Universidade de Aveiro Departamento de Engenharia Mecânica
2012

Fábio António Oliveira Fernandes

Análise de lesões decorrentes de impactos com capacetes

Analysis of injuries resulting from impacts with motorcycle helmets

Universidade de Aveiro Departamento de Engenharia Mecânica
2012

Fábio António Oliveira Fernandes

Análise de lesões decorrentes de impactos com capacetes

Analysis of injuries resulting from impacts with motorcycle helmets

Dissertação apresentada à Universidade de Aveiro para cumprimento dos requisitos necessários à obtenção do grau de Mestre em Engenharia Mecânica, realizada sob a orientação científica de Prof. Dr. Ricardo José Alves de Sousa / Professor Auxiliar Convidado,Departamento de Engenharia Mecânica, Universidade de Aveiro.

Dissertation presented to the University of Aveiro as a requirement to obtain the Master Degree in Mechanical Engineering, and carried out under the scientific supervision of Professor Dr. Ricardo José Alves de Sousa/ Invited Assistant Professor , Department of Mechanical Engineering, University of Aveiro.

O júri

Presidente	**Prof. Doutor António Manuel Godinho Completo** Professor Auxiliar do Departamento de Engenharia Mecânica da Universidade de Aveiro
Arguente	**Prof. Doutor Renato Manuel Natal Jorge** Professor Associado do Departamento de Engenharia Mecânica da Faculdade de Engenharia da Universidade do Porto
Orientador	**Prof. Doutor Ricardo José Alves de Sousa** Professor Auxiliar Convidado do Departamento de Engenharia Mecânica da Universidade de Aveiro

agradecimentos / acknowledgements

Ao completar mais uma fase do meu percurso académico, é com o maior orgulho que aqui presto os meus sinceros agradecimentos às pessoas que me ajudaram na realização desta dissertação, a qual não estaria terminada sem o seu apoio e ajuda.

Ao Professor Doutor Ricardo Sousa pela sua orientação, disponibilidade, recursos disponibilizados, críticas construtivas e ideias importantes na concretização deste trabalho, não esquecendo a sua enorme paciência.

Gostaria de agradecer aos colegas do GRIDS pelo excelente ambiente de trabalho proporcionado. Um especial obrigado ao Rodrigo Coelho, ao João Caseiro e também ao Miguel Martins pela pronta disponibilidade na resolução de qualquer problema.

Ao meus amigos que também foram importantes nesta fase.

Este projecto contou com a colaboração de outras entidades, que foram fundamentais para alcançar os objectivos. Um obrigado à CMS, em especial ao Mário Rui Santiago por todo material e documentação fornecida.

A special thanks to Professor Dr. Rémy Willinger for the cooperation and knowledge transmitted.

Um muito obrigado aos meus pais, por me terem proporcionado todas as condições e incentivado ao longo do meu percurso académico, que sem o vosso esforço e dedicação nada disto tería sido possível. Um especial obrigado também à minha avó.

Finalmente, um agradecimento especial à Cátia pela compreensão e enorme paciência, apoio, carinho e motivação que foram fundamentais nesta fase da minha vida.

"Even a journey of 1,000 miles begins with a single step"

– Lao Tzu

Palavras-chave

segurança passiva, traumatismo craniano, capacete, simulação numérica, movimento rotacional, modelo biomecânico da cabeça em elementos finitos, cérebro, normas

Resumo

Neste trabalho efetua-se uma avaliação do desempenho de um capacete rodoviário já comercializado e aprovado pela maioria das normas de segurança atuais. Este desempenho é avaliado através da reprodução fidedigna de impactos semelhantes aos que ocorrem comumente em acidentes reais, onde ambos os movimentos, translacionais e rotacionais estão presentes. Duas validações foram realizadas por comparação com resultados experimentais: uma relativa ao modelo constitutivo do poliestireno expandido, que integra a camada de absorção de energia do capacete e outra relativa aos valores das acelerações do centro de gravidade da cabeça após os impactos definidos pela norma de segurança ECE R22. Após validação, um impacto oblíquo foi simulado e os resultados foram comparados com os valores limites de traumatismo craniano, a fim de prever as lesões na cabeça resultantes de acelerações rotacionais, não previstas na norma referida. A partir desta comparação, concluiu-se que lesões cerebrais, tais como concussão e lesão axonal difusa podem ocorrer mesmo com um capacete rodoviário que foi aprovado pela maioria das normas atuais, e apenas replicando um impacto que vulgarmente é observado em colisões reais. As mesmas lesões foram previstas após avaliação das lesões num impacto da norma ECE R22 com uma cabeça biomecânica modelada em elementos finitos. As conclusões apontam para uma recomendação assertiva no sentido de que os efeitos decorrentes de desacelerações rotacionais devem também ser contemplados pelas normas de segurança vigentes e que os procedimentos de teste actuais devem ser melhorados, especialmente a cabeça de teste, a qual não é capaz de prever lesões, para promover a segurança entre os motociclistas.

Keywords passive safety, head injuries, helmet, numerical simulation, rotational motion, biomechanical finite element head model, brain, standards

Abstract

In this work it is carried out the performance assessment of a motorcycle helmet, approved by the majority of current standards and already placed on the market. The evaluation is based on accurate reproduction of impacts that are similar to the ones that commonly occur in real crashes, where both motions, translational and rotational are considered. The numerical framework is validated against two different set of experimental results. The first concerns the constitutive model of the expanded polystyrene, the material responsible for energy absorption during impact; the second related to the head's centre of mass acceleration after the impacts defined in the European ECE R22 standard. Both were validated against experimental data. Doing so, an oblique impact was simulated and the results were compared against head injury thresholds in order to predict the resultant head injuries. From this comparison, it was concluded that brain injuries such as concussion and diffuse axonal injury can occur even with a helmet that was approved by the majority of the helmet standards, that unfortunately do not contemplate rotational components of acceleration. The same lesions were predicted after assessing injuries resulting from an impact defined by the ECE R22 standard with a biomechanical FE head model. At the end, conclusion points out a strong recommendation on the necessity of including rotational motion in forthcoming motorcycle helmet standards and improving the the actual test procedures, especially the test headform, which is not able to predict lesions, to improve the safety between the motorcyclists.

Contents

List of Figures

List of Tables

Acronyms

ABPT	Applied Brain Pressure Tolerance
AIS	Abbreviated Injury Scale
ANSR	Autoridade Nacional Segurança Rodoviária (Portuguese Road Authority)
CNS	Central Nervous System
COG	Centre Of Gravity
CPU	Central Processing Unit
CSDM	Cumulative Strain Damage Measure criteria
CSF	Cerebrospinal fluid
CT	Computer Tomography
DAI	Diffuse Axonal Injury
DBI	Diffuse Brain Injury
DOT	Department Of Transportation (from USA)
EDH	Epidural Haematoma
EPS	Expanded Polystyrene Foam
FEA	Finite Element Analysis
FEM	Finite Element Method
FEHM	Finite Element Head Models
FMVSS	Federal Motor Vehicle Safety Standard
GAMBIT	Generalized Acceleration Model for Brain Injury Threshold
GSI	Gadd Severity Index
HIC	Head Injury Criterion
HIP	Head Impact Power
HUMOS	Human Model for Safety
ICH	IntraCerebral Haematoma
KTH	Kungliga Tekniska Högskola (FEHM of Royal Institute of Technology)
MIPS	Multi-direction Impact Protection System
MTBI	Mild Traumatic Brain Injury
MRI	Magnetic Resonance Imaging
NHTSA	National Highway Traffic Safety Administration
PHPS	Phillips Head Protection System
PLA	Peak Linear Acceleration
PTW	Powered Two Wheelers
RMDM	Relative Motion Damage Measure
SDH	Subdural Haematoma
SFC	Skull Fracture Criterion
SIMon	Simulated Injury Monitor
SMF	Snell Memorial Foundation
SUFEHM	Strasbourg University Finite Element Head Model
TBI	Traumatic Brain Injury
THUMS	Total Human Model for Safety
UCDBTM	University College Dublin Brain Trauma Model
UCDBTM	University Louis Pasteur
VMSS	Von Mises Shear Stress
WSTC	Wayne State Tolerance Curve
WSU	Wayne State University
WSUHIM	Wayne State University Head Injury Model

Chapter 1

Introduction

Reading guide

The work here presented is divided in 4 chapters. In order to supply the reader with a practical reading guide, it is prepared in the next lines a small description of all chapters and their contents.

Chapter 1 - Introduction
Chapter one presents the objectives and motivation for the base of this thesis. Also, it is provided is this reading guide.

Chapter 2 - State of Art
This first part presents the topics involved and treated during the development of the thesis. Explains concepts related to the impact protection and energy absorption. Refers to the development of helmets and different types existing today as well as the materials typically used in impact situations. It is made a briefly introduction to head anatomy with following description of head injuries mechanisms and associated thresholds. It includes a literature review concerning the research carried out, on this field and also the suitability of FEM in the prediction of head injury, where it is made a literature review about FE head models. Also, it is given a literature review of helmet test standards most commonly used and some drawbacks of them are highlighted. The recent attempts to overcome these drawbacks are also reviewed where it is presented new solutions to overcome problems like the not assessed rotational motion.

Chapter 3 - Finite element modelling - A framework for simulating helmeted impacts
This chapter describes the procedures and methods used to characterize the mechanical behaviour of materials involved in this work. It describes also the energy absorption tests simulated and the comparison of the results of these simulations against experimental data, validating the helmet model. This chapter also presents the results obtained from the energy absorption tests and oblique impacts simulated in Abaqus and these results are compared with head injury thresholds, in order to predict the resultant head injuries. At the end, the impact point P of ECE R22.05 is performed with a biomechanical finite element head model and again, the prediction of head injuries is made.

Chapter 4 - Conclusions and future work
This chapter presents general conclusions and discuss the results obtained in this work. It also has some suggestions and ideas to implement in future works related to this study.

1.1 Motivation

One of the most important parts of the human body is the head, where are the eyes, the ears, the nose, the mouth and especially the brain which is one of the most vital organs of the human body. An impact to the head can have serious consequences and could even be fatal. For these reasons, head protection and safety are very important.

The brain is among the most vital organs of the human body. From a mechanical point of view, the natural evolution of the head has lead to a number of integrated protection devices. The scalp and the skull, but also to a certain extent the pressurized subarachnoidal space and the dura matter, are natural protections for the brain.

Since a long time ago that helmets are used as a protective system, first to protect the head in combats. More recently, it was noticed that head injuries could easily occur without penetration, such as brain injuries and since then many researchers have tried to improve helmets by studying the head injury mechanisms. Nowadays, the helmet is the most current protection system of the head and helmets are used in a large number of different applications such as military, in emergency and protective services, in different types of work, in different types of sports and in vehicles like bicycles and motorcycles among others. As each of these applications have different technical requirements, helmets were evolved according to their specific applications and the evolution of science, such as material science, helmet standards, injury mechanisms and criteria.

In fact, road accidents are the main cause of head injuries and motorcyclists greatly contribute to this [1, 2, 3, 4, 5, 6, 7, 14]. Motorcyclists are less protected against road accidents than the users of some other vehicles and only have one effective mean of protection, the safety helmet. Thus, the current motorcycle helmet standards are responsible for the improvement of helmet quality and effectiveness by restricting the market just for the ones that are able to fulfil all the requirements of a certain standard, making thus safer helmets. However, there is a delay between the evolution of standards and the evolution of science, for example in injury mechanisms and criteria. For example, the current standards do not take into account the rotational acceleration and its effects, though many researchers have already performed some studies where they concluded that rotational acceleration has an important effect and contribution in the occurrence of some brain injuries, such as diffuse axonal injury (DAI). However, this area is still an active area of research, and more in-depth investigations are still required to determine the injury tolerance of the brain to rotational acceleration. This and other effects were and still are studied with the goal of establishing thresholds to the different head components, such as brain, bridging veins, etc.

There are experimental studies [116] made in human corpses and animals such as monkeys and pigs in order to obtain data about the mechanical properties of head/brain components. This data, allied with the increase of computer power and recent advances on computational modelling [7], particularly with the finite element method, allow to perform tests and assess the injuries that result from an impact. The modelling of biomechanical human head models using finite elements provide a strong basis for helmet design improvements over the current headforms used by the current standards. These FE models, based on realistic geometric features of a motorcycle helmet, head model and known material properties allow researchers to discover the critical parameters, biomechanical limits and relate them with impact forces and accelerations. This allows a further accurate, computational-based prediction of the brain injury, relating it to the medical investigations observed in autopsy of real accidents. Some researchers attempted and proposed head level injury predictors (biomechanical criteria).

Helmet design by helmeted-head impact simulation is a complex problem. This complexity arises from the nature of the individual components constituting this system. For example, it is easy to model the metallic headform for impact simulation but the actual human head modelling for the same purpose is still under gradual progression. Moreover, new developments in helmet technology are noticeably increasing to overcome some of the helmet outstanding problems such as weight, comfort, etc. However, at present, helmet manufacturers rely solely on experimental testing to verify their helmet design and these tests are usually made according to standards requirements. This is of course a costly procedure and have limited flexibility.

Motorcycle helmet standards take a long period of time to be upgraded, especially the ECE R22.05 [8], which means that currently helmets are assessed by outdated methods and criteria, using headforms and assessing the peak translational acceleration on the center of it and using HIC as criteria.

Thus, the main goal of this work is to analyse the injuries to the head level resulting from impacts with motorcycle helmets, by the implementation of a numerical model for simulations with a helmet-head system and even the inclusion of a biomechanical FE head model for the head with a previous validation with a headform and experimental data. The major motivation behind this thesis resides in assessing current standards according to the results of this work and based on those results establish a solid ground for further current standards and thus promote safety in motorcycle helmets.

1.2 Objectives

The main goal of this work is the development and testing of a finite element model of a commercially available helmet. The test conditions for the simulations are those prescribed by the ECE 22.05 standard. After the finite element helmet model is fully developed, the results are compared with experimental data provided by the manufacturer, and possible reasons for discrepancies are analysed, in order to validate the finite element model and simulate the energy absorbing tests of the ECE R22.05 standard accurately.

In recent years, the availability of increasingly powerful computers has made it possible to use advanced calculus techniques (i.e. finite element methods) even for problems characterised by severe non-linearity and material failures.

Moreover, the use of virtual models gives superior flexibility during the design process owing to the simplicity with which the model is modified and re-analysed. Indeed, by virtual testing, it is possible to assess the influence of a large number of parameters in a way that would be extremely costly and less flexible for experimental testing.

A first approach to this work will be modelling the behaviour of expanded polystyrene, so the results obtained are in agreement with the experimental data. After the validation of the polystyrene model, proceeds to the validation of the helmet model by using a headform according to ECE R 22.05 requirements. After the validation of the results obtained with the headform, these same results will be validated against experimental data provided by the helmet manufacturer. Nonetheless, it would be pleasant the inclusion of an advanced biomechanical FE head model to simulate an impact closer to the real conditions instead of a rigid head (headform). This biomechanical head model will allow a further accurate computational-based prediction of the brain injury.

For the simulations it will be used the Explicit version of the commercial FE package Abaqus [9] because it is suitable for realistic simulations, for modelling highly nonlinear behaviour including complex materials, severe deformation and contact and for it analysis capabilities. Thus, it is suitable for modelling complex models such as this helmet-head model.

The desire of implementing a FE human head model is justified by the expectation that even if a helmet passes shock absorption tests required by the European standard ECE R 22.05, simulations performed with an anatomical head could show that severe injuries could happen.

Therefore, assessing the current standards according to the results of this work and based on those same results try to propose improvements in current standards and thus promote safety in motorcycle helmets, which is a clear objective. Moreover, it is also an objective the creation of some kind of predictor or threshold, relating any kind of damage such as a brain injury like DAI or subdural haematoma (SDH), with impact forces, accelerations, intra-cranial pressure and von Mises stress, etc. This would provide a strong basis for helmet design improvements.

Chapter 2

State of Art

This first part presents the topics involved and treated during the development of the thesis. Explains concepts related to the impact protection and energy absorption. Refers to the development of helmets and different types existing today as well as the materials typically used in impact situations and new concepts. It includes a briefly introduction to the head anatomy, injuries and their mechanisms and also a literature review concerning the head injury criteria and finite element human head models. Also, it is given a literature review of current helmet test standards most commonly used and new tests and helmet concepts to assess and overcome the not evaluated rotational motion.

2.1 Motorcycle helmet

Road Accidents are one of the major causes of death in the World [1]. About 31 thousand people die and 1.6 million people are injured every year in the European Union as a direct result of road accidents [2]. Motorcyclists are less protected against road accidents than the users of some other vehicles, such as car occupants that are protected by safety belts, airbags and even by the body structure of the car, while motorcyclists only have one effective mean of protection, the safety helmet. This is also confirmed by Penden [3] and by Koornstra *et al.* [4], that the risk of motorcyclists getting killed in a road accident is higher than other vehicles and also by Lin and Kraus [5] that reports motorcycle's riders are over 30 times more likely to die in a traffic crash than car occupants. Thus, motorcycle crash victims form a high proportion of those killed and injured in road traffic crashes, as shown in table 2.1. In Portugal, at the year of 2010, 7603 of powered two wheelers (PTW) drivers suffer injuries, 6848 of which were minor injuries, 578 were severe injuries and 177 of the victims died, which results in 24% of all road accident fatalities and 16% of all road accident injuries [10].

Other statistics show that motorcycles comprise only 6.1%, 5.1% 2.9% and 2.4% of all motorized vehicles in European Union, Japan, Australia and USA respectively [11]. Nonetheless, motorcyclists account for 14.6% of total road-user fatalities in European Union, 12.1% in Australia, 9.4% in the USA and 9.2% of total traffic fatalities in Japan [12]. These statistics demonstrate again the inferior capability of protection of this type of transportation.

Table 2.1: Fatalities and sever injures of each vehicle category in Portugal at the year of 2010 (Source: ANSR [10]).

Vehicle Category	Deaths (%)	Seriously Injured (%)
Heavy Vehicles	2	2
Cars	53	53
PTW	24	22
Bicycles	4	3
Pedestrians	15	19

A more recent study shows that in the European Union, road accident fatalities increased 17,7% relatively to the PTW involved in all traffic accidents and the associated number of fatalities was almost 6000 in 2008 [13]. In the developing countries, where the motorcycle is the main source of transportation, the contribution to the total road traffic fatalities was about 90% [1].

As already shown, motorcyclists are at high risk in traffic crashes, where the head is one of the areas subjected to sever and fatal injuries. Head injury is one of the most frequent injuries that result from motorcycle accidents, as shown in figure 2.1, where head injuries occurred in 66,7% of the cases of COST database [14]. This study also reports that the majority of these injuries were severe.

Other statistics on motorcycle accidents shows that between 2000 and 2002 in the USA, about 51% of unhelmeted riders suffered head injuries as compared to 35% of helmeted riders [12], showing thus the importance of wearing helmet. In the same study, is shown that in 27% of fatalities, the only injury present was head injury. In 2008, 42% of fatally-injured motorcyclists (822 deaths) were not wearing helmets and NHTSA estimates that the majority of these unhelmeted motorcyclists would have survived if they had worn helmets [15] and also estimates that motorcycle helmets are 37% effective in preventing fatal injuries [16]. This effectiveness has increased over the years possibly due to improvements in helmet design and materials [17]. Brown *et al.* [18] concluded also that riding and crashing a motorcycle while unhelmeted, is associated with more frequent and more severe injuries and increased mortality. King *et al.* [19], showed that the acceleration transmitted to the head is almost superior in unhelmeted cases, with a few exceptions in angular acceleration.

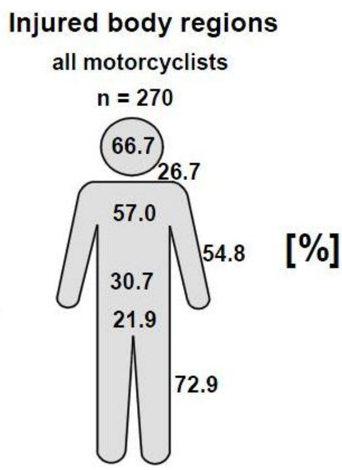


Figure 2.1: *Injured body regions of motorcyclists (Source: COST database [14]).*

Other studies show that helmets reduce the risk of death in motorcycle collisions by approximately 42% and reducing the risk of head injury by 69% [20, 21].

As a result of motorcycle accidents, head injury is considered a major cause of death, accounting for 70% of the total deaths where helmet usage is the most important factor in preventing it and in reducing the risk of head injuries from motorcycle crashes [1, 22, 23, 21]. Hence, a motorcycle helmet is the best protective gear that is possible to wear while riding a motorcycle because in road accidents this is the only effective mean of protection offered to motorcyclists, which is intended to protect motorcyclists from sever or even fatal injuries. Motorcycle helmets provide the best protection from head injury for motorcyclists involved in traffic crashes, which is a main cause of death and disability associated with other serious injuries.

