Biological and Medical Physics, Biomedical Engineering

Sergey Ermakov
Alexandr Beletskii
Oleg Eismont
Vladimir Nikolaev

Liquid Crystals in Biotribology

Synovial Joint Treatment

Springer

Liquid Crystals in Biotribology

BIOLOGICAL AND MEDICAL PHYSICS, BIOMEDICAL ENGINEERING

The fields of biological and medical physics and biomedical engineering are broad, multidisciplinary and dynamic. They lie at the crossroads of frontier research in physics, biology, chemistry, and medicine. The Biological and Medical Physics, Biomedical Engineering Series is intended to be comprehensive, covering a broad range of topics important to the study of the physical, chemical and biological sciences. Its goal is to provide scientists and engineers with textbooks, monographs, and reference works to address the growing need for information.

Books in the series emphasize established and emergent areas of science including molecular, membrane, and mathematical biophysics; photosynthetic energy harvesting and conversion; information processing; physical principles of genetics; sensory communications; automata networks, neural networks, and cellular automata. Equally important will be coverage of applied aspects of biological and medical physics and biomedical engineering such as molecular electronic components and devices, biosensors, medicine, imaging, physical principles of renewable energy production, advanced prostheses, and environmental control and engineering.

More information about this series at http://www.springer.com/series/3740

Sergey Ermakov · Alexandr Beletskii
Oleg Eismont · Vladimir Nikolaev

Liquid Crystals in Biotribology

Synovial Joint Treatment

Springer

Sergey Ermakov
V.A. Belyi Metal-Polymer Research Institute
National Academy of Sciences of Belarus
Gomel
Belarus

Oleg Eismont
Republican Scientific Center for
Traumatology and Orthopedy
Minsk
Belarus

Alexandr Beletskii
Republican Scientific Center for
Traumatology and Orthopedy
Minsk
Belarus

Vladimir Nikolaev
Department of Traumatology and Orthopedy
Gomel State Medical University
Gomel
Belarus

ISSN 1618-7210 ISSN 2197-5647 (electronic)
Biological and Medical Physics, Biomedical Engineering
ISBN 978-3-319-20348-5 ISBN 978-3-319-20349-2 (eBook)
DOI 10.1007/978-3-319-20349-2

Library of Congress Control Number: 2015943358

Springer Cham Heidelberg New York Dordrecht London

Printed on acid-free paper

Springer International Publishing AG Switzerland is part of Springer Science+Business Media (www.springer.com)

Preface

The book is given over to study of nature of abnormally low friction and wear of human and animal joints. Trends in joint function and lubrication, possibilities of rheological correction and cartilage mechanodestruction prophylaxis during arthropathies are generalized.

The complex researches of cartilage friction process in natural and artificial lubricants that lead to breakthrough understanding of joint boundary lubrication nature take a large part of the book. It has been proved that liquid-crystal state of synovial fluid, such as cholesteric-nematic crystals, plays an essential role in intra-articular friction decrease. The results of this discovery have fundamental and applied meaning. They greatly expand up-to-date understanding of the role of liquid crystal in biological tribosystem function and expose a completely new trend of joint lubrication properties research.

Creation of novel pharmaceuticals—artificial synovial fluid reproducing lubrication mechanism inherent to natural synovia—a new and important balance of the work.

Experimental and clinical data on chondroprotective efficiency of preparation are of practical interest for further research in the field. It should be noted that only a combination of biological, technical, physical, chemical and medical knowledge makes it possible to investigate the lubrication mechanism of joint cartilage and the prophylaxis methods of premature wear.

The book should be of interest to communities of scientific workers, practicians and students of medicobiological and technical specialties. It will draw the attention of researches to problems of biotribology, chondroprotection, liquid-crystal biological environment function and creation high-performance materials for endoprosthetics.

Gomel, Belarus — Sergey Ermakov
Minsk, Belarus — Alexandr Beletskii
Minsk, Belarus — Oleg Eismont
Gomel, Belarus — Vladimir Nikolaev

Contents

Acronyms and Nomenclature

ANS	8-aniline-1-naphthylsulphonate (fluorescent probe)
BS	Blood serum
C	Cholesterol
CE	Cholesterol ester
CL	Common lipids
DMSO	Dimethylsulfoxide
HUA	Hyaluronic acid
LC	Liquid crystal
LCCC	Liquid-crystal cholesterol compound
MP	Medicinal preparations
Na-CMC	Sodium carboxymethyl cellulose
NCID	Non-steroid counter inflammatory drugs
NSA	Non-steroidal anti-inflammatory drugs
PVP	Polyvinyl pyrrolidone
RF	Rheumatoid factor
RSR	Relative specific radioactivity
SAA	Surface active agent
SCMC	Sodium carboxymethyl cellulose modulus of tension
SF	Synovial fluid
SR	Specific radioactivity
TG	Triglycerides
E	Modulus of tension
F_a	Adhesion component of friction
F_c	Cohesion component of friction ия
F_f	Friction force
G	Coefficient of rigidity
I_{fl}	Fluorescence strength
K	Forces of action of body weight part

K_{22}	Elastic constant
M	Forces of action of lateral abductor muscles
P	Reliability
Q	Liquid volume flow
R	Resulting compressive forces
S	Liquid crystal order parameter
S_5	Body center of gravity
T	Temperature
W	Free energy of liquid crystal
W_s	Surface energy of liquid crystal
W_e	Elastic energy liquid crystal
a	Cholesterics molecular layer thickness
b	Extrapolative path
d	Thickness of liquid crystal
h	Thickness of lubricant layer
h_i	Range of sinusoidal relief
h_l	Thickness ratio of lubricant layer
Δh	The thickness of the specimen
k	The coefficient of permeability
Δl	Deformation
$\overrightarrow{n}$	Unit vector (director)
p	Pressure
$\tilde{p}$	Dimensionless pressure
Δp	Differential pressure
S	Pitch of the helically-coiled cholesteric
t	Time
v	Speed
Φ	Pressure in the fluid
Θ	Calorific endothermal effect
Ψ	Pressure in the articular cartilage
α	Angle
η	Viscosity
φ	The angle of molecular orientation in cholesteric
λ	Wave-length of sinusoidal relief
θ	The angle of director orientation $\vec{n}$ relatively to the instantaneous direction of the cholesteric major molecule axis
ρ	The radial measure in the cartilage
τ	Dimensionless time parameter

Introduction

Synovial joints is a unique biological body of movement (motion). They can function under significant live loads, at high and low speeds for a long time and not wear for lifelong. A study that deals with the design, friction, wear and lubrication of interacting surfaces in relative motion is called tribology. In spite of considerable progress in tribology in recent years, neither technical unit exhibits the same friction characteristics as natural joints do. Therefore, synovial joints are of great interest to present and future tribology. Investigation of such complicated processes can be made at the intersection of disciplines about the physical and chemical properties of rub surfaces, the interaction of a solid phase and its field with atoms and molecules of various substances, mechanics of solids and liquids and others. Thus, tribology is an up-to-date cross-disciplinary science that encompasses various aspects of the world.

Biotribology is a medical-biological branch of tribology. It studies friction interaction of biopolymer composites having unique structure—cartilages in the synovial joint of humans and animals. Comprehension of joint tribology opens up great perspectives in the destruction pathogenesis during trauma and arthropathies. Thus, one can also speak of development of new methods for effective cartilage protection against premature destruction and wear.

Biotribology arose with swift advance in the past five decades. A significant amount of information about friction and wear features of joint cartilage in health and disease, importance of joint structure specificity, joint surface macro- and microgeometry, biochemical composition and rheology of synovial fluid for load transfer and decrease, joint kinematics and lubrication has been collected.

It is doubtless that the biotribology developmental pathway is closely associated with researches in frictional interaction of various materials and units in engineering. Simultaneously, researches in engineering cannot be automatically used in tribology of synovial joints. It is impossible to describe their abnormally low friction and wear nowadays.

Meantime, the authors think that abnormally low friction of joints is conditional upon synovial fluid lubricity. The last achievements in physics of liquid crystals and

the obtained data on their unique lubricant properties are very promising and encouraging.

It has been found that liquid-crystal state is inherent to a number of biological tissues and matrixes, but there were no similar evidence for synovial fluid. Therefore, research of cholesteric liquid crystals and definition of their role in natural lubrication is considered to be a brand new scientific approach to further knowledge of the nature of articulate cartilages abnormally low friction, and also to development of new arthropathy treatment mode.

Intensive biotribological researches are carried out mainly in England, the USA, Germany and some other countries in recent years. Similar works are also conducted at research institutes and laboratories in Gomel, Minsk, Moscow, Vilnius, Kiev and Ivanovo. The primary goal of this study is to create artificial synovial fluid for destruction prophylaxis and new effective biopolymer self-lubricating materials for joints endoprosthetics. However, the available scientific works and publications on biomechanics and tribology of the joint synovial have incidental character, and are insufficiently generalized.

In the work the authors made an attempt to analyze the available literature data on structure and function features of synovial joints, lubrication mechanisms, pathogenesis of cartilage mechanodestruction and opportunities of its prevention with known pharmaceuticals and physical factors.

Our own study of cartilage friction and wear, experimental and theoretical proofs of interfacial liquid-crystal state of synovial fluid, justification for new intraarticular friction decrease concept and a new scientific field for creation of artificial synovial fluid and new arthropathy treatment mode take up a large part of the book.

The authors hope that this study will be of interest and use for not only arthrology, biomechanical and tribology experts, but also the engineering, medical and biological communities.

Chapter 1
Biomechanics of Joint Synovia

Abstract In this section summarizes the results of research into the field of living synovial joints and structural elements of the environment. Construction features of synovial joints are analyzed as of kinematic pairs capable of operation under considerable loads. Molecular and supermolecular structural arrangements of cartilage matrix and their influence on the deformation properties of cartilage tissue are considered. The latter is underlined to act as a specific damper of dynamic loads which participates in uniform distribution of pressure in joints. An analysis is presented of synovial fluid both as a lubricating medium and an active structural element of joints performing important biomechanical functions together with the cartilage tissue.

Everyday life of human and animals is known to be continuous motion and it is inseparably linked with normal functioning of the synovial joints [1–3]. Exceptional reliability and durability of joints as natural friction elements depends directly on the nature of their structure, mechanical properties and principles of functioning. Their main structural elements are two mated bone contacts coated with hyaline cartilage that is enclosed in the articular lined with the synovial membrane and filled with the synovial fluid [1].

According to modern ideas, the entire of complex of structurally differentiated, yet morphologically integral and functionally combined tissues is conventionally termed as “the synovial medium in joints”. Only interaction and interdependence of the elements of the articulation medium, like the synovial membrane, synovial fluid, and articular cartilage, are capable to produce optimal biophysical conditions, perform metabolic processes between the articular cavity and blood vessels and frequently to maintain high performance of joints in extreme conditions and for prolonged periods. This idea about the synovial medium of as an integral organ specific system is fundamental in studying any matters of functioning of the joints.

S. Ermakov et al., *Liquid Crystals in Biotribology*,
Biological and Medical Physics, Biomedical Engineering,
DOI 10.1007/978-3-319-20349-2_1

1.1 Kinematics of Synovial Joints

At present, the anatomic structure of synovial joints is rather well known (Fig. 1.1). According to modern ideas, the synovial joints are dynamically loaded movable articular members of the skeleton in which (translational and rotational) motions take place various in scopes and degrees of freedom. The variety forms of joints of humans and animals makes it impossible to imagine their functional features by a single-unified model [4].

The hip joint is a classic example of the ball articular joint with three degrees of freedom. It is a compact union of the femoral head almost spherically shaped ball and the pelvic acetabulum shaped hemispherically that form a multiaxial system with an extensive scope of motions. Therefore, the joint in the average physiological position is capable to contract and extend in the sagittal plane within 140°, to retract and extend the extremity in the frontal plane within e 50°, to rotate it around the hip longitudinal axis through 50° in the horizontal plane. The motions are limited by the tension in the capsule, ligaments and muscles and can vary considerably in response to the joint spatial position, e.g., when the hip is bent, moved away, etc.

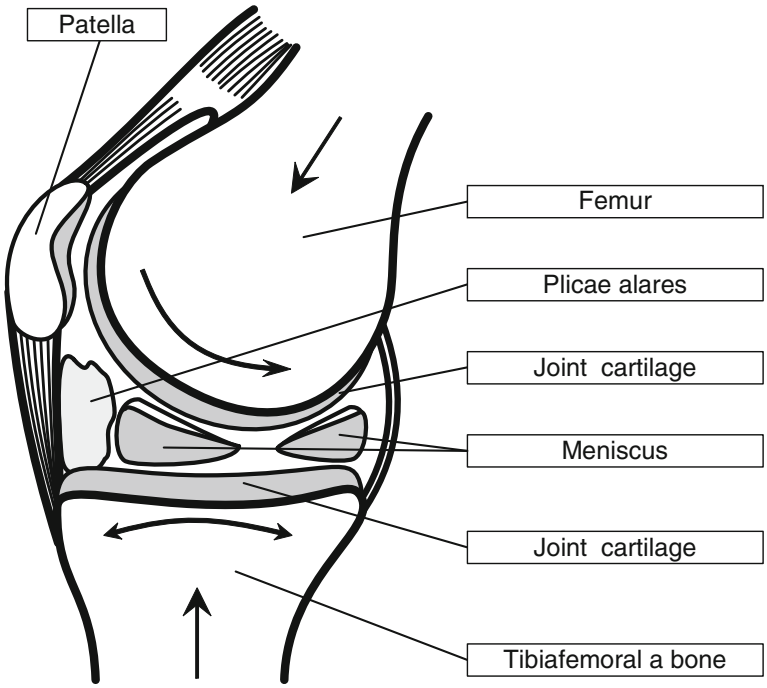


Fig. 1.1 Anatomic structure of knee joint

The knee joint performs two main independent motions: flexion-extension movements of the tibiofemoral joint within a sufficiently large range (up to 140°) in the в sagittal plane and limited rotational movements (up to 10°) of the shinbone in the horizontal plane at the terminal extension stage. When deflecting the knee through over 45°, the shin relaxes and the limiting functions of the capsule joint apparatus can increase sizably.

In addition to these complicated joints, other joints almost coincide with a sheave. The talocrural joint can be likened to two cylinders resembling a well-fitted radial bearing with a single degree of freedom practically and limited rotation limited by the articular apparatus. The joint between the shoulder and elbow bones has a similar structure serving just to bend and extend the forearm in the elbow. In this case, the extra lateral movement of bones is impossible because the collateral ligaments are close to the axis of rotation and the sliding surfaces of this sheave-like "pivot" have a specific anatomic structure.

This different natural design of the joints is certainly dictated by their specific locomotor functions and simultaneously the need to maintain high mechanical properties [5]. Therefore, from the viewpoint of the mechanics, the joints of animal and human extremities are very effective kinematic pairs capable to withstand considerable loads during prolonged period.

Investigation of the principles of the structure and functioning of different elements of the human and animal locomotorium revealed that both the position and the resulting forces affecting on partial elements of joints and neighboring bones [6, 7] determine their physiological loading in natural conditions. The resulting force pressing on the head of the hip joint is composed of the body weight and muscles that stabilize the pelvis normally in case the body center of gravity displaces (Fig. 1.2).

The line of the resulting effect passes medially across the hip head and laterally to it at a vertical angle 15–20°. As a result, the diagram of loading of the hip joint has the peak loadings acting on the hip head and exceeding the body weight three to nine times, i.e. the order of magnitude of 700–6300 N. These peaks appear at extreme amplitudes of oscillations, then the loading drops to the minimum in the sweep middle. Thus, the hip joint has to withstand during a year up to 2.5 million of cyclic loads and accomplish a considerable scope of movement, such as bending-unbending within a range of 45–60°, extension-contraction at the same time within a range of 11–12°, rotation through 6–14°, at a relative sliding speed of articular surfaces in different periods of a walking cycle 0.01–0.05 m/s.

The biomechanics of other elements of the human and animal locomotorium have its own specific features [8, 9]. The hip joint bends to almost 160° [10, 11], however, during normal ambulation the bending ranges just within 60–67° [12]. The shinbone in the unsupported phase is in the position of external rotation in respect to the hip condyle. At the moment the foot reaches support when the knee fully unbends, it acquires active and passive stabilization by the articular, capsular, and muscular apparatus. When the hip condyle reaches the terminal unbending stage the condyles rotate internally in respect to the tibial plateau.

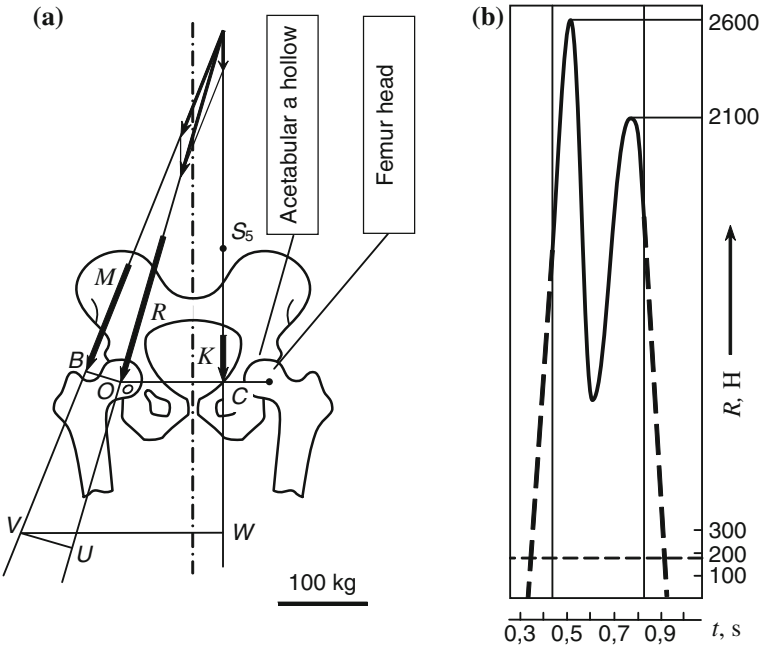


Fig. 1.2 Main elements of hip joint pivot unit **a** and diagram changes of compressive forces in hipbone head within step cycle **b** [8]. S_5 body center of gravity; *K* forces of action of body weight part; *M* forces of action of lateral abductor muscles; *R* resulting compressive forces

Rotation of the shinbone is limited by collateral joints actively and by a specific configuration of the articular surfaces passively [1, 13]. The sections show that axial compression is capable independently to stabilize the unbent knee after transversing all the ligaments, including the cruciform ones. The cruciform joints limit actively the displacement of mated articular surfaces in motion in the ventrodorsal direction. The posterior cruciform joint is the main knee stabilizer. Relaxation of all the ligaments during rotation in the terminal unbending stage is a prerequisite of the normal functioning of the knee joint because it reduces the contact pressure on the medial portions of the shinbone. It is noteworthy than unbending of the knee increases the curvature of the hip condyles, normally more of the internal always than the external one.

Moreover, the essential role in the knee stability belongs to the capsular-articular elements and meniscuses, especially in the retromedial portion of the joint [14].

As the knee joint motion is exposed to five forces: axial, ventrodorsal in the sagittal plane, medial lateral in the horizontal plane and two moments of rotation relatively to the shinbone bone axis in the ventrodorsal axis, the ventrodorsal and medial lateral forces play just a secondary role as compensating forces to balance movements in these directions. At the same time, the forces of rotation in relation to the shinbone axis bone and ventrodorsal axis can redistribute the forces of flow

from the lateral portion of the joint towards the medial portion. It is believed that the exterior hip condyle in the physiological conditions is loaded three or four times heavier than the interior condyle [14]. The knee joint may experience considerable forces that depend on the dynamic loading conditions. For example, it is noted in [15] that its elements when ascending a stairway may be affected by forces of the order of magnitude 150 Nm; when standing up after low crouching this joint can exposed to the loading five times exceeding the body weight.

Nevertheless, the time dependence of the resulting forces in the knee joint can have a similar pattern to that in the hip joint: it is determined by a significant straining of muscles and ligaments in addition to the load bearing force.

The latter play quite an important role in the knee joint biomechanics, controlling the physiologically comfortable positions, redistributing and reducing the compressive forces affecting the articular contact [8]. The paper [16] that changes of the length of ligaments redistribute contact forces, while shifts in the points of attachment of muscles do not practically affect them.

The effects in motion of the knee joint due to the spatial position of the centers of its rotation contribute sizably to the biomechanical process. It is known that the knee due to the ellipsoid form of the hip condyles does not have any permanent center of rotation typical for common pivots. Therefore, when it unbends and the radii of curvature of the condyles grow the center of rotation move along an arcuate line. However, the latter deviates from the line of arrangement of momentary centers of rotation of the joint. It can be traced vividly if one imagines the joint as a closed crossing four-link kinematic chain with two parts having the same degree of relative motion freedom, i.e. ignoring conventionally rotation of the shin bone in the terminal unbending stage and assuming that the cruciform joints are constantly stretched (Fig. 1.3). Modeling of rotation of links *AD*, *BC* (the cruciform joints) and *AB* (the tibial bone) around the fixed hip link *DC* enables to obtain a dorsal movement of momentary centers of rotation best reproducing the functional features of the natural knee joint. The diagram makes it ultimately clear how to interpret many features of functioning of joints in physiological norm and in pathology.

The analysis in [17] proves that application of a longitudinal compressive force via a joint to a joint boosts its stability, on the one hand, but limits its degrees of freedom and that can change the joint mechanical properties due to movement of the momentary centers of rotation in space. A position of the knee joint center of rotation depends on the extent of initial compression and can displace by over 0.02 m. An essential fact too that the momentary centers of rotation in the hip intercondyle pit (the fixed element) of the lie mutually closer than in the unloaded joint. As a result, the sliding speeds under these conditions differ little and their directions coincide with the articulation slot line. A significant scatter of the momentary centers of rotation, e.g., in case of injury of the meniscus, ligaments, or changes in the configuration of mating surfaces, induce extra forces with the speed vector different from the articulation slot line and directs it a certain angle to it. Friction in the joint intensifies joint, the cartilage is continuously microtraumatized, degeneration centers appear, and mechanodestruction follows.

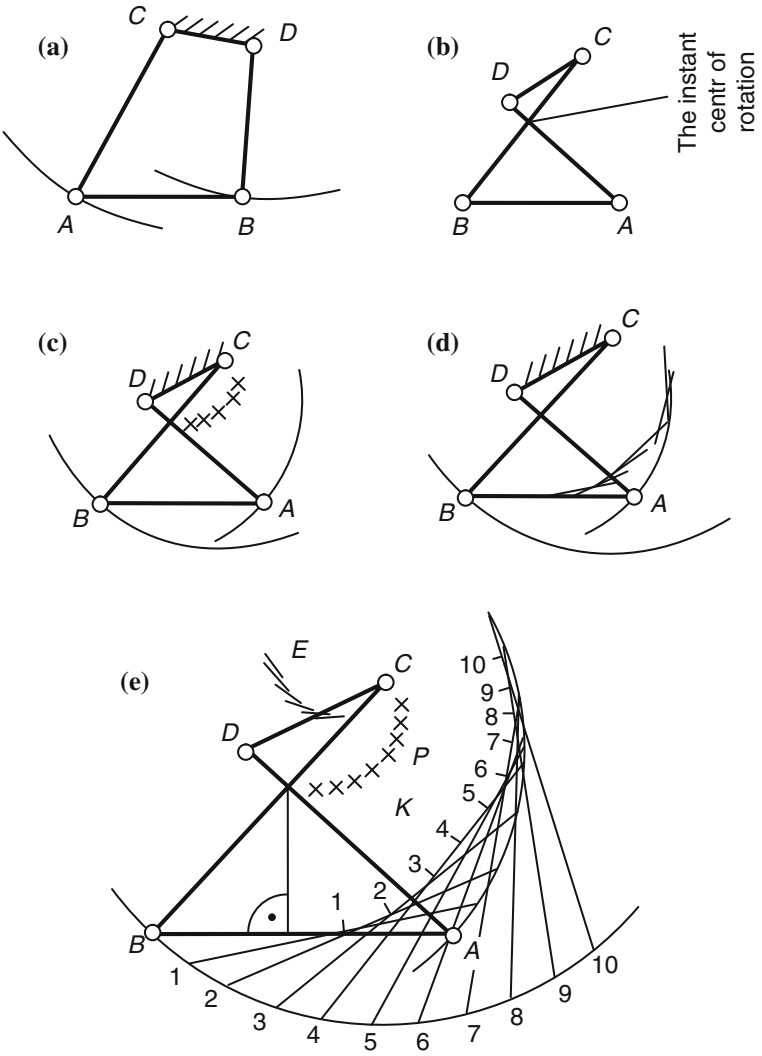


Fig. 1.3 Four-link knee joint kinematic chain: **a** closed Four-link kinematic chain; **b** crossed kinematic chain; **c** crossed kinematic chain showing passage of axis of motion of knee through momentary center of rotation in point of intersection of cruciform ligaments when hip component is fixed; **d** crossed kinematic chain showing curvature of contours of hip when shinbone moves; **e** crossed kinematic chain showing geometrical arrangement of tibiofemoral contact points during knee bending [13]. *AB* intercondyle shinbone hill; *BC* crossed kinematic chain of anterior cruciform joint; *AD* posterior cruciform joint

Therefore, operative interventions are biomechanically justified when it is necessary to correct the centers of rotation of pathologically altered joints and can restore their normal functioning in a number of cases.

From the viewpoint of transmission of the supporting loading the geometrical characteristics of joints are essential, such as forms of dynamically contacting surfaces and their mutual compliance (congruence) that determine the extent of mobility of articular ends, the amplitude of their motion and the number of axis around which they move. It is believed that working contact joints are fully loaded in only one position corresponding to transmission of the maximum force via the joint [18]. The articular contacts in other positions are incompatible, i.e. incongruent. As a result, loads of various magnitudes and directions in the hip joint can produce four contact types: a usual contact that occurs under inconsiderable loads; a contact in response to a position of joints; a contact in response to loading and occurring only under considerable loads; no contact between surfaces at all at any joint position and load.

The data about the actual contact area in different joints are controversial. Paper assumes that the actual contact area in each knee joint condyle under load is 1–2 cm^2. However the results in [12] show that the mean supporting area of the interior condyle of the unbent knee is 2.5–6.7 cm^2, that of the exterior condyle is 1.7–5.1 cm^2, i.e. 1.6 times less than that of the interior one. It is believed that the surface contact reaches 4 cm^2 in the hip joint under full loading. Thus, it is apparent that the articular surfaces have sufficiently high contact pressures of 60–70 kg/cm^2 under maximum loads exceeding the body weight three to nine times [19].

Mechanical interactions between articular surfaces are known to occur always through the layer of a compliant composite material—the hyaline cartilage of irregular thickness. The cartilage layer gradually thins away in contact zones in the direction towards the edges of articular surfaces. It is probable that the geometry of the articular cartilage in the contact zone that affect considerably the mechanism of transmission of mechanical forces in the joint. In fact, analysis of the contact interaction of one solid body with another through the elastic interlayer from a homogenous material shows that the geometric characteristics of the interlayer strongly affect both the specific pressure and the pattern of its distribution (Fig. 1.4).

Constant thickness of the elastic interlayer changes the specific pressure in response to the contact angle according to the sinusoidal law, meanwhile the thickness of the elastic layer is variable according to the law of mass and the specific pressure distribution in the contact region is regular. The fact that the cartilage layer in the contact zone between articular surfaces narrows towards the edges permits to assume a similar pattern of distribution of contact forces in joints. Hence, the thickness distribution of the cartilage in the joint is such that during ambulation the pressure on the articular surfaces distributes regularly in the contact region. Whence an essential conclusion is that the cartilaginous layer is a peculiar shock absorber of dynamic loads favoring regular distribution of pressure in the joint. Table 1.1 shows examples of the moduli of elasticity of some materials most frequently used for prosthetic arthroplasty.

Table 1.1 shows clearly that just few possess the mechanical properties comparable with the properties of the joint cartilage as a natural composite material.

Some human and animal joints have fibrous cartilaginous structures or the interarticular cartilage known as the meniscus in addition to the hyaline cartilage

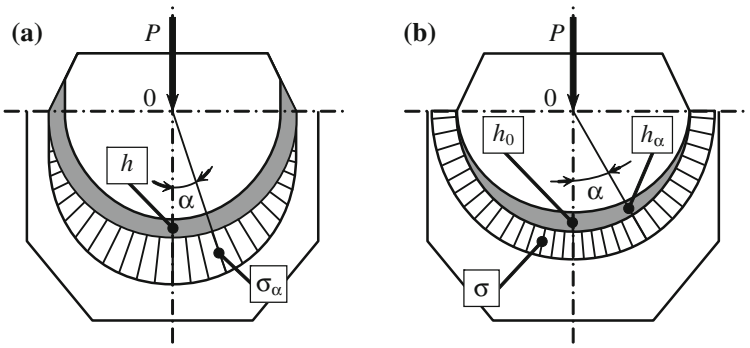


Fig. 1.4 Diagram of contact interaction between two solid bodies separated by an elastic interlayer of inhomogeneous material in response to angle of contact α: **a** specific pressure sinusoidal distribution; **b** regular specific pressure distribution

Table 1.1 Moduli of elasticity of composite materials [19]

Material	Modulus of elasticity × 10^3, MPa
Joint cartilage	0.001–0.17
Natural rubber	0.0025–0.1
Silicon rubber	0.01
Super high-molecular polyethylene	0.5
Bone cement	3.0
Bone	10.0–30.0
Alloy Ti–Al–Va	106.0
Stainless steel	205.0
Alloy Co–Cr–Mo	230.0
Aluminium	350.0

coating the bone [1, 2]. For example, there is a medial meniscus and a lateral meniscus in the human knee joint. In the joint they balance the mismatch between the hip condyle and the shinbone, expand the actual contact area. Their configuration demonstrates it (Fig. 1.5).

The peripheral part of each meniscus is thicker and convex; the central part has a fine free edge. The sagittal section shows an edge-like shape. The medial meniscus resembles a hemisphere that arranges laterally over the entire circumference of the respective shinbone condyle. As a result, the external edges of the meniscus repeat the configuration of the external edges of the tibial plateaus. It is believed that a high congruence of articular surfaces provided by meniscuses strongly contributes to the joint stability, even if anterior cruciform joint is injured [20, 21].

The unique dumping properties of the cartilaginous tissue in the meniscus, their good adaptability to articular surfaces and extensive contact area with hip condyles play an essential role in undertaking and redistributing continuously variable

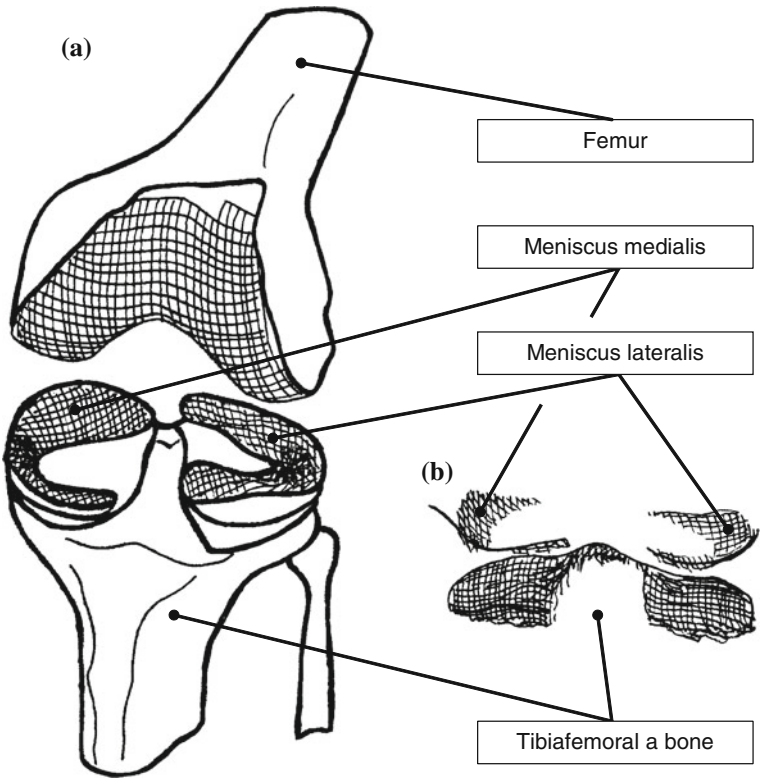


Fig. 1.5 Diagram illustrating contact interaction in knee joint, (**a**) and stereogram representing contact zones between hip condyles, tibial plateaus and meniscus, (**b**)

specific pressure in the joint [1]. According to [20], the meniscus bears up to 50 % of the compressive loading when the knee is unbent and up to 85 % when the knee is 90° bent. It is significantly because that the meniscus is capable to change its configuration in response to the angle of bending of the knee joint assisted by the joint capsule along the edges of the tibial plateaus that arrest the meniscus peripherally resisting the effect of the forcing out forces appearing under rather heavy axial loads [1].

A total or subtotal removal of the internal meniscus shrinks the contact in the medial portion of the joint by 50–70 %, hence, the specific pressure on jumps sharply [22]. It may lead to gradual impairment of the dumping properties, degeneration, and mechanical destruction of the cartilage. Numerous clinical observations confirm this fact [23]. A number of experimental models of the early osteoarthrosis are based on the meniscus resection [21, 24]. It is noted that degenerative changes in the joint are pronounced and proportional to the area of the removed meniscus segment [20, 22]. That is why in recent years the concept of partial or so-called intermeniscus resection is gaining recognition in case meniscus

damage; the progress of the concept is due to adoption of modern arthroscopic medical surgical equipment [25].

The experiments have shown that partial meniscus ectomy leads in the majority of cases to a significant dysfunction of joints [25]. A probable explanation of this behavior of the locomotorium is that, if the cartilage is normal, and just a slight of the affected meniscus is removed, distribution of stresses in the joint remains practically unchanged and regular [1]. However, it is possible in such cases that arthrosis may reappear [26]. One of the causes of its development may be a disorder of the congruence of the articular surfaces after an operative intervention. It can produce zones the contact of cartilages is fully dependent of the loading magnitude and occurs only in case the forces are large and the zone also, where there is no contact at all under any load at any position of the joints. This boundary region between the zone of permanent contact and its absence is the most sensitive and vulnerable to appearance and development of really centers of degenerative changes in the cartilage. Other more detailed studies confirm it too [1, 22, 23].

Thus, the accomplished analysis of joints as versatile kinematic pairs capable to withstand significant variable loads shows that, alongside with their features, structure, functioning, and biomechanics, the articular cartilage plays an essential role in transmission of mechanical interactions between mated bone surfaces. To perform this function, the cartilaginous tissue should possess specific biomechanical properties that differ from those of other tissues [27]. Their unique structural arrangement in the cartilage can explain it.

1.2 Structure and Functions of Joint Cartilage

The first works in the sphere of chemical composition and structural organization of the cartilage matrix relate to the mid of the last century when J. Hunter studied the morphology of a pathologically altered cartilage in vivo. In 1837, E. Müller exposed the cartilage to heavy pressure and obtained a solution of a substance that he called chondrine [28]. Later K. Krukenberg separated the main chondrine component and identified it as chondroitin sulfate. In 1953 E. Davidson, K. Meyer, A. Linker and B. Weissmann discovered one more substance that they believed initially a constituent cartilage component and called keratan sulfate. In these years K. Meyer with his disciples showed that chondroitin sulfates were a variety of polysaccharides; their molecules were chains of disaccharide links alternating with the glucuronic acid and *N*–acetylgalactosamine—a sulfate derivative (Fig. 1.6a, b). Later it was established that keratan sulfate is the same polysaccharide, but it consisted of alternating molecules of galactose and the sulfate derivative *N*–acetylglucosamine (Fig. 1.6c).

Though certain results were achieved by the studies of the chemical composition of the constituent components of the cartilaginous tissue, the first full idea about its structure appeared only in 1969 when the methods used to study of nucleic acids were successfully applied to extraction of undamaged giant aggregates of

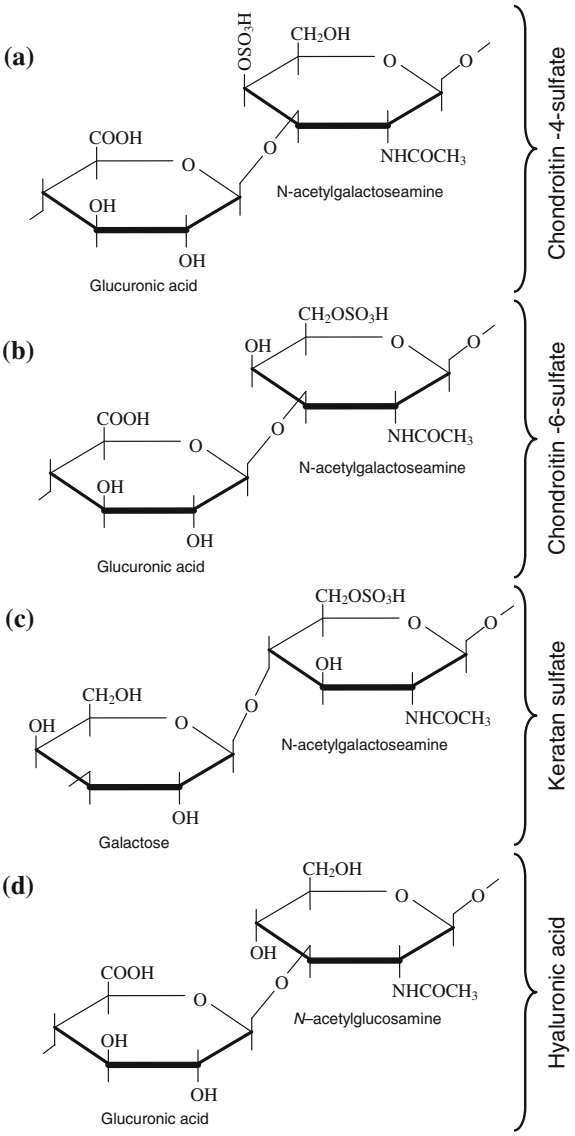


Fig. 1.6 Chemical structure of glucose aminoglycans in the joint cartilage

macromolecules of proteoglycans from the cartilage [27, 28]. In the same year, an assumption was made that the structure protein polysaccharide complexes in the cartilage matrix was not limited to appearance of the aggregates of proteoglycans, that there existed larger complexes contributing to stabilization of the structural organization of the joint cartilage and maintenance of its biomechanical properties. In 1983 this assumption was confirmed in [29] showing that the aggregates of

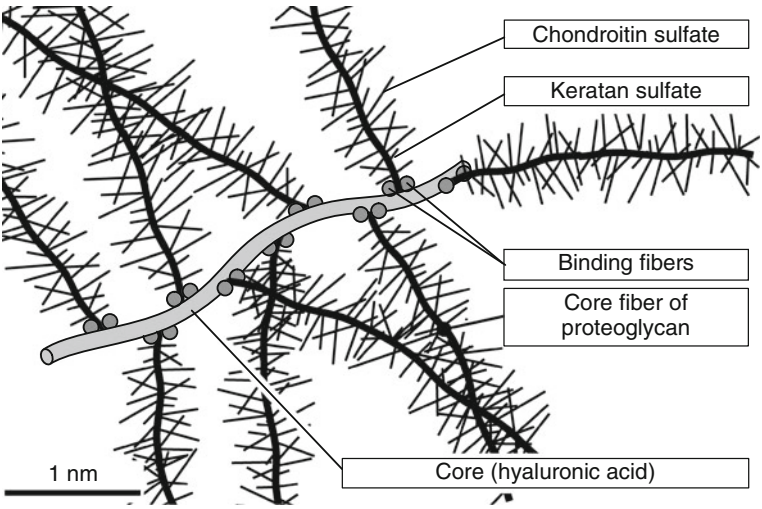


Fig. 1.7 Fragment of aggregate of proteoglycans. Proteoglycans (subunits) are attached in a non-covalent manner with two special binding proteins to hyaluronic acid molecule [30]

proteoglycans were combined by two special glycoproteins from the cartilage matrix into still larger structures or superaggregates.

Thus, so numerous biochemical experiments, application of electron microscopy enabled to produce a structural characteristic and to visualize the protein polysaccharide complexes and to get general ideas about the fine structure of the proteoglycans and their aggregates (Fig. 1.7) [27, 30].

The latter account for 50 % of the dry weight of the joint cartilage and represent giant macromolecules that include the hyaluronic acid (HUA), Glycoproteins and stem or medullar proteins with numerous lateral chains of polysaccharides (glucose aminoglycans) play an essential role in their composition. At present these ideas are generally accepted because they provide fuller explanation of the functions of proteoglycans in creating the physico-chemical and biomechanical properties of cartilaginous tissue, though some details are continuously verified and added [28].

According to the generally accepted ideas, the main proteoglycan aggregate in the e cartilage matrix is the HUA molecule (Fig. 1.7). It belong to unsulfated glucose aminoglycans and consists of alternating remnants of the *D*-glucuronic acid and *N*-acetylglucosamine (Fig. 1.6d). Compared with other glucose aminoglycans and the HUA it has a larger molecular weight and can combine up to 200 proteoglycan subunits with their points of joining being spaced by at least 50–60 disaccharide links [27]. The averaged composition of a proteoglycan subunit (a macromolecule) is shown in Table 1.2. Since practically the entire HUA molecule in the cartilage matrix is involved in production of proteoglycan aggregates, it determines their sizes.

Polypeptide chains of medullar proteins branch away from the HUA like from a stem and form the base of the proteoglycan subunits (Fig. 1.7). The proteoglycan

Table 1.2 Proteoglycan macromolecule (a subunit) composition [27]

Components	Number of chains	Molecular mass	Part of total macromolecular mass, %
Protein	1	200000–300000	7–12
Chondroitin sulfates	100	20000	80–85
Keratan sulfate	50	5500	7
Oligosaccharides	50	1200–2000	1–3
Charged groups			
Sulfate	4500		
Carboxylate	4200		

subunits attach to the HUA in a non-covalent manner with a specific terminal region having a globular conformation [31]. Moreover, the so-called combining proteins typical only for the joint cartilage matrix [32, 33] additionally stabilize these portions of the proteoglycan aggregates. A multitude of polysaccharide chains attach, in their turn, to each long medullar protein molecule [28]. They are chiefly sulfated varieties of glucose aminoglycans: chondroitine-4-sulfate, chondroitine-6-sulfate, and keratan sulfate (Fig. 1.6).

Both chondroitin sulfates consist of alternating remnants of the glucuronic acid and galactose amine and are alternating copolymers of the *D*-glucuronic acid and *N*-acetylgalactosamine. The *O*-sulfate group attaches to them and localizes in the molecule determining the difference between these isomers having a practically identical structure. The length of the chains of the chondroitin sulfates in the cartilaginous tissue is made up by 30–60 disaccharides on the average [1]. Quite often, the molecules have a hybrid structure when the same chain has disaccharide units, such as both condroitine-4- and condroitine-6-sulfate [27]. However, condroitine-6-sulfate dominates in the cartilage matrix.

Keratan sulfate is the only glucose aminoglycans free of hexauronic acid. Its molecule is a polymer composed of a repeating disaccharide unit of *N*-acetylglucosamine containing a sulfate group and *D*-galactose near the sixth carbon atom [27]. The keratan sulfate chain is somewhat shorter than the chondroitin sulfate chain and usually it is about 15–30 disaccharide units.

All glicoaminoglycans in the cartilage matrix have a significant number of anion groups. Each disaccharide link, e.g., in the chondroitin sulfates one the saccharides, vis. the glucuronic acid, contains a carboxyl group ($COOH^-$), while another is an *N*-acetylgalactoamine-sulfatene group (SO_4^{2-}). Therefore, the polyanionic pattern of glucose aminoglycans is the main factor governing their physico-chemical and biomechanical properties. Due to this polyanionic nature and saturation with free acid (carboxyl and sulfatene) groups with similarly (negatively) charged glucose aminoglycans, they are straightened in the proteoglycan subunit and mutually repel. Hence, they are oriented perpendicularly to the medullar proteoglycan subunit in the shape of a peculiar wire brush (Fig. 1.7). Due to the same electrostatic

interactions still, the proteoglycan monomers, in their turn, form an analogous structure of proteoglycan macromolecules arranging perpendicularly to the HUA chain.

Because of such arrangement features, the fine structure of proteoglycans and their aggregates acquire the properties of the so-called diffuse macromolecules because voids appear in the intervals between their glucose aminoglycans units that act like a sponge and can selectively accumulate low- molecular components, such as water molecules and other comparatively fine ingredients of the proteinopolysaccharide complex. Therefore, the proteoglycans in the joint cartilage matrix possess a huge structuring capability concentrating a significant quantity of water whose mass exceed much it own mass. Hence, the articular cartilage is a highly hydrated with water being one of its main components.

The ratio between proteoglycans and water determines the degree of its possible loss by the cartilage, i.e. the higher the concentration of proteoglycans the less water can leave the cartilage under the effect of contact pressure in locomotion, and vice versa, lack of proteoglycans due to reduction of the concentration of glucose aminoglycans, e.g., in osteoarthrosis, leads inevitably to the softening of the cartilage matrix and its stronger permeability followed by development of pathological changes [34, 35].

The osmotic pressure due to the protein polysaccharide complexes is known to be responsible for the ability of the cartilage to withstand high mechanical pressure without losing water and for the fluid balance. Until quite recently, it was believed that the concentration of free water in the cartilage matrix was about 94 %. However, more recent studies give different values of interstitial water concentration in cartilages—70–100 %, it is apparently due to concentrations of proteoglycans in them [27].

This free behavior of the interstitial water and its compressibility add essential biomechanical properties to the articular cartilage; their sense is the following. When the cartilage experiences pressure, water is forced out from the regions of sulfate and carboxyl groups where the charge arrested them. At the same time these groups converge and the forces of repulsion between their negative charges inhibits further compression of the tissue. When the compression disappears, water returns and makes deformation under load reversible [36]. Hence, the interstitial water together with the proteinopolysaccharide complex with its inherently high osmotic pressureм and huge structuring ability due to the high density of the negatively charged group of sulfonated glucose aminoglycans maintain the elasticity and stiffness of the joint cartilage, i.e. it is responsible for the shock-absorbing properties in joints.

Apparently, the high strength properties of the cartilaginous tissue would be impossible without its highly ordered supermolecular organization. The intercellular matrix or collagen as the third component plays an essential role. It represents specific proteinic macromolecules or their aggregates forming the connective tissue in the fibrous structure.

At present, it is believed that there are up to 25 types of collagens [1, 35, 37]. The structural or fibril forming collagens (types *I–III*) are the main in the connective tissue. They typically perform mechanical functions. It is known that the collagen

Table 1.3 Mechanical properties of biological and non-biological materials [36]

Material	Young modulus, MPa	Breaking strength, MPa
Collagen	10^3	50–100
Elastin	0.6	–
Bone	10^4	10^2
Cured rubber	1.4	–
Oak	10^4	10^2
Mild steel	$2 \cdot 10^5$	$5 \cdot 10^2$

type *I* dominates in bones, skin, tendons, joints, and meniscus. The collagen type *III* dominates in muscles, blood vessels, visceral organs [1, 36–38]. The collagen type *II* dominates in the articular cartilage. Its concentration in the intercellular matrix is 20–25 % exceeding the concentration of proteoglycans (5–10 %) and ranks second after water (65–80 %).

All structural collagens possess high mechanical properties (Table 1.3) [39].

Their fivers are capable to stretch 10–20 % without rupturing. Transverse glycoprotein interfibrillar links have an essential significance in the mechanical properties of the structural collagens. They prevent mutual sliding of collagenous fibers, water presence is also essential [1]. It typical for water to change the protein density without affects its shear modulus.

The methods of X-ray structural analysis and electron microscopy revealed that portion of regular transverse striation repeat periodically after every 60–70 nm of the longitudinal fibrous collagenous structure. There are grounds to assume that it is due alternation of highly ordered crystalline portions of fibrils with zones having aminoacidic composition and less regular arrangement of molecules. The collagen in fibrils probably stretches predominantly [39].

To understand the role of the collagen individually in maintaining the main biomechanical functions of the cartilage integrally the data known at present about its molecular and supermolecular structural organization are essential. Their essence is the following (Fig. 1.8).

The mature collagen in the connective tissue is preceded by procollagen macromolecules that are its intracellular form consisting of left-handed α-chains. Both covalent hydrogen bonds combine the latter and into an extended right-handed spiral. Terminal extensions split from the procollagen macromolecules in the extracellular space, this results in tropocollagen macromolecules 300 nm long and 1.4 nm in diameter. They combine by cross-intermolecular covalent links and form microfibrils 3–5 nm in diameter. The following stage is that several microfibrils form collagenous fibrils 20–100 nm in diameter with the participation of polysaccharide components—proteoglycans and glycoproteins. There are intermolecular transverse (cross) links in the portions of adjacent macromolecules of the fibrils. The integrity of these intra- and intermolecular cross-links together wit hydrogen bonds stabilizing the spiral configuration of α-chains maintains the high structural and mechanical strength of the collagenous fibers.

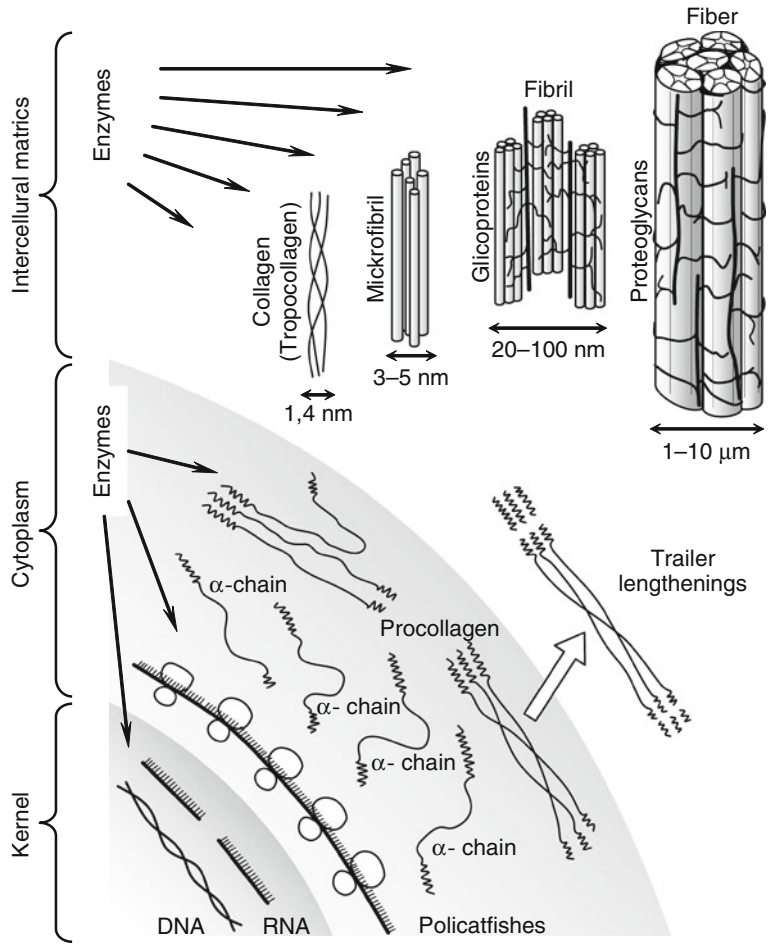


Fig. 1.8 Diagram био synthesis collagen and forms of collagenous structures

The collagens of type *II* in the cartilage matrix have different diameters (1–10 μm) and, what is most important; they do not produce any dense bundles that the collagens of other types usually do. This supramolecular organization favors, first, the bonding of water and, second, boosts interaction of the fibers with the aggregates of proteoglycans (Fig. 1.9).

The latter intercombine and join the finest collagenous fibrils by glucose aminoglycans freely protruding in all directions from the medullar proteinic chains arresting the interstitial fluid in their cells. Sometimes they imply a mechanic arrest of aggregates of proteoglycans by a collagenic net since their links are predominantly mechanical [4, 27]. The interaction between the proteoglycans and collagen, in its turn, reinforces the highly ordered structural organization of the cartilage matrix both in the ordinary conditions and during its resistance to shear forces [40].

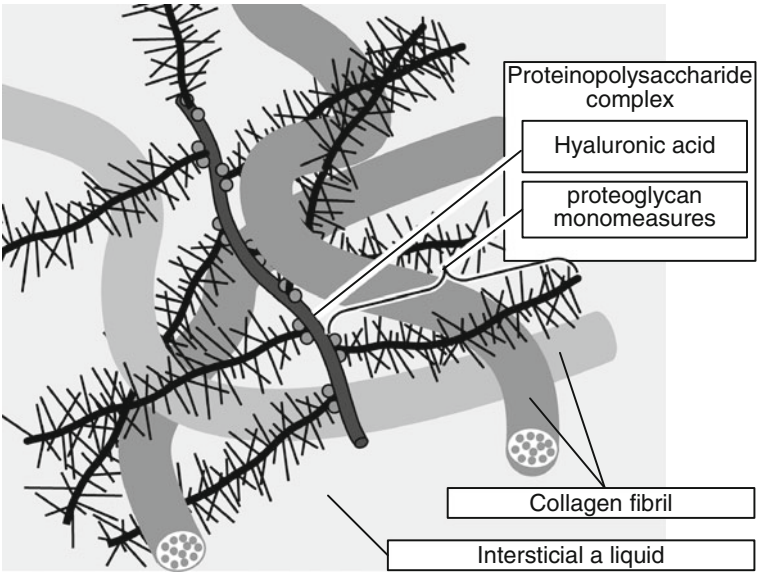


Fig. 1.9 Diagram of structural cartilage organization [27]

In fact, the tensioning of the collagenous net in the unloaded articular cartilage is balanced by osmotic pressureм in proteoglycans, or, in other words, by the collagenous skeleton counteracting the expending pressure generated by proteoglycans. Thus, the net is in a stressed inflated condition [41]. Hence, the collagenous skeleton, while performing a structural, stabilizing function in respect to proteoglycans and water and controlling the extent of hydration of the cartilaginous tissue, posses much elasticity in respect to the tensioning forces. It is evidenced by the fact too that the strength of the cartilaginous tissue in tension depends significantly on the concentration of the collagen; if it is affected the defibering of fibrils occurs. It leads inevitably to the swelling of the cartilaginous tissue. At the same time, while the collagenous skeleton is very strong in tension, it resists compression poorly. As a result, compression deforms the cartilaginous tissue immediately.

1.3 Biomechanical Properties of Joint Cartilage Matrix

The resulting tension of the fibrous structure and incompressibility of the intercellular substance are the main cause of the spring-like elasticity of the cartilage. The matrix remains as it is at the initial hydration stage because the osmotic pressure keeps the water inside, specifically the intrafibrilouls water [27, 28]. However, when the load applied to the cartilage exceeds the osmotic pressure, the interstitial fluid can exude from the cartilage matrix, i.e. the creep phase is observed.

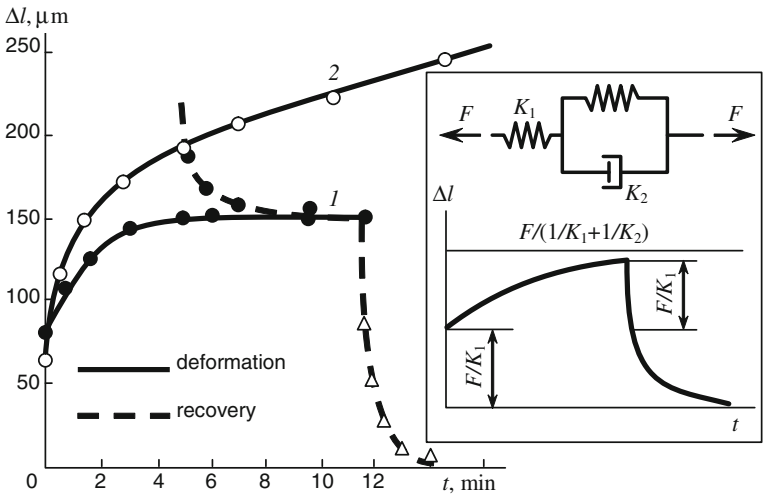


Fig. 1.10 Deformation and restoration of cartilage immersed into 0.9 % NaCl solution: oscillating (*1*) and non-oscillating (*2*) shinbone joints and their simple elastoviscous analogy (element of Kelvin-Foigt)

In particular, a constant load applied to the shinbone causes initial elastic deformation that later develops into the creep deformation with a receding rate during 24 h (Fig. 1.10) [42, 43].

However, when the joint oscillates, the deformation becomes approximately constant reaching the value typical for the applied loading during 5–10 min. If oscillations begin several minutes after the loading is applied to the immobile joint, the deformation quickly diminishes to the value typical for the applied loading in oscillatory conditions, i.e. it becomes much less than in the conditions of static loading. When the loading is relieved, an opposite phenomenon occurs when the cartilage absorbs fluid and restores its transverse dimensions. However, the cartilage is unable to restore fully unless there is fluid in the contact. It is shown in [44] that the cartilage immersed into the saline solution is elastic, otherwise it is not. Introduction of some fluid onto the deformed surface of the cartilage restores its original condition immediately. It indicates convincingly that the cartilage elasticity depends on the fluid availability. This elastoviscous behavior of the cartilage immersed into the fluid under normal effective loads is approximated by the viscoelastic models representing the elements of Foigt connected in series and by the linearly elastic spring.

When the loading is removed, the oscillatory motion favors also more effective restoration of the initial thickness of the cartilage [42]. One of the probable causes of this fast restoration of the cartilage under dynamic loading is shown in [42] as fluid filtration through its matrix due to the difference between the osmotic and hydrostatic pressure because of subsequent immersion of different portions of the

cartilaginous surfaces into the fluid followed by their leaving. This pattern of the fluid balance most apparently produces relatively constant deformation of the cartilage in the creep phase and its faster restoration once the loading is removed.

Other experimental data [42] evidence in favor of this assumption too. It is shown that the thickness of the deformed cartilage restores as oscillations of the joint under the applied load become more frequent, i.e. its average thickness grows as the frequency of oscillations increases up to 40 cycles per minute and remains comparatively constant. It is noteworthy that the frequency of oscillations above which the dimensions of the deformed cartilage under the applied load cease to restore, i.e. 40 cycles per min, corresponds to normal walking [45]. This fact fully apparently confirms the idea that the cartilage should be considered a labile system capable to adapt to definite biochemical conditions.

Thus, deformation and consecutive restoration of the cartilage upon unloading result, primarily, from mechanical forcing out and absorption of the fluid, respectively. Note that deformation of the cartilage when compressed in the fluid is weaker than in the air [43].

There are different points of view regarding the effect of the fluid and the cartilage matrix on its susceptibility to deformation. According to [42], the ionic environment can produce a strong effect on the cartilage deformability. Apparently, the reaction of the cartilaginous tissue to the ionic medium relating to the washing away of cations so that the negative charges of the fixed sulfate groups of protein polysaccharides become exposed and the colloid chains stiffen because of their mutual repulsion. Addition of cations produces an opposite effect. It proves that the density of the main substance is osmotically active when the matrix retains. The results in confirm it showing that the modulus of elasticity in compression of the joint cartilage is proportional to the concentration of proteoglycans, while is primarily determined by the stiffness due to the sulfonated glucose aminoglycans, i.e. chondroitin sulfates and keratan sulfate. Since the electrostatic and osmotic properties of protein polysaccharide complexes determine the elastic characteristics of the cartilage, it is apparent that the compliance of the cartilaginous tissue substantially depends on both the properties of proteoglycans and the factors capable to change these properties. The considered cases are exactly vivid examples of this active effect of the fluid on the biomechanical properties of the cartilaginous tissue matrix.

This behavior is not typical for a majority of the liquid media. Insignificant difference between their effects on the elastic properties of cartilages proves it. Paper [46] present a study of the viscoelastic properties of the cartilaginous tissue under the effect of low-voltage sinusoidally variable compressive loading. Under these conditions the synovial fluid and the Ringer solution influence practically similarly the modulus of elasticity of the cartilage. The authors of the present work obtained similar results when they studied the compressive deformation of the cartilage in the silicon organic, synovial, and pseudosynovial fluid with addition of liquid-crystalline compounds under dynamic loading conditions, vis. during friction of the cartilage on glass [47, 48]. However, notwithstanding the seemingly identical effect of these media during compression of the cartilage, their effect on its restoration is different once the

loading is removed. It was discovered particularly that the effect on the cartilage after unloading is much stronger in the synovial and pseudosynovial fluids with liquid-crystalline compounds than in the silicon organic medium.

In our view, one of the probable causes of such unambiguous behavior of the media is that the elastic post-effect in the cartilaginous tissue upon unloading is due to the presence of liquid-crystalline compounds in the synovial and pseudosynovial fluids, i.e. the components with low shear resistance that is inherent to liquid-crystalline states [49, 50]. It has significant importance under dynamic loading conditions approaching to the real conditions in joint, i.e. when tangential forces in addition to and concurrently with the normal compressive loading affect the cartilaginous tissue. These components in the liquid medium in motion favor shear localization in the fluid layer rather than in the cartilaginous tissue significantly reducing deformation of the cartilage matrix. Thus, its damage is prevented and the deformation appearing under loading remains reversible [35, 51].

Another cause of such behavior of the cartilage matrix upon unloading in various liquid media in dynamic conditions can be due to poor lubricating properties of the fluid surrounding the cartilaginous tissue; it may be due to possible evolution of rivaling processes between its molecules and the molecules of the interstitial fluid in question. The cartilage tissue is known as a highly porous material (with the mean sizes of pores being 6–10 nm). The pores actively participate in the mechanism of oozing of the interstitial fluid on the cartilage surface during compression and vice versa during cartilage restoration in the process of removal of the compressive loading [52].

As a result, quite strong shear deformation shear appearing in these media in при motion localizes in the cartilaginous tissue rather than in the fluid leading to the closing of pores in the cartilage matrix, on the one hand. On the other hand, the molecules of the surrounding fluid have the inherent physico-chemical properties (e.g., higher surface activity in respect to the cartilage, their dimensions exceeding the sizes of pores in the matrix, etc.) that enable them to prevent contact with the interstitial fluid exuding in compression. All these factors prevent the oozing of the fluid into the matrix and reduce considerably its swelling or restoring upon unloading. The experiments, described in [53] confirm the mechanism of evolution of these processes: no restoration of the cartilaginous tissue is observed during friction of the cartilage on glass if there is no surrounding fluid.

The process of wear in friction can irreversibly change the viscoelastic properties of the cartilage eliminating it from the surface layer. The latter is known [54] to be one of the main factors that control fluid transfer, hence, the deformation of the cartilage matrix as a whole. For example, it is established in [55] that, after elimination of the surface layer 100 μm thick from the cartilage the fluid exchange during loading grows two-three times and the creep deformation reaches an equilibrium condition two times quicker compared with the normal cartilage.

It is believed that there are two biologically inactive forms of mechanical fluid transfer through the cartilaginous matrix [36, 56].

In the first case, the interstitial fluid is transferred through the porous cartilage matrix under the effect of pressure gradient in the tissue itself and the law of Darcy governs fluid consumption:

$$Q = k \cdot \Delta p / \Delta h, \tag{1.1}$$

where Q—volume consumption per unit of cross-sectional area of the porous substance; Δp—a pressure drop across the thickness Δh of the specimen; k—the coefficient of permeability.

The first experiments of determination of the coefficient of permeability of the cartilage showed already that its mean value in the direction normal to the contact amounts to $5.8 \cdot 10^{-16}$ m^4/N s. it proves that the cartilaginous tissue strongly resists to the fluid's movement. It means that it is a porous material with an extremely poor permeability. It was observed that the permeability varied across the specimen and impairs with deepness. The coefficient of permeability on the contact of the cartilage is $7.65 \cdot 10^{-17}$ m^4/N s versus $4.3 \cdot 10^{-16}$ m^4/N s in its body.

Further studies yielded a more detailed picture of the changes in the permeability of the cartilaginous tissue. Tests of its Sects. 200–400 μm thick revealed that the permeability coefficient through the body of the cartilage specimen would initially grow and then decline. Its value would range from $(2.3–2.8) \cdot 10^{-16}$ $m^4/N \cdot s$ near the contact to $(3.3–3.8) \cdot 10^{-16}$ m^4/N s in the middle zone and $(1.5–1.8) \cdot 10^{-16}$ m^4/N s near the specimen base.

This irregular permeability of the cartilaginous tissue was attributed to a higher density of the fixed middle portion, orientation, and denser arrangement of collagenous fibers near the contact determined by both the morphological features and the structure of the cartilage matrix.

The *surface* (200–600 μm) zone among five now functionally identifiable morphological zones in the cartilage presents most interest for formulating ideas about the specifics of fluid transfer and process of friction on cartilaginous surfaces (Fig. 1.11) [1, 35, 57, 58]. Domination of collagen, insignificant water quantity and glucose aminoglycans characterize it [27, 35, 51, 59]. The bundles of collagenous fibers in it 30–32 nm in diameter with a periodicity 64 nm orient tangentially, parallel to the contact. The outer layer in the tangential zone is a cell-free membrane (3–4 μm) that is rather strongly selectively permeable for low-molecular substances [27, 35, 58]. From the viewpoint of functional specifics, the surface tangential zone performs during loading of the cartilage the function of a protective cushion resting on the middle and deeper zones.

Nevertheless, numerous studies of the structural organization and distribution of main components over the zones of the cartilage matrix have led to two opposite points of view of the permeability in different directions. Some researchers believe that the cartilage behaves in respect to the fluid flow as an isotropic porous material [27], while others believe that its behavior is anisotropic [56].

Alongside with the internal properties inherent to the constituting components of the cartilage matrix, external factors too affect considerably its permeability, e.g., the fluid around them, with the effect of the nature of the latter being quite sizeable.

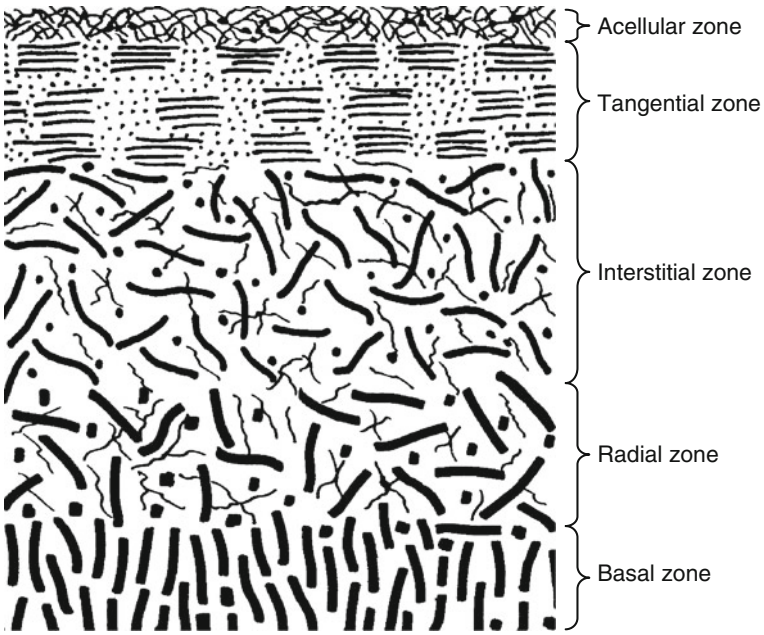


Fig. 1.11 Structure of joint cartilage [1]

According to [36], while the coefficient of permeability of the cartilage in the solution of Ringer is $33 \cdot 10^{-16}$ m^4/N s, it is $1.1 \cdot 10^{-16}$ m^4/N s in the normal saline solution or three times less.

The second form of fluid transfer is attributed to compression of the cartilage matrix [60]. Its changes in volume produce a pressure gradient in the interstitial fluid. This mechanism is known as flow due to consolidation [56]. According to this idea, there is a non-linear relation between deformation of the cartilaginous tissue and its resistance to fluid flow [61]. The permeability normal to the contact of the cartilage matrix drops exponentially as its deformation increases [62]. A higher density of the fixed cartilage matrix explains it because the pores close during compression.

The dependence of cartilage permeability normal to the contact on the compressive pressure shows the same non-linear pattern [56]. It is established that the lateral permeability also depends on deformation and pressure within the cartilage matrix: the compression reduces it [28]. These experiments led to an important conclusion that the ability of the cartilaginous tissue to resist fluid flow in response to deformation causes a reverse reaction in compression manifesting itself by retarded consolidation of the cartilage matrix in the creep phase [56].

Modern data about the molecular composition and the supramolecular organization of the cartilage matrix enable to assess its resistance to the vector forces of compression and elongation active in real joints faced normal and parallel towards the joint cavity cartilaginous surfaces, respectively. The electron microscope studies revealed the inhomogeneous structure of the cartilaginous tissue [63]. For example,

Table 1.4 Moduli of elasticity of the normal joint cartilage in tension (E) and shear (G)

Elasticity modulus	Values, MPa	Information sources
Short-term E_0	12.20	[81]
	2.35	[44]
Equilibrium E_∞	7.24	[81]
	0.71	[44]
	3.67	[19]
Short-term G_0	4.18	[81]
	11.20	[45]
Equilibrium G_∞	2.65	[81]
	0.90	[19]

the collagen fibers arrange in the surface layer parallel to the cartilage contact and in the internal layers at an angle to it (Fig. 1.11). They orient in the surface layer in the direction of predominant movements of the joint; they are thinner and more densely packed than in the deeper zones [63, 64]. According to the above inhomogeneous structure, the cartilage has anisotropic properties along each of the main axes: two axes in the horizontal plane oriented along and across the fibers, and one in the vertical plane, i.e. through the cartilage body.

Hence, it is apparent that exactly the clearly pronounced dependence of the cartilage viscoelastic properties on the time of effect of the forces, on the one hand, and its observed inhomogeneous structure, on the other hand, cause variation of the elasticity moduli both during short and prolonged loads in compression, elongation and shear of the cartilaginous tissue. Moreover, the experimental determination of the latter shows that a significant role belongs also to a subjective factor. For example, the analysis in of the short-lasting and balanced elasticity moduli of the cartilaginous tissue during elongation and compression shows that the same elasticity moduli obtained by different authors have quite different values (Table 1.4). It is apparently due to application of different methods and unlike testing conditions.

The results of studies of the elastoviscous behavior of the cartilaginous tissue provide evidence in favor of this assumption. It is commonly believed that the moduli characterizing the elastic properties of the cartilage matrix in compression are within 0.7–3.5 MPa [44].

Nevertheless, the paper [65] shows that, in case the side surface of the cartilage specimen is enclosed into a rigid body [66] at a frequency within the interval 0.2–5 Hz, the dynamic modulus of compression of the human and animal joint cartilage is close to 100 MPa, i.e. the cartilaginous tissue in the conditions strongly restraining deformation behaves as a much stiffer material than in the free state.

Other studies point to the fact that the elasticity modulus in compression estimated by the method of the cartilage pile affects significantly the ratio between the diameter of the cartilage pile and its height [67, 68]. When the cartilage pile is larger than its diameter the elasticity modulus is 2.0–2.5 MPa. It reaches 1,000 MPa in the cartilage under load or close to that in natural joints, i.e. when the specimen diameter is several times larger than its height. It characterizes the elastic properties

in compression. It is apparent that the elasticity modulus in compression depends also on the deformation of the cartilage matrix. In fact, the data in [46] show than any change in the cartilage relative deformation in the range <0.6 boosts the dynamic elasticity modulus from 12 to 45 MPa. We observed the same regularities studying the deformation of cartilage specimens during friction on glass in the presence of various media [48], vis.: the elasticity modulus grows from 10.08 to 21.51 MPa as the relative deformation increases from 0.06 to 0.28.

