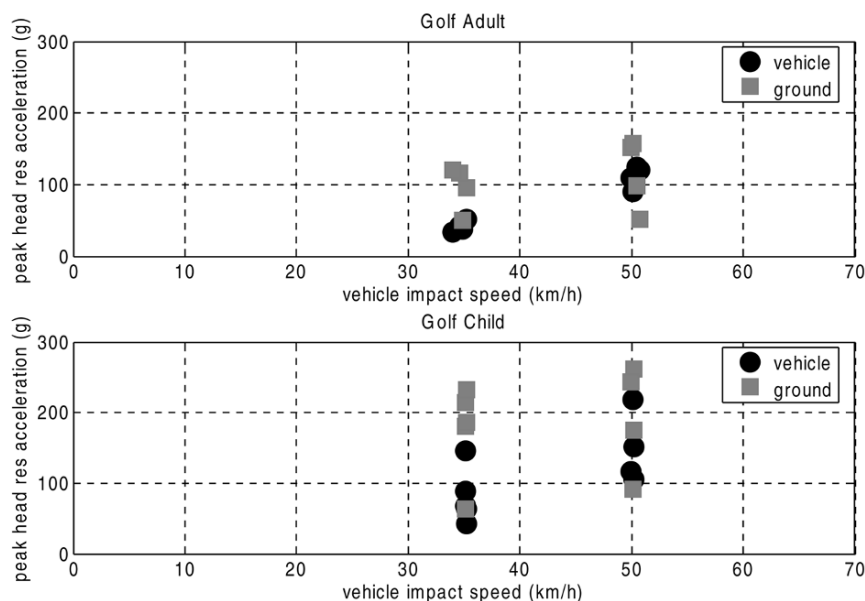


Fig. 9.5 Comparison of head/vehicle and head/ground accelerations as a function of vehicle impact speed, adapted from [24].

The pelvis, torso and shoulders can sometimes act to shield the head during the ground contact and head injuries from contact with the ground should therefore not be used to estimate vehicle impact severity.

Similarly, chest loading from ground impact is only weakly dependent on vehicle impact speed and the chest is well shielded in ground impacts. The pelvis 3 ms acceleration response from the ground contact increased with speed, but there was some variability. In some cases, the lower legs strike the ground first, then the pelvis and lastly the head, and in these cases a harsher leg and pelvis impact with the ground resulted in a less severe head/ground impact. In another case the ground had a direct hit on the pelvis and the head, with resulting high accelerations in these body regions [19].

Influence of Vehicle Design

Although some dummy tests have shown no significant influence of vehicle design on the severity of head impact with the ground [18], there is a considerable body of evidence to the contrary [2, 7, 12, 19, 25]. Pedestrian kinematics are far more influenced by vehicle shape than by stiffness [26], and therefore the principal vehicle design parameter for ground related pedestrian injuries is shape. Accident analysis

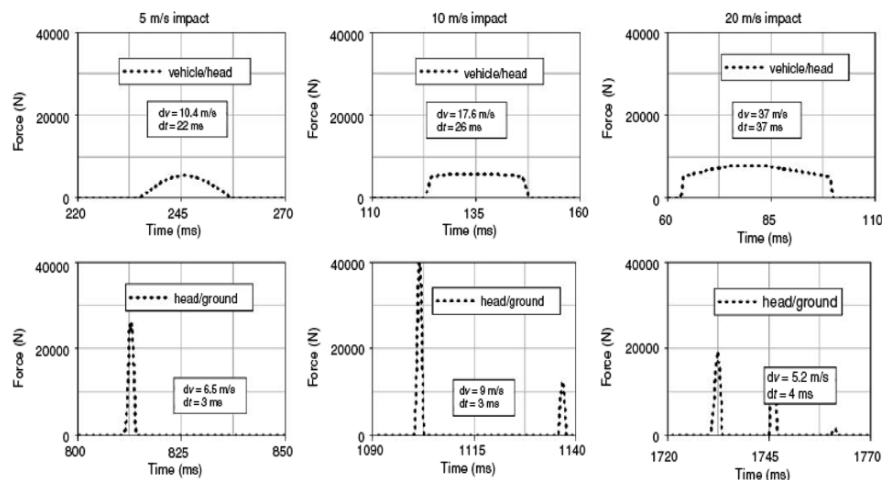


Fig. 9.6 Comparison of head/vehicle and head/ground contact force time-histories derived from Madymo multi body modelling showing head velocity change (dv) and contact time duration (dt) for different vehicle impact speeds, adapted from [19].

in Germany showed that the length of the bonnet and the windscreen inclination have little influence on ground contact injuries, but head injuries were observed more frequently for bonnet leading edge heights at or above 70 cm, while thorax injuries were more frequent for bonnet leading edge heights below 70 cm [4]. There is some corroboration of these results in a review of the recent Pedestrian Crash Data Study compared to the older Pedestrian Injury Crash Study which found that ground contact injuries from accidents involving newer cars are much less significant than for older cars [2].

However, there is also directly conflicting evidence regarding the influence of bonnet leading edge height on head injuries from ground contact. Two concurrent dummy test series in the 1970s concluded respectively that adult head injuries are reduced [12] and are increased [27] for passenger cars with higher bonnet leading edges. The former test series also investigated child dummy HIC scores from ground contact and found that higher bonnet leading edges resulted in higher child HIC scores. Considering that bonnet leading edge height relative to the body centre of gravity is more significant than absolute bonnet leading edge height, the findings in [12] for adults and children appear somewhat contradictory.

Significant further evidence of the influence of bonnet leading edge height on pedestrian head injuries from ground contact has come from comparing pedestrian injuries from passenger cars and from Light Trucks and Vans (LTVs). LTVs generally have significantly higher bonnet leading edges than passenger cars and accident analysis has shown that for passenger cars the ground contact accounts for only 7% of head injuries compared to 39% for LTVs [7]. Additionally, multibody modelling of SUV/pedestrian and car/pedestrian impacts found that side-facing adult pedestri-

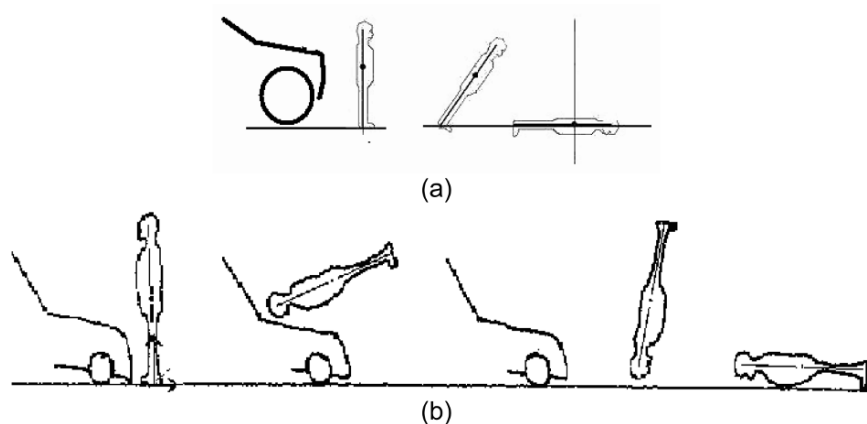


Fig. 9.7 Possible mechanisms of increased ground head injuries due to whole body rotation for (a) very high or (b) very low vehicle impact heights, adapted from [30].

ans struck by SUVs seemed to suffer head/ground contact before the upper body hit the ground and, despite the overall high level of variability, this was not found in the car impact cases [28].

We therefore propose two possible mechanisms for increased head impact severity with the ground as a result of vehicle shape, both related to whole body rotation:

1. In cases where the principal line of vehicle impact force is considerably above the pedestrian centre of gravity, upper body rotation away from the vehicle could result in a severe head impact with the ground, see Figure 9.7a. A similar hypothesis was proposed by [29], following multibody modelling to reconstruct a low velocity fatal SUV impact against a female pedestrian. They concluded that a high centre of contact force results in the pedestrian being rotated about their heels.
2. In cases where the principal line of vehicle impact force is considerably below the pedestrian centre of gravity (for example impact by a very low fronted sports car, see Chapter 3), whole body rotation towards the vehicle (ie the well-established wrap trajectory, see Chapter 3) may result in a severe head strike with the ground following the flight phase at certain impact speeds, see Figure 9.7b. However, due to the variability, this increased rotational velocity may also reduce the head impact severity with the ground, depending on the body orientation at the instant of ground impact.

However, these proposed mechanisms for ground contact head injuries are untested and require further analysis.

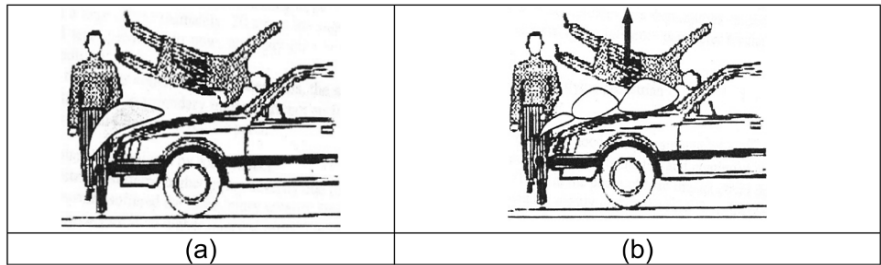


Fig. 9.8 Proposed airbag solution for retaining pedestrian on vehicle bonnet following impact (a) principle of action of pedestrian airbag on the front of the bonnet (b) technical solution for preventing pedestrian from separating from the bonnet, adapted from [4].

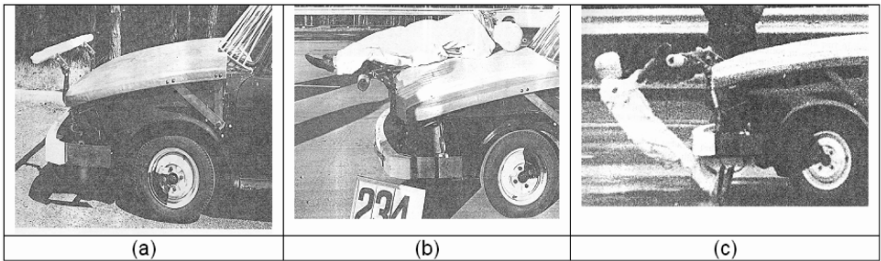


Fig. 9.9 Proposed mechanical catching device to maintain pedestrian on bonnet top following impact (a) device statically deployed, (b) retaining an adult on the bonnet at 16 km per hour (c) deployed catcher knocking down a child, adapted from [31].

Possible Methods to Reduce Ground Contact Injuries

A number of solutions have been proposed to reduce the severity of ground contact injuries. Apart from discouraging vehicles with high bonnet leading edges that appear to increase the severity of head contact with the ground [28], a number of active solutions have been proposed. These involve some form of containment method which prevents separation of the pedestrian from the vehicle and thereby prevents ground contact [4, 31], see Figures 9.8 and 9.9. Unfortunately, none of these solutions appear practical and they have not been adopted in production vehicles, possibly due to the potential risks of obscuring the driver's view (Figure 9.8b) or causing severe injury in case of malfunction (Figure 9.9c).

It remains unclear whether future development of containment devices can overcome these practical difficulties. However, since the relative severity of ground contact compared to vehicle contact injuries is higher at low impact speeds, these devices are likely to be of most practical benefit for low velocity impacts. This is encouraging since pedestrian containment on the vehicle is more realistic at low speed than at high speed [19].

Concluding Remarks

There is a high degree of variability associated with pedestrian and cyclist ground impact. Nonetheless, although the severity of injuries is generally lower than from vehicle contact, pedestrian accident data shows that a considerable proportion of non-minor injuries are caused by ground impact, and there are therefore large potential benefits from reducing these injuries. Pedestrian injuries from ground impact are only weakly dependent on vehicle impact speed, and are subject to substantial variations resulting from minor changes in pedestrian stance and motion at impact. However, despite the variability, there is growing evidence showing that high fronted vehicles result in more severe head injuries from ground contact. The obvious solution for reducing ground related injuries is to contain the pedestrian on the vehicle front and thereby eliminate the ground impact. However, so far the disadvantages associated with proposed containment devices have outweighed the potential disadvantages, and they have not been implemented in any production vehicles. This area is therefore open to further research.

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Chapter 10

The Influence of Vehicle Design on Pedestrian and Cyclist Injuries

Introduction

Despite a popular view that pedestrian and cyclist safety cannot be significantly improved due to the mass and stiffness disparity between unprotected humans and motorised vehicles [1, 2], researchers have known since the 1970s that vehicle design has a significant effect on the severity and distribution of pedestrian injuries arising from vehicle impact [3–7]. Since then, hundreds of articles have addressed this topic. In contrast, there has been far less focus on the influence of vehicle design on cyclist injuries and although it is possible that there are some conflicting requirements for cyclist safety compared to pedestrian safety [8], the more general consensus is that pedestrian friendly measures on car fronts will also reduce injuries to cyclists [9]. In this chapter the focus is therefore on the influence of vehicle shape on pedestrian injuries but, where cyclist specific research findings are available, references to cyclists are made. The most established findings on vehicle design are presented with frequent reference to the injury data (Chapter 2), the impact kinematics (Chapter 3), injury criteria (Chapter 5), and the fundamental impact theory (Chapter 7) and modelling methods from the literature (Chapter 8) are used to demonstrate research findings. The chapter concludes with a description of the current state-of-the-art of pedestrian protection as assessed using the pedestrian sub-system impactors (Chapter 6).

Real-world vehicle pedestrian accidents range from large trucks striking young children to compact city cars striking large adults, and assessment of vehicle design for pedestrian safety must take account of this variability. This will be done by defining the principal vehicle related geometric parameters (bumper and bonnet leading edge height etc) relative to the height of critical anatomical landmarks (knee, hip, etc.) for a standing pedestrian.