Although some argues about motorcycle helmets, for example, their usage decreases motorcyclist vision and increases neck injuries, motorcycle helmets were found to reduce the risk of death and head injury in motorcyclists that crashed, therefore helmet's benefits and it's usage is advised by several studies [1, 12, 15, 16, 17, 18, 22, 24, 21]. Therefore, the main goal of a motorcycle helmet is the protection of the head during an impact, by preventing or reducing the occurrence and the severity of head injury and thus saving the rider's life.

2.1.1 Origins

Since a long time ago that helmets have been used as a primary form of protection, by protecting the head against weapon's strikes, preventing against any of penetration. Thus, the primary helmet's function was the reducing or the preventing of the likelihood of head injury in combats. An example is the helmet represented in figure 2.2, which dates back to the 5^{th} century B.C. Following the evolution of ancient societies, the materials and the construction techniques of helmets become more advanced. However, the helmets evolved and diversified with the emerging of new needs of protecting the head against any kind of impact.

Figure 2.2: *Ancient Greek Corinthian bronze helmet - 5^{th} century B.C. [25].*

In the early 1900s, with the widespread introduction of the motorcycle arises the need of a crash helmet. Initially, motorcycle helmets were no more than leather bonnets, first used in racing and usually worn with goggles. These helmets were adapted from earlier aviators which main goal was keeping the head comfort and so, almost no protection was provided to the head. Thus, the problem of the non-existence of a crash helmet persisted. One example is shown in figure 2.3.

From this point, helmets evolved based on the understanding of what a helmet should do, in other words, understanding the biophysical characteristics of the head and the development of kinematic head injury assessment functions [26]. Therefore, it was realized the necessity of a hard outer shell to distribute the applied force and thereby reduce the localization of the impact load and thus the likelihood of skull fracture. Moreover, the evolution of materials science was also crucial in the helmet's evolution.

The concept of a hard shell dates back to ancient Greek time, which objective was the force distribution to the head, which would reduce the probability of skull fracture, or maybe it was basically seen as a simple better way to deflect objects from the head [26]. However, as already referred, these first bonnets used as racing motorcycle helmets have no hard shell.

A few years later, it was created a helmet that was constituted by some individual hard leather pieces, usually sewn to a hard fibre material crown section and lined with felt or fleece which a few years later was replaced by an inner suspension. This new feature increased the capacity to absorb and distribute the impact's energy to the head more effectively than the previously ones [26]. This new device was the solution to the necessity of introducing a good absorbing impact energy material to reduce the inertial loading on the head and thus reduce the likelihood of injuries due to induced accelerations.

Figure 2.3: *Leather bonnet [27].*

In the early 1930's, the first hard shell of modern motorcycle helmets was constructed which was made of several layers of cardboard glued and later, it was constructed by impregnating linen with varnish resins, that allowed the cure into the desired solid shape [26]. Due to the shape, the early helmets were called by "pudding bowls".

In 1939, it was introduced by Riddell the first helmet with moulded plastic shell in football [26]. However, until 1941, when Riddell solved the cracking problem of the plastic shells, these helmets had a bad reputation. Despite the application is not the same, there was almost no difference between these two types of helmets until the middle of the 20^{th} century, when it was recognized that in the case of motorcyclists, they deal with one-time life threatening blow that could occur easily in a fall [26].

A few years later, Holbourn [37, 38] performed an important study where it was understood that non-penetrating head injuries are caused by short-duration accelerations acting on the head and its contents. These acceleration injuries are the most common and dangerous form of injuries for motorcyclists and are often caused by blunt impact rather than penetration [36].

In 1953, Turner *et al.* [28] introduced the padding of modern helmets, again in football helmets, which consisted in resilient closed cell rubber foam that was placed in the interior of the shell to dissipate impact energy effectively. However, this design was very heavy and as a result suspension design continued to prevail.

At the same time, Roth and Lombard [29] presented the modern helmet as known today, represented in figure 2.4. Its hard shell was constructed by 4 layers of fibre glass and several materials were used as padding material, such as expanded polystyrene foam or polyurethane foam. At first, polyurethane foam was mostly used, but due to the better properties of the expanded polystyrene foam (EPS) (cheap, readily available, relatively easy to manufacture and a good crushable energy absorbing material), it was the most used as padding material.

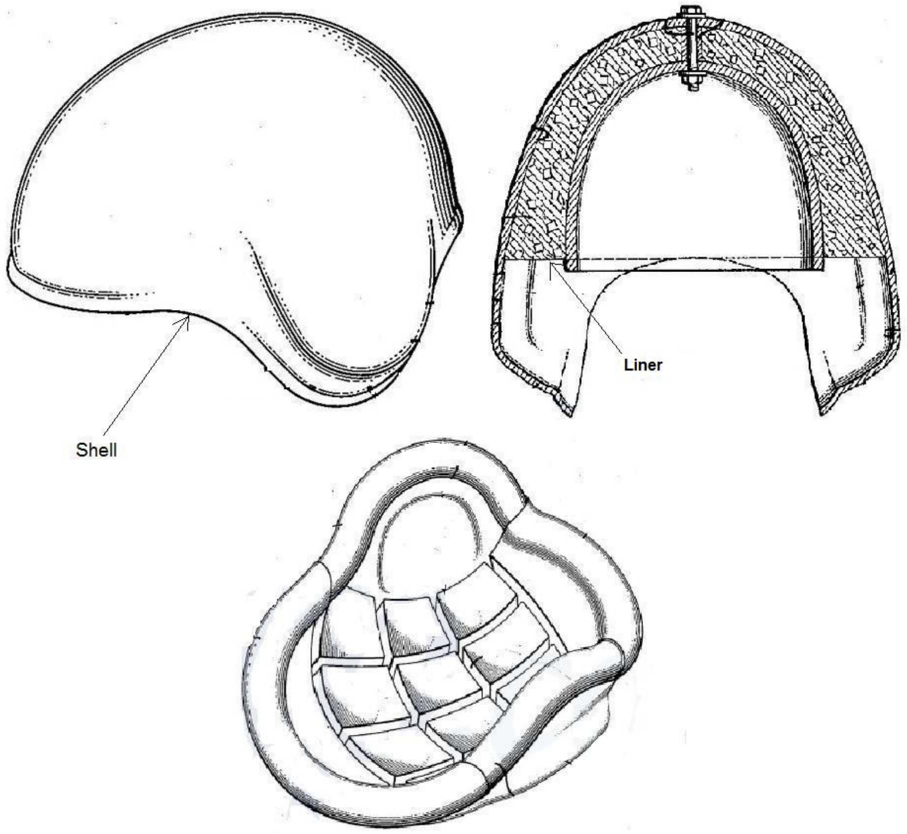


Figure 2.4: *Roth's and Lombard's crash helmet [29].*

One year after, some auto racing enthusiasts began to manufacture helmets in a garage behind the Bell Auto Parts store in Southern California [26]. The initial helmet's shell was hand laminated fibreglass resin composites with a thick semi-rigid foam polyurethane liner. They were the first to extend the pudding bowl to cover more area of the head, as observed in figure 2.5. Along with their extended coverage of the head, they were believed at the time to be among the most protective race helmets ever designed.

From here, the helmets evolution was restricted through the creation of standards. In 1957, a biomechanics head injury study by Snively [31], founder of the Snell Memorial

Figure 2.5: *One of the first open face Bell helmets, the 500-TX [30].*

Foundation, had a profound impact on modern helmet design and performance. This study showed that the only helmet that not result in a life threatening skull fracture, was the helmet made by Lombard and Roth's company, the only helmet that had EPS as padding material. Since then until today, this material is the most used material as foam liner. Other important studies in the evolution of motorcycles helmets were Cairns [32, 33], where in the last one, Cairns concluded that there was a need of adoption of a crash helmet standard, to compel all the civilian motorcyclists wearing helmets, what would result in considerable number of lives saved. Thus, modern helmets are capable of distribute the impact loading over a large area of the head as possible and to reduce the total force on the motorcyclist's head as much as possible. However, the evolution of actual helmets is not side by side with the evolution on the understanding of head injury mechanisms, but follows the evolution of standards, which means that if a standard is outdated, nothing requires improvements in helmets. Thus, an improved standard means improvements to helmets [14]. Newman [26] highlighted these same issues, such as the lack of progress, the actual utilization of old fashioned test methods that do not properly reflect the real life circumstances of accidents, like the biofidelity of the head form, the nature of the failure criteria, as well as the manner by which the movement of a test helmet is constrained.

Besides that, modern helmets developed for motorcycles are able to resist to very strong impacts and have helped the human head become less and less vulnerable [34], although some substantial improvements are still possible [14].

However, Fraga *et al.* [35] performed a development of a motorcycle model with focus in head and neck biofidelity and concluded that for low speed urban collisions, the protection offered by helmets has come to a point where the head tends to lose its place as one of the most fragile parts of the body.

2.1.2 Function

A motorcycle helmet is the most common and best protective headgear for prevention of head injuries caused by direct cranial impact [39].

Generally, the helmet purpose is understood as head protection against skull fractures. However, the main purpose of a helmet is not only the protection of the head from skull fractures or any kind of penetration which modern helmets are usually efficient to prevent it. Other main purpose of motorcycle helmets is the prevention of brain injury, because brain injuries are often very severe and result in permanent disability or even death. Thus, the purpose of protective helmets is to prevent head injury, by decreasing the amount of impact energy that reaches the head, reducing the severity or probability of injury [6, 40].

Beyond motorcycle crash helmets keep the head comfort, by cutting down on the wind noise and acting like a shield against the wind blasts, the bad weather conditions and any kind of object, they protect the head in case of accident by absorbing the impact and cushion the head to extend the time of impact. In order to have a perception of how a helmet behaves during an impact, it is necessary to understand all the mechanisms involved.

Helmets can be divided into two major parts depending on the main role of each one. There is a hard outer shell that distributes the impact force on a wider foam area which reduces the localization of the impact load and thus increasing the capacity of foam liner energy absorption and thus, reducing the total force that reaches the head and the likelihood of injuries like skull fractures [41]. Besides the resistance to penetration, it is the initial shock absorber in an accident [40]. The other main part is the inner liner, which is generally made of an excellent absorbing impact energy material to reduce the inertial loading on the head (by slowly collapsing under impact) and thus reduce the likelihood of injuries, especially brain injuries due to the induced accelerations. An example of these type injuries is the closed head injury type which is the most common type of head injury in motorcycle accidents, where the skull is not fractured and results from a great acceleration of the head causing brain injuries due to the movement of brain inside the skull. For example, when an impact to the back of the head occurs, the brain moves forward inside the skull, squeezing the tissue near the impact site and stretching the tissue on the opposite side of the head. Immediately after, brain rebounds in the opposite direction, stretching the tissue near the impact site and squeezing the tissue on the other side of the head. Figure 2.6 shows the mechanism behind the closed head injury as explained above.

This behaviour of the brain is explained by its consistency that make it able to move inside the skull, within the cerebrospinal fluid (CSF) and so, when an impact occurs and the helmet's energy absorption capacity is not enough, the skull stops suddenly but the brain keeps going (the inertia effect), as predicted by Newton's first law, until it collides against the skull's interior. From these collisions and other relative motions of the brain, it may sustain severe injuries such as shearing of the brain tissue to bleeding in the brain, or between it and the dura mater, or even between the dura mater and the skull. This bleeding and consequent inflammation causes brain swelling, which makes it presses harder against the inside of the skull and more damage is done to some very vital regions.

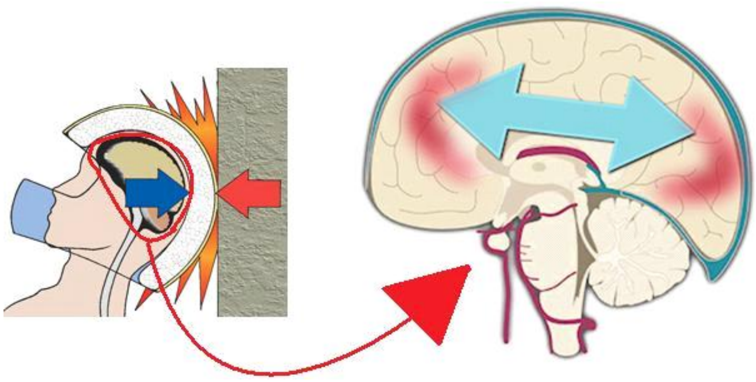


Figure 2.6: *Closed Head Injury (Adapted from [42, 43]).*

2.1.3 Design

The helmet's behaviour during an impact is mostly affected by its design [44].From what was seen in the section 2.1.2, it is important that the inner liner of a motorcycle helmet is soft and also thick so the head decelerates at a mild rate as it crushes the liner during the impact, because thicker foams remain in the plateau regime for longer compression lengths [48]. However, a helmet cannot be too thick due to practical and esthetic constraints [41], which has implications in the softness of the liner, so its thickness (typically between 20 and 50 mm) is limited by comfort and shape constraints [45]. In addition, the utilization of a thicker liner increases both the volume and mass of the helmet, with obvious disadvantages with respect to loading of the cervical spine [46, 47].

Shuaeib *et al.* [93] indicated the foam density and foam thickness as the most contributing factors in preventing head injury. Therefore, it is important to find the perfect balance between the softness and the thickness of the inner liner taking into account their limits, for example when the liner is too soft, the head will may crush it completely upon impact and as outside the liner is the hard shell, the head suddenly stops, which results in high accelerations induced to the brain and causing then brain injury. Contrary, if the impact speed is lower than the one for which it was designed, the head will be decelerated a little more abruptly than was actually necessary given the available space between the inside and outside of the helmet. Thus, an ideal helmet liner is stiff enough to decelerate the impact to the head in a smooth and uniform manner just before it completely crushes the liner. However, the required stiffness depends greatly on the impact speed of the head [39, 50, 51, 45, 52] and also in criteria used to optimise the protective padding liner [36]. Shortly, the best protection guaranteed by a helmet is for the impact speed which it was designed. Mills [53] carried out a simple mathematical approach about helmet foam liner thickness design based on the impact velocity.

Thus, one of the issues of helmet's design is the doubt of how strong a helmet should be to provide the best possible protection, where the shell stiffness and the liner density are important parameters. In practice, motorcycle helmet manufacturers design the helmets based on the speed used in energy absorbing tests in order to meet the specifications set out in the standards (experimental data is a costly choice [41, 53, 54, 55]). For example, the energy absorbing tests made up by ECE R 22.05, are done at the velocity of 7.5 m/s. Richter *et al.* [56] reported that the range of the most common head impact speed in real crashes is 5.83-8.33 m/s, which means that helmets are current designed to the most common impact speed reported. Thus, it can be said that motorcycle helmet standards try to prevent more injuries as possible. Mills [57] agree that real crashes occur at a range of impact velocities, most frequently at relatively low velocities and helmets cannot prevent all injuries, as some crashes are too severe for any wearable helmet. Bourdet *et al.* [58] reported that the current motorcycle helmets are very effective for moderate speed impacts, but its protection reaches its limits at higher energies, where the helmet deformation reaches its limits. This is supported by the analysis conducted in the COST 327 project [14], which shows that serious injuries occur at impact speeds above 13.89 m/s, almost the double of those considered in the standard tests. Bourdet *et al.* [58] and Mellor and StClair [70] postulated that if helmets could be made to absorb more energy, the number of injuries and its severity can be reduced.

Furthermore, it is shown by Bosch [36] that the optimal protective padding liner density depends on the impact site, where the protective padding liner density should be lower for the front and rear regions and should be higher for top region impact. This is justified by the indications of Gilchrist and Mills [51], where the shell geometry has influence in the shell stiffness, helmet shells are stiffest when loaded at the crown, since that site has a doubly-convex curvature and is distant from any free edges. Thence, the placing of softer liner in the crown region with the objective of compensates the high shell stiffness and attempts to make the helmet impact response independent of site [69]. Besides the geometry, the exterior finish of the shell is also important, because of the friction between it and the impact surface, which has a tremendous effect on the rotational acceleration [70, 114, 115].

Therefore, the motorcycle helmets design is affected by the requirements of each standard, which is reported in several studies, such as those performed by Hopes and Chinn [59], Chang *et al.* [60], Gilchrist and Mills [61], Kostoupulus *et al.* [62] and Yettram *et al.* [45] which have shown that because of some standards' requirements such as the penetration test, for example in Snell M2010 [63] and in BSI 6658 [71], where the helmets have to be designed with an enough stiffer shell to pass the test which may lead to higher accelerations values. In fact, this could result in a helmet with a thicker shell that typically weights about 6-8 times more as compared to the foam liner [41]. A motorcycle helmet shell, is typically 3-5 mm thick, for the current materials used [57]. This test has been also criticized by Hume *et al.* [64] since the frequency of motorcycle accidents involving spike objects is extremely small, and this test causes the outer shell of the helmet to be excessively thick which results in a heavy weight helmet. Otte *et al.* [65] conducted statistical study and his findings supported the conclusions of Hume *et al.* [64]. Mills [57] concluded exactly the same. However, some standards have no type of regulation to this type of test, for example ECE R 22.05 [8].

Helmet improvement is also achieved by defining an adequate material behaviour [58]. The force generated when a helmeted head strikes something, or as the head strikes a padded surface, depends on the crushing characteristics of impacted material [66] and also on the inherent strength of material and the size of the loading area. Therefore one of the primary objectives of a good helmet design is to maximize the area of padding that can interact with the head during impact.

The modern motorcycle helmets are capable of providing an effective protection, especially in reducing the direct force effect to the head and statistical results pointed out that helmets are effective in reducing fatalities and sever injuries [67]. However, the injuries that result from accelerations or decelerations are still a problem, mostly the rotational acceleration that remains underestimated [56], especially by the principal helmet standards. Nowadays, some researchers criticize this position of the principle standards and also some of their outdated requirements. In a helmet optimization study, Deck *et al.* [6] affirmed that today helmets are designed to reduce the headform deceleration, or in other words, helmets are designed to pass the standard requirements and not optimized to reduce head injury. Thus, there are still needs of improvements respecting to helmet design. A helmet designer must have a through and comprehensive understanding of the head impact biomechanics and a helmet should be defined in terms of how it should function rather than how it was styled or manufactured. The main way by which biomechanics has influenced helmet design is not so much in the understanding of different injury mechanisms, but rather in a better appreciation of the biophysical characteristics of the head and the development of kinematic head injury assessment functions. This insight has provided better ways to test the impact capabilities of a helmet without first placing it on a human being and a means to judge how well one might expect it to work in actual use [26].

Nevertheless, what a helmet designer normally changes to affect helmet response is the foam thickness, foam material and shell material [72].

2.1.4 Components and Materials

A typical modern motorcycle helmet is composed by six basic components:

- a very thin and hard outer shell,
- a thick and soft impact-absorbing inner liner,

- a comfort padding,
- a retention system,
- a visor,
- a ventilation system.

These and other components are represented in figure 2.7 and their functions in figure 2.8.

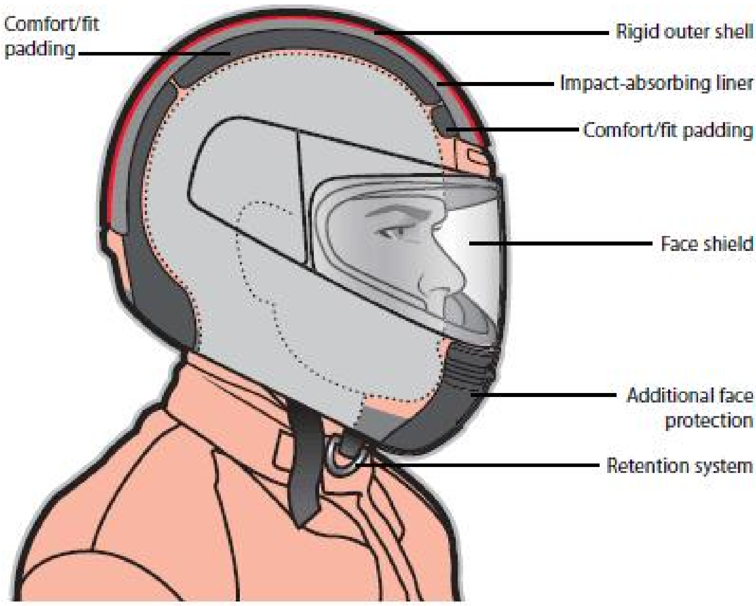


Figure 2.7: *Helmet components - Basic construction [68].*

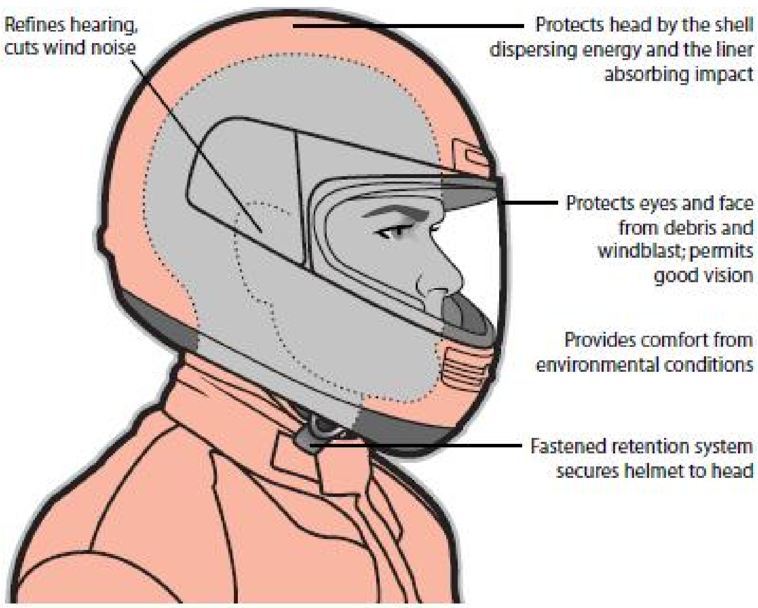


Figure 2.8: *Helmet components - Protective/Comfort functions. [68].*

Outer shell

In general, the hard outer shell is made from thermoplastic materials such as polycarbonate (PC) or acrylonitrile-butadiene-styrene (ABS), or even by composite materials such as fibre reinforced plastics (FRP) like glass reinforced plastic (GRP) or carbon reinforced plastic (CRP) or just carbon fibre or Kevlar®. The shells made of thermoplastics materials are isotropic while the FRP shells show an anisotropic material behaviour in the plane of the shell [73]. The most common FRP is GRP, which consists typically in epoxy resin reinforced with glass fibre. Commonly, thermoplastic shells are relatively cheap when compared against the composite ones, however the GPR is a relatively low cost with fairly good mechanical performance [74]. Carbon fibre and Kevlar are normally used for the most advanced helmets [75].

