Kai-Uwe Schmitt · Peter F. Niederer
Duane S. Cronin · Markus H. Muser
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Trauma Biomechanics

An Introduction to Injury Biomechanics

Fourth Edition

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Preface

Injury is arguably one of the most under-recognized health problems facing society today. Potential injury hazards exist everywhere in our daily environments including the workplace, home, transportation, and sports and recreational settings. From traffic-related injuries alone, the World Health Organization estimates that 1.2 million people die each year worldwide and as many as 50 million are injured or disabled. The associated expenditures, lost productivity, legal, and medical costs resulting from these injuries and fatalities are staggering. More importantly, the personal losses resulting from serious injury and death are incalculable.

While this trauma often results from catastrophic events that are deemed accidents, the mechanisms of these injuries are both understandable and preventable. Trauma biomechanics uses engineering principles to explore the physical response of the human body to applied forces that produce failure of the tissues. With a firm understanding of trauma biomechanics, researchers and designers are able to apply existing knowledge and to generate new data for the development of improved injury prevention strategies.

The book, *Trauma Biomechanics*, provides a comprehensive overview and introduction to the subject for these researchers and designers. While countless examples of protective equipment such as seat belts, air bags, and helmets have been designed using the principles of trauma biomechanics, the prevalence and severity of the injuries necessitate that we continue to educate and train the next generation of engineers, scientists, and medical professionals.

I have taught injury biomechanics for more than two decades and this book on trauma biomechanics has become a mainstay as a supplemental text in my graduate engineering courses. The book deftly and succinctly covers the background, tools, methods, and resources of trauma biomechanics before delving into a systematic review of the anatomy, injury classification, injury mechanisms, and injury criteria of each body region. Each chapter includes summaries of the most relevant scientific literature and biomechanical research that provide background and context for the interpretation of the graphical and tabular information.

The book concludes with a chapter on injury prevention that combines both collision avoidance as well as passive injury countermeasures. While the intended audience of the book is primarily scientists and clinicians working in the area of trauma, the straightforward writing style and graphical depictions make even the most technical engineering and medical concepts approachable for a broad array of injury prevention professionals including epidemiologists, social scientists, and policy makers.

Jeff Crandall
President, International Research Council
on Biomechanics of Injury

Nancy and Neal Wade
Professor of Engineering and Applied Sciences
Director, Center for Applied Biomechanics
University of Virginia

Preface: 3rd Edition

Injury is a leading cause of death, hospitalization, and disability worldwide. The World Health Organization predicts that unintentional injuries arising from road traffic incidents will rise to take third place in the rank order of international disease burden by the year 2030. Although these statistics and the associated economic costs are staggering, the effect of unintentional injury and death from trauma is more apparent, and more disturbing, when seen personally. By a young age, nearly everyone in the world, regardless of region, wealth, or education, has had a relative or someone that they know killed or disabled in an “accident.” The quality of life and financial effects on the injured person and their families and friends are plainly evident and clearly devastating. Many unintentional injuries are in reality not accidents; they could be prevented with changes in policy, education, or through improved safety devices. Arrayed against these preventable injuries, a diverse group of injury prevention researchers and practitioners work to decrease the incidence of unintentional injury.

In trauma biomechanics, the principles of mechanics are used to understand how injuries happen at the level of the bones, joints, organs, and tissues of the body. This knowledge is central in the development, characterization, and improvement of safety devices such as helmets and seat belts and in the safe design of vehicles and equipment used for transportation, occupation, and recreation. The field of trauma biomechanics is highly interdisciplinary, with engineers and physicists being centrally involved with medical practitioners and many other experts. This book, Trauma Biomechanics, is organized as a short primer of this subject and it provides a logical overview of the field. It is written to be accessible to a range of students or practitioners, while still providing considerable detail in each section. Each chapter contains plentiful and up-to-date references to guide readers who require more information on a particular topic.

In contrast to the relative abundance of texts that describe basic biomechanics, sports biomechanics, gait analysis, and orthopaedic biomechanics, this is one of only two or three texts focused on trauma biomechanics that I am aware of. I have used a previous version of the book as a required text for a combined senior undergraduate- and graduate-level Mechanical Engineering class called the “Fundamentals of Injury Biomechanics” at the University of British Columbia. The students commented positively on the layout and accessibility of the book and they used it as a key reference in the assigned problems and project work in the class. I think the short primer structure of the book helped to make it accessible to the students. It is possible to start reading at the beginning of any chapter and quickly come up to speed with the most important basic knowledge about the anatomy, tolerance, and injury prevention techniques for that region of the body. This is of great utility for students but also for people working in injury research contexts where they can be asked to rapidly switch their focus from injury in one area of the body or from one mechanism to another. This can occur not only while studying in university but also in many industrial and academic research contexts. For example, this is frequently required of people working on government-sponsored injury reconstruction teams or who are engaged in reconstructing injuries in the litigation context.

I recommend this book as a key basic resource for anyone interested in injury prevention. Everyone, from graduate students working in an academic injury biomechanics setting to engineers, physicists, clinicians, surgeons, kinesiologists, biologists, statisticians, and social scientists working in the broad field of injury prevention, frequently has questions about how injuries happen in various parts of the body. This book is an essential and accessible resource to anyone with these questions.

Peter A. Cripton
Associate Professor of Mechanical Engineering
and Associate Faculty Member of the Department
of Orthopaedics, The University of British Columbia

Preface: 2nd Edition

Everyday more than 140,000 people are injured, 3000 killed, and 15,000 disabled for life on the world's roads. Likewise, sports-related injuries are numerous and have a significant socioeconomic impact. The field of trauma biomechanics, or injury biomechanics, uses the principles of mechanics to study the response and tolerance level of biological tissues under extreme loading conditions. Through an understanding of mechanical factors that influence the function and structure of human tissues, countermeasures can be developed to alleviate or even eliminate such injuries.

This book, Trauma Biomechanics, surveys a wide variety of topics in injury biomechanics including anatomy, injury classification, injury mechanism, and injury criteria. It is the first collection I am aware of that lists regional injury reference values, or injury criterion, either currently in use or proposed by both U.S. and European communities. Although the book is meant to be an introduction for medical doctors and engineers who are beginners in the field of injury biomechanics, sufficient references are provided for those who wish to conduct further research, and even established researchers will find it useful as a reference for finding the biomechanical background of each proposed injury mechanism and injury criterion. As more people become aware of and understand this subject, it will someday lead to better mitigation and prevention of automotive and sports-related injuries. I like this book very much and believe that you will find the same.

King H. Yang
Professor of Biomedical Engineering and Mechanical Engineering
Director of Bioengineering Center, Wayne State University

Acknowledgments

With this fourth edition we expand the scope of the book toward blast injury. The biomechanics of high energy trauma differs from the mechanisms relevant in accidental injuries as sustained in road traffic and sports and thus deserves to be presented in an own chapter. We are very grateful for the input with regard to blast injury provided by Prof. Duane Cronin, who joins our team of authors with this edition.

Although the expansion brings in new aspects of trauma biomechanics, the general intention of the book remains unchanged. It is a short primer for everyone interested in the basics of trauma biomechanics and injury prevention. We thank all readers for support and feedback and hope you will also appreciate this latest edition.

Kai-Uwe Schmitt

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Introduction

1

The human body is exposed to mechanical loads throughout its life. Besides forces deriving from ubiquitous and penetrating fields such as gravity or forces due to electromagnetic fields which are non-contact in nature and as such effective over distances, there is a great variety of forces acting on the human body from contacts with the surrounding. In addition, numerous forces are generated in the course of physiological processes inside the body in the different organs and tissues. Throughout evolution, all forms of life adapted their physiology to mechanical interactions; some of them to the extent that a proper function in fact requires the influence of forces, for example bone remodelling. Mechanical forces are known to modulate cell development even in utero (Knothe Tate et al. 2008).

The science of biomechanics is devoted to the analysis, measurement and modelling of the effects that are observed under the various mechanical loading situations primarily in humans, but also in animals and plants. As this definition suggests, a quantitative approach is thereby in the foreground. The range of forces which is of interest is enormous: Internal forces may originate from the action of molecules, contractile fibres on a cellular level or muscles on a macroscopic scale, moreover, pressures and shear stresses may be generated by biological fluid flows or active biological transport processes including osmosis. External forces, in turn, occurring in everyday life may span a virtually unlimited extent. Accordingly, the forces of interest in biomechanics cover typically a range from pN to MN (lower or higher forces, respectively, are hardly considered because of lack of biological effect on the lower side or complete devastation on the upper), and they may vary with time within picoseconds to years.

An inevitable consequence of forces acting in- or outside the human body consists of the possibility that they may cause injury. Such adverse consequences are usually associated with the action of excessive external forces impinging during unfavourable events, in particular accidents, with which we may be confronted in daily life. In fact, accidents of all kinds represent the leading cause of death at young age. In Table 1.1 numbers are shown from the US that may be regarded as representative for industrialized countries. Internal forces, in contrast, are mostly thought to be governed by anatomical or physiological constraints which prevent

K.-U. Schmitt et al., *Trauma Biomechanics*, DOI: 10.1007/978-3-642-53920-6_1,

Table 1.1 Reported numbers of leading causes of death in the USA (age 15–24)

Cause		Number
Accidents (unintentional injuries)		12,032
• Motor vehicle accidents	6,984	
• All other accidents	5,048	
Intentional self-harm (suicide)		4,688
Assault (homicide)		4,508
Malignant neoplasms		1,609
Diseases of the heart		948
Congenital anomalies, deformations and chromosomal abnormalities		429
Influenza and pneumonia		213
Cerebrovascular disease		186
Pregnancy, childbirth and the puerperium		166
Chronic lower respiratory disease		160
All other causes		4,666
Total		29,605

Adopted from the National Vital Statistics Report Vol. 61, 2012. This report is updated regularly

the occurrence of injury. Yet, broken ribs due to intense coughing, rupture of muscle fibres because of tetanic contraction or endocardial bleeding in cases of hypovolaemic shock are injuries resulting from forces produced by the body itself.

The special discipline of biomechanics that is concerned with injury caused by mechanical interaction is denoted as biomechanics of injuries or trauma biomechanics and is the subject of this book. Since there are a great many types of injuries, injury mechanisms and activities which are prone to cause injury, a large variety of human activities and situations where excessive loads may occur has to be considered. When performing a thorough analysis of such circumstances, however, it becomes evident, that trauma biomechanics is strongly dominated by its interdisciplinary character. First of all, the field of biomechanics itself covers a wide range of areas of interest, from macroscopic motion analysis in sports, for example, to the sub-microscopic modelling of molecular transmembrane transport. Many basic biological aspects are furthermore involved for we are dealing with living matter associated with intrinsic active processes such as muscle contraction or electrochemical activities. The wide-spread knowledge obtained during the recent decades in the different fields covering mechanics and biology in general greatly contributes to trauma biomechanics, in that for an in-depth understanding of injury processes all aspects from the macroscopic scale to the sub-cellular level may have to be taken into account. Therefore, many subjects of importance for trauma biomechanics relating to basic mechanics, anatomy and physiology have to be covered in order to be able to treat the entire field in a systematic approach; nevertheless, in view of the great diversity that exists, a selection has to be made and comprehensiveness is not reached.

1.1 About the Contents of This Book

Several preliminary remarks are useful in order to elucidate the extent and limitations of the subjects treated in this book:

1. A distinction has to be made between injury resulting from unexpected, sudden and singular events, i.e., accidents in a strict sense, and injury caused by the chronic exposure to unfavourable loads over extended periods of time. A head injury of a pedestrian that is sustained from an impact on the hood of an automobile during a collision, or the gradual destruction of hair cells in the inner ear as the result of a chronic exposure to loud music—both examples are associated with injury, yet, the type of injury, injury mechanisms, tolerance levels, injury criteria, reconstruction and analysis methods as well as protection measures differ fundamentally. Also with respect to insurance and liability issues, procedures are greatly different.
2. The injury causing period in the course of a traffic accident has a duration of 100–200 ms typically whereby the early part of this period is often decisive. In many cases the person involved is not aware of the event and does not (cannot) react prematurely to the imminent danger. Accordingly, muscular reactions which set in with a time delay of 60–80 ms are often of secondary importance only and can therefore be disregarded. The situation is basically different in case of chronic overloading, where physiological and also psychic reactions are always in the foreground.
3. A further important aspect is related to age. The mechanical properties of human tissue, organs and body as a whole change, in particular with respect to injury tolerance, decisively during aging towards unfavourable levels. There are a number of reasons for this, among them, a reduction of tissue compliance due to a decrease of body water content along with a stiffening of soft tissue and a gradual demineralization of bone at ages above 30–40 years. As a result, the advent of injury, primarily bone fractures, increases dramatically with age. The incidence even of spontaneous fractures, occurring under normal physiological loads, is well known. In view of the aging population in the industrialized countries, this aspect has to be given particular attention.
4. Adolescence, at the other end of the age scale, poses likewise important problems for trauma biomechanics in that mechanical and biological properties undergo dramatic changes from birth to adult age. Experiments with children are hardly conceivable; also work with adolescent cadavers is not a routine matter. Downscaling from adult characteristics to children needs a careful analysis ("children are not small adults"). The development of child dummies, in particular (see Sect. 2.6.1) is therefore not straightforward. Due to the lack of experimental approaches, statistics represent the main method for the analysis of child injury. A significant contribution in this direction is made by The Center for Injury Research and Prevention at The Children's Hospital of Philadelphia (http://injury.research.chop.edu/).

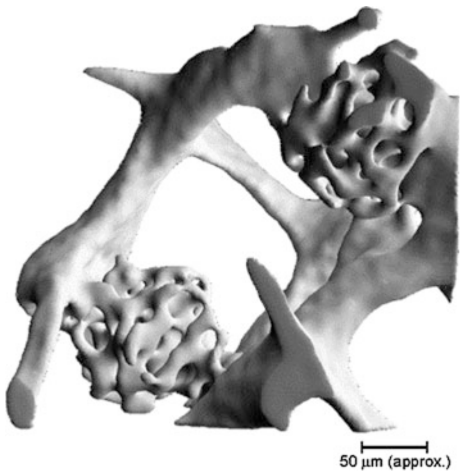

Fig. 1.1 Microcallus formation. The picture exhibits a 3D micro computed tomography (micro CT) scan of the excised portion (biopsy) of a human iliac crest where microfractures induced new bone formation (*Courtesy* Prof. R. Müller, ETH Zurich)

5. The mechanical response of the body in case of pathological alterations may furthermore be significant. Renal trauma as a result of stress concentration around a cyst has been observed in urology or the aggravation of the effects of a whiplash-type event due to pre-existing neck impairment is a well known complication.
6. Under very restrictive conditions, however, micro-injury on a cellular level may to some extent be advantageous for tissue regeneration. Figure 1.1 shows the micro callus formation following micro-damage in spongy bone which may serve as an example of an injury that stimulates bone remodelling. After a long and strenuous hiking tour such micro-injuries in the bones of a healthy foot are quite common. Chronic overexposure, in contrast, may lead to quite adverse developments. Figure 1.2 exhibits a marathon runner whose skeleton was largely demineralised due to excessive training.
7. Injuries are suffered mostly in connection with motion (sports, household, etc.) or mobility (traffic). While in general biology the use of animal models (under restrictive regulations) is widespread, the non-linearity inherent in motions and related injury mechanisms prevent scaling up from, say, rats to humans almost completely. Accordingly, except for aspects of basic physiology, only scarce information in trauma biomechanics derives from animal experiments.
8. When the entire spectrum of "injury" including causation, frequency, prevention, mitigation, rehabilitation, long-term sequelae and socio-economic consequences is considered, clinical medicine cannot be disregarded, since the treatment of injury is made by medical doctors providing in- and outpatient service. It is thereby often forgotten that the overall reduction of specific mortality (i.e., mortality by case) which is observed in most activities associated with a risk of injury, is partly due to dramatic developments of emergency rescue services, first aid procedures and intensive care treatment. A drawback however is that an analysis of injury mechanisms and accidental

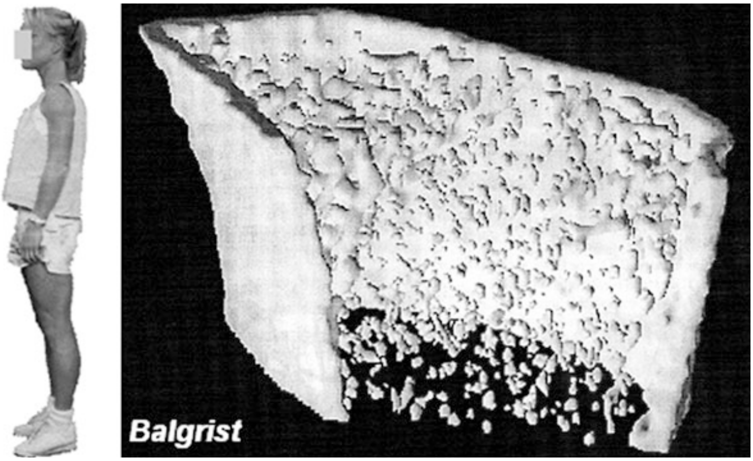

Fig. 1.2 Twenty-eight year old woman (*left*) and micro-CT scan (non-invasive) of radius close to the wrist (*right*). The extreme demineralisation of the radius is to be attributed to excessive training as a Marathon runner (*Courtesy* Prof. M. Dambacher M.D, Balgrist Orthopaedic Hospital, University of Zurich)

events is sometimes made by physicians without a complete knowledge of the relevant facts. This may be a consequence of their intense and highly demanding work with patients. Yet, an objective assessment on a scientific basis of the severity and causality aspects of accidents resulting in injuries requires a multidisciplinary approach. In addition to the medical information retrieved by clinicians, all technical and biomechanical circumstances have to be taken into account in accident analysis and reconstruction. This is especially important in cases of forensic expert witnessing. A specialized education and extensive experience is required for this purpose.

9. This book is primarily focused on the mechanics of injuries which are inflicted without intention. However, injuries may also be conveyed intentionally in a criminal, terroristic, or battlefield environment. Related subjects include wound ballistics, protective garment for soldiers or low-injury producing police weapons. The chapter on ballistic and blast trauma partly includes aspects associated with injuries that exhibit an intentional background. The reader further interested in such issues is referred among others to several publications by the International Committee of the Red Cross (http://www.icrc.org) where also further references can be found. The overall significance of intentional injuries (under non-belligerent circumstances) should in any event not be underestimated; e.g., there were around 11,000 deaths in the US (2010) due to the use of firearms (not including accidental firearm incidents and suicide) in comparison with some 12,000 persons who were killed in the same year as passengers of an automobile. The situation is thereby quite diverse: According to UN statistics, in the United States in 2009, 3.0

intentional homicides committed with a firearm per 100,000 inhabitants were recorded, while the figure for the United Kingdom, for example, was 0.07 per 100,000, about 40 times lower, and for Germany 0.2. Gun homicides in Switzerland are similarly low, at 0.52 in 2010 even though Switzerland ranks third in the world for the number of guns per citizen. Suicide is a further important cause (Table 1.1). Nontechnical aspects (social, political, psychological, general society-related) are particularly important in this context. Investigations, e.g., on the influence of physical violence in childhood (Paradis et al. 2009) involve large-scale socio-psychological cohort analysis.

10. The least serious injuries are of course those that do not occur. Accordingly, injury prevention is given a high priority in all situations where injury may happen. The prevention of traffic accidents has been recognized and implemented as an important governmental task since decades. In contrast, injury prevention in sports has primarily been perceived by international and national sports associations within the framework of sports medicine, mostly in the form of rigorous regimentation, ban of certain particularly violent variants of sports, development of protective devices such as helmets or shin guards and trainer education. Insurance companies furthermore support all injury prevention campaigns as part of their mission whereby these efforts include especially also workplace and household accidents. While all of these preventive activities are oriented towards pre-accident conditions, extensive rehabilitation is often required after injury has occurred and healed. Again, government agencies, sports federations, professional work associations, clinical medicine as well as insurance companies develop extensive efforts. Since this book is devoted and limited to trauma biomechanics, aspects of prevention and rehabilitation are only included in so far as there is a direct relation with the occurrence of injury.

Most systematic and quantitative research in trauma biomechanics has been made in connection with traffic accidents, although injuries sustained in sports, at the workplace or during household activities are likewise prominent (see the statistics issued by the International Labor Organization with respect to injury associated with work; http://laborsta.ilo.org). There are mainly two reasons for this:

First, serious and fatal injuries, mostly sustained in traffic accidents, represent the leading cause of premature death (Table 1.1) and, as a result, enormous social costs are involved. Accordingly, liability problems along with political interventions and government rulemaking put the automobile industry under enormous pressure (the public response to the 1965 book by Ralph Nader, "Unsafe at Any Speed" was overwhelming) and stimulated comprehensive research and development activities. Death occurring at young age is particularly deplorable and extensive countermeasures are justified; if all ages are included (Table 1.2) however, diseases outnumber accidents by far. This is not astonishing, since the (inevitable) end of life occurs mostly at an advanced age where age-related deterioration of health is in the foreground. Second, although traffic accidents, like all other types of accident, exhibit a wide variety and variability, it is nevertheless

Table 1.2 Reported causes of death (all ages)

Cause	Number
Diseases of the heart	596,339
Malignant neoplasms	575,313
Chronic lower respiratory diseases	143,382
Cerebrovascular diseases	128,931
Accidents (unintentional injuries)	122,777
Alzheimer's disease	84,691
Diabetes mellitus	73,282
Influenza and pneumonia	53,667
Nephritis, nephritic syndrome and nephrosis	45,731
Intentional self-harm (suicide)	38,285
Septicemia	35,539
Chronic liver diseases and cirrhosis	33,539
Essential hypertension and hypertensive renal disease	27,477
Parkinson's disease	23,107
Pneumonitis due to solids and liquids	18,090
All other causes	512,723
Total	2,512,873

Adopted from the National Vital Statistics Report, Vol. 61, 2012

possible to identify some "typical" or "representative" types of accidents, e.g., a frontal collision of an automobile against a barrier or a 90° side impact which lend themselves for a well defined testing protocol and quantitative analysis. In a sports, workplace or household related environment it is in contrast difficult to find typical situations which cover a substantial amount of injury-producing situations. From Table 1.3 the wide range of injury-producing circumstances, e.g., at the workplace is evident.

In comparison with traffic accidents, the literature on injury in sports, etc. is—although abundant—less stringent from a biomechanics' point of view in that general statistics, mostly descriptive explanations of injury mechanisms, medical treatment strategies and practical recommendations for trainers or people responsible for workplace safety dominate over quantitative analysis. Whenever quantitative information is sought with respect to human tolerance or injury criteria derived thereof, the literature on sports injuries refers almost exclusively to results from traffic accident analysis. It is also remarkable that most investigations on sports accidents are made in disciplines associated with an enormous financial background such as soccer, American football or skiing, while less prominent activities, e.g., orienteering, receive much less attention.

Table 1.3 Workplace fatalities 2011

Cause of fatality	%
Fall, slip, trip	33
Transportation	24
Contact with objects and equipment	18
Exposure to harmful substances and environment	17
Violence and other injuries by persons or animals	4
Fire or explosion	4

Adopted from the US Bureau of Labor Statistics

An even greater variety from an anatomical and physiological point of view in comparison with accidents exists with respect to the occurrence of injuries due to chronic mechanical (over-) exposure. A distinction between impairment due to chronic exposure and disablement resulting from diseases which are unrelated to the exposure in question is often difficult or impossible. Psychic influences are particularly important. Quantitative information is scarce. Vibrations of construction machinery, for instance, or noise levels in factories and entertainment facilities are limited by regulations which are derived from long-term statistical evidence rather than from physiological experiments.

From the considerations above it becomes understandable that the backbone of this book is essentially based on the trauma biomechanics of traffic accidents. After a general chapter describing basic definitions and methods, a sequence of chapters is presented that deal with the different body regions. These chapters are arranged systematically by starting with a brief outline of the anatomy of the body region under consideration, limited to those aspects which are of special importance in view of injury mechanisms. Furthermore, the range of possible injuries, underlying injury mechanisms and the biomechanical response to loading of this body region are described. Known injury tolerance values and injury criteria based thereon to assess the likeliness of injury are discussed. Following injuries sustained in traffic accidents, sports injuries are treated whereby selected special aspects of anatomy relevant for the understanding of related injury mechanisms, injury analysis and tolerance criteria are included. An important part in each chapter is devoted to protection measures which are recommended or regulated in order to mitigate injuries. Where appropriate, additional information on injury prevention measures or other special subjects is presented. For further reading, references are given at the end of each chapter. Problems to be solved as exercises are added in order to allow the reader to test and deepen his or her understanding and to stimulate further studies.

Two chapters are added in order to enlarge the scope of this book into further important, closely related areas. First, a chapter is devoted to blast injuries, where also aspects related with the intentional causation of injuries are relevant. Second, a section on injuries due to chronic exposure to mechanical loading is included. Many practical issues in connection with such injuries fall within the framework of ergonomics, general workplace safety and management of occupational safety hazards, however. For example a thromboembolism occurring during a long air voyage is a problem of ergonomic seat design and passenger behaviour rather than of trauma biomechanics. As this book is limited to the latter field, such subjects are not included. E.g., over the home page of the US Occupational Safety and Health Administration (http://www.osha.gov) relevant information can be found.

1.2 Historical Remarks

Biomechanics as a science is as old as mechanics. While one of the first scientists with a profound activity in biomechanics, Giovanni Alfonso Borelli (1608–1679; sometimes referred to as "The Father of Biomechanics") devoted much of his time to the analysis of bird flight and swimming of fishes, Leonhard Euler (1707–1783), the creator of continuum mechanics, wrote an extensive treatise on the principles of the motion of blood in the arteries ("Principia pro motu sanguinis per arterias determinando", op. posth.). Until the mid 19th century, however, the mechanics of injury or trauma biomechanics was not the subject of systematic research. This might be attributed to the fact that dangers were ubiquitously imminent and injury must have been considered a natural feature associated with life. The reader should not forget that through 2000 years of history up to 1945, there has never been a period longer than 15 years without war in Europe. Injury prevention was rather straightforward and pragmatic, e.g. in the form of cuirasses for knights.

The first known systematic and scientific approach towards trauma biomechanics was taken by the German anatomist Otto Messerer in Munich who published his results in the year 1880 under the heading "On the Elasticity and Strength of Human Bones" (in German). His activity was however rather isolated at that time. Nevertheless, the "Messerer-wedge" is well known in forensic science and still serves as a reminder for his seminal work.

As mentioned before, the field of trauma biomechanics today is mostly centred on injuries sustained in traffic accidents. Yet, historically, its roots are in aviation. During the 1st National Conference on Street and Highway Safety (USA 1924) simple and practical aspects of traffic safety such as the colours of traffic lights or driver education dominated while biomechanics was not (yet) of concern. In contrast, trauma biomechanics was already at that time a significant issue in the field of military aviation where the human body is exposed to extreme mechanical loading conditions. After having observed many accidents with aeroplanes, Hugh DeHaven, who can be considered, following Borelli, as the "Father of Trauma Biomechanics", started an analysis of the underlying injury mechanisms. In 1942, he published a first work titled "Mechanical Analysis of Survival in Falls from Heights of 50–100 Feet". In the following years, military aviation remained the focus of research into trauma biomechanics. Transsonic flight and ejection seat dynamics were among the problems that stimulated this research. Basic experimental methods like subtraumatic volunteer experiments to determine the biomechanical response of the human body or the development of anthropomorphic test devices (dummies) followed.

Fig. 1.3 Colonel Stapp sitting on the rocket sled "Sonic Wind No. 1" with which he was subjected to a deceleration of approximately 40 g (http://www.stapp.org)

Probably the most famous pioneer in aviation-related trauma biomechanics was Colonel John Paul Stapp. He became particularly well-known for his experimental work, including several experiments subjecting himself to various impacts. In one of the most spectacular series of these experiments conducted during the early fifties, Stapp was seated on a rocket sled. From a velocity of approximately 1000 km/h, the sled was stopped within 1.4 s in a water bath resulting in a sled deceleration of 40 times gravity (Fig. 1.3). No serious injuries were reported from this experiment. Stapp, whom the Time Magazine called "the fastest man on earth and No. 1 hero of the Air Force" (Time, September 12/1955), also founded an annual conference for the discussion of trauma biomechanics' related subjects—the Stapp Car Crash Conference. John P. Stapp died in 1999 at the age of 89 years.

Later on, astronautics necessitated the investigation of human physiology under totally opposite conditions as considered here, namely zero gravity. There were nevertheless developments which were also of interest in trauma research. E.g., the first computer model for the simulation of 3D human motion (R.D. Young, Texas A&M, 1970) was developed in connection with the analysis of human motion patterns in case of absence of external forces. With respect to traffic accidents, McHenry (Calspan Corp., Buffalo) wrote the first computer model for motions of humans involved in a frontal crash. Since in this case the influence of external forces is of importance, a large portion of the simulation was devoted to the modelling of interactions of body parts with surrounding structures. As this complicated the computational and numerical complexity decisively, first models were limited to planar (2D) motions.

During the early days of automotive transportation, safety issues were primarily considered to be a domain of the driver who was assumed to be responsible for driving in a manner that would safeguard drivers, passengers as well as occupants in other vehicles, bicyclists and pedestrians. Restraint systems were thought of (Fig. 1.4) but not widely implemented until after World War II. Nevertheless, during the 1920s and 1930s, car manufacturers gradually improved the vehicle design also with respect to safety. Reliable and durable four-wheel breaking systems were introduced and laminated safety glass replaced the plate glass that

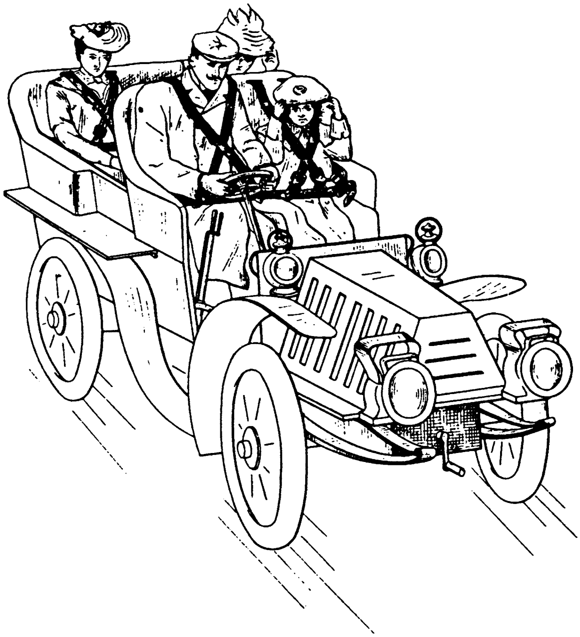

Fig. 1.4 The seat belt patent (1903) by Gustave D. Lebau. Rather than providing protection in case of a collision, however, these belts were mainly intended to keep the passenger in his or her seat in view of bumpy roads and lacking wheel suspension technology

was used for windscreens. Further developments focused on lighting such as sealed-beam head lamps and the wheels by introducing tubeless tires. All-steel car bodies were used instead of wooden structures and thus increased the stiffness of the vehicles.

It was the consequence of the rapidly increasing mobility after World War II along with a dramatic increase in numbers of injuries sustained in traffic accidents that in-depth emphasis was finally given to these problems. The Automotive Crash Injury Research programme (ACIR, Cornell University, 1951) represented an early systematic approach with respect to injury analysis in traffic accidents. An important development set subsequently in when the concept of a stiff passenger compartment combined with a defined crush zone was realised. At the same time, the steering column as a possible source of injury received attention, too, leading to the development of multi-element and energy-absorbing steering columns. Further improvements included the crashworthiness of the instrument panel, the development of restraint systems like the three-point belt and the airbag. Furthermore, the terms "passive" and "active" safety were established and systematic crash testing along with numerical simulation was introduced by automobile manufacturers. A comprehensive overview over the research made in automotive safety up to 1970 can be found in the 1970 International Automobile Safety Conference Compendium (published by SAE, New York).

Crash injury management, i.e., passive safety can be approached from three different levels. First, a reduction of injuries can be achieved by improved crashworthiness of the vehicle. This includes in the first place the design of energy

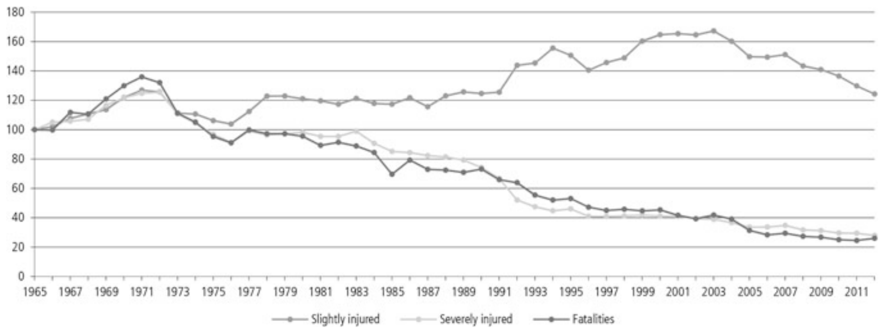

Fig. 1.5 Development of the number of casualties from traffic accidents in Switzerland (indexed on 1965). While the numbers of killed and seriously injured persons were clearly reduced over the last decades, the number of minor injuries remains on a high level (bfu 2013). This circumstance can be regarded as typical for the situation in an industrialized country

absorbing structures. Second, the occupant motion in case of impact can be controlled. Restraint systems like the seat belt emphasise this aspect by keeping the occupant in the designated area and also connecting the occupant motion with that of the vehicle. Third, the actual impact, i.e. the contact between the human body and its environment, is addressed by way of energy absorption and distribution of the impact load over a greater contact area. Active safety, in turn, is a matter of braking systems, vehicle handling properties, pre-crash control (distance radar), and in an increasing fashion, computer-controlled driver assistance.

In addition to such technical advancements, also governmental bodies became increasingly aware of the safety issue in road traffic after World War II. By introducing first programmes for driver education, enacting driving regulations and developing new highway concepts, a higher safety standard was aimed at. Road planning and construction being furthermore within the responsibility of government agencies, management of accident risk by a favourable lay-out of the road environment, general traffic and speed control, guard rails, etc. is an important part of their task.

The reduction of casualties in road traffic as it is documented by accident statistics in many countries over the last decades (Fig. 1.5) can partly be attributed to the fact that trauma biomechanics' activities have predominantly been concerned with life-threatening injuries sustained in traffic accidents. However, as mentioned before, the traffic environment is only one field in which accidents occur. Injuries sustained in accidents at work, at sports or during daily life activities are also of importance. In an industrialised country (USA), the accumulated number of non-traffic (motor vehicle) related accidental deaths was in fact almost as high as the number of fatalities due to motor vehicle accidents in the age group 15–24 (Table 1.1). With increasing age, however, health-related problems prevail (Table 1.2).

In Table 1.4 the number for accidents causing injury in sports (without specification of severity) are given for the USA (2007) together with the corresponding numbers for Switzerland; a comparison which highlights on the one hand the

Table 1.4 Number of reported sports-related injuries in the USA

Sport	USA 1997	Switzerland 2003
Basketball	644,921	5,880
American football	344,420	na
Baseball, softball	326,569	na
Soccer	148,912	55,040
Trampolines	82,722	na
Skateboards	48,186	10,330
Golf, golf carts	47,777	na
Skiing	na	49,660
Snowboard	na	28,890
Sled, bobsleigh	na	10,800

Adopted from: Charles W. Nuttall, 5th Int. High Energy Physics Laboratories Technical Safety Forum, SLAC 2005) and Switzerland (adopted from: Swiss Council for Accident Prevention 2005. Note that favourite sports differ in the two countries: While American football and baseball are hardly played in Switzerland, soccer and skiing are quite popular

Table 1.5 Estimated fatal accident rate (FAR) per time spent for the associated activity and individual risk per person and year

Activity	FAR per 10^8 h exposure	Individual risk of death per person and year ($\times 10^4$)
Travel		
Air	na	0.02
Train	3–5	0.03
Bus	4	2
Car	50–60	2
Occupation		
Chemical industry	4	0.5
Manufacturing	8	na
Shipping	8	9
Coal mining	10	2
Agriculture	10	na
Boxing	20,000	na
Rock climbing	4,000	1.4

Adopted from: Practical industrial safety, risk assessment and shutdown systems, Dave McDonald, Elsevier 2004 situation in a large and a small industrialised country, but which also reflects the enormous differences between countries due to local habits and preferences on the other hand. The large number of injuries occurring in sports furthermore indicates that the vast majority of these injuries are not life threatening.

Considering worldwide statistics, it is nevertheless found that traffic accidents account for the highest number of fatalities: While the World Health Organisation (WHO) estimated a total of 1.2 Mio. fatalities from traffic accidents worldwide for the year 2002, the International Labour Organisation (ILO) extrapolated the number of fatal work accidents as 335,000 in 1998. The number of fatalities per time spent for certain activities may serve as an indicator for the danger associated with these activities. Table 1.5 demonstrates that traffic participation per se is not particularly dangerous in comparison with other activities (to a great extent due to the enormous efforts made in traffic safety), yet, this effect is by far surpassed by the time spent in traffic.

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Methods in Trauma Biomechanics 2

Work in trauma biomechanics is subjected to a number of limitations which are less stringent or even totally absent in other fields of the technical and life sciences. First of all, experiments involving loading situations with humans which are prone to cause injury are excluded. Second, animal models are of limited use because of the difficulty to scale trauma events reliably from animals up or down to humans. Questionable representativeness with respect to human biomechanics in spite of some similarity, furthermore, cost and above all ethical considerations along with public awareness limit however such experiments to special circumstances today.

Accordingly, methods applied in trauma biomechanics are to a great extent indirect and include mainly approaches based on

- statistics, field studies, databases (Sect. 2.1)
- basic concepts of biomechanics (Sect. 2.2)
- injury criteria, injury scales and injury risk (Sect. 2.3)
- accident reconstruction (Sect. 2.4)
- experimental models (Sect. 2.5)
- impact tests performed in the laboratory (Sect. 2.6)
- numerical simulation (Sect. 2.7).

2.1 Statistics, Field Studies, Databases

Epidemiology is of fundamental importance in trauma biomechanics and it also represents the oldest methodological approach. The identification of injury risks and the analysis of causative factors are largely based on epidemiological evidence which in turn stimulates the development of intervention strategies as well as of technical and legal countermeasures with the aim of accident prevention and injury reduction. Whether such countermeasures are indeed effective can again only be decided on the basis of statistical surveys which often require long-term studies. Hence, when working in the field of trauma biomechanics, in particular towards

K.-U. Schmitt et al., *Trauma Biomechanics*, DOI: 10.1007/978-3-642-53920-6_2,

issues related to injury mitigation and prevention, the acquisition and in-depth analysis of real world accident data is an indispensable prerequisite and research tool.

The collection, classification and interpretation of accident data have to be subjected to a careful assessment with respect to the sampling process in that in most cases the available data set is not exhaustive but is limited to a selected sample. One should always be aware of the fact that major limitations on the applicability of the results of any statistical evaluation are already incorporated in decisions on how and what data are collected. In contrast to fully controlled laboratory experiments, uncertainties arise for example due to the fact that many important parameters in real accident situations are not monitored and may exhibit a large variability. In addition, the memory of those involved in an accident or acting as witnesses may be inaccurate about the details or influenced by legal or insurance related considerations. Other factors such as the current composition of the vehicle fleet in case of traffic accidents, the price of gasoline, changes of legislation, adaptation of rules in contact sports, or changes with respect to insurance coverage of workplace accidents have to be considered when attempting to analyse the influence and effectiveness of newly introduced safety measures. A sound statistical evaluation may also fail because of an insufficient number of cases available for a representative analysis.

With respect to methodology, two types of accident data bases or injury surveillance systems can be distinguished, viz., general accident data collections involving a large, possibly complete coverage of accidental events on the one hand, and in-depth studies of selected cases on the other hand. General large-scale accident files are typically collected by the police, other government bodies or insurance companies and are presented in annual accident statistics. They usually contain a large number of cases but only limited information per case. In turn, in-depth case analyses are performed by specialised teams which attempt to recover as much detail as possible of each case under scrutiny—which somewhat cynically can be regarded as an involuntary experiment—on the basis of investigation of the accident scene, workplace or household locations and installations, vehicles, sports accessories, furthermore, police reports, witness depositions, interviews, medical records, weather reports, video coverage of sports events and on-site reconstruction with original vehicles or installations. Numerical simulation is then often applied to elucidate loading conditions and to relate them to injury patterns. Needless to say that such investigations are associated with a high expense and only a limited number of cases can be evaluated in this fashion. Representativeness is a particularly critical aspect in this approach.

Insurance companies often have larger collections than governmental bodies because accidents are reported to insurance companies for financial reasons while more reluctance is present with respect to involving the police, in particular in case of self-accidents without the involvement of a second party. Yet, insurance data are often not accessible, and if so, biased or not detailed enough. For example, insurance companies tend to quantify vehicle damage more in terms of repair cost than in terms of the biomechanically more important deformation energy.

Cases included in large-scale data collections are moreover often not collected and analysed by specialists in accidentology and may contain significant errors and be selected according to criteria which are not applied uniformly. Accordingly, the results obtained from different data bases are often difficult to compare due to differences in the data collection schemes. Even within one specific data base type, e.g. police records, differences in basic definitions, data set volume or privacy policies may vary considerably from source to source. Whether e.g. an elderly patient who dies in a hospital from pneumonia two weeks after a severe traffic accident is indeed a traffic accident victim and included in the statistics or not may depend simply on the reporting practice of the hospital.

In most industrialised countries, accidents associated with traffic, workplace, household and sports fall within the competence of different government agencies, foundations, private institutions, sports associations, insurance companies, etc. with little mutual interaction. Reporting and investigation practises may differ along with injury prevention strategies such that comparisons between various types of injury-producing circumstances have to be made with great care. Uniform statistics are mostly available from small countries like Switzerland where the Swiss Council for Accident Prevention (bfu) provides a comprehensive coverage of accident data.

The largest systematic collections and statistics on traffic accidents are provided by the US National Highway Traffic Safety Administration (NHTSA). They include general data with respect to vehicles, crashworthiness and trends (National Automotive Sampling System, NASS) as well as information on traffic fatalities in the Fatal Accident Reporting System (FARS). An overview of these activities can, for example, be found in Compton (2002). Similar, although sometimes less systematic, information is available from most other countries worldwide. Work place safety issues are comprehensively addressed in the statistics of the US Occupational Safety & Health Administration (OSHA). In most industrialised countries, furthermore, workplace accidents are covered by government controlled insurance organisations. General statistics are regularly available from such sources.

The situation with respect to sports accidents and injuries is somewhat different. Sports activities are largely voluntary and leisure-based (with the exception of mandatory participation in schools), are mostly covered by special insurance programs (in particular when competitive events or contact sports are involved), and product liability is highly diverse and selective (e.g., trampolines, diving boards in swimming pools, American football helmets, ski bindings). Specific, let alone general statistics involving comprehensive coverage over years, e.g. to analyse trends are largely missing. General awareness with respect to sports injuries has only recently increased. The Olympic Committee established in 1990 a Medical Commission and Library involving a Special Collection of Sports Medicine and Sports Science where the injury problem is partially included. While some sports associations release no systematic information with respect to accidents and injuries, others record and analyse injuries very comprehensively. The Fédération Internationale de Ski (FIS), for example, has—in collaboration with the Oslo Sports Trauma Research Center—developed an Injury Surveillance System

(ISS) for the FIS disciplines of alpine skiing, cross-country skiing, ski jumping, nordic combined, freestyle skiing and snowboarding.

In-depth case studies are made by specialised teams, usually with a specific aim or involving a limited geographical area. In order to be useful, such efforts have to be maintained over years and a sufficiently large number of cases has to be collected observing uniform procedures. Most projects of this type which are documented in the literature are performed in connection with traffic accidents. For example, a team working at the Medical University Hannover (Germany) has been collecting data of collisions occurring in the area of the city of Hannover over many years. Since 1999, an additional research team also collects data in the city area of Dresden, the data of the two sites is combined in the GIDAS data base (www.gidas.org). Because the data was collected systematically and following a uniform protocol for a long time, it is for instance possible to analyse factors related to changes in vehicle design.

Another example is the data base on whiplash associated disorders causing a sick leave of more than four weeks duration which is hosted by AGU Zurich (Switzerland, http://www.agu.ch). The collection includes cases from the entire country of Switzerland. Due to the large amount of available data, specific topics concerning technical, medical as well as biomechanical aspects of soft tissue neck injuries can be addressed (e.g. Schmitt et al. 2003; Linder et al. 2013). Yet other in-depth investigations are made by vehicle manufacturers where specialised teams investigate cases in which vehicles of their own production are involved in order to assess the effectiveness of safety measures and identify needs for improvements. Some of these latter accident data bases also include cases where vehicle damage occurred, but no injury was recorded. Such data are particularly helpful for statistical analysis, as they offer the possibility of well-defined control groups, which are not necessarily available in other types of data bases.

Having recognised that an adequate supply of road accident and injury records is perceived to be important for the selection, implementation and evaluation of road safety measures, several approaches such as for example the European STAIRS project (Standardisation of Accident and Injury Registration Systems, 1997–1999) or, more recently, the European Road Safety Observatory (www.erso.eu), that are aimed at harmonising accident data collections in order to allow more comprehensive and comparable studies, were conducted.

Little such efforts are underway for workplace, household or sports injuries which in view of increasing globalization and international mobility may cause, among other, problems with liability and insurance coverage.

2.2 Basic Concepts of Biomechanics

In the following paragraphs, a number of basic mechanical concepts which are of importance in trauma biomechanics are shortly reviewed. A more general overview over the mathematical methods used in biomechanics can be found in

Niederer (2010). In general mechanics, a distinction is made between rigid body mechanics and continuum mechanics. In real applications, both formulations are associated with assumptions and approximations such that their applicability, validity and limitations have to be carefully assessed in each problem to be approached, in particular, when applications in biomechanics are considered.

The aim of mechanics consists of a quantitative description of the effects that forces exert on the motion and deformation of bodies, in case of biomechanics, primarily living objects. To this end, mass, time, position are the fundamental independent quantities as function of which all other mechanical quantities are expressed.

Rigid body mechanics: Basic quantities are mass m, time t, position $\vec{r}(t)$, associated quantities are moment of inertia I, angular velocity $\vec{\omega}(t)$. The position vector $\vec{r}(t)$ denotes the location of the centre of mass of a rigid body as function of time. Further quantities derived thereof are the velocity of the centre of mass $\vec{v}(t) = \frac{d}{dt}\vec{r}(t)$, and the acceleration $\vec{a}(t) = \frac{d^2}{dt^2}\vec{r}(t)$.

The linear motion of the rigid body is described by Newton's second law of motion:

$$m \cdot \vec{a}(t) = \sum_i \vec{F}_i(t) \tag{2.1}$$

where the sum extends over all forces $\vec{F}_i(t)$ acting on the body. The spatial orientation of the body, in turn, is obtained from the angular momentum equation,

$$I \cdot \frac{d}{dt}\vec{\omega}(t) = \sum_i \vec{M}_i(t) \tag{2.2}$$

with the angular acceleration $\frac{d}{dt}\vec{\omega}(t)$ and the sum over all moments $\vec{M}_i(t)$ acting on the body. Because of the solidification principle, these equations also hold for deformable bodies, however, the centre of mass is not at a constant location with respect to the contour of the body in such cases. Variational principles, which can be derived within the framework of Newtonian mechanics, lead to Lagrange or Hamiltonian formulations which may be useful depending on the application under consideration.

Continuum mechanics: Basic quantities are field-oriented, viz., density $\rho(\vec{r}, t)$, time t, velocity field $\vec{v}(\vec{r}, t)$. The density $\rho(\vec{r}, t)$ as well as the velocity field $\vec{v}(\vec{r}, t)$ refer to a specific, fixed location $\vec{r}$ in space (this approach is often denoted as Euler representation of the continuum). The equation of motion reads (the independent variables are omitted for brevity)

$$\frac{\partial}{\partial t}(\rho\vec{v}) + \left[\vec{v}, \vec{\nabla}\right] \cdot (\rho\vec{v}) = \vec{k} + \left[\vec{\nabla}, \hat{\sigma}\right] \tag{2.3}$$

where $\vec{k}(\vec{r}, t)$ denotes field forces, e.g., gravity, while the stress tensor $\hat{\sigma}(\vec{r}, t)$ describes the internal state of loading (i.e., forces per unit area as normal and shear stresses) of the continuum and includes in addition forces which are due to external contact. $\vec{\nabla}$ is the Nabla operator and vectorial quantities in brackets separated by comma denote a scalar product. The angular momentum relation requires that the stress tensor $\hat{\sigma}$ be symmetric. Conservation of mass furthermore yields the continuity equation

$$\frac{\partial}{\partial t}\rho + \left[\vec{\nabla}, (\rho\vec{v})\right] = 0 \tag{2.4}$$

These equations are non-linear and the velocity field can be obtained as a solution providing that the mechanical characteristics of the continuum in the form of a constitutive equation (see below) are introduced (a textbook on continuum mechanics is, e.g., Liu 2002).

In order to proceed, a distinction between a solid and a fluid continuum has to be made. In case of a solid, the velocity field follows readily from the displacements that the particles making up the continuum undergo as function of time. A wide variety of constitutive equations relating the displacements (or the deformations of the continuum resulting thereof) to the stress state can be found in the literature. For fluids, in turn, the stress tensor can be formulated in terms of the velocity field and its gradients.

While rigid body models are characterised by a finite number of degrees of freedom associated with a set of ordinary differential equations, in continuum mechanics partial differential equations prevail and the number of degrees of freedom is infinite. For numerical treatment, the partial differential equations have to be approximated in special formulations involving in particular discretisation, of which the finite element approximation is most often used in trauma biomechanics (see Sect. 2.7).

Constitutive Properties of Biological Tissues: Stress–strain characteristics of solid biological tissues are typically non-linear, anisotropic and visco-elastic. The non-linearity is mainly due to the large tissue deformations that are observed in biomechanics, the anisotropy to the fibrous character of biological tissues and the visco-elasticity to the internal friction inherent in the fibre-extracellular matrix composition. There are furthermore active elements (muscle fibres) whose tone influences the mechanical properties. In tests made under ex vivo conditions, the state of muscle activation has to be taken into account (muscle fibres can be activated chemically, e.g., by Barium compounds). Likewise, embalming of cadavers changes their mechanical behaviour. For biological fluids, non-Newtonian characteristics may be important (for a comprehensive treatment of constitutive properties in biomechanics see Holzapfel and Ogden 2006).

A distinction is often made in biomechanics between “soft” and “hard” tissues. In order to specify this difference more quantitatively, the non-linear, anisotropic, partly active (muscles) properties of biological tissues have to be characterised by a simplified linear approximation. Under uniaxial loading of a long and thin specimen, a piecewise linear stress–strain relation in the form of Hooke’s law can be adapted and a local modulus of elasticity or Young’s modulus E can be defined.

For "soft" tissues, E varies typically between some 10 and 10^5 kPa, whereas the values for "hard" tissues are on the order of several GPa.

While there are numerous kinds of soft tissues, hard tissue in humans appears essentially in the form of calcified tissue, in particular bone. Thereby, the calcium is contained in hydroxyapatite crystals [Ca_5 $(PO_4)_3$ OH] which are embedded in a collageneous matrix. Aside from the integrity and mechanical loading capacity of bones, a physiological calcium balance is eminently important for the overall homeostasis of the human body in that calcium is essential for many physiological processes, among them, the action of muscles, the transmission of nerve signals or the coagulation of blood. As such, calcium is by far the most abundant bone mineral material ("calcium reservoir"), others, e.g., phosphor, being much less concentrated. Therefore, the terms "calcification" and "mineralization" of bone are often used synonymously. Not surprisingly, bone mineral density (BMD) has been found to be a significant determinant with respect to fracture risk (Beason et al. 2003). Low calcium content in bones, as in the case of osteoporosis, increases the risk of bone fracture and lowers injury tolerance.

Essential constituents of soft tissues from a biomechanics' point of view are elastin, collagen, and smooth muscle fibres. Modulus of elasticity (again under the simplified approximation of a piecewise linearised treatment of uniaxial loading) elastin (a globular, highly extensible polypeptide) has a Young's modulus of 10^2–10^3 kPa, collagen (a stiff three-fold triple helix molecule) a Young's modulus of up to 10^5 kPa, while smooth muscle fibres cover a wide range of stiffness characteristics between elastin and collagen depending on the state of activation. The anatomy of the organs consisting of soft tissue is mostly determined by their physiological function, therefore, according to the great variety of physiological functions, the composition of soft tissues varies greatly and so does their mechanical behaviour under load.

Bone, in turn, is less variable, although it exists in various forms: Cortical bone makes up the shaft (metaphysis) of the long bones as well as the outer layer of other bones while trabecular or cancellous bone is located mostly in the medullary canal of long bones, particularly in regions close to joints (epiphysis) as well as in the spine and in bones whose primary task is not to support loads (e.g., skull, iliac crest).

Since injury is basically associated with deformations beyond yield, linear approximations of the mechanical behaviour are generally questionable and great care has to be exercised when such procedures are applied. In reality, prior to irreversible, injurious tissue destruction, a mostly non-destructive, non-linear visco-elastic deformation behaviour sets in, followed by a plastic deformation phase. In soft tissues, plasticity is mainly due to an in general reversible rearrangement of tissue fibres. In the case of hard tissues, the processes underlying plastic deformation are less clear, they can however be visualised experimentally (Fig. 2.1). It has in fact been estimated that peak stresses in bone may significantly be reduced due to plasticity (Stitzel et al. 2003).

Age dependence of constitutive properties is prominent. While soft tissues in young children are highly deformable, with increasing age, stiffening sets in. This effect is mainly due to decreasing water content and increasing fibre cross-linking.

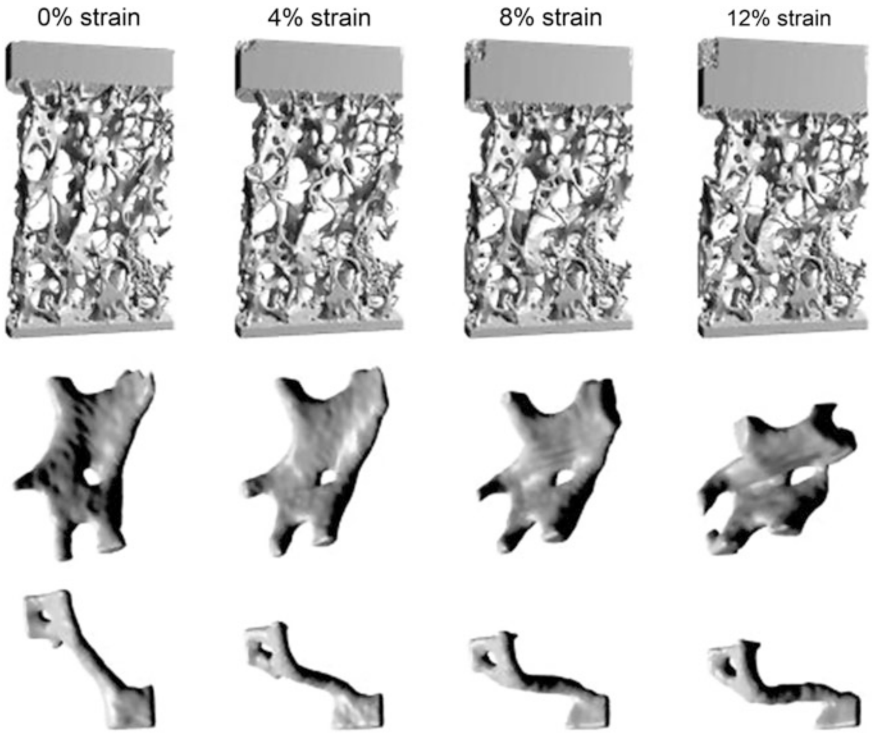


Fig. 2.1 Image-guided failure assessment of human spine samples with the aid of micro-CT (edge length of cross section 4 mm). The upper row exhibits a compressed specimen, imaged in steps of 4 % strain. The middle and lower rows show that the typical constituents of trabecular bone, viz., plates and rods, respectively, can undergo large plastic deformation before ultimate failure. Since the basic material of bone, i.e., hydroxyapatite crystals in a collageneous matrix is the same also in cortical bone (although the microstructure is quite different), local plastic deformation may also occur in cortical bone sections under load (From: R. Müller et al. Functional Microimaging at the Interface of Bone Mechanics and Biology, in: Holzapfel and Ogden, op. cit.)

While the total body water during adolescence amounts up to 70 % body weight, it decreases down to almost 50 % in old age. The younger a child, furthermore, the more bendable a bone is because of the gradual development of mineralisation. Accordingly, fractures denoted as "greenstick" fractures are observed in children in contrast to adults where fractures tend to exhibit a more brittle appearance.

Two major failure criteria are often applied in general mechanics in that it is assumed that failure sets in if a limit value of one of the following parameters is exceeded:

- absorbed energy (von Mises criterion, applied in trauma biomechanics, e.g., in the thorax).
- shear stress (criterion according to Tresca, usually not applied in trauma biomechanics).

In trauma biomechanics, in addition, the following quantities are used for the formulation of failure criteria, i.e., onset of injury (see Sect. 2.3), viz.

- acceleration (applied, e.g., in head injury).
- deformation (applied, e.g., for the assessment of bone fractures).

2.3 Injury Criteria, Injury Scales and Injury Risk

Injury criteria are important tools to assess the severity of accidental loading and the risk of sustaining injury thereof. By definition, an injury criterion correlates a function of physical parameters (e.g., acceleration, force) with a probability of a certain body region to be injured in a specific fashion (e.g., concussion, fracture). Injury criteria are generally derived from experimental studies in combination with empirical evidence, and their formulation and validation requires an extensive stepwise extrapolation procedure, since, as mentioned above, experiments on living humans at traumatic levels are excluded.

First, in addition to the concept of "injury criterion", two further expressions have to be introduced, viz., "damage criterion" and "protection criterion". While an injury criterion is intended to describe the property with respect to injury tolerance of living tissue, a damage criterion normally relates to post mortem test objects as surrogates for the living human. In both cases, a threshold value for the exposure to a quantity calculated from physical parameters is established above which, i.e., if the exposure exceeds the threshold, the test tissue in question is injured with respect to its anatomical or physiological structure in a specific fashion in more than 50 % of all experiments made or accidental exposures under comparable conditions. A protection criterion is obtained when postulating a threshold value on the basis of measurements performed with an anthropomorphic test device (see Sect. 2.6.1) as a human surrogate. In the latter case, the relation to human injury tolerance levels is mainly derived from empirical investigations. It is thereby assumed that a healthy middle-aged adult does on average not sustain injuries of the kind addressed by the particular criterion if he or she is exposed to loading conditions which are comparable to the ones defined in the protection criterion. The actual risk of injury can then be estimated with a risk function which relates the probability to be injured to the criterion developed (i.e. the underlying mechanical properties measured). A threshold value is defined such that, given a certain loading scenario represented by a certain value for the criterion, the risk of sustaining injury does not exceed a percentage of 50 %. Depending on the type of injury, this threshold may also be selected at a lower value of e.g. 20 %.

However, the definitions of injury, damage, and protection criteria are often not clearly differentiated and thus the term injury criterion is widely used for any index meant to quantify impact or accidental loading severity. Protection criteria, in turn, are determined in internationally standardised test procedures, mostly for use in automotive laboratories. These procedures are listed in Sect. 2.6. In the Chaps. 3–8 specific injury criteria for each body region are presented.

Table 2.1 The AIS classification

AIS code	Injury
0	Non-injured
1	Minor
2	Moderate
3	Serious
4	Severe
5	Critical
6	Untreatable

Scales to classify the type of an injury are based on medical diagnosis and were developed for injuries sustained in traffic accidents. The most widely used such scale is the Abbreviated Injury Scale (AIS), which was first developed in 1971 as a system to define the severity of injuries throughout the body and which is regularly revised and up-dated by the Association for the Advancement of Automotive Medicine (AAAM). AIS is a standardised system for categorising the type and severity of injuries arising from vehicular crashes (Table 2.1) and is oriented towards the survivability of an injury, i.e., each category represents a certain threat-to-life associated with an injury. Thus, AIS is an anatomically based, global severity scoring system that classifies each injury in every body region by assigning a code which ranges from AIS0 to AIS6. Higher AIS levels indicate an increased threat-to-life. AIS0 means "non-injured" and AIS6 "currently untreatable/maximum injury".

As a result, the AIS severity score is a single, time independent value for each injury and every body region. The severity is described regarding its importance to the whole body, assuming that the described injury occurs to an otherwise healthy adult. However, it has to be noted that the AIS considers only the injury and not its consequences. Clinical complexity, cost of surgical treatment, and long-term sequelae are in particular not taken into account. Hence, severe impairments such as loss of eyesight or life-threatening complications due to nosocomial infections occurring in a hospital are not coded as severe injuries, because they do not represent an initial threat-to-life.

Moreover, the AIS code is not a linear scale in the sense that the difference between AIS1 and AIS2 is comparable to the one between AIS5 and AIS6. It does therefore not make sense to calculate average AIS codes (AIS 3.7, e.g., is a meaningless number). To describe an overall injury severity for one person with multiple injuries, the maximum AIS (MAIS) is used. The MAIS represents the highest AIS code sustained by one person on any part of the body, even if the person in question sustained several injuries of the same severity level at different body parts. If, for example, a car occupant sustained AIS2 injuries on the head and the legs but no injuries classified higher, the MAIS will still be MAIS2.

To account for a better representation of patients with multiple injuries, the Injury Severity Score (ISS) was introduced which is regularly updated like the AIS scale (latest version: AAAM 2005 Update 2008). The ISS distinguishes six different body regions: head/neck, face, chest, abdomen, extremities including pelvis, external (i.e. burns, lacerations, abrasions, and contusions independent of their location on the body surface). For each of these regions the highest AIS code is determined. Then the ISS is calculated by the sum of the squares of the AIS codes of the three most severely injured body regions. Thus the minimum ISS is 0 and the maximum ISS is 75 (i.e. three AIS5 injuries). If an AIS6 injury is recorded, the ISS is automatically assigned to 75. ISS values higher than 15 are regarded as major trauma. Several studies have shown that the ISS correlates quite well with several measurement systems such as mortality (e.g. Baker and O'Neill 1976) or long-term impairment (e.g. Campbell et al. 1994). However, there are also limitations associated with the ISS such as the maximum of three contributing injuries. Consequently an further developments like the New Injury Severity Score (NISS) as well as other injury scales were presented (see e.g. Chawla et al. 2004).

In addition to the AIS, other scales are used to specify injuries of particular body regions in more detail. The Quebec Task Force (Spitzer et al. 1995), for example, established a scaling scheme to categorise soft tissue neck injuries (see Chap. 4). A classification scheme for head injuries often seen in emergency medical reports is the Glasgow Coma Scale (GCS) (Teasdale and Jennett 1974). GCS aims at describing the state of consciousness and some neurological signs (e.g. reflexes) of the injured person after a traumatic incident, and may thus allow the inclusion/exclusion of potential injury mechanisms. The scale ranges from 3 (deep coma) to 15 (fully awake).