The main body regions for pedestrian fatality are the head (31.4%) and midbody (pelvis, chest and abdomen) (22%). However, a large proportion of the non-fatal, but seriously debilitating, injuries are to the lower limbs (32.6%), see Table 2.3 [10]. Similar patterns have been found for cyclist injuries, see Tables 2.6 and 2.7.

Therefore, the design of vehicles for pedestrian and cyclist safety must focus on reducing injuries to these three body regions.

The three main vehicle related considerations for pedestrian and cyclist injuries are mass, shape and stiffness. These will be addressed separately using simple models to demonstrate the physics which underpin the findings from staged tests and more advanced models reported in the literature. Vehicle mass will first be shown not to be very significant. In contrast, vehicle stiffness will be seen from first principles to be a fundamental parameter for pedestrian and cyclist injury and the importance of vehicle shape will also be demonstrated since this determines the impact locations of the vehicle surfaces on the body. Furthermore, it will be seen that shape influences the energy associated with each impact, while stiffness determines the corresponding force. Therefore shape and stiffness combine to determine injury likelihood for pedestrians and cyclists when struck by a particular vehicle type.

Some reported findings are directly contradictory. For example, early cadaver tests indicated that stiffness is less important than vehicle shape [6], but recent multibody modelling concluded that stiffness has a greater influence on pedestrian injury than shape [11]. Contradictory evidence should not be surprising given the volume of the published research and the fact that injuries are also strongly influenced by factors other than vehicle design, such as impact speed, age and relative pedestrian/cyclist orientation relative to the vehicle at impact. Furthermore, in the analysis of accident data and even within controlled experiments, it has often been difficult to decouple the influences of shape and stiffness on pedestrian and cyclist injuries. However, the principles governing the individual influences of vehicle shape and stiffness on pedestrian injuries are relatively straightforward and can be summarised by Ashton's statement in 1979 that a smooth and compliant front profile may reduce overall pedestrian injury severity [3]. This chapter aims to bridge the gap between this broadly accurate but simplistic statement and the many more detailed findings that have been reported by researchers around the world over four decades.

Definitions

There are hundreds of vehicle types which can be potentially involved in a pedestrian accident. These include cars, buses, vans, sport utility vehicles and trucks (see Figure 10.1), and significant variations are found even within these categories.

Snedeker et al. [12] recently published average bumper height, bonnet leading edge height and bumper lead (Figure 10.2) for cars, SUVs and vans, see Table 10.1. Great variations in commercial truck sizes are found and these are therefore not included here. There are also significant differences in pedestrian stature with age, gender and ethnicity, nutrition, etc., see Figure 10.3 [13].

Pedestrian injuries are mainly caused by the front part of a vehicle [14], and early research found that the bumper, bonnet leading edge and lower windscreen corner were the most hazardous vehicle regions for both pedestrians and cyclists



Fig. 10.1 Variation of common vehicle shapes.

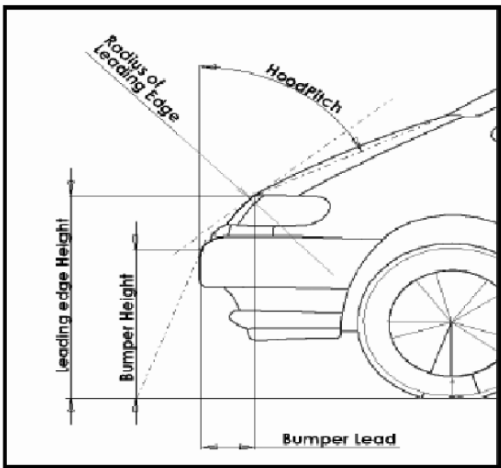


Fig. 10.2 Principal vehicle geometrical features showing bumper height, bonnet leading edge height and radius, bumper lead and bonnet (hood) pitch, adapted from [12].

Table 10.1 Average vehicle dimensions for cars, SUVs and vans [12].

Height above ground	Car	Van	SUV
Bumper height	500 mm	580 mm	640 mm
Bumper Lead	140 mm	160 mm	140 mm
Bonnet Leading edge	740 mm	860 mm	1020 mm

[15]. Analysis of pedestrian kinematics (Chapter 3) shows that the bumper typically strikes the lower limbs, while the bonnet leading edge strikes in the region of the hip and head impact subsequently occurs with either the bonnet top or the windscreen area, see Figure 10.4.

Therefore, a comparison of bumper height (h_{bumper}) with knee height (h_{knee}) and bonnet leading edge height (h_{ble}) with hip height (h_{hip}) (see Figure 10.4) for the different vehicle classes is instructive, see Figure 10.5. The average and 95%ile limits are given for the adult knee and hip heights as well as for the bumper and bonnet leading edge for each vehicle type.

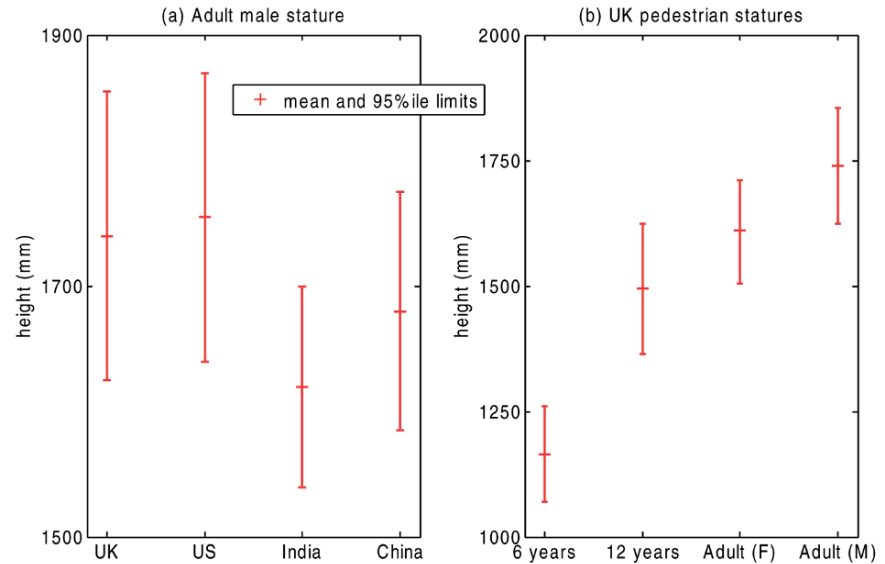


Fig. 10.3 Male pedestrian statures in different countries and in the UK separated by age and gender [13].

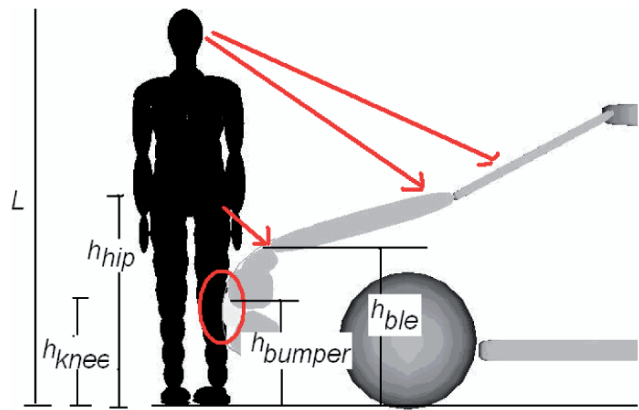


Fig. 10.4 Typical contact locations in vehicle pedestrian accidents.

Figure 10.5 shows that it is difficult to assess the influence of bumper and bonnet leading edge height for a particular vehicle for all kinds of pedestrians [16]. For example, the variability in bonnet leading edge height for vans means that there are cases where contact occurs both above and below the hip height, depending on pedestrian stature. To facilitate analysis, the following normalised dimensions are therefore defined:

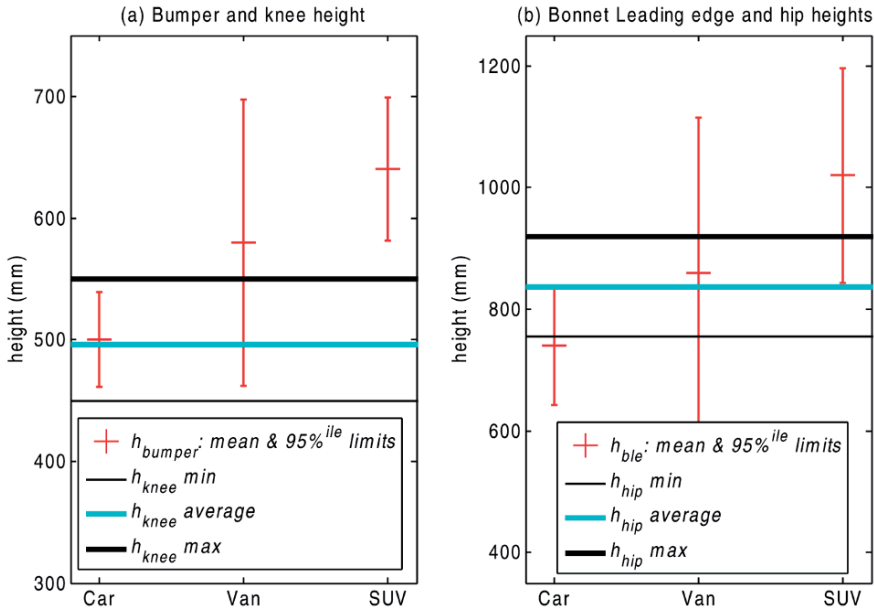


Fig. 10.5 Bumper height (h_{bumper}) and adult knee height (h_{knee}) and bonnet leading edge (h_{ble}) and adult hip heights (h_{hip}) for different vehicle classes [12, 13].

- relative bumper height:

$$\bar{h}_{bumper} = \frac{h_{bumper}}{h_{knee}},$$

- relative bonnet leading edge height:

$$\bar{h}_{ble} = \frac{h_{ble}}{h_{hip}}.$$

The location of head contact on the vehicle is frequently measured using the Wrap Around Distance (WAD) [17, 18], which is the length measured along the vehicle front profile from the ground to the head impact location [19, 20], see also Chapter 3.

Influence of Vehicle Mass

It can be shown from fundamental considerations that the mass difference between most motorised vehicles and pedestrians is sufficiently large to make the considerable mass disparity between different vehicle types striking pedestrians and cyclists

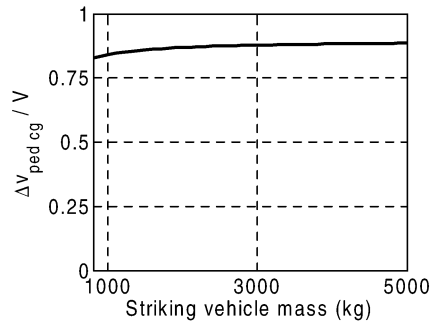


Fig. 10.6 Ratio of pedestrian centre of gravity velocity change ($\Delta v_{\text{ped_cg}}$) to collision velocity (V) estimated using momentum conservation and a single segment pedestrian representation.

largely insignificant. For example, consider a pedestrian modelled as a rigid rod¹ struck by the bonnet leading-edge (ignoring the bumper contact) of vehicles of the same shape but ranging in mass from 800 to 5000 kg. Assuming a common post impact velocity at the impact location, and a vertical height difference (h) between the pedestrian cg and the bonnet leading edge of 0.15 m, the ratio of pedestrian centre of gravity velocity change ($\Delta v_{\text{ped_cg}}$) to vehicle impact velocity (V) can be used as a crude measure of impact severity normalised with respect to the vehicle impact speed. This ratio can be found from consideration of the conservation of momentum (see Chapter 7) using the following equation:

$$\frac{\Delta v_{\text{ped_cg}}}{V} = \left[\frac{M_{\text{veh}} k^2}{k^2 (M_{\text{veh}} + M_{\text{ped}}) + h^2 M_{\text{veh}}} \right]. \quad (10.1)$$

Figure 10.6 shows the very low sensitivity of this ratio to striking vehicle mass over the large mass range considered.

This indicates that the substantial mass difference between passenger cars (mean mass 1275 kg) and heavier vehicles such as SUVs (mean mass 1625 kg [21]) and even larger trucks does not greatly influence pedestrian injury risk. A similar result was obtained with detailed Madymo multibody modelling [22] and analysis of accident data [23, 24], and it can be concluded that mass variation alone is not a very important vehicle design parameter for most pedestrian and cyclist injuries.

Influence of Vehicle Stiffness

In contrast to the mass, the stiffness of the vehicle structure significantly influences pedestrian injury severity. This is obvious from the fundamental relationship

¹ Assumed mass $M_{\text{ped}} = 75$ kg, radius of gyration $k = 0.4$ m to represent a 50th percentile adult male.

Table 10.2 Method to grade EuroNCAP pedestrian safety tests, adapted from [25].

Test configuration	Red score	Green score	Yellow score
Headforms	HIC > 1350	HIC < 1000	Between red and green values
Upper legform ^a	Max. bending > 380 Nm Total forces > 6.0 kN	Max bending < 300 Nm Total forces > 5.0 kN	Between red and green values
Legform ^b	Max. shear > 7 mm Max. bending > 20° Max. tibia acc. > 200 g	Max. shear < 6 mm Max. bending > 15° Max. tibia accel. < 150 g	Between red and green values

^aAs the total force is the parameter considered in the process to get to force-deflection, the rating procedure has only been based on results regarding total force criteria.

^bAs the impactor has been considered rigid in the process to get to force-deflection, the rating procedure has only been based on results regarding the maximum tibia acceleration criteria.

between maximum crush ($d_{\max}$) and mean acceleration ($\bar{a}$) for the idealised case of a plastic, concentric impact between a vehicle mass M striking a pedestrian mass m at a velocity V (see Chapter 7):

$$\bar{a} = \left[\frac{1}{2} \frac{V^2}{d_{\max}} \right] \left[\frac{M}{m + M} \right]. \quad (10.2)$$

Equation (10.2) shows that impact severity assessed by mean acceleration $\bar{a}$ (see Chapter 5) increases with the square of the impact velocity for a fixed crush depth, and is inversely proportional to deformation for a fixed impact velocity. Therefore, maximising crush depth and minimising impact speed are essential for pedestrian and cyclist protection.