The outer shell is responsible for:

- spreading the impact load over a large area of the helmet, therefore reducing the concentrated stresses during an impact that reaches the head and increasing the amount of energy absorbed, by having a larger area of effective energy absorbing liner;
- the prevention of penetration of the helmet by a pointed or a sharp object that might otherwise puncture the skull;
- ensuring a structure to the inner liner so it does not disintegrate upon abrasive contact with pavement or other impacting surfaces. This is important because the foams used as liner materials have very little resistance to penetration and abrasion as showed by Richter *et al.* [56], all the helmets that showed damage to the internal lining also had a cracked shell. Also, it is necessary that it does not fracture due to the higher risk of injury when the helmet fractures [76, 77]. Thus, it can be said that one of the shell's primary roles is to provide integrity against multiple impacts, what makes it an indispensable helmet component. Also, if protects the foam against abrasion, it protects the head too.
- absorbing the initial shock in an accident. However, just a little amount of energy is absorbed. From the literature, there are several values determined, such as 30% of the total impact energy [78], 10-30% of the total energy [50, 79, 51], 12-15% in a study performed by Ghajari *et al.* [80] and 34% of the total impact energy dissipated [81]. This value is not consistent between the studies due to the differences in the tests, such as impact velocity, the materials and their properties, etc.

Inchmeal, the shells made from advanced composite materials are substituting the thermoplastics ones. However, the helmet shells made from composite materials are generally more expensive than the conventional thermoplastic material and these ones are still investigated to discover if they effectively are better than the thermoplastic ones.

During an impact, when the liner foam crushed completely, the unabsorbed energy will be transferred to the head and the impact forces developed will be very high. These impact forces will be reduced if there is another mechanism that could absorb energy, for example, if the outer shell absorbs some additional energy during impact [82]. This fact makes composite materials desirable for helmet's shell application, plus absorbing energy by deformation, the composite shell also absorbs energy through the damage mechanisms, such as fibre breakage, matrix cracking and delamination. The main advantage of using composite outer shells lies in their capability of absorbing more energy by rupture in comparison with thermoplastic outer shells. Thermoplastics shells can also absorb energy by both buckling and permanent plastic deformation [75], however a relative little amount compared to the composite shells, if the energy absorption mechanisms that relies mainly to fibre breakage of the composite are activated.

The shell stiffness has an important influence in the overall dynamic performance of the helmet. The stiffness of FRP shells is higher than the stiffness of thermoplastic shells as demonstrated by Beusenberg *et al.* [77], by comparing both experimentally, where the stiff FRP outer shell showed only minor deformation, where the energy was predominantly absorbed by foam deformation, as shown in figure 2.9. For this reason Brands [83] considered FRP shells more preferable. Gilchrist and Mills [51] studied the deformation mechanisms of ABS and GRP and concluded the same as Beusenberg *et al.* [77], the shell deformation is less in composite shell than in thermoplastic one. However, it was also reported that the impact forces with fibre composite helmet shells are much greater than those with thermoplastic shells.

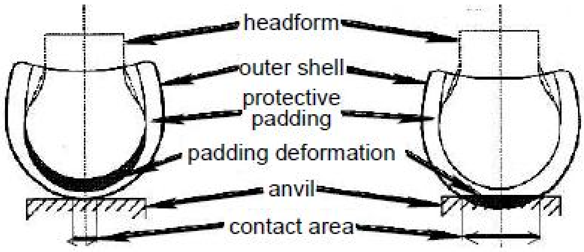


Figure 2.9: *Helmet deformation modes with FRP (left) and PC (right) shells [77].*

This could be explained by these energy absorption mechanisms that relies mainly to fibre breakage, which make composite shells so desirable. However, such behaviour cannot be achieved at low energy impacts, showing a dependence of composite shells on the impact velocity, which is greater than thermoplastic ones [50]. Also, the composite shells are much stiffer, which could leads to substantial accelerations at low energy impacts because their energy absorbing capacity relies mainly to fibre breakage. However, at higher energy impacts, composite shells provide substantial protection to the motorcyclist due to the large amount of impact energy absorbed by the helmet system until its final failure (fibre fracture) [62]. Therefore, at high energy impacts, composite shells are more effective. Gilchrist and Mills [51] also showed that to occur delamination it is necessary a great amount of impact energy, and they also reported that the impact forces with fibre composite helmet shells are much greater than those with thermoplastic shells (in a flat anvil). Mellor and Dixon [84] carried out experiments on GRP shell motorcycle helmets with various anvil shapes to investigate the impact characteristics. They concluded that GRP shell effectively spreads the load of anvils when they are of kerbstone and edge type compared to flat type.

On the other hand, at low energy impacts, a thermoplastic shell like a polycarbo-nate one might be more effective, having better protective characteristics with lower stiff shells, as demonstrated by Markopoulos *et al.* [85]. This finding is also present in other studies [50, 51, 45, 52, 39]. Despite of delamination mechanism is responsible for a good amount of energy absorbed by composite helmets, which make them particularly desirable, the thermoplastic-shelled helmets may actually performed better than for example the fibre-glass ones, because the inner liner is better at absorbing energy than the shell and plastic shells like polycarbonate flex rather than crush and delaminate, and this flexibility, lets EPS absorb more energy. The stiffer FRP outer shell is often used in combination with a low-density EPS foam, whereas the softer PC and ABS outer shells (relatively poor shock absorbing capacity) compensate their compliance with a stiffer, high-density EPS foam [36].

Moreover, the fibre-based materials had a much lower rate of fracturing, whereas plastic shells fractured more often and the rebound of a helmet with a thermoplastic shell is much higher than a fibreglass helmet, which makes the thermoplastic one less effective and thus less safe [44].

In order to assess the impact behaviour and better understood the energy absorbing mechanisms of composite shells, several finite element models of helmet with composite outer shell were proposed. The first one was proposed by Brands [83], a simplified model in which outer shell was composed of resin reinforced with glass fibres and aimed to explain the dynamical behaviour of a helmet during impact. The composite material was modelled with an elastic law considering no damage during impact and random orientation of fibres in the material, without delamination and thus no realistic behaviour was reproduced by this model.

Kostopoulos *et al.* [62] with a more realistic model that considered different composite layers, modelled with an eleastoplastic law and rupture and delamination mechanisms, studied the influence of the complex behaviour of a composite shell on the helmet's shock absorption capability. From the results of this study Kostopoulos *et al.* [62] indicates that what makes composite materials ideal for production of safety helmets is the ability to sustain extensive damage without compromising the integrity structure. In the same study, Kostopoulos *et al.* [62] also showed that composite shell systems exhibiting lower shear performance provide additional energy absorbing mechanisms and result to better crashworthiness helmet behaviour and thus, from different composite materials tested, Kevlar® shell was the one that exhibit longer impact duration and an associated lower peak acceleration value, in other words, Kevlar® fibre shell exhibits much higher absorbed energy and the energy absorbed by the foam liner was also higher. However mechanical properties of the composite materials used in the study were based on literature data and not experimental tests.

After this one, other models were proposed by Pinnoji and Mahajan [86] and Mills *et al.* [69], where the outer shell was made of resin reinforced with glass fibres. However, the outer shell modelled with an elastoplastic [86] and an elastic [69] law respectively, did not take into account delamination or rupture.

Pinnoji and Mahajan [82] performed a study with the aim of analysis the damage and delamination mechanisms of different composite outer shells of a helmet occurring during impact and compared the results with those obtained with ABS [87]. The results showed that the energy absorbed by the composite shell in helmets during damage and delamination is smaller than the energy absorbed by the plastic deformation of ABS shell. The composite shell is stiffer as compared to ABS shell in the direction of impact and gives higher impact forces on the head. Nevertheless the composite model proposed by the authors is based on the study of Kostopoulos *et al.* [62] which was not validated.

The model proposed by Kostopoulos *et al.* [62] was until recently the most advanced helmet finite element model with composite outer shell with a major drawback, the model has not been validated.

Recently, it was performed a study, divided in three parts, by Tinard *et al.* [88, 74, 7] where it was developed and validated a new finite element model of composite outer shell for a motorcyclist helmet and it was assessed and optimized regarding to biomechanical criteria.

In the first one [88], it was proposed a realistic model of a composite outer shell of commercial helmet based on experimental tests, such as modal analysis, to obtain the elastic and rupture properties of each layer, identifying the constitutive law of the composite material used. In the second one [74], the helmet model was validated against experimental data under normative conditions as prescribed by standard ECE 22.05 [8]. Nevertheless, the delamination mechanism has not been considered, which is a drawback of this model due to the importance of delamination mechanism during the crash of composite materials [62, 74]. In the third and last one [7], it is evaluated the real injury risk sustained by a detailed and validated FE head model during impacts with the approved motorcycle helmet [74] and this one is optimised against biomechanical criteria rather than standards criteria. The results showed that even if a helmet passes the tests of shock absorption required by the standard ECE 22.05, injury risks remain high.

Other studies with models of helmets with composite outer shell coupled with a FE human head model were proposed by Pinnoji and Mahajan [86], Pinnoji and Mahajan [90] and Ghajari *et al.* [89]. However, the aim of these studies was not the shell.

Recently, Pinnoji *et al.* [91] tested the possibility of outer shells made of aluminium foams, which have high strength, light weight and good energy absorption capabilities. The aim of Pinnoji was reduce the helmet weight without changing its dynamic performance. As results, it was observed that the resultant force on head is less with metal foam shell and the helmet weight is reduced by 30% as compared to ABS helmet. The headform acceleration was also lower than the ABS outer shell. However, due to the permanent deformation (deforms plastically) of metal foam, it might not behave well to a second impact in the same region. Though, this is only a possibility that was tested and the motorcycle helmets' market still is dominated by thermoplastics and fibre reinforced shells. More recently, further developments were done in this issue by Pinnoji *et al.* [437]. Different metal foam shell densities were tested and the helmet was validated. The ULP FE head model was used too to assess the helmet against biomechanical criteria. The better results were obtained with the metal foam shell of density 150 kg/m^3 which corresponds to approximately 73% reduction in mass compared to that of ABS shell and also the impact forces on the head are lower in both front and top impacts. The von Mises stress in the brain with all helmets was within the injury tolerance limits at 7.5 m/s impact velocity except for ABS helmet in top impact. In front and top impacts, the von Mises stress in the brain was reduced by approximately 25% and 22%, respectively for helmet with low-density metal foam shell compared to the ABS helmet. It was also observed that the resultant force on head was less with lower density metal foam helmet as compared to the ABS helmet.

Inner liner

The majority of motorcyclists and helmet manufacturers give a lot of attention to the outer shell and its material, because it is the helmet's outer part. However, the helmet component that absorbs most of the energy in a crash is the inner liner. The purpose of the inner liner foam is to absorb the remaining force of the impact that was partially absorbed (a small amount of energy) and dispersed by the outer shell, by crushing during the impact and thereby increasing the distance and period of time over which the head stops, reducing its deceleration, absorbing most of the impact energy and so reducing the load transmitted to the head. In the study performed by Deck *et al.* [6], one of the conclusions was that the elastic limit of the foam used as inner liner has the most important influence on HIC response but its Young's modulus has the most important influence on biomechanical head response. The liner density is also an important property because the yielding stress at which the foam crushes is directly related to it [96]. Currently, the most common liner material in protective helmets is expanded polystyrene foam (EPS), which is a synthetic cellular material with excellent shock absorbing properties and a convenient cost-benefit ratio [81], whose mass density applied in helmets varies from approximately 30 to 90 kg.m^{-3} [36, 83]. EPS absorbs the energy during the impact of the helmet, through its ability to develop permanent deformation, by crushing (foam collapsing), providing the required protection to the motorcyclist. Again, the impact velocity is an important variable since the normal velocity component largely determines the amount of EPS liner crushing [57]. It can be concluded that high-density EPS are able to absorb larger amounts of energy than low-density EPS can do, but transfer higher accelerations and forces localised at the impact point [81]. Although this type of foam has an excellent first impact performance, a motorcycle accident is typically a multi-impact situation which means that after a first impact there is no efficient shock absorber around the impacted area, because EPS has almost none elastic springback [41, 92, 93, 79]. Thus, its energy absorption capability is significantly decreased after a first impact, particularly in high energy impacts. This is one of the reasons why, if a helmet is damaged in an accident, it will has little protective value in the occurrence of a subsequent event [40]. To overcome this issue, some materials were proposed, such as:

- expanded polypropylene foam (EPP) by Shuaeib *et al.* [93];
- micro-agglomerate cork (MAC) by Alves de Sousa *et al.* [94].

The EPP is very similar to the EPS, presenting similar peak accelerations and impact durations for a same helmet with EPS, as verified by Shuaeib *et al.* [93]. The micro-agglomerate cork has a good energy absorption capacity and high visco-elastic return and its capacity to keep absorbing energy is almost unchanged after the first impact, mainly due to its viscoelastic behaviour, which is a characteristic desired in multiple impact situations. This characteristic is also important for helmets' approval by some well accepted standards that requires a test with two impacts on the same helmet point, for example Snell M2010 [63]. However, for the same volume of EPS, the MAC is a heavy solution, which is a problem for helmet approval and increase the risk of injury. However, Pedder [95] found that multiple impacts do not occur on the same helmet site in crashes, occurring at different sites as helmets rotate between the impacts. Also, the ECE R22.05 standard does not demand double impacts to same site as some of its previous versions, such as ECE R22.03 [57]

Novel configurations

In addition to new materials, several configurations have been proposed in order to enhance the energy absorption properties of motorcycle helmets, which can lead to an improvement of the safety levels provided by current commercial helmets.

Caserta *et al.* [97] replaced part of the helmet's liner by layers of hexagonal aluminium honeycombs as reinforcement material to the energy absorbing liner of a commercial helmet, as shown in figure 2.10. The results showed that this new configuration provides better protection to the head from impacts against specific surfaces, than the original EPS liner. Best results were obtained from impacts against the kerbstone anvil. However,the results obtained from impacts against the most impacted surface, the flat anvil, revealed some limitations, where at some impact points the results were even worst than the commercial helmet.

Also, the thickness of the liner necessary to accommodate honeycomb layers is extremely limited, where in a real accident scenario excessively thin layers of EPS foam could be easily broken by the honeycombs during impact and thus, the honeycomb could penetrate the scalp causing head injuries.

Recently, Blanco *et al.* [98] proposed a innovative helmet liner that consists of an ABS lamina with deformable cones in it, as shown in figure 2.11. Energy is absorbed via a

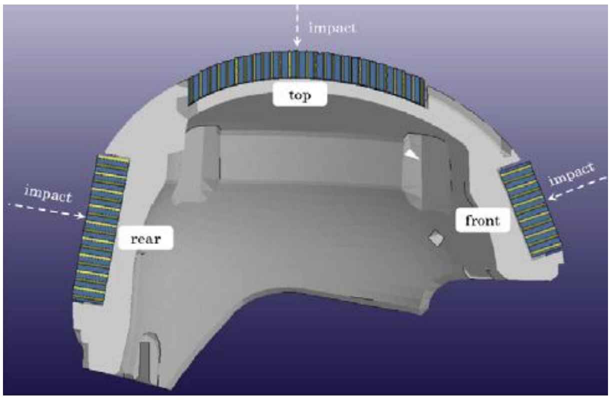


Figure 2.10: *Schematic section of the prototype liner proposed by Caserta et al. [97].*

combination of folding and collapsing of the cones. The main advantage that such liner may introduce over common EPS pads is that it allows a better optimization of energy absorption for different impact sites and configurations. Experimental and numerical tests were performed and the model was validated. No optimization was done, leaving a gap to further improvement, but the results from the model validation shows high accelerations induced to the head. Although this concept was developed to the ski helmets, it could easily be applied to motorcycle helmets.

Figure 2.11: *ABS cone liner proposed by Blanco et al. [98].*

Other configuration is the cone-head shock absorbing foam liner, developed to absorb impact force more effectively and thus, protecting the head more effectively from intra-cranial injury. This concept proposed by Morgan [99] consists in a motorcycle helmet foam liner made of two density layers, as shown in figure 2.12. The outer layer, which is the black part, is made of high density foam and has truncated cones facing inwards. The inner layer, the grey one, which is close to the head, is made of softer low density foam and has cones facing outwards.

When an impact occurs, the impact force is pushing towards the head and causes the lower density cones to compress. The collapsing of the cones causes the energy to spread sideways within the thickness of the foam liner instead of towards the head. The dispersion of the energy sideways prevents the impact energy to be translated through, until it reaches the brain and the area of effective energy absorbing liner is increased by this mechanism. As a result, the head causes the bases of low density foam cones to compress. Also, the head will experience a gradual deceleration because of the crushing/compression of the cones, minimizing the energy induced to the head. The cones reduce the deceleration of the head and the impact time of interaction is longer or the head stopping time is longer. Hence there is a reduction in the forces translated across the thickness of the new shock absorbing liner to the skull.

This concept is the most promising from the ones presented and is already used in commercial helmets by Kali protective helmets. This lighter liner helps also to reduce rotational acceleration of the head during impact.

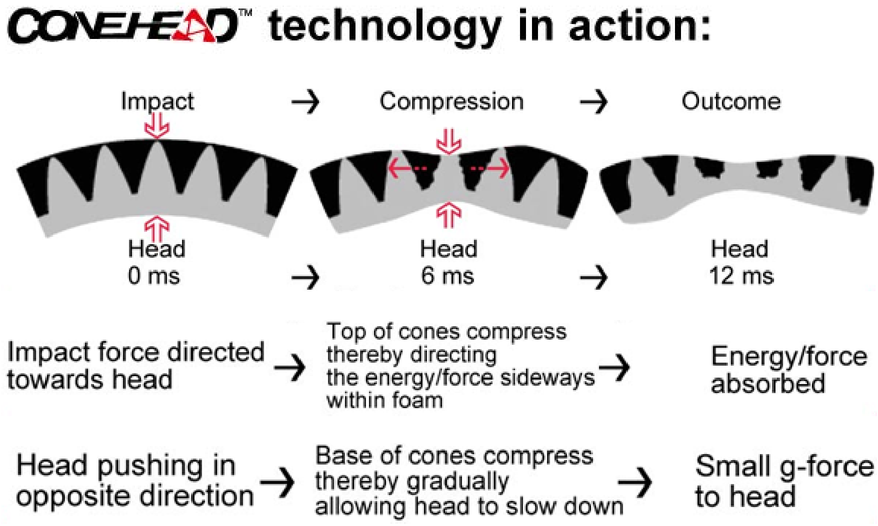


Figure 2.12: *The mechanism of Cone-head compression liner [99].*

Comfort liner

The comfort padding consists in sufficiently firm foam covered by a fabric layer that contacts and surrounds the head. This inner comfort foam is generally made of soft and flexible foams with low density as open-cell polyurethane (PU) or polyvinyl chloride (PVC) [39, 36, 83, 100, 57]. It keeps the comfort and the suitable helmet fitting by distributing the static contact forces [101, 100, 102]. The static contact force distribution is important to avoid headaches [101]. As a result of the low stiffness, the comfort foam doesn't contribute significantly to the energy absorbing properties as it crushes completely without absorbing any relevant amount of energy and, therefore, has no injury reducing effect [77, 75]. Manufacturers generally produce different sizes for every model adding different thicknesses of comfort liner to two different sizes of shell and energy-absorbing liner. This is important, as showed by Chang *et al.* [103], that assessed the effect of the fit between the head and the energy-absorbing liner and concluded that the fitting influences the acceleration induced to the head.