Further scales address impairment, disability and societal loss through ratings of the long-term consequences of the injury by assigning an economic value. An example is the Injury Cost Scale ICS (Zeidler et al. 1989), by which the average costs for an injury is determined taking into account the costs for medical treatment and rehabilitation, loss of income and disability. Further economic scales are the Injury Priority Rating IPR (Carsten and Day 1988) and the HARM concept (Malliaris 1985) applied by the US government. One of the most crucial problems in trauma biomechanics is the assessment of the relationship between injury severity and a mechanical load which causes this injury, i.e. to find a relationship that allows assigning probabilities which describe the likeliness that a certain mechanical load (e.g. determined by an injury criterion) will cause a particular injury. This is important because without such correlations, it is useless trying to interpret any results obtained, for instance, in crash tests. Hence, it is necessary to perform well-equipped laboratory experiments using human surrogates to determine the biomechanical response and corresponding injury tolerance levels and consequently establish so-called injury risk functions.

For the determination of injury risk curves basic statistical methods are applied of which the maximum likelihood method, the cumulative frequency distributions, and the Weibull distribution are most often used. In Chap. 3, an example with respect to head injury is presented. For in-depth information however with respect to the application of statistical methods to the often complex and difficult analysis of accident and injury data the reader is referred to statistical text books. Great care has to be exercised in such analyses; among the various problems which may arise when transforming experimental results to (real world) injury risk functions, are

- a small number of tests performed,
- differences in the biomechanical response between the human surrogates used in testing (e.g. cadavers) and living humans,
- anthropometric differences between the test subjects and the real world population at risk,
- a large spread of data due to different test conditions used by different researchers,
- a large number of possible injury mechanisms and injuries that might occur.

Basically the same limitations apply when using data from accident statistics instead of experimental results to fit injury risk curves. Nonetheless, decades of trauma biomechanics' research have provided a sufficiently large number of sources that allow establishing a number of well-founded relationships that link mechanical loads to injury probability—at least for certain injuries and injury mechanisms, respectively. However, work is this area is by far not finished and revisions of existing criteria on the basis of new findings are not uncommon.

2.4 Accident Reconstruction

The reconstruction of accidents is an indispensable procedure in the field of trauma biomechanics because relations between loading and injury under physiological conditions manifest themselves only in real-life accidents. Likewise, accident reconstructions are often required for forensic purposes likewise in criminal and in civil cases.

The reconstruction of an accident consists of the mathematical analysis of the event in question on the basis of the laws of classical mechanics as outlined in Sect. 2.2. Other than laboratory experiments, however, accidents in everyday life occur under largely uncontrolled and unmonitored conditions. Depending on the extent, quality and accuracy of the available documentation, therefore, the specialist in accident reconstruction has to apply assumptions and approximations at quite different levels of complexity. While an accident in a skiing competition may be covered by various video recordings or the traces in a traffic accident may accurately be documented by the police, a fall from a ladder during household activities is hardly documented. All information is of importance in a reconstruction process. Much as in a puzzle, various sources of information have to be combined in order to produce a reliable and conclusive account of the events; this may include facts as different as the sequence of traffic lights in a vehicle–pedestrian impact and the bending stiffness of a pole in case of a sports incident. A scrutiny of the accident scene is always indispensable. Experience from formerly performed tests under laboratory conditions or the results from well documented "comparable" accidents may furthermore be of help. Of paramount importance is often the collaboration with the medical forensic expert in that injury patterns may provide useful clues for the purpose of accident reconstruction; for example, from the particular appearance of street dirt under the skin, the direction of a fall can be deduced.

Missing documentation or missing visible evidence may pose problems in accident reconstruction. In case of vehicle collisions, uncertainties might arise e.g. if due to anti-locking systems no skid marks are produced. Furthermore, reconstruction becomes more difficult when no or only marginal vehicle deformation occurs. In order to reduce the repair cost, modern vehicles are designed such that in collisions of low intensity nearly no damage is caused (or at least it is not visible from the outside and therefore often mistaken as not existing by laymen). However, missing visible damage does neither mean that there was no collision at all nor that the energy transmitted might not have been sufficient to cause injury to the occupant.

Within the framework of a rigid body approximation (Eqs. 2.1 and 2.2) for the description of an impact event, empirical investigations and laboratory experiments have shown that the acceleration of the centre of mass experienced by a body limb under the influence of impact forces is an important parameter to assess the severity of an impact. In many practical cases, the modulus of the acceleration is thereby often related to the acceleration due to gravity, g ($1\ \mathrm{g} = 9.81\ \mathrm{m/s^2}$), because we are constantly exposed to gravity such that we can relate a given acceleration level to everyday experience. Yet, the acceleration which a body undergoes during the course of an accident varies with time, such that the quantities "peak acceleration" and "mean acceleration" along with the corresponding intervals in time should always be clearly distinguished in order to prevent misunderstandings.

Reconstruction techniques have mostly been developed systematically for traffic accidents. In such cases, a number of specific parameters relating to an involved vehicle have found to be useful for an assessment of the loading situation of occupants.

- The collision or impact velocity of a vehicle is probably the parameter most frequently quoted in the public. In accident reconstruction, the travelling speed or, more accurately, the speed before the beginning of any braking action, is sometimes of importance when investigating whether or under which circumstances a collision could have been avoided, or whether a speed limit was exceeded.
- The collision-induced velocity change (delta-v) of the vehicle under consideration is, however, in most cases more useful for describing the collision severity where the effects of the collision on the occupants are concerned. The delta-v corresponds approximately to the integral of the translational vehicle deceleration over the collision time for collisions which are characterised by a single impact without significant rotation of the vehicle. Yet, in complex collision situations (roll over, fall over the roadside, etc.) delta-v may not be a well-defined parameter.

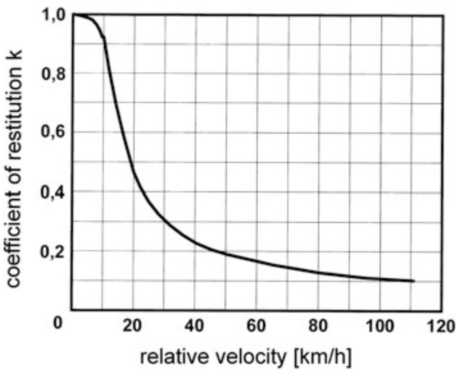


Fig. 2.2 Schematic representation of the relation between the coefficient of restitution and the relative velocity for a frontal impact on a rigid barrier for a passenger car [adapted from Appel et al. 2002]. Newer cars will generally exhibit higher coefficients of restitution in the low-speed range than shown in this figure

- The energy equivalent speed (EES) characterises the amount of energy needed to deform a vehicle. In fact, the EES represents the impact velocity into a rigid barrier that would have been necessary to cause the same permanent deformation as observed in the real world accident. The EES is given in [km/h] and can be obtained for many vehicle types from so-called EES catalogues. These catalogues are established on the basis of crash tests conducted under well-defined test conditions.
- A further parameter used to describe impact conditions is the vehicle overlap. This is the extent to which the vehicle and the collision partner (e.g. another vehicle or a barrier in a crash test) overlap. The overlap is generally presented as the percentage of the total width of the vehicle under consideration covered by the opposing vehicle (or wall).
- From basic mechanics, the principles of elastic and plastic impact and the accompanying coefficient of restitution (k-factor) are used to characterise the elastic and plastic (i.e. permanent) components of the deformation suffered in the impact. Figure 2.2 shows, as an example, the dependency of the coefficient of restitution on the impact velocity (against a rigid wall). The k-factor heavily depends on the design of the car front structure, in particular the bumpers and underlying absorbers for low-energy collisions. Due to the requirements for no or little dam-age cost in these collisions, bumpers have been designed to be stiffer and more elastic, thus, for newer cars, higher coefficients of restitution must be assumed in the low-speed area. Furthermore, some impact absorber concepts involve designs or materials whose deformation recovers slowly after an impact. Since this restitution does not occur during the impact itself, the vehicle deforms in a fully plastic way although the accident investigator may not find any deformation after the collision.

Today, most traffic accident reconstructions are performed with facilitating computer programmes which mainly employ rigid body dynamics (Eqs. 2.1 and 2.2). Using such programmes, two methods can be distinguished in principle: “forward” and “backward” calculation. In the first case, the kinematics before the collision are assumed, i.e. initial directions of motion, velocities etc. are assigned to the collision partners. Then, the actual collision and the final positions of the collision partners after the collision are determined by integration of the rigid body equations taking into account tyre and collision forces. Finally, the positions and traces that were recorded on the actual accident scene are compared with the results of the calculation. In an iterative process, the input parameters are adjusted and the procedure is repeated until a satisfactory match between the results obtained in the calculation and the available accident data is reached. The backward calculation method starts by investigating the final positions of the collision partners. Next, the motions after the impact are reconciled with the traces found (e.g. skid marks) giving the positions at impact, again utilizing rigid body approximations. Eventually, the initial parameters that lead to the determined course are obtained. Graphics are finally used to give a visual account of the reconstructed accident.

Because of the large mass ratio car occupant/vehicle, the influence of car occupants, likewise of other objects which are not rigidly connected with the vehicle can be taken into account in an approximate fashion. This is not the case in motorcycle or bicycle accidents, where the programmes mentioned above can only be applied under restricted conditions and the results have to be interpreted carefully.

Collision phases, not only in traffic accidents, are usually associated with deformation processes for which the application of approximations based on continuum mechanics (Eqs. 2.3 and 2.4 and associated constitutive relations) are required. Because of liability issues mostly, car manufacturers are reluctant to publish the finite element models which they use to assess the crashworthiness of their vehicles. Various types of simplification are therefore made in general purpose reconstruction programmes. One way is to assume a segmented stiffness distribution of the vehicle's front, and then to integrate the equations of motion of the two vehicles over the collision duration. Another way, often employed in European reconstruction programmes, is to assume the collision duration to be infinitely short (in comparison to the pre- and post-crash motion of the vehicles) and to calculate only the transfer of the (linear and rotational) momentum from one vehicle to the other. The EES values mentioned above may, for both approaches, be used as control values to obtain not only conservation of momentum, but also the energy balance over the collision.

Once a vehicle motion is reconstructed, the motion of the occupants or of an impacted external victim (pedestrian, two-wheeler) during impact can be estimated, again using rigid body models. Furthermore, indications with respect to the occupant loading can be obtained. Further extrapolations, in particular concerning injuries, however require expertise beyond the classical (mechanical) accident reconstruction. The same holds true for accidents occurring at the workplace, household or in sports. Given appropriate circumstances and a careful adaptation to the situation in question, traffic accident reconstruction models and computer programmes can also be utilised in other accidents. For the purpose of injury analysis, the subsequent application of a finite element model of the human body may yield useful clues.

Finally, accidents are sometimes reconstructed by a one-to-one reproduction on location or in the laboratory with the original installations, vehicles, sports accessories, etc. This procedure is particularly important in non-traffic related accidents as well as in the course of legal procedures where large claims justifying the often considerable expense of such tests are involved.

2.5 Experimental Models

All mechanical characteristics relating to the behaviour in time of the human body, of a part of it, of an organ or tissue when it is subjected to dynamic mechanical loading is subsumed under the term "biomechanical response". The head-neck kinematics as observed in a rugby scrummage or the force–deflection characteristics of the chest due to a frontal vehicle impact are examples for the biomechanical response of the human body. Besides such mechanical changes, the biomechanical response can also lead to physiological changes like neck pain, oedema of the lung or aberrations of the ECG.

A thorough knowledge of the biomechanical response is indispensable for the development of measures for injury prevention and mitigation. Since accident situations as such are highly dynamic by their nature, relevant tests to investigate the biomechanical response of the human body have generally to be conducted under corresponding loading conditions. Nevertheless, whenever extrapolations to dynamic conditions are possible, quasi-static tests are made because of the much simpler installations needed for such tests.

The analysis of the biomechanical response of the human body is not only crucial for an understanding of injury mechanisms, but it is also needed for the definition and verification of injury tolerance thresholds. An important aspect thereby is the biological variability. In particular, age-related changes are prominent. For a reliable measurement of an injury risk function, a large amount of experimental data is therefore required. As biological material for testing purposes is not readily available, a careful examination of statistics is of primary importance. Response data may also be restricted by the impossibility to install instrumentation at the desired location. Bearing in mind that many of the relevant studies represent pioneering work in trauma biomechanics research dating back to the 1940s, some of these shortcomings can be explained with the lack of adequate measurement instrumentation and the lack of knowledge at that time. In the chapters dealing with the biomechanical response of the different body regions these problems are discussed in more detail. Furthermore, Sect. 2.6.1 is devoted to the utilisation of human surrogates (dummies) used in impact testing where the response data obtained from the surrogate have to be interpreted in light of biological verisimilitude.

In the following, experimental models used to determine the biomechanical response of the human body are briefly discussed. Five different models can thereby be distinguished, viz., human volunteers, human cadavers, animals, mechanical human surrogates and mathematical models.

Volunteer experiments are, for obvious reasons, restricted to the low severity range only, i.e. well below any level thought to be possibly injurious. The pain threshold is often taken as the upper limit up to which mechanical loads are applied. Advantages related to volunteer tests are first of all the use of the "correct" anatomy and physiological state. Moreover, the influence of the muscle tone can be studied and effects like the bracing prior to a collision can be considered. However, the cohorts used for volunteer tests are usually not statistically representative for the population at risk. Particularly, females, children and the elderly are strongly under-represented in the available volunteer data. Difficulties also arise with the instrumentation as load cells can often not be brought to the location of interest (e.g. the centre of gravity of the head or the first thoracic vertebra), even a rigid external fixation is difficult to obtain because of the skin. Advances in high-speed video camera technology along with sophisticated mathematical post-processing have considerably contributed to the improvement of such results. Even cineradiography has sometimes been used to monitor the response of the skeleton to impact, e.g. by Ono and Kaneoka (1997) to investigate the motion of the vertebrae of the cervical spine. As the number of subjects tested in this fashion is particularly small, questions of scaling to other groups of humans as well as to a higher impact severity are all the more critical.

Human cadavers (usually denoted as post mortem human subjects (PMHS) or post mortem test objects (PMTO)) are the second type of model used to determine human biomechanical response. Despite the great anatomical similarity to the living human (a PMTO may to some extent be compared with a sleeping human), several influencing factors have to be considered. First, the age of the PMHS is often high. Age-related degeneration is therefore often prevalent in the cadaver cohort available for a test series. For example, in case of osteoporosis, fracture is observed too frequently. Second, the lack of pressure in the lungs and the blood vessels, the absence of muscle tone, as well as differences due to preparation techniques used (i.e. embalmed vs. non-embalmed cadavers) significantly influence the biomechanical response. Fresh cadavers, however, were shown to be good models for the detection of fractures, vessel ruptures and lacerations. Nonetheless, physiological responses (e.g. the neck pain or ECG aberrations) cannot be addressed with such models. For the investigation of the response of a single body part only, for instance of the leg (see Chap. 7), isolated cadaver parts are used. Here the connection to the rest of the body has to be mimicked in the test set-up in an appropriate way.

Animal models have a limited significance for human trauma biomechanics. Nevertheless, anaesthetised animals offer the only possibility to investigate physiological reactions to severe mechanical loading. Animal experiments also allow a comparison between living and dead tissue and thus give important input to the proper interpretation of cadaver tests. However, due to differences in anatomy and physiology, the possibilities of scaling the results obtained, particularly with respect to injury thresholds, are limited.

Further models used in trauma biomechanics include mechanical human surrogates, i.e. anthropomorphic test devices (ATD) as well as mathematical (computational) models. Because of their importance (e.g., all regulations on vehicle occupant safety are formulated in terms of measurements made on an ATD), these models are discussed in separate sections below.

The objective of impact testing in the laboratory consists of a realistic simulation of accident scenarios and of the determination of the mechanical loading that a human victim possibly sustains in such an accident. Most laboratory test setups are thereby made for vehicle crash testing mostly because of the comprehensive regulatory coverage of vehicle safety. In the automotive industry, extensive usage of crash facilities is made for the assessment of restraint systems as well as for the development of new measures in passive safety to reduce the number and severity of injuries sustained in automotive accidents. Yet, laboratory tests are also used to certify football helmets or ski bindings, etc.

Real world accident scenarios are manifold. Thus, only selected impact conditions which are thought to be of relevance are simulated in crash testing. Bearing in mind the need of repeatability and comparability of test results along with the cost and time related to crash testing, several standards were developed that define the exact test protocols, the evaluation process, as well as the protection criteria to be derived thereof. In Sect. 2.6 such standardised test procedures are described in detail.

Three different categories of automotive crash tests can be distinguished, viz., full scale tests, sled tests, and component tests (Fig. 2.3). The basic principles with respect to laboratory practice, evaluation of results and documentation also apply to non-automotive testing and certification procedures as, for example, the closing force of elevator doors or the strength of nets used by the fire brigade.

In full scale impact tests, a vehicle impacts an obstacle or another vehicle or is impacted by a moveable object (e.g. a barrier as used in side impact tests). Anthropomorphic test devices (i.e. crash test dummies) represent occupants located in the vehicle under consideration, and the kinematics and the mechanical loadings of the dummy are recorded during impact. Full scale crash tests have the advantage that the actual vehicle properties, e.g. the deformation characteristics, are inherent in the results. These properties influence the acceleration response of the vehicle and consequently the loading of the occupants. In addition to passive safety issues, full scale tests also provide information about the repair costs to be faced after a collision and are therefore performed by insurance companies with respect to the rating of the insurance premium. Full scale tests are also used for non-biomechanical purposes, e.g. to check the fuel system integrity or the braking system.

While in full scale tests the interaction between the restraint systems and the deformation characteristics is investigated, sled tests are primarily used to analyse the isolated behaviour of restraint systems or vehicle components (e.g. a front seat). For this purpose, parts of the vehicle or the components of interest are mounted on a sled. The sled is accelerated or decelerated, respectively, in a controlled manner without damaging the test rig. Consequently, the sled including parts of the rig can be re-used, thereby significantly reducing the associated cost. The disadvantages of this type of test are, among other, the restriction that the vehicle loading may only be unidirectional, and that the vehicle acceleration pulse must be established by a prior full-scale test or, in prototyping, by e.g. computer simulations.

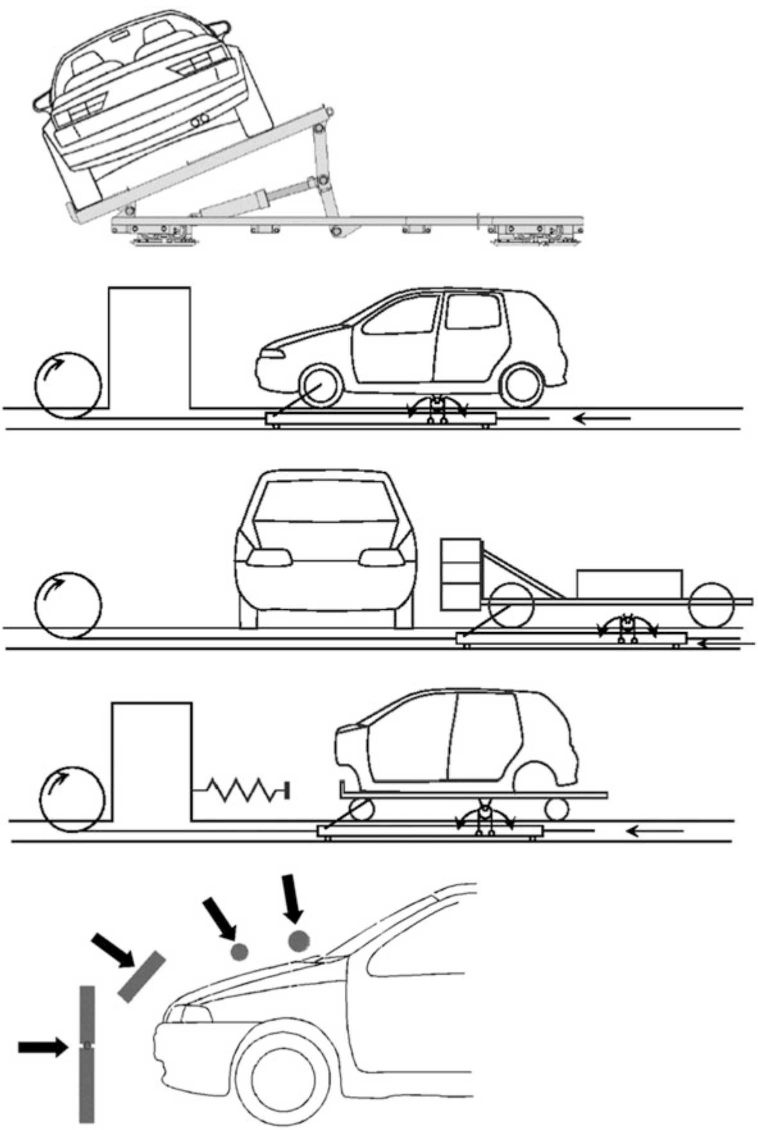


Fig. 2.3 Different methods of crash testing. From top to bottom: full scale testing (rollover test, frontal and lateral impact), sled testing and different impactors used in pedestrian safety testing of the front of a car

Component tests form a third type of testing. Here, in quasi-static as well as dynamic tests various aspects concerning single parts of the car body may be investigated. In quasi-static tension tests, for instance, the strength of the seat belt attachment points is examined. Furthermore, using devices such as the free motion head form (FMH) the compliance and the energy dissipation properties of the vehicle interior are assessed. The FMH is a head form mounted on a propelling device such that it can be projected onto the vehicle structure in question under different angles. Using other dummy parts (e.g. lower and upper limb surrogates and head forms simulating children and adult heads), pedestrian safety is assessed by evaluating the deformation properties of the vehicle front. As opposed to e.g. full scale tests, component tests offer the advantage that the point of impact of e.g. a head impactor on the bonnet may be specified with millimetre-accuracy. Thus, along with the fact that the cost of component tests is an order of magnitude lower, a large number of impact points may be assessed.

2.6 Standardised Test Procedures

All new car models are required to pass numerous tests related to occupant safety before they may be brought into circulation. These tests often differ in different regions of the world; the most important regional standards are those of the U.S. and of the European Community. In Europe, the corresponding procedures are laid down in the regulations of the UN Economic Commission for Europe (ECE). ECE R94, for example, describes the test procedure for frontal impact protection, while in ECE R95 the side impact test is defined. These regulations have been incorporated in the EC directives, where e.g. 96/27/EC contains ECE R95 and 96/79/EC includes ECE R94. For the sake of simplicity, we will refer to the older ECE Rxx designation in the following chapters. In the United States, the Federal Motor Vehicle Safety Standards (FMVSS) are incorporated in the Federal Register 49 CFR part 571. Since most car makers aim to sell their cars on a global market, the differing safety standards in different parts of the world constitute a considerable problem. International harmonisation of tests and the international recognition of test results obtained in a certified laboratory are important aspects in worldwide trade. To this end, numerous bilateral trade agreements between countries, furthermore free trade initiatives, UN, US and EU ("Cassis de Dijon" principle) activities were made or are under way. Therefore, the UN/ECE/WP.29 has been designated to develop harmonised regulations, called GTR (Global Technical Regulations).

Where aircraft crashes are concerned, the U.S. Federal Aviation Administration (FAA) has laid down some crash tests procedures in the FAR (Federal Aviation Regulations) parts, these are largely identical to the regulations by the European Aviation Safety Agency (EASA). In addition, instruments, machines, installations, sports accessories, etc. which are in daily use are subjected to a myriad of regulations, guidelines and recommendations made by government bodies, manufacturers, insurance companies, sports associations and consumer organisations. In different countries, quite different regulations and practises can be found. A general overview can hardly be made. In Europe, however, most safety requirements are tested in conjunction with product liability and constitute a part of a product certification process (CE mark).

Table 2.2 ECE regulations

Regulation	Collision type	Impact velocity [km/h]	Test conditions	Comments
R94	Frontal	56	40 % overlap, deformable barrier	2 Hybrid III dummies
R12	Frontal	48–53	Rigid wall	Concerning deformation of the steering assembly
R33	Frontal	48–53	Rigid wall	Concerning stability of passenger compartment
R12	Frontal	24	Impactor test	Determining force on body block impactor
R95	Side	50	Moveable, deformable barrier, 90° angle	1 EuroSID II at driver position
R32–34	Rear-end	35–38	Moveable, rigid barrier (mass: 1100 kg)	Integrity of the petrol system
R42	Minor collisions	2.5, 4	Pendulum	Checking safety in operation only
R44	Child restraint systems (CRS)	50	Sled tests	Different dummies used depending on CRS
R16	Seats	–	Static	Recliner moment, deformation
R17	Seats	–	Sled tests, 20 g	Seat anchorage to vehicle body, head restraint geometry
R14	Belts	–	Static	e.g. deformation

For details see http://www.unece.org

As can be seen in Tables 2.2 and 2.3, the ECE regulations and the FMVSS are quite similar and include many corresponding regulations. However, differences arise for the types of dummies requested, the test conditions prescribed or the evaluation of the tests (Fig. 2.4).

Furthermore, different protection criteria apply in some cases. The requirements stated in both the ECE regulations and FMVSS are also often adapted in other countries and therefore can be considered the most powerful safety regulations worldwide. For complete and up-to-date information the reader is advised to consult the corresponding internet sites, as these regulations are amended or changed on a regular basis.

Tables 2.4 and 2.5 summarise the requirements for occupant protection as defined in ECE R94 and FMVSS 208 for frontal impact and in ECE R95 and FMVSS 214 for lateral impact.

More details on the protection criteria mentioned and their threshold values are given in Chaps. 3–8 for the respective parts of the human body. It has to be noted that neither the ECE nor the FMVSS include regulations concerning occupant safety in low speed rear-end collisions although these occur frequently and cause

Table 2.3 FMVSS regulations

Regulation	Collision type	Impact velocity [mph]	Test conditions	Comments
571.208 (latest version phase 2)	Frontal	25	100 % overlap, 0–30° rigid barrier	2 unbelted Hybrid III dummies (50 % male)
		35	100 % overlap, 0° rigid barrier	2 belted Hybrid III dummies (50 % male)
		25	100 % overlap, 0° rigid barrier (max. 5° oblique)	2 unbelted Hybrid III dummies (5 % female)
		35	100 % overlap, 0° rigid barrier (max. 5° oblique)	2 belted Hybrid III dummies (5 % female)
		25	40 % overlap, 0° deformable barrier	2 belted Hybrid III dummies (5 % female)
		–	Various configurations, firing of airbags	Various dummies in OOP situations
571.204	Frontal	30	100 % overlap, rigid barrier	Steering assembly rearward displacement
571.212	Frontal	30	100 % overlap, rigid barrier	Concerning the mounting of the windscreen
571.203	Frontal	15	Impactor test	Determining force on body block impactor
571.214	Side	33.5	Moveable, deformable barrier, oblique impact	Old: 2 SID dummies, new: 1 ES-2re front and SID IIs rear
571.214	Side	20	Moveable pole, oblique impact	2 ES-2re or SID IIs
571.301 + 303	Rear-end, front, side	30	Moveable, rigid barrier (mass: 1800 kg)	Fuel system integrity
581	Minor collisions	2.5 (rear), 5 (front)	Pendulum/barrier	Checking safety in operation only
571.213	Child restraint systems (CRS)	30	Sled tests	Different dummies used depending on CRS
571.210	Seats	–	Static tests	e.g. deformation
571.209	Seat belts	–	Static tests	e.g. deformation

For details see http://www.nhtsa.dot.gov

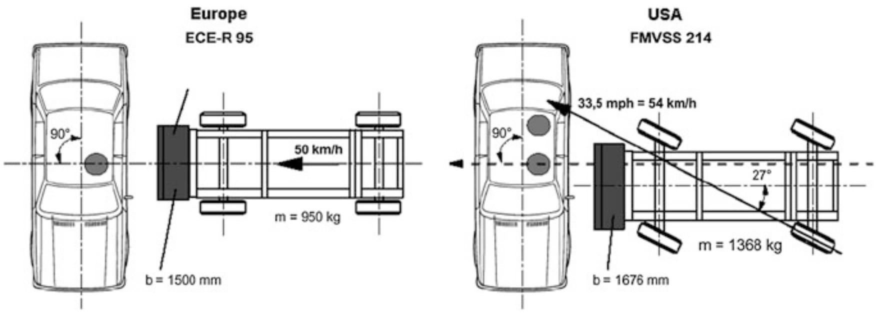


Fig. 2.4 Different test conditions for lateral impact are required by the ECE (*left*) and the FMVSS (*right*)

Table 2.4 Frontal impact threshold values

	FMVSS 208	ECE R94
Dummies	Hybrid III 50 % male, 5 % female	2 Hybrid III 50 % male
Head	HIC 15 < 700	HPC < 1000
		a3 ms < 80 g
Neck	$N_{ij} \leq 1.0$, {-4.17kN < Fz < 4.0kN} (Hybrid III 50 % male) {-2.62 kN < Fz < 2.52 kN} (Hybrid III 5 % female)	M_{ext} < 57 Nm
Thorax	a3 ms ≤ 60 g, deflection ≤ 63 mm (Hybrid III 50 % male)/ deflection ≤ 52 mm (Hybrid III 5 % female)	Deflection < 50 mm VC < 1.0
Femur	Axial force < 10 kN	Not exceeding defined force corridor
Knee	–	Deflection < 15 mm
Tibia	–	Axial force < 8 kN
		TI ≤ 1.3

Table 2.5 Side impact threshold values

	FMVSS 214	ECE R95
Dummies	ES-2re, SIDIIs	1 EuroSID II
Head	HIC 36 < 1000 (both dummy types)	HPC < 1000
Thorax	A max < 82 g (both dummy types) d max < 44 mm (ES-2)	VC < 1.0 d < 42 mm
Abdomen	F < 2.5 kN (ES-2)	Internal force < 2.5 kN
Pelvis	F < 5.1 kN (SIDIIs)/F < 6 kN (ES-2)	Pubic force < 6 kN

eminent health problems and associated cost. To fill this gap, a new test procedure was developed by AGU Zurich in collaboration with Autoliv GmbH Germany, GDV Munich and the University of Graz (Muser et al. 1999). A modified version of this procedure has been incorporated into an ISO standard by ISO/TC22/SC10.

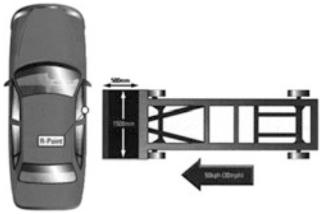
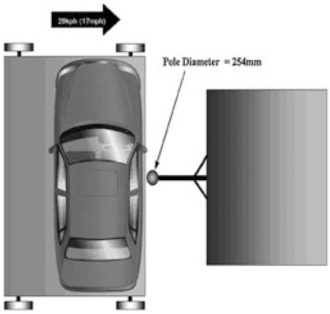
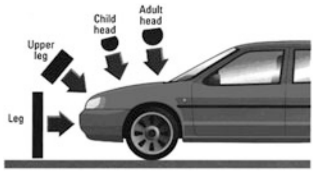
In addition to the crash tests required by governmental regulations, consumer tests are performed. As legislation provides a minimum statutory standard of safety for new cars only, and because the results from the governmental tests are not necessarily published, it is the aim of consumer tests to encourage car manufacturers to exceed these minimum requirements and make the results of these consumer tests publicly available. Thus, consumers can obtain reliable and accurate comparative information regarding the safety performance of individual car models.

In Europe, dummies have been used in consumer tests to determine the occupant loading even before regulations demanded such tests. Thus, the public was made aware of the importance of passive safety issues. Moreover, consumer tests are characterised by ranking systems which are intended to give the consumers the possibility to assess and compare the occupant protection potential of different vehicle types. Such rating schemes often include dummy symbols with coloured body regions ranging from green (i.e. low loading) to red, and final star ratings where the number of stars correlates with the total number of credits gained in the assessment. This total number of credits cannot only be obtained from vehicle performance during the crash tests, but credits are also given for safety features concerning prevention or driver education (e.g. an acoustic "fasten seat belt" reminder or an electronic stability control (ESC) will influence the final grade positively in EuroNCAP tests).

To date, the most important consumer tests are the so-called New Car Assessment Programs (NCAP). NCAP testing is performed in Europe (EuroNCAP), Australia, Japan and the US. The test conditions and ranking systems differ for different NCAP agencies. Table 2.6 lists the tests performed by EuroNCAP. It should be noted that in the US and e.g. Australia, NCAP tests were performed by government agencies long before non-government institutions like EuroNCAP came into play.

It should also be borne in mind that consumer test ratings do not necessarily reflect the biomechanical performance of a car in a crash in an absolute way, but rather relative to the other cars tested under the same conditions. Threshold levels or rating scales are normally selected such that e.g. a certain percentage of cars in a test series will be rated 'good' and another percentage will be rated 'bad', even if, in a hypothetical case, all cars of a series would exhibit biomechanically subcritical results. Furthermore, as consumer test programmes are not impeded by sometimes lengthy legislative processes, changes in the rating scheme and the test conditions may occur quickly, such that e.g. a vehicle that would have been rated good at the time when it was designed receives a lower rating at the time when it first appears on the market.

Table 2.6 Test conditions applied by the Euro-NCAP (http://www.euroncap.com)

Impact	Test conditions	
Frontal impact	64 km/h, deformable barrier, 40 % overlap plus knee-mapping sled test, where applicable 2 Hybrid III in driver and passenger seats, TNO P1/2 and P3 dummies in CRS in rear seats	
Side impact	50 km/h, Trolley fitted with a deformable front is towed into the driver's side of the car ES-2 in driver seat, TNO P1/2 and P3 dummies in CRS in rear seats	
Pole test (head protection)	29 km/h, car is propelled sideways into a rigid pole ES-2 in driver seat	
Rear impact (whiplash protection)	3 sled tests with driver seat using Low, Medium, and High Severity pulse BioRID in driver seat	
Pedestrian impact	40 km/h or variable impactor speed, various impacts on front structure upper legform, legform, adult and child head impactors	

Note Impact is performed on the driver side, i.e. the illustrations show a right-hand drive vehicle. Further tests related to active safety systems such as electronic stability control (ESC) or speed assistance systems are conducted

2.6.1 Anthropomorphic Test Devices

Standardised tests require the usage of well-defined and validated test objects. An anthropomorphic test device (ATD) is a mechanical model of the human body used as a human surrogate in crash testing. ATDs are in particular designed such that mechanical loading parameters can be measured at impact levels which would be injurious for a living human. To this end, a dummy is made of steel or aluminium (e.g. skeleton), polymers (joint surfaces, skin) and foam (flesh) and is equipped with several accelerometers and load cells to record acceleration, force or deformation. To date various types of ATDs—commonly called crash test dummies—are available whereas each ATD is designed for one specific type of impact only.

In automotive engineering, ATDs are used in the homologation tests required for new vehicles, and in safety device testing to evaluate the occupant protection potential. To a somewhat smaller extent, dummies are also used in the aircraft industry for similar purposes. Historically, the first dummies were developed for the use in aviation, to test parachutes and ejection seats.

Test devices and especially devices embodied in official regulations are expected to fulfil a given set of requirements:

- Anthropometry and biofidelity. An ATD should on the one hand represent a human in terms of size, mass, mass distribution, moments of inertia and (sitting) posture and on the other hand display a human-like biomechanical response to impact. The 50th percentile adult male of which the underlying anthropometric data were established in the 1960s from the US population (standing height: 1.75 m, total weight: 78.2 kg) is the most commonly used dummy in automotive crash testing. Other dummy types include the 5th percentile female (h: 1.51 m, w: 49.1 kg) and the 95th percentile male (h: 1.87 m, w: 101.2 kg). 3, 6 and 10 year old child dummies are furthermore available. The biofidelity is assessed on the basis of cadaver and volunteer studies.
- Instrumentation. The crash test dummy should be sensitive to and allow the measurement of parameters that are related to the injury or the injury mechanism to be examined.
- Repeatability and durability. It should be borne in mind that a dummy must continue to record data for later evaluation even if a critical threshold is exceeded during the test, i.e. it should not or only rarely be damaged.

Repeatability (performing the same test repeatedly with the same dummy) and reproducibility (comparing results obtained under the same test conditions with different dummies) require that an ATD be calibrated regularly. Moreover, practical considerations play an important role in dummy design. Dummies should be robust enough to withstand a high number of tests (even with overload) and they should allow easy handling (up to 102 kg!) and adjustment of the posture.

Currently, over 20 different dummy types are available most of which are, however, not included in regulations. Table 2.7 gives an overview of available ATDs.

Table 2.7 Dummies available and their field of application

Application	Anthropomorphic test devices
Frontal impact	Hybrid III family, THOR
Lateral impact	EuroSID, EuroSID2, SID, SID-HIII, SID IIs, BioSID, WorldSID
Rear-end impact	BioRID, RID2
Pedestrian	POLAR
Children	P0, P3/4, P3, P6, P10, Q-dummies, CRABI
Belt	TNO-10
Impactor	free motion head impactor, head/hip/leg impactor for pedestrian impact

The Hybrid III family of dummies consists of a 3-year-old, 6-year-old, 10-year-old, small adult female (5th percentile), mid-sized adult male (50th percentile) and large adult male (95th percentile). These dummies are designed for use in frontal impact tests. The Hybrid III 50th percentile male dummy (Fig. 2.5) is the most widely used crash test dummy for the evaluation of automotive restraint systems in frontal crash testing. The dummy is defined in the US Federal Motor Vehicle Safety Standards (FMVSS, contained in the US Federal Register) as well as in the European directives. The skull and skull cap of the Hybrid III 50th percentile male dummy are made of cast aluminium parts with removable vinyl skins. The neck is a segmented rubber and aluminium construction with a centre cable. It accurately simulates the human dynamic moment/rotation flexion and extension response in situations involving high neck loading. The rib cage, in turn, is represented by six high-strength steel ribs with polymer based damping material to simulate human chest force–deflection characteristics. Each rib unit comprises left and right anatomical ribs in one continuous part which is open at the sternum and anchored to the back of the thoracic spine. A sternum assembly connects to the front of the ribs and includes a slider for the chest deflection rotary potentiometer. The angle between the neck and upper torso is determined by the construction of the neck bracket, in which a six-axis lower neck transducer can be incorporated. A two-piece aluminium clavicle and clavicle link assemblies have cast integral scapulae to interface with shoulder belts. A curved cylindrical rubber lumbar spine mount provides human-like slouch of a seated person and mounts to the pelvis through an optional three axis lumbar load cell. The pelvis is made of a vinyl skin/urethane foam moulded over an aluminium casting in the seated position. The ball-jointed femur attachments carry bump stops to reproduce the upper leg to hip moment/rotation characteristics. While the femur, tibia and ankle can be instrumented to predict bone fracture, the knee is designed to evaluate tibia to femur ligament injury. The foot and ankle simulates heel compression and ankle range of motion.

A further frontal impact dummy called THOR (Test device for Human Occupant Restraint) (Fig. 2.6) was developed in recent years. This dummy is also based on the anthropometry of the 50th percentile male. Compared to the design of the Hybrid III, all dummy components were improved except the arms, which are

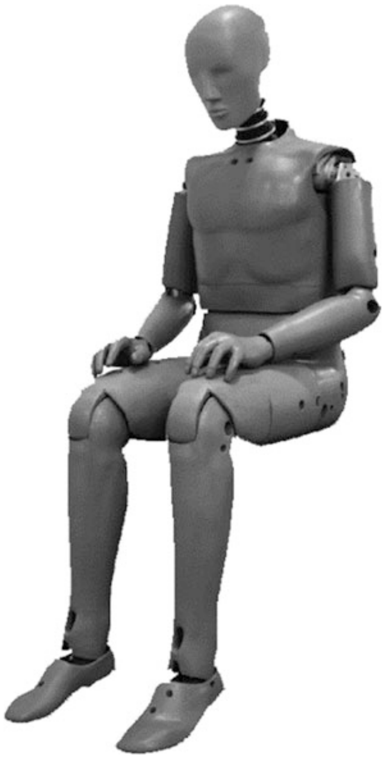


Fig. 2.5 50th percentile male Hybrid III dummy (Humanetics 2013)

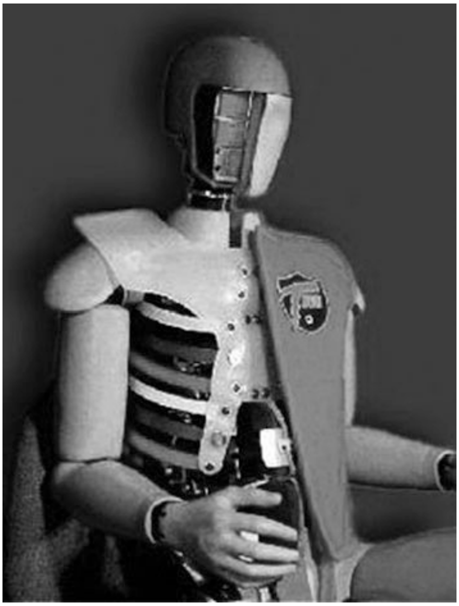


Fig. 2.6 The THOR dummy (Gesac 2013)

identical to those of the Hybrid III. The facial region of the dummy is, for example, instrumented with unidirectional load cells to asses the probability of facial skull fracture. Furthermore, the biofidelity and geometry of the rib cage was enhanced by the use of elliptical ribs and by improving instrumentation such that the dynamic three-dimensional compression of the rib cage can be determined at four distinct points. A new abdominal assembly was developed to allow for the measurement of belt intrusion and compressive displacement at the upper abdomen that might possibly result from an airbag. Changes to the pelvis and the lower limbs increased the sensing capabilities and in addition, the ankle joint was rendered more human-like.

The first side impact dummy (SID) was developed in the late 1970s at the University of Michigan. SID is based on the predecessor of the Hybrid III (the Hybrid II) with an adapted thorax, but without arms and shoulder structures. SID is also sized corresponding to the 50th percentile male and is used in US government-required side-impact testing of new cars (FMVSS 214). The dummy primarily allows a measurement of the injury risk to the head, chest and pelvis. To account for a better head-neck biofidelity, a SID dummy equipped with a Hybrid III head and neck is available (called SID-HIII). It is applied in compliance testing of side-impact head airbags. Additionally, the SID II, i.e. a side-impact dummy representing a 5th percentile female, became commercially available in 2000. The newer version of FMVSS 214 whose phase-in is scheduled to be complete in 2013 specifies the SID IIs on the rear seat, and the Euro-SID IIre (cf. below) on the front seat of the vehicle side struck by the barrier.

European lateral impact regulations (ECE R95) require the use of the Euro-SID1, the European side impact dummy. In Australian and Japanese regulations the Euro-SID1 is likewise prescribed. An updated version, today also accepted for homologation testing, is denoted as ES-2. The original Euro-SID, which was finalised in 1989, represents a 50th percentile adult male. Euro-SID basically consists of a metal and plastic skeleton, covered by flesh-simulating materials. The sitting height is 0.904 m. The total body mass is 72 kg. The dummy which has no lower arms is shown in Fig. 2.7. While the head and the legs are that of the Hybrid III, the thorax was developed to analyse lateral impact. Three separate identical ribs covered with flesh-simulating foam are attached to a rigid steel spine box through a system consisting of a piston/cylinder assembly, springs and a damper. A special shoulder construction allows the arms to rotate realistically and expose the ribs to direct impacts. The pelvis is designed to allow for a measurement of the pubic symphysis force specified in ECE R95. The dummy can be used for side impacts from its left- as well as from its right-hand side.

Further developments in side impact dummies include the Biofidelic Side Impact Test Dummy (BioSID) intended to improve the performance of the current US standard SID series. Although available since 1990, the BioSID was not included in FMVSS 214. BioSID has more sensors and a more biofidelic body than SID/Hybrid III, such that it allows the measurement of the thoracic, abdominal and pelvic injury potential as well as the rib deflection and other

Fig. 2.7 ES-2 dummy (Humanetics 2013)

compression-based injury criteria. By rotating the upper torso by 180 degrees, the dummy can be converted from a left side to a right side impact dummy.

As the automotive industry becomes more global, a harmonised side-impact dummy, denoted as World-SID was developed by a worldwide consortium under the umbrella of the International Standardisation Organisation (ISO). In a comprehensive approach, a mid-sized male side impact dummy for improved assessment of injury risk to car occupants in lateral collisions was developed within the framework of the World-SID programme. Besides an improved biofidelity (e.g. Damm et al. 2006), the World-SID is intended to lead to a worldwide harmonisation in safety regulations and is meant to be incorporated in the Global Technical Regulation initiative (GTR) which was created to this end.

So far only dummies for frontal and lateral impact were presented. This is not surprising, as current occupant safety regulations are restricted to these impact directions. Because the assessment of occupant protection in other than frontal and lateral impacts (rear-end impacts are most prominently absent) was introduced in a relatively late stage, there was no need to develop suitable test devices. However, since injuries sustained in rear-end collisions, especially neck injuries sustained in low-speed rear-end collisions, constitute a major problem in road traffic (see Chap. 4), the need emerged to develop anthropomorphic test devices that allow the investigation of these impact conditions.

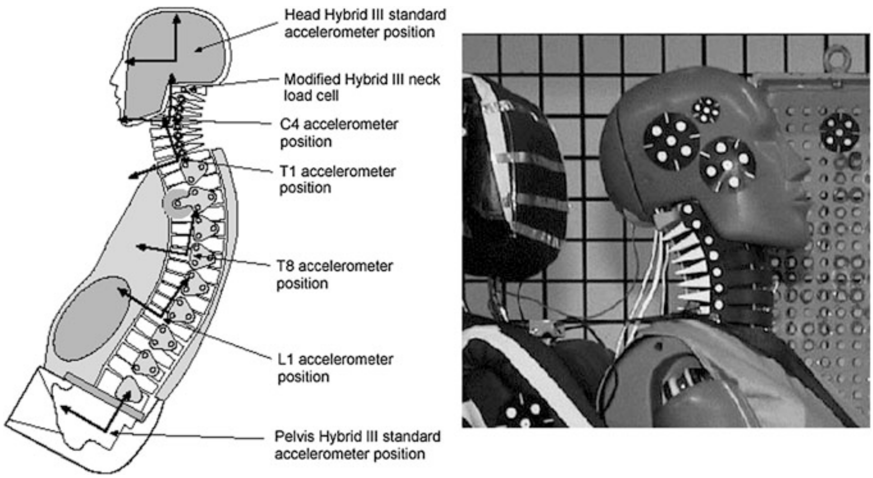


Fig. 2.8 The BioRID makes use of a fully segmented spine (Humanetics 2013)

To date, two different dummies for posterior impact are available, the BioRID and the RID2. Both are mid-sized male dummies have been developed in Europe for assessing the risk of "whiplash" injuries in low-speed rear-end impacts. The main feature of the biofidelic rear-end dummy (BioRID) is its fully segmented spine consisting of 24 segments. Each human spinal pivot point is reproduced. Due to such a detailed representation, a biofidelic spinal movement is observed (Fig. 2.8). The rear impact dummy (RID2), in turn, is based on the THOR frontal impact dummy. However, several modifications were made of which the new design of the neck, which consists of seven aluminium discs interspaced with rubber stops, and of the flexible thoracic and lumbar spine are most relevant in view of an the analysis of the neck injury risk. Both the RID2 and the BioRID were developed and validated for pure rear-end impacts with a movement of the spine exclusively in the anterior-posterior plane. More recently, an improved neck for the RID2, called RID3D, was presented, allowing also oblique rear-end and even low speed frontal impacts to be analysed. Although these dummies offer the possibility for better investigation of the head-neck kinematics, difficulties in handling arise due to the increased flexibility of the spine. The seating procedure, for example, is quite an intricate task compared to a Hybrid III.

In addition to the dummies described above, several specially designed test devices exist. These test devices are generally used for one particular test purpose only.

- The TNO-10 dummy is a loading device for testing vehicle safety belts in a frontal crash situation. The dummy represents a 50th percentile male adult with respect to size and weight distribution. For reasons of simplicity the dummy has no lower arms and only one lower leg assembly combining the two human legs.

- The Child Restraint Air Bag Interaction dummy (CRABI) is used to evaluate air bag exposure to infants restrained in child safety seats that are placed in the front seat. CRABI dummies come in three sizes: six-month-old, 12-month-old and 18-month-old. Further child dummies like the Q-dummies, or infant dummies representing the newborn (P0) and the nine-month-old (P3/4) are available in addition to the child dummies of the Hybrid III family.
- The POLAR dummy (current version: POLAR II) has been designed to simulate more accurately the kinematics of the human body during car-pedestrian collisions. Standing 175 cm tall and weighing 75 kg, the new dummy will help to gather more accurate data on injuries sustained by pedestrians.
- Test devices representing only parts of a dummy are used. The free motion head form (FMH) models a human adult head. Mounted on a propelling device, parts of the vehicle interior may be subjected to a simulated head impact. These tests are required by some safety regulations, e.g. FMVSS 201. Other impactors are used to test the behaviour of a car front with respect to pedestrian safety. These impactors, representing an adult head, a child head, an upper leg and a lower leg are, for example, used in the EC directives and the Euro-NCAP (New Car Assessment Programme) test scheme.
- A 50th percentile torso-shaped body block which is solely used to test the deformation characteristics of the steering assembly, is required for testing in e.g. ECE R12. Parts of ECE R12 have, however, been superseded under certain conditions by ECE R94 and are therefore not required any more in Europe.

Given the fact that a considerable variety of dummies exist which represent different levels of development and which apply in part to the same test conditions, efforts are being made to scale measured values in order to allow for comparisons. To this end, Injury Assessment Reference Values (IRAV) are determined which are dummy specific and can be used for scaling purposes (Mertz et al. 2003).

2.7 Numerical Methods

Thanks to the continuous advancements in computer technology as well as in numerical methods, mathematical modelling has become gradually more detailed and more powerful. Today, computer simulations are an important tool in trauma biomechanics and are applied in all areas of safety engineering such as vehicle crashworthiness design and accident reconstruction; in addition, computer models are successfully used in human body modelling, addressing in particular biomechanical response and possible injury mechanisms.

The most widely used simulation techniques are the multi body system (MBS) approach based on rigid body dynamics (Eqs. 2.1 and 2.2) and the finite element (FE) method, a particular formulation of continuum mechanics (Eqs. 2.3 and 2.4). Multi body systems are sometimes also referred to as lumped mass models in that complex structures such as a human organ or a vehicle are condensed into one or more rigid units connected by mass-less elements like springs and dampers (see

Fig. 2.9 MBS model showing a BioRID dummy seated (adapted from Schmitt et al. 2004)

e.g. the Lobdell thorax model, Chap. 5). Besides, the solidification principle of basic mechanics as well as Saint–Venant's principle of continuum mechanics are always in the background. Multi body systems and FE representations of subunits are furthermore often combined. Likewise, a multi body system can contain flexible subunits, e.g., a cantilever or a plate which can be approximated with models having only few degrees of freedom.

In a multi body system the various elements are connected by kinematic joints. The presence of the kinematic joints restricts the relative motion between adjacent bodies and hence reduces the degrees of freedom of the system. Different types of joints are available, for example translational, revolute and spherical joints, of which each is characterised by a specific number of degrees of freedom. Additional kinematic constraints (e.g. spring/damper elements) can be applied. The rigid bodies themselves are characterised by their inertial properties and by the location of the above mentioned joints only. For the modelling of contact interactions (e.g. head-windscreen impact) and for visualisation purposes, geometrical shapes may be associated with rigid bodies. For the modelling of human body or dummy parts, ellipsoids are often used. Other geometrical primitives include planes and cylinders.

The behaviour of a MBS system is analysed by subjecting the system to external forces such as an acceleration field corresponding to a crash pulse or the forces associated with a fall from a window. This technique has proved its strength especially in whole body response modelling. Approximating the human body by various rigid bodies that are linked by joints and by assigning inertia and mass properties to those bodies, the gross human body kinematic behaviour during impact can be simulated. First models were presented already in the 1970s. To date, a wide range of extensively validated models is available. In particular, dummies are well suited to be modelled as a MBS, because the geometrical and mechanical properties (inertia, mass, joint properties) of the dummy components are clearly defined. Figure 2.9 illustrates an example of a MBS that includes a model of a BioRID.

In the finite element (FE) method, originally derived from Galerkin's theorem, a continuous system is reduced to a discrete numerical model consisting of well-defined elements (e.g. hexahedrons, quadrilaterals, bars). Each element consists of a fixed number of nodes. The degree of freedom of the whole FE model is therefore restricted by the number of nodes. Depending on the boundary conditions applied and the geometry of the mesh, in particular for those elements that share common nodes, the degree of freedom of the whole FE model is given. A detailed description of the finite element method can be found, among other, in Bathe (2007) and Zienkiewicz and Taylor (1994). However, it should be noted that the nature of the problems to be solved in trauma biomechanics (e.g. non-linear material behaviour, large deformations in short time intervals) require specialised approaches to the solution of the models. In general, FE programmes used in this field are based on explicit time integration formulations. These formulations are based on the differential equations of motion of the nodes rather than on the equilibrium of inertial, field and contact forces (implicit formulation). This approach requires less calculation effort and easily lends itself for implementation in parallel computers. On the other hand, more care must be taken to control the numerical stability than in implicit formulations.

The FE method offers the possibility of detailed analysis of the response to impact of both vehicle structure and the human body or ATD, respectively (Fig. 2.10). For example, regarding the response of the head and brain, FE models offer the possibility to investigate the stress distribution in the brain during impact. Such results are important with respect to the understanding of diffuse brain injuries (see Chap. 3), but can hardly be addressed in experiments. There are, however, promising approaches to apply e.g. the external loading conditions on the head as measured on an ATD in a crash test to a FE model of the head, thus allowing insight into complex damage mechanisms in the brain. Other complex biomechanical phenomena, for instance the influence of muscle activity or the interaction of fluid flow and the changing geometry of the surrounding tissue, can be approached by the FE method as well (e.g. Schmitt et al. 2002).

In summary, both the MBS and the FE technique offer their specific advantages and disadvantages in the field of crash simulations. The FE method allows for detailed studies of complex geometries and problems concerned with contact interactions. With respect to crash simulations, the study of local deformations and stress distributions are important advantages of this method. As such, this method can also be used for the analysis of injury mechanisms by modelling a specific part of the human body. However, a detailed representation of a complex geometry leads to an enormous number of elements and therefore a large number of unknowns to be calculated. In case of non-linear constitutive properties of the involved materials as well as large deformations, the enormous computational cost often associated with the FE method represents a major limitation. Parallel processing is suited to alleviate this problem and thus computer systems are able to handle FE models with millions of degrees of freedom today. In contrast, its capability to represent complex kinematic connections efficiently makes the MBS approach particularly attractive. Additionally, computation times required are

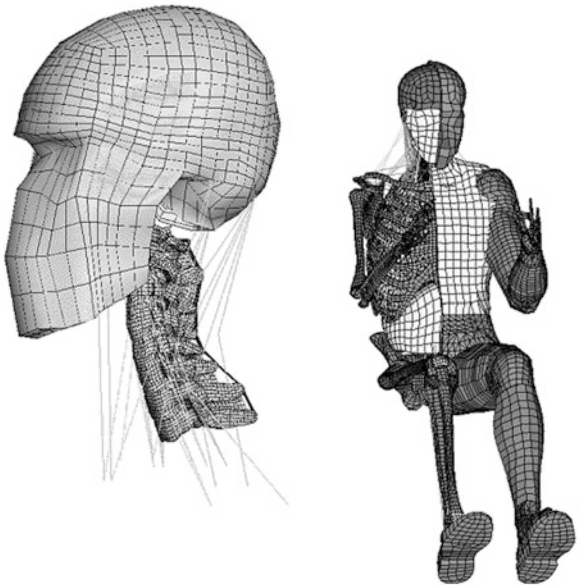


Fig. 2.10 The FE technique used in human body modelling. A detailed model of the head-neck complex on the left (adapted from Schmitt et al. 2002) and a whole body model on the right (adapted from Iwamoto et al. 2002)

generally much shorter than for FE calculations since usually only a comparably small number of ordinary differential equations, though mostly stiff, are to be treated. Hence the MBS methods are widely used as design tools as they are well suited for optimisation studies involving many design parameters.

With respect to human body modelling, general problems arise that both techniques have to cope with. The choice of parameters to describe the material behaviour of the living human tissue requires the availability of experimental data with respect to the deformation characteristics of living tissues. Such data are hardly available, and, if so, often associated with a large uncertainty because of general biological variability on the one hand, and limitations of the particular experimental procedure chosen for the constitutive tests on the other. Furthermore, the validation of human body models, especially those intended for use in several different impact conditions, is crucial but remains a complex task.

To conclude, both methodologies can be reasonably used in the field of general impact and injury analysis. Depending on the purpose, either the best suited technique has to be chosen, or a combination of both methods can be considered. Such an integrated approach can, for instance, be realised in simulations of interactions of a car occupant and a deploying air bag. In this case an FE model is used to model the airbag while the human (or an ATD) is represented by a MBS. Various other studies are presented where a MBS is used to model the gross motion while FE models are included for detailed analysis of single structures, for example, an ice hockey player crashing into a rink board. As of today, numerical models are included at basically all stages of the development process of safety devices.

Fig. 2.11 Virtual testing: the FE model represents a 50 percentile female rear-end impact dummy. Such a dummy is not (yet) available as a physical ATD (adapted from Carlsson et al. 2012)

Despite the widespread and rapidly increasing use of simulation techniques and their potential to reduce the number (and associated cost) of crash tests, numerical simulations are not yet included in the general vehicle safety standards. This can partly be attributed to the fact that general guidelines for simulations and especially for quality control are only emerging today, but would be required if crash simulations were embodied in safety regulations. The complexity of even moderate MBS simulation data sets makes it a difficult task for e.g. an external reviewer to efficiently validate simulation results, whereas the result of a real crash test is, in most cases, obvious. However, the introduction of virtual testing, i.e. the use of computer simulations in addition to physical testing, is subject of intense debate. Performing virtual tests would, for example, allow for a much wider range of test conditions (also with respect to anthropometric sizes of virtual ATDs (Fig. 2.11)) and thus contribute to a broader safety assessment. Gutsche et al. (2013) presented an approach for introducing virtual testing with respect to seat design and soft tissue neck injury.

2.8 Summary

Statistics and databases are tools to map the real-life situation with respect to accidents and injuries. They also allow for the analysis of trends, e.g. related to new vehicle designs or the use of safety gear in sports. The most prominent injury scaling system in trauma biomechanics is the Abbreviated Injury Scale (AIS). Injury risk curves relate the level of a given criterion to the risk of sustaining an injury. Accident reconstruction allows investigating an accident in detail to reconstruct e.g. velocity changes (delta-v) and other parameters characterising the accident. The transfer of the parameters into biomechanical loadings of the persons involved, however, is much more complex.

To determine the biomechanical response, cadaver tests, animal models, or, where justifiable, volunteer tests are used. The data obtained allows investigating the injury risks and serves as important input for the development and validation of ATDs or computer models. Relatively simple multi- (or rigid-) body-systems (MBS) simulations, complex finite element (FE) models, or combinations thereof assume an increasingly important role in the design of e.g. safety devices and car structures.

Full-scale tests, sled tests and impactor tests are common experimental procedures in trauma biomechanics. Full scale tests are expensive, but necessitate fewer assumptions. Sled and impactor tests, on the other hand, allow for parameter variation studies due to their lower cost. For such tests various ATDs are available whereas usually an ATD is designed for a special impact type. The evaluation of standardised test procedures is prescribed in regulations such as ECE or FMVSS or by consumer tests (e.g. EuroNCAP).

2.9 Exercises

E2.1: A driver seat with integrated seat belts and a special device to prevent submarining (sliding of the pelvis underneath the lap belt in a frontal collision, see e.g. Fig. 7.18) has been developed. Plan a test to verify the efficiency of this seat (test method, crash pulse/velocity, dummy type).

E2.2: Describe the various parameters used to describe the “violence” of a collision. Which parameters are important for trauma-biomechanics?

E2.3: A free motion head form impacts a deformable surface, whose force–deformation characteristic is (a) known (b) not known (only the material properties and the geometry of the surface are known). Choose a numerical method to simulate this impact, give reasons for your choice.

P2.1: In Europe and the U.S. different crash test procedures are required for homologation of new cars. Also, consumer tests employ different crash test procedures. Discuss the effects of these disparities on the car maker and on the consumer.

P2.2: ATDs that are more biofidelic and offer more measurement possibilities than e.g. the Hybrid III or the Euro-SID have been available for quite some time now. Why are they neither specified in the relevant regulations nor used in consumer crash tests?

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3 Head Injuries

Head injury sustained in accidents continues to be a leading cause of death and disability even though considerable advancement in the understanding of head injury mechanisms and the introduction of different measures to prevent such injury has resulted in the reduction of the number and severity of head injuries.

A brief review of the head anatomy is followed by the description and classification of possible head injuries and their underlying injury mechanisms. Further, the biomechanical response of the head as investigated in various experimental studies and injury criteria that were derived thereof to quantify the impact response of the head in crash testing are discussed.

Aspects of head injuries in sports are considered in an own section. Finally, principles of head injury protection are presented. Seat belts and airbags may prevent head injury by avoiding a head impact altogether, while helmets and e.g. deformable structures inside and outside a car (pedestrian head impact) mitigate the consequences of an impact by distributing the impact force over a larger area, and by absorbing impact energy.

3.1 Anatomy of the Head

The human head (cranium) can be regarded as a multi-layered structure with the scalp being the outermost layer followed by the skull, the meninges and eventually the central nervous system that represents the innermost tissue.

The scalp is about 5–7 mm thick and consists of the hair-bearing skin, a subcutaneous connective tissue layer, and a muscle and fascial layer. Upon application of a traction force to skin of the head, these layers move together as one.

Below the scalp there is a loose connective tissue and the periosteum (i.e. a fibrous membrane) that covers the bony skull.

The adult skull is a complex structure consisting of several bones fused together and associated suture lines (Fig. 3.1). The only facial bone connected to the skull through freely moveable joints is the mandible. Thickness and curvature of the bones can vary substantially.

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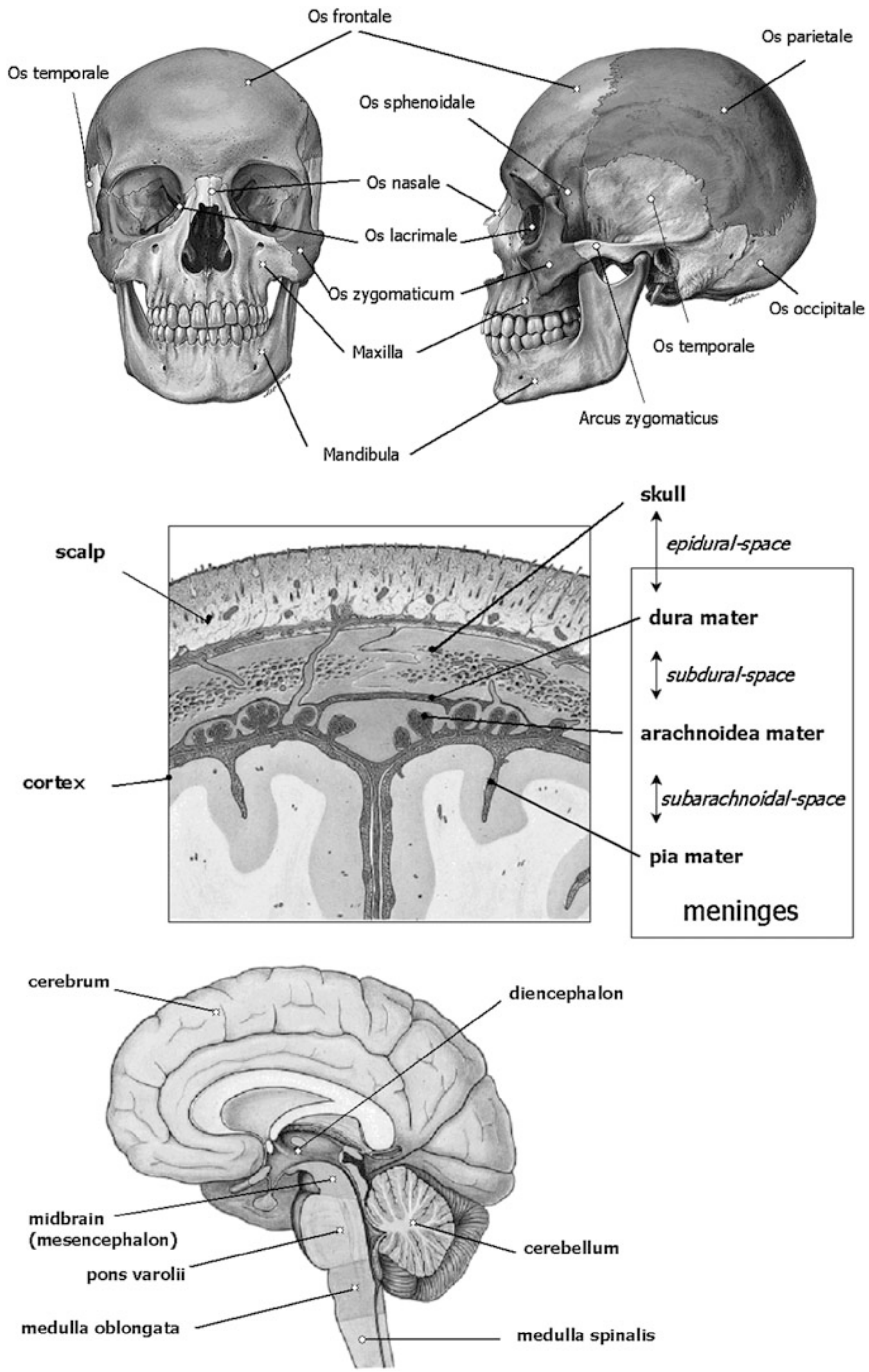


Fig. 3.1 Anatomy of the head: bony structures of the skull (*top*), the meninges (*middle*), and the brain (*bottom*) (adapted from Sobotta 1997)

The inner surface of the cranial vault is concave with an irregular plate of bone forming the base. This base plate contains several small holes for arteries, veins and nerves as well as a large hole (foramen magnum) through which the brainstem passes into the spinal cord.