The biomechanical properties of the joint cartilage depend too on the condition of the cartilaginous tissue proper. It is believed that the pathologically altered tissue has poorer viscoelastic properties than a normal articular cartilage. The results in confirm it showing that removal of the fine surface layer 33–53 μm thick leads to a rather noticeable change in the cartilage elasticity modulus from 8.2–10.3 to 6.8–9.3 MPa.

1.4 Mechanical Properties of Thin Layers of Cartilages at Microindentation in Various Environments

It was shown previously, that the cartilage represents an adaptive physically nonlinear material and its functional properties are controlled automatically by biochemical and biomechanical factors [69]. Behavior of the cartilage under load is possible due to interphase mobility—dynamically optimal relation of the cartilage matrix and synovia fluid.

The cartilage as two-phase composite is characterized by large gradient of deformational properties in depth, which is connected with morphological and biochemical changing of this structure including relation of solid and liquid phase. To determine the gradient of mechanical characteristics, for example, the modulus of elasticity, the measurements should be carry out for thin, as possible, layers of this biotissue. There are distinguished four zones in the articular cartilage that are different in matrix morphology and biochemistry—superficial, intermediate, radial, and calcified.

Each zone has its specific function but all they have to resist to compression when the joint is under load. Structural features of each zone define their contribution to the cartilage deformation and recovery. For example, the superficial zone important for the compressive strength of the matrix has thickness 0.2–0.6 mm in human cartilages. It is the thinnest cartilage zone and therefore it would be reasonable to test 0.2 mm thick cartilage slices. However, the thinner indented sample, the more influence of the substrate on the measurement results.

Taking into account the joint functioning (compression of conformal bodies), it is suitable to use contact diagnostics with indenters having the similar curvature as joint [70]. Elasticity modulus may be determined according the Hertzian solution for sphere and elastic half space compression. However, because of small thickness

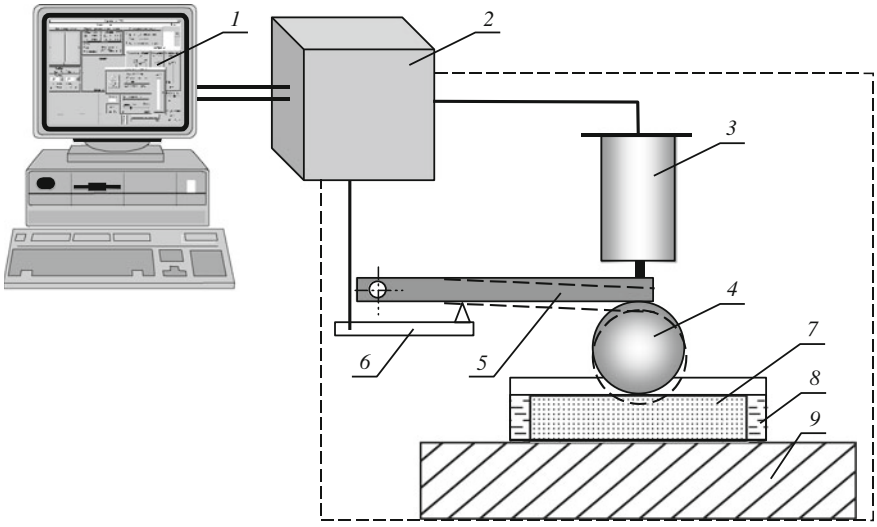


Fig. 1.12 Schematic diagram of the device for cartilage microindentation: *1* is computer; *2* is electronic unit; *3* is electromagnet; *4* is spherical indentor; *5* is arm for the indentor supporting; *6* is piezoelectric atomic force microscope probe; *7* is sample; *8* is liquid medium; *9* is lifting platform

of sample the influence of substrate is increasing and reducing the accuracy of this estimate.

Obviously, smaller indentation depth can reduce influence of rigid substrate. In this case, reasonable correlation between the sample thickness and indentation depth can help to obtain reliable data. These conditions may be provided at microindentation of thin cartilage slices.

To realize such scheme, a specialized instrument was developed (Fig. 1.12).

The instrument used stainless steel ball indentor of diameter 4 mm pressed to the examined sample in static or dynamic mode under computer control. A measurement circuit estimated indentation depth with nanometer accuracy that allowed applying minimum loads to the sample. Small loads and indentation depth enabled examination of thin layers under pressure below the yield stress.

Nanometer accuracy of the indentation depth measurement was reached with the help of atomic force microscope probe and corresponding system of the displacement detection. Calibration of the indentation depth measurement circuit was fulfilled according the routine accepted for surface probe microscopes.

Loading system used an electromagnet controlled by host computer via specialized electronic unit (Fig. 1.12). The instrument measurement and feedback circuits allowed indentation either under constant load or for the preset depth. The loading system was calibrated under the scheme of dead weight compensation.

For static loading, the sample was subjected to single indentation cycle: load was applied gradually so that indentation rate was about 100 micron/min, then it was kept at maximum level for 60 s and after that the load was gradually decreased. In the case of dynamic loading, cycles of the load increase and successive sharp drops

were repeated necessary number of times. Frequency of a loading/unloading cycle was about 0.1 Hz. Maximum indentation depth under both operation modes did not exceed 200 microns. Force applied by indentor to the sample reached 16 N under static loading and 3.5 N under dynamic mode.

The instrument design provided also for the sample placing in liquid medium that enabled investigations of influence of different media on the sample deformational characteristics.

Host computer registered indentation depth and load and saved the data in a file on hard disk. The results were processed and analyzed with help of software.

Using the instrument, cartilage samples of pork knee joint were tested for compression and recovery in different media. The cartilage samples were taken not later than 12 h post mortem and tested without storing. The samples were prepared as follows. Cylinder of diameter 6 mm was cut off medial menisci in the thickest area (about 3 mm). Then each cylinder was sliced parallel to articular surface for samples with thickness 0.5 mm starting from the surface. So, cartilage slices from the depth 0, 0.5, 1.0, 1.5, and 2.0 mm under the articular surface were tested.

In the experiments, the following fluids were used as liquid media: (1) the natural synovia taken not later than 12 h post mortem and tested without storing; (2) the pseudosynovia prepared according [71] and representing 2 wt% water solution of carboxymethyl cellulose sodium salt with addition of inorganic salts; (3) the pseudosynovia as in medium 2 with addition of 2 wt% liquid-crystalline cholesterol esters [72]; (4) the blood serum; (5) the physiological solution. Tested samples were immersed in medium so that the fluid did not cover cartilage slice. Experiments were conducted at ambient temperature of 22–24 °C.

Indentation of the cartilage samples under static mode was fulfilled for the preset depth. The applied load and indentation depth were registered and then were to calculate elastic modulus. Elastic modulus for the case of spherical indenter of radius r was defined according to Hertzian equation:

$$E = \frac{3}{4}\frac{P}{\sqrt{r\delta^3}} \tag{1.2}$$

where P is applied load; δ is indentation depth.

Analysis of the experimental results (Fig. 1.13) shows that rheological characteristics of the surrounding medium effect the cartilage matrix reaction on the applied load.

Among the tested fluids, the natural synovia imparted the highest resistance to the samples. The pseudosynovia doped with liquid crystals showed the closest to the synovia properties. That may be a confirmation for the theory proposed in [72] stating that liquid-crystalline compounds of the natural synovia are responsible for the cartilage matrix deformational and tribological properties.

Highest elastic modulus was observed for the cartilage surface layers. Deeper in the cartilage, stiffness of the matrix was lower than in surface layers. This result agrees well with known data on the elastic properties of the cartilage zones [72, 73]. Collagen fibres and flattened chondrocytes forming the superficial zone are oriented

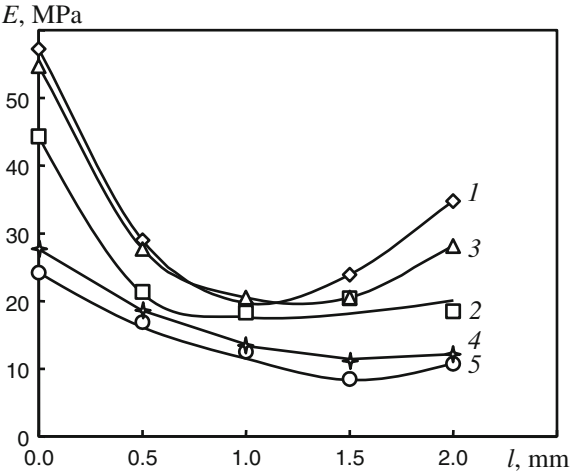


Fig. 1.13 Distribution of elastic modulus values across the cartilage depth under static loading for different media: *1* is the natural synovia; *2* is the pseudosynovia; *3* is the pseudosynovia doped with liquid crystals; *4* is the blood serum; *5* is the physiological solution

in the cartilage along the articular surface. Such structure imparts high tensile and compressive resistance to this area of the matrix. In contrast to superficial layers, collagen fibres and other matrix components in inner cartilage zones are either randomly arranged or oriented perpendicular to the articular surface [27]. It is seemed that heterogeneous structure of the cartilage besides interface mobility has marked dependence elastic modulus versus pressure which was observed previously in [72].

Measurement of the sample recovery immediately after unloading showed that the least hysteresis was obtained when the synovial fluid washed the cartilage slices (5–10 %). In medium of the pseudosynovia doped with liquid crystals, the closest result was reached (10–15 % hysteresis). Hysteresis of the cartilage slices in the pseudosynovial fluid was 20–30 %. The highest values of the parameter were obtained with the physiological solution: some samples showed more than 50 % hysteresis.

So, the best compression characteristics for the cartilage slices were obtained with the natural synovial fluid as medium. Natural components of the cartilage synovial medium work together and provide the best performances to the entire joint.

Indentation of the cartilage slices under dynamic mode was done at constant load. Indentation depth increased with each loading/unloading cycle as the cartilage matrix recovery lagged and electronics traced these changes to maintain constant load. Experiments were stopped when indentation depth reached 200 microns.

Results of the cartilage slice compression under dynamic mode were practically the same as under static loading (Fig. 1.14).

However, calculated values of the elastic modulus were lower, especially for the surface layers. That may be caused by slow recovery of the cartilage matrix after the unloading resulting in deeper and deeper indentation with each cycle. So, total

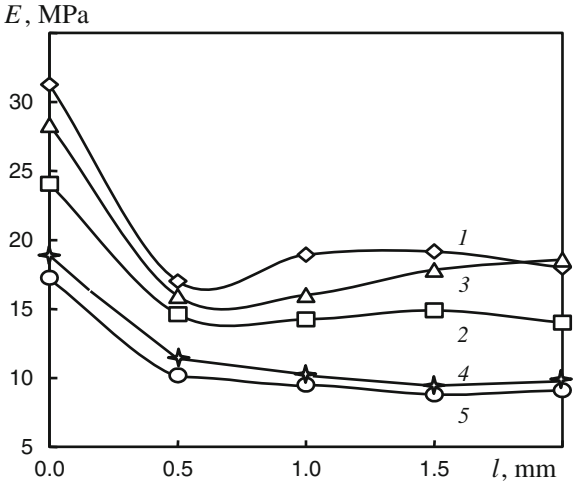


Fig. 1.14 Distribution of the elastic modulus values across the cartilage depth under cyclic loading in: *1* is the natural synovia; *2* is the pseudosynovia; *3* is the pseudosynovia doped with liquid crystals; *4* is the blood serum; *5* is the physiological solution

hysteresis accumulated during all previous cycles contributed to the final depth used then for calculations of the elastic modulus values.

It should be also mentioned that distribution of the elastic modulus values across inner layers of the cartilage differs from that observed under static loading: the deepest zones showed slower recovery of the matrix in this area.

For the surface layers, a ratio between the maximum indentation depth δ_{max} and the pit depth after the sample unloading δ_{rec} is plotted in Fig. 1.15.

The natural synovia provided faster and more full recovery of the surface cartilage layers under cyclic loading and unloading than other media. The pseudosynovia doped with liquid crystals gave the closest results again. It should be mentioned that $\delta_{max}/\delta_{rec}$ ratio increased with time that means that resistance of the cartilage rose with compression. However, this regularity was not kept when the physiological solution and blood serum served as a medium. Probably, high viscosity and non-Newtonian properties of first three fluids contributed significantly to the observed behavior of the cartilage samples.

So, analysis of the cartilage behavior under loading/unloading cycles allows checking the artificial lubricants adequacy to the natural synovia. Such analysis may be done from the data obtained at microindentation of the cartilage samples in tested medium.

From all the viewpoints, the synovia remains ideal fluid for the joint and should be used as a standard for model media. Analysis of the cartilage compression and recovery under static and cyclic loading in different media allows comparison of artificial lubricants with the natural synovial fluid.

Thus, the analysis asserts that the articular cartilage is a composite material. Mechanical, biological and biochemical factors influence strongly its biomechanical

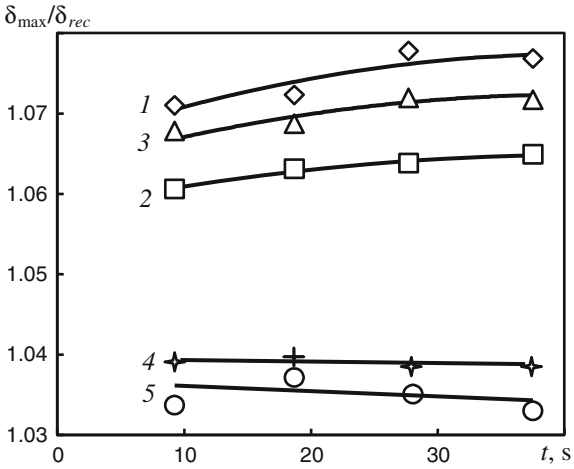


Fig. 1.15 Kinetics of ratio of the maximum indentation depth and indentation depth after unloading under cyclic loading in different media: *1* is the natural synovia; *2* is the pseudosynovia; *3* is the pseudosynovia doped with liquid crystals; *4* is the blood serum; *5* is the physiological solution

properties. It is quite apparent that integral mechanical functions of the cartilage are possible only providing the optimal qualitative and structural interrelation between all the components of the cartilage matrix is maintained. Any disorder of this interrelation upsets inevitably the biomechanical properties of the joint cartilage. It relates inseparably to still another main element of the synovial medium in joints, vis. the synovial fluid. It is a sole visible source of nutrition of cells (chondrocytes) in the cartilage matrix that are known to produce proteins, collagen, and glicosaminoglycans, i.e. the necessary construction material that governs the main parameters of the cartilage, such as elasticity, shock-absorbing ability [4, 44].

1.5 Role of Synovia as Protector of Joint Friction Surfaces

According to modern ideas about the synovial medium in the joints, the biomechanical effect of their normal functioning implies circulation of the interstitial fluid in the articular cartilage in motion [27]. The released interstitial fluid mixes up with the surrounding synovial fluid before being re-absorbed by the articular cartilage. This process is an essential tool of maintaining the cartilaginous tissue viability because the absorbed fluid contains the nutritive substances vital for chondrocytes. A number of experiments in vivo confirm this concept when the synovial fluid was exposed to continuous compressive loading during a long time. The paper shows that continuous compression of the cartilage both prevents access of the synovial fluid to the contact under pressure and interrupts the interstitial water diffusion through the intercellular substance of the cartilage matrix in the contact region. As a

result, the chondrocytes perish; the intercellular substance degenerates because any deformation of the cartilage upsets the normal mechanism of fluid transfer.

Restriction of motions of joints in osteoarthritis can also lead to generative changes in the articular cartilage. The earliest centers of destruction of the tissues appear in the regions of the joints that cease to experience the mechanical effect of the body weight normal for load-carrying surfaces [74]. These clinical observations evidence in favor of the above-proposed concepts.

According to the data in [4, 75], the mechanical role of the synovial fluid is that, by possessing sufficient variable viscosity, it absorbs extreme loads and protects the articular surfaces. The synovial fluid is known to be an aqueous solution of many substances (Table 1.5). It contains in addition to protein a biopolymer that is very essential to maintain the pseudoelastic properties, vis. hyaluronic acid (HUA) [1].

The conformation of the HUA molecules is commonly determined as a loose chaotic ball with a radius 150–400 nm; its volume exceeds 10 times the volume of the molecule proper. Some researchers believe that it is more justified to view the confirmation of the HUA molecules as a "cell" formed by a branched polymer with cross-linked branches [1]. In other words, the HUA produces a complex spatial structure in solutions capable to bind significant quantities of water.

The HUA molecules in solutions can combine with protein. The HUA complex in the synovial fluid with proteins of various natures is known as "mucin", a substance that precipitates quickly when acetic acid is added. It is believed that the bond between the protein and the HUA is weak [2]. The electric charge of this proteinopolysaccharide complex is strongly negative so that the solution tends to spherical configuration. According to the existing idea, mucin contains 66 % albumins and 34 % HUA. Later data show that mucin has a considerably more complex composition and the relationships between protein and polysaccharide components. However, notwithstanding that it is quite hard to separate protein and HUA methodologically, one should always bear in mind that the HUA molecules have non-valence bonds with glycoproteins. Exactly this fact governs many physico-mechanical properties of hyaluronate and those tissue components that contain it.

The most essential property of the proteinopolysaccharide complex in the synovial fluid is its viscosity. Earlier studies determined the relative viscosity of the synovial fluid (i.e. in respect to the water viscosity) with a very large scatter, within a range from 5.7 to 1160 cPs.[1] Later it was established that this scatter does not in fact exist because the HUA in the proteinopolysaccharide complex responsible for the viscosity of the synovial fluid and exactly its concentration determines the rheological properties of the synovial fluid. The fact that heating with tripsin and addition of the protease fail to reduce the synovial fluid viscosity proves it; meanwhile addition of hyaluronidase, a HUA depolymerizing enzyme, does reduce the synovial fluid viscosity down to the water viscosity [40, 76]. However, since the synovial fluid contains the long-chain HUA macromolecules, it turns out

[1]For comparison, the water viscosity at 20 °C is equal to 1 cPs.

Table 1.5 Comparative composition and physical properties of the synovial fluid and blood plasma

Indicators	Synovial fluid	Blood plasma
Water (%)	94.0–95.0	90.0–91.0; 93.0–95.0
Solid substance (%)	5.0–6.0; 3.4(1.2–4.83)	5.0–7.0; 9.0–10.0
Inorganic		
Salts (%)	0.7–0.8	0.78–0.95
Organic		
Protein (g\l)	2.5; [1.72(0.45–3.15) g%]	6.5
Inc. of the total quantity		
Albumin (%)	67.0 [1.89 g%]	52.0; 63.6 ± 4.0
Globulin (%)	[0.91 g%]	48.0; 36.4 ± 1.63
% of these		
α_1	8.0	5.0; 4.1 ± 0.98
α_2	4.0	11.0; 7.4 ± 1.12
β	11.0	14.0; 9.9 ± 1.92
γ	10.0	18.0; 15.0 ± 2.52
Fibrinogen	Absent	Present always, 270 mg%
Cartilage protein-polysaccharide	Present in the α_2 and β – globulin zone	Absent
Hyaluronate (g\l)	3.0	Absent
Hyaluronic acid (mg%)	157.0 (4.0–297.0)	Absent
Enzymes	Diastase, protease, acid phosphatase, etc. (up to 30)	Amilase, protease lipophosphatase. Esterase, etc.
Urea (g\l)	0.2–0.4	0.2–0.4
Uric acid (g\l)	0.015 [3.9 (3.3–4.7) mg%]	0.015
Lipids (g\l)	0.9–11.3	4.0–8.0
Glucose (g\l)	0.66	0.7–1.0
Physical properties		
Appearance	Light yellow. Transparent	
Quantity	1.0–2.0 cm^3	
Viscosity, conv. units	5.0–7.0	4.7
Friction coefficient	0.001–0.03	
Osmotic pressure, mm water column	120.0–140.0	[7.5–8.0 atm]
Medium pH	7.768; 7.39 (7.29–7.45)	7.41; 7.35–7.4

pesudoelastic or thixotropic, i.e. it possesses the properties of a fluid with the viscosity that diminishes from the uppermost limit corresponding to slow shear rates to the lowermost limit corresponding to quick shear rates [74, 76].

Good lubricity of the synovial fluid was attributed in early 50 s exactly to the thixotropic properties. However, this conclusion turned out premature. The fact is

that addition of hyaluronidase to depolymerize the HUA, on the one hand, failed to eliminate the thixotropic properties of the synovial fluid, hence it did not practically affect its lubricity [1, 40, 76]. On the other hand, the purified HUA itself does not manifest any lubricating effect [77], so the idea is deduced that the proteinopolysaccharide complex in the synovial fluid performs apparently a somewhat different function rather than playing the role of lubrication of synovial joints.

The results of studies of the elastic properties of the synovial fluid confirm this assumption. Initially these studies were performed in 1953 when the synovial fluid was compressed between a flat convex lens and an optic plane and it was shown that it was impossible to force the fluid out from the space between the surfaces even under a pressure 10 MPa.

Some time later, in 1963, it was assumed that the molecular structural complex of HUA can explain the viscoelastic properties of the synovial fluid and proteins the elasticity of which exactly can create the shock absorbing effect. The results of the studies reported in [4, 78] supported this hypothesis too. It was established that the polyelectrolyte spheres of the proteinopolysaccharide complex in the synovial fluid tend to interact producing three-dimensional nets that turn the synovial fluid into a gel in the static conditions. Later these assumptions re-appeared in. The synovial fluid can be modeled as a five-element viscoelastic material of Kelvin-Foigt. Moreover, it was shown that the cartilage surface could adsorb HUA molecules producing a necessary protective layer in the process of joint operation. Whence a conclusion follows that any heavier loading on the articular contact enables the synovial fluid to form a three-dimensional molecular complex on the rubbing surfaces. This complex has the consistency of the gel and elasticity due to the squeezing out of low-molecular components when the hyaluronate concentration is high [79].

At present it is believed that that exactly the hyaluronate in the synovial fluid as such gel forms the necessary leveling interlayer between the articular ends of articulated bones and its physico-mechanical properties in combination with the joint cartilage properties ensure the functional congruence of rubbing surfaces in locomotion [42, 79]. In other words, together with the tribological and biological functions, the synovial fluid performs essential biomechanical functions jointly with the cartilage matrix that are needed to maintain the optimal conditions for the functioning of synovial joints.

Therefore, it is quite apparent that the synovial fluid with abnormal rheological properties is unable to perform sufficiently its functions of the lubricant, shock absorber, and protector in joints. This disorder of many parameters of the articular medium is observed during rheumatic afflictions of joints: the rheumatoid arthritis (RA) and deforming osteoarthrosis (DOA). In the case of DOA pathogenesis an essential role belongs to heavier mechanical loading on the cartilage contact, depolymerization and reduction of the concentration of proteoglycans resulting in changes in the elastohydrodynamic properties of the cartilage matrix and damage of the diffusion processes [28]. The synovial fluid secretion becomes deficient and its rheological properties change. In motion and loading on the articular cartilage, the synovial fluid leaks out like water from the sponge without performing its functions

of the protector and shock absorber [80]. It leads to continuous microtraumas of the cartilages, which become less elastic, dry, and rough; they lose their viscoelastic properties and undergo mechanodestruction.

In the RA the catabolic phase of the metabolism and the activity of Proteolytic enzymes, disintegration of the proteoglycans in the cartilage intensify, HUA depolymerizes and its concentration in the synovial fluid drops [34, 77]. The rheumatoid process makes the synovial fluid less viscous and deprives it of a significant part of its protective functions.

Thus, it follows from the above discussion that the synovial medium in joints is an extremely intricate dynamically balanced system of a whole complex of structurally differentiated, functionally integrated elements with essential biomechanical properties.

References

1. V.N. Pavlova, *Synovial Medium in Joints* (Meditsina, Moscow, 1980), p. 296. (in Russian)
2. P. Kumar, M. Oka, J. Toguchida et al., Role of uppermost superficial surface layer of articular cartilage in the lubrication mechanism of joints. J. Anat. **199**, 241–250 (2001)
3. M. Kobayashi, M. Oka, The lubricative function of artificial joint material surfaces by confocal laser scanning microscopy: comparison with natural synovial joint surface. Biomed. Mater. Eng. **13**, 429–437 (2003)
4. V.C. Mow, A. Ratcliffe, S.L.-Y. Woo (eds.), *Biomechanics of Diarthrodial Joints*, vol. 1 (Springer-Verlag, New York, 1990), p. 451
5. R. Zhang, M. Lu, Z. Lan, Biochemical research of joint. 3. An experimental research on the Femur's knee joint of pongidae. Acta. Mech. Sin. **18**(2), 181–185 (1986)
6. R. Alexander, Forces in animal joints. Eng. Med. **9**(2), 93–97 (1980)
7. C.J. Snijders, Einfuhrende Biomechanik der Gelenke. Theor. Fortschr. Praht. Erfahrungen Manuellen Med. 234–243 (1980)
8. F. Pauwels, *Biomechanics of the Locomotor Apparatus* (Springer, Berlin, 1989), pp. 329–374
9. K. Fujikawa, B.B. Seedham, V. Wright, Biomechanism of the patello-femoral joint. Part 2: a study of the effect of simulated Femoro-Tibial Varus deformity on the congruity of the patello-femoral compartment and movement of the patella. Eng. Med. **12**(1), 13–21 (1983)
10. H.H. Huberti, W.C. Hayes, Tendon patello-femoral contact stress at large flexion angle. *Proceedings 35th Annual Conference English Medicine Biology*, Philadelphia, vol. 24, p. 46 (1982)
11. K. Fujikawa, B.B. Seedhom, V. Wright, Biomechanism of the patello-femoral joint. Part 1: a study of the contact and the congruity of the patello-femoral compartment and movement of the patella. Eng. Med. **12**(1), 3–11 (1983)
12. D.B. Kattelkamp, Chemical implications of knee biomechanics. Arch. Surg. **107**(3), 406–410 (1973)
13. D.E. Hastings (ed.), *The Knee: Ligament and Articular Cartilage Injuries* (Springer, New York, 1978), p. 191
14. A.H. Hofman, P. Grigg, D.M. Flinn, A biomechanical model of the posterior knee capsule of the cat. Trans. ASME. J. Biomech. Eng. **107**(3), 140–146 (1985)
15. K. Schuldt, J. Ekholm, G. Nemeth et al., Knee load and muscle activity during exercises in rising. Scand. J. Rehabilit. Med. **9**, 174–188 (1983)

16. S. Hirokawa, Analytical study on three-dimensional model of femoro-patellar joint. Trajectory pattern and load bearing capacity on the contact points. Trans. Soc. Instrum. Contr. Eng. **22**(5), 580–587 (1986)
17. T.P. Andriachi, R.P. Mikosz, S.J. Hampton et al., Model studies of the stiffness characteristics of the human knee joint. J. Biomech. **16**(1), 23–29 (1983)
18. C.H. Barnett, D.V. Davies, M.A. McConaill, *Synovial Joints* (Longmans, London, 1961), p. 304
19. D. Dowson, Medical engineering—the multi-disciplinary challenge. Proc. Inst. Mech. Eng. **205**, 1–10 (1991)
20. A.M. Ahmed, D.L. Burke, In vitro measurement of static pressure distribution in synovial joints. Part 1: tibial surface of the knee. Trans. ASME. J. Biomech. Eng. **105**(3), 216–225 (1983)
21. P.G.J. Maquet, Biomechanics of the knee, 2nd edn. (Springer-Verlag, Berlin, 1984), p. 305
22. A.S. Voloshin, J. Wosk, Shock Absorption of meniscectomized and painful knees: a comparative in vivo study. J. Biomed. Eng. **5**, 157–161 (1983)
23. C.J. Wirth, M. Rodriguez, K.A. Milachwski, *Meniskusnaht – Meniskusersatz* (New York, Georg Thieme, Stuttgart, 1988), p. 144
24. P. Ghosh, J. Sutherland, Ch. Bellenger et al., The influence of weight-bearing exercise on articular cartilage of meniscectomized joints. Clin. Orthop. Related. Res. **252**, 101–113 (1990)
25. L.L. Johnson, *Arthroscopic Surgery: Principles and Practice* (Mosby, St. Louis, 1986), p. 286
26. W. Glotzer, K.P. Benedetto, Ch. Rangger, Progredienz des Knorpelschadens (Arthroserate) Nach Partieeller Arthroscopischer Meniskusteilresektion, ed. by W. Glinz Arthroskopie bei Knorpelschaden und bei Arthrose. Stuttgart: Enke, pp. 18–25 (1990)
27. V.N. Pavlova, T.N. Kopjeva, L.I. Slutskii, G.G. Pavlov, *Cartilage* (Meditsina, Moscow, 1988), p. 320. (in Russian)
28. J.P. Fulkerson, C.C. Edwards, O.D. Chrisman, Articular Cartilage, Chap. 12. The Scientific Basis of Orthopedics, 2nd edn. Los Altos, pp. 347–371 (1987)
29. D.H. Manicourt, D.C. Pite, D.S. Howell, New glycoproteins associated with proteoglycan aggregates. Biol. Cartilage, 34–35 (1983)
30. B. Alberts, D. Bray, J. Lewis, et al., *Cell Molecular Biology*. In 3 Volumes (English trans.), Moscow, Mir, p. 234 (1987)
31. T. Hardingham, Proteoglycans. Biomech. Soc. Trans. **9**(6), 489–497 (1981)
32. J.R. Baker, B. Caterson, J.E. Christner, The link proteins of cartilage proteoglycan aggregates: structure and function. Ala. J. Med. Sci. **18**(1), 46–52 (1981)
33. J.W. Lash, N.S. Vasan, Glycosaminoglycans of Cartilage, ed. by B.K. Hall. *Cartilage. Structure, Function, and Biochemistry*, vol. 1 (Springer-Verlag, New York, 1983), p. 215
34. R.D. Altman, D.S. Howell, N.L. Cottlieb, Introduction: new directions in therapy of osteoarthritis. Sem. Arthritis Rheumatism **17**(2), 1–2 (1987)
35. B.I. Kupchinov, S.F. Ermakov, E.D. Beloenko, *Biotribology of synovial joints* (Vedy, Minsk, 1997), p. 272. (in Russian)
36. J. Edwards, Physical characteristics of articular cartilage. Proc. Inst. Mech. Eng. **181**, 16–24 (1967)
37. R. Mayne, R.E. Burgeson (eds.), *Structure and Function of Collagen Types* (Academic Press Inc, Orlando Florida, 1987), p. 318
38. A.D. Wagget, C.M. Kielty, C.A. Shuttleworth, Microfibrillar elements in the synovial joint: presence of Type VI collagen and fibrillin—containing microfibrils. Ann. Rheum. Dis. **52**(6), 449–453 (1993)
39. R. Alexander, *Biomechanics* (Moscow, Mir, 1970), pp. 73–103. (in Russian)
40. V. Wright, D. Dowson, J. Kerr, The structure of joints. IV. Articular Cartilage. Int. Rev. Connect. Tissue Res. 6, 105–124 (1973)
41. A.J. Grodzinsky, V. Roth, E. Myers et al., The significance of electromechanical forces in the nonequilibrium swelling behavior of articular cartilage in tension. Trans ASME. J. Biomech. Eng. **103**(4), 221–231 (1981)

42. F.K. Linn, Lubrication of joints in animals. Problems of friction ad lubrication. ASME Trans. **2**, 141–153 (1969)
43. A.N. Romanovskaja, L.G. Voskresenskii, Comparative study of rheological properties of synthetic and biological polymeric materials: an example of filled silicon rubber and human articular cartilage. Mech. Compos. Mater. **5**, 906–909 (1984) (in Russian)
44. L. Sokoloff, Elasticity of aging cartilage. *Proceedings of Fed*, vol. 25, no 3, pp. 1089–1095 (1966)
45. I.C. Clarke, R. Contini, R.M. Kenedi, Friction and wear studies of articular cartilage: a scanning electron microscope study. Trans. ASME **F97**(3), 358–368 (1975)
46. C.R. Johnson, D. Dowson, V. Wright, The elastic behavior of articular cartilage under a sinusoidally varying compressive stress. Int. J. Mech. Sci. **19**(5), 301–308 (1977)
47. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov, E.D. Beloenko, Structure-mechanical and antifriction properties relationship in synovial fluid of joints. Mech. Compos. Mater. **2**, 240–246 (1988)
48. S.F. Ermakov, E.D. Beloenko, O.L. Eismont, Role of liquid crystals in tribological behavior of joint cartilages. J. Friction Wear **25**(5), 31–35 (2004)
49. E. Beloyenko, S. Ermakov, O. Eismont, Cholesterol Liquid Crystals (ChLC) effect on cartilage matrix structures and cells. Abstr. Symposium SIROT, Haifa, p. 93 (1997)
50. Y.M. Pleskachevsky, S.V. Shilko, S.F. Ermakov, Methods of wear redaction based on bioprototypes of tribojoints. J. Synth. Lubr. **22**, 225–236, (2005)
51. S.F. Ermakov, B.I. Kupchinov, V.G. Rodnenkov, E.D. Beloenko, O.L. Eismont, Influence of nature of rubbing surfaces and lubricant on articular cartilage friction. Acta Bioeng. Biomech. **3**(supl. 1), 65–71 (2001)
52. Z.M. Jin, D. Dowson, J. Fisher, The effect of porosity of articular cartilage on the lubrication of a normal human hip joint. Proc. Inst. Mech. Eng. **206**, 117–124 (1992)
53. S. Uchids, M. Morita, H. Fujimasu, Frictional behaviour and viscoelastic properties of articular cartilage. J. Jap. Soc. Lubric. Eng. **23**(2), 117–123 (1978)
54. P.A. Torzilli, D.A. Dethmers, D.E. Rose, H.F. Schryuer, Movement of interstitial water through loaded articular cartilage. J. Biochem. **16**(3), 169–179 (1983)
55. P.A. Torzilli, Mechanical response of articular cartilage to an oscillating load. Mech. Res. Commun. **11**(1), 75–82 (1984)
56. V.C. Mow, J.M. Mansour, The nonlinear interaction between cartilage deformation and interstitial fluid flow. J. Biomech. **10**(1), 31–39 (1977)
57. C.R. Flannery, C.R. Hughes, B.C. Schumacher et al., Articular cartilage superficial zone protein (SZP) is homologous to megakaryocyte stimulating factor precursor and is a multifunctional proteoglycan with potential growth promoting, cytoprotective and lubricating properties in cartilage metabolism. Biochem. Biophys. Res. Commun. **234**, 535–541 (1999)
58. Z. Zea-Aragon, N. Terada, N. Ohno et al., Replica immunoelectron microscopic study of the upper surface layer in rat mandibular condylar cartilage by a quick- freezing method. Histochem. Cell Biol. **121**, 255–259 (2004)
59. V.C. Mow, W.C. Hayes (eds.), *Basic Orthopaedic Biomechanics* (Raven Press, New York, 1991), pp. 245–293
60. P.A. Torzilli, V.C. Mow, On the fundamental fluid transport mechanisms through normal and pathological articular cartilage during function: the formulation. J. Biomech. **9**(8), 541–552 (1976)
61. J.M. Mansour, V.C. Mow, The Permeability of articular cartilage under compressive strain and at high pressures. J. Bone Joint Surg. **58A**, 509–516 (1976)
62. V.C. Mow, W.M. Lai, Mechanics of animal joints. Ann. Rev. Fluid Mech. **11**, 247–288 (1979)
63. C. Weiss, L. Rosenberg, A.J. Helfet, An ultrastructural study of normal young adult human articular cartilage. J. Bone Joint Surg. **50A**, 663–664 (1968)
64. G. Meachim, D. Denham, I.H. Emery, P.H. Wilkinson, Collagen alignments and artificial splits at the surface of human articular cartilage. J. Anat. **118**, 101–102 (1974)

65. A.M. Romanovskaja, Y.S. Zuev, M.N. Khotimskii, Shock absorbing properties of some tissues and structures of mammal locomotorium. Mech. Comp. Mat. **5**, 911–915 (1985). (in Russian)
66. G.R. Higgison, J.E. Snaith, The Mechanical stiffness of articular cartilage in confined oscillating compression. Eng. Med. **8**(8), 11–14 (1979)
67. T.A. Prokhorova, The effect of normal pressure and temperature on the friction properties of articular cartilages, Deposited with VINITI, no 1080–B, p. 15 (1986) (in Russian)
68. T.A. Prokhorova, Friction properties of articular cartilages in case of sudden normal loading. Deposited with VINITI, no 1081–B, p. 26 (1986) (in Russian)
69. S.V. Shilko, S.F. Ermakov, The role of liquid phase and porous structure of cartilage in formation of biomechanical properties of joints. Part 1. Russian. J. Biomech. **12**(2), 31–39 (2008)
70. A.A. Suslov, S.F. Ermakov, A.V. Beletzky, S.V. Shilko, V.I. Nikolaev, The role of liquid phase and porous structure of the cartilage in formation of biomechanical properties of the joints. Part 2. Russian. J. Biomech. **12**(4), 31–37 (2008)
71. Patent GB1391577, Int. Cl. C08L 1/28, A61 K 317/15. Pseudo-synovial plastic body fluids and method of preparing same / C.A. Homsy. - № 11352/73, Filed 8.03.73, Complete Specification published 23.04.75
72. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov, S.N. Bobrysheva, E.D. Beloenko, The effect of liquid crystals on joint lubrication. Wear **171**, 7–12 (1994)
73. G.E. Kempson, M.A.R. Freeman, S.A.V. Swanson, The determination of creep modulus for articular cartilage from indentation tests on the human femoral head. J. Biomech. **4**, 239–250 (1971)
74. J.M. Mansour, V.C. Mow, On the natural lubrication of synovial joints: normal and degenerate. Trans. ASME **F99**(2), 163–173 (1977)
75. V.V. Vasilenkajtis, Biopolymers, functional molecules of the rheumatoid and arthritic articular medium and justification of some methods of its rheological drug correction. Trans. Riga Res. Inst. Traumatol. Orthop. **13**, 525–536 (1975) (in Russian)
76. H. Lipshitz, R. Etheredge, M.J. Climcher, In vitro studies of the wear of articular cartilage. the wear characteristics of chemically modified articular cartilage when worn against a highly polished characterized stainless steel surface. J. Biomech. **13**, 423–436 (1980)
77. V. Rejholec, Long term studies of antiosteoarthritic drugs: an assessment. Sem. Arthritis Rheumatism **17**(2), 35–53 (1987)
78. S.F. Ermakov, Modern conceptions on biomechanics of human synovial joints. Mech. Compos. Mater. **4**, 539–556 (1992)
79. R.A. Brand, Joint lubrication. Ch. 13. *The Scientific Basis of Orthopaedics*, 2nd edn. (1987), pp. 373–386
80. V.V. Vasilenkajtis, Artificial synovial fluid for joints orthoped. Traumatol **10**, 11–15 (1989). (in Russian)
81. W.G. Hayers, L.F. Mocros, Viscoelastic properties of human articular cartilage. J. Appl. Physiol. **31**, 562–568 (1971)
82. A.I. Kaplan, Cartilage. World Sci. **12**, 26–35 (1984)

Chapter 2
Brief Review of Liquid Crystals

Abstract Examined and summarized the types of liquid crystals. Analyzed structural characteristics of smectic, nematic and cholesteric liquid crystals. It is noted that cholesteric liquid crystals are helically twisted structure, the pitch of the helix which is temperature dependent and individual chemical properties of liquid-crystalline compounds of cholesterol. Results on the influence of temperature on rheological properties of cholesteric liquid crystals are presents.

This chapter describes briefly main ideas of liquid crystals, their structure and properties. Such approach seems reasonable since it does not require searching corresponding publications.

2.1 Liquid-Crystal State of Matter and Kinds of Liquid Crystals

There are three phase states of the matter, namely crystalline, liquid, and gaseous. In each of them a given substance has certain characteristics typical only for this substance. For crystalline substances these characteristics are the temperature and heat of melting, for liquids—the boiling temperature and the heat of vaporization. These and other characteristics of substances in different phase states are used to identify them.

2.1.1 History of Liquid Crystal Discovery

At the end of the 19th century Austrian scientist Fridrich Reinitzer from the Institute of Plant Physiology studied two substances extracted from carrot, namely light-colored hydrocarotin and dark-red carotin. He knew from publications that hydrocarotin was similar to cholesterol; at least they had like compositions and chemical properties. However, Reinitzer did not decide to consider hydrocarotin as

S. Ermakov et al., *Liquid Crystals in Biotribology*,
Biological and Medical Physics, Biomedical Engineering,
DOI 10.1007/978-3-319-20349-2_2

pure cholesterol without a thorough study of its chemical and physical properties. Most probably, hydrocarotin was a mixture of cholesterol compounds or one of cholesterol derivatives, as Reinitzer believed. To clarify the situation, he decided to study some pure cholesterol derivatives especially as their physical and chemical properties were studied poorly. Moreover, the chemical formula of cholesterol had not been determined certainly. Reinitzer hoped to solve this problem at the same time. If all or almost all would be known about cholesterol it would be easier to identify components extracted from hydrocarotin at least by comparison.

This work took one and a half year. Reinitzer used industrial cholesterol. He refined it thoroughly and determined its physical and chemical properties. Using element analysis methods he derived the precise chemical formula of cholesterol—$C_{27}H_{46}O$.

This result was a great achievement in the chemistry of natural substances. Reinitzer also described in detail cholesterol derivatives. In addition to cholesterol acetate and bromeacetate, he synthesized and studied nitrocholesterol and cholesterolbenzoate. Two esters, namely cholesterolacetate and cholesterolbenzoate, were of special interest.

Both substances in the solid phase are white fine-crystalline powders. When Reinitzer determined the melting temperature by melting in glass capillaries he saw that they became colored. At that time nobody understood that strange fact of coloring of colorless compounds in melting. Reinitzer described the coloring of cholesterolacetate as follows: “When cooling melted cholesterolacetate one is surprised by very nice color appeared before cholesterolacetate thickens. This phenomenon is seen in long capillaries used to determine the melting temperature, yet it is better if the substance is melted between a table and cover glasses. In reflected light bright emerald color appears in some areas which is rapidly spread over the whole specimen bulk and then transforms to blue-green, in places to dark-blue, then to yellow-green, orange, red, and, finally, dark-red”.

Cholesterolbenzoate demonstrated slightly different behavior: two colored areas appeared in it, one of them was one-color and the other—multicolor just as in case of cholesterolacetate. It appeared at lower temperatures. No such effect was observed up that time.

Another interesting phenomenon also was not observed. Reinitzer found two melting points of cholesterolbenzoate.

Many chemists, Reinitzer’s contemporaries, studied cholesterol derivatives encountered serious difficulties when determining their melting points. The researchers failed to determine the melting point of cholesterol derivatives and established the presence of an amorphous, tar-like phase. Reinitzer observed just this amorphous tar-like phase appearing in cholesterolbenzoate melting. But unlike his colleagues, Reinitzer studied it more thoroughly. He managed to find that at a temperature of 145.5 °C the white fine cholesterolbenzoate powder transformed into a turbid, hardly transparent liquid. Initially he attributed turbidity to the presence of impurities but additional studies shown no impurities. As cholesterolbenzoate was heated further, the turbidity disappeared suddenly. This occurred at 178.5 °C.

Reinitzer was first who found that a new turbid phase appeared between the solid phase and the usual melt in cholesterolbenzoate melting. The color of that phase depended on the temperature and vanished in transition to the usual melt but sometimes it remained in the solid phase if the latter was rapidly overcooled.

Reinitzer was not only a chemist, he was also proficient in technical microscopy. For this reason he used an opportunity to examine the colored phase with common and polarizing microscopes. He observed a very interesting phenomenon. The turbid phase appeared from the transparent melt as drops with jagged edges seen at a great magnification. This gave Reinitzer cause to call the drops star-like aggregates. In further cooling of the substance the star-like aggregates expanded and a multiphase system formed. Brooklets of a liquid flew over a brightly colored background and played. Reinitzer called them "oily grooves".

The oily groves became transparent if the polarizers were crossed, i.e. they shown marked double refraction. The colored background rotated the light polarization plane. In other words, it possessed optical activity. The rotation angle of the polarization plane was very high and depended on the light wavelength.

All these facts arrived Reinitzer to a conclusion that in case of cholesterolacetate and cholesterolbenzoate he dealt with a rare variety of physical isomerism when several substances of the same chemical composition had different structures and physical properties.

Reinitzer attempted to separate these two isomers but he failed. Then he decided to send the substances to Professor Otto Leman being a famous researcher in physical methods of crystal study. He was a skilful experimenter got excellent training in experimental physics and crystallography at the Strasbourg University and examined the growth and solution of crystals with a polarizing microscope.

The samples were sent on March 14, 1888. At that time Leman studied unusual mechanical properties of silver iodide. At temperatures above 146 °C this cubic crystal did not cleave and crumble to small pieces in compression. It changed its shape easily like melted sealing-wax or rosin. It could be rolled into a thin sheet by the slight touch of a common needle.

Such behavior of silver iodide crystals was beyond concepts of those days. The ductility of metals was explained by their polycrystalline structure but in that case a monocrystal was considered whose lattice could not deform so greatly, as it was believed.

Attempting to answer the questions arisen Leman received the samples from Reinitzer. The study of properties of the substances sharpened still more the issues of a high ductility of the compounds showing the crystal-like behavior. The thorough investigation of cholesterolbenzoate properties with exchange of opinions concerning different aspects of the problem with Reinitzer and other specialists and the discovery of some nitrooxyphenol esters by Hatterman with properties similar to those of cholesterolbenzoate gave rise to new ideas on the behavior of crystals. Their essence was expressed in the Leman's letter to Reinitzer: "Physicists always show great interest to the fact that there are crystals whose ductility allows one to call them liquid crystals". However, Leman avoided using this term in his papers. He gingerly called cholesterol compounds fluid crystals and nitrooxide compounds—drop-liquid

crystals. The latter definition resulted from the fact that these compounds had a less viscosity and were shaped as spherical drops in mixtures with fluids.

Analyzing the situation with silver iodide, cholesterolbenzoate, and nitrooxide compounds Leman supposed that if such ductile crystals as silver iodide existed which hardly retained their weight why crystals totally incapable of retaining their shape should not exist? They will run into a fluid while their physical properties should be anisotropic like those of other crystals and they should grow from oversaturated solutions. Cholesterolbenzoate was undoubtedly one of such liquid crystals.

That reasoning of Leman resulted in the discovery of liquid crystals. Solid crystals having a right spatial arrangement and totally disordered isotropic liquids were bridged by viscous-fluid and absolutely liquid drop-fluid crystals. Leman believed that the solid crystals retained well their weight that caused their polyhedral shapes. Liquid crystals of silver iodide, cholesterolacetate, and cholesterolbenzoate hardly retained their weight and for this reason sometimes they had a polyhedral shape while smoothed and deformed. Finally, drop-fluid crystals of nitrooxide compounds were absolutely incapable of retaining their weight and their equilibrium shape was a spherical drop.

This Leman's understanding formed finally in 1890.

Leman really considered both liquid and drop-fluid crystals as crystals while realizing clearly that these substances differed from solid crystals in their structure. Anisotropy and homogeneity are the main attributes of real crystals. Anisotropy is the dependence of crystal properties on the direction. It results in optical anisotropy appearing as double refraction. Homogeneity implies equal properties along parallel directions. It causes a geometrically right, periodical arrangement of particles making up the crystal, i.e. the existence of a lattice.

Liquid crystals are anisotropic but their fluidity is incompatible with the existence of the lattice. For this reason Leman supposed that those phenomena (anisotropy and fluidity) could be explained satisfactory if the right spatial arrangement of molecules was considered as a minor factor, as a mode of stacking implemented only in slow crystallization. This stacking mode can be disordered by an enough high deforming force but this does not affect the nature of crystal. According to Leman, the essence of crystal is the anisotropy resulted from the anisotropy of molecules themselves. Leman wrote: "The proper anisotropy of molecules which produces indirectly the right structure of a body is an essential and main attribute of crystal rather than the right arrangement of the molecules in the right system of points".

Another equally important feature of crystal is its ability to grow in oversaturated solutions. As Leman noted, both non-deformed and deformed (without a lattice) crystals grow with the same rates while amorphous bodies do not grow.

Afterwards this view of Leman arrived him to some erroneous conclusions which luckily did not influence scientific progress but hampered the recognition of liquid crystals.

Reinitzer's and Leman's results introduced new, revolutionary ideas in science which encountered a suspicious reaction of most physicists and chemists rather than

made a sensation in the scientific community. The discovery of liquid crystals followed by a long period of their study and recognition.

This is a brief history of the discovery of liquid crystals. The development of liquid crystal research is described in more detail in book [1] specially devoted to this problem.

It took more than two decades to recognize the existence of the liquid-crystal state of the matter. This is not surprising since, according to the understanding of many physicists and chemists of the beginning of the 20th century, the phrase "liquid crystals" was nonsensical. This is Leman's blame. It was he who proposed the expression "liquid crystal" unfortunate at that time.

Afterwards French physicist Fridel corrected Leman's mistake and proposed the term "mesomorphous state" as a synonym of the term "liquid-crystal state" formed from the Greek word "mezos" meaning intermediate. Indeed, the liquid-crystal state is the fourth phase state of the matter and takes an intermediate position between the crystalline and liquid states.

2.1.2 Classes of Liquid Crystals

At present *thermotropic* and *liotropic* liquid crystals are distinguished. The class of *thermotropic* liquid crystals is formed by the substances which transit to the liquid-crystal state with elevating temperature. In this case the following *phase transitions* are considered: solid–liquid crystal–*isotropic liquid*. They are phase transitions of the first kind and can be successfully registered by the differential thermometry method (Fig. 2.1). When the isotropic liquid is cooled down the phase transitions are repeated in the reverse sequence.

The temperature of the transition solid–liquid crystal is called the *melting temperature* T_m. The temperature of the transition liquid crystal–isotropic liquid is called the *elucidation temperature* T_e. The term results from the fact that many crystals in the mesophase are turbid liquids scattering light considerably. The scattering disappears and the melt becomes transparent in transition to the isotropic liquid.

Liotropic liquid crystals are the substances which form the liquid-crystal phase when dissolving them in water or other solvents. As a rule, the phase appears within a strictly certain range of the concentration of the substance being dissolved in the solvent. Potassium *n*-alconate soap is a typical example of liotropic liquid crystals. Nowadays liotropic liquid crystals are very attractive for researchers. This is due to the fact that liotropic liquid-crystal phases are typical for more materials than thermotropic ones and to a great role of liotropic liquid crystals in biology. Note that myosin being an albumen contained in the contracting substance of the muscular tissue, desoxyribose nucleic acid playing a leading role in the transfer of heritable data, many polypeptides, ferments and other substances of biological origin are capable of forming the liotropic liquid-crystal phase [2, 3]. It is also found that liquid crystals are of great importance in metabolic processes in living organisms.

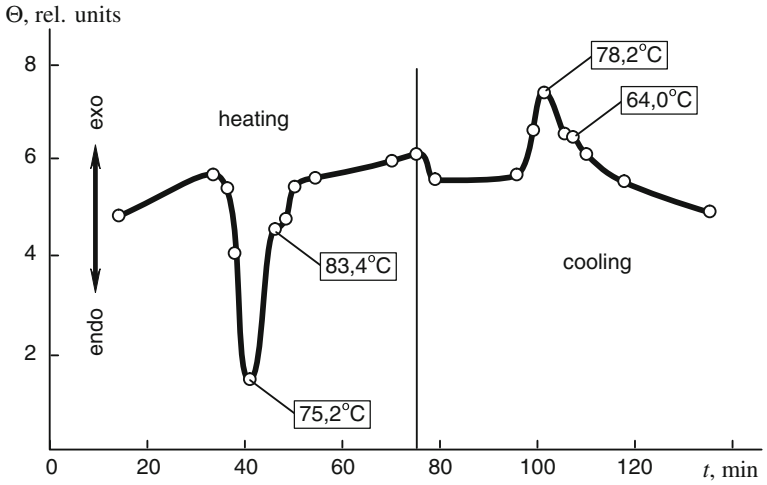


Fig. 2.1 Phase transition temperatures of cholesterolpelargonate

Like solid crystals, liquid crystals show the *anisotropy of properties* which is always closely related to the *anisotropy of the matter structure* [4]. Apparently, proceeding from this fact German chemist Vorlender supposed that liquid crystals were formed mainly from the organic compounds whose molecules had an elongated cigar-like shape, i.e. shown the anisotropy of the structure. His predictions had come true. It should be noted that earlier Leman assumed the structure of molecules of liquid-crystal substances to be anisotropic.

It is found that the substances with flat *disc-shaped molecules* also form the liquid-crystal phase. Such substances are called *discotic liquid crystals* or *discotics*.

Elongated cigar-shaped molecules of the liquid crystal are always in thermal motion. The direction of their major axes continuously changes relatively to a certain direction. To mark the direction of predominant orientation the unit vector n is used which is called "*director*". The director characterizes phenomenologically the long-range order of molecules.

Because of the thermodynamic least free energy principle the elongated or disc-shaped structure of liquid crystal molecules a certain arrangement of the molecules appears in the mesophase. The pattern of the arrangement is governed by both the structure of the liquid crystal molecules and thermodynamic conditions.

2.2 Classification and Structure of Liquid Crystals

The classification of liquid crystals was proposed by French physicist Fridel in the beginning of the 19th century and is based on molecular arrangement. Like Reinitzer and Leman, he examined microscopically thin layers of liquid crystals formed on a table glass in melting. He found that irrespective of the chemical structure of the

compounds liquid crystals formed a limited number of kinds of optical pictures. Fridel related the kind of the optical picture observed with the arrangement of molecules in the samples and proposed the classification of liquid crystals on this basis.

He divided liquid crystals into *smectic* (from the Greek word "smegma" meaning soap) and *nematic* (from the Greek word "nema" meaning thread) crystals. Nematic liquid crystals are divided into properly nematic and *cholesteric* crystals. Two following facts resulted in such division. The first fact was that no mesogens were found passing consecutively the nematic and cholesteric mesophases. The second fact was that it was possible to transform a properly nematic liquid crystal into a cholesteric one and vice versa by external effects such as mechanical deformation, electric and magnetic fields.

In smectic liquid crystals molecules are arranged so that their major axes are parallel and their centers of mass lie in one plane. Major axes of the smectic molecules or the director n form an angle β with the normal to this plane (Fig. 2.2).

The location of the centers of mass of the molecules results in the formation of *smectic layers*. It is apparently that the thickness of the layers depends on the molecule length and the angle between the director n and the location plane of the centers of mass.

By the pattern of arrangement in the layers smectic liquid crystals are divided into two groups: with *structured* and *non-structured layers*. In smectic crystals with structured layers the centers of mass of the molecules form a two-dimensional lattice in the layer. The director can be oriented relatively to the layer both normally (Fig. 2.2a, b) and at an angle (Fig. 2.2c, d). Smectic crystals with structured layers are ordered better than crystals with non-structured layers. In smectic crystals of both groups the mutual slip of layers can occur and in most cases the rotation of molecules about their major axes is possible.

In non-structured layers molecules of the smectic crystal are distributed chaotically (Fig. 2.2a, c). This group consists of several subgroups of smectic crystals which differ in the angle of director orientation relatively to the normal to the layer.

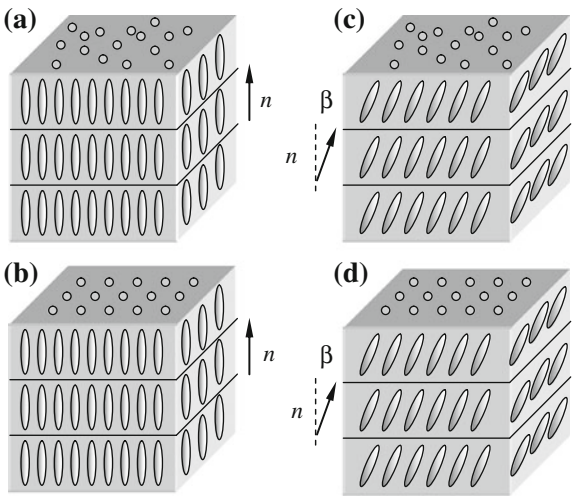


Fig. 2.2 Structure diagram of smectic liquid crystals with non-structured (**a**) and structured (**b**) layers

Smectic crystals for which $\beta = 0$ (Fig. 2.2a) belong to the first subgroup. Therefore, in these crystals major axes of the molecules are perpendicular to smectic layers. Such crystals are commonly called *smectics A*. They are most widespread among smectic crystals.

The second subgroup consists of non-structured smectic crystals for which $\beta \neq 0$. They are called *smectics C* (Fig. 2.2c). Among them are smectics *C* with a small ($\beta < 30°$) and great ($\beta \approx 45°$) angle between the director and the normal to the layer. With elevating temperature smectics *C* with the small slope angle transform into smectics *A*. However, the smectics *C* are known for which the angle β depends on the temperature within a certain temperature range and can decrease down to zero while other smectics with the small slope angle show no temperature dependence of the slope angle β.

Smectics *C* with the great angle β do not transform into smectics *A* and their angle β is independent of the temperature.

If molecules of the smectic substance are chiral, i.e. twisted about their major axes, the twisted *smectic mesophase* C^* is formed (Fig. 2.3).

In such smectics the director rotates along the taper generatrix when passing from one layer to another.

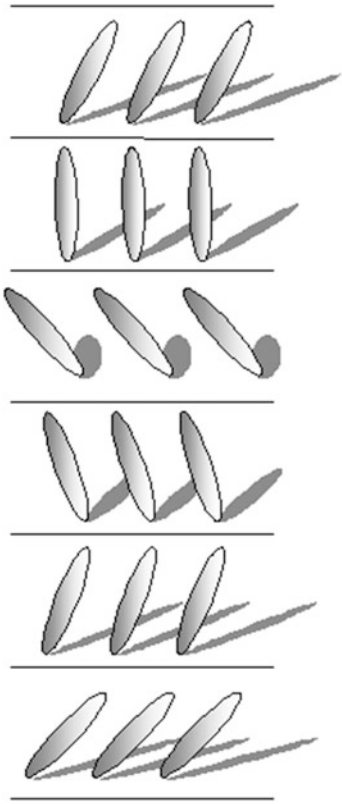


Fig. 2.3 Smectic C^* mesophase

In *smectics B* centers of mass of molecules in layers are located at points of the hexagonal lattice and the director is perpendicular to layers (Fig. 2.4).

A similar location of centers of mass is typical for *smectics H* but the director *n* is inclined to layers (Fig. 2.5).

If the smectic *H* forms a compound with chiral molecules the quasi-hexagonal arrangement remains but when passing from layer to layer the director rotates along the taper generatrix as in case of the smectic C^*. This smectic is called *smectic* H^*.

The quasi-hexagonal arrangement of centers of mass of molecules in layers is also typical for *smectics E* but unlike other smectics with structured layers the rotation of the molecules about their major axes is hampered. As the director is perpendicular to the layers, rhombic arrangement occurs (Fig. 2.6).

In *smectics G* the arrangement of centers of mass of molecules and the hampering of rotation about major axes are the same as in smectics *E* but the director is inclined to the layer plane (Fig. 2.7).

Smectics D have the cubic body-centered structure (Fig. 2.8).

It consists of structure units containing many molecules; they are called micelles (from the Greek word *mica* meaning grain). In nematic liquid crystals molecules are parallel to each other and their centers of mass are arranged chaotically unlike smectic crystals (Fig. 2.9).

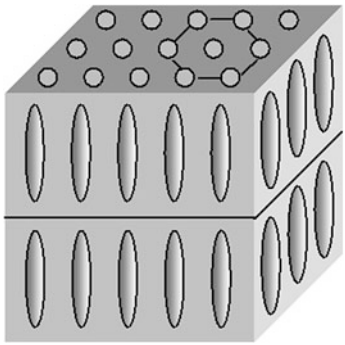


Fig. 2.4 Smectic *B* mesophase

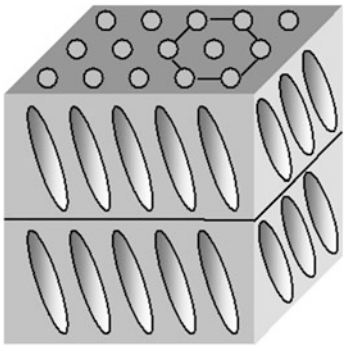


Fig. 2.5 Smectic *H* mesophase

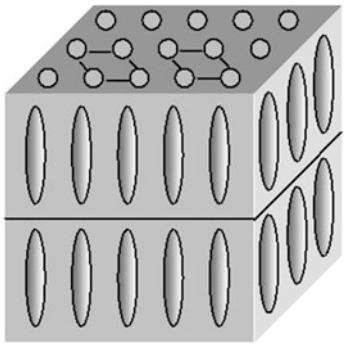


Fig. 2.6 Smectic *E* mesophase

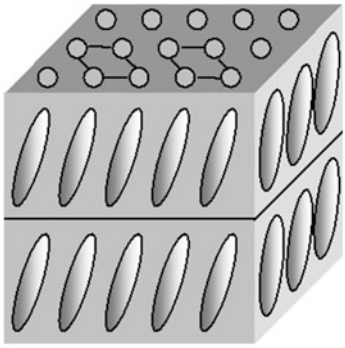


Fig. 2.7 Smectic *G* mesophase

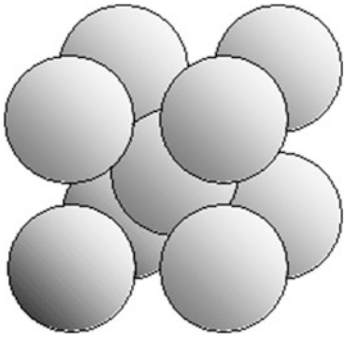


Fig. 2.8 Smectic *D* mesophase

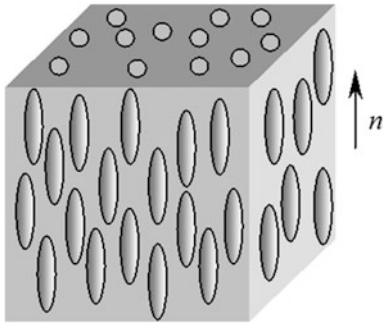


Fig. 2.9 Structure diagram of nematic liquid crystal

In this case layers typical for smectics are not formed and long-range order exists only in respect to the orientation of major axes of the nematic crystal.

Cholesteric liquid crystals are formed by cholesterol derivatives (esters), yet they may contain substances of other classes.

Cholesteric liquid crystals typically consist of molecular layers; each of them demonstrates the molecular arrangement specific for nematics when centers of mass lie in the molecular plane chaotically. The director also lies in the layer plane [5]. The thickness of the molecular layer called also a quasi-nematic layer [6] is 0.5–0.6 nm. In transition from one molecular layer to another the director rotates by a small angle relatively to the director of the underlying layer. This angular shift accumulates in the sequence of the layers and, generally, ends of the molecules or the director end move along a spiral (Fig. 2.10). Apparently, the lead P of the spiral is

$$P = \frac{2\pi}{\alpha} d \tag{2.1}$$

where α is the shift (twist) angle of the director in transition from layer to layer; d is the thickness of the molecular layer of the cholesteric crystal.

According to the results reported in [7], the angular shift of the director in transition from layer to layer in cholesteric crystals is 15 angular minutes on the average. Other authors give the value of about 0.5° [8]. Further thorough studies have shown that the twist angle α depends on both the temperature and the molecular structure of the cholesteric liquid crystal as well as some other factors [9–11].

Taking $\alpha = 0.5°$ and $d = 0.6$ nm we obtain that the lead of the cholesteric spiral is 400 nm.

It is apparently that in the molecular layers corresponding to the director twist angles of 0°, π, and 2π the major axes of liquid crystal molecules have the same orientation (Fig. 2.10) that makes the layers indistinguishable. Then the distance $S = P/2$ is the period of the specific lattice of the cholesteric liquid crystal. It is easy to note that in cholesteric crystals the molecular orientation is repeated at a distance equal to the period S. However, the molecules can move freely and exchange their positions in each molecular layer but their spiral arrangement with the period S remains unchanged.

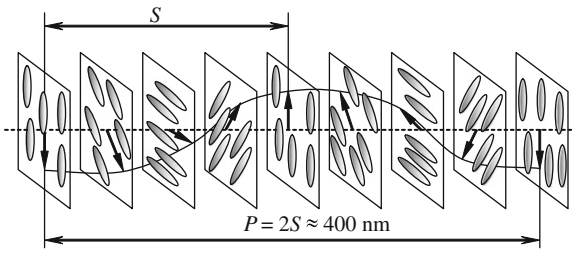


Fig. 2.10 Structure diagram of cholesteric liquid crystal

Macrolayers consisting of many molecular or *quasi-nematic* layers in which the spiral arrangement is retained along the certain direction at a distance of one or more lattice periods are called monocrystal layers of cholesteric liquid crystals (Fig. 2.11). Such layers can be formed between bearing surfaces.

We presented in general outline the classification of liquid crystals. Apparently, it is far from the total scientific classification whose main attribute is predictability. The presented classification does not possess predictability and is mainly of the descriptive character. Therefore, often it may occur that newly synthesized liquid crystals show mesophases which can not be classified according to this scheme. Such mesophases are called exotic. Future studies will allow researchers to refer many of them to the already known kinds and groups. Yet, the possibility of appearing new arrangement modes among exotic mesophases can not be ruled out.

This substance forms the *discotic mesophase* within the temperature range 81.2–87 °C. One of these new arrangement modes has been found recently. It appears in the so-called disc-shape crystals or *discotics* (Fig. 2.12). Their molecules have the symmetric branched formula which can be approximated by a flat disc. For example, benzol-hexa-*n*-heptanoate contains such molecules.

Discotics demonstrate the layered molecular arrangement like smectic crystals [3, 12]. Their molecules lie in the layer planes forming close hexagonal packing shown in Fig. 2.13.

The classification presented above shows that though some liquid crystals have the solely mesophase typical only for the given crystal in many liquid crystals

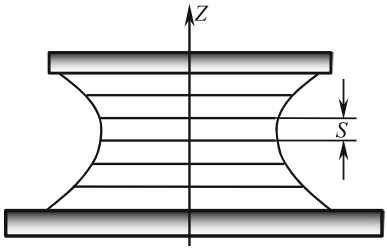


Fig. 2.11 Monocrystal layers of cholesteric liquid crystal between bearing surfaces

Fig. 2.12 The molecule of disc-shape crystals

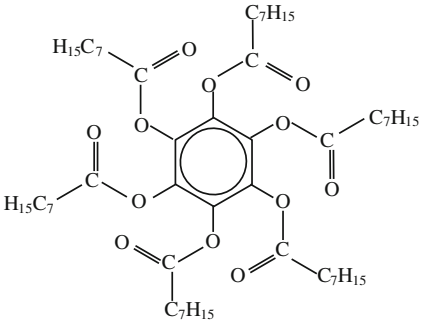


Fig. 2.13 Discotic mesophase

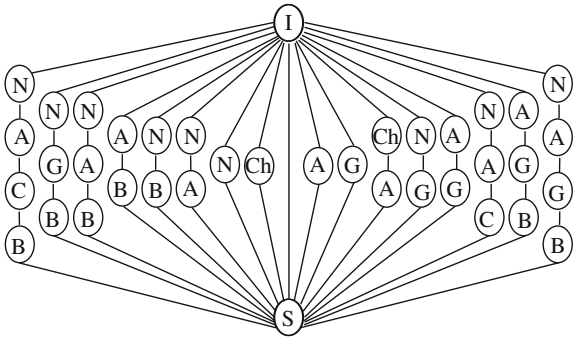


Fig. 2.14 Some cases of polymesomorphism of liquid crystals: *I* isotropic fluid, *S* solid crystal, *N* nematic, *A*, *B*, *C*, and *G* corresponding smectics, *Ch* cholesteric

different phase transformations occur with varying temperature and several mesophases change consecutively. As experiments have shown, there are certain regularities of changing mesophases. Figure 2.14 illustrates some cases of the *polymesomorphism* of liquid crystals known presently.

It is seen in the figure that in transition from the solid crystal to isotropic liquid through different mesophases a sequence of the mesophases occurs with decreasing ordering degree: the smectic, cholesteric, and nematic mesophases.

The same regularity is typical for the smectic mesophase. If the solid crystal melts initially the smectic crystals appear with structured layers and only at higher temperatures the smectics with non-structured layers are formed.

The smectic *C* always exists at lower temperatures than the smectic *A*. As for polymesomorphism with the mesophase *G*, Fig. 2.14 shows that it behaves like the mesophase *C* and often precedes the mesophase *B* with decreasing temperature.

2.3 Properties of Liquid Crystals

2.3.1 Optical and Electrical Properties of Liquid Crystals

An original molecular architecture of monocrystalline layers of cholesteric crystals causes some specific optical properties. First of all, this is *optical activity*, i.e. the ability of the crystals to rotate the polarization plane of linearly polarized light passing the crystal layer. Alpha-quartz crystals considered quite optically active rotate the polarization plane of light by approximately 20° per one millimeter of the crystal thickness while the cholesteric monocrystalline layer of the same thickness rotates the polarization plane by an angle of about 18,000° [7]. Such gross optical activity is typical only for cholesteric liquid crystals. Nematic and smectic crystals do not possess optical activity.

However, one can give the structure of the cholesteric to the nematic by twisting it between the bearing surfaces. Additives of cholesteric liquid crystals to nematics result in the formation of the cholesteric structure.

Circular dichroism is another typically crystal property of cholesteric liquid crystals. It causes the division of natural white light into two components by the cholesteric crystal. The electric vector of one of the components rotates clockwise and that of the other component rotates counter-clockwise. Depending of the structure of liquid crystal molecules one of the components passes the crystal while another component reflects from it causing the certain *color of the cholesteric substance*. This phenomenon is called the selective reflection of light and results from changes in the spiral lead of cholesterics with varying temperature. It should be noted that the selective reflection of light governs the action of liquid-crystal thermoindicating films applied in medicine, engineering and life, for example, in liquid-crystal room thermometers [13, 14]. The systematic description of optical properties of cholesteric liquid crystals is presented in monograph [15].

Like cholesteric liquid crystals, nematic and smectic crystals possess double refraction. However, it occurs at a certain orientation of the director relatively to the light propagation direction.

Double refraction proves the anisotropy of the refractive index that is caused by the anisotropy of the permittivity of liquid crystals. There are refractive indices of the light polarized along ($n_{\|}$) and perpendicularly to ($n_{\perp}$) the director. Similarly, there are permittivities for the orientation of the director along ($\varepsilon_{\|}$) and perpendicularly to ($\varepsilon_{\perp}$) the electric field direction. The difference $\Delta\varepsilon = \varepsilon_{\|} - \varepsilon_{\perp}$ is called the *dielectric anisotropy of the liquid crystal*. It can be both positive and negative. The dielectric anisotropy results from the anisotropy of liquid crystal molecules, their structure, and the nature of atoms contained in the molecules.