The distribution of vehicle stiffness has recently been determined from analysis of EuroNCAP pedestrian impactor tests [25], see Chapter 6. Results from 425 tests were divided into categories according to headform HIC score and legform shear force and bending moment, see Table 10.2 (and Chapter 5). The categories were coded green for good, red for poor and yellow for intermediate, and the mean and ± 1 standard deviation force versus deformation corridors were derived. The results for the bumper, bonnet (front, middle and rear) and windscreen base are shown in Figure 10.7. Significant differences between vehicle regions and vehicle types are present, and therefore the current vehicle fleet presents a broad range of pedestrian and cyclist injury risk based on stiffness alone. Not surprisingly, therefore, adjusting the stiffness of a small production car without influencing the styling showed that substantial improvements could be made to the EEVC test results scores [26].

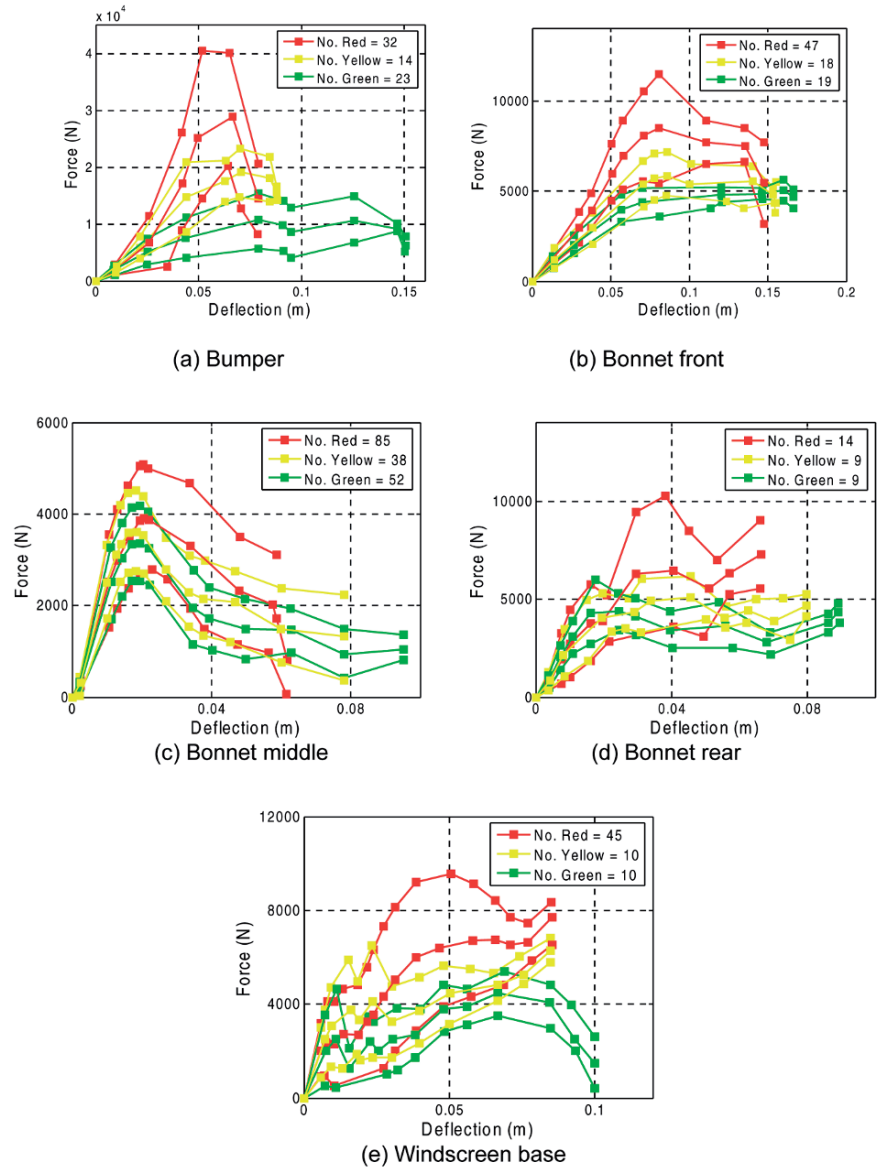


Fig. 10.7 Mean and ± 1 standard deviation stiffness plots for (a) the bumper, (b–d) bonnet (front, middle and rear) and (e) windscreen base derived from EuroNCAP pedestrian impactor tests, adapted from [25]: colours represent vehicle rating: green = good, red = poor, yellow is intermediate.

Influence of Overall Vehicle Shape

Vehicle shape relative to pedestrian/cyclist height determines which body regions are struck and the impact energy associated with each body region impact. Chapter 2

Table 10.3 Primary impact height for cars, SUVs and vans determined using simple average of bumper and bonnet leading edge heights in Table 10.1.

	Cars	Van	SUV
Average of bumper and bonnet leading edge height	620 mm	720 mm	830 mm

showed that there are major differences in injury patterns for different classes of vehicle. The bonnet surface and windshield are the most important injury inducing contact regions for passenger vehicles, while for light trucks and vans (including SUVs) the main injury sources are the bonnet surface and the bonnet leading edge. The importance of vehicle shape was recognised as early as 1961 by Henry Wakeland who proposed the somewhat tongue-in-cheek ‘Automobile stylists Keel-Hauling test’ in which the vehicle designer was to be pulled over the front of the vehicle using ropes to provide a measure of the pedestrian injury potential of the vehicle [2]. However, for the most part changes in the overall frontal shapes of passenger cars have been prompted more by attempts to reduce the aerodynamic drag factor and thereby decrease fuel consumption [27], and pedestrian safety considerations have only influenced overall vehicle shape much more recently.

Vehicle shape is substantially dependent on vehicle class and the majority of vehicles striking pedestrians and cyclists can be broadly classified as cars, SUVs and trucks/vans. Using the mean dimensions for cars, SUVs and vans (see Table 10.1, [12]), the conservation of momentum applied to a single segment pedestrian model (see Chapter 7) can again be used to predict the generalised momentum transfer in the primary vehicle impact associated with different vehicle shapes. Analysis of pedestrian kinematics in Chapter 3 showed that the impact sequence is bumper-to-leg, bonnet-leading-edge-to-midbody and bonnet-or-windscreen-to-head for all three categories of vehicles. However, almost negligible upper body motion occurs in the very short time interval between bumper and bonnet leading edge contacts (see Figure 3.6). Therefore, as a first approximation when assessing overall pedestrian kinematics as a function of vehicle class, it is reasonable to consider the bumper and bonnet leading edge contacts as a single contact occurring at the average height of the bumper and bonnet, see Table 10.3.

Using the data in Table 10.3 and assuming as before a 50th percentile male mass and height,² a mean car mass of 1265 kg and mean SUV/van³ masses of 1625 kg, the relative severity of the primary impact for each of the vehicle types can be assessed by considering the linear and angular post impact velocities of the pedestrian segment. The centre of gravity linear velocity ($\Delta v_{\text{ped_cg}}$) normalised by vehicle impact speed (V) is given by equation (10.1) and the normalised body angular velocity change (ω) is given by

$$\frac{\omega}{V} = \frac{Mh}{(M + m)k^2 + Mh^2}. \quad (10.3)$$

² That is, assuming a mass $m = 75$ kg, radius of gyration $k = 0.4$ m as before.

³ That is, assuming that average van and SUV masses are equal.

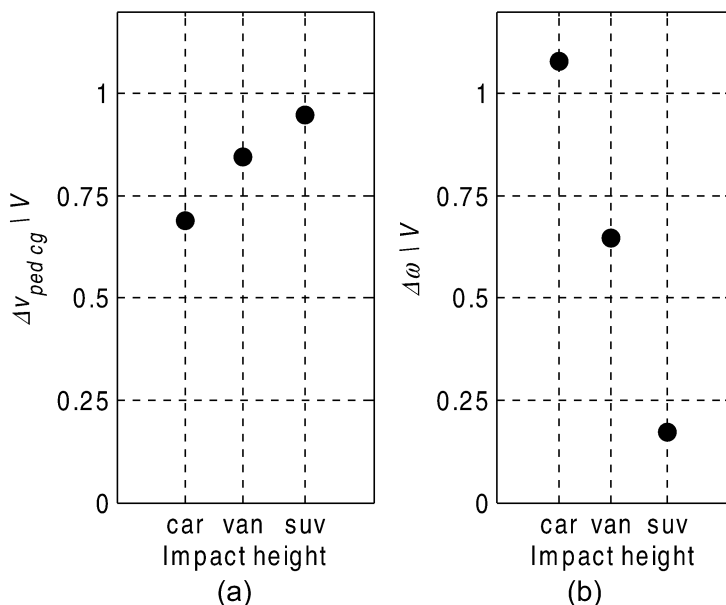


Fig. 10.8 Single segment momentum model predictions for the influence of vehicle type (car, van or SUV based on dimensions from [12]) on pedestrian kinematics in primary impact (bumper and bonnet leading edge combined): (a) normalised linear velocity change of pedestrian body centre of gravity ($\Delta v_{ped_cg} / V$) and (b) normalised angular velocity change of the pedestrian body ($\Delta \omega / V$).

These kinematic quantities are a measure of the linear and angular impulse transmitted to the struck pedestrian. Their variation for the average impact height of cars, vans and SUVs are shown in Figure 10.8. This conservation of momentum approach clearly shows that the lower impact height of cars results in a significantly lower linear impulse in the primary impact compared to SUVs, but this is accompanied by a significantly higher angular impulse for car impacts compared to SUVs. These trends are even more dominant on an energy basis. The effective impact height and hence the corresponding linear and angular impulses for vans lies between that of cars and SUVs (see Table 10.3 and Figure 10.8). We will see that the velocity of the subsequent head contact with the bonnet or windscreen is substantially influenced by whole body rotation following primary impact and therefore it can be concluded from this simple analysis that the generalised shapes associated with cars, vans and SUVs result in substantial variations in head and midbody loading for pedestrians.

The highly simplified single segment model used in this analysis ignored the influence of the ankle, knee and hip joints but it provides a clear overview of the influence of vehicle shape on pedestrian loading patterns: pedestrian impacts from high fronted vehicles result in more severe midbody loading and lower head impact velocity on the vehicle due to reduced rotation. Bearing these results in mind, we now assess the influences of vehicle design in more detail.

Bumper Shape

Bumper shape is defined by three main measures: height from the ground relative to the pedestrian height, protrusion from the vehicle (or bumper lead), and the vertical spread of the bumper (ie distance from upper to lower bumper edges), see Figure 10.2. The relative bumper height determines the initial contact location on the body, and the relative bumper height, shape and stiffness are the main determinants of the magnitude and distribution of the resulting bumper force. Apart from soft tissue cuts and abrasions, the principal injuries from bumper impact are fracture of the tibia, fibula and femur and tearing of the ligaments in the knee, see Chapter 2. These injuries occur due to high stresses in the contact region and the distribution of shear force and bending moment due to the inertia of the leg/thigh and the hip, knee and ankle joint constraint forces and moments. The bumper height relative to knee height is significant and can be classified as either $h_{\text{bumper}}/h_{\text{knee}} \leq 1$ or $h_{\text{bumper}}/h_{\text{knee}} > 1$, resulting in different loading patterns of the knee joint and the long bones of the leg.

Analysis of accident data, mathematical models and cadaver and dummy tests has shown that, for vehicle bumpers striking adult pedestrians, the maximum bending moment in the leg coincides approximately with the height of the bumper, and the knee shear force and bending moment are highest for impact close to the knee level due to the inertia of the leg below the knee. As a result, a bumper striking close to the knee causes large lateral deflections in the knee joint, posing a significant risk of knee ligament injury. Conversely, for bumper impacts in the mid-tibia region,⁴ the bending moment in the knee is minimised as the upper and lower leg segments rotate together and lateral bending of the knee joint is prevented. This implies an optimum bumper contact height with the leg for minimising knee injuries at about 35 and 24 cm above the ground for adults and children respectively. Furthermore, lowering the impact height below the knee not only reduces knee injuries, but limits involvement of the critical pelvic and abdominal areas. Bumpers that strike above the knee level, common in off-road vehicles and multipurpose vehicles, cause the upper body and leg to rotate in opposite directions and can result in severe knee trauma. These vehicles have a requirement for high ground clearance which make it difficult to introduce methods of pushing the pedestrian legs forward [1, 3, 6, 28–39].

The physical reasons for these findings can be illustrated using a two segment model of the pedestrian hinged at the knee⁵ and struck by a vehicle bumper from the side at a point location [32, 40, 41]. The knee hinge can be considered as locked/free to represent the intact/damaged knee respectively, see Figure 10.9. From momentum conservation, the knee joint constraint and assuming plastic impact, rigid body relations and a nominal bumper contact duration and ignoring the subsequent bonnet leading edge contact, average normalised knee shear force and bending moment as a function of bumper height are shown in Figure 10.10. For low severity impacts,

⁴ At about 70% of knee height above the ground.

⁵ 50th percentile male height = 1.74 m, mass = 77 kg, knee height = 0.5 m; vehicle mass = 1200 kg.