Three membranes called the meninges protect and support the spinal cord and the brain and separate them from the surrounding bones (Fig. 3.1). From outside to inside, we find the dura mater, the arachnoidea mater, and the pia mater. The dura mater is a tough, fibrous membrane, while the arachnoidea mater resembles a spider-web. Both membranes are separated by a narrow space, the subdural space. Analogously, the subarachnoidal space separates the arachnoidea mater and the pia mater. The pia mater covers the surface of the brain, dipping well into its fissures. Cerebrospinal fluid (CSF) fills the subarachnoidal space and the ventricles of the brain and thus cushions the brain (and the spinal cord) from mechanical shock. As CSF constantly circulates and surrounds the brain on all sides, it serves as a buffer and helps to support the brain's weight.

Several blood vessels cross the meninges supplying the brain and the scalp. The so-called bridging veins, i.e. the veins that bridge the subdural space, are of particular interest as they may be subject to injury through tearing (see Sect. 3.2.).

Finally, the central nervous system consisting of the brain and the spinal cord is located at the centre of the head. Structurally and functionally the brain can be divided into five parts: cerebrum, cerebellum, midbrain, pons and medulla oblongata (Fig. 3.1).

3.2 Injuries and Injury Mechanisms

The most important injuries to the head are those to the skull and the brain including the meninges. Figure 3.2 gives an schematic overview on possible head injuries. In principle, head injuries are characterised as open or closed depending on whether the dura mater is injured (open) or not (closed). Soft tissue injuries to the scalp and face commonly occur in automotive accidents. The resulting injuries include contusion and laceration but are generally regarded to be of minor importance. Likewise facial injuries, to the eyes or ears for example are considered minor injuries and therefore are mainly rated as AIS1 or AIS2. These injuries will not be discussed here.

More severe head injuries can arise from fractures. Facial fractures include fracture of the nasal bone, which occurs most frequently, and maxillary fractures. The latter are considered serious with AIS grades of up to 3. Figure 3.3 shows the LeFort classification that is used to categorise maxillary fractures. Examples of head injuries classified according to the AIS scale are presented in Table 3.1.

With respect to the skull, fractures are divided into basilar and vault fractures (i.e. all other fractures not occurring at the basis of the skull). It should be noted that basilar fractures may be difficult to visualise using conventional radiographic methods, so that diagnosis can be difficult.

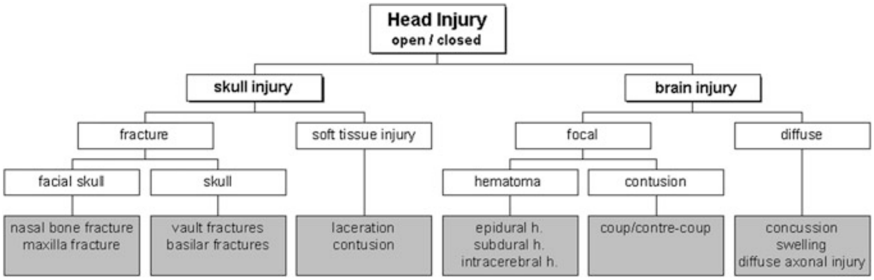


Fig. 3.2 Possible injuries to the head

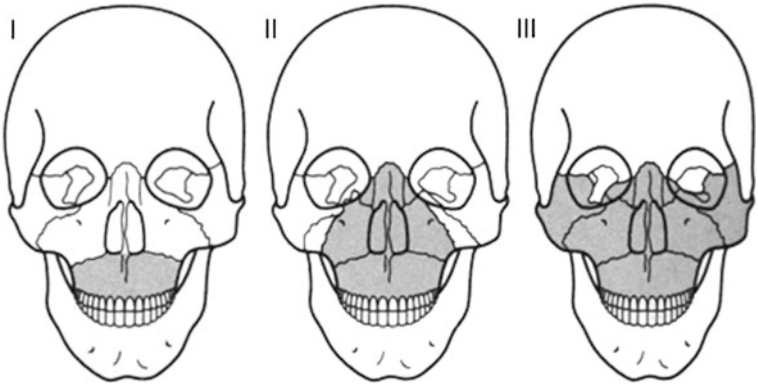


Fig. 3.3 Three types of facial fractures as classified by LeFort (adapted from Vetter 2000)

Table 3.1 AIS classified head injury (AAAM 2005)

AIS code	Description
1	Skin/scalp: abrasion, superficial laceration Face: nose fracture
2	Skin: major avulsion Vault fracture: simple, undisplaced Mandible fracture: open, displaced Maxilla fracture: LeFort I and II
3	Basilar fracture Maxilla fracture: LeFort III Total scalp loss Single contusion cerebellum
4	Vault fracture: complex, open with torn, exposed or loss of brain tissue Small epidural or subdural hematoma
5	Major penetrating injury (>2 cm) Brain stem compression Large epidural or subdural hematoma Diffuse axonal injury (DAI)
6	Massive destruction of both cranium and brain (crush injury)

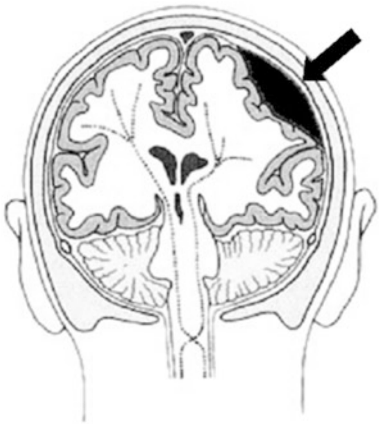
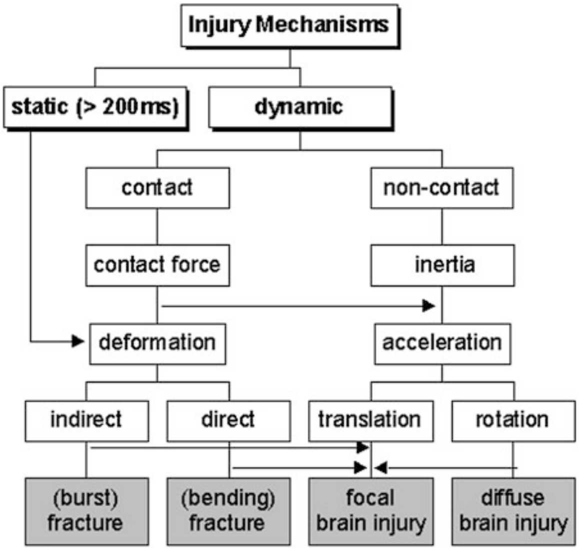


Fig. 3.4 Bleeding into the epidural space is called an epidural hematoma and can cause brain contusion (adapted from Vetter 2000)

Injuries to the brain are clinically classified into two broad categories: diffuse injuries and focal injuries. Diffuse brain injuries form a spectrum ranging from mild concussion to diffuse white matter injuries. The most common form of such brain injury is mild concussion (fully reversible, no loss of consciousness). Particularly in sports, mild traumatic brain injury (mTBI) is often diagnosed (see Sect. 3.5). A more severe form of concussion is cerebral concussion which is characterised by immediate loss of consciousness. The outcome of patients suffering from cerebral concussion strongly depends on whether there are associated brain injuries or not (Melvin and Lighthall 2002). Diffuse axonal injury (DAI) describes disruption to the axons in the cerebral hemispheres and the subcortical white matter.

Focal brain injuries are lesions where the damage is locally well-defined. Possible focal injuries are hematoma and contusions. Contusion is the most frequently found lesion following head impact. Generally, contusion occurs at the site of impact (coup contusion) and at site opposite the impact (contre-coup contusion). Contre-coup contusions are considered more significant than coup-contusions (Melvin and Lighthall 2002). As for hematoma, three different types are distinguished depending on the site of the bleeding: epidural hematoma, subdural hematoma and intracerebral hematoma (Fig. 3.4). Epidural hematoma, i.e. bleeding above the dura mater, is observed as a result of trauma to the skull and the underlying meningeal vessels. It is therefore not due to brain injury. Usually skull fracture is associated, but an epidural hematoma may also occur in the absence of fracture. If the hematoma is found below the dura mater, it is called a subdural hematoma. Three sources were identified for subdural hematoma: lacerations of cortical veins and arteries by penetrating wounds, large-contusion bleeding into the subdural space, and tearing of bridging veins between the brain's surface and the dural sinuses. The mortality rate of this type of hematoma exceeds 30 % in most studies (Melvin and Lighthall 2002). Intracerebral hematomae are well-defined homogeneous collections of blood within the brain and can be

Fig. 3.5 Possible mechanisms of head injury



distinguished from contusions by e.g. computer tomography. An increased intra-cranial pressure due to swelling of the brain may be the consequence of diffuse as well as focal injuries. This secondary effect may lead to a reduced cerebral blood flow and a reduced oxygen supply, whose consequences may well exceed those of the primary injury itself.

The mechanisms causing head injuries are manifold. In principle, injuries can result from static and dynamic loading (Fig. 3.5). For our purpose, static loading is defined as a load lasting for more than 200 ms. Under such static loading the head deforms until it reaches a maximum deformation. Then the skull fractures, often leading to multiple fractures. In accidents, however, this type of loading is rare. Dynamic loading is the predominately loading scenario. Two types, contact and non-contact loading, are distinguished, each resulting in a different head response. Direct contact of the head to (or from) an object can cause the skull to deform, possibly resulting in direct fractures (mostly due to bending and often close to the impact location) or in indirect fractures (burst fractures oriented in the direction of the force vector). Furthermore, after deformation of the head local brain injury (even without fractures) like epidural hematoma or contusion as well as scalp injuries are observed. Additionally, rapid contact loading produces stress waves that propagate in the skull or the brain (Fig. 3.6). Wave propagation in the brain may lead to a pressure gradient with positive pressure at the site of impact (coup) and negative pressure on the opposite side of the impact (contre-coup). Such a mechanism is proposed for the generation of intracranial compression which causes focal injuries of the brain tissue and bruising. However, it is not yet fully understood whether the injury is due to negative pressure (tensile loading causing e.g. bleeding or tissue disruption) or due to a cavitation phenomenon (Viano

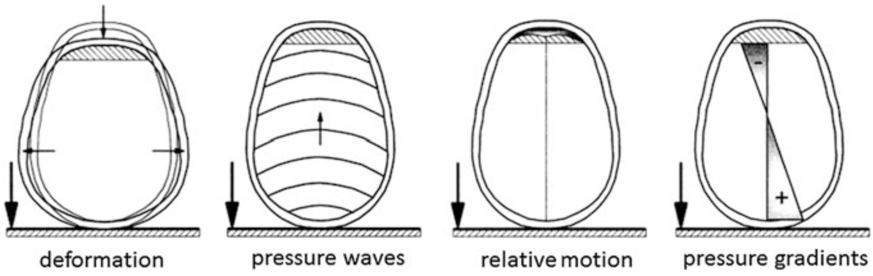


Fig. 3.6 Different injury mechanisms for contact impact; fractures do not necessarily occur (adapted from Vetter 2000)

2001). In addition, the pressure gradient can give rise to shear strain within the deep structures of the brain.

Contact loading may also result in a relative motion of the brain surface with respect to the inner surface of the skull base. Surface contusions on the brain (so called gliding contusions) and tearing of the bridging veins (causing subdural hematoma) can be the consequences.

In non-contact situations, the head is loaded exclusively due to inertial forces, i.e. acceleration (or deceleration) of the head. Acceleration can either be translational or rotational. Translational acceleration generally results in focal brain injury while rotational acceleration also causes diffuse brain injury. As an exception, subdural hematoma, i.e. a focal skull injury, may arise due to acceleration induced relative motion between brain and skull tearing the bridging veins. Furthermore, acceleration response of the head does, of course, also occur in contact loading. Thus the mechanisms described above apply in the same way.

It should be noted that headaches, which are often erroneously thought to be due to a "head injury", caused either by contact or non-contact mechanisms, may also be initiated by lesions in the upper area of the cervical spine. Therefore, a comprehensive analysis of the occupant dynamics must be undertaken in order to prevent premature diagnoses of cerebral concussion or mild traumatic brain injury (mTBI).

3.3 Mechanical Response of the Head

Many cadaver studies on head impact have been carried out to investigate the mechanical response properties of the head. In general, the impact responses were described in terms of head acceleration and impact force and therefore depend on the inertial properties of the head and impacted surface. For a 50th percentile male, the average head mass is 4.54 kg and the average mass moments of inertia are $I_{xx} = 22.0 \times 10^{-3}$ kgm^2, $I_{yy} = 24.2 \times 10^{-3}$ kgm^2, and $I_{zz} = 15.9 \times 10^{-3}$ kgm^2 (e.g. Beier et al. 1980). For the paediatric head only few data is available (see e.g. Prange et al. 2004).

Table 3.2 Peak force for fracture at different regions of the skull

Impact area	Force (kN)	References
Frontal	4.2	Nahum et al. (1968)
	5.5	Hodgson and Thomas (1971)
	4.0	Schneider and Nahum (1972)
	6.2	Advani et al. (1975)
	4.7	Allsop et al. (1988)
Lateral	3.6	Nahum et al. (1968)
	2.0	Schneider and Nahum (1972)
	5.2	Allsop et al. (1991)
Occipital	12.5	Advani et al. (1982)

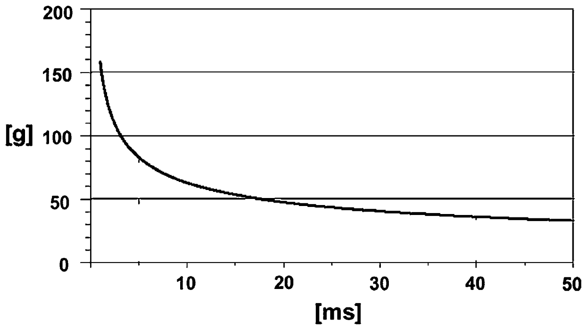


Fig. 3.7 The Wayne State Tolerance Curve (acceleration vs. duration of acceleration pulse) (adapted from Krabbel 1997)

In these cadaveric studies, mainly drop tests against a rigid flat surface were performed. Table 3.2 summarises the peak force values reported for fracture at different sites of the head. Furthermore the acceleration response of the head was investigated. When measuring the acceleration of the head, two problems arise: firstly, accelerometers cannot be mounted at the centre of gravity of the head and secondly, the head is not a rigid body. Therefore several methods for measuring the acceleration have been proposed (e.g. Padgaonka et al. 1975). It is also recommended to measure the head rotational acceleration so that the acceleration of the head's centre of gravity can be computed thereof. Nonetheless there remain some uncertainties as the exact stiffness distribution of the skull is generally not known.

As a result of extensive cadaver tests focusing on head acceleration, the Wayne State University Cerebral Concussion Tolerance Curve, abbreviated as the *Wayne State Tolerance Curve (WSTC)*, was established (Gurdjian et al. 1953; Lissner et al. 1960; Gurdjian et al. 1966). The WSTC indicates a relationship between the duration and the average antero-posterior translational acceleration level of the pulse that accounts for similar head injury severity in head contact impact (Fig. 3.7). Clinically observed prevalence of concomitant concussion in skull

Table 3.3 Test conditions of the experiments the WSTC is based upon

Pulse duration	Test objects	Test set-up	Response measured	Injury criterion
2–6 ms	Cadavers	Drop test	Acceleration at the back of the head	Skull fracture
6–20 ms	Cadavers and animals	Impact test	Acceleration of skull, brain pressure	Pathological changes
>20 ms	Volunteers	Sled tests	Whole body acceleration without head impact	Concussion, state of consciousness

fracture cases was used to relate cadaver impacts to brain injury. In fact, 80 % of all concussion cases also had linear skull fractures (Melvin and Lighthall 2002). Gurdjian and colleagues assumed that by measuring the tolerance of the skull to fracture loads, one is effectively inferring the tolerance to brain injury.

Combinations of acceleration level and pulse duration that lie above the curve are thought to exceed the human tolerance, i.e. they cause severe, irreversible brain injury. Combinations below the curve do not exceed human tolerance, but may result in reversible injury. As the original WSTC covers a time duration range of 6 ms only, the curve was extended for durations longer than 6 ms using animal and volunteer data. Figure 3.7 presents the modified curve; the test conditions used to obtain the data are given in Table 3.3. As can be seen, the head can withstand higher acceleration for shorter durations.

The WSTC is supported by experiments conducted in Japan which led to the Japan Head Tolerance Curve (JHTC) (Ono et al. 1980). JHTC was mainly obtained from experiments with primates and scaling of results to humans. Differences between the WSTC and JHTC are negligible for time intervals up to 10 ms, and only minor differences exist for longer durations.

When the WSTC is plotted in a logarithmic scale, it becomes a straight line with a slope of −2.5. Based on this finding, Gadd (1961) proposed a first head injury criterion, the severity index (SI). A modified form of this criterion is still in use today (see Sect. 3.4.1).

Using the WSTC or any criterion developed thereof, restrictions that arise from the test conditions have to be considered. The paucity of data points, the position of the accelerometer (back of the head), the fact that rotational acceleration is not considered, and the techniques used to scale the animal data are, for instance, major limitations. However, from a biomechanical point of view the main criticism concerns the correspondence of skull fracture and brain injury that was assumed. This hypothesis remains to be verified, as there was no direct demonstration of functional brain damage in an experiment in which biomechanical parameters sufficient to determine a failure mechanism in the tissue were measured (Melvin and Lighthall 2002).

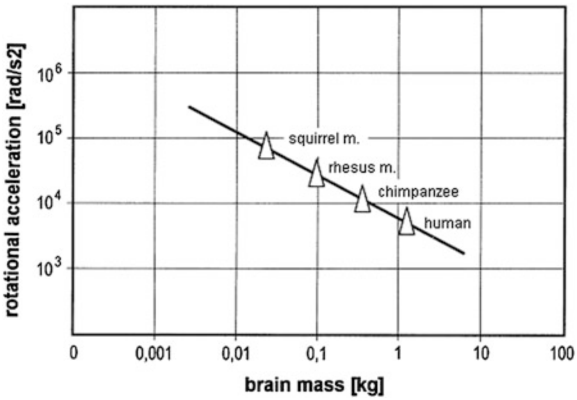


Fig. 3.8 Results from experiments and scaling addressing tolerance towards rotational acceleration (adapted from Krabbel 1997)

Bearing in mind that the WSTC is based on direct frontal impact tests, the results can, strictly speaking, not be applied to non-contact loading conditions and to other impact directions, respectively. Nonetheless, WSTC is still the most important data source with respect to the linear acceleration response of the head.

Further experimental studies addressed rotational acceleration which may cause diffuse brain injury and subdural hematoma. Besides volunteers and cadavers, primates were subjected to head rotation, where the rotational acceleration was measured and the resulting degree of injury was assessed (e.g. Ommaya et al. 1967; Hirsch et al. 1968; Gennarelli et al. 1972). It was found that the angular acceleration and the according injury thresholds are related to the mass of the brain. Thus, the tolerance limit for the human was obtained by scaling the results from the primate tests (Fig. 3.8). Table 3.4 gives tolerance values that are commonly used. However, additional studies on volunteers suggest that much higher tolerance values up to 25000 rad/s^2 may be possible for a short duration (Tarriere 1987).

In this section several experimental studies were presented that aimed at predicting head injury from one specific input parameter, i.e. the translational or the rotational acceleration, respectively. A more extensive review of these studies is given by Goldsmith and Monson (2005). However, in the vast majority of head impact situations it can be expected that both translational and rotational acceleration are present and combine to cause brain injury. Accordingly, comprehensive brain injury prediction requires taking into account the various responses of the brain tissue for any combination of mechanical loading. The development of sophisticated mathematical models of the head using, for example, the finite element method addresses this task and aims at determining measures for prediction of the head's mechanical response to impact. When combined with results of detailed investigation of the response of the living human, such models promise to contribute substantially to today's understanding of head injury mechanisms and the impact tolerance of the head.

Table 3.4 Tolerance thresholds for rotational acceleration and velocity of the brain

Tolerance threshold	Type of brain injury	References
50 % probability: $\ddot{\alpha} = 1800$ rad/s^2 for $t > 20$ ms $\dot{\alpha} = 30$ rad/s for $t \leq 20$ ms	Cerebral concussion	Ommaya et al. (1967)
$\ddot{\alpha} = 4500$ rad/s^2 and/or $\dot{\alpha} = 70$ rad/s	Rupture of bridging vein	Löwenhielm (1975)
2000 rad/s^2 $< \ddot{\alpha} <$ 3000 rad/s^2	Brain surface shearing	Advani et al. (1982)
$\dot{\alpha} < 30$ rad/s: AIS 5: $\ddot{\alpha} = 4500$ rad/s^2 $\dot{\alpha} > 30$ rad/s: AIS 2: $\ddot{\alpha} = 1700$ rad/s^2 AIS 3: $\ddot{\alpha} = 3000$ rad/s^2 AIS 4: $\ddot{\alpha} = 3900$ rad/s^2 AIS 5: $\ddot{\alpha} = 4500$ rad/s^2	(General)	Ommaya (1984)

3.4 Injury Criteria for Head Injuries

Although great progress in passive safety, such as the introduction of advanced restraint systems, was made in the last couple of years to reduce the number and severity of head injuries, there is only one injury criterion in wide use, the Head Injury Criterion (HIC). Besides the HIC and its European equivalent, the Head Protection Criterion (HPC), the “3 ms criterion” and the Generalised Acceleration Model for Brain Injury Threshold (GAMBIT) are presented. However, it should be noted that all these criteria are based on acceleration response only. Consequently, injuries that are related to impact force rather than acceleration are not addressed by these criteria. In other words, those criteria do not allow an evaluation of the injury risk of sustaining fractures of the bony structures of the head. The only dummy capable of measuring a force response to facial impact is the THOR dummy (see Sect. 2.6.1), but this dummy is not included in recent crash test standards. Further attempts on improving head injury criteria include a criterion based on the total change of kinetic energy of the head during impact (HIP, Newman et al. 2000), or employ e.g. finite element models to predict shear strain in the brain tissue, thus bypassing the discussion on whether rotational or translational acceleration is more important (Willinger and Baumgartner 2001; Takhounts et al. 2003, 2008; Patton et al. 2012). Such criteria, however, require extensive calculation and modelling steps after e.g. a crash test.

3.4.1 Head Injury Criterion

The Head Injury Criterion has a historical basis in the work of Gadd (1961), who used the Wayne State Tolerance Curve (WSTC) (see Sect. 3.3) to develop the so-called severity index (SI). In 1971, Versace (1971) proposed a version of the HIC

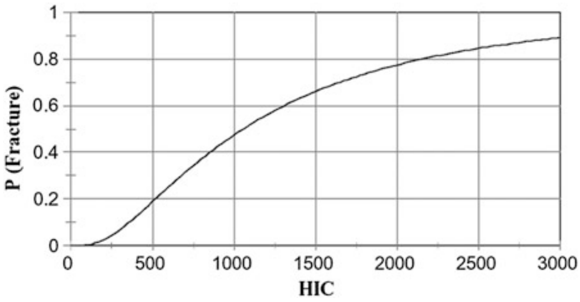


Fig. 3.9 Probability of skull fracture (AIS $\geq$ 2) in relation to the HIC as determined by Hertz (1993)

as a measure of average acceleration that correlates with the WSTC. The actual version of HIC was then proposed by the US National Highway Traffic Safety Administration (NHTSA) and is included in FMVSS 208. HIC is computed based on the following expression:

$$HIC = \max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)dt\right]^{2.5} (t_2 - t_1) \tag{3.1}$$

where t_2 and t_1 are any two arbitrary time points during the acceleration pulse. Acceleration is measured in multiples of the acceleration of gravity (g) and time is measured in seconds. The resultant acceleration is used for the calculation. FMVSS 208 requires t_2 and t_1 not to be more than 36 ms apart (thus called HIC_{36}) and the maximum HIC_{36} not to exceed a value of 1000 for the 50th percentile male. In 1998 NHTSA also introduced the HIC_{15}, i.e. the HIC evaluated over a maximum time interval of 15 ms (Kleinberger et al. 1998). For the according threshold value, a maximum of 700 was suggested for the 50th percentile male.

To determine the relationship between HIC and injuries of the skull and brain, available test data was analysed statistically by fitting normal, log normal, and two-parameter Weibull cumulative distributions to the data set, using the Maximum Likelihood method to achieve the best fit for each function (Hertz 1993). The best fit of the data was achieved with the log normal curve (Fig. 3.9).

The probability of skull fracture (AIS $\geq$ 2) is given by the formula

$$p(fracture) = N\left(\frac{\ln(HIC) - \mu}{\sigma}\right) \tag{3.2}$$

where N() is the cumulative normal distribution, $\mu = 6.96352$ and $\sigma = 0.84664$.

Since the data used to establish this risk analysis consists of short duration impacts of typically less than 12 ms, the HIC curve is applicable to both HIC_{15} and HIC_{36}. Thus, the probability of skull fracture (AIS $\geq$ 2) associated with a HIC_{15} threshold value of 700 for a mid-sized male is 31 % and for a limit of 1000 for HIC_{36} (50th percentile male) it is approximately 48 %.

Basically the limitations as described for the WSTC itself apply (see Sect. 3.3). Not taking into account rotational acceleration is often criticised. A further drawback is the lack of a functional relationship between human head injury and the acceleration response measured in an anthropomorphic test device. Despite these limitations, HIC is still the most commonly used criterion for head injury in automotive research.

3.4.2 Head Protection Criterion

The determination of the Head Performance Criterion HPC is required in regulations ECE R94 and R95. Hence, the HPC is used to quantify head impact in both frontal and lateral impact. The definition and the calculation procedure to obtain the HPC are identical to the HIC_{36}. Thus, the corresponding maximum time interval is 36 ms; the according threshold level for frontal and lateral direction is 1000.

If no head contact occurs, the HPC is fulfilled regardless of the acceleration level reached. If the beginning of the head contact can be determined satisfactorily, t_1 and t_2 (cf. Eq. 3.1) are the two time points which define a period between the beginning of the head contact and the end of the recording, at which the HPC is at its maximum.

3.4.3 3 ms Criterion (a_{3ms})

The "3 ms criterion" is also based on the WSTC. It is defined as the acceleration level exceeded for a duration of 3 ms and should not exceed 80 g (Got et al. 1978). This criterion is also incorporated in ECE R21 and R25, the regulations dealing with impact of the occupant to interior structures of a vehicle and the impact to the head restraints, respectively. The analogous US regulation, FMVSS 201, as well as the frontal impact regulation FMVSS 208 also require fulfilment of this criterion.

Furthermore, a modification of the a_{3ms} criterion is used in helmet testing. Choosing a duration of 5 ms, the acceleration level shall be less than or equal to 150 g. ECE R22 describes the details of this so-called a_{5ms} criterion.

3.4.4 Generalized Acceleration Model for Brain Injury Threshold

In an attempt to combine translational and rotational acceleration, Newman (1986) proposed the *Generalized Acceleration Model for Brain Injury Threshold*, abbreviated GAMBIT. Assuming that a combined load case of translational and rotational acceleration can cause head injury, the following relationship was proposed:

$$GAMBIT = \left[\left(\frac{a(t)}{a_c} \right)^n + \left(\frac{\ddot{\varphi}(t)}{\ddot{\varphi}_c} \right)^m \right]^{\frac{1}{k}} \tag{3.3}$$

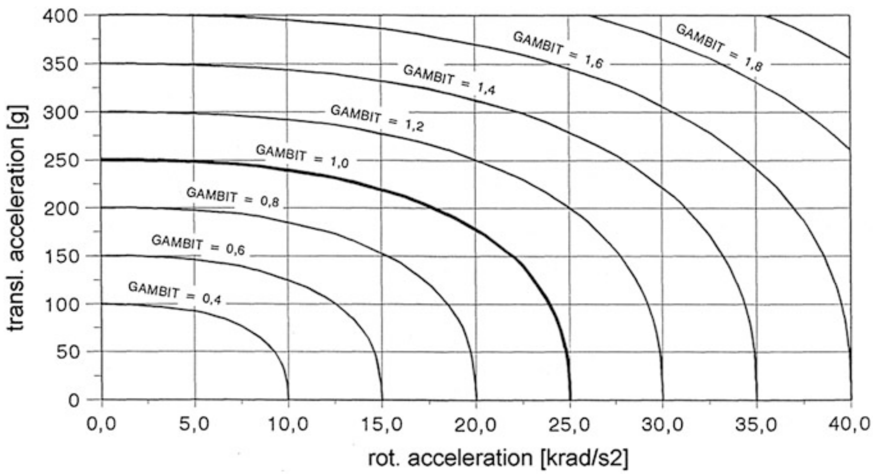


Fig. 3.10 GAMBIT curves for constant GAMBIT values (adapted from Kramer 1998/2006)

Here *a(t)* and $\ddot{\varphi}(t)$ denote the translational and rotational acceleration, respectively. a_c and $\ddot{\varphi}_c$ represent critical tolerance levels for those accelerations and n, m and k are constants. Fitting the constants by means of statistical analysis and computer simulations to available data, Kramer (Kramer 1998/2006) presents a solution which reads

$$GAMBIT = \left[\left(\frac{a(t)}{250}\right)^{2.5} + \left(\frac{\ddot{\varphi}(t)}{25}\right)^{2.5}\right]^{\frac{1}{2.5}} \tag{3.4}$$

with *a(t)* and $\ddot{\varphi}(t)$ given in [g] and [krad/s^2]. Figure 3.10 shows curves of constant GAMBIT obtained by using Eq. 3.1. The curve for a GAMBIT of 1.0 was determined to represent a probability of 50 % for irreversible head injury. Non-contact head impact accounted for GAMBIT values below 0.62.

Assuming that translational and rotational accelerations contribute equally to the probability of head injury and assuming that the tolerances derived in experiments with either translational or rotational acceleration are also valid in a combined loading scenario, Eq. 3.3 was simplified to

$$GAMBIT = \frac{a_m}{250} + \frac{\ddot{\varphi}_m}{10} \tag{3.5}$$

with a_m [g] and $\ddot{\varphi}_m$ [krad/s^2] being the mean translation and mean rotational acceleration, respectively, and considering 250 g the maximum tolerable translational acceleration and taking 10 krad/s^2 for the limit for rotational acceleration (Newman 1986). Thus, GAMBIT of 1.0 represents the overall tolerance value.

To date the GAMBIT still lacks validation and is therefore hardly ever used, nor is it included in any regulations so far.

3.5 Head Injuries in Sports

The incidence rate for head injuries depends strongly on the type of sports. Upon team sports at the Olympic Games 2004, Junge et al. (2006) found that 24 % of all injuries reported were head injuries. Mild concussion was sustained most often (11 %) followed by lacerations (4 %), fractures (2 %) and contusions (2 %). Handball accounted for 42 % of the head injuries, soccer for 20 %, basketball and hockey for 13 %. Also in other sports head injuries account for a significant portion of the injuries sustained: skiing/snowboarding 15–30 %, ice hockey 4–18 %, baseball 11 % head only (28 % facial injuries), equestrian sports 19–48 %, boxing 16 % for concussion (Hunter 1999; Yen and Metzl 2000; Levy et al. 2002; Zarzyn et al. 2003; McIntosh and McCrory 2005; McIntosh et al. 2011). Direct impact to the head (or helmet), e.g. due to collisions, contact with opponents or falls, is the predominant cause for athletic head trauma.

Although concussions reported in sports are often classified as minor or mild, they are, particularly in professional sports, a major concern. Possible long-term consequences on brain function and degeneration of brain tissue after repeatedly sustained concussion have raised the profile of the disorder (e.g. Bailes and Cantu 2001; Biasca et al. 2006a, b; Delaney et al. 2006; Meaney and Smith 2011). Thus it must be ensured that an athlete returning to play after a (mild) concussion is fully recovered (see also Chap. 9 for injury due to chronic mechanical exposure).

Mild traumatic brain injury (mTBI) is defined as a complex patho-physiologic process induced by mechanical loading of the brain. Typically mTBI is associated with a range of clinical symptoms that are common with those observed in mild diffuse cerebral injury (see Sect. 3.2) and can include temporary impairment of neurological functions. Although patients usually recover after a few days, mTBI requires treatment and monitoring. This holds particularly true since repeated mTBI is believed to result in chronic degenerative brain damage. Therefore it is recommended to document every mTBI.

Several studies are presented that investigate head loading and address possible injury criteria and thresholds for concussion and mTBI, respectively. The techniques used to investigate the loading of the head range include video analysis, reconstruction of head impacts using crash test dummies, measuring loads by instrumented helmets and computer simulations. Table 3.5 summarizes results of several studies.

As can be seen the results vary significantly illustrating the difficulty to define reasonable injury threshold values. A tolerance limit of 200 g for translational acceleration and 4500 rad/s^2 for rotational acceleration was proposed by Ommaya (1984, see also Sect. 3.3). Analysing impacts observed in the US National Football League NFL, Pellman et al. (2003) suggested the use of a concussion threshold of a HIC of 250. King et al. (2003) estimated a 50 % risk of mTBI for a HIC of 235, a linear acceleration of 79 g and an angular acceleration resultant of 5757 rad/s^2 and a 75 % risk for a HIC of 333, a linear acceleration of 98 g and an angular acceleration resultant of 7130 rad/s^2. Rowson et al. (2012) evaluated a large

Table 3.5 Head impact data related to the risk of sustaining concussion/mTBI

Sport	Translational acceleration (g)	Rotational acceleration (rad/s^2)	HIC (-)	Delta-v (m/s)	Comment/references
Football (professional)	60 $\pm$ 23	4235 $\pm$ 1716	121 $\pm$ 64	5.0 $\pm$ 1.1	No injury, data from video analysis/reconstruction using Hybrid III dummies Pellman et al. (2003)
Football (professional)	<85	<6000	240 (HIC15)	-	Threshold values for reversible brain injury, computer simulation Zhang et al. (2004)
Football (professional)	98 $\pm$ 28	6432 $\pm$ 1813	381 $\pm$ 197	7.2 $\pm$ 1.8	Concussion, data from video analysis/reconstruction using Hybrid III dummies Pellman et al. (2003) and Viano et al. (2007)
Football (college)	81	-	200	-	Concussion (one observation), in-helmet accelerometers Duma et al. (2005)
Football (college)	32 $\pm$ 25	2020 $\pm$ 2042 y-axis	26 $\pm$ 64	-	Average of 3311 observations, in-helmet accelerometers Duma et al. (2005)
Football (college)	21-23	-	-	-	No concussion, accelerometers embedded in helmets Mihalik et al. (2007)
Football (college)	60.5−168.7	-	-	-	Concussion, accelerometers embedded in helmets Guskiewicz et al. (2007)
Football (college)	-	1230 (subconcussive) 5022 (concussive)		-	Over 300'000 observations, instrumented helmet injury risk curves established Rowson et al. (2012)
Football Hockey Soccer	29.2 $\pm$ 1.0 35.0 $\pm$ 1.7 54.7 $\pm$ 4.1	-	22.5 $\pm$ 3.6 13.5 $\pm$ 1.8 48.5 $\pm$ 7.0	-	No incidents of concussion, accelerometer placed on helmet Naunheim et al. (2000)
Boxing	58 $\pm$ 13	6343 $\pm$ 1789	71 $\pm$ 49	-	Punches to a Hybrid III dummy head Walilko et al. (2005)
Boxing	<43.6	<675.9	-	-	Below mTBI injury level, values determined for different punches Smith et al. (1988)
Boxing (hook)	71.2 $\pm$ 32.2	9306 $\pm$ 4485	-	-	Non-injurious, punches to a Hybrid III dummy head Viano et al. (2005)

amount of subconcussive and concussive head impacts and found that an average subconcussive impact had a rotational acceleration of 1230 rad/s^2 (rotational velocity: 5.5 rad/s) while an average concussive impact had a rotational acceleration of 5022 rad/s^2 (rotational velocity: 22.3 rad/s). In addition an injury risk curve was established indicating that a nominal injury value of 6383 rad/s^2 associated with 28.3 rad/s accounts for a 50 % risk of concussion. In a general approach, using a finite element model of the head, Zhang et al. (2004) analysed injury levels based on resulting brain tissue responses. Predictions indicated that shear stress around the brainstem region could be an injury predictor for concussion. The induced shear stress may alter brain function leading to injury. A shear stress of 7.8 kPa was proposed as the tolerance level for a 50 % probability of sustaining a mTBI. Furthermore the model indicated that intracranial pressure can serve as a global response indicator for head injury. It was found that intracranial pressure was more influenced by translational acceleration while shear stress in the central part of the brain was more sensitive to rotational acceleration.

In summary, the various approaches addressing the definition of a concussion injury criterion are not yet conclusive. Because of varying magnitudes and locations of impacts resulting in concussion, as well as other factors such as the number of sustained subconcussive impacts and the number of previous concussions, it may be difficult to establish a general threshold. Particularly in helmet sports it was observed that the threshold values proposed are not always in line with medical findings after impact (Guskiewicz and Mihalik, 2011). Additionally a coupling of head and neck injuries must be considered. On the one hand a head impact can also be associated with a potential for neck injury and on the other hand also neck strength influences measures like the HIC and may thus help explain different concussion risks in professional and youth athletes, women and children (Viano et al. 2007).

With respect to boxing (where facial injuries, particularly eye injuries, are the most common injuries) several studies estimated the loading transferred to the head by punches. Peak punch forces are reported to range from 1666 N to 6860 N whereas the figures vary strongly depending on the body weight of the boxer (Walilko et al. 2005). For a heavyweight boxer, Atha et al. (1985) performed experiments using a ballistic pendulum. Targeting a 7 kg cylindrical metal mass, the boxer's fist reached impact velocities up to 8.9 m/s with a resulting peak impact force of 4096 N. The peak acceleration of the pendulum was 53 g. Smith et al. (2000) determined peak loads of 4800 N for elite, 3722 N for intermediate and 2381 N for novice English boxers. Using an instrumented head form, Smith et al. (1988) measured acceleration for different types of punches to reach up to 43.6 g for translational acceleration and 675.9 rad/s^2 for rotational acceleration. Since these values are below the threshold values proposed by Ommaya (200 g, 4500 rad/s^2), it was suggested that repeated sub-concussive blows were the cause for mTBI.

Walilko et al. (2005) conducted experiments in which Olympic boxers of different weight classes delivered punches to the face of an instrumented Hybrid III dummy. It emerged that the average punch force was 3427 $\pm$ 811 N, the hand velocity reached 9.14 $\pm$ 2.06 m/s, the effective punch mass 2.9 $\pm$ 2 kg and the neck shear force was 994 $\pm$ 318 N. The punch force was higher for the heavier weight class due to a higher effective mass of the punch. It was concluded that the risk of traumatic brain injury from straight blows inducing translational acceleration is low (less than 2 %). However, the high rotational acceleration (exceeding the limit of 4500 rad/s^2) suggested an injury risk due to rotation.

For more severe brain injury like diffuse axonal injury in the white matter (DAI, cf. Sect. 3.2), Margulies and Thibault (1992) determined the threshold for head rotational acceleration to be 9000 rad/s^2. This value is somewhat below the range proposed by Ommaya et al. (2002) who suggested 12500 rad/s^2 for mild DAI, 15500 rad/s^2 for moderate DAI and 1800 rad/s^2 for severe DAI. Comparing the load experienced by human volunteers in a boxing match to different injury thresholds, Smith and Meany (2002) concluded that boxing is unlikely to result in DAI.

Generally it can be noted that in the context of sports head injuries, rotational acceleration is suspected to play a major role with respect to the injury mechanisms for diffuse brain injury and therefore receives considerable attention. Also the consequences of heading in soccer or football are controversially discussed from this point of view (e.g. Kirkendall et al. 2001). It is argued that the translational component of the acceleration in heading is less injurious and can more easily be resisted (e.g. by adequate neck muscle strength). Rotational acceleration in contrast is associated with a higher injury risk and should be prevented, for instance, by good technique. To reduce the risk of concussion from heading, several measures are propagated by the various sports associations ranging from appropriate exercise to prevent rotation to the use of smaller and lighter balls for juvenile players or even a ban of heading for young players.

From a biomechanical perspective, however, it seems unlikely that in sports activities either isolated linear or isolated angular acceleration is sustained. Yang (2007) for instance analysed data of the NFL and found an almost linear correlation between translational and rotational acceleration. Thus, both components have to be considered when investigating the injury mechanism for athletic head injuries.

3.6 Head Injury Prevention

To protect the head against injury a variety of approaches are proposed, all of which, in principle, aim at padding, load distribution and preventing head contact to an object (e.g. Newman 2002).

As described above, impact force and acceleration (both translational and rotational) are relevant physical parameters determining head loading. Most head protection devices primarily aim at reducing translational accelerations through reduction of the forces acting upon the head during an impact; however, this often leads to a reduction of the rotational acceleration as well, since the latter are generated by impact forces acting off the centre of gravity of the head.

One possibility to protect the head is through damping of the forces that could injure the head using deformable padding materials. The stiffness of these materials, together with the available deformation distance, defines the peak force (and, therefore, acceleration) expected to act on the head, while the capability of the padding to absorb energy is the deciding factor for the duration of the acceleration pulse. Such energy absorption can be achieved by deformation or by destruction of material, like in the padding of helmets. If the moving head strikes an object that deforms and thereby allows a longer deceleration distance for the head, the forces generated will be lower. Consequently, the acceleration will be reduced and also the respective injury criteria, for example the HIC, will yield better results. The extent of energy absorption depends strongly on the material properties, the thickness, and the shape of the padding (see also Sect. 3.6.1). Thus to effectively protect the head by padding, a compromise between the following requirements has to be found: allowable padding thickness, maximum padding area, uniformity of the crushing strength of the padding material, and weight.

The first two requirements address the energy absorption capabilities, while the force developed during the impact is controlled by the third. Lastly, the duration of the impact acceleration pulse is also controlled by the elasticity of the material.

For practical reasons, the range of most of these parameters is limited. Basically the same principles as for padding of the car interior apply to helmet design, but additional requirements, for instance concerned with comfort or ventilation, have to be taken into account. Additionally, controlling the weight of helmets through selection of weight-effective padding materials is paramount. More recent helmet designs also address head rotation at impact by implementing several liners that can shift relative to the outer helmet shell. Various guidelines and regulations are currently available defining the requirements for different types of helmets, like motorcycle or sports helmets.

In addition to energy absorbing interior structures as mentioned, all modern vehicles are equipped with restraint systems such as seat belts and airbag systems for head injury protection. The seat belt (i.e. a three-point belt rather than a lap belt) aims at preventing head contact by restraining an occupant. In frontal impact, the belt effectively reduces the risk of the head impacting parts of the vehicle interior like for example the steering assembly, the A-pillar or the dashboard. Also airbags contribute to the reduction of head injury, particularly of severe brain injury. By distributing the restraint force over large body areas, including the head, high force concentrations are reduced. The head benefits from smaller acceleration. Furthermore, airbags allow a better control of the deceleration of the occupant within the occupant compartment and also reduce relative motions between adjacent body parts. However, in special impact scenarios, airbags are associated with additional loading to the occupant. See Chap. 5 for comments on such airbag inflation induced injuries.

3.6.1 Head Injury Prevention in Pedestrians

Head injuries are the prevalent cause of death in car-pedestrian collisions. In order to prevent or at least mitigate the consequences of a head impact onto a car front, the deformation characteristics of the bonnet and fenders must be adapted. With the

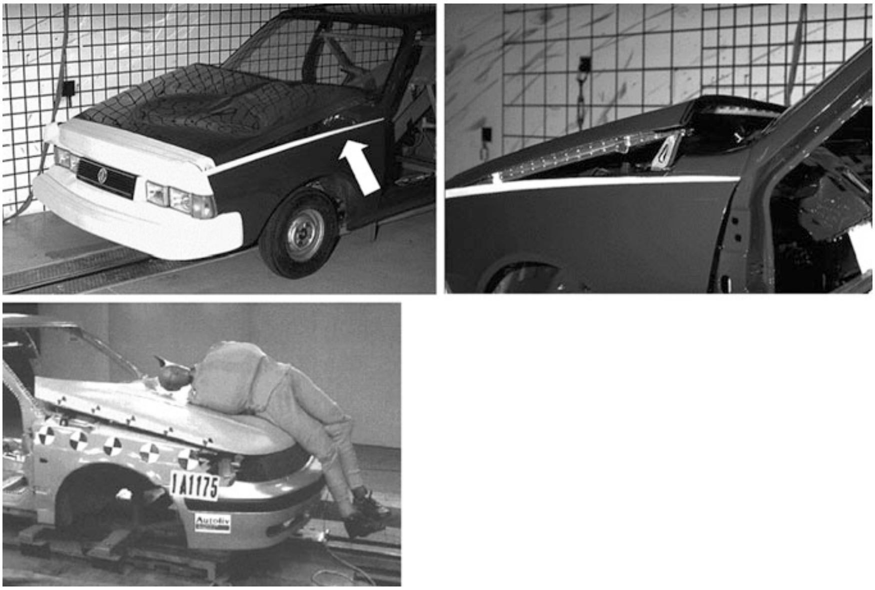


Fig. 3.11 Already in 1983 the Working Group on Accident Mechanics at ETH Zurich developed a rear-rising bonnet and showed its effectiveness (*top row*). The picture on the *bottom row* illustrates a model as it is introduced in current vehicles (Autoliv 2003)

advent of homologation testing requirements in Europe, implementing compliant bonnet and fender designs has become an important task within the car industry.

Critical points are the stiff bonnet frame as well as the inner reinforcement structures. Especially, the structure of the reinforcements at the underside of the bonnet affects the severity of pedestrian head impact. In case the pedestrian head hits any of these stiff structures, a high head acceleration is to be expected. Lighter reinforcement structures and optimised bonnet design along with the use of bonnet material that allows sufficient deformation were shown to be beneficial.

Generally, the under-bonnet clearance is the measure that determines the maximum deformation possible. Besides the re-design of the bonnet itself, rear-rising bonnet systems (also called “pop-up” bonnets) were developed (Fig. 3.11). By lifting the rear portion of the bonnet additional space underneath the bonnet is gained which can be used for deformation and energy absorption. Depending on the sensor and trigger unit used, such systems can either be activated if a pedestrian impact is likely or if it actually occurs.

Airbags covering parts of the vehicle front represent another injury countermeasure and may be used in addition to rear-rising bonnet systems. Over-the-bonnet airbags were presented that deploy above the bumper to cover the front of the vehicle and most of the bonnet. Cowl airbags on the other hand deploy from the base of the wind shield and cover this area from A-pillar to A-pillar; they are triggered by an impact sensor at the front of the vehicle (Fig. 3.12). Today first vehicles equipped with a cowl airbag are on the market. It should be noted that

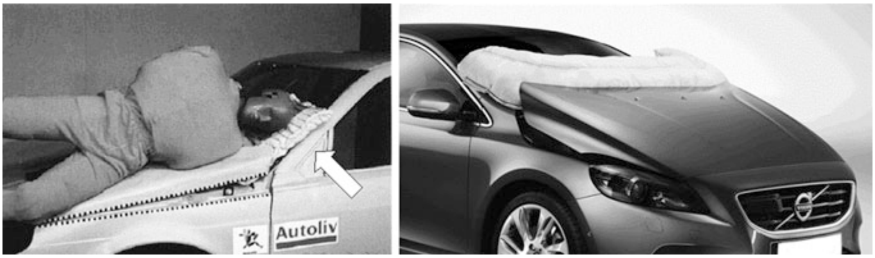


Fig. 3.12 Cowl airbags combined with rear-rising bonnets used to prevent pedestrian head injuries (Autoliv 2003; Volvo 2013)

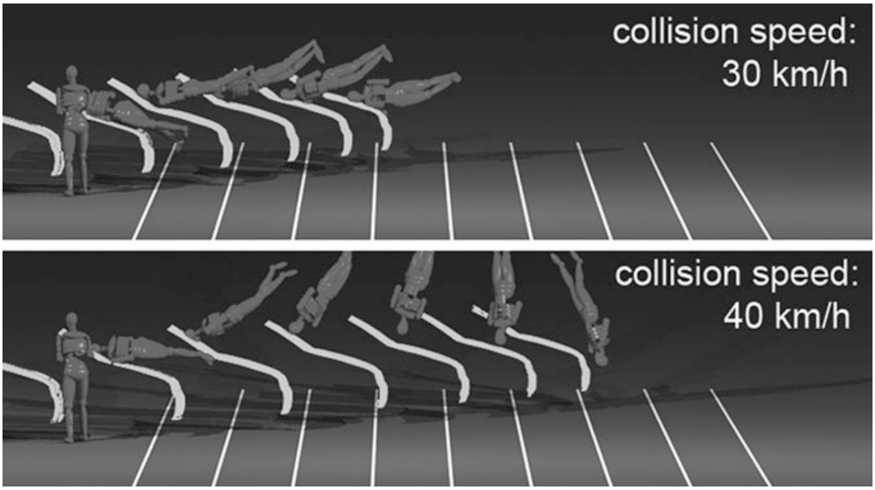


Fig. 3.13 Computer simulation of a pedestrian impact: the kinematics of the pedestrian change dramatically with increasing collision speed of the vehicle. This indicates that the secondary road impact has also to be considered (AGU 2003)

detecting a pedestrian and determining whether a collision is unavoidable is difficult. Different strategies are used to tackle this task using various approaches in sensing technology. Consequently airbag systems are usually accompanied by further active safety systems such as warning systems that detect a pedestrian whose movement might result in a critical situation and alert the driver. If the driver fails to react, autonomous braking systems are available that aim at avoiding a collision (with pedestrians as well as other road users). The airbag comes last in this cascade and is only deployed if the collision could not be avoided.

Finally, it is important to also consider the kinematics after the primary impact of the pedestrian on the vehicle with regard to secondary road impact (Fig. 3.13). Otherwise, an optimised vehicle design to prevent injury from the primary impact might well increase the risk of sustaining injuries from the secondary impact, which would ultimately prove to be counter-productive.

3.7 Summary

Concussion and mTBI are of major concern in sports. Several studies address possible injury thresholds, but a general threshold is difficult to achieve due to various parameters influencing the injury risk (including the effect of repeated subconcussive impacts). In the automotive field different injury criteria are in use. The Wayne State Tolerance Curve is a major landmark in this context since it serves as the basis for several head injury criteria. Currently, HIC is the most widely used criterion, although it is often criticised that only translational acceleration is taken into account.

3.8 Exercises

E3.1: A belted occupant (no airbag) of normal stature is involved in a frontal crash with a delta-v of about 40 km/h. He loses consciousness immediately after the collision for a short period of time. No externally visible lesions. Classify the head injury.

E3.2: Explain the two principal purposes of a motorcycle helmet.
E3.3: "*HIC_{15} is always lower or equal to HIC_{36} for a given head acceleration*". Is this statement true? Why/why not?

P3.1: A head impactor (m = 4.5 kg, v = 40 km/h) is impacted vertically on a deformable surface. The surface exhibits a spring-like characteristic, i.e. the reaction force on the head is directly proportional to deformation depth. The "spring constant" of the surface is 80 kN/m. Calculate the acceleration vs. time function for the impactor, assuming that

(a) the surface is purely elastic, i.e. acts like a spring, and
(b) the surface is completely inelastic, i.e. its coefficient of restitution is zero.

Compare peak, average, and the duration of the two head acceleration curves. Which curve would exhibit the higher HIC value?

P3.2: The same head impactor (m = 4.5 kg, v = 40 km/h) is impacted vertically on a deformable surface. Design the force–deformation characteristic of this surface such that HIC is below 1000 and deformation distance is minimal. You may assume that the reactive force of the surface depends only on deformation depth, i.e. independent of the shape of the impactor. Use and compare simple characteristics as e.g. the triangular characteristic, a rectangular characteristic (reaction force is constant, irrelevant of deformation depth), an inverse triangular characteristic (reaction force is highest at the beginning of deformation, lowest at the end).

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Spinal Injuries

4

The potential for long-term impairment, including para- and quadriplegia, is always inherent in injuries to the spine and particular to the spinal cord. Of all spinal segments, the cervical spine is most frequently injured. Considering that the head and the neck form one functional entity, head loading often also implies neck loading.

With respect to traffic accidents, severe (head-contact) cervical injuries are recorded from unbelted car occupants and motorcyclists (with or without helmet). The vast majority of cervical spine injuries, however, are minor soft tissue neck injuries usually graded as AIS1 and often not exhibiting a morphological manifestation. These injuries, while not associated with overt structural injury to the cervical spine or the central nervous system, are both common and potentially debilitating. It should be noted that several expressions are found in the literature to describe such soft tissue neck injuries. Cervical spine disorders (CSD), whiplash injuries and whiplash associated disorders (WAD) are also commonly used. Although the term "whiplash" is probably used most often, it is misleading and incorrect, because it evokes a certain injury mechanism (i.e. a forward–backward movement like during the development of the crack of a whip). However, the underlying mechanism is not yet fully established (Sect. 4.2). Generally, it might not even be correct to automatically use the term "injury" in that context since "injury" implies the prevalence of a morphological finding. The terms "injury" on the one hand and "neck pain", "complaint" or "symptom" on the other hand have to be differentiated; they are often erroneously used as synonyms.

Despite the uncertainties regarding the injury mechanism, one should be aware that soft tissue neck injuries are among the most frequently occurring injuries in automotive accidents. Although most sufferers make a complete recovery within a short period of time, some cases develop prolonged medical problems. Hence, the socioeconomic significance of these injuries is tremendous. This relates first of all to health aspects, but concerns for example also legal aspects as these injuries play a major role in insurance and court cases. Consequently, a better understanding of the vehicle, collision and occupant parameters that are prevalent in soft tissue neck injuries is needed in order to develop preventive measures.

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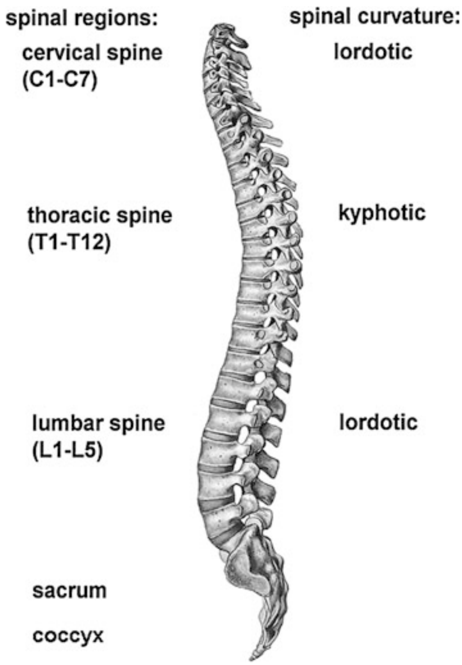


Fig. 4.1 Human spine (adapted from Sobotta 1997)

4.1 Anatomy of the Spine

The human vertebral column is the principal load-bearing structure of the head and the torso (Fig. 4.1). It is divided into 7 cervical, 12 thoracic, and 5 lumbar vertebrae. The vertebrae are numbered C1 to C7 for the cervical spine (with C1 being the uppermost vertebra which is connected via the occipital condyles to the skull), T1 to T12 for the thoracic vertebrae, and L1 to L5 for the lumbar vertebrae. The entire column is supported by the sacrum and the coccyx which are anatomically a part of the pelvic girdle. The size of the vertebrae increases from cranial to caudal. Adjacent vertebrae are separated by intervertebral discs. The lateral view of the entire spine shows the principal spine curves: the lordotic cervical and lumbar curves and the kyphotic thoracic curve (Fig. 4.1). When viewed frontally, the normal spine is straight.

In general, each vertebra consists of a cylindrically shaped body, a vertebral (or neural) arch, the (dorsal) spinous process and transverse processes at each side (Fig. 4.2). The spinous and transverse processes serve as attachment points for muscles and ligaments. Such muscles and ligaments account for stability and movements, especially of the head and neck. There are three spinal ligaments that run along the entire length of the spine: the anterior and posterior longitudinal ligaments which line the anterior and posterior aspects of the vertebral bodies, and the supraspinous ligament which joins the tips of the spinous processes (Fig. 4.3). Spinal

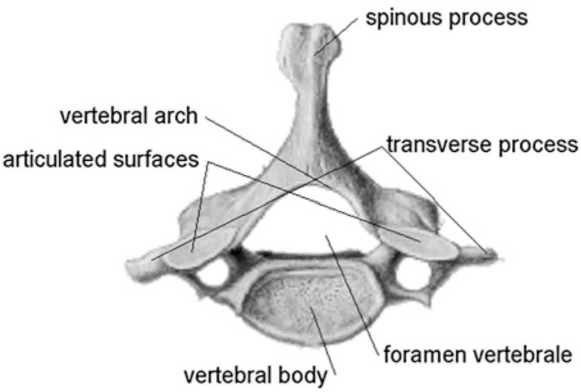


Fig. 4.2 Different regions of a vertebral body (adapted from Sobotta 1997)

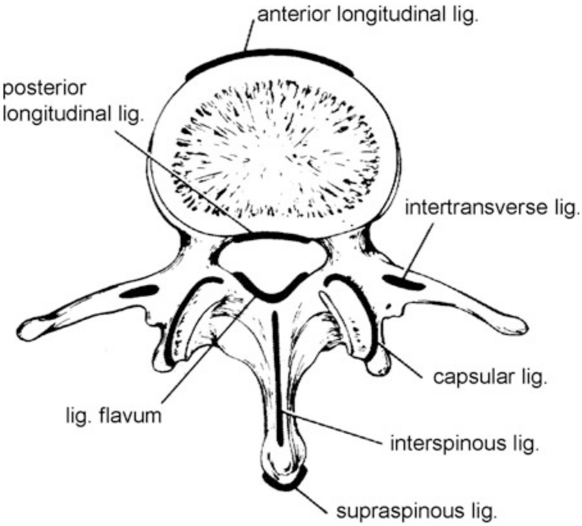


Fig. 4.3 Major spinal ligaments and their attachment points (adapted from Sobotta 1997)

ligaments are in general pre-stressed in neutral position. As for the muscles, superficial, intermediate and deep muscles are distinguished. They are symmetrical about the sagittal plane, i.e. all muscles appear in pairs. The deep muscles are closely attached to the vertebrae, while the intermediate muscles account for longer distances connecting, for example, the neck to the thorax and the skull. The superficial muscles, on the other hand, have no direct attachments to the spinal column.

The foramina vertebralia of all vertebrae form the spinal canal that includes the spinal cord and the associated soft tissues (Fig. 4.4). The spinal cord itself is surrounded by cerebrospinal fluid (CSF). Blood vessels, especially venous vessels, are also present within the spinal canal.

In the cervical spine, it has to be noted that vertebrae C1 and C2 are anatomically different from other vertebrae. C1, also called atlas, comprises a bony ring with large articulated surfaces only. Together with the second cervical vertebra,

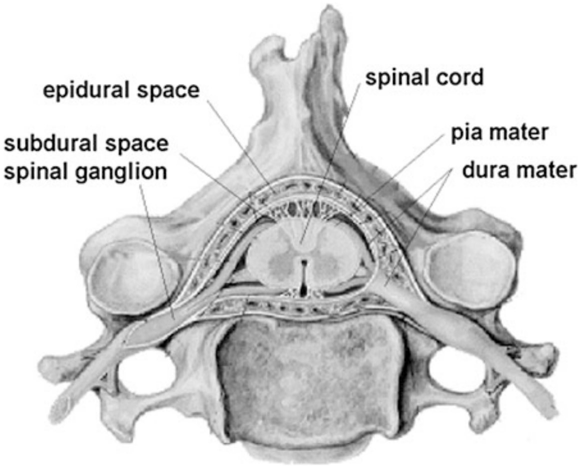


Fig. 4.4 Spinal canal and associated soft tissue (adapted from Sobotta 1997)

which is characterized by a dominant process (dens axis) on its upper side, they form the atlanto-axial joint. Consequently, there is no intervertebral disc between C1 and C2 (Fig. 4.5).

With respect to physiological neck motion, four basic movements are possible: flexion, extension, lateral bending and (axial) rotation (Fig. 4.6) and combinations thereof. To allow this motion, different joints can be identified. Apart from the atlanto-axial joint which is responsible for head rotation, the intervertebral joints, in particular the intervertebral discs, by nature of their fibre-enforced annulus and the viscous nucleus, transmit (compression and shear) forces and moments. Furthermore, the motion is guided by the two sets of facet joints (also called zygapophyseal joints) of each vertebra.

4.2 Injury Mechanisms

Assessing the threat to life, the AIS (Abbreviated Injury Scale) grades several spinal injuries (Table 4.1). Injuries to the upper cervical spine are generally considered to be more serious and life threatening than those at a lower level (Viano 2001).