Liquid crystals also possess the anisotropy of magnetic properties. The difference of the diamagnetic susceptibilities measured along major axes ($\chi_{\|}$) of liquid crystal molecules and perpendicularly to the axes ($\chi_{\perp}$) is called the *anisotropy of the diamagnetic susceptibility* ($\Delta\chi$). It can be both positive and negative. For nematic

and smectic crystals $\Delta\chi > 0$ and for pure cholesterol ethers $\Delta\chi < 0$. However, for some cholesterics $\Delta\chi > 0$ [16].

The anisotropy of electrical and magnetic properties of liquid crystals provides the possibility of the orientation of their molecules under the effect of electric and magnetic fields. The direction of major axes of liquid crystal molecules in electric and magnetic fields is governed by the sign of the dielectric and diamagnetic anisotropies, respectively. If the anisotropy is positive the major axes of molecules are parallel to the field direction and if it is negative the orientation is perpendicular.

Frideriks and his colleagues observed first the effect of changes in the nematic orientation in electric field. For this reason this effect is generally called the Frideriks transition. These experiments have shown that there is a limiting thickness of the layer between the bearing surfaces, for example between flat and convex glasses, at which the re-orientation of molecules occurs under a given magnetic field strength [17]. The threshold character of the Frideriks transition is caused by a quite strong adhesion between liquid crystal molecules and the bearing surfaces. It occurs only when the magnetic field strength is enough high to overcome the adhesion of the nematic crystal to the bearing surfaces.

The orientation effect of electric field is complicated by the electric conductivity of liquid crystals. The electric conductivity is anisotropic and can cause substance motion and the redistribution of electric fields. All this induces a number of new effects such as magnetoelectricity, electrohydrodynamic instability etc. whose nature is described in papers [8, 14, 18–21].

2.3.2 *Peculiarities of Formation of Liquid-Crystal Layers on Solid Surfaces*

In liquid crystals the director can rotate not only in electric and magnetic fields but also under the effect of the bearing surfaces. They are, for example, the walls of a flat capillary formed by glasses.

The bearing surfaces can orient the director parallel, normally, or at an angle to them. In the first case when major axes of molecules are parallel to the surface the orientation is called *planar* or plane. Their optical properties are similar to those of a plate of the optically positive monocrystal cut parallel to the crystal optical axis. The optical axis is the direction in the monocrystal along which light propagates without double refraction. In the second case when the major axes of molecules are perpendicular to the bearing surfaces the orientation is *homeotropic*. In its optical properties such structure is similar to a plate of the optically positive monocrystal cut perpendicularly to the optical axis.

The orientation mode on the bearing surface is governed by its nature, preparation technique, and the liquid crystal nature. Under given conditions the orientation is implemented at which the free energy of liquid crystal molecules on the bearing surface is minimal. It is apparently that the orientation effects are possible

only when liquid crystal molecules are attracted to the solid surface by certain intermolecular forces. These forces, hence the energy of interaction, are governed by a number of factors. The surface preparation technique is believed the main factor [16].

The energy of interaction between the liquid crystal and bearing surface can be: (a) comparable with the energy of intermolecular interaction in the liquid crystal; (b) much less than the intermolecular interaction energy in the crystal.

In the first case the *strong adhesion* of liquid crystal molecules to the bearing surface occurs; in the second case the adhesion is *weak*. It is difficult to predict which of the cases appears in the experiment. In practice empirical rules are used. It is found that the strong adhesion is easier to implement than the weak one.

A number of methods of providing a certain orientation of liquid crystal molecules on the bearing surface have been developed. They are divided into *mechanical* and *chemical* methods. The mechanical methods involve the producing of the *orienting microrelief* on the bearing surface. This is achieved by surface machining with polishing pastes in the certain direction. The appeared microgrooves orient molecules of nematic and cholesteric crystals parallel to the surface (Fig. 2.15).

Liquid crystals are also oriented on the surfaces rubbed with soft materials like wool, paper, and leather.

The required microrelief can be also produced by *depositing* thin, about 20 nm thick, metal or oxide films onto the bearing surfaces at a certain angle. If the flow of the substance under deposition is perpendicular to the surface liquid crystal molecules are oriented mainly parallel to it. If the substance is deposited at blunt angles (more than 80°) the molecule orientation is "oblique". In this case the director **n** is inclined to the bearing surface at an angle of 20°.

As a rule, the mechanical methods provide a strong adhesion of liquid crystal molecules to the bearing surface. The orientation of the molecules along microrelief grooves is more profitable from the energy viewpoint that the transverse orientation.

According the results reported in [16, 22], for the sinusoidal profile with the wavelength of 20 nm and the amplitude of 1 nm the energy gain at the parallel orientation in comparison with the perpendicular one is $8 \cdot 10^{-5}$ J/m^2. This provides so strong adhesion that to rotate liquid crystal molecules normally to microrelief grooves it is necessary to apply electric field with the strength of $6 \cdot 10^8$ V/m or magnetic field with the strength of $6 \cdot 10^8$ A/m.

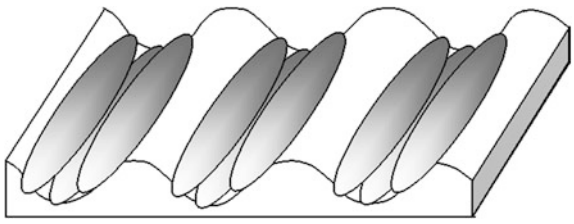


Fig. 2.15 Orientation of liquid crystal molecules along microrelief grooves

As a rule, major axes of smectic liquid crystals are oriented normally to the bearing surface. In its close vicinity a smectic film copies the surface microrelief. As the distance from the bearing surface increases, the distortion of smectic films decreases and, finally, disappears at all. The thickness of the distorted smectic film on finely polished surfaces reaches 1 mm and more [14].

The *chemical methods of orientation* are based on the use of chemical compounds which are deposited onto the bearing surface or introduced directly into the liquid crystal and produce a certain surface structure because of physisorption and chemisorption on the bearing surfaces. These substances are called orientants. As a rule, their molecules contain active end groups and a hydrocarbon radical. The end groups interact with the surface and the hydrocarbon radicals are oriented either normally or parallel to the substrate depending on their length. Liquid crystal molecules are placed between the hydrocarbon radicals and oriented parallel to them. Thus, the corresponding orientation of liquid crystal molecules appears.

It should be noted that the great number of orientants providing a strong adhesion in most cases includes surfactants. The surfactants initiate the homeotropic orientation of nematic and cholesteric liquid crystals [22].

The analysis of the described orientation methods has shown that, as a rule, their application varies the energy state of the surface. Interactions on the solid–liquid interface are described phenomenologically by the surface tension. The use of this approach for the liquid crystal–solid system yielded the empirical rule according to which the homeotropic orientation occurs in the cases when the critical surface tension (σ_c) of the solid is less than the surface tension of the liquid crystal (σ_{lc}). If $\sigma_c > \sigma_{lc}$ the planar orientation of liquid crystal molecules appears [22, 23]. The author of [22] proposed to call it the rule of Fridel-Kreig-Kmetz (FKK rule). Though the rule was verified by a great number of experiments it is not true in some cases.

The studies have shown that when using FKK rule one should consider the sign of the dielectric anisotropy, the nature of forces (polar or dispersion) causing the surface tension, and the pattern of the solid surface microrelief. Paper [22] summarizes the study results which are presented in Table 2.1.

Here we should mention the ability of liquid crystals to orient molecules of dissolved substances parallel to their major axes. This effect is called the "*guest-host*" effect. Dichroic dyes which absorb light depending on the molecule orientation and light polarization make the "*guest-host*" effect visible. If a small amount of the dye (the guest) is added to the nematic crystal (the host) dye molecules are oriented by nematic molecules parallel to their major axes. The nematic arrangement is retained in this case.

If the nematic crystal with the dye additive is placed into a flat capillary the liquid crystal has certain color at the planar orientation. In electric field normal to the capillary walls the nematic changes its color. This proves that the Frideriks transition resulted in the re-orientation of both nematic and dye molecules. The behavior of the "guest" is governed by the behavior of the "host".

Table 2.1 Orientation possible in substrate—liquid crystal interaction substrate

Подложка		Подложка — жидкий кристалл			
Material	Microrelief pattern	$\sigma_к > \sigma_{жк}$		$\sigma_к < \sigma_{жк}$	
		$\Delta\varepsilon > 0$[a]	$\Delta\varepsilon < 0$	$\Delta\varepsilon > 0$	$\Delta\varepsilon < 0$
Activated oxides and diacids (polar surface)	Smooth surface	$\perp$[2]	x[b]	Нет случаев	
	Symmetrical grooves	$\perp$	$\parallel$	Есть $\sigma_к > \sigma_{жк}$	
	Non-symmetrical grooves	 a	 a		
	3D valleys[d]	$\parallel$	$\perp$		
Wet oxides, polymers, and fatty acid derivatives	Smooth surface	x	x	$\perp$[5]	$\perp$[5]
	Symmetrical grooves	$\parallel$	$\parallel$	$\perp$	$\perp$
	Non-symmetrical grooves				
	3D valleys[d]	$\perp$[e]	$\perp$[e]	$\parallel$	$\parallel$

[a]$\Delta\varepsilon$ characterizes the polarity of liquid crystal
[b]this case is poorly confirmed by experiments
[c]$\perp$, $\parallel$,  , homeotropic, planar, and inclined orientations
[d]regular set of 3D alternating peaks and valleys
[e]homeotropic orientation resulted from 3D relief independent of liquid crystal nature and appearing due to zero interactions

2.3.3 *Rheological Properties of Liquid Crystals*

The anisometric structure of molecules of liquid crystals influences their rheological behavior.

The viscosity of liquid crystals, like that of common fluids, decreases with temperature elevation. However, nematics and some cholesterics demonstrate a substantial viscosity variation at temperatures of the transition liquid crystal phase—isotropic fluid. In most cases the viscosity rises. The viscosity increment of nematics is insignificant [24–26].

Cholesteric liquid crystals are characterized by an almost jump-like viscosity rise by several orders of magnitude. According to the results of [16], the viscosity of cholesterolacetate increases by approximately six orders of magnitude at the elucidation point. However, the measurements of the viscosity of cholesterolformiate with varying temperature carried out in a rotation viscometer did not show so great viscosity rise at the elucidation point (Fig. 2.16). This can result from errors of the measuring techniques or the discrete mode of temperature variation in the experiments.

There are liquid crystals whose viscosity decreases under similar conditions [1]. The described viscosity variations occur at low shear rates. The rheological nature of liquid crystals changes depending on the shear rate. According to the results of [24], at shear rates less than 10^{-3} c^{-1} the cholesteric liquid crystal shows the non-Newtonian flow mode. Within the range from 10 to 10^3 c^{-1} the hydrodynamic flow orients liquid crystal molecules and the viscosity does not depend on the shear rate similarly Newtonian fluids. The isotropic fluid of cholesteric liquid crystals behaves in the same manner as the Newtonian fluid.

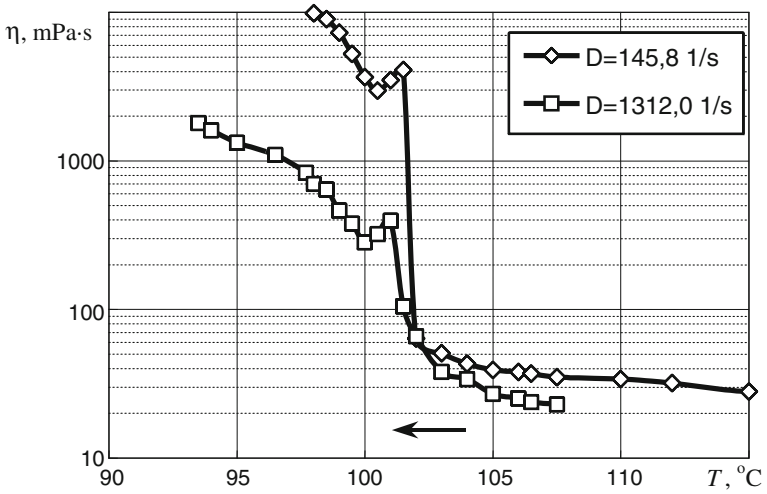


Fig. 2.16 Temperature dependence of viscosity of cholesterolformiate

In the liquid crystal phase the viscosity of liquid crystals depends on the orientation of the director relatively to the flow. Tsvetkov [24] and other researchers have shown that the orientation of nematics by magnetic field perpendicular to the axis of the capillary in which the liquid crystal flows increases the effusion time, i.e. the viscosity. If the magnetic field direction coincides with the capillary axis direction the time of liquid crystal effusion decreases. This proves that liquid crystals possess *the viscosity anisotropy*. The viscosity is less if molecules are oriented along the flow. The viscosity anisotropy of liquid crystals appears only at a slow flow velocity. Otherwise the orientation effect of magnetic or electric field is too weak since the orientation in the liquid crystal flow becomes dominating.

According to the results reported in [16, 27], the direction of major axes of nematic molecules appearing in the flow does not coincide with the flow direction. As I.Ya. Frenkel shown [27], the major axes of anisometric molecules are oriented at an angle of 45° to the flow direction. This angle can be less than 45° at high shear rates and it decreases with increasing shear rate. However, according to the calculation results from [16], the orientation angle does not depend on the shear rate but is governed by the ratio of anisotropic viscosities. Experimental verification confirmed the theoretical predictions.

References

1. A.S. Sonin, *A Way Century Long. History of Discovery and Study of Liquid Crystals* (Nauka, Moscow, 1988), p. 224 (in Russian)
2. G.H. Brown, J.J. Wolken, *Liquid Crystals and Biological Structures* (Academic Press, New York, 1979), p. 316

3. N.V. Usol'tseva, O.B. Okopova, V.V. Bikova, A.I. Smirnova, S.A. Pikin, *Liquid Crystals: Discotic Mesogens* (University Publishers, Ivanovo, 2004), p. 546 (in Russian)
4. S. Kumar (ed.), *Liquid Crystals: Experimental Study of Physical Properties and Phase Transitions* (University Press, Cambridge, 2000), p. 494
5. H. Kitzerow, C Bahr, *Chirality in Liquid Crystals. Series: Partially Ordered Systems* (Springer Verlag, Berlin-Heidelberg-New York, 2001), p. 234
6. *Cholesteric Liquid Crystals* (Collection of Papers) (University, Novosibirsk, 1976), p. 100 (in Russian)
7. E. Fergason, *Liquid crystals, in "Solid State Physics"* (Moscow, Nauka, 1972), pp. 127–135. (in Russian)
8. S.A. Pikin, L.M. Blinov, *Liquid Crystals* (Nauka, Moscow, 1982), p. 208. (in Russian)
9. S. Morishita, K. Nakano, Y. Kimura, Electroviscous effect of nematic liquid crystals. Tribol. Int. **26**(6), 399–403 (1993)
10. P.G. de Gennes, *The Physics of Liquid Crystals* (Clarendon Press, Oxford, 1974), p. 400
11. S.F. Ermakov, V.G. Rodnenkov, E.D. Beloenko, B.I. Kupchinov, *Liquid Crystals in Engineering and Medicine* (Asar, Moscow, CheRo, Minsk, 2002), p. 412. (in Russian)
12. R. Eidenschink, G. Konrath, H. Kretzschmann, M. Rombach, Unusual lift by shearing mesogenic fluids. Proceedings of the 17th international liquid crystal conference, Strasburg, pp. 1−8, 1998
13. N.K. Vistin, J. D.I. Mendeleev All-Union Chem. Soc. **28**(2), 141−148 (1983) (in Russian)
14. V.A. Belyakov, *Liquid Crystals* (Nauka, Moscow, 1986), p. 160. (in Russian)
15. V.A. Belyakov, A.S. Sonin, *Optics of Cholesteric Liquid Crystals* (Nauka, Moscow, 1982), p. 360. (in Russian)
16. A.S. Sonin, *Introduction to Physics of Liquid Crystals* (Nauka, Moscow, 1983), p. 320. (in Russian)
17. V.K. Frederiks, Modern understanding of structure of anisotropic fluid and its substantiation. J. Phys. Chem. **7**(6), 889−895 (1936) (in Russian)
18. S. Pestov, V. Vill, physical properties of liquid crystals. *Series: Landolt-Börnstein: Numerical Data and Functional Relationships in Science and Technology-New Series* (Springer Verlag, Berlin-Heidelberg-New York, 2003), p. 300
19. A.V. Makarevich, L.S. Pinchuk, V.A. Goldade, *Electric Fields and Electroactive Materials in Biotechnology and Medicine* (Infotribo, Gomel, 1998), p. 106. (in Russian)
20. S. Chandrasekhar, *Liquid Crystals* (Cambridge University Press, Cambridge, 1977), p. 266
21. L.M. Blinov, V.G. Chigrinov, Electrooptic effects in liquid crystal materials. *Series: Partially Ordered Systems* (Springer Verlag, Berlin-Heidelberg-New York, 1996), p. 296
22. J. Cognard, *Alignment of Nematic Liquid Crystals and Their Mixtures* (Science Publishers, London-New York-Paris, 1982), p. 104
23. L.T. Creagh, A.R. Kmetz, Mechanism of surface alignment in nematic liquid crystals. Mol. Cryst. Ug. Cryst. **24**, 59–60 (1973)
24. V.N. Tsvetkov, G.M. Mikhailov. J. Exp. Tech. Phys. (7), 1399−1408, (1939) (in Russian)
25. A.S. Babaev, B.G. Temaev, V.I. Stafeev, Electrophysical properties of some liquid-crystal substances. J. Phys. Chem. **52**(6), 1412−1415, (1978) (in Russian)
26. J.F. Johnson, R.S. Porter, *Liquid Crystals and Ordered Fluids*, vol. 2 (London, Plenum Press, New York, 1974), p. 342
27. Ya.I. Frenkel, *Kinetic Theory of Fluids* (Publishing house AN of the USSR, Moscow-Leningrad, 1945), p. 402 (in Russian)

Chapter 3
Modern Concepts of Friction and Lubrication of Solids

Abstract When the engine of a moving automobile, locomotive or any other transport vehicles is switched off, these vehicles stop after some time. A bullet or a projectile sent with a huge speed quickly loses it after several seconds of flight. The common clock stops if its spring is not wound up. There are may such examples. Friction is the common cause why they stop. The moving bodies expend energy to overcome friction. Friction is a complex of phenomena evolving on the surfaces of contact of bodies or their structural elements between themselves or with the environment that cause resistance to their relative motion [1, 2]. Because friction is due interactions between moving bodies or with the environment, it is accompanied by energy dissipation.

3.1 General Information on Tribology. Basic Notions and Terms

Man is unconsciously related to friction throughout his existence. Walking, running, moving objects all relate to friction between feet and soil. Conscious use of friction by man dates back to the time of getting fire by friction. Later man learned to use harrows and sledges to move heavy objects. Further experience showed that if harrows or sledges ran on rollers the gain was much more. In this way man switched over from sliding friction to rolling friction. Discovery of the wheel was a revolutionary benchmark in the technological progress. It is because that the rolling friction is much lighter than the sliding friction.

Ways to reduce friction brought about the development of sliding bearings and then rolling bearings. The excavated elements of the bearings relate to the 1st century BC.

The use of liquids to reduce friction was progressive and quite promising, in particular vegetable oils and animal fats. Egyptians used plastic lubricating materials from mixtures of olive oil and lime to oil the axles of wooden chariots already in the 14th century BC. This lubrication was in fact a prototype of the earlier well-known solid oil.

S. Ermakov et al., *Liquid Crystals in Biotribology*,
Biological and Medical Physics, Biomedical Engineering,
DOI 10.1007/978-3-319-20349-2_3

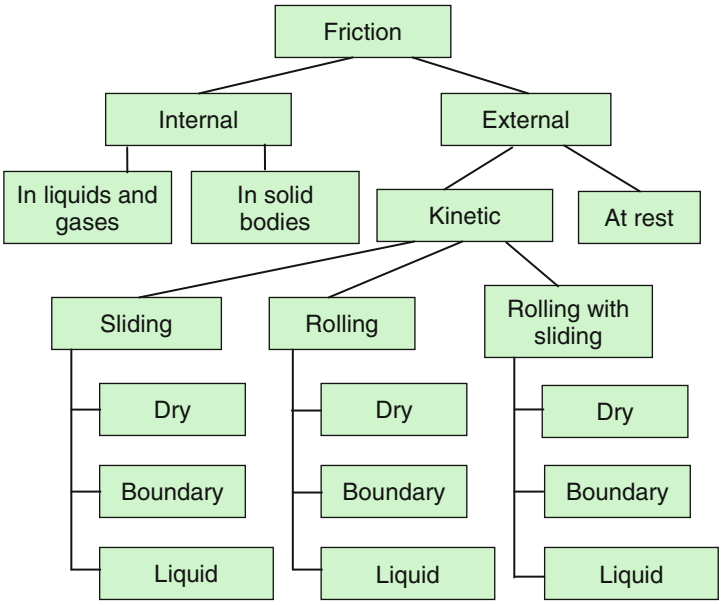


Fig. 3.1 Classification of friction types

As carts, chariots, wooden tooth gears, (1000 BC, Mesopotamia), iron bearings in presses (400 BC, Greece), bronze inserts and bearings (300–200 BC, China, Rome) appeared, greases came into use obtained by evaporation of volatile petroleum products during protracted heating. Mineral oils came into use in the 19th century; synthetic oils and various additives to them, mineral oils and solid lubricating materials came into use in the first half of the last century.

It is believed that the first scientist to ask "What causes friction?" and tried to answer, i.e. began to study friction, was great Leonardo da Vinci (1452–1519). In 1508 he conducted experiments and concluded that "Any rubbing body exerts resistance in friction equal to one fourth of its weight". Thus he formulated first the idea of the coefficient of friction.

In later years the ideas of the nature of friction developed and enlarged. Different types of friction were established: external and internal (friction in liquids, in gases during their flow and in solid bodies during their plastic deformation), dry, boundary and liquid friction, kinetic friction (rubbing of bodies when they move one in relation to the other), and friction at rest. The latter notion is hardly true. When bodies contact and remain immobile one in relation to the other, there is no energy dissipation, hence there is no friction. It is more justified to speak about the resistance to starting when the forces to overcome inertia of bodies were still unknown. Figure 3.1 shows a detailed classification of the types of friction.

At present the integrity of knowledge about friction and phenomena accompanying it are integrated by the science that came to be named *tribology*. This word originates from two Greek words: *tribos*—friction and *logos*—science [3].

Tribology as a scientific discipline appeared in the early 60s and covers a broad range of problems including experimental and theoretical studies of physical (mechanical, electric, magnetic, thermal), chemical, biological and other phenomena connected with friction [4]. Tribology in the recent years has acquired new categories *tribochemistry*, *tribophysics and tribomechanics.*

Tribochemistry studies interactions between contacting surfaces and a chemically active environment. It studies the problems of corrosion in friction, chemical principles of selective transfer and the effect of chemically active substances appears in friction due to destruction of polymers or lubricating materials on the surfaces of parts.

Tribophysics studies the physical aspects of interaction between contacting surfaces during mutual displacement.

Tribomechanics studies the mechanics of interactions between contacting surfaces in friction. It deals with the laws of energy dissipation, impulse as well as the mechanical semblance, relaxing vibrations in friction, reversing friction, equations of hydrodynamics and other aspects in relation to the problems of friction, wear and lubrication.

A handbook glossary of over 1,200 terms of friction, wear and lubrication of machine parts was published in 1979 in the USSR. A number of terms relating to Tribology are standardized in the CIS [6] and the far-abroad countries [7]. For example, GOST 27674-88 contains 107 terms classified according to the types of friction, wear, lubrication, methods of lubrication and lubricating materials. We quote some most frequently used terms.

External friction is the phenomenon of resistance to relative displacement appearing between two bodies in the zones of tangential contact between surfaces accompanied by energy dissipation.

Internal friction is the phenomenon of resistance to relative displacement of particles of one and the same body.

Wear is the process of destruction and separation of the material from the solid body surface and (or) accumulation of its residual deformation in friction manifest in gradual change of the size and (or) shape of the body.

Fatigue wear is the mechanical wear due to fatigue fracture during repeated deformation of microvolumes of the material surface layer.

Wear is the wear result determined in established units. Wear can be expressed in the units of length, volume, mass, etc.

Wear resistance are the properties of a material to resist wear under definite conditions of friction estimated by the value inverse to the speed of wear or wear rate.

Friction at rest is the friction of two bodies during microdisplacements to transition to relative motion.

Friction of motion is the friction of two bodies in relative motion.

Sliding friction is the friction of motion of two solid bodies at which the speeds of the bodies in contact points have different magnitudes and directions or different by the magnitude and direction.

Rolling friction is the friction of motion of two solid bodies at which their speeds in contact points have the same magnitude and direction.

Force of friction is the force of resistance during relative displacement of one body over the surface of the other body under the external force tangential to the common boundary between these bodies.

The maximum force friction at rest is the force of friction at rest that induces motion if exceeded.

Sliding velocity is the difference between speeds of bodies in contact points during sliding.

Friction surface are the surfaces of bodies involved in friction.

Coefficient of friction is the relation of the force of friction between two bodies to the normal force holding these bodies together.

Friction without lubricating material is the friction of two when their friction surfaces have no lubricating material of any type.

Friction with lubricating material is the friction of two bodies when their friction surfaces are lubricated with a material of any type.

Lubricating material is the material delivered to the friction surface to reduce the force of friction and (or) the wear rate.

Lubrication is the action of the lubricating material reducing the force of surface friction and (or) the wear rate.

Lubrication is delivery of the lubricating material to the friction surface.

Liquid lubrication is the lubrication when friction surfaces of parts are fully separated by a liquid lubricating material.

Solid lubrication is the lubrication when friction surfaces of parts in relative motion are separated by a solid lubricating material.

Hydrodynamic (gas dynamic) lubrication is the liquid (gaseous) lubrication when the friction surfaces separate fully under pressure that spontaneously appears in the liquid (gas) in relative motion of surfaces.

Hydrostatic (aerostatic) lubrication is the liquid (gaseous) lubrication when the friction surfaces of parts in relative motion or at rest separate fully due to delivery of a liquid (a gas) into the clearance between friction surface under external pressure.

Elastohydrodynamic lubrication is the lubrication when the characteristics of friction and thickness of the liquid lubricating film between the two surfaces in relative motion are determined by the elastic properties of the materials of two bodies and by the rheological properties of the latter.

Boundary lubrication is the lubrication when friction between the surfaces in relative motion and their wear are determined by the properties of the lubricating material different from the volume viscosity.

Semiliquid (mixed) lubrication is the lubrication when partially hydrodynamic and partially boundary lubrication appears.

Viscosity is the volume properties of the liquid, semiliquid or semisolid substance to exert resistance in friction.

Lubricity is the properties of a lubricating material to reduce wear and the force independently of its viscosity.

3.2 Friction of Lubricated Solids

3.2.1 Molecular-Mechanical Theory of Friction

The modern ideas about the nature of friction of solid bodies are based on the molecular mechanical (adhesion deformation) theory of friction [8–12]. According to this theory, the force of friction F_{fr} both with and without a lubricating material is due to the forces of intermolecular interaction and the forces of resistance to viscoelastic and plastic deformation of near-surface layers in solid bodies in the zone of contact. Both the first and the second components of the forces of friction are directed opposite to the motion of rubbing bodies. The component due to the forces of intermolecular interaction are termed the molecular or adhesive component of the forces of friction F_a, while the component relating to deformation of near-surface layers in rubbing bodies is termed mechanical or cohesive component of the forces of friction F_κ. With these designations in mind the force of friction according to the molecular mechanical theory is a sum of the above forces:

$$F_{fr} = F_a + F_\kappa. \tag{3.1}$$

It is the most general tribological law. Depending on the type and conditions of friction, the structure of the bodies, and bonds in them, individual components can increase or reduce or even disappear altogether. For example, in case of internal friction in a liquid the adhesive component F_a is close to zero, while in case of external friction of perfectly smooth surfaces the cohesive component F_k would have been equal to zero.

Adhesion and cohesion are known to be due to one and the same forces, their localization is different [13]. If the forces act within the body, they produce cohesion, if they act on the surface; they produce adhesion (Fig. 3.2).

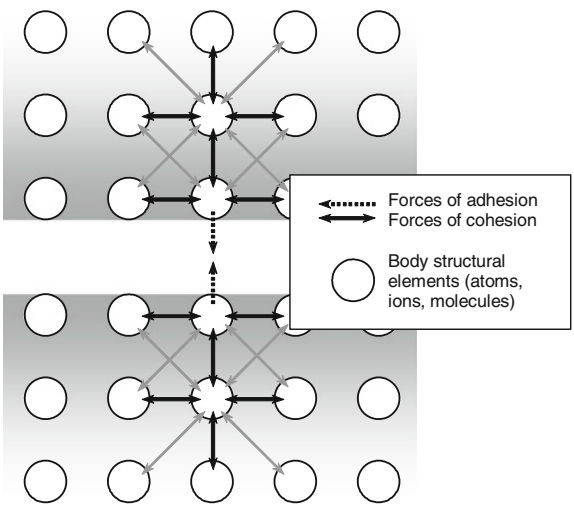


Fig. 3.2 Forces of adhesion and cohesion [3]

Meanwhile, in the real conditions these forces result from formation and rupture of bonds (often termed as friction bonding) in the contact zone during tangential displacement of contacting solid bodies. The magnitude of the forces appearing between rubbing bodies is hugely affected by the physico-mechanical and chemical properties of surface layers and by the properties of the lubricating medium. The notion of lubricity serves to assess the latter.

According to [11], lubricity is the integrity of physical and chemical properties of a lubricating material causing reduction of the adhesive and cohesive interaction between rubbing solid bodies. The higher the lubricity, the weaker are the forces of friction bonds between rubbing surfaces.

Wear is not always due to the parameters of lubricity; it can be due to other factors [13]. That is why investigation of lubricity aims at identifying specifically the physical and chemical properties producing the integral or lubricating effect and at establishing how these properties on the physical state and chemical composition of a lubricating material [11–14].

The thickness and the nature of the latter in the zone of dynamic contact determine the type of lubrication and thus the type of friction. The notion of relative thickness is used to characterize the thickness of the lubricating layer. The lubricating layer relative thickness h_l is determined as a ratio between the lubricating layer thickness h and the sum of mean arithmetic deviations of profiles R_{a1} and R_{a2} of contacting surfaces [3]:

$$h_l = \frac{h}{R_{a1} + R_{a2}}, \tag{3.2}$$

The lubricating layer relative thickness is an essential criterion of classification of types of lubrication.

The following types of lubrication are identified: boundary, liquid (hydrodynamic, elastohydrodynamic, hydrostatic, etc.) and mixed (partially boundary, partially liquid) [3, 15]. Figure 3.3 shows the classification of the types of lubrication based on the criterion h_l.

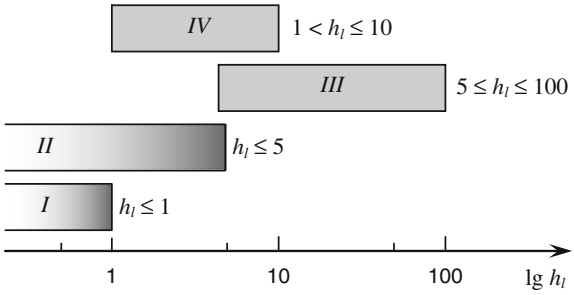


Fig. 3.3 Classification of types of lubrication based on criterion of lubricating layer *relative* thickness h_l: *I* boundary; *II* mixed; *III* liquid (hydrodynamic, hydrostatic, aerostatic); *IV* elastohydrodynamic (liquid) [3]

Because among the above-listed types of lubrication the boundary lubrication challenges the tribologists most, below the relevant processes will be considered.

3.2.2 Boundary Lubrication of Solids

According to international standard ISO 4378/3, the boundary lubrication implies the type of lubrication to which the properties of the lubricating material in the volume cannot be attributed and which is determined by the properties of the boundary layers that appear during interaction between the surface and the lubricating material as a result of physical and chemical adsorption [7].

The boundary lubrication of surfaces of interfaced bodies separates them with a rather thin layer of the lubricating material thickness (the thickness is from one molecule to 0.1 μm) (Fig. 3.3). The protective boundary films in the process of friction prevent seizure of solid bodies, and it is a decisive factor when hydrodynamic lubrication is impossible due to small thickness of the lubricating layer [8–12, 16]. These films can appear continuously in friction due to the effective environment performing the lubrication action.

The surface layers of solid bodies are known to possess the structure and properties different from those of the material lying deeper [5, 8, 11]. Each particle within the body (the atom or molecule making it up) is exposed to the forces of interaction (repulsion or attraction) from the part of surrounding particles (Fig. 3.2). The resultant of these forces is equal to zero in this case.

The force directed away from the surface inwards the solid body acts in the layer with thickness comparable with the radius of molecular interactions due to their asymmetric effect on each particle. It induces excessive potential energy termed the surface energy [12].

Thus, the particles in the surface layer create the field of surface forces. Due to them the molecules of gases, liquids and other substances are adsorbed by the solid body surface when they approach it and reduce the surface energy [17]. The substances with definite properties, structures and adsorbability tend most to be adsorbed [18]. These substances are called surfactants. There are two large classes of surfactants with different absorption patterns and some other properties.

The first class is low-molecular compounds of biophilic nature. These compounds have the hydrophilic "head" (one or several polar groups, for example, –OH, –COOH, $–SO_3H$, $–OSO_3H$, –COOMe, $–N^+(CH_3)_3I$, $–NH_2$) and the hydrophobic "tail". As a rule, it is an aliphatic chain that includes sometimes an aromatic group.

The second class is high-molecular surfactants in which hydrophilic and hydrophobic groups alternate and distribute regularly throughout the polymeric chain.

The molecules of a low-molecular surfactant, for example, are organic acids, their metallic soaps, alcohols, etc., that orient during adsorption perpendicular to the surface. The active "head" of the surfactant molecule disposes on the surface, the hydrophobic "tail" orients normally to it.

The adsorbed surfactant layer can be both monomolecular or polymolecular (Fig. 3.4). Each layer of the polymolecular film consists of the similarly oriented

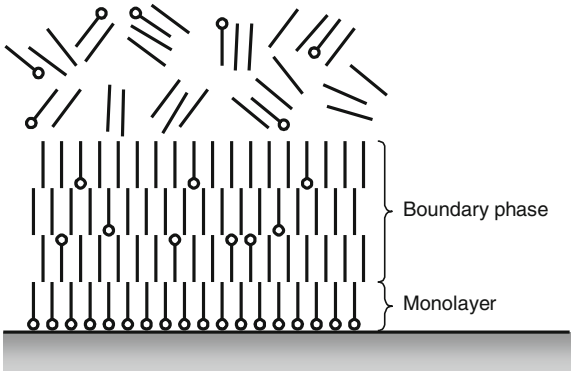


Fig. 3.4 Structure of boundary lubricating layer [12]

molecules. The active parts of surfactant molecules of the first row attach to the solid body surface, the molecules of the second row attach to them and then the molecules of the hydrophobic "tail" facing the hydrophobic "tails" of the first row. The orientation of the molecules of the first group, of the second row of the fourth group, etc., repeats [11, 12]. Even if non-polar molecules occur among surfactant molecules in some layer, they orient parallel to the surfactant molecule. As a result the boundary layer appears on the surface in which surfactant molecules are oriented properly rather than chaotically like in a liquid.

The boundary layers are in a specific aggregate state having the quasicrystalline structure that justifies to assert their specific liquid phase, or the boundary phase [11, 15]. Here physical adsorption and chemosorption can take place. In case of physical adsorption, that is very fast-evolving reversible process, the molecules of the medium on the solid body surface retain their individuality [19]. The physical adsorption is typical when the molecules of the adsorbed substance stay for some time on the surface, then they can desorb in case of suitable conditions, for example temperature rise [11, 20].

The adsorption of the surfactant on the solid body surface can result in the effect of strength drop due to adsorption (the Rebinder effect) [13]. There are external and internal adsorption effects.

The external effect is observed during adsorption of the surfactant on the external surface of the deformable solid body and it is typical for comparatively large surfactant molecules (the molecules of organic acids, alcohols, etc.). As a result of the external adsorptive effect the surface energy of the solid body diminishes leading to a drop of the flow limit of the material in the surface layer. In this case it is termed as the plasticizing effect of the surfactant.

The internal adsorptive effect is due to adsorption of comparatively small molecules of the surfactant (the molecules of wear, etc.) on the internal surfaces of the interface, or nucleation of fracture microcracking. The walls of microcracks tend to cover the entire surface, the surfactant molecules travel to its top. But they are unable to penetrate to the very top because of the comparability between the surfactant molecules and the microcrack width near the top (Fig. 3.5).

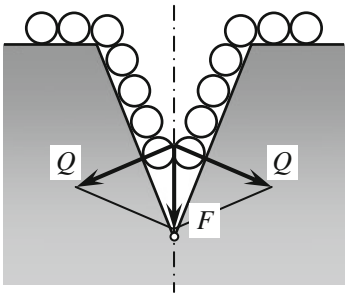


Fig. 3.5 Scheme of wedging effect of adsorption of polar molecules of lubricating material. *F* Pressure of adsorbed layer; *Q* wedging forces

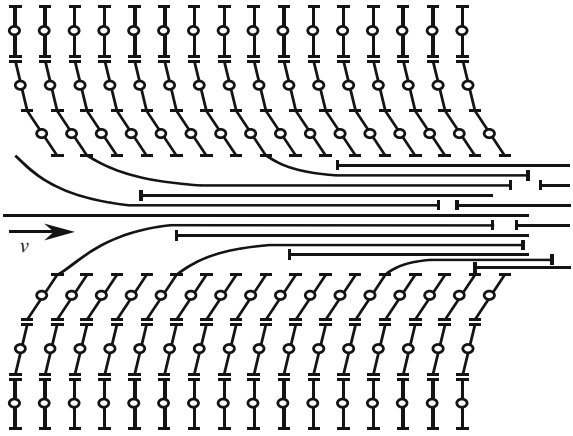


Fig. 3.6 Diagram of lamellar sliding boundary layers during boundary lubrication [11]

The forces of repulsion between the molecules and the surfactant begin to wedge the microcrack walls out assisting further growth of microcracking. Reduction of the surface energy of the microcrack walls favors it during adsorption of surfactant molecules. These processes result in the embrittlement of the surface layer on solid bodies.

Thus, adsorption of the surfactant alters significantly the mechanical properties of surface layers on solid bodies that can play a decisive role in controlling the process of friction and attendant wear [10–12, 21].

Adsorption of polar molecules of organic acids exerts a significant effect on the surface properties. When adsorbed on the solid body surface these molecules attach to it with carboxyl groups and orient perpendicularly [11]. The structure of the first layers of molecules adsorbed on the solid surface resembles the crystalline structure. The higher disposed layers have a lamellar structure (Fig. 3.6). Then the mechanical strength of the boundary layers of molecules grows as they approach the solid body surface [11].

The surface layer changes still more during chemical adsorption (chemosorption). When the molecules are adsorbed on the surface of the solid adsorbent the

adsorbate enters into chemical interactions with the surface layer material and form chemical compounds.

It is used in modern lubricating oils in which additives are applied with the action based on the chemical interaction between their active components and solid bodies. As a result of chemosorption, the friction surfaces are modified chemically and acquire the surface compounds reducing friction and wear [22, 23]. The chemical mechanism of the lubricating effect relates directly in this case to the original lubricity of the lubricant. It acts as a carrier of components in the chemical reaction, while friction unit acts as a reactor in which the processes are controlled both by the composition of the lubricant and the nature of the rubbing surface and friction conditions [11].

The significance of the chemical factors of the lubricating effect is considerably greater at elevated temperature when lubricating materials or their components are exposed to mechanochemical transformations. The resulting products prevent seizure and reduce wear. Specific antiwear and antiscoring additives boost these effects [22–24]. Among them the most popular compounds containing chemically active groups including atoms of chlorine, sulfur, phosphorus, nitrogen. The chemical interaction of such additives, for example, with metallic friction surfaces is affected considerably by elevated temperature and the catalytic effect of the metal. Temperature measurements in the contact spots in friction have revealed that it can reach 700–800 °C in separate contact spots. Meanwhile, the recent data justify to believe that some active additives interact chemically with the rubbing surface already at temperatures of the order of 20–30 °C [23–27].

Other methods of applying antiwear and antiscoring films are determined and extensively studied recently. Specific attention is paid to investigation of the "wearless" friction characterized by abnormally low coefficients of friction and wear. the so-called "third body", or a fine separating layer with low mechanical properties acting as a lubricant appears and regenerates in the process friction over the interphase boundary between contacting bodies [35, 36] under the specific conditions of the effect of the medium when there is the effect of selective transfer [28–31], grafted polymerization [32–34] or irradiation. The positive gradient of mechanical properties appears in the near-surface contact layer the lubricating effect in the boundary lubrication mode is known to have no relation with the viscosity or chemical transformations in the zone of dynamic contact [10–12]. It has a more intricate pattern and is due to appearing of the specific boundary layer because of the effect of active lubricating components on the mechanical properties of surfaces of solid bodies [11, 37]. Hence, on the one hand, an adsorptional layer appears with its specific structure [11, 38–40] and, on the other hand, adsorption plasticizes the rubbing surface significantly contributing to the lubricity of the friction system [41]. The process of boundary friction evolves in the surface layers on interfaced bodies and the boundary layer on the lubricating substance with the comparable thicknesses of these separating layers.

Lubricity in boundary lubrication relates to the state of the surface of the "solid body—lubricant" interface. The interface manifests a sharp leap of shear resistance.

Due to formation of the boundary layer and adsorptional plasticization the leap is somewhat smoothed [8, 11].

Since the surfaces of solid bodies are rough, the conditions of formational of adsorptional layers are different. There are three types of portions of contact surfaces [11]: (a) the portion separated by the polymolecular boundary layer; (b) the portion separated by mono- and bimolecular layers; (c) the portion over which projections come into contact and crumble by shear deformation. As a result, the total resistance to mutual shear of the surfaces at low shear rates when temperature and tribochemical effects van be ignored is a sum of their resistances: (a) the shear in the polymolecular layer; (b) the shear of the surfaces separated by monomolecular layers; (c) the shear of the surface layer on the solid body [11, 14].

Formation of the polymolecular boundary layer and the surfactant it contains is observed by different methods [11, 38–40]. The most comprehensive review of the modern methods of studying liquid boundary layers on solid surfaces is made [11]. These methods are based on determination of thicknesses of boundary layers under different conditions. They have revealed sufficiently reliable characteristics of the lubricating effect of various materials during boundary lubrication.

Application of the methods to determination of the thickness of boundary layers has enabled to establish that formation and thickness of the polymolecular boundary layer depends much on the composition and structure of surfactant molecules. The molecules with short or ramified alkyl radicals yield just fine boundary layers or do not produce them at all [11]. The decisive role in producing thicker boundary layers plays the alkyl radical length [11, 42, 43]. Solution of higher alcohols and amines are capable too to form polymolecular boundary layers [40], but they are weaker than the boundary layers from the fatty acids with the same alkyl radical length [14].

The shear resistance in the polymolecular boundary layer from the solutions of fatty acids and alcohols declines as the alkyl radical gets longer. If the clearance between rubbing surfaces is stabilized by external, the alkyl radical length stops affecting the shear resistance [11].

Moreover, it is a typical feature that the polymolecular boundary layer becomes stronger as it gets thinner [11–13].

Any experimental study of the lubricating effect of monomolecular surfactant layers is impeded due to the inhomogeneity of the real solid body surface and practically impossible shear localization between adsorbed layers of molecules since flow of the substrate destroying these layers cannot be prevented. Such flow can occur if the contact zone is loaded even without shear [11, 14].

The force of the contact interaction in case of the monomolecularгo adsorptional layer depends primarily on the degree of filling of the layer and relates correspondingly to the adsorbability of the surfactant in this case, unlike the polymolecular boundary layer, the length of the alkyl radical of the adsorbed surfactant has little significance. If the adsorptional layer is not filled, short-chain molecules can be more effective. It is explained by their easier orientation in the unfilled layer and their considerable role in reducing the surface energy [11]. Therefore the regularities relating the lubricity to the chemical composition of the adsorbed surfactants are the same for the cases of the poly- and monomolecular boundary layers.

It is indicated above that plasticization of the surface layers of rubbing solid bodies by adsorption matters significantly for the lubricating effect of various substances. The degree of plasticization is determined by the pattern of deformation and the nature of dynamically contacting bodies and lubricating media.

Adsorptional plasticization affects the friction contact in two ways and hence the friction. First, the shear resistance in the surface layer on the solid body drops. Second, resistance to normal stresses declines thus the contact surface smoothes and local stresses on this surface are relieved. On the whole, surface shear resistance reduction with the related deformable volume decrease diminish the forces of friction, while the surface smoothness and smaller brittleness reduce wear [41, 44].

Differentiated approach to the adsorptional effects governing the lubricity of various materials in boundary friction shows that the optimal additives to lubricants ensuring higher lubricity are different for each of the three types of the surface contact sites.

In case of the polymolecular boundary layer, the decisive role belongs to the molecular-kinetic factors determining resistance to thinning of the layer. In case of the monomolecular adsorptional layer, they are the strength of the adsorptional bonding and the kinetics of restoration of the adsorptional layer on the juvenile surface. In case of the adsorptional plasticization the whole surface activity and the kinetics of diffusion of molecules into the defects in the surface layer on the solid body are determined.

The analysis of the said above leads to the conclusion that high lubricity is ensured providing shear localizes fully or in its considerable portion in the polymolecular boundary layer in which the shear resistance is minimal or absent practically. Such high lubricity can be achieved if the lubricating layers in the friction zone acquire the liquid crystalline structure [42, 45, 46].

So far it is established that the seizure of surfaces in friction during boundary lubrication can occur only when the temperature of the friction contact exceeds some critical value T_{cr} for the given combination of the lubricating material and the friction couple material [13, 24]. The effect of the friction couple material manifests itself in the weakening of the latter with other unchanged conditions leading to a sharp drop of the critical temperature. disintegration of the lubricating layer and seizure onset take places most probably due to desorption of the surfactant agents the lubricating material contains [20].

It is shown that the properties of boundary lubrication can change in time. This boundary lubrication feature relates closely to the kinetics of adsorption of polar molecules. The time interval within which the polar molecules orient properly on the rubbing surfaces has been termed the latent period. Therefore, the friction process should stabilize within the latent period and influence considerably the wear resistance of dynamically contacting materials. Sometimes the surfactants the lubricating material contains rival with the molecules and the lubricant base during adsorption on the surface of the solid body. It prolongs the latent period that affects negatively also the wear resistance of friction couples [47].

3.2.3 Solid Lubricants

One effective way to promote lubricity developed in the recent years is introduction into the lubricant the additives of layered (lamellar) solid substances [48]. These substances include molybdenum disulfide, graphite that produce colloid solutions in the lubricant and oil soluble compounds. They are specifically promising because they assist in forming stable lubricating systems [49]. The specificity of the lubricants with the compounds of this type is that they decompose at elevated temperatures and undergo a sequence of transformations forming simpler products than the original.

These transformations occur primarily in boundary layers, as a result of the surface compounds appear in the zone of friction ensuring their effectiveness.

The appeal of the lamellar solid substances is to their some specific properties. For example, graphite is capable of oriented adsorption on the metallic surface [11]. The adsorptional graphite layer represents seemingly a single endlessly long molecule in two measurements. The bonding between these two planar layers of molecules is very weak, so that graphite splits up easily into numerous very fine flat scales. These scales orientate tangentially to the solid body surface and make its microgeometrical profile more like a plane. Secondary adsorption occurs exactly on this surface rejuvenated by graphite, for example, adsorption of fatty acid molecules that in combination yields a positive effect on the boundary friction process.

Many hypotheses and assumptions have been advanced to explain this low friction of lubricating materials with solid lamellar substances. The hypothesis belied most justified is advanced in [50]. From its standpoint the mechanism of the lubricating effect of both lamellar solid substances and high-molecular linear polymers can be explained. Its essence is the following.

To ensure low friction of solid lubricating materials the surface energy should be low over the planes of their cleavage or over the surfaces of bidimensional molecules (lammels), while the interaction between the planes should be weak enough. The additional factor is lack of high enough potential barriers and relative constancy of potential energies of the supposed sliding planes. This requirement presumes that the atoms or molecules adsorbed by substances can under certain conditions easily travel over the surface representing the so-called bidimensional gas (or the bidimensional liquid). This behavior is possible for physically adsorbed molecules and little probable in case of chemosorption. Hence, if physical adsorption of the surface energy declines and molecules locate in interlayer spaces, the spacing between the planes grows so that the friction coefficient should drop. While, if the adsorbed molecules or atoms enter into chemical interactions with contacting contact e surfaces, they inhibit their easy sliding relative one another and increase the coefficient of friction and wear.

This approach permits to explain rather well both high and low friction of lamellar solid substances when they operate under different conditions. The experience shows that these conditions affect considerably both the energy of bonding

between sliding planes and the pattern of adsorptional interactions between molecules and their ability to migrate easily towards rubbing surfaces, i.e. the antifriction properties of lamellar lubricating materials in general [46–50].

3.2.4 Hydrodynamic Lubrication of Solids

It has been noted before that the lubricity during boundary lubrication of solid bodies does not depend on the viscosity of lubricating materials and it is rather due to formation of interphase boundary layers with the specific structure in the contact zone.

In case of liquid lubrication it is different when the lubricating layers are not enough thick and viscosity determines resistance of the lubricating material to motion (the internal friction) and it does depend on the nature of the interfaced surfaces [5–7, 12]. The ribbing surfaces are separated by a film of the liquid lubricating material under pressure. Since the pressure on the lubricating material equalizes the external loading, the force of friction in liquid lubrication is a sum of resistance to sliding of its layers over the thickness of the lubricating interlayer and results therefore from the rheological properties of the latter.

Hydrostatic and hydrodynamic lubrications are discriminated in response to pressure in the lubricating film and the way of its formation.

The hydrostatic lubrication implies formation of liquid films between interfaced surfaces in which static pressure results from continuously externally maintained circulation of the liquid and balancing of external loading so that the rubbing surfaces are separated [5]. Special methods serve to produce the hydrostatic pressure sufficient to separate friction surfaces (Fig. 3.7), while leak of the lubricant through the ends of the hydrostatic bearing is compensated, as a rule, by relevant external delivery.

During hydrodynamic lubrication the pressure in the lubricating film develops and continues automatically. It is ensured by suitable design means in friction systems and selection of the lubricating material in response to the speed of sliding [12]. The mechanism of creating pressure in the bearing layer is most vividly examplified by the flat bearing (Fig. 3.8).

Assume the plate *A*–*A* travels with a speed v at an angle α to the stationary substrate *B*–*B*. The space between the plate and the substrate is filled with a viscous liquid. The liquid layer wets the plate *A*–*A*, while the forces of viscous friction move the adjacent as high layer. The motion is thus transmitted between the layers, excepting the layer wetting the stationary substrate *B*–*B*. As a result the lubrication is sucked into the narrowing wedge clearance and definite pressure is maintained in it that determines the load bearing capacity or the load lifting capacity of the lubricating layer.

So, the wedge clearance is the necessary condition to maintain the hydrodynamic lubrication mode.

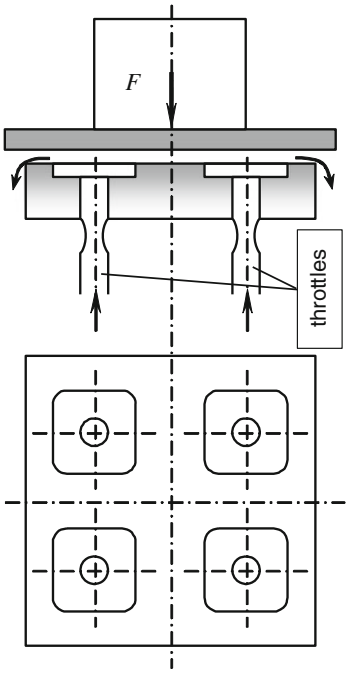


Fig. 3.7 Diagram of operation of hydrostatic bearing [12]

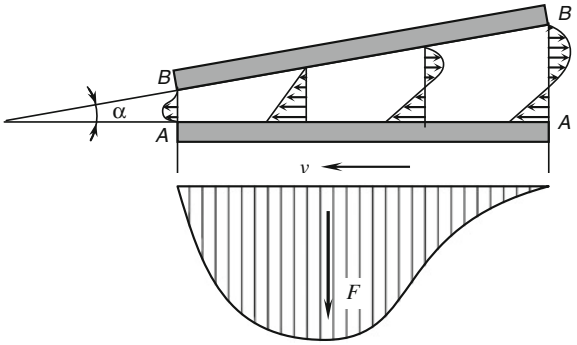


Fig. 3.8 Diagram of formation of bearing lubrication layer [12]

It is apparent that a definite thickness of the lubricating interlayer is as essential for existence of both hydrostatic and hydrodynamic lubrication when the interlayer manifests the volume properties of the liquid. Because of that the pressure in the lubricating film in real conditions should be maintained such that the least thickness of the lubricating layer is not equal to the sum of mean projections on interfaced surfaces. It is achieved in friction units with special technological and design methods and techniques.

However, in the majority of cases, there are several or at least two types of effective friction. In this case the portion of the rubbing surfaces contact directly (friction without the lubricating material) and there is a portion separated by the

boundary film (boundary friction) or by the film of the lubricating material (hydrodynamic or elastohydrodynamic friction). Frequently microwedges appear due to interactions between individual projections so that the load carrying capacity of the films increases additionally. The simultaneous integrity of the concurrent phenomena is called mixed friction.

When analyzing mixed the physical (viscosity, density, compressibility, etc.) and physico-chemical (the affinity to adsorption and chemosorption, chemical affinity of additives with the surface) properties of the liquid and rubbing bodies should be taken into account. The lubricating films appearing during mixed are 2–5 times thicker than the sum of projections on contacting surfaces (Fig. 3.3).

The type of friction depends mainly on the speed, load in the friction zone and lubricating material quality. The lubricating material properties are characterized by its rheological parameters.

Thus, in the presence of the lubricating medium the performance of tribosystems is determined by the friction mode: liquid, boundary or mixed. The diagram of Gercy-Schtriebek provides a vivid idea about the conditions of transition from one friction mode to another when the coefficient of friction f relates the dimensionless parameter $\eta v/p$ that enables to determines the regions of relevant lubrication mode (Fig. 3.9).

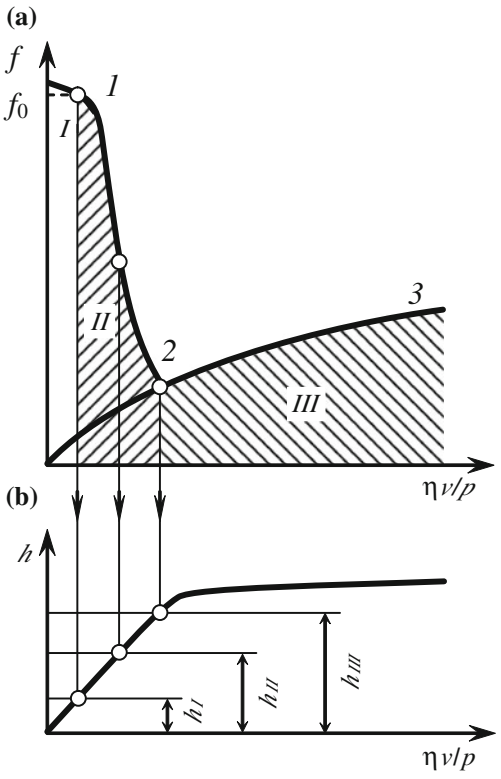


Fig. 3.9 Diagram of friction modes: **a** dependence of coefficient of friction f on Gercy-Schtriebek parameter $\eta v/p$; **b** dependence of lubricating layer thickness h on Gercy-Schtriebek parameter $\eta v/p$ [3]

The parameter $\eta v/p$ is the characteristic of friction modes during mixed and liquid lubrication. During other types of friction the lubricating material viscosity can be taken into account.

According to the diagram of Gercy-Schtriebek, the boundary friction occurs at quite slow speed of sliding of the order of 0.1 mm/s and a very fine lubricating layer of the order of 0.1 μm. The coefficient of friction f remains almost constant when the speed grows to some level. This period is shown by the curve within the portion f_0–1. If acceleration continues, the friction coefficient declines rapidly. The sliding surfaces move away one from another, but not so much as to exclude possible contact between individual projections on the rough surfaces. In this situation the boundary friction is rather possible. That is why such friction is called semiliquid friction by convention (portion *1–2*).

The coefficient of friction f reaches the minimum at the moment when the lubricating layer covers just the projections on the sliding surfaces. Further course of Gercy-Schtriebek parameter is determined by the dependencies on the dimensionless parameter

$$\lambda = \frac{\mu v}{p}, \tag{3.3}$$

where μ—the dynamic viscosity; v—the shaft angular speed; p—the mean specific load on the bearing : $p = P/(dl)$, where P—the radial load on the bearing; d and l—its diameter and length.

When the value λ grows, the lubricating layer gets thicker and covers with excess all the projections on the on the sliding surfaces, excepting those in direct contact. The resistance is wholly determined by the viscosity of the liquid. This friction is called liquid friction (portion *2–3*). Figure 3.9 shows that, when λ and the lubricating layer thickness grows, the coefficient of somewhat increases friction and correspondingly grows heat generation in the working zone of the bearing. Theoretically the most favorable conditions of operation of the bearing would be at point *2* at the minimal *f*. but here the lubricating layer thickness has n margin. The tiniest drop of the value λ, for example, due to the lowering of the viscosity of the lubricant or the shaft angular speed shaft would entail a leap of the coefficient of friction. Therefore for stable operation of the sliding bearing a point is selected in the middle of portion *2–3*. In this case some changes in the viscosity of the lubricant and speed of the shaft do not cause any significant rise of the friction coefficient or intensify heat generation and wear.

Irrespective significant achievement in the sphere of Tribology, nature goes much ahead of the researcher in development of energy saving mobile couples. For example, the synovial joints of man and animals have the friction coefficient 0.001–0.03 or much less than the high-precision rolling bearings in the hydrodynamic friction mode [43]. Nevertheless, it is apparent that it is impossible to stuffy the mass produced friction machines, therefore unique devices and equipment should be developed to conduct such research. Hence some problems of practical adoption of our new methods will be discussed as well as the means of

triboengineering tests and their application to studies of the friction processes in natural and artificial tribosystems.

3.3 Methods and Means of Tribotesting

Triboengineering tests are carried out to get information about triboengineering properties of oils, lubricating and other materials used in mobile couples. According to [51], they include an assessment of their antiwear, antiscoring and antifriction properties using the following means:

(1) laboratory instruments or installations with tested specimens of simple geometrical shape (planes, cylinders, spheres);
(2) simulators with specimens of standard machine parts or specially fabricated similar parts (tooth wheels, parts of the piston groups of the internal combustion engine, sliding or rolling bearing, etc.);
(3) full-scale machine and mechanisms units in operation.

Machine service conditions (frequent stops and starts, variable loads and speeds, moisture and other corrosive agents, abrasive particles in environment, etc.) influence the obtained results during running tests in the same way as main parameters (relative speed of rubbing surfaces motion, pressure, temperature). It is hard to identify the most essential parameter affecting behavior of lubricating material. To diminish these effects the tests should be protracted and conducted simultaneously in several single-type machine that takes much time and means. Therefore, most operation tests are performed for final check of the optimum lubricating materials selected as a result of a series of laboratory and bench tests.

Bench tests with simulators enable to determine the triboengineering characteristics of lubricating materials during friction of real machine parts and mechanisms checking all influencing parameters. However, the tests with simulators are also timeconsuming and costly and used mainly to determine the antiscoring and antiwear properties of oils in teeth wheels, comprehensive tests of motor oils in single- and multicylinder installations, benches for tests of rolling and sliding bearings.

Unlike the field and bench tests of lubricating materials, the *laboratory tests* save much time, they allow the conditions on friction surfaces to be changed and the main parameter affecting the triboengineering characteristics of lubricating materials to be discriminated. The conditions of tests with instruments are different from the real work conditions of the lubricating materials in real machines, but the advantages of the laboratory tests favor their broad application, specifically for developing new additives and lubricating compositions (load, speed and temperature check, measurement capabilities of friction, wear, low volumes of experimental lubricating materials).

The following *types of laboratory tests* of lubricating materials are discriminated:

(1) direct tests (with special machines and instruments to test oils during friction of solids);
(2) indirect tests (lubricity is rated by various physico-chemical parameters without simulating friction between lubricants surfaces).

3.3.1 Laboratory Tribometers

Surface contact geometry of their working elements (the specimens) determining the load in contact area is the main distinctive feature of the laboratory testing machines of various types. The following *schemes of surface contact* are discriminated.

Surface contacts over the area are full (Fig. 3.10a, b) or partial sliding bearing (shaft and cylindrical bearing, shaft and bearing as separate parts of the cylinder) (Fig. 3.10c, d) and mutually sliding plane surfaces. The nominal pressure on the bearing projection can reach 80–100 MPa with these schemes of friction.

Machines and instruments with c contact of friction surfaces over areas.

The Ollman-Viland machine is a steel roll 6.35 mm rotating at 600 min^{-1} (0.2 m/s) r.p.m to which two inserts made of soft steel or bronze are pressed under load (Fig. 3.10c). A hydraulic or a mechanical device with 500 N steps at 30 s intervals transmits the load. The ultimate load is 20,000 N. The shaft and the lower insert do not come into contact. The roll is driven by the motor-scale and the angle of rotation of the stator serves determine the friction moment. The load under which the moment of friction increase abruptly (up to breaking of the safety pin characterizing appearance of seizure) determines the lubricating material quality. The total wear of specimens is determined after tests by mass loss of the roll and inserts.

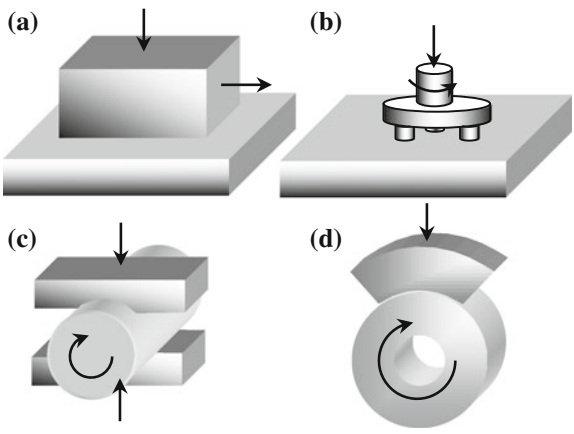


Fig. 3.10 Diagram of friction units of laboratory tribometers simulating contact between friction surfaces over area

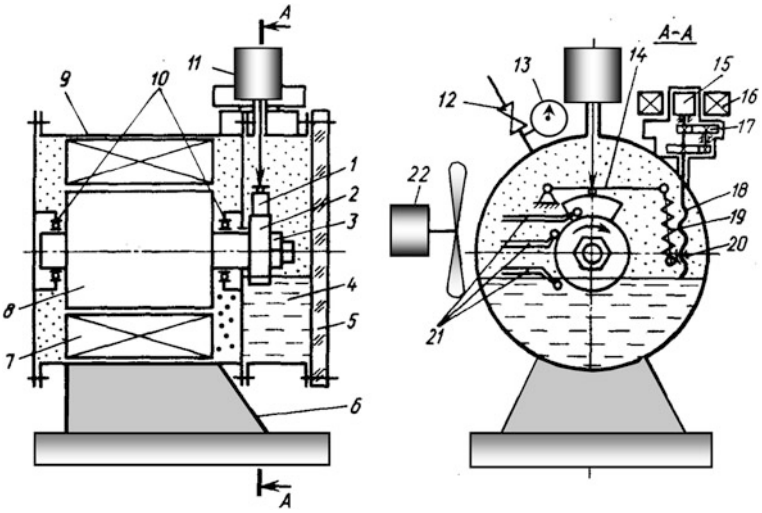


Fig. 3.11 Instrument "rotating roller—partial insert"

The instrument "rotating roller—partial insert". The working specimens roller *2* and insert 1 (block) (Fig. 3.10d) are placed into cylindrical chamber *9* (Fig. 3.11). The roller is secured with nut *3* on the shaft of the synchronous motor connected with the chamber via rolling bearings *10.* Stator 7 and rotor 8 of the motor are arranged in the chamber too.

Self-aligning insert *1* is retained against the roller by lever *14.* The lever is loaded by spring *19* adjustable by nut *20* on screw *18*. The screw is rotated via reducing gear *17* by rotor *15* driven by the reversible motor with its stator *16* outside the working chamber. Airtight chamber 9 is fixed on support *6*. Fan *22* serves to cool the motor windings.

Lubrication of specimens of friction surfaces is performed by immersing the rotating roller into tested oil *4* filling directly into the chamber. Valve *12* serves to fill the chamber with gas. Gauge *13* measures gas pressure in the chamber. Sensor *11* registers continuously the total wear of the specimens. The moment of friction is determined by the current consumed by the motor. The temperature of the block, roller and oil is measured by thermocouples *21.* The friction unit can be monitored through transparent cover *5* during test.

The machine with reciprocating motion of specimens. Figure 3.10a shows the diagram of the working unit of the machine. Slider *1* is moved along the half-cylindrical guide by the crank mechanism in Fig. 3.12. Lower specimen *5* is fixed with clamps *6* on steel platform *4* isolated from slider body with asbestos-cement plate *2*. Heating device *3* built-in between steel platform *4* and plate *2* heats specimen *5* to 350 °C.

Upper specimen *7* is fixed in the socket of force-measuring beam *8* shaped as a parallelogram ensuring constant contact area within one motion cycle. The force-measuring beam is rigidly connected to head *9* moving along guides *10*. The

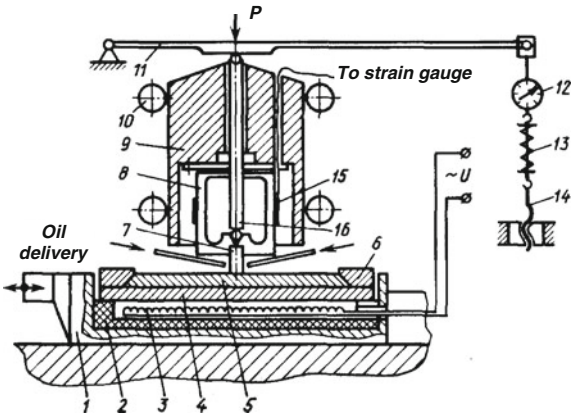


Fig. 3.12 Machine with reprocicating motion of specimens

resilient elements of the force-measuring beam carry welded high-temperature resistors *15* recording friction (along the specimen stroke) with the installed oscillograph. The specimens are loaded by a spring mechanism consisting of unequal-arm lever *11,* load gauge *12*, spring *13* and jack *14*. Loading from the lever is transmitted to the upper specimen through rod *16*.

Tests are conducted at the 260 stroke/min frequency. The sizes of the specimens in mm: the lower one is 130 × 40 × 10, the upper—40 × 5× 25. The contact area is 0.8 cm^2. The oil is fed to the surface of the lower specimen by a micrometric dozer seven drops per minute.

The following parameters are registered during tests: friction force, behavior along the specimen stroke, the temperature near the contact area; the r.p.m. of the lower specimen. Bern-in by step loading is conducted pretest.

The contact load is incremented to 300 N in steps of 160 N for antiscoring property evaluation of oils at a preset temperature of the lower specimen. Transition from one load step to another is conducted after temperature regulation in the contact area. The ultimate seizure load and the temperature in that point are the evaluation criteria of oil antiscoring properties.

Linear contacts between rubbing surfaces are the cylinder and plane, the contact between two cylinders over the generatrix and the contact between the elastic cylindrical wire and the cylinder (Fig. 3.13). In these cases, the initial contact occurs over the surface generated as a result of elastic strain of both working elements. Besides, the initial pressure in the contact can be calculated.

During operation in sliding if wear occurs, a contact area of working elements expands, but it can be determined by the wear track on a stationary specimen. This friction scheme simulates operation of tooth wheels and roller bearings. The maximum pressure in the contact of working elements can reach 1000–1500 MPa.

Machines and instruments with linear contact. The Falex Machine consists of a steel roll 6.35 mm in diameter with the hardness ∼HB 100 and two V-shaped blocks of hardened steel with the hardness HRC 20–24 (Fig. 3.13b). The apex angle of the V-shaped notch is 96 °. The roll rotates at the constant r.p.m. 290 min^{-1}

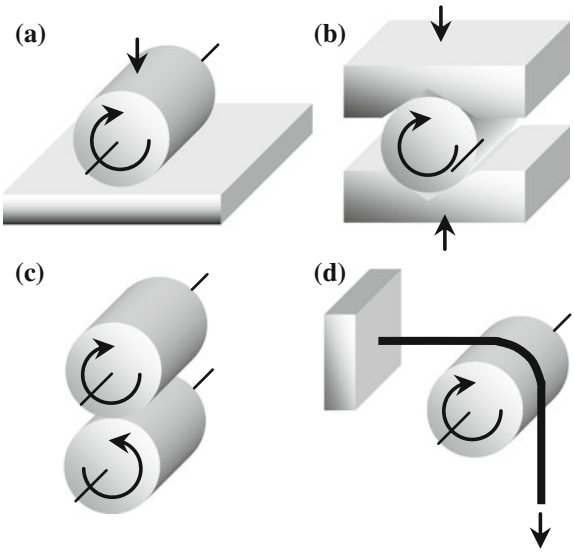


Fig. 3.13 Schemes of friction units of laboratory tribometers simulating contact between friction surfaces along line

(0.1 m/s). When the surfaces of the V-shaped notches touch in two blocks, a linear contact with the roll working surface occurs. Tests are carried out in an oil bath. The temperature of oil can vary from the room temperature to 200 °C by external heating. The step loading is performed. The criterion of ultimate operability of the lubricating material is the moment when the rod connecting the working roll with the machine spindle breaks at a definite step of loading. The maximum load step is 18,000 N.

The Falex Machine consists of a steel roll 6.35 mm in diameter with the hardness ~HB 100 and two V-shaped blocks of hardened steel with the hardness HRC 20–24 (Fig. 3.13b). The apex angle of the V-shaped notch is 96 °. The roll rotates at the constant r.p.m. 290 min^{-1} (0.1 m/s). When the surfaces of the V-shaped notches touch in two blocks, a linear contact with the roll working surface occurs. Tests are carried out in an oil bath. The temperature of oil can vary from the room temperature to 200 °C by external heating. The step loading is performed. The criterion of ultimate operability of the lubricating material is the moment when the rod connecting the working roll with the machine spindle breaks at a definite step of loading. The maximum load step is 18,000 N.

The Timken machine is a rotating roller 49.2 mm in diameter with a flat specimen clamped to it (Fig. 3.13a). The sliding speed is up to 18 m/s. The maximum load is 520 N. The initial pressure calculated by Herzian is up to 570 MPa. The temperature of the specimens can be varied to 260 °C with heaters. The force of friction is registered during tests. Wear is measured with a special electronic system and displayed on the monitor. The measurement accuracy of linear wear is 2.5 μm.