Retention system

The retention system or chin strap keeps the helmet attached to the head all the time. However, there are records of a considerable number of roll off helmets even with the chin strap intact and closed [56] after a crash, leaving the head unprotected from any following impact. Thus, the strap must be pulled as tight as possible against the soft tissue under the chin. All types of helmets have a retention system and the chin-strap is usually made of polyethylene terephthalate (PET) or nylon. The retention system generally consists of a strap bolted to each side of the outer shell. Mills *et al.* [69] concluded that chin-straps and also the foam inside the chin bar affect helmet rotation on the head.

Visor

The visor or face shield is made of a strong and transparent material like PC and is designed to protect the face of the rider from wind, dust, rain, insects, ultraviolet radiation and also from any object that impacts the face region. In addition, the majority of the visors are equipped with water proof coating and scratch proof coating and also it must be free of optical distortion providing a clear vision.

Ventilation system

The ventilation system ensures that fresh air is ducted into the helmet and exhaled air and humidity are vented out. One proper ventilation system not only prevents the visor from steaming up, but also ensures a pleasant climate within the helmet. A ventilation system is represented in figure 2.13.

Besides having a multi-impact protection performance, the EPP foam is a resilient material, which is pointed by Shuaeib *et al.* [93] as a material that has potential as liner material for ventilation system improvement because its resiliency allows for the ease of ventilation holes and channels moulding without the foam breakage at the stage of mould extraction. Moreover, EPS foam is brittle in its nature difficulting the introduction of ventilation channels in the foam.

A study performed by Pinnoji and Mahajan [86] indicates that the ventilation channels grooved in liner foam is not detrimental to the dynamic performance of the two-wheeler helmet.

Figure 2.13: *Ventilation system [104].*

A complete description of the manufacturing process of each component can be found in [92].

2.1.5 Types of helmets

Nowadays, there are several configurations of helmets available in the market that can be classified into five basic types of helmets for motorcyclists. From the most to the least protective, the helmet types are:

- Full face helmet;

- Modular Helmet (also known as "flip-up" helmet);
- Open face helmet (also known as "three-quarters" helmet);
- Half Helmet.

Full Face

Full face motorcycle helmets are by far the most common type of helmet [20], being the most worn type of helmet [56]. A full face helmet covers the entire head, with a rear that covers the rear of the skull at the top of the neck, and a protective section along the cheekbones to encompass the jaw and the chin, denominated chin bar. The fact of full face helmets cover the entire head means that they are the safest option between the all types of helmets, by reducing the risk of head injury providing extra strength around the entire skull. Such helmet has an opening in its shell across the eyes and nose, where is a visor that is movable and blocks out the wind, rain, dust, insects and road debris, protecting the eyes from sunbeams and also allows the assess to the face. Full face helmets usually include vents to increase the airflow to provide ventilation to the motorcyclist and help decreasing temperature inside the helmet. The figure 2.14 shows a full face helmet.

Figure 2.14: *Full face helmet by CMS [105].*

The significant attraction of these helmets is due to their protective capacity as already referred. However, the fact of them involve the entire head has some disadvantages like the increased interior heat, the sense of isolation and the reduced peripheral vision. Also, they are one of the heavier types of motorcycle helmets due to the padded interior and mainly due to the shell, that covers a larger area comparing against other types of helmets, which could be detrimental in a crash because it can cause injuries on the neck and on the brain due to acceleration or just increase neck fatigue in an ordinary ride [46, 47].

Nevertheless, the COST 327 final report [14] and Richter *et al.* [56] showed that 15.4% and 16% respectively, of total helmet damages were located at the chin guard, which shows the important protection offered by full face helmets at this area. Otte [106] concluded that impacts on the face and jaw areas are common in motorcycle crashes. In addition, Chang *et al.* [107, 52], concluded that the chin bar provided by these type of helmets offer an essential protection and the energy-absorbing capability of these could be improved by the introduction of the energy-absorbing liner in this area, plus the comfort liner. Actually, the chin bar contains a rigid foam to absorb energy [57]. Also, Mills *et al.* [69] for impacts on the front of the helmet, the chin bar foam came into play, protecting the face [108]. Thus, wearing a helmet with less coverage eliminates that protection and so the less coverage the helmet offers the less protection is provided to motorcyclist's head. Mills [57] emphasizes that chin bar prevents the lower part of the forehead and temple being struck as the helmet rotates.

According to the COST 327 final report [14] and [44], full-face helmet appears to offer better protection than the others to the entire head.

However, Shuaeib *et al.* [67] alerted that the extent of coverage in helmets like these might led to helmets with weaker lateral protection represented in helmets with thin shells at the sides, which may constitute a weak point on helmet lateral protection. Also, the side is the weaker area as compared to other helmet areas due to the edge flexibility resulted from lower stiffness associated with the larger shell curvature at the edge. The impacts to temporal regions are an important issue in real motorcyclist accidents because impacts to this region represent a considerable number of total impacts (39,5%, 12,8% and 18,3%, respectively [59, 109, 65] and the side of the skull represents the weaker area as regarding human tolerance for skull fracture due to the lower skull thickness.

Off-Road/Motocross

A motocross or off-road helmet is a lighter version of full face helmet intended for the off-road use. The motocross helmet has an extended chin bar to provide further protection. The figure 2.15 shows a motocross helmet.

Figure 2.15: *Motocross helmet by CMS [105].*

Modular helmet

A modular helmet or "flip-up" helmet is basically a combination between full face and open face helmets. It combines the safety of full face helmets, with the openness of open face helmets.

When fully assembled and closed, it resembles full face helmet by having a chin-bar for absorbing impacts in that area. Its chin-bar may be pivoted upwards (or in some cases removable and possible to wear as an open face helmet) by a special lever to allow assess to the face, as in an open face helmet, which is a great advantage in terms of comfort and practicability. Most of modular helmets also offer interlocking mechanisms, so that motorcyclist can adjust the face of the helmet at varying heights according to his preference. Some even offer a dual visor option, which is a tinted visor to shield the motorcyclist's eyes from the sunbeams, as shown in figure 2.16. However, this same mechanism makes this type of helmets the heaviest type.

There are modular helmets designed to be worn only in the closed position while riding a motorcycle and there are some that can be used in either positions, the closed one and the opened one, but the last one loses the protection offered by the chin-bar. Other disadvantage of riding a motorcycle wearing this helmet in the opened position is that the chin-bar section also protrudes further from the forehead while the is open and so riding a motorcycle with the helmet in the opened position increase the risk of neck injury in a crash.

Although, modular helmets do look the same as full face helmets, even when the front is down, they might offer a little less protection in the chin area. Nevertheless, there aren't wide scientific studies that assess the protective capacity of the pivoting or removable chin bar of modular helmets. Thus, the doubt of how protective are these helmets is still an issue, leaving an opportunity for future work. The actual state of the standards contributes somehow to this. The DOT standard does not require chin bar testing. The ECE 22.05 allows the certification of modular helmets with or without chin bar tests, where is only indicated if the helmet protects or not the chin area. However, the Snell tests the helmet's chin bar, and the modular helmets are not exception. Recently, Snell certified a modular helmet for the first time, the Zeus ZS-3000, in 2009 [110].

Figure 2.16: *Modular helmet by CMS [105].*

Open face helmet

The open face helmet covers almost the entire head, except part of the face specially the lower chin-bar, leaving this area unprotected. Thus, an open face helmet provides the same rear protection as a full face helmet, but little protection to the face and none to the chin [14], even from non-crash events like eye injuries due to some of this helmets don't have a visor to protect the users from dust.

However, many open face helmets have visors that may be used to reduce sunlight glare and also to protect the eyes from the wind, dust and even bugs, as shown in figure 2.17. It also Although allows natural air circulation, but the noise levels can be very high. Hitosugi *et al.* [111] observed that persons with open-face helmets were significantly more likely to have sustained severe head injuries, especially brain contusions, than were persons with full-face helmets.

Figure 2.17: *Open face helmet by CMS [105] .*

Half helmet

Half helmet has essentially the same front design as an open face helmet but without a lowered rear in the shape of a bowl, as shown in figure 2.18. The half helmet barely provides the minimum coverage generally allowed by some standards, by covering only the top half of the cranium and offers no protection for the face from the ears down. This issue is also highlighted by Shuaeib *et al.* [67], where the half-shell helmet is considered the most vulnerable to impacts at lateral and back head regions. Thus, a half-shell helmet offers less protection simply because it covers the least area and also does not contain much padding, absorbing less energy. In addition, this type of helmets is recognized for coming off of the motorcyclist's head in some accidents which allied with all other factors that proves the inferiority of these helmets, led to prohibit the use of half helmets in some countries [72]. A recent study evaluated the effectiveness of different styles of helmets, including half-coverage, open-face and full-face [112]. The riders wearing half helmets involved in crashes, were twice more likely to have head injuries and brain injuries than riders wearing full face helmets or even open face helmets.

Figure 2.18: *Half helmet [113].*

2.2 Biomechanics of Head Injury

Head injury sustained in traffic accidents is one of the main and common causes of death and disability despite of considerable advancements in the understanding of head injury mechanisms, especially neurotrauma and the introduction of restraint systems which reduced the number and the severity of head injuries over the past decades, as motorcycle helmets. Some studies like the one carried out by Allsop *et al.* [117], shows that 30% of all vehicular injuries are head injuries. The human head is a natural complex set of bones and several soft tissues. Any injury to these head components is considered a head injury. Before present the injury mechanisms it is presented a briefly description of human head anatomy to better understand some terminologies.

Shuaeib *et al.* [67] concluded that head injury contributes more to the fatal cases than injuries to other body parts.

2.2.1 Head Anatomy

The human head is a natural complex set of bones and several soft tissues [118]. However, a simple description is presented highlighting just the main components.

Human head can be described as a multi-layered structure, where the scalp is the outermost layer followed by skull bone (cranial and facial bones), dural, arachnoidal and pial membranes as well as CSF [119]. All together cover the brain. This multi-layered structure is shown in figure 2.19.

The external layer is the scalp that normally is 5 to 7 mm thick and consists of five layers of soft tissue that covers the skull:

- the hair-bearing skin (cutaneous layer),

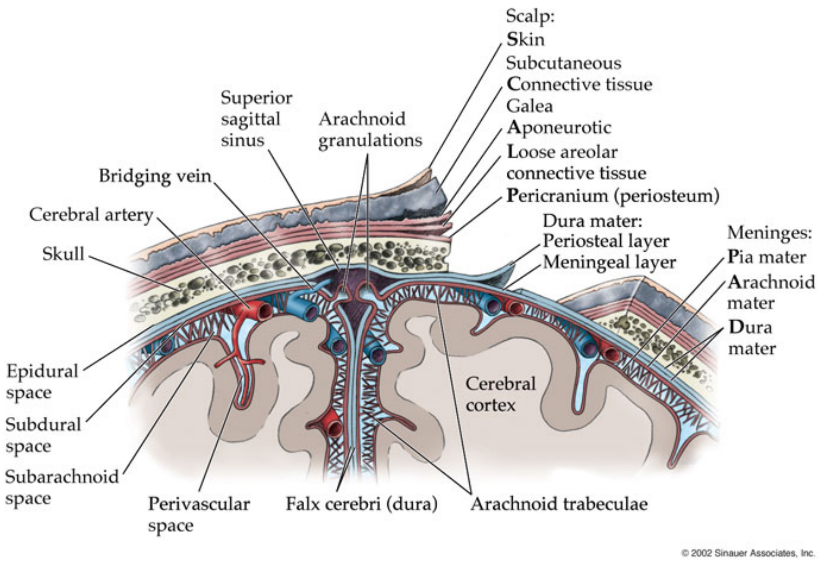


Figure 2.19: *Human head - Multi-layered structure [120].*

- a subcutaneous connective tissue layer,
- a muscle and facial layer aponeurosis,
- a loose connective tissue,
- a fibrous membrane pericranium.

The loose connective tissue layer forms a shear plane between the skin and the cranium [138].

Underneath the scalp, is the bony skull that can be viewed as a three-layered sandwich structure with an inner and outer shell of compact bone and a diploë of spongy bone sandwiched between them as a core [121]. Cranial bone is an anisotropic, viscoelastic and porous structure (properties in [145]). Nevertheless, different bone locations on the cranium show a very different bone layer thickness and therefore different mechanical properties [301]. The thickness of the skull varies between 4 and 7 mm [121],being thinner at the sides and the lower rear of the head [122]. Shuaeib *et al.* [67] highlighted that more consideration should be given to the side of the head due to the weakness from the head tolerance and helmet performance. The skull can be divided into two different sections, the cranium and the face. The cranium consists of eight bones, and the face consists of 14 bones [44]. These bones are connected at lines called sutures that joint all together into one structural unit, as shown in figure 2.20. The only facial bone connected to the skull through free movable joints is the mandible. The cranium can also be divided in four regions, the frontal, left and right parietal, and the occipital.

The base of the braincase is an irregular plate of bone containing depressions and ridges plus small holes (foramen) for arteries, veins, and nerves, as well as the large hole (foramen magnum) that is the transition area between the spinal cord and the brainstem [122], as represented in figure 2.21. In conclusion, the skull protects the brain, working as a stiff braincase.

In figure 2.19, it is possible to observe that below the skull there are three membranes called the meninges which protect and support the spinal cord and the brain, and also

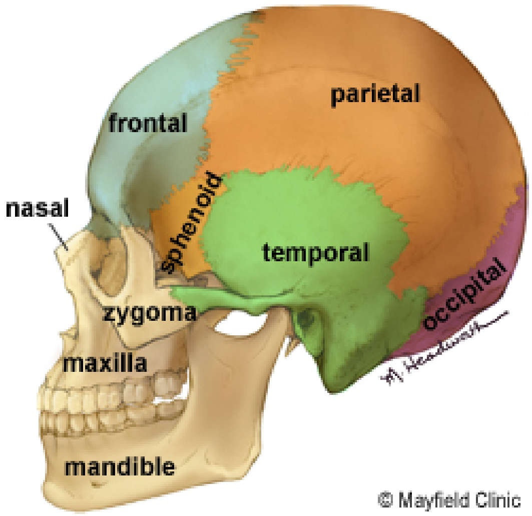


Figure 2.20: *Lateral skull view [123].*

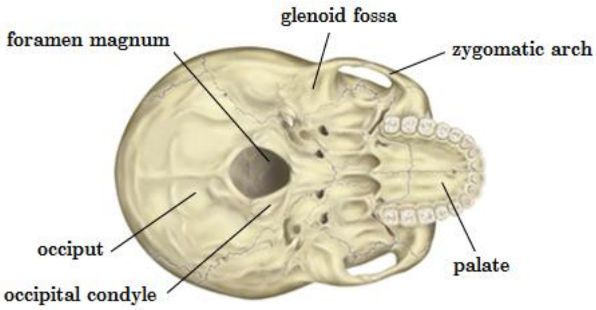


Figure 2.21: *Bottom skull view (Adapted from [124]).*

separate them from surrounding bones and consist primarily of connective tissue, and they also form part of the walls of blood vessels and the sheaths of nerves as they enter the brain and as they emerge from the skull [121].

The meninges consist of three layers:

- the dura mater,
- the arachnoid,
- the pia mater.

This set plays an important role, by protecting the brain from the irregularities of the skull, as previous referred. Brain tissue, having the consistency of a heavy pudding, is the most delicate of all body tissues. For protection, this vital organ is located in a sealed bony chamber, the skull. To protect it further from the rough bone and from blows and shocks to the head, the brain is enveloped by the meninges.

The space between the skull and the outer most layer, the dura mater is called epidural space and it is the region where are the major arteries, taking blood to meninges. The outermost layer, dura mater is a tough fibrous membrane adherent or close to the inner surface of the bone that surrounds the spinal cord and the inner surface of the skull whereas the arachnoidea mater resembles a spider-web, as the name indicates. Beneath the dura mater is the middle covering, the thin and fibrous arachnoid. Between the dura mater and the underlying arachnoid is a narrow subdural space filled with a small amount of fluid that acts as a lubricant, preventing adhesion between the two membranes [121]. The third and innermost layer is the very thin, delicate, and capillary-rich pia-mater, which is attached to the brain and dips down into the sulci and fissures covering the brain surface and acquiring its shape.

The subarachnoid space is a large gap that separates the arachnoid from the pia mater, which is filled with CSF, the lymphlike fluid that constantly circulates and surrounds the whole brain, acting as a shock absorber and thereby, protecting the brain. It also helps supporting the brain's weight. Also, as a further mean of protection, there are fibrous filaments known as arachnoid trabeculations, which extend from the arachnoid to the pia and help hold the brain to prevent it from excessive movement in cases of sudden acceleration or deceleration, acting as a natural brain protection. However, this is not enough to prevent injuries in neurotrauma cases, especially in a motorcycle accident.

In figure 2.19, it is also possible to see the folds of the dura mater that form the falx cerebri, which projects into the longitudinal fissure between the right and left cerebral hemispheres. A sagittal dural partition membrane, the falx cerebri, partly separates the left and right hemispheres of the brain. Another dural fold forms is the tentorium cerebelli, a membrane that separates the cerebrum from the cerebellum and brain stem.The falx and tentorium cerebelli, the two most important meninges in the head, inhibit the movements of the brain inside the head and thus influence the dynamics of the human head as a whole. The lower separating membrane, the tentorium cerebelli, resides on the inferior wall of the skull, and separates the cerebrum from the cerebellum and brain stem.

The meninges are also crossed by blood vessels that supply the brain and the scalp. These blood vessels are the veins that bridge the subdural space and so they are called the bridging veins. This veins are associated to frequent and severe injuries due to the fact that acute subdural hematoma together with diffuse axonal injury account for more head injury deaths than all other lesions combined [125].

The brain is a vital organ made of a fragile soft visco-elastic material and is main part of the central nervous. The brain is the most important component of the head, with respect to head injury. To protect it, it is covered by a strong and stiff skull as already referred. It can be divided into:

- the cerebrum,
- the brainstem
- the cerebellum.

The cerebrum is the largest and most complex part of the brain, as shown in figure 2.22. It is composed of the right and left hemispheres, connected by the corpus callosum. These hemispheres can be divided again into four lobes:

- the frontal,
- the parietal,
- the temporal,
- the occipital.

The outermost layer of the cerebrum is called the cortex and consists of grey matter. Beneath the cortex is a thick layer of white matter. The diencephalon connects the cerebrum with the brainstem.

The brainstem includes the midbrain, the pons, and the medulla oblongata.

The cerebellum is located in the posterior part of the head and includes two hemispheres.

The brain, covered by the membranes and CSF, is connected to the spinal cord through the foramen magnum. In conclusion, all this set of bones and soft tissues that involve the whole brain, actuate as a natural and complex mechanism that protects the brain.

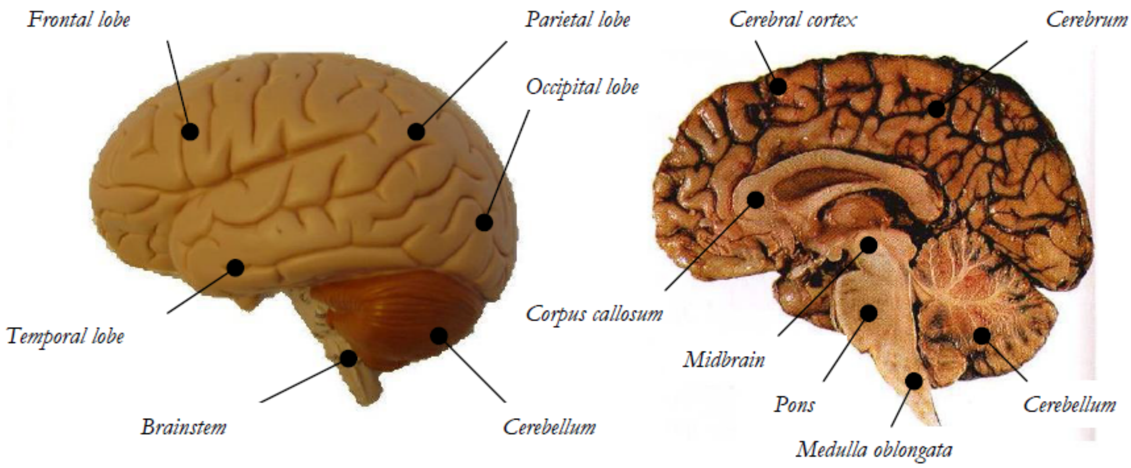


Figure 2.22: *Brain [44].*

2.2.2 Head injuries

In many accidents, the human head is exposed to loads exceeding the loading capacities of its natural protection features (bone, tissue, etc), which results in severe head injuries that could cause disability or even be fatal. Crash accidents are on of the main causes of head injuries [126].