In principle, injuries to the cervical spine can be classified according to the possible motion of the neck and possible mechanical loading (Figs. 4.6 and 4.7). Shear in antero-posterior direction and axial torsion may cause dislocation of the atlanto-occipital joint, while large compression might result in fracture of the atlas (C1) thereby breaking C1 into two to four sections (Jefferson's fracture, Fig. 4.8). Isolated atlas fractures are common after axial loading; however, 40–44 % of atlas fractures have concomitant axis fractures (Kakarla et al. 2010). If axial compression is combined with extension of the neck, C2 fractures, commonly known as hangman's fractures, can occur. In a vehicle accident this type of fracture is

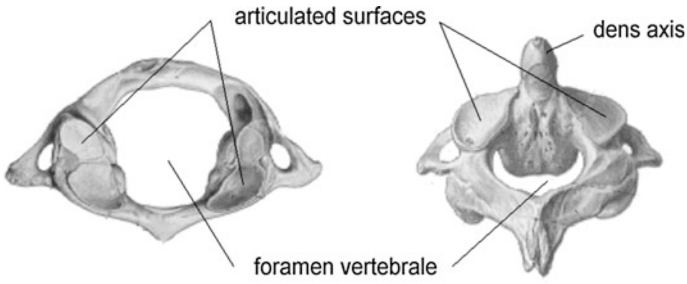


Fig. 4.5 Cervical vertebrae C1 (altas, *left*) and C2 (axis, *right*) (adapted from Sobotta 1997)

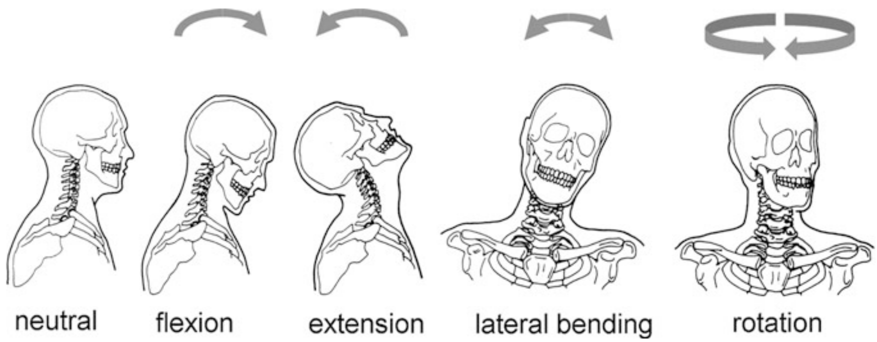


Fig. 4.6 The four basic movements of head and neck (adapted from Sances et al. 1984)

Table 4.1 Examples of spinal injuries according to AIS scale (AAAM 2005)

AIS code	Description
1	Skin, muscle: abrasion, contusion (hematoma), minor laceration
2	Vertebral artery: minor laceration
	Cervical/thoracic spine: dislocation without fracture
	Thoracic/lumbar spine: disc herniation
3	Vertebral artery: major laceration
	Cervical/thoracic spine: multiple nerve root laceration
4	Cervical/thoracic spine: spinal cord contusion incomplete
5	Cervical/thoracic spine: spinal cord laceration without fracture
6	Decapitation
	Cervical spine: spinal cord laceration at C3 or higher with fracture

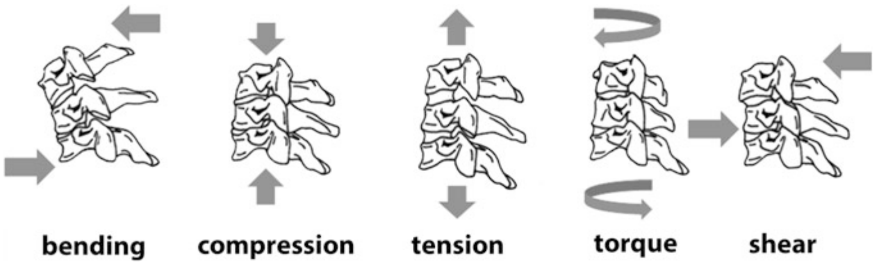


Fig. 4.7 Possible loading of the neck includes compression of the neck, tension (force stretching the neck), shear (force perpendicular to the neck column), flexion moment (forward bending of the neck), extension moment (rearward bending of the neck) and axial torsion (adapted from McElhaney et al. 2002)

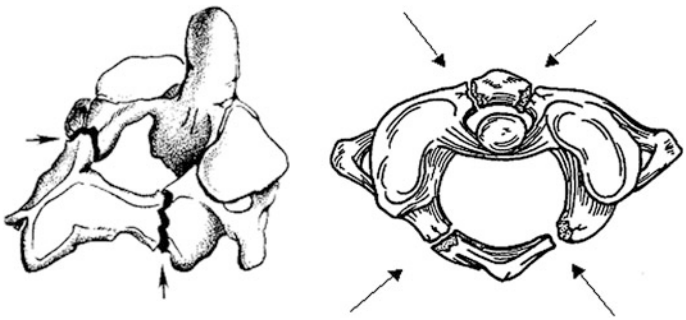


Fig. 4.8 Hangman's fracture (*left*) and Jefferson's fracture (*right*) (adapted from Vetter 2000)

often related to an unrestrained occupant whose forehead or face impacts, for example, the windscreen.

In automotive crashes, loading of the neck is generally due to head contact forces and combined axial or shear load with bending. Because of the anatomical curvature of the cervical spine, bending in whichever direction is almost always present. While pure compression may result in fractures as described above, non-contact head acceleration and airbag deployment are suspected to be the cause of pure tension injuries of the upper cervical spine (e.g. McElhaney et al. 2002). Lesions resulting from tensile loading include dislocation of the occipital condyles, ligamentous injury and fractures, e.g. dens fracture (e.g. Nightingale et al. 1998).

However, as head contact is frequently observed, the following neck injury modes are considered predominant: compression-flexion, compression-extension, tension-flexion, tension-extension and lateral bending.

Wedge fractures of the anterior vertebral bodies result from a combination of flexion bending moment and axial compression force of the neck. Head rotation often accompanies this loading scenario, but is not essential. With increasing load, burst fractures and fracture dislocation of the facets can occur (Fig. 4.9). The latter

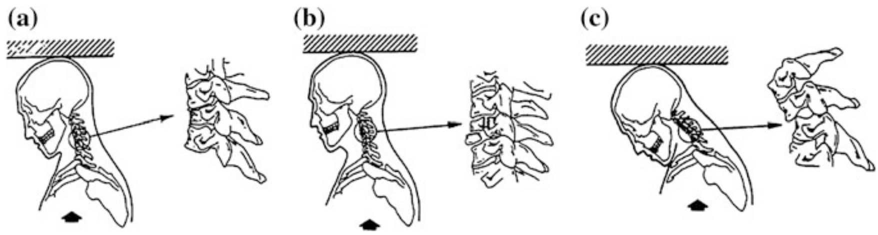
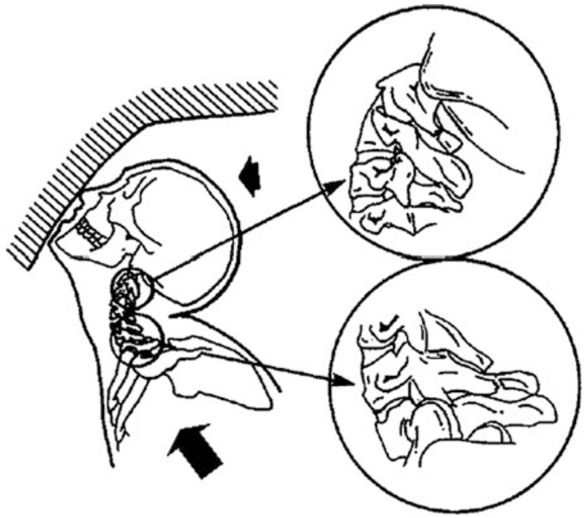


Fig. 4.9 Compression-flexion injury mechanism can result in wedge fractures (**a**), burst fractures (**b**), or bilateral facet dislocation (**c**). Although the figure illustrates head rotation as part of the injury mechanism, research has shown that head rotation needs not accompany this injury (adapted from McElhaney et al. 2002)

Fig. 4.10 Compression-extension mechanism: compression of the cervical spine can be enforced by inertia forces due to the body being moving towards the head (adapted from Goldsmith and Ommaya 1984)



two conditions are unstable and potentially disrupt or injure the spinal cord. Hereby the extent of the injury depends on the penetration of the vertebral body or its fragments into the spinal canal.

Compression-extension loading produces fractures of the posterior structures of the neck in both the upper and lower regions (e.g. Pintar et al. 1995; Nightingale et al. 1997). As indicated in Fig. 4.10, frontal impact to the head with the neck in extension is likely to cause compression-extension loading.

Frontal impact in which the torso is restrained and the neck is meant to stop head movement can result in flexion of the cervical spine while being subjected to tension. Bilateral facet dislocation was observed after such loading. However, it should be noted that such injury may also result from compression-flexion loading, suggesting that the magnitude of the bending moment rather than the axial load seems to be the determining factor.

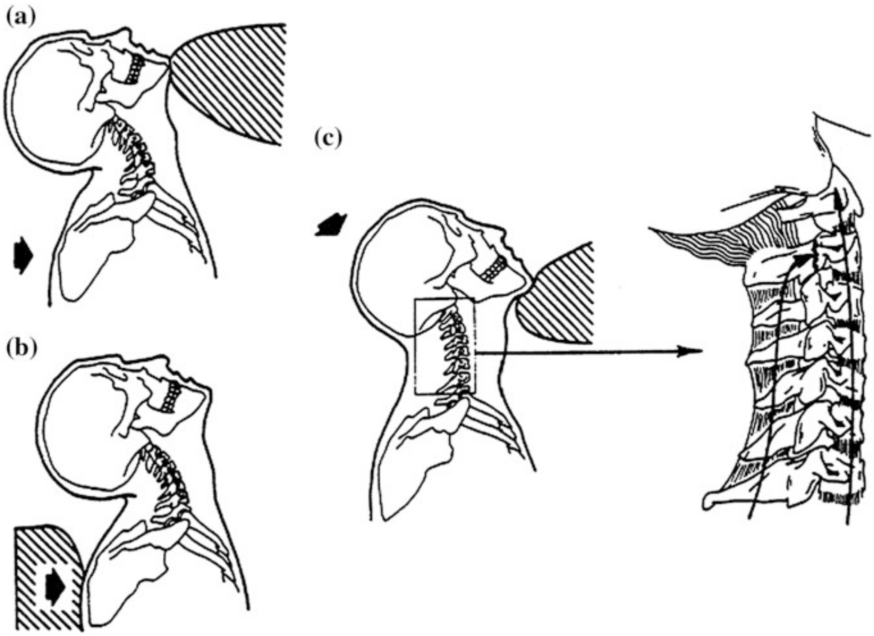


Fig. 4.11 Tension-extension loading primarily occurs through (**a**) fixation of the head with continued forward displacement of the body, **b** inertial loading of the neck following an abrupt forward acceleration of the torso, and **c** forceful loading below the chin directed postero-superiorly (adapted from McElhaney et al. 2002)

Tension-extension loading is the underlying mechanism for several injuries (Fig. 4.11). It commonly occurs when unbelted occupants hit the windscreen or when the chin impacts the dashboard. In both cases the head rotates rearward and tensile force and an extension moment are applied on the neck. In case of hitting the windscreen with the head, also a hangman's fracture of C2 can occur. Furthermore, soft tissue neck injuries are hypothesised to result from tension-extension loading. A more detailed discussion of this type of injury is presented further down in this section.

Injuries due to lateral bending are, for instance, observed after automotive side impact. Axial loads (e.g. compression) or shear loads are often associated with lateral bending (Fig. 4.12). Lateral wedge fractures of the vertebral body and fractures to the posterior structures on one side of the vertebral column can result. Additionally, lateral bending in combination with torsional loads may occur. Such cases possibly lead to unilateral facet dislocations or unilateral locked facets (Moffat et al. 1978). However, pure torsional loads on the neck are rarely encountered in automotive accidents (Viano 2001).

Soft tissue neck injuries are by far the most frequent injuries of the spine that are sustained in automotive accidents. Epidemiological data shows that females have a higher risk of sustaining such injuries than males (e.g. Viano 2003; Linder

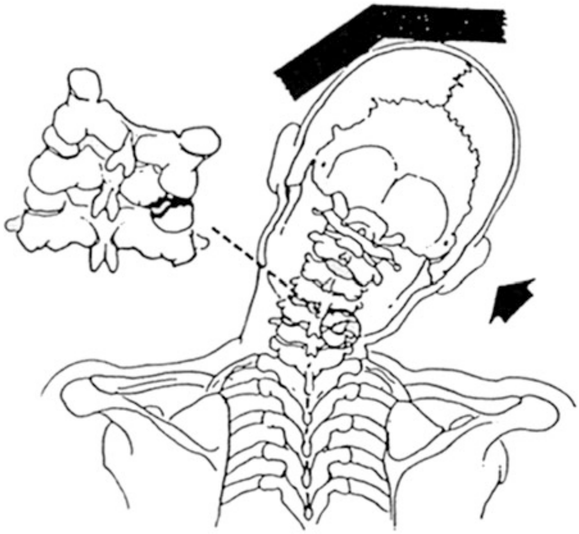


Fig. 4.12 Lateral bending and compression resulting in fracture on the compressed side (adapted from Vetter 2000)

et al. 2008, 2012). Soft tissue neck injuries are reported from low-speed rear-end collisions, as well as from frontal and frontal-oblique collisions involving belted occupants, with and without head contact. The symptoms are diverse ranging from neck pain, headache, numbness, dizziness to visual disorders and neurological deficiencies (e.g. Ferrari 1999). In many cases no lesions are evident even if advanced diagnostic measures are applied, and therefore such injuries are most often classified as minor (AIS1). Such a classification using the AIS scale is rather broad. To allow for a more detailed assessment of soft tissue neck injuries, the Quebec Task Force (Spitzer et al. 1995) established another injury scale, categorising the symptoms and signs into four grades according to the clinical presentation (Table 4.2).

Regarding the injury mechanism, hypotheses have to rely on experiments (see Sect. 4.3) or on symptomatic clinical observations. Due to the complex anatomy of the cervical spine various vulnerable structures are assembled in a quite small area. Hence, different tissues were proposed to be the cause for neck pain. Many studies have suggested that ligaments and muscles are injured. The zygapophyseal joint is also often suspected to cause complaints. Also nerve tissue injuries, in particular in the vicinity of the spinal ganglia, are reported. Further hypotheses can be found identifying other tissues (e.g. vertebral arteries, intervertebral discs), but these suggestions are discussed very controversially (cf. e.g. Ferrari 1999; Curatolo et al. 2011).

Analysis of the motion of the neck during rear-end or frontal collisions (both may to result in soft tissue neck injuries) reveals complex kinematic sequences including various mechanical loading conditions. The resulting movement of a car occupant during a rear-end collision (i.e. the vehicle is struck from behind and thus

Table 4.2 Classification of WAD as proposed by the Quebec Task Force (adapted from Spitzer et al. 1995)

Grade	Clinical presentation
0	No complaint about the neck
	No physical sign(s)
1	Neck complaint of pain, stiffness or tenderness only
	No physical sign(s)
2	Neck complaint and musculoskeletal sign(s)
3	Neck complaint and neurologic sign(s)
4	Neck complaint and fracture or dislocation

is accelerated forward), can for example be divided into three different phases (Fig. 4.13). In the retraction phase, an occupant who is sitting upright in the driver's position is pushed forward by the seat back. The contact and force transmission occurs primarily in the shoulder area. Due to its inertia, the head, which is not in contact with any structure of the car, has a tendency to keep its state of motion. Relative to the occupant, the head lags behind the torso. It moves backwards without any rotation about the lateral axis, i.e. it retracts. Hence, the upper cervical spine is forced into flexion mode and the lower cervical spine into extension. This deformation of the neck, called S-shape formation, is regarded as crucial for the injury mechanism. The presence of the S-shape is well established today and was observed in various experiments using dummies equipped with specially designed necks as well as volunteers and cadavers (cf. e.g. Ono and Kaneoka 1997; Ono et al. 1998; Eichberger et al. 1998; Grauer et al. 1997; Svensson et al. 1993; Wheeler et al. 1998; Yoganandan and Pintar 2000; Ivancic and Xiao 2011). Following the S-shape formation, the head starts rotating backwards which in turn leads to the extension of the entire cervical spine. Eventually, the retraction phase is concluded by maximum extension which depends on impact severity, the presence of a head restraint and the individual physiological limitations of the occupant.

The next phase is a forward movement which is characterised by a change in the direction of the movement, i.e. head, neck and torso move now forward. This phase is strongly influenced by the seat design, in particular by the elasticity of the seat and the corresponding rebound effect (Muser et al. 2000; Viano et al. 2013). When the occupant returns to a position which, in sagittal direction, equals the initial position prior to the impact, the forward motion phase is completed and the belt restraint phase follows. The belt system restrains the occupant such that the thoracic spine is stopped while the head continues to move forward. Consequently, an inverse S-shape of the neck can be observed. However, due to damping of the restraining forces by the thoracic cage, this effect is less pronounced than the S-shape experienced first. Additionally, since the (ventral) belt contact point is lower than the (dorsal) shoulder contact point, more vertebrae participate in this second S-shape formation, thus allowing for smaller loads on the individual vertebrae.

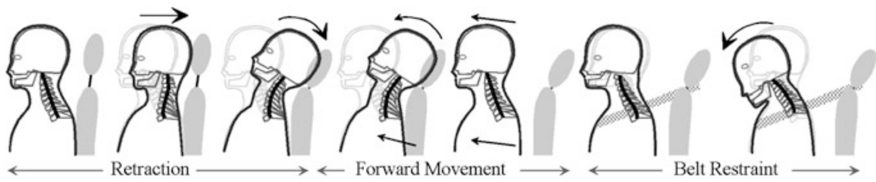


Fig. 4.13 Different phases of a rear-end collision (adapted from Muser et al. 2000)

Finally, an overall flexion of the cervical spine concludes the kinematic phases specifically related to rear-end collision.

For frontal collisions (without head contact) similar kinematic phases including the inverse S-shape apply.

Considering the complex occupant kinematics, it is not surprising that different mechanisms are discussed as causes for soft tissue neck injuries (e.g. Walz and Muser 1995). A shearing movement of the vertebrae has, for instance, been related to lesions of the facets of the intervertebral joints (Yang et al. 1997). Hyperextension of the neck, i.e. an exceedance of maximum neck moments and head excursion angles (Mertz and Patrick 1971), is also considered to be a possibility. However, due to head restraints such a mechanism has become rare. Additionally, a hypothesis proposed by Aldman (1986) takes into consideration a pressure gradient that develops in the venous and cerebrospinal fluid of the spinal canal causing cellular injuries (Svensson et al. 1993; Schmitt 2001).

In summary, many causes for soft tissue neck injuries are hypothesised, but the underlying mechanisms are difficult to elucidate and are therefore not fully understood, let alone scientifically proven. In practice, the S-shape deformation is regarded to play a crucial role when discussing possible injury mechanisms. Consequently, injury criteria based on the S-shape deformation have been established (see Sect. 4.4) and new dummies have been developed that are capable of a biofidelic reproduction of the relative motion between the head and the torso (see Sect. 2.6.1).

Injuries to the thoracolumbar spine sustained in automotive crashes are rare and play a minor role compared to cervical spine injuries. However, back pain is commonly reported after collisions and severe injuries to the spinal cord may, of course, also occur. King (2002) identifies seven different types of thoracolumbar spine injuries: anterior wedge fractures of the vertebral bodies, burst fractures of vertebral bodies, dislocations and fracture-dislocations, rotational injuries, Chance fractures, hyperextension injuries and soft tissue injuries. With respect to automotive accidents, anterior wedge fractures result when combined flexion and axial compression loading arise. This may happen, for example, in severe frontal impacts when the shoulder harness imposes a large load across the torso, causing the curved thoracic spine to straighten out. Experiments conducted with cadavers and volunteers that were restrained by a three-point belt showed that a compressive force is generated in the thoracolumbar spine which might cause wedge fractures (Begeman et al. 1973). Principally these injuries occur at all levels of the thoracolumbar spine, but are most likely between T10 and L2 (King 2002). Anterior wedge fractures are also observed in aircraft accidents, particularly when ejection of the pilot is involved. Historically, the analysis of the pilot ejection problem was the primary motivation to study injuries of the thoracolumbar spine.

Chance fractures (named after G. O. Chance who first described this type of fracture in 1948) are due to improper wearing of the lap belt in case of a frontal collision. If the angle of the lap belt relative to the horizontal plane is too flat, the belt may slide over the iliac crest, thereby compressing the abdominal organs (see also Sect. 6.2). This also causes the lumbar spine to flex which can result in separation of the posterior elements of the spine, for instance, by ruptures of the supra- and interspinous ligaments. Furthermore, the spinal cord is stretched and might be injured.

Injuries of the soft tissue of the thoracolumbar spine are also often reported after automotive accidents. The soft tissues involved are the intervertebral discs, the various ligaments, the facet joints, the muscles and tendons attached to the vertebral column. A usual complaint of this type of injury is low-back pain. Incidents provoking this complaint are manifold, ranging from minor rear-end collisions to severe frontal impacts. In some cases the back pain is associated with disc rupture or disc bulge. However, a causal relationship between an impact and a rupture does usually not exist (King 2002). Disc ruptures are generally the result of a slow degenerative process.

4.3 Biomechanical Response and Tolerances

The mechanical performance of the human spine was subject to numerous volunteer, cadaver, animal and dummy tests. Experiments were conducted statically and dynamically (both with and without head impact) utilising different test setups (see also Sect. 2.5). Further, the use of so-called functional units is common in spine testing. Here a functional unit usually consists of a motion segment of two or three vertebrae. Tissue that is not of interest in the study performed (e.g. muscles) is dissected. Analysing the head-neck kinematics, some studies also use larger units that are made of a cadaver head and neck whereas the neck is fixed at its lower end and then mounted on a (mini-) sled. However, it has to be noted that the use of functional units can influence the kinematics significantly. This has to be considered when drawing conclusions from results obtained in such experiments. A thorough review on biomechanical experiments facilitating functional units under various loading conditions is provided by Jaumard et al. (2011).

Muscle activity can often not be simulated in experiments, because they are either removed (functional units) or without tonus (cadaver experiments). Only volunteer experiments offer the possibility to measure muscle activity to some extent. Tolerances to other spinal injuries, like for instance vertebral artery lesions, are difficult to assess because a physiological limit of the loading of the structure of interest cannot be defined properly.

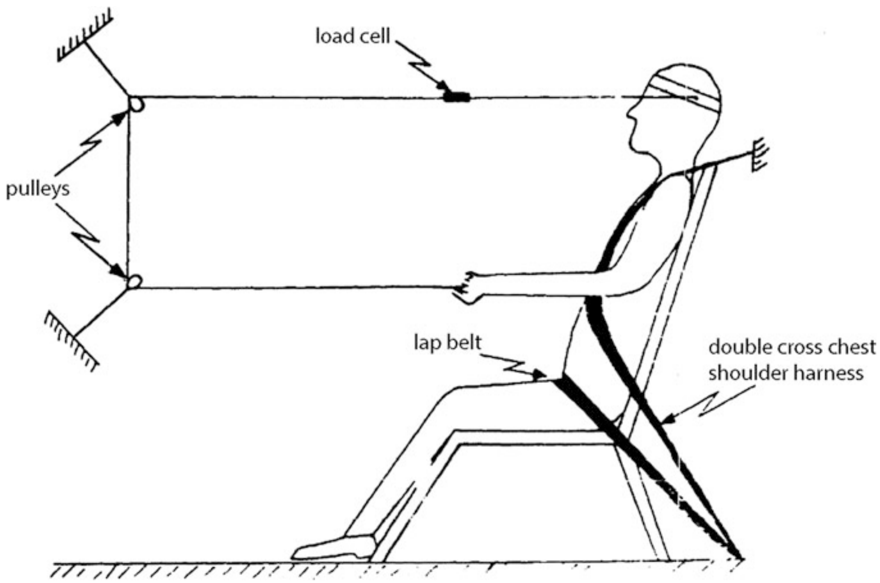


Fig. 4.14 Set-up used by Mertz and Patrick (1967) to perform static volunteer tests

Investigating the biomechanical response of the cervical spine, many studies still refer to and rely on tolerance levels based on volunteer and cadaver experiments that were performed in the late nineteen sixties and the early seventies, for instance by Mertz and Patrick (1967, 1971). Figure 4.14 shows a test set-up used to determine static properties of the neck.

Further, sled tests were conducted to account for the dynamic effects when loading the neck (Goldsmith and Ommaya 1984). Volunteer tests provided data up to the pain threshold, and cadaver tests extended the limits for serious injuries (Fig. 4.15).

More recent studies investigated the relative motion of each vertebra in volunteer sled test experiments by using X-ray based techniques like cineradiography (e.g. Ono and Kaneoka 1997, 2001; Ono et al. 2006). This way the motion pattern of each vertebra could be assessed.

For the lumbar spine, fracture thresholds were determined by different experiments using functional spinal units. Compression fractures were observed for loads ranging from approximately 2–6 kN (e.g. Hutton and Adams 1982; Myklebust et al. 1983; Yoganandan et al. 1988; Myers et al. 1994; Belwadi and Yang 2008). Regulations in aviation safety require the maximum compression of the lumbar spine not to exceed 6672 (N = 1500 pounds). For more details see regulations CS 23–29 of the European Aviation Safety Agency (EASA 2013) or the corresponding U.S. standards, 14 CFR 23–29, of the Federal Aviation Agency (FAA 2013). Limits for frontal, rear-end and downward acceleration of the lumbar spine were proposed such that for impact durations of less than 100 ms, a 40 g threshold shall not be exceeded for well-restrained seated passengers (Viano 2001).

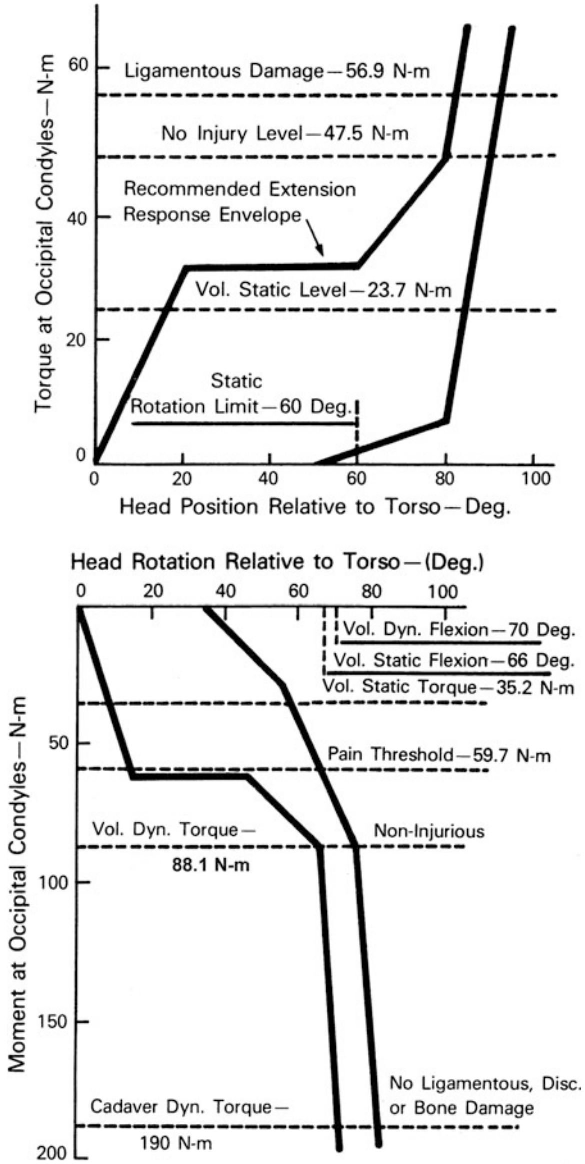


Fig. 4.15 Head-neck response envelope for extension (*top*) and flexion (*bottom*) as determined by Goldsmith and Ommaya (1984)

Table 4.3 summarises tolerance values of the cervical spine derived from the various experiments reported in the literature; however, due to differences in experimental techniques and test conditions, the data shows a considerable spread. Furthermore, one has always to bear in mind that tolerance, in addition to being a function of the loading environments, is related to a variety of factors including variability of the anatomical structures (e.g. in terms of geometry or properties

Table 4.3 Tolerance of the cervical spine to injury

Response measured	Test objects	Threshold criterion	Threshold value	Reference
Extension	Volunteers	No-injury (static)	23.7 Nm	Goldsmith and Ommaya (1984)
		Pain	47.3 Nm	Mertz and Patrick (1971)
		No-injury	47.5 Nm	Goldsmith and Ommaya (1984)
	Cadavers	AIS2, ligamentous injury	56.7 Nm	Goldsmith and Ommaya (1984)
Flexion	Volunteers	Pain	59.4 Nm	Mertz and Patrick (1971)
			59.7 Nm	Goldsmith and Ommaya (1984)
		Maximum voluntary loading	87.8 Nm	Mertz and Patrick (1971)
			88.1 Nm	Goldsmith and Ommaya (1984)
	Cadavers	AIS2 (no fractures)	189 Nm	Mertz and Patrick (1971)
			190 Nm	Goldsmith and Ommaya (1984)
Compression	Cadavers	Bilateral facet dislocation	1.72 kN	Myers et al. (1991)
		Compression injuries	4.8–5.9 kN	Maiman et al. (1983)
Tension	Volunteers	No-injury (static)	1.1 kN	Mertz and Patrick (1971)
	Cadavers	Failure	3.1 kN	Shea et al. (1991)
Shear (a-p)	Volunteers	No-injury	845 N	Mertz and Patrick (1971)
	Cadavers	Irreversible damage	2 kN	Goldsmith and Ommaya (1984)
	Functional unit	(Odontoid) fractures	1.5 kN	Doherty et al. (1993)
	Functional unit	Ligament rupture	824 N	Fielding et al. (1974)

such as bone density) and presence of degeneration. Hence, it is not surprising that differences for male and female tolerance values are observed. Nightingale et al. (1997) report significant differences when testing compressive failure. For compression tolerance McElhaney et al. (2002) conclude that for the cervical spine average forces of 1.68 and 3.03 kN result for females and males, respectively.

Further differences in the dynamic response of female and male volunteers (e.g. differences in head acceleration) were found in sled test experiments mimicking rear-end collisions (Linder et al. 2008; Carlsson et al. 2011, 2012a). Also geometric measures like a smaller circumference of the female neck might influence the kinematics in rear-end collisions resulting in higher angular acceleration

(Dehner et al. 2007). For other factors such as muscle activation, it remains unclear whether (or how) they affect kinematics and injury outcome (e.g. Siegmund 2011).

Age, in contrast, does account for different biomechanical response and tolerance values (cf. e.g. Yoganandan and Pintar 2000; Yoganandan et al. 2002). Higher tolerance values for compression were, for example, reported by McElhaney et al. (2002) for young males compared to adults (tolerance ranges from 3.64 to 3.94 kN compared to the values for adults as mentioned above). Concerning the influence of spinal degeneration such as osteochondrosis, spondylosis, spondylarthrosis or spinal disc height reduction (without affection of the spinal cord) the literature is not conclusive. Related to neck pain, for instance, the literature does not report a clear link between degeneration and extent of clinical complaints (e.g. Meenen et al. 1994; Marchiori and Henderson 1996; Nykänen et al. 2007; Carroll et al. 2008).

4.4 Injury Criteria

In addition to the tolerance values for spine loading described in Sect. 4.3, several neck injury criteria are defined. Besides the rather simple load limits included in current regulations, more complex criteria are proposed, particularly regarding "whiplash" injury. Even criteria used to assess the same type of collision focus on different effects, thus the assessment of an impact based on one criterion may be useful; yet it might not be sufficient. Different criteria relate to different phases and, by their definition, emphasise different aspects of the occupant motion. Hence, neck injury criteria reveal important information which can be used to describe injury risk, but sometimes also allow conclusions about associated subjects like seat design or the injury mechanism.

In general, it is important to note that injury criteria are restricted to the conditions specified in their definitions. Application to other conditions, for example to other impact directions, has to be addressed carefully. Adjustments in the test procedure and/or the evaluation and the interpretation of the results obtained might be necessary. This holds, of course, also true for the choice of the anthropomorphic test devices. As described in Sect. 2.6.1, the designs of a dummy, and especially of a dummy neck, are very different. Therefore, when assessing neck injury by determination of neck loads and injury criteria, the influence of the dummy is always inherent (e.g. Muser et al. 2000, 2002; Bortenschlager et al. 2007). In this context it should be pointed out that there is currently no rear impact dummy available representing a female. Likewise specific injury criteria addressing the risk of female occupants are lacking, although the high incidence rate of soft tissue neck injuries in females is established. First steps towards female injury criteria and a female rear-end dummy were undertaking in recent years, but are not yet concluded (e.g. Schmitt et al. 2012; Carlsson et al. 2012b; Linder et al. 2013).

With respect to neck injury, the neck injury criteria NIC (Boström et al. 1996), N_{ij} (Klinich et al. 1996; Kleinberger et al. 1998), and N_{km} (Schmitt 2001; Schmitt et al. 2002a) are often used. While N_{ij} was designed to detect severe neck injuries in frontal impact, the other two were developed with regard to soft tissue neck injuries in rear-end impacts. Work by Kullgren et al. (2003), Muser et al. (2003) and Davidsson and Kullgren (2011) show that NIC and N_{km} correlate well with the risk of AIS1 neck injury sustained in rear-end collisions.

Consequently, these criteria are also included in the recently introduced "whiplash" related seat test procedures by Euro NCAP (see Sect. 2.6). Although the biomechanical foundation of this Euro NCAP assessment scheme is weak in several aspects (e.g. Bortenschlager et al. 2007; Farmer et al. 2008; Schmitt and Muser 2009), a correlation between such consumer tests and real-life outcome is not necessarily excluded; Kullgren et al. (2010) did in fact find a good correlation between Euro NCAP results and real-world crash and injury ratings. Further studies have also shown that cars fitted with more advanced whiplash protection systems performed better in dynamic testing and had a lower injury risk in real-life (e.g. Kullgren et al. 2007; Farmer et al. 2008).

A general limit of injury criteria is the fact that they can be determined under controlled conditions, i.e. in experiments, only. Real world crashes cannot be assessed retrospectively through those criteria, because there is no possibility to measure the neck loads. With respect to soft tissue neck injuries, this poses a problem as those cases often result in legal procedures requiring an assessment by an expert witness to clarify the likeliness whether the individual injury claimed is causally linked to an accident. Neither tests using crash test dummies nor purely statistical approaches are sufficient to assess an individual case. Therefore special schemes were developed to biomechanically assess this causality (e.g. Walz and Muser 2000; Schmitt et al. 2002b, 2003a). The change of velocity (delta-v) of the vehicle under question can be determined by accident reconstruction and can thus serve as an estimate of the impact severity which is then related to the injury risk. For frontal and lateral impacts, injury thresholds ranging from a minimum delta-v of 16–20 km/h are found in the literature (e.g. Ferrari 1999; Kornhauser 1996; Watts et al. 1996; Kullgren et al. 2000). For rear-end collisions, delta-v values of 8–15 km/h are reported (e.g. Ferrari 1999; Schuller 2001). However, as pointed out by various researchers, use of delta-v is an approximation; it is not sufficient to take into account solely the delta-v value, other vehicle specific factors such as the vehicle acceleration, stiffness as well as the individual physique of an occupant have to be considered when assessing "whiplash" injury. The wider use of event data recorders in vehicles which record the vehicle dynamics during a crash allow more precise determination of parameters to characterize the impact (such as the vehicle acceleration) and thus improve the basis for the assessment of specific collisions.

4.4.1 Neck Injury Criterion NIC

Assuming that pressure gradients caused by a sudden change of the fluid flow inside the fluid compartments of the cervical spine are related to neck injuries, the neck injury criterion NIC was developed by Boström et al. (1996). The definition of the NIC as a function of time was validated based on animal experiments. A relation to predict injury caused by pressure gradients (Eq. 4.1) was found between the acceleration in the anterior-posterior direction (i.e. x-direction when using SAE J211/2) of the centre of gravity of the head relative to the first thoracic vertebra (T1) and the velocity derived thereof.

$$NIC(t) = 0.2a_{rel}(t) + v_{rel}(t)^2 \tag{4.1}$$

The threshold value above which a significant risk of sustaining minor (AIS1) neck injury is assumed to be inherent was set to be 15 m^2/s^2. This value has served well in accident logical studies and is still used. However, it has emerged that reasonable values are only obtained for the retraction phase of a rear-end impact, i.e. when, in a vehicle fixed reference system, both acceleration and velocity are directed backwards. In addition, it turned out that a considerable error is introduced to the NIC(t)-curve as soon as the head is no longer parallel to T1, i.e. the head extension angle reaches values of about 20–30°. Thus, the NIC_{max} was introduced, which indicates that the maximum value of the NIC(t)-curve found within the time interval between the beginning of the collision and the point in time where the head, relative to the neck, reverses its direction of motion.

A modification of the NIC for low-speed frontal impact—called $NIC_{protraction}$—has been proposed (Boström et al. 2000) and was related to long-term AIS1 neck injuries, i.e. for AIS1 injuries with symptoms for more than 6 months. As a threshold for 50 % injury risk, 25 m^2/s^2 was proposed. Bohmann et al. (2000) reduced this value to 15 m^2/s^2 extending the injury assessment to short-term and long-term consequences. The following equations are used to determine $NIC_{protraction}$:

$$NIC_{generic}(t) = 0.2a_{rel}(t) + v_{rel}(t)|v_{rel}(t)| \tag{4.2}$$

$$NIC_{protraction}(t) = \left|Min(NIC_{generic}(t))\right| \tag{4.3}$$

4.4.2 N_{ij} Neck Injury Criterion

This criterion was proposed by the US National Highway Traffic Safety Administration (NHTSA) (Klinich et al. 1996; Kleinberger et al. 1998) to assess severe neck injuries in frontal impacts, including those with airbag deployment and thus accounting for more severe impact conditions at higher Δv. Recently, the N_{ij} criterion was included as part of FMVSS 208.

Table 4.4 Intercept values for calculating N_{ij} as included in FMVSS 208

Dummy	M_y (flexion/extension) [Nm]	F_z (compression/ tension) [N]
HIII 50 %	310/135	6160/6806
HIII 5 %	155/67	3880/4287
HIII 5 % (out of position)	155/61	3880/3880
HIII 6 year	93/37	2800/2800
HIII 3 year	68/27	2120/2120

The underlying concept for the N_{ij} can be found in a study by Prasad and Daniel (1984) who performed crash tests using piglets as child surrogates. As a result with respect to neck injuries, they suggested to combine axial forces with moments for a composite neck injury indicator. The N_{ij} criterion developed implies a linear combination of the axial forces and the flexion/extension bending moment, both normalised by critical intercept values:

$$N_{ij} = \frac{F_z}{F_{\text{int}}} + \frac{M_y}{M_{\text{int}}} \tag{4.4}$$

where F_z and M_y are the axial force and the sagittal bending moment, respectively. F_{int} and M_{int} indicate the according critical intercept values. These intercept values were established and validated for a three-year-old dummy. Scaling techniques were used to obtain the according intercept values for other dummy sizes and thus making the N_{ij} eligible for those dummies, too. The actual intercept values as suggested by the NTHSA are shown in Table 4.4. Hence, evaluating the criterion for all possible load cases, four different values are obtained: N_{te} for tension and extension, N_{tf} for tension and flexion as well as N_{ce} and N_{cf} giving analogue values for compression. An injury threshold value of 1.0 applies for each load case.

Adopting the N_{ij} to analyse the effect of deploying side airbags, Duma et al. (1999) replaced the sagittal bending moment by the total bending moment.

To assess AIS1 neck injury, reduced threshold values of 0.2 and 0.16 for long-term and short-term injury, respectively, were proposed (Boström et al. 2000; Bohmann et al. 2000).

However, evaluating the N_{ij} in its original form for rear-end collisions (for which it was not designed) produced difficulties in the interpretation of the results obtained (Linder et al. 2000). Therefore a modification of the N_{ij} criterion—called the N_{km}—which is suitable for the assessment of low-speed rear-end collisions was developed.

4.4.3 Neck Protection Criterion N_{km}

The neck protection criterion N_{km} (Schmitt 2001; Schmitt et al. 2002a) is based on the hypothesis that a linear combination of loads and moments describes best the relevant neck loading. A similar approach led to the definition of the N_{ij} criterion for frontal impact (Sect. 4.4.2) and thus the N_{km} can be regarded as a modification thereof.

However, with respect to possible injury mechanisms in rear-end collisions, sagittal shear forces rather than axial forces are regarded as the critical load case. A combination of shear and the sagittal bending moment accounts for a constellation often found in the cervical spine also during S-shape formation (e.g. Deng et al. 2000). To date, the S-shape formation is mainly associated with the retraction phase, but, looking at the kinematics, an opposite S-shape, i.e. the torso lagging behind the head, could result in a similar deformation and therefore also incorporate an injury risk (Boström et al. 2000). This opposite S-shape which can for instance be observed during the rebound phase is not assessed by the maximum NIC due to its limitations mentioned above.

Furthermore, it is assumed that shear forces could potentially be harmful to the facet joints, in particular in the upper neck region (Yang et al. 1997; Deng et al. 2000; Winkelstein et al. 2000). Although the actual injury mechanism is unknown, the load cases of shear and extension/flexion moment seem to be relevant for neck injuries. Therefore the N_{km} does not address a single injury mechanism but takes into account a potential injury risk caused by the combination of loads and moments.

To combine shear and moment linearly seems straightforward, as for the calculation of a resulting load on a certain structure of the neck a linear combination of the existing forces and moments follows the understanding of simple mechanics. Additionally, the interpretation of the N_{km} results becomes more obvious when implementing a linear combination—a practical consideration which is important for the use of the criterion.

In the human, axial compression/tension forces are considered to influence the amount of shear (Yang et al. 1997) and are as such included. However, difficulties arise when measuring such axial forces. Performing crash tests using an ATD with standard instrumentation and an additional load cell at the upper neck position, the latter will measure the occurring axial forces. Different reasons causing an inaccuracy in such measurements can be identified:

- the centripetal force from rotation of the dummy around the pelvis is measured as an axial force.
- due to the fact that most dummies of today (except the BioRID dummy) do not represent the thoracic kyphosis, they consequently do not account for compression forces resulting from the straightening (ramp-ing) effect.
- recent dummy designs do not allow physiological backward movement of the head during the retraction phase, i.e. when the torso is pushed forward by the seat back, the head is expected to lag behind due to its inertia. However, as the neck is connected with joints to the head and to the torso, a rotation of the head is originated which creates axial forces (although the head is intended to move relative to the torso without any rotation in the sagittal plane).
- in cases where the head in extension reaches above the head restraint and causes it to be pushed into a lower position ("hammer effect"), axial forces from this extension are also measured.

Table 4.5 Intercept values for calculating N_{km}

Load case	Value	Reference
Extension	47.5 Nm	Goldsmith and Ommaya 1984
		Mertz and Patrick 1993
Flexion	88.1 Nm	
Negative and positive shear	845 N	

Due to these inaccuracies concerning the measurement of the axial forces, they were not explicitly included in the N_{km}. Hence, the N_{km} criterion was defined according to the following equation:

$$N_{km}(t) = \frac{F_x(t)}{F_{\text{int}}} + \frac{M_y(t)}{M_{\text{int}}} \tag{4.5}$$

where $F_x(t)$ and $M_y(t)$ are the shear force and the flexion/extension bending moment, respectively; both values should be obtained from the load cell positioned at the upper neck. F_{int} and M_{int} represent critical intercept values used for normalisation.

Distinguishing positive shear, negative shear, flexion and extension, the N_{km} criterion identifies four different load cases: N_{fa}, N_{ea}, N_{fp} and N_{ep}. The first index represents the bending moment (f: flexion, e: extension) and the second indicates the direction of the shear force (a: anterior, i.e. in positive x-direction, p: posterior, i.e. in negative x-direction). The sign convention according to SAE J211/2 was used. Consequently, positive shear forces measured at the upper neck load cell indicate that the head is moved backwards relative to the uppermost cervical vertebra.

The intercept values used to calculate the criterion are shown in Table 4.5 which exhibits the human tolerance levels for the causation of AIS 1 injuries (Goldsmith and Ommaya 1984). These values were determined on the basis of volunteer experiments (Mertz and Patrick 1993) and suggest tolerance levels up to which no injury is expected. The tests revealed no difference for the maximum shear tolerated in anterior and posterior direction. Sled tests as well as computational simulations were used (Schmitt et al. 2002a) to validate the proposed criterion.

For computation of the N_{km} values, the load curves measured are divided by the according intercept values, then the bending modes and the load types under investigation are identified. Finally the N_{km} values are obtained by adding the adequate shear force and moment curves, while keeping the time scale unchanged, and determining the maximum of the resulting curve.. Hence, the N_{ep} for instance represents the maximum value in time when extension and negative shear occur simultaneously. If a certain combination of loads and moments is not observed within the time interval analysed, the N_{km} quadruples may be incomplete.

With regard to a critical N_{km} value, 1.0 was used taking into account that either a moment or a shear force exceeding the intercept value produces a risk of sustaining neck injuries.

To date, the N_{km} has shown its usefulness to assess low speed rear-end collisions in various tests. The criterion is included in seat test procedures such as those used by Euro NCAP to assess the risk of soft tissue neck injuries in new vehicles. In particular, it was shown that N_{km} values allow for the characterisation of the crash phase of forward movement and as such the N_{km} gives additional information to that gained by the NIC_{max}, which accounts for the earlier phase only. As for the correlation of the N_{km} and the risk of sustaining neck injuries, Muser et al. (2003) found the N_{ea} load case to be the strongest predictor. Furthermore, it was shown that N_{km} values are capable to quantify different characteristics of seat design (Muser et al. 2002). With respect to the ongoing discussion about the design principles for improved car seats, i.e. the conflict on allowing deformation (plasticity) versus elasticity (Parkin et al. 1995), the N_{km} was found to be a helpful tool since minimising both values simultaneously indicates a balanced seat design.

4.4.4 Lower Neck Load Index

A further neck injury criterion to assess the risk of soft tissue neck injuries, called Lower Neck Load Index (LNL) was proposed by Heitplatz et al. (2003). The LNL takes into account three force components and two of the moment components measured at the base of the neck (Eq. 4.7). Hence, evaluation of this criterion requires a dummy that is equipped with a lower neck load cell.

$$LNL(t) = \left|\frac{\sqrt{(My_{lower}(t))^2 + (Mx_{lower}(t))^2}}{C_{moment}}\right| + \left|\frac{\sqrt{(Fy_{lower}(t))^2 + (Fx_{lower}(t))^2}}{C_{shear}}\right| + \left|\frac{Fz_{lower}(t)}{C_{tension}}\right| \tag{4.6}$$

where *Fi(t)* and *Mi(t)* are the force and moment components, respectively. The denominators represent intercept values which are proposed to be $C_{moment} = 15$, $C_{shear} = 250$ and $C_{tension} = 900$ for a RID dummy (Heitplatz et al. 2003). For other dummies intercept values are not yet proposed.

With respect to rear-end collisions, the definition of the LNL becomes very similar to the N_{km} definition, apart from the additional term for the tension force and the fact that the data is recorded at the lower neck load cell. To date experience with the LNL is very limited. At the current stage, the LNL also comprises shortcomings such as no established biomechanical connection to a possible injury mechanism and no correlation to real world injury outcome (Bortenschlager et al. 2003).

4.4.5 Neck Injury Criteria in ECE and FMVSS

Current regulations specify maximum spinal loads for frontal impact (ECE R94, FMVSS 208). For low speed rear-end impact there are no homologation tests defined.

ECE R94 requires the neck extension moment not to exceed 57 Nm. Furthermore, the shear forces and the axial tension force measured should be below the values indicated in Fig. 4.16.

The current FMVSS 208 includes injury criteria for the neck, consisting of individual tolerance limits for compression, tension and four N_{ij} components (Table 4.6). The tolerance values are based on volunteer, cadaver and dummy tests and apply to the 50th percentile male and the 5th percentile female.

4.4.6 Further Neck Injury Criteria

Assuming that neck pain sustained in a rear-end collision is caused by an intervertebral rotation exceeding the limit of physiological intervertebral motion, Panjabi et al. (1999) proposed the intervertebral neck injury criterion (IV-NIC). The IV-NIC is defined as the ratio of the intervertebral motion under traumatic loading θ_{trauma} and the physiological range of motion θ_{physio} (Eq. 4.7). The criterion is defined for each intervertebral joint i and is calculated separately for flexion and extension.

$$IV - NIC_i = \frac{\theta_{trauma,i}}{\theta_{physio.i}} \tag{4.7}$$

Hence the maximum IV-NIC value identifies the time, location and bending mode of the maximal intervertebral rotation and for values greater than 1.0, it indicates that the physiological range is exceeded.

The IV-NIC is still neither validated nor is there a threshold value proposed. Due to the fact that in all dummy types, pin joints are used to connect the vertebrae, the intervertebral motion cannot be mimicked, and therefore the evaluation of the IV-NIC is impossible in ATD experiments. Difficulties also arise in defining the physiological range of motion, which was, in the study by Panjabi et al. (1999), solely defined on the basis of a single human cadaver specimen.

The neck displacement criterion (NDC) has been proposed to assess the risk of soft tissue neck injury (Viano and Davidsson 2002). It addresses the S-shape of the neck by taking into account the extension moment, displacement in z (axial) direction and displacement in x (sagittal) direction. By plotting the head rotation versus the x-displacement and plotting the z-displacement versus the x-displacement, two NDC diagrams are obtained. Sled tests utilising volunteers, the BioRID and the Hybrid III dummy were performed to define tolerance corridors for the NDC diagrams. However, these corridors cannot yet be regarded as definitely set. A study presented by Kullgren et al. (2003) concluded that the NDC does not

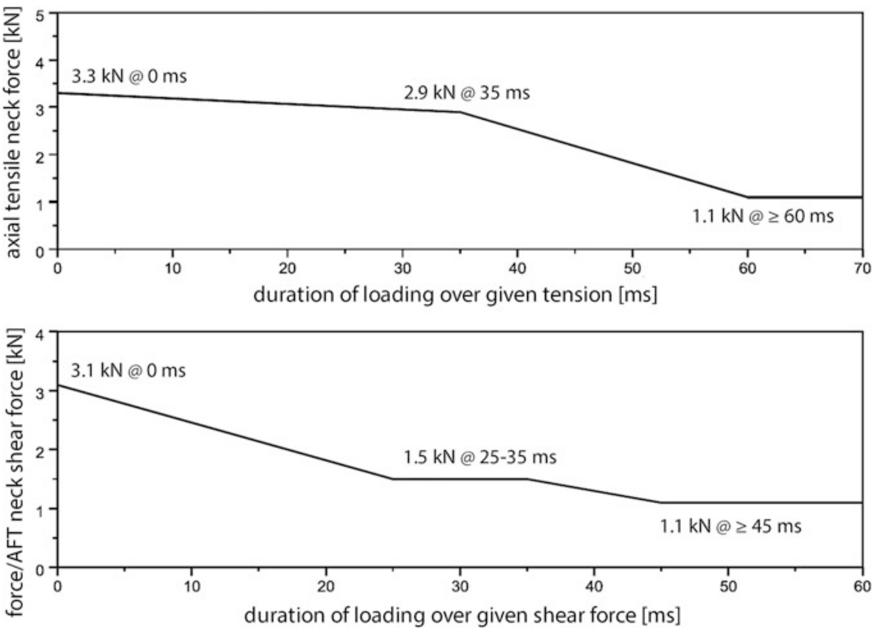


Fig. 4.16 Duration dependent limits for neck forces specified in ECE R94. *Top* tension, *bottom* shear

Table 4.6 Threshold values for neck load included in FMVSS 208

Load case	Threshold
Flexion	190 Nm
Extension	57 Nm
Axial tension	3300 N
Axial compression	4000 N
Shear (Anterior and posterior)	3100 N

correlate well with the real life risk of sustaining "whiplash" injury. No additional substantial work concerning the NDC is published yet and the criterion is rarely used.

Taking into account the head-to-torso rotation as a criterion for soft tissue neck injury was proposed by Kuppa et al. (2005). Magnetohydrodynamic angular rate sensors were used to determine the criterion in rear impact tests using a Hybrid III dummy.

Munoz et al. (2005) suggest a whiplash injury criterion (WIC) which considers the upper and lower neck moment around the y-axis (extension/flexion). Based on experiments using a BioRID, WIC was defined as the difference between the y-moment around the occipital condyles and the y-moment determined at the lower neck (T1) load cell. The criterion is meant to be related to the S-shape. So far, however, the criterion did not receive much attention.

4.4.7 Correlating Neck Injury Criteria to the Injury Risk

As to date no AIS1 neck injury mechanism is clearly identified, it is difficult to validate the neck injury criteria proposed. Validation methods that are not related to an injury mechanism must be used to investigate the strength with which an injury criteria correlates with the real world injury risk. The information gained from such correlation analyses can then also be used for designing crash test procedures with which the risk of sustaining neck injury can be assessed.

Exemplary two studies using different methodologies are presented here that investigated the predictive quality of neck injury criteria with respect to AIS1 neck injury. Until today further studies of these kinds were conducted using more recent data and also similar studies for other body regions but the neck can be conducted by comparable methodologies.

A study by Kullgren et al. (2003) aimed at validating different proposed neck injury criteria with reconstructed real-life crashes of vehicles that were equipped with crash pulse recorders. A large car fleet of more than 40,000 vehicles fitted with crash pulse recorders has been monitored for several years and all crashes with these cars, irrespective of repair cost and injury outcome were reported. The vehicle crash data is therefore available for each case (e.g. in terms of vehicle acceleration). The results can be expected to be more precise as if the accident was reconstructed retrospectively. In addition the medical outcome of all occupants was recorded, i.e. there were also cases in which a crash happened, but no neck pain was claimed. These cases served as a control group for the analysis. In contrast, other studies that focus on patients only, lack a control group of healthy occupants.

As a result of this type of studies, the injury risk can be described in different ways. Injury risk curves can, for example be developed. Concerning vehicle acceleration as an injury predictor, Kullgren et al. found that below 5 g mean acceleration, the risk to sustain a long-term neck injury (i.e. the symptoms last longer than one month) appeared to be very low. At a mean acceleration above 7 g the risk seems to approach 100 %. No one was, however, observed to have symptoms for more than one month as long as the mean acceleration was below 3 g.

In a further step, the crash recorder data served as a basis for a numerical simulation study in which the seats of some car models were exposed to the recorded crash pulses. A model of a BioRID was used to represent the occupant. The dummy readings were correlated to the real-life injury outcome. The effectiveness to predict AIS1 neck injury was assessed for different neck injury criteria such as NIC_{max}, N_{km}, NDC and the lower neck moment. Consequently it was possible to link data from real-life accident and injury outcome to measures that can only be determined in testing or simulation. The results indicated that NIC_{max} and N_{km} are applicable to predict risk of AIS1 neck injury when using a BioRID as human surrogate. Further statistical analysis indicated that also a combination of both criteria has a high predictive power with respect to injury risk.

The second study was performed by Muser et al. (2003). Here the correlation of neck injury criteria and the real world injury risk was assessed on the basis of results from sled test experiments and an accident data base of a large insurance company. The data base was used to develop a seat performance list, i.e. seat that perform well (few cases reporting neck pain after a crash) and poor were identified. The sled tests were performed with various seats from car models that performed good/poor according to the real-life insurance data. Two different anthropomorphic test devices were used: a BioRID and a RID2. After evaluating the sled tests, the results obtained were correlated with the risk to sustain whiplash injury as determined from an accident data base. The correlation between the dummy measures and the real-life accident data was evaluated for each dummy type and for each parameter analysed. With this method it was possible to predict the protective potential of a specific seat system. As a result N_{km}, when applied with a BioRID dummy, showed the strongest correlation to the injury risk. Also NIC_{max} showed in all cases a positive correlation between injury risk and values measured in the sled tests. In summary, this study combined real-life data and laboratory experiments to derive a correlation between biomechanical measures in terms of injury criteria and the actual injury risk.

4.5 Spinal Injuries in Sports

Spinal injuries that result from sports accidents comply with the same principles as mentioned above. Additionally, direct blows to the spine are observed. Muscle strains and sprains of the ligamentous structures may be the most common (minor) neck injuries in sports. Also compression fractures are frequently reported athletic injuries. Fractures from repetitive loading such as sacral stress fractures (as reported almost exclusively in high-level running sports) are, in contrast, quite rare (cf. Chap. 9).

The cervical spine is the most vulnerable part; often the underlying injury mechanism is a compression-flexion mechanism (Fig. 4.9). In neutral position the cervical spine exhibits an extension due to normal lordosis. When flexing the neck (to approx. 30°), the cervical spine straightens. If a force is now applied to the vertex, the load is transmitted along the longitudinal axis of the cervical spine without much energy being dissipated by the paravertebral muscles. Hence the cervical spine is compressed between head and torso. Fracture, luxation or dislocation may result. Examples for such a mechanism include headfirst techniques in American football and contact sports as well as diving accidents. Diving into shallow water, often in conjunction with a head impact, can result in compression induced neck injury. Commonly such cervical axial compression injuries are observed with the cervical spine between C5 and C7 level being at particular risk (Aito et al. 2005; Boden and Jarvis 2008; Wennberg et al. 2008).

Fortunately, catastrophic cervical spinal cord injuries are relatively uncommon during athletic activities. However, they are much more likely to result than thoracic or lumbar trauma. Instable fractures and dislocation are frequent causes for catastrophic cervical spinal cord injuries often resulting in permanent neurological sequelae. Usually the injury is observed in the lower cervical spine (e.g. McIntosh and McCrory 2005; Boden and Jarvis 2008).

Stinger and transient quadriplegia/paresis, in contrast, are more frequent injuries that have a wide spectrum of clinical severity and disabilities (e.g. Vaccaro et al. 2002). Episodes of transient quadriplegia related to the cervical spinal cord are reported whereas the episode is usually followed by complete recovery occurring in ten to 15 min, but sometimes taking up to 2 days (e.g. Torg et al. 2002). Neuropraxia is classified according to the type of neurological deficit. The term plegia is used for episodes with complete paralysis; paresis for episodes with motor weakness and paresthesia for episodes that involved only sensory changes without motor involvement.

In athletes with diminution of the antero-posterior diameter of the spinal canal, the cord can, on forced hyperextension or hyperflexion, be compressed, causing such transient motor and sensory manifestations. The mechanics of cervical spinal cord compression were described by Penning (1962) as “pincer mechanism”. Pavlov et al. (1987) devised the measurement of the spinal canal to vertebral body ratio to determine whether an athlete has a narrow spinal canal and therefore exhibits a higher risk for cord compression. The spinal canal to vertebral body ratio is described by the distance from the midpoint of the posterior aspect of the vertebral body to the nearest point on the corresponding spin laminar line divided by the antero-posterior width of the vertebral body. Normally, regardless of gender or age, the spinal canal to vertebral body ratio is close to one. A ratio of smaller or equal 0.8 was recorded at one or more levels in patients who experienced cervical cord neuropraxia.

With respect to thoracic and lumbar segments of the spine, similar injury patterns as for the cervical spine can be observed. Compression fractures of the lumbar vertebrae are, for instance, also reported from snow sports (e.g. Yamakawa et al. 2001; Franz et al. 2008). In addition, low-back pain is a symptom for which a life time prevalence in the general adult population of 85–90 % is estimated (Trainor and Wiesel 2002). Hence, also athletes suffer from low-back pain although it is not clear whether they are at higher risk. Some studies suggest that for certain athletes (like wrestlers and elite gymnasts) there might be a higher risk, but results are not yet conclusive. Various risk factors are investigated taking into account lumbar flexibility, lower-extremity function or the footwear used. So far, a history of low-back pain was found to be the greatest predictor of future occurrences of low-back pain in athletes (e.g. Bono 2004).

Participation in sports appears to be a risk factor for the development of disc degeneration with disc degeneration being influenced by the type and intensity of the sport (e.g. Sward et al. 1991). The prevalence of spondylolysis in athletes was not found to be higher than in the general population. However, some studies suggest that there seems to be a higher prevalence in some sports like weight lifting, diving, wrestling (Bono 2004).

4.6 Prevention of Soft Tissue Neck Injury

Since causes for neck injuries sustained in sports are manifold depending, for instance, on the type of sports, the condition of the athlete and the actual situation, it is hardly possible to apply a general strategy for prevention. For some sports like football neck collars are available to reduce neck load (Rowson et al. 2008). However, the applicability of such devices is limited in certain sports since the reduction of load transmission correlates with a restriction of the head and neck motion.

In the automotive environment soft tissue neck injuries are a major concern. Therefore, vehicle seat design aims at providing seats that offer good "whiplash" protection. This proves to be a rather difficult task, given that the underlying injury mechanisms are not known. In a holistic approach to prevent "whiplash" injury, general design guidelines for seat development were established (Walz and Muser 1995; Lundell et al. 1998). Such guidelines attempt to address all existing hypotheses concerning the injury mechanism by minimising relative motion between head and thorax and thus reducing all kinds of biomechanical loading that might cause injury. There is a certain risk in this approach that much time and effort might be spent in reducing certain neck loads that are not responsible for WAD at all, or in reducing loads that were sub-critical already before the improvement process started. The societal impact demands, however, that measures be taken despite this risk, since it might still take a long time for researchers to solve the biomechanical problems associated with soft tissue neck injury.

It is assumed that without relative acceleration between head and torso, no soft tissue neck injury will be sustained. Bearing in mind that the neck injury criterion NIC is widely used to assess the risk of soft tissue neck injuries, it is especially this relative acceleration that has to be reduced to obtain good NIC values. Taking into account hypotheses that claim the relative movement between adjacent vertebrae to be causal for "whiplash" injury, such motion must be avoided. Hence, the curvature of the spine must be kept unchanged during the impact. Additionally, the rebound phase has to be considered. To minimise the interaction with the seat belt, rebound must be reduced.

The potential of a seat to prevent soft tissue neck injury is assessed by performing experiments where acceleration, forces, and moments of torque and determining various neck injury criteria thereof are measured. To ensure a broad basis for analysis and assessment, none of the measures mentioned can account for all factors alone, but all should be reduced. Due to the uncertainty with respect to the injury mechanism, an increase of any response related to the biomechanical guidelines—even if accompanied by a clear reduction of another criterion—is to be avoided.

Until today several seat systems to prevent "whiplash" injury were presented, some of which were introduced on the market. Further measures like the introduction of a specific seat test procedure as part of the Euro NCAP assessment also helped to increase awareness and to stipulate the optimization of seat design. Several studies suggest that such systems have the capability to significantly reduce the risk of sustaining soft tissue neck injury (Boström and Kullgren 2007; Kullgren et al. 2007; Farmer et al. 2008). However, it was also shown that male occupants benefit much more from current activities to improve seat design than females. Partly this can be explained by the fact that there is no female dummy available that can be used for testing, so far a computer model of a female dummy was presented which can be used for corresponding simulations (Carlsson et al. 2012b). Further endeavours are needed to prevent optimization for males only.

Basically all major structures of a vehicle seat such as the head restraint, the seat back including the recliner, as well as the seat base and the seat slide serve as the basis for the development of different "whiplash" protection devices. The following sections summarise some of the advancements. However, it must be stated that the future direction of such developments is currently not clear. The development of active safety systems to avoid collisions (from warning systems to autonomous braking systems) and their broad introduction into the vehicle fleet appears to be of higher priority to some as improving the seat design by means of passive safety.

4.6.1 Head Restraint Geometry and Padding Material

Head restraints, originally introduced to prevent severe injuries to the neck related to hyperextension, may also prevent the sudden relative movement of head versus torso (cf. S-shape). This protection potential can only be exploited if the head restraint is positioned correctly.

The influence of the head restraint geometry on the protective potential of the head restraint was investigated in several studies. A decrease in "whiplash" injury incidence was found with increasing head restraint height (e.g. Eichberger et al. 1996; Hell et al. 1998; Ferrari 1999). Similarly it was found that the head to head restraint distance is associated with injury potential (e.g. Wiklund and Larsson 1998; Ferrari 1999; Hofinger et al. 1999). The shorter the head to head restraint distance, the more effectively the S-shape is prevented.

Not only the geometry, but also the inner structure, in particular the padding material, might be modified to prevent soft tissue neck injury. The potential benefit of different foam material such as automotive visco-elastic foam that absorbs energy was demonstrated in different studies (e.g. Szabo et al. 2002; Schmitt et al. 2003a).