The SAE machine consists of two cylindrical rollers mounted on separate shafts arranged one under another in such a manner that the rollers roll one towards

another with a different circumferential speed (Fig. 3.13c). The lower shaft is connected to the upper driving shaft with a set of tooth gears with different gar ratios. The r.p.m. of the driving shaft can changes in three stages: 500, 700 and 1000 min^{-1}. It corresponds to changing the speed of sliding of the upper roller from 1.31 to 2.62 m/s. The ratio of the sliding speeds of the upper and lower rollers are equal to 14.6:1. The tested oil is placed into an oil tank into which the lower operating roller is immersed. The oil delivered by the lower roller lubricates the upper roller. The load from 0 to 1800 N is transmitted to the lower roller by a lever (lever arms ratio is 10:1) at a continuous rate of loading 310 N/s. The contact pressure can reach 2000 MPa. The tests are carried out until detectable visually seizure occurs on the surface of the upper roller. Tests of lubricating materials with type of the machine approaches by the operating conditions (combination of sliding and rolling, the method of oil delivery to the contact between friction surfaces) to the real operating conditions of tooth gears. Lacking measurements of friction moment, sufficiently accurate measurements of roller wear and some uncertainty of determination of the seizure onset (the subjective assessment of the working surface damage) are the disadvantages of the machine.

It should be notes that the contact plot of the specimens tested in SAE machines is implemented on the friction machines MI-1, SMC-2, 2070 SMT-1 developed and produced in Russia.

The pinpoint contact. It appears when two sphery working elements with mutually perpendicular axes like ball and plane, cone and cylinder contact (Fig. 3.14).

This contact of surfaces is comparable with the contact of balls and rolling bearings cages. The maximum pressure in the contact in that case can reach 5000 MPa.

Machines and instruments with pinpoint contact. The FMT-1 four-ball machine (Fig. 3.14e). The machine spindly with the chuck to secure the upper ball rotates with r.p.m. of 1420–1450 min^{-1}. Standard ball 12.7 mm in diameter are used according to GOST 3722–81 not lower than precision degree II in steel CHX15. Three lower stationary balls are fixed in the cup with the lubricating material. Electric heaters heat the lubricating material; there is a device switching off the

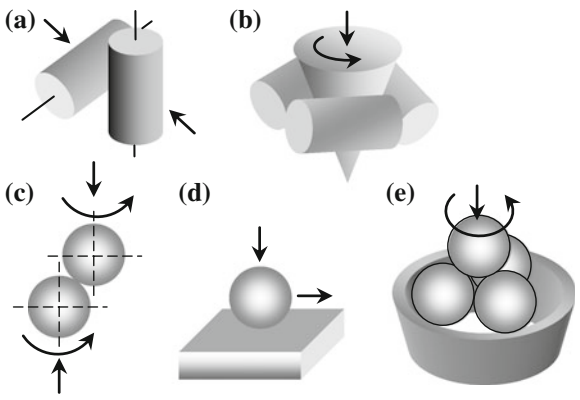


Fig. 3.14 Schemes of friction units of laboratory tribometers simulating pinpoint contacts between friction surfaces

heaters automatically once the ultimate moment of friction between the balls is reached. The method of determination of the lubricating properties is standardized (GOST 9490–75) and covers liquid and plastic lubricating materials. Two types of tests are conducted: short-time with axial load step-up and long-time (60 min) under a constant load. A short-time test lasts 10 ± 0.2 s under each assigned load. Each test under the given load is performed with a fresh sample of the tested lubricating material and with four new balls. After each test, the diameter of the wear spot on each of the three lower balls is measured and an average number is calculated. The tests under the given load are repeated twice.

The short-time tests serve to determine the capability of the lubricating material to prevent seizure of the friction surface and to determine its ultimate load carrying capacity. Critical load P_c at which the wear spot diameter on the lower balls sharply increases (by the value over 0.1 mm); the load of welding P_w at which the machine stops automatically once the moment of friction 1200 N m is reached or the balls weld; the index of scoring I_s is a dimensionless value based on the wear measurements of the balls from the initial load to the load of welding according to GOST 9490–75 are *evaluation criteria.*

The long-time tests serve to determine a wear factor D_w characterizing the effect of the lubricating material on the wear of the rubbing surfaces under constant load less than the critical. The average wear spot diameter on the lower balls after an 1h test is a wear factor.

The four-ball friction machine serves to test lubricating materials according to other methods from methodological guides RD 50-531–85

The Falex-Roxana four-ball machine. A rotating ball 12.7 mm in diameter contacts three stationary balls of the same diameter and made of the same material

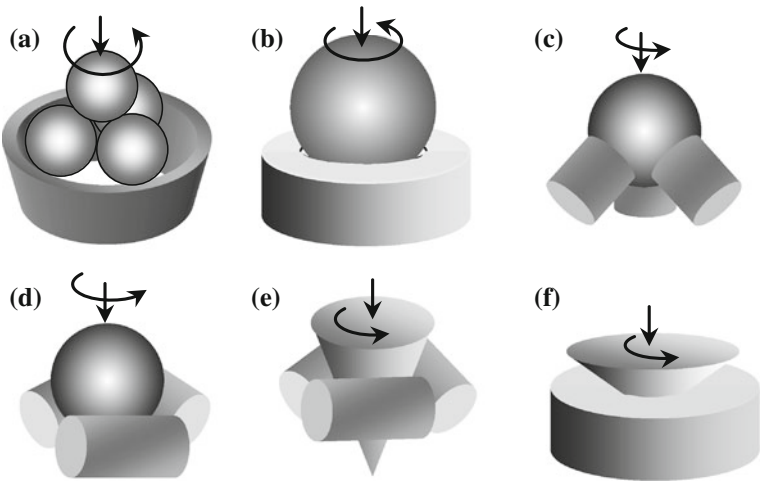


Fig. 3.15 Schemes of tests with machines KT-2 and MACT-1 for determination of temperature of stableness of lubricating materials

(Fig. 3.14e). The upper ball has three r.p.m. 600, 1200 and 1800 min^{-1}. The load is from 1 to 500 N transmitted by the lever to the upper ball. The temperature of the specimens and the tested oil is variable by the heaters up to 300 °C. The wear of the lower balls under different loads and friction force is measured. The duration of the tests under each load is constant (10 s).

The KT-2 and MAST-1 machines serve to determine the temperature stableness of the lubricants in friction. There are six schemes of the tests (Fig. 3.15): *a*—rotating ball—three stationary balls 8 mm in diameter; *b*—rotating ball 12.7 mm in diameter—spherical fillet at a ring end; *c*—rotating ball 12.7 mm in diameter—three specimens with flat surfaces; *d*—rotating ball 12.7 mm in diameter—three cylindrical rings rollers; *e*—rotating conical specimen—three cylindrical rollers; *f*—rotating conical specimen—conical fillet at a ring end.

The schemes *a, c, d, e,* are applicable to tests of lubricating materials for tooth gears, rolling bearings, cam—pusher pairs, etc.; the schemes *b* and *f* are for tests of sliding bearings and piston ring—liner of the internal combustion engine.

The operating principle of the machines is the following: the vertical spindle with the r.p.m. 1 min^{-1} has a collar to secure the upper rotating specimen (a ball or a cylindrical specimen with the cone working surface). The upper specimen contacts with the stationary specimens (in one of the listed schemes of friction) secured in the cup with the tested lubricating material. The axial force can be 10–150 N. The electric heaters can change the temperature of the friction unit and the surrounding layer of the lubricating material within the range 20–300 °C and maintain within the preset level with the error ±3 °C.

The tests are conducted at a constant sliding speed, under a constant axial load and at an incremental temperature. The tests at a preset temperature last 60 s. The test at the next temperature step is conducted with a fresh lubricating material portion and new specimens. The tests according to schemes *b* and *f* are conducted at all temperatures with the same specimens, but the lubricating material is renewed periodically.

In the process of 1-min tests the temperature in the lubricating material volume and the moment of friction between the specimens are registered by a dynamometric recorder with the error ~ 1 %. The test method is standardized (GOST 23.221–84).

Various *evaluation criteria of their triboengineering characteristics* are used when testing lubricating materials, such as:

(1) Frictional losses determined by the torque value, forces or the coefficient of friction.
(2) Load under which critical conditions on the friction surface (seizing, scoring) or a given value of friction or temperature is achieved.

The higher load causes any above-noted critical conditions, the more load range in which the lubricating layer prevents the contact surface.

The wear of friction surfaces is determined at the reference test conditions by load, speed, temperature.

The critical temperature characterizes fracture resistance of boundary lubricating layers and the lubricating material in case temperature changes in the friction unit at

a constant load and speed (the critical breaking temperature of the lubricating layer). In some case of comparative tests the quality of the lubricating material is estimated by the level of heating lubricating layer close to the contact area.

It is better to assess the triboengineering properties of oils and greases with laboratory machines and instruments according to one-at-a-time parameter-variation principle that affects the behavior of the lubricating layer in the friction contact. Wear of interfaced parts or any other damage of friction surfaces occurs when the lubricating layer disintegrates and the surfaces come into direct contact. When the lubricating layer disintegrates partially, the parts wear gradually, if the layer disintegrates completely, disastrous wear, scoring and seizure occur. To reach the first type of damage the tests are conducted at relative moderate loads, speeds and temperatures. To reach the second type of fracture one of the above parameters is increased in steps during the tests.

3.3.2 Pendulum Tribometers

Highly sensitive and accurate triboengineering measurements are implemented, as a rule, with the methods based on oscillatory processes [11, 41, 52–56]. These oscillatory processes are enabled by the physical pendulum (Fig. 3.16a) in which the tested friction unit is positioned either in the pendulum support point (Fig. 3.16b) or out of it when friction interaction of the pendulum proper or some its point with environment is checked (Fig. 3.16c).

During measurements the amplitude of pendulum oscillations are registered and its kinetic dependencies are derived bearing information about the friction interactions.

This method has yielded fundamental results about friction interactions of solid bodies and the mechanism of the lubricating effect of some substances [11, 41]. It is established that the method of inclined pendulum is highly sensitive in studying of the processes of friction both with and without lubrication [11] (Fig. 3.16c) when the plane of pendulum oscillations is at some angle φ to the vertical. Deviation of the plane of pendulum oscillations from the vertical is achieved by contact of the

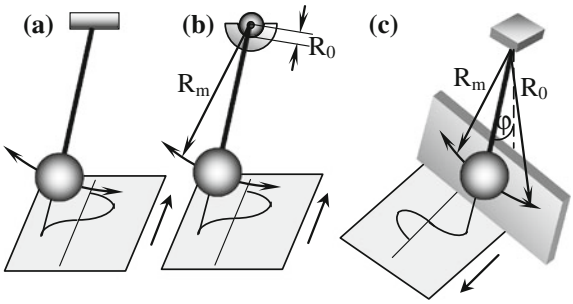


Fig. 3.16 Schemes of pendulum tribometers

pendulum with the inclined surface of some body. This inclined surface can carry a lubricating substance or be free of it.

It is shown that in response to the thickness of lubricating layers of polar molecules and the roughness of rubbing surfaces the attenuation of oscillations of the inclined pendulum can be either linear or exponential. The linear attenuation is typical for Coulomb friction during dry friction of smooth surfaces and during boundary lubrication. The exponential curve of attenuation of pendulum oscillations is typical for viscous friction, i.e. in case of friction interactions when both rubbing surfaces and lubricating materials begin to show viscoelastic properties [11, 57].

It is also established that the shape of the curve of attenuation of pendulum oscillations is affected considerably by the ratio between the radius of the generatrix of the friction surface R_0 (Fig. 3.16b) and the radius from suspension center to the pendulum center of mass R_m (Fig. 3.16b). According to [52], their ratio determines quantity in exponent. When the ratio between these radii is a unity, the highest method sensitivity is achieved.

Registration of oscillation amplitude in operation of the pendulum tribometer is labor-consuming. It was measured directly with a rule in the first models of pendulum tribometers. Graphic registration of pendulum oscillations on the chart moving with constant speed, for example, with a stylus attached to the pendulum, introduces significant error because of the friction of the stylus against the chart. Electromagnetic methods of registration of the amplitude of pendulum oscillations are also poorly accurate.

An optical method based on changes of the light beam cross section by wedge-shaped diaphragm that the pendulum carries in the plane normal to the beam is the best acceptable method for these purposes. The light beam is aimed at a photocell. Its current varies proportionally to the light beam section.

The authors used this approach of amplitude recording in the developed integral pendulum tribometer [58–63].

Figure 3.17 shows the alternative implementation of the integral method of assessment the oscillation amplitude; Fig. 3.18 shows the diagram of stresses, where the abscissa axis is time t, the ordinate axis is deviation of the pendulum from the equilibrium, while the level of stresses represented as the logical zero and the logical one.

The mechanical part of the device (Fig. 3.17) includes base *1* of the pendulum mechanism and fixtures *2* and *6* of stationary *3* and mobile *5* elements of the tested friction unit, pendulum *4* rigidly connected to mobile element *5* of the friction unit and complementary with sensor *7* and pendulum winding mechanism *8*. The electronic part of the device is of high-precision analog and digital integrated microcircuits.

The device operates in the following manner [58]. Winding mechanism *8* deviates pendulum *4* away from the equilibrium position through some angle. At the next moment the pendulum is released and performs oscillatory motion in respect to the equilibrium position (Fig. 3.18a).

Sensor *7* transforms mechanical oscillations of the pendulum mechanism into the sinusoidal analog signal that is sent to the input of voltage—frequency transducer

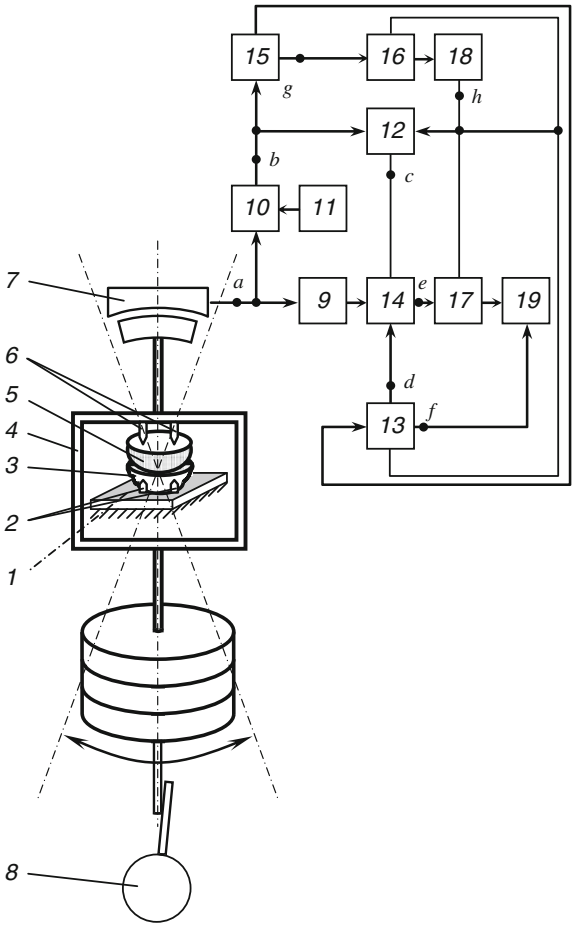


Fig. 3.17 Alternative implementation of integral calculus method of assessment of oscillation amplitudes and their variations per period (Copyright certificate 1630033 A1 SU)

(VFT) *9* and to the first input of comparator 10. Source *11* sends reference voltage to the second input of comparator 10 that it compares with the tested analog signal and shapes at the outlet pulses (Fig. 3.18b) synchronous with the mechanical oscillations. Then these pulses proceed to the *S*-input of trigger 12 and through invertor 15 to the *S*-input of trigger *13* and the counting input of pulse counter *16*.

When the pendulum passes the equilibrium position just the first time the direct output of trigger *12* transforms the logical zero into the logical one (Fig. 3.18c) corresponding to the onset of the first half-wave of oscillations (Fig. 3.18a); element AND *14* connects VFT *9* to the counting input of pulse counter *17*.

The next passage of the pendulum through the equilibrium position corresponding to the end of the first half-wave of oscillations (Fig. 3.18a) switches over trigger *1;* the logical one at its inverse output changes into the logical zero (Fig. 3.18d) and interrupts through element AND *14* the communication between VFT *9* and pulse counter *17*. As a result the sinusoidal electric signal passes from

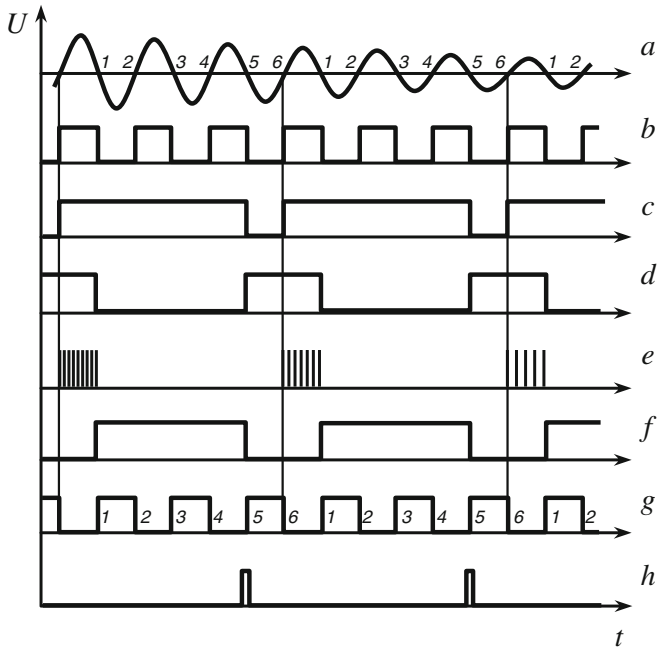


Fig. 3.18 Epures of stresses in characteristic points in schemes shown in Fig. 3.17

the output of sensor *7* into the input of VFT *9* so that its output shapes a pulse packet (Fig. 3.18e) with the number of pulses proportional to the area of the first half-wave of oscillations. These pulses proceed through element AND *14* into the input of pulse counter *17* where they are summed up and converted into the parallel digital code also corresponding to the area of the first half-wave of oscillations. Transformation of the logical zero into the logical one at the direct output of trigger *13* (Fig. 3.18f) shapes the pulse of parallel recording so that the counting result is updated from the output of counter *17* into microprocessor unit *19* in order for saving and further processing. Simultaneously the signal of transformation of the logical zero into the logical one inverted by invertor *15* (Fig. 3.18g) actuates pulse counter *16* with its output connected to the input of decoder *18*.

As a result after the number of passages of the pendulum through the equilibrium position assigned after the cycle of measurements the respective output of decoder *18* generates the zero-to-one differential (Fig. 3.18h) that resets the inputs and returns triggers *12–13* and pulse counters *16–17* into the initial zero position ready for further measurements. The next passage of the through the equilibrium position corresponding to the even number after the cycle of measurements is over (Fig. 3.18a, h) turns trigger *12* into the logical one state (Fig. 3.18c) and the described processes repeat.

Microprocessor unit *19* accumulates the results from pulse counter *17* and respective areas of even or odd half-waves of oscillations. The above formulas use

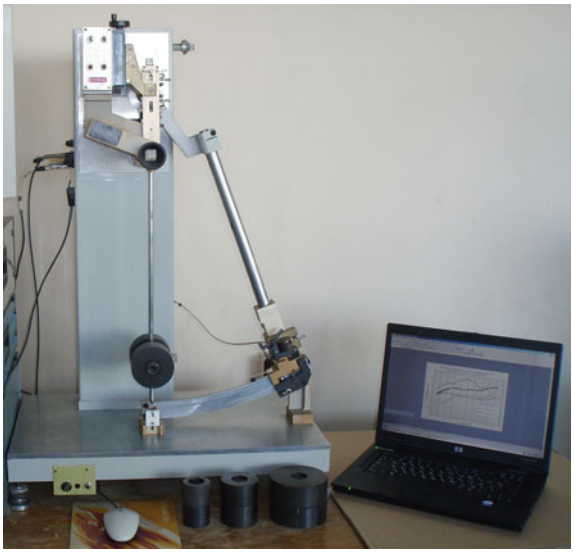


Fig. 3.19 External type of new electron digital pendulum tribometer

them to determine the oscillation amplitudes and their variations per period. Then, with the account of the features of the earlier descried processes, the microprocessor unit performs further processing of the obtained experimental data. The result is determination of the friction coefficients in the bearing unit during each cycle of oscillations of the pendulum and their dependencies on the number of cycle or on time.

Figure 3.19 shows design of new pendulum tribometer.

Possibility to change the configuration of the tested friction unit is the peculiarity of this instrument. In particular, the device can be adjusted to study friction interactions in the cartilage. For this purpose, the joint is partitioned. An articular cavity is secured on the support platform of the tribometer; an articular tubercle is secured on the platform of the specimens of the pendulum. Then the articular elements are connected together and the natural synovial fluid is injected. The pendulum oscillates and the areas of oscillations due to deviations of the pendulum sideward from the equilibrium position are determined. The difference between two adjacent areas indicates the friction magnitude in the joint. Then the synovial fluid is removed and replaced with the tested fluid in the joint, measurements are repeated. The lubricity of the tested fluids is estimated by the difference between the obtained results and the results for natural synovial fluid.

3.3.3 Indirect Methods of Lubricant Testing

The wetting angle. Adsorption of liquid lubricants can be rated by the value of the wetting angle formed by the tangent to interface between the liquid droplet and the solid surface. When the surface is hydrophobic tending to repulse the lubricating

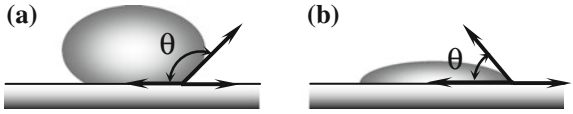


Fig. 3.20 Drop of liquid on hydrophobic (**a**) and hydrophilic (**b**) surfaces

fluid, the edge angle θ is large, while it is small when the fluid is hydrophilic and attracting (Fig. 3.20).

The wetting angle is measured in the following manner. A drop of the test fluid is applied with a pipette on the surface placed on the horizontal table of the measuring microscope MMI-2. First the microscope is turned so that the working table surface becomes vertical; the eyepiece of the measuring tube faces the researcher. Then a piece of the metallic angle with the profile number from 2 to 3.2 is attached to the table center, its second shelf forms a platform under the tested specimens.

For measuring the drop is aligned with the microscope field of vision and the sighting intersecting lines are aligned with the point where the contact between the drop on the tested fluid surface and the air is visible (Fig. 3.20). Then the angle measuring by the microscope is adjusted to measure the wetting angle. The degree of wetting of the solid surfaces is inversely proportional to the value of the wetting angle. The measure of wetting is assumed to be equal to wetting angle cosine.

Determination of the contact potential difference. The method of determination of lubricant—solid adsorption is determination of variation of the electronic work function with the contact potential difference method (CPD). The electronic work function is an energy characteristic of solid surfaces and represents the work to be expended by the electron to leave a metal. Under other equal conditions this function can change when the surfaces of metals contains adsorbed molecules of various lubricating liquid characterized how much the surface is filled with molecules and fluids or the number of adsorbed molecules.

The CPD serves to rate changes in the electronic work function of two metals, one being the reference metal bodies (its electronic work function remains constant during tests). Usually a golden plate serves as a reference. For measuring the CPD the instrument with the dynamic capacitor is used (Fig. 3.21). The sensor is the surface of the measured A and reference B specimens shaped as sheets forming the variable capacitance with the air dielectric to register vibration of reference specimen B [64].

If the capacitor sheets have different electronic work functions φ, the electric charge Q appears determined by its capacitance $C_к$. Because the charge is constant $(\varphi_A - \varphi_B) = \Delta\varphi = const$ when the reference plate B vibrates, the output voltage $U_в$ changes because $Q = U_к C_к = U_в C_в$ and $U_в = U_к C_к / U_к C_к$, where $C_в$—the capacity of the amplifier (a constant value). The indicator receives a signal proportional to the value $U_к = \varphi_A - \varphi_B = \mathrm{CPD}_{AB}$.

In order to measure the signal the compensation method is used according to which the potentiometer R_1 receives the counter voltage equal to $U_к = \mathrm{CPD}_{AB}$.

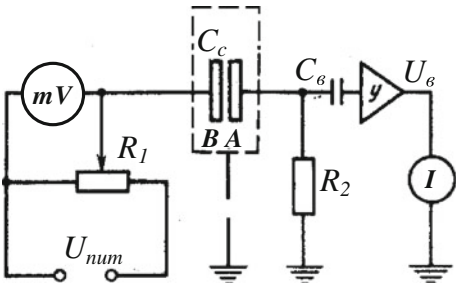


Fig. 3.21 Diagram of measurement of electron work function by method of dynamic capacitor

Under these conditions and when the electronic work function of the reference specimen (gold having in atmospheric conditions a practically constant electronic work), the numeric value of the electronic work function of the tested metal can be determined from the equation

$$\varphi_x = \varphi_B - U_к, \tag{3.4}$$

where e—the electron charge.

The values $U_к$ = CPD are usually expressed in millivolts.

To assess the adsorption properties of the lubricating material on the solids the CPD of a pure plate surface (the specimen) is measured at first, then the surface is covered with a thin layer of the lubricating material and the CPD is determined again. The difference between the CPD of the pure and lubricated surfaces (ΔCPD) characterizes the degree of lubricant—solid adsorption. The larger the difference, the stronger the adsorbed molecules adhere to the solid body surface.

Determination of electrode potential. For comparative assessment of interactions of lubricating oils and plastic greases with metallic surfaces the electrode potential of the metallic specimen (the electrode) is determined after exposure in the tested lubricating composition at a given temperature [65]. A standard pH-meter serves to implement the method. The temperature at which the metallic specimen is thermostatted should be close to the temperature of operation of the tested lubricating material. The electrode potential is changed by adhesion of the active components of the lubricating composition to the metallic surface or its chemical modification.

The adsorption heat measurement method. During contact between the solid body surface and the lubricating material the concentration of molecules of the surfactant in the surfaces of bodies is always higher than in the lubricating material volume. This higher concentration is accompanied by saturation of certain portion of the unbalanced forces on the solid body surface and a drop of its surface energy.

The process of adsorption is accompanied by reduction of the free energy and entropy, hence by the receding enthalpy of the system that evidences the exothermal nature of the process. The released heat called adsorption heat is an essential characteristic of the process of adsorption and a number of processes depending on adsorption.

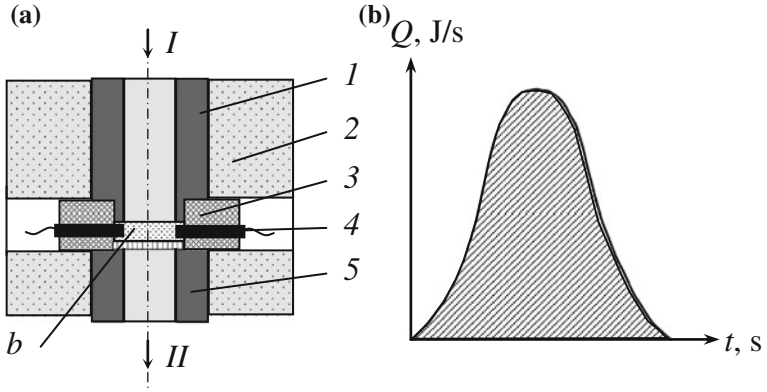


Fig. 3.22 Principal diagram of flow-through microcalorimeter cell **a** *1* input tube; *2* metallic body; *3* heat insulating jacket; *4* thermistors; *5* output tube; *6* adsorbent. Flows of carrying fluid: *I* input; *II* output. **b** Typical diagram produced by microcalorimeter

The adsorption heat of lubricating substances on metals is usually measured with flow-through microcalorimeters; their principle of operation is that the thermal effect is registered when the tested solution is passed through a special measuring cell with an adsorbent that is in the thermal equilibrium with the carrier neutral fluid, for example, h-heptane. Figure 3.22 shows the diagram of the microcalorimeter.

Figure 3.22b, shows a typical microcalorimeter recorder trace during adsorption. The adsorption heat is calculated from the area limited by the curve and the abscissa axis (dashed).

The hot wire technique. At elevated temperatures generated in the contact between friction surfaces the chemically active components of lubricating compositions can react with the metallic surface producing chemically modified protective layers of metal chlorides, sulfides, and phosphides. To estimate qualitatively the reactivity of the lubricating medium with chemically active additives the hot wire technique is applied when a high temperature of the metal (up to 600 °C) can be reached by passing electric current through a thin wire immersed into test lubricating composition. Periodic (pulsing) current passage through the wire simulates indirectly temperature flashes that occur in the contact surface in friction.

The wire resistance when the electric current passes through it is determined by comparing the voltage drop in the wire with the voltage drop when the current passes through the reference resistance. During chemical reaction between the wire and the additive the wire diameter diminishes and its electrical resistance grows. The measured resistance is proportional to the chemical reaction rate. The wire temperature changes by varying the input voltage so that the reaction rate between the wire and the additive can be measured at different temperatures.

Another method of assessing the chemical reactivity of additives is used when the wire burns due heated by the electric current when it is periodically dipped into the tested oil with a chemically active additive. The shorter the time needed for the wire to burn, the additive is chemically active [65].

The thermogravimetric method. To investigate the reactivity of the chemically active additives and their behavior at elevated temperatures a thermogravimetric analysis is used with a derivatograph that registers simultaneously the heat effects when heated substance (characterizing exo- and endothermic reactions), the substance weight loss and the mass loss speed changes in time.

The main instrument is a registering pyrometer that automatically records two heating temperatures: from a simple thermocouple that registers the heating temperature of weighed sample and from a differential thermocouple that registers endo- and exoeffects. The thermocouples are mounted on one arm of the analytical scale beam so that the derivatograph is able to register continuously the mass changes of the weighed sample and the rate of changes. The derivatograph records these temperatures, mass changes and their rate as functions of time for both pure additives and their mixtures of the reduced metal (iron usually). The temperature of melting, boiling and decomposition is determined by the curve recorded by the differential thermocouple.

Assessment of protective properties of oils in the corrosive medium [66]. Mass loss of the steel ball 9.5 mm in diameter, steel CHX15, after exposure at a given temperature in test oil during 7, 14 and 24 h and further treatment with spirit of salt or sulphur. Also it is possible to determine how mass losses by the specimen depend on thermostatting point within 20–250 °C.

Assessment of copper corrosive wear caused by motor oils [66]. Mass loss of the cylindrical copper rod (copper M1) when it rotates in a cup from corrosion resistant steel filled with the tested motor oil that contains abrasive powder (the electrocorundum 14A6p). The electric heater within the range of temperatures 130–250 °C heats the oil in the cup. Tests are conducted at the preset during 30 min. The oil temperature is adjusted by the thyristor device and registered by the electron potentiometer.

3.3.4 Operating Tests

These tests serve to select finally the lubricating materials suitable for the design features of an object, the conditions of its operation and service maintenance.

The duration of these tests depends on many factors, mainly on the wear rate of interfaced units and parts of a tribosystem. As a rule, the tests are conducted until parts begin to wear to the extent exceeding the error of measurements or until the tribocouple fails to operate (fracture, scoring, seizure, etc.). The operation tests serve to register the magnitude of wear of parts of the friction units during a definite period of operation or to register the number of failures of some tribocouple. The results serve to decide which lubricating material is suitable for application.

In the process of the tests it is necessary to monitor the kinetics of wear either by disassembling the tribosystem and apply micrometry or by technique of cutting holes, or without disassembling it by analyzing the concentration in the lubricating oil of different elements that the structural materials of the rubbing couple contain

(iron, copper, aluminum, etc.). When necessary the concentration of active elements in the oil is determined that are included into the composition of additives to the oil (antiwear, antiscoring, antifriction additives) and depletion of the additives is determined by their loss.

Comparison of the data on oil samples during a certain time enables to estimate the wear rate (by metal accumulation in the oil) and consumption of additives (by accumulation of active elements) that, in its turn, permits to estimate the tribosystem condition, to predict reliability of friction units, to schedule current and overhaul repairs.

Control methods of the friction unit dynamic wear without disassembling are more economical (cutting test time).

There are many methods of the quantitative determination of the elementary composition of the wear products in oil samples. They can be conventionally split into the following main groups: physico-chemical, spectral analysis, nuclear-physical and photometric.

The physico-chemical methods are mass, volume, calorimetric and polarographic methods. They are sufficiently sensitive (up to 10^{-6} g of metal per 1 ml of oil), yet they are quite labor-consuming and little efficient. The polarographic method serves to study oil samples during electrolysis. The oil is first incinerated and the residue is treated with acids. Then they are evaporated and the obtained mineral residue is dissolved in the electrolyte. The voltage of the electrolytic cell is changed and the current is measured. The results serve to plot the volt-ampere characteristic (the polarogram) and the elementary composition of the sample and the concentration of substances are determined.

The group of the spectral analysis methods includes the following.

The spectral atomic absorption method analyzes the elementary (quantitive and qualitative) composition by atomic emission spectra. The concentration of the element is question is judged by the intensity of attenuation of lines in the oil sample.

The MPS spectrometers of different modifications are used, in particular the spectral analyzer MPS-5, produced by the Leningrad Optic-Mechanical Group (LOMG). The argon plasma as a source of excitation in the spectrometer is excited by high voltage from the high-frequency generator (25 MHz). Some quantity of the oil sample ($\sim$0.8 g) is diluted with xylene and injected into the argon plasma. The emission of excited atoms when the oil incinerates in the plasma is registered with optical system consisting of the concave diffraction lattice and a set of photoelectric devices, deciphered by the computer and sent to a print. The method is efficient (300–400 samples per day) and highly sensitive (up to 10^{-8} g per ml of oil). The main drawback is a lack of registration of the relative contribution of wear particles in oil samples with the size over 10 μm cause these comparatively large particles do not manage to disintegrate into atoms in the spectrometer plasma and can introduce error into the analysis. The method is applied to assess wear of combustion engines and to diagnose condition of friction units of locomotives, gas engine compressors and other machines [18].

The infrared spectroscopy determines the molecular composition of the test oil sample and serves to determine the concentration of oxidation products in the oil,

sulfates, inorganic nitrites and salts of carbonic acid by the spectra of absorption in the infrared range. The so-called "ageing number" serves to determine the extent of oil ageing quantitatively. The method is implemented by means of the IR instrument produced by the LOMG and other manufacturers.

The emission spectrophotometry uses the spectra of emission of atoms and ions under the effect of electromagnetic excitation from an electric light source (the electric arc, sparkle). Preliminarily the oil sample is centrifuged, filtered or incinerated to separate the residue. The residue is burnt by the electric arc. The beams of light emitted by the elements the sample contains are separated from the spectrum by the output slots in the instrument and projected by the mirrors to photoreceivers. The spectrophotometry serves to measure the ratio between the intensity of lines of the analyzed element and the reference line. The wear rate of a rubbing couple of some unit or mechanism is characterized indirectly by the concentration of metals in the oil sample that the rubbing parts contain.

The nuclear physical methods are scintillation damping, radiotracer method, surface activation (the differential method of radioactive indicators), neutron activation analysis, nuclear magnetic resonance (NMR).

Scintillations or glow of oils used in engineering appears under the effect of ultraviolet light or ionizing radiation. When there are mechanical impurities, the glow intensity of oils recedes. This is the principle of the method of scintillation damping. The proper glow of petroleum and synthetic lubricating oils is weak, that is why activators are added to obtain sufficient light output and register scintillation with the radiometric instruments. Usually a scintillation sensor with a photoelectron multiplier serves to register flashes. The method provides information about wear products without disassembling or stopping machinery to examine oil samples and does not require preliminary activation of parts or wear products. The sensitivity of the method is 10^{-5} g. it is applicable for laboratory studies of antiwear properties of lubricating materials.

The radioindication method. According to this method the wearable part is activated either directly in the reactor or by electrolytic application of a radioactive coating by injecting a radioactive isotope during metal melting, or, finally by making radioactive inserts in the studied part. In operation the activated parts wears, the wear products penetrate into the oil. The activity of oil samples is determined by special counters of radioactivity. To ensure high sensitivity of the method strong radioactivity of parts is required, abode 1 mCi. In its turn, it necessitates special protection of the operation personnel from radiation and premises from radioactive contamination. That is why the radioindication method can be used in special laboratories equipped with protective devices in accordance with the sanitary norms.

The method of nuclear magnetic resonance (NMR) is based on the resonance absorption of electromagnetic energy by ferromagnetic particles the oil sample contains. The NMR spectra are read of the pure oil, waste oil and the oil that operated after removal of mechanical impurities. The concentration of particles in the oil sample is determined by comparison of the broadening of the spectral line of the respective element. The NMR method can also determine the concentration of paramagnetic and diamagnetic particles.

Photometric methods include the dispersion analysis of particles of wear products and ferrography. The automatic analyzers are extensively used to determine the dispersion composition of wear particles in oil. They are based on the photometric method of counting of particles of definite size groups [67]. The instruments of type like PC-112 are compact, have light weight and can be used during operation of various machines and mechanisms.

The analyzer contains a glass flow-through cuvette and light source producing a well-lit zone in the center of the cuvette. When the liquid flows through the suspended particles cross the lit zone and attenuate the light flux by the value proportional to the area of cross section of the particles towards the optical axis. Changes of the light flux are registered by the photoreceiver and transformed into. The electric pulse after amplification are directed from the output of the photodetector to the input of the electronic unit that counts the pulse the amplitude of which is between the lower and per levels of discrimination. Only the particles of a definite size group are registered and the results are printed out.

The analyzer has the following sub-ranges of registration of particle sizes:

I—5–10 μm; II—10–25 μm; III—25–50 μm; IV—50–100 μm; V—100–500 μm.

The analyzed oil sample is diluted depending on the level of contamination in order to obtain the sample with the concentration of particles below the ultimately tolerable value 1,000 particles/cm^3. The solvent benzene "galosh" intended to dilute oil samples is pre-filtered through a paper or nitrocellulose filter. Then the diluted sample is thoroughly mixed in an electromechanical stirrer during 15 min. to count a number of particles within one sub-range 25 ml of the mixture is needed.

A new PC-151 photometric analyzer has been developed. It is a six-channel analyzer that needs no solvent to operate. A single analysis of the studied oil prints out the results of all five sub-ranges of sizes of wear particles. The method of dispersion analysis of wear particles in oil samples with PC-112 instruments serves to diagnose the condition of friction units of centrifugal and piston liquid pups at compressor stations of main gas pipelines. The method is also used to diagnose other mechanisms (internal combustion engines, hydraulic motors, etc.).

The method of analyzing wear products based on extraction of ferromagnetic and paramagnetic particles from oil sample by magnetic field has become popular and received the name of ferrography.

Foxborough Co (USA) developed several modifications of ferrographs, analytical and direct displaying to implement the method.

This instrument has a rotation pump to make test oil sample drippled from the test tube to a thin glass or plexiglass plate with a trough directing oil drops into the oil receiver. The plate is placed under magnets. The particles of wear products extracted by magnetic force fields precipitate in the trough on the plate and distribute into larger particles in the upper portion of the plate and into smaller particles in its lower portion (as the magnetic field intensifies). After the tested oil sample has drippled off the plate it is pumped through the "fixative" that fixes the wear particles settled on the plate. When the traces of distribution of particles along the late are photographed the ferrograms are obtained and compared for different oil samples to get the patter of wear of the studied tribosystem in time. For more

detailed wear investigation, the plate with the fixed ferrogram is studied under an optical microscope with a set of short-distance objectives of various magnifications or with a scanning electron microscope. The following technique is used for approximate determination of the composition of wear particles. The glass plate with precipitated wear particles is heated to 320 °C during 90–120 s. After heating, the metal wear particles, depending on the nature of the metal, acquire distinctive colors: blue (high-carbon steel); yellow-brown (pig iron); white-gray (chrome, lead, aluminum). The color microphotographs of the particles enable to identify approximately their nature.

The direct displaying ferrograph is a plastic oil conduit through which the tested oil sample flows with a solvent under a set of magnets. The oil sample is transilluminated by the light flux at two portions of the oil conduit: at the inlet above the first magnet and at the outlet above the last magnet. The intensity of the flux is registered by an electronic system. As the oil sample flows above the magnetic fields the wear particles are extracted and trapped together with other condensed products (for example, polymers). The wear products obstruct passage of the light flux through the oil conduit; the more the particles are there, the more light is absorbed by the oil sample. The electronic system displays digital data on number of large and small particles counted by the electronic units using the principle of definite attenuation of light fluxes. The fewer particles are detected in the oil sample, the less is the wear rate of the friction unit.

The direct displaying ferrograph can be built-in directly into the main oil pipe of a machine or a bench. In this case, the instrument is equipped with a device for periodic with short intervals passage of a portion of the oil flow from the man oil pipeline into the oil-measuring conduit of the instrument. After the electronic units process the sample and large and small wear particles are counted, this oil portion can be returned into the main pipeline. In this manner, the friction unit can be monitored continuously.

References

1. B.N.J. Persson, *Sliding Friction: Physical Principles and Applications* (Springer, Berlin-Heidelberg-New York, 2000), p. 515
2. K.C. Ludema, *Friction, Wear, Lubrication: A Textbook in Tribology* (CRC Press LLC, Florida, 2000), p. 257
3. M. Hebda, A.V. Chichinadze (eds.), *Handbook of Triboengineering: Theoretical Principles in,* vols. 1, 4. (Mashinostroenie, Moscow, 1989), p. 400 (in Russian)
4. P.N. Bogdanovich, V.Y. Prushak, S.P. Bogdanovich, *Friction and Wear in Machines* (Texnologia. Minsk, 2011), p. 527 (in Russian)
5. E.L. Shvedkov, D.Y. Rovinskii, *Glossary of Friction, Wear and Lubrication of Machine Parts* (Naukova Dymka, Kiev, 1979), p. 185 (in Russian)
6. GOST 27674-88, *Friction, Wear and, Terms and Definitions* (Moscow, 1988), p. 20 (in Russian)
7. International Standard ISO 4378/1, 2, 3
8. I.V. Kragelskii, *Friction and Wear* (Moscow, Mashinostroenie, 1968), p. 480 (in Russian)

9. N.M. Mikhin, *External Friction of Solid Bodies* (Nauka, Moscow, 1977), p. 222 (in Russian)
10. F.P. Bowden, D. Tabor, *Friction and Lubrication of Solids* (Mashinostroenie, Moscow, 1968), p. 544
11. A.S. Akhmatov, *Molecular Physics of Boundary Friction* (Fizmatgiz, Moscow, 1963), p. 472 (in Russian)
12. D.N. Garkunov, *Triboengineering* (Mashinostroenie, Moscow, 1989), p. 328 (in Russian)
13. P.A. Rebinder, E.D. Shchukin, Achievements of physical sciences. Phys. Ushekhi. **8**(1), 3–42 (1972) (in Russian)
14. V.A. Belyi, A.I. Sviridenok, Present trends in development of tribological research. J. Frict. Wear **8**(1), 5–24 (1987) (in Russian)
15. M. Hebda, A.V. Chichinadze (eds.), *Handbook of Triboengineering: Lubricating Materials, Lubrication Engineering, Sliding and Rolling Bearings*, vols. 2, 3. (Mashinostroenie, Moscow, 1990), p. 416 (in Russian)
16. V. Latyshev, A.E. Krylosov, R.I. Karabanov, *Role of Chemical Compounds at a Friction of Metals. Physico-Chemical Mechanics of the Friction Process* (Ivanovo University, Ivanovo, 1977), pp. 3–18 (in Russian)
17. G. Parfit, C. Rochester (eds.), *Adsorption from Solutions on Surfaces of Solid Bodies* (Mir, Moscow, 1986), p. 488
18. A.A. Abramzon, *Surfactants. Properties and Applications* (Khimiya, Leningrad, 1975), p. 248 (in Russian)
19. *Physical Encyclopedic Dictionary* (Nauka, Moscow, 1984), p. 944 (in Russian)
20. A.V. Chichinadze (ed.), *Fundamentals of Tribology (Friction, Wear, Lubrication)* (Mashinostroenie, Moscow, 1995), p. 779 (in Russian)
21. E.D. Shchukin, E.A. Amelina, L.A. Kochanova et al., Physico-chemical mechanics of contact interactions. J. Frict. Wear **1**(2), 247–262 (1980) (in Russian)
22. S.G. Arabyan, A.B. Vipper, I.A. Kholomonov, *Oil and Additives for Tractor and Harvester Engines* (Mashinostroenie, Moscow, 1984), p. 208 (in Russian)
23. P.I. Sanin. Chemical Aspects of Boundary Lubrication // Journal of Friction and Wear (in Russian), vol. 1, no 1, pp. 45–57, 1980
24. R.M. Matveevskii, V.L. Lashkhi, I.A. Buyanovskii et al., *Lubricating Materials: Antifriction and Antiwear Properties, Methods of Tests, Handbook* (Mashinostroenie, Moscow, 1989), p. 224 (in Russian)
25. K.M. Badyshtova, Y.A. Bershadt, S.K. Bogdanov et al, *Fuels, Lubricating Materials, Technical Fluids. Range and Application*, Reference Edition (Khimiya, Moscow, 1989), p. 432 (in Russian)
26. Y. Drozdov, V.G. Pavlov, V.N. Puchkov, *Friction and Wear in Extreme Conditions: Handbook* , (Mashinostroenie, Moscow, 1986), p. 224 (in Russian)
27. S.G. Entelias, E.M. Berliner (ed.) *Process Lubricant Coolants for Cutting Metals, Handbook* (Mashinostroenie, Moscow, 1986), p. 352 (in Russian)
28. D.N. Garkunov, *Selective Transfer in Heavily Loaded Friction Units* (Mashinostroenie, Moscow, 1982), p. 208 (in Russian)
29. V.N. Litvinov, I.K. Mikhin, N.K. Myshkin, *Physico-Mechanical Mechanics of Selective Transfer in Friction* (Nauka, Moscow, 1979), p. 187 (in Russian)
30. M.L. Rybakova, L.I. Kuksenova, *Structural and Wear Resistance of Metals* (Mashinostroenie. Moscow, 1982), p. 212 (in Russian)
31. V.F. Pichugin, On the selective transfer mechanism in wear of copper Alloy-Stell units. J. Frict. Wear **5**(2), 284–294 (1984) (in Russian)
32. R.N. Zaslavskii, V.D. Asrieva, Y.S. Zaslavskii et al., The mechanism of antiwear action of a friction polymer-forming lubricating grease. J. Frict. Wear **2**(1), 125–133 (1981) (in Russian)
33. V.G. Lapteva, E.N. Dokuchaeva, V.F. Kaplina, Wear resistance of friction units in manufacturing equipment when using tribopolymer forming lubricants. J. Frict. Wear **6**(1), 98–106 (1985) (in Russian)
34. L.S. Pinchuk, V.A. Goldade, *Electret Materials in Machine Building* (Infotribo, Gomel, 1998), p. 288 (in Russian)

35. A.A. Silin, Friction in space vacuum. J. Frict. Wear **1**(1), 168–178 (1980) (in Russian)
36. A.A. Silin, To a behaviour and stability question it is artificial rased tribosystem. Frict. Wear Lubr. Mater. **2**, 269–299 (1985) Moscow (in Russian)
37. I.V. Kragelskii, M.N. Dobychin, V.S. Kombalov, *Principles of Calculation of Friction and Wear* (Mashinostroenie, Moscow, 1977), p. 526 (in Russian)
38. S.M. Ayupov, O.F. Kondrashov, I.L. Markhasin et al., Definition Rheological parameters of boundary layers of liquids on an example of solutions of stearin acid in oil. Colloid J. **38**(1), 3–7 (1976) (in Russian)
39. V.A. Smurugov, I.O. Delikatnaya, On the structurization of the thin oil films on the polymer surfaces. J. Frict. Wear **4**(6), 1108–1110 (1983) (in Russian)
40. A.A. Kut'kov, About the mechanism of a friction of the polymers greased with surface-active greasings. Mech. Polym. **1**(1), 128–135 (1965) (in Russian)
41. P.A. Rebinder, *Physico-Chemical Mechanics. Selected Works* (Nauka, Moscow, 1979), p. 381 (in Russian)
42. S.F. Ermakov, V.P. Parkalov, V.A. Shardin, R.A. Shuldykov, Effect of liquid-crystal additives on tribological performance of dynamically contacting surfaces and mechanism of their friction. J. Frict. Wear **25**(2), 87–91 (2004)
43. D. Moor, Principles and Application of Tribonics, Moscow, Mir, 1978, p. 488
44. P.G. Alekseev, A.V. Shcheglova, Effect of surface active medium on the deformation hardening and wear resistance of solid surface. J. Frict. Wear **4**(2), 189–193 (1983) (in Russian)
45. G. Biresaw (ed.), Tribology and the liquid-crystalline state. Am. Chem. Soc. Symp. Ser. **441**, 130 (1990)
46. J. Cognard, *Alignment of Nematic Liquid Crystals and Their Mixtures* (Science Publishers, London-New York-Paris, 1982), p. 104
47. A.A. Markov, Y.V. Lun'kov, T.N. Nazarova, V.K. Gusev, Test studies effect of adsorption of the oils on the wear resistance of metals. J. Frict. Wear **5**(3), 538–541 (1984) (in Russian)
48. A.B. Vipper, V.L. Lakhshi, Y.A. Mikutenok, The effect of friction modifiers on the engine oil performances. J. Frict. Wear **2**(5), 935–937 (1981) (in Russian)
49. V.L. Lakhshi, A.B. Vipper, V.V. Kulagin, Oil-soluble organic connections of molybdenum—additives to lubricant oils. Chem. Technol. Fuels Oils. **1**, 56–58 (1984) (in Russian)
50. A.P. Semenov, M.V. Nozhenkov, On the nature of lubrication by the solid antifriction materials. J. Frict. Wear **5**(3), 408–416 (1984) (in Russian)
51. R.M. Matveevskii, V.L. Lashkhi, I.A. Buyanovskii et al., *Lubricating Materials: Antifriction and Antiwear Properties, Methods of Tests, Handbook* (Mashinostroenie, Moscow, 1989), p. 224 (in Russian)
52. A. Unsworth, D. Dowson, V. Wright. The frictional behavior of human synovial joints. Part 1. Natural joints. Trans ASME **F97**(3), 369–376 (1975)
53. B.V. Budanov, V.A. Kudinov, D.M. Tolstoy, Interrelation of friction and vibration. J. Frict. Wear **1**(1), 79–89 (1980) (in Russian)
54. R.M. Akhmatova, M.S. Ostrovskii, Statistic method for measuring the coefficients of friction and wear of precise instruments and machines operating under vibration conditions. J. Frict. Wear **9**(4), 719–723 (1988) (in Russian)
55. P.L. Krupkin, K.V. Tsivanyuk, Study of regular variations in friction coefficient. J. Frict. Wear **4**(2), 277–284 (1993) (in Russian)
56. A.L. Zharin, *The Method Contact Difference of Potentials and its Application in Tribology* (Nauka, Minsk, 1996), p. 236. (in Russian)
57. R.A. Brand, Joint Lubrication. Chapter 13. Sci. Basics Orthop. 373–386 (1987)
58. S.F. Ermakov, V.I. Nikolaev, A.V. Beletzky, L.A. Pashkevich, O.L. Eismont, A computerized system for investigating friction processes in synovial joints. J. Frict. Wear **30**(3), 164–168 (2009)
59. S.F. Ermakov, V.P. Parkalov, V.A. Shardin, Study of screening effect of cholesterol liquid crystalline compounds by pendulum sclerometry. J. Frict. Wear **24**(2), 44–50 (2003)

60. B.I. Kupchinov, S.F. Ermakov, V.P. Parkalov, V.G. Rodnenkov, USSR Invention Certificate 1326902 CCCP, Int. Cl.[4] G 01 H 11/06, no 4012129/24-28; Application 09. 12. 85; Published 30. 07. 87, Bulletin of Inventions and Discoveries no 28, p. 142 (1987)
61. B.I. Kupchinov, S.F. Ermakov, V.P. Parkalov, V.G. Rodnenkov, USSR Invention Certificate 1326903, Int. Cl.[4] G 01 H 11/06, no 4012129/24-28; Application 09. 12. 85; Published 30. 07. 87, Bulletin of Inventions and Discoveries no 28, p. 143 (1987)
62. S.F. Ermakov, B.I. Kupchinov, E.D. Beloenko, USSR Invention Certificate 1630033, Int. Cl.[5] A 61 F 2/76, no 4455100/14; Application 05. 07. 88; Published 22. 10. 90
63. S.F. Ermakov, B.I. Kupchinov, USSR Invention Certificate 1420480, Int. Cl.[4] G 01 N 19/02, no 4195328/25-28; Application 16. 02. 87; Published 30. 08. 88, Bulletin of Inventions and Discoveries no 32, p. 202 (1988)
64. Y.N. Shekhter, V.M. Shkol'nikov, T.I. Bogdanova, V.D. Milovanov, *Preservative Lubricating Materials* (Khimiya, Moscow, 1979), p. 256 (in Russian)
65. D.N. Garkunov, *Triboengineering* (Mashinostroenie, Moscow, 1985), p. 424. (in Russian)
66. G.I. Shor, Effect of additives on volume and surface properties of oils, additives to lubricating oils. Addit. Lubr. Oils 87–104, 1981 (in Russian)
67. A.V. Chichinadze, R.M. Matveevskii, E.D. Braun et al., *Materials in Triboengineering of Non-Stationary Processes* (Nauka, Moscow, 1986), p. 248. (in Russian)

Chapter 4
Modern Concepts of Friction, Wear and Lubrication of Joints

Abstract The results of research devoted to problems of friction, wear and lubrication of living joints were reviewed. The theoretical and experimental results of different authors were analyzed to see that hydrodynamic, elastohydrodynamic, hydrostatic and boundary modes of lubrication can be detected in living joints, and also frictional interaction in cartilages, by the mechanism of rolling friction. Different models of hydrostatic joints lubrication are compared with models of rolling friction and boundary lubrication of joints.

Publications of the recent years contain much information both about the biomechanics of joints and about the results of investigations of specific features of articular lubrication, the role of the synovial fluid and joint cartilages. The problem of the mechanism of lubrication of joints as effective friction pairs remains in the focus of arthrology and tribology [1–6]. It is apparent that the variety and intricacy of the physico-chemical processes evolving in the synovial medium in the joints is also a complex of biomechanical properties inherent to its individual components. They play a vital role in the mechanism of lubrication of joints. These processes and properties of the human and animal joints make them unbelievably effective kinematic pairs that have extremely low friction coefficients and wear during prolonged service life. However, notwithstanding great scientific value of the information accumulated during the studies of friction and wear of synovial joints, the nature of this phenomenon remains a mystery in many respects. To understand the causes of this state-of-the-art of arthrology, let us address the known postulates and ideas about the lubrication of joints.

4.1 General Ideas of Friction in Joints

Efforts to explain the low friction in the joints have a long history. J. Hunter as far back as in 1743 made a presentation to the Royal Society about the synovial fluid that should be treated as a lubricant proper for joints [7]. Later studies in this respect were furthered in the early 20th century only when A. Benninghoff put forward an

S. Ermakov et al., *Liquid Crystals in Biotribology*,
Biological and Medical Physics, Biomedical Engineering,
DOI 10.1007/978-3-319-20349-2_4

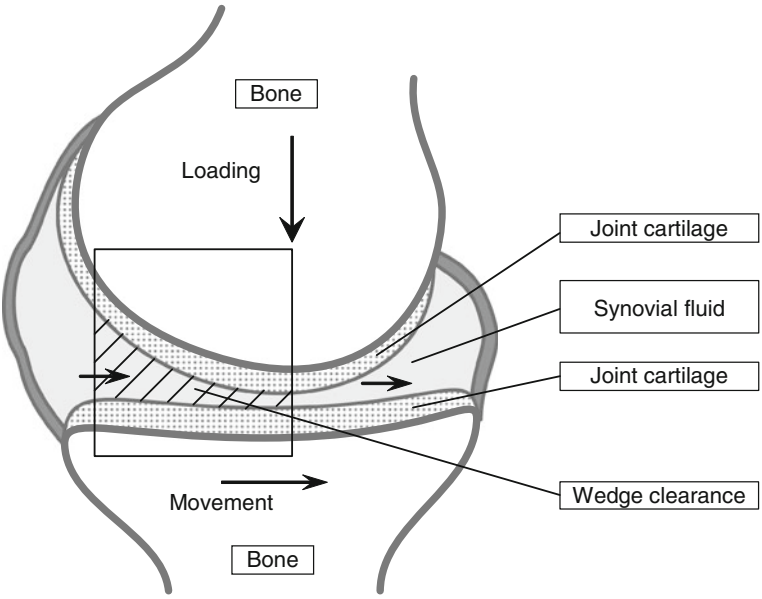


Fig. 4.1 Diagram of hydrodynamic lubrication in joints

assumption in 1929 that the cartilage elasticity might be due to secretion and re-absorption of the synovial fluid [5]. However, this rather important conclusion remained neglected for many years.

A search for most probable explanations of the mechanism of friction in joints in the early 30 s compelled the researchers to resort to the theories of lubrication quite reputable already in engineering. Using the hydrodynamic theory of lubrication of O. Reynolds, M.A. McConail made an assumption in 1932 that the geometry of cartilage surfaces and the viscosity of the synovial fluid produce a hydrodynamic type of lubrication in joints (Fig. 4.1) [7, 8].

However, it became shortly apparent that this hypothesis was unable to explain the extremely small friction coefficients that are observed in real joints. It is because that, on the one hand, the relative sliding velocity of bone surfaces would never exceed several centimeters per second. The calculation of the least thickness of the lubricating film following the theory of Reynolds ignores the viscoelastic properties of rubbing surfaces, on the other hand, and yields an extremely small friction value, i.e. of the order of magnitude of the cartilage roughness [8]. Irrespective these shortcomings, this hypothesis enjoyed enormous import because it was a first effort of application of achievements of other sciences relating to the functioning of joints and pushed consecutive studies in this direction.

McConnell's study truly provoked sizeable interest to the problem of lubrication in joints. T.E. Stanton proposed at that time a gravitation pendulum for experimental assessment of friction in joints [7], because the curve of attenuation enabled to judge both about the friction value and about the pattern of friction in the

specimens of biological and non-biological materials that served as load-carrying components. The linear curve of attenuation of oscillations of the pendulum, i.e. independent of the sliding friction of the load-carrying unit, was considered typical for boundary lubrication, while the exponential attenuation curve, i.e. dependent on the sliding velocity, was considered typical for hydrodynamic or film lubrication of [7, 9].

E.S. Jones was the first in 1934 to measure experimentally the friction coefficient in animal joints [7, 10]. He established that the friction coefficient of knee joints in the lubricating synovial fluid or saline solution was within 0.02, meanwhile the dry friction coefficient increased by an order of magnitude. It was also noted the rubbing cartilage surfaces in the latter case would quickly disintegrate. Later studies confirmed the obtained data convincingly by showing that the friction coefficient in animal joints ranged from 0.005 to 0.02 [9, 10].

Jones some time later, in 1936, continued his studies by experimenting with the pendulum workbench in which an interphalangeal joint served as a load-carrying unit [7, 11]. These experiments revealed that the amplitude of pendulum oscillations was gradually diminishing exponentially; so he concluded that viscous dumping existed in the joint, hence, a fluid film lubricates the latter. However, disregarding these results, Jones recognized the fact of hydrodynamic lubrication of the joints just partly. The circumstance guided him that joints are not always in motion, on the one hand, so no high velocities develop in it, on the other hand. Even at the moment after the onset of very slow movements, the rubbing cartilaginous surfaces produce a rather low friction coefficient. Whence Jones concluded that there is mixed lubrication in the joints, i.e. hydrodynamic or film lubrication in the process of motion and boundary now when motion begins or ends. Nevertheless, he failed to demonstrate that the synovial fluid possessed the necessary lubricity to satisfy common [5, 7].

Almost 20 years later no new theories of lubrication of joints ever appeared. At that time, the synovial fluid and the hyaluronic acid discovered in it, as well as the characteristics of their physico-chemical properties arrested attention of the researchers entirely [5]. Only in the beginning of the 50 s, the researchers resumed studies of the mechanism of lubrication of joints. McConnell was the first to advance a possibility of the hydrodynamic theory to the joints. He began to doubt if the assumption was right that there is a single mechanism that produces an effective system of their lubrication [7]. Nevertheless, he continued to believe that the additional structures, like the meniscus and the fatty body, produce converging wedges of lubrication at slow sliding speeds when joints reciprocate. E. Gardner developed his point of view about how essential were the hydrodynamic principles of lubrication of joints. Like Jones, he admitted that boundary lubrication could exist in certain situations.

In that epoch, it became evident that the hydrodynamic theory of lubrication of Reynolds assuming stiff contacting surfaces has limitations when explaining the features of lubrication of articular surfaces easily deformable in the dynamic contact. Grubin removed these limitations in 1949 when he included elastic deformation of coupled surfaces into his calculations and demonstrated theoretically that

the lubricating had a larger thickness than it had assumed before. Hence, the hydrodynamic lubrication is possible in the conditions that had presumably permitted only the boundary lubrication [5]. In 1959, D. Dowson and G.R. Higginson found a justification for the theory of Grubin and applied the term elastohydrodynamic lubrication to description of tribological systems.

They realized that the elastohydrodynamic lubrication can exist in the transition region from the boundary to the hydrodynamic lubrication and that the thickness of the lubricating film can vary in response to the conditions. Thus, the elastohydrodynamic theory was supplemented with the concept that a lubricated load-carrying system could theoretically functions in case of different friction mechanisms, from boundary (fine filmed) to mixed (film), from elastohydrodynamic to hydrodynamic (full film) in response to the operating conditions (load, speed, temperature), the geometry of rubbing surfaces, the characteristics of and load-carrying ability of tribosystems.

L. Dintenfass and then R.I. Tanner used the main concepts from this paper and endeavored to find a more sound proof of applicability of the elastohydrodynamic theory to explanation of mechanism of lubrication of joints [12]. They approached the problem from the analytical viewpoint and calculated the synovial fluid viscosity and lubricating film thickness using the equation of classic hydrodynamics as the latest elastohydrodynamic theory. They analyzed discrepancy between the calculation and experimental results and formed the idea the elastohydrodynamic model of lubrication of joints is most probable. Moreover, Dintenfass assumed that elastic deformation of the cartilage together with the viscoelastic properties of the synovial fluid increase the contact area of the articular surfaces and maintain thus its larger load-carrying ability. However, irrespective of the importance of the postulates of the elastohydrodynamic theory of lubrication of joints and primarily those relating to the viscoelastic behavior of articular surfaces in the dynamic contact, P. Marnell and R.K. White providing later a proof evidencing that no pure elastohydrodynamic lubrication of joints can ever exist [13]. In other words, the elastohydrodynamic theory of lubrication of joints, like the hydrodynamic theory in its time, was put under doubt and did not gain any confirmation.

Other researchers doubted too that the film lubrication of joints could exist. J. Charnley put under doubt in the late 50 s experimentally obtained the data that Jones obtained about the exponential pattern of attenuation of pendulum oscillations during friction in the interphalangeal joint proving in favor of the film lubrication in the load-carrying unit. Referring to his own experimental results with the pendulum workbench, he demonstrated in particular that the amplitude of pendulum oscillations as a load-carrying unit of the shinbone joint recedes in time fully linearly rather than exponentially. Hence, it evidences the boundary lubrication of the joint in the pendulum workbench that he demonstrated in particular [11]. Charnley tried to explain the exponential results of Jones by the specific features of his experiments in which an intact joint was instead of a sectioned joint block, so the connective tissues influenced the pattern of attenuation of pendulum oscillations.

C.H. Barnett and A.F. Cobbold later studied this problem more exhaustively. Using the shinbone joints of a dog as load-carrying units they demonstrated that the

ligaments and tendons were intact, while the amplitude time characteristics had really an exponential form [11, 14, 15]. When the ligaments and tendons were sectioned, the dependence of the amplitude on time would become linear.

The results fully confirmed the reasoning of Charnley about the effect of the connective tissues of joints on the pattern of attenuation of pendulum oscillations, on the one hand, and quite convincingly prove the assumption that boundary friction can occur in the joints, on the other hand. He believed that it was confirmed by the fact the synovial fluid had the friction coefficient that remained low until the sliding velocity of the rubbing surfaces was very low or nearly zero. Hence, in opposition to the hydrodynamic theory with it main parameter being the viscosity of the synovial fluid, Charney supported the theory of boundary lubrication of joints and believed that another as important parameter was the lubricity or "oiliness" of the synovial fluid alongside with configuration and the quality of rubbing cartilaginous surfaces [7].

In other words, Charney insisted on domination of the boundary lubrication over the hydrodynamic thus assuming that exactly it was the main mechanism providing effective functioning of the joints. Still, he and other researchers admitted that the deformability of articular cartilages plays in all probability a quite essential role in the mechanism of their lubrication together with the fact that this mechanism was most probably intricate or complex [7].

Subsequent experiments with pendulum tribometers confirmed justness of earlier obtained linear amplitude-time dependencies both for animal and human joints. It is shown in [11, 15] that the human hip joint would demonstrate similar regularities as a load-carrying support of the pendulum. Whence it was concluded that the mechanism of boundary lubrication governs the functioning of the human hip joint. Besides, it was also conjured that the cartilage surface layer rich in lipids created this mechanism. Since the studies of dissected joints with the pendulum would invariable yield linear dependences [14, 15], it led to the conclusion that the mechanism of boundary rather than film lubrication was effective in this case.

However, Barnett doubted it in 1962 and showed replaced of the animal joint with a hydrostatic bearing with a liquid film created beforehand would also yield a linear pattern of the ratio between the amplitude of oscillations and time [11, 14, 15]. Moreover, he determined that growth of loading reduces the friction coefficient, as it should have been expected in case of film lubrication. The authors [11] made the same conclusion when they analyzed the equation charactering the motion of the pendulum load-carrying unit as the coulomb one, i.e. as constant typical in the lubrication, on the other hand, and responsive to the viscous resistance determined only by the rheological properties lubrication. Proceeding from their assumptions about the possibility of the lubrication film of regular thickness over the entire contact of the joint head, they calculated a very low viscous resistance. It followed that its effect in the in the load-carrying unit on the exponential pattern of attenuation of pendulum oscillations would be so little manifest that very sensitive instruments are necessary to register it. In other words, it is possible that any effect of viscous forces can remain undetected by the system measuring the amplitude of pendulum oscillations. Therefore, a conclusion was made that the obvious linear type of the

pendulum amplitude-time dependencies should not indicated obligatorily the mechanism boundary of lubrication in the load-carrying unit, specifically if other types of lubrication are probable too [11, 15].

However, oppositely to the preceding calculations, S.A.V. Swanson used the assumption about the possibility of fully film lubrication proved theoretically that to explain the observable characteristics of friction in real joints the synovial fluid viscosity should be 100–1000 times than any other its known value [15]. From this fact, he concluded that the lubrication of joints was predominantly boundary, though admitted some possible share of the film mechanism.

Thus, the majority of authors who used the pendulum friction machine in their experimentx share the view about the mechanism of lubrication of joints believing it predominantly boundary by nature.

Meanwhile, other researchers who used the reciprocating friction machines believed that the major role in the functioning of human and animal joints belonged to the liquid film lubrication [7]. Hence, new theories of lubrication of joints were advanced based on the regularities typical for hydrostatic friction, unlike the hydrodynamic or elastohydrodynamic lubrication. An essential role was attributed to the interstitial fluid that is known to make up a considerable portion (up to 60–80 %) of the intercellular substance of the cartilage matrix. The role in the mechanism of motion of joints remained long a subject of debates [10]. Already in 1904, R. Fick considered that the articular cartilage deformed and its volume thus did not change at all. Hence, the cartilaginous tissue secreted no fluid. However, some time later a different point of view appeared according to which the changes in the cartilage in the process of motion caused alternating compression and decompression that favored transfer of the interstitial fluid. The ability to exude and absorb the fluid was also studied by Beninghoff who investigated the cartilage elasticity in his series of experiments [7]. However, the properties of exudation and absorption were known long ago and nobody but C.W. McCutchen emphasized the fact that these properties might play an essential role in the mechanism of lubrication of synovial joints [5, 7].

4.2 Conceptual Models of Joints Lubrication

McCachen in 1959 used a system in which rubbing materials were cartilage and glass and demonstrated that after each period of fluid the friction coefficient between them would drop significantly. In case absorption was suppressed and its concentration in the cartilage would reach the ultimate minimum, the friction coefficient would grow strongly reaching 0.35. Therefore, he assumed that lubrication of synovial joints resulted from secretion of the interstitial fluid when the cartilaginous tissue was compressed. McCachen used the term “effusion lubrication” (“weeping lubrication”) to describe the system in which the cartilage released the interstitial fluid and created hydrostatic pressure maintaining contact at a microscopic distance and favoring in this way the general mechanism of lubrication of joints [7].

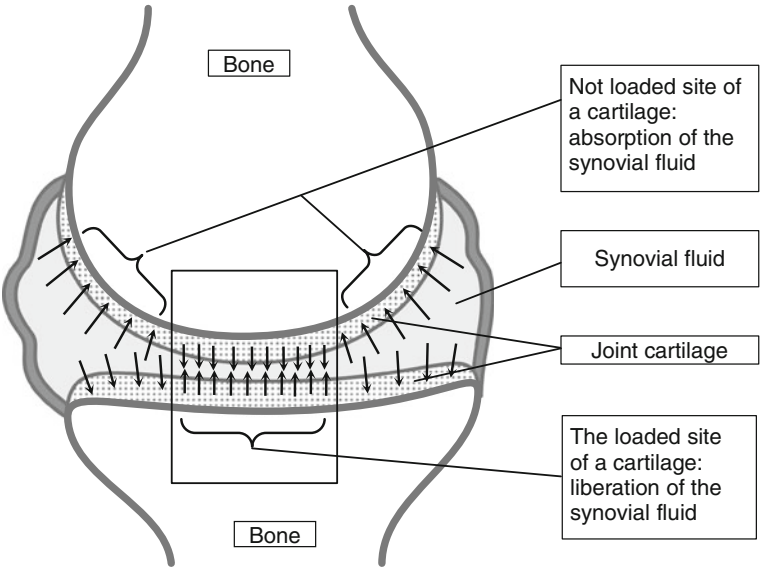


Fig. 4.2 Diagram of mechanism "effusion lubrication" ("weeping lubrication") [7]

It is assumed in the theory of lubrication by effusion that the synovial fluid moves between cartilage surfaces and in cartilaginous bodies proper like a compressible fluid (Fig. 4.2).

In other words, according to [7], when the dynamic contact affects mutually rubbing articular surfaces the contact bears the hydrostatic pressure of the fluid exuded by the cartilaginous tissue, therefore, no friction appears. This interpretation of the theory of lubrication by effusion enabled for the first time to explain soundly the low friction in the joints, in particular when rubbing cartilaginous surfaces ceased to move one in respect of another. However, notwithstanding the positive aspect of this theory, the problem if the cartilaginous tissue would exude enough fluid to produce a liquid film thick enough for the joints to function effectively sprang up to be debated so far.