As relative bumper height increases from $h_{\text{bumper}}/h_{\text{knee}} < 1$ to $h_{\text{bumper}}/h_{\text{knee}} > 1$, the source of inertial resistance at the knee switches from the upper to the lower body and the knee shear force changes sign, Figure 10.10a. For the intact (locked) knee, the magnitude of the knee shear force for bumper contact just above the knee is higher and opposite in direction compared to bumper contact just below the knee. The bending moment in the intact knee is minimised for $h_{\text{bumper}}/h_{\text{knee}} \approx 0.6$. Conversely, as bumper height approaches knee height ($h_{\text{bumper}}/h_{\text{knee}} \approx 1$) there is a sharp increase in knee bending moment for the intact knee model, especially for $1 < h_{\text{bumper}}/h_{\text{knee}} < 1.4$ (Figure 10.10b).⁶

The bumper height also influences the magnitude of the bumper contact force. Cadaver tests have shown that the peak bumper force was larger for a relative bumper height of 0.9 compared to 0.6 when the bumper was almost rigid, but this was not seen for a compliant bumper and no clear correlation between bumper force and injuries in the leg area was observed [6, 31]. Similarly, multibody modelling has shown that raising the relative bumper height from 0.5 to 0.9 increases the bumper force [42]. However, a two segment model and dummy tests have indicated a reduction in bumper force for contact close to the knee compared to the mid-tibia region [32, 38]. These conflicting results can be partly explained by recognising that the relationship between bumper height and bumper force depends on the lateral bending stiffness of the knee. In the two segment momentum model with the intact (rigid) knee joint, the bumper force increases almost linearly for $0.6 < h_{\text{bumper}}/h_{\text{knee}} < 1.6$ due to the reduced eccentricity of the impact with respect to the body's centre of gravity, see Figure 10.11a. However, in the damaged (freely pivoting) knee model the bumper force is minimised for $h_{\text{bumper}}/h_{\text{knee}} = 1$, due to the reduced effective mass resulting from the freely pivoting knee formulation. The actual knee lateral bending stiffness lies between free and locked and damage to the ligaments limiting lateral bending in the knee will therefore reduce the bumper force.

Bumper height also influences which bones in the leg are injured. Analyses of light trucks and vans compared to cars have shown a greater risk of injuries above compared to below the knee [21, 43]. A detailed review of leg injuries from high fronted vehicles in the Pedestrian Crash Data Study showed that a higher bumper is more likely to cause femur fracture while a lower bumper is likely to cause tibia fracture and/or knee ligament injury [44]. The average relative bumper lower heights⁷ for femur fracture, ligament injury and tibia fracture were 0.95, 0.79 and 0.78 respectively. Furthermore, the impact velocity was statistically different for the femur fracture group compared to the knee ligament injury group [44, 45]. For bumper impacts above the knee at about 40 km/h femur fracture is normal, but at 20–30 km/h ligament injuries usually occur instead. For bumper impact below the knee at close to 40 km/h tibia fracture is normal but ligament injuries usually occur at impact velocities of 20–30 km/h. These and other accident data and experimental tests therefore

⁶ The knee bending moment is by definition zero in the damaged/free knee model.

⁷ The bumper upper height was surprisingly not reported.

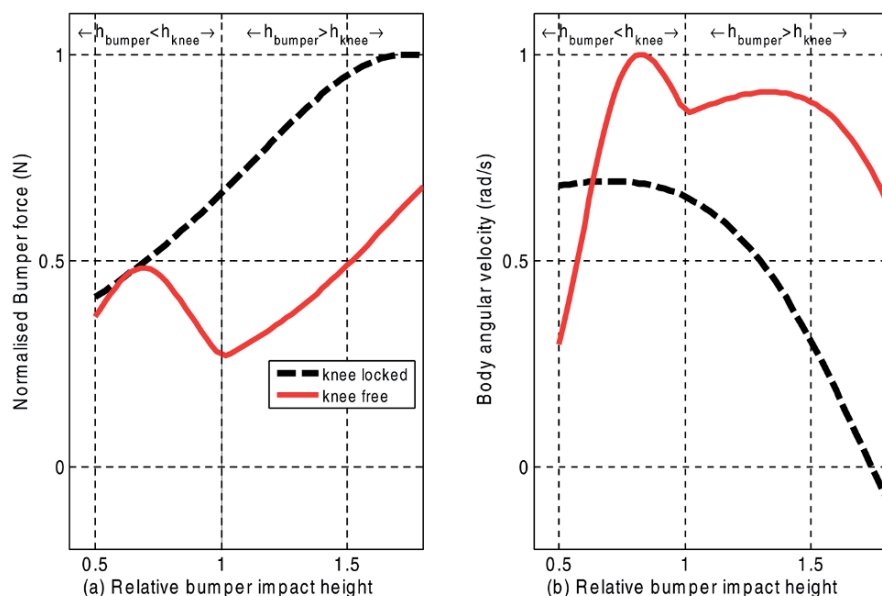


Fig. 10.11 Two segment model predictions for a side struck pedestrian (a) bumper force and (b) upper body angular velocity as a function of relative bumper height.

show that in impacts with sufficient energy to cause bone fracture a protective effect for the knee ligaments is found [45–49].

Early evaluation of cyclist impact using dummy and cadaver tests showed similar peak bumper loads obtained in cyclist and pedestrian impacts [50]. However, a more recent analysis of Japanese accident data showed that the bumper caused fewer tibia fractures in cyclists than pedestrians due to the bent knee posture of cyclists and their higher leg and pelvis position [51].

Influence of Bumper Height on Whole-Body Kinematics

The influence of bumper height on whole-body rotational kinematics has been investigated because of the potential influence of bumper height on the velocity of subsequent head impact with the bonnet or windscreen [32]. However in cadaver and dummy tests, no significant influence of bumper height (32.5 or 45 cm) on leg angular velocity was found [31, 52] and, in two-segment and multibody mathematical modelling and dummy tests, only a weak influence of bumper height on leg angular velocity was found [32, 42]. Consideration of the two segment momentum model with the rigid knee shows that, for relative bumper heights ranging between 0.5 and 1.0, the influence of bumper height on body angular velocity is small, see Figure 10.11b. For the free knee model, whole-body angular velocity is reasonably

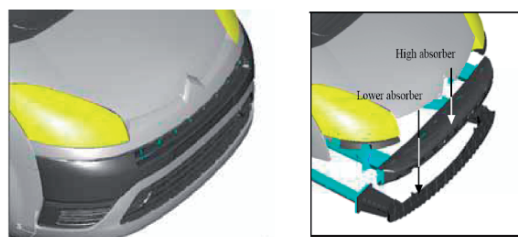


Fig. 10.12 Upper and lower bumper in the Citroen C6, adapted from [58].



Fig. 10.13 Development of vehicle front end style, adapted from [57].

constant for relative bumper heights ranging between 0.7 and 1.5. However, we will see that the dominant vehicle shape characteristic for head impact velocity is the relative bonnet leading edge height.

Secondary Bumper

A significant reduction in knee injuries can be achieved by the provision of a secondary bumper close to the ground which supports the lower leg, see Figure 10.12, and vehicle designs have developed accordingly, see Figure 10.13 [1, 53–58]. For the double bumper system compared to the standard bumper, dummy tests showed a 20% lower knee bending moment, and the shear force and lateral bending angle at the knee and the tibia acceleration were more than 50% lower [35], and only simple long bone injuries were reported [56].

Using the THUMS finite element model (see Chapter 8) the distribution of bending moment in the leg was shown to be shifted downward due to the force exerted by the lower absorber. A high bending moment was produced at the proximal tibia without the lower absorber due to the load concentrated at the bumper absorber, and therefore high loading on the knee joint was generated. The lower bumper absorber was effective in reducing risks of medial collateral ligament rupture, but the tibia fracture risk can increase. The tibia fracture risk is reduced by lowering the height of the lower absorber because this reduces the tibia bending moment. Conversely, raising the lower absorber can increase the risk of medial collateral ligament rupture

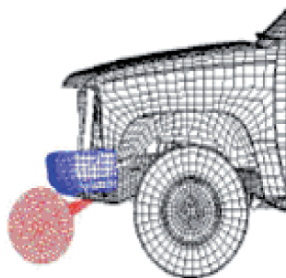


Fig. 10.14 Possibility to reduce the lateral knee loading by an inflating bumper airbag [60].

[59]. However, cadavers tests have shown that ankle injuries occur more frequently at low impact levels due to violent tilting at the ankle joint when the leg is hit low-down by the bumper [36].

For vehicles such as SUVs designed for off road usage, the secondary lower bumper for pedestrian protection frequently conflicts with ground clearance requirements, and an inflating bumper airbag has been proposed to provide leg protection by reducing lateral knee loading [60], see Figure 10.14.

The overall influence of bumper height on leg loading can be summarised using the two segment pedestrian model. Figure 10.15 shows the normalised shear force and bending moment distributions as a function of vertical position along the leg normalised with respect to knee height for both the locked (intact) and free (damaged) knee cases. The bumper heights for an average car, van and SUV from [12] are shown along with a theoretically ideal bumper height of 0.3 m (= 60% of knee height), which is close to the average bumper height for a car fitted with a secondary lower bumper. Figures 10.15a&c show that for bumper contacts well below the knee – for example from a car with a secondary bumper – the femur shear and bending loads are low and this explains why the femur is rarely fractured in these cases. For bumper contact close to the knee, the upper tibia and knee joint shear and bending loading are high for the locked (intact) knee model, and this explains why tibia and knee ligament injuries frequently occur together in these cases. For bumper contact above the knee (i.e. from a high fronted vehicle without a secondary bumper), the shear and bending in the knee and particularly the femur are high in the locked knee model and this is likely to result in femur fracture somewhat above the knee. When the femur fractures close to the knee, the free knee model applies and the knee joint and tibia shear and bending loads are greatly reduced. This explains why femur fracture does not generally occur together with tibia and ligament injuries.

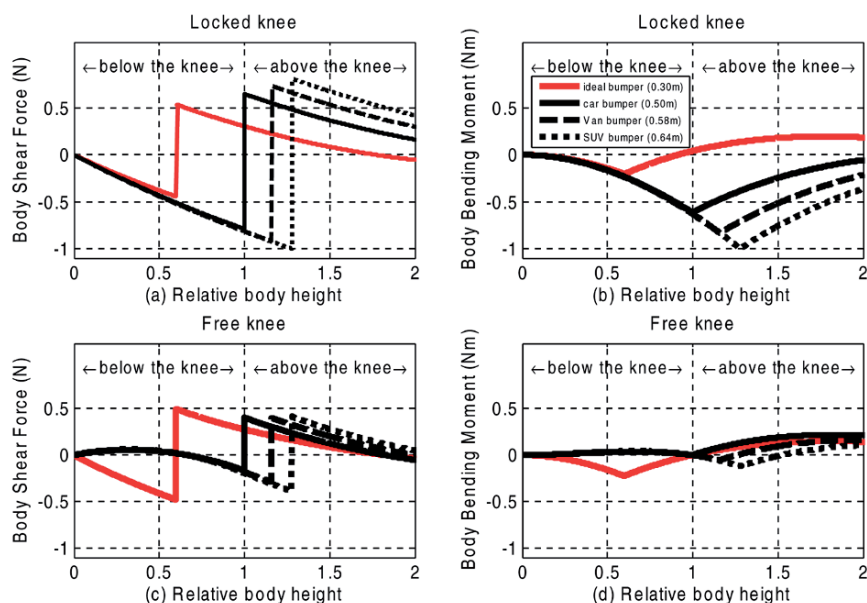


Fig. 10.15 Normalised distribution of shear force and bending moment in single segment pedestrian model versus relative body height (knee height = 1) for various bumper heights.

Bumper Lead

Bumper lead is a measure of how far the bumper protrudes in front of the vehicle (see Figure 10.2), and traditionally bumper lead was used as a means to protect the vehicle. Bumper lead largely determines the time delay between bumper and bonnet leading edge contact with the pedestrian. A short bumper lead has been recommended to minimise the injuries to the lower leg and knee [3], and modern cars generally feature very short or entirely absent bumper leads, see Figure 10.13. Dummy tests show that the influence of bumper lead on the knee bending moment is slight, but increasing the bumper lead from 100 to 200 mm resulted in a 10% increase in knee shear force [35].

Bumper Stiffness

Equation (10.2) showed that average acceleration is inversely proportional to deformation and early investigation of pedestrian accidents quickly led to the recommendation of compliant vehicle front profiles [3, 32]. The ideal vehicle bumper design for pedestrian protection allows sufficient compliance to cushion the impact while limiting knee lateral bending [54]. However, cadaver tests showed that



Fig. 10.16 Honda Civic bumper tests, adapted from [65].

although reduced bumper stiffness greatly lowers the contact force, the ground reaction force is increased and leg injuries are not always reduced by more compliant bumpers [6]. Dummy tests with a softer bumper showed highly decreased knee shear force and tibia acceleration but no significant reduction in the knee bending moment [35]. Cadaver tests showed that stiff bumpers cause more severe tibia fractures, but reduced injuries to the knee and ankle due to the protective effect of bone fracture at the impact site [56]. To combat the effect of ground reaction forces, an energy absorbing bumper with efficient force-deflection characteristics and the ability to raise the leg slightly was developed [29].

However, a significant problem for the optimisation of vehicle front bumper stiffness is caused by bumper regulations designed to minimise vehicle repair costs [55, 61, 62]. To comply with bumper standards in many countries, the bumper structure must be able to absorb considerable energy, and this prevents large decreases in bumper stiffness [53]. Recent research has shown that to provide adequate leg protection, free space between the bumper skin and aluminium bumper is required [63, 64], and the European Traffic Safety Council recommend that the front face of the bumper needs to crush by 5–7.5 cm to protect the leg [64]. Current plastic bumpers are very flexible, and do not need significant change in their ability to deform [65]. The deep bumper and the integrated air dam or spoiler common in modern cars are beneficial in distributing the contact force in legform impacts. Honda uses a combination of deforming loop and crush cans at the front face of the bumper armature and a two-stage energy management system has been implemented in the bumper which helps absorb leg impact loading, without compromising bumper energy absorption in low speed vehicle impacts, see Figure 10.16 [65]. Opel have used low density pedestrian protection foam in front of the stiff aluminium bumper crossmember to absorb the impact energy together with sufficient deformation space to avoid the impactor hitting the stiff aluminium bumper crossmember or the foam bottoming out [66].