Controlling the occupant kinematics by using optimised seat back geometry together with a carefully adjusted foam stiffness distribution along the seat back and the head restraint might well prove a viable way of injury protection without any additional technical effort; such an approach was used in the so-called WIL (Whiplash Injury Lessening) system (Sekizuka 1998; Sawada and Hasegawa 2005).

4.6.2 Controlling Head Restraint Position

Theoretically, according to the 'no relative motion' paradigm presented above, WAD injuries could be avoided by simply ensuring a zero distance between head and head restraint during the collision. Several ideas have been proposed to achieve this goal. Muser et al. presented in 1994 a head restraint equipped with capacitive sensors and electrical actuators, always keeping the head restraint at a pre-programmed distance. Mainly due to cost considerations, such a system was not introduced in the market at that time, but is available today in luxury cars. Based on a similar head restraint design Matsubayashi et al. (2007) suggest to add a radar system to detect unavoidable rear-end impacts. If there is a high risk of impact, the systems warns the vehicle occupant and ensures that the head restraint is moved closer to the head.

The SAHR system represents a self-aligning head restraint (Wiklund and Larsson 1998). It is a re-active head restraint system, i.e. it moves the head restraint upwards and closer to the occupants' head during the impact. Thus the distance between head and head restraint is reduced only when needed. The rearward motion of the torso towards the seat back is used to load a plate, which in turn is connected through a lever to the head restraint. This seesaw mechanism rotates the head restraint forward, resulting in earlier head contact, and upward with respect to the occupant. Various studies demonstrated the ability of the system to prevent "whiplash" injury (e.g. Viano and Olsen 2001; Muser et al. 2002). The SAHR system was one of the first whiplash protection systems to be included in serial production in 1997. Since then, various other, similar systems have been presented, and the design principle has found widespread use.

Various systems that move the head restraint forward during the collision without relying on the force interaction of the occupant to the seat back were also presented. Some approaches make use of more or less classical airbags fitted in the head restraint, albeit inflated much slower by pressurised air. Alternatively pre-tensioned springs were implemented as an energy source. It is often argued that these active systems offer the advantage over the SAHR-types that, with the latter, an occupant having a lower body mass than e.g. the 50 % male may not be able to move the head restraint far enough forward to be of any use at all. Preliminary results of seat tests using a prototype female dummy did not confirm this concern (i.e. the female dummy managed to activate the SAHR system), but further endeavours are needed to investigate and improve the interaction of such anti-whiplash systems and different dummy anthropometries (Schmitt et al. 2012).

Aftermarket devices, essentially consisting of cushions placed between the head restraint surface and the head, also serve to reduce head to head restraint distance. Such systems were available in the 1990s, but are hardly found today.

4.6.3 Controlling Seat Back Motion

Another approach is based on the insight that not the relative motion as such, but rather its 'violence' in terms of relative velocity and acceleration, generates the risk of WAD injury. It should therefore be possible to lower injury risk by e.g. damping devices in the seat back that, first, lower the acceleration at the upper torso level and, consequently, also the relative velocity between head and torso.

The WHIPS seat (Lundell et al. 1998) is equipped with a recliner that allows controlled backward movement of the seatback during rear-end impact. If a critical load is exceeded, the motion is performed in two steps: a translational rearward movement of the seatback is followed by a rotational motion reclining the seatback. Advantageous neck injury criteria values were observed for the WHIPS seat tested under low-speed rear-end conditions (e.g. Hell et al. 1999; Muser et al. 2000; Langwieder et al. 2000). The WHIPS system is also among the first systems that appeared on the market, allowing a statistical analysis of its efficiency in preventing WAD (Jakobsson and Norin 2004; Jakobsson 2005; Jakobsson et al. 2008).

Similar effects may also be achieved by interfering with other parts of the seat. Assuming that the relative acceleration between head and T1 is to be reduced to prevent whiplash injury, the seat slide was modified such that it allows controlled translational motion of the seat relative to the car while damping this motion. This leads to a delay in the torso acceleration, thereby synchronising the loading of the head and the upper torso (Schmitt et al. 2003c). The system WipGARD (Zellmer et al. 2001), an aftermarket device which can be fitted into certain Volkswagen models, is mounted between the seat slide and the floor. Like WHIPS, WipGARD enables the seat back to perform a translation followed by a rotation once a critical load is reached.

4.7 Summary

The cervical spine is the most vulnerable part of the spine in sports as well as in automotive accidents. Particularly soft tissue neck injuries are of major concern in automotive safety. Although the injury mechanism is not yet fully established, several advances have been undertaken to reduce the injury risk by improving seat design. Consumer tests encourage such seat design by introducing schemes to assess and rate the neck injury risk. In that respect the correlation of such dynamic tests to the real-life injury risk is crucial. Today several injury criteria are suggested of which NIC and N_{km} are most widely used. However, future research is needed to improve the biomechanical foundation, for example, regarding the higher injury risk of females.

4.8 Exercises

E4.1: Compression of the spine can result in injury. Give examples of different loading scenarios involving compression. Which injuries can result thereof?

E4.2: Which scale is suitable to classify the degree of neck injury/neck pain after a rear-end collision?

E4.3: Imagine a rear-end collision. Describe the motion of an occupant sitting in the car struck from behind. How does the seat belt interact with the occupant motion?

E4.4: In American football head first impact can occur. How is the risk of neck injury influenced by the degree of neck flexion at the moment of impact?

P4.1: Discuss whether a static assessment of a seat (e.g. the determination of the head-to-head restraint distance) is sufficient to determine the soft tissue injury risk or whether the performance of dynamic tests is mandatory.

P4.2: Standard EN1621-2 describes a procedure to test motorcycle back (spine) protectors. In the prescribed experiments the protector is basically impacted by a wedge-shaped drop weight of 5 kg such that the kinetic energy at impact is 50 J. The standard is also often applied to assess the protective potential of back protectors used in sports (e.g. snowboarding). Discuss which spinal injuries are addressed by the standard and whether it should be applied for both motorcycle and sports protectors.

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Thoracic Injuries

5

Injury to the thorax commonly occurs in impact from the front and the side as well as in all impact directions intermediate to these two. Impact to the thorax is frequently observed due to contact, for example, with various components of the vehicle interior (like the steering assembly, safety belt, the door or the dashboard). In case of sports, thorax impacts may be due to contact with an opponent player (soccer, ice hockey) or due to direct blows (boxing, taekwondo).

Most thorax injuries caused by contact mechanisms are caused by blunt impact. In automotive accidents, sharp impact to the thorax is rare, occurring only due to obstacles inside the passenger compartment or when the occupant is ejected from the vehicle. Consequently, this chapter focuses on blunt impact.

5.1 Anatomy of the Thorax

The thorax consists of the rib cage and the underlying soft tissue organs. It extends from the base of the neck to the diaphragm which inferiorly bounds the thorax and separates the thoracic cavity from the abdominal cavity (Fig. 5.1).

The rib cage is formed by twelve pairs of ribs which are posterior connected to the thoracic vertebrae of the vertebral column. At the anterior side of the thorax the sternum fixes the upper seven ribs. The lower ribs are either connected indirectly to the sternum or are attached to muscles and the abdominal wall (so-called floating ribs). The ribs are interconnected with each other by the internal and external intercostal muscles. As the connections of the ribs to the vertebrae, the intercostal muscles, and the sternum are flexible, the rib cage represents a quite stiff though deformable cover of the internal organs and facilitates respiration.

While the rib cage is very flexible in a newborn, its stiffness increases with growth but still retains a certain flexibility. In the elderly the joints between the ribs and the sternum and the vertebrae, respectively, become stiffer. In addition, the ribs become more brittle due to changes of the bone properties. This increases the likeliness of rib fractures and reduces the protective potential of the rib cage.

K.-U. Schmitt et al., *Trauma Biomechanics*, DOI: 10.1007/978-3-642-53920-6_5,

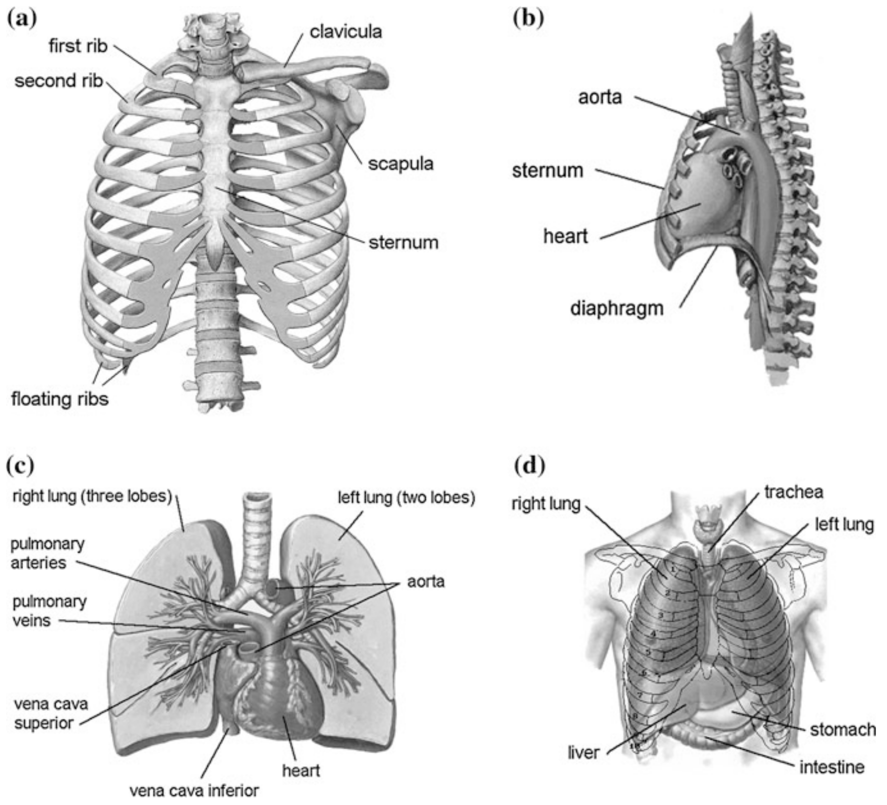

Fig. 5.1 The thoracic anatomy (adapted from Sobotta 1997; Netter 2003)

The interior volume covered by the rib cage can be divided into three areas. The right and left outer regions contain the lungs. The centre section, called mediastinum, hosts among others the heart, the trachea and large vessels.

The left lung consists of two lobes, while the right lung consists of three lobes. Two layers of membranes surround the lung: the visceral pleura, which encloses the lung tissue, and the parietal pleura, which covers the entire inside of the rib cage (including the cranial side of the diaphragm and the vertebral bodies). The visceral and the parietal pleura are not connected to each other, but form a small cavity. This pleural cavity is an enclosed space. To keep the lung in their inflated state, a continuous underpressure is maintained in the pleural cavity. If this underpressure cannot be maintained (for example due to a perforation of the chest), the lungs deflate and the pleural cavity is filled with air. This phenomenon is called pneumothorax (see Sect. 5.2.2).

For respiration the diaphragm, the rib cage and the intercostal muscles function as a pump by drawing air into the lung (inspiration) and expelling air from the lung (expiration). For inspiration the thoracic volume is increased by lifting the rib cage and by lowering the diaphragm. Consequently the lung will expand and air is sucked in. To normally expel the air (expiration) the thoracic structures and the diaphragm are relaxed.

The mediastinum is located between the two lungs, the thoracic vertebrae and the sternum. Large vessels included are the aorta, vena cava, the pulmonary arteries and veins (Fig. 5.1). Due to the restricted space available in the mediastinum, a compression of the anterior rib cage may easily cause injuries to internal structures.

5.2 Injury Mechanisms

This description of thoracic injuries and the according injury mechanisms focuses on blunt impacts in traffic accidents. Hence only scenarios where a flat or blunt object strikes the chest without penetration are regarded. This type of impact is most often seen in automotive accidents with an occupant contacting, for example, the steering wheel, the dashboard or components of restraint systems. Special attention has to be given in this context to the problem of car occupants who are seated in an unusual position ("out-of-position", e.g., front passengers keeping the feet on the dashboard) during a crash situation. Likewise, elderly passengers are of concern (Yoganandan et al. 2007).

If the thorax is suddenly decelerated due to a blunt impact, three different injury mechanisms can be distinguished: compression, viscous loading, and inertial loading of the internal organs. Furthermore, any combination of these three basic phenomena can occur.

The resulting injuries can be categorised as skeletal injury and soft tissue injury. Most often the thoracic wall and the lung are injured together with rib fractures, fractures of the sternum, and pleura ruptures. In case of fractures of the vertebral column, injuries of the spinal cord may also occur, possibly leading to transverse lesion (in case of motorcyclists this sometimes results in quadriplegia). Fortunately such injuries are less frequently recorded, as are injuries on the aorta, the heart, the oesophagus and the diaphragm. Table 5.1 provides an overview of different injuries and their according AIS (Abbreviated Injury Scale) rating.

To date, the mechanisms of rib fractures and some of the lung injuries are reasonably well understood whereas some mechanisms leading to other injuries of the thoracic organs still merit further research. Figure 5.2 summarises possible soft tissue thoracic injuries.

5.2.1 Rib Fractures

According to the AIS, a single rib fracture can be graded as AIS 1. If 2-3 ribs are broken, the grade increases to AIS 2. Hence this type of injury is usually not severe and most single rib fractures are in fact self-healing. However, if multiple

Table 5.1 AIS rating for skeletal and soft tissue thoracic injuries (AAAM 2005)

AIS skeletal injury		AIS soft tissue injury	
1	One rib fracture	1	Contusion of bronchus
2	2–3 rib fractures; sternum fracture	2	Partial thickness bronchus tear
3	4 or more rib fractures on one side; 2–3 rib fractures with hemothorax or pneumothorax	3	Lung contusion; minor heart contusion
4	Flail chest; 4 or more rib fractures on each of two sides; 4 or more rib fractures with hemo- or pneumothorax	4	Bilateral lung laceration; minor aortic laceration; major heart contusion
5	Bilateral flail chest	5	Major aortic laceration; lung laceration with tension pneumothorax
–	–	6	Aortic laceration with hemorrhage not confined to mediastinum

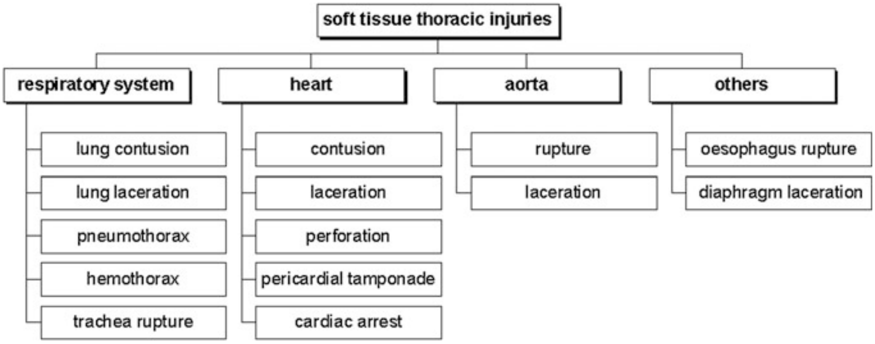

Fig. 5.2 Possible soft tissue thoracic injuries

fractures occur, life threatening complications may arise. If the skin and the soft tissue overlaying the fracture remain intact, the fracture is called a closed fracture. If, on the other hand, sharp edges of broken ribs perforate the chest wall, the fracture is called an open fracture. Such open fractures are of particular concern because they can lead to a pneumothorax, lung collapse and infections. Broken ribs may also perforate the visceral or parietal pleura, causing respiratory problems.

Generally, sagittal loading of the thorax is more likely to cause single rib fracture, while lateral impact more often results in multi rib fracture. In principle, ribs can fracture at any point, but most likely they break at the point of maximum curvature or at the location where a force is applied. Hence, given the fact that the ribs are stronger curved laterally, together with fewer muscle tissue that covers and thus protects the ribs in that area, lateral fractures are more likely. The site of lateral rib fracture(s) depends on the shape of the impacting body (Fig. 5.3).

In case of multi rib fracture the thorax wall may lose its overall stability. This can result in a thorax motion that is contrary to normal: during inspiration the

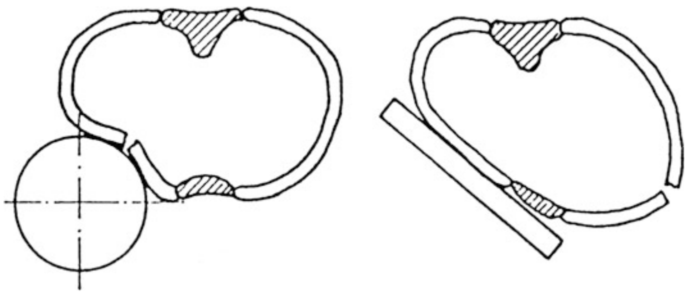

Fig. 5.3 Site of rib fracture depending on impact body (adapted from Kramer 1998)

disrupted thorax wall is sucked in and thus reduces the volume of the lung. On expiration the thorax wall moves outwards making it difficult to expel the air out of the lung. The greater the area of thorax wall damaged, the lesser the amount of air which can be exchanged. This condition is called a flail chest which eventually results in hypoxemia.

According to cadaver studies (e.g. Stalnaker and Mohan 1974; Melvin et al. 1975) the number of rib fractures depends on the magnitude of rib deflection rather than on the rate of deflection. Due to the viscous nature of the thorax, the amount of force, however, depends on the rate at which the force is applied. Hence, force appears to be related to the number of rib fractures for a given loading rate. Plastic deformation properties of bone may furthermore have an important influence on peak stresses in ribs (see Sect. 2.2). Whenever at a certain location the stress reaches a plastic limit, there is no further (or only a minimal) stress increase until a final failure level is attained. If plastic deformation processes are disregarded, therefore, stresses may be overestimated.

The occurrence of rib fractures is strongly age dependent. While the rib cage can be compressed frontally in a young person until it contacts the spine without fracturing a rib (but compressing the organs in between), the rib cage of individuals of more than 50 years of age break at much lower loads, often, for example, during the cardiac rescue procedure.

5.2.2 Lung Injuries

As indicated in Fig. 5.2, injuries to the respiratory system mainly concern lung injuries. Due to thorax compression (both with and without rib fracture) a lung contusion can occur. This often happens in combination with a flail chest.

Unlike rib fractures, lung contusion is rate dependent (Fung and Yen 1984). At high velocities, a compression or pressure wave is transmitted through the thorax wall to the lung tissue, causing damage to the capillary bed of the alveoli. Sometimes also central lung contusion without damage of the surrounding tissue is

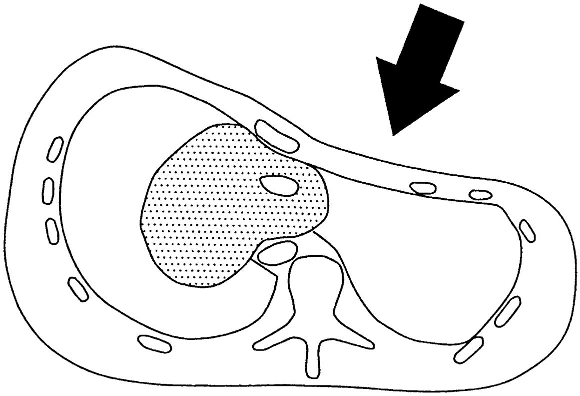

Fig. 5.4 Compression of the heart (adapted from Kramer 1998)

observed. As a serious complication, lung contusion also increases the risk of pneumonia, i.e. an inflammation of the lung tissue.

Laceration and sometimes also perforation of the lung tissue can be observed at sites of rib fractures. This may result in a pneumothorax or a hemothorax. In the first case the pleural cavity is filled with air, in the second case with blood. A combined situation where the pleural cavity contains both blood and air is called a hemo-pneumothorax.

A pneumothorax results from a perforation of the pleura, i.e. a hole is created in the pleural sac between the lung and the rib cage, caused, for instance, by broken ribs. On inspiration the intrapleural pressure is reduced and air is sucked into the pleural cavity through the leak in the lung. During expiration the laceration in the lung tissue is compressed preventing the air in the pleural cavity to be expelled. Hence, while breathing, the amount of air inside the pleural cavity increases, eventually compressing the lung.

A hemothorax also reduces the effective lung volume, but due to blood in the pleural cavity. Hereby a laceration of blood vessels (e.g. in the lung tissue) may cause blood to accumulate in the pleural cavity.

5.2.3 Injuries to Other Thoracic Organs

From thoracic impact, the heart can be subjected to several injuries including contusion and laceration (Fig. 5.2). Contusion occurs due to compression and depends on the associated velocity, while laceration may be due to high magnitude of compression over the sternum. At high rates of loading, the heart may undergo arrhythmia, fibrillation or arrest. High speed blunt impacts (15–20 m/s) appear to interrupt the electromechanical transduction of the heart wall—a condition called commotio cordis (Maron and Estes III 2010). Figure 5.4 illustrates a thoracic impact, with the heart under compression between the sternum and the vertebral column.

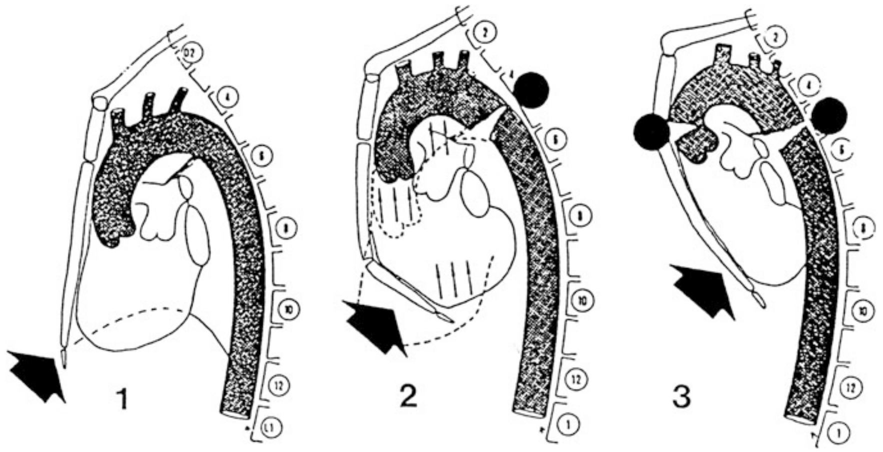

Fig. 5.5 Compression of the heart and possible sites of aortic rupture (adapted from Viano 1990)

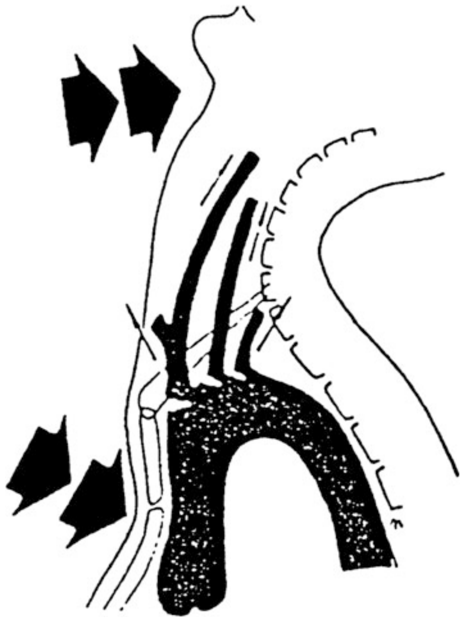

Fig. 5.6 Thorax compression in combination with hyperextension of the neck can result in laceration of the aorta (adapted from Viano 1990)

Furthermore, major thoracic blood vessels like the aorta may be injured. Rupture and laceration are the most likely mechanisms resulting from blunt trauma to the thorax. Cavanaugh (2002) reports that arterial injuries account for 6–8 % of AIS > 2 only, but represent 27–30 % of the estimated harm. It is also remarkable that 80–85 % of the victims sustaining an aortic trauma in an automotive accident die at the scene of the accident (e.g. Smith and Chang 1986; Butler et al. 1996).

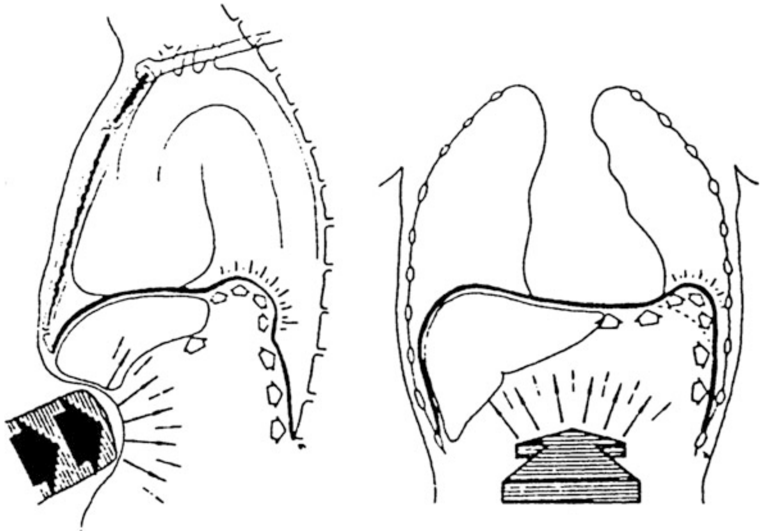

Fig. 5.7 Laceration of the diaphragm due to blunt impact on the abdomen (adapted from Viano 1990)

Mechanisms of injury were found to be predominately high speed motor vehicle crashes followed by falls and pedestrians being struck (e.g. Ochsner et al. 1989).

Aortic rupture is thought to occur either from traction or shear forces generated between relatively mobile portions of the vessel and points of fixation or, secondly, due to direct compression over the vertebral column or, thirdly, caused by an excessive sudden increase of intraluminal pressure. Aortic rupture after thorax compression is shown schematically in Fig. 5.5. Additionally, Viano (1983) reported that the inertial loading of the blood-filled heart can cause the heart to displace in the thoracic cavity and thus stretch points of attachment of the aortic arch, such as the superior arteries or the ligamentum arteriosum. This may occur if the heart is displaced vertically, laterally, or obliquely. Further, Viano considered the possibility of aortic laceration in combination with hyperextension of the neck in high speed loading (Fig. 5.6). As for the site of thoracic aortic injury, it was found that the region of the aortic isthmus, just distal to the origin of the left subclavian artery, is most vulnerable. It accounts for the vast majority of such injury (e.g. Butler et al. 1996; Creasy et al. 1997).

Other injuries of thoracic organs include rupture of the oesophagus and laceration of the diaphragm. The latter possibly results in a hernia. However, as Fig. 5.7 shows, a laceration of the diaphragm is most probably a consequence from blunt impact to the abdomen (see Chap. 6).

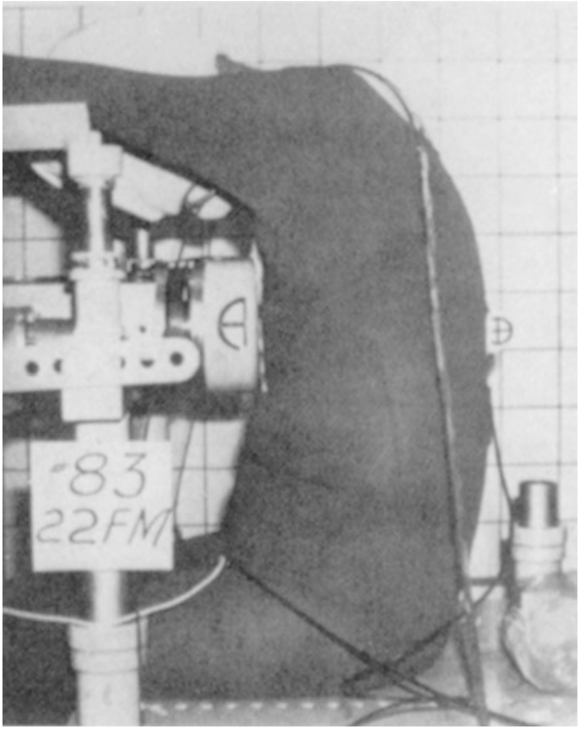

Fig. 5.8 Cadaver test using an impactor to apply load on the sternum (from Kroell et al. 1971)

5.3 Biomechanical Response

Many biomechanical tests have been performed under controlled laboratory conditions to measure the biomechanical response of the human thorax in terms of accelerations, forces, deformations and pressures. In particular cadaver tests, extensively conducted in the 1970s, were performed to obtain details of resulting injury to the body after impact. The data was then used to develop frontal and side impact dummies as well as to develop injury criteria. Furthermore the data was used to establish and validate mathematical models of the thorax.

In terms of test conditions, mainly pendulum and sled tests were used. Additionally, quasi-static tests—some with volunteers—were performed to determine the stiffness of the thorax.

5.3.1 Frontal Loading

To investigate the biomechanical response of the thorax to frontal loading, extensive test series were performed. Human cadavers were impacted in pendulum tests using a 6-inch-diameter rigid pendulum (Fig. 5.8). Measuring the deflection

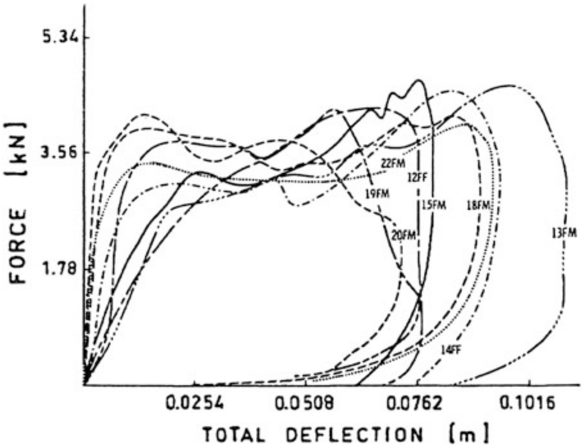

Fig. 5.9 Force-deflection characteristics of the thorax in frontal impact (from Kroell et al. 1974)

of the sternum, force-deflection characteristics for the thorax were determined (e.g. Kroell et al. 1971, 1974; Stalnaker and Mohan 1974). Figure 5.9 shows a representative force-deflection curve as obtained from such experiments. The hysteresis curve can be divided into a loading and an unloading phase. The loading phase is characterised by an initial rapid rise which is mainly due to the viscous properties of the thorax, and a plateau region that is also due to a viscous response. At maximum deflection, the impactor and the test subject are moving at a common velocity. The forces measured at this point are due to inertial forces caused by whole-body acceleration, and the elastic forces due to tissue compression. The unloading phase of the curve represents the unloading of the compressed tissues and follows the elastic non-linear unloading of the thorax. Analysing the relationship between the force plateau and impactor velocity, it was found that the force plateau increases with impactor velocity except for impactors with low mass but high velocity, which do not at all exhibit a force plateau. Furthermore, it was shown that lower impactor masses resulted in lower deflections (Lobdell et al. 1973).

Based on such cadaver tests, force-deflection corridors for different combinations of impactor mass and velocity were developed which are used for performance requirements for dummies.

In addition to dynamic pendulum tests focusing on sternal impacts, quasi-static tests have been performed. Since three-point belts and airbags are frequently used today, lower rate loading has become more important in frontal impact. Distributed loading to the ribs due to airbags as well as rib and clavicle loading due to the shoulder belt make quasi-static thorax loading data also necessary. Performing such tests, the sternum of volunteers or cadavers is loaded with a plate with the subject's back against a rigid structure. The applied load and the anterior-posterior deflection of the thorax are recorded. Reviewing the data available, Melvin et al. (1985) concluded that for deflections of up to 41 mm the thorax has an approximate linear stiffness of 26.3 N/mm, and for deflections greater than 76 mm, the

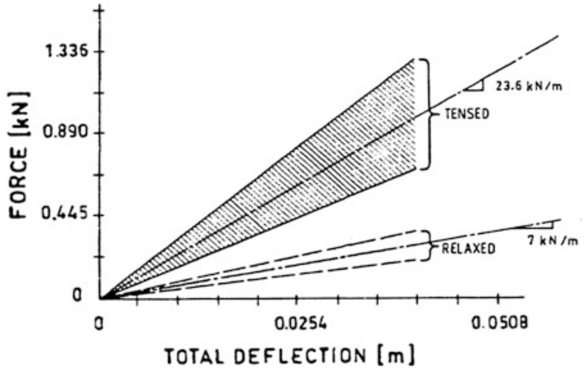

Fig. 5.10 Results from quasi-static volunteer tests showing the influence of a tensed and relaxed state of the thorax (from Lobdell 1973)

stiffness increases to 120 N/mm. However, the results are influenced by the individual physique of the test subject and differ remarkably for different conditions of the test subject such as embalmed and unembalmed cadavers and relaxed and tensed volunteers. In Fig. 5.10 results obtained by Lobdell et al. (1973) are presented, clearly indicating the difference between a relaxed and a tensed volunteer. The fact that the stiffness of the thorax is increased in a tensed state can be regarded as beneficial in terms of injury tolerance.

The influence of wearing a seat belt, particularly a diagonal shoulder belt, on the occupant loading has been investigated since the late 1970s. It was observed that the thorax is more vulnerable to injury under the more concentrated belt loading. Injury due to belt loading appeared to be caused by thorax compression. More recent cadaver experiments deepened the knowledge on thorax compression due to the shoulder belt also investigating effects of the mechanical coupling of the ribs and the sternum (e.g. Shaw et al. 2007). Further studies analysed the effect of advanced belt systems that incorporate, for example, load limiters or pretensioners. From an analysis of accident data, Bendjellal et al. (1997) concluded that the shoulder belt force should be limited to 4 kN. Also Foret-Bruno et al. (1998) suggested a belt load limitation of 4 kN combined with a specially designed airbag system. They estimated that 95 % of AIS3+ thorax injuries could be prevented in frontal impacts.

Additionally, the loads to the thorax that are applied by a deploying airbag were assessed in several studies (e.g. Cavanaugh 2002). In general, injury was related to the internal airbag pressure. When at any time during the deployment process the available volume of the airbag is smaller than the gas volume generated, high forces on the subject can arise. If an occupant is for instance "out of position", i.e. if the subject is in the path of the airbag, a load on the subject's thorax due to airbag pressure occurs. Figure 5.11 illustrates such loading caused by punch-out forces. Here the punch-out interaction is due to the proximity of the chest to the airbag system. Punch-out also occurs due to contact with the airbag module restricting the normal deployment for the airbag as it breaks out of the module, and begins to unfold. In contrast, membrane loading (Fig. 5.11) occurs later in the

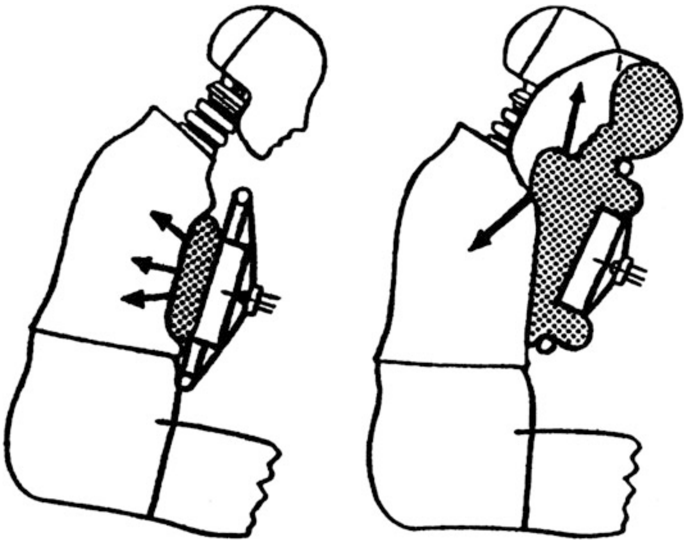

Fig. 5.11 Mechanisms for airbag inflation-induced injury. Punch-out loading mechanism causing pressure on the thorax (*left*) and membrane loading mechanism resulting in pressure to the thorax and the head-neck complex (*right*) (from Melvin and Mertz 2002)

deployment process after the airbag has left the module but with the occupant nonetheless being too close to the airbag. Consequently, the airbag wraps around the occupant, particularly in the head/neck and chest region. In addition to thorax injuries, head injuries, e.g. basilar fractures, are observed (McElhaney et al. 2002). It appeared that the harmonisation of the airbag and the belt system is an important factor to maximise the benefits from such systems.

From analysing accidents with deployment of the frontal airbag, Otte (1995) concluded that the protective effect of a three-point belt system was sufficient up to a delta-v of 35–40 km/h. For a higher delta-v, a supplementary airbag system should be activated. In contrast, Kallieris et al. (1995) reported that bag-like compression to the thorax is favourable, as the forces are distributed more evenly. Performing frontal impact cadaver tests (at 48 km/h impact velocity), they recommended that an overall restraint of the occupant should be achieved by a belt, but that thorax injury mitigation should be aimed at using airbag systems. The combined thoracic index CTI was developed to assess both airbag and belt loading in crash tests (see Sect. 5.4.5). Various tolerance values for frontal loading are presented in Table 5.2.

Based on experimental results, a mathematical model describing the behaviour of the thorax in frontal impact was developed (Lobdell et al. 1973). The lumped-mass model utilises springs, masses and dampers (Fig. 5.12) and the model's force-deflection response was tuned to match the low and high velocity corridors determined experimentally (Kroell et al. 1971, 1974). In the meantime the model

Table 5.2 Frontal impact tolerances of the thorax

Tolerance level	Injury level	Reference
Force		
3.3 kN to sternum	Minor injury	Patrick et al. (1969)
8.8 kN to chest and shoulders	Minor injury	Patrick et al. (1969)
Acceleration		
60 g	3 ms value for Hybrid III	FMVSS 208 (old version)
Deflection		
58 mm	No rib fracture	Stalnaker and Mohan (1974)
52 mm	Limit for Hybrid III (5 %)	FMVSS 208
63 mm	Limit for Hybrid III (50 %)	FMVSS 208
Compression		
20 %	Onset of rib fracture	Kroell et al. (1971, 1974)
40 %	Flail chest	Kroell et al. (1971, 1974)
VC_{max}		
1.0 m/s	25 % probability of AIS $\geq$ 4	Viano and Lau (1985)
1.3 m/s	50 % probability of AIS $\geq$ 4	Viano and Lau (1985)
Combined thoracic index (CTI)		
A_{max}/60 g + D_{max}/76 mm	50 % probability of AIS > 3 in cadavers	Kleinberger et al. (1998)

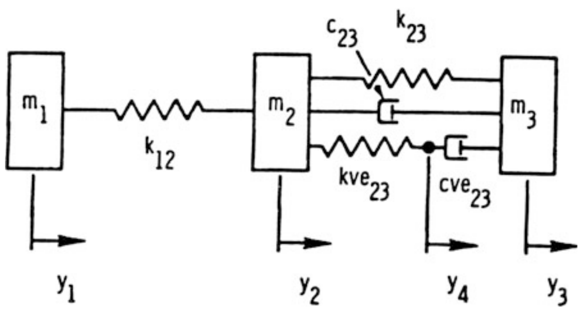

Fig. 5.12 Viscous thorax model (from Lobdell 1973)

has been modified by various researchers and validated with additional test results. It is still used within the scope of dummy thorax design studies although finite element models are becoming more sophisticated (see Chap. 2).

5.3.2 Lateral Loading

To investigate the biomechanical response of the human body in side impact, the same methods used to analyse frontal loading were applied. Cadaveric studies were also the method of choice to address the force-deflection characteristics of the thorax due to lateral impact. As a result of such impactor tests, hysteresis curves were presented that were similar to those obtained for frontal loading except for the fact that no or less apparent force plateau regions were determined. Furthermore, it was shown that the resistance of the thorax to lateral impact is smaller than to frontal loading. The arm of the test subject also has an effect on the test results depending on its position during impact. The arm can partly or completely be placed between the impacting mass and the thorax or it can be raised. Cesari et al. (1981) demonstrated this influence in a test series with cadavers. They concluded that the arm can have a protective effect when positioned between the striking object and the thorax.

In addition to the impactor tests, so-called drop tests were performed to analyse the force-deflection characteristics of the struck-side half-thorax. Cadavers were dropped from a height of 1–3 m onto an unpadded or padded force plate (e.g. Stalnaker et al. 1979; Tarriere et al. 1979). Results are summarised in Table 5.3. As a further result of these studies, a corridor for the development of a side impact dummy was proposed.

To further investigate side impacts, sled tests were conducted at the University of Heidelberg (Kallieris et al. 1981). A seat with a low friction coefficient was mounted on a sled. The sled was suddenly decelerated from a specified velocity so that the test subjects (cadavers) sitting on the seat slid across the seat and impacted a padded or unpadded wall. The acceleration of the ribs, the sternum and the thoracic vertebrae was measured. It was noticed that besides the acceleration, physical parameters of the test subjects had a significant influence on the injury outcome.

Consequently, the Thoracic Trauma Index (TTI) was proposed (Eppinger et al. 1984), which includes among others an age factor (see also Sect. 5.4). Nowadays, side impact dummies allow to measure upper and lower rib acceleration so that the TTI can be calculated to assess side impact crash worthiness.

5.4 Injury Tolerances and Criteria

As described in the previous sections, injuries to the thorax occur due to compression, viscous or inertial loading or combinations thereof. Using different kinds of experiments, the biomechanical response of the thorax in terms of tolerance values for various load cases was determined. Furthermore, injury criteria were developed to relate a certain loading of the thorax with an according injury risk. This chapter presents the most commonly used thoracic injury criteria and tolerance thresholds (Tables 5.2, 5.3).

Table 5.3 Lateral impact tolerances of the thorax

Tolerance level	Injury level	Reference
Force		
7.4 kN	AIS0	Tarriere et al. (1979)
10.2 kN	AIS3	Tarriere et al. (1979)
5.5 kN	25 % probability of AIS $\geq$ 4	Viano (1989)
Acceleration		
T8-Y 45.2 g	25 % probability of AIS $\geq$ 4	Viano (1989)
T12-Y 31.6 g	25 % probability of AIS $\geq$ 4	Viano (1989)
60 g	25 % probability of AIS $\geq$ 4	Cavanaugh et al. (1993)
TTI(d)		
TTI(d) 85 g	Max. in SID dummy for 4-door cars	FMVSS 214
TTI(d) 90 g	Max. in SID dummy for 2-door cars	FMVSS 214
TTI 145 g	25 % probability of AIS $\geq$ 4	Cavanaugh et al. (1993)
TTI 151 g	25 % probability of AIS $\geq$ 4	Pintar et al. (1997)
Compression to half thorax		
35 %	AIS3	Stalnaker et al. (1979), Tarriere et al. (1979)
33 %	25 % probability of AIS $\geq$ 4	Cavanaugh et al. (1993)
Compression to whole thorax		
38.4 %	25 % probability of AIS $\geq$ 4	Viano (1989)
VC_{max} to half thorax		
0.85 m/s	25 % probability of AIS $\geq$ 4	Cavanaugh et al. (1993)
VC_{max} to whole thorax		
1.0 m/s	50 % probability of AIS $\geq$ 3	Viano (1989)
1.47 m/s	25 % probability of AIS $\geq$ 4	Viano (1989)

5.4.1 Acceleration and Force

Early attempts to quantify thoracic loading focused on acceleration. As of today, the human tolerance for severe thorax injuries is considered as peak spinal acceleration sustained for 3 ms or longer not to exceed 60 g in a frontal impact. This value is also embodied in FMVSS 208 to assess frontal impact crash worthiness. For lateral impact, different thresholds are proposed (see Table 5.2).

Closely related to acceleration is the definition of force tolerance values. Assuming an effective thorax mass of 30 kg, a force limit of 17.6 kN corresponds to the 60 g acceleration level. However, cadaver tests by Patrick et al. (1969) observed minor skeletal injuries already at 3.3 kN for impacts to the sternum and 8.0 kN for distributed loads to the shoulders and the thorax. These results show that the reliability of a single acceleration or force criterion as a general injury parameter for thoracic injuries is rather limited. Neither of the two criteria takes the viscous nature of the thorax into account. Consequently, more complex criteria were developed to obtain a better correlation with experimental results.

5.4.2 Thoracic Trauma Index

The Thoracic Trauma Index (TTI) is an injury criterion for the thorax in the case of a side impact. It assumes that the occurrence of injuries is related to the mean of the maximum lateral acceleration experienced by the struck side rib cage and the lower thoracic spine. Furthermore, the TTI takes into account the weight and the age of the test subject and thus combines information on the kinematics with parameters of the subject's individual physique. The TTI (dimension [g]) is defined as follows:

$$TTI = 1.4AGE + 0.5(RIB_y + T12_y)(M/M_{std}) \tag{5.1}$$

with AGE being the age of the test subject (in years); RIB_y (g) represents the maximum of the absolute value of the lateral acceleration of the 4th and 8th rib on the struck side; $T12_y$ (g) gives the maximum of the absolute value of the lateral acceleration of the 12th thoracic vertebra; M denotes the subject's mass (kg) and M_{std} refers to a standard mass of 75 kg.

When using a 50th percentile Hybrid III dummy to perform crash tests, a different version of the TTI, called the TTI(d), can be calculated. To obtain TTI(d) values, the age related term in Eq. 5.1 is omitted and the mass ratio becomes 1.0. It is important to note that the acceleration signals needed to determine the TTI and TTI(d), respectively, have to be preprocessed, i.e. filtered and sampled, according to a prescribed procedure (defined in FMVSS 214 and SAE J1727).

To relate TTI values to thoracic injuries, a large number of cadaver tests were performed (e.g. Kallieris et al. 1981) and injury risk functions were established statistically. Hence, the TTI reflects a statistical correlation rather than a biomechanical one. It cannot directly be related to any injury mechanism involved.

5.4.3 Compression Criterion (C)

Analysing blunt impact tests, Kroell et al. (1971, 1974) concluded that the maximum thorax compression correlated well with AIS while force and acceleration did not. Defining compression (C) as the chest deformation divided by the thickness of the thorax the following relationship was established:

$$AIS = -3.78 + 19.56C \tag{5.2}$$

Thus measuring 92 mm thorax deflection for the 230 mm chest of the 50th percentile male results in a compression C of 40 % and predicts AIS4. 30 % compression leads to AIS2. Performing statistical analysis of the injury risk shows that in frontal impact a thorax compression of 35 % results in a 25 % probability of severe injuries rated AIS4 or higher. FMVSS 208 allows a maximum deflection between 63 mm for the 50 % Hybrid III dummy and 52 mm for the 5 % Hybrid III dummy in frontal impact.

5.4.4 Viscous Criterion

The viscous criterion (velocity of compression), also called the soft tissue criterion, is an injury criterion for the chest area taking into account that soft tissue injury is compression-dependent and rate-dependent. The VC value [m/s] is the maximum of the momentary product of the thorax deformation speed and the thorax deformation. Both quantities are determined by measuring the rib deflection (side impact) or the chest deflection (frontal impact). Hence:

$$VC = V(t) \times C(t) = \frac{d[D(t)]}{dt} \times \frac{D(t)}{b} \tag{5.3}$$

where $V(t)$ (m/s) is the velocity of the deformation calculated by differentiation of the deformation $D(t)$, and $C(t)$ denotes for the instantaneous compression function which is defined as the ratio of the deformation $D(t)$ and the initial torso thickness b. Details on how the deformation data must be filtered is given in ECE R94 for side impact and SAE J1727 for frontal impact, respectively. Often the maximum VC, VC_{max}, which was found to correlate well with the risk of thoracic injuries (Viano and Lau 1985), is reported. Using the Lobdell model (see Sect. 5.3.1), a relationship between the VC and energy absorbed in the thorax can be established. As for the critical values, both ECE R95 (lateral impact) and ECE R94 (frontal impact) require the VC to be less or equal to 1.0 m/s.

5.4.5 Combined Thoracic Index

The Combined Thoracic Index (CTI) represents an injury criterion for the chest area in case of frontal impact (Kleinberger et al. 1998). Combining compression and acceleration responses, the CTI particularly addresses both airbag and belt loading. The CTI is defined as the evaluated 3 ms value from the resultant acceleration of the spine and the deflection of the chest. The calculation of the CTI value is based on the following equation:

$$CTI = \frac{A\text{max}}{A_{\text{int}}} + \frac{D\text{max}}{D_{\text{int}}} \tag{5.4}$$

where

$A_{\max}$ = 3 ms value (single peak) of the resultant acceleration of the spine (g)
A_{int} = critical 3 ms intercept value (g)
$D_{\max}$ = deflection of the chest (mm)
D_{int} = critical intercept value for deflection (mm)

Intercept values are defined for different dummy types. For the 50th percentile Hybrid III, for example, they read 85 g for A_{int} and 102 mm for D_{int}.

The combined compression and acceleration criterion accounts for the differences in loading of the thorax by belt versus airbag systems. It is based on the assumption that, for a given load, a belt system would apply greater pressure along its contact area than an airbag system, which has a larger contact area. With the combined belt/airbag system, the predominant loading could be a line load, i.e. the load from the belt is larger than the one from the airbag, or a distributed load in the opposite case. The CTI is meant to reflect the whole range of possible loading scenarios between these two extremes. While the maximum thorax acceleration is a measure of the magnitude of total forces applied to the torso in proportion to its mass, the thorax deflection is an indicator of the belt loading. A higher deflection per unit of acceleration thus leads to a greater relative contribution of the belt system (Cavanaugh 2002).

The CTI was developed based on cadaver tests and correlated to the AIS by logistic regression analysis. To date, the CTI was included in FMVSS 208, where details on data acquisition and the different intercept values are given.

5.4.6 Other Criteria

The Rib Deflection Criterion (RDC) is the criterion for the deflection of the ribs, expressed in mm, in a side impact collision. According to ECE R95 the RDC shall be less than or equal 42 mm (side impact dummy).

ThCC (or TCC) is the abbreviation for Thoracic Compression Criterion. ThCC is the criterion of the compression of the thorax in frontal impact between the sternum and the spine and is determined using the absolute value of the thorax compression, expressed in mm. Today a maximum threshold value of 50 mm is defined in ECE R94.

5.5 Thoracic Injuries in Sports

In some sports thoracic injuries are a serious concern due to the high severity of the injuries sustained (e.g. in equestrian sports). Also direct trauma such as blows to the chest by projectiles (e.g. baseballs, balls) as well as indirect impact is observed in sports. Generally, the aforementioned descriptions of injuries and injury mechanisms also apply for traumatic injuries in sports. In contrast to the automotive field, systematic and sports specific investigations on thorax impact scenarios or injury threshold levels are rare. In an exemplary study on the injury potential of taekwondo kicks in the thoracic region, Serina and Lieu (1991) determined peak values of chest deflection of 5 cm while the VC values were up to 1.4 m/s.

Additionally overuse injuries, for instance, in the form of stress fractures of the sternum or the rib are reported from different sports including rowing (e.g. Coris and Higgins 2005; McDonnell et al. 2011; Karlson 2012).

5.6 Summary

Injuries of the thorax are most often related to blunt impact. Compression, viscous or inertial loading are frequent causes for injury. Particularly the visco-elastic properties of the thorax are reflected in its biomechanical response. Consequently some injury criteria also consider the velocity with which the thorax is deformed.

5.7 Exercises

E5.1: Depict an ideal linear elastic-plastic stress-strain curve in one dimension.

E5.2: VC-criterion: A model thorax is exposed to a frontal impact. The maximal measured chest deflection is 45 mm while the diameter of the thorax (front-back) is 20 cm. Determine the VC value under the assumption that the deformation rate is constant during the impact and the 45 mm deflection are reached within 40 ms (duration of impact).

E5.3: Combined Thoracic Index: The CTI consists of two additive components. Explain the physical background of the two components.

P5.1: Work out the set of equations which correspond to the lumped parameter lung model according to Lobdell (Fig. 5.12).

Write a computer programme using e.g. Matlab (or a similar software) and discuss the solutions:

(a) Parametric study: impacting mass $m_1 = 10, \ldots, 40$ kg, impact velocity $dy_1/dt = 5, \ldots, 15$ m/s

(b) Sensitivity of the parameters: which parameter has the highest influence?

(c) Comparison with Kroell's measurements (Fig. 5.9).

Use the following parameters as a reference:

mass: $m_2 = 0.45$ kg, $m_3 = 27.2$ kg

spring constants: $k_{12} = 281$ kN/m, $k_{23} = 26.3$ kN/m, $kve_{23} = 13.2$ kN/m

damping constants: c_{23}(compression) $= 0.52$ kNs/m,

c_{23}(extension) $= 1.23$ kNs/m, $cve_{23} = 0.18$ kNs/m

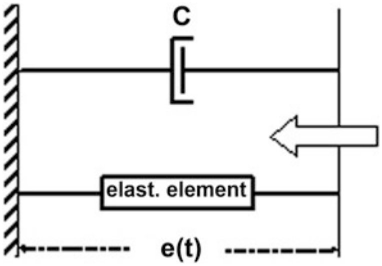

Fig. 5.13 Mechanical model of a viscous element

P5.2: Qualitatively "derive" the VC criterion on the basis of the energy absorbed in the viscous element of the simplified model shown below (Fig. 5.13).

The absorbed energy can be estimated from the following proportional relation

$$E \propto \int_{\Delta t} \dot{e}(t)de$$

The integral extends over the time of impact when deformation occurs. Use the fact that under realistic conditions, $(\dot{e})^2 >> e\ddot{e}$

Finally, e is identified with the relative compression, C, while $\ddot{e}$ is analogous to V.

P5.3: Assume that a belt equipped with a force limiter interacts with the thorax. Use Lobdell's model to discuss the maximal thorax deflection with/without force limitation. Within the framework of the model, the force in the spring element k_{12} is limited in case of force limitation. What is a reasonable limit?

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Abdominal Injuries 6

The abdominal cavity is a vulnerable region of the human body. In general, trauma to the abdomen is caused by blunt impact or by penetration. In automotive accidents, blunt impact is frequently observed although the injury might not be apparent initially.

In terms of investigating the biomechanical response of the abdomen, experimental studies turned out to be particularly difficult to perform and the results obtained are not easy to interpret. Thus, we still lack sufficient knowledge of injury mechanisms and appropriate injury predictors. "The reader needs to be keenly aware of the wide variation in human response and tolerance. This is due primarily to the large biological variations among humans and to the effects of aging. Average values that are useful in design cannot be applied to individuals" (King 2001). This lack of knowledge is also evident in human surrogates used in crash testing (see Chap. 2). An excellent review on abdominal injuries was, for example, presented by Rouhana (2002).

6.1 Anatomy of the Abdomen

Cranially, the abdominal cavity is bounded by the diaphragm and caudally by the pelvic bones and the muscles attached to it. The lumbar vertebral column, which itself is usually not considered as part of the abdomen, forms, together with the sacrum and the pelvis, the posterior boundary of the abdomen. Anteriorly and laterally the upper abdomen is bound by the lower rib cage. The lower abdomen is surrounded anteriorly and laterally by musculature. Because of the ribs, the upper abdominal region, sometimes called the "hard thorax" (Eppinger et al. 1982), shows a different behaviour with respect to impact response and tolerance than the lower abdomen. The presence of the lower ribs (although not directly attached to the sternum, see Chap. 5) comes particularly into effect in rear-end and side impacts. For frontal impact, however, it appears that organs directly in front of the

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vertebral column are at higher risk of being compressed than organs lateral to the spine.

The abdominal cavity hosts several organs that are generally divided into "solid" and "hollow" organs. The main characteristic to divide the organs into these two groups is the gross density of an organ (not the tissue density). Solid organs like the liver, spleen, pancreas, kidneys, ovaries and adrenal glands have a higher density than hollow organs such as the stomach, large and small intestines, bladder and uterus. The lesser density of the hollow organs is due to the presence of a relatively large cavity within the organ itself. Those cavities are, for example, filled with "air" or digestive matter. The solid organs, in contrast, contain fluid-filled vessels and therefore exhibit a higher density.

Major blood vessels of the abdomen are the abdominal aorta, the inferior vena cava, the hip artery (arteria iliaca communis) and the hip vein (vena iliaca communis). The abdominal aorta and the vena cava enter the abdomen from the cranial side through separate openings in the diaphragm. Figure 6.1 illustrates the abdominal organs in situ. With respect to the biomechanical response of the abdomen on traumatic impact, it is important to note that the organs inside the abdominal cavity inhere a relatively high degree of mobility. They are neither rigidly fixed to the abdominal wall nor to each other. Partly they are embedded in fat (e.g. the kidneys) or are tethered by folds of the peritoneum (e.g. the intestines). The peritoneum is a serous membrane that covers the inner abdominal walls and surrounds each organ. As this membrane is smooth and moist, it acts as a lubricant and thus also adds to the mobility of the organs. Consequently, the abdominal organs can adjust to different postures such as sitting or standing. Furthermore, the position of the liver, for instance, changes during respiration, because it moves with the diaphragm. This mobility has therefore a great influence on the biomechanical response and, of course, also on the outcome of experimental studies examining the injury mechanism.

In summary from an anatomical point of view, possible injury mechanisms seem to be dependent on the complex structure of the abdomen together with physical properties like the density, structure and the material within the organs.

6.2 Injury Mechanisms

Due to the complex structure of the abdomen, there are many factors influencing the location, likeliness and severity of a blunt impact. First of all, anatomy suggests that the position of an organ contributes to its injury risk. Organs that are located anterior of the vertebral column are, in case of a frontal impact, more likely to be compressed against the spine than those lying laterally. Additionally, the upper abdomen is in part covered by the lower rib cage, which also has a protective effect in frontal impact. Obviously, the non-symmetric organisation of the abdominal organs accounts for different injury risk depending on the impact direction. If the abdomen is struck from the right side, liver injury is more probable

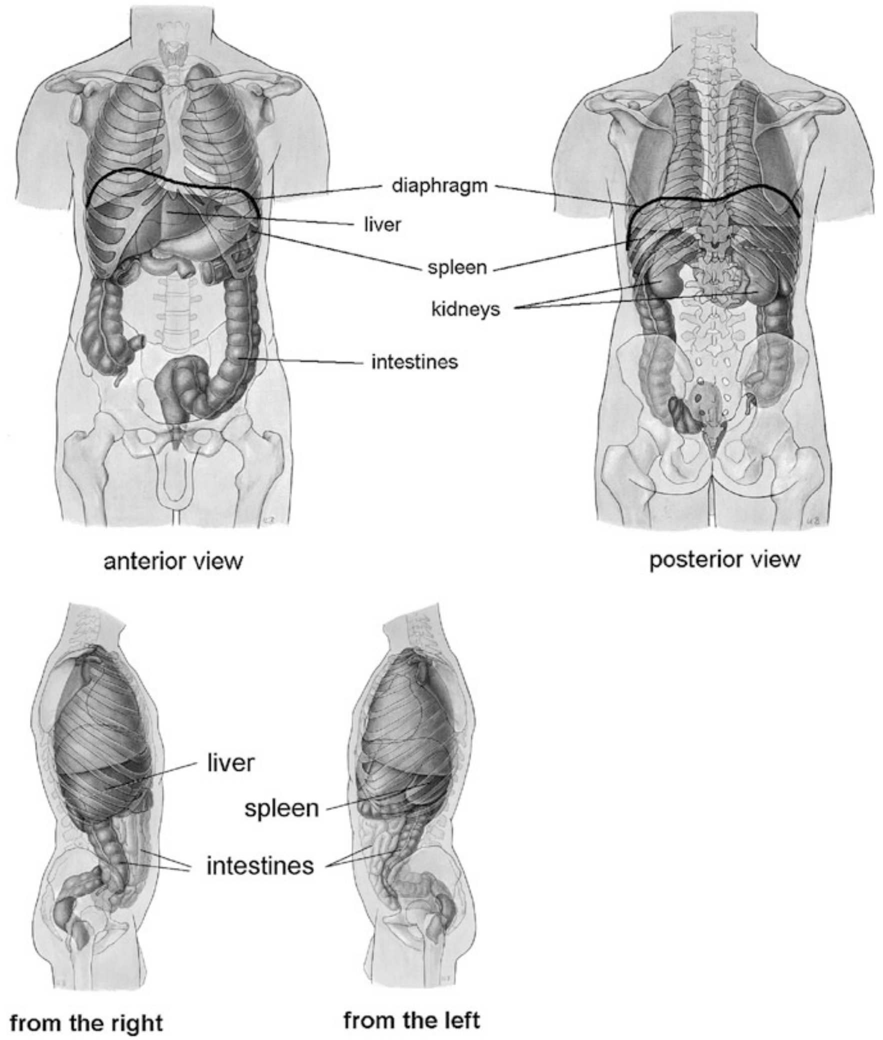

Fig. 6.1 The abdominal organs as a projection on the body surface (adapted from Sobotta 1997)

than if struck from the left (Fig. 6.2). As the lung, the liver can experience a central rupture where the tissue around the damaged part is not altered.

Examples of possible abdominal injuries and their classification according to the Abbreviated Injury Scale (AIS) are given in Table 6.1.

In automotive crashes, the vehicle interior offers several contact areas that strongly influence the injury outcome when hit. Possible contact points include the steering assembly, the side door, the arm rest, the dashboard and the glove compartment, whereas unbelted occupants are, of course, at higher risk to contact such structures than belted ones.

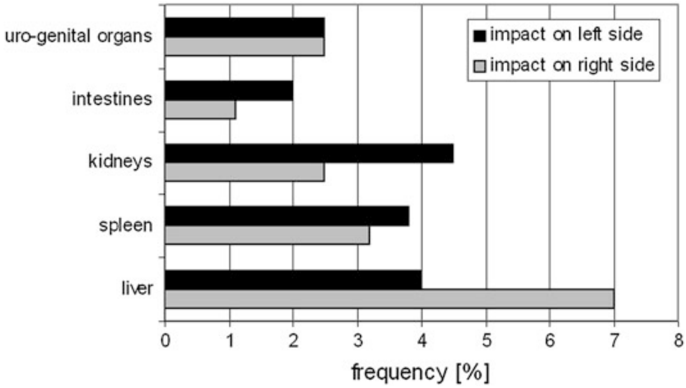

Fig. 6.2 Frequency of AIS >3 abdominal injury for different organs due to lateral impact on the *right* or *left* side, respectively (adapted from Rouhana et al. 1985)

Table 6.1 Abdominal injuries (AAAM 2005)

AIS code	Description
1	Skin, muscle: contusion (hematoma)
2	Spleen or liver contusion (<50 % surface area)
3	Major kidney contusion spleen: rupture
4	Abdominal aorta: minor laceration kidney/liver: rupture
5	Kidney: total destruction of organ and its vascular system
6	Hepatic avulsion (total separation of all vascular attachments)

Also the structure of the organs themselves is important in terms of abdominal injury. Solid organs were found to be injured more often than hollow organs. Furthermore, the pathological state of an organ can have a marked influence on the injury tolerance due to changes of the material properties (e.g. stiffness). Among other medical conditions, previous surgery resulting in adhesions inside the abdominal cavity is also suspected to predispose the subject to injury. The age of a patient was shown to affect the injury outcome in blunt trauma, with children and the elderly being at higher risk. Particularly for children anatomical reasons have to be considered as, for example, the abdominal region is proportionally larger than in an adult, the liver is less protected by the rib cage and thus at higher risk. However, analysing cases of children that sustained abdominal injury due to seat belt compression it was observed that first of all incorrect and/or inappropriate use of the restraint system lead to the injury (e.g. Arbogast et al. 2007).

6.3 Testing the Biomechanical Response

In the same way as for other regions of the human body, the biomechanical response of the abdomen is addressed in experimental studies utilising cadavers or animals. Several studies used impactors that completely covered the abdomen (in front as well as in lateral impact). However, the overall response curves obtained do not account for the non-homogenous nature of the abdomen. The location of impact (e.g. left side or right side) and the posture of the object during impact play important roles.