While admitting the role of the cartilage and synovial fluid in lubrication of joints, new modern structural methods stimulated further interest of researchers to investigate the functional anatomy of articular tissues. Considerable progress of scanning electron microscopy in mid 60 s enabled to penetrate into the ultrastructural characteristics of the articular cartilage and general morphological concepts issued [16]. More detailed studies illustrated that the surface layer of the cartilage matrix differs by its structure and properties remarkably from other subsurface zones. Most of the researchers, in their turn, began to attribute to it an essential role in lubrication of joints [17]. C. Weiss made as important assumption in 1968 that the cartilage surface layer acted as a membrane [18]. Other authors shared this concept then after evaluating the studies of the chemical and morphological characteristics of the articular cartilage. This concept asserts the surface

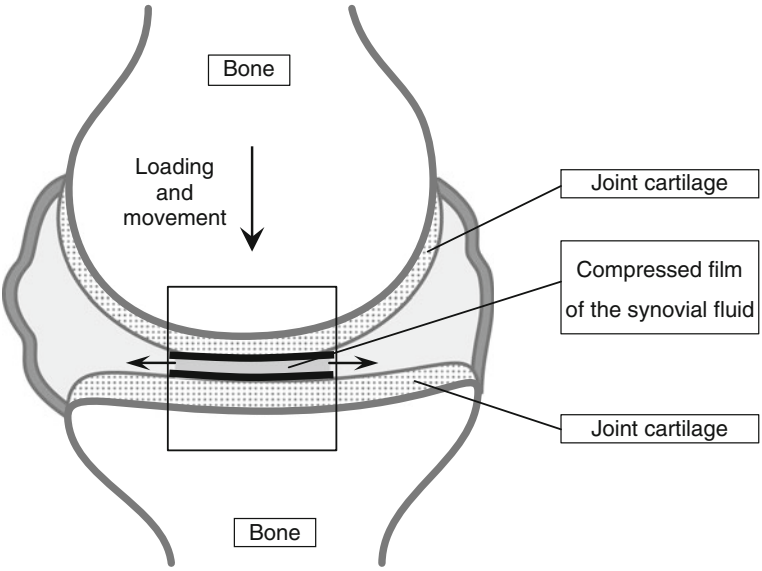


Fig. 4.3 Diagram of mechanism of joint by "compressed liquid film" («squeezed film lubrication») [24]

layer of the cartilage matrix is highly permeable for water, nutritive substances, and some other compounds with low molecular weights [19–21]. It was concluded that migration of fluid between the synovial cavity and articular cartilage was essential in the process of lubrication and nutrition of the joint cartilage.

The data about the morphological, chemical, physical, and mechanical properties of the cartilage and synovial fluid emerged in late 60 s gave rise to new theories of lubrication of joints. The essential role played primarily the observations showing that: 1—the cartilage surface can adsorb the hyaluronic acid from the synovial fluid and produce the required protective layer in the process of functioning of the [22, 23]; 2—the synovial fluid viscosity can change in response to the shear rate and pressure on the articular contact; 3—the compliant cartilage contact can support the compressible liquid film (Fig. 4.3), so that when it approaches the articular surfaces its thickness may reach an order of tens of micrometer so that the rubbing surfaces become almost fully separated.

P.S. Walker with colleagues published in the late 60 s a number of articles. They studied cartilaginous hyaline specimens dried by freezing using the SAM method and established that the synovial fluid was entrapped and arrested between projections on the cartilage surface [25, 26]. They reported that the liquid film thickness ranged from 0.25 to 10 μm [26, 27].

Accordingly, the fluid viscosity was calculated theoretically under the preset experimental conditions and it was established that this parameter in the contact between the synovial fluid and the cartilage had much higher values than inside the fluid. This was also postulated in the experimental friction studies. More profound

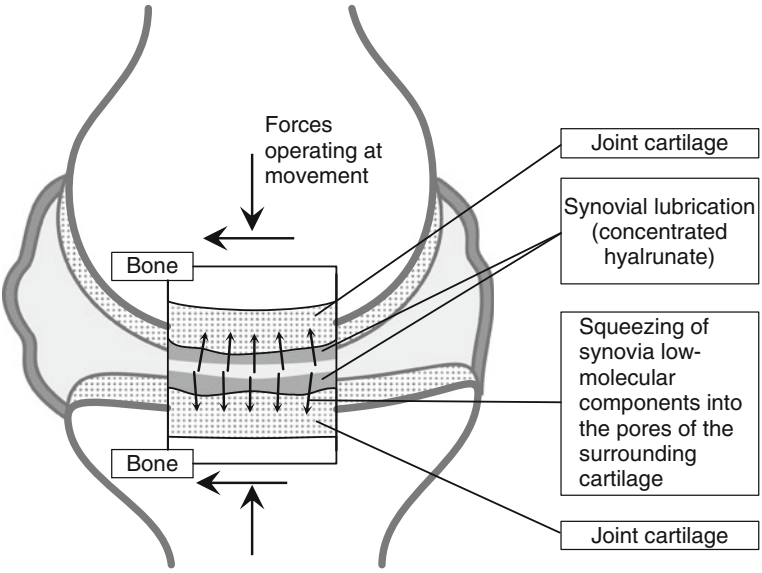


Fig. 4.4 Diagram of mechanism of "boosted joint lubrication" [11]

studies of the effects demonstrated that compression of the lubricating film would intensify during a longer period of time [28]. A conclusion followed from the studies that the synovial fluid was trapped during functioning of the joints, and then it would concentrate or enrich by squeezing of its low-molecular components into the pores of the surrounding cartilage (Fig. 4.4).

Thus, Walker and other researchers proposed a new explanation of the mechanism of lubrication of joints stating that the molecular structure of the synovial fluid, its elasticity, porosity, and characteristics of the articular cartilage contact favor longer existence of separating contact-lubricating films [25, 28]. These lubrication types are known as "reinforced or "boosted" lubrication.

It is believed that, when articular cartilages come into contact, the spaces between surface projections on cartilaginous bodies entrap the synovial fluid, squeeze the low-molecular components through the pores of the matrix of cartilaginous bodies boosting the concentration of high-molecular components and the synovial fluid viscosity in the compressed film, hence, the joints are better able to carry the loading.

Thus, even a cursory comparison of the theory of boosted lubrication and the theory of lubrication by effusion shows that there is a distinct difference between these two postulated type lubrication of joints. It is true that lubrication by effusion requires secretion of the interstitial fluid, boosted lubrication needs just the opposite—fluid absorption by the cartilage when it loaded in the process of motion, i.e. these two types of lubrication of joints seem to exclude one another.

Trying to resolve this contradiction between two types of lubrication, F.C. Linn proposed a new joint cartilage model using the known mechanical and физических

properties and the results obtained with the SAM about the cartilaginous tissue ultrastructure [15]. This model of the synovial type joint includes two porous elastic disks from a composite material under compression that a fluid film separates. The composite material can absorb and release the fluid through its surface providing resistance in elongation or compression, while it retains still substantial non-linearity. If it is assumed that τ—a dimensionless time parameter времени, ρ—the radial measure in the cartilage, Ψ and Φ—pressure in the articular cartilage and pressure in the fluid, then the fluid flow through the joint cartilage surface is proportional to the difference between these pressures, i.e. Ψ–Φ.

According to parametric solution of Linn for this system, the synovial joints demonstrate two types of lubrication in response to period of time (Fig. 4.5). Initially (τ = 0) the fluid flow characterized by the value Ψ–Φ goes from the clearance into the cartilaginous bodies in their middle zone; the external circular region shows a reverse pattern, i.e. cartilaginous bodies are compressed in the initial period, the fluid is absorbed in the central part of the cartilage contact and released through the pores over their periphery. Linn notes that this situation persists during the next period too (τ = 3). However, in the later period (τ = 6), as compression of the cartilaginous bodies accumulates, the fluid flow is fully directed from the cartilaginous bodies into the clearance between them.

According to many works, both the mechanism of boosted lubrication and that of lubrication by effusion are active because the fluid circulates in the contact region in reciprocating directions between the clearance and the cartilage bodies. After some time an asymptotic condition of the self-regulating hydrostatic mode sets and only the mechanism of lubrication by effusion remains active, i.e. the fluid flow proceeds only in direction, namely, the cartilaginous bodies into the intermediate clearance.

Thus, Linn used the obtained data employing model ideas about joints of the synovial type in order to compromise the theories of boosted lubrication and lubrication by effusion in joints.

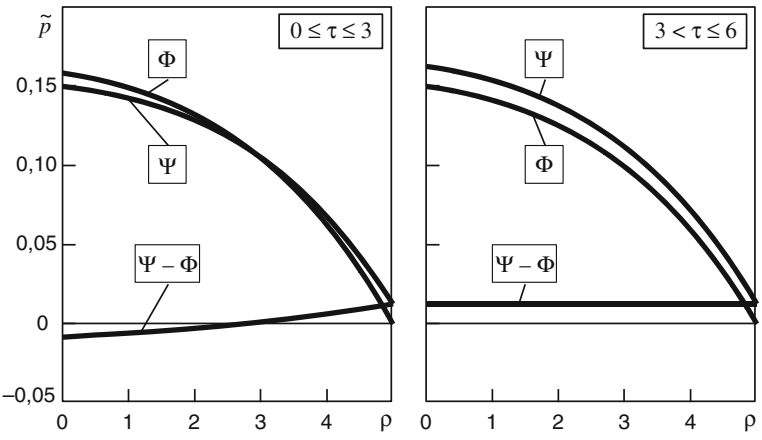


Fig. 4.5 Dimensionless pressure in fluid film Φ and model cartilage Ψ as function of radius ρ

Nevertheless, a number of researchers continued to believe that that the precise manner by which the fluid film appeared between cartilage surfaces, did not conform the hypothesis of McCuchen about lubrication by effusion or the postulate of Walker about the booster mechanism [10, 29]. The following facts evidenced in favor of these statements. On the one hand, a number of publications [22, 30, 31] note that decomposition of the hyaluronic acid was due to the selective effect of hyaluronidase on the synovial fluid so that would lose its thixotropic properties and its viscosity would approach to that of water without influencing the tribological properties of cartilaginous surfaces. That put naturally the main postulates of the theory of boosted lubrication of joints under doubt. On the other hand, the analytical studies of transitional oscillatory processes in dynamic joint motion under the effect of a fixed normal load on the laminar porous, permeable elastic medium demonstrated that McMuchen's forms of lubrication by effusion could be most probable only in the case if the joint did not slide. At the same time, simulation of the process sliding of the synovial joint after the motion stabilized under normal loading in the contact similarly to the laminar porous and elastic system enabled to determine that, when the properties of materials corresponded to those of the healthy cartilage and subcartilage bone substances, the interstitial fluid exudes over the traveling loading front and absorbs in joint cartilage contact in front and behind of the traveling loading rear line [10]. Other authors observed the same flow field too. The assumption ensued that the natural mechanism of lubrication of normal joints was conditioned by the circular fluid flow over the rubbing articular surfaces.

This assumption was based on the model that treated the articular cartilage as a single layer of a homogeneous porous material. The properties of the cartilaginous tissue in fact change considerably with depth. Considering it, J.M. Mansour and V. C. Mow developed the hypothesis further [10, 32] by describing the joint cartilage behavior with a three-layer model of the porous water-saturated permeable bi-phase material. They believe that this three-dimensional model permits to show the frictional behavior of both normal and pathologically changed cartilaginous tissue because after that the biochemical and ultrastructural changes can be taken into account by the parameters of porosity, permeability, and stiffness of a three-layered composite material in the model cartilage.

According to this hypothesis, joints are supposed to have new system of fluid lubrication in which there is a bi-direction fluid flow between the synovial cavity and articular cartilage. This system depends on the mechanical fluid flow in the mated cartilaginous bodies due the pressure gradient in motion caused, according to [32, 33], by consolidation of the cartilaginous tissue. The calculated interstitial fluid flow fields showed that, providing the properties of the material corresponded to the healthy articular cartilage, the latter produces a surface lubricated in the natural way. In case of normal sliding conditions the interstitial fluid moves in the cartilage contact circularly, i.e. the fluid is squeezed in front and underneath the leading edge of the opposite joint cartilage, it is absorbed behind and underneath its tail edge (Fig. 4.6a). The pathological conditions reverse the fluid flow pattern in some portions upsetting the appearance of the self-supplying fluid edge (Fig. 4.6b).

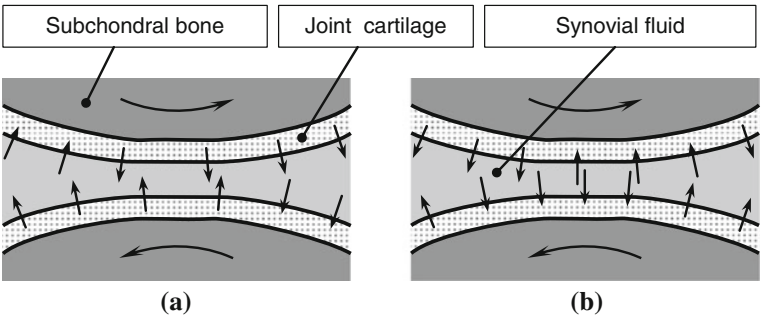
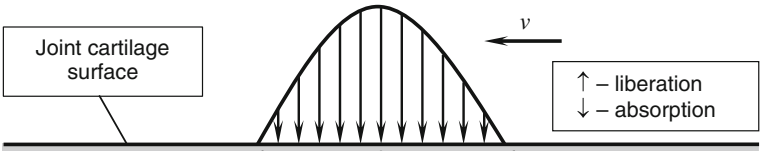


Fig. 4.6 Fluid flow in moving normal articular cartilage (**a**) and in pathologically changed cartilage (**b**) [7]



Case	Consumption of fluid through joint surface				Total consumption
1	0.0080 ↑	0.0492 ↑	0.0178 ↓	0.0197 ↓	0.0198 ↑
2	0.0064 ↑	0.0265 ↑	0.0084 ↓	0.0243 ↓	0.0003 ↑
3	0.0056 ↑	0.0231↑	0.0044 ↓	0.0215 ↓	0.0027 ↑
4	0.0079 ↑	0.0205 ↑	0.0126 ↓	0.0145 ↓	0.0013 ↑
5	0.0097 ↑	0.1046 ↓	0.0411 ↑	0.0545 ↑	0.0006 ↑
6	0.0097 ↑	0.1046 ↓	0.0411 ↑	0.0545 ↑	0.0006 ↑

Fig. 4.7 Consumption of interstitial fluid through joint cartilage surface: *1* normal cartilage, v = 0.0254 m/s; *2* ditto, v = 0.0508 m/s; *3* ditto, v = 0.0762 m/s; *4* 60 % of cartilage porosity, v = 0.0254 m/s; *5, 6* pathologically changed cartilage. v = 0.0254 m/s [10]

The described processes are well illustrated by the results of calculation of the interstitial fluid flow through the joint cartilage surface shown in на Fig. 4.7 [10]. Accordingly, cases *1–4*, corresponding to different loads on the moving contact of the normal joint cartilage the interstitial fluid is released in regions *I* and *II* in a quantity sufficient to supply the lubrication film the zones of contact between the rubbing surfaces. The needed quantity of the fluid absorbed in region *III* is less than the quantity of the released fluid. Hence, the normal articular cartilage seems capable to produce and maintain the fluid film for lubrication.

It is assumed in *4* that only 60 % of the contact porosity remains when the surface is damaged, but no changes of the properties of the surface tissue layer

occur. From the point of view of quality, the fluid flow or the fluid consumption in the cartilage contact do not change considerably.

Meanwhile, in case *5*, when the porosity of the surface layer 60 %, its stiffness changes concurrently. In case *6*, when the modulus of elasticity of the subchondrial bone grows when the properties of the cartilaginous tissue change, the quantity of the interstitial fluid released in region *I* is insufficient to maintain the lubricating film in region *II*. In other words, the pathological conditions determine the apparent mechanism of disorder of natural lubrication in the joint cartilage.

Thus, according to the above concepts the natural mechanism of lubrication of the joint of the synovial type is fully determined by the process of circular flow of the interstitial fluid, the normal healthy cartilage tissue being capable to generate and maintain a liquid film on the rubbing surfaces, meanwhile the pathological tissues preclude it.

It is quite apparent that this concept stands out from those that existed earlier because it offers a more detailed picture of lubrication in joints. However, like many other theoretical works, it fails to reflect fully the processes really evolving in synovial joints. In particular, it describes rather well the properties of the cartilage matrix, its ability to release and absorb the fluid, but it ignores the physical nature of lubrication of rubbing cartilaginous bodies. Meanwhile, numerous studies show that the latter plays as essential role in reducing friction of articular surfaces. It is proven by the experimental work in the first place according to which the level of friction of coefficient of cartilaginous bodies is much lower in the synovial fluid than in any other lubricating medium, e.g., in the buffer saline solution under the same experimental conditions [15].

Nevertheless, more profound studies revealed more other features of friction of articular surfaces in liquid lubricating media. The contribution of the sophisticated tribometric methods of studies was significant. Application of the pendulum apparatus enabled direct measurements of the friction characteristics of the load-carrying unit. It contained a device to apply sudden loads in the initial period of pendulum oscillations. Unlike the preceding methods of measuring the amplitude of oscillations followed by their transformation into the torque, it was revealed that sudden application of loading to the natural hipbone lubricated by the synovial fluid produced an extreme behavior of the friction coefficient in response to the frequency of pendulum oscillations [11]. In other words, when pendulum oscillations grow the friction coefficient of the joint would initially increase reaching a definite value for each specific load and then decline. In case of sudden loading application to a dry loading or pre-loaded joints, both lubricated and unlubricated with the synovial fluid, the friction coefficient would gradually diminish as the number of cycles of pendulum oscillations grows.

It was endeavored in [11] to explain this behavior of the friction coefficient by different conditions of loading application to the joints. Summarizing the results obtained during sudden loading application, they concluded it cases the conditions typical for the effect of hydrostatic lubrication with film compression. In particular, when initially the hip head is immersed into the lubricating fluid and while the film was thinning, friction in the joint grows. After some time interval the moment is

reached when the film thickness would be would be determined by the speed of sliding and load. It might have been expected that the film would disintegrate completely; the friction would rise and end the test. However, it would not occur in the experiments with natural joints. A possible explanation of such behavior was the forces of friction at some moment of the test would equal the forces developing during shear displacement of the cartilaginous tissue concurrently with the pendulum deviation. The authors believed that the fact of the friction coefficient rise evidences it exactly and it was observed in tests of a metal-polymer friction unit of an artificial joint that would commonly have a higher shear modulus than the cartilaginous tissue [34].

It is exhaustively demonstrated in [35] that it is absolutely untrue to assume any relationship between the maximum with transition from external friction to elastic shear of the cartilaginous tissue and that registered forces reduce in this case. These researchers believe that the forces of resistance in the hip joint in the region of the maximum and over the portion of reduction of the dependencies in question are still due to the processes of sliding and to the action of the so-called mechanism of «live friction». According to this mechanism, normal loading levels the macroroughness of rubbing cartilages (Fig. 4.8a, b), the real contact spots appear in the zones of their contour contact formed by two components (Fig. 4.8c, d): the real contact area between the elements of the elastic matrix through the boundary film (the real contact of type *I)* and the real contact area formed by the microstabs of the interstitial fluid (the real contact of type *II)* [37]. They demonstrated that contacting microstabs of the interstitial fluid during the life of contour contact zones could both compress elastically and bear the applied load, while boundary friction takes place in the case of the microfriction bonds of the second type and it accounts for a negligible loading share of loading.

The authors of the idea of the mechanism of live friction believe that the film separating the cartilage, the hip head, and the acetabulum was quite thick in the experiments with hip joints by application of sudden loading, so the friction is fully governed by the resistance of the synovial fluid shear film. In these conditions, the mechanism of live friction cannot exist because the compressible synovial fluid film actually supports the microstabs of the interstitial fluid and their stiffness at this period differs from the modulus of volume compression of the interstitial fluid [35].

It is assumed that a larger number of test cycles when the synovial fluid become thinner approaching to the boundary, the stiffness of the microstabs of the interstitial fluid becomes already corresponding the modulus of volume compression of this fluid and they practically undertake all the load applied to the joint thus creating a liquid contact. The forces of friction stop growing and the friction coefficient goes down. However, according to the results of these authors, it would be more justifiable to indicate apparently that no contact at all would be possible between the compressible film and the microstabs of the interstitial fluid in the first interval because there be no fluid in the micropores of the cartilaginous matrix (Fig. 4.8d). The observed rise of the friction coefficient would still relate to compression of the synovial fluid film and it would finally lead to an entirely boundary contact between the elements of the elastic cartilaginous matrix that, because of the creep of the

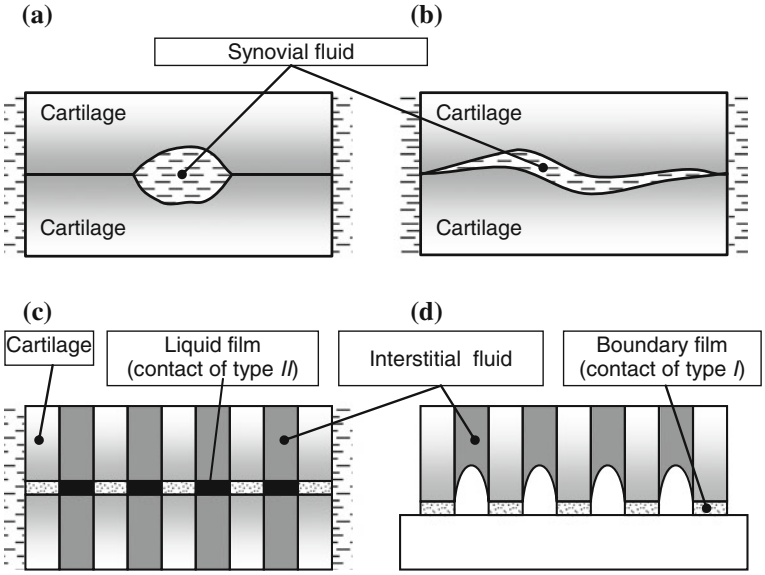


Fig. 4.8 Schematic presentation of idea of mechanism of «live friction»: **a** cartilage-to-cartilage pairs at moment of loading of macroprojections; **b** cartilage-to-cartilage pairs at moment of rest; **c** contour contact spot of cartilage-to-cartilage pairs under heavy loads; d contour contact spot of cartilage-to-smooth surface under light loads [35, 36]

cartilaginous tissue, would come into contact with the microstabs of the interstitial fluid and actuate the mechanism of live friction. However, it would contradict to the main postulate of the mechanism of live friction that asserts that the proportion of boundary lubrication is negligibly small. The latter is apparently the cause why the authors of the mechanism of live friction would rather neglect the effect of the physical nature of lubrication of rubbing cartilaginous bodies, notwithstanding the fact that they obtained different results of lubrication of the cartilage in the synovial fluid and in the saline solution. Meanwhile, it is perfectly apparent that these experimental data are unable to explain the mechanism of live friction, like many other concepts of hydrostatic lubrication of joints.

4.3 Molecular Models of Joint Lubrication

A strong ambiguity of the theories of lubrication of joints compelled researchers to go back to a more substantial analysis of rubbing cartilaginous surfaces and to treat them as mobile dynamic structures rather than inert structureless incapable to interact actively with the synovial fluid components. A substantial role was the discovery of proteinopolysaccharide complexes in the synovial fluid that are giant clusters of hyaluronic acid (HUA) molecules and proteins [1, 38]. Two different

vies existed in recent years explaining formation of these complexes in the synovial fluid [39]. On the one hand, it was assumed that proteins served as a link that enabled HUA adsorption of the cartilage contact; on the other hand, the proteins were believed to serve only to adsorb hydrogen molecules in the HUA disaccharide chains.

It is doubtless still that the protein-polysaccharide complexes had strong negative charges. In the late 60 s, the structure of this shape 100–1000 nm in size was first discovered both in the synovial fluid and in the contacting cartilage after their specimens were specially prepared [25]. However, their nature and the role of the mechanism of friction in joints remained rather obscure for a long time.

It was only in the mid 80 s that both scanning and transmission electron microscope permitted more detailed studies in this direction [40, 41]. It is shown in [41] that pre-treatment with 2 % glutarol aldehyde followed by rinsing in the phosphate buffer and running water covers the rubbing meniscus and the articular cartilage with drop-like, oval structures or globules. Their dimensions (100–2800 μm) and arrangement density strongly depend on the age. The authors believe that appearance of the globular structures on the meniscus and joint cartilage surfaces was due to their properties and their histological structure. Due to the diffusive nature of metabolism of substances, according to the data in [40], the products of metabolism come from these tissues to the surface adjacent to the synovial fluid. These products aggregate and form globules at the interface between these two media with different chemical and physical properties. Apparently, the latter consist of a denser substance than the synovial fluid, and they retain their shape under the forces of surface tensioning. No globules of proteoglycans appear when the studied material is fixed with the ruthenium red. Incubation of the material with pyridine dissolving all the fats and fatty substances does not destroy the structure of the globules. Incubation with hyaluronidase does not destroy the globules either, though it would sometimes make their environment loose. The authors of the experiments reported in [40] advanced an assumption that the globules on the rubbing cartilage surfaces had a protein origin and they play an important role in the friction interaction between the articular cartilages following the mechanism of rolling friction with reciprocating displacement.

This hypothesis was further expanded some time later in [41] in which the SAM results were supplemented with tribological studies both performed with a pendulum tribometer and a reciprocating friction machine. This data demonstrated a tripsin additive would not affect the synovial fluid viscosity, but would impair several times its lubricity. The hyaluronidase additive is known [30] to lead to the enzymic decomposition of the HUA molecules and to reduce strongly the synovial fluid viscosity noticeably affecting the lubrication only after quite a long time. The SAM studies revealed that, in addition to the HUA having a shapeless mass 200–500 μm in size, the studied synovial fluid specimens had spherical particles 20–30 μm in size. The author of these studies that deep dying of the specimens with the ruthenium red and their disappearance after the synovial fluid was treated with the tripsin enzyme, prove that the detected particles consisted primarily of protein. Since the latter had spherical shapes and fermenting treatment of the studied

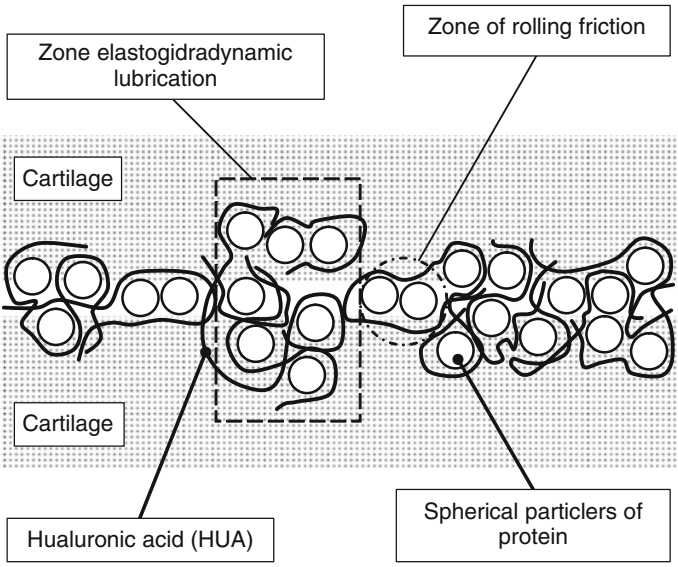


Fig. 4.9 Schematic idea of model of lubrication of joints of rolling friction type [41]

material with tripsin would lead to their full disappearance and impairment of the lubricating ability of the synovial fluid, a model was proposed in [41] of lubrication of joints of rolling friction type (Fig. 4.9).

According to this model, a net of HUA molecules surrounds the spherical protein particles much like a race of a ball bearing, so that the protein particles can rotate freely like rotating elements of a ball bearing.

Nevertheless, the author of these studies, when proposing his own model, did not reject the possibility elastohydrodynamic lubrication of joints. Still, notwithstanding the extensive material presented in [40, 41] in favor of the above hypothesis, it is noteworthy that the problem of formation of spherical structures in the studied specimens remains debatable. It is because aggregation of the synovial fluid components into the particles of spherical configuration can strongly depend on the manner of preparation of the material for electron microscopic studies, and that is pointed out in [25]. Nevertheless, the results of the studies [40, 41] have principal significance because they indicate the essential role of proteins in lubrication of joints.

There is a different view of the lubricating effect of proteins [42]. It is based on the facts of numerous studies [30, 41, 43] showing that the effectiveness of lubrication of cartilages was independent of the synovial fluid viscosity and it coincides the majority of the known theories of boundary lubrication.

On the basis of the experimental data about friction of various surfaces in the synovial fluid and 0.15 M NaCl solution and with the account of the fact that the glycoprotein (LGP-1) that D.A. Swann identified earlier in the synovial fluid, produces a strong lubricating effect [44–46], a model is proposed in [44] of boundary friction of cartilages based on the hygrophilous properties of cartilage

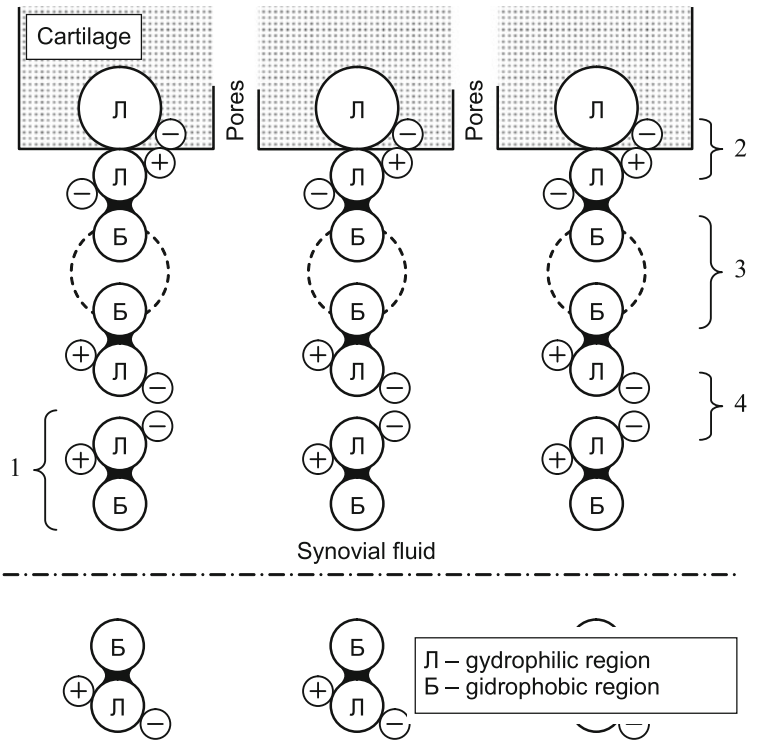


Fig. 4.10 Model of boundary friction of articular cartilages: *1* subunit of lubricating glycoprotein LGP-1; *2* zone of ionic attraction of hydrophilic regions of glycoproteins by hydrophilic regions of cartilage surfaces; *3* zone of attraction hydrophobic regions of glycoproteins (high shear resistance); *4* zone of attraction hydrophilic regions of glycoproteins (low shear resistance) [44]

surfaces and the structuring of overlying boundary layers of molecules of water and glycoproteins (Fig. 4.10). According to this model, like in any system with boundary lubrication, the lubricating substance should have the properties of adhesion to rubbing surfaces that can be created by ion, covalent, hydrophobic-hydrophilic, or hydrophilic-hydrophilic bonds. Because the glycoprotein identified by Swan [44] glycoprotein has hydrophobic and hydrophilic regions, while the radioactive LGP-1 shows the properties of effective adhesion to the cartilage surfaces [7], so the authors of this model believe that such bonds are theoretically possible in the articular cartilage. As a result, according to the data in [42], formation of the boundary layers of glycoproteins and their electrostatic repulsion ensures effectively separation of rubbing cartilaginous surfaces.

However, if it is true, then, from the negative charge of the cartilage surfaces due to the macromolecules of proteoglycans with fixed negatively charged groups, it should have been expected that the lubricating layers of cation surfactant substances would dominate over the anion groups in friction of cartilages. Yet, according to the studies presented in [47], an opposite phenomenon is observed. Hence, low friction

of joints cannot be explained solely by the effect of electrostatic forces. Therefore, when analyzing these processes other mechanism of intermolecular interaction should be also taken into account.

One of the alternative mechanisms can be based on the consideration of the liquid-crystallin condition of the synovial fluid [38, 48–51]. The studies of the recent years are convincingly in favor of this approach. Now a liquid-crystalline organization was identified in many fluids and tissues of living creatures and the fundamental role of this condition in the process of metabolism was demonstrated [52–54]. Many biochemical reactions evolve in the liquid-crystalline phase; the diseases like the sickle-cell disease and atherosclerosis are attributed to changes in the parameters and state of liquid-crystalline structures in cells, tissues, and organs [54]. Therefore, it is natural to suppose that liquid-crystalline compounds in the synovial fluid can play an essential role in the mechanism of friction and lubrication of joints.

According to modern data, the articular lubrication is a selective blood plasma dialysate in which 60–80 % of cholesterol is esterified [1]. The lipid components are sufficiently wee present in the synovial fluid (0.9–11.3 g\l) and articular cartilage [1, 55–57]. There are all reasons to assume that the major portion of the cholesterol compounds in the latter is esters. Nevertheless, the biological role and composition of lipids in the synovial fluid in joints remain obscure.

At present it is known that cholesterol in the bodies of man and animals can transform into bile acids, steroid hormones and витамин D_3 [58]. There are hypotheses about the transport functions of cholesterol for the ATF and its esters for unsaturated fatty acids, and assumptions about the cholesterol energy value under the effect of enzymes [59]. Though some of them have no direct experimental confirmation so far, now it is certain that cholesterol compounds play a biological role in the process of life and activity of various tissues and organs both in case of norm and in case of pathology. This view makes no exception for the human and animal locomotorium.

It is known in particular that higher permeability of the synovial membrane for plasma lipoproteins in rheumatoid arthritis is accompanied by rising concentration of lipids in the synovial fluid, but the concentration of cholesterol usually stays below its level in the blood. The methods of chromatography revealed that major part of part of cholesterol during synovial effusion is in the free state rather than esterified [60]. Some patients with the rheumatoid and such like arthritic conditions showed cholesterol crystals in the synovial fluid. Though it was established that the latter could cause destructive synovitis in animals after intraarticular administration, still they are unstudied as a cause of arthropathia in man [61, 62]. It is proven by the fact of absence of cholesterol crystallization after hemarthrosis, there is no apparent link between this fact and disorders of the lipoprotein composition in the blood plasma, also such formations are extremely rare in case of inflammatory afflictions of joints [62]. There is no proof that cholesterol crystals may form in the synovial effusion of patients after prolonged corticosteroid therapy or in case of impassability of lymphatic vessels.

Though the clinical observations can be hardly helpful in explaining the role of lipids in maintaining the homeostasis in the synovial medium of joints, they clearly

indicate the link between changes in the lipid composition and development of an articular pathology. It is apparent that these changes are local and secondary in respect to the disorder of metabolism of the articular medium. The following facts prove it.

The evidence of intraarticular cholesterol biosynthesis is obtained [63]. The marked cholesterol quite quickly accumulates in the synovial fluid after intraarticular administration of acetate-1^{14}C.

It is established that cholesterol can be "trapped" by the joint in case of inflammatory processes [64]. It is believed that plasmatic lipoproteins, when they degrade in the articular cavity, are capable to release lipids, a part of them recombines with dissolved lipoproteins, the others aggregate and form crystals.

It is discovered that in case of the rheumatoid arthritis, it can intensify the lipoprotein metabolism and produce immune complexes [65]. It is facilitated by intraarticular accumulation of lipids if their evacuation is upset and it prolongs the effect of lysosomal enzymes in the articular effusion liberating cholesterol. Accumulation of lipids in the subintimal layer of the synovial membrane is considered a typical symptom of this disease [66].

Because it is known that normal control of the cholesterol concentration in the body proceeds by partial cholesterol resorption from perspiration through the skin cover, a number of researchers believe that it partly explains the positive effect of mud therapy on articular pathologies [67].

The specific trophicity of the inflammation center and ability to gradual destruction because of changes in the phase state of phosphilipid membranes at the elevated temperature of the medium stimulated the efforts to achieve a targeted and prolonged effect of therapeutical drugs (hydrocortisone, etc.) enclosed into liposomes in the articular tissue in case of rheumatoid and similar arthrites [68, 69].

Thus, though a considerable scope of information has been accumulated so far, the phenomenon of friction in joints remains a riddle in many respects. Along with the variety of hypotheses, aspiring to explain it, there is none relating to the possibility of formation of a liquid-crystalline structure in interphase layers of the cartilage during boundary friction. Both local and foreign publications contain no information about the role of liquid-crystalline cholesterol compounds in the functioning of joints.

Therefore, a hypothesis was advanced [48, 70] that low friction of articular cartilages is due to the liquid-crystalline state of the fluid and possibly due to the esterified cholesterol. This assumption is discussed in more detail in the next chapter.

References

1. V.N. Pavlova, *Synovial Medium in Joints* (Meditsina, Moscow, 1980), p. 296 (in Russian)
2. P. Kumar, M. Oka, J. Toguchida et al., Role of uppermost superficial surface layer of articular cartilage in the lubrication mechanism of joints. J. Anat. **199**, 241–250 (2001)

3. M. Kobayashi, M. Oka, The lubricative function of artificial joint material surfaces by confocal laser scanning microscopy: comparison with natural synovial joint surface. Biomed. Mater. Eng. **13**, 429–437 (2003)
4. B.N.J. Persson, *Sliding Friction: Physical Principles and Applications* (Springer, Berlin, Heidelberg, New York, 2000), p. 515
5. S.F. Ermakov, Biomechanics of synovia in living joints. 1. Modern concepts of living joints friction, wear and lubrication. J. Friction Wear **14**(6), 97–109 (1993) (in Russian)
6. G.D. Jay, Lubricin and surfacing of articular joints. Curr. Opin. Orthop. **15**, 355–359 (2004)
7. R.A. Brand, Joint lubrication. Ch. 13, in *The Scientific Basis of Orthopaedics,* 2nd edn. (1987), pp. 373–386
8. V.C. Mow, A. Ratcliffe, S.L.-Y. Woo (eds.), *Biomechanics of Diarthrodial Joints*, vol. 1. (Springer, New York, Berlin, Heidelberg, London, Paris, Tokyo, Hong Kong, 1990), p. 451
9. A. Unsworth, D. Dowson, V. Wright, Some new evidence on human joint lubrication. Ann. Rheum. Dis. **34**, 277–281 (1975)
10. J.M. Mansour, V.C. Mow, On the natural lubrication of synovial joints: normal and degenerate. Trans. ASME **F99**(2), 163–173 (1977)
11. A. Unsworth, D. Dowson, V. Wright, The frictional behavior of human synovial joints. Part 1. Natural joints. Trans. ASME **97**(3), 369–376 (1975)
12. R.I. Tanner, An alternative mechanism for the lubrication of synovial joints. Phys. Med. Biol. **11**, 119 (1966)
13. P. Marnell, R.K. White, Quantitative analysis of joint lubrication. Wear **61**, 203 (1980)
14. C.H. Barnett, D.V. Davies, M.A. McConaill, *Synovial Joints* (Longmans, London, 1961), p. 304
15. I.C. Clarke, R. Contini, R.M. Kenedi, Friction and wear studies of articular cartilage: a scanning electron microscope study. Trans. ASME **F97**(3), 358–368 (1975)
16. C.R. Flannery, C.R. Hughes, B.C. Schumacher et al., Articular cartilage superficial zone protein (SZP) is homologous to megakaryocyte stimulating factor precursor and is a multifunctional proteoglycan with potential growthpromoting, cytoprotective and lubricating properties in cartilage metabolism. Biochem. Biophys. Res. Commun. **234**, 535–541 (1999)
17. H. Lipshitz, R. Etheredge, M.J. Climcher, In vitro studies of the wear of articular cartilage. The wear characteristics of chemically modified articular cartilage when worn against a highly polished characterized stainless steel surface. J. Biomech. **13**, 423–436 (1980)
18. C. Weiss, L. Rosenberg, A.J. Helfet, An ultrastructural study of normal young adult human articular cartilage. J. Bone Joint Surg. **50A**, 663–664 (1968)
19. A. Maroudas, P. Bullough, S.A.V. Swanson, M.A.R. Freeman, The permeability of articular cartilage. J. Bone Joint Surg. **50B**, 166–167 (1968)
20. H. Muir, P. Bullough, A. Maroudas, The distribution of collagen in human articular cartilage with some of its physiological implications. J. Bone Joint Surg. **52**, 554–563 (1970)
21. F.F. Jaffe, H.J. Mankin, C. Weiss, A. Zarins, Water binding in the articular cartilage of rabbits. J. Bone Joint Surg. **56A**, 1031–1035 (1974)
22. F.K. Linn, Lubrication of joints in animals. Problems of friction ad lubrication. ASME Trans. **2**, 141–153 (1969)
23. S.F. Ermakov, Modern conceptions on biomechanics of human synovial joints. Mech. Compos. Mater. **4**, 539–556 (1992)
24. R.S. Fein, Are synovial joints squeeze-film lubricated. Proc. Inst. Mech. Eng. **181**, 125–129 (1967)
25. P.S. Walker, J. Sikorski, D. Dowson et al., Behavior of synovial fluid on surfaces of articular cartilage: a scanning electron microscope study. Ann. Rheum. Dis. **28**(1), 1–14 (1969)
26. P.S. Walker, A. Unsworth, D. Dowson et al., Mode of aggregation of hyaluronic acid protein complex on the surface of articular cartilage. Ann. Rheum. Dis. **29**, 591–602 (1970)
27. P.S. Walker, B.L. Gold, Comparison of the bearing performance of normal and artificial human joints. Trans. ASME **F95**(3), 333–341 (1973)
28. D. Dowson, A. Unsworth, V. Wright, Analysis of boosted lubrication in human joints. J. Mech. Eng. Sci. **12**, 364–369 (1970)

29. P.A. Torzilli, Mechanical response of articular cartilage to an oscillating load. Mech. Res. Commun. **11**(1), 75–82 (1984)
30. V. Wright, D. Dowson, J. Kerr, The structure of joints. IV. Articular cartilage. Int. Rev. Connect. Tissue Res. **6**, 105–124 (1973)
31. B. Kupchinov, S. Ermakov, V. Rodnenkov, S. Bobrycheva, E. Beloenko, The effect of liquid crystals on joints lubrication. Wear **171**, 7–12 (1994)
32. V.C. Mow, J.M. Mansour, The nonlinear interaction between cartilage deformation and interstitial fluid flow. J. Biomech. **10**(1), 31–39 (1977)
33. P.A. Torzilli, V.C. Mow, On the fundamental fluid transport mechanisms through normal and pathological articular cartilage during function: the formulation. J. Biomech. **9**(8), 541–552 (1976)
34. A. Unsworth, D. Dowson, V. Wright, D. Koshal, The frictional behavior of human synovial joints. 2. Artificial joints. Trans. ASME **97**(3), 377–382 (1975)
35. T.A. Prokhorova, Friction properties of articular cartilages in case of sudden normal loading, in *Deposited with VINITI*, vol. 1081–B. (1986), p. 26 (in Russian)
36. T.A. Prokhorova, The effect of normal pressure and temperature on the friction properties of articular cartilages, in *Deposited with VINITI*, vol. 1080–B. (1986), p. 15 (in Russian)
37. T.A. Prokhorova. O.V. Oganesjan, V.K. Mikhajlov, Problem of the mechanism of low friction of articular cartilages, in *Friction, Wear and Lubricants. Proceedings of International Science Conference* (Moscow, 1985), pp. 15–16 (in Russian)
38. B.I. Kupchinov, S.F. Ermakov, E.D. Beloenko, *Biotribology of Synovial Joints* (Vedy, Minsk, 1997), p. 272 (in Russian)
39. P.C. Seller, D. Dowson, V. Wright, The rheology of synovial fluid. Rheol. Acta **10**, 2–7 (1971)
40. M.N. Pavlova, B.N. Kumanin, The ultrastructure of rubbing surfaces in a joint. Anat. Histol. Embryol. Arch. **8**, 38–42 (1983) (in Russian)
41. H. Chikama, The role of the protein and the hyaluronic acid in the synovial fluid in animal joint lubrication. J. Jpn. Orthop. Ass. **59**(5), 559–572 (1985)
42. W.H.J. Davis, S.L. Lee, L. Sokoloff, A proposed model of boundary lubrication by synovial fluid: structuring of boundary water. Trans. ASME J. Biomech. Eng. **101**(3), 185–192 (1979)
43. B. Kupchinov, S. Ermakov, V. Rodnenkov, S. Bobrysheva, E. Beloenko, Role of liquid crystals in the lubrication of living joints. Smart Mater. Struct. **2**, 7–12 (1993)
44. D.A. Swann, Macromolecules of synovial fluid, in *The Joints and Synovial Fluid*, ed. by L. Sokoloff (Academic Press, New York, 1978), p. 374
45. D.A. Swann, E.L. Radin, M. Nazimiec et al., Role of hyaluronic acid in joint lubrication. Ann. Rheum. Dis. **33**, 318–328 (1974)
46. D.A. Swann, R.B. Hendren, E.L. Radin, The lubricating activity of synovial fluid glycoproteins. Arthritis Rheum. **24**, 22–23 (1981)
47. D. Gvozdanovic, V. Wright, D. Dowson, Formation of lubricating monolayers at the cartilage surface. Ann. Rheum. Dis. **34**, 100–106 (1975)
48. B.I. Kupchinov, V.G. Rodnenkov, S.F. Ermakov et al., The problem of the mechanism of functioning of the joint as a rubbing body. Trans. Belarusian Acad. Sci. **29**(5), 463–465 (1985) (in Russian)
49. S.F. Ermakov, E.D. Beloenko, O.L. Eismont, Role of liquid crystals in tribological behavior of joint cartilages. J. Friction Wear **25**(5), 31–35 (2004)
50. E.D. Beloenko, O.L. Eismont, L.A. Pashkevich, I.A. Chved, S.F. Ermakov, Effectiveness of medication containing bioactive cholesteric-nematic liquid crystals substance in treatment of experimental osteoarthritis. Proc. Nat. Acad. Sci. Belarus: Med. Ser. **1**, 5–8 (2005)
51. E.D. Beloenko, B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov, O.L. Eismont, Liquid-cristal state of joint synovial lubricating medium. Experimental substantiation. Acta Bioeng. Biomech. **3**(1), 24–32 (2001)

52. N.V. Usoltseva, Chemical characteristic, biological and medicinal significance of lyotropic liquid crystals. J. D.I. Chem. Soc. 28(2), 36–45 (1983) (in Russian)
53. S.S. Khalatov, *Holesterol Affliction* (Medgiz, Moscow, 1946), p. 127 (in Russian)
54. G. Brown, D. Walken, in *Liquid Crystals and Biological Structures*, ed. by J.M. Varshavskii (Mir, Moscow, 1982), p. 420
55. N.V. Semenov, *Biochemical Components and Constants of Live Media and Human Tissues* (Meditsina, Moscow, 1971), p. 114 (in Russian)
56. A.V. Sarma, G.L. Powell, M. LeBerge, Phosphoiipid composition of articular cartilage boundary lubricant. J. Orthop. Res. **19**, 671–676 (2001)
57. B.A. Hills, Surface-active phospholipids: a pandora's box of clinical applications. Part II. Barrier and lubricating properties. Int. Med. J. **32**, 242–251 (2002)
58. A.K. Finagin, *Cholesterol Metabolism and Its Regulation* (Vishcha Shkola, Moscow, 1980), p. 168 (in Russian)
59. J.F. Mead, R.B. Alfin-Slater, D.R. Howton, G. Popjak (eds.), *Lipids: Chemistry, Biochemistry and Nutrition* (Plenum Press, New York, London, 1986), p. 486
60. G.C. Bole, Synovial fluid lipids in normal individuals and patients with rheumatoid arthritis. Arthritis Rheum. **5**, 589–601 (1962)
61. J.H. Bland, J.F. Gierthy, E.D. Suhre, Cholesterol in connective tissue of joints. Scand. J. Rheumatol. **3**, 199–203 (1974)
62. J. Zuckner, J. Uddin, G.E. Gantner, R.W. Dorner, Cholesterol crystals in synovial fluid. Ann. Int. Med. **60**, 436–446 (1964)
63. D.S. Newcombe, A.S. Cohen, Chylous synovial effusion in rheumatoid arthritis: clinical and pathogenetic significance. Am. J. Med. **38**, 156–164 (1965)
64. W.H.R. Nye, R. Terry, D.L. Rosenbaum, Two forms of crystalline lipid in «Cholesterol» effusion. Am. J. Clin. Pathol. **49**, 718–728 (1968)
65. G. Noseda, Anti-lipoproteine aütoantikörper mit hypolipidamie bei entzündlichem rheumatismus. Schweiz. Med. Wochensch. **105**, 1–58 (1975)
66. M.S. Rusakova, PJa Muldijarov, A.M. Satabaldiev, Changes of the cover layer of the synovial membrane at an early rheumatoid process stage (the data of light and electron microscopy). Arch. Pathol. **54**(7), 41–47 (1982)
67. W. Naucke, Balneotherapeutische Wirkung von Torfen und Einiger Essentieller Torf-Inhaltsstoffe. Zeitschrift für Baderund Klimaheilkunde **3**, 230–246 (1980)
68. R.K. Ternovoj, Intraarticular corticosteroids therapy: background, approaches to dosing and the problem of therapeutic action duration. Rheumatology **2**, 55–57 (1986) (in Russian)
69. I.M. Gandzha, L.J. Babynina, R.K. Ternovoj, Treatment of rheumatoid synovitis by intraarticular administration of liposome-immobilized hydrocortisone, in *Abstracts of the All-Union Congress of Rheumatologists "Local Therapy of Rheumatic Disorders"* (Moscow, 1988), pp. 6–7 (in Russian)
70. V.A. Belyi, B.I. Kupchinov, V.G. Rodnenkov, S.N. Bobrycheva, S.F. Ermakov, Study of synovial fluid lubricity. J. Friction Wear **5**(6), 16–19 (1984)

Chapter 5
Effect of Liquid Crystals on Biological Mechanisms of Reducing Joint Friction

Abstract The results of studying the friction of joint cartilages lubricated by synovia and lubricants without and with additives of liquid crystal compounds are presented. Based on the experimental data, the structural-mechanical and antifriction properties of joint synovia are compared. It is shown that, having in mind the regular structure of surface layers and deformation behavior of cartilages, the present results confirm the assumption on liquid crystal compounds as a cause of low friction in joints. The best tribological characteristics for the combinations with blood serum are demonstrated by drugs based on chondroitin sulfate, which display characteristics close to natural synovial fluid during frictional interaction of cartilage.

The liquid crystal state seems to be an ideal medium for the occurrence of many biochemical reactions and the provision of the reproduction of biological structures. It is characteristic of cell membranes and organelles, photoreceptors of retina, microfibrils, collagen albumens, myelin nerve fibers, nucleic acids, saliva, and other natural objects [1–4]. The joint synovia also belongs to these objects. Its mesomorphic state may play an important role in reducing joint friction.

5.1 Effect of Friction Surfaces and Lubricant on Joint Cartilage Friction

Proceeding from modern ideas on anatomical, physiological, and biochemical features of members of the human and animal locomotor system including mechanisms of joint lubrication it is important to study the boundary lubrication of cartilages which occurs under heavy loads because collagen fibers are brought together. According to the generally accepted concept, low friction under these conditions should result from the formation of adsorbed films on friction surfaces. The latter are effectively separated by the films acting like boundary films of fatty acids, i.e. surfactants whose molecules are elongated and produce polymolecular adsorbed films normal to the friction surfaces [5–7]. Such structures are known to be characteristic of smectic liquid crystals. Since synovia contains acid components

S. Ermakov et al., *Liquid Crystals in Biotribology*,
Biological and Medical Physics, Biomedical Engineering,
DOI 10.1007/978-3-319-20349-2_5

[8] it can be assumed that low friction in the boundary lubrication of cartilages is due to the mechanism of the action of fatty acids, i.e. the substances capable of forming smectic liquid crystal structures on bearing surfaces.

Results of these studies are of interest for the following reasons. The lubricating behavior of the systems containing fatty acids is known to be governed by features of the structure of lubricating films in the zone of dynamic contact between solids rather than the lubricant rheological behavior [9, 10]. Fatty acids of the biological origin belong to smectic liquid crystals characterized by a low shear resistance because shear in friction is localized in adsorbed lubricant films [10–15]. The use of fatty acids as surfactants to reduce friction is most effective when the nature of the friction surfaces provides chemical reactions between them and the acids [10, 16].

The assumption that synovia demonstrates the lubricating behavior similar to that of fatty acids was verified with a number of model lubricants [2, 17, 18].

The friction of cartilages were studied in pair with the surfaces reactive (copper, steel 45) and inactive (glass) for fatty acids under lubrication with synovia, water, and vaseline oil without and with additives of rapic acid. Synovia was taken from a knee joint of large animals in 2 h *post mortem*. Human synovia was taken if it was necessary to treat an injured knee joint. Cylindrical specimens of joint cartilages were made of medial and lateral menisci of knee joints of large animals. Glass specimens were cut from Petri dishes. Glass was chosen as the counterbody material for its low activity to fatty acids [9, 19] and the gained experience of their use when simulating the friction of joint cartilages [6, 7, 20].

The tests were carried out in an end friction machine (Fig. 5.1) using the pin-on-disc geometry at the sliding velocity of 0.1 m/s under pressures 0.2–10.0 MPa.

The pin was made of cartilages and the disc served as the counterbody. The machine was equipped with the system of friction torque measurement comprising induction gages and excluding the influence of angular and vertical displacements of the torsion device on the accuracy of friction force measurement (Fig. 5.2). The friction force, cartilage specimen deformation, and sliding velocity were recorded automatically.

To improve the statistical significance of experimental data forty preliminary experiments were carried out with fresh synovia samples and cartilage specimens and the dependence of the friction coefficient on the load in the cartilage–glass pair was determined based on the averaged data. It was for reference in the subsequent selection of new cartilage specimens. For these specimens the similar dependence was determined and if the deviation between these and the reference data was statistically insignificant with the reliability of 99 % the specimens were used for tests with other lubricants. The experiments were repeated five times and their results were processed statistically.

Table 5.1 contains the results of the comparative tests of the cartilage–glass pair in different lubricants. The data show that the additive of rapic acid in the amount of 3 wt% into vaseline oil and water does not change the friction coefficient. This is caused by the inactivity of glass to fatty acids. According to the results of [9, 19], no stable adsorbed films of fatty acids appear on the glass surface. Therefore, under these conditions fatty acids are ineffective.

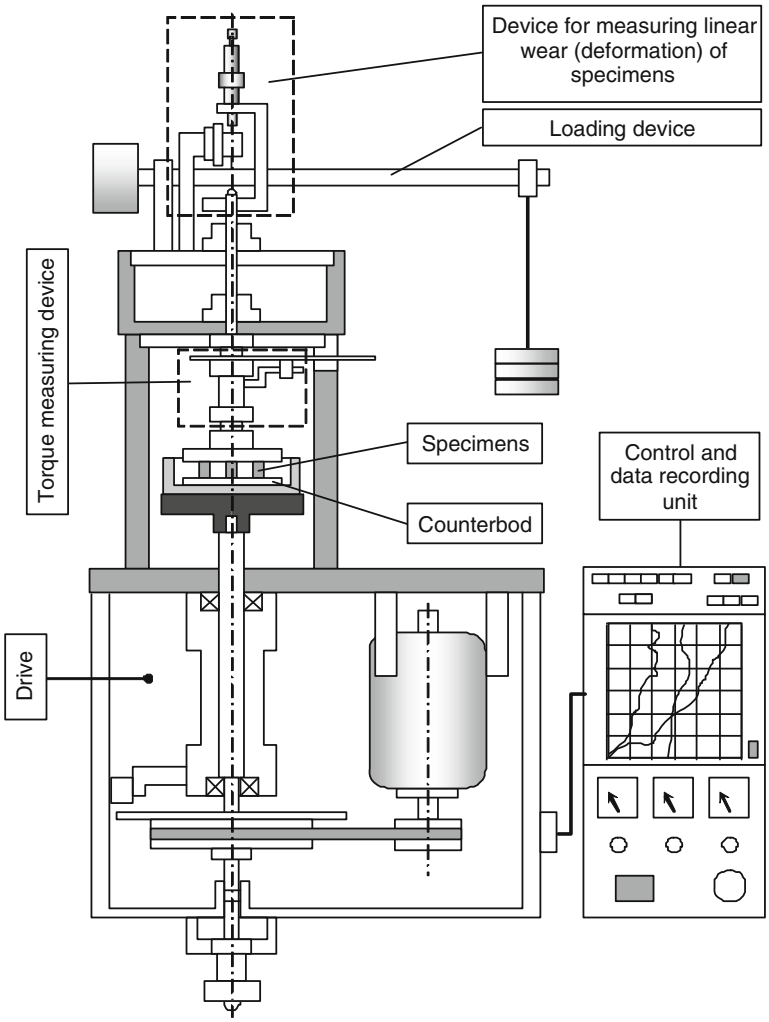


Fig. 5.1 Design of end friction machine

The lubricant can also influence the formation of fatty acid adsorbed films. This influence consists in the concurring ability of molecules of the fluid base (water) in respect to fatty acid molecules [9]. As a result, water molecules react with the glass surface forming very stable silica gel films [10] which exclude the appearance of fatty acid adsorbed films on the surface.

Since in these experiments the introduction of rapic acid into the lubricants did not influence the friction coefficient and one of the friction surfaces was surely free of adsorbed fatty acid films we could suppose that other surface, i.e. cartilage one, showed a weak activity to fatty acids. Otherwise the friction coefficient should decrease because of interaction between fatty acid molecules and the cartilage

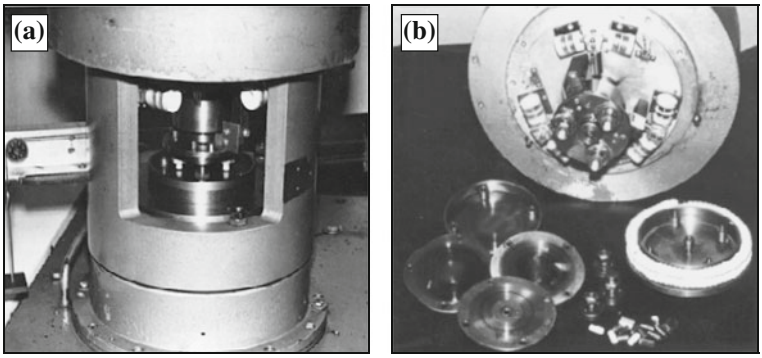


Fig. 5.2 Device for measuring friction force (**a**) and test specimens (**b**)

Table 5.1 Tribological characteristics of cartilage–glass (inactive surface) pair lubricated with synovia and model lubricants

Load (MPa)	Friction coefficient				
	Synovia	Water	Water + 3 wt% of rapic acid	Vaseline oil	Vaseline oil + 3 wt% of rapic acid
0.2	0.015	0.35	0.35	0.05	0.05
0.6	0.009	0.25	0.20	0.04	0.035
1.0	0.007	0.20	0.20	0.03	0.025
1.5	0.008	0.18	0.18	–	–
2.0	0.009	Unstable	Unstable	0.025	0.025
3.0	0.010	Unstable	Unstable	Unstable	Unstable
4.0	0.010	–	–	–	–
5.0	0.012	–	–	–	–
6.0	0.013	–	–	–	–

surface forming stable boundary films on it [19]. This is confirmed by the study results of the friction of the cartilage specimen against active materials such as copper and steel 45 under lubrication with vaseline oil containing rapic acid (Table 5.2). Owing to the nature of these metals their surfaces can react physically and chemically with fatty acids which form adsorbed films of metallic soaps with a low shear resistance. This results in friction coefficient decrease.

Along with this it should be noted that the introduction of rapic acid into water in friction of the cartilage against active surfaces induces more complex processes. In this case the friction coefficient increases 2–3 times compared with cartilage friction in pure water.

However, the results of model experiments with copper–copper and polymer–copper pairs lubricated with water containing rapic acid (Fig. 5.3, curves 1–4) show the rapic acid additive to reduce the friction coefficient. This proves a higher adsorption capacity of rapic acid molecules compared with water molecules that favors the appearance of adsorbed fatty acid films on the surfaces.

Table 5.2 Tribological characteristics of cartilage–metal (active surfaces) pair lubricated with synovia and model lubricants

Friction pair	Load (MPa)	Friction coefficient				
		Synovia	Water	Water + 3 wt% of rapic acid	Vaseline oil	Vaseline oil + 3 wt% of rapic acid
Cartilage–copper	0.2	0.03	0.30	0.75	0.06	0.025
	1.0	0.015	0.10	0.25	0.02	0.01
	2.0	0.01	0.08	0.17	0.015	0.01
	3.0	0.01	0.07	0.13	0.013	0.01
	4.0	0.013	0.05	–	0.01	0.008
	5.0	0.015	0.05	–	0.01	0.009
	6.0	0.02	0.05	–	–	–
Cartilage–steel	0.2	0.045	0.45	1.00	0.08	0.04
	1.0	0.02	0.12	0.50	0.04	0.02
	2.0	0.01	0.07	0.30	0.03	0.01
	3.0	0.01	0.06	0.23	0.025	0.009
	4.0	0.01	0.05	–	0.02	0.008
	5.0	0.015	0.06	–	0.015	0.006
	6.0	0.02	0.05	–	0.015	0.006

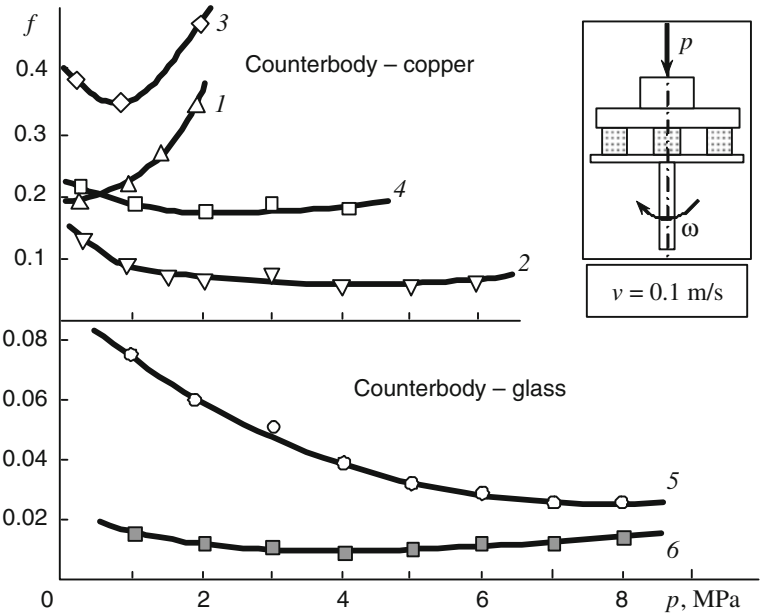


Fig. 5.3 Dependencies of friction coefficient of pairs copper–copper (*1*, *2*), polyamide *6* copper (*3*, *4*), cartilage–glass (*5*), and copper–glass (*6*) on pressure when lubricating with: *1*, *3* distilled water; *2*, *4* distilled water + 3 wt% of rapic acid; *5*, *6* rapic acid

Therefore, when the cartilage rubs against active surfaces it contacts fatty acid lubricating films. Yet, as the results show (Table 5.2) such contact causes the friction coefficient to rise rather than decrease that is confirmed by the data on the friction of the cartilage–glass pair in pure rapic acid (Fig. 5.3, curve 5).

The above regularities can be explained in the following manner.

Since rapic acid molecules have a higher adsorption capacity compared with water molecules fatty acid boundary films are formed on the metal surfaces. The cartilage rubbing against the metal surface contacts the films. The cartilage has a porous structure [8, 21] with pores whose dimensions are comparable with those of rapic acid molecules. Therefore, fatty acid molecules penetrate into the pores and loosen them. This makes the wear of the cartilage surface layers severer and increases the friction coefficient due to the occurrence of the internal Rehbinder effect [18]. This mechanism is proven by the examination of the cartilage surfaces by scanning electron microscopy.

It has been found that in friction of cartilages against active surfaces in water containing rapic acid the structure anisotropy of the cartilage surface layers disappears (Fig. 5.4a, b). According to the data reported in [8, 22, 23] the thickness of the cartilage surface layer showing marked anisotropy is 200–600 μm. Therefore, anisotropy disappearance confirms certainly the catastrophic wear of cartilages in this case. In friction of the cartilage against the inactive surface with the same lubricants surface layer anisotropy does not disappear but wear particles are formed (Figs. 5.4c, d and 5.5a–f).

The friction of the cartilage against glass in water causes flake-shaped particles to appear (Fig. 5.4c, d) and when adding rapic acid spherical particles have been found (Fig. 5.4a–f).

The observed wear of the cartilages in friction is apparently induced by the adsorption-fatigue damage of their surface layers typical for elastic materials [19, 23]. Different shapes of wear particles are explained by different effects of rapic acid and water on the cartilage surfaces. The introduction of rapic acid into water can make the plasticization of the cartilage surface layers severer compared with pure water and the material elongates at single sites (Fig. 5.5a, b). As friction continues, these sites experience repeated alternating deformations favoring the accumulation of damages (Fig. 5.5c, d) and the origination of microcracks followed by their merging and the appearance of spherical wear particles (Fig. 5.5e, f). However, in case of synovia consisting mainly of water (94 %) the friction coefficient for the active and inactive materials almost did not change (Tables 5.1 and 5.2) and no wear particles appeared (Fig. 5.4e, f). These results prove that synovia does not possess the properties typical for fatty acids.

It was believed for a long time that one of mucopolysaccharides, i.e. hyaluronic acid (HA), is the main lubricating component of synovia [8, 24, 25]. It has the structure typical for chiral biopolymers. Water solutions of HA show optical activity [26]. Solutions of such polysaccharides as cellulose and its derivatives whose structure is similar to that of HA have the mesophase [26]. Finally, the lubricating behavior of HA including the ability of changing concentration, flowability etc. proves the relation between the structure and functions of this biopolymer. All this

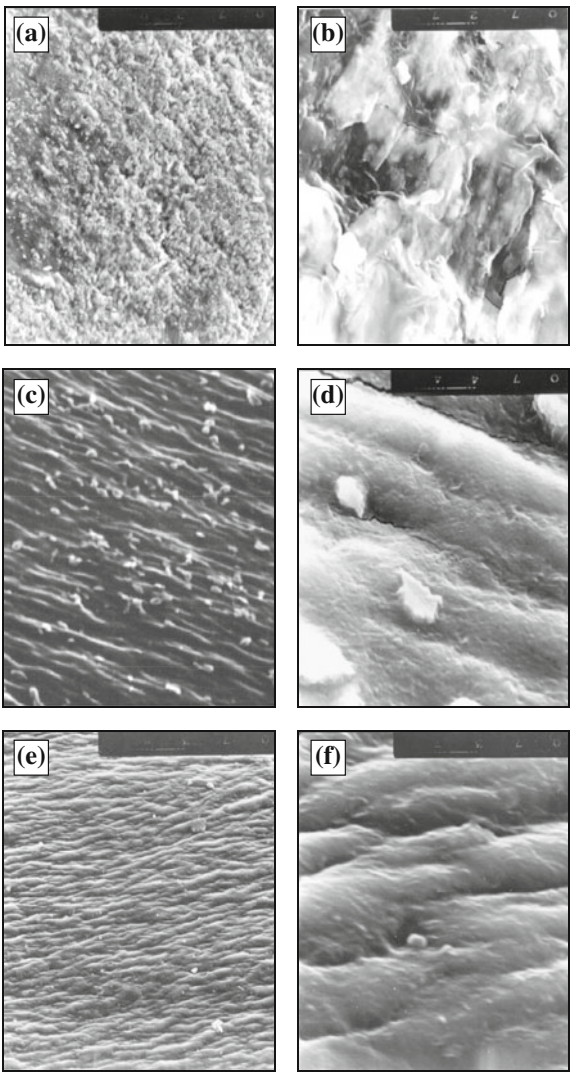


Fig. 5.4 Cartilage surface after friction against copper (**a**, **b**) and glass (**c**–**f**) in presence of: **a**, **b** distilled water + 3 wt% of rapic acid; **c**, **d** distilled water; **e**, **f** synovia. 270 × 330 μm (**a**, **c**, **e**); 32 × 40 μm (**b**, **d**, **f**)

indicates that HA may play a role in producing the mesomorphic state of the joint lubricant. Yet, the fact that the selective enzymatic depolymerization of HA does not affect significantly the lubricity of synovia [8, 27] requires the latter to be studied more thoroughly to find its other components capable of forming lubricating films with a low shear resistance and high load-bearing capacity.

From this viewpoint the ability of liquid crystal molecules to orient molecules of dissolved substances parallel to their major axes is of great interest for understanding the specific features of the lubricating behavior of joint synovia. This effect is called "guest–host" [11]. In paper [20] the influence of low-molecular liquid crystal additives on properties of the chiral polymers similar to HA was studied.

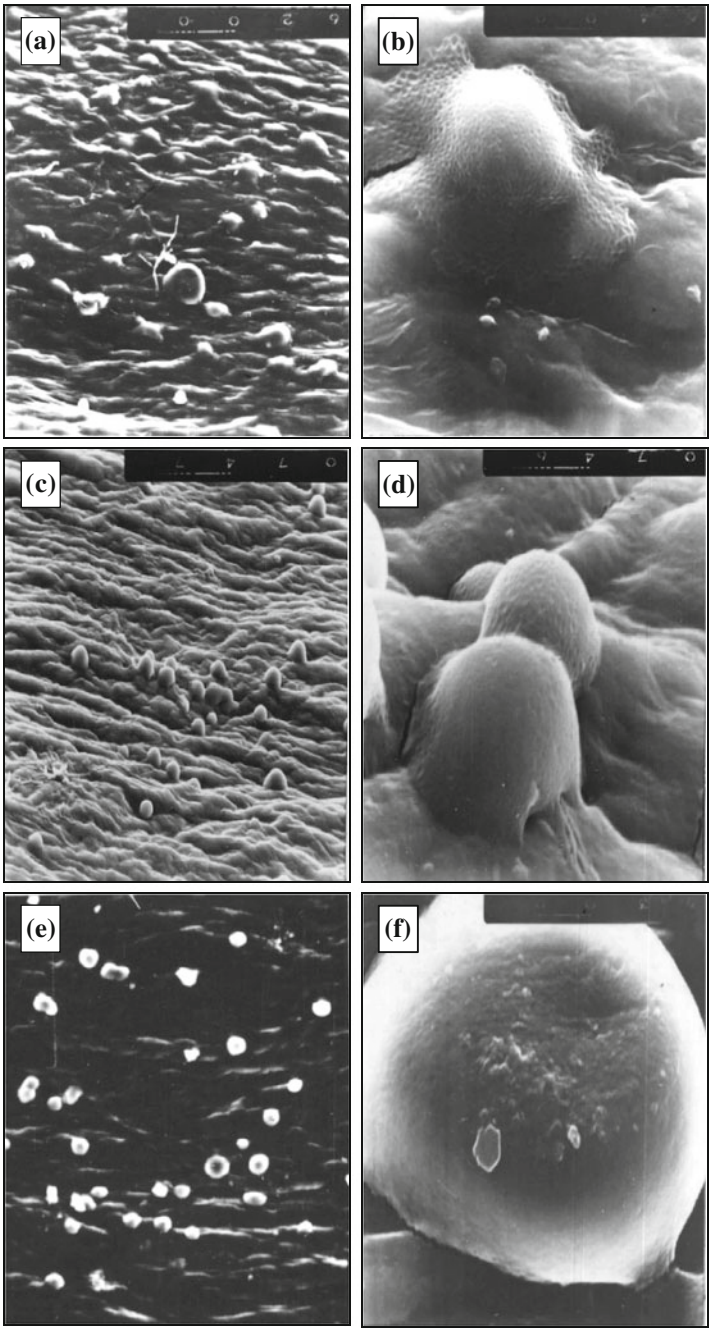


Fig. 5.5 Kinetics of formation of spherical wear particles in friction of cartilage against glass in distilled water + 3 wt% of rapic acid. 270 × 330 μm (**a**, **c**, **e**); 32 × 40 μm (**b**, **d**, **f**)

Among substances capable of producing this effect esters of cholesterol and fatty acids being cholesteric liquid crystals are of primary interest. They belong to steroids and lipids which are widespread in biological structures. Therefore, it is natural to suppose that these compounds are contained in joint synovia and can play a significant role in the formation of its liquid crystal state [28, 29].