The mechanisms of head injuries are still not fully understood, mainly due to brain injuries. The brain is a complex structure made of complex materials. Moreover, each individual has its own characteristics, which means that individual mechanisms of trauma produce very specific types of head injury [127].

Nevertheless, the most relevant injuries to the head are those to the skull and the brain, including the meninges [119]. Head injuries can be divided into cranial injuries (skull fractures) and intracranial injuries (focal injuries and the diffuse brain injuries). The intracranial injuries are mostly brain injuries.

Other studies [109, 64, 140, 141] include scalp damage besides skull fractures, focal injuries and diffuse injuries in their head injuries classification. However, only the last ones will be addressed in this work because are more important than scalp damage [67].

Comparing skull fracture and brain damage, brain damage is much more serious than skull injury, particularly in helmeted-head cases, due to the protective effect of the helmet from head direct impact. This is proved by the statistical studies performed by Otte *et al.* [65] and Hurt *et al.* [109] where skull fractures account for circa 13,1% and 16% respectively and 38,2% and 58,4% for brain injuries. Moreover, Kraus *et al.* [142] reported that the largest source of traumatic brain injury (TBI) in the world is traffic accidents and Kleiven *et al.* [143] highlighted the significant number of road accidents that influence the central nervous system in a devastating way by transferring high kinetic energy to the nervous tissue.

Independently of the severity of the injuries, there are many different types of injuries that can occur in a crash.

Skull fractures

A skull fracture is a break in one or more bones of the skull that usually occurs as a result of blunt force trauma, which impact force is excessive enough to fracture bone at the impact site or near that because there are bones more fragile than others which absorbs less energy until fracture.

Skull fractures can be either open or closed. A closed fracture is a bone fracture without substantial injury to the surrounding skin. An open fracture on the other hand, is more serious than a closed one, because of the accompanying risk of infections caused by damage to the surrounding tissues and exposure to pathogens.

Skull fractures can occur with and without brain damage, but is in itself not an important cause of neurological injury [135, 136]. However, when bone fragments penetrate blood vessels or brain tissue, the complications may be mild, moderate or severe. Also, even when skull fracture does not occur, bending of the skull may be sufficient to damage underlying blood vessels and brain tissue [122].

Skull fractures can also be divided accordingly to the fracture location into basilar skull fractures (fractures to the base of the skull) and vault fractures (fractures to the non-base part of the skull). Basilar fractures are considered clinically significant, because the dura may be torn adjacent to the fracture site and increasing highly the probability of contamination of the central nervous system (CNS) [102].

Skull fractures can also be classified accordingly to the type of fracture into linear and depressed fractures. The linear fractures are the most common, and usually the less severe type of skull fracture. These fractures are breaks in the bone that transverse the entire thickness of the skull, but no displacement is involved and usually results from low-energy transfer due to blunt trauma over a wide surface area of the skull and usually, it does not have much significance on the course of brain injury, although dangerous complications may occur [137, 130]. The depressed fractures results usually in portions of bone displaced inward which may damage the underlying tissues. Thus, despite skull fractures do not necessarily cause neurological disability, bone fragments may penetrate brain tissue or blood vessels when a depressed fracture occurs, probably resulting in neural injury and intracranial haematoma, especially when the depression is deeper than the thickness of the skull [136]. Depressed skull fractures are frequent mechanisms of head injury and are often associated with traumatic brain injury [145]. An example of both types is shown in figure 2.23.

Furthermore, some minor skull fracture does not cause brain injury, and it could be argued that this is one of the natural mechanisms to absorb energy [144].

The soft tissue injuries to the scalp and face commonly occur with skull fractures, which include contusion and laceration. However, these injuries are considered injuries with minor severity, compared with other injuries.

During a real accident, the skull fracture may be caused by rigid object penetrating the skull such as road posts, tree branch, motorcycle parts, etc. If a helmet is not worn, an impact with a rigid convex object, would cause localised high pressures on the skull, hence possibly a depressed skull fracture, at a force less than that needed to cause brain injury by excessive acceleration [57].

Depending on the extent of helmet coverage, the outer shell of the helmet may prevent

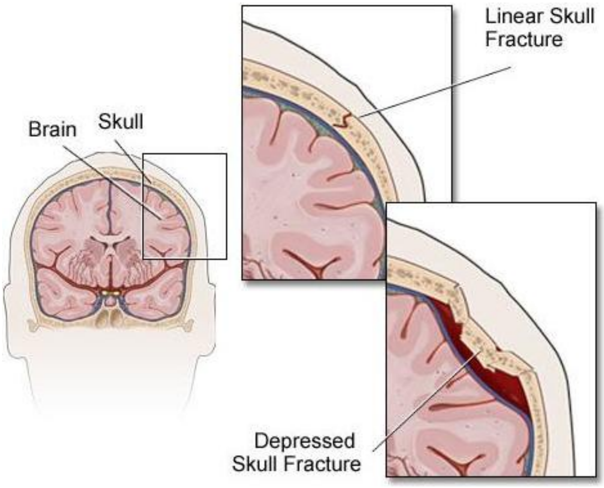


Figure 2.23: *Skull fractures: Linear skull fracture and depressed skull fracture [146].*

such penetration by spreading out the force applied to the head. The coverage area provided by the helmet's shell is an important factor in the protection capacity of the helmet, where less coverage area means less protection. Shuaeib *et al.* [67] highlighted the temporal head region as the weaker region (also the weaker region of a helmet), being the region of higher probability of fracturing. The temporal region is on of the most impacted regions in a motorcycle accident as shown by Hopes and Chinn [59] and Otte *et al.* [65] 39,5% and 18,3%.

Today's helmets are an effective mean of protection against skull fractures, as described by Chinn *et al.* [147], where none of the cases of motorcyclist accidents have sustained skull fractures. The helmet's shell distributes the applied force and thereby reduces the concentration of loading, and thus the propensity for skull fracture. However, an impact to the head can deform the skull, even if it does not fracture, and thus the underlying brain tissue can be injured as it distorts under the influence of the deforming skull *et al.* [26].

Although skull fracture can be a seriously injury, skull fracture is not considered as a major criterion for helmeted-head impact studies due to the fact that brain damage by acceleration will take the precedence, before the impact load will cause a depressed skull fracture for the helmeted-head [67].

Brain injuries are much more serious. They frequently result in death, permanent disability or personality change and, unlike bone, neurological tissue has very limited ability to recover after an injury. Therefore, the primary purpose of a helmet is to prevent traumatic brain injury while skull and face injuries are a significant secondary concern.

Focal brain injuries

The focal brain injury is the type of injury that occurs in localized regions of the brain subjected to tensile or compressive stresses. Approximately two thirds of the deaths associated with head injuries are attributable to focal injuries [44]. Focal brain injuries are reported to be highly correlated with fatality [130].

The focal brain injury is a lesion that corresponds to local damage. The focal injuries consist of epidural hematomas (EDH), subdural hematomas (SDH), intracerebral hematomas (ICH) and contusions (coup and contrecoup).

Most focal injuries are due to direct contact with bone fragments from skull fractures, or to relative motion between different parts of the skull and the brain. Such relative motion may be due to linear or rotational acceleration of the skull [122]. An example of this type of injuries dependent on the brain relative movements to the cranial cavity is the subdural hematomas that are focal injuries [144, 229].

Epidural haematoma

An epidural haematoma (EDH) is the result of trauma to the skull (skull deformation) or to the underlying meninges and is not due to injury to the actual brain. Generally, bleeding above dura mater is a result of this type of injury. Usually skull fracture is associated, but an epidural haematoma may also occur in the absence of skull fracture.

Epidural haematoma is a relatively infrequently occurring sequel to head trauma, circa 0.2-6% [131].

EDH are not as lethal as subdural haematomas [44]. If the haematoma is found below the dura mater, it is called a subdural haematoma. This and other focal injuries are shown is figure 2.24.

Subdural haematoma

In motor vehicle accidents, one of the most frequent injuries to the brain that results in fatality or the need for long-term rehabilitation is subdural haematoma (SDH) [128]. A SDH is caused by a rupture to an artery or bridging veins. Examples of such injuries are 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 most common mechanism of subdural hematoma is tearing of veins that bridge the subdural space as they go from the brain surface to the various dural sinuses [121]. This injury arises from tangential force against the skull, and is directly related to rotational effects on the brain [128]. The most common type of SDH is the disruption of brain's surface vessels. This is entirely the result of inertial and not contact forces. A SDH is caused by short duration and high strain rate loading [127].

Gennarelli and Thibault *et al.* [139] reported an incidence of acute subdural haematoma of 30% with an associated mortality rate of 60%. More recently, it was reported the mortality rate of this type of injury is greater than 30% [134].

In a study performed by Richter *et al.* [56], of a total of 409 head lesion cases in a group of study of 81 motorcyclists, the author observed that more than half of the total of the 409 lesions were brain lesions. From those brain injuries, a major part was subdural hematoma (SDH).

For nearly one third of the acute SDH cases are directly related to a bridging vein rupture [353]. Following a head impact, the brain lags behind the skull which leads to a longitudinal strain in the veins and this can further lead to vein rupture, as shown in figure 2.27.

Contusion

Contusion is the most frequently found lesion following head impact [121, 44, 119]. This consists of heterogeneous areas of necrosis, pulping, infarction, haemorrhage or oedema. There are two types of contusions: the coup and the contrecoup. Coup contusion occurs at the site of impact while contrecoup contusion occurs at remote sites the impact. Contrecoup contusions are considered more significant than coup contusions [134]. This type of injury is shown in figure 2.26.

Coup contusions are contact injuries [135, 326] and are produced by compressive forces operating beneath an area of skull inbending or tensile forces generated by the negative pressure produced beneath an area of skull inbending.

Intracerebral hematomas (ICH)

Usually, this type of focal injury is distinguished from contusions by a more pronounced localisation of the haematoma [102]. Intracerebral haematomas are well defined homogeneous amounts of blood within the brain. They are most commonly caused by sudden acceleration/deceleration of the head. Other causes are penetrating wounds and blows to the head. This type of injuries are generally regarded to be of minor importance [121].

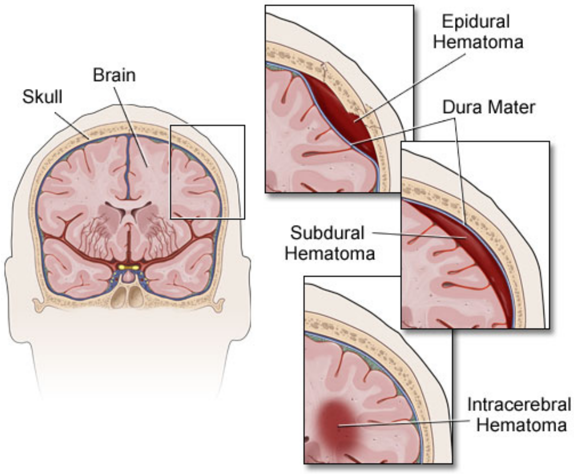


Figure 2.24: *Focal brain injuries [172].*

Diffuse injuries

Diffuse brain injuries are fundamentally different from focal injuries. The diffuse brain injury is associated with global disruption of brain tissue. This class of injury is usually a consequence of distributed loading conditions that generally induce relatively low energy damage affecting substantial volumes [229]. A situation such that encountered on helmeted-head impact where the both the shell and liner work to distribute the load on as large area of the head as possible.

DBI generally occurs via impact often without skull fracture and is referred to as closed head injury [139, 308, 309]. It is primarily involved with dynamic non-contact loading, although it is believed to occur in closed head impacts as well [492, 351].

Diffuse brain injuries account for approximately 40% of patients with severe brain injuries, and one third of deaths due to head injury [129].

Concussion

Concussion is the most common head injury diagnosis resulting from motorcycle and moped accidents [133]. It is not a severe injury, where the time of recovery is short [126].

Mild concussion is a less severe type of concussion, where disruption of the brain tissue doesn't happen contrary to the others types of diffuse injuries.

In a study performed by Richter *et al.* [56], concussion was too, one of most observed injuries.

Diffuse axonal injury

Diffuse axonal injury (DAI) is caused by the disruption or elongation of neuronal axons in the brain tissue, more specifically in the cerebral hemispheres, midbrain and brainstem [119]. This injury arises from the same mechanisms as SDH, which are tangential forces applied to the skull. DAI is produced by a longer duration and more gradual onset of acceleration than SDH [44]. A frequently occurring result of blunt head impact is an injury to the axonal structure referred as DAI. It can be said that DAI is a distribution of focal lesions in the axonal components of the neural structure.

It is considered the most severe injury between the diffuse ones [125]. It is reported that this particular type of injury constitutes a large amount of the total brain injuries. At the end of 1 month, 55% of the patients are likely to have died, 3% may have vegetative survival and 9% may have severe deficit [125, 132, 130]. More recently, Bandak [229] observed that this type of injury constitutes about more than 50% of all head injuries.

DAI is a frequently occurring brain injury resulting from motor vehicle accidents, and often results in fatality or the need for long-term rehabilitation [349, 128]. This type of damage is caused from uniform pressure loading on the head that is most likely to occur in helmeted-head impacts, as the helmet will tend to protect against local injury effects but not the uniform pressure loading.

It is primarily involved with dynamic non-contact loading, although it is believed to occur in closed head impacts as well [492, 351].

Brain Swelling

Brain swelling due to an increase in intravascular blood within the brain may worsen the effects of primary injury by increasing intracranial pressure. This increased pressure may force the brain and brainstem downwards through the foramen magnum causing further damage to the tissues, what may greatly increase the risk of fatality for patients with others injuries, such as DAI [122].

2.2.3 Head Injury Mechanisms

Head injuries can be explained by the associated injury mechanisms. An injury mechanism explains the immediate mechanical and physiological changes that result in functional and anatomical damage [286].

Head injury typically results from either a direct impact to the head or from an indirect impact applied to the head-neck system when the torso is rapidly accelerated or decelerated. In either case, the head sustains a combined linear and angular acceleration which can product injury.

These mechanisms can be divided into static and dynamic loading. Any loading with a duration superior to 200 ms is considered a static loading [119]. Under such load the head deforms until it reaches a maximum deformation. This type of loading often leads to skull fractures. Moreover, this type of loading is rare in motorcycle accidents, where the duration of impacts is typically inferior to 200 ms, which is considered a dynamic loading.

This type of loading can be divided in contact and non-contact loading. Direct contact of the head to an object can cause the skull to deform, possibly resulting in skull fractures, due to skull bending for example. According to Gurdjian [174] and Thomas [175], skull bending is the cause of linear skull fracture. When the skull is deformed beyond its loading capacity, a fracture occurs. Other examples of contact injuries beyond skull fractures are EDH, coup/contrecoup contusions and lacerations.

Furthermore, after deformation of the head, local brain injury like epidural haematoma or contusion, as well as scalp injuries can occur, even without skull fractures. Contact phenomena typically cause focal brain injuries. Additionally, rapid contact loading produces stress waves (pressure waves) that propagate in the skull or the brain, as show in figure 2.25, which 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 (coutrecoup), as demonstrated by Nahum *et al.* [192] and shown in figure 2.26. Such a mechanism is proposed for the generation of intracranial compression which causes focal injuries of the brain tissue and bruising [119]. In addition, the pressure gradient may originate shear strains 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, causing surface contusions on the brain and tearing of the bridging veins leading to SDH [176, 131, 177, 135], as shown in figure 2.27. The same applies to the penetration wounds.

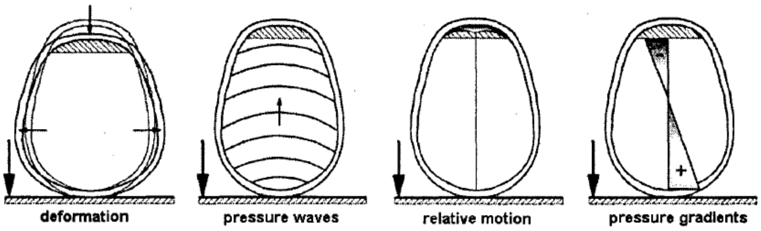


Figure 2.25: *Different injury mechanisms for contact impact [119].*

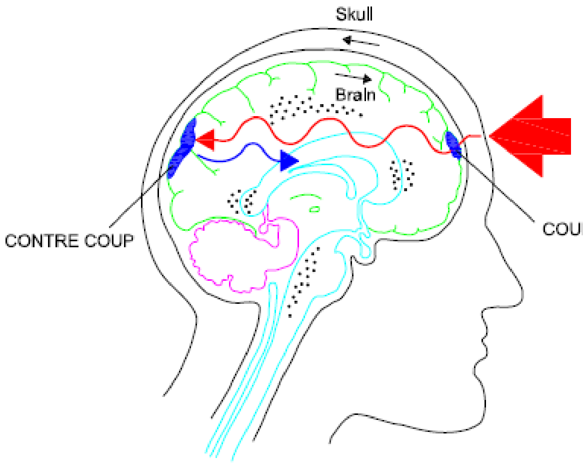


Figure 2.26: *Coup-contrecoup injury (adapted from Kleiven [121]).*

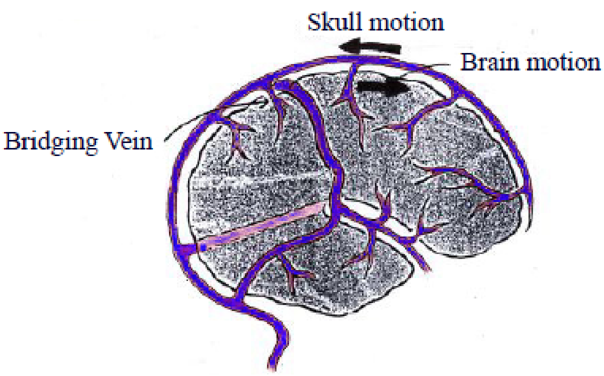


Figure 2.27: *SDH (adapted from Kleiven [121]).*

In non-contact situations, the head is loaded exclusively due to inertial effects, as accelerations or decelerations of the head. Inertial injuries do not necessarily involve a direct impact to the head, but they are caused by acceleration of the head. Generally, an impact to the head results in acceleration of the head, which leads to inertial loading of the intracranial structures, such as brain. Examples of inertial injuries are concussion, SDH, contrecoup contusions, DAI and ICH.

Acceleration can be divided in translational or rotational. Generally, translational acceleration generally results in focal brain injury while rotational acceleration also causes diffuse brain injury. In other words, pure translational acceleration creates intracranial pressure gradients, while pure rotational acceleration produces rotational of the skull relative to the brain and is particularly likely to tear bridging veins [229], as shown in figure 2.27, and can even produce shearing of brain tissue through the mechanism shown in figure 2.28. If the blow is directed eccentrically, the result is a combined translational and rotational acceleration type of injury.

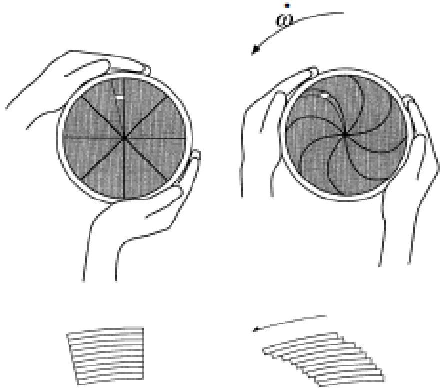


Figure 2.28: *Angular acceleration of the bowl produces shear strains in the contents, as illustrated by the layers sliding across each other [447].*

Purely translational or rotational loading to the human head is uncommon in reality, as these types of movements are not physiologically possible, mainly because of the effects of the head-to-neck connections in reality [44]. Normally, both types are present in any head impact.

Currently, rotational acceleration is seen by many researchers as the principal cause of brain injury, causing SDH and tearing brain tissue and bridging veins, on other words, every known type of head injury can be produced by rotational force to the head, except skull fractures and EDH [125, 157, 173, 164, 197, 198, 19, 285]. EDH is not related to head motion or acceleration, but to skull deformation. Figure 2.29 shows the occurrence of the most severe head injury by a qualitatively relationship between angular acceleration amplitude and time duration of this acceleration. The trend is that at short pulse durations, cerebral concussion can be produced. But as acceleration magnitude increases, strain rate sensitive bridging veins may be torn and SDH occurs. At longer pulse duration, cerebral concussion can be achieved at lower acceleration levels, but it takes considerably more acceleration to cause subdural bridging vein rupture. Shearing brain tissue are thought to be caused by high angular acceleration at longer pulse durations [102].