The test conditions chosen do, of course, also influence the outcome of the experiments. Analysing the biomechanical response of the abdomen, the so-called "fixed back" condition (in contrast to the "free back" condition) is often used. By fixing the back of the object during impact the influence of the spine is eliminated.

In terms of experimental procedure it turned out to be quite difficult to accurately determine the deflection of the abdomen. Mostly high speed videos were evaluated, sometimes determining the intrusion with respect to a fixed point (e.g. the spine) or measuring the compression relative to the other side of the object. Due to the poor quality of most of these videos, many studies were not able to determine the deflection and therefore reported the force history only. Others, as pointed out by Rouhana (2002), presented results that are not reliable.

Today, force-deflection curves obtained from cadaver tests are available for the lower abdomen in frontal impact (Cavanaugh et al. 1986; Nusholtz et al. 1988) (Fig. 6.3). Also Hardy et al. (2001) and Foster et al. (2006) reported response data from cadaver testing for different test conditions including loading due to airbag and seat belt.

For the upper abdomen in frontal impact, it is suggested to use the same data as for the lower abdomen until better data is available (Rouhana 2002). With respect to lateral impact, several cadaveric studies were presented performing sled tests, pendulum tests and drop tests. Drop tests especially addressed the impact on an arm rest, dropping the cadaver from a certain height on the arm rest (e.g. Walfisch et al. 1980). Force–time curves instead of force-deflection curves were presented (see above).

For blunt impact on kidneys, Schmitt and Snedeker (2006), Schmitt et al. (2006) carried out pendulum tests on human as well as porcine kidneys. It was shown that the kidney tissue failure is a predominately energy driven phenomenon. The visco-elastic material properties were described and force–deformation characteristics are provided.

In Fig. 6.4 possible injury mechanisms and the resulting injury are presented for the three organs injured most often.

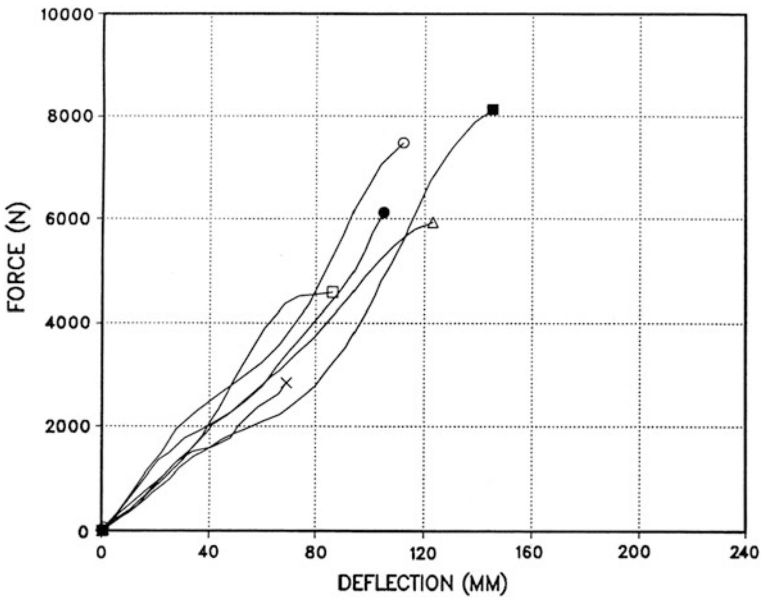

Fig. 6.3 Force-deflection characteristics for the lower abdomen obtained in frontal impact tests using a rigid impactor (adapted from Nusholtz et al. 1988)

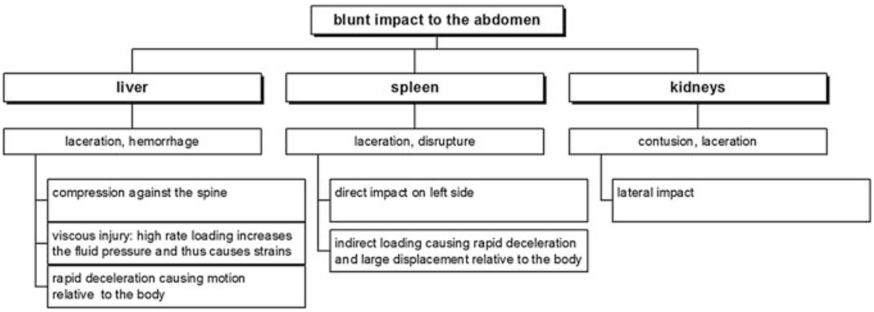

Fig. 6.4 Possible injuries and injury mechanisms for different abdominal organs

6.4 Injury Tolerance

Several experimental studies were performed to find ways to quantify the biomechanical response of the abdomen to impact and consequently to develop suitable injury criteria. The strength of various mechanical parameters as predictors of abdominal injury was investigated. Although various possibilities were addressed, not many conclusions in terms of tolerance thresholds could be drawn.

This, again, reflects the complex nature of the abdomen and the difficulties associated with the performance of adequate experiments. Consequently, further basic research is needed to elucidate the injury mechanism and appropriate injury tolerance levels. This paragraph summarises the most important attempts to quantify predictors for abdominal injury.

In general it is assumed that the force on an occupant should well correlate with injury outcome. Experiments with anaesthetised rabbits under condition of lateral impact, for example, confirmed this hypothesis by showing that the peak force correlated well with the probability of AIS $\geq$3 renal injury (Rouhana et al. 1986). However, it did not correlate with the probability of hepatic injury. Testing swine, Miller (1989) found that the peak force correlated well with the likeliness of AIS $\geq$3 and AIS $\geq$4 lower abdominal injury in belt loading. While no threshold values could be determined from the animal experiments, the results from human cadaver studies proposed a maximum tolerable force value of 4.4 kN (Talantikite et al. 1993).

In contrast, acceleration was found not to be a good measure for abdominal injury (Rouhana 2002). A major problem is the actual measurement of the acceleration. Typically the acceleration was obtained from accelerometers attached to the spine and the rib cage, respectively. Attaching accelerometers to those structures, they basically record the whole body acceleration. Thus a good correlation to injuries of the abdomen cannot necessarily be expected.

Bearing in mind that the solid organs of the abdomen are “fluid-filled”, a rate dependent behaviour was suspected. In fact, several studies including one by Mertz and Weber (1982), who performed tests on pigs, found a strong influence of the rate of abdominal compression and injury. Analogously to the thoracic impact, a good correlation to the injury severity was obtained when calculating the product of the maximum impact velocity V and the maximum abdominal compression C (e.g. Rouhana et al. 1984; Stalnaker and Ulman 1985). It was shown that for very low loading velocities (e.g. seat belt loading), the maximum compression was a better predictor of abdominal injury. For high loading velocities (e.g. airbag loading), the maximum velocity was a better injury predictor. For compressions and velocities in between, the product of V*C was found to be a better predictor than either the maximum velocity or the maximum compression separately. In addition, it was also shown that the product of maximum force F and maximum abdominal compression C correlate well with the probability of AIS $\geq$4 injury (Rouhana 1987).

With respect to the liver, Kallieris and Mattern (1984) observed laceration of the liver tissue of belted car occupants in lateral impact (near side impact) at an acceleration of 75 g or more. Analysing blunt impact of kidneys an impact energy threshold of 4 J, or a corresponding strain energy density of 25 kJ/m^3, were found to cause moderate to severe renal injury (Schmitt and Snedeker 2006a, b).

6.4.1 Injury Criteria

To date, only the European regulation for side impact testing (ECE R95) proposes a threshold level for abdominal loading. The abdominal peak force (APF) as determined by use of the EuroSID dummy is required to be less than or equal to 2.5 kN internal force (which is equivalent to 4.5 kN external force). ECE R44 (child restraint systems) requires a qualitative check of the belt position. Furthermore, visible signs of abdominal penetration are assessed after dynamic testing (using clay positioned in the abdominal cavity of the test dummy). In contrast to side impact dummies, current paediatric anthropometric test devices are not capable of accurately determining abdominal loading. The efforts to accurately model the visco-elastic behaviour of interior organs in dummies have largely remained futile; difficulties range from the reproducibility of mechanical properties of plastics to the problem of measuring e.g. deformations and deformation velocities inside plastic (e.g. foam) materials without influencing the material properties through the sensors (Rouhana et al. 2010). Apart from the APF criterion mentioned above, none of the currently available ATD's offers the possibility of reliably evaluating abdominal injuries.

6.5 Influence of Seat Belt Use

Within the scope of abdominal injuries, the influence of the seat belt is often discussed. Since the 1960s the so-called "seat belt syndrome" has been reported in the literature. Blunt impact to the abdomen is assumed to be caused by the seat belt either because of submarining and/or because of misplacement of the belt. Both submarining as well as misplacement are primarily related to lap belts or the lap part of 3-point belts, respectively. Submarining occurs in crashes with high change of velocity (Δv) when the occupant's pelvis manages to slip underneath the lap belt such that the lap belt then loads the abdomen. Hence, the structure of the seat strongly influences the probability of submarining. To prevent this movement, the seat base cushion often exhibits a wedge like shape at the frontal end, anti-sliding-airbags or other devices retaining the lower pelvis parts might be introduced. If the lap belt is not positioned properly (misplacement), i.e. if the belt is placed above the pelvis (spina iliaca superior anterior), it also loads the abdomen instead of the more stable pelvis. The correct placement of the belt is particularly crucial for children (e.g. Arbogast et al. 2004, 2007) and pregnant women. Nonetheless, it is important to state that pregnant women should definitely wear the belt. Moreover, the US National Highway Traffic Safety Administration advises pregnant women to position the lap belt low across the pelvis and not to disable the airbag. Restraints protect the foetus by protecting the mother, because maternal injury is predictive of foetal outcome and proper restraint reduces maternal injury (e.g. Klinich et al. 2008). Special devices to enable a correct path of the belt are commercially available. However, research is not yet fully conclusive about the injury risk of pregnant vehicle occupants (see e.g. Manoogian et al. 2007).

Besides submarining and misplacement as possible reasons for abdominal injury, the overall effectiveness of the seat belt was proven in many studies, showing that unbelted occupants are twice as likely to sustain fatal injuries as belted occupants (e.g. Langwieder et al. 1990; Lane 1994; Rouhana 2002). Furthermore, Langwieder et al. (1990) reported that up to 90 % of seat belt associated injuries are AIS1 injuries. Nonetheless, it can be suspected that the pattern of injury changes due to seat belt use. While the belt effectively reduces head, neck and thorax injury, it might possibly be responsible for more frequent but minor abdominal injuries if worn incorrectly (e.g. Harms et al. 1987; Rutledge et al. 1991). Likewise the potential influence of changes in belt design such as the introduction of new seat belt pretensioners should be considered and the systems must be checked to ensure that the induced abdominal loading is acceptable (Rouhana et al. 2010).

6.6 Abdominal Injuries in Sports

Blunt or penetrating trauma to the abdomen is rare. Despite being uncommon, the spleen is the most frequently injured organ in blunt abdominal trauma (e.g. after a collision with another play or after a blow to the abdomen) (Gannon and Howard 2010). Due to the high vascularity of the organ the consequences of such trauma can be severe. Much more often the literature reports cases of hernia and groin injury. The “sports hernia” can occur in athletes who participate in sports that require repetitive twisting and turning at speed (e.g. ice hockey, soccer, tennis, field hockey). However, in many cases, an actual hernia is not seen. Several theories exist in the literature regarding the causes of the sports hernia most of which implicate an overuse syndrome. Hip abduction, adduction, and flexion–extension with the resultant pelvic motion produce a shearing force across the pubic symphysis, leading to stress on the inguinal wall musculature perpendicular to the fibres of the fascia and muscle. Pull from the adductor musculature against a fixed lower extremity can cause significant shear forces across the hemipelvis. Subsequent attenuation or tearing of the fascia transversalis or conjoined tendon has been suggested as the source of pain. A systematic review on sports hernias can, for example, be found in Caudill et al. (2008). Other sources for—often chronic—groin pain include pubic bone oedema or entrapment neuropathies (e.g. Macintyre et al. 2006; Harmon 2007).

6.7 Summary

In the automotive environment, abdominal injuries are most often discussed with regard to seat belt (mis-) use and/or in lateral impacts. However, the biomechanical foundation concerning injury mechanisms and injury threshold values is still rather weak. Some test procedures such as ECE R95 (side impacts) consider the abdominal peak force as an injury criteria.

6.8 Exercises

E6.1: Why is the abdomen a particular “white spot” with respect to known biomechanics tolerance levels?

E6.2: Discuss whether a pregnant woman sitting on the front passenger seat should deactivate the airbag.

P6.1: Develop suggestions how the abdominal forces can be measured in a child dummy.

P6.2: Imagine you want to perform a cadaver test to determine the biomechanical response of the abdomen. One possibility is performing a pendulum impact test on a sitting cadaver (similar to the set-up shown in Fig. 5.8), the other option is performing a drop test on a lying cadaver (i.e. drop a weight from a certain height). In both cases an accelerometer is attached to the impactor such that the force–deformation characteristics can be determined. Discuss the pros and cons of each set up.

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Injuries of the Pelvis and the Lower Extremities

7

Injuries to the lower extremities play a major role in sports (soccer, skiing, etc.). They have furthermore emerged as the most frequent non-minor injury resulting from frontal vehicle crashes since restraint systems (belts, airbag in frontal impacts) are not particularly designed for the protection of the legs (and arms). Yet, injuries of the extremities are often the reason for long-term impairment (Håland et al. 1998; Crandall 2001).

7.1 Anatomy of the Lower Limbs

The lower limbs are commonly divided into pelvis, thigh, knee, lower leg, ankle and foot (Fig. 7.1). The pelvis which links the lower extremities to the spine is a ring of bones basically composed of four bones: two hipbones form the side and front walls while the sacrum and the coccyx form the rear wall (Fig. 7.2). Mechanically the pelvis represents the only load path to transmit the weight of the torso to the ground. Hence, its structure is quite massive. The hipbones consist of three fused bones (ilium, ischium, os pubis) and also host the acetabulum, a cup-shaped articular cavity forming one part of the hip joint. The pubic bone and the pubic symphysis, i.e. the joint connecting the right and the left pubic bone, form the frontal part of the pelvis. Particularly the thinner, frame-like parts of these pubic bones, the superior and inferior pubic rami, are often subjected to injury. At the rear wall of the pelvis, the sacrum is a fusion of the sacral vertebrae with sacral nerves (e.g. the sciatic nerve) that arise from the spinal cord passing the sacrum. Major blood vessels are also located near the sacrum and the coccyx.

As for the orientation of the pelvis, Fig. 7.3 illustrates the position of the bony structures in different postures. It is obvious that injury resulting for example from impact to the knee depends on the posture at the moment of impact.

Differences between the male and female pelvis are also apparent, but will not be discussed here. Although the shape and the mechanical properties of the male and female bones are slightly different, their influence on injury mechanisms concerning automotive impacts are regarded minor—in contrast to various

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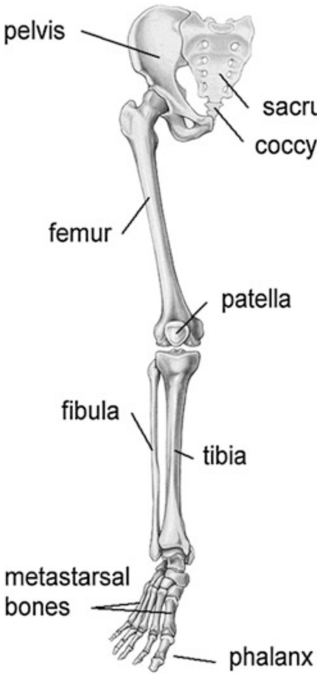
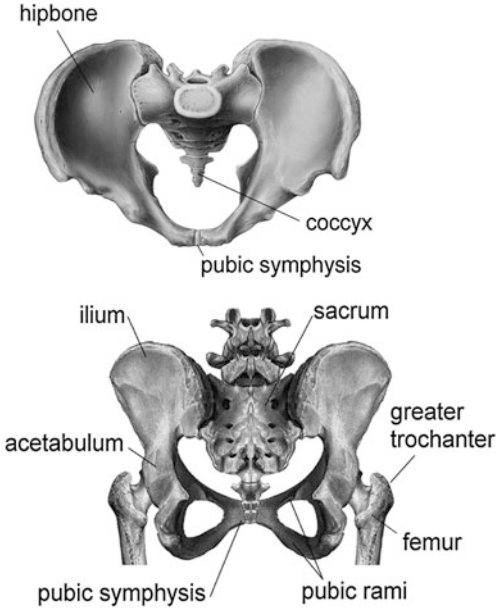


Fig. 7.1 Anatomy of the lower limbs (adapted from Sobotta 1997)

Fig. 7.2 The bony structures of the pelvic girdle (adapted from Sobotta 1997)



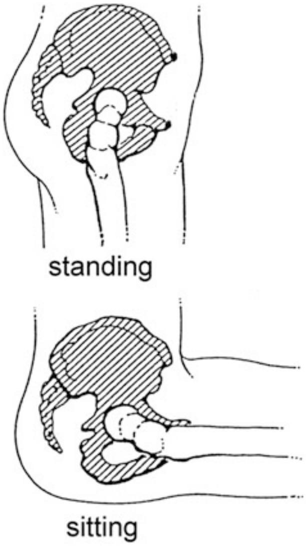


Fig. 7.3 Orientation of the hip in different postures: standing (*top*) and sitting (*bottom*) (adapted from Kramer 1998)

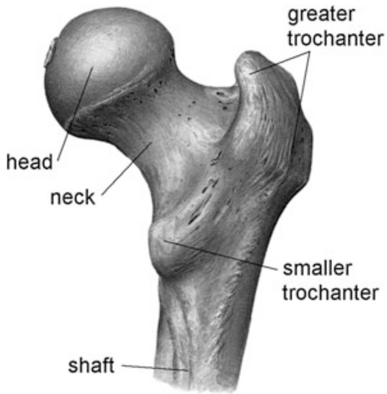


Fig. 7.4 Regions and landmarks of the femur (adapted from Sobotta 1997)

anatomical and physiological factors that might lead to different injury rates in sports injuries (e.g. Boles and Ferguson 2010).

The femur is the long bone of the thigh and is proximally connected by the hip joint to the pelvis and distally linked to the knee. The different regions and landmarks of the femur are shown in Fig. 7.4. Two bones, the tibia and the fibula, form the lower leg between the knee and the ankle. The knee is the joint that connects the femur and the lower leg (Fig. 7.1). It is an anatomically dense area involving several muscles, tendons, ligaments and menisci. Moreover, vulnerable structures of the knee like the patella are often subjected to direct impact. A strong

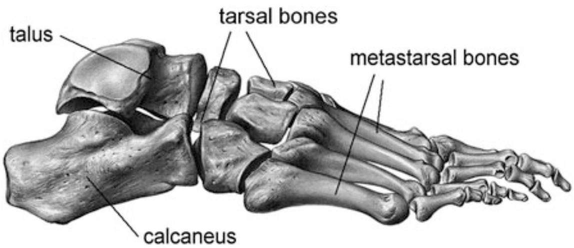


Fig. 7.5 Bony structures of the foot (adapted from Sobotta 1997)

musculature which can create considerable forces and thus may influence the injury mechanisms (see Sect. 7.2.2) surrounds the legs.

Finally the foot is adjoined to the lower leg. A foot consists of several bones: calcaneus and talus are located at the proximal end, the metastarsal bones and phalanx at the dorsal end (Figs. 7.1 and 7.5).

7.2 Injury Mechanisms

Regarding the pelvis and the lower extremities, fractures are the most common injuries sustained in accidents. Such fractures result from sports accidents or falls, respectively, rather than from automotive accidents. Hip fractures, for example, which are often caused by falls, particularly concerning the elderly, are a major concern in public health (Majumder et al. 2008). In contrast, pelvis fractures sustained in automotive accidents are quite rare. They contribute to only about 1 % the total Injury Priority Rating (IPR) (King 2002). Analysing frontal impacts of passenger cars, Kramer (1998) found that, while head injuries were the most common injuries sustained by 35 % of all occupants, pelvis and hip injuries were present in only about 7 % of all cases. However, 25 % of the occupants had leg and foot injuries. Similar results were obtained when evaluating the NASS data base (Crandall 2001). Moreover, checking for AIS $\geq$ 2 injuries (AIS, i.e. Abbreviated Injury Scale) in frontal impact, a strong influence of the restraint systems on the likeliness of injuries of the lower extremities was apparent. It was observed that the percentage of lower limb injuries is about twice as high than that for head injuries when the occupant is belted and the vehicle is equipped with airbags. The analysis of the different regions of the lower extremities showed that the feet and the ankles are at highest risk for AIS $\geq$ 2. Additionally, Morris et al. (2006) found that, based on UK accident data, lower extremity AIS $\geq$ 2 injuries are by far the most costly injuries and account for some 43 % of injury costs in both front and struck-side crashes.

Due to the fact that the pelvis and the proximal femur are often simultaneously injured, such injuries are commonly referred to as hip injury whereas the word “hip” does neither describe a particular anatomical structure that is injured nor is “hip injury” related to a particular injury mechanism. In the strict sense, the hip is the bony structure around the hip joint (femur head, pelvis, acetabulum). However, fractures involving the proximal part of the femur are commonly called hip fractures as well.

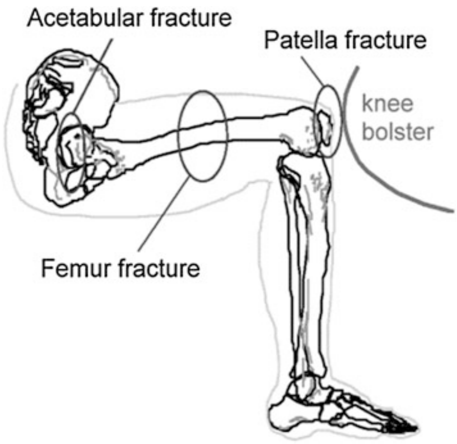


Fig. 7.6 Possible fractures due to impact to the knee (adapted from Crandall 2001)

Generally, fractures are either open or closed. While the skin and soft tissues overlaying the fracture are intact in closed fractures, the bone is exposed to direct outside contamination in open fractures. Further characteristics to classify fractures include the position of the fractured segments (displaced/undisplaced), the location of the fracture along the bone (intraarticular, metaphyseal, diaphyseal) and others (see e.g. Levine 2002). Concerning fracture of long bones in general and with particular respect to the bony structures of the legs, fracture patterns are differentiated depending on the loading condition that caused the fracture, i.e. the injury mechanism. There are four possible types of fracture mechanisms: direct loading, indirect loading, repetitive loading and penetration. In motor vehicle accidents direct and indirect loading are the most frequent types. If in a frontal collision the knee of an occupant hits, for example, the dashboard, direct loading can cause fracture of the patella, indirect loading may lead to fractures of the femur shaft or the acetabulum (Fig. 7.6). Different fracture patterns that can arise from direct and indirect loading, respectively, are presented in Fig. 7.7.

Like other injuries, injuries of the pelvis and the lower extremities are categorised in the AIS (Table 7.1). In the following sections possible injuries and their injury mechanisms are discussed; they are limited, however, to impact related injures only.

7.2.1 Injuries of the Pelvis and the Proximal Femur

Injuries to the pelvis are categorised clinically as isolated fracture of the pelvic ring, multiple fractures of the pelvic ring, sacrum fracture and associated injuries. An isolated fracture of the pelvic ring consists of a single fracture around the pelvic ring (for example of the pubic ramus or the ilium). Pubic ramus fractures are frequently observed in lateral impacts when the greater trochanter is hit. The pelvic ring remains stable after isolated fractures, i.e. significant displacement of the fractured

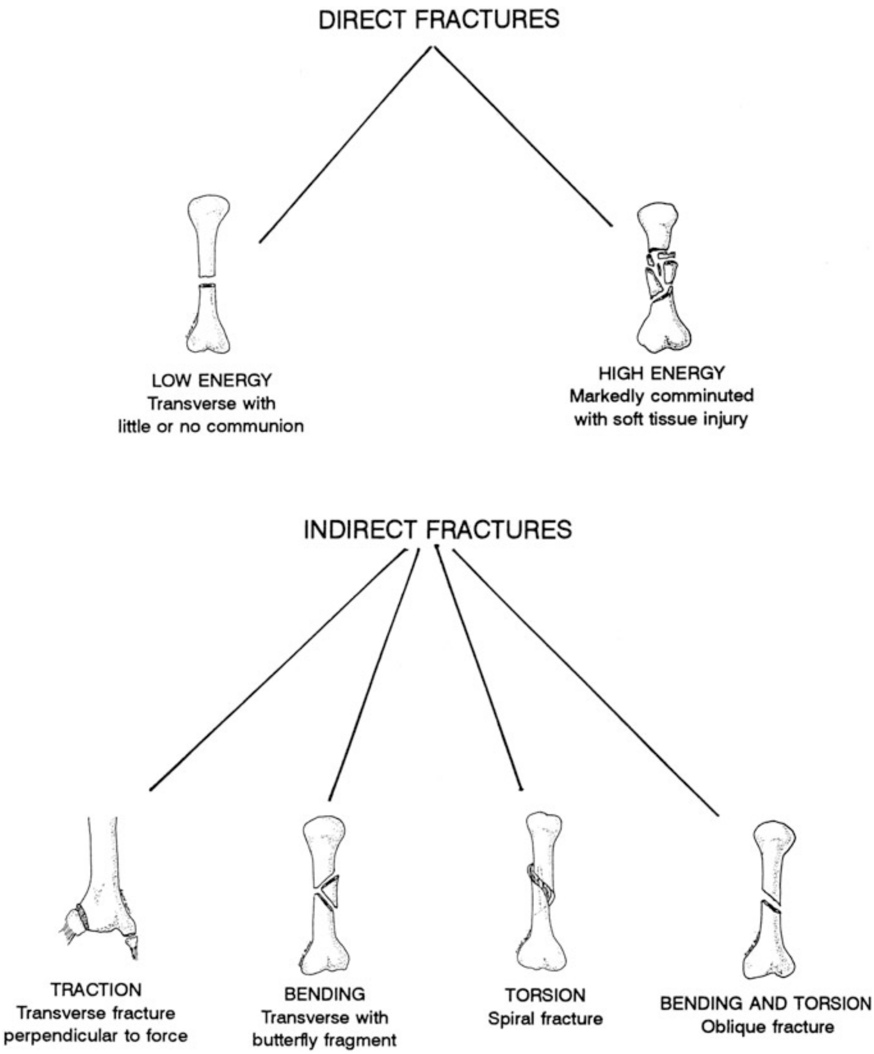


Fig. 7.7 Types of fracture according to loading (from Levine 2002). It should be noted that bending fractures can also occur directly, i.e. they are also a form of direct fracture

segments is not found. This is different if multiple fractures occur. Here the pelvic ring becomes unstable enabling large displacement of the fractured segments. Urogenital injuries can arise in combination with multiple pelvic fractures.

Sacrum fracture occurs in extensive pelvic injuries, fracturing usually across the foramina or in the vicinity of the holes through which the sacral nerves pass. Obviously, these nerves are also at danger in case of such injury. Moreover additional injury, especially haemorrhage, can be associated with pelvic fractures.

Table 7.1 Examples of AIS rated injuries of the pelvis and the lower extremities (AAAM 2005)

AIS code	Description
1	ankle, hip: sprain, contusion
2	patella, tibia, fibula, calcaneus, metatarsal: fracture pelvis: fracture (closed, undisplaced) toe: amputation, crush hip, knee dislocation muscles, tendons: laceration (rupture, tear, avulsion)
3	femur: fracture pelvis: fracture (open, displaced) traumatic amputation below knee
4	pelvis: "open book" fracture traumatic amputation above knee
5	pelvis: substantial deformation with associated vascular disruption and blood loss >20 % by volume
6	–

Excessive bleeding from the large blood vessel in the pelvic wall as well as from the fractured surfaces themselves can be life-threatening (even if proper surgery is applied).

From a biomechanical point of view, the underlying mechanisms of pelvic fracture are either compression, vertical shear or a combination thereof. Compression of the pelvis can further be differentiated into lateral and frontal (i.e. anterior-posterior) compression. Figure 7.8 illustrates the possible locations of fracture in case of lateral compression. If an anterior-to-posterior directed force compresses the pelvis centrally, so-called straddle fractures occur, i.e. multiple fractures of the pubic rami. Anterior-posterior (a-p) compression with impact forces on the right and left iliac crest can result in a hinge or "open book" fracture (Fig. 7.8). Thus in a-p compression the pelvic diameter is acutely increased causing tensile forces to act on tissue hosted by the pelvis and possibly violating ligaments. If the pelvis is subjected to vertical loading, shear can cause fracture as well as rupture of ligaments (Fig. 7.9).

Pelvic fractures used to be predominant in pedestrian collisions up to the early 1990s. When the pedestrian was struck by a car laterally on the pelvis, fractures of the pubic rami occurred (prevailing on the non-struck side). This type of injury has nearly vanished in today's pedestrian collisions. Due to changes in the front shapes and the front structures of recent cars, pedestrian kinematics seem to be different such that pelvic fractures are prevented (Otte 2002; Snedeker et al. 2003; Simms and Wood 2009).

As described above, the hip is frequently injured in falls. Such lateral loading of the hip commonly causes the femur neck to fracture. Femoral neck fractures in geriatric patients tend to be very low energy injuries (Levine 2002) in particular, as processes such as osteoporosis increase the fracture risk. Falls are furthermore of concern with epileptic patients whereby in addition to the hip also the head is

Fig. 7.8 Possible locations for fracture arising from lateral compression (adapted from Vetter 2000)

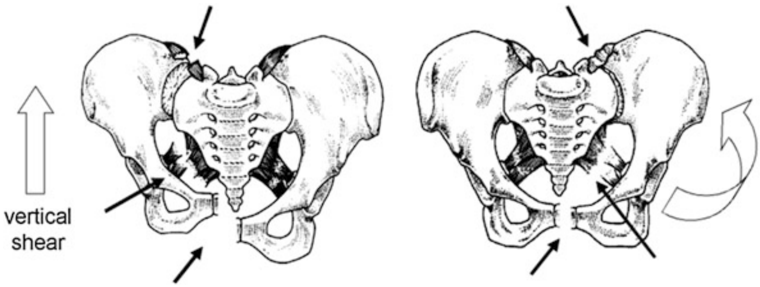


Fig. 7.9 Fracture and ligament rupture due to vertical shear forces (*left*). "Open book" fracture (*right*) (adapted from Vetter 2000)

primarily involved; yet, severe injuries are rare (e.g. Lawn et al. 2004). In turn, goalkeepers in soccer are frequently exposed to sideways falls due to their task (Schmitt et al. 2008a, 2009). Various protection devices in the form of (visco-elastic) padding are in use for hip protection of elderly as well as of athletes (e.g. Schmid Daners et al. 2008; Schmitt et al. 2008b).

In automotive accidents, however, fracture of the proximal femur is rare. Lateral impact is more likely to cause fracture of the pubic rami instead. Luxation and dislocation of the hip joint (Fig. 7.10), possibly in combination with fracture of the acetabulum, can also result from lateral impact. In general, the hip joint is dislocated when the hip is flexed and adducted and simultaneously loaded along the femur in rearward direction. These conditions apply, for instance, to occupants experiencing a frontal collision while sitting with their legs crossed.

7.2.2 Leg, Knee and Foot Injury

Addressing the injury mechanism for femur fractures sustained in frontal automotive collisions by interaction with the dashboard, it was found that axial compression accounts for the vast majority of fractures (62 %), followed by bending (24 %), torsion and shear (each 5 %) (Crandall 2001). In addition to the mechanical loading conditions, the fracture pattern of femur shaft fractures is

Fig. 7.10 *Left* Dislocation of the right hip. *Right* Dislocation and acetabulum fracture of the left hip after lateral impact. The head of the femur broke through the medial wall of the acetabulum (*courtesy* Prof. F. Walz)

influenced by the fact that the femur is not perfectly straight, i.e. it is slightly curved so that the convex side faces forward. This plays a role especially in indirect loading of the femur (Fig. 7.6).

Impact to the knee can cause the patella to fracture (direct loading of the patella). Indirect loading of the patella can likewise occur from strong muscle contraction (quadriceps) on the partially flexed knee, possibly resulting in patella fracture (Levine 2002). Of importance is furthermore the impact angle with respect to the upper leg at which the knee is hit (Meyer and Haut 2003). Unlike the hip joint which has a stable ball-and-socket joint (Sect. 7.2.1), the bone anatomy of the knee imparts little support to the joint's stability. This makes the knee ligaments prone to injury (see also Sect. 7.5). When the knee is bent and an object forcefully hits the tibia backwards, the posterior cruciate ligament can be torn (this injury is commonly called "dashboard injury"). In lateral impacts (e.g. pedestrian impacts) also collateral ligament rupture is seen. Generally complete dislocation of the knee joint can result in tears of the major four knee ligaments.

In analogy to femur fractures, tibia fractures may be caused through direct or indirect loading of the leg. Tibia fractures are the most common fractures in long bones, and because the tibia lies subcutaneously (i.e. the bone is only covered by the skin), such fractures are often open fractures. Most fractures occur between the mid-shaft region and the distal third of the tibia, where the smallest cross-section and the smallest cross-sectional moment of inertia are found. As for the impact conditions, Crandall (2001) reported that for tibia as well as for fibula injuries, axial loading and direct impact are equally probable. If both the tibia and the fibula are fractured, stability depends on the level of the fractures, i.e. the fracture is more unstable if both bones are broken on the same shaft level. A more severe type of tibia fracture is the tibia plateau fracture, i.e. a fracture of the tibia including a disrupture of the articular surface of the knee joint. Tibial plateau fractures make up for only 1 % of all fractures seen clinically, but account for 10 % of all below-knee AIS ≥ 2 injuries sustained in vehicle crashes (Crandall 2001). Mechanisms resulting in tibia plateau fracture include direct impact to the knee, axial loading

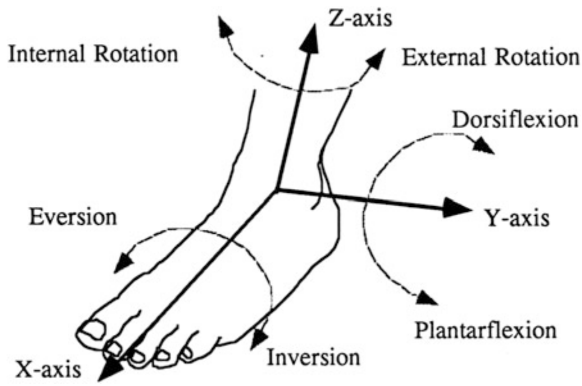


Fig. 7.11 Anatomical motions of hindfoot joints (adapted from Crandall et al. 1996)

and axial loading plus hyperextension of the knee. Axial loading might occur due to simultaneous floorboard intrusion and knee entrapment (e.g. if the knee gets caught under the dashboard).

The injury mechanisms of ankle and foot injuries are closely related to the possible motion range of the ankle and the hindfoot (Fig. 7.11). According to Crandall (2001), ankle injuries sustained in frontal automotive collisions are primarily due to axial load (58 %), inversion (15 %) and eversion (11 %). Only 5 % resulted from direct contact. Metatarsal injuries, however, are solely caused by direct impact and 100 % of all calcaneus injuries are due to axial loading. Pure axial loading can also result in talus fractures and so-called pylon fractures, i.e. intra-articular fractures of the distal tibia involving the calcaneus, talus and the distal tibia, for instance by driving the talus into the tibia. Dorsiflexion (e.g. forced in crashes with toepan intrusion) and entrapment of the knee can increase axial forces.

Inversion and eversion account for the vast majority of malleolar injuries, especially ankle fractures, making the ankle the most frequently injured major joint of the human body (see also Sect. 7.5). Furthermore, the rate of rotation, the orientation of the ankle and individual factors of the occupant like age or pre-existing damage were found to influence the likeliness of malleolar injury (Crandall 2001).

Foot injuries, particularly metatarsal fractures that were sustained in automotive accidents, result mainly from contact to the foot pedals. If, in addition, the heel strikes the floorboard, local loading to the calcaneus occurs, possibly leading to its fracture (Levine 2002).

Investigating the effect of preload in the Achilles tendon with respect to ankle injuries, cadaver experiments showed that a 2 kN Achilles preload affects pylon fracture by decreasing the external force needed to cause this fracture (Kitagawa et al. 1998a). Achilles preload also leads to tension fractures of the calcaneus. Additionally, injury mechanisms are influenced considerably by the musculature of the lower extremities. Cappon et al. (1999) report the posterior muscles (plantarflexors) to be able to generate more than 3.5 kN and for the anterior muscles (dorsiflexors) 1.4 kN were determined (Petit et al. 1996). Likewise, in pre-impact bracing muscles can contribute more than 2.8 kN to the load of the lower

Table 7.2 Mechanical strength (average values) of the bones of the lower limbs as reported by Levine (2002)

	Femur		Tibia		Fibula	
	Male	Female	Male	Female	Male	Female
Torque (Nm)	175	136	89	56	9	10
Bending (kN)	3.92	2.58	3.36	2.24	0.44	0.30
Average maximum moment (Nm)	310	180	207	124	27	17
Long axis compression (kN)	7.72	7.11	10.36	7.49	0.60	0.48

extremities (Crandall et al. 1995). Sled testing and computer simulations showed that muscle tension largely increases the axial forces, increases the effective mass and stiffness of the leg, accounts for redistribution of the stress within the bones, and alters the kinematics with less forward excursion and small joint rotation (Crandall et al. 1995). However, to gain a better understanding of the injury mechanisms of lower limb injuries, more studies addressing the role of muscle force are necessary.

7.3 Impact Tolerance of the Pelvis and the Lower Extremities

The response to mechanical loading of isolated bones such as the femur, the tibia or the fibula was measured from experiments similar to those used in material testing (e.g. bending and tension tests). Table 7.2 summarises such results.

To determine the biomechanical response of the pelvis cadaver tests subjecting the test objects to vertical, frontal and lateral loading were performed. Because of the use of very different test set-ups (e.g. rigid versus padded impactors) and different test procedures (e.g. instrumentation and posture of the test object), the results obtained for each load case are difficult to compare. As in some tests hardly any fractures were produced, the evaluation of the results is not at all conclusive.

To mimic the loading conditions of a automotive frontal collisions, Nusholtz et al. (1982) performed pendulum tests impacting a sitting cadaver at the knee. No pelvic or hip fractures were observed up to 37 kN impact force. Brun-Cassan et al. (1982) conducted whole-body impacts on unrestrained cadavers with peak knee forces ranging from 3.7 to 11.4 kN. No pelvic fractures were found, except for one case where the right patella and the iliac crest fractured at a knee load of 8.8 kN. Despite the differences found in the cadaver tests, a maximum of 10 kN for axial loading of the femur is included in FMVSS 208.

Concerning lateral impact, tests results reported in the literature are also diverse. In summary, it appears that neither the peak pelvic acceleration nor the peak pelvic deformation correlate well with the probability of pelvic fracture. However, Viano et al. (1989) found that the ratio of pelvic deformation to pelvic width (presented in percent compression) was a reliable measure for fractures of the pubic rami. 27 % pelvic compression corresponded to a 25 % probability of serious injury. This finding was confirmed by Cavanaugh et al. (1990), who also

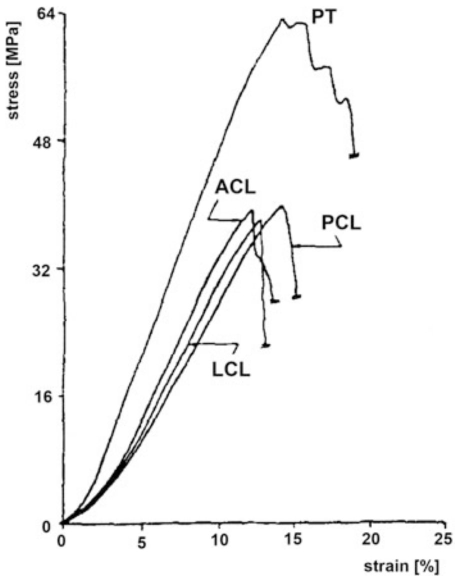


Fig. 7.12 Stress-strain curves for the patella tendon (PT), the anterior (ACL) and posterior (PCL) cruciate ligaments and the lateral collateral ligament (LCL) fascicle-bone units (adapted from Butler et al. 1986)

performed lateral impact on cadavers. A slightly higher percentage of tolerable compression (32.6 %) was determined to account for a 25 % probability of fracture of the pubic rami.

Also the (visco-elastic) material properties and the failure thresholds of soft tissue of the lower extremities were characterized by various experimental studies leading to a large spread of results. Since for tendons and ligaments the ultimate load is related to the cross-sectional area such variations are to be expected. The average ultimate tensile stress of tendons and ligaments as described in the literature ranges from 50 to 100 MPa. Depending on the experimental setup threshold values for failure of the knee ligaments due to strain are reported from 7 to 40 % (e.g. Butler et al. 1986; Kerrigan et al. 2003; Arnoux et al. 2006, Robinson et al. 2005). Figure 7.12 presents some typical stress–strain-curves for knee ligaments. Dynamic tests performed on cadaveric knees suggest a 50 % risk of the collateral ligament injury to be associated with an applied bending moment of 117 to 134 Nm (Ivarsson et al. 2004). However, in several other studies a wide range of failure thresholds is reported so that general conclusions can not be drawn.

With respect to the lower leg and the foot, various cadaver and volunteer experiments were conducted. The static response to axial loading was analysed for both cadavers and volunteers (Hirsch and White 1965). Force–deflection response was obtained. A comparison between the cadaver and volunteer results suggests that there is no difference between the compressive foot and ankle stiffness of the living human and the cadaver specimens (Crandall et al. 1996). Dynamic response in axial loading of the tibia and the ankle was also determined from cadaver studies (e.g. Petit et al. 1996; McMaster et al. 2000). Figure 7.13 illustrates a test set-up using a pendulum to impact the foot and the tibia. Reported failure loads for tibia

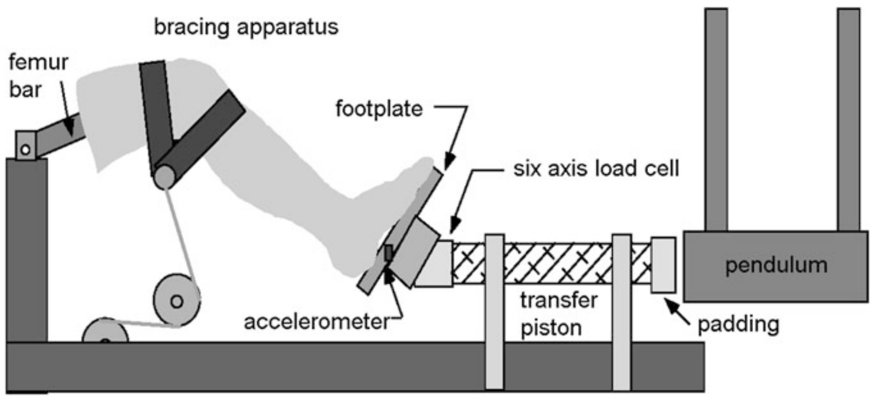


Fig. 7.13 Test apparatus to perform dynamic impact experiments (adapted from Crandall 2001)

fracture are for example 7.8 kN (Begeman and Prasad 1990) and 8.0 kN (Yoganandan et al. 1996), 7.3 kN are given for pylon fractures (Kitagawa et al. 1998b), and 8.1 kN were found for calcaneal fractures (Begeman and Prasad 1990).

For dorsiflexion/plantarflexion the biomechanical response was investigated in static and dynamic tests. From cadaveric studies it emerged that after 45° of forced dorsiflexion, there is a 50 % probability of ankle injury (Levine 2002). Rudd et al. (2004) concluded that an ankle joint moment of 59 Nm represents a 25 % risk of ankle fracture in dorsiflexion for a 50 percentile male. Parenteau et al. (1998), who performed quasi-static experiments, reported injury at $47.0 \pm 5.3°$ and 36.2 ± 14.8 Nm in dorsi-flexion, and at $68.7 \pm 5.9°$ and 36.7 ± 2.5 Nm in plantarflexion. Dynamic experiments resulted in injury at 138 Nm but exhibited a large variability of $\pm$ 50 Nm (Begeman and Prasad 1990).

Likewise eversion and inversion of the foot was analysed statically and dynamically. For quasi-static loading, Parenteau et al. (1998), for example, found a failure threshold of approximately 34 and 48 Nm for inversion and eversion, respectively. Figure 7.14 presents the according response corridors as reported by Crandall et al. (1996).

Bearing in mind the additional influence from active and passive musculature, preload as well as individual intrinsic factors, further experiments are necessary to improve the definition of impact tolerances of the lower limbs and thus to provide information for the development of accurate injury criteria, improved mathematical models and biofidelic dummies.

7.4 Injury Criteria

To assess loading of the pelvis and the lower extremities in crash tests, few criteria are established in existing regulations. This is not surprising as the total number of specimens tested in the underlying investigations concerning loading mechanisms

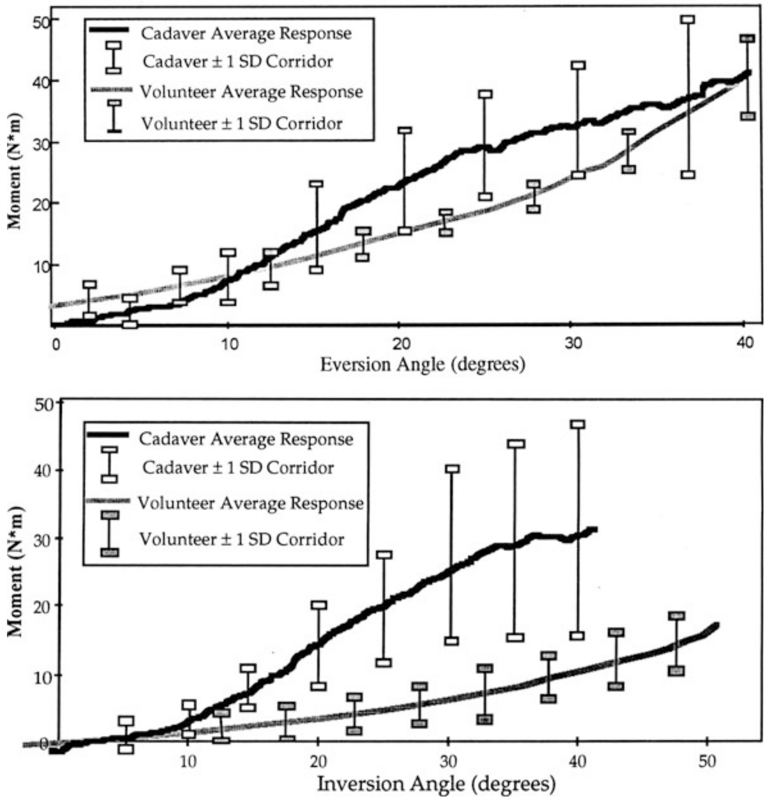


Fig. 7.14 Response corridors for eversion (*top*) and inversion (*bottom*) as obtained from quasi-static volunteer and cadaver tests (from Crandall et al. 1996)

is still small. However, additional criteria are proposed but not yet included into test standards. Complementary to the test procedures focusing on vehicle occupants, European regulations and consumer tests such as EuroNCAP address the leg injury risk of pedestrians (see also Sect. 2.6).

7.4.1 Compression Force

To protect the hip-thigh-knee complex a maximum compression force of 10 kN for axial loading of the femur is defined in FMVSS 208.

The tibia compression force criterion (TCFC) as defined in ECE R94 determines the force axially transmitted to each tibia of a test dummy. To date, the maximum threshold value for TCFC is 8 kN.

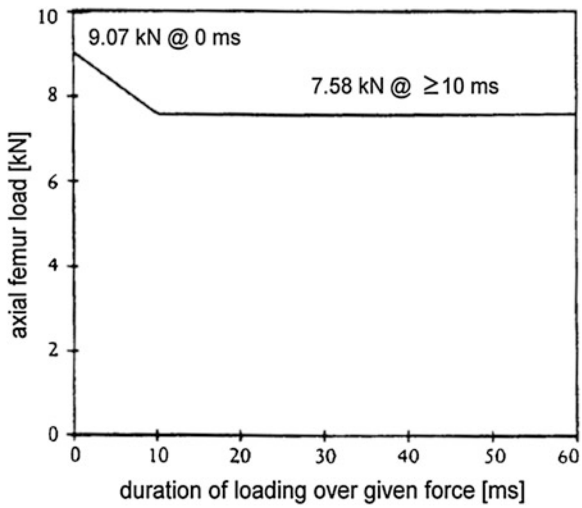


Fig. 7.15 Femur force criterion as defined in ECE R94

7.4.2 Femur Force Criterion (FFC)

FFC as defined in ECE R94 assesses the compression force acting on the femur as well as the duration (ms) for which the force is applied. The FFC is determined by the compression force (kN) that is transmitted axially on each femur. Figure 7.15 presents the force limits that must not be exceeded in a test.

7.4.3 Tibia Index (TI)

The Tibia Index (TI) involves the bending moments as well as the axial force in the tibia. The underlying idea of the TI is to prevent tibia shaft fractures. The TI is calculated according to the following equation:

$$TI = \frac{M}{M_{crit}} + \frac{F}{F_{crit}} \quad (7.1)$$

with M being the bending moment and F the compressive force. M_{crit} and F_{crit} represent critical intercept values and read 225 Nm and 35.9 kN, respectively, for the 50th percentile male. These critical values were obtained in static bending tests of the tibia (Yamada 1970). The maximum TI measured at the top and bottom of each tibia shall not exceed 1.3 at either location (ECE R94). As a further restriction, a maximum compression force was defined, i.e. the maximum compression force measured has to be smaller than 8.0 kN. Using scaling techniques, according critical values were determined for a 5th percentile Hybrid III female and for the 95th percentile Hybrid III male dummy. The detailed evaluation procedure including the required filtering is described in ECE R94.

7.4.4 Other Criteria

For side impact, the maximum strain on the pubic symphysis is taken as a measure for pelvic strain. The according criterion (cf. ECE R95) is called pubic symphysis peak force (PSPF) and shall not exceed 6 kN.

A maximum tibial displacement of 15 mm to protect the knee ligaments is laid down in ECE R94 (frontal impact). Moreover, a maximum ankle and foot load of 7.5 kN to protect the hindfoot and the ankle is discussed.

7.5 Pelvic and Lower Extremity Injuries in Sports

With respect to sports injuries of the lower extremity, several extrinsic and intrinsic factors and their influence on the injury risk are discussed (see e.g. Murphy et al. 2003). While several studies indicate that the injury incidence is greater during competition than in training sessions, controversial results are presented for the influence of skill level, shoe type or ankle bracing. As for intrinsic factors there is strong evidence that previous injury, especially when followed by inadequate rehabilitation, increases the injury risk while a correlation between injury risk and, for example, limb dominance, fitness status, body size or flexibility is not generally established or depends strongly on the study design. This inconsistency in the literature reflects the large variability represented by the different sports and the individual athlete and makes it difficult, if not impossible, to systematically present injury mechanisms and injury thresholds.

Concerning the lower extremities, injury to the muscle is frequent. Blunt impact, e.g. caused by an impactor's knee or by a fall, compresses the muscle and is the predominant mechanism for muscle contusion. Up to now the literature lacks suitable values for contusion injury risk. This might indicate that the prevalence of such injury depends strongly on individual factors. However, in a first and surely not yet conclusive attempt, Schmitt et al. 2009 estimated the risk to sustain hip contusion in soccer goalkeepers performing a dive to the side and landing on the hip to be in the order of approximately 110–125 N/cm^2.

In approximately 9–17 % after a direct blow to a muscle myositis ossificans traumatica develops whereas the incidence is also thought to be related to the severity of injury. Myositis ossificans traumatica is a nonneoplastic proliferation of bone and cartilage in an area previously exposed to trauma and haematoma. Its origins, relationship to other forms of bone proliferation (after surgery, congenital), and treatment are less than clear (e.g. Beiner and Jokl 2002).

Hamstring strains are common in various sports that involve running or sprinting and jumping, but are also common in dancing and waterskiing. A problem with this injury is the high recurrence incidence of 13.9–63.3 % (de Visser et al. 2012). It is suspected that hamstring strains develop during the later part of a swing phase when the hamstrings are working to decelerate knee extension, i.e. the muscle develops tension while lengthening. Consequently, the hamstrings must change from functioning eccentrically to concentrically which is suggested to make the muscle vulnerable to injury (Peterson and Hölmich 2005). Further risk factors such as age, previous injury, strength imbalance, flexibility of fatigue are discussed to contribute to the multifactorial nature of hamstring strain injuries (Opar et al. 2012).

The bony structures of the lower extremity can be subjected to various scenarios that result in direct or indirect fracture due to mechanical loads exceeding the fracture threshold (see Sects. 7.2 and 7.3). Additionally, repeated, but subcritical loading induces cumulation of microtrauma which can result in stress fractures (cf. Chap. 9). Stress fractures of tibia, femur or the metastarsal bones are observed in long distance runners or ballet dancers.

The high stability of the hip joint ensures that dislocation and subluxation are infrequently observed in sports, though not impossible. Anterior dislocations are sometimes reported from high-energy collisions in skiing and contact sports. Pelvic fractures, however, are unusual injuries in athletes (Anderson et al. 2001). The proximal femur is prone to fracture from direct loading (see Sect. 7.2.1) as well as to overuse injury in terms of stress fractures. Under normal conditions, the downward bending moment (force on femoral head times length of femoral neck) induces tension stresses and strains in the superior aspect of the femoral neck. These are counteracted by contraction of the abductor muscles producing a compensatory compressive strain on the superior aspect of the femoral neck. Now, if the gluteus medius muscle is fatigued, this neutralizing effect is minimized and tensile strains are experienced by the superior aspect of the femoral neck (Egol et al. 1998). Hence, in particular if loaded repeatedly stress fracture can occur. Other sources of hip pain include apophyseal avulsion injuries or tears of ligaments (e.g. Blankenbaker and De Smet 2010).

The various structures of the knee are prone to injury from direct or indirect loading. As already mentioned in Sect. 7.2.2 fracture of the patella can occur. Further patella disorders include patella tendon ruptures, patellofemoral pain syndrome or patellar tendiopathy (formerly known as “jumper’s knee”). The majority of knee injuries sustained in sports, however, concern the ligaments and the menisci. Evaluating over 7000 knee injury records, Majewski et al. (2006) found that most injuries were related to the anterior cruciate ligament (ACL) (20.3 %) followed by medial meniscus lesion (10.8 %), lateral meniscus lesion (3.7 %), medial collateral ligament (MCL) lesion (7.9 %), lateral collateral ligament (LCL) lesion (1.1 %), and posterior cruciate ligament (PCL) lesion (0.65 %). The activities leading to most injuries were soccer (35 %) and skiing (26 %). LCL injury was associated with tennis and gymnastics, MCL with judo and skiing, ACL with handball and volleyball, PCL with handball, lateral meniscus lesions with gymnastics and dancing, and medial meniscus lesions with tennis and jogging.

The mechanical function of the menisci is related to weight bearing, shock absorption, stabilisation and rotational facilitation. Failure of the menisci generally involves shear and compression. Meniscus tears are caused by a body rotation around the fixed and weight bearing knee. This can occur either in a combination of flexion and rotation or extension and rotation during weight bearing.

ACL rupture occurs most often in response to valgus loading in combination with external tibial rotation or to hyperextension with internal tibial rotation (Whiting and Zernicke 1998). The first mechanism can for instance be observed in a rugby or American football when the foot is on the ground, bearing weight, and another player contacts the lateral aspect of the knee increasing the valgus loading and rotation. Based on the second mechanism, ACL ruptures in basketball or gymnastics are recorded with the injury occurring after the athlete lands following a jump. With respect to skiing different mechanisms were reported that potentially result in ACL injury (e.g. Hunter 1999; Bere et al. 2011). Hunter (1999) differentiated: (1) valgus-external rotation (catching an inside edge and falling forward between the skis), (2) the boot-induced ACL injury (landing on the back of the ski with an extended knee, resulting in the boot forcing the tibia anterior as the front of the ski hits the ground) and (3) the phantom-foot phenomenon (falling backward between the skis, catching the inside edge of the downhill ski, driving the leg into forced internal rotation). Bere et al. (2011) who analysed injuries in world-cup alpine skiing identified: (1) slip-catch mechanism (losing balance while turning and leaning backward and/or inward causing internal rotation of the knee plus valgus), (2) landing back-weighted after jumping (tibeofemoral compression and anterior drawer of the tibia relative to the femur and (3) dynamic snowplow mechanism (internal rotation of the knee and/or valgus after losing balance and the unweighted ski drifting away from the athlete). Despite the different descriptions it is apparent that situations resulting in knee rotation and valgus play a major role with respect to ACL tears.

PCL rupture is less common in isolation and more commonly occurs in conjunction with injury to further knee ligaments (Voos et al. 2012). PCL injury is, for instance, observed when the tibia is forced posterior relative to the femur (see Sect. 7.2.2 "dashboard injury"). Figure 7.16 illustrates an example. Tearing of the PCL is also possible in a fall on the flexed knee pushing the tibia rearwards, if the knee is forced in flexion with the foot plantar-flexed or in rapid cutting on a minimally flexed knee. However, while it is generally accepted that PCL injury typically occurs in a flexed or hyperflexed knee position, the effects of the application of a tibial torque remain controversial (Whiting and Zernicke 1998; Senter and Hame 2006).

It is interesting to note that in many sports female athletes are at higher risk for injury than their male counterparts. This holds particularly true for non-contact ACL tears; the overall rate of ACL rupture was reported three times higher in female athletes than in male athletes (Sutton and Bullock 2013). Several causes are discussed for this higher incidence rate including differences in anatomy and physiology (e.g. quadriceps angle, ACL size, femoral notch dimensions, muscle strength, ligament dominance, hormonal factors) as well as dynamic neuromuscular imbalance. Reviews on this topic can be found in Dugan (2005), Boles and Ferguson (2010) or Sutton and Bullock (2013).

The ankle joint is among the most commonly injured joint in sports. 75 % of all ankle injuries are ankle ligament injuries, with the vast majority (85 %) of those ankle sprains caused by inversion injuries (also called supination) typically due to

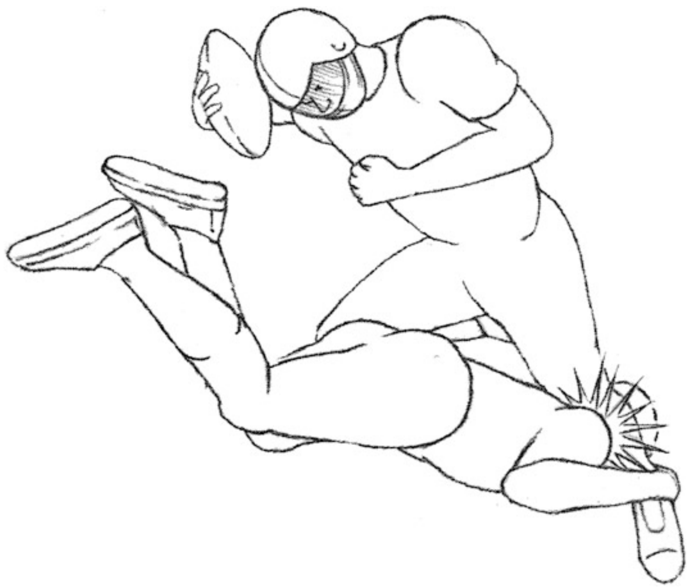


Fig. 7.16 PCL tears if the tibia is forced rearwards relative to the femur (adapted from Peterson and Renström 2002)

rolling the ankle while the foot is in contact with ground (Whiting and Zernicke 1998). Factors that strongly identify at-risk individuals include static foot measures (e.g. high longitudinal arch, greater foot width) and gait characteristics (e.g. more laterally situated gait). However, these findings on injury risk factors are still discussed controversially in the literature (Morrison and Kaminski 2007). With respect to further injuries of the ankle, the foot and the Achilles tendon, the reader is referred to Sect. 7.2.2. In addition overuse injuries to the foot (stress fracture) and the Achilles tendon (degeneration possibly even leading to rupture) occur and, similarly to the overuse injuries mentioned above, afflict mainly endurance athletes or military recruits (see Chap. 9).

7.6 Prevention of Lower Extremity Injuries

As already pointed out in previous chapters, the strategies to prevent injuries in sports are very much depending on the type of sports and the individual circumstances. Thus it is not surprising that the protective potential of several measures is discussed controversially. The use of prophylactic knee braces (e.g. Rishiraj et al. 2009) as well as the effect of shin guards to prevent tibia fractures (e.g. Francisco et al. 2000; Gorissen et al. 2012) are such examples.

With respect to the hip, various hip protectors are available for elderly and athletes. In principle such protectors make use of a hard shell to distribute the impact loading and/or padding to absorb energy (see also Sect. 3.6). Depending on

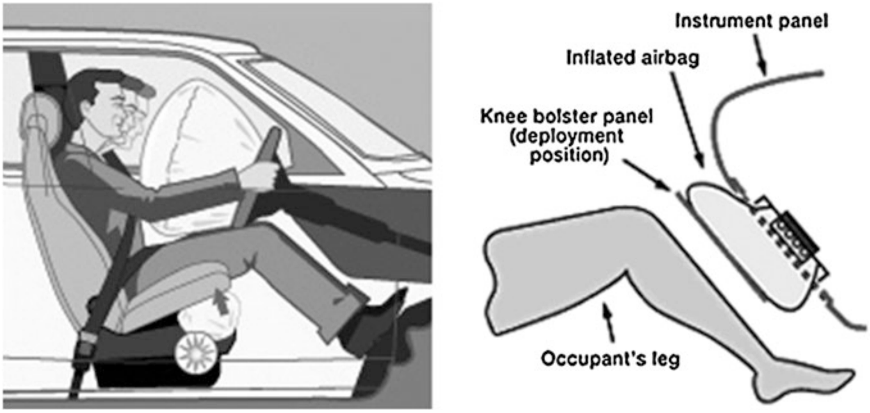


Fig. 7.17 Measures to prevent lower limb injuries: knee airbag (*right*) and anti-sliding-bag (*left*). The anti-sliding-bag also prevents submarining (see Chap. 6) (Autoliv 2003)

the design and the materials used the protective potential of the hip protectors varies significantly (e.g. Schmid Daners et al. 2008; Schmitt et al. 2008b).

In the automotive environment, in contrast, more agreement exists concerning countermeasures to reduce the number of injuries of the pelvis and the lower extremities sustained in automotive crashes. Knee bolsters that are located in the lower part of the dashboard were introduced to protect the knee from impact and also to provide an additional load path for the deceleration of the lower body in the absence of seat belts. To prevent knee ligament injury, it is important that the knee bolsters also contact the upper patella area in order to ensure that forces are induced axially to the femur. Likewise, knee airbags were developed to reduce loading on the knee in frontal collisions (Fig. 7.17).

To keep the occupant's knees and legs at a safe distance from the instrument panel and to prevent submarining (see Chap. 6), a system called anti-sliding-bag was presented. This system is installed in the front edge of the seat cushion and it prevents the occupant from sliding under the seat belt in a crash when activated.

Furthermore, the structure of the passenger compartment can be designed such that intrusion and intrusion velocity, respectively, are significantly reduced in the region of the lower limbs. With respect to foot injuries, a re-design of foot pedals is thought to reduce the loads transmitted. To help prevent debilitating leg and ankle injuries caused by deformation of the footwell, foot airbags to be placed underneath the carpet were developed. By lifting the heel in case of a frontal impact, dorsiflexion of the foot is reduced. Håland et al. (1998) showed that such airbags have the potential to reduce the foot acceleration by up to 65 %, the tibia force by up to 50 % and the tibia index by 30 % to 60 %.