5.2 Synovia as Liquid Crystal Biological Fluid

It is known that esters of cholesterol and fatty acids presents in blood plasma in the liquid crystal state [30]. Cholesterol esters prevent arteries against damages due to blood pressure [31]. Since synovia originates from blood plasma they have similar compositions (Table 1.5). Yet, no data on the presence of cholesterol and its liquid crystal compounds in synovia are reported in publications and an experimental attempt to find them was made [29, 32].

The microstructure and chemical analyses of synovia were carried out in 2 h after the samples were taken [33].

The presence of cholesterol and its compounds in synovia was determined by the accelerated Ilka's method based on the Liberman-Byrxard reaction [32]. Dried synovia samples were examined by polarization microscopy within the 20–50 °C range using a warm table. To exclude synovia decomposition the samples were dried by storing them for a long time (over 10 days and nights) at a negative temperature.

Synovia was taken from knee joints of fifteen animals and forty six patients (twenty eight men and eighteen women 11–80 years old) including twenty patients with rheumatic arthritis, eight patients with *osteoarthrosis deformans*, fourteen patients with traumatic synovitis, two patients with psoriatic arthritis, and two patients with infectious gonitis. Experiments were repeated two or three times for each sample.

The experiments have shown that the total content of cholesterol is 1.0–0.2 mmol/l (0.04–0.06 wt%) in animal and 3.8–0.4 mmol/l (0.16–0.20 wt%) in human synovia. The varying concentration of cholesterol in human synovia is apparently due to both the different character of joint pathology and age and individual peculiarities of the patients under study. However, no valid dependence of these changes on the nosologic form of the diseases was found.

The Ilka's method is known to be capable of determining only the total content of cholesterol irrespective of the fact if the latter presents in synovia in the free or bound state. To obtain more accurate lipid composition of synovia the lipid fraction of sixteen synovia samples was examined by thin-film chromatography. Synovia was taken from three patients with rheumatic, four patients with traumatic, and nine patients with pigmented villonodular synovitis [20, 34]. The chromatograms of the synovia of the patients with pigmented villonodular and traumatic synovitis are shown in Fig. 5.6.

The studies have shown that synovia contains cholesterol esters. Free cholesterol precipitated due to the addition of digitonin and it was found that the concentration of the esters in total cholesterol is 25–35 wt%.

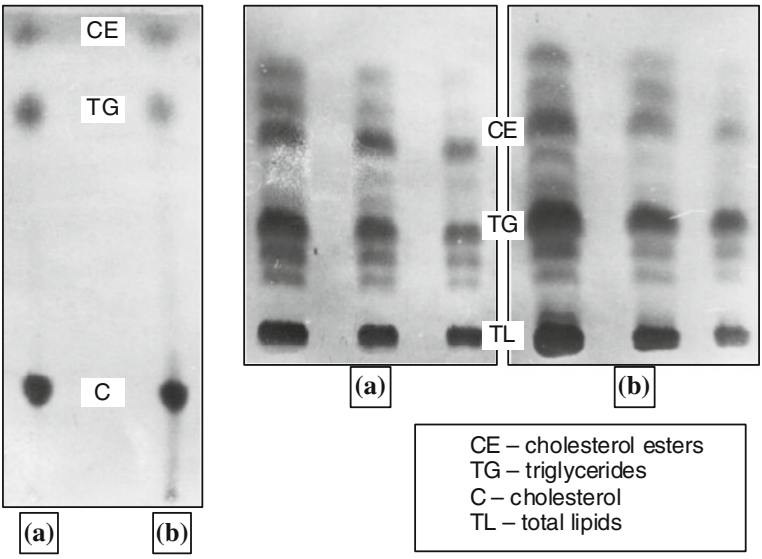


Fig. 5.6 Chromatograms of synovial fluids: **a** pigmented villonodular synovitis; **b** traumatic synovitis

Since cholesterol esters possess optical activity ten randomly taken synovia samples were examined by a polarizing microscope.

The tests have shown that dried human and animal synovia contains optically active inclusions. If the polarizers are crossed the inclusions are seen as luminous points or regions changing their color and the transmitted light intensity when rotating the sample. The thorough study has shown that single luminous points are radial aggregates resembling spherical crystallites found by S.S. Khalatov in blood plasma [35].

When the samples were heated to the temperature of 40–43 °C the inclusions became dark due to the transition of the liquid crystal phase to the isotropic fluid. The subsequent cooling of the samples to similar temperatures resulted to the appearance and retaining of the luminescence at lower temperatures because of the transition of the isotropic liquid back to the liquid crystal phase.

Figure 5.7 shows a luminous inclusion located near the crack edge appeared in the sample in drying. The heating of the sample to 24–26 °C causes the inclusion material to melt and spread over the crack that is accompanied by luminescence near the crack edges. When elevating temperature above 43 °C the luminescence disappears and it appears again when cooling the sample down to 40–43 °C. Subsequent cycles of heating and cooling result in the disappearance and appearance of the luminescence, respectively. Such behavior of the substance showing the luminescence has lead us to a conclusion that it possesses optical activity within the temperature range 25–41 °C and demonstrates features of a thermotropic liquid crystal compound.

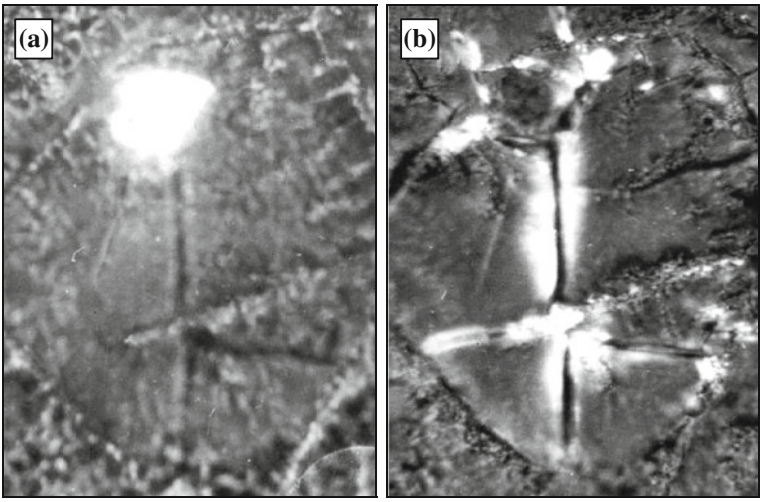


Fig. 5.7 Optically active elements of dried synovia before (**a**) and after (**b**) heating. 240 × 330 μm. Crossed polaroids

Table 5.3 Composition of cholesterol esters identified in synovia of patients with arthritis of various aetiology ($M \pm m$, %)

Cholesterol esters	Molecular weight, conventional units	Synovitis		
		Pigmented villonodular (n = 9)	Rheumatoid (n = 3)	Traumatic (n = 4)
Cholesterol palmitate	624	25.1 ± 1.9	26.2 ± 1.1	24.1 ± 2.3
Cholesterol palmitoleate	622	1.5 ± 0.5	2.1 ± 0.1	3.4 ± 0.9
Cholesterol stearate	636	10.5 ± 2.6	9.1 ± 0.3	8.7 ± 0.8
Cholesterol oleate	650	22.0 ± 0.9	23.6 ± 2.0	25.1 ± 0.7
Cholesterol linoleate	649	32.0 ± 1.2	32.1 ± 1.9	31.6 ± 2.8
Cholesterol arachidonate	672	8.9 ± 0.7	6.9 ± 0.3	7.1 ± 1.9

Therefore, the presence of cholesterol and its ester forms as well as a substance being mesomorphic at the human organism temperature in synovia prove that these compounds are cholesterol esters.

However, the methods used are incapable of identifying the compounds. The total fraction of cholesterol esters extracted from synovia by the chloroform-ethanol blend was examined by chromatography. The analysis has shown that synovia contains esters of cholesterol and the following acids (Table 5.3): palmitic (21.8–27.3 %), palmitic-rapic (1.0–4.3 %), stearic (7.9–13.1 %), rapic (21.1–25.8 %), linoleic (28.8–34.4 %), and arachidonic (5.2–9.5 %). As the data of Table 5.3 show, cholesterol esters of rapic, palmitic, and linoleic acids prevail in the composition of the total fraction of cholesterol esters.

So, it has been shown that the natural joint lubricant contains the mixture of cholesterol esters which are known to be thermotropic liquid crystal compounds possessing the mesophase within a broad temperature range and having the cholesteric structure. Therefore, it is naturally to assume that a certain qualitative and quantitative combination of the esters produces a liquid crystal mixture with the mesophase within the physiological temperature range. Hence, one can expect that the joint lubricant itself possesses the liquid crystal state at the same temperatures.

The above assumption was verified experimentally [29, 36]. The occurrence of corresponding thermodynamic transitions in such systems is the most reliable criterion of belonging them to liquid crystal systems [28, 37]. For this reason the differential scanning calorimetry and probe fluorescence were used to estimate the temperature ranges of the mesophase of natural synovia and lipids extracted from it. The samples were taken from ten patients including five patients with rheumatic arthritis, two patients with pigmented villonodular synovitis, and three patients with traumatic synovitis.

The lipid fraction of synovia was extracted by the modified technique of Folch et al. [38]. Liposomes were obtained from the lipid suspension on an ultrasonic generator.

8-aniline-1-naphthyl sulfonate (ANS) served as the fluorescent probe. It is an amphiphilic molecule adsorbed well by lipid bi-layers with the orientation of charged sulfate units in the polar group plane. Since the quantum yield of ANS fluorescence depends on the surround polarity and can vary from 0.004 in water to 1.0 in non-polar media and the binding of ANS leads to the shift of the fluorescence maximum wavelength the probe fluorescence method is not only structure sensitive but also provides data on the molecular motion of some components of the liotropic biological fluid. In these experiments the 1×10^{-3} M water solution of ANS with the fluorescence maximum at 520 nm (the excitation wavelength was 360 nm) was used. The studies have shown that if ANS was added to synovia or liposomes (the final probe concentration was 1×10^{-5} M) its fluorescence maximum was registered at 485–490 nm (the excitation wavelength was 360 nm). This proves that ANS molecules are in the non-polar surround. The ANS fluorescence intensity reached its maximum at the following composition of the samples: 150–400 μl of synovia + 3.0 ml of the isotonic NaCl solution + 1×10^{-5} M of the ANS solution. The fluorescent analysis was performed on Jobin-Yvon and Fica instruments.

Figure 5.8 shows the results of the differential thermal analysis of synovia samples taken from patients with rheumatic arthritis (curve 1), traumatic (curve 2) and pigmented villonodular (curve 3) synovitis. The main phase transition in these biological systems occurs at 63–64 °C, the second transition—at 70–73 °C, and the third transition—at 86–88 °C. The transition temperatures vary within the 2–3 °C range depending on the nosologic form of joint pathology that is apparently due to the different content of water, cholesterol and other lipid components in the synovia samples under study.

This assumption is confirmed by the following facts. In addition to hydrocarbon groups not reacting with water (lipophilic) lipid molecules contain one or more polar hydrophilic groups easily reacting with water. Such molecules are called amphiphilic to emphasize their dual nature. Owing to this dual nature and the

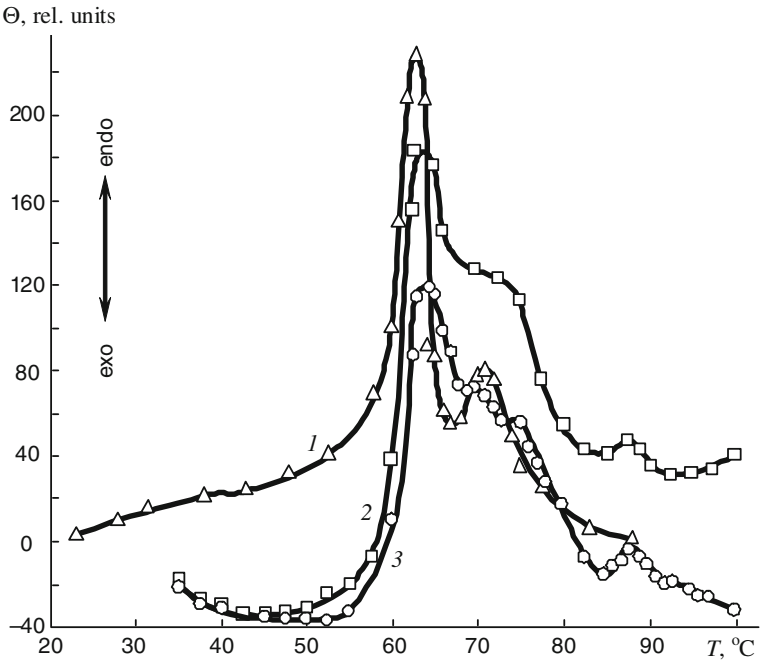


Fig. 5.8 Thermograms of synovial fluids of patients with rheumatoid (*1*) traumatic (*2*), and pigmented villonodular (*3*) synovitis

selective interactions of two and more types of molecules the lyotropic mesomorphic phase may appear [1, 2]. Water solutions of cholesterol and its compounds in the presence of phospholipids are the examples of these systems [1, 38].

Lyotropic liquid crystal systems, like thermotropic ones, are characterized by a very high degree of molecular structure organization. They are very sensitive to changes in the temperature and concentration, e.g. water, cholesterol etc. [1, 2]. Variations in these characteristics induce phase transitions in the lyotropic systems accompanied by structure disordering. The high temperature of the transition in natural synovia (Fig. 5.8) allows one to assume that at these temperatures (63–65 and 70–73 °C) the phase transition of synovia components like lyotropic liquid crystal—isotropic fluid occurs. The third endothermic peak at 86–88 °C is apparently caused by the melting of desoxyribose nucleic acid which may appear in synovia due to cell disintegration.

To determine thermal transitions within the physiologic temperature range lipids of synovia were studied. Lipids extracted from synovia taken from a patient with pigmented villonodular synovitis were heated and the endothermic peak at 28.8 °C was found (Fig. 5.9).

Since the reversible endothermic process occurred at that temperature it can be assumed that the latter is the temperature of the phase transition of the lipids from the crystal to mesomorphic state.

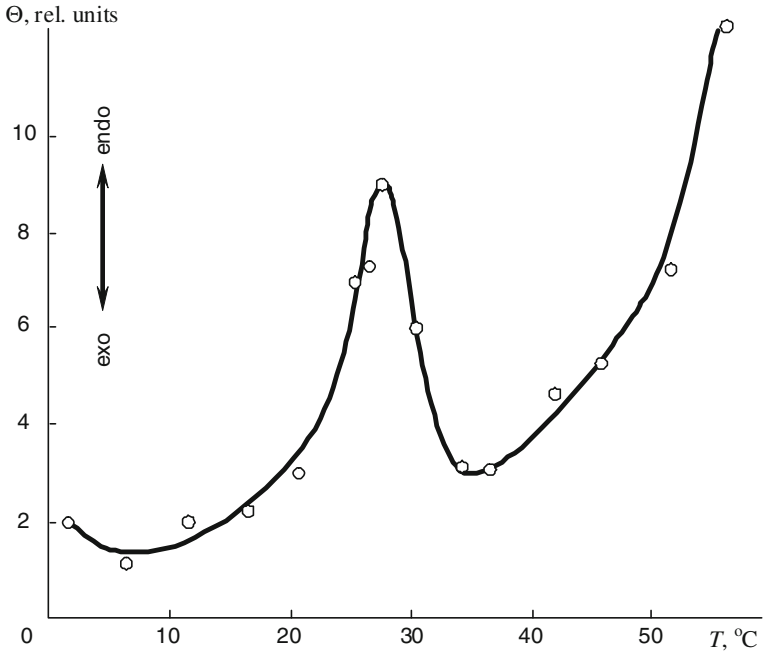


Fig. 5.9 Thermogram of lipids extracted from synovia of patient with pigmented villonodular synovitis

The comparison of the results obtained show that synovia components may have phase transitions within the range of physiological or higher temperatures that indicates the possibility of their liquid crystal state. It is also confirmed by the study results of synovia within a broad temperature range by the ANS fluorescence probe. The experiments have shown (Fig. 5.10) that the total temperature dependence of the intensity of the fluorescence of the ANS related to components of natural synovia has two inflections, i.e. at 31 and at 59 °C.

Figure 5.11 exemplifies individual temperature dependencies of the intensity of the fluorescence of the ANS related to components of natural synovia taken from patients with rheumatic arthritis and traumatic synovitis.

The temperature dependence curves may show individual differences depending on the nosologic form of disease. The temperatures at which the inflections appear on the curves may wary by 2–3 °C (29–32 °C) within the physiological temperature range while at higher temperatures no variations occur and the ranges of temperature transitions for different diseases coincide exactly.

Than it was supposed that the inflections on the temperature dependencies of the natural synovia fluorescence intensity determined by using ANS result from the physical state, i.e. phase transitions, of lipids and cholesterol esters. To confirm this supposition, fluorescence characteristics of lipids and cholesterol fractions extracted from synovia were studied.

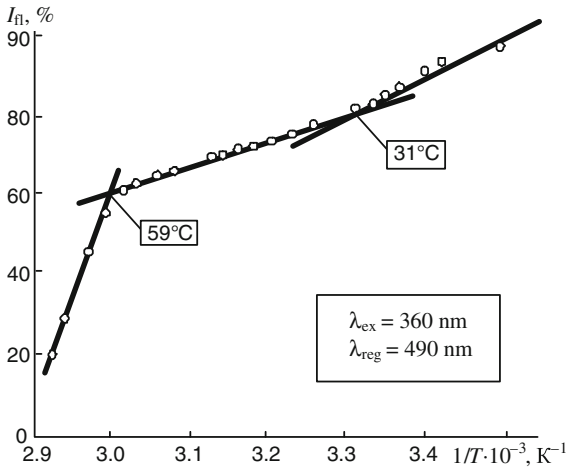


Fig. 5.10 Dependence of intensity of fluorescence of ANS bound to components of patient synovia on reverse temperature. The fluorescence intensity typical for tested samples at 20 °C is taken as 100 %

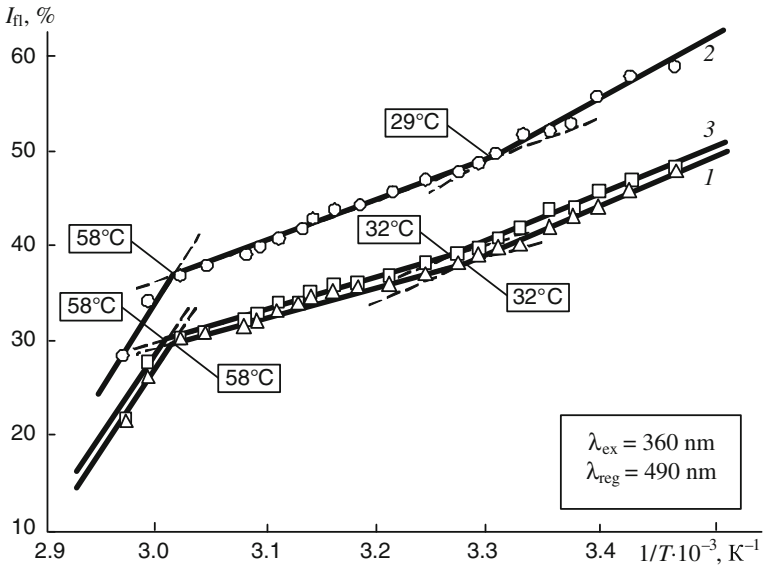


Fig. 5.11 Dependence of intensity of fluorescence of ANS bound to components of synovia of patients with rheumatoid (*1*, *3*) and traumatic (*2*) synovitis on reverse temperature

Figures 5.12 and 5.13 illustrate the temperature dependencies of the intensity of the fluorescence of the ANS related to the liposomes obtained from lipids and total cholesterol fractions.

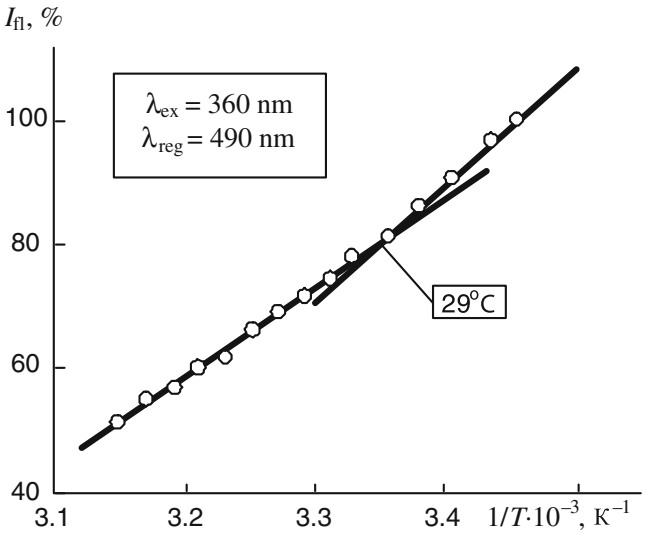


Fig. 5.12 Dependence of intensity of fluorescence of ANS bound to liposomes of total lipid fraction of synovia of patients with rheumatoid synovitis on reverse temperature (results of five experiments). The fluorescence intensity typical for tested samples at 20 °C is taken as 100 %

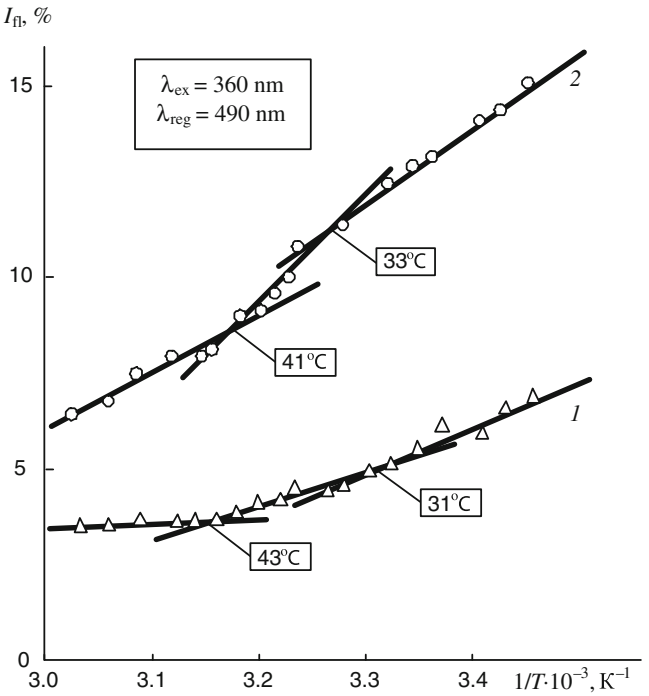


Fig. 5.13 Dependence of intensity of fluorescence of ANS bound to total cholesterol fraction of synovia of patients with rheumatoid (*1*) and pigmented villonodular (*2*) synovitis on reverse temperature

The analysis of the experimental data shows that the temperatures of the first phase transition determined for the lipid fraction of synovia by the differential scanning calorimetry (Fig. 5.9) are the same as the temperatures of the inflections on the temperature dependencies of the intensity of the fluorescence of the ANS related to both natural synovia (Fig. 5.11) and liposomes obtained from synovia lipids (Fig. 5.12) as well as to total cholesterol fractions (Fig. 5.13). This proves that the transitions result from changes in the physical state of one and the same biological substrates from the crystal to liquid crystal state.

Slight temperature variations observed for the specimens under study depending on the nosologic form of the disease can be explained by the fact that synovia is a dynamical system containing, in addition to water, phospholipids, and cholesterol, other components. The concentration of the latter can change due to the disturbance of metabolism in diseases, hence, it can influence the temperature transitions of the biological system as a whole.

At the same time two phase transitions occur within the physiological temperature range on the temperature dependence of the fluorescence of the ANS bound to the total cholesterol extracted from synovia. The first transition appears at 31–33 °C and the second one at 41–43 °C (Fig. 5.13). The first transition is typical for both lipids and total cholesterol fraction while the second one only for the latter fraction.

This fact can be explained only by phase transitions of the cholesterol compounds. Therefore, within the mentioned temperature range (41–43 °C) these compounds among them, according to the above chromatography data, cholesterol esters prevail undergo the phase transition similar to the liquid crystal—isotropic fluid transition. This conclusion is confirmed by the comparison of the obtained experimental data with the results of thermopolarization microscopic studies described above.

Thus, the combination of the results obtained by three independent methods (Table 5.4) proves that synovia has phase transitions at both physiological and higher temperatures. It should be noted that the temperatures determined for lipids

Table 5.4 Temperature ranges of changes in physical state of natural synovia and components extracted from it

Study techniques	Phase transition	Temperature ranges of phase transitions of synovia and its components, °C			
		Native synovia	Liposomes from lipids extracted from synovia	Lipids extracted from synovia	Total cholesterol fraction extracted from synovia
Thermopolarization microscopy	I II	25 41	–	–	–
Scanning differential microcalorimetry	I III IV V	62–64 70–73 86–88	–	28–29	–
Probe fluorescence of ANS	I II III	29–32 58–59	29	–	31–33 41–43

of synovia by the differential scanning calorimetry practically coincide with the temperatures of the inflections on temperature dependencies of the fluorescence intensity of the ANS bound to both natural synovia and its total cholesterol fraction. The variability of the temperatures registered for these samples by different methods is within 2–3 °C. Therefore, the close temperatures determined for the synovia samples by the independent methods are due to changes in the structure state of one and the same biological objects.

5.3 Interrelation Between Structure-Mechanical and Antifriction Properties of Joint Synovia

The assumption that liquid-crystal cholesterol compounds are capable of reducing the friction of mated cartilages was verified experimentally using a pendulum tribometer most suitable for this application since it implements the locomotions typical for joints of living organisms (Fig. 5.14) [23, 39].

A humeral joint is the most convenient test object for the pendulum tribometer from the anatomical viewpoint. Specimens for tribotests were made of the head of the humerus (the convex specimen) and the glenoid cavity (the concave specimen) of animals (pigs) in 1–2 h *post mortem* (Fig. 5.15). At least three specimens were prepared for each test and the cartilage was not separated from the bone.

The head of the humerus was fixed to the pendulum and mated to the glenoid cavity mounted on a rigid base. The pair was loaded by changing the pendulum mass using calibrated weighs.

To make test results comparable in all experiments the cartilage contact area was provided constant that was checked by the transfer of a water-soluble dye from the previously dyed surface of one cartilage to the non-dyed surface of another one. The friction force was determined by the attenuation time of pendulum oscillations and by the slope of the time dependence of the oscillation amplitude. The initial amplitude was the same for all experiments. The attenuation time and its standard error were found for each lubricant [23].

Liquid crystal cholesterol compounds with the mesophase within the temperature range including the temperature of living organisms were studied.

Prior to the tests 1 cm^3 of a lubricant was injected into the joint under study. The following lubricants were tested: distilled water, organosilicon fluid ПМС-300 with the viscosity of 300 cs. used when treating knee joints of patients [40], pseudo-synovia being the water solution of Na-carboxymethylcellulose prepared according to patent [41], organosilicon fluid and pseudo-synovia with 2 wt% of a liquid crystal, and synovia with the cholesterol content of 0.9 and 4.09 mmol/l.

Figure 5.16 shows the kinetic dependencies of the oscillation amplitude of the tribometer pendulum when lubricating the joint under testing by the above lubricants.

The analysis of the data presented in Fig. 5.16 shows that the additive of the liquid crystal to pseudo-synovia yields the same friction force as synovia with the minimal content of total cholesterol (curves 5, 6).

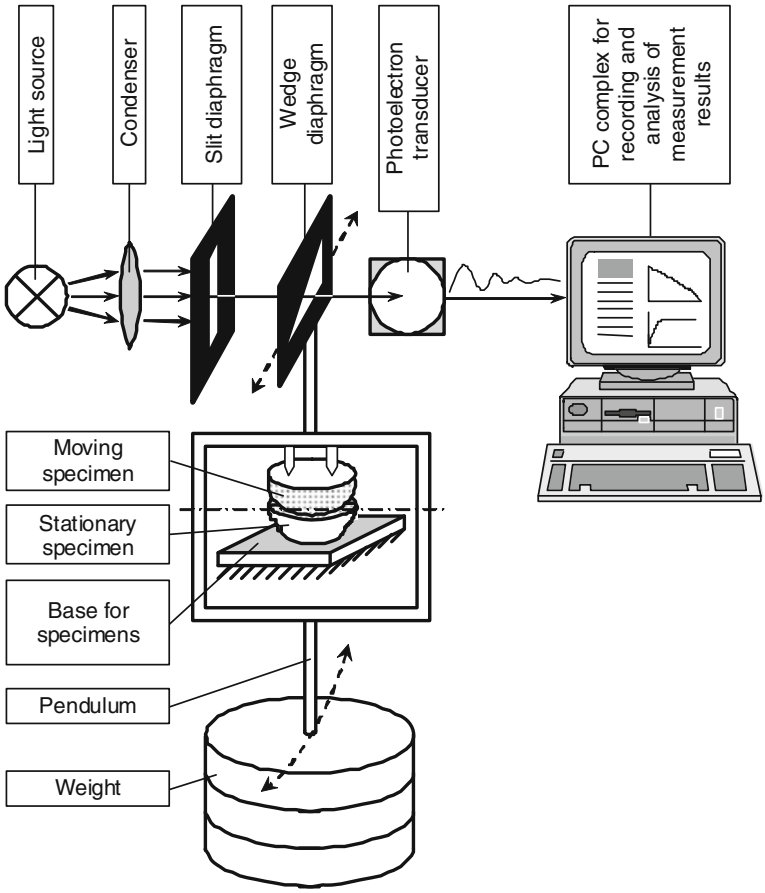


Fig. 5.14 Block diagram of pendulum tribometer

For synovia with a higher cholesterol content the attenuation time of pendulum oscillations is longer (curve 7).

The friction of the cartilages also decreases significantly in case of organosilicon fluid with liquid crystals (curve 4).

Thus, the studies carried out have shown that the ability of joint synovia to provide the low friction of the mated cartilages is attributed to the presence of cholesterol compounds in it. This conclusion is also confirmed by the comparative study of the lubricity of synovia and pseudo-synovia with the cartilage–glass pair at the sliding velocity of 0.1 m/s performed on the end friction machine [42].

Figure 5.17 shows the load dependence of the friction coefficient of the cartilage–glass pair lubricated with pseudo-synovia (1), synovia (2), and pseudo-synovia with 2 wt% of liquid crystals (3). The dependencies were plotted by the results of the statistical processing of data of forty tests which were carried out directly after synovia was punctured from animal knee joints.

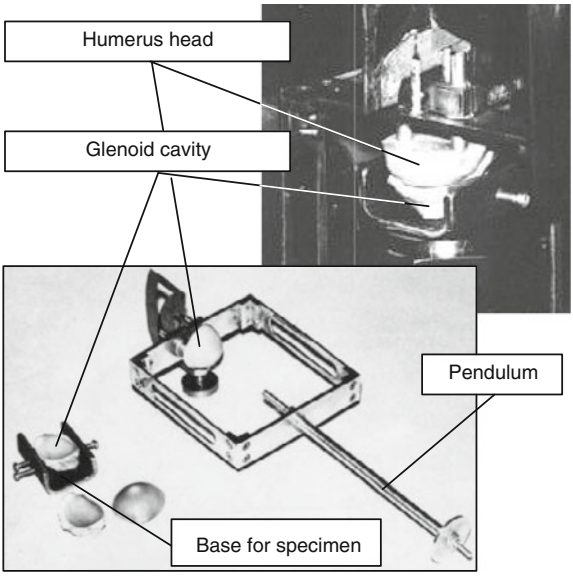


Fig. 5.15 General view of friction unit of pendulum tribometer and test specimens

The analysis of the results shows that the transition of cholesterol compounds into the liquid crystal state on cartilage surfaces is apparently one of the most likely processes responsible for a low friction of the joint cartilages under these conditions. Under the constant friction regimes and lubricant the replacement of the cartilage specimens by new ones in the cartilage–glass pair reduces the friction coefficient (Fig. 5.18). This effect is observed in lubrication with synovia, vaseline oil, and organosilicon fluid. Therefore, it can be assumed that such cartilage replacement increases the concentration of liquid crystal compounds in the contact zone because of their secretion from the porous cartilage structure under the effect of the load.

A higher content of the liquid crystal compounds in the contact zone reduces the friction coefficient of the cartilage–glass pair.

The examination of friction tracks on glass by the method of polarization-optical microscopy has shown the presence of optically active inclusions (Fig. 5.19).

The study results obtained demonstrate that the good antifriction behavior of cartilages lubricated with synovia results from the presence of cholesterol compounds in the latter possessing liquid crystal properties within the temperature range including the living organism temperature.

However, it is obvious that these studies are incomplete without taking into account relation between the morphological and mechanical properties of the joint cartilage and its antifriction behavior in different lubricants. A few attempts in this field have already been made. For example, the comprehensive understanding of the joint synovia morphology allowed the authors of [43, 44] to assume that the

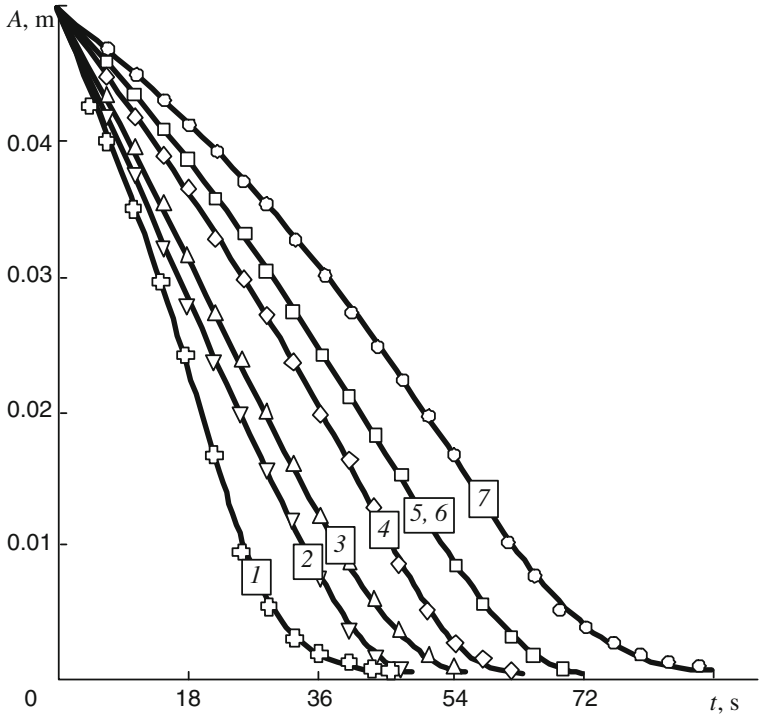


Fig. 5.16 Amplitude of tribometer pendulum oscillations when lubricating joint under testing with: *1* distilled water; *2* organosilicon fluid; *3* pseudo-synovia; *4* organosilicon fluid with 2 wt% of liquid crystals; *5* pseudo-synovia with 2 wt% of liquid crystals; *6* synovia with cholesterol content 0.9 mmol/l; *7* synovia with cholesterol content 4.09 mmol/l

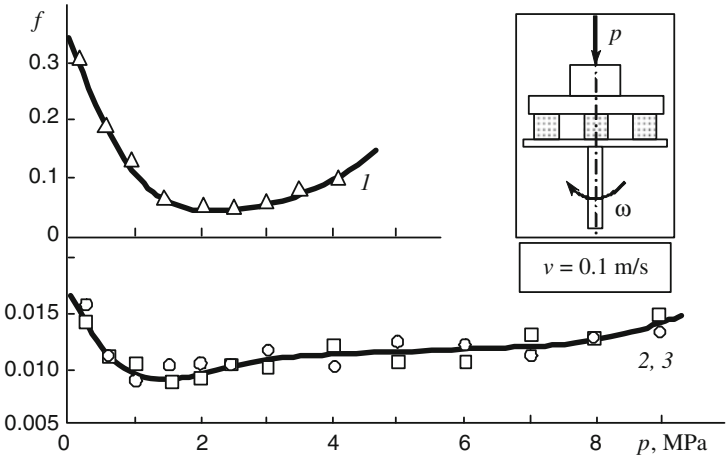


Fig. 5.17 Dependence of friction coefficient on load for cartilage–glass pair lubricated with: *1* pseudo-synovia; *2* synovia (*square*); *3* pseudo-synovia with 2 wt% of liquid crystals (*circle*)

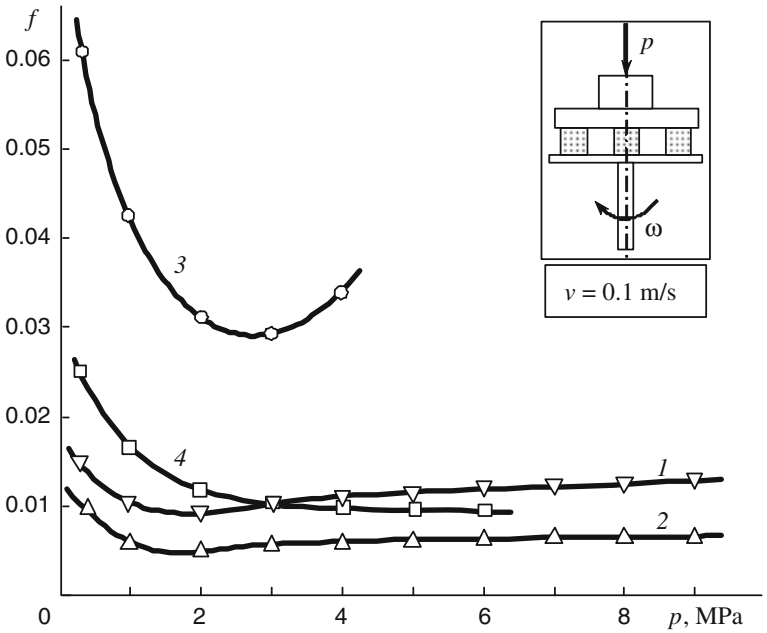


Fig. 5.18 Dependence of friction coefficient on load for cartilage–glass pair lubricated with: *1* synovia; *2* synovia with cartilage replacement repeated three times; *3* vaseline oil; *4* vaseline oil with cartilage replacement repeated three times

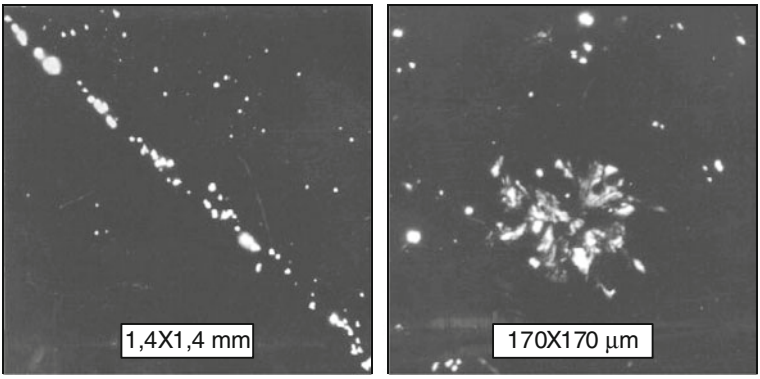


Fig. 5.19 Optically active elements on glass surface after friction against cartilage in vaseline oil

concepts of the elastohydrodynamic lubrication theory can be used to elucidate mechanisms of joint friction and lubrication. According to these concepts, the mechanism of joint lubrication may be a specific mode of the elastohydrodynamic effect governed by the processes of sliding or the squeezing of the synovia film out of the matrix between collagen fibers in the zone of cartilage contact. This protects the cartilages in case of severe loading or restricted motion.

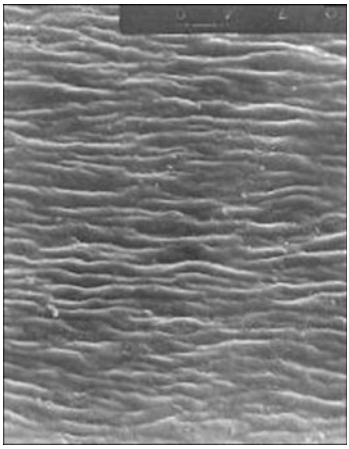


Fig. 5.20 Anisotropy of cartilage surface microrelief. Orientation of microgrooves coincides with direction of prevailing locomotions in joint. 300 × 375 μm

The regular relief of rubbing surfaces is known to influence considerably the elastohydrodynamic effect [45]. The best tribological behavior occurs when microgrooves are perpendicular to the sliding direction while the tribological characteristics are poor with their longitudinal orientation and intermediate with the isotropic microrelief.

Since the elastohydrodynamic effect is governed substantially by the orientation of microrelief grooves relatively to the sliding direction the effect of the regular relief of the cartilage surface on its friction in synovia was studied.

The SEM examination of the cartilage surfaces has shown that bundles of collagen fibers and microgrooves between them are parallel to the direction of preliminary locomotions occurring in the joint (Fig. 5.20) [46].

The study of the friction of cartilages with the collagen fiber orientation parallel and perpendicular to the sliding direction in pair with glass has shown that the regular structure of the cartilage surface layer influences greatly the friction coefficient of the cartilage–glass pair lubricated with synovia. The friction coefficient is minimal if sliding runs along collagen fibers (Fig. 5.21). It increases if the fibers are perpendicular to the sliding direction [42, 47, 48]. The results obtained together with the data reported in [45] prove an insignificant role of the elastohydrodynamic effect in friction of cartilages lubricated with synovia.

However, the above effect of the regular structure of cartilage surfaces on their friction can be explained proceeding from the properties of the liquid crystal compounds contained in synovia [32, 36]. According to the data of [49], there is a close correlation between the regular microrelief of solid surfaces and the formation of planar-oriented liquid crystal films.

The orientation of major axes of molecules of liquid crystal cholesterol compounds along microrelief grooves corresponds to the absolute minimum of the molecule elastic energy and is characterized by the thermodynamically stable, i.e. more energetically profitable, state. When the molecules are perpendicular to the grooves they are bent and possess a higher elastic energy. Moreover, the adhesion of

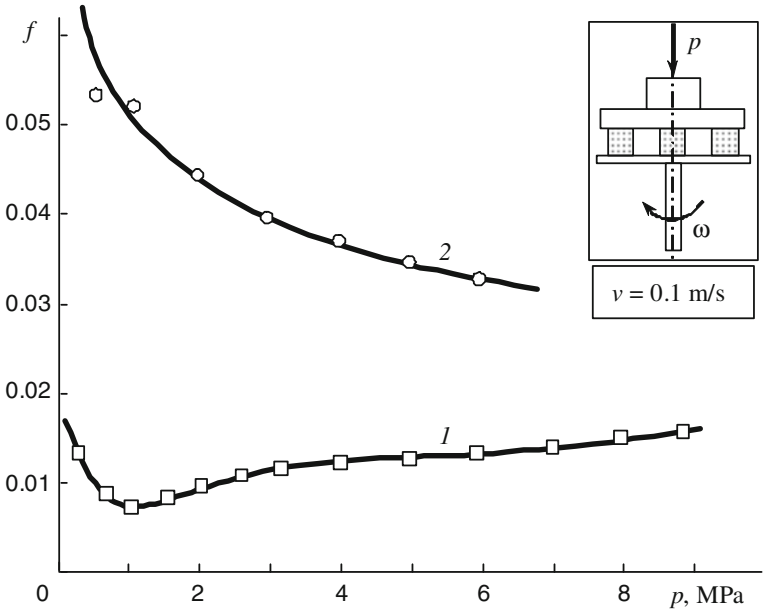


Fig. 5.21 Dependence of friction coefficient on load for cartilage–glass pair lubricated with synovia for sliding of cartilage along (*1*) and across (*2*) direction of prevailing locomotions in joint

liquid crystal molecules to the surface depends on the microrelief characteristics and, as it is shown in [29], is maximal when the groove width is comparable with the molecule size (1–10 nm), i.e. when the microrelief appears as submicroroughness.

The recent data on the macro- and microroughness of the cartilage surface [19, 50] describe incompletely the surface microrelief since they were obtained by the stylus methods with a poor resolution.

To describe the cartilage surface roughness in more detail let us use the analysis results of the biosynthesis of collagen structures (see Paragraph 1.1). In addition to the macro- and microroughness, the cartilage surface has the submicroroughness caused by the microfibrillar (3–5 nm) and fibrillar (20–100 nm) structure of collagen fibers [8]. The strict ordering of the micro- and macromorphology of the cartilage matrix provides the regular structure of the cartilage surface at the submicrolevel. Such topography of the cartilage surface favors the strong adhesion of liquid crystal molecules oriented parallel to surface microrelief grooves and the formation of planar-oriented liquid crystal films on the surface [29, 49]. The structure of the films is similar to that of solid lubricants such as graphite and molybdenum disulfide, yet it possesses a much weaker intermolecular interaction due to its liquid crystal state. As a result, shear localizes in inter-layer regions of the liquid crystal structure and a weak interaction of the layers apparently provides a low friction force in the joint that is confirmed experimentally. The lubricating behavior of model lubricants with liquid crystal cholesterol compounds similar to that of natural synovia also shows that the above explanation is reasonable.

Thus, there is a correlation between the microstructure of the cartilage surface and the antifriction behavior of the cartilage–synovia system. The quantitative and qualitative characteristics of this correlation are apparently dependant on the cartilage deformation properties since the cartilage–synovia system is known to be labile, i.e. capable of responding quickly and precisely to changes in the joint including load and velocity variations. Yet, we failed to find papers describing the deformation properties of cartilages under variable loads in relation to their antifriction characteristics. Results of these studies will allow researchers to improve their understanding of the lability of joint cartilages in friction.

To determine these regularities we compared the friction of the cartilage against glass in synovia, pseudo-synovia with and without the additive of liquid crystal cholesterol compounds, and organosilicon fluid ПМС-300 used to lubricate joints [23]. The tribotests involved the stepwise loading of the cartilage from 0.2 to 6.0 MPa followed by quick unloading down to 0.2 MPa [46].

The experimental results are presented in Table 5.5. The cartilage deformation in loading does not depend on the composition and antifriction properties of the lubricant that agrees well with the data of paper [51].

Similarly to the experiments described elsewhere [52] at each loading stage initially an instantaneous elastic deformation occurs followed by creep. This is explained by the properties of collagen at the initial stage and by the secretion of the liquid phase out of it at the creep stage.

The time dependencies of the cartilage deformation and antifriction properties in unloading depend on the lubricant tested (Table 5.6).

It has been found that for synovia and pseudo-synovia with liquid crystal cholesterol compounds when unloading the specimens the friction coefficient rises initially then after 18–24 min (1.08–1.44 ks) it decreases to the initial value.

Practically same values and the variation pattern of the friction coefficient in unloading prove that the labilities of the cartilage lubricated with synovia and pseudo-synovia with liquid crystal cholesterol compounds are similar. At the same time in friction of cartilages against glass with lubricants without liquid crystal compounds no lability is found. For example, in friction of the cartilage lubricated with organosilicon fluid the friction coefficient increases 1.5–2 times compared to the initial value. The cartilage size recovers gradually to 90–110 μm in synovia and pseudo-synovia with liquid crystals and only to 10–20 μm in organosilicon fluid.

The results obtained allow us to conclude that under varying loads the structure-tribological properties of the cartilage in the synovia recover in the same manner as in pseudo-synovia with liquid crystal cholesterol compounds that proves a significant role of the latter in the mechanism of cartilage lability.

Thus, the found relationship between the structural-mechanical and antifriction properties of synovia gives a new explanation of the friction mechanism of joints and the pathogenesis of their destruction in collagen tissue diseases. The experimental data can form a real basis for developing artificial synovial fluids for the medical correction of disturbed tribological characteristics of synovia in joint pathology.

Table 5.5 Dependence of friction coefficient and compressive deformation of cartilage on pressure

p (MPa)	Synovia		Pseudo-synovia + liquid crystal cholesterol compounds		Organosilicon fluid	
	f (10^{-3})	Δl (μm)	f (10^{-3})	Δl (μm)	f (10^{-3})	Δl (μm)
0.2*	14	0	15	0	85	0
0.6	9	240	9	240	40	250
1.0	7	424	8	450	30	470
2.0	9	634	10	630	20	650
4.0	10	888	10	890	20	820
6.0	13	1117	13	1125	20	1100

Note *Deformation of cartilage (Δl) at p = 0.2 MPa was taken as reference point

Table 5.6 Dependence of friction coefficient and deformation of cartilage on time in unloading

t (ks)	Synovia		Pseudo-synovia + liquid crystal cholesterol compounds		Organosilicon fluid	
	f (10^{-3})	Δl (μm)	f (10^{-3})	Δl (μm)	f (10^{-3})	Δl (μm)
0	45	680	50	680	140	680
0.36	25	652	25	665	150	670
0.72	20	638	20	650	140	668
1.08	18	625	20	628	130	667
1.44	17	611	17	618	150	666
1.80	15	602	15	610	150	665
2.52	15	592	15	598	150	665
3.24	15	583	15	593	150	663
3.60	15	583	15	590	150	663

5.4 Conceptual Model of Lubricity of Liquid Crystal Compounds in Intra-articular Friction

The unique lubricating behavior of synovia results from the presence of cholesteric and nematic liquid crystals and the anisotropy of cartilage friction surfaces. This has been proven by the experimental results discovered a new capability of synovia of providing a good antifriction behavior of joint cartilages by implementing the lubricant liquid crystal state in the friction zone. Results of these studies change radically the current understanding of biological mechanisms of reducing intra-articular friction.

This concept is corroborated by the following facts. First, mesogen compounds of cholesterol were found in human and animal synovia and the liquid crystal state of synovia within the range of physiological temperatures was proven [23, 36, 53]. Second, it was shown that the model lubricants containing liquid crystal cholesterol compounds provide the same qualitatively and quantitatively characteristics of the cartilage–cartilage and cartilage–glass pairs as synovia [48, 54]. Third, a correlation was found between the features of the microstructure of the cartilage surface caused by microrelief anisotropy and the antifriction behavior of the cartilage–synovia system resulted from the presence of liquid crystal compounds [42, 55]. Fourth, it was shown that the deformation properties of cartilages in friction under variable loads relate to the presence of the liquid crystal compounds in synovia [20, 46]. Fifth, synovia with a high content of the cholesterol compounds demonstrates the best antifriction behavior [28, 56]. And, finally, a low friction force in friction of solids lubricated with liquid crystal cholesterol compounds is shown [23, 57–59].

These scientific ideas are confirmed by both good agreement of the results obtained by different independent highly sensitive biochemical, physical-chemical, and tribological techniques and the methodology used to solve the problem.

The experimental results prove a good lubricity of liquid crystal cholesterol compounds in friction of biopolymers. The analysis shows that these results can be explained quite satisfactorily from the position of current ideas of the phenomenological theory of liquid crystals being a basis for creating the conceptual model of liquid crystal lubricity in intra-articular friction. It is founded on the theory of elasticity of liquid crystals dealing with the influence of the microrelief of bearing surfaces on the orientation of liquid crystal molecules and their interaction with the surfaces [29, 49].

According to this concept, the total free energy W of the liquid crystal equals to the sum of the free surface energy W_n and the elastic energy W_x of the liquid crystal:

$$W = W_n + W_x \tag{5.1}$$

For simplicity let us consider the liquid crystal in a flat capillary (Fig. 5.22) in which the torsional deformation described by the elasticity constant K_{22} is the only deformation distorting the field of the liquid crystal director.

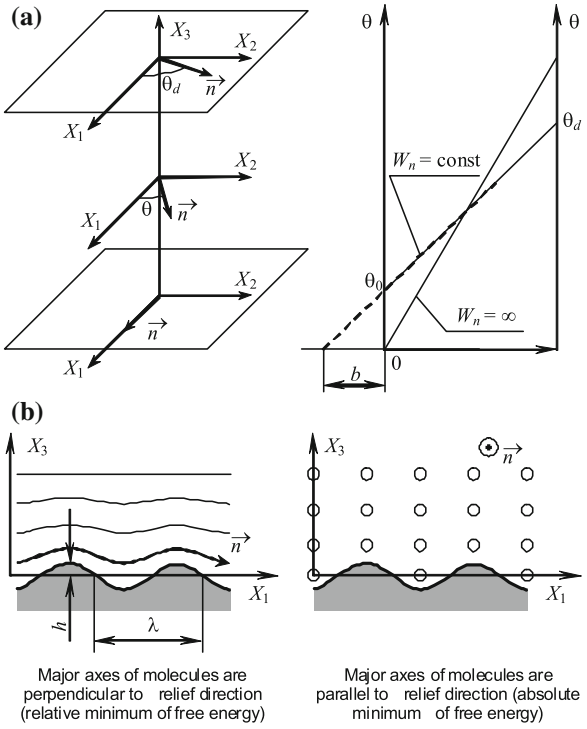


Fig. 5.22 Influence of boundary conditions on orientation of director $\vec{n}$ in bulk of cholesteric (**a**) and sinusoidal bearing surface on ordering of cholesteric (**b**)

In the one-constant approximation W_x is written as

$$W_x = \frac{1}{2} K_{22} \int_0^d \left(\frac{\partial \theta}{\partial X_3} \right)^2 dX_3 \tag{5.2}$$

The state with the minimal free energy is known as a thermodynamically stable state. This condition corresponds to a solution of the Euler-Lagrange equation:

$$\partial^2 \theta / \partial X_3^2 = 0 \tag{5.3}$$

Integrating equation (5.3) we obtain:

$$\partial \theta / \partial X_3 = C \tag{5.4}$$

Expression (5.4) describes the twisted structure of the cholesteric liquid crystal in bulk provided that the adhesion of its molecules to the bearing surface is infinitely large. Yet, in the vicinity of the bearing surface the dependence of θ on X_3 is non-linear. Therefore, we can introduce an extrapolation length (Fig. 5.22a) determined by the condition $\frac{d\theta}{dX_3} = \frac{\theta(0)}{b}$ and approximately equal to the size l of the

near-surface region or exceeding it considerably depending on the adhesion of the cholesteric liquid crystal to the bearing surface. To assess the extrapolation length we assume that at the *cholesteric—bearing surface* interface at $\theta(0) \rightarrow 0$ the total surface energy is

$$W_n = \frac{1}{2} A\theta^2(0) \tag{5.5}$$

where A is a positive constant having the dimension of the surface tension.

The elastic energy W_x of the cholesteric can be found by substituting (5.3) into (5.2) and using the boundary conditions ($C = \left(\mathrm{C} = \frac{\theta(d)-\theta(0)}{d}\right)$). Then we have

$$W_x = \frac{1}{2} K_{22} d \left[\frac{\theta(d) - \theta(0)}{d}\right]^2 \tag{5.6}$$

Solving (5.1) for $\theta(0)$ we find

$$\theta(0) = \frac{K_{22}}{A} \frac{d\theta}{dX_3} \tag{5.7}$$

It follows from the definition of the extrapolation length that

$$\frac{d\theta}{dX_3} = \frac{\theta(0)}{b} \tag{5.8}$$

The comparison of (5.7) and (5.8) yields

$$b = \frac{K_{22}}{A} \tag{5.9}$$

The constant A can be estimated as

$$A \approx \frac{U_n}{a^2}$$

where U_n is the energy of the interaction of the bearing surface with a single cholesteric molecule located in its vicinity; a is the average size of the molecule.

The elasticity constant K_{22} is comparable to U/a where U is the energy of the interaction of cholesteric molecules with each other. Then we found

$$b \approx a \cdot \frac{U}{U_n} \tag{5.10}$$

Equation (5.10) is the basic equation when describing boundary effects related to the influence of bearing surfaces on the behavior of liquid crystals. The analysis of

the equation for the extrapolation length b shows the existence of the following two alternatives:

(a) the energy of the interaction of liquid crystal molecules with the bearing surface is comparable with the energy of the interaction of the molecules with each other—$U_n = U$. This case corresponds to a strong adhesion and it follows from formula (5.10) that $b \approx a$, i.e. the extrapolation length is comparable with the molecule dimensions;
(b) the energy of the interaction of liquid crystal molecules with the bearing surface is much less than the energy of the interaction of the molecules with each other—$U_n << U$. This case corresponds to a weak adhesion. Then it follows from (5.10) that the extrapolation length exceeds considerably the molecule dimensions: $b >> a$.

The authors of [29, 49] assumed for simplicity that the bearing surface microrelief was sinusoidal and the free energy was minimal. They obtained the following relation between the extrapolation length and the microrelief characteristics:

$$b = \frac{2\lambda^3}{4\pi^3 h^2} \tag{5.11}$$

where λ and h are the wavelength and amplitude of the sinusoidal microrelief, respectively.

The analysis of (5.1) and (5.11) shows that the orientation of cholesterol ester molecules along microrelief grooves corresponds to the absolute minimum of the elastic energy while their transversal orientation corresponds to the relative minimum of the elastic energy (Fig. 5.22b). Therefore, the first case is characterized by a thermodynamically stable state and it is more energetically profitable for cholesterol ester molecules. However, as expressions (5.10) and (5.11) show, the adhesion of the cholesteric to the bearing surface depends on the surface microrelief characteristics.

If a cartilage surface is assumed to be the bearing surface it follows from the analysis of (5.11) that the available data on cartilage surface roughness characteristics [19, 50] are incomplete because they do not describe the third roughness level—*submicroroughness* (Table 5.7). The presence of submicroroughness on the cartilage surface together with macro- and microroughnesses is generally recognized since it results from the microfibrillar and fibrillar structure of collagen fibers (Fig. 5.23) [8, 23]. The analysis of characteristics of collagen structures yields the submicroroughness characteristics of the cartilage surface presented in Table 5.7.

Table 5.7 Roughness characteristics of joint cartilage

Roughness type	Level I	Amplitude h_i (μm)	Wavelength λ_i (μm)	References
Macroroughness	1	1.0–2.0	30.0–50.0	[17]
Microroughness	2	0.10–0.20	0.50	[17, 48]
Submicroroughness	3	0.01–0.05	0.02–0.10	–

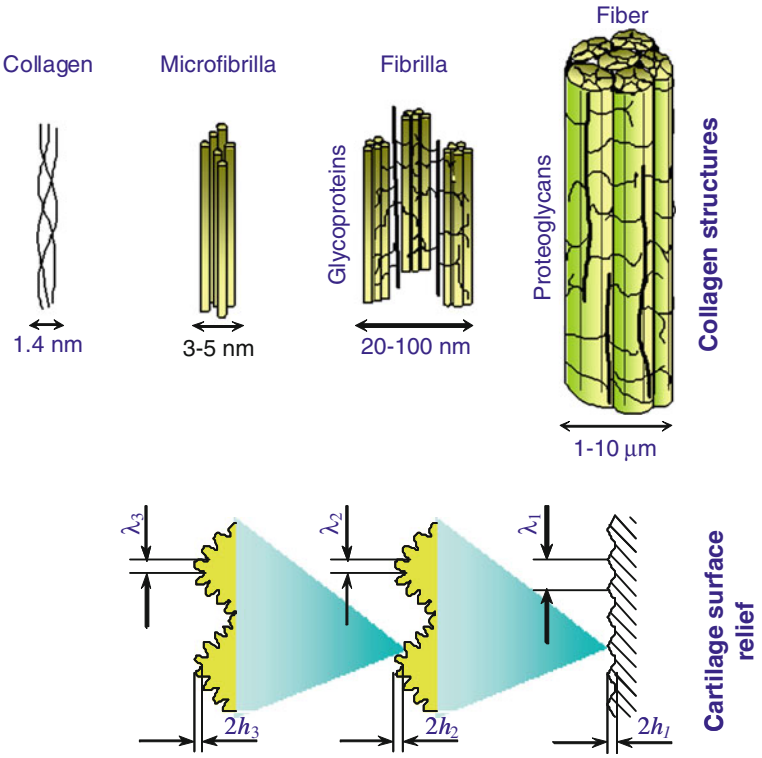


Fig. 5.23 Interrelation between texture of collagen structure and organization of cartilage surface relief

Substituting the submicroroughness characteristics of joint cartilages from Table 5.7 into expression (5.11) we obtain the extrapolation length b = 1.2…6.4 nm. It is comparable with the dimensions of cholesterol ester molecules (a = 1.0…7.0 nm) [29]. Therefore, if molecules of liquid crystal cholesterol compounds interact with the cartilage surface a strong adhesion occurs.

It can be assumed that molecules of the cholesterol compounds contained in synovia orient along surface microgrooves which are parallel to the direction of the prevailing locomotions; the molecules form a planar-oriented liquid crystal structure consisting of a lump of spirally twisted nematic layers (Fig. 5.24).

The spacing between the layers equals to the cross size of the molecules. Owing to the physical nature of molecules of liquid crystal cholesterol compounds in the zones of the maximal pressure or real contact zones of collagen fibers the minimal thickness of the liquid crystal lubricating film, i.e. the monocrystal cholesteric film, equals to half spiral pitch (200–550 nm) (Fig. 5.24b).

In lighter loaded zones or at edges of the real contact of collagen fibers the thickness of the liquid crystal film is also multiple half spiral pitch but exceeds considerably the above value for the loaded zones (Fig. 5.24c).

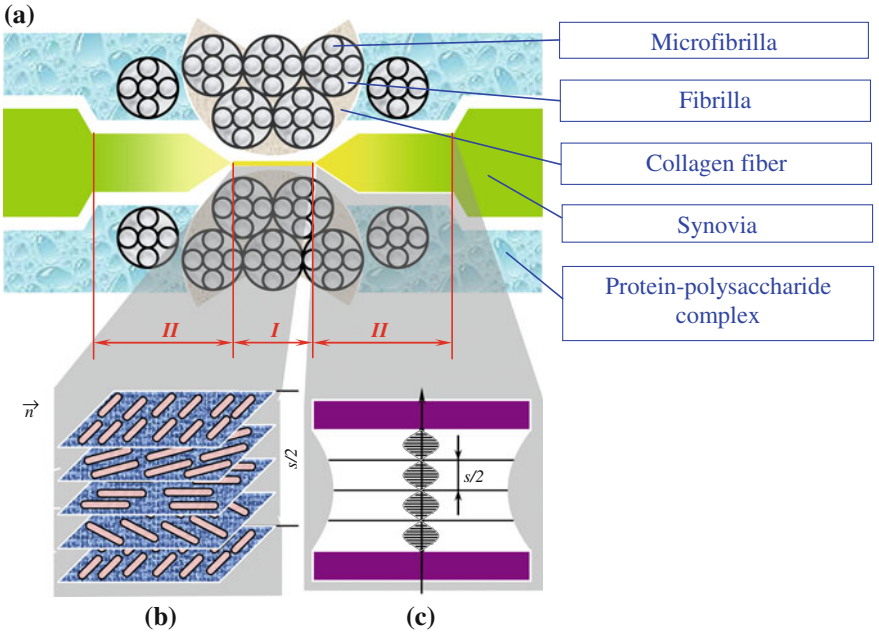


Fig. 5.24 Model of lubricity of liquid crystal cholesterol compounds in intra-articular friction of cartilages: **a** schematic of contact in "cartilage–synovia–cartilage" system; **b** volumetric organization of molecules in monocrystal film in zone of real contact of collagen fibers (*I*); **c** organization of liquid crystal layers at edges of zone of real contact of collagen fibers (*II*)

Such planar liquid crystal texture of cholesterol esters resembles the structure of lamellar solid lubricants, yet it possesses a much weaker interlayer interaction due to the liquid crystal state. Therefore, shear localizes between the layers and their easy mutual sliding provides a low friction force in the joint that is confirmed experimentally.

Thus, it can be concluded that the role of the microrelief of the cartilage surface consists in the orientation effect leading to the formation of liquid crystal lubricating films in the friction zone with the primary orientation of molecules parallel to the locomotions. As a result, energy dissipation during the relative motion of the joint cartilages minimizes.

This conceptual model of the lubricity of liquid crystal cholesterol compounds is considered from the physical, more exactly energetic, viewpoint without account for the nature of the solids in dynamic contact. For this reason it is a general model applicable to both joint and any surfaces and friction pairs.

So, synovia is a liotropic biological fluid containing a mixture of cholesteric liquid crystals. In the presence of cholesterol soluble in the mixture their certain combination and concentration provide the liquid crystal state of synovia within the temperature range including physiological temperatures. It is apparent that qualitative and quantitative changes of this component of synovia and significant

temperature variations in inflammation can shift the temperature range of the mesophase and worsen the lubricating behavior of the joint fluid. Along with this, the disappearance of the anisotropy of the cartilage surface microrelief due to severe wear hampers the formation of stable liquid crystal lubricating films that also plays an important role in the pathogenesis of cartilage destruction in collagen diseases and is confirmed by both clinical practice and experimental results.

The results of the studies of joint friction prove the effectiveness of the lubricity of liquid crystal cholesterol compounds and open a new direction in the development of means and methods of the biological control of the antifriction and structural-mechanical properties of joint synovia.

5.5 Tribological Principles of Developing Medicinal Preparations Based on Blood Serum as a Liquid-Crystalline Medium for Therapeutic Correction of Synovial Joints

Therapeutic problems concerning osteoarthrosis are important in view of its frequent occurrence and the aftereffects of this disease for patients.

The various factors leading to this disease may affect differently the synovial fluid in the joints, leading to both particular and similar changes in the joints. Osteoarthrosis is, as a rule, a local disease that mainly affects the lower extremities as the main load-bearing elements [60].

Most modern researchers hold the opinion that the articular cartilage undergoes degeneration due to two main reasons: overloading of the cartilage and reduction of its resistance to usual physical loads [60, 61]. Nevertheless, independently of etiopathogenic peculiarities, the main histopathological, biochemical, and metabolic changes arising from osteoarthrosis are analogous and ultimately lead to damage to the cell structures of the cartilage, i.e., chondrocytes [61]. This results in violation of the hydrodynamic equilibrium between synthesis and damage of the cartilage tissue. In the case of deficient synthesis, the chondrocytes fail to compensate the lack of their matrix in sulfated mucopolysaccharides (chondroitin sulfate, etc.).

On the other hand, osteoarthrosis in joints is accompanied by abrupt intensification of degradation and depolymerization of hyaluronic acid (HUA). Since HUA is a natural polymer responsible for viscoelastic properties, these properties in the synovial fluid (SF) undergo a considerable degradation. As a result, the mechanical load on the articular cartilage increases, leading to failure of the cartilage and, consequently, to pain and limited mobility in the joint.

It is known from previous history that to inhibit and cure osteoarthrosis it is necessary to fulfill certain necessary conditions; i.e., it is necessary to externally replenish the deficit of both sulfated mucopolysaccharides and HUA [29, 62, 63]. A wide spectrum of medicinal preparations (MP) have been developed for these aims; they are subdivided, correspondingly, into chondroprotectors and hyaluronates. Most efficient, of course, is intra-articular injection of these MP. Their

therapeutic effect is connected with the involvement of their ingredients in the formation of the main substance of the cartilage tissue, together with improved viscoelastic properties of the SF and restoration of its shock-absorbing function. The reduction of mechanical loading on the affected joint via restoration of homeostasis in it alleviates pain and provides improved mobility that may last for a few months after the therapeutic course.

It is, however, well known that a liquid-crystalline state is intrinsic to living matter in all its manifestations and is an ideal medium for a number of biochemical reactions and the provision of the reproduction function of biological structures. This state is characteristic of cell membranes and organelles, photoreceptors of the eye retina, myofibrillas, collagen proteins, myelin nerve fibers, nucleic acid, saliva, and other natural objects [1, 4, 29]. The synovial fluid of joints is not an exception. Its mesomorphous state plays an essential role in reducing intra-articular friction [53]. It is evident that for successful substitution of synovia, its analogue should possess the same biochemical properties and liquid-crystalline composition of the cholesterol esters as natural SF has.

From the biological viewpoint, blood serum (BS) is perfect for this purpose as an absolutely compatible biological medium that displays properties very similar to natural synovia [29, 63, 64]. There is no doubt that cholesteric LC esters are present in BS [30]. It should be noted, however, that the compositions of the cholesteric ester in BS and those in natural synovia have not been compared so far. Conclusions by analogy can be derived in studies of the tribological properties of both BS and known MP because all investigations of this type use metal-polymer friction pairs. Such experiments are indispensable but will prove to be unbiased only in the case of examination of a natural cartilage–cartilage tribopair. The present authors are of the opinion that the described approach is vital to the problems of creation and investigation of medicinal products used to cure arthropathy, especially those intended for intra-articular injection [65].

Consequently, identifying of cholesteric esters in BS and investigation of effect of chondroitin sulfate and HUA containing MP on BS lubricity for natural cartilage–cartilage pair under friction interaction are important for arthronosos medication development.

We used puncture samples from the knee joint and autoserum of patients' blood for the present investigations. The cholesteric LC compounds in the SF and BS samples were studied by chromatographic methods. For this purpose, the SF and BS samples were extracted with a mixture of chloroform and ethanol. Mixtures of the cholesterol esters were then extracted from the lipid extracts using the methods of thin-layer chromatography. Thin-layer chromatography was conducted in a developing solution of heptane–ester–acetic acid in the proportion 80:20:1, respectively. Cholesteric esters were eluted from the plate by heptane and were then identified by a Hitachi Co. gas chromatograph against the corresponding carbonic acids.

Tribological investigations were conducted on a pendulum tribometer, which is most suitable for these aims as it realizes locomotion characteristic of living articulations [29, 64]. The test samples were made from a shoulder head bone (convex) and a scapula glenoid cavity (concave part) of animals 1–2 h post mortem.