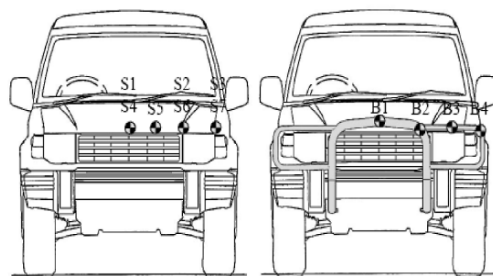


Fig. 10.17 Impact locations of SUV with and without steel bullbar using child headform [69].

Bullbars

Bullbars are frequently fitted to off-road style vehicles, even in cases where the vehicle is predominantly used in an urban environment. Bullbars are traditionally constructed from circular steel tubing and consequently adult legform and child headform impact tests at 30 to 40 km/h have shown a very high risk of serious leg injuries in adults and life-threatening brain injuries in children struck by vehicles with steel bull bars, see Figure 10.17. However, the vehicles to which bullbars are fitted generally have poor pedestrian protection even without bullbars, and it has been predicted that deformable plastic bullbars actually reduced the risk of child head injury and adult leg injury. However, there is insufficient accident data to confirm these predictions [67–70].

Shape of Bonnet and Bonnet Leading Edge

The height and shape of the bonnet leading edge are important parameters for vehicle/pedestrian and vehicle/cyclist collisions because they strongly influence injuries caused directly by the bonnet edge contact as well as the subsequent head kinematics. Adult midbody injuries from bumper or bonnet leading edge contact are significant because pelvis and upper thigh trauma often cause permanent disablement [71] and a review of pedestrian accidents has found that up to 17% of pedestrian fatalities involved only a chest, spine or abdomen injury [72]. Similarly, the kinematic motion of the head is important because this determines the location and velocity (and hence energy) of the head impact on the bonnet or windscreen. For children, head contact can occur directly with the bonnet leading edge [73].

An early indication of the importance of bonnet leading edge height was given by a 1969 comparison of pedestrian injuries from impacts with an E-type Jaguar (low smooth bonnet) or a Rolls Royce (high square bonnet), which showed that because the bonnet leading edge of the Rolls Royce was above the hip height of a 50th percentile male, the resulting impact forces were very high [74]. Similarly, a

1972 study comparing pedestrian accidents involving Volkswagen Beetles (rounded bonnets) and Cadillacs (square bonnets) concluded that there would be a 30% reduction in pedestrian fatalities if all cars striking pedestrians had Beetle fronts [75]. Subsequent analysis of accidents, and tests with dummies and cadavers and detailed multibody/finite element models have all shown that pelvic loading is much greater for higher profile vehicles and therefore bonnet leading edge height is a significant factor for femur and pelvis injuries [6, 12, 22, 76–80]. Partly in response to these findings, the bonnet leading edge of most modern cars is substantially lower and more rounded than for older cars (see Figure 10.13), where high bonnet leading edges resulted in abdominal injuries accounting for almost 20% of all injuries and 7% of serious/fatal injuries [71]. Conversely, no serious injuries to the pelvic area were produced in cadaver tests with a lower vehicle profile [6]. Bonnet slope was not found to be significant for pelvis injuries [78].

The bonnet leading edge contact force increases substantially with bonnet leading edge height [35, 38] because of the increased effective mass of the pedestrian as the bonnet leading edge strikes closer to the pedestrian centre of gravity. This can be illustrated with the single segment representation of a pedestrian struck by a vehicle with a bonnet leading edge vertically offset by a distance h from the pedestrian centre of gravity.⁸ Ignoring the bumper contact, the normalised bonnet leading edge impulse is given by equation (10.1), and this can be seen as a crude measure of impact severity, see Chapter 7. We have already seen that the influence of h is strong and Figure 10.18a shows the relationship between normalised bonnet leading edge impulse and bonnet leading edge height normalised with respect to the pedestrian centre of gravity height. This relationship is in direct proportion to the effective mass of the single segment representation of the pedestrian (m_{eff}) which is

$$m_{\text{eff}} = \left[\frac{k^2}{k^2 + h^2} \right] m, \quad (10.4)$$

and the dependence of effective mass normalised by actual pedestrian mass on relative bonnet leading edge height is shown in Figure 10.18b. For impacts close to the centre of gravity height, the effective mass approaches the actual mass of the pedestrian. However, as the bonnet leading edge height reduces there is a substantial reduction in effective mass, so that for a relative bonnet height of 0.7 the effective mass is reduced by 30%.

Pelvis height largely coincides with the pedestrian centre of gravity height and these results are therefore doubly significant for pedestrian injury: not only does the impulse increase with relative bonnet leading edge height, the relative bonnet leading edge height also determines whether contact is made with the femur or directly with the critical pelvis/abdomen region. Therefore, bonnet height has a direct influence on vehicle aggressivity for pedestrians.

Analysis of pedestrian accidents showed that pelvic fractures predominantly occur when there is a direct impact with the bonnet leading edge [81]. This is par-

⁸ Assumed pedestrian mass $m = 75$ kg, height $L = 1.74$ m, radius of gyration $k = 0.4$ m, as before, $M = 1265$ kg.

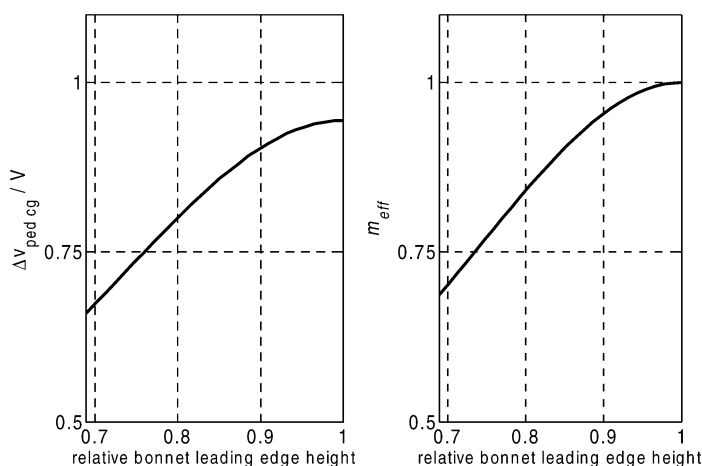


Fig. 10.18 Single segment representation of normalised bonnet impulse ($\Delta v_{ped_cg} / V$) and effective pedestrian mass (m_{eff}) as a function of bonnet leading-edge height normalised with respect to pedestrian centre of gravity height.

ticularly important in the context of the proportion of vehicles classified as light trucks or vans (including SUVs), most of which have a high bonnet leading edge. By 2004, 40% of vehicle sales in the United States belonged to this category, and analyses of accident data showed that pedestrian fatality risk in collisions with these vehicles is substantially higher than for conventional passenger cars [21, 24, 82], see Chapter 2. Multibody modelling has shown that the primary reason for the increased hazard to pedestrians from high fronted vehicles is the increased height of the bonnet leading edge for these vehicles compared to conventional cars. The location of the bumper and bonnet edge contacts is such that the midbody region is directly struck in an SUV/pedestrian collision, allowing less rotation of the body. This means that for pedestrians struck by high fronted vehicles there is the combination of a harder primary impact which occurs directly with a critical midbody region. Lowering the bumper and bonnet and reducing the stiffness for SUVs would help to reduce injuries to these mid-body regions [22].

Analysis using the THUMS finite element model (see Chapter 8) has shown that the bonnet leading edge radius influences pelvis and femur injury risk since it alters the effective closing speed of contact of the bonnet edge with the midbody regions. The closing speed of contact between the thigh and the car bonnet can be made significantly smaller than the vehicle impact speed, depending on the roundness of the bonnet leading edge. In an ideal design, acceleration of the distal femur by the bumper and rolling motion imparted to the thigh by the bonnet radius can greatly reduce the energy of the hip impact [12, 79]. It was concluded that a car with a sufficiently low bonnet leading edge height, large bonnet edge radius, moderate bumper lead and high bumper edge height would practically exclude the possibility of femur

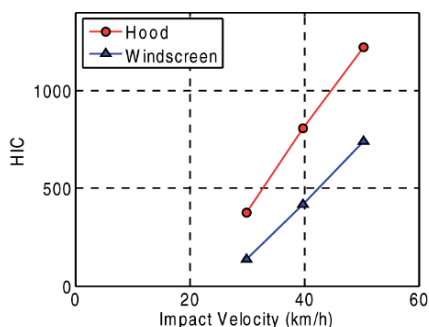


Fig. 10.19 Relationship between HIC score and head impact velocity from dummy tests for impact with the bonnet (hood) and windscreen [69].

fracture in primary lateral impact of a 50th percentile male at impact speeds less than 40 km/h [12].

Real-world injuries and dummy and cadaver tests show a strong interactive effect between the bumper and the bonnet leading edge [38, 49, 79, 83, 84]. Since the pedestrian generally attains a substantially common velocity with the vehicle during the impact process, bonnet leading edge impulse is greatest for a short bumper lead and a high bonnet. Conversely, the bonnet leading edge impulse is reduced when the bumper lead is increased and the bonnet edge is lowered. In dummy tests, it was found that the bumper tends to shield low bonnet leading edges from impact, and conversely, the bumper has little effect on bonnet leading-edge force for high bonnet leading edges. For a median bonnet height the shielding by the bumper of the bonnet leading edge increases with bumper lead [38]. In modern cars, the risk of chest injury due to passenger car impact has been reported as extremely low, but this was partly because the elbow to bonnet impact provides cushioning [85].

We have already seen (Figure 10.8b) that the height of the bonnet leading edge significantly influences the subsequent rate of upper body rotation for both adults and children in vehicle pedestrian impacts, and this affects the speed of the subsequent head impact on the bonnet/windscreen. The large significance of increased head impact speed is apparent from equation 2, and has been shown experimentally using dummy tests, where the HIC score (see Chapter 5) strongly depended on the head impact speed, see Figure 10.19. It is evident from this data that the overall effective stiffness of the windscreen is lower than for the bonnet (hood).

Accident analysis and tests using cadavers and dummies and multibody models showed that increasing bonnet leading-edge height reduces the subsequent head impact speed on the bonnet or windscreen [6, 11, 38, 42, 77, 79, 80, 86]. More recently, the dependence of head impact speed (normalised by vehicle impact speed) on bonnet leading edge height for different vehicle types has been reported using multibody modelling and dummy tests [11, 85], see Table 10.4. The multibody simulation results show higher speed ratios than the dummy tests, but both modelling approaches indicate increased head impact velocities for cars compared to SUVs.

Table 10.4 Head impact speed normalised by vehicle impact speed for 40 km/h dummy tests with different vehicle type impact for dummy tests and multibody simulations [11, 85].

		Compact car	Midsized car	Large car	Minivan truck	SUV/light
Head impact	Dummy tests [85]	0.80	0.88	–	–	0.72
speed/vehicle impact speed	Multibody simulations [11]	1.18	–	0.92	0.97	0.9

Similarly, simplified multibody modelling was used to predict that vehicle shape has a considerable influence on cyclist head impact velocity: the predicted head impact velocity for an adult cyclist seated in a cruising position was doubled for simulated impacts with a low fronted car compared to a high fronted car [87].

Analysis of accidents has shown that the pedestrian slips over the front edge of the vehicle, because the pedestrian's body does not fully attain the vehicle speed due to the eccentric impact. The amount relative motion, and hence the WAD to head contact, increases with increasing impact speed [88], but there are confounding effects due to vehicle deceleration and bonnet dive if braking occurred during impact. However, the WAD is also influenced by bonnet leading edge height, which can therefore change the vehicle region with which the head impact occurs. Cadaver tests have shown that raising the bonnet leading edge height results in shorter WAD values and conversely dropping the bonnet leading-edge height from 800 to 600 mm increased the WAD from 1.78–2.05 m due to a considerable slipping effect of the pedestrian's body [79].

The WAD is influenced by vehicle speed, pedestrian stature and bumper height and bonnet leading edge height and length [89, 90]. Consequently, the variability in WAD is broad: analysis of cases in the Pedestrian Injury Causation Study showed that in 80% of cases the WAD was between 115 and 240 cm [19]. More recent accident data from Japan have shown that the WAD for cars is generally greater than the pedestrian height due to sliding of the pedestrian over the bonnet [20], but not for minivan type vehicles, see Figure 10.20.

The WAD for children is lower than for adults, and accident analysis performed by the IHRA showed that the transition from a child to adult WAD starts at 1400 mm and ends at 1700 mm [91]. Multibody modelling reconstructing child pedestrian accidents showed that the average ratio of WAD to child pedestrian height is 0.91 with a standard deviation of 0.06 [92].

In an analysis of serious injuries in accidents in which passenger cars were classified as wedge fronted (lower bonnet leading edge) or box fronted (higher bonnet leading edge) (see Figures 10.21 and 10.22), 52% of wedge-shaped cases involved windscreen damage compared to only 33% for the box-shaped cars, and damage to the A-pillar and roof was much more frequent for wedge-shaped cars, implying a different WAD [71]. For box-shaped cars contact occurs more frequently in the rear bonnet area as force transmission in the upper thighs/pelvis area causes torso rotation and secondary impact on the bonnet area. In contrast, in accidents involving

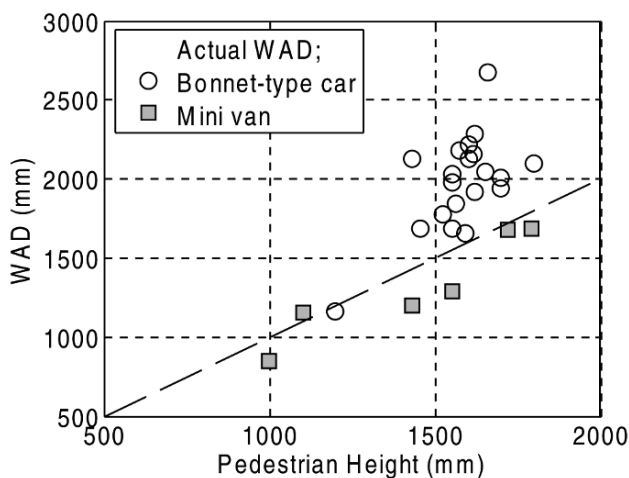


Fig. 10.20 Relationship between WAD [17] and pedestrian height for cars and minivans, adapted from [20].