While pedestrian sensing and warning technologies as well as autonomous braking systems are designed to prevent accidents, further measures must be introduced to reduce lower extremity injuries when an impact is unavoidable. The design of the front geometry contributes greatly to the injury outcome. Various studies indicated that pelvic and upper leg injuries are only prevalent in cars with rather sharp and high bonnet leading edges (e.g. Otte 1999, 2002; Simms and Wood 2009). Also the degree of bonnet leading edge roundness was identified as a relevant factor with regard to upper leg kinematics of pedestrian impact. An optimised contour of the car front including the bumper and the bonnet leading edge might thus help to avoid a large percentage of lower extremity injuries (Snedeker et al. 2003).

Bonnet ornaments can obviously be dangerous in case of an impact. Therefore, today, bonnet ornaments have to be mounted such they can flap or rotate and their shape should be designed such that no additional (focal) loading is transmitted to the pedestrian.

Another relevant aspect in pedestrian-vehicle collisions is the structure of the car front, particularly its deformation properties. The design of bumpers may be optimised to allow a higher degree of energy absorption. Several designs were presented showing that bumpers incorporating multi-density foams and a structural undertray (secondary load path) to support the legs of a pedestrian reduce impact forces and thus decrease the risk of leg and knee injuries. Moreover, the head lamp surrounding can be re-designed using a deformable housing which allows pushing the headlamps back into the car body to reduce the risk of injury.

The closing speed of contact between the thigh and car bonnet is a further critical factor determining injury risk. The closing speed is often not equivalent to the collision speed and may heavily depend on the radius of the bonnet leading edge. Reducing the closing speed by all possible means should therefore be aimed at.

7.7 Summary

Injury to the lower extremities is particularly frequent in sports where ligament ruptures or joint injuries are often observed. Elderly are at risk of hip injury due to falls. Furthermore pedestrians hit by a car often sustain lower extremity injuries and are therefore of special concern in the automotive field. Consequently the assessment of the leg injury risk is also part of European regulations and consumer tests (e.g. EuroNCAP). In addition current regulations related to the injury risk of vehicle occupants focus on limiting the maximum compression force.

7.8 Exercises

E7.1: Determine the bending moment M which acts in a femur due to a frontal impacting force F (e.g. dashboard impact) in the middle (i.e. at location a/2) as function of the angle α (see Fig. 7.18).

E7.2: Sideways fall: A hip protector in the form of a deformable visco-elastic lining is integrated in the underwear. The thickness of the lining d is 2 cm. As a worst case scenario, a person having a mass m = 65 kg falls onto the protecting

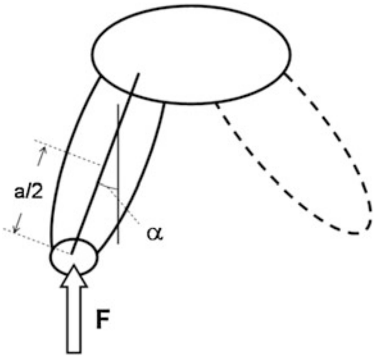


Fig. 7.18 Schematic of the force F impacting the femur

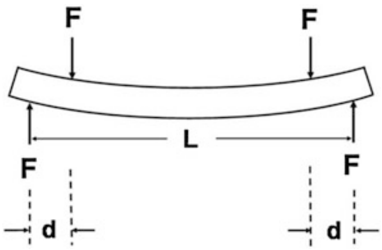
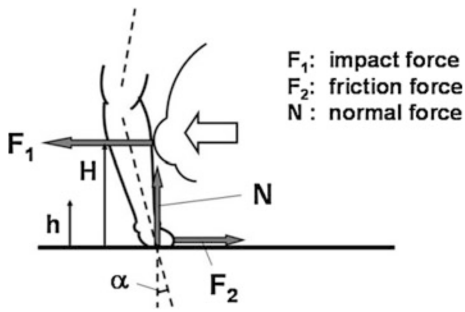


Fig. 7.19 Schematic of a four-point bending test of a bone specimen

Fig. 7.20 Schematic illustrating the forces acting on a lower leg of a pedestrian in case of an impact



layer from a height of h = 90 cm. Determine the average deceleration during the impact on the (undeformable) ground *a* as well as the average force *F* under the assumption of a pure translational motion of the body and no other contact, i.e., the body is in a horizontal position already as initial condition and touches the ground only with the protector.

P7.1: A bone specimen is tested using a four-point bending arrangement (Fig. 7.19).

(a) Determine the maximal deflection at the middle of the bone under the assumption that it exhibits a uniform cylindrical shape with circular cross section (radius r), and that the deformations are small such that linear elasticity (Young's modulus E) can be applied. Calculate furthermore the maximal tensile stress that occurs in the specimen.

(b) Compare the stresses due to compression and bending when, in addition to the four-point bending given in a), the specimen is loaded longitudinally by the force F1. The deformations can again be assumed to be small.

(c) How does the distribution of the normal stresses change in the cross section in case that the limit for plastic deformation is exceeded? Does this provide a possible explanation for the "Messerer wedge"?

P7.2: With increasing age the thickness of the cortical shaft of long bones decreases on the average. In turn, the diameter of the shaft tends to increase. Assume that the thickness of the shaft decreases by 20 % from the inside, how much diameter increase is necessary to compensate such that the area moment of inertia remains constant?

P7.3: A pedestrian is hit by the bumper of an older car (Fig. 7.20). Determine the bending moment in the lower leg up to the knee as function of the vertical coordinate h.

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Injuries of the Upper Extremities

8

In automotive accidents, injuries to the upper extremities have probably received least of all attention compared to other body segments. This lack of interest can partly be explained by the fact that such injuries are generally not life-threatening. Nonetheless, they may cause long-term impairment associated with significant societal cost.

In contrast to the automotive environment, injuries to the upper extremities are common in sports and therefore received considerable attention. Various studies addressed the kinematics of the upper extremities in different motion patterns like throwing, a golf swing or a tennis stroke. Many studies can also be found on the diagnosis and treatment of upper extremity sports injuries. With respect to injury mechanisms, however, many questions remain unanswered and concerning injury criteria and injury threshold levels basically no conclusive literature is available. In their systematic review on prevention of sports injury, McBain et al. (2012) confirmed the lack of research related to upper extremity injury, but also noted a change in research focus from equipment interventions to training interventions. However, to develop reliable measures to prevent injury, knowledge of the underlying injury mechanisms is needed.

8.1 Anatomy of the Upper Limbs

Generally, the upper extremities can be divided into four different parts: the shoulder (or shoulder girdle), the arm, the forearm and the hand. Figures 8.1 and 8.2 illustrate the corresponding bony structures.

The shoulder comprises scapula, clavicula and the joint articulations that attach the upper extremities to the torso. The arm is formed by the humerus and is linked to the shoulder by the shoulder joint which is probably the most mobile joint in the human body. The movement of the clavicula and scapula allows translation of the shoulder in horizontal and frontal planes. Additionally rotations about the three anatomical axes are provided by the shoulder joint.

K.-U. Schmitt et al., *Trauma Biomechanics*, DOI: 10.1007/978-3-642-53920-6_8,

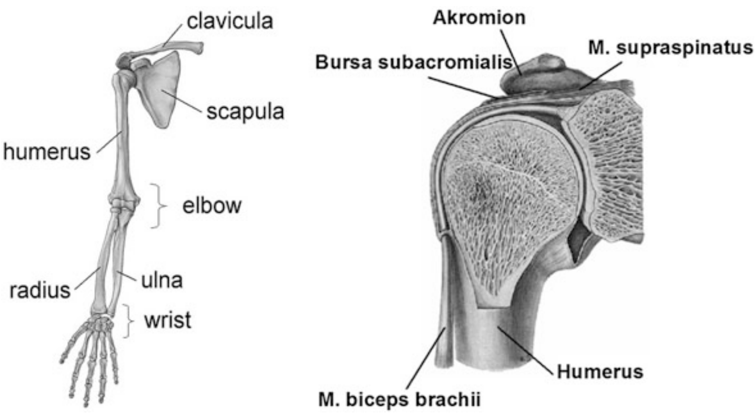

Fig. 8.1 Bones and joints of the *upper* extremities (adapted from Sobotta 1997)

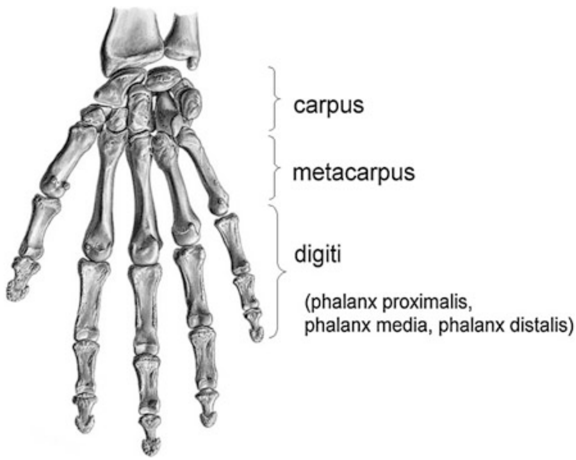

Fig. 8.2 Skeleton of the hand (adapted from Sobotta 1997)

The elbow joint connects the arm to the forearm which consists of the ulna and the radius. A much simpler joint than the shoulder joint, the elbow joint allows flexion of the forearm towards the humerus, extension of the forearm away from the humerus and one half of the forearm pronation/supination rotations. These pronation/supination rotations are completed by the ulna rotating at the wrist. The wrist joint, finally, connects the forearm to the hand. Associated muscles and soft tissues complete the four parts of the upper extremities.

It should be noted that there are differences between the upper extremities of males and females. Within the scope of trauma biomechanics, the most relevant differences are the mass and the bone mineral density. Both are lower in women with the bone mineral density also being further reduced with age.

8.2 Injury Incidence and Injury Mechanisms

Injuries of the upper limbs focus on fractures of the long bones. Of course, the soft tissues and the muscles may also be injured (e.g. skin abrasions due to airbag contact (Duma et al. 2003; Rath et al. 2005)), but these injuries play a minor role in the field of automotive accidents.

As for fractures, the classification presented in Chap. 7 also applies. Most common are clavicula fractures which occur, for instance, in direct blows, through compression during lateral impact of the shoulder or in falls on the outstretched arm. A typical fracture of the ulna is the so-called nightstick fracture which is a diaphyseal ulna fracture unaccompanied by a radius fracture. It results from low-energy direct impact (e.g. caused by an airbag) and is characterised by a transverse fracture through the ulna (see Fig. 7.8). Humerus fractures result mainly from direct impact, but can also occur without any contact. Some cases are reported in which muscle forces as involved in overhand throwing caused the humerus to fracture (Levine 2002).

Surveying vehicle crashes in the UK, Frampton et al. (1997) analysed upper extremity injuries of car occupants whose vehicle were not equipped with airbags. It was found that 86 % of all upper extremity injuries were at AIS1 level (minor abrasions, contusions, lacerations). Hence, 14 % formed AIS2+ injuries of which most injuries were fractures whatever the collision type. In frontal collision, forearm fractures were observed most frequently. Shoulder injuries were mainly found in struck-side crashes and rollovers. Clavicle fractures were identified to be the most frequent shoulder injuries. Humerus fractures were found in struck-side crashes but were not common in frontal and rollover crashes. Hand injuries were recorded in some frontal collisions.

Investigating a sample of 540 crashes where the driver airbag deployed, Huelke et al. (1997) found a total of 34 % of drivers sustained AIS1 upper extremity injuries and 3 % sustained AIS2+ injuries to the upper limb.

Kuppa et al. (1997) found an increase from 1.1 to 4.4 % in the occurrence of upper extremity injuries of severity AIS2+ as a result of airbag deployment. A study by Goldman et al. (2005) suggests that injuries to the upper extremity might become more common as a result of an increasing portion of the vehicle fleet being equipped with airbags. In contrast, Segui-Gomez and Baker (2002) who compared vehicles from model years 1993–1997 to vehicles from model year 1998–2001, noted a reduction of upper limb injuries in frontal crashes since the introduction of depowered airbags.

Analysing the US Crash Injury Research Engineering Network (CIREN) database, Conroy et al. (2007) found that the injury pattern differs for drivers and passengers. Only 24.8 % of all occupants had upper extremity injuries. One-half of the injuries to drivers were forearm fractures compared to one-third for passengers. Occupants in side impacts were more likely to have clavicle fractures (29.5 % for passengers vs. 17.1 % for drivers). Airbags were more likely to be a source of forearm fractures, but only 10 % of driver arm fractures with airbag deployment in frontal impacts were associated with airbag fling.

Airbag induced upper extremity injuries in side impacts were also noted by McGwin et al. 2008. Although they did not find an association between side airbag availability and the overall risk of upper extremity, an increased risk for dislocation and AIS2+ injury was observed. In the risk of fracture, however, there was no difference.

In summary, the following causes for upper extremity injuries were identified in the studies mentioned above:

- direct contact to airbag
- contact to interior of vehicle (including intrusions, e.g. in side impacts)
- contact of the arm with an interior part of the vehicle as a result of the arm being flung by the airbag
- inboard limb injuries due to contact with another occupant sitting next.

Furthermore, it was observed that clavicula fractures may be caused by the seat belt diagonal section lying across the outboard shoulder and thus transmitting the belt loads transversely across the clavicula.

Additional studies indicate that women are at higher risk in sustaining a AIS2+ upper extremity injury (e.g. Bass et al. 1997; Schneider et al. 1998; Atkinson et al. 2002). It is hypothesised that this is caused by the following factors: (1) women have generally smaller bones resulting in lower ultimate bone strength, (2) women experience an age-related loss of bone mineral density, (3) women are generally shorter in stature and therefore sit closer to the airbag system incorporated in the steering wheel, (4) young adult women tend to add bone to the endocortical surface, in contrast to men who add bone to the periosteal side, resulting in a lower resistance to bending (Schoenau 2001).

Finally, it should be noted that the occurrence of airbag induced upper extremity injuries depends, of course, on the characteristics of the airbag. The term "aggressiveness" is used to describe the influence of airbag design related parameters such as module (cover) design, pressure–time history, seam design and bag folding pattern. The aggressiveness is determined on a relative basis to assess the injury risk between different systems using devices like the Research Arm Injury Device (see Sect. 8.4).

8.3 Impact Tolerance

Early work by Weber (1859) and Messerer (1880) determined the load and moment required to produce failure in the bones of the human upper extremities. These studies remained the major reference data until upper extremity injuries received more attention again in the late 1990s. Several research groups addressed the biomechanical response of upper limbs gaining additional data by performing further impact testing. Table 8.1 summarises tolerance values for the humerus reported in the literature.

Table 8.1 Failure tolerances for the humerus

Humerus				Reference
Bending moment		Shear force		
Male [Nm]	Female [Nm]	Male [kN]	Female [kN]	
115	73	–	–	Weber (1859)
151	85	–	–	Messerer (1880)
157	84	1.96 (overall)		Kirkish et al. (1996)
230	130	2.5	1.7	Kirkish et al. (1996), scaled to 50 %ile male and 5 %ile female
138	–	–	–	Kallieris et al. (1997)
–	154	–	–	Duma et al. (1998a)
217	128	–	–	Duma et al. (1998b), scaled to 50 %ile male and 5 %ile female

Concerning forearm fractures, Bass et al. (1997) performed cadaver tests in which ulna nightstick fractures and multiple fractures were observed. The results suggest that the humerus position, the forearm pronation angle and the forearm position relative to the airbag module affect the risk of injury from airbag deployment. Furthermore, it was concluded that there is a forearm strength above which the risk of injury is low, even if the forearm is positioned in front of the airbag module. These findings support the hypotheses that women are at higher risk of sustaining upper extremity injuries.

Investigating the human forearm under a dynamic bending mode, Pintar et al. (1998) determined that the mean failure bending moment for all (male and female) specimens was 94 Nm. However, the bending tolerance of the forearm was found to be highly correlated to bone mineral density, bone area and forearm mass. Consequently, the study suggests that any occupant with lower bone mineral density and lower forearm mass is at higher risk to sustain a fracture.

Cadaver tests by Duma et al. (1998b) addressing the influence of the impact direction showed the forearm to be 21 % stronger in supinated position (91 Nm) than in a pronated position (75 Nm). Conducting additional tests with female forearms in the pronated position and scaling those results to match the 5th percentile female geometry, a tolerance value of 58 Nm was obtained. Given that the forearm is typically pronated in the driving position, the value obtained from this weaker pronation position is meant to represent a conservative injury threshold.

The difference between static and dynamic impact was analysed by Begeman et al. (1999). Bending tests of the forearm were performed both quasi-static and dynamic by using a drop weight which resulted in a loading rate of approximately 3 m/s. Fracture of the ulna or the radius occurred with an average dynamic peak force of 1370 N and an average moment of 89 Nm. Static fracture loads and moments were approximately 20 % lower. Nightstick or simple fractures were the most common type of failure. Differences between the radius and the ulna were not significant. In contrast to the work by Pintar et al. (1998) a correlation of the failure moment with age, cross-sectional properties, bone mineral content or moment of inertia was not found. As tests with one broken bone still showed a high failure moment, the authors suggest that other tissues may play a significant supportive role.

Regarding the elbow, Duma et al. (2001) observed that elbow injuries are caused not only by an axial force but also by a force that acts vertically relative to the horizontal forearm. Hence, a linear combination of the elbow axial and shear force showed a significant correlation to elbow injuries. Carrying out further cadaver tests, Duma et al. (2002) predicted, for the 5th percentile female, a 50 % risk of elbow fracture at a compressive load of 1780 N at an elbow angle of 30° superior to the longitudinal axis of the forearm.

Concerning the shoulder complex, several experimental studies using post mortem human subjects or volunteers investigated the response due to mechanical loading (e.g. Bolte et al. 2003; Compigne et al. 2004) as well as the mechanical properties of the entire shoulder complex (e.g. range of motion and stiffness, Davidsson 2013) or individual anatomical structures (e.g. the shoulder ligaments, Koh et al. 2004). Lateral and oblique impacts as observed in side or frontal-oblique collisions were the focus of these studies. The results reported provide important data, particularly with respect to the definition of response corridors for evaluation and improvement of the biofidelity of current anthropometric test devices. Furthermore, injury threshold values (especially for clavicle fractures) are given, but due to the complexity of the shoulder region the results are not yet conclusive.

8.4 Injury Criteria and Evaluation of Injury Risk from Airbags

To date neither governmental bodies nor consumer test organisations have released guidelines or regulations on how the risk of sustaining upper limb injuries in automotive accidents is to be assessed. There are no conclusive injury criteria or test porticoes implemented yet.

However, Hardy et al. (1997, 2001) presented the concept of the Average Distal Forearm Speed (ADFS) to assess the risk of forearm fractures. Based on static and dynamics airbag deployment tests using cadavers as well as other test devices (i.e. Hybrid III dummy, RAID, SAE arm, for details see below), it was concluded that the distal speed of the forearm which is a function of both the forearm mass and the forearm proximity to the airbag module is a good predictor of the likelihood of forearm fracture. Scaling the results measured to the 5th percentile female geometry, it was found that an ADFS value of 10.5 m/s corresponds to the 50 % probability of fracture. Furthermore, the ADFS decreased linearly with increased distance from the airbag module.

Further studies investigated the possibilities on how upper extremity injuries can be considered in the evaluation of crash tests and on how the aggressiveness of airbags with respect to upper limb injuries may be assessed.

To determine potential forearm fractures due to static deployment of driver airbags, Saul et al. (1996) found that the measurement of meaningful acceleration and bending moments was feasible using a specially instrumented 50th percentile male Hybrid III dummy arm. The arm correlated with airbag aggressiveness, proximity to the airbag module and relative position of the arm with respect to the airbag module.

Mounting the so-called SAE arm, which represents a specially designed and instrumented 5th percentile female upper extremity, to a Hybrid III dummy, Bass et al. (1997) determined that a forearm moment of 61 $\pm$ 13 Nm represents a 50 % risk of a ulna/radius fracture. The 50 % risk of both a ulna and radius fracture corresponded to a dummy forearm moment of 91 $\pm$ 14 Nm.

To enable reproducible evaluation of airbag aggressiveness, the Research Arm Injury Device (RAID) was developed (Kuppa et al. 1997). Using the device to investigate the interaction between the deploying airbag and an upper limb close to it, it was found that the orientation of the forearm with respect to the airbag module and the distance between forearm and airbag module were significant parameters with respect to the measured peak bending moment.

Despite the possibilities offered by the different arm test devices, inconsistencies between test objects were noted. Differences in the interaction of the shoulder-arm region with the side airbag were recognised by Kallieris et al. (1997) when performing static deployment test using cadavers and the Hybrid III dummy. Comparing the results of static side airbag deployment tests conducted with the Eurosid-1, an instrumented Hybrid III arm and cadavers, also Sokol Jafredo et al. (1998) observed great differences in the kinematics of the upper limbs. No correlation between the forces measured on the dummy and the cadaver could be established. Duma et al. (1998a) recorded kinematic differences between the SAE arm mounted to a Hybrid III female dummy and cadavers, but the moments recorded in the cadaver and the dummy were similar.

Several suggestions were made to improve current ATD design such that the kinematics of the upper extremities and the associated loading can be considered more accurately. Törnvall et al. (2007, 2008), for example, developed a new shoulder design for the THOR dummy which includes a humanlike clavicle and an improved joint.

8.5 Upper Extremity Injuries in Sports

Competitive and recreational athletes sustain a wide variety of soft tissue, bone, ligament, tendon and nerve injury to their upper extremities. As already described in previous sections, also in sports the upper extremities are prone to fracture, luxation, dislocation, (partial) rupture of tendons and ligaments as well as injury to muscles and nerves.

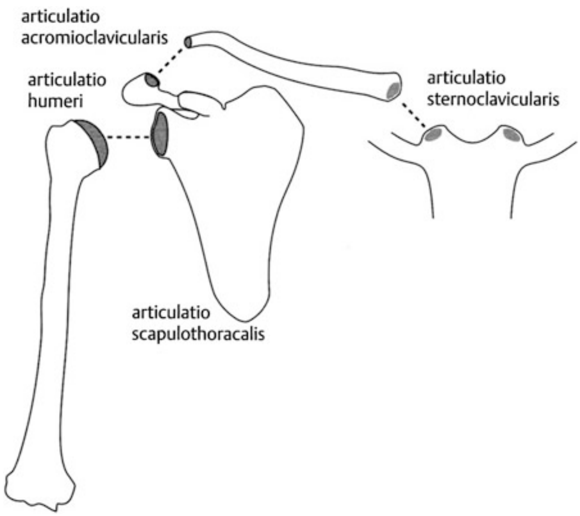

Fig. 8.3 Joints of the shoulder girdle (adapted from Brinckmann et al. 2002)

Particularly the various joints of the upper extremities are at risk. The four joints of the shoulder girdle (Fig. 8.3) together with elbow, wrist and hand allow for a wide range of motion and complex motion patterns to be performed. The ability of a joint to resist dislocation is directly related to its inherent stability. What a joint gains in mobility, it sacrifices in stability. This holds especially true for the shoulder which is prone to dislocation due to its relatively poor bony fit and limited supporting musculature.

Injuries involving the shoulder are a common consequence of sports participation. Sports with significant overhead demands such as tennis, baseball, volleyball and swimming often produce repetitive overuse syndromes. In contrast, injuries encountered in American football, hockey and other contact sports are often the result of direct trauma, e.g. fracture of the clavicle caused by a fall on the shoulder.

Traumatic shoulder injuries further include anterior glenohumeral instability (dislocation) due to a blow to the shoulder in the abducted and externally rotated position and, although less frequent, posterior glenohumeral instability which can, for instance, result from a heavy frontal shoulder charge in field games. Acromioclavicular sprain may be initiated by direct or indirect forces that tend to displace the scapular acromion process from the distal end of the clavicle. Furthermore, injury to the rotator cuff muscles or the acromion is observed as a result from force transmitted along an adducted arm pushing the head of the humerus against the coracoacromial arch.

Particularly in sports with overhead activities shoulder injuries, particularly overuse injuries, are common, frequently involving to the tendons of the rotator cuff muscles (e.g. Anderson and Alford 2010). It is suspected that the shape of the acromion, i.e. whether it is flat, curved or hooked, influences the prevalence of such injuries. Similarly it is suspected that impingement syndrome is associated

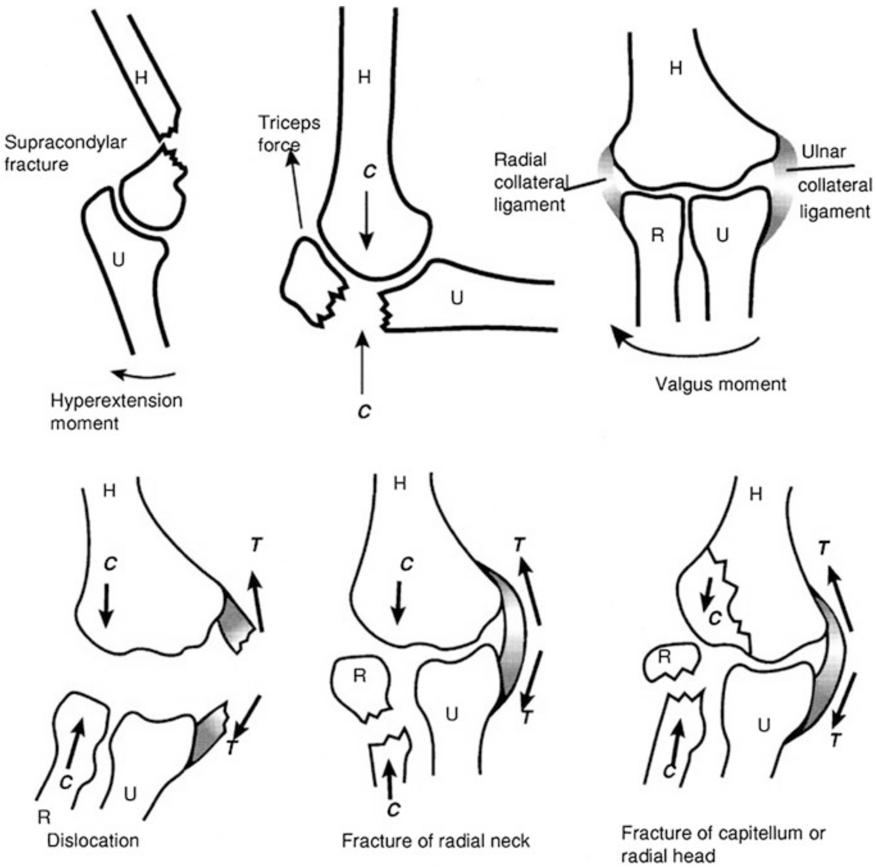

Fig. 8.4 Elbow fractures due to hyperextension moment, axial compression abduction moment (*top* from *left* to *right*), dislocations and fractures from a combination of abduction and hyperextension loading (*bottom* from *left* to *right*). *H* denotes for humerus, *R* for radius, *U* for ulna, *C* for compression and *T* for tension (adapted from Bartlett 1999)

with acromion shape. Here impingement syndrome refers to arm abduction that results in suprahumeral structures (most notable the supraspinatus tendon and the subacrominal bursae) being forcibly pressed against the anterior surface of the acromion and the coracoacromial arch (Whiting and Zernicke 1998). Further causes for shoulder pain can be related to biceps tendon disorders (e.g. rupture).

Due to the fact that the elbow is much more stable as the shoulder it is less likely to become dislocated. The bony structures of the elbow, however, are prone to fracture from direct blows as well as from indirect loading. Figure 8.4 summarizes different types of fracture depending on the mechanical impact.

Furthermore, particularly in racquet sports and sports involving throwing, the elbow often sustains nerve injuries (Keefe and Lintner 2004) and overuse injuries such as epicondylitis, tendonitis, myotendious strain and osteochondrosis of which epicondylitis is the most common. As a result of repeated loading existing microdamage increases, progressive tissue degeneration is observed until the tissue eventually exhibits inflammatory response. Lateral and medial epicondylitis are differentiated. Lateral epicondylitis is a degenerative condition of the tendon fibres that attach on the bony prominence (epicondyle) on the outside (lateral side) of the elbow. These tendons are responsible for anchoring the muscles that extend or lift the wrist and hand. A large number of tennis players are affected by lateral epicondylitis (hence it is also called tennis elbow). In racquet sports several causes are suspected to abet lateral epicondylitis including bad technique (particularly in backhand strokes), off-centre ball contact, grip tightness and racquet vibration (Whiting and Zernicke 1998). Medial epicondylitis in comparison affects the flexor tendon origin on the medical epicondyle and is experienced by throwers, golfers and tennis players (forehand and service stroke).

Concerning forearm and wrist, fractures of the distal radius are most common in sports. Loading on the outstretched arm with (hyper-) extended hand, e.g. a fall in inline skating or snowboarding, induces compression forces that can lead to fractures. Several classification systems have been proposed to describe distal radial fracture using either clinical (radiological) criteria or characteristics based on injury mechanism similar to the classification described in Sect. 7.2. Additionally Bancroft (2013) differentiated wrist injuries caused by high-impact-sports (e.g. inline or ice skating, snowboarding, alpine skiing) and low-impact-sports such as tennis, golf or basketball. Using this systematic, high-impact injuries range from displaced fractures and dislocations to ligamentous and acute tendinous tears while, for example, nondisplaced fractures, contusions or ligamentous strain represent low-impact injuries.

In the context of radius (and ulna) fractures, the ulnar variance is often referred to. Ulnar variance is defined as the ratio between the length of ulna and radius, i.e. it characterises the length difference between the articular surfaces at the distal radial-ulnar joint. If the two bones are of the same length, the ulnar variance is zero. Positive ulnar variance implies that the ulna is relatively longer than the radius, as determined from a radiograph at neutral rotation. Most often a person shows a small negative ulnar variance, with the radius taking approx. 80 % of compressive forces submitted via the hand (Whiting and Zernicke 1998). A general correlation of the ulnar variance to the fracture risk is, however, not (yet) established. Particularly in young athletes, repetitive injury to the radial epiphysis prior to skeletal maturity is suspected to result in premature closure of the growth plate leading to an ulnar variance. This phenomenon is, for example, observed in gymnasts whose wrists are subjected to high loading during exercises often leading to injury and overuse syndromes (called "gymnast's wrist") (Markolf et al. 1990; Bancroft 2013). Despite these observations, general guidelines on how much loading the wrist may withstand are still not available due to the influence of too many individual factors.

Table 8.2 Studies using cadaveric arms to investigate the wrist injury risk and the efficiency of wrist guards

Reference	Set-Up/Impact conditions	Positive effect of wrist guard?
Giacobetti et al. (1997)	75° dorsiflexion impact velocity: 25 mm/s Average fracture load = 2245 N	No
Lewis et al. (1997)	60–70° dorsiflexion 63.5–152.4 cm drop height	Yes
Moore et al. (1997)	75° dorsiflexion 78–104 cm drop height 16 kg impactor mass	Yes
Greenwald et al. (1998)	75° dorsiflexion 40 cm drop height 23 kg impactor mass	Little effect under high loads, possibly prophylactic effect in low-energy impacts
Staebler et al. (1999)	75° dorsiflexion loading 100 N/s	Yes
McGrady et al. (2001)	30.5/76 cm drop height	Yes

In contrast, there are several studies available analysing the loading of the wrist and the forearm due to forward or backward falls. In experimental studies the incidence rate of fracture was investigated by loading cadaveric arms (with and without wrist guard) in conditions representing a fall on the outstretched arm. Generally a wide spread of data was observed. Giacobetti et al. (1997), for instance, report an average force to fracture of an unprotected arm of 2245 N (ranging from 1470–4116 N). Table 8.2 summarizes further cadaver studies on the loading of the wrist. As can be seen, a variety of experimental set-ups and testing conditions were used which makes it difficult to compare the results. In addition experiments using a mechanical model of the forearm (similar to the arm of a crash test dummy) (e.g. Kim et al. 2006) or volunteers to investigate the loading of a wrist (e.g. Hwang et al. 2006; Schmitt et al. 2012a) are presented. Also computer models were used to analyse the fracture risk, for example with respect to influence of the impact direction (Troy et al. 2007).

Several parameters were identified that influence the risk of sustaining wrist fracture including the degree of elbow flexion, the impact direction, the impact velocity and the effective mass (Chiu and Robinovitch 1998; Chou et al. 2001; DeGoede et al. 2002a, b; Kim and Ashton-Miller 2003; Schmitt et al. 2012a). The elbow flexion, for instance, has a significant influence on the effective mass that loads the forearm at impact. Flexing the elbow when landing reduces the effective mass and thus also the impact force, i.e. the injury risk is reduced. However, depending on the set-up of the underlying experiments the effective masses reported differ greatly. While Kim et al. (2006) or Schmitt et al. (2012a) found the effective mass (for one arm) to be approximately 5 % of the body weight or 3–5 kg, respectively, Greenwald et al. (1998) report more than 20 kg.

To prevent wrist fractures various designs of wrist guards are available. These guards mainly aim at transferring load from the hand to a larger area of the lower arm in case of a fall; additionally some prevent abrasions. The evaluation of the protective potential of wrist guards has received significant attention (e.g. Schmitt et al. 2012b). Depending on the study design, some works report a benefit from wearing wrist guards while others did not find a significant injury reducing effect (see also Table 8.2 for studies using cadaveric arms). For throughout reviews on this topic the reader is referred to Russel et al. (2007) or Michel et al. (2013).

With respect to the hand, mostly fractures of the metacarpals and phalanges along with sprain and rupture of the collateral metacarpophalangeal and interphalangeal ligaments occur, particularly in contact sports (e.g. Bartlett 1999). Although injuries to the hand diminish an athlete's ability to perform and often result in a loss of playing time, many of these injuries may be treated effectively nonoperatively such that the athlete can return to sport rapidly (Snead and Rettig 2001).

8.6 Summary

In sports the upper extremities are often injured. In addition to fractures and ruptures, dislocations of the different joints as well as overuse injuries are frequently observed. In automotive accidents injuries to the upper extremities are discussed in the scope of airbag induced injuries. Currently no standardized test procedure includes assessment of the corresponding injury risk and there are no commonly accepted injury criteria available. The biomechanical response with respect to fracture of bony structures or rupture of ligaments, however, was established in various experiments.

8.7 Exercises

E8.1: List and discuss the most common causes for arm injury in automotive accidents.

E8.2: How is the risk of forearm injuries assessed in side impact tests according to ECE R95?

E8.3: What do you regard as the most likely arm injury in a swimmer?

E8.4: The shoulder joint is designed similar to the hip joint. Why is the shoulder more prone to injury than the pelvis (hip) joint?

P8.1: Assume a side impact. Vehicle A strikes the driver side of vehicle B with a speed of 14 m/s (50 km/h). To prevent injury of the driver, vehicle B is equipped with a side airbag. Estimate the time available to analyse the impact characteristics, make the fire/no-fire decision and eventually inflate the airbag. What kind of sensor(s) would you use to trigger a side airbag?

P8.2: Imagine you have developed a new wrist guard for snowboarding. Design a suitable experimental set-up to prove the effectiveness of the wrist guard to prevent radius fractures. Discuss whether such wrist guards have the potential to increase the injury risk for shoulder injury.

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9 Impairment and Injuries Resulting from Chronic Mechanical Exposure

An accident is defined as a violent, unusual and possibly harmful event that occurs suddenly and unexpectedly and is mostly of a short duration. Persons involved in an accident can in general not or only insufficiently react in order to prevent injury. The term "chronic", in contrast, implies that a process extends over durations which are long in comparison with typical accidental time intervals. Accordingly, physical and mental reactions of the persons involved always occur and cannot be neglected. Of primary importance is however the fact that under conditions of chronic mechanical overexposure, impairment of health and injury may result from an amount of mechanical load to which the body is exposed or of functional misuse in the course of some physical activity that is well below an acute tolerance limit for an individual (such as is described in the other chapters of this book), but whose effect is aggravated and outperformed by an extended duration during which it acts. The exposure may thereby often be interrupted and limited to regularly or irregularly occurring time intervals, e.g., to training periods (sports) or work assignments (jackhammer) which may however extend over years. A single, isolated exposure as such is usually mostly harmless or allows for a straightforward recovery. As a result, causation, mechanisms, patterns, tolerances, prevention and mitigation of adverse health effects are quite different from what is observed in accidents.

A distinction between injury and disease is often not well defined. Long-term sequelae of exposure to potentially harmful mechanical loads associated with a certain profession, e.g., in construction work, are considered as occupational illnesses and treated as such. Or, the Repetitive Strain Injury (RSI) syndrome that is observed among other in tennis, is denoted a disease rather than an injury despite its name because it manifests itself as a form of impairment representing the final clinical result of a long sequence of numerous microscopic injuries. Objective, quantitative diagnostic procedures are furthermore often difficult to develop (see, e.g., Gouttebarge et al. 2009).

Mechanical exposures which may cause health problems appear in quite various forms. The spectrum of mechanical loading scenarios thereby reaches from stochastic sequences of single, isolated impacts during certain periods of work, training or games, such as occurs for instance in boxing, to continuous vibrations,

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e.g., loud sound. Of comparable variety are the characteristics, professions, activities and social environments of the involved persons. Accordingly, reporting and documentation practices, intervention of authorities, legal subsumption, regulations, liability issues and insurance coverage are quite diverse and different from procedures that come into play after an accident with injury. Likewise, government agencies in charge of chronic disease control—regardless of the origin of the impairment—have in general little relation with traffic safety or general accident and risk management.

Chronic mechanical overexposure, overuse or functional misuse can manifest itself in a variety of different forms and severity:

- A worker using a drill hammer routinely can suffer from TFCC (triangular fibrocartilage complex) lesions.
- After a strenuous hiking tour, our feet may be covered by painful, but harmless blisters.
- A promising career in sport may be terminated by too much and inappropriate training.
- Repeated exposure to loud music in a recreational facility can induce a permanent hearing loss.
- Lower back pain leading to partial disability can result from routine household work in an unphysiological position.
- Hypertension forces the heart to perform an increased, ultimately harmful work load to maintain a physiological cardiac output.
- The heart of a bicycle or rowing champion who develops physiological hypertrophy resulting from intensive training needs careful attention and a well planned period of regression after discontinuation of the competitive activity.

In traffic, safety is almost exclusively devoted to acute situations, while potentially harmful chronic loading is generally controlled by comfort-oriented and ergonomic design of the user environment. Occupational illnesses and injuries, in particular those which are due to chronic mechanical exposure, have in turn extensively been examined in the past because of their high socio-economic significance, political implications, insurance claims and workman's compensation programs. In industrialised countries, workplace safety is therefore governed by numerous regulations and subjected to rigorous control by government bodies and insurance companies. The Occupational Safety and Health Act (OSHA) was passed in the USA in 1970. Since then, workplace safety has been given increased attention and impressive results were reached, although de-industrialisation may be a significant collateral cause (Table 9.1). On an international level, the International Labor Organization (ILO, www.ilo.org/) has a monitoring task and issues, among other, recommendations in view of workplace safety.

A Pandora's box opens itself when the conditions in less developed countries are considered: Mine workers in insufficiently equipped mines, farm workers exposed to hazardous pesticides, factory workers (e.g. textiles) in dangerous factories, etc. are of concern. Countermeasures are evident.

The situation with respect to sports-related impairment which is likewise associated with enormous health cost (Table 9.2), is however quite different.

Table 9.1 Development of workplace safety from 1970 until 2005 in the USA

Year	Fatalities	Employment ($\times 10^3$)	Fatality rate (fatalities per 100,000 workers)
1970	13,800	77,000	18.0
1975	13,000	85,200	15.0
1980	13,200	98,800	13.0
1985	11,500	106,400	11.0
1990	10,500	117,400	9.0
1995	6,275	126,200	5.0
2000	5,920	136,377	4.3
2005	5,734	142,894	4.0

Until 1990 numbers are estimations compiled in the National Safety Council Accident Facts, from 1990 they are based on a census of the Bureau of Labor Statistics

Public awareness is mostly related to the health condition of star players in prominent teams rather than to health impairment resulting from sports activities of the general population. In professional sports, safety issues are primarily reflected in game rules, referee work and trainer education, while in general sports, countermeasures are often of an ad hoc nature, little systematic, limited to general recommendations or promoted by manufacturers of sports accessories. The reason might derive from the fact that most sports activities are of a recreational and voluntary nature. Again quite different, finally, are circumstances regarding health problems due to continued strenuous household work: This area is virtually unexplored from a scientific point of view.

Medical disciplines mostly involved in chronic diseases are rheumatology, orthopaedics, neurology, sports medicine, radiology, pain management. Due to the significant involvement of psychic and social factors, psychiatry is likewise of importance. While diseases considered here such as bursitis (inflammation of a bursa, a friction-reducing layer between skin, ligament or tendons and bony structures) are mostly due to external mechanical overloading over extended periods of time, mechanical overload can also be caused by the body itself, in that obesity, hypertension or muscular dysbalance represent long-term risks in view of health impairment and premature death. Functional misuse, e.g. continued work in a bent position, is a further source for health impairment. The application of biomechanics in the quantitative sense treated in this book is difficult under circumstances exhibiting such a variety. “Simple” statistics, i.e., collection of numbers and monitoring of data over years prevail. This is not surprising since realistic experiments on long-term mechanical exposure can hardly be performed, analytical procedures suffer from missing basic knowledge and physiological and psychical reactions under conditions of continued pain exhibit enormous individual variations which cannot be disregarded. Accordingly, workplace safety issues are to a large extent limited to the prevention of acute injury.

Table 9.2 Number of accidents/injuries and illnesses along with cost due to sports, estimated for Switzerland in 2001 by the Swiss Federal Health Office

	Number of accidents/injuries	Number of diseases	Direct cost (CHF)	Indirect cost (CHF)
Accidents/injuries and cost caused by sports	300,000	–	1.1 10^9	2.3 10^9
Benefit from sports-related physical activity	–	2.3 10^6	2.7 10^9	1.4 10^9
Illnesses due to lack of physical activity	–	1.4 10^6	1.6 10^9	0.8 10^9

While direct cost are related to medical treatment, indirect cost are caused by absence from work, insurance administration, forensic expenditures, etc. The figures have to be considered in relation to the population of 7.5 million and a GNP of around 400 $\times$ 10^9 US$

Rehabilitation is an important issue likewise in cases of acute and chronic illnesses. Biomechanics are of important in this context insofar as physiotherapy, physical therapy, orthopaedic aids, wheelchairs, training devices etc. rely on the application of methods and on the design of apparatus where biomechanics have to be taken into account mostly on a quite basic, albeit important level. Since this is not directly related with trauma biomechanics, however, these subjects are not further considered here.

9.1 Occupational Health

Professions and working conditions which are associated with heavy and strenuous mechanical exposure are numerous and cover a wide variety of forms ranging from as different activities as underground mining to ballet dancing. The problem of long-term impairment appears in all of these professions, but it has to be treated differently in each case. Moreover, working conditions and standards to be maintained differ considerably from country to country. Activities of labour unions and NGOs as well as legal procedures with respect to early retirement, inability to work and workman's compensation are furthermore quite influential and subjected to different local political conditions. Gender differences have furthermore been observed. In an extensive report prepared by the WHO (Kane 1999), the particular risks that women are exposed to are discussed.

A large number of regulations and status of health reporting plans in connection with strenuous and hazardous work is maintained by the Occupational Safety and Health Administration of the US Department of Labor (http://www.osha.gov/). In other countries, similar organizations exist. In Switzerland, e.g., it is the partly government-controlled Accident Insurance Company (suva) with mandatory coverage of all "blue collar" employees (i.e., including also low-risk professions) whose tasks includes workplace as well as leisure accident issues. From the many professions, a few exemplary remarks are made in the following:

Table 9.3 Observed maximal sound levels in various circumstances

	Noise (dBA)		Noise (dBA)
Firecrackers	125–155	Lawn mower	90–110
Concerts	120	Car horn	110
Gun shots	150–167	Jack hammer	113
Movie theatres	118	Hair dryer	90
Sporting events	127	Chain saw	110
Health club, aerobics studies	150	Personal stereos	105–120
Motor boats	85–115	Children's toys	135–150
Motor cycles	95–120		

Source The Safe Side, Wisconsin, USA, Vol. VII, 2004

- Construction work, mining, lumber and woodwork: These are the professions that come to mind first when a hard and strenuous work environment is addressed. Early retirement due to health problems and health-related absenteeism are in fact well known (Brenner and Ahern 2000) and have been recognized as such for a long time. While the risk of acute injury is approached by numerous regulations (helmets, ear protection, gloves, etc.) long-range impairment is difficult to control. For practical reasons, e.g., weights to be lifted cannot be limited because these are given by the materials to be used. Accordingly, lower back pain is widespread (Latza et al. 2002). Cost associated with health problems along with the need for a continued increase in efficiency causes therefore a growing application of machinery.
- Nursing: Nursing is known as a profession which is partly associated with strenuous work and impairment. The situation was analysed comprehensively in the EU sponsored NEXT study (Nurses Early Exit Study). The physical environment, i.e., lifting, bending or work with non-cooperative and aggressive patients was thereby recognized as a major problem.
- Professional dancing (classical ballet, break dance): Amenorrhea is a well-known complication in females who are exposed to continued strenuous or irregular work environments such as professional ballet dancing or long-haul airline work. Associated loss of bone mineral content cannot systematically be compensated by physical training (Warren et al. 2002). Not surprisingly, the ankle joint is mostly prone to chronic impairment in ballet dancers (Rand et al. 1999). A comprehensive overview over risks associated with professional dancing is given in (Dance Medicine–The Dancer's Workplace, Unfallkasse Berlin).The particular risks associated with break dancing were investigated by Kauther et al. (2009).
- Noise: Hearing loss is a common outcome of extended exposure to loud noise. In some cases, enormous sound levels are reached even using tiny music players with ear phones (Table 9.3). Since the advent of high power music equipment with widespread use the problem aggravated substantially because a large

number of young people is affected. It should however not be forgotten that problems exist also under traditional conditions: The maximal sound level to which a musician in the Wagner orchestra is exposed approaches 140 dBA. The risk of hearing loss is strongly correlated with the exposure time. According to the Swiss Accident Insurance, 4 h/week of disco-music at 93 dBA (closed room) or 2 h/week of outdoor concert at 100 dBA are considered tolerable; for lack of precise knowledge, a linear scale is assumed for extrapolation.

9.2 Sports

9.2.1 Non Contact Sports

Overuse injuries account for approximately 50 % of all injuries in sports (Wilder and Sethi 2004). They are mostly due to physical overuse (overuse syndrome). Thereby, repeated micro-trauma beyond the reparative abilities of the musculo-skeletal system will eventually lead to macroscopic injury and clinical symptoms. "Repetitive Strain Injury (RSI) syndrome" is a further expression which describes this phenomenon.

Most often clinically observed health problems include:

- Tendinitis (painful inflammation of a tendon) is among the most common problems diagnosed in this context. It is furthermore aggravated by age-related degeneration that affects many of the large tendons in both the upper and lower extremity (Karamanidis and Arampatzis 2007). This leads to an increased predisposition with respect to painful lesions during athletic activity. Changes of collagen composition have been identified as a major reason (Riley et al. 1994). Specific examples are rotator cuff tendinitis, tennis elbow (epicondylitis), and Achilles tendinitis, observed in activities such as running, overhead throwing and serving balls in tennis. It is important to note that these lesions are ultimately associated with unrecoverable degenerative processes rather than "only" inflammatory changes. The term tendinopathy may in such cases be more appropriate than tendinitis, which implies only inflammatory changes.
- Medial tibial stress syndrome (MTSS) is a typical overuse leg problem seen in athletes (Madeley et al. 2007). Specialised training and rehabilitation procedures are recommended in this case in order to improve the endurance of the ankle musculature.
- Stress fractures are characterized by tiny cracks in bone often caused by repetitive overloading (such as in the feet of a basketball player who is continuously jumping on the court) (Snyder et al. 2006; Wilder and Sethi 2004; Egol et al. 1998; Fredericson et al. 1997). A reaction to such injury is demonstrated in Fig. 1.2 where microcallus formation is shown. While such microscopic injuries—if sufficiently scarce—favour bone remodelling (our skeleton is totally replaced within 4–6 years under healthy conditions e.g. Martin 2003), continued extensive microcrack formation is deleterious. In case

of adolescent athletes, furthermore, epiphysitis or apophysitis, i.e., growth plate overload injuries such as the Osgood-Schlatter disease (swelling and pain below the knee) or a process denoted as "shin splint" (similar symptoms) are observed (Wilder and Sethi 2004).

Due to the various, for sports typical extensive jumping and running activities, the lower extremities, in particular the feet, ankle (Valderrabano et al. 2006), sport shoes and the structure of the ground (stiff, elastic or energy absorbing) are of importance (e.g. Bartlett 1999). Since to some extent typical and systematic loading scenarios in various sports with respect to feet and ankle can be defined, a quantitative analysis in the design of sport shoes is possible (Reinschmidt and Nigg 2000; Stefanyshyn and Nigg 2000). In case of running shoes, major aspects include pronation control and cushioning. For court shoes, in turn, lateral stability, torsional flexibility, cushioning and traction control have been identified as significant design parameters in view of injury prevention. A further problem associated (not only but especially) with sport shoes is onychodystrophy, i.e. mycosis (fungal infection) due to chronic mechanical irritation (Romano et al. 2005). The knee, in particular with jumpers, is likewise of concern (Tibesku and Pässler 2005).

Extensive recommendations can be found in order to prevent sports-related overuse injury (Niams 2013; NCSS 2013; bfu 2013). Additionally, trainer education and training methods (Ezechieli et al. 2013), mostly in professional sports, represents an important task.

9.2.2 Contact Sports

Some forms of contact sports (boxing, kick-boxing, wrestling, martial arts) are potentially prone to cause injury. This aspect has mostly been covered in the earlier chapters of this book and is not further worked out here. It is the primary task of the referee(s) to enforce regulations and prevent acute serious injury (a ruptured eyebrow or lip does not represent a serious injury in this environment). Nevertheless, chronic exposure may particularly be problematic in contact sports: While a knock-out associated with unconsciousness as a single event may be harmless and recovery is, in general, rapid, continued exposure to violent blows is not. This is reflected, e.g., in the somewhat cynical expression "slap-happy" which refers to an (unsuccessful) boxer who suffers from partial dementia due to repeated blows to the head.

Boxing, kick-boxing, wrestling and martial arts are partly covered by strict regulations such as the ones issued by the World Boxing Association (WBA 2013) or the Fédération Internationale de la Lutte Amateur (FILA 2013). Nevertheless, long-term sequelae of such sports are not primarily taken into account in these initiatives. Likewise, an investigation on Taekwondo kicks under condition of competition (Serina and Lieu 1991) revealed that injurious levels in the thorax region are sometimes reached (see also Sect. 5.5).

9.3 Household Work

As mentioned above, long-term effects of household work are mostly unexplored in spite of the fact that unphysiological postures during kitchen work (low tables causing back pain) or cleaning activities (bent position over extended periods of time) are not uncommon. Impairments due to such circumstances are mostly treated by family doctors individually without significant amounts of insurance compensation. Nevertheless, some consumer organizations try to analyse the problem and maintain a consulting and recommendation service for the public. Children (Thompson et al. 2011) and pregnant women (Virk et al. 2012) are particularly exposed. Likewise, in home healthcare household safety is of concern (Gershon et al. 2012), in particular as in an aging population home healthcare is of rapidly increasing importance.

Ageing is associated with particular injury hazards. Not only, as mentioned in Chap. 2, do biomechanical characteristics deteriorate with age, but coordination of motion even in ordinary daily life without specific threats may pose problems. This is highlighted, e.g., by the increase of accidental falls where the injury potential is high (Boelens et al. 2013).

9.4 Summary

Chronic diseases and injuries which are due to mechanical overloading, overuse or functional misuse represent a major social and socio-economic problem. In the industrialised countries, this problem is further aggravated by the increasing life expectancy associated with a growing percentage of aged persons. Yet, biomechanics as treated in this book has a limited significance in this area. Quantitative tolerance limits with respect to loading which would have to be defined as function of frequency and duration of loading hardly exist. This is primarily due to the fact that a quantitative procedure is mostly not possible in that the basic principles of biomechanics, i.e., measurement, analysis and modelling of long-term mechanical exposure have hitherto not been applied widely and systematically. Accordingly, only qualitative recommendations, general rules, monitoring and statistical approaches prevail. In view of the high significance of chronic impairment, however, a great challenge exists for future biomechanics' research in this particular field.

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Ballistic and Blast Trauma

10

Exposure to blast and ballistic threats occurs in both defence and civilian environments and is, unfortunately, still common today (see also Chap. 1). Ballistic injury refers to the interaction of a projectile and the human body leading to penetrating or blunt trauma, while blast injury refers to detonation of an explosive and the subsequent complex interaction of the blast, fragments and debris with the human body. There is, of course, overlap between ballistic injury and blast fragmentation. Some common areas of blast injury include the lower extremity, thorax, and head, although all body regions may be exposed to blast loading. Similarly, all body regions are susceptible to ballistic injury where protection is often first focused on life sustaining organs (heart, lungs and brain). Active areas of research include head injury in blast, vehicle and vehicle occupant blast protection, and fragmentation and ballistic protection of the head, face and thorax including behind armour blunt trauma.

Blast and ballistic events may share some commonalities in terms of the time scale over which events occur and the resulting injuries; however, the actual time and injuries sustained strongly depend on distance between the threat and target, and the use of Personal Protective Equipment (PPE). Close proximity blast events (i.e. anti-personnel landmine) can occur over hundreds of microseconds, while free-field blast loads are on the order of 1–10 ms for conventional explosives. Ballistic impacts on unprotected tissue occur over durations of up to 2 ms and the projectile impact velocity can vary significantly for large distances to the target. For reference, automotive impacts such as frontal crash typically occur over timeframes of approximately 100 ms.

The primary challenges in addressing injury biomechanics problems for blast and ballistic scenarios include accurate definition of boundary or loading conditions and high deformation rate material properties. Additionally, we often observe significant variability in experimental test results, even in blast or ballistic studies where the loading conditions applied to a structure or body are highly controlled. In blast scenarios, this variability can be related to incomplete detonation of the explosive, or variability of the soil or materials with which the explosive interacts. Furthermore in the modelling environment, it is challenging to represent all of the

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physics of a blast, particularly for close proximity blast exposure. Similarly for ballistic impacts, the presence of large deformations with possible failure of the projectile and impacted material, may account for the large degree of variability in the test data. These issues are exacerbated by the short duration of these events and the need for material mechanical properties at appropriate deformation rates.

While ballistic and severe blast injuries are often readily observable, many aspects of these injuries may not be easily visualized and in some cases may not be obvious from external viewing or examination. Such cases include blast lung, mild traumatic brain injury (mTBI) resulting from blast exposure, and behind armour blunt trauma that may result in lung contusion. The goal of this chapter is to provide a brief background on blast and ballistic threats, the resulting injuries, and methods to mitigate these injuries.

10.1 Ballistic Injury and Protection

Exposure to ballistic threats in the civilian environment is primarily through small arms violence. Total homicide rates are substantially associated with firearms, and in countries with higher homicide rates, firearms account for a larger portion of total homicides (Small Arms Survey 2012). Homicide rates associated with firearms range up to 32 per 100,000 population, compared to the total homicide rate of approximately 60 per 100,000 population in the same region (Krug et al. 2002). In contrast, deaths in direct conflict (conservatively estimated at 52,000 per year from 2004 to 2007) are estimated to be 10–20 % of non-conflict deaths and are primarily associated with small arms and light weapons (Small Arms Survey 2013a). However, indirect conflict deaths (disease, destruction of infrastructure and other changes associated with a conflict) may be as high as four times the direct conflict deaths (Small Arms Survey 2013b).

Ballistics is generally classified in terms of interior ballistics (processes taking place to accelerate a projectile, typically within the barrel of a firearm), exterior ballistics (projectile flight and trajectory) and terminal ballistics (interaction of the projectile with the target), with wound ballistics referring to the interaction between a projectile and the human body (Sellier and Kneubuehl 1994). This discussion is focused on terminal and wound ballistics with a brief discussion of projectiles. In modern firearms, the projectile is mounted in a cartridge containing propellant (gun powder) and a primer that is initiated by impact from a firing pin. The cartridge and projectile are confined within the breach and barrel of the gun and when the firing pin strikes the primer it ignites the gunpowder which burns and generates pressures ranging up to a few thousand bar, which accelerate the projectile down the barrel. Barrels with rifling (spiral grooves in the barrel) impart a spin to stabilize the projectile during free flight to the target. For additional information on interior and exterior ballistics, the reader is referred to Sellier and Kneubuehl (1994).

projectile	0.50 BMG	7.62x51 NATO	9 mm Luger
mass [g]	45.00	9.53	8.04
muzzle velocity [m/s]	908	833	360
kinetic energy [kJ]	18.6	3.3	0.52

Fig. 10.1 Examples of jacketed projectiles and energies at muzzle velocity

Ballistic threats cover a wide range of sizes or calibres (projectile diameters), constructions, and velocities, including fragmentation from blast, shaped charge rounds, long rod penetrators, and many other implementations. Early gun projectiles were spherical in shape, while recent projectiles are cylindrical with an ogive-shaped nose that are spin-stabilized, and include a metal jacket (e.g. gilding metal) (Fig. 10.1). Black powder or gunpowder was used in the Far East as early as the ninth century, and was first documented in the West in the thirteenth century; however, significant development of small arms did not occur until around the nineteenth century with smooth bore barrels. Spherical lead projectiles were followed by advances in rifled barrels which impart spin to the projectile to improve accuracy, and metal cartridges to improve loading speed and consistency. At that time, projectile velocities were on the order of 450 m/s whereas modern projectile velocities vary from ∼200 m/s for small handguns to ∼1000 m/s for some rifle projectiles. Small arms refers to revolvers and pistols that are designed to be accurate to ranges of about 40 m while rifles may be accurate to ranges up to 1000 m or more (Mahoney et al. 2005). Common small arms and rifle projectiles range in size, dimensions and construction where the projectile core may be soft and deformable (lead) that may be designed to deform significantly on impact, or hard (e.g. steel) for penetrating armour. Some specific examples include full metal jacket (reliable non-expanding and deep penetrating round), jacketed hollow point

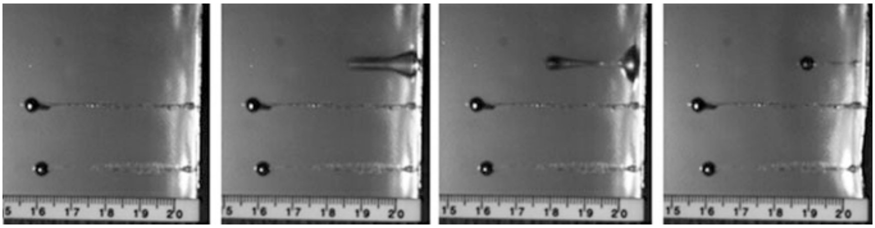


Fig. 10.2 Spherical 4.5 mm BB impacting gelatin at 65 m/s, 2 previous impacts are evident in the images. Projectile travelling from *right* to *left*

and soft point (designed to expand on impact for increased soft tissue damage) (Mahoney et al. 2005). Projectiles are often characterized by their kinetic energy at impact: $E_k = (1/2)\ \mathrm{mv}^2$, where the velocity refers to the linear velocity of the projectile. The energy associated with projectile rotation necessary for spin stabilization is typically neglected, being a small component of the total energy.

10.1.1 Wound Ballistics and Penetrating Ballistic Injuries

The terminal effects of projectiles have been observed via field experience since the invention of firearms, and were initially investigated using post-mortem human subjects and animal models. Some of the first detailed investigations on wound ballistics were published in the 1800s, identifying the importance of projectile velocity on the resulting wound and the expansion of the tissue during an impact (Cooper and Dudley 1997). Today, the effects of projectiles are often investigated using tissue surrogate materials such as ballistic soap or ballistic gelatin (Sellier and Kneubuehl 1994) since they provide a higher degree of consistency and improved opportunities to observe the transient effects during an impact. One of the most widely used surrogates is ballistic gelatin, which is produced from biological materials (e.g. skin, bone and tendons) through extraction with hot water in an acidic (type A) or alkaline (type B) environment, and then combined with water, heated and mixed, and conditioned at specific temperatures for a period of 2–3 days before use (Jussila 2004). An impact between a steel sphere and 10 %, 4C gelatin (Fig. 10.2) progresses with the development of a temporary or expanded cavity, followed by maximum dynamic penetration and finally static penetration with the wound tract identifiable.

Two aspects of this impact scenario are of note: the formation and size of the temporary cavity, which is related to the amount of energy imparted to the tissue or tissue surrogate by the projectile; and the permanent cavity corresponding to localized cell necrosis (Mahoney et al. 2005). Penetrating impacts in soft tissues leads to the formation of a temporary cavity, followed by the collapse of this cavity to the wound tract or permanent cavity (Fig. 10.3). The relative size of these cavities depends on the projectile construction and velocity, and has been investigated for different projectiles (Fackler 1987) with hand guns and rifles generally

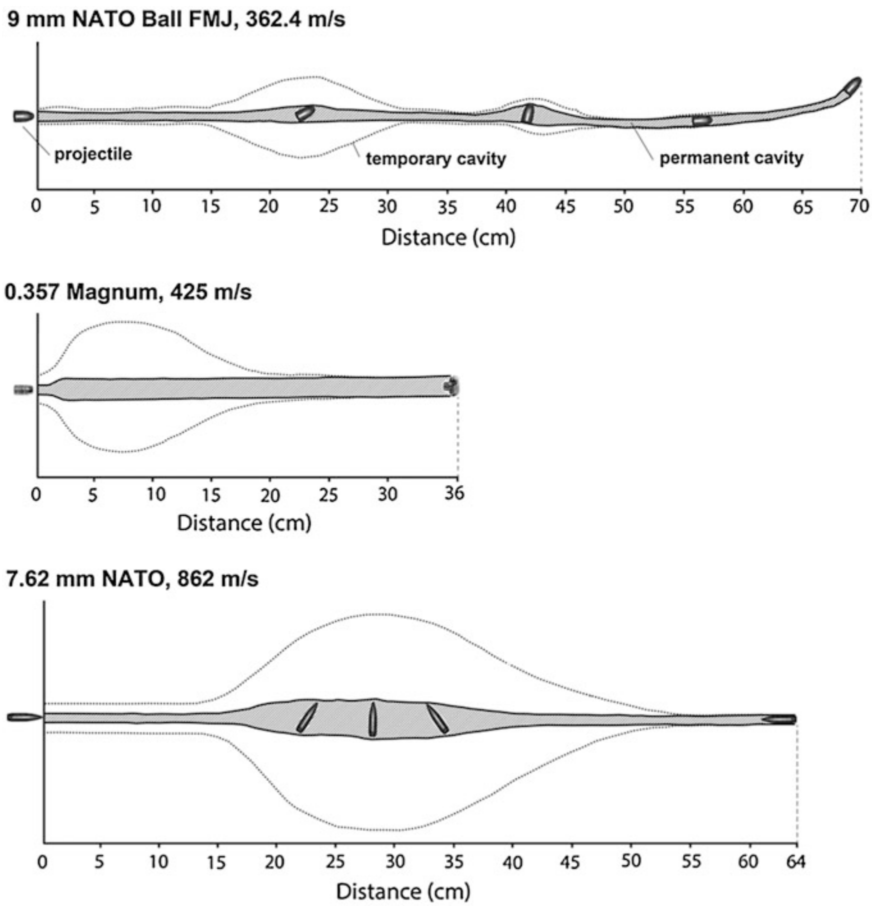


Fig. 10.3 Schematic of permanent and temporary cavities in ballistic gelatin for different projectiles (adapted from Fackler and Malinowski 1988)

treated separately due to the significant difference in projectile energy. In addition, non-deforming projectiles often yaw or tumble as they pass through tissue or tissue surrogates, generating a moderate sized temporary cavity with large penetration. In contrast, projectiles that fragment or deform significantly on impact typically result in shallower penetration but typically with a larger temporary and permanent cavity. The size of the permanent cavity is essentially related to tissue interaction directly with the projectile and is on the order of a few projectile diameters for deforming projectiles, the projectile length for non-deforming projectiles that yaw, and many projectile diameters for fragmenting projectiles. It should be emphasized that the effects are very dependent on the projectile construction and impact velocity. The reader is referred to Fackler (1987) and Fackler and Malinowski (1988) and Sellier and Kneubuehl (1994) for data on various projectiles and the experimentally measured temporary and permanent cavities.