Not less than three samples (the cartilage was not separated from the underlying bone) were made for each test. The scapula head was fixed on the pendulum and installed in the glenoid cavity, which in turn was fastened on a stationary base. The radius of the articular head under study was 0.015 m and the pendulum length 0.5 m. The initial deviation angle of the pendulum was 15° and remained constant in all experiments. The studied tribopair was loaded by the pendulum mass using calibrated weights chosen according to the weight of the animal, which in these experiments was 44.8 N. To ensure the correlation between the investigation results in all tests, we tried to make the contact area constant, which was checked by transferring water-soluble paint from the preliminarily painted surface of one cartilage to the conjugated surface. The lubricating material studied was a natural SF or a mixture of it with a hyaluronate-based MP such as Vitreous Body, produced by Belmedpreparaty Co. (Belarus) (MP_1); Ostenila, produced by Chemedica AG (Germany) (MP_2); and mixtures based on chondroitin sulfate (MP_3). The mentioned MP differ in their mechanism of action and are known as preparations that enrich the viscoelastic properties of the SF and protect the cartilage from mechanical and chemical damage and are highly efficient in stimulating regenerative processes in the cartilage tissue. The first two MP differ in their HUA content, which in MP2 (10 mg/ml) is almost two orders higher than in MP1 (not less than 0.13 mg/nl). These MP are widely used in various forms to cure arthropathy [29, 62, 63, 66]. We determined the friction force in the studied load-bearing joint with a lubricating material judging by the time of oscillation attenuation of the pendulum, as well as from the dependences of the friction coefficient on time and concentration of the MP in BS.

Chromatographic analysis of the general ester fractions in cholesterol isolated from the SF and BS by extraction with a chloroform-ethanol mixture showed that the above-named biological media contain cholesterol esters of the such fatty acids (see the Table 5.8) as palmitoleic, palmitic, stearic, linoleic, oleic, and others.

It is seen from the Table that cholesterol esters of fatty acids whose molecular mass is 622–650 au form the basis of the general fraction of the cholesterol ester of the SF and BS. The cholesteric esters of palmitic and oleic acids are present in the highest concentration in both the SF and BS. The experiments also showed that the cholesteric ester content in the SF is about 3.8 ± 0.4 mol/l and in the BS 4.3 ± 0.3 mol/l, which is, in fact, similar for the given biological media.

Table 5.8 Composition of the main cholesterol esters identified in natural SF and BS ($M \pm m$, %)

Cholesterol esters	Molecular weight, conventional units	Mediums	
		SF	BS
Cholesterol palmitoleate	622	3 ± 2	4 ± 1
Cholesterol palmitate	624	24 ± 5	20 ± 2
Cholesterol stearate	636	9 ± 2	8 ± 1
Cholesterol linoleate	649	8 ± 4	7 ± 5
Cholesterol oleate	650	24 ± 2	25 ± 3
Other cholesterol esters	–	Up to 100	Up to 100

Consequently, the experiments show that the qualitative and quantitative composition of the cholesterol ester in the BS is correlated with the analogous composition of these compounds in the SF.

Previous investigations conducted by three independent methods, namely by thermopolarization microscopy, differential scanning microcalorimetry, and probe fluorescence of different SF samples, have shown that the above-mentioned composition and concentration of cholesterol esters in the fatty acids promote the formation of an LC mixture whose mesophase interval is found within the physiological temperatures [29, 53].

It follows from the above procedure that the BS, like the natural articular lubricant SF, contains a mixture of thermotropic LC compounds having a cholesteric type of structural ordering and a mesomorphous state within the region of physiological temperatures. Taking into account previous tribological investigations of SF, we may anticipate that cholesteric LC compounds in the BS will exert a similar positive effect on its antifriction properties as in the case with SF.

This supposition was validated experimentally using a pendulum tribometer with a natural cartilage–cartilage friction pair (Fig. 5.25). The investigation results visually demonstrate that the BS (Fig. 5.25, line 3) provides for a two-to-three times lower friction coefficient in contrast to, e.g., distilled water (line 1) or physiological solutions (line 2) that are known to be devoid of cholesterol esters.

Notice that in contrast to the natural synovia, the BS has in fact the same viscosity as water and is categorized as a Newtonian liquid, i.e., a liquid whose

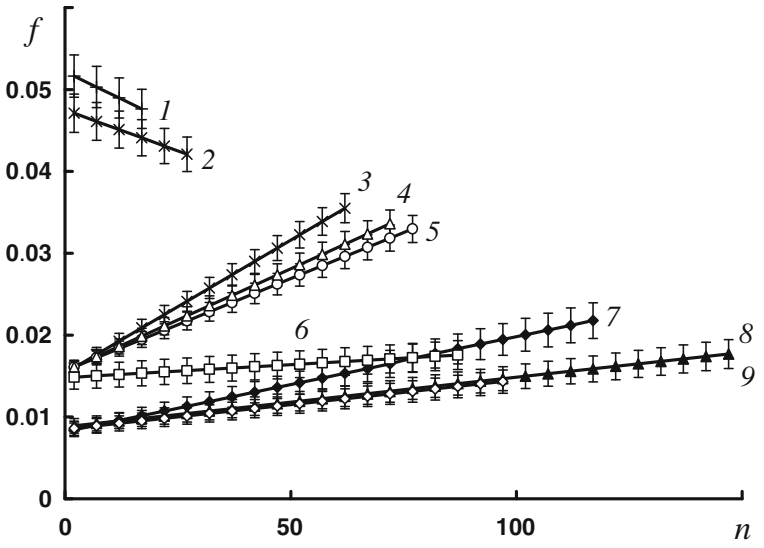


Fig. 5.25 Friction coefficient dependence of a natural cartilage–cartilage pair versus number of oscillations of pendulum tribometer under lubrication with: *1* distilled water; *2* physiological solution; *3* BS; *4* BS = 5 % MP; *5*, *6* MP containing HUA in solution of MP_1 "Vitreous Body" *5* and MP_2 "Ostenil" 6; *7* MP_3 containing cholesterol sulfate; *8* BS + 50 % MP_3; *9* SF

viscosity is independent of the shear rate [67]. It is quite probable, however, that the viscosity of the liquids under study plays an important role in frictional interactions between cartilages.

The experimental data confirm that BS doping with MP_1 containing HUA believed to be responsible for synovial viscosity [29, 61] results in improved tribological properties of the serum (Fig. 5.25, line 4).

This is also supported by experiments on friction of cartilage against cartilage immediately in MP_1 (line 5) that contains not less than 0.13 mg/ml HUA, and is, therefore, more viscous than BS. Thus, the presence of HUA in MP_1 generates characteristic 3D skeletons on the cartilage surface and promotes thereby the formation of a stronger separating lubrication film during rubbing as compared to that of BS, which is confirmed experimentally. Another confirmation of this behavior of the lubricating media containing hyaluronate preparations is the frictional interaction of cartilage in MP_2 (Fig. 5.25, line 6) with a higher HUA concentration (10 mg/ml), and consequently, a higher viscosity of the preparations. Figure 5.26 illustrates the rheological properties of hyaluronate preparations with different HUA content. Depending on the HUA concentration, the viscosity of the preparations is seen to vary within fairly broad limits, i.e., by several orders of magnitude. The less expressed slope of the friction coefficient vs. time dependence for the MP_2 with increased HUA concentration (Fig. 5.25, line 6) undoubtedly speaks in favor of the considerable effect of the viscosity of the named preparations on their antifriction properties. The above results also confirm the formation of an optimal lubrication layer in the considered test conditions. Note that the thickness of this layer remains constant independently of time due to the rather high content of HUA molecules, which is characteristic for this type of hyaluronate preparation MP_2.

Nevertheless, because of their high viscosity, it was problematic to estimate the tribological efficiency of such batch-produced hyaluronate preparations containing 10 mg/ml HUA or more in combination with cholesterol esters incorporated in the BS; as a consequence, it was difficult to achieve a homogeneous mixture with the serum under usual conditions. Despite the fact that the therapeutic effect of these MP on the base of hyaluronate derivatives is related to the improvement of the rheological properties of articular lubrication, hyaluronate derivatives are unable to remain in the articular cavity for long after injection and could probably not independently maintain optimal rheological properties in the synovia for a long time. They may also undergo enzymatic and free-radical degradation in articular cavities affected by osteoarthrosis [60, 68]. Therefore, the positive clinical effect of the above-described MP is attributed largely to the response of synoviocytes to their introduction, which causes them to produce a superior hyaluronate of their own [61, 68].

Since the above indicated curves for the considered water-based media in fact begin at similar friction coefficient values (≈0.015), we can advance the important conclusion that all these compositions can be classified in one and the same group of lubricating media according to their antifrictional characteristics. The best tribological characteristics are, however, displayed by the compositions that contain a natural polymer HUA. It follows from the experimental data Fig. 5.25, lines 3–6)

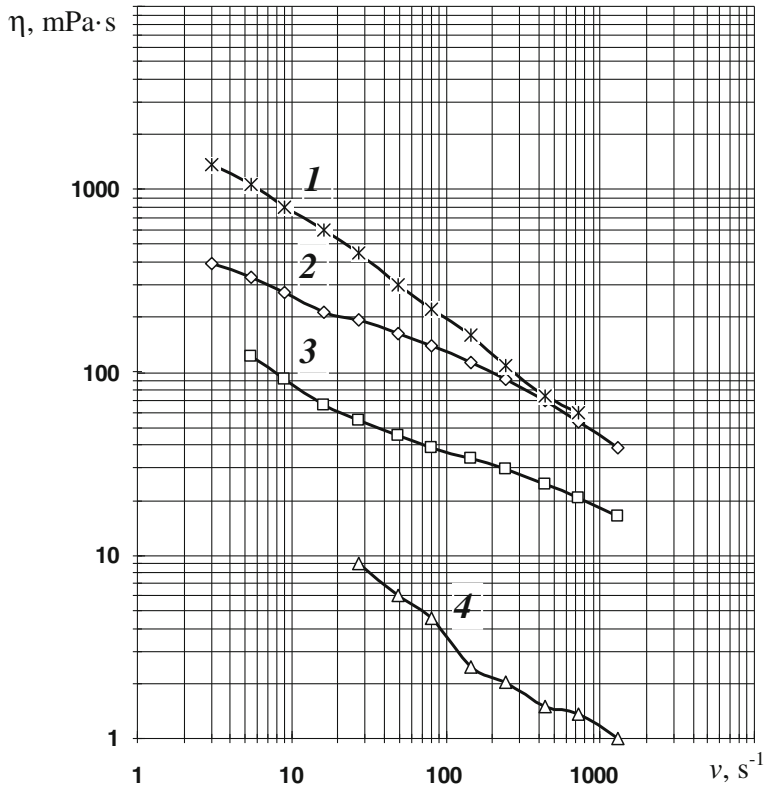


Fig. 5.26 Dependence of viscosity of hyaluronate preparations on shear rate at different HUA concentrations: *1* C = 10 mg/ml; *2* 7.5; *3* 5; *4* C = 2.5 mg/ml

that the friction coefficient values of about 0.015 for this group of liquid media turn out to be limiting values due to the lubricating mechanism in which, as was noted previously, an important role is played by the viscoelastic properties of the lubricating medium.

On the other hand, all MP containing chondroitin sulfate are known to contribute much to the construction of the main bone stock and cartilage tissue and to normalization of the production of articular liquid; in addition, they play the role of a lubricant, improving the mobility of the articular surfaces. The experimental results comply fully with the above conclusion.

It is shown that preparation MP3 containing chondroitin sulfate generates considerably lower friction coefficient values during rubbing of the cartilage–cartilage pair (Fig. 5.25, line 7) as compared to hyaluronate preparations MP_1 and MP_2. Evidently, in contrast to the latter hyaluronate preparations, the MP_3 is perfectly compatible with BS in any ratio. In combination with BS, preparation MP3 was found to ensure a synergetic effect (Fig. 5.25, line 8; Fig. 5.26, lines 3 and 4), visualized in the even better lubricity of these compositions as compared to the

lubricating effect of the initial components, namely MP_3 itself or BS. Under a certain concentration of MP3 in the BS, the friction coefficient values actually coincide with those found with the use of SF (Fig. 5.25, lines 8 and 9).

The essential difference in the mechanisms of interaction between the MP based on chondroitin sulfate or hyaluronates and cartilage surfaces is supported by the different behavior of the concentration dependences of the friction coefficient upon the number of pendulum oscillations $n = 60$; i.e., in stationary friction of cartilage the concentration dependence in the former case is of a clearly extremal character, while in the latter case it is monotonously descending.

Indeed, according to works [29, 69, 70] the extremal friction curves are typical of lubricating media that contain additives, e.g., in the form of surfactants. Since the extreme points are close to the critical concentration of micelle formation during friction [69, 70], under the optimal concentration, BS is characterized by an increasing amount of micelles based on chondroitin sulfate macromolecules and cholesterol ester molecules whose lubricity is much higher than that of the chondroitin sulfate macromolecules themselves. This ultimately leads to the extreme character of the dependence obtained experimentally additive MP3 concentration 20–50 % corresponds to the formation of complex compounds based on chondroitin sulfate and cholesterol esters in BS.

At the same time, in the case of monotonous dependences of the friction coefficient, the concentration curves can be considered as a type of adsorption isotherm [69, 70]. Given that, in compliance with [70], the friction coefficient reaches its minimum at saturation of the adsorption layer with structured molecules, then in our case (Fig. 5.27, curve 1), we may accept that this process of adsorptive structurization on the cartilage surface takes place with HUA molecules.

Thus, it has been established experimentally by pendulum tribometry during friction of cartilage against cartilage that MP based on hyaluronates and chondroitin sulfate affect differently the lubricity of BS.

It has been shown that combined use of preparations based on chondroitin sulfate and BS enhances the tribological properties, which practically coincide with those of natural SF (Fig. 5.25, line 9). Therefore, the above-described MP can be effectively used as optimal biological new-generation substitutes of SF.

It is also evident from Fig. 5.25 that these MP in combination with SF form a new original group of lubricating media for articular cartilage, containing chondroitin sulfate and possessing unique lubricating properties.

The dependences of the friction coefficient for these lubricating media, as in the case with the medium based on hyaluronates, in fact begin at the same point, though the former are found at a lower level (0.008–0.015), whereas at optimal concentrations of the components they almost coincide with the corresponding dependence of the natural lubricant SF.

The results of the above-discussed investigations form the basis for further development of new more efficient methods of regulating the tribological properties of BS during frictional interaction of cartilage and for their effective use as a potential water-based liquid-crystalline synovial medium for treatment of osteoarthrosis and analogous pathologies [71].

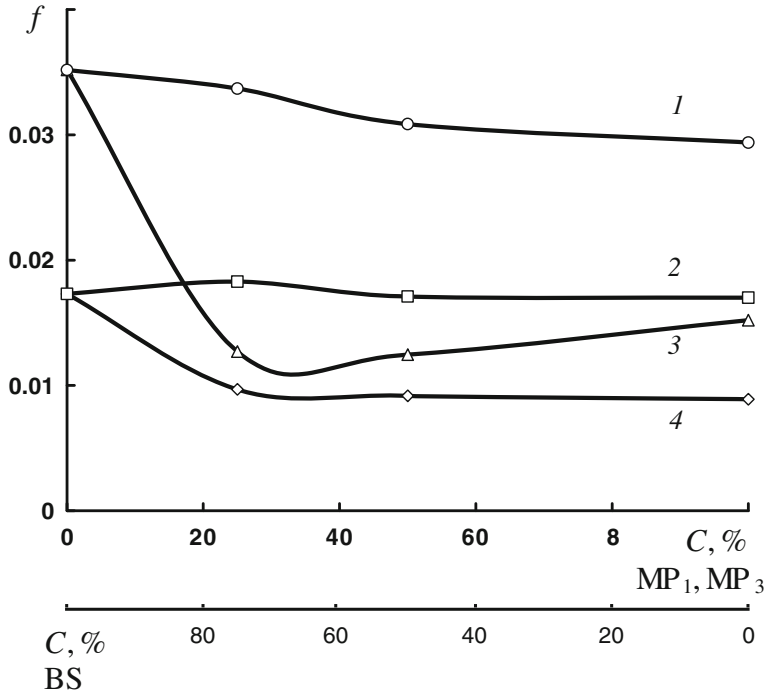


Fig. 5.27 Concentration dependences of the friction coefficient of cartilage–cartilage pair under lubrication with BS doped with MP containing: *1*, *2* hyaluronate "Vitreous body" (MP_1); *3*, *4* chondroitin sulfate (MP_3) under number of pendulum oscillations, respectively, $n = 5$ (*2*, *4*) and $n = 60$ (*1*, *3*)

References

1. N.V. Usol'tseva, O.B. Okopova, V.V. Bikova, A.I. Smirnova, S.A. Pikin, *Liquid Crystals: Discotic Mesogens* (University Publishers, Ivanovo, 2004), p. 546 (in Russian)
2. G. Brown, D. Walken, in *Liquid Crystals and Biological Structures*, ed. by Ja.M. Varshavskii (Mir, Moscow, 1982), p. 420
3. A.V. Sarma, G.L. Powell, M. LeBerge, Phosphoiipid composition of articular cartilage boundary lubricant. J. Orthop. Res. **19**, 671–676 (2001)
4. B.A. Hills, Surface-active phospholipids: a Pandora's box of clinical applications. Part II. Barrier and lubricating properties. Int. Med. J. **32**, 242–251 (2002)
5. D. Dowson, A. Unsworth, V. Wright, Analysis of boosted lubrication in human joints. J. Mech. Eng. Sci. **12**, 364–369 (1970)
6. P.C. Seller, D. Dowson, V. Wright, The rheology of synovial fluid. Rheol. Acta **10**, 2–7 (1971)
7. D. Gvozdanovic, V. Wright, D. Dowson, Formation of lubricating monolayers at the cartilage surface. Ann Rheum. Dis. **34**, 100–106 (1975)
8. V.N. Pavlova, *Joint Synovia* (Meditsina, Moscow, 1980), p. 296. (in Russian)
9. F.P. Bowden, D. Tabor, *Friction and Lubrication of Solids* (Mashinostroenie, Moscow, 1968), p. 544

10. A.S. Akhmatov, *Molecular Physics of Boundary Friction* (Fizmatgiz, Moscow, 1963), p. 472. (in Russian)
11. P.G. de Gennes, *The Physics of Liquid Crystals* (Clarendon Press, Oxford, 1974), p. 400
12. A. Blumshtein (ed.), *Liquid-Crystal Order in Polymers* (Mir, Moscow, 1981), p. 392 (in Russian)
13. V.G. Rodnenkov, B.I. Kupchinov, Tribological behavior of liquid crystals. Part 1. Tribology of pure liquid crystals and their mixtures (Review). J. Friction Wear **16**(3), 59–65 (1995)
14. V.R. Ivanov, N.V. Usol'tseva, I.I. Gorina. Thermographic diagnostics in medicine involving liquid crystals. J. D.I. Mendeleev All-Union Chem. Soc. **28**(2), 68–75 (1983) (in Russian)
15. G.P. Barchan, Liquid crystal state of interphase layers in sliding. Proc. USSR Acad. Sci. **258** (1), 86–88 (1981). (in Russian)
16. A.A. Markov, YuV Lunkov, T.N. Nazarova, V.K. Gusev, Experimental study of lubricating oil influence on metal wear resistance. J. Friction Wear **5**(3), 123–126 (1984)
17. S.F. Ermakov, Influence of nature of counterbody and lubricant on articular cartilage friction. J. Friction Wear **9**(2), 101–105 (1988)
18. S.F. Ermakov, B.I. Kupchinov, V.G. Rodnenkov, E.D. Beloenko, O.L. Eismont, Influence of nature of rubbing surfaces and lubricant on articular cartilage friction. Acta Bioeng. Biomech. **3**(suppl. 1), 65–71 (2001)
19. D.F. Moore, *Principles and Applications of Tribology* (Pergamon Press, Oxford, 1975)
20. S.F. Ermakov, E.D. Beloenko, O.L. Eismont, Role of liquid crystals in tribological behavior of joint cartilages. J. Friction Wear **25**(5), 31–35 (2004)
21. Z.M. Jin, D. Dowson, J. Fisher, The effect of porosity of articular cartilage on the lubrication of a normal human hip joint. Proc. Inst. Mech. Eng. **206**, 117–124 (1992)
22. P.S. Walker, J. Sikorski, D. Dowson, M.D. Longfield, V. Wright, Features of the synovial fluid film in human joint lubrication. Nature **225**(5236), 956–957 (1970)
23. B.I. Kupchinov, S.F. Ermakov, E.D. Beloenko, *Biotribology of Synovial Joints* (Vedy, Minsk, 1997), p. 272. (in Russian)
24. V.K. Mow, Role of lubrication in biomechanical joints. J. Lubr. Technol. **91F**(2), 320–328 (1969)
25. F.C. Linn, Lubrication of animal joints 1. The Arthrotripsometer. J. Bone Joint Surg. **49**, 1079–1098 (1967)
26. A.V. Volokhina, Yu.K. Godovskii, G.I. Kudriavtseva, et al., *Liquid Crystal Polymers*, ed. by N.A. Plate (Khimiya, Moscow, 1988), p. 182 (in Russian)
27. V. Wright, D. Dowson, J. Kerr, The structure of joints. IV Articular cartilage. Int. Rev. Connect. Tissue Res. **6**, 105–124 (1973)
28. E.D. Beloenko, S.F. Ermakov, B.I. Kupchinov, V.G. Rodnenkov, O.L. Eismont, Liquid crystal state of joint synovial lubricating medium Experimental substantiation. Acta Bioeng. Biomech. **3**(suppl. 1), 24–32 (2001)
29. S.F. Ermakov, V.G. Rodnenkov, E.D. Beloenko, B.I. Kupchinov, *Liquid Crystals in Engineering and Medicine* (Asar, Moscow, CheRo, Minsk, 2002), p. 412. (in Russian)
30. G.I. Sidorenko, V.G. Tsapaev, Diagnostic significance of phase transition temperatures of blood serum cholesterol in atherosclerosis. Cardiology **23**(10), 92–95 (1983)
31. J. Patsak, *Organic Chemistry* (Mir, Moscow, 1986), p. 366 (Czech translation)
32. B.I. Kupchinov, V.G. Rodnenkov, S.N. Bobrysheva et al., The mechanism of functioning of the joints as frictional elements. Dokl. Academii Nauk Belarusi **29**(5), 463–465 (1985)
33. G. Kerolus, G. Clayburne, H.R. Schumacher, Is it mandatory to examine synovial fluids promptly after arthrocentesis? Arthritis Rheum. **32**(3), 271–278 (1989)
34. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov, S.N. Bobrysheva, E.D. Beloenko, The effect of liquid crystals on joint lubrication. Wear **171**, 7–12 (1994)
35. S.S. Khalatov, *Holesterol Affliction* (Medgiz, Moscow, 1946), p. 127. (in Russian)
36. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov et al., Role of liquid crystals in the lubrication of living joints. Smart Mater. Struct. **2**, 7–12 (1993)
37. N.A. Plate, V.P. Shibaev, *Ridge-Like Polymers and Liquid Crystals* (Khimiya, Moscow, 1980), p. 304. (in Russian)

38. J. Folch, M. Lees, G.H. Sloane-Stanley, A simple method for isolation and purification of total lipids from animal tissue. J. Biol. Chem. **226**, 497–509 (1967)
39. S.F. Ermakov, B.I. Kupchinov, A.A. Suslov, et al., Synthetic synovial fluids for correction of living joint tribology. Abstracts of 6th international symposium "INTERTRIBO-93", Bratislava, pp. 30–31, 1993
40. F. Pauwels, *Biomechanics of the Locomotor Apparatus* (Springer-Verlag, Berlin-Heidelberg-New York, 1989), p. 374
41. Patent GB1391577, Int. Cl. C08L 1/28, A61 K 317/15. Pseudo-synovial plastic body fluids and method of preparing same/C.A. Homsy—No 11352/73, Filed 8.03.73, Complete Specification published 23.04.75
42. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov et al., Study of cartilage tribological properties. J. Friction Wear **9**(4), 73–77 (1988)
43. L. Dintenfass, Lubrication in synovial joints: a theoretical analysis. J. Bone Joint Surg. **45A**, 1241–1243 (1963)
44. R.I. Tanner, An alternative mechanism for the lubrication of synovial joints. Phys. Med. Biol. **11**, 119–121 (1966)
45. V. Pekoshevski, Effect of surface roughness and its orientation on oil film strength. J. Friction Wear **21**(5), 88–94 (2000)
46. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov, E.D. Beloenko, Relation of the structural-mechanical and antifriction properties of the synovial medium of the joints. Mech. Compos. Mater. **24**(2), 188–194 (1988)
47. S.F. Ermakov, B.I. Kupchinov, E.D. Beloenko, A.A. Suslov, Anisotropy of the cartilage surface and its effect on the joint tribology. Abstracts of 7th European conference on applications of surface and interface analysis, Göteborg, pp. 48–49, 1997
48. S.F. Ermakov, B.I. Kupchinov, E.D. Beloenko, et al., Liquid crystalline components of synovia and their role in the joint tribology. Proceedings of symposium "Inzynieria Ortopedyczna i Protetyczna", Belostok, pp. 12–131, 1997
49. A.S. Sonin, *Introduction to Physics of Liquid Crystals* (Nauka, Moscow, 1983), p. 320. (in Russian)
50. P. Walker, J. Sikorski, D. Dowson, et al., Lubrication mechanism in human joints. *Bio-Engineering Group on Human Joints,* University of Leeds, pp. 49–56, 1966-67
51. C.R. Johnson, D. Dowson, V. Wright, A new approach to the determination of the elastic modulus of articular cartilage. Ann. Rheum. Dis. **34**, 116–117 (1975)
52. A.N. Romanovskaya, G.L. Voskresenskii, Comparative study of rheological behavior of synthetic and biological polymeric materials on example of filled silicon rubber and human and animal joint cartilage. Mech. Compos. Mater. **5**, 906–909 (1984) (in Russian)
53. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov, S.N. Bobrysheva, E.D. Beloenko, I.R. Voronovich, Yu.M. Pleskachevskii, V.A. Belyi, Property of Synovial Medium to Ensure the High Antifriction of Cartilages in Joints of Humans and Animals, in *Nauchnye otkrytiya* (Scientific Discoveries), Moscow, pp. 14–16, 1999
54. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov, E.D. Beloenko, O.L. Eismont, Some results of studies in liquid-crystalline state of synovial lubricant in joints. J. Friction Wear **23**(3), 69–75 (2002)
55. S. Ermakov, B. Kupchinov, E. Beloyenko, A. Suslov, O. Eismont, The effect of liquid crystals on tribomechanical properties of cartilages. Inzynieria Ortopedyczna i Protetyczna—IOP 99: Proc. of the Sympozjum, Bialystok, pp. 93–99, 1999
56. E.D. Beloenko, O.L. Eismont, L.A. Pashkevich, I.A. Chved, S.F. Ermakov, Effectiveness of medication containing bioactive cholesteric-nematic liquid crystals substance in treatment of experimental osteoarthritis. Proc. Natl. Acad. Sci. Belarus: Medical Series **1**, 5–8 (2005)
57. S.F. Ermakov, V.P. Parkalov, V.A. Shardin, R.A. Shuldykov, Effect of liquid-crystal additives on tribological performance of dynamically contacting surfaces and mechanism of their friction. J. Friction Wear **25**(2), 87–91 (2004)
58. S.F. Ermakov, R.A. Shuldykov, A.V. Mikyalenis, The tribological aspects reversible regulations of frictional interaction of solid bodies at presence liquid crystal. Proceedings of

the 6th international symposium of friction products and materials «YAROFRI-2006», Yaroslavl, pp. 59–65, 2006
59. G. Biresaw (ed.), Tribology and the Liquid-Crystalline State, American Chemical Society, Symposium Series, no 441, p. 130, 1990
60. E.S. Tsvetkova, in *Book Rheumatic Deceases*, ed. by V.A. Nasonov, N.V. Bunchuk. Osteoarthrosis (Meditsina, Moscow, 1997), pp. 385–396 (in Russian)
61. V.N. Pavlova, T.N. Kop'eva, L.I. Slutskii, G.G. Pavlov, Cartilage (Meditsina, Moscow, 1988), p. 320 (in Russian)
62. D.T. Felson, Osteoarthritis of the knee. N. Eng. J. Med. **354**, 841–848 (2006)
63. L.S. Pinchuk, YuM Chernyakova, V.A. Gol'dade, The tribology of joints and problems of modern orthopedics. J. Friction Wear **29**(3), 224–233 (2008)
64. S.F. Ermakov, A.V. Beletskii, V.I. Nikolaev, Tribological principles of developing medicinal preparations based on blood serum as a liquid-crystalline medium for therapeutic correction of synovial joints. J. Friction Wear **32**(1), 49–53 (2011)
65. V.I. Nikolaev, A.V. Beletskii, D.V. Charnashtan, S.F. Ermakov, Formation and estimation of friction processes of synovial joints. Probl. Zdorov. Ekol. **16**(2), 100–104 (2008)
66. G. Berbuggen, Chondroprotective drugs in degenerative joint diseases. Rheumatology **45**, 129–138 (2005)
67. A.A. Suslov, S.F. Ermakov, A.V. Beletskii, S.V. Shil'ko, V.I. Nikolaev, The role of liquid phase and porous structure of cartilage in formation of biomechanical properties of joints. Russ. J. Biomech. **12**(4), 33–39 (2008)
68. J.E. Scott, Hyaluronan, multum in parvo. Eur. J. Rheumatol. Inflamm. **1**, 3–8 (1995)
69. S.F. Ermakov, *Tribology of Liquid- Crystalline Nanomaterials and Systems* (Minsk, 2012) (in Russian)
70. G.I. Fuks, Adsorption and lubricating ability of oils. Trenie Iznos **4**(3), 398–414 (1983)
71. A.V. Beletzky, V.I. Nikolaev, S.F. Ermakov, Medico-biological aspects of regulating tribological properties of blood serum as a potential liquid-crystalline synovial medium fur treating osteoarthrosis. Dokl. Natl. Acad. Sci. Belarus **25**(2), 87–91 (2004)

Chapter 6
Liquid Crystals as Effective Drugs for Treatment of Articular Disorders and Similar Pathologies

Abstract The problem of treatment of rheumatic diseases has pronounced medicosocial significance. It is certain that similarity of the tribological, rheological and structural properties of artificial lubricants with the natural synovial fluid is the most essential provision of successful replacement of the fluid in case of joint pathologies. The results of structural, rheological and tribological researches on the creation of artificial synovial liquids which contain cholesteric liquid crystals typical to natural synovial liquids have been described. The experimental data on high chondroprotective efficiency of preparation, checked on osteoarthritis models and during clinical approval, are proof that liquid crystals play an essential role in intraarticular friction decrease. It can be a real prerequisite for development of new pharmaceuticals for cartilage mechanodestruction prophylaxis and therapy during arthropathies.

Rheumatic diseases are a basic and most common group of disorders of joints. At present over 80 nosological units have been classified [1]. Notwithstanding the differences of their origins, evolution and development, the basis of the existing rheumatic diseases are systemic or local injuries of the collagenic tissue resulting, as a rule, in a progressing process of destruction of joints. The diseases are observed in all climatic and geographical zones. About 4 % of the global population is affected; at present there is not any remarkable indication that the figure will reduce [2, 3]. The WHO data evidence that 30 % of cases is temporary disability and 10 % is total disability due to rheumatic diseases. A significant percent in the structure of disablement belongs to individuals under 45 years of age [4–6].

Because of this, the problem of treatment of rheumatic diseases has pronounced medicosocial significance. It is natural that such treatment is impossible unless the mechanism of destruction of joints is studied and understood comprehensively. Now it is certain that similarity of the tribological, rheological and structural properties of artificial lubricants with the natural synovial fluid is the most essential provision of successful replacement of the fluid in case of joint pathologies.

S. Ermakov et al., *Liquid Crystals in Biotribology*,
Biological and Medical Physics, Biomedical Engineering,
DOI 10.1007/978-3-319-20349-2_6

6.1 Mechanisms of Connective Tissue Destruction at Rheumatic Diseases

Numerous causal factors of rheumatic diseases can differently affect the synovial medium in joints inducing both individually pronounced and practically alike changes in its elements. Therefore, studies and analysis of destruction of the cartilage demand to subdivide articular diseases into inflammatory and non-inflammatory, or degenerative and dystrophic [6, 7].

Among various clinical manifestations of rheumatic disorders, the leader is the degenerative and dystrophic affection of joints and primarily the deforming osteoarthrosis. It happens four times more often than inflammatory processes or arthritis (e.g., rheumatoid, etc.) and makes up 60–80 % of all forms of articulate pathology in the countries of Europe and America [2, 3, 6]. Women and persons over 40 years dominate among the patients. Therefore, the degenerative-dystrophic processes were believed the symptoms of ageing. In fact, some semblance between the morphological changes in the cartilage in these diseases and during ageing so far leave the problem too whether it is a disease or a reaction of compensation to a loss of functional adaptation. Still, these processes are not identical [8–10].

The deforming osteoarthrosis is, as a rule, a local disease affecting predominantly the joints in the lower extremities that support the maximum load. It is believed that a pathological process can appear in a healthy or in an afflicted cartilage. Respectively, they conventionally differentiate between the primary (idiopathic) arthrosis, or "structural chondrosis", and secondary arthrosis, or "mechanical chondrosis" that appears as a result of suffered traumas of joint elements, arthritis, dysplastic and static disorders.

A majority of researchers at present share the view that degeneration of the joint cartilage is due to two main causes: excessive loading on the cartilage and weakening of its resistance to common physiological loads [6, 11]. A substantial significance is attached to exo- and endogenic etiological factors, such as chronic static and dynamic overloads, microtraumization and extreme exposure, obesity, hereditary predisposition and nervous endocrine disorders, the ischemia due to changes in toe general and regional blood flow [6, 12]. Yet, most of the causes are quite conventional because they base primarily on the data of epidemiological studies that preclude controversial conclusions.

Thus, the deforming osteoarthrosis is a multifactor disease of joints. Irrespective of ethiopathogenetic features; the main histopathological, biochemical and metabolic changes in osteoarthrosis are the same and lead, in a final account, to damage of the cellular structures of the cartilages, or chondrites [13].

An essential function of chondrocytes is reproduction and maintenance of the structural integrity of the cartilage matrix. At present, it is known that, if normal, these cells are responsible for collagen synthesis of predominantly type II, sulfonated glucose aminoglycans and hyaluronic acid, glycoproteins and other non-collagenous proteins and enzymes [8, 9, 13, 14]. In contrast to other proteins, the collagen type II has better physico-chemical and mechanical stability that

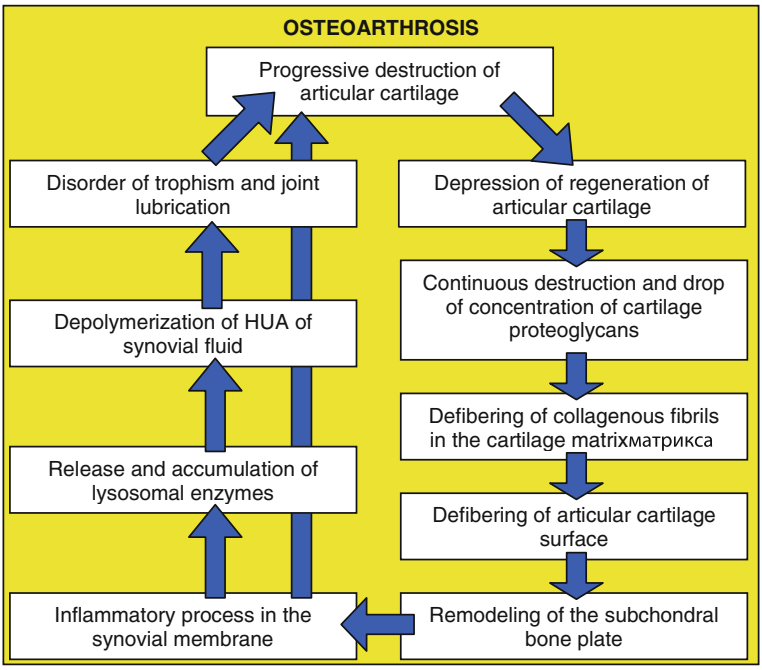


Fig. 6.1 Diagram of pathogenesis of deforming osteoarthrosis

corresponds to the biophysical and functional features of the cartilage component in the joint. The functional activity of chondrocytes depends directly on the condition of the synovial medium in the joints. Therefore, any change in the homeostasis of the articulation medium under the effect of any of the above casual factors can be a source of a disorder of the synthetic function of cells. The synthesized products can remain collagens and glicoaminoglycans, yet their structure will be different and their resistance to mechanical loads will be less.

Though the pathogenetic mechanism of osteoarthrosis is in many respects unclear and is subject of discussion, it can in its general features represented in the following manner (Fig. 6.1).

Most researchers believe that a disorder of metabolism of chondrocytes when catabolic processes dominate over anabolic processes starts cartilage degeneration in osteoarthrosis [11–13, 15]. Appearance of the products of the altered cellular synthesis subject to fast decay and elimination favors leakage of proteoglycans from the cartilage matrix. Proteolytic enzymes released alongside play en essential role in maintaining the process of degradation of proteoglycans and secondary damage of ultrastructures of chondrocytes [11].

As a result, the dynamic equilibrium between the synthesis and degradation of the cartilage tissue is upset. Therefore, in case of a deficit of normal synthesis and excessive pathologically altered synthesis the chondrocytes become unable to compensate its lacking matrix in sulfonated mycopolysaccharides. The damage

appearing in response morphologically cause proliferative responses of chondrocytes and intensify their functional activity so that their deficient reparative potential cannot stop the already started process [6, 12].

These facts have been convincingly proven in experimental works. At present, a number of models exist of the osteoarthrosis induced by changes of external and internal factors of the joint: immobilization, a trauma of the cartilage, the meniscus and ligaments, penetration of foreign bodies, enzymes etc. into the joint cavity.

The cartilage is known to swell and its elastic properties depend on the concentration of mucopolysaccharides in the cartilage tissue. E.g., any change in the mean molecular mass and osmosis of glucose aminoglycans under the effect of hyaluronidase destroys the elastiviscous characteristics of the cartilage [8, 9, 15–17]. The most pronounced physico-chemical damage of the matrix appears initially in the intermediate cartilage zone. Microfractures of the collagenous skeleton and depletion of the matrix by proteoglycans lead to its irregular hiperhydration proportional to the residual concentration of glucose aminoglycans [8, 9, 15, 17]. This process is accompanied by the thinning and splitting of collagenous fibers.

As a result of this change in the optimal structural and quantitative interrelations between the matrix components the biomechanical and tribological cartilage properties are inevitably disordered followed by further development of the process of mechanodestruction of the cartilaginous tissue.

In case of the idiopathic osteoarthrosis, this damage has secondary nature in respect to the preceding change in the ultrastructure and resistance of collagenous fibers to loading. However, a primary rupture of collagenous fibrils is possible in the healthy cartilage in case a tolerable physiological load is exceeded significantly, e.g., in case of the impact traumatic mechanism.

In the physiological conditions, the joint functions properly like in an isolated system. During osteoarthrosis, any damage of the collagenous skeleton and reduction of the barrier role of the cartilage matrix open up unrestricted two-way transport of larger protein molecules. Therefore, the pathological changes and the processes of mechanodestruction fetch masses of products of wear and degradation of the collagen, proteoglycans, and cellular membranes into the synovial fluid. These proteins, absent in the normal synovial fluid, acquire autoantigenic properties, and induce immunal mechanisms of cartilage destruction. The immunopathological process like a reaction of retarded hypersensitivity develops parallel both in the cartilage and in the synovial membrane [15].

Its activation is strongly influenced by the appearance of a significant quantity of the so-called inflammation mediators and modulators (histamine, kinins, prostaglandins, chemotactic and other substances), intensification of free-radical oxidation. The bone tissue is believed to influence this process in a certain manner. Remodeling is observed and consolidation of the subchondrial bone plate with disorders of the blood flow and cartilage trophism. Any rise of specific loads is responded by a peculiar protective reaction of articular bone surfaces as an intensification of reparative processes over the periphery and expansion of the contact area, i.e. osteophytes appear (Fig. 6.2). These zones reveal deposition of crystals of

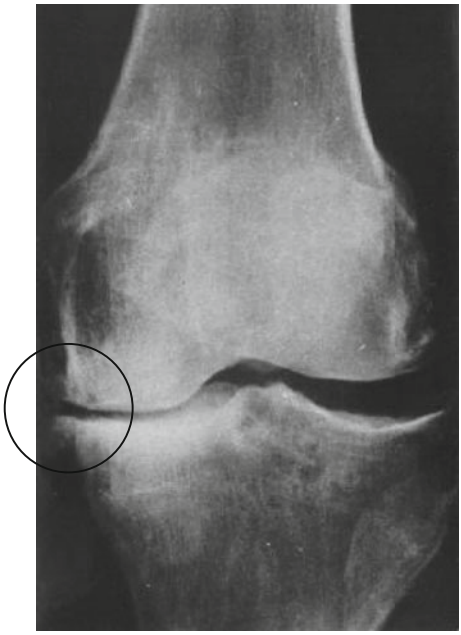


Fig. 6.2 X-ray pattern of deforming osteoarthrosis of knee joint demonstrating appearance of osteophytes over edges of the tibial plateau

calcium dihydropyrophosphate and hydroxyapatite [6, 15]. Obduced by the protein in the synovial fluid they activate the complementary system.

A complex of cellular and vascular reactions appears in the synovial membrane typical for the hyperplastic inflammatory process. The developing synovitis and disorder of the functions of the synovial membrane are accompanied by accumulation of inflammatory or effusion in the joint cavity. Its qualitative and quantitative characteristics are strongly different from those of the normal synovial fluid. The effusion typically changes the pH of the medium, leads to accumulation of protein components and products of degradation of proteoglycans, activation of destructive enzymes, and reduction of concentration of the hyaluronic acid [15]. As a result, the rheological, trophic, and lubricating functions of the synovial fluid become strongly impaired.

On the other hand, when the inflammation mediators cause liberation of proteolytic and lysosomal enzymes macrophages, lymphocytes and plasmocytes in the composition of cellular infiltrates of the synovial membrane, the latter facilitate turning the synovitis into chronic and the processes of destruction of joint elements [6].

The neglected osteoarthrosis stages manifest how the synovial membrane gets sclerosed, secretion of the synovial lubricating fluid becomes deficient [8]. Cracks, fragmentation portions appear in the cartilage, fibers become loose and hydrophily drops (Fig. 6.3a). The process of destruction of the cartilage ends in its mechanodestruction (Fig. 6.3b).

The rheumatoid arthritis occupies a dominating place in the group of inflammatory disorders in joints. The rheumatoid arthritis is a systemic immunal

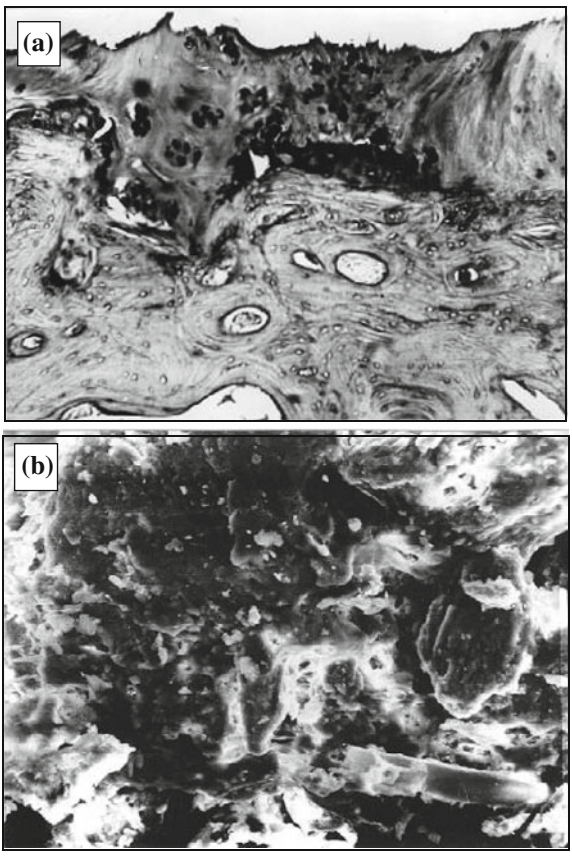


Fig. 6.3 Deforming osteoarthrosis: **a** pathomorphological changes in joint cartilage (1.5 × 1.2 mm, dyed with hematoxyl and eosin); **b** joint cartilage scan (300 × 250 μm)

inflammatory disease of the connective tissue; its main pathological manifestation is affliction of the locomotorium. It features fast evolution of osseal and cartilaginous destruction of joints among predominantly young patients (20–40 years) that leads to early and stable disability. Fifty percent of them become invalids 2–3 years after the rheumatoid arthritis begins [2, 6].

According to modern ideas, the inflammatory process during rheumatoid arthritis relates to immunopathological disorders of cellular and humoral immunity. However, its etiology and many pathogenetic mechanisms of development remain obscure. A triggering role of exogenic or endogenic antigens in this process is debated (the structurally damaged collagen, immunoglobulin, fragments of bacteria, micoplasmas, and viruses) [18]. Using these concepts, the experiment yielded a number of chronic arthritis among animals resulting from various immunological effects, but they reflect just individual links of the pathogenesis of the diseases [6, 18]. There is an adequate arthritis model identical to the rheumatoid process in man. That is why it seems impossible to formulate a single scheme of the rheumatoid arthritis pathogenesis and the like disorders at present.

The cartilage destruction in rheumatoid arthritis seems an intricate and multifactor process. The leading role belongs to the cellular and humoral immunopathological reactions that evolve in the so-called "locus minoris resistencia" [6]. These are borderline zones between the cartilage and the synovial membrane. Active destruction of the cartilage and then of the subchondrial bone develops through invasion by the degenerative pannus which is the vascular and fibrous tissue of the synovial membrane perichondrial portion. Degradation of the collagen and proteoglycans in the matrix occurs under the effect of free-radical oxidation of tissues, collagenolytic and lysosomal enzymes, such as macrophages and neutrophils, chondrocytes and synovicytes. A similar effect is produced in the cartilage by synovial effusion that the same causes turn into a medium "aggressive" to it. This process is maintained by persistence of antigens in all synovial tissues, including the cartilage. The products of degradation of proteoglycans released by the cartilage and acquiring antigenicity play apparently a definite role in evolution of immunopathological reactions.

A disorder of the hematosynovial barrier permeability and pathochemical processes in the synovial fluid change drastically its basic biophysical properties. It is facilitated by dystrophic changes in the cartilage, profound impairment of its biomechanical characteristics. Under these conditions, the cartilage easily undergoes both enzymic and mechanical destruction. Joint stiffness, one of early symptoms of the rheumatoid arthritis, is attributed to the disorder of lubricating parameters of tissues in articular pairs. The friction coefficient of a sick joint cartilage is supposed to increase sharply to 0.4 [19]. Therefore, the process terminates in premature wear of the cartilage, severe deformation of joints, appearance of a fibrous and an osseous anchylosis.

The known stereotype of development of inflammatory changes in articular tissues, irrespective of etiological factors, makes all chronic arthritis alike because of commonness of many link of the pathogenesis of destruction of joints. Proceeding from this fact a certain similarity of the synovitis during the osteoarthrosis and rheumatoid arthritis has been remarked.

Notwithstanding the variety of forms and features of evolution of rheumatic disorders, the articular syndrome a number of similar clinical symptoms; it is a combination of pain, symptoms of synovitis, joint stiffness and limited functions of joints.

Sensations of pain have a complicated and variegated genesis. A degenerative damage of the joint cartilage in the deforming osteoarthrosis manifests itself clinically by dull, aching pain of the "mechanical" type. It appears during ambulation and after exertion in the afternoon, as a rule. Friction and crackling in joints, restriction of motions due to pain appear in the late stages of osteoarthrosis. It is due to deficient lubrication in the joint, cracking and fissuring of contacting cartilage surfaces frequently reaching the subchondrial bone plate (Fig. 6.4).

The rheumatoid arthritis and similar inflammatory disorders have the pattern of «inflammatory rhythm» of articular pain coinciding with arthritic aggravation, it is permanent and pain intensifies in ambulation. Accumulation of excessive effusion, inflammation mediators, and products of the upset tissue metabolism, higher

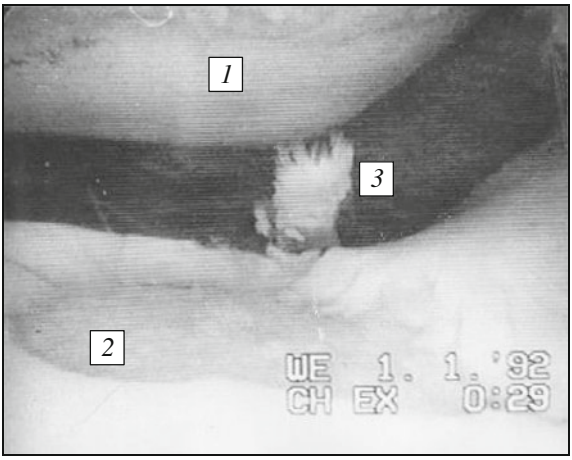


Fig. 6.4 Arthroscopic pattern of mechanodestruction of joint cartilage in knee joint medial portion during post-traumatic deforming osteoarthrosis: *1* pronounced cartilage thinning on sliding contact of hip condyle reaching subchondrial bone plate; *2* deep extensive cartilage erosion stopping at subchondrial bone plate contacting with hip shin-bone condyle; *3* products of wear of joint cartilage in joint cavity

intraarticular pressure and appearance of deep and extensive erosion of the cartilage integument play large role in pain induction. Pain receptors of vessels, the periosteum, the fibrous capsule, ligaments, and adipose bodies in the joint are irritated mechanically and chemically.

6.2 Basic Directions of Treating Joint Rheumatic Deceases

Treatment of rheumatic diseases is an intricate task because due to the above reasons no etiotrophic therapy is feasible so far. The therapeutic schemes are based on the known ideas about partial links of the pathogenesis of the disorders. Widely employed means and methods of therapy of rheumatic diseases are aimed predominantly at achieving the counterinflammatory and analgesic effect. That is why this symptomatic treatment achieves just a temporary inhibition of the process without eliminating its causes, let alone development and progress of cartilage destruction.

From this standpoint, one of the main principles of therapy should be effective protection of the cartilage against enzymatic destruction and premature wear. This trend of arthrology acquired the name of "chondroprotection" [20]. The main task for the next decade is to find means stimulating the reparative functions, correcting metabolic disorders and preventing cartilage destruction [20, 21].

Since rheumatic diseases evolve in a chronic relapsing manner, the known means of their therapy are palliative, hence their treatment should be combinatory

and long-term. It implies combined administration of medicamentous drugs, physiotherapy, preventive, and corrective orthopedics, physical exercises, massage, sanatoria, and spas. The main principles and features of the therapy of individual nosological forms of rheumatic diseases are quite well disclosed in the recent publications [6, 12, 22]. Therefore, it seems expedient to consider the most general means of treatment of rheumatic diseases from the viewpoint of evaluation of their chondroprotective effectiveness.

At present steroid and non-steroid counterinflammatory drugs are widely administered. Among them glucocorticoids are administered more often as natural hormones of the adrenal cortex or their synthetic analogs. They possess a broad spectrum of pharmacological effects; their mechanism is dictated by the physico-chemical properties, high trophicity, and their classification as lipids. The main therapeutical effect of the steroids is a pronounced counterinflammatory, antiallergic, and immunodepressive action [23].

The main way of administration of corticosteroids in rheumatoid arthritis is enteral. However, log administration of the drugs may cause complications due to immunity depression, disorders of the adrenal cortex functions and water—electrolyte metabolism. Moreover, glucocorticoids can inhibit the biosynthetic function of chondrocytes suppressing production of collagen and proteoglycans [23–25]. Clinical observations of the progress of cartilage destruction after prolonged steroid therapy confirm that these drugs produce no chondroprotective effect.

Another large group of medicamentous means is non-steroid counterinflammatory drugs (NCID). Their non-specific pharmacological effect favors their administration in arthrosis arthritis of any etiology. The NCID produce no hormonal effect of glucocorticoids and feature relatively good tolerance. They produced pronounced counterinflammatory, pain and temperature relieving effect by direct inhibition of the synthesis of prostaglandins, mono- and lymphoquins, histamine, etc. [6, 22]. These drugs can reduce the permeability of capillaries and restrain exhudative manifestations of inflammatory reactions; they produce a cytotoxic effect and depress proliferative processes tissues [26]. Some of them are capable to stabilize release of destructive enzymes.

Still, since the NCID are derivatives of various acids, they may cause side effects or provoke aggravation of chronic gastrointestinal disorders [27]. It has been demonstrated that prolonged administration of the NCID produces a clinical effect, but they impair metabolism in the cartilage and reduce the synthesis of proteoglycans [21, 28, 29]. Hence, it has been concluded that these drugs fail to prevent cartilage destruction during rheumatic afflictions [30].

An alternative approach to solving the problem was discovered in the early 60 s. Adoption of the so-called “true chondroprotectors” into the clinical practice (rumalonum, arteparonum and their analogs) containing proteoglycan complexes as the main biologically active substance has become a definite achievement of modern pharmacotherapy of rheumatic diseases.

Long-term clinical and experimental studies revealed that the drugs, due to their biochemical affinity with the sulfated proteoglycan matrix, possess special trophism to the arthrosis-altered cartilage structure and a capability to propagate in it.

Destruction of the main substance of the cartilage matrix is inhibited by rivaling inhibition of the enzymic system of lysosomes and activation of the synthetic functions of chondro- and synovicytes [20, 21, 25, 31, 32]. As a result, the trophism and metabolism of the cartilage normalize, the biosynthesis of mucopolysaccharides is stimulated, the processes of their depolymerization and elimination are inhibited. The drugs of these groups have shown correlation between the number of injections and higher concentration of the HYA in the synovial fluid [20, 30]. This fact favors both improvement of the tribomechanical properties of the cartilage and the synovial fluid and the functional abilities of the joint in general.

Prolonged rumalonum therapy of osteoarthrotic patients stabilizes annual workday losses, significantly reduces disablement cases compared with the patients' control group [20, 21, 30]. The most optimum is to treat early stages of disorders when the process of leakage of proteoglycans is still reversible. In such cases the positive results of administration of the drugs is observed in 63–70 % cases [21].

These data permitted to administer for a long period rumalonum and arteparum as basic therapy of the deforming osteoarthrosis. However, notwithstanding the relatively good tolerance of the drugs, side effects and complications can be observed. Since the therapeutical drugs contain biologically active parts of the cartilage matrix (peptides, glicoaminoglycans, nucleotides and nucleosides, chondrocytes, bone marrow cells, etc.) possessing antigenic properties, a severe complication is possible, such as an allergic reaction. That is why it is difficult to administer these therapeutical drugs in case of the rheumatoid arthritis [25]. Because the anticoagulation effect is similar to that of reparinum, they are counter indicated in case of stronger risk of a hemorrhage (hemorrhagic diatheses), diabetes mellitus, chronic afflictions of the cardiovascular system (hypertension, endocarditis, cardiosclerosis), ulcer diseases of the gastrointestinal tract, hepatic and renal pathology [20, 30].

In recent years, new therapeutical drugs appeared containing the cartilage matrix components and improving the cartilaginous tissue metabolism. In this connection, a new term emerged—the "drugs structurally modifying the cartilage". These therapeutical drugs include, e.g., structum or its analogs. The effective component of the drug is the sodium salt of chondroitin sulfate or the basic component of proteoglycans in the cartilage matrix. Now the chondroitin sulfate designates a group of structurally similar polysaccharides, as a rule, consisting of the sulfonated and unsulfonated remnants of glucuronic acid and N-acetylglucosamine. The chondroitin sulfate mechanism of effect is obscure. It is assumed that it is most probably due to the physico-chemical properties of the drug rather than to the biological ones. Experimentally it is established that the structum therapeutic effect is due to substitution of the chondroitin sulfate destroyed by catabolism in the cartilage and activation of the synthetic functions of chondrocytes inhibiting the destructive enzymes and further development inflammatory reactions in articular tissues. Experiments in vitro revealed that chondroitin sulfate is combined by chondrocytes and is included into the cartilage matrix composition. Chondroitin sulfate in the cultures of chondrocytes is capable to prevent the destructive effect of

interleukin-1 in the cartilage [33]. Nevertheless, it has yet been proven if incorporation of chondroitin sulfate can restore normal sulfonation of glucose aminoglycans in osteoarthritis [34]. The mechanism of counterinflammatory effect of chondroitin sulfate observed experimentally as protease inhibition of протеаз and reduction of decay of the cartilage matrix has not been explained so far [35, 36].

The true bioavailability of chondroitin sulfate is unknown. It is believed that in case of peroral administration the absorption of the intact low-molecular chondroitin sulfate is about 5 % and does not exceed 10–13 % [37–39]. According to the meta-analyses of over 20 studies the chondroitin sulfate drug has moderate effectiveness of pain relief and improvement of functions of joints in osteoarthritis among approximately 65 % of patients during 4 months of administration [40–42]. No convincing proof of structurally modifying properties of chondroitin sulfate has yet been obtained.

Structum as a therapeutic drug is produce in pills and intended for ingestion. It has been established that the drug can accumulate in the articular cartilage and synovial fluid preserving residual therapeutic effectiveness for 2 months after it is withdrawn. The structum is a harmless drug that practically produces no side effects. Therefore, it can be administered for prolonged therapy of osteoarthroses. Clinical research has revealed that administration of structum during 4 months reduces substantially the pain syndrome and improves functions of injured joints among most patients. Glucosamine is becoming attractive in recent years. A number of randomized studies have revealed that this compound produces a structure modifying effect in respect to the articular cartilage and symptomatic effects during therapy of osteoarthritis [40]. Most of the research of glucosamine dealt with its sulfate salt, some used glucosamine hydrochloride. The problem of the sulfate role in the pathogenesis of osteoarthritis has been discussed in publications. It is established that chondrocytes are very sensitive even to the least reduction of sulfate concentration, in this case the synthesis of glucose aminoglycans is upset [43, 44]. A positive effect of exogenic sulfate cannot be fully excluded.

Glucosamine is a monoaminosaccharide. It is synthesized in the body from glucose [45, 46]. The glucosamine molecular weight is 179.17, that of glucosamine hydrochloride is 215.63. Chemical concern Sigma-Aldrich produces glucosamine drugs as both hydrochloride and sulfate 98–99 % pure. The commercially produced crystalline drugs of glucosamine sulfate contain 78.5 % glucosamine on the average.

Unlike chondroitin sulfate, the mechanisms of effect of glucosamine sulfate are better known. It is believed that exogenically administered glucosamine sulfate intensifies the synthesis of proteoglycans being a substrate for the synthesis of chondroitin sulfate [47]. Glucosamine is a competitive inhibitor of glucose in the synthesis of chondroitin sulfate [48]. Glucosamine possesses immunity modulating properties. It prevents activation of T-lymphocytes in vitro [49]. Glucosamine produces a counterinflammatory effect on neutrophils, inhibits appearance of active forms of oxygen, hemotaxis and phagocytosis [50]. Thus, a certain progress of studies of the mechanism of effect of glucosamine sulfate has enabled to explain both the symptomatic and structure modifying effect of the drug.

According to the detailed review of pharmakinetics of glucosamine the bioavailability of glucosamine sulfate during peroral administration is 26 %, it approaches to 95 % after intramuscular administration [51]. It is believed that high doses of glucosamine are safe. It has been established that the therapeutic doses of the drug do not produce any significant effect on the metabolism of glucose [52, 53]. In the majority of clinical studies, the incidence of side effects during administration of glucosamine sulfate did not exceed that induced by a placebo [40].

Administration of glucosamine began in Europe in the 60s of the last century. A number of reviews dealing the clinical application of glucosamine report that the drug is an effective pain reliever and improves functions of knee joints in osteoarthrosis [41, 54–56]. The extent of improvement is comparable with that of non-steroid counterinflammatory therapeutical drugs with fewer side effects. It has been established in a number of recent studies that glucosamine sulfate inhibits the progress of osteoarthrosis, prevents restriction of articulation of the knee fissure [57–60]. This result is commonly achieved during administration of glucosamine sulfate in a dose 1,500 mg/day during 3 years [40, 57, 58]. Nevertheless, some researchers remark absence of any symptomatic effectiveness of glucosamine during prolonged administration [61].

Local therapy of the joint process is another necessary component of treatment of rheumatic diseases. In this connection, the effective component of ointments and gels administered for this purpose (butadionum, indometicinum, bruphenum, ketaprophenum, etc.) is attributed to the presence of NCID that both produces analgesic and counterinflammatory effects [6, 30].

A selection of drugs for intraarticular treatment of rheumatic diseases is highly limited and fails to satisfy the present-day requirements. Administration of corticosteroids is indicated primarily for stopping the rheumatoid or similar synovitis attacks and unacceptable for “dry” joints.

In case of the rheumatoid arthritis the drugs are administered intraarticularly that produce the immunodepressive effect (chlorineochinum, resochinum, delagilum) and cytostatics (cyclophosphanum) that perform the synoviortesis too. In the first case, the exhudative proliferative changes in the joint are reduced due to the cytotoxic and cytolytic effect on the synovial membrane. Synoviortesis (chemical synovectomy) relates to bloodless methods of destruction of the pathologically altered synovial membrane by intraarticular administration of radioactive colloids of gold, yttrium, radium, osmium oxides, etc. [62]. The arthritis is stopped by coagulating the synovial tissues followed b their restoration and partial sclerotherapy. Recently this method of treatment has lost its popularity because it has shown a significant destructive effect on the articular cartilage and a high incidence of arthritis relapses [63, 64].

Thus, at present there is a deficit of means and methods of general and local treatment of rheumatic diseases that would effectively protect the cartilage from destruction. Nevertheless, the features of enzymatic injury of the cartilage in rheumatic diseases are better studied and ways of overcoming this injury have been determined, meanwhile the processes of mechanodestruction of the cartilaginous

tissue and possibilities of its prevention have been studied insufficiently. From this viewpoint there are worthwhile efforts to create artificial synovial fluids (lubricants) intended to produce effect in this link of pathogenesis disorders.

6.3 Problems of Developing Artificial Lubricants for Local Therapy of Joint Deceases

The study of the problems of therapeutic correction of the articular tribological parameters in case of pathologies has a comparatively short background. The first efforts to develop artificial lubricants can be referred to the late 60s, the period of intensive studies of the mechanism of friction and lubrication of joints. By this time, the idea existed already that the lubricating effect of the synovial fluid was due to the hyaluronic acid [65, 66]. In case of rheumatic diseases, because of enzymatic destruction of the latter, the rheological characteristics of the synovial fluid become strongly impaired and its secretion becomes deficient [67]. Therefore, in order to compensate temporarily the lacking synovial fluid and to protect the cartilage against mechanodestruction, it was proposed to test several biocompatible synthetic materials simulating some properties of the natural synovial fluid, in particular silicon organic (silicone) fluids and water-soluble polymers [68–71].

The experimental study of the silicon organic fluid as artificial lubrication of joints yielded no significant results because it was discovered that there is no statistically significant difference between its friction coefficient and that of the saline solution [68, 70]. It can be explained by its extremely high (300 cSt) and constant viscosity that would cause considerable shear forces during relative motion of cartilages. It is quite apparent that such behavior of the silicon organic fluid affected negatively the process of friction itself and was intolerable during osteoarthrosis when the muscular apparatus is incapable to overcome any extra exertion of joints.

A possibility was also considered of using polymers with the characteristics resembling those of the compounds of HUA with protein, such as polyethylene oxide (Fig. 6.5) [69]. The latter dissolves in water easily, it has a high molecular weight (100,000–6,000,000 c.u.) that permit to obtain rather viscous solutions with small concentrations. A microscopic study of the polymer frozen on thee slide under loading revealed that its structure is similar to the structure of the natural synovial fluid under the same conditions. However, the poor lubricity and thermomechanodestruction in the tribological tests did not permit to use polyethylene oxide as an artificial lubricant for treatment of joint pathologies. The experiments demonstrated a sharp viscosity drop leading to impairment of the antifriction behavior of this substance. Sterilization of the polymer also presented certain difficulties.

Since the assumption that it would be possible to use polyethylene oxide, as an artificial lubricating material for joints did not prove true experimentally,

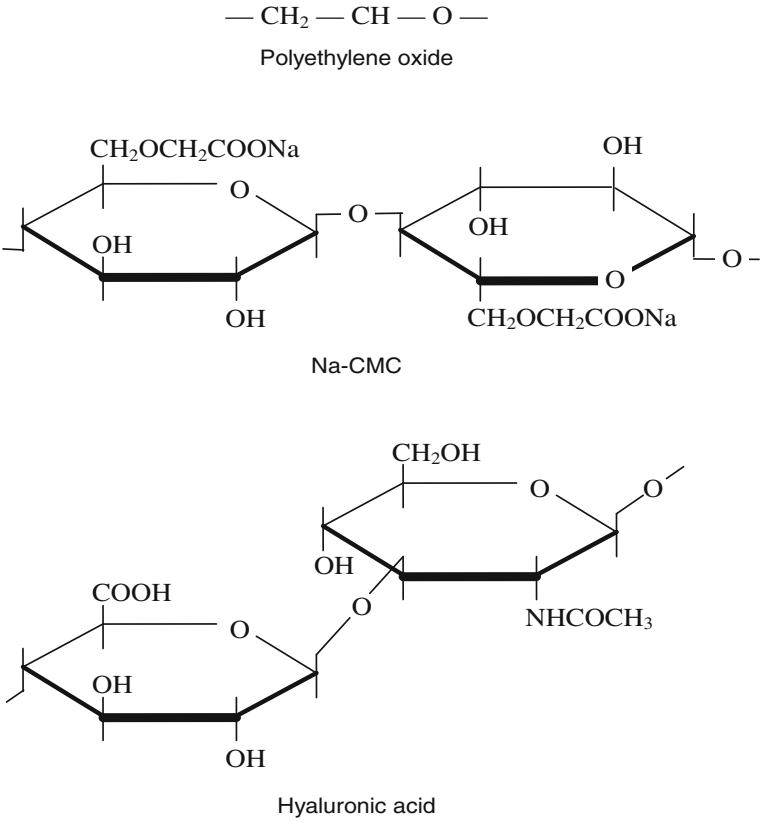


Fig. 6.5 Chemical structure of polyethylene oxide, sodium salts of carboxymethylcellulose and hyaluronic acid

researchers looked at another more complex compound. It was a sodium salt of carboxymethylcellulose (Na-CMC). It dissolves readily in water, biologically inert and low toxic, after a certain quality of purification it serves as an additive into foodstuffs. Aqueous solutions of Na-CMC have the alkaline medium and viscosity similar to the synovial fluid. Like all other compounds containing pyranose cycles, the structure of this natural polymer is similar to the HUA chemical structures (Fig. 6.5). Moreover, it was shown experimentally that Na-CMC yields insignificantly to mechanodestruction in friction in contrast to polyethylene oxide. As a result, the Na-CMC aqueous solutions served to develop a pseudosynovial fluid that, according to the patent [72], contained additionally saline and other components intrinsic for the blood plasma. However, the assumption of the researchers that it would be possible to use the latter expensively in the clinical practice as an artificial lubricant was not implemented because of its rather poor antifriction properties.

There had also been an effort of clinical application of polyvinylpyrrolidone (PVP) as a lubricating medium for therapeutic correction of articulation parameters in rheumatic diseases. It is a polymer of N-vinylpyrrolidone (γ-vinyllactam of N-aminobutyric acid). The PVP in the practical medicine serves as a blood substitute because of its low toxicity and absence of antigenic properties. This drug is known to be an agent prolonging the effect of antibiotics and hormones, a good desintoxicating and counterinflammatory means, a preservative of blood cells. Yet, they are not solely these properties that pointed to the possibility of PVP application for a new purpose.

It was shown experimentally that a 15 % PVP solution could simulate to some extent the rheological properties of the synovial fluid. It is believed that the latter can be improved significantly through joint application of this drug with HUA solutions. In this case other positive PVP properties are apparently implemented, such as the action of HUA as a protector against enzymic destruction in the pathological joint cavity. Whence followed an assumption that this drug can be used effectively as a lubricant and a means of suppression of negative effects of corticosteroids and an immunosuppressant during intraarticular administration.

However, the PP-containing drugs have negative in addition to positive properties. Their friction coefficient is an order of magnitude larger (0.05–0.08) than the extreme ultimate values typical for healthy joints. Besides, the PP-containing drugs, while having constant rheological characteristics, belong to Newtonian fluids; their viscosity does not depend in the shear rate, meanwhile natural synovial fluids belong to the thyrotrophic fluids; their viscosity, on the contrary, changes as the shear rate varies. Therefore, while the viscosity of the 15 % aqueous PVP solution is 1.8 MPa c and is determined exceptionally by the PVP concentration, the rheological characteristics of the natural synovial fluid depends typically both on the HUA concentration and the shear rate; therefore, the viscosity of the synovial fluid can vary within a rather broad range, vis. from 5–7 MPa s to several hundreds. Some difference is noteworthy between the pH values of PVP solutions (pH = 5.2…7.0) and the synovial fluid (pH = 7.4…8.2).

Hence, it can be assumed that the obtained experimental data about inhibition of cartilage degeneration in osteoarthrosis among animals are due rather to the counterinflammatory and immunoregulatory effect than the PVP lubricating properties. It is confirmed too in the work showing with the same experimental model a similar effect of PVP and dimethyl sulfoxide (DMSO) suppressing inflammation and inhibiting cartilage degeneration. The results of clinical tests that PVP intraarticular administration yields temporary improvement of functions of joints among 70–80 % osteoarthrotic patients, while it is less effective in rheumatoid arthritis.

The clinical practice has adopted in recent years the drugs based on the derivatives of hyaluronan (hyalhan, hyalart, synvisc, orthovisc) [73–79]. Hyaluronan is a polysaccharide (glycane) the molecules of which are long chains containing up to several hundreds or thousands of alternating disaccharides of two types: N-acetyl glycosamine glycane and glycouronate. Each glycouronate subunit has a negatively charged carboxyl group neutralized by Na^+, K^+, $Ca^{++,}$ and Mg^{++} ions [74]. The term ‘hyaluronan’ was introduced into biology and medicine for more precise

definition of the complex compound that is called hyaluronic acid in chemical scientific publications.

The drug hyalhan was among the first to be tested; its biologically active substance is a derivative of hyaluronan—hylan (a sodium salt of hyaluronic acid) with a moderate molecular weight (0.5–1.0 mln) [76]. The representatives of the new generation of these drugs are synvisc and orthovisc. The drugs contain in the saline solution (pH = 7.2) liquid hylan A and gel-like hylan B with a high molecular weight (6 mln.), hence the elasticity and viscosity excel these characteristics of the natural synovial fluid.

The drugs are intended for intraarticular administration in order to substitute and replenish the synovial fluid in osteoarthrosis. In fact, it is hard to refer them to truly therapeutical means because they are just biological analogs of those substances with the inherent properties of the synovial fluid of a healthy person. After injections, the hyaluronan derivatives cannot stay in the joint cavity for a long time and can hardly maintain the optimal rheological properties synovial fluid independently for a long term. Moreover, in case of osteoarthrosis they can be vulnerable to enzymic and free radical degradation in the joint cavity [74]. Therefore, the positive clinical effect of the drugs is significantly attributed to the response of synovicytes to their administration and production of a fuller proper hyaluronan [74, 75, 78–80]. The pharmacological activity of the hyaluronan derivatives is explained also by the fact that many cells have receptors of these substances maintaining interrelations between and protection of cellular membranes from inflammation mediators and free radicals [74]. Hyaluronan in experiments inhibits hemotaxis and migration of leukocytes, reactions of phagocytosis, proliferation of lymphocytes and adhesion of monocytes, reduces the synthesis of prostaglandin E2 and bradykinin, modulates secretion of cytokines and inflammation mediators [75].