Fig. 10.21 Example of a wedge shaped vehicle.

wedge-shaped cars pedestrian head contact is more often with the windscreen, the A-pillar or the roof, and this explains the high degree of damage in these vehicle regions [71]. Using a 1500 HIC tolerance limit and a 10 kN pelvis force, a lower limit for the bonnet leading edge of 75 cm to protect the head and an upper limit of 85 cm to protect the pelvis has been proposed [80].

Bonnet Leading Edge Stiffness

Multibody simulations have reported no clear relationship between stiffness of the bonnet leading edge and HIC score [93–95] but have shown that the thigh impact force is almost linearly dependent on the bonnet leading edge stiffness [11]. This finding is clearly explained by equation (10.2), which predicts that injuries associated with the bonnet leading edge contact will be reduced by increasing deforma-



Fig. 10.22 Example of a box shaped vehicle.



Fig. 10.23 Moving the bonnet leading edge rearward: Volvo S40 and Citroën C6, adapted from [65].

tion, i.e. for reduced stiffness. This is particularly relevant for impact with children, where a direct head impact with the bonnet leading edge can occur [73], and HIC scores of over 3000 have been found for headform contact with the bonnet latch [69]. Headlamps and their housing can be very stiff and should be designed as energy absorbing components [1, 58, 66]. In general, the bonnet edges near the wings are stiffer than the central region and accident data showed the risk of AIS2+ femur or pelvis injury was higher at the leading edge of the wing than at the leading edge of the bonnet [76]. For cyclists, the bonnet leading edge caused femur, tibia and fibula fractures, whereas it caused only femur fractures in pedestrians. In contrast, we have already seen that tibia and fibula fractures in pedestrians are associated with contact with the bumper and apron front [51].

Detailed changes to the sheet metal bodywork of the bonnet leading edge are required to reduce stiffness and provide sufficient crush depth. This can be achieved by weakening or moving the under-bonnet reinforcement to facilitate deformation. Round profile vehicles require less alteration than vehicles with high bonnet leading edges [64]. To reduce the stiffness of the bonnet leading edge, some manufacturers have moved the bonnet front-edge rearward, for example the Volvo S40, see Figure 10.23 [65]. In some cases, the headlamps effectively form the bonnet leading edge and these can be made less stiff by using plastic instead of glass and/or by making the lamp housing deformable.

Recently, a bonnet leading edge airbag for pedestrian protection in SUV impacts has been developed [96], see Figure 10.24. This design passed the EuroNCAP upper

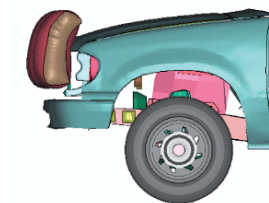


Fig. 10.24 SUV bonnet leading edge airbag, adapted from [96].

legform impactor test (see Chapter 6) and in full scale tests the airbag decreased the risk of chest/pelvis injuries considerably, especially in the chest and abdomen area. Furthermore, simulations of the device showed that it was largely insensitive to the underlying bonnet leading edge stiffness.

Head Impact on the Bonnet/Windscreen

The relative bonnet leading edge height and the relative bonnet length strongly influence where on the vehicle the head contact occurs. For adult pedestrians, high fronted vehicles or passenger cars with long bonnets usually result in a head contact on the bonnet top [82]. In contrast, for adults struck by passenger cars with short bonnets and/or lower bonnet leading edges, head contact usually occurs in the windscreen region [33, 71]. For some combinations of pedestrian height and vehicle shape, head contact with the cowl/windscreen or with the roof can occur. For children, head contact is generally with the bonnet top or the bonnet leading edge.

Head injury likelihood is strongly influenced by the stiffness of the vehicle surface, and the available deformation space before almost rigid components are contacted (see equation (10.2)). The latter is particularly relevant for head impact on the bonnet top, which is a relatively compliant structure, but bottoming out can lead to head contact with stiff engine components [1, 97]. Plastic bonnets are thicker and therefore effectively stiffer than sheet metal bonnets and have a higher risk of head injury [98]. Aluminium bonnets are less stiff than steel, but they can bottom out and this results in a similar HIC score [99]. Collapsible bonnet arresters have been developed [58] and bonnet design has evolved to create a more uniform stiffness profile [66, 100], see Figure 10.25.

Headform impacts on the car body and windscreen [69] have shown a clear dependency of HIC score on dynamic deformation, see Figure 10.26. Confusingly, the data presented in Figure 10.26 regarding the stiffness of the bonnet (hood) relative to the windscreen appears to conflict with the data in Figure 10.19, although both graphs are derived from the same source [69]. Older headform impactor tests at 43km/h showed that approximately 12.5 cm of under-bonnet clearance were required to maintain HIC scores less than 1000 [101], but Figure 10.26 indicates that

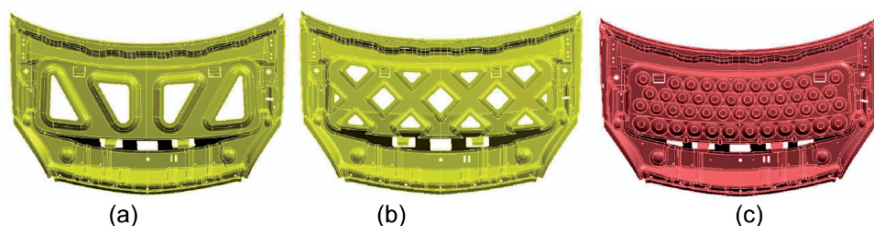


Fig. 10.25 Design of bonnets: (a) traditional, (b) increased number of ribs, (c) multicone design, adapted from [100].

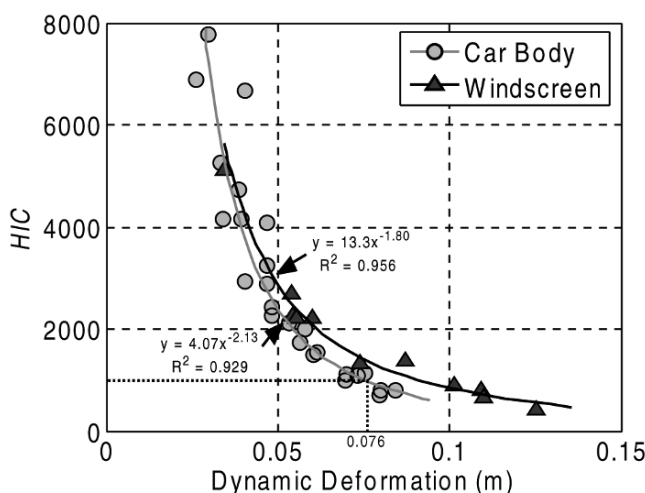


Fig. 10.26 Dependency of HIC score on dynamic deformation from 40 km/h headform tests, adapted from [69].

for more recent cars only about 6.6 cm are required to maintain HIC scores less than 1000. The variation in HIC score for various head impact locations on the bonnet/windscreen of Japanese cars determined using the EEVC headform is shown in Figure 10.27 [69]. However, the HIC score only accounts for injuries based on linear acceleration (see Chapter 5) and recent finite element analysis of pedestrian head impacts on windscreens has demonstrated the importance of assessing angular head kinematics to account for shear strain induced diffuse axonal head injuries [102].

Multibody simulations have been used to investigate the differences between pedestrian and cyclist head impact velocities and angles on car fronts [103]. Figure 10.28 shows the ratios of the cyclist head impact speed and angle to those of pedestrians when struck by the same vehicle. In most cases, cyclist head impact speed and angle are lower than for pedestrians due to the kinematic differences between pedestrians and cyclists described in Chapter 3.

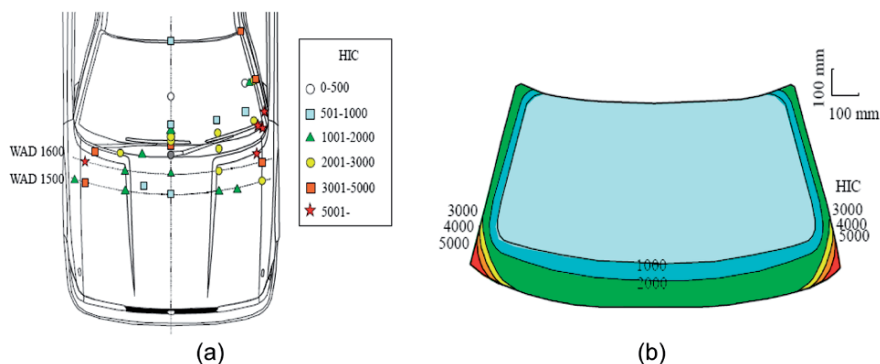


Fig. 10.27 HIC scores using EEVC adult headform for Japanese passenger cars at different levels of WAD, adapted from [69].

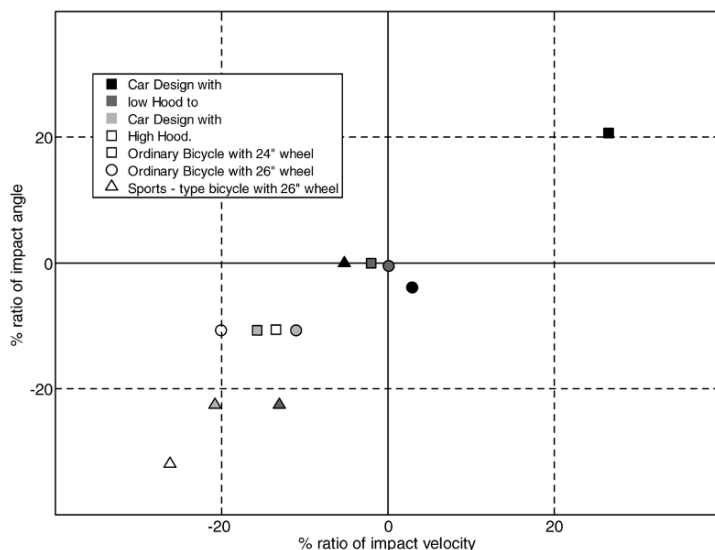


Fig. 10.28 Multibody predictions of the ratio of head impact angle and velocity for cyclists compared to pedestrians when struck by the same vehicle type, adapted from [103]: ratio of impact angle = (impact angle of cyclist/impact angle of pedestrian - 1) $\times$ 100, ratio of impact velocity = (impact velocity of cyclist/impact velocity of pedestrian - 1) $\times$ 100).

The detailed design of the vehicle strongly influences the HIC score, with softer parts often immediately adjoining stiffer elements. The bonnet/fender seam, bonnet hinge and stopper, cowl, corner of the windscreen frame and bottom of the A-pillar are very stiff and lead to high HIC scores [99, 104]. A full cover bonnet has been proposed to reduce head injury risks at the edges [101] and for the cowl area a bonnet overhang has recently shown good protective capability, see Figure 10.29 [105]. For the fender area the Honda CR-V has deformable flanges under the fender-



Fig. 10.29 Exposed cowl and covered cowl, adapted from [105].

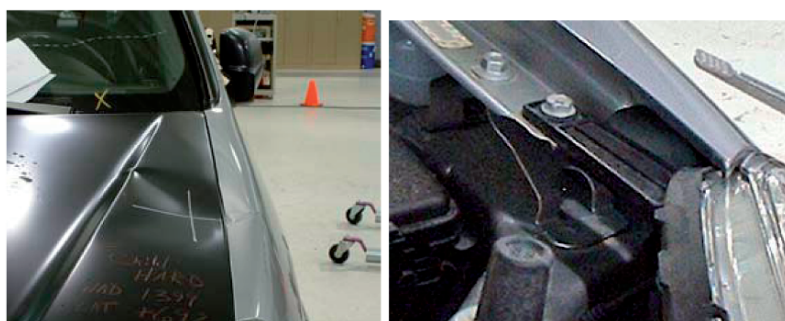


Fig. 10.30 Honda CR-V fender countermeasures, adapted from [105].

hood junction and crush space in adjacent areas of the bonnet reinforcement, see Figure 10.30 [105]. The stiffness of the A-pillar of newer vehicles for occupant compartment protection can result in very high HIC scores unless specific pedestrian protection features are implemented. In contrast, the centre of the windscreen pane is a relatively compliant area for head contact.

Deformation space is crucial for head protection and this is traditionally achieved through under bonnet clearance. However, modern cars are densely packed under the bonnet and several designs for a lifting bonnet in the event of a pedestrian impact have been proposed [58, 106–109] and staged tests with dummies have indicated their ability to maintain HIC scores in head bonnet impacts below 1000 for a 40 km/h impact [107, 110]. This approach is now featured on the recent Citroen C6 and Jaguar XJR production vehicles. In recognition of the high injury potential from head contact with the A-pillars, an airbag has been proposed to lift the bonnet and then cover the A-pillar regions [111, 112], see Figure 10.31. Although there is not yet enough accident data to assess the effectiveness of these devices, there is clearly a strong potential for improved pedestrian protection with these methods [113].