Penetrating ballistic injuries in humans can result in fatalities due to collapse of the circulatory system (damage to the major vessels or heart) and/or damage to vital regions of the brain. The treatment and outcome of a penetrating ballistic injury depends greatly on timely access to appropriate care, which may be available in civilian environments but may be limited in armed conflicts, for example. Several publications addressing the treatment of ballistic injuries are available in the literature, e.g. Coupland (1993) and Molde et al. (2001). In general, soft tissue injuries are always assumed to be contaminated and are treated through debridement or opening of the wound for removal of contaminants (e.g. bullets, fragments, clothing and detached tissues). Dead or nonviable tissue is then removed or excised, the wound is irrigated and closure is delayed to ensure all contamination and nonviable tissues are removed to mitigate infection and the possibility of compartment syndrome (Mahoney et al. 2005). Impact on hard tissues often results in local bone fracture and many small bone fragments, where unattached fragments should be removed from the wound. Deformation or yaw (tumbling) of the projectile can lead to internal damage and a large exit wound relative to the entrance wound. As in other areas of injury biomechanics, ballistic injury is often discussed in terms of the body region(s) affected: head, neck, thorax, spine, abdomen and pelvis, and the limbs. Penetrating head injuries are classified as tangential (a glancing impact that may fracture the skull and cause laceration or contusion of the brain tissue), penetrating (projectile penetrates the skull resulting in contusion, laceration and hematoma) and perforating (project penetrates and exits the head), where the latter generally results in the largest amount of damage. Penetrating neck injuries can be serious, particularly if major vessels or the airway are affected, and most often require immediate treatment. Penetrating wounds to the thorax are often serious due to the potential for pneumothorax, hemothorax, pulmonary laceration and contusion, cardiac injury, and large vessel injury. The organs of the abdomen (e.g. kidney, spleen, liver) may experience large amounts of disruption and damage from penetrating gunshot wounds and require surgical intervention. Although limb wounds are one of the most common gunshot wounds, they are often not life threatening unless a major vessel (e.g. femoral artery) or a large bone (e.g. femur) is significantly damaged; however, there is a possibility for amputation in this type of injury depending on the extent of the wound and damage to vessels and/or nerves. Given the severity of penetrating injuries, anticipated exposure by police or military personnel is mitigated through the use of Personal Protective Equipment.

10.1.2 Personal Protective Equipment (Ballistic Protection and BABT)

Personal Protective Equipment (PPE) refers to worn apparel or carried items designed to protect the user from injury or trauma. Protection for the human body has historically evolved to meet changing threats to the body, including blunt

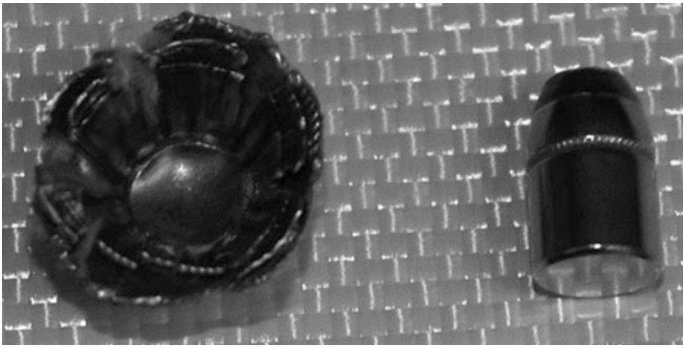


Fig. 10.4 9 mm projectile deformed after impact (*left*) and original (*right*), impact on flexible body armour

impact protection using thick layers of animal hides followed by metallic armour to protect against piercing weapons. However, existing protection was rendered ineffective following the development of firearms and significant efforts were undertaken to address this new threat using various materials including metals, silk and nylon. PPE intended for protection from ballistic impacts must often protect against other forms of injury such as blunt impact from objects or falls, and other penetration threats such as knives. For the purposes of this section, the discussion is focused on soft or flexible body armour (ballistic vests) and hard body armour that may be worn separately or in conjunction with soft body armour to protect the vital organs in the thorax. However, ballistic head protection is often worn, as well as additional armour to protect the lateral aspects of the thorax, neck, groin, extremities and eyes/face.

The primary trade-offs in PPE performance include weight (mobility, ergonomics and coverage) and protection level. For armour, weight is normally expressed in terms of areal density (e.g. kg/m^2) or weight per unit area so that different protections can be compared on a coverage basis. Thoracic protection is typically evaluated based on the ability to mitigate projectile perforation, and the amount of dynamic deformation that occurs during an impact, which can lead to blunt trauma. This is known as Behind Armour Blunt Trauma (BABT) (Knudsen 2010). Perforation describes a projectile that has completely passed through a target whereas projectiles that are stopped within the target are termed partial penetrations.

Soft or flexible body armour (vest) typically consists of multiple layers of ballistic fabric that allow for some flexibility and is often worn as a vest covering the thorax with areal densities up to 8 kg/m^2 depending on the material used and level of protection. Modern flexible armour originated with the development of aramid fibres in the mid-1970s (e.g. Kevlar®) which was woven into fabric layers (see background of (Fig. 10.4) and used in the form of multiple stacked layers to provide ballistic protection. Various forms of aramid fibres and weaves, including uni-directional fibres have been used in flexible armour. More recently,

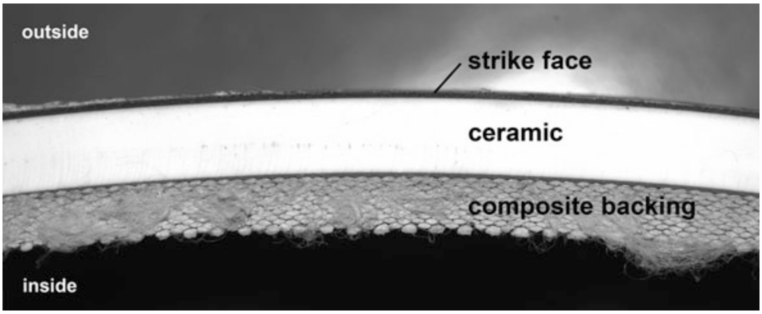


Fig. 10.5 Conventional hybrid hard armour cross-section

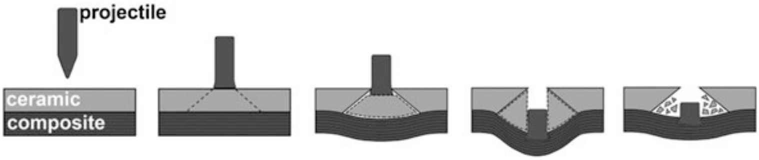


Fig. 10.6 Projectile impact on hard armour (adapted from Singh et al. 2012)

polyethylene-based fibres (e.g. Dyneema®) have been used to provide lower areal density solutions.

Flexible protection is intended for relatively low energy projectiles including handgun or pistol projectiles, and works by decelerating a projectile over a very small distance. An example of a 9 mm projectile pre and post impact is shown in Fig. 10.4. The deformation of projectiles is often very large with material failure occurring at high deformation rates, making modelling of these impacts challenging.

Hard or plate armour is designed to stop high energy threats that may include armour piercing cores. One example is the M2 AP projectile that comprises a gilding copper jacket, hardened steel core and lead tip filler, which is widely used in armour testing and research studies. When impacting at high velocity, it is challenging to stop this type of projectile with traditional flexible armour while maintaining acceptable weight and mobility. Hard armour is commonly constructed (Fig. 10.5) with a high hardness material on the impact or strike face of the armour to disrupt the projectile, and backed ductile or deformable material to decelerate and stop the impact debris.

The impact of a projectile on ceramic with a thin backing has been described by several authors (Wilkins 1978; den Reijer 1991) and begins with the initial contact between the projectile and ceramic blunting the projectile and initiating a shock wave in the ceramic (Fig. 10.6). This is known as the dwell stage and occurs over a time of 6–8 μs. As the impact progresses, damage is initiated in the ceramic at the impact location and fractures propagate outwards creating a fracture conoid

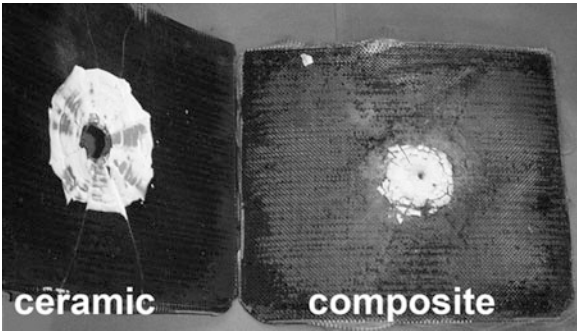


Fig. 10.7 Conventional hybrid armour opened to show the fracture conoid on the back of the ceramic and composite backing which has stopped the impact debris

Fig. 10.8 M2 AP projectile (*left*), intact steel core (*centre*), and steel core following impact with hard armour (*right*)

(Fig. 10.7). Movement or flow of the comminuted ceramic allows for ingress of the projectile while continuing to erode and deform the projectile (Fig. 10.8). As the impact proceeds, the backing material begins to deform, decelerating the projectile and impact debris. Depending on the armour construction, projectile construction, and impact velocity the projectile and debris may be stopped in the armour or may perforate the armour.

The impact or strike face is often constructed from ceramic materials due to their high hardness and strength, with a covering or spall cover to mitigate the expulsion of impact debris and improve the ceramic performance. The purpose of this hard strike face is to break up or disrupt the projectile and the ceramic is often fractured or damaged during the impact event. To a large extent, the amount of damage in the ceramic determines the viability of the armour to withstand multiple projectile impacts, also known as multi-hit capability. Common ballistic ceramics include Aluminum Oxide, Boron Carbide and Silicon Carbide. The backing material typically consists of ballistic fabric layers that are laminated or bonded together using a thermoset (e.g. PVB phenolic) or thermoplastic (e.g. polyethylene-based) matrix to enhance penetration resistance.

At typical projectile impact velocities the pressures generated at the impact site are immense and the equation of state (relationship between pressure, volume and energy) must be considered to describe the material response (Meyers 1994). Interesting phenomena may occur at the impact site such as spallation, where a compressive stress wave encounters an interface of lower impedance and is reflected as a tensile wave causing material damage, and bulking where the fractured or comminuted ceramic volume is larger than the intact material volume.

10.1.3 Armour Performance and Testing

The performance of body armour is assessed on the ability to resist perforation and the amount of dynamic deformation during the impact, also known as Back Face Signature (BFS). An example of a widely used test standard is the National Institute of Justice (NIJ) Standard-0101.06 (NIJ 2008) Ballistic Resistance of Body Armour, which provides testing and performance standards for body armour. In this standard, body armour is classified according to types depending on the anticipated threat or projectile type and velocity. Note that armour is not defined as "bulletproof" but as ballistic resistant based on the specified projectile type and velocity. Armour may be tested in the as-manufactured or "as-new" state (Table 10.1), and in a conditioned state (Table 10.1) where it has been exposed to environmental conditioning (for example 65 °C temperature, exposure to 80 % relative humidity, and tumbling exposure of 72,000 rotations) to simulate the effect of time and exposure for soft or flexible armour. Similarly, hard armour conditioning requires exposure to temperature and humidity as well as thermal cycling (−15–90 °C) and drop testing, which may affect bonding within the plate or the integrity of the ceramic component.

The obvious goal of ballistic protection is to prevent armour perforation, which can then lead to penetrating injury for the wearer. However, even when the projectile is stopped in the armour (partial penetration) the dynamic deformation of the armour during the impact must be minimized to protect the vital organs in the thorax. Significant dynamic deformation can lead to blunt thoracic trauma (Behind Armour Blunt Trauma, BABT). The extent of BABT will depend on the rate of impact and extent of deformation.

The performance of body armour is typically rated using the National Institute of Justice standard which places limits on the penetration resistance of the armour, measured by V_{50}, the velocity at which 50 % of the projectiles are expected to penetrate the armour. Other measures, such as V_{05} corresponding to lower probability of perforation may also be specified in test standards. Ballistic testing may be undertaken with the armour mounted on a special type of clay backing (Fig. 10.9) which is used to record the dynamic deformation of the armour. The clay is specified as Roma Plastilina No.1 oil-based modelling clay and must be

Table 10.1 NIJ Test standard for as-new armour and conditioned armour (NIJ 2008)

NIJ level	Projectile	Mass (g)	Impact velocity (m/s)
For as-new armour:			
Type IIA	9 mm FMJ RN	8.0	373 ± 9.1
	0.40 S &W FMJ	11.7	352 ± 91
Type II	9 mm FMJ RN	8.0	398 ± 9.1
	0.357 Magnum JSP	10.2	436 ± 9.1
Type IIIA	0.357 SIG FMJ FN	8.1	448 ± 9.1
	0.44 Magnum SJHP	15.6	436 ± 9.1
Type III (rifles), hard armour[a]	–	–	–
Type III (rifles), flexible armour	7.62 mm FMJ (M80)	9.6	847 ± 9.1
Type IV (armour piercing rifle), hard armour[a]	–	–	–
Type IV (armour piercing rifle), flexible armour	0.30 calibre armour piercing (M2 AP)	10.8	878 ± 9.1
Special type[b]	Defined by purchaser in the case of special requirements based on the threat or protection performance.		
For conditioned armour:			
Type IIA	9 mm FMJ RN	8.0	355 ± 9.1
	0.40 S & W FMJ	11.7	325 ± 9.1
Type II	9 mm FMJ RN	8.0	379 ± 9.1
	0.357 Magnum JSP	10.2	408 ± 9.1
Type IIIA	0.357 SIG FMJ FN	8.1	430 ± 9.1
	0.44 Magnum SJHP	15.6	408 ± 9.1
Type III (rifles), hard armour[a]	7.62 mm FMJ (M80)	9.6	847 ± 9.1
Type III (rifles), flexible armour	7.62 mm FMJ (M80)	9.6	847 ± 9.1
Type IV (armour piercing rifle), hard armour[a]	0.30 calibre armour piercing (M2 AP)	10.8	878 ± 9.1
Type IV (armour piercing rifle), flexible armour	0.30 calibre armour piercing (M2 AP)	10.8	878 ± 9.1
Special type[b]	Defined by purchaser in the case of special requirements based on the threat or protection performance.		

[a]Hard armour may be worn in conjunction with soft armour to achieve the designated protection level. In this case the flexible armour is tested at the specified threat level, and then the hard armour in conjunction with the flexible armour is tested
[b]Examples of threats provided in (NIJ 2008)

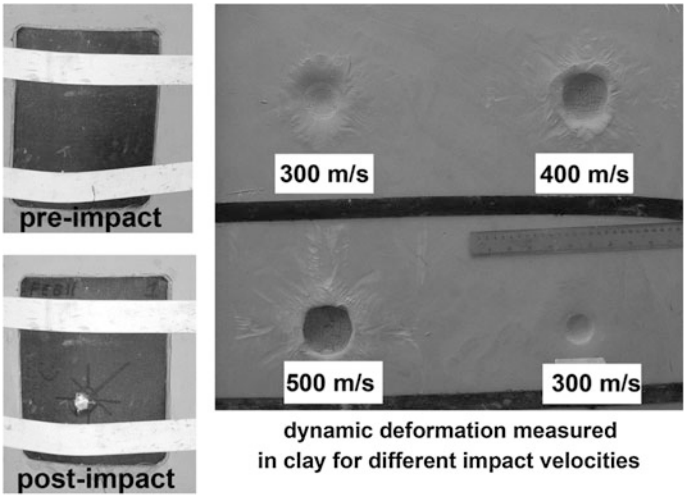


Fig. 10.9 Impact testing of armour on clay (*left*) and resulting deformation in clay (*right*)

at a specific temperature and calibrated with a steel ball drop test to be used for ballistic testing. In the NIJ standard, the measured dynamic deformation in the clay is limited to 44 mm while other specifications may use different values, depending on the protection level. For example, the UK police test standard uses an upper prediction limit of 25 mm for the Back Face Signature (BFS) (Croft and Longhurst 2007). It should be noted that the methods for calculating V_{50} and BFS may differ for different test standards, and may include statistical methods using multiple impacts to determine the likelihood of perforation or maximum BFS.

10.2 Blast Injury and Protection

Exposure to blast in the civilian environment is associated with accidental exposure (e.g. dust, fertilizer plant, natural gas explosions) and intentional exposures (e.g. Oklahoma city bombing 1995, Boston Marathon bombing 2013). Although many different types of explosives exist the output of an explosion is often characterized in terms of Trinitrotoluene (TNT) equivalent with values for conventional high explosives ranging from a few grams to several kilotons. For example, the 1917 Halifax harbour explosion of approximately 2.9 kilotons TNT equivalent (Glasner 2007) was the largest man-made explosion prior to the development of nuclear weapons. In comparison, the Hiroshima nuclear bomb yield was approximately 13 kilotons. In conflicts, blast exposure from conventional weapons and Improvised Explosive Devices (IEDs) has become more common with many IEDs in the range of 12–23 kg (Flynn 2009) and some exceeding 45 kg. However, typical human exposures to blast are on the order of tens of kilograms of TNT equivalent (Haladuick et al. 2012) and the severity of exposure depends on the distance between the explosive and target, level of protection worn, and surrounding environment.

10.2.1 Explosives and Detonation

An explosion is a rapid release of energy, which is typically adiabatic over the short duration of the explosion. Explosions can be classified as physical (e.g. pressure vessel failure), nuclear and chemical. A chemical explosion occurs by fast oxidation of an explosive material producing heat and gases which expand rapidly and generate shock waves in the expansion medium. Similar to ballistics, the development of chemical explosives began with the invention of gunpowder or black powder (potassium nitrate, carbon and sulphur). This was followed by the creation of various explosives based on nitric acid, the invention of nitroglycerine, the invention of dynamite by Alfred Nobel, and finally the development of TNT, the latter of which could be manufactured relatively safely and cheaply and became the standard explosive during WWI. More recent explosives include RDX (cyclonite or cyclotrimethylenetrinitramine), which is a common component used in many explosives such as C-4, which is often described as a plastic explosive. C-4 (91 % RDX) has a TNT equivalent of approximately 1.34 (Department of the Army 1967), based on pressure. It should be noted that the equivalence values may vary between literature sources and depend on the specific equivalence method (e.g. pressure or impulse).

Rapid reactions that convert explosive material into gases can be classified as deflagration or detonation. Deflagration or burning is a slow chemical conversion of explosive material into products and commonly refers to a reaction between fuel and atmospheric air but can produce an explosion if contained. An example is a dust explosion where a material such as coal, sawdust, grain and flour can produce dust clouds from manufacturing or handling processes. If a combustible dust material in high concentration, with particles generally less than 75 μm in size (Hetherington and Smith 1994), is suspended in the air with sufficient oxygen an ignition source can result in deflagration of the suspended material. In a case where the material is contained, such as in a building, the resulting increase in pressure can lead to an explosion and destruction of the structure. This same principle is often used in thermobaric weapons, which can generate relatively long duration overpressures using a fuel (e.g. powdered metal) dispersed over a large area which is ignited and relies on oxygen in the atmosphere for combustion.

In contrast, detonation requires the propagation of a shock wave through the explosive to provide the necessary activation energy for the chemical reaction (Fig. 10.10), and the explosive typically incorporates the necessary amount of oxygen required for the chemical reaction. This shock wave is typically provided by a detonator, which may utilize the vaporization of a wire to initiate the reaction. Explosives that require a significant amount of energy to detonate are known as high explosives (HE).

As a shock wave passes through a high explosive (HE), a chemical reaction is initiated. This reaction converts the explosive to gaseous products at very high

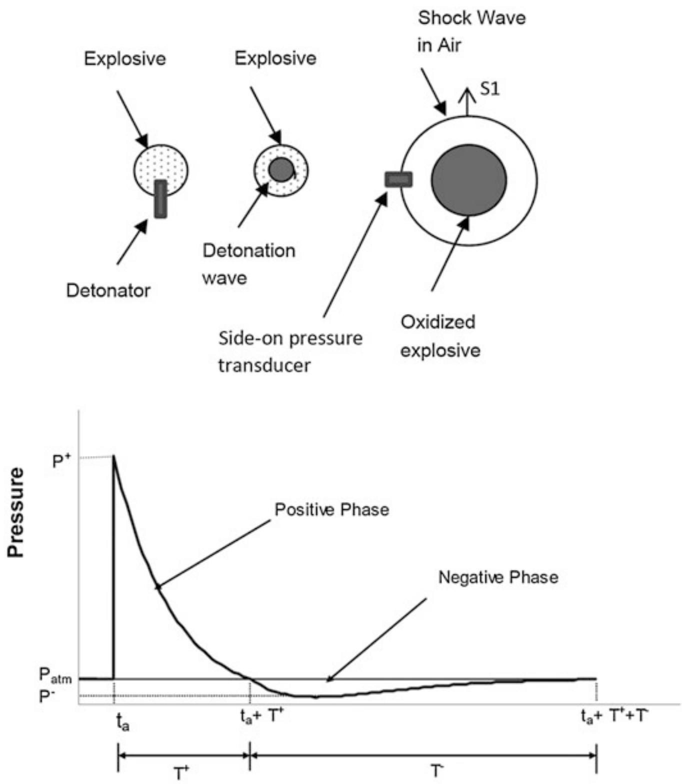


Fig. 10.10 Detonation of an explosive (*top*) and resulting free-field pressure wave (*bottom*)

temperatures and pressures occurring in a zone of finite, constant thickness known as the reaction zone, which propagates through the explosive at the detonation velocity (D). Detonation velocities are on the order of multiple kilometres per second (Dobratz and Crawford 1985). The highest pressure in the explosive, known as the von Neumann spike (PVN), is achieved at the leading edge of the reaction zone. The trailing edge of the reaction zone, where the chemical reaction is complete, is known as the Chapman-Jouguet point and the associated pressure is known as the Chapman-Jouguet pressure (PCJ). A common term used to describe the effectiveness or shattering power of an explosive in contact with a target material is brisance. This effect is due to the interaction of the detonation wave in the explosive with another material. As with the treatment of shock waves in solids, the detonation wave travelling through a high explosive can be described with the conservation equations with the exception that the energy equation must include the chemical energy (Q) of the explosive Eq. (10.1).

$$E_n - E_{n0} - Q = \frac{1}{2} \cdot (P + P_0) \cdot (V_0 - V) \tag{10.1}$$

Table 10.2 Examples of longitudinal wave speeds. Values are approximate since the density and/or stiffness of some materials may vary

Material	Speed (m/s)
Air	340
Water	1480
Aluminium	6100
Steel	5800
Plexiglass	2600
Polystyrene	2300
Muscle	1550–1630
Fat	1450
Cortical bone	3000–4000
Cancellous bone	1450–1800

After detonation, the behaviour of the explosive gases is often represented using a pressure–volume-energy relationship, also known as an equation of state. A common representation for high explosives is the Jones-Wilkins-Lee (JWL) equation of state (Eq. 10.2).

$$P = A \cdot \left[1 - \frac{\omega}{R_1 V}\right] \cdot e^{-R_1 V} + B \cdot \left[1 - \frac{\omega}{R_2 V}\right] \cdot e^{-R_2 V} + \frac{\omega Q}{V} \tag{10.2}$$

with A, B, R_1, R_2 and ω: equation constants, V: ratio of detonated to undetonated explosive volume, P: pressure and Q: explosive energy.

10.2.2 Waves and Impedance

Disturbances or perturbations (e.g. changes in load) are communicated through materials via stress waves (longitudinal, shear, bending and surface), which travel at the acoustic speed of the material when the rate of loading is small (i.e. infinitesimal disturbances). Different types of waves travel at different speeds. Acoustic speeds depend on the stiffness and density of a material, with some examples listed in Table 10.2.

If a wave encounters a boundary, part of the wave will be transmitted and part will be reflected. The amount of energy transmitted depends on the relative impedance between the two materials, corresponding to the material properties and geometry (area change) of the two contacting materials. Material impedance is defined as the product of the density (ρ) and the acoustic wave speed (U). When travelling into a higher impedance material, an incident wave is reflected and transmitted in the same sense, that is, a compressive wave remains a compressive wave (Fig. 10.11, top). In the case of a wave travelling from a high impedance to a

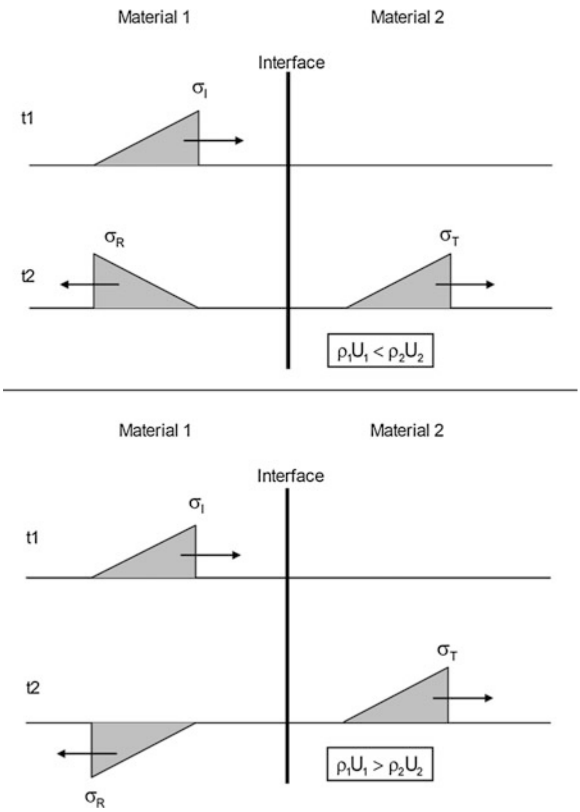


Fig. 10.11 Incident, reflected, and transmitted waves

lower impedance material (Fig. 10.11, bottom), the wave is reflected in the opposite sense. For example, a compressive wave becomes a tensile wave and this explains how spallation may occur on the opposite to struck side in a ceramic, which has a lower strength in tension than compression. The amount of the wave transmitted and reflected is described by Eqs. 10.3 and 10.4.

$$\frac{\sigma_T}{\sigma_I} = \frac{2\rho_2 U_2}{\rho_1 U_1 + \rho_2 U_2} \tag{10.3}$$

$$\frac{\sigma_R}{\sigma_I} = \frac{\rho_2 U_2 - \rho_1 U_1}{\rho_1 U_1 + \rho_2 U_2} \tag{10.4}$$

The mathematics is somewhat more complicated for non-normal angles of incidence leading to wave refraction (change of angle at the boundary). For a more detailed treatment of this topic, the reader is directed to Meyers (1994).

Explosive detonations invariably produce waves in media with which they interact. In the case of close proximity detonation, the resulting large disturbance in a medium such as air, soil, metal or biological tissues can generate shock waves

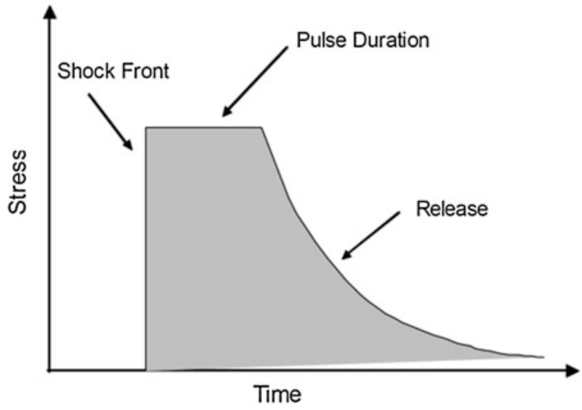


Fig. 10.12 Characteristics of a shock wave (idealized)

in that medium. A shock wave is generated by a violent disturbance of the material which can be created by high velocity impact, supersonic flow, or an explosion. A shock wave is defined as a discontinuity in pressure, temperature and density across a very small dimension, which propagates at a speed greater than the acoustic speed in the material. This is a nonlinear phenomenon that can occur when the material bulk modulus increases with increasing pressure (necessary to generate a shock wave), and requires an equation of state to describe the material pressure–volume-energy response. Several common equations of state (EOS) exist including the Gruneisen EOS (Eq. 10.5). In this equation, the pressure is a function of the non-linear material bulk modulus (first term in Eq. 10.5) and the volumetric work (second term in Eq. 10.5). Parameters for this equation are available for a wide range of materials (Meyers 1994), and the rule of mixtures can be used to determine the constants for combinations of materials (Wilbeck 1978).

$$p = \frac{\rho_0 C^2 \mu [1 + (1 - \frac{\gamma_0}{2})\mu - \frac{a}{2}\mu^2]}{\left[1 - (S_1 - 1)\mu - S_2 \frac{\mu^2}{(\mu+1)^2}\right]^2} + (\gamma_0 + a\mu)E \tag{10.5}$$

where P: pressure [Pa], ρ_0: initial density [kg/m^3], ρ: current density [kg/m^3], C: acoustic wave speed [m/s], $\mu = \rho/(\rho_0 - 1)$, γ_0: Gruneisen Gamma, E: specific internal energy [J/m^3], a: first order correction, S_1, S_2: slope coefficients for the particle-shock velocity curve.

Shock waves are characterized by a rapid increase in pressure, a short duration, and a release (Fig. 10.12). Shock waves generated by explosions in air are on the order of 1–10 ms for several kg of explosive, 100 ms for 1 ton of TNT, and 1–5 s for large nuclear blasts.

10.2.3 Blast in Air

Detonation of a spherical charge in air produces a spherically expanding shock wave travelling at supersonic velocity S1 (Fig. 10.10) driven by the expansion of the high pressure and high temperature combustion products. With no other influence, the combustion products or fireball will over-expand due to the outwards velocity of the detonation products, after which it will contract producing a negative phase or under-pressure. The static pressure, shown in Fig. 10.10, corresponds to the pressure as observed in the unobstructed flow of the blast and is often described using the Friedlander Eq. 10.6.

$$P = P^{+} \cdot \left(1 - \frac{t}{T^{*}}\right) \cdot e^{-\frac{bt}{T^{*}}} \tag{10.6}$$

where P: pressure as a function of time, P^{+}: peak static or incident pressure, T^{*}: positive phase duration, t: time and b: decay constant.

This pressure is also known as the side-on or incident pressure, since it is the pressure wave that may interact with surrounding structures. This is distinguished from the dynamic pressure or specific kinetic energy determined by the local density and flow velocity (Ritzel et al. 2011). The sum of the free-stream static and dynamic pressures is known as the total or stagnation pressure, which is the corresponding pressure if the flow velocity were zero. This is distinguished from the reflected pressure, which is defined when the blast wave interacts with a structure or body as described below.

Three regimes are often identified to describe the blast event following detonation of the explosive (Ritzel et al. 2011):

- Near-field where incident overpressures may exceed 10 atmospheres (1 MPa), and this area is typically within the expansion zone of the fireball where the ratio of static and dynamic pressures changes and is not spatially constant. Additionally, there is a large change in density and temperature across the contact surface of the fireball.
- Mid-field is defined as the zone beyond the expansion of the fireball where the overpressure varies between 1 and 10 atmospheres and follows a traditional exponential decay; however, the ratio of static and dynamic pressures can vary, nonlinear effects such as after burning may be present, and the negative phase may not be correctly predicted by Eq. 10.6.
- Far-field refers to a zone where the shock can be considered as 1-D and the overpressures are less than approx.1 atmosphere.

When a blast wave interacts with a structure, the resultant pressure on the structure is known as the reflected pressure, and is higher than the static pressure depending on the strength of the incident shock wave, angle of incidence, the shape of the structure, and compliance of the structure. In the case of a blast wave normal to a rigid wall, the reflected pressure ranges from 2 to 8 times the incident pressure (Eq. 10.7). The reflected wave moves at a faster speed than the incident wave due to the higher pressure of the reflected wave.

$$\frac{P_R}{P^+} = 2 \cdot \left[\frac{7P_{atm} + 4P^+}{7P_{atm} + P^+}\right] \tag{10.7}$$

where P^+: static or incident pressure, P_R: reflected pressure and P_{atm}: atmospheric pressure.

The pressure, positive phase duration, impulse and shock velocity resulting from detonation of an explosive can be determined using several methods including empirical equations and tables based on measured data, semi-empirical approaches where the explosive event may be modelled in a simplified manner based on descriptive equations and the explosive properties, or first-principle approaches to describe the physics of the detonation, resulting products and expansion in a medium in detail.

A common empirical method is the Conventional Weapons (CONWEP) formulations, which is a computer implementation (Hyde 1998) based on equations (curve fits) developed by Kingery and Bulmash (1984) for free air and surface blasts. Manuals also exist that provide tables and charts (e.g. U.S. Department of the Army 1990), which can be expressed in terms of a given scenario through scaling parameters, based on the charge weight. In general, distance from the explosive is expressed as a scaled distance, which is a function of the charge weight (TNT equivalent) to the power of 1/3 ($W^{1/3}$) (Baker 1973). Semi-empirical methods may include simplified numerical methods and codes to apply pressure loading to structures in a coupled manner, which incorporate the shape and compliance of the structure as well as the spherical shape of the blast wave that is important in close proximity blast (Lockhart and Cronin 2013; Singh et al. 2012), although much more computationally expensive. First principle models may be used in a similar manner, but incorporate descriptions of the explosive oxidation, products, heat generated and thermal effects. However, these very detailed models are often computationally expensive and may not be fully coupled to the target structure (e.g. rigid boundaries in computational fluid dynamics codes).

In cases where an explosive is detonated above the ground, the blast wave reflects off the ground and, travelling through the pre-compressed or pre-shocked air, catches up to the primary shock front to generate a Mach stem that intersects the original shock front at the triple point. Due to the interaction of the reflected wave with the primary wave, the Mach stem results in a localized region of higher pressures. The presence and size of the Mach stem depends on the charge size, height of burst (HOB) and standoff or distance from the explosive to the target. The effects of surface bursts (zero HOB) can be described by free air bursts which are augmented by an enhancement factor. For a perfectly rigid reflecting surface, this factor should be 2.0, but due to energy lost in crater formation and interaction with the ground, this enhancement factor is often approx. 1.8 (Hetherington and Smith 1994).

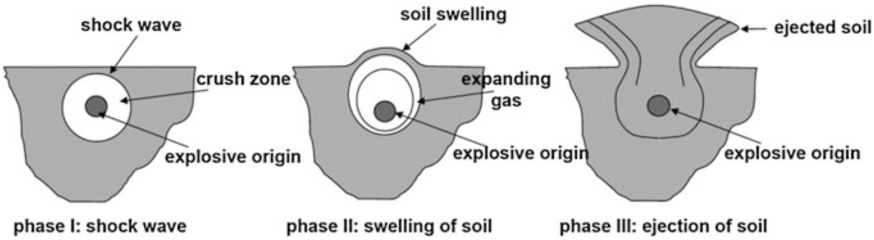


Fig. 10.13 Detonation of a buried explosive

The detonation of a high explosive buried in soil transmits a shock wave into the surrounding soil (e.g. loose sand, clay, limestone, granite), crushing the nearby material and absorbing some of the explosive energy. In general, the passage of the shock wave creates three zones in the soil: the crushed zone, the rupture zone, and the elastic zone (Bergeron et al. 1998). Within the crush zone, 2–3 times the charge radius, the soil can be liquefied due to the high pressures. The rupture zone, in which fissures are created due to wave rarefactions, extends to approximately 5 or 6 times the radius of the charge (Bangash 1993). For a deeply buried explosive, a spherical charge will create a spherical cavity, known as a camouflet, with the cavity radius depending on the mass of the charge. In the case of anti-personnel (AP) mines and many IEDs, they are typically buried at or near the surface of the ground and this depth of burial (DOB) has a significant effect on the behaviour of the explosion. A deeply buried explosive will transmit little or no energy to the free surface and target above it whereas an explosive that is surface buried (zero DOB) will transmit the maximum energy to the target. Assuming a constant charge type and weight, the amount of energy transferred to the target is primarily related to the soil material properties. In the limit, a rigid soil material would not undergo any permanent deformation and the energy of the explosive would be directed towards the target. Alternatively, a very soft soil could absorb a large portion of the explosive energy through deformation. Soil deformation resulting from an AP mine detonation can be described in three phases (Fig. 10.13):

- Phase I—Soil adjacent to the mine is crushed as the shock wave passes through the material.
- Phase II—Deformation (swelling) of the soil surface begins due to expansion of the detonation products and reflection of the compressive stress wave at the soil and air interface. For mines with a DOB greater than zero, a small volume soil cap is eventually ejected at high velocity.
- Phase III—A large volume of soil is ejected due to the continued expansion of the detonation products. This soil is ejected upwards in a conical shape, with the cone angle increasing with increasing DOB and decreasing soil density. The cone angle is typically less than 90°. In general, a shallower DOB results in a smaller amount of ejected soil with a higher velocity. The denser the soil, the more energy is directed upwards to the target.

10.2.4 Blast Injury

Injuries and fatalities from blast exposure have been identified and documented since the invention and first uses of explosives. Blast injury is often characterized using four levels, where primary blast injury (PBI) refers to injury sustained from the effect and interaction of the blast wave with the body commonly leading to auditory injury (tympanic membrane rupture), pulmonary contusion and injuries to the gastro-intestinal tract (Cooper and Dudley 1997). Blast exposure from conventional weapons and Improvised Explosive Devices (IEDs) has become more common and recently has been associated with an increased incidence of mild traumatic brain injury (mTBI), attributed to primary interaction with the blast wave (Gupta and Przekwas 2013). Secondary blast injury is the result of fragments and debris accelerated by the explosion impacting the body. Although this is a significant component of blast injury, fragment and ballistic injury to the body has been studied extensively by many authors and is relatively well characterized via existing injury codes such as Fragman (Netherlands), Computerman (USA) and Modèle Informatique du Combattant (France). These models require detailed characterization of the fragment size, velocity and grouping at specific distances and are specific to each explosive device type. Tertiary blast injury is the result of global body accelerations (i.e. flailing) where injury may be sustained as components of the body are subjected to extreme loading, or the body impacts surrounding structures or the ground. Quaternary blast injury refers to all other types of injury including burns, blindness and toxic gas inhalation. Primary, tertiary and quaternary injury may occur in the near and mid field regions, while there is possibility for primary injury and secondary injury in the far field (Fig. 10.14, Table 10.2). It should be noted that the concepts of primary, secondary, tertiary and quaternary blast do not describe the rate of incidence or severity of the injury, but rather relate to the origin of the insult. For example primary refers to interaction with the leading blast wave, secondary describes projectiles and debris that are accelerated secondarily by the blast wave, and tertiary describes the effects of the blast wind and fireball interaction with a body and the body interaction with the surroundings. The ordering of these injuries typically corresponds to the time frames on which the interaction occurs; that is, the first interaction is between the blast wave and the target, followed by fragmentation and lastly whole-body translation. There are some exceptions to this timing for very close proximity blast events. In general, secondary injuries from fragments are one of the most common injuries, due to the larger distances (Fig. 10.14); however, this is also one of the most understood mechanisms from our experience with ballistics, wound treatment and ballistic protection. It should be noted that this chapter provides only a brief overview of injuries resulting from blast injury. The reader is directed to the references in the particular sections as a link to the bodies of work on specific types of blast injury (Table 10.3).

The prediction of blast injury requires a detailed knowledge of the blast load conditions and resulting trauma mechanisms. Blast injury resulting from simple or

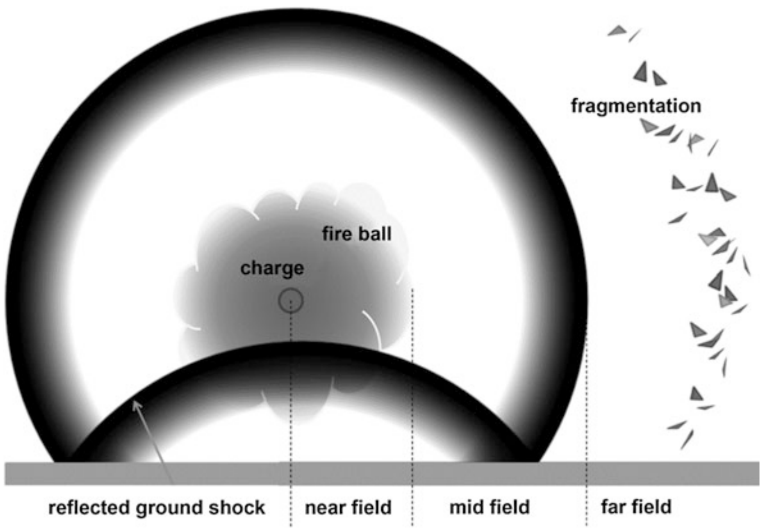


Fig. 10.14 Qualitative differences in maximum distance of threat propagation (adapted from Thom and Cronin 2009)

free-field blast has been widely studied in the literature (Bowen et al. 1968; Stuhmiller et al. 1996; Axelsson and Yelverton 1994) but still presents challenges in terms of protection. This is in part due to the complicated nature of experimental testing and the physiological nature of many associated injury types. One of the most widely referenced studies is that which led to the Bowen curves (Bowen et al. 1968) (Fig. 10.15), predicting the probability of fatality as a function of blast overpressure and duration. Additional challenges are found when considering explosive blast in urban environments or enclosed spaces. For example, occupants inside vehicles exposed to IEDs will be exposed to primary blast effects if the vehicle compartment is breached, and these effects may be amplified by the enclosed space resulting in higher levels of injury. In this case there is potential for complex blast loading (Fig. 10.16) due to wave reflections from surrounding structures and the possibility of other injuries related to the proximity of structures.

In experimental testing, shock tubes are often used to represent blast loading since the conditions are more controlled, repeatable and more conducive to experimental measurements (Clemedson 1956). In the Bowen curves (Fig. 10.15), scaled durations greater than 20 ms were achieved using a shock tube. However, there are several requirements to produce interpretable data including locating the target at an appropriate distance within the tube and ensuring the target does not obstruct more than 10 % of the tube cross section (Needham et al. 2013). Shock tubes typically produce long duration pulses (Fig. 10.17) although some designs (blast tubes) can achieve a decaying pulse.

Table 10.3 Possible injuries to unprotected victims (high-explosive detonation in open air) (adapted from Wightman and Gladish 2001)

Injury	Near field → Far field
Total body disruption	
Burns and inhalation injuries	
Toxic inhalations	
Traumatic amputations	
PBI of the lung and bowel	
Tertiary blast injury	
PBI of the ear	
Secondary blast injury	

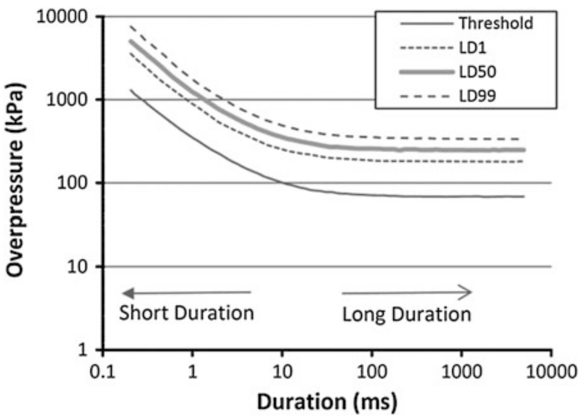


Fig. 10.15 Bowen pulmonary iso-injury curves (70 kg man with long axis of the body perpendicular to the blast wave propagation) (adapted from Bowen et al. 1968)

Many different approaches have been used to measure blast exposure for the purposes of predicting injury including: side-on or static pressure measurements to record incident pressure, live animal testing, and physical surrogates. For specific conditions, the incident pressure magnitude and duration can be used to predict the likelihood of pulmonary injury (Fig. 10.15), where these curves were derived from animal tests conducted on mammals ranging from small to large. It was noted that small animals such as mice and rabbits exhibited a lower threshold to pulmonary injury than larger mammals. For large animals, one of the largest bodies of information is on the testing of sheep, where the results were scaled (Bass et al. 2006) to be applicable to a human. Physical devices for testing include surrogates such as the Blast Test Device (BTD), a high stiffness cylinder with pressure gauges to measure the pressure field in four directions used in injury models (Stuhmiller et al. 1996;

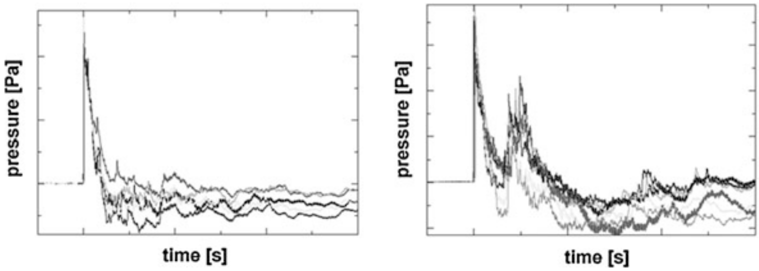
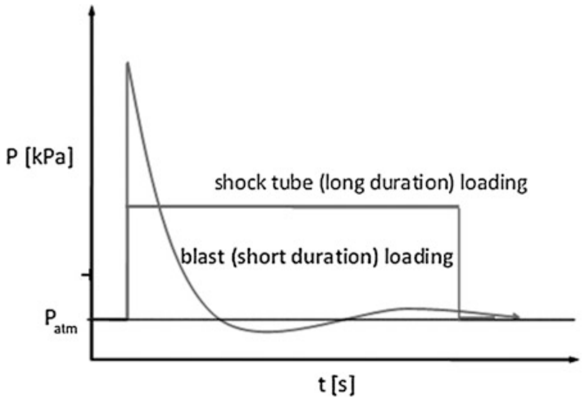


Fig. 10.16 Simple blast loading (*left*) and an example of complex blast loading (*right*)

Fig. 10.17 Comparison of blast (*short duration*) loading and shock tube (*long duration*) loading



Axelsson and Yelverton 1994) and the use of various instrumented anthropometric test devices to measure different kinetic and kinematic responses. Although these techniques measure important parameters in the blast field and response parameters, the relation of these parameters to injury mechanisms and performance metrics when protection is worn have still not been fully resolved in the literature.

Primary Blast Injury

PBI refers to injuries caused by interaction of the blast wave with the human body, and affects gas-containing organs such as the lungs, tympanic membrane and gastro-intestinal tract. Since the blast wave magnitude decreases with distance from the explosive, injury may occur in the near and mid-field regions (Fig. 10.14). The injury threshold of the tympanic membrane is much lower than that of the lungs and gastro-intestinal tract (Mayorga 1997). Exposure to blast overpressure can cause contusion of lung tissue, commonly described as (Bowen et al. 1968) ranging from serious (AIS 3) to critical (AIS 5) (AAAM 2005). Although the mode of loading varies between blunt impact and blast, the resulting pulmonary injury is generally accepted to be contusion, described as parenchymal damage occurring at the tissue level resulting in interstitial edema and capillary hemorrhage. Early experimental studies were undertaken to understand blast injury from large scale nuclear weapons and led to the widely used Bowen iso-injury curves (Fig. 10.16) (Bowen et al. 1968) which relate simple free-field blast peak (P^+) and duration (T^+) (Fig. 10.10), to the expected level of lethality. These curves have since been revised based on further analysis of the data by Bowen (Bass et al. 2006; Rafaels et al. 2010) providing an estimate of pulmonary injury under very specific conditions. Several methods currently exist to predict the potential for injury including correlations based on the blast wave transient pressure (Bowen et al. 1968), analytical models based on experimentally measured data (Stuhmiller et al. 1996; Axelsson and Yelverton 1994), and numerical approaches (Cronin et al. 2004). These methods were developed for single exposure to overpressure; however, studies have been conducted to demonstrate the effect of repeated exposures to blast, which is generally accepted to result in reduced tolerance to injury, and has been the focus of study for exposed populations such as breachers and operators of large artillery.

Injury from interaction with the blast or shock wave may also occur in underwater detonations. Detonation of an explosive underwater produces a bubble of combustion products and corresponding overpressure that exceeds the overpressure, impulse, and range that would be generated in air for the same charge size. Additionally, a series of smaller pressure waves are generated as the over-expanded bubble contracts and oscillates. In an underwater blast, fragments do not travel as far due to the viscosity of the water, and much of the wave is transmitted to a body as opposed to being reflected due to the closer impedance match between water and human tissues. For submerged bodies, injuries primarily occur to the air containing organs including the lungs and gastro-intestinal tract (Nelson 1970).

A marked increase in blast injury has been noted in recent conflicts, such as Iraq and Afghanistan, where a large percentage of the injuries are caused by blast. Further, a high incidence of mild traumatic brain injury (mTBI), evidenced by cognitive deficits often without loss of consciousness, has been identified in these recent conflicts (Gupta and Przekwas 2013). Although the mechanism of injury has not been determined conclusively, it is believed that mTBI may result from primary interaction with the blast wave leading to some damage or change in function at the tissue or cellular level. This is a very active area in blast research at present.

As may be expected, protection from primary blast injury can be achieved by increased standoff from the explosive so that the experienced overpressure decays or decreases to a tolerable level. In cases where this is not possible, protection requires decoupling of the body from the blast wave, which may be accomplished through the use of selected materials of appropriate density and mass. For example, hard armour plates attenuate blast waves due to the high impedance of the ceramic material and the mass of the plate mitigates acceleration and impact with the thorax to protect the lungs. In contrast, flexible or soft armour may enhance or mitigate blast injury depending on the material type and areal density. It has been demonstrated that low areal density flexible armour, comprising ballistic fibres with little or no permeability, could amplify the blast wave and resulting lung injury while high areal density flexible armour, due to the increased mass, attenuated the blast wave (Thom and Cronin 2009). Primary blast protection for the ear can be achieved through the use of hearing protection, which mitigates the blast wave from interacting with the tympanic membrane (Cooper and Dudley 1997), while a more complete level of protection for the head and body exists in explosive ordinance disposal (EOD) suits that include full coverage of the head and body (Makris et al. 2004; Nerenberg et al. 2002) with materials selected to isolate the body from a blast wave, even at close proximity to a detonating explosive.

Secondary Blast Injury
In most cases, fragments or projectiles are accelerated to high velocity from the detonation of an explosive. These projectiles can include fragments of a shell or bomb casing, intentional projectiles packed around an explosive or debris from the surrounding environment leading to penetration injury, or blunt injury in some cases (e.g. large fragments impacting a body wearing ballistic protection). This is known as secondary blast injury, describing that the projectiles are accelerated by the blast and typically interact with the body after the blast wave. Fragments are often characterized with distributions of mass and velocity, where initial velocities can be on the order of kilometres per second depending on the fragment size and explosive charge weight. As with all projectiles, velocity decreases with distance from the source due to drag and protection can be addressed as described in the ballistics discussion. However, in certain circumstances such EOD applications, full body coverage including extremities is required to protect the body. The required V_{50} ratings vary depending on body region and size of projectile used. For example, the required V_{50} for the chest (front) is 1433 m/s for a 16 grain Fragment Simulating Projectile (FSP) and 1036 m/s for a 64 grain FSP, while the requirement is 655 m/s (16 grain FSP) for the front of the upper legs and 701 m/s for the visor protection the head (Nerenberg et al. 2002). Details are often specified in test standards for specific applications, such as MIL-STD-662F (V50 Ballistic Test for Armour).

Tertiary Blast Injury
Tertiary Blast Injury (TBI), also termed "displacement injury" in some references, describes the effects of translation and disruption injuries to the body. Tertiary injuries range from blunt impact (e.g. blunt injuries to the thorax and head) to traumatic amputation of the extremities or complete disruption of the body in extreme cases. The injuries incurred will depend on the size of the charge, standoff between the body and charge, and the amount of protection worn. For example, tests with a Hybrid III ATD demonstrated 0.5–1.0 m horizontal translation before falling when exposed to a 20 kg C-4 charge (3 m HOB, 7 m standoff) (Manseau et al. 2006). It was noted that the head accelerations resulting from the fall were significant compared to those experienced during the primary interaction of the blast wave. The importance of tertiary injuries was recognized in many of the first studies on blast where traumatic amputations and head injuries were identified.

Injuries from anti-personnel (AP) landmines (approximately 28–500 grams of explosive) can also be considered as tertiary effects, where the detonation typically occurs in very close proximity to a limb. Exposure to a landmine blast can result in varying degrees of injury to lower extremities depending on proximity and size of the landmine, variability in explosive output, soil conditions (Nechaev et al. 1995), and protection worn. The effects of a close-proximity explosive detonation can be considered to originate from two sources: the short term loading from the initial detonation (shock transmission into the leg), and the long term loading due to expansion of the explosive gases. Clinical experience with typical landmine injuries indicates that injuries primarily occur in the foot, ankle and lower tibia, with some secondary effects from debris and soil ejecta that may be driven into the tissues if no protection is worn or the protection worn is compromised (Coupland and Korver 1991). Existing tools for the evaluation of blast mine injury include simple metal columns, mechanical legs, frangible synthetic legs and biological specimens (Cronin et al. 2011) with numerical techniques providing insights into the physics of injuries (Cronin et al. 2003).

Occupants inside vehicles exposed to IEDs may also experience tertiary injury. This scenario has become common in recent conflicts for occupants of military vehicles. In this case, the IED is typically detonated under the vehicle (under body blast) leading to an intense upwards acceleration of the vehicle and local deformation of the vehicle floor. The primary areas of injury include the lower extremity, head, neck and spine. This is a very active area in blast research at present with a focus on injury mitigation.

Protection from tertiary injury is often best achieved through increased standoff from the explosive, or the use of protective equipment that can isolate the body from the effects of blast and impact of the body with the surroundings.

Quaternary Blast Injury

This category of injury includes the effects of exposure to the very high temperatures in the near field that may lead to burns, the intense light from the explosive detonation that can cause blindness or damage to the eyes, toxic gas inhalation from the explosive combustion products or surrounding materials, and other longer term effects such as radiation in the case of nuclear bomb exposures. The latter are classified by some authors as a fifth category of injury (Quintary).

10.3 Summary

Intentional and unintentional exposure to ballistic and blast threats can result in unique serious or fatal injuries. The primary trade-offs in Personal Protective Equipment performance include weight (mobility, ergonomics and coverage) and protection level. Methods and materials have been developed to provide ballistic protection for critical areas of the body such as the thorax and head, although it must be emphasized that the protection level corresponds to a particular projectile and impact velocity. Further, these protections levels are probabilistic in nature (e.g. V_{50}) so that protective body armour is defined as bullet resistant, not bullet proof.

A principal element of protection in blast is to avoid perforation of the protection so there is no direct interaction between the blast and the body. Protection from blast injury is best achieved through increased standoff from the explosive, or the use of protective equipment that can isolate the body from the effects of blast and impact of the body with the surroundings. Protection from ballistic and blast threats presents many ongoing challenges due to the constantly evolving threats such as explosive charge size, and projectile construction and velocity. This is further complicated by new environments such as blast exposure in urban environments or closed spaces where effects may be amplified.

10.4 Exercises

E10.1: Verify the kinetic energies of the projectiles given in Fig. 10.1. Assume a 9 mm projectile is fired into a block of ballistic gelatin. Referring to Fig. 10.3, for the 9 mm projectile, at what distance is the temporary cavity the largest? Comment on what is happening at this location to cause a larger temporary cavity.

E10.2: For the following cases, determine the reflected pressure (P_R) and the reflected pressure ratio ($\frac{P_R}{P^+}$) assuming the incident wave impinges on an infinite rigid wall at standard temperature and pressure (STP).

(a) 1000 kPa static overpressure (equivalent to a near field 30 kg TNT explosion at 3 m)
(b) 750 kPa static overpressure (equivalent to a near field 20 kg TNT explosion at 3 m)
(c) 230 kPa static overpressure (equivalent to a mid-field 10 kg TNT explosion at 4 m)
(d) 85 kPa static overpressure (equivalent to a far field 5 kg TNT explosion at 5 m)

E10.3: A 70 kg man is exposed to a blast wave with an overpressure of 1000 kPa. Using the Bowen curves (Fig. 10.15), at what duration does is the 50 % lethal dose (LD) occur? What about at 500 kPa? 100 kPa?

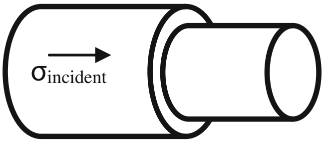
P10.1: Explain the concept of spalling in an armour impact scenario; how does it occur, what are its consequences in terms of injury and how can they be prevented?

P10.2: An elastic stress wave is travelling through steel and encounters an interface, beyond which is muscle tissue, as shown. Assume both materials are cylindrical. Calculate the ratio of the transmitted stress to the incident stress, and the ratio of the reflected stress to the incident stress. Use the given equations below, which account for the geometric impedance as well as the material impedance. Refer to Table 10.2 for the wave speed of materials.

$$\sigma_T = \frac{2 \cdot A_1 \cdot \rho_2 \cdot C_2}{A_1 \cdot \rho_1 \cdot C_1 + A_2 \cdot \rho_2 \cdot C_2} \cdot \sigma_1$$

$$\sigma_R = \frac{A_2 \cdot \rho_2 \cdot C_2 - A_1 \cdot \rho_1 \cdot C_1}{A_1 \cdot \rho_1 \cdot C_1 + A_2 \cdot \rho_2 \cdot C_2} \cdot \sigma_1$$

steel:
diameter: 0.5 m
density: 7800 kg/m^3



muscle tissue:
diameter: 0.35 m
density: 1050 kg/m^3

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11 Solutions to Exercises

Most chapters include questions which allow the reader to test and deepen his/her understanding. Exercises start with the letter E. In the following solutions to these exercises are presented. In addition problems (labelled with P) are provided. Solutions of those problems are available for lecturers only and can be obtained directly from the publishers.

E2.1: Test method: sled test, it allows for a variation of parameters at lower cost than full scale tests. Furthermore, dummy motion can be better monitored in a sled test. Crash pulse: choose a crash pulse measured in the target vehicle in e.g. a FMVSS or European homologation test for frontal impacts. If successful, further tests may be considered at more severe pulses, e.g. Euro-NCAP. ATD: use a Hybrid III dummy for comparability with the aforementioned full size tests.

E2.2: (a) EES: related to the deformation, (b) pre- and post-collision velocities, (c) delta-v, (mean, peak) acceleration.
Normally, only c. related to the injured person's car is significant. Of course, if a. and b. are known, c. may be derived thereof. Qualitative parameters e.g. integrity of the passenger compartment are also important.

E2.3: (a) Since the force-deformation characteristic may be directly input into the calculation, a single-mass system is the quickest approach. The problem could even be solved using e.g. a Matlab or Excel programme.
(b) FE-method. Explicit (forward time-integration) approach is preferred, since large deformations are expected.

E3.1: AIS 1, non-contact injury, (minimal) diffuse (non-focal) injury, GCS 15 (after re-gaining consciousness).

E3.2: (a) Padding is used to absorb impact energy, thereby attaining a lower peak head acceleration.
(b) The hard shell distributes the impact load on a larger surface. Concentrated loads would cause e.g. skull fractures.

E3.3: True. If the time interval where the HIC formula evaluates to its maximum is:

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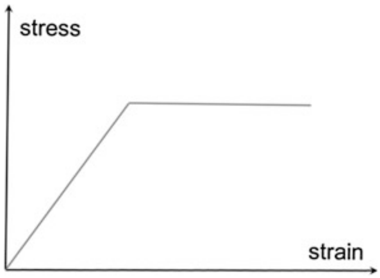
(a) Shorter than or equal to 15 ms: HIC_{36} and HIC_{15} yield the same value
(b) Longer than 15 ms: HIC_{36} is higher

E4.1: See figures in Sect. 4.2. Possible injuries involving compression are fractures or facet dislocation due to compression-flexion and compression-extension loading. Axial compression also seen in sports can result in fracture of the atlas.

E4.2: The Quebec Task Force (QTF) has developed a suitable scale for soft tissue neck injury (Table 4.2). AIS is less suitable since complaints under question are hardly life-threatening.

E4.3: As shown in Fig. 4.13 the occupant movement can be divided into three parts: backward movement, forward movement (rebound) and belt restraint phase. The belt interacts with the occupant in the last phase.

E4.4: A slight flexion straightens the neck and thus increases the injury risk when the neck is impacted axially.

E5.1: 

E5.2: C = 45/200; V = 0.045/0.04; VC = 0.25

E5.3: In the CTI the effects of both belt and airbag on the thorax in a frontal crash are included in combined form. The first term (A_{max}/A_{int}) thereby reflects the action of the airbag, in that it is postulated that the airbag mainly decelerates the thorax by application of a load distributed over large parts of its surface, while the second term (D_{max}/D_{int}) describes a more localized loading due to the belt which causes primarily a deflection.

E6.1: Difficulties to define tolerance levels include the complex anatomy (several organs of different mechanical properties) as well as problems related to the design of suitable experiments to determine the biomechanical response.

E6.2: NHTSA recommends not to disable the airbag. Furthermore it is important that the (lap) belt is positioned properly over the pelvis.

E7.1: M = F (a/2) sin(α)

E7.2: Velocity at impact: $v = \sqrt{2gh}$, $v = 4.2$ m/s, deceleration: $a = v^2/2d$, $a = 441$ m/s^2 = 44.9 g, force: F = m a, $F = 28.7$ kN.

E8.1: Common causes for arm injury include (a) direct contact to airbag, (b) contact to interior of vehicle, (c) contact of the arm with an interior part of the vehicle as a result of the arm being flung by the airbag, (d) inboard limb injuries due to contact with another occupant sitting next.

E8.2: Arm injuries are not assessed in ECE R95. The SID dummy that is used in R95 is not equipped with forearms.

E8.3: Overuse injury, e.g. in the shoulder.

E8.4: The shoulder is mechanically less stable than the hip joint (compare e.g. the ball-and-socket joint as shown in Figs. 7.1 and 8.3).

E10.1: The kinetic energy is given by: $E = 1/2\ \mathrm{mv}^2$
Substituting the values for the three projectiles in Fig. 10.1:
0.50 BMG: $E = 0.5\ (0.045)\ (908)^2 = 18550$ J $= 18.6$ kJ
7.62×51 NATO: $E = 0.5\ (0.00953)\ (833)^2 = 3306$ J $= 3.3$ kJ
9 mm Luger: $E = 0.5\ (0.00804)\ (360)^2 = 521$ J $= 0.52$ kJ
For the 9 mm projectile in ballistic gelatin, the temporary cavity is largest at approximately 23 cm. This is caused by the projectile tumbling in the ballistic gelatin.

E10.2: Using Eq. 10.7 with $P_{\mathrm{atm}} = 101.325$ kPa

$$\frac{P_R}{P^+} = 2 \cdot \left[\frac{7P_{atm} + 4P^+}{7P_{atm} + P^+}\right]$$

(a) 5510 kPa, 5.51
(b) 3812 kPa, 5.08
(c) 798 kPa, 3.47
(d) 225 kPa, 2.64

E10.3: From the Bowen curves for pulmonary injury (Fig. 10.15), the duration at which a 1000 kPa overpressure will result in LD50 is about 1.4 ms. At 500 kPa, the duration is about 2.2 ms. At 100 kPa, the LD50 threshold is not reached.

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