The therapeutical effect of the drugs based on hyaluronan derivatives is attributed to improvement of the rheological properties of articular lubrication and the physiological status of the intraarticular tissues, their capability to protect the articular cartilage from mechanical and enzymic destruction. The effect is observed after administration of five doses of hyalhan or three doses of synvisc with a 1-week interval. The positive clinical effect of the drugs persists up to 6–8 months. It is observed as reduction of the intensive pain syndrome and improvement of the joint functional status among 76–77 % patients. Better results are achieved at early stages of osteoarthrosis [76].

The drugs are capable to induce allergic reactions. Therefore, they should not be administered during aggravations of osteoarthrosis accompanied by the intraarticular effusion, in case of the arthritis of the inflammatory genesis, or injected into the circumarticular tissue, to administer to the patients with allergic reactions to proteins in the anamnesis, to pregnants or children. Cases are known when short-term pain syndromes, discomfort in joints and intraarticular exudation occurred after injection of the drugs.

Thus, the following facts characterize the early stages of development of pseudosynovial fluids to protect cartilages from destruction by rheumatic diseases.

All the proposed models of the synovial fluid replicate just in part its chemical (the pH of the medium) or physical (rheological and structural) properties. They can appear only under certain specific conditions (loads, shear rates, etc.); it is not enough to achieve low friction during a dynamic contact of cartilages in the joint where these conditions are variable within a rather broad range. The artificial lubricants are known that contain none of the components inherent to the natural synovial fluid; they are, in the first place, those that are responsible for the anti-friction characteristics. So they can be considered only as temporary materials in the interface between rubbing cartilage surfaces devoid of the natural lubricating properties. Therefore, the only more realistic way of development of effective artificial lubricants is to create them on the basis of comprehensive analysis of the synovial medium in joints in order to identify and later adopt exactly those components that are truly capable to affect actively the biological mechanisms of reduction of the intraarticular friction and processes of mechanical and enzymatic destruction of the joint cartilage.

From this standpoint, it should be acknowledged a significant achievement of modern pharmacology and medicine that the drugs have been developed and comprehensively studied that are based on the hyaluronan derivatives similar to the natural synovial fluid. This trend of studies coincides with our trend and they both enable to expand the arsenal of means containing biologically and functionally active components of the synovial fluid for effective local therapy of disorders of joints [80, 81].

6.4 Liquid-Crystal Lubricants for Treating Joint Deceases and Similar Pathologies

Now it is certain that similarity of the tribological, rheological and structural properties of artificial lubricants with the natural synovial fluid is the most essential provision of successful replacement of the fluid in case of joint pathologies [82, 83]. This similarity of the properties when the deficit of the synovial fluid is to be replenished is capable both to ensure the same pattern of artificial and natural lubrication of joint cartilage surfaces in static and dynamic conditions and to influence actively the tribomechanical parameters of joints in rheumatic diseases. In this connection, using the accumulated experimental material and its theoretical explanation, it has been assumed that introduction of medicamentous means based on aqueous solutions of high-molecular polymers with thixotropic properties into the zone of friction of cartilages when articular lubrication is lacking, i.e. such that resemble the rheological properties of the natural synovial fluid and liquid crystalline cholesterol compounds capable of planar orientation in the joint cartilage contact, producing the mesophase within the range of physiological temperatures, the friction of cartilaginous surfaces will be identical to that under natural physiological conditions. This assumption was validated experimentally.

Artificial lubricants were studied simulating the structural, rheological, and antifriction properties of the natural synovial fluid by introducing liquid crystalline substances into the aqueous Na-CMC solution in ratios determined empirically. A natural joint lubricant or a mixture of the liquid-crystal cholesterol compound (LCCC) with the mesophase within the range of physiological temperatures was used as a compound that would produce a mesomorphous condition of the artificial synovial fluid.

The analysis of the above model systems with the method of polarizing microscopy revealed that the pattern of their behavior is analogous to that of the synovial fluid. The structure of the dehydrated specimens of the synovial fluid and the artificial lubricants produced in the described manner turned out identical and were dendrites regularly distributed over the contact glass (Fig. 6.6).

The rheological properties of the drugs based on Na-CMC and LCCC aqueous solutions demonstrated that variations of the Na-CMC concentration enables to achieve the viscosity practically similar to the viscosity typical for the specimens of the natural synovial fluid. The optimum Na-CMC concentration in the compositions in question and the viscosities at different shear rates are given in Table 6.1 in comparison with the natural synovial fluid.

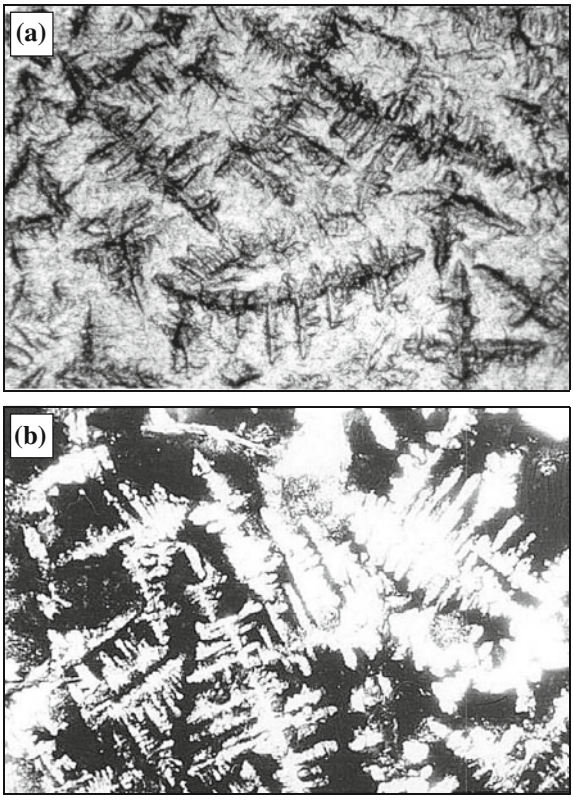


Fig. 6.6 Structure of dehydrated specimens: **a** synovial fluid; **b** artificial synovial fluid (410 × 300 μm)

Table 6.1 Rheological properties of drugs based on Na-CMC and LCCC solutions at 37 °C

Studied lubricating media	Viscosity of studied specimens, MPa s, at shear rates (s^{-1})		
	3.0	145.8	1312.0
2 % Na-CMC solution	54.7	15.9	7.0
2 % Na-CMC solution with addition of LCCC	54.7	16.9	7.2
Synovial fluid	54.7	15.3	7.0

Table 6.2 Tribological properties of the drugs based on Na-CMC and LCCC aqueous solutions in friction of articular cartilages

Lub medium	Lubricating medium composition	Time of full attenuation of pendulum oscillations *t*, c	Mean friction coefficient *f*
1	2 % Na-CMC solution	57	0.049
2	2 % Na-CMC solution with addition of LCCC	72	0.027
3	Natural synovial fluids with general cholesterol concentration 0.9 mol/l	70	0.031

These drugs were tested in cartilage-on-cartilage friction pairs with a pendulum tribometer [81, 84, 85]. Results were compared with antifriction properties of the natural synovial fluid and the 2 % Na-CMC solution.

Table 6.2 shows the obtained experimental data. It is apparent that the drugs with the Na-CMC aqueous solution and the LCCC additive yield the friction coefficients comparable with those of the friction natural synovial fluid (variants of lubricating media 2 and 3).

Any reduction or increase of the experimentally established concentration of the liquid crystalline compounds would impair the tribological properties of the concerned artificial lubricant. Hence, the experimentally selected composition of Na-CMC and LCCC aqueous solutions was optimum in the tribological and rheological respects and it was used in further experiments.

A positive effect of the LCCC on friction interaction was confirmed by the study of cartilage friction on glass in an end-friction unit at a sliding velocity 0.1 m/s within a range of loads 0.2–10.0 MPa (Table 6.3) [81, 84, 86, 87].

The results of the experiments in Table 6.3 clearly show that the 2 % Na-CMC aqueous solution is characterized during friction of the cartilage on the glass by sufficiently high and unstable friction coefficients. Introduction of the LCCC into the Na-CMC aqueous solution improves substantially the antifriction properties of the compositions in question. It is a typical fact that the cartilage-glass friction pairs behave identically in the model lubricating systems with the addition of LCCC and like the natural synovial fluid both with statistically indiscriminate friction coefficients and when the friction coefficients depend similarly on loading and the coefficients in both lubricating media coincide when the loading is the same

Table 6.3 Tribological properties of the drugs based on Na-CMC and LCCC aqueous solutions in friction on glass

Load p, MPa	Variants of lubricating media								
	Synovial fluid			2 % Na-CMC solution with addition of LCCC			2 % Na-CMC solution		
	f	Δf	Δf (%)	f	Δf	Δf (%)	f	Δf	Δf (%)
0.2	0.014	0.002	17.9	0.015	0.004	30.3	0.33	0.12	36.2
0.6	0.009	0.0016	16.8	0.009	0.0023	24.3	0.21	0.06	29.3
1.0	0.009	0.0015	16.7	0.009	0.0017	20.1	0.15	0.09	66.2
1.5	0.008	0.0006	7.7	0.008	0.0007	8.1	0.08	0.06	79.7
2.0	0.009	0.0025	27.3	0.010	0.0011	11.6	0.07	0.05	66.2
2.5	0.010	0.0045	44.7	0.010	0.0027	27.4	0.07	0.06	95.1
3.0	0.011	0.0012	11.8	0.011	0.0040	35.8	0.08	0.07	92.7
4.0	0.011	0.0025	23.4	0.011	0.0032	31.6	0.11	0.09	90.4
5.0	0.011	0.0032	29.3	0.011	0.0032	29.3	–	–	–
7.0	0.013	0.0028	22.1	0.012	0.0016	13.7	–	–	–
9.0	0.014	0.0037	27.7	0.014	0.0045	32.7	–	–	–

p = 1.5 MPa. This abnormal behavior of the friction coefficients when they pass through the minimum in the cartilage—glass pairs in the presence o the new artificial lubricant has been achieved for the first time because the artificial synovial fluids are known not to produce this effect.

The assumption that partial wear of the cartilage surface zone while retaining the inherent regular structure should not produce any significant effect on its antifriction properties, was validated experimentally during friction of the articular cartilage on glass in the presence of distilled water, natural and model (2 % Na-CMC aqueous solution with LCCC) synovial fluids [82].

The tests were performed in the following manner. First, the antifriction properties of cartilage specimens were assessed in the presence of the synovial fluid. If the friction coefficients did not exceed 0.015, the specimens were accepted for further tests. Otherwise new specimens were made. Then the synovial fluid was replaced with distilled water and the specimens were tested for friction under a constant load p = 1.0 MPa during 30 min. The distilled water initiates wear of cartilaginous surfaces simulating their lubrication in pathological conditions. After that, the cartilage rubbed on the glass in the natural and model synovial fluids. Values of friction coefficients were compared with those obtained at the initial testing stage. Figure 6.7 presents the results.

Their analysis indicates that substitution of the synovial fluid with distilled water sharply boosted the friction coefficient to f = 0.2…0.3, crackling appeared typical for joints with rheumatic diseases. The friction coefficient changed insignificantly in time. The cartilage specimens demonstrated deformation in the direction of sliding. The rubbing surfaces showed debris in the form of scales (0.3–0.6) × (0.3–4.0) × (3.0–4.0) μm in size (Fig. 6.8). However, the structure of surface zones with

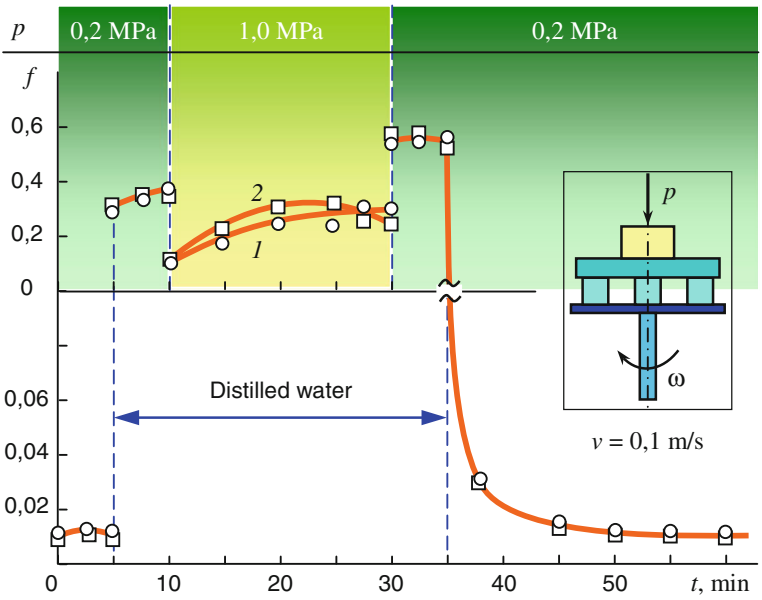


Fig. 6.7 Friction coefficient of cartilage on glass during lubrication with natural (*1* circle) and model (*2* square) synovial fluids. Distilled water; MPa

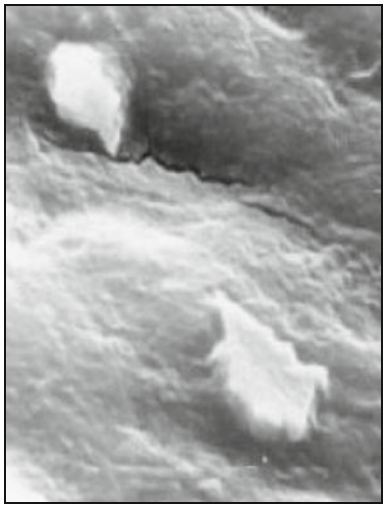


Fig. 6.8 Surface of cartilage after friction in presence of distilled water (15 × 20 μm)

the inherent tangential arrangement of collagenous fibers persisted proving that the wear of the cartilaginous surfaces was insignificant. Substitution of distilled water with the natural synovial fluid would reduce the friction coefficient back to the level

Table 6.4 Tribological and rheological properties of drugs based on Na-CMC solutions in PVP and LCCC

Relevant lubricating compositions	Viscosity, MPa s, at shear rate (s^{-1})			Friction coefficient under load (N)	
	3.0	145.8	1312.0	6.7	67
15 % PVP aqueous solution	1.8	1.5	1.3	0.05	0.05
2 % Na-CMC solution in hemodesis with addition of LCCC	55.0	17.5	8.0	0.025	0.015
Synovial fluid	54.7	15.3	7.0	0.025	0.01

observed among the unworn cartilage specimens. The new artificial lubricant based on the 2 % aqueous Na-CMC and LCCC solution of yielded similar results.

Hence, an important conclusion follows that timely administration of these medicamentous drugs simulating the lubricating behavior of the natural synovial fluid when the cartilages are partly worn or the joints are afflicted by rheumatic diseases will favor restoration of their tribological properties.

Since development of many rheumatic diseases, rheumatoid arthritis in particular, is accompanied with aggravations with very intensive inflammatory and immune reactions in joints, an additional variant of the artificial synovial fluid was developed. It is a mixture of the LCCC with 2 % Na-CMC solution in the PV, pp. The latter possesses counterinflammatory, detoxicating and immunomodulating and is capable to prolong the effect of corticosteroids.

Table 6.4 presents the antifriction and rheological characteristics of this variant of the artificial synovial fluid.

Table 6.4 shows that this variant of the artificial synovial fluid like the previous one has the rheological and antifriction characteristics similar to the natural synovial fluid.

Figures 6.9 and 6.10 present the temperature dependencies of the AHC fluorescence intensity of these two forms of artificial lubricants containing LCCC and intended for intraarticular application.

They show clearly that the new model lubricating system undergo phase transformations within the range of temperatures adequate to the interval of high and low temperatures of the specimens of the natural synovial fluid (20–47 °C). Because it is known that the structure of liotropic liquid crystals in biological fluids is governed by the concentration of water and cholesterol compounds, the obtained experimental data prove that the new drugs after dilution behave similarly to the natural synovial fluid. The new artificial synovial fluids are intended primarily for local intraarticular therapy of joint disorders.

However, it was assumed that in addition to the intraarticular application of the above types of artificial lubricants LCCC can be without invasion administered into the injured zone of the tendomuscular apparatus and joints by diffusion through the coverlets from the LC solutions. The assumption was based on the publications reporting that there was this type of transport and about the positive effect of local

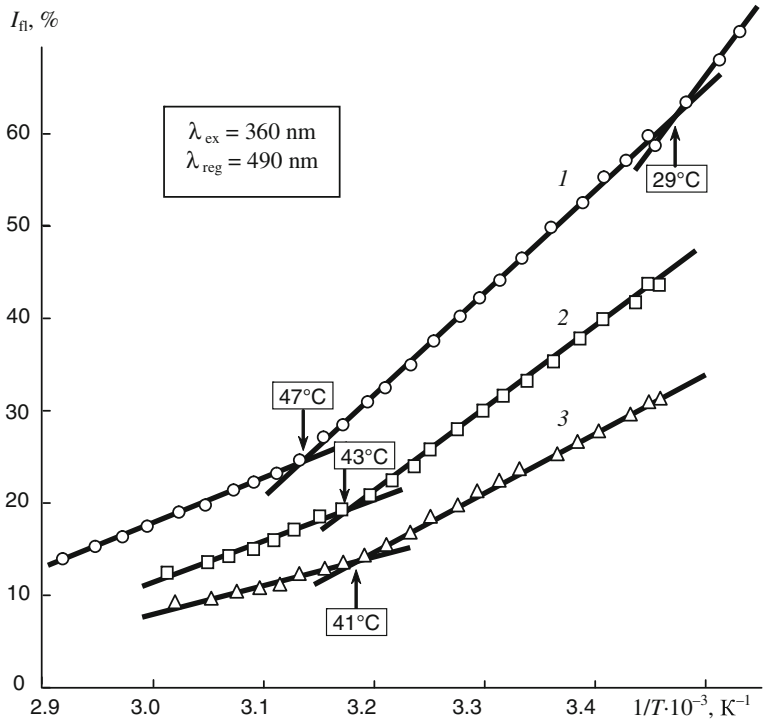


Fig. 6.9 Dependence of intensity of fluorescence of the drug-combined artificial compound containing Na-CMC aqueous solution with LCCC addition, on reverse temperature: *1* 100 µl of preparation + 3 ml of 0.15 M NaCl; *2* 50 µl of preparation + 3 ml of 0.15 M NaCl; *3* 25 µl of preparation + 3 ml of 0.15 M NaCl

curative mud application that contain cholesterol compounds in addition to some biologically active compounds [88].

Since it is a simple application of medicamentous means and has large advantages over out-patient therapy, a drug and a method of arthrosisoarthritis treatment were developed by application of LCCC solutions in the therapeutical vaseline on the coverlets of the afflicted joint [83]. A LCCC mixture with the mesophase within the range of physiological temperatures served as the base of the liquid-crystalline preparation. These liquid-crystalline substances quite available in the animate nature; they are found in the composition of some food products and serve in medicine for temperature diagnostics. The experimental data quoted above have demonstrated that introduction of these LCCC into artificial synovial fluids and synthetic oils improve their antifriction characteristics.

The capability of diffusion of the new composition through the human skin was investigated using skin specimens of a corpse or a patient after amputation of an afflicted extremity. A skin patch was fixed to the bottom of a glass cylinder placed in a beaker to test for diffusion. The cylinder was spaced a bit from the beaker

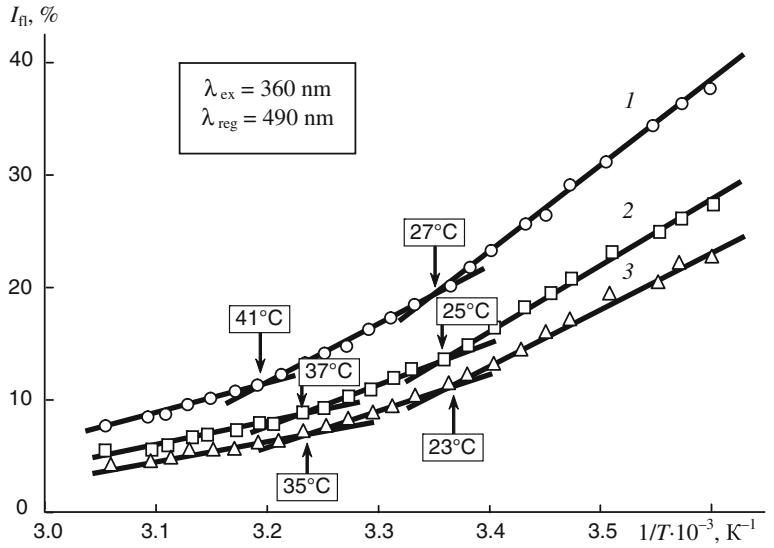


Fig. 6.10 Dependence of fluorescence intensity of drug-combined AHC containing Na-CMC solution in PVP with LCCC additive on reverse temperature: *1* 100 µl of preparation + 3 ml of 0.15 M NaCl; *2* 50 µl of preparation + 3 ml of 0.15 M NaCl; *3* 25 µl of preparation + 3 ml of 0.15 M NaCl

bottom so that it was immersed into the LCCC solution in the vaseline oil. Vaseline oil without LCCC was placed in the cylinder above the skin patch, it was periodically sampled and checked if it had any liquid crystals with a UR-20 spectrophotometer by variations of the optical density of the absorption band within the range 1750 cm^{-1} typical for the oxygen atom oscillations in the ester link. The diffusion was checked at 36.7 °C. The results yielded satisfactory diffusive capability of the LCCC in question through the skin (Table 6.5).

Thus, the established role of the LCCC in the mechanism of reduction of intra-articular friction served a plausible ground for development and creation of both invasive and non-invasive forms of LCCC-based drugs for the therapy of various articular pathologies together with the method of their preparation. A thorough review of scientific publications and patents has revealed that the above intra-articular forms of the new LCCC-based drugs have an unprecedented composition and mechanism of the lubricating effect. In addition, an active general biological role of the experimentally discovered and selected LCCC and a possibility of a positive effect of their application to treatment of articular disorders were supposed. In order to validate these assumptions preclinical tests of the drugs in question were performed with animals.

The capability of the LCCC to diffuse the coverlets and a possibility to be transported in joints were studied experimentally using two groups of albino common male rats weighing 265–290 g [83, 85, 89]. The first (12 animals) group was tested for LCCC diffusion in the state of the physiological norm. The adjuvant

Table 6.5 Results of study of LCCC diffusion through the skin

Time of sampling of tested compositions (h)	LCCC concentration в in sample, % by mass
24	0.53
48	3.71
72	7.75
96	9.61

arthritis of the left extremity joints was modeled in the second group (12 animals) by administration of 0.1 ml of the Freind adjuvant (Grand Island Biological Company, USA) beneath the plantar aponeurosis. Examination was performed between the 14th and 21st days when inflammatory and degenerative changes in the region of the joints became pronounced. Implementation of this and all subsequent experiments, euthanasias of ether-narcotized animals were governed by the generally practiced requirements.

The tested drug was a mixture of 1,2^{3}H-labeled of the oleic acid cholesteryl ester (the specific activity of the preparation is $7.5 \pm 0.5 \cdot 10^{6}$ decays/min mg) with the vaseline oil. Daily applications of the fresh prepared preparation to the injured left knee joint in a quantity 0.5 ml with a 60 min exposure were performed under narcosis. A 20 % sodium oxybutyrate solution in a dose 1.5 g/kg served for narcotization. To improve accuracy due to penetration of the preparation into the gastric tract after the narcosis ended the joint was rubbed with chloroform and water.

The animals were withdrawn from the test by three in each group after 1, 3, 5 and 7 days and after 3, 5, and 7 applications, respectively.

The following specimens of tissues were studied: blood, liver, kidneys, anterolateral portions of the joint capsule, hip joint cartilage condyles. The radioactivity of the drugs was measured three times with a liquid scintillation method using a Mark-3 instrument produced by Searle Co. (Netherlands). The count efficiency was checked with the method of external standard and it was 40–50 %. The results were analyzed by the specific radioactivity (SR) of the drugs determined by the ratio between the true radioactivity of the preparation and the tissue weighed sample in decays/min mg. The quantity of the isotope label in the tissue was determined by the relative specific radioactivity (RSR); to calculate it the SR of the tissue preparation was compared with the SR of the introduced radioactive substance and the ratio was expressed in percentage.

Table 6.6 present the results of the radioisotope studies. Their analysis pointed to the following.

The healthy animals revealed diffusion of the preparation through the coverlets marked by appearance of the radioactive substance in the blood in inconsiderably growing concentrations. Its accumulation is observed in the joint cartilage and the joint capsule, liver and kidneys depending on the duration of the experiment.

The obtained data prove convincingly that the LCCC moves intraarticularly by diffusing through the coverlets.

Table 6.6 Results of radioisotope studies of diffusion of 1,2[3]H-labeled cholesterol esters through coverlets of pronounced

Experi-ment duration (days)	Group[a]	Radioactivity of drug, decays/min per mg tissue				
		Hip condyle cartilage surface	Joint capsule	Liver	Kidneys	Blood
1	1	3.4 ± 1.0	4.6 ± 1.0	5.9 ± 1.0	3.7 ± 0.4	559 ± 150
	2	20 ± 15	37 ± 20	35 ± 17	34 ± 13	972 ± 420
3	1	4.2 ± 3.0	3.9 ± 2.0	7.6 ± 0.5	3.0 ± 0.4	621 ± 210
	2	32 ± 17	46 ± 21	53 ± 9	32 ± 8	287 ± 135
5	1	7.5 ± 4.0	9.0 ± 3.0	25 ± 2	26 ± 14	722 ± 365
	2	33 ± 17	28 ± 1	80.0 ± 0.4	43 ± 17	294 ± 147
7	1	14 ± 6	5.0 ± 3.0	25 ± 5	71 ± 40	1267 ± 620
	2	5.5 ± 3.0	29 ± 13	36 ± 2	25 ± 7	294 ± 147

[a]*Group 1* Physiological norm; *group 2* adjuvant arthritis

In adjuvant arthritis, the diffusive ability of the preparation is stronger than that of the animals in the control group. The blood SR leaps sharply after 1 day, then it declines and stays practically unchanged throughout the experiment. The radioactive substance accumulates faster and in larger quantities in the joint cartilage and capsule, liver and kidneys. In this case, acceleration of the LCCC transport can be quite explained by a higher permeability of the joint capsule determined by the arthritis pathogenetic features.

The analysis of the drug RSR in the cartilage and capsule has revealed a statistically valid figure of the percent of incorporation of the radioactive LCCC into these tissues in adjuvant arthritis compared with the physiologically normal animals. The maximum label accumulation is observed after 5 days in adjuvant arthritis and after 7 days in healthy animals, after 3 days in the joint capsule in the animals of both groups (Fig. 6.11).

Thus, the established fact LCCC diffusion through coverlets and the discovered phenomenon of their selective intraarticular transport confirm the feasibility of LCCC application to the region of afflicted joint. Judging by the results of the experiment, the optimum number of applications is 5–7 days.

The dynamics and paths of elimination of the active substance of the new artificial lubricating LCCC-based drugs from the joint cavity after intraarticular administration were studied in 69 common male rats weighing 240–285 g with adjuvant arthritis in the left extremity joints. On the 21st day after the beginning of the joint process all the animals were injected 0.1 ml of the artificial lubricant containing the Na-CMC aqueous solution with a LCCC additive into the afflicted left knee joint. The liquid-crystalline preparation contained the oleic acid cholesteryl ester labeled with 1-^{14}C (the SR of the preparation was $2.3 \pm 0.2 \times 10^6$ decays/min mg).

The animals were withdrawn from the experiment in series of five after 1, 2, 3, 4, 5, 6 and 7 days, in a series of six after 10 days, and in a series of seven after 14, 21,

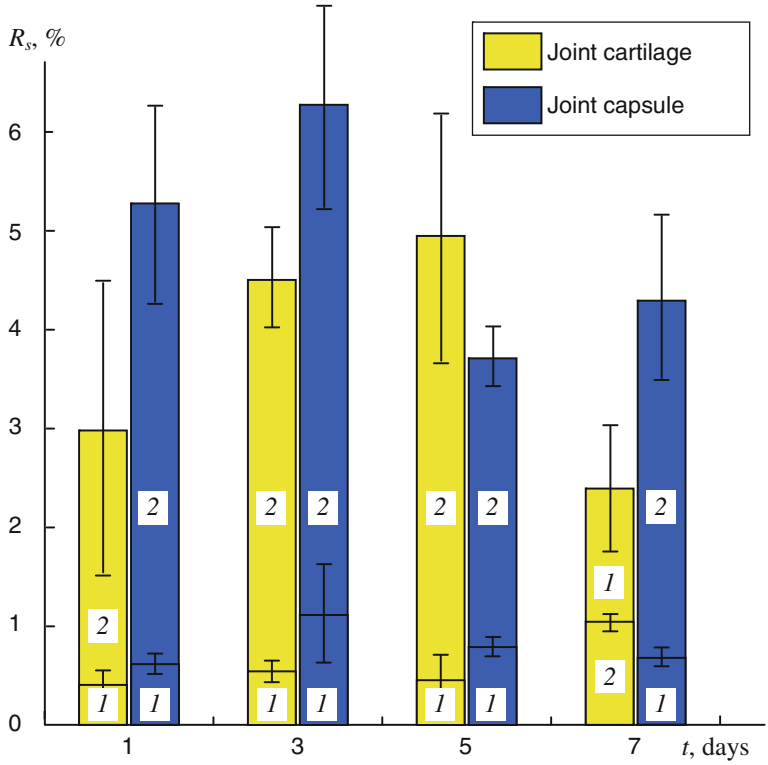


Fig. 6.11 Relative specific radioactivity of joint cartilage and joint capsule in animals: *1* physiologically normal; *2* adjuvant arthritis of articular cartilage

30 and 45 days. The specimens of the following tissues were examined: blood, liver, kidneys, joint capsule, and cartilage integument of the hip condyles. The radioactivity of the drugs was measured with the above-described method. The count efficiency was 80 %. Table 6.7 presents the results.

The accumulation of the isotope label was observed in the hip articular contact condyles where its elevated concentrations were remarked during the first 14 days. The maximum SR in the tissue specimens was registered between the 3rd and 6th days. The SR in the cartilaginous tissue begins to reduce on the 4th day, yet small radioisotope label inclusions were traced during 45 days.

The drug was eliminated from the joint cavity through the capsule where it accumulated during the first 6 days. Afterwards its concentration declined gradually. Appearance of the radioactive label in the blood was registered since the first days. Its relatively constant concentration was observed during 10 days and it was rising beginning on the 14th day of the experiment.

The drug was utilized in the liver where the maximum incorporation of the radioactive label was registered during the first 5 days, and then its concentration

Table 6.7 Results of radioisotope studies of routes and rates of elimination of the artificial lubricant based on LCCC labeled with 1-^{14}C from the knee joint cavity in animals with adjuvant arthritis

Experiment duration (days)	Number of animals (*n*)	Radioactivity of drug, decays/min per mg tissue				
		Hip condyle cartilage surface	Joint capsule	Blood	Liver	Kidneys
1	5	89.7 ± 18.9	9451 ± 2215	11.7 ± 1.4	9.3 ± 4.1	4.6 ± 1.4
2	5	317.8 ± 122.6	3153 ± 2190	12.2 ± 1.6	4.5 ± 0.6	7.2 ± 1.3
3	5	229.9 ± 77.3	13,559 ± 3507	32.3 ± 12.6	4.8 ± 1.2	5.7 ± 2.1
4	5	259.4 ± 16.0	5862 ± 2184	30.5 ± 13.0	6.7 ± 1.2	2.9 ± 0.2
5	5	303.3 ± 20.4	11,243 ± 1790	18.1 ± 3.9	9.6 ± 0.7	9.3 ± 1.4
6	5	399.8 ± 25.5	10,420 ± 3103	9.0 ± 1.5	3.4 ± 0.8	11.0 ± 3.9
7	5	339.1 ± 21.8	5183 ± 1793	9.4 ± 1.2	3.3 ± 0.6	6.6 ± 0.9
10	6	222.2 ± 52.0	4136 ± 400	78.7 ± 9.2	4.0 ± 0.5	2.7 ± 0.4
14	7	72.4 ± 9.7	5233 ± 1144	58.2 ± 9.8	3.7 ± 0.7	4.1 ± 0.5
21	7	35.8 ± 8.6	3880 ± 1180	84.4 ± 8.0	2.8 ± 0.3	3.8 ± 0.5
30	7	53.4 ± 21.6	7471 ± 1512	73.4 ± 10.4	6.1 ± 2.5	12.7 ± 7.7
45	7	64.0 ± 15.9	3554 ± 1233	43.8 ± 5.0	2.5 ± 0.5	6.0 ± 1.0

gradually reduced. The preparation was eliminated via the kidneys where the SR in the tissue was maximum during the first 7 days and on the 14–30th days.

As a result, it has been established that the artificial LCCC-based lubricant drugs after intraarticular administration into the knee joint with a pronounced affliction of articular surfaces can accumulate and persist in elevated concentrations during 12–14 days. This fact guided investigation of the local biological effect of the new LCCC-based artificial lubricants drugs for intraarticular application.

The effect of the drugs in the cartilaginous tissue in adjuvant arthritis was studied in 60 common male rats weighing 349–366 g.

The experiments were carried out in the chronic arthritis stage after 21 days since it started until rats destructive changes appeared in joints. To this end 18 animals were injected a dose of 0.1 ml of the artificial lubricant containing the Na-CMC aqueous solution with the LCCC additive (12 rats) once and three times (6 rats) with an interval of 12 days into the cavity of the afflicted left knee joint. Six more rats were injected three times doses of 0.1 ml of another artificial lubricant containing an aqueous LCCC solution in Na-CMC and PVP.

The skin LCCC applications in the Vaseline oil to the afflicted knee joint during 7 days were tried in 12 experimental animals.

Six animals more served to study the effect of the 15 % PVP solution regularly on the joint tissue used in clinical practice. The PVP was injected intraarticularly in a dose of 0.1 ml four times once in a week.

The control groups of animals numbered 18 rats that where withdrawn the experiment by 6 animals together with the experimental rats after 21, 28 and 51 days.

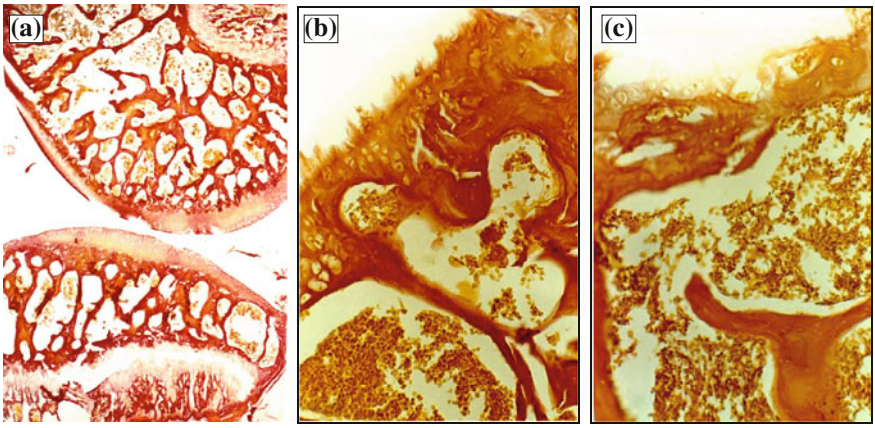


Fig. 6.12 Adjuvant arthritis of tibiafemoral joint of rats (control group of animals): **a** hip articular contact (1.2 × 1.7 mm). Hip articular cartilages of rats on days 21–28 (**b**) and 51 (**c**) after beginning of adjuvant arthritis (500 × 700 μm). Dying of Van Gison

The knee joints of the rats were studied with total histological preparations of tibiafemoral joints in the saggital plane passing through the zones of maximum contact between articulating joint cartilages. The drugs were dyed with hematoxylin and eosin, illuminated and studied under microscope. The results were the following.

Minor changes in the control group of animals were on the hip condyles and kneecap. The cartilage had a rough surface and irregular thickness. There were spots without the coverlet, only a deep zone remained (Fig. 6.12a).

At later stages of the arthritis there were spots of rough loose fibers and pronounced fissure of the cartilage, sometimes the subchondrial bone plate was exposed (Fig. 6.12b, c). Erosion in some spots was filled with the connective tissue. The cartilage matrix had irregular pink coloring. The majority of cartilage cells (chondrocytes) had tiny poorly colored nuclei and disordered nucleus-cytoplasmatic ratio. Some cells missed nuclei. Single isogenous groups of chondrocytes were observed just in few spots.

This histological pattern evidenced a pronounced dystrophic process, intensified enzymic and mechanical destruction of the cartilage typical for chronic arthritis.

The control group after a single and triple administration intraarticularly of the artificial lubricant containing the Na-CMC aqueous solution with the LCCC additive in the chronic adjuvant arthritis stage (Figs. 6.13 and 6.14a, respectively) the integumentary hyaline cartilage manifested a regular thickness; a sufficiently wide surface layer was preserved.

The joint surface cartilage was smooth with clear-cut edgesa. The interstitial substance of the cartilage was colored more homogeneously. The cartilage cells were large, with a high nucleus-cytoplasmatic ratio.

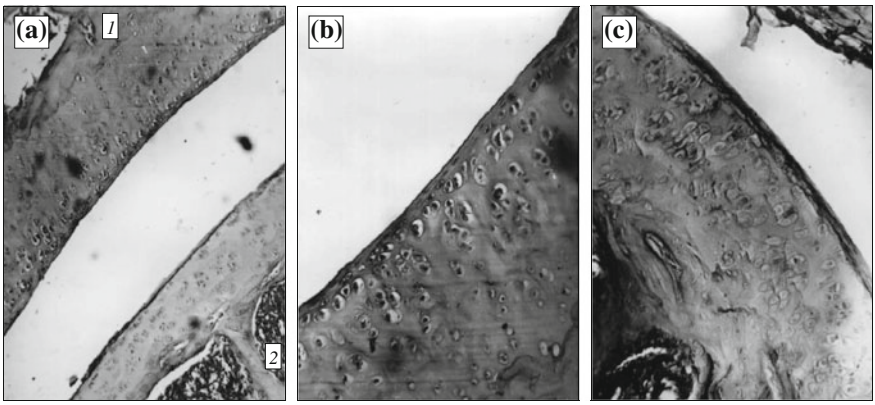


Fig. 6.13 Articular surface of patellofemoral joint of rats after single intra-articular administration of artificial lubricant containing Na-CMC aqueous solution with LCCC additive in chronic adjuvant arthritis stage: **a** articular contact of kneecap (*1*) and hip (*2*) (1.2 × 1.6 mm); **b** articular kneecap surface (540 × 750 μm); **c** hip articular surface (540 × 750 μm). Dying with hematoxylin and eosin

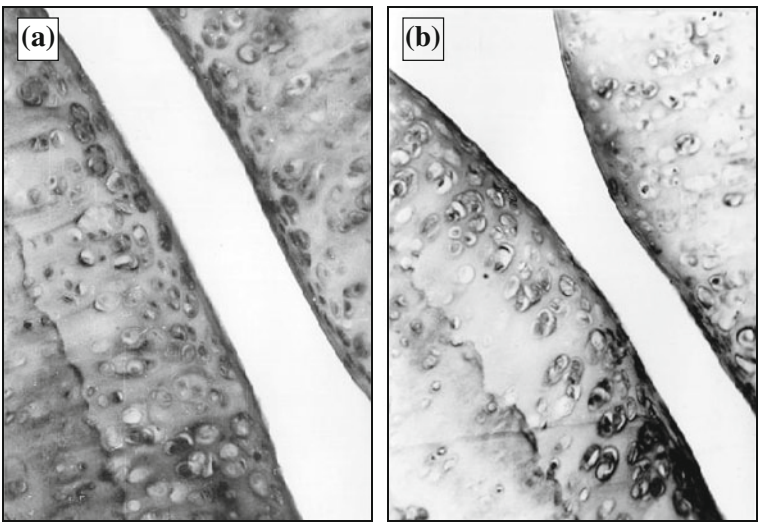


Fig. 6.14 Articular surface of tibiafemoral joint of rats after triple intraarticular administration in chronic adjuvant arthritis stage: **a** artificial lubricant containing Na-CMC aqueous solution with LCCC additive (570 × 750 μm); **b** artificial lubricant containing LCCC in Na-CMC and PVP aqueous solution (570 × 750 μm. Dying with hematoxylin and eosin

There was a multitude of isogeneous groups of regenerating chondrocytes with clearly colored nuclei. A triple administration of the preparation intensified the chondroprotective effect.

The same histological pattern was also observed after administration of the second artificial lubricant for intraarticular application containing LCCC in Na-CMC and PVP solutions (Fig. 6.14b).

After administration of the 15 % PVP solution into the afflicted joints of rats in the chronic adjuvant arthritis stage the histological changes were the following. The cartilage articular surface was irregular, though there were fewer cracks and they were smaller than in the control group of animals. A pronounced metachromasia of the interstitial cartilage substance and its thinning were observed. The cell-free zone remained to a smaller extent than in the control group, though to a lesser extent than in the experimental groups of animals after administration of the new artificial LCCC-based lubricants. The joint capsule still had signs of inflammation, though less pronounced than in the control group of animals.

The histological drugs were dyed additionally with picrofyxin of Van Gison to assess the condition of the collagenous structures of the surface and deeper zones of the cartilage matrix. Figure 6.15 shows the results.

A comparative analysis has revealed that the new artificial lubricant containing the Na-CMC aqueous solution with the LCCC additive produces a stronger protective effect of the cartilage collagen in adjuvant arthritis (Fig. 6.15a) compared with the effect of the 15 % PVP solution (Fig. 6.15b). Later it would become manifest after more intensive and homogenous coloring of these tissue structures and preserving them in the tangential zone of the cartilage. A similar protective effect was observed also after application of the second variant of the new artificial synovial fluid that also contains LCCC, but now in the Na-CMC and PVP aqueous solution. Similar results were obtained after skin application of the LCCC in vaseline oil.

In summary, the experimental tests of the new artificial LCCC-based synovial fluids have demonstrated their pronounced local effect of protection of the articular cartilage from mechanodestruction in adjuvant arthritis. The artificial lubricants reduce the dystrophic changes and preserve main cartilaginous structures from enzymatic destruction. In the experiments with animals, it was established that the new liquid-crystalline drugs did not produce mutagenous, allergenic, or cumulative effect, nor did they cause acute or chronic toxicity.

The tests served to develop a process of preparation of the mixtures with the features inherent to the natural synovial fluid, liquid crystalline cholesterol compounds, and methods of monitoring of its basic characteristics. The therapeutical and preventive drug Diasinol is now produced and serves for external application; it contains a liquid-crystalline biosubstance (the producer is the factory of diagnostic and therapeutical drugs Belbiopharm of the Republic of Belarus).

According to a resolution of the Pharmacological Committee of the Republic of Belarus, clinical tests of Diasinol as a therapeutical drug for external application have been carried out at four clinical hospitals here. It has been established that the drug is therapeutically effective for treatment of degenerative dystrophic disorders

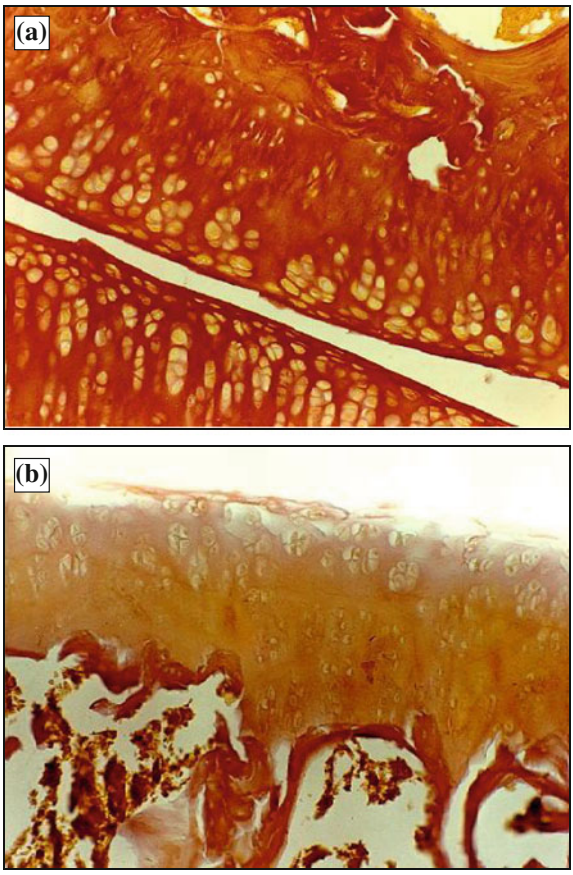


Fig. 6.15 Articular cartilage and tibiafemoral joint (**a**) and hip (**b**) of rats in chronic adjuvant arthritis stage: **a** after triple intraarticular administration of artificial lubricant containing Na-CMC aqueous solution with LCCC additive; **b** after quadruple intraarticular administration of 15 % aqueous solutions of PVP. Dying of Van Gison (780 × 600 μm)

of joints shown by 85.9 % observations on the average. The pharmacological effect of the preparation is due to the modification of the lubricating properties of the synovial fluid and protection of the cartilage from mechanodestruction.

The therapeutic effect manifests itself by reducing the pain intensity syndrome, facilitation of ambulation of joints, reduction of crackling in joints and their stiffness, and, in the result, improvement of the functional status of patients. Better results are produced by 7–10 applications of Diasinol to the patients with the osteoarthrosis of knee joints of I–II degree, including those after sports traumas.

After submission of the materials of clinical tests, the Pharmacological Committee of the Republic of Belarus was licensed to adopt Diasinol for medical practice.

Thus, the results of the accomplished studies show that the LCCC-based therapeutical drugs for treatment of joints, because of their versatility of application, can be administered as injections or applications, both with therapeutic effectiveness. They are identical to the natural synovial fluid by their basic physico-chemical

and antifriction properties and strong chondroprotective action. The latter fact is still one more proof of the biological role of the LCCC in reduction of intraarticular friction and protection of the cartilage from mechanodestruction in case of an articular pathology.

References

1. D. Symmons, S. Sand, The language of rheumatology II. Classification and grouping. Ann. Rheum. Dis. **55**, 83–86 (1996)
2. D.T. Felson, Epidemiology of the rheumatic diseases, in *Arthritis and Allied Conditions. A Textbook of Rheumatology*, ed. by W.J. Koopman (Williams & Wilkins, Baltimore, 1997), pp. 3–34
3. B. Swoboda, Aspekte der Epidemiologischen Arthroseforschung. Orthopäde **30**, 834–840 (2001)
4. W. Mau, M. Bornmann, H. Weber et al., Prediction of permanent work disability in a follow-up study of rheumatoid arthritis: results of tree structured analysis using RECPAM. Br. J. Rheumatol. **35**, 652–659 (1996)
5. E. Fex, B.-M. Larsson, K. Nived, K. Eberhardt, Effects of rheumatoid arthritis in patients followed 8 years from onset. J. Rheumatol. **25**, 44–50 (1998)
6. H. Zeidler, J. Zacher, F. Hiepe (Hrsg), *Interdisziplinäre klinische Rheumatologie* (Springer, Berlin, Heidel-berg, New York, 2001), p. 1254
7. F. Hartmann, A. Wittenborg, H. Zeidler, in *Klinische Rheumatologie. Teil I. Allgemeine Grundlagen*, ed. by H.E. Bock (Klinik der Gegenwart, Band XII, München Wien Baltimore Urban & Schwarzenberg, 1987), p. 751
8. V.C. Mow, A. Ratcliffe, S.L.-Y. Woo (eds.), *Biomechanics of Diarthrodial Joints* (Springer, New York, Berlin, Heidelberg, London, Paris, Tokyo, Hong Kong, 1990), p. 451
9. D.R. Eyre, Collagen structure and function in articular cartilage: metabolic changes in the development of osteoarthritis, in *Osteoarthritic Disorders*, ed. by K.E. Kuettner, V.M. Goldberg (The American Academy of Orthopaedic Surgeons, Rosemont, 1995), pp. 219–229
10. J.A. Martin, J.A. Buckwalter, Articular cartilage aging and degeneration. Sports Med. Arthrosc. Rev. **4**, 263–275 (1996)
11. J.A. Buckwalter, H.J. Mankin, Articular cartilage. Part II. Degeneration and osteoarthrosis, repair, regeneration and transplantation. J. Bone Joint Surg. **79A**, 612–632 (1997)
12. J.-Y. Reginster, J.-P. Pelletier, J. Martel-Pelletier, Y. Hentorin (eds.), *Osteoarthritis: Clinical and Experimental Aspects* (Springer, Berlin, Heidel-berg, New York, 1999), p. 521
13. A.R. Poole, Imbalances of anabolism and catabolism of cartilage matrix components in osteoarthritis, in *Osteoarthritic Disorders*, ed. by K.E. Kuettner, V.M. Goldberg (The American Academy of Orthopaedic Surgeons, Rosemont, 1995), pp. 247–260
14. L.J. Sandell, Molecular biology of collagens in normal and osteoarthritic cartilage, in *Osteoarthritic Disorders*, ed. by K.E. Kuettner, V.M. Goldberg (The American Academy of Orthopaedic Surgeons, Rosemont, 1995), pp. 131–146
15. J.P. Fulkerson, C.C. Edwards, O.D. Chrisman, Articular cartilage. Chapter 12, in *The Scientific Basis of Orthopaedics*, 2nd edn. (Los Altos, 1987), pp. 347–371
16. A.L. Schiller, Pathology of osteoarthritis, in *Osteoarthritic Disorders*, ed. by K.E. Kuettner, V. M. Goldberg (The American Academy of Orthopaedic Surgeons, Rosemont, 1995), pp. 95–101
17. V. Roth, V.C. Mow, D.R. Lai et al., Correlation of intrinsic compressive properties of bovine articular cartilage with its uronic acid water content. Proc. Orth. Res. Soc. **6**, 49–57 (1981)
18. J.M. Stuart, A.S. Townes, A.H. Kang, The role of collagen autoimmunity in animal models and human diseases. J. Invest. Derm. **79**(Suppl. 1), 121–127 (1982)

19. A. Unsworth, D. Dowson, V. Wright, Some new evidence of human joint lubrication. Ann. Rheum. Dis. **34**, 277–281 (1975)
20. J.H. Talbott, R.D. Altman, N.L. Gottlieb, D.S. Howell, Chondroprotection. Sem. Arthritis Rheum. **17**(2), 1–2 (1987)
21. E. Schacht, Chondroprotection—a perspective. EULAR Bull. **15**(4), 128–132 (1986)
22. J. Steinmeyer, Medikamentöse Therapie der Arthrose. Orthopäde **30**, 856–865 (2001)
23. S.S. Leopold, B.B. Redd, W.J. Warme et al., Corticosteroid compared with hyaluronic acid injections for the treatment of osteoarthritis of the knee. J. Bone Joint Surg. **85A**, 1197–1203 (2003)
24. M. Annefeld, Ultrastructural and morphometrical studies on the articular cartilage of rats: the destructive effect of dexamethasone and the chondroprotective effect of rumalon. Agents Actions **17**(3/4), 320–321 (1985)
25. S.A. Jimenez, Biochemical aspects of repair and its cellular control, in *Osteoarthritis: A Suitable Case for Treatment: Proc. of the Symp. Held at the XI Europ. Congr. of Rheumatology in Athens*, *July 1987* (EULAR Publishers, Basel, Schweiz, 1988), pp. 19–24
26. M. Haataja, I.E. Fraki, E. Vainio, Effect of antirheumatic drugs on proteinases in synovial fluid of patients with rheumatoid arthritis. Int. J. Clin. Pharmac. Biopharm. **16**(9), 417–419 (1979)
27. G.F.B. Birdwood, J.V. Gantmacher, *Further experience SEAPAL Congr. of Rheumatology, Bangkok* (Hans Huber Publishers, Berne, Stuttgart, Vien, 1984), pp. 72–73
28. J.H. Herman, E.V. Hess, Nonsteroidal anti-inflammatory drugs and modulation of cartilaginous changes in osteoarthritis and rheumatoid arthritis. Am. J. Med. **77**(4b), 16–25 (1984)
29. M. Bouakka, G. Loyau, J. Bocquet, Effect of a glycosaminoglycanpeptid complex (GP-C) on the biosynthesis of proteoglycans in articular chondrocytes treated with Interleukin-1. Curr. Therap. Res. **43**(4), 588–589 (1988)
30. V. Rejholec, Long-term studies of antiosteoarthritic drugs: an assessment. Sem. Arthritis Rheum. **17**(Suppl. 1), 35–53 (1987)
31. C. Bassleer, P. Gysen, R. Bassleer et al., Effects of peptidic glicoaminoglycans complex on human chondrocytes cultivated in three dimensions. Biochem. Pharvacol. **37**(10), 1939–1945 (1988)
32. P.G. Bulloough, Cartilage repair osteoarthritis, in *Osteoarthritis: A Suitable Case for Treatment: Proceedings of a Symposium held at the XI Europ. Congr. of Rheumatology in Athens, July 1987* (EULAR Publishers, Basel, Schweiz, 1988), pp. 13–18
33. J.P. Bali, H. Cousse, E. Neuzil, Biochemical basis of the pharmacological action of chondroitin sulfates on the osteoarticular system. Sem. Arthritis Rheum. **31**, 58–68 (2001)
34. K. Sugahara, H. Kitagawa, Recent advances in the study of the biosynthesis and functions of sulfated glycosaminoglycans. Curr. Opin. Struct. Biol. **10**, 518–527 (2000)
35. F. Ronca, L. Palmieri, P. Panicucci et al., Anti-inflammatory activity of chondroitin sulfate. Osteoarthr. Cartil. **6**(A), 14–21 (1998)
36. G.C. De los Reyes, R.T. Koda, E.J. Lien, Glucosamine and chondroitin sulfates in the treatment of osteoarthritis. Survey. Prog. Drug. Res. **55**, 81–103 (2000)
37. A. Conte, N. Volpi, L. Palmieri et al., Biochemical and pharmacokinetic aspects of oral treatment with chondroitin sulfate. Arzneimittel-Forsch **45**, 918–925 (1995)
38. A. Adebowale, J. Du, Z. Liang et al., The bioavailability and pharmacokinetics of glucosamine hydrochloride and low molecular weight chondroitin sulfate after single and multiple doses to beagle dogs. Biopharm. Drug. Dispos. **23**, 217–225 (2002)
39. N. Volpi, Oral bioavailability of chondroitin sulfate (chondrosulfate (R)) and its constituents in healthy male volunteers. Osteoarthr. Cartilage **10**, 768–771 (2002)
40. F. Richy, O. Bruyere, O. Ethgen et al., Structural and symptomatic efficacy of glucosamine and chondroitin in knee osteoarthritis: a comprehensive meta-analysis. Arch. Intern. Med. Jul. **163**(13), 1514–1522 (2003)
41. T.E. McAlindon, M.P. LaValley, J. Gulin et al., Glucosamine and chondroitin for treatment of osteoarthritis. A systematic quality assessment and meta-analysis. JAMA **283**(11), 1469–1475 (2000)

42. B.F. Leeb, H. Schweitzer, K. Montag, J.S. Smolen, A meta-analysis of chondroitin sulfate in the treatment of osteoarthritis. J. Rheumatol. **27**(1), 205–211 (2000)
43. P.M. van der Kraan, B.J. de Vries, E.L. Vitters et al., The effect of low sulfate concentrations on the glycosaminoglycan synthesis in anatomically intact articular cartilage of the mouse. J. Orthop. Res. **7**(5), 645–653 (1989)
44. P.M. van der Kraan, E.L. Vitters, B.J. de Vries et al., High susceptibility of human articular cartilage glycosaminoglycan synthesis to changes in inorganic sulfate availability. Orthop. Res. Jul. **8**(4), 565–571 (1990)
45. E.D. Schleicher, C. Weigert, Role of the hexosamine biosynthetic pathway in diabetic nephropathy. Kidney Int. **58**(Suppl. 77), 8–13 (2000)
46. D.R. Runkel, M.J. Cupp, Glucosamine sulfate use in osteoarthritis. Am. J. Health Syst. Pharm. **56**(3), 267–269 (1999)
47. T.R. Oegema Jr, L.B. Deloria, J.D. Sandy et al., Effect of oral glucosamine on cartilage and meniscus in normal and chymopapain-injected knees of young rabbits. Arthritis Rheum. **46**, 2495–2503 (2002)
48. R.A. Windhaber, R.J. Wilkins, D. Meredith, Functional characterization of glucose transport in bovine articular chondrocytes. Pflugers Arch. **446**(5), 572–577 (2003)
49. L. Ma, W.A. Rudert, J. Harnaha et al., Immunosuppressive effects of glucosamine. J. Biol. Chem. **277**(42), 39343–39349 (2002)
50. J. Hua, K. Sakamoto, I. Nagaoka, Inhibitory actions of glucosamine, a therapeutic agent for osteoarthritis on the functions of neutrophils. J. Leukoc. Biol. **71**(4), 632–640 (2002)
51. I. Setnikar, L.C. Rovati, Absorption, distribution, metabolism and excretion of glucosamine sulfate. Rev. Arzneimittel-Forsch **51**, 699–725 (2001)
52. A. Almada, P. Harvey, K. Platt, Effects of chronic oral glucosamine sulfate on fasting insulin resistance index (FIRI) in non-diabetic individuals. FASEB J. **14**(4), A750–A751 (2000)
53. J.G. Yu, S.M. Boies, J.M. Olefsky, The effect of oral glucosamine sulfate on insulin sensitivity in human subjects. Diabetes Care **26**, 1941–1942 (2003)
54. T.E. Towheed, M.C. Hochberg, A systematic review of randomized controlled trials of pharmacological therapy in osteoarthritis of the knee with an emphasis on trial methodology. Sem. Arthritis Rheum. **26**(5), 755–770 (1997)
55. T.S. Barclay, C. Tsourounis, G.M. McCart, Glucosamine. Ann. Pharmacother. **32**(5), 574–579 (1998)
56. S.B. Kayne, K. Wadeson, A. MacAdam, Is glucosamine an effective treatment for osteoarthritis? A meta-analysis. Pharm. J. **265**, 759–763 (2000)
57. J.Y. Reginster, R. Deroisy, L.C. Rovati et al., Long-term effects of glucosamine sulfate on osteoarthritis progression: a randomized, placebo-controlled clinical trial. Lancet **357**, 251–256 (2001)
58. K. Pavelka, J. Gatterova, M. Olejarova et al., Glucosamine sulfate use and delay of progression of knee osteoarthritis: a 3 year, randomized, placebo-controlled, double-blind study. Arch. Int. Med. **162**(18), 2113–2123 (2002)
59. O. Bruyere, A. Honore, O. Ethgen et al., Correlation between radiographic severity of knee osteoarthritis and future disease progression. results from a 3-year prospective, placebo-controlled study evaluating the effect of glucosamine sulfate. Osteoarthr. Cartilage **11**, 1–5 (2003)
60. O. Bruyre, K. Pavelka, L.C. Rovati et al., Glucosamine sulfate reduces osteoarthritis progression in postmenopausal women with knee osteoarthritis: evidence from two 3-year studies. Menopause **11**(2), 138–143 (2004)
61. J. Cibere, J.A. Kopec, A. Thorne et al., Randomize, double-blind, placebo-controlled glucosamine discontinuation trial in knee osteoarthritis. Arthritis Rheum. (Arthritis Care Res.) **51**(5), 738–745 (2004)
62. N. Gschwend, *Surgical Treatment of Rheumatoid Arthritis* (George Thieme Verlag, Stuttgart, New York, 1980), p. 310
63. R. Bauer, F. Kerschbaumer, Comparative follow-up of synovectome and synoviorthesis, in *10-th European Congress of Rheumatology* (Abstr., Moscow, 1983), pp. 242–243

64. F. Kerschbaumer, R. Bauer, Side effects of radiosynoviorthesis of the knee joint, in 10-th *European Congress of Rheumatology* (Abstr., Moscow, 1983), pp. 243–244
65. A. Maroudas, Hyaluronic acid films. Proc. Inst. Mech. Eng. **181**(3J), 122–129 (1967)
66. P.S. Walker, J. Sikorski, D. Dowson et al., Behavior of synovial fluid on surfaces of articular cartilage: a scanning electron microscope study. Ann. Rheum. Dis. **28**(1), 1–14 (1969)
67. S. Huttl, O. Greguska, *Hyaluronic Acid—Its Degradation and Protection in Processes, Induced by Oxygen Free Radicals* (Abstr. 16-th Sempos. ESOA. Sochy, 1987), pp. 6–11
68. B. Hell, B.S. Kapadi, Artificial lubrication of joints: use of silicone oil. Ann. Phys. Mech. **9**, 334–340 (1986)
69. P. Seller, D. Dowson, M. Longfield et al., Requirement of an artificial lubricants for joints, in *Bio-Engineering Group on Human Joints* (University of Leeds, 1967), pp. 142–148
70. V. Wright, D.J. Haslock, D. Dowson et al., Evaluation of silicone as an artificial lubricant in osteoarthrotic joints. Br. Med. J. **2**, 370–377 (1971)
71. V.O. Ribitsch, Viscoelastic behavior of synovial fluids and artificial replacement, in *Biomechanics of Diarthrodial Joints*, vol. 2, ed. by V.C. Mow, A. Ratcliffe, S.L.-Y. Woo (Springer, New York, Berlin, Heidelberg, London, Paris, Tokyo, Hon Kong, 1990), pp. 287–304
72. Patent No. 1391577 USA. Int. Cl. CO8L 1/28, A61 K 317/15. Pseudo-Synovial Plastic Body Fluids and Method of Preparing Some/C.A. Homsy, no 11352/73. Filed 8. 03. 73. Complete Specification published 23.04.75
73. J.P. Pelletier, J. Martel-Pelletier, The pathophysiology of osteoarthritis and the implication of the use of hyaluronan and hylan as therapeutic agents in viscosupplementation. J. Rheumatol. **39**, 19–24 (1993)
74. J.E. Scott, Hyaluronan, multum in parvo. Eur. J. Rheumatol. Inflamm. **15**, 3–8 (1995)
75. G. Abatangelo, M. O'Regan, Hyaluronan: biological role and function in articular joints. Eur. J. Rheumatol. Inflamm. **15**, 9–16 (1995)
76. A. Engstrom-Laurent, Hyaluronan in joint disease. J. Int. Med. **242**, 57–60 (1997)
77. L.S. Simon, Viscosupplementation therapy with intra-articular hyaluronic acid. Fact or fantasy? Rheum. Dis. Clin. North Am. **25**, 345–357 (1999)
78. J.D. Evanich, C.J. Evanich, M.B. Wright, J.A. Rydlewicz, Efficacy of intraarticular hyaluronic acid injections in knee osteoarthritis. Clin. Orthop. **390**, 173–181 (2001)
79. K.D. Brandt, J.A. Block, J. Michalski et al., Efficacy and safety of intraarticular sodium hyaluronate in knee osteoarthritis. Clin. Orthop. **385**, 130–143 (2001)
80. S.F. Ermakov, A.V. Beletskii, V.I. Nikolaev, Tribological principles of developing medicinal preparations based on blood serum as a liquid-crystalline medium for therapeutic correction of synovial joints. J. Friction Wear **32**(1), 49–53 (2011)
81. E.D. Beloenko, S.F. Ermakov, B.I. Kupchinov, V.G. Rodnenkov, O.L. Eismont, Liquid-cristal state of joint synovial lubricating medium. Experimental substantiation. Acta Bioeng. Biomech. **3**(1), 24–32 (2001)
82. S.F. Ermakov, E.D. Beloenko, O.L. Eismont, Role of liquid crystals in tribological behavior of joint cartilages. J. Friction Wear **25**(5), 31–35 (2004)
83. B.I. Koupchinov, S.F. Ermakov, E.D. Belojenko, V.G. Rodnenkov, V.N. Kestelman, Pat. 5,238,929 US, A 61 K 31/56. Correction of tribology of arthritis-affected joints and medicine for its implementation—No. 779,490. Filed 22.10.91. Published 24.08.93
84. B.I. Kupchinov, S.F. Ermakov, V.G. Rodnenkov, E.D. Beloenko, O.L. Eismont, Some results of studies in liquid-crystalline state of synovial lubricant in joints //. J. Friction Wear **23**(3), 69–75 (2002)
85. B.I. Kupchinov, S.F. Ermakov, E.D. Beloyenko, A.A. Suslov, Liquid crystalline components of synovia and their role in the joint tribology, in *I Sympozjum "Inzynieria Ortopedyczna i Protetyczna—IOP 97"* (Bialystok, 1997), pp. 125–131
86. S. Ermakov, B. Kupchinov, E. Beloyenko, A. Suslov, O. Eismont, The effect of liquid crystals on tribomechanical properties of cartilages, in *Inzynieria Ortopedyczna i Protetyczna—IOP 99: Proceedings of the Sympozjum* (Bialystok, 1999), pp. 93–99

87. S.F. Ermakov, B.I. Kupchinov, V.G. Rodnenkov, E.D. Beloenko, O.L. Eismont, Influence of nature of rubbing surfaces and lubricant on articular cartilage friction. Acta Bioeng. Biomech. **3**(1), 65–71 (2001)
88. W. Naucke, Balneotherapeutische Wirkung von Torfen und Einiger Essentieller Torf-Inhaltsstoffe. Zeitschrift fur Baderund Klimaheilkunde **27**(3), 230–246 (1980)
89. B. Kupchinov, S. Ermakov, V. Rodnenkov et al., Role of luquid crystals in the lubrication of living joints. Smart Mater. Struct. **2**, 7–12 (1993)

Conclusions

Even after the known and own ideas of friction and wear of the human and animal synovial joints have been analyzed and extended, it is impossible to explain the nature of their tribological properties. This phenomenon is given rise by the unique and unrecognized mechanism of articulate cartilage lubrication. The numerous data on frictional interaction features received on experimental joint models, of course, are insufficient to understanding the real mechanisms of intraarticular friction decrease in natural conditions. The following conclusion can be drawn from the above analysis.

According to modern views, the cartilage is bi-phase model. A strong porous collagenic framework and mainly sulfated proteoglycan units form its solid phase. Polyanionic nature and the specific structural organization of the last promote cartilage overhydratation and provide its main biomechanical properties: elasticity and shock-absorbing ability. The liquid phase of a cartilage represents low-molecular organic substances and electrolytes in water.

Viscoelastic deformation of a cartilage matrix and pumping out the interstitial liquid in dynamic contact area occur during joint movement. A liquid returns back to a cartilage and its dimension is restored at unloading. In this case, the lubricant layer is formed from components of synovial and interstitial liquids bleeding from cartilage. Joint trophic function is carried out the same way. Efficiency of joint lubrication is promoted also by the main biomechanical property of synovial fluid due to hyalurate, namely, thixotropic behavior pattern, that is ability to be dissolved as gradient of cartilage sliding speed increases.

This standard simple partly explains the mechanisms of cartilage friction decrease at rather fast movements and low nominal pressure in dynamic contact area, but along with this the nature of their boundary lubrication at the low, almost vanishing, sliding speeds and high pressure, remains unclean. It is supposed that in that case there is a formation of monomolecular lubrication layers between the cartilage-conjugated surfaces. Hypotheses of formation layers from hyalurate molecules of a without and with interstitial liquid, a specific lubricant protein —"lubricin" is not sufficiently justified. Joint boundary lubrication is recently assumed to have a compound nature.

S. Ermakov et al., *Liquid Crystals in Biotribology*,
Biological and Medical Physics, Biomedical Engineering,
DOI 10.1007/978-3-319-20349-2

Formation of joint effective layer rubbing down and dividing the cartilage surfaces is tried to be explained with a hypothesis of micropolar liquid.

This the least known division of a biotribology has aroused the most interest and become the primary goal of our research.

Modern achievements in the field of physics of liquid crystals have come into notice. Thanks to them the liquid-crystal state of living systems became a subject of biological investigations in the last decades. There were evidence for effective lubricant action of liquid crystals during solid friction. At the same time, similar data on synovial fluid haven't been found. For this reason, the main body of the book is given over to proofs of a role of cholesteric-nematic liquid crystals in intraarticular friction decrease. This situation was confirmed with results of the physical-chemical analysis of synovial fluid and its components, and also studying of their lubricant properties by numerous research techniques of liquid crystals and liquid-crystal biological media.

A groundbreaking science-based concept of joint cartilage boundary friction became a natural result of this experimental work. It is consistent with known ideas of mechanisms of joint lubrication, and significantly extends and complements knowledge of a biotribology and biophysics of liquid-crystal biosystems. It should be noted that synovial fluid is the complex multicomponent liotropic liquid-crystal media. Therefore lubricating action of a synovial fluid can be realized by the cholesterol esters and cholesteric fatty acids together with proteins, many of which show properties of nematic liquid crystals.

Liotropic liquid-crystal media are least studied today. For that matter the discovery of a liquid-crystal state of the joint synovial fluid has become a new development in biotribology and pathogenetic mechanisms of cartilage destruction during rheumatic afflictions.

The comprehensive description of known understanding of molecular structure, the structural organization and biomechanical features the synovial fluid elements, the analysis of concepts of joint cartilage lubrication and the cause of destruction during trauma and arthropathies, data on efficiency of the existing methods of treatment and prophylaxis of intraarticular damages occupy an important place in the book.

Data on biophysical characteristics of joint cartilages as the rub natural biopolymers will be useful for the material engineer and experts in creation of new composite materials for joint endoprosthesis.

According to D. Dawson, the materials used for joint endoprosthesis possess properties distinct from cartilaginous tissue. That leads to complication of joint replacement.

Creation of materials for endoprosthesis friction couples having comparable characteristics as cartilages and micropolar liquid is the most important endoprosthesis objective. Creation of a new development in arthrology and pharmacotherapy of cartilage destruction during rheumatic afflictions is other essential result of the research work.

The universal and second to none artificial lubricants, having properties of natural synovial fluid, used for both invasive and noninvasive treatment and novel arthropathy therapy have been developed on the basis of the received results.

The experimental data on high chondroprotective efficiency of preparation, checked on osteoarthritis models and during clinical approval, are proof that liquid crystals play an essential role in intraarticular friction decrease. It can be a real prerequisite for development of new pharmaceuticals for cartilage mechanodestruction prophylaxis and therapy during arthropathies.

Summarizing all the aforesaid, authors hope that the their research work will draw attention of arthrology, tribology and biophysics specialists and will serve as a stimulus to further investigation of artificial materials and medias for joints containing liquid crystals.

Index

S. Ermakov et al., *Liquid Crystals in Biotribology*,
Biological and Medical Physics, Biomedical Engineering,
DOI 10.1007/978-3-319-20349-2