Apart from the stiffness of the corresponding vehicle regions, the head impact severity also depends on the obliquity of the contact, which is determined by the orientation of the bonnet/windscreen relative to the head velocity vector at the instant of contact. Together with neck reaction forces, the impact obliquity and the



Fig. 10.31 Bonnet lift by airbag to protect head from A-pillar and lower windscreen frame contact, adapted from [111, 112].

contact friction determine the effective mass of the head in the impact and the ratio of linear to rotational impulse imparted to the head. For more oblique impacts, the angular impulse to the head is increased and the linear impulse is decreased. Linear accelerations are well correlated with skull fracture and focal brain injuries, while angular acceleration and change of angular velocity of the head of the head are responsible for diffuse axonal injuries [102] (see Chapter 5), and therefore head impact obliquity may influence the resulting head injuries. A 1994 review of passenger cars involved in pedestrian accidents showed windscreen angles ranging between 29° and 54° from the horizontal, and higher injury severity was established for more vertical windscreens [114]. Furthermore, it has been shown in dummy tests comparing head kinematics that the effective mass of the head is higher in SUV impacts compared to passenger car impacts due to the combination of contact angle and neck reaction forces, though the SUV had lower HIC scores due to the reduced head impact velocity for these vehicles [115]. However, this is an area that requires further research.

Evaluation of Production Vehicles Via Proposed Pedestrian Safety Regulations

A 1998 program of 269 tests using the EEVC impactors applied to five passenger cars, an MPV and an SUV showed that most vehicles failed the EEVC Working Group 10 tests [116]. The upper legform and legform impactor performances were particularly poor, and it was concluded that a major redesign of vehicles would be required to pass the tests. The best results were achieved with the head impactor tests, where about one quarter of head locations achieved compliance. By 2001, most cars tested according to the EEVC tests performed badly in the pedestrian tests [64]. The vehicles that did best were Japanese manufacturers, but even they did not pass by 2001.

In a subsequent evaluation of EuroNCAP pedestrian tests (which use the EEVC impactors), it was found that although over 54% of the child headform impacts

Table 10.5 EuroNCAP test results, adapted from [117], # for seven small cars the number of adult headform test was reduced because the test area was small.

Test type	Number of tests	Number of tests passing	Proportion of tests passing	Number of tests passing or within 25%	Proportion of tests passing or within 25%
Child headform	246	91	37%	132	54%
Adult headform #	225	25	11%	56	25%
Upper legform	123	0	0%	1	1%
Legform	122	9	7%	17	14%

Table 10.6 Summary of HIC results by location within tests zone, adapted from [105].

Zone	Average HIC	Standard deviation	Min	Max	<i>N</i>	Pass < 1000	Pass only if in relaxation zone (1000–1700)	Fail (> 1700)
Rear adult	1943	1005	864	4302	21	3	7	11
Middle adult	989	348	536	1443	6	3	3	9
Front adult	698	220	415	1220	14	13	1	9
Rear child	721	506	379	1302	3	2	1	9
Middle child	779	519	309	2307	16	12	3	1
Front child	1846	1472	671	6773	24	6	13	5
Adult	1378	942	415	4302	41	19	11	11
Child	1205	1134	309	6773	43	20	17	6
Total	1374	1110	309	6773	84	39	28	17

passed, only 25% of adult headforms tests were compliant. For the legform tests, only 14% of cases passed and only 1% of the upper legform tests were compliant, see Table 10.5 [117].

More recently, Mallory et al. [105] evaluated 11 vehicles selected to represent the US car fleet with a focus on large passenger vehicles. They performed head impact tests according to the draft Global Technical Regulation (GTR) (see Chapter 6) and found that the peripheral areas of the head impact test zone produced the most severe impact, especially the hinges, cowl, wiper spindles and the wings, see Tables 10.6, 10.7 and 10.8.

The results of the GTR testing of US vehicles do not show a clear connection between vehicle size and head injury risk [105]. HIC scores measured centrally were lower than those measured at the sides, rear and front of the test area. The bonnet hinge location had the highest average HIC value, in particular where there was an exposed hinge with no bonnet covering to dissipate energy. The area adjacent to the cowl, including the wiper spindles, is challenging for pedestrian safety design. The worst performers did not have bonnet overhang over the cowl and one vehicle tested allowed direct contact of the headform with the cowl. The area of the test zone adjacent to the fender was also problematic for many vehicles: none of the four fender area impacts had HIC scores below 1000. The best performer in the

Table 10.7 Summary of HIC results by vehicle model, adapted from [105].

Vehicle	Avg. HIC	Std. Dev.	Min.	Max.	N	Pass (<1000)	Pass only if in relaxation zone (1000-1700)	Fail (>1700)
Hummer H2	2846	2125	909	6773	8	1	3 (none in estimated relax zone)	4
Jeep Wrangler	1945	1505	379	4302	8	3	2	3
Ford E350	1772	1001	868	3993	8	1	5 (3 in estimated relax zone)	2
Dodge Dakota	1303	600	448	2276	8	3	3	2
Toyota 4Runner	1208	685	356	2288	8	4	1	3
Ford Crown Victoria	1063	1052	481	3583	8	5	2	1
Honda CR-V	1044	329	671	1660	7	3	4	0
Dodge Durango	973	519	343	1766	8	5	2	1
Chevrolet Silverado	964	176	740	1274	8	5	3	0
Toyota Tacoma	964	544	309	1814	8	5	2	1
Volkswagen Passat	763	348	378	1302	5	4	1	0

Table 10.8 Summary of HIC results by impacted structure, adapted from [105].

	Impacted Structure	Avg. HIC	Std. Dev.	Min.	Max.	N	Pass (<1000)	Pass only if in relaxation zone (1000-1700)	Fail (>1700)
Peripheral	Hinge	2301	1140	1133	4302	11	0	4	7
	Hood Leading Edge	1892	1578	671	6773	19	5	10	4
	Latch	1815	1020	929	3574	5	1	3	1
	Cowl Area	1500	731	836	2902	9	3	3	3
	Fender Area	1330	349	1048	1774	4	0	3	1
Central	Engine Cover	1186	NA	1186	1186	1	0	1	0
	Hood Ridge	1125	1024	2307	2307	3	2	0	1
	Radiator Cap	1020	258	1203	1203	2	1	1	0
	Handle	909	NA	909	909	1	1	0	0
	Air Intake Box Area	859	NA	859	859	1	1	0	0
	Battery	837	153	729	859	3	2	1	0
	Fluid Cap	833	NA	833	833	1	1	0	0
	Throttle Box	763	NA	763	763	1	1	0	0
	Open Area	603	260	309	1220	22	20	2	0
	Insulator Bracket	536	NA	536	536	1	1	0	0

fender area impact was the Honda CR-V, which has deformable flanges under the fender-hood junction, as well built-in crush space in adjacent areas of the bonnet reinforcement.

Concluding Remarks

This chapter has shown that the mechanisms of pedestrian and cyclist injuries from vehicle impact are now largely understood. Fundamental considerations, accident data and a variety of modelling approaches have shown that the combination of vehicle shape and stiffness have a significant influence on pedestrian and cyclist injury outcome, and it has been comprehensively shown how detailed front-end design can be used to reduce injury risks for pedestrians and cyclists.

Although the basic principles governing the effects of vehicle shape and stiffness on injury outcome are relatively straightforward, there are many confounding factors which complicate the implementation of vehicle design for pedestrian and cyclist safety, and isolating the individual influences of shape and stiffness is a difficult task. Nonetheless, vehicle designers now have the scientific and technical knowledge required to alter most vehicle designs to radically reduce pedestrian and cyclist injury severity in the event of a vehicle collision. This can be achieved through a combination of vehicle shape configuration, structural stiffness reductions in the probable impact zones and the use of safety devices such as lifting bonnets and airbags in regions where sufficient structural compliance cannot be achieved. However, it is clear from the variety of vehicle shapes and stiffness properties present in current production vehicles that there is a large variability in the aggressivity of the current vehicle fleet for pedestrians and cyclists, and an optimum design has not yet been achieved.

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Chapter 11

Conclusions and Future Perspectives

Introduction

This book has provided detailed descriptions of pedestrian and cyclist injuries from vehicle impacts, and has reviewed the principal biomechanical analysis methods that can be applied to reduce the scale and severity of this problem. However, many gaps in our understanding remain, and it is appropriate to summarise the overall level of knowledge in pedestrian and cyclist impacts and to provide perspectives for future research. This is done here under the following headings:

- Epidemiology and in-depth crash injury studies
- Kinematics
- Population variations to injury and disability outcomes
- Modelling and simulation
- Cost benefit calculations of future gains
- Immediate problems of pedestrians and cyclists in low income countries
- Design requirements for future vehicles
- Requirements for a global pedestrian/cyclist friendly standard

Epidemiology and In-Depth Crash Injury Studies

The World Health Organisation estimated in 2004 that 1.2 million people are killed in road traffic crashes annually and that up to 50 million are injured [1]. They also predicted that by 2020 road traffic fatalities will be the third leading contributor to the global burden of disease. These startling statistics are heavily influenced by the high proportion of pedestrians and other vulnerable road users in fatal road accidents in less motorised countries. As shown in Chapter 2, in many African and Asian countries pedestrians account for over 40% of road accident fatalities. Therefore, attempts to reduce global road accident fatalities must include a focus on pedestrian, cyclist and other vulnerable road user protection. This is especially so for countries

like India and China which have very high populations and rapidly increasing levels of motorisation.

This book has assessed the biomechanical aspects of pedestrian and cyclist impact, but there are also significant road engineering, legislation and policy factors which need to be addressed, especially with reference to requirements in less motorised countries. This cause has been championed in recent years by Dinesh Mohan and Murray Mackay [2–4], who have pointed out that most existing pedestrian safety research has focused on the front-end design of passenger cars, as these are the principal collision partners for vulnerable road users in highly motorised countries. However the majority of vulnerable road user fatalities in countries like India result from impact with vehicles other than passenger cars. These include buses and trucks, but also motorcycles and indigenous vehicles which have not been designed for pedestrian safety. Therefore, implementation of a Global Technical Regulation for pedestrian safety (Chapter 6) for all motorised vehicles should have a significant protective effect for both pedestrians and cyclists.

Detailed accident databases in highly motorised countries have shown that it is generally possible to attribute pedestrian and cyclist injuries to contact with specific vehicle parts and to distinguish between vehicle and ground contact as the source of injury. These databases are invaluable for directing legislative and vehicle design efforts to reduce the severity of pedestrian injuries in highly motorised countries. However, the disparity between these research findings and the needs in less motorised countries is compounded by the much poorer availability of detailed accident data in the latter. In fact, the majority of pedestrian fatalities occur in countries which do not routinely collect detailed road accident statistics, and traffic conditions in these countries are often totally different from those pertaining in highly motorised countries, either today or at any stage in the past. This information deficit in countries where it is most needed must urgently be addressed if worldwide pedestrian and cyclist fatalities are to be significantly reduced. Simply importing safety strategies from Europe or America in Asia and Africa is a false economy. Even within Europe there are large variations in road accident fatality rates and it is no coincidence that Ireland, which does not routinely collect detailed accident information, has a significantly poorer road safety record than the UK, which does.

For those countries which do collect detailed pedestrian accident data, the manner in which the injury statistics are collated and analysed has the tendency to mask the diversity of injury mechanisms. For example, the IHRA database of AIS2–6 pedestrian injuries from the US, Germany, Japan and Australia (Chapter 2) is generally used to conclude that the principal body regions of interest for pedestrian impact are the head and legs [5, 6]. However, the same data also shows that the midbody (pelvis, abdomen and chest) account for over 20% of AIS2–6 pedestrian injuries. Therefore, although pedestrian midbody injuries from impacts with passenger cars have been reducing due to improvements in bonnet leading edge height and stiffness, the midbody region remains a critical aspect of pedestrian injury when all pedestrian impacts with different vehicle types are considered. A consequence of this is that the upper legform subsystem test should definitely be included in the current Global Technical Regulation.

Similarly, the wide variety in vehicle pedestrian impact configurations is generally not considered and staged tests with dummies and cadavers and even computer simulations have overwhelmingly studied the idealised case of a stationary adult pedestrian struck from the side by a vehicle in the mid-bonnet/bumper region (Chapter 8). In these cases, depending on the shape of the vehicle, lower/upper leg injuries arise from contact with the bumper, upper leg/pelvis injuries from contact with the bonnet leading edge and head injuries occur through contact with the windscreen glass or bonnet top. In fact, the IHRA data shows that over 30% of head injuries occur from other contacts, of which the majority involve the windscreen frame or A-pillars. Therefore a significant proportion of accidents occur with peripheral regions of the vehicle and result in injuries to body regions that are not the head and legs, and further focus on the diversity of actual pedestrian impact cases is required. Such an approach should also yield additional safety benefits compared to further focus on the idealised case of a stationary side-struck pedestrian contacted in the mid-bonnet/bumper region.

Accident databases need to be carefully assessed to fully understand the influence of protective measures. For example, the evidence from Japan is that, although pedestrian fatalities reduced significantly between 1990 and 2000 [7], the number of seriously injured pedestrians increased at a comparable rate. This shows that many people are now surviving accidents which were previously fatal. This may be seen as a positive development, but further assessment of the long-term prognosis of these serious injuries and the resulting quality of life of survivors is needed. Estimates from high income countries suggest that 80% of the total cost of motor vehicle crashes is attributable to non-fatal events [8], and this is largely due to the very high societal cost of permanent disabilities.

Kinematics

The trajectories of pedestrian impact in the case of a stationary pedestrian struck from the side by a passenger car in the mid-bumper/bonnet region are now very well understood, in large part due to staged tests performed using cadavers. However, the primary reason for focussing on pedestrian/cyclist kinematics is injury evaluation and most established injury criteria relate to force or acceleration of specific body regions (head, pelvis etc) during impact. However, in the majority of staged tests only trajectories are measured using high-speed video. Therefore, further staged pedestrian and cyclist impact tests in which the principal cadaver body regions are instrumented with triaxial accelerometers and strain gauges and the vehicle contact regions are instrumented with load cells would provide very significant additional experimental data for the purpose of understanding pedestrian and cyclist impact kinematics. Furthermore, analysis of accident configurations (Chapter 3) shows that the ‘idealised’ case of a passenger car striking a pedestrian from the side by the mid bumper/bonnet region represents only a small number of actual accidents. Therefore, although it is known from staged tests using crash dummies and from compu-

tational models that significant variations in pedestrian trajectories occur for different vehicle types and initial pedestrian stance and position, further staged tests using cadavers are required to quantify the effect of these differences on injury outcome in a robust manner. These staged tests would also facilitate further corroboration of detailed computational models (see Chapter 8), and also verify the ability of optimisation algorithms to predict pedestrian and vehicle pre-impact conditions for accident reconstruction purposes [9].