The main requisite for countercoup damage is rotational movement of the head in the coronal, sagittal or horizontal plane, or even a combination of these, where the movement is translated to the brain which collides with its dural compartment, which surface may

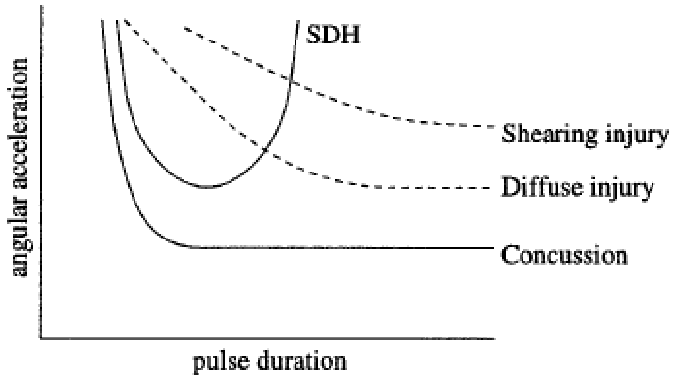


Figure 2.29: *Relationship between angular acceleration and head injury [137].*

be rough, especially at the orbital and temporal areas [67]. At these sites shear strains develop to cause contusion to the brain and tearing off blood vessels. Brain damage may also occur as a result of translational because of the short duration reduction in intracranial pressure [67]. It have been reported that skull distortion and rotation of the head are more important in the production of coup and countercoup injuries than either rotation alone or translation [229]. Concussion is produced much more readily by rotational acceleration than by translational acceleration. It is reported that rotational acceleration of the head can cause DAI to the white matter of the brain in animal models [198]. Other researchers have been able to cause DAI in the brain of animals by application of direct impact to the brain without an associated head angular acceleration [199, 200, 201, 202]. In fact this area is still an active area of research, and more in-depth investigations are still required. However, it could be concluded that DAI brain damage is a critical parameter in helmeted-head biomechanics, and relationship relating this type of damage with impact forces or accelerations would provide a strong basis for helmet design improvements.

To improve the understanding of head injury mechanisms, several researchers, performed experimental works at the beginning. The studies point to brain deformation or strain as a principal cause of injury. Unfortunately, the measurement of strain is almost impossible during an impact, particularly in vivo [19]. Nevertheless, clinical studies, physical modelling and finite element analysis have hypothesized that resulting strains are the primary cause of neurological deficiencies. Later, with the increasing of computational power, finite element method (FEM) has started being used. A review of some of these studies is here presented.

Since from the beginning, that the doubt of what type of acceleration causes severe injuries or is responsible for brain injuries is established, by the creation of two theories. In a pioneering work, Holbourn [37] was the first to cite angular acceleration, with or without direct impact, as an important mechanism in head injury, mainly in the appearance of cerebral concussion. It was hypothesized that shear strain and tensile strain induced by rotation acceleration could cause cerebral concussion as well as contrecoup contusion, tearing cerebral blood vessels sucha as bridging veins and brain tissue causing haematoma and DAI. It was also claimed that translation is not injurious, while rotation could explain the majority of traumatic brain injuries due to the nearly incompressible properties of brain tissue. Later, Ommaya [315] confirmed this hypothesis in a study subjecting primates to linear or angular head accelerations, where specimens were concussed more frequently under head angular acceleration.

On the other hand, Gurdjian, Lissner and other co-workers [178, 179, 180, 181] attributed intracranial damage to deformation of the skull and pressure gradients caused by skull deformation and acceleration of the head due to direct impacts to the head. Linear acceleration was considered to be the most important mechanism, while rotational acceleration, negative pressure and cavitation were of minimal or no significance.

Strich [290] found diffuse degeneration of white matter in the cerebral hemispheres, as well as in the brain stem and corpus callosum areas in patients who have endured severe head trauma. This indicates that high shear strain in the white matter adjacent to the cortex is likely to occur in a real life accident.

A few years later of Ommaya supported the Houlboun hypothesis, Ommaya [182] indicated that rotation alone could not produce the levels of injury caused by direct impact. Ommaya *et al.* [246] proposed a method in order to extend the results of experiments on concussion producing head rotations on lower primate subjects to predict the rotations required to produce concussions in man and the results of concussion in the monkeys indicates that an acceleration of 40000 rad/s^2 will have a 99% probability of producing concussion which corresponds to an angular acceleration of 7500 rad/s^2 for humans. One year later, Ommaya *et al.* [330] studied the effect of whiplash injury on monkeys and showed that if the head was subjected to a rotational acceleration above a threshold value, subdural and subarachnoid injuries were probably occur. Later, Ommaya and Hirsch [150] suggested that rotation could account for approximately 50% of the potential for brain injury, while the remainder was attributed to direct impact.

Rotational effects were tested by Unterharnscheidt and Higgins [149] that applied controlled angular accelerations to monkeys, where these suffered SDH, torn bridging veins, and brain damage. Later, Unterharnscheidt [183] studied the effects of translational and rotational acceleration of the brain in closed head injury. Pure translational acceleration creates pressure gradients while rotational acceleration produces rotation of the skull relative to the brain (shear stress).

Gennarelli *et al.* [151, 152] demonstrated that translation of the head in the horizontal plane produced essentially only focal effects, resulting in contusions and ICH, while diffuse injuries were seen only when a rotational component was present. The principal mechanism of purely linear acceleration appears to be pressure gradient, while that for purely rotational acceleration appears to be shear stress, which results from differential motion between the skull and brain. Gennarelli *et al.* [152] subjected monkeys to controlled sagittal plane head motions and it was found that pure translation and rotation can cause concussion and other brain lesions, where the frequency and severity was greater after rotational motion. This is consistent with the hypothesis presented by Holbourn [37].

Lowenheilm [184] proposed angular acceleration as the cause of contusion and concluded that the site of maximum shear occurred at a constant distance from the surface of the brain. It was also stated that the deep brain could be injured while the surface was not injured and that the zone of maximum shear became deeper as the angular acceleration pulse duration increased.

In most of these studies it was assumed that the brain tissue is incompressible [283, 284] and is therefore most likely to fail in shear. McElhaney *et al.* [284] concluded that the bulk modulus of brain tissue is roughly 10^5 times larger than the shear modulus. Thus, the brain tissue can be assumed to deform in shear. Therefore, distortional strain is used as an indicator of the risk of traumatic brain injury.

Abel *et al.* [153] studied the incidence and severity of cerebral concussion in monkeys following sagittal plane angular acceleration, where the results showed SDH as a threshold phenomenon.

Hodgson and Thomas [154] used monkey brain models and subjected those to translation, pure rotation and a combined motion, separately. Pure rotation produced the highest, most diffuse and long lasting shear strain and brain displacement, while translation produced very low shear strain.

Dirnhofer *et al.* [291] concluded that axial accelerations are usually caused by accidents due to fall and clinical observations shows that this may lead to DAI in the brainstem as well as tearing injuries to the posterior fossa tentorium. The findings of high strain in the central parts of the brain and lower strains in the brainstem for the axial rotational impulse supports the findings of Gennarelli *et al.* [157] that horizontal impulses produce almost exclusively DAI in the central parts of the brain.

Ono *et al.* [156] found that the occurrence of concussion and cerebral contusion in monkeys correlated highly with the and rotational translational acceleration of the head from a direct impact. The authors suggested that a rotational component is necessary for the occurrence of brain contusions but concerning the occurrence of concussion, the authors showed no correlation with the rotational acceleration of the head.

At the same time, Ward and Chan [158] developed a 3D FE model of a primate head. The model was subjected to sagittal plane, non-centroidal, and biphasic angular acceleration pulse reported by Abel *et al.* [153]. They found that the maximal shear stresses were around 50% lower when simulating without applying the rotational component of the acceleration, showing the importance of rotational acceleration component.

The importance of impact direction in causing SDH was realized by Fruin *et al.* [187] in their clinical study of interhemispheric SDH. They found that six out of eight cases with known trauma sites were due to occipital impacts. The larger amount of motion between the brain and the skull for the occipital impact compared to results of the frontal impact might also be explained by the anatomical difference between the frontal and occipital region of the skull.

In a series of studies with the aim of investigate the influence of rotational acceleration in causing brain injury, Gennarelli, Thibault, and co-workers, in a series of studies Gennarelli *et al.* [151, 185, 132], Gennarelli and Thibault [139] and Thibault and Gennarelli [186], by using live monkeys and physical models concluded that angular acceleration contributes more than linear acceleration to the generation of concussive injuries, DAI and SDH. The authors hypothesized that these injuries were induced by the shear strain generated by angular acceleration and claimed that virtually every known type of head injury can be produced by angular acceleration.

Gennarelli *et al.* [132], in experiments with monkeys, shown that the incidence and degree of DAI correlated, although indirectly, with the direction of the head acceleration. Coronal plane angular acceleration was the direction that caused the longest lasting coma, while sagittal plane angular accelerations and oblique accelerations produced coma for a shorter period.

Clinical studies report that diffuse brain injuries occur most commonly due to impact with deformable or padded surfaces [128]. This loading typically results in high rates of head angular acceleration. The author concluded that SDH was mainly produced by short duration and high amplitude rotational accelerations, while DAI was mainly produced by longer duration and low amplitude rotational accelerations in coronal plane.

Gennarelli [135] concluded that the main cause of contusions is the contact interfaces between the brain and the membranes that leads to deformations because the surrounding neurological tissues are much stiffer than brain and acceleration of the head differentially loads these different parts of the brain, where there is relative movement that leads to deformation between the various parts.

Gennarelli *et al.* [132, 157] also investigated the influence of certain impact directions for DAI, using primates, where loads in the lateral direction were considered more likely to cause DAI compared to impulses in the sagittal plane.

Hodgson *et al.* [155] investigated the influence of certain impact directions for cerebral concussion, using primates. These animal models of concussion and diffuse axonal injury studies, such as [155, 157] showed that animals sustained a more severe form of brain injury from lateral impacts than from other impact directions. Low tolerance of the head to lateral impact in comparison with frontal impact was also observed in cadaver tests conducted by [288]. Also, it was concluded in these and in other studies that diffuse brain injury (DBI) results from head angular acceleration and focal-type injuries result from linear acceleration [152, 157, 312, 155, 150, 156, 314, 183].

Nevertheless, later studies on volunteers Pincemaille *et al.* [247] suggests that the human tolerance is largely underestimated using primate experiments and simplistic scaling rules.

Adams [190] concluded that rupture of bridging veins due to these high strains are considered to be the main cause of SDH, as concluded too in other studies [37, 191, 153, 135]. Indeed, brain injuries can be associated to the large strains during which the brain's tolerance threshold is exceeded and the cerebral tissue is torn [282].

Margulies *et al.* [306] studied the influence of the falx cerebri on intracranial motion and deformation by using this brain surrogate in human and primate skull models. Later, Ivarsson *et al.* [307] used the same brain surrogate to study the influence of lateral ventricles and irregular skull base on brain kinematics under sagittal plane rotation acceleration.

Other studies developed concussive injury tolerance levels highlighting a relationship between angular acceleration magnitude and angular velocity, with lower tolerance at higher angular velocities [335, 243, 282]. However, due to its dependence on both magnitude and duration of angular acceleration, angular velocity may not be the best indicator of injury tolerance [334].

DAI is commonly a result of inertial induced loads, because intracranial motions arise when the skull is accelerated and the brain mass, due to its inertia, lags behind or continues its motion relative the skull. Hence, the risk of DAI is highly dependent on brain mass [243].

Contrary to the studies performed by Gennarelli and co-workers, McLean [193] argued that there were no cases of brain injury without head impact in his investigation of a series of more than 400 fatally injured road users and that is not possible the human neck transmit enough energy to the head to cause brain injury without a direct impact to the head.

Zhou *et al.* [169] found higher strains in the bridging veins during the acceleration than in the deceleration phase while applying the acceleration pulse from Abel *et al.* [153]. Since the acceleration pulse is directed in the posterior-anterior direction, it was suggested that SDH is more easily produced in an occipital impact than a corresponding frontal one. Later, the same researchers [287] found that anterior-posterior (A-P) motion causes higher strain in the bridging veins than a corresponding lateral motion. However, in all of these first numerical studies, a tied interface was imposed between the skull and the brain leaving out any possibility of evaluating relative motion induced injuries such as SDH.

DiMasi [167] simulated a crash test using only the rotational motion, and the translational kinematics only. It was found a higher brain cumulative volume fraction that has experienced a specific level of maximal principal strain for pure rotation, than pure translation, while a combination of the full kinematics gave the highest values. In the same study, DiMasi showed that pure translational acceleration of the head would induce minimal strain, while a pure rotational acceleration would produce considerably greater strain. Also, a combination of translational and rotational acceleration would induce more strain than rotational acceleration alone.

In a similar study, Ueno and Melvin [168] used the kinematics from a Hybrid III dummy, but applying it to a 2D head model. They found that the use of either translation or rotation alone may underestimate the severity of an injury. The results also indicated that translational acceleration is related to pressure while rotational acceleration has a dominant effect on shear deformation. However, the nodes at the skull-brain boundary were rigidly connected, as in the previous numerical studies carried out by other researchers. Kinematic head injury predictors are usually based on the assumption that either linear acceleration or rotational acceleration is the main cause of head injury while it has been shown that their combination increases the injury risk [167].

Rotation of the skull relative to the brain, presses the highly irregular skull base towards the brain. This leads to a combined compression and shearing of the meningeal and cortical tissues in this area, which increases the effects of the sliding of the brain over the skull base [194]. The effects of this relative rotation are most severe when the head is subjected to a backward non-centroidal rotational acceleration, or in case of a forward non-centroidal rotational deceleration.

Acute SDH is caused by three sources: hemorrhagic contusions that break through the arachnoids, the rupture of bridging veins, and rarely by laceration of cortical arteries or veins. In autopsy series, two thirds of the SDH were associated with contusions [327].

Lee *et al.* [161] used a 2D sagittal model and [162] used a 3D model, presented by [403], to study the mechanisms of SDH. They found that the contribution of angular acceleration to tearing of bridging veins was greater than the translational acceleration.

Bain and Meaney [195] have shown that DAI is a function of distortional strain and not dilatation. Until today, several predictors of CNS (central nervous system) injuries has been used in different studies. The maximal principal strain was chosen as a predictor of CNS injuries since it has shown to correlate with DAI [195, 294, 295, 282, 296, 358], as well as for mechanical injury to the blood-brain barrier [297]. Other local tissue injury measures have also been proposed and evaluated, such as von Mises stresses [298, 223, 297], product of strain and strain rate [299, 300, 19], strain energy [297] and the accumulative volume of brain tissue enduring a specific level of strain, the Cumulative Strain Damage Measure (CSDM) [164, 167]. For instance, Miller *et al.* [223] showed in a 2D FE study that the maximal von Mises stress predicts comparable patterns of axonal and macroscopic hemorrhagic cortical contusions in the miniature pig.

Zhang *et al.* [170], in a three-dimensional numerical study, compared brain responses between frontal and lateral impacts, and found higher shear stress in the core of the brain during a lateral impact. This confirmed earlier results by Gennarelli [132, 157] that loads in the lateral direction are more likely to cause DAI than impulses in the sagittal plane. However, a tied interface was again imposed between the skull and the brain leaving out any possibility of evaluating relative motion between the skull and the brain. In other study using a FE human head model too, it was shown that the human head had a decreased tolerance to lateral impact in comparison with an impact from the frontal direction [281]. Later, Zhang *et al.* [263] concluded that both linear and angular accelerations are significant causes of mild traumatic brain injuries. Besides this study, others have hypothesized that the resulting strains in the brain tissue are the primary cause of neurological deficiencies following DBI [360, 243, 306, 361, 362].

Nevertheless, some experimental studies have identified angular acceleration magnitude as the sole determinant of DBI severity, with increasing magnitudes associated with more severe injuries [317, 150].

In some studies is suggested that maximal principal strain higher than 18-21% leads to DAI [195, 358] while the vascular rupture is expected at strain levels above 30-60% [191, 358]. Thibault [359] suggested a maximal principal strain of around 10% to cause reversible injury to the axons which could be used as an approximate threshold for concussion an DAI. Shreiber *et al.* [297] who derived a threshold of 19% in principal strain in the cortex for a 50% risk of cerebral contusions.

Zhang *et al.* [446] using an anatomical detailed finite head model and reconstructing real neurotrauma cases, inducing both translational and rotational acceleration, concluded that strain rate and product between strain and strain rate in the midbrain region appeared to be the best injury predictor for concussion.

Franklyn *et al.* [396] proposed that the localized strain and strain rate variables were the most relevant brain injury indicators for the concussion and axonal injury seen in real-world accident.

Kleiven [275] performed a numerical study using a FE human head model and concluded that the horizontal impulses produced almost exclusively DAI in the central parts of the brain (region that endured higher stresses and strains for a rotational impulse compared to a translational one with the same impact power), supporting the findings of Gennarelli [157] and that the use of either translation or rotation alone may underestimate the severity of an injury, as concluded previously by DiMasi [167] and Ueno and Melvin [168]. Kleiven [275] and Kleiven and von Holst [196] found the largest strains for the centrally or frontally located bridging veins for all impact directions which supports the experimental studies made by Hirakawa *et al.* [188] and Jamieson and Yelland [189] where SDH was found in the occipital region. It was found too that the influence of impact direction had a substantial effect in the prediction of SDH [275].

Kleiven [121] developed and parameterized a detailed FE model of the human head to evaluate the effects of head size, brain size and impact directions. Simulations with various brain sizes indicated that the increased risk of SDH in elderly people may to a part be explained by the reduced brain size resulting in a larger relative motion between the skull and the brain with distension of bridging veins. Later, associated with this study, Kleiven and von Holst [171] found for rotational impulses of short duration, that the change in angular velocity has been shown to correspond best with the intracranial strains, which is in agreement with hypothesis suggested by Holbourn [37]. Kleiven [121] also concluded that larger relative motion between the skull and the brain is more apparent for an occipital impact than for a frontal one in both experiments and FE model.

Aare *et al.* [464] compared the results from different oblique impacts simulated and concluded that the rotational effects have a major influence on the strain levels in the human brain where the maximum strain in the brain was usually found in the brain white matter.

Later, Kleiven [285], using a FE human model,found that low levels of strain can be seen in the vicinity of the ventricles in the FE human model, which supports the hypothesis of Ivarsson *et al.* [289] that a strain relief is present around the ventricles. Kleiven [285] also compared the influences of translational and angular impulses in different directions and concluded that the largest strain in the brain appears for the lateral and axial rotational impulses, while substantially smaller strain is found for the translational impulses. Almost a tenfold increase in the intracranial strains is found when changing from a lateral translational to a lateral rotational motion. Thus, Kleiven concluded that the worst case was the lateral rotation where the highest strain appears in the cortex, corpus callosum and brain stem. In this study is also clear that rotational motion is responsible for high levels of strain close to the vertex of the skull as well as close to the irregularities in skull base, mainly due to the relative motion between the different parts. The findings of larger stresses and strains in the corpus callosum for the lateral angular acceleration impulse as well as the lateral translational impulse support the conclusions drawn by Gennarelli *et al.* [132, 157] that loads in the lateral direction is more likely to cause DAI compared to impulses in the sagittal plane. The largest strains, on the other hand, occurred in the surface of the cortex area. However, large stresses and strains in the surface of the cortex area are related to cortical contusions, and such injuries are usually less critical DAI associated with shear strains and effective stresses in the corpus callosum and brain stem areas [130].

As already referred, it has been suggested that the severity of the injury correlates with the amplitude of the angular acceleration [153, 243, 156] or also with the resulting angular velocity [363]. Duration of the impact has been reported to affect the injury type. Short duration results in focal injury, while long duration impacts result in DBI [243, 156].

Deck *et al.* [332] found that the effect of angular acceleration increase of the order of 50% the intra cerebral shearing stress for all accident cases considered what ever the impact severity was. Considering the strain energy computed within the CSF layer leading to SDH prediction, the rotational component has alternatively increasing or decreasing effect depending on the direction of the impact. It was found also that no specific trend was observed when impact direction was taken into account even if only a limited number of cases were available for occipital and vertex impacts.

Tamura *et al.* [344] examined the relationship of strains measured between the axon and brain tissue. The results revealed that the strain level experienced by each axonal element was only one third of the total strain experienced by the brain tissue. This finding implies that directly incorporating a cellular level axonal threshold into an FE brain model could result in substantial over-prediction of injury occurrence.

Fijalkowski *et al.* [310] indicated that severity of injury is directly attributed to biomechanical factors such as magnitude and duration of angular acceleration, brain mass, and plane of rotation. One year later Fijalkowski *et al.* [334] exposed the subjects to mean angular acceleration durations. All other rotational kinematics were held constant. This study demonstrated angular acceleration duration was an influencing factor in DBI severity. However, this nonexclusive indicator has important limitations. For example, as angular acceleration duration decreased, it was suggested injury type variation from diffuse to focal [139]. DBI determinants previously associated with injury severity include brain mass [355, 246], plane of rotation [157, 156, 313], angular acceleration magnitude [153, 312, 326, 149] and angular velocity [335, 356, 357]. An inverse relationship between brain mass and peak angular acceleration exists, wherein decreasing brain mass requires increasing levels of angular acceleration for similar injury levels [355, 243, 246, 150]. A separate study demonstrated that coronal plane rotations are most injurious due to decreased inertial properties and geometric constraints [157, 155, 243, 306, 282]. Other experimentation has shown a positive correlation between magnitude of angular acceleration and severity of injury [153, 312, 155]. Furthermore, duration of angular acceleration is a determinant of injury type, wherein short duration impacts result in focal injury while long duration impacts result in DBI [243, 156, 313, 314].

Mordaka *et al.* [336], using FEM simulations, showed that head rotational kinematics had great influence on injury severity, particularly on DAI and should therefore not be neglected. Moreover, the severity of injury was found to be related not to angular acceleration but to peak change in angular velocity. The results showed that increased peak change in angular velocity caused higher maximal principal strain in the brain and in consequence higher probability of DAI, and acute SDH. A dramatic, three-fold increase in the strain levels in the brain was found when doubling the impact velocity. This variation in strain in the brain for a difference in impact speed is varying similarly to the peak change in angular velocity. This corresponds to Holbourn's hypothesis [37] that the strain is proportional to the change in angular velocity for rotational impulses of short durations.

In a study performed by Deck *et al.* [332], the effect of angular acceleration was found to increase of the order of 50% the intra cerebral shearing stress for all accident cases considered what ever the impact severity was.

In a numeric study performed by Zhang *et al.* [341], using an anatomically detailed Finite Element Model of human head, model predictions indicated that different regions of the brain were susceptible to higher strain responses after applied rotational acceleration loading. Such regional strain response patterns were further influenced by the direction in which the head rotated. Coronal rotation induced mutifocal high strain in the midbrain, thalamus and caudate regions perhaps indicating an effect of the falx on the strain propagation. In the case of head rotation in the sagittal plane, the critical strain was mainly located in the hippocampus and upper brainstem region. This strain localization was likely dictated by the presence of the tentorium opening and its transverse orientation affecting the tissue deformation in the sagittal plane.

Yogananda *et al.* [428] reported that different injuries can be created by changing the characteristics of the linear and angular acceleration loading curve. More recently, Post *et al.* [427] performed a study in order to analysis the predictors that influence the most the loading curve shape in the brain and indicated that higher maximum principal strains and Von Mises stresses are the predictors that have most influence in the shape of the curve.

Cloots *et al.* [348] show that the axonal strains cannot be trivially correlated to the tissue strain without taking into account the axonal orientations, which indicates that the heterogeneities at the cellular level play an important role in brain injury and reliable predictions of it.

Asiminei *et al.* [353] observed that high sensitivity of the rate of head acceleration-deceleration was the major determinant of bridging vein failure where no strain rate sensitivity could be observed up to 20 s^{-1}.

Krave *et al.* [440] induced rotational acceleration to animals' head and conclude that a sagittal rotational acceleration trauma of the head and neck induced DAI.

Many of these studies have presented thresholds to assess the occurrence of injuries. These predictors are described in the next section.

2.2.4 Head Injury Criteria - Injury tolerance

In the previous section (section 2.2.3), it was shown that head injuries can be divided into contact injuries and inertial injuries [135]. Contact injuries occur in case of a direct impact to the head, for example, skull fractures, EDH, coup/contrecoup contusions and lacerations. Inertial injuries do not necessarily involve a direct impact to the head, but they are caused by the head acceleration. Examples of inertial injuries are concussion, SDH, contrecoup contusions, DAI and ICH.

For over half a century, research has been undertaken to assess the injury mechanisms causing inertial head injury during impact and to establish associated tolerance levels of the human head. The development of injury criteria has been a major goal among the researchers, in order to be able to assess the risk of sustaining a head injury and to assess the effectiveness of potential protection gears such as a motorcycle helmet.

Injury criteria for inertially induced head injuries can roughly be divided into three categories, as described by Bosch [36]:

- Injury criteria based on translational accelerations of the head's centre of gravity,
- Injury criteria based on translational and rotational accelerations of the head's centre of gravity,
- Injury criteria based on stresses and strains in the brain tissue.

Each category is discussed in the following. All these injury criteria are mainly developed to consider closed head brain injury. However, there are studies were it is presented the thresholds for skull fracture, due to contact load. The localized load, which could be considered as the suitable criteria to the skull fracture, depends on the shape of the impactor and on the thickness of the skull at the impact site. As already referred, Shuaeib *et al.* [67] highlighted the temporal region of the skull as the less thick one and as the weaker. Table 2.2 presents a summary of fracture peak forces at different regions of the skull.

Table 2.2: Peak force for fracture at different regions of the skull.

Impact area	**Force [kN]**	**Author**
Frontal	4.2	Nahum *et al.* [250]
	5.5	Hodgson and Thomas [251]
	4.0	Schneider and Nahum [252]
	6.2	Advani *et al.* [253]
	4.7	Allsop *et al.* [254]
	4.3-4.5	Yoganandan *et al.* [255]
	15.6	Voo *et al.* [140]
Temporal	3.6	Nahum *et al.* [250]
	2.0	Schneider and Nahum [252]
	5.2	Advani *et al.* [256]
	3.4-4.4	Yoganandan *et al.* [255]
	6.2	Voo *et al.* [140]
Occipital	12.5	Nahum *et al.* [257]
	11.7-11.9	Yoganandan *et al.* [255]
Parietal	3.5	Hume *et al.* [64]
Vertex	3.5	Yoganandan *et al.* [255]

Yoganandan *et al.* [255] have carried out a detailed review on the skull fracture experimental values against a hemispherical anvil, while in the other studies, mainly drop tests against a rigid flat surface were performed. Hume *et al.* [64] stated that a depressed skull fracture would be likely in the temporal area if the impacted area was less than 5 cm^2 and the localized pressure exceeds 4 MPa.

McElhaney *et al.* [338], Melvin *et al.* [339] and Robbins *et al.* [340] have reported cranial bone stress thresholds. According to the mentioned references, compact cranial bone breaks in tension at 48-128 MPa, while the cancellous bone breaks in compression at 32-74 MPa.

Raul *et al.* [448] proposed a global strain energy of the skull reaching 2.2 J as an indicator for skull fractures, with a 50% risk.

Translational acceleration based injury criteria

Several head injury criteria have been proposed. These methods try to relate the head injury to a single parameter, which could be measured and used more easily by protection systems researchers/designers. Several acceleration based injury predictors have been emerged.

Peak linear acceleration (PLA)

The PLA is basically the maximum acceleration value measured at the centre of gravity of the test headform during impact and is used by the current standards to assess the helmets. Usually, it is stated as a number multiplied by the gravitational acceleration constant g (1g

= 9.81 m/s^2). This method ignores the duration of the impact. However, the standards take into account the impact duration through the head injury criterion and also limits the duration of the impact. Moreover, some studies present a limit of 80 g for a duration that shall not pass 3 ms [209, 242, 261] to not occur any type of head injury.

Mertz *et al.* [241] estimates a 5% risk of skull fractures for a peak acceleration of about 180g, and a 40% risk of fractures for a peak acceleration of 250g.

More recently, King *et al.* [19], in a numerical study, estimated the MTBI (mild traumatic brain injury) tolerance for head linear acceleration, where there is a probability of MTBI occurrence of 25%, 50% and 75% for head linear acceleration of 559 m/s^2, 778 m/s^2 and 965 m/s^2, respectively.

Head Injury Criterion (HIC)

The most commonly acknowledged and widely applied head injury criterion is the HIC, which is based on the assumption that the linear acceleration of the head is a valid indicator of head injury thresholds. However, it does not take into account head kinematics nor direction of impact and rotational acceleration [210, 156, 132], even though rotational acceleration is believed to be the cause of several head injuries as demonstrated in the previous section, section 2.2.3. In consequence, the validity of HIC is intensively debated and there is reason to believe that the safety development could be made more efficiently by taking into account the effect of rotational kinematics into current safety procedures.

The HIC is the result of the evolution from the Wayne State Tolerance Curve (WSTC), developed in the pioneering work of Gurdjian and his co-workers [204, 179] and was first presented by Lissner [205], which established the relationship between the average translational acceleration and duration of the average acceleration pulse and it also created a boundary that separates the "skull fracture" zone from the "no skull fracture" zone [203], being used as the criterion for determination of concussion and the onset of brain injury. Further works were also developed [181], until the final form of WSTC was published by Gurdjian *et al.* [259], shown in figure 2.30, where skull fracture and concussion were used as the failure criterion. This relation between concussion and skull fracture was also observed by Melvin and Lighthall [134], where 80% of all observed concussion cases also had linear skull fractures. In the final form, the WSTC was developed by combining results from a wide variety of pulse shapes, cadavers, animals, human volunteers, clinical research, and injury mechanisms.

Therefore, the head can withstand higher accelerations for shorter durations and any exposure above the curve is considered an injury, while below the curve it does not exceed human tolerance. The WSTC is also supported by experiments conducted by Ono in primates and scaled to humans, which led to the Japan Head Tolerance Curve (JHTC), that is very similar to the WSTC. Nevertheless, the WSTC is based only on direct frontal impact tests, it was not applied to non-contact loading conditions and to other impact directions.

By plotting the WSTC in a logarithmic scale, it becomes a straight line with a slope of -2.5, which was used by Gadd in his proposed severity index called Gadd severity index (GSI) [206, 207]. Gadd [208] introduced the concept of a severity index to provide a rational and consistent basis for comparing the severity of various head impacts, based on the WSTC and on the long pulse duration tolerance data by means of the [237] test data and given by this empirical expression:

$$GSI = \int a(t)^{2.5} dt \tag{2.1}$$

where a is the instantaneous head acceleration in g's and t is the time duration of the acceleration pulse in seconds. The initial value to this failure criterion from Gadd's point

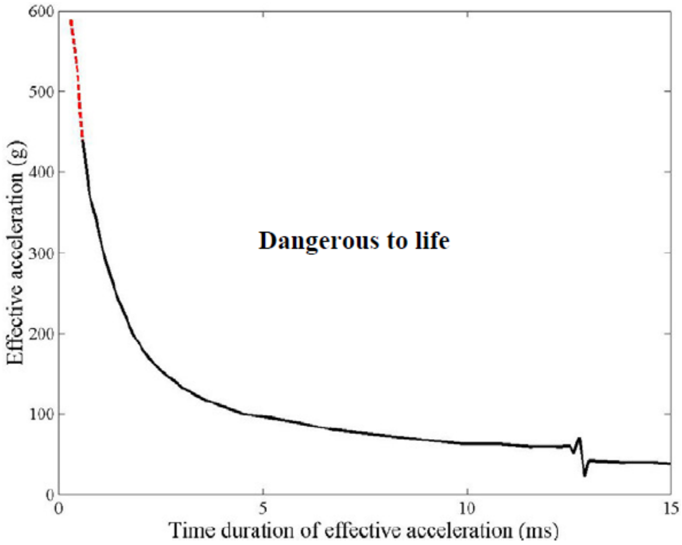


Figure 2.30: *The Wayne State Tolerance Curve [121].*

of view , was initially set to a 1000, as a threshold for concussion for frontal impact. Later, Gadd [234] suggested a threshold of 1500 for non-contact loads on the head.

Over the years, this criterion was reviewed and several modified forms were proposed. One of those reviews, was made by Versace [209], who analysed the relationship between the WSTC and GSI and proposed a mathematical approximation of the WSTC that is based on average acceleration, which expression is represented by the equation 2.2.

$$VSI = \left[\frac{1}{T}\int_{t}^{t} a(t)dt\right]^{2.5} \tag{2.2}$$

Later, the National Highway Traffic Safety Administration (NHTSA) proposed the head injury criterion (HIC) [235], a new criterion to identify the most damaging part of the acceleration pulse by finding the maximum value of the same function. This form is known at present as HIC:

$$HIC = \left(\left[\frac{1}{t_2 - t_1}\int_{t1}^{t2} a(t)dt\right]^{2.5}(t_2 - t_1)\right)_{max} \tag{2.3}$$

Where $a(t)$ is a resultant head acceleration in g's, the interval $t_2 - t_1$ are the bounds of all possible time intervals defining the total duration of impact that must be less or equal to 36 ms and t_1 and t_2 are any two points of the acceleration pulse in time, in seconds.

A HIC value of more than 1000 is considered to result in severe head injury (however a helmet could be approved by a standard with higher HIC values). Hopes and Chinn [59] indicated that there is an 8.5% probability of death at an HIC value of 1000, 31% at 2000 and 65% at 4000.

The HIC takes into account both acceleration and duration. The linear acceleration, a, is the resultant acceleration measured by a triaxial accelerometer array positioned in the centre of gravity of the dummy head/headform, which has similar inertial properties than the human head. This approach claims that two parameters (acceleration and duration of the acceleration over the time of impact) rather than a single parameter (peak acceleration) are suitable for the definition of the limitation, which consisted in an improvement to the criteria assessment [210]. However, this does not take into account variations in human tolerance and is based on the assumption that the brain is a visco-elastic medium [48].

Other researchers have criticized the use of the HIC as a suitable predictor for head injury. Viano [141] has showed analytical results for skull fracture and vascular brain damage where high degree of overlap exists. In figure 2.31, it is also possible to observe the occurrence of head injuries though the HIC values for these cases are lower than the limit. In addition, Viano [141] added that reliable predictions should not be expected from a measurement of a resultant translational acceleration of the head and analysis by a mathematical routine that gives results in a single HIC value. However, Hopes and Chinn [59] also reviewed HIC drawbacks made by other researchers and concluded that HIC still could be an useful predictor for comparing energy absorbing safety devices in impacts where the death frequently occurs without skull collapse. Also, Deck *et al.* [6] concluded that HIC is able to represent the global severity level of an impact and the potential head injury level, however HIC is unable to predict diffuse brain injuries and SDH that are linked to the angular acceleration sustained by the head during the impact. It was also highlighted that an optimisation based on biomechanical criteria is different than the optimisation with HIC criterion [272]. In a study about HIC, Fenner *et al.* [6] criticised HIC for not being sensitive to impact direction. Newman [210] has the same opinion. Kleiven and von Holst [196] criticized HIC too, because it does not predict the size dependence of the intracranial stresses associated with injury and it does not take into account the head size. However, limits for some sizes of the human head were proposed for HIC_{36} by Kleinberger [239] and for HIC_{15} by Eppinger [240] using the HIC scale factor proposed by Melvin [238]. The higher proposed limits were the ones relative to adults, 1000 and 700 for HIC_{36} and HIC_{15}, respectively. However, these limits only take into account the skull material properties.

In overall, HIC is considered to be not enough to predict head injuries because it does not take into account the injury type, the rotational motion and the impact direction and also has nonsensical units [450].

Thus, HIC only treats the resultant translational acceleration and the duration of the impulse and no consideration is made for the direction of the impulse or rotational acceleration components [274, 275, 285].

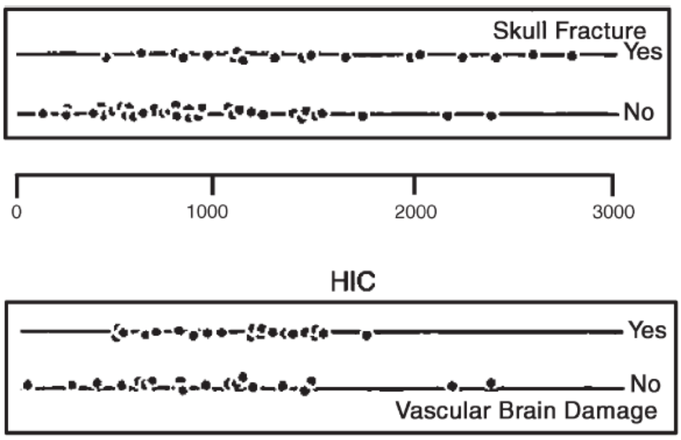


Figure 2.31: *Relationship between measured HIC and the occurrence of the skull fracture or the extravasations of fluid from blood vessels [141].*

Actually, the HIC is the most disseminated injury criterion mainly due the fact that it is the head injury criterion adopted by the current helmet's standards for helmet homologation, such as the ECE R22.05, to assess the safety of motorcycle helmets. Thus, all helmets available in the market were assessed according to this criterion and the HIC values were inferior relatively to the imposed limit. The PLA is used together with the HIC by the majority of the standards. ECE R22.05 [8] is an example of these standards that use both criteria. In the case of ECE 22.05, the peak linear acceleration is limited to 275 g and the HIC value should be inferior to 2400 in order to be approved. However, Shuaeib *et al.* [67] concluded that a sever but not life-threatening injury can occur if HIC reaches or exceeds 1000. Limits for HIC were suggested by Horgan [442] for HIC values of 1000 and 3000 which were defined as 16% and 99% probability of life threatening injuries, respectively.

King *et al.* [19], in a numerical study, estimated the MTBI (mild traumatic brain injury) tolerance for HIC_{15}, where there is a probability of MTBI occurrence of 25%, 50% and 75% for HIC values of 136, 235 and 333, respectively.

Other assessing head injury method based on probability laws and on the HIC was also proposed [270]. This criterion is widely used in automotive safety testing and evaluation of protective equipment for the head [211, 212, 213, 214]. Further refinement related to helmet testing were made when a headform is used, the HIC was revised to the following equation:

$$HIC_d = 0.75446HIC + 166.4 \tag{2.4}$$

where HIC_d refers to the HIC for the hybrid III head only without a body [215].

Thus, HIC and proposed acceleration thresholds do neither take into consideration rotational and translational loads, nor directional dependency. There is therefore a need for more complex injury assessment functions, accounting for both translational and angular acceleration components as well as changes in the direction of the loading [285].

Combined rotational and translational acceleration based injury criteria

The brain is composed by biological tissue which is a natural viscoelastic material. The response of this tissue is dependent on the magnitude and rate of the acceleration or on the change of rotational velocity.

Purely translational or rotational loading to the human head is uncommon in reality, as these types of movements are not physiologically possible, mainly because of the effects of the head-to-neck connections in reality. Rotation is the most injurious loading mechanism to the brain and in the vast majority of head impact situations it can be expected that both translational and rotational accelerations are present and combine to cause brain injury.

Aare *et al.* [273], performed a FE study, simulating several oblique impacts for different impact sites and directions and when measured the strains in the brain tissue in the Finite Element model of the human head subject to oblique impacts, it could be seen that injury thresholds should include rotational parameters as well as translational parameters.

Vezin and Verriest [278] suggested that high stress rates in the brain arise in multiple complex configurations, where both rotational and linear accelerations appearing in all directions.

Injury criteria that takes into account rotational kinematics or both rotational and translational have been proposed by some researchers, by performing tests on volunteers, cadavers, primates, where it is assessed the rotational acceleration effect. Global kinematic measures such as magnitude in angular and linear acceleration, change in angular and translational velocity, as well as predictors HIP and GAMBIT, were investigated by some researchers with regard to their ability to take into account consequences of different impact directions and durations for the prediction of intracranial strains associated with injury [285].

Rotational acceleration thresholds

The importance of the rotational acceleration in head impact brain injury particularly for DAI have been emphasized by many researchers, as already seen in section 2.2.3. However, the brain injury tolerance to rotational acceleration is not fully understood, which still is an active area of research.

Glaister [198] has presented the work carried out by Lowenhielm [217], which resulted in the tolerance curve shown in figure 2.32. Lowenhielm [217] developed mathematical models based upon the anatomy and has also used mechanical models and accident analysis to assess the tolerance of the human brain to angular acceleration. It was also shown that both critical angular acceleration and angular velocity must be exceeded, to result in an injury. In another words, the angular acceleration must be applied long enough to attain a critical angular velocity and excessive displacement between brain and skull.

Kleiven [337] found that HIC manage to predict the strain level in the brain of a FE model for purely translational impulses of short duration, while the peak change in angular velocity showed the best correlation with the strain levels in an FE head model for purely rotational impulses. Thus, as already referred, in a real crash both components of acceleration are present and HIC are not able to predict the rotational one.

The values for bridging vein disruption (solid line) and concussion tolerance are illustrated in figure 2.32 as shaded areas on a plot of angular acceleration against angular velocity. It may be noted that no axis of rotation (or plane) is specified but it is obvious from the complex anatomy of the brain and its non-isotropic constraint within the cranium, that differences in sites of injury, as well as on tolerance level will be expected depending upon the axis about which the skull is rotated [67]. It was pointed out that family of tolerance curves would be expected for the three axis of rotation as well as for three major axis of linear acceleration [198].