For wrap trajectory pedestrian and cyclist impact cases, the possible further interaction of the body with the vehicle following head contact is significant for the total pedestrian/cyclist projection distance, and further research is required to clarify this. Due to the generally increased cyclist speed relative to pedestrians (Chapter 3), interactive effects between cyclist speed and the level of impact engagement with the vehicle are also possible. There are currently conflicting findings about the influence of cycling speed on projection distance from vehicle impact, and further research is required.

The kinematics of ground contact are known to be highly variable, and significant additional work using analysis of accident data and staged tests using cadavers is required to understand ground contact injuries more fully. In particular, the relationship between the vertical and horizontal head velocity at the instant of ground impact is unknown, and this is significant for the kind of head injuries which result. There is no evidence of differences between pedestrian and cyclist ground contact mechanisms, but there is growing evidence indicating that high fronted vehicles result in more severe head injuries in the ground impact. This needs to be evaluated urgently in the context of the growing number of SUVs in many highly motorised countries. The obvious solution for reducing ground related injuries is to contain the pedestrian on the vehicle front and thereby eliminate the ground impact. However, although this concept was first proposed several decades ago, so far the disadvantages associated with proposed containment devices have outweighed the potential advantages, and they have not been implemented in any production vehicles. Further effort in this area is desirable.

Population Variations to Injury and Disability Outcomes

The relationship between mechanical loading and injury and disability outcome is crucial for the science of impact biomechanics. We now have a generalised understanding of hard and soft tissue trauma, and injury criteria for the main body regions based on global force and acceleration measures were reviewed in Chapter 5. However, significant further work is required for the formulation of injury criteria which are more specific to age and sex and the state of health of a person, and which correlate injury to tissue level strains. Prediction of tissue level injuries will require a breakthrough in experimental techniques, or a continuation of the existing trend of inverse prediction using finite element methods. In the meantime, the available models for injury prediction are crude, and prediction of injuries for individual cases

cannot be made with any confidence. More reliable injury criteria are therefore required for future evaluation of specific injury reduction measures (such as headlamp layout and A-pillar trim) for pedestrians and cyclists.

Modelling and Simulation

The Polar pedestrian dummy is a much more useful research tool than the various pedestrian subsystem component tests, because tests using a whole body dummy implicitly capture the influence of overall vehicle shape on pedestrian impact dynamics. The focus to date has been on improving the ability of the Polar pedestrian dummy to reproduce cadaver impact trajectories and head velocity time-histories. While a good dummy trajectory is clearly a prerequisite, validation of parameters such as head acceleration and knee shear force/bending moment used in injury criteria evaluation is a necessary next step. Considering the simplifications required, physical dummy testing is unlikely to be able to predict tissue stresses/strains, and therefore injury predictions using a dummy are limited to rigid body parameters. Nonetheless, a pedestrian dummy which reproduces not only the movement but also the principal rigid body dynamics of pedestrian and cyclist impacts would be a very powerful tool to assess the influence of speed, shape and stiffness on pedestrian injuries as well as the effectiveness of proposed injury countermeasures. Such a dummy could also be used in the validation of multibody computer simulations and in the refinement of subsystem component tests.

Detailed finite element models of pedestrian and cyclist impact have the potential to predict internal body loading parameters such as tissue level strain. These models require a description of the geometry and the structural/constitutive behaviour of the body. Advances in medical imaging techniques have facilitated the development of very sophisticated geometrical models, but representation of the dynamic compressive characteristics of biological soft tissues, especially muscles, remains a topic of scientific pursuit. There are significant differences between the tensile and compressive properties for many biological tissues and, together with the need for damage algorithms, these complexities mean there is a real risk of incorrect predictions of tissue stress/strain resulting from simplifications such as the frequent assumption of linear elastic isotropy for the behaviour of human soft tissues.

Furthermore, validation of existing finite element pedestrian models has largely been limited to prediction of whole body trajectories measured in cadaver tests and component level acceleration and force time-histories. These are necessary but not sufficient conditions for validation of a finite element model which will be used for tissue level stress and strain predictions. Therefore, to validate a finite element model as a tool which can tell us more about pedestrian or cyclist impact than a multibody model can, finite element model should be able to predict some experimental stress or strain measurements. Furthermore, due to the very large number of degrees of freedom in finite element models, the input parameters (material constants, etc.) should not be based on the same experimental dataset used to validate

the model. This is currently not generally the case and modelling of vehicle specific and pedestrian specific collisions using finite element models will not really be possible with any degree of confidence until this is rectified. Therefore, pedestrian and cyclist experiments recording tissue strains in human subjects are urgently needed for further model validation.

Cost Benefit Calculations of Future Gains for Pedestrians and Cyclists

Pedestrian and cyclist accidents need to be studied and documented like any other disease to facilitate the application of appropriate countermeasures, especially in less motorised countries where many of these accidents occur [7]. In general, the benefits to society of road safety research are large and outweigh the costs of the research and the road safety measures themselves [10, 11]. However, cost benefit analyses specific to pedestrian and cyclist safety measures are limited [8] and complicated by the need to account for changes in highly motorised countries in the amount and the time spent in walking and cycling and changes in road user sense of security and state of health [12]. For example, some of the reduction of pedestrian fatalities in motorised countries in recent decades simply reflects the lesser amount of walking that takes place [13], rather than intrinsic improvements in safety.

Due to the very strong relationship between pedestrian and cyclist impact speed and injury/fatality risk, the biggest benefits in all world regions would be achieved by reducing vehicle impact speeds. Ultimately, separation of vulnerable road users from fast moving motorised traffic is a key priority and the highly successful Swedish Vision Zero policy recommends that vehicle travel speeds in areas with pedestrians should be limited to 30 km/h [14]. In cases where separation is not possible, the introduction of automatic braking systems and intelligent speed adaptation should assist with reducing impact speeds [15]. However, a review of policy in Norway and Sweden found that the main source of inefficiency in current road safety policies relates to vehicle safety standards [16]. Therefore, regulation of vehicle front-end design in all countries would also bring significant further reductions in pedestrian and cyclist injuries.

Immediate Problems of Pedestrians and Cyclists in Low Income Countries

Richer countries can clearly afford to spend more on road traffic accident prevention. However, there is a strong irony in this as the health and economic burden of road traffic injury is significantly higher in low and middle-income countries than in high income countries [17]. It is therefore imperative that the strategies which are implemented are cost effective, and preventive measures in low income countries

should therefore focus on high-risk demographic groups. In particular, despite the huge increase in the level of motorisation in China and India, walking and cycling remain the dominant modes of transport, and over 60% of traffic-related deaths in these countries are pedestrians and cyclists [3, 18].

The three main priorities should be (1) separation of fast flowing traffic from pedestrian and cyclists, (2) reduction of vehicle travel speeds (and hence impact speeds with pedestrians and cyclists) where this separation is not possible and (3) implementation of pedestrian friendly front-end design in those vehicles which are most likely to collide with pedestrians and cyclists.

Separation of Fast Flowing Traffic from Pedestrians and Cyclists

The risks associated with importing road safety strategies from Europe and the USA into less motorised countries have already been described. In China, re-engineering of the road infrastructure systems based on best practice in high-income countries led to the elimination of roads that once provided separate lanes for cyclists and other vulnerable road users to accommodate the increasing motorised vehicle volumes. This is particularly unfortunate as the high-income countries that originally advocated rapid motorisation are now attempting to re-engineer their own road systems to separate the vulnerable road users from motorised traffic [18].

Reduction of Vehicle Impact Speeds with Pedestrians and Cyclists

The overwhelming evidence from fundamental mechanics and accident investigation is that pedestrian and cyclist injury risk depends strongly on vehicle impact speed (Chapters 2 and 7). Therefore, in cases where separation of motorised traffic from vulnerable road users is not possible, policies which reduce the impact speed are likely to be most effective. As recommended by the Swedish Vision Zero policy, vehicle travel speeds in areas with pedestrians should be limited to 30 km/h [14]. The benefits of reducing vehicle impact speeds on pedestrian fatalities has been clearly shown in Figure 2.6 which shows a sharp increase in fatality risk for pedestrians at vehicle impact speeds in excess of 40 km/h. Conversely, fatality risk is less than about 10% for impact speeds below 30 km/h.

Implementation of Pedestrian Friendly Front-End Design in Those Vehicles Which Are Most Likely to Collide with Pedestrians and Cyclists

This book has shown in detail that vehicle front-end design has a significant influence on pedestrian and cyclist injury risk (chapter 10), and significant advances have been made in the design of passenger cars. However, in less motorised countries pedestrians and cyclists are much more frequently struck by vans, pickup trucks, buses and indigenous transport vehicles [19]. Therefore, improving the front-end design of these kinds of vehicles would be enormously beneficial in the cities of developing countries [2]. In particular, urban buses operate in a road environment full of vulnerable road users and have working lives that are substantially longer than cars. Therefore, a specific design for urban buses and trucks is needed [2].

Design Requirements for Future Vehicles

This book has shown that the combination of shape and stiffness of vehicle front-ends is crucial for pedestrian and cyclist injuries. However, although the basic principles governing the effects of vehicle shape and stiffness on injury outcome are relatively straightforward, there are many confounding factors which complicate the implementation of vehicle design for pedestrian and cyclist safety, and isolating the individual influences of shape and stiffness is a difficult task. Nonetheless, we now have a significant understanding of the influence of bumper geometry on leg injuries, the effect of the bonnet leading edge height on midbody loading and on overall body kinematics, and we know from evaluations of production vehicles that significant differences in head injury risks arise due to localised hard spots in joints and seams.

Designers of future vehicles clearly need to take account of these findings to create vehicles with rounded shapes and minimised hard spots in the likely contact zones. This can be partly achieved by setting structural members back from actual contact surfaces, especially in the bonnet leading edge, bonnet top and A-pillar areas. However, vehicle designers should also harness emerging technologies to provide maximum protection for both adults and children across the broad range of real collision configurations. In particular, there is the potential to use active components such as airbags on the vehicle exterior and lifting bonnets in combination with advanced sensor technologies to provide an ‘accident specific’ response which accounts for impact speed, pedestrian/cyclist anthropometry and stance and can thereby effectively minimise pedestrian and cyclist injuries. This approach could make the vast majority of pedestrian and cyclist accidents non-fatal. However, it is clear from the variety of vehicle shapes and stiffness properties present in current production vehicles that there is a large variability in the aggressivity of the current vehicle fleet for pedestrians and cyclists, and an optimum front-end design has not

yet been achieved. It is hoped that this book has provided the means to substantially address this problem.

For minimising ground contact injuries, further work on the influence of vehicle design is required and it remains unclear whether the future development of containment devices can overcome the practical difficulties which have so far been encountered. However, since the relative severity of ground contact compared to vehicle contact injuries is higher at low impact speeds, devices aimed at containing struck pedestrians and cyclists on the vehicle front are likely to be of most practical benefit for low velocity collisions. This is encouraging since pedestrian containment on the vehicle is more realistic at low speed than at high speed [20].

For flat fronted vehicles, apart from reducing front-end stiffness, an important requirement is low ground clearance, as this reduces the risk of overrun by the vehicle. This is effectively achieved with many urban buses in highly motorised countries, and is in direct contrast to commercial lorries.

Requirements for a Global Pedestrian/Cyclist Friendly Standard

A motor vehicle standard governing front-end design for pedestrian/cyclist protection which is adopted globally has the potential to significantly reduce the number of vulnerable road users killed or seriously injured in traffic accidents. Primarily for economic reasons, all existing pedestrian safety standards (including the proposed Global Technical Regulation [21]) are based on subsystem component tests. However, the choice of leg and headform impactors in lieu of a whole-body dummy makes it almost impossible to determine the net change in pedestrian or cyclist risk resulting from a specific vehicle design alteration [6]. Furthermore, although accident data from the USA shows that a large number of mid-body injuries occurs due to the prevalence there of light trucks and vans [22, 23], there is currently no ISO or IHRA upper legform test. Therefore, in the long term a Global Technical Regulation for pedestrian and cyclist safety should be based on whole body tests. This will require further development of the biofidelity and validation of the Polar or an alternative pedestrian dummy, and should include development of a family of dummies representing the range of human anthropometrics. A more detailed assessment of the range of pedestrian and cyclist impact configurations is also needed, as it is known that the idealised side struck case does not adequately cover the variation occurring in real accidents.

The ability of computational models to capture the dynamic response of crash test dummies is now very well established and, due to their lower costs compared to physical tests, an economically viable Global Technical Regulation for pedestrian and cyclist safety might be developed in which a combination of physical tests and numerical simulations are used to assess vehicle front-end safety. This might be done in the following manner: a full-scale crash test at 40 km/h is performed in which either an adult male or female pedestrian dummy is symmetrically side-struck by the vehicle and injury criteria in the lower limbs, midbody region and head are

evaluated. This full-scale test is then used to provide vehicle validation data for benchmarked computational models which can then be used to simulate a matrix of different impact tests involving adults and children in a variety of initial collision configurations. Of course, significant further work to assess the feasibility of this approach would be required.

Conclusion

It is clear from this summary that there are two factors which would yield the most significant improvements in pedestrian and cyclist impacts. The first is restricting vehicle speeds to below 30km/h in urban areas where the volume of vulnerable road users is high. The second is a mechanism on the front-end of vehicles which could retain pedestrians and cyclists on the vehicle in the event of a collision and thereby prevent subsequent ground contact injuries. These two factors, combined with existing efforts at creating pedestrian friendly vehicle front-ends, would significantly reduce the risk of serious injuries or fatalities for pedestrians and cyclists.

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