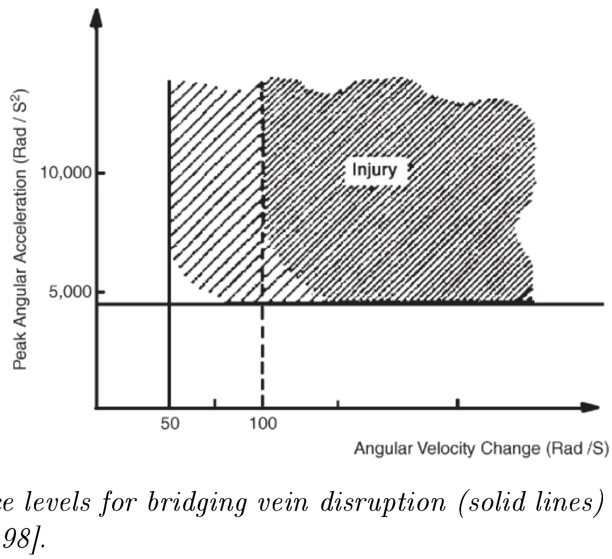


Figure 2.32: *Tolerance levels for bridging vein disruption (solid lines) and for gliding contusions (dashed line) [198].*

In a work performed by COST-327 Motorcyclist's helmet working group [14], it was found that a 50% probability of head injury with AIS 3 is resulting from 10 krad/s^2 and 0 g linear acceleration down to 0 krad/s^2 and 190 g, whereas AIS 4-5 from 14 krad/s^2 and 0 g down to 0 krad/s^2 and 230 g. A 10% probability of head injury with AIS 2-5 is given from rotational acceleration varying between 5000 and 6000 rad/s^2. These values agree well with the tolerance curve presented in the work made by Glaister [198]. The forgoing results combines both the linear and rotational acceleration effects to a single parameter which is the Abbreviated Injury Scale (AIS) [385], index obtained through statistical approach [67]. The AIS 1 means a minor injury and the AIS 6 means a very high lethal injury. The Abbreviated Injury Scale is an empirically-based categorical scale that assigns an injury severity rating on the basis of the observed injuries sustained by the experimental subject following the test.This index ranks injury severity as follows:

Table 2.3: AIS head injury classification [258].

AIS code	Injury Severity	Injury description
0	No Injury	
1	Minor Injury	Scalp abrasion or superficial laceration Nose Fracture
2	Moderate Injury	Vault and mandible fractures
3	Serious Injury	Basilar fracture Total scalp loss Single contusion cerebellum
4	Severe Injury	Brain damage : small EDH and SDH
5	Critical Injury	Penetrating injuries Brain stem compression Large EDH and SDH DAI
6	Fatal Injury	Massive destruction of both cranium and brain

More recently, Gennarelli *et al.* [317] hypothesized the magnitude of angular acceleration required to induce increasing levels of diffuse brain injuries (Concussion and DAI) in the human correlated with the abbreviated injury scale (AIS) (at a certain angular velocity). To further define the present injury level, peak angular accelerations obtained in this study were scaled to the human according to brain mass [150], and the corresponding injury level was determined 2.4.

Table 2.4: Categories of DBI based on biomechanical injury [317].

AIS level	Injury Severity	Angular Acceleration (rad/s^2)	Angular Velocity (rad/s)
1	Mild Cerebral Concussion	2877.8	25
2	Classical Cerebral Concussion	5755.6	50
3	Sever Cerebral Concussion	8633,4	75
4	Mild DAI	11511,2	100
5	Moderate DAI	14389	125
6	Severe DAI	17266,8	150

The values indicated in table 2.4 were obtained through this relationship for rotational acceleration [rad/s^2] = 2877.8 $\times$ AIS.

The biomechanical response of the head during impact also includes rotational motion, which is believed to cause injury [38, 236, 157], in particular acute SDH and DAI that are considered more lethal than most other brain lesions [139]. This gives a special interest in deriving injury criteria for SDH and DAI. A summary of various tolerances of the human brain to rotational acceleration (and rotational velocity) is given in table 2.5.

The great majority of these studies were performed by inducing rotational motion to the sagittal plane.

Zhang *et al.* [263], using an advanced finite element model of the head, suggested tolerance for reversible brain injury level was less than 85 g for translational acceleration and less than 60000 rad/s^2, for a head exposed to combined translational and rotational acceleration (impact duration between 10 and 30 ms). It was also 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.

Before that, Margulies and Thibault [243] presented a criterion for DAI. It is developed using experiments on primates in combination with gel physical models and analytical scaling procedures. The criterion is represented by curves that represents equal strain in the analytical model as a function of the angular acceleration and peak change of angular velocity. Judging from figure 2.33, rotational accelerations exceeding 10 krad/s^2 combined with a rotational velocity of 100 rad/s or higher, gives a risk of DAI for an adult individual. These curves show that for small changes in angular velocities the injury is less dependent on the peak angular acceleration, while for high values of peak change in angular velocity, the injury is sensitive to the peak angular acceleration. This is in agreement with the hypothesis of [37], that stated the shear strain, for long duration impulses (large peak change in rotational velocity) is proportional to the acceleration, while the injury is proportional to the change of velocity of the head for short duration impacts. The full line represents the limit for an average adult (brain mass of 1,4 kg).

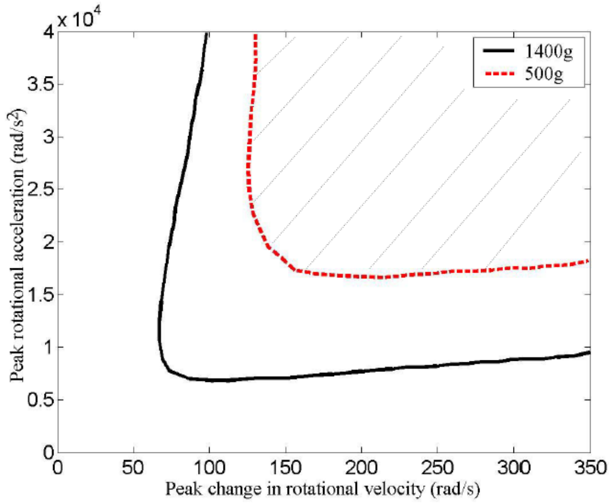


Figure 2.33: *Angular threshold for DAI prediction [243].*

In the study performed by Gennarelli and Thibault [139] on primates, it was proposed that an angular acceleration exceeding 175 krad/s^2 combined with an impulse time exceeding 5 ms, would produce SDH in the monkey (Figure 2.34). Interestingly, the threshold for SDH in monkey seemed to increase with the duration of the impulse, which is the opposite of other head injuries such as skull fracture and concussion as shown in figure 2.30. This phenomenon

Table 2.5: Human brain tolerance to rotational acceleration and velocity.

Injury	Tolerance	Author
Cerebral concussion	50% probability: α = 1800 rad/s^2 for t<20 ms ω = 30 rad/s for t $\geq$ 20 ms 99% probability: α >7500 rad/s^2 for t>6.5 ms	Ommaya *et al.* [246] and Ommaya and Hirsch [150]
Cerebral concussion	α = 14000 rad/s^2 for 11 ms	Unterharnscheidt and Higgins [149]
Bridging vein rupture	α=4500 rad/s^2 or ω= 50 - 70 rad/s	Löwenhielm [244, 364]
Several brain injuries	α= 1700 rad/s^2 and ω = 60 - 70 rad/s	Ewing *et al.* [268]
Cerebral concussion	α = 13000 rad/s^2 for 11 ms	Ono *et al.* [156]
Brain surface shearing	α = 2000 - 3000 rad/s^2	Advani *et al.* [257]
Cerebral concussion DAI SDH	α = 20 krad/s^2 for 18 ms α = 20 krad/s^2 for 18 ms α = 32 krad/s^2 for 14 ms	Gennarelli and Thibault [139] Gennarelli and Thibault [139] Gennarelli and Thibault [139]
Several brain injuries	ω<30 rad/s: safe: α < 4500 rad/s^2 AIS 5: α > 4500 rad/s^2 ω>30 rad/s: AIS 2: α = 1700 rad/s^2 AIS 3: α > 3000 rad/s^2 AIS 4: α > 3900 rad/s^2 AIS 5: α > 4500 rad/s^2	Ommaya [365]
DAI	α = 19 krad/s^2 for 20 ms	Gennarelli and Thibault [157]
Several brain injuries	α= 25000 rad/s^2 for short durations	Tarriere [260]
Cerebral concussion	α= 13600 - 16000 rad/s^2 and ω= 25 - 48 rad/s	Pincemaille *et al.* [247]
Cerebral concussion	α = 18 krad/s^2 for 18 ms	Thibault and Gennarelli [282]
DAI	α=10000 rad/s^2 and ω= 100 rad/s	Margulies and Thibault [243] (similar values of Gennarelli and Thibault [276])
Cerebral concussion	50% of probability: α=6200 rad/s^2	Newman [226]
Brain injuries	α>5000 rad/s^2	Thomson *et al.* [267]
Several brain injuries	4500< α <5000 rad/s^2 and ω= 60 rad/s	Shuaeib *et al.* [67]
Mild DAI Moderate DAI Severe DAI	α=12500 rad/s^2 α=15500 rad/s^2 α=18000 rad/s^2	Ommaya *et al.* [264]
MTBI	25% of probability: α=4384 rad/s^2 50% of probability: α=5757 rad/s^2 75% of probability: α=7130 rad/s^2	King *et al.* [19]
MTBI	α= 6000 rad/s^2 for 10<t<30 ms 25% of probability: α=4600 rad/s^2 50% of probability: α=5900 rad/s^2 80% of probability: α=7900 rad/s^2	Zhang *et al.* [263]
SDH	α= 10000 rad/s^2	Yoganandan *et al.* [466]

Cerebral concussion	α= 6400 rad/s^2 and ω= 35 rad/s	Viano *et al.* [343]
SDH	α= 10000 rad/s^2	Depreitere *et al.* [324]
Cerebral concussion	α= 6200 rad/s^2	Fijalkowski *et al.* [316]
Cerebral concussion	α= 1800 rad/s^2	Kleiven [441]
DAI	α= 8000 rad/s^2 or ω= 70 rad/s	Kleiven [441]
Cerebral concussion	α= 7600 rad/s^2 for 15 ms α= 7300 rad/s^2 for 23 ms	Fijalkowski *et al.* [334]
DAI	α= 10000 rad/s^2 for t>4 ms or ω= 19 rad/s	Davidsson *et al.* [347]

can also be seen in experiments on human cadavers reported by [244] and experiments on monkeys reported by Unterharnscheidt and Higgins [149]. The results from the experiments on primates by Gennarelli and Thibault [139] are presented in figure 2.34. The circles in this figure show the forces that lead to concussion, the dots show the forces that lead to diffuse brain injuries and the crosses show the forces that lead to acute SDH. This can contribute to a good prediction for concussion, due to the repeatability of the points for concussion in figure 2.34.

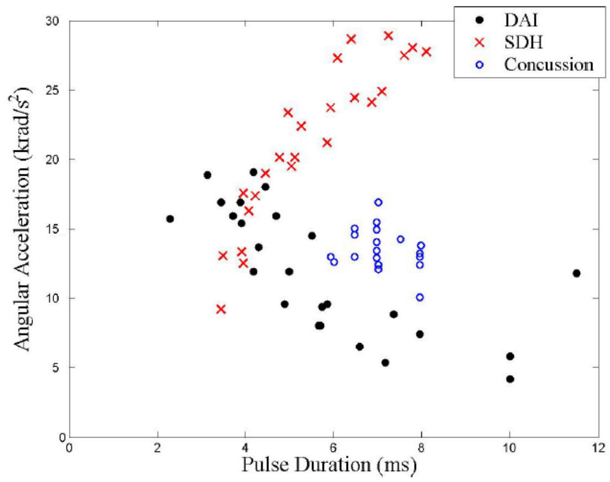


Figure 2.34: *Angular threshold for injury prediction (Adapted from [139]).*

Margulies and Thibault [323] proposed a lesion parameter, generalized from DAI research, the peak change in rotational velocity of the human head $\Delta\omega_p$, defined as:

$$\Delta\omega_p = max\left[\sqrt{\left(\int_{t1}^{t2}\omega_x(t).dt\right)^2+\left(\int_{t1}^{t2}\omega_y(t).dt\right)^2+\left(\int_{t1}^{t2}\omega_z(t).dt\right)^2}\right] \forall t_1 < t_2 \quad (2.5)$$

$\Delta\omega_p$ is hypothesized to correlate to brain injury. The change in rotational velocity is directly related to the moment of the impulses exerted by the tangential forces working upon the brain mass over the duration of the impact. These impulses are responsible for the compression of the brain mass in the tangential direction. Because the brain is a viscoelastic material and contains blood-filled veins, the time duration will be a relevant factor in the amount of brain compression. Hence the change in angular velocity, and the tangential impulses related to it, as they are a combination of forces and time duration, should be related to brain injury. A second new head lesion parameter is the peak angular acceleration of the head after triangulisation α_p, which is defined as the peak angular acceleration value of the triangulated pulse. It has been used in literature in acute SDH tests by Löwenhielm [217] and later by Depreitere *et al.* [324].

Therefore, Deck *et al.* [332] strongly suggest to consider head rotation in any future standard evolution, whatever the accident type is, pedestrian motorcyclist or sport. Actually, is evident that a better injury criterion should therefore include both rotational and translational parameters, such as the change in rotational velocity and HIC and should probably be dependent of direction.

Generalized Acceleration Model for Brain Injury Threshold (GAMBIT)
Newman [218] introduced a head injury assessment function that takes into account both translational and rotational acceleration. Newman [218] attempted to combine translational and rotational head acceleration response into one injury criterion by considering both these accelerations as the cause for stresses generated in the brain and resulting in brain injury. On the assumption that translational and rotational acceleration equally and independently contribute to head injury, the GAMBIT expression is:

$$G(t) = \left[\left(\frac{a(t)}{a_c} \right)^n + \left(\frac{\alpha(t)}{\alpha_c} \right)^m \right]^{1/s} \tag{2.6}$$

where $a(t)$ and $\alpha(t)$ are the instantaneous values of translational expressed in [g] and rotational acceleration respectively expressed in [rad/s^2], n, m and s are empirical constants selected to fit available data; a_c and α_c represent critical tolerance levels for those accelerations.

The GAMBIT requires to establish the maximum value of the function $G(t)$. G=1 is normally set to correspond to a 50% probability of AIS>3. Some versions of $G(t)$ have been presented [218, 226].

Proposed values for the constants are $n = m = s = 2$, a_c=250 g and $\alpha_c = 25000$ rad/s^2 [218] and more recently, COST 327 group [14] proposed a reduction in the rotational threshold to $\alpha_c = 10000$ rad/s^2 that was also used more recently by Mellor and StClair [70].

Figure 2.35 shows curves of constant GAMBIT obtained by using equation 2.6. 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 acceleration contribute equally to the probability of head injury and assuming that tolerances derived in experiments with either translational or rotational acceleration are also valid in a combined loading scenario.

In overall, the GAMBIT predicts injury when G > 1 and no injury when G < 1. However, the GAMBIT was never extensively validated as an injury criterion. For example, the maximum time interval for the acceleration pulse has never been set. Although a step in the right direction, it had some limitations in the lack of data used to derive the thresholds, lack of impulse duration dependency, as well as in lack of accounting for directional sensitivity [121]. More recently, Newman [226] reported a 50% probability of concussion for a GAMBIT value superior to 0.4.

GAMBIT has recently been employed in some studies in an effort to better understand the nature of protective headgear [227, 14].

Head Injury Power (HIP)
Newman *et al.* [226] reasoned that the rate of change of translational and rotational kinetic energy, i.e. power, could be a viable biomechanical function for the assessment of head injury.

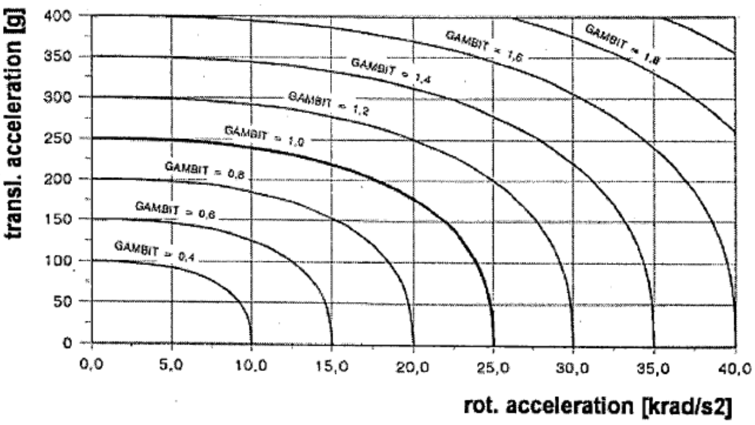


Figure 2.35: *GAMBIT curves for constant GAMBIT values [262].*

Newman *et al.* [226] introduced a more general head injury assessment function, the Head Impact Power (HIP). This function, which considers the maximum rate of translational and rotational energy transfer to be the controlling element in inertially induced brain injury, which has been successfully used in the development of a new North American football helmet ("The Riddell Revolution"). HIP was also employed to help quantify head injury threats in soccer [219]. Still, the failure criteria for every published helmet performance standard in the world continue to be based solely on linear acceleration of a test headform.

Newman *et al.* [226] proposed coefficients for the different directions that could be chosen to normalize the HIP with respect to some selected failure levels for a specific direction, in other words, proposed a scaling of the impact power for different directions, depending on the tolerance level for the actual direction. However, values of the coefficients were not presented by Newman and information regarding directional sensitivity was lacking.

HIP is expressed by an empirical expression that relates a measure of power to head injury:

$$HIP = Aa_x \int a_x \, dt + Ba_y \int a_y \, dt + Ca_z \int a_z \, dt + D\alpha_x \int \alpha_x \, dt + E\alpha_x \int \alpha_x \, dt + F\alpha_x \int \alpha_x \, dt \quad (2.7)$$

The development of the HIP function is described in Newman *et al.* [227] and HIP value is expressed in Watt unit. Each term in this expression represents the change in kinetic energy for one degree of freedom, where the first half represents the linear contribution and the second one the angular contribution. HIP criterion needs individual scaling coefficients for the different terms to account for differences in intracranial response due to a variation in load direction. The directional sensitivity of the human head is not known and values of the coefficients were not presented and information regarding directional sensitivity was lacking. Therefore, the coefficients of equation 2.7 are currently set to reflect mass and mass moments of inertia of the human head. The coefficients A, B and C represent the mass of the human head, m [kg] and D, E and F represent the appropriate moments of inertia for the human head I_{xx}, I_{yy} and I_{zz} [kg.m^2] respectively, which denote the injury sensitivity for each of the six degrees of freedom of the head. Newman *et al.* [226] proposed the utilization of average values for a 50th percentile adult male to compute HIP.

The time dependent accelerations in each of the six degrees of freedom can be separated in the linear components a_x, a_y and a_z, expressed in m/s^2 and also in the rotational components α_x, α_y and α_z expressed in rad/s^2.

Newman *et al.* [227] validated the HIP, using experimental data from carefully analysed collisions of American Football players during the game. Nevertheless, the author himself suggested that HIP need further validation. The value taken as an injury predictor candidate is the maximum value reached by this function. However, the HIP_{max} is only validated for mild traumatic brain injury. A 50% probability of concussion at a maximum Head Impact Power (HIP_{max}) of 12.8 kW was found. The HIP_{max} is not validated for higher severity head injuries, since such experimental data is not yet available and then the tolerance level needs to be adjusted compared to the tolerance established for mild traumatic brain injury. From their results, the authors concluded that HIP_{max} better correlates with mild traumatic brain injury than HIC. The authors give three advantages of HIP_{max} over HIC to backup this conclusion: Besides translational acceleration, HIP_{max} can also incorporate directional sensitivity, sensitivity for rotational acceleration and sensitivity for angular and translational velocities. However, as already referred, it has only been validated for mild traumatic brain injury.

More recently, Marjoux *et al.* [322] tested the prediction capability of HIC for SDH and skull fractures. The HIP results for skull fractures are moreover as good as the ones with HIC. When considering moderate brain injuries, the results for HIP are slightly better than HIC. This was expected by the authors since the HIP calculation takes rotational acceleration fields into account and neurological injuries are supposed to be more correlated with angular accelerations than linear accelerations as suggested in King *et al.* [19]. However for more violent cases, the rotational accelerations were found negligible in this study compared to the linear ones. This could explain why the results of the HIC, which provides a more elaborate way to take linear accelerations into account, become better than the HIP for severe neurological injuries.

Some investigation regarding this issue was done by Kleiven and von Holst [143] where it was stated that when it comes to relative motion and strains in the bridging veins, the HIP criterion should give a better prediction of the risk of SDH than other criterion like HIC. This is evident since the HIP takes into account the load direction and the rotational components of the acceleration. However, the only factors that differentiate between directions in the original HIP are the variations in the mass and moments of inertia. Nevertheless, Kleiven and von Holst [143] suggested a modification of the HIP to account for differences induced by reduction of brain size due to atrophy.

Newman *et al.* [226] proposed a scaling of the impact power for different directions, depending on the tolerance level for the direction in question. The maximal relative motion and strain in the bridging veins appear about 6 ms later than the maximal values of the HIC and HIP. This delay between injury and prediction value are apparent for all directions according to Kleiven and von Holst [143]. Newman *et al.* [226] estimated a 50% probability of concussion for a HIP of 12.8 kW, and a HIC value of 240 for a 50^{th} percentile head.

Kleiven [285], using a detailed FE human head model, evaluated the various global kinematic-based injury measures for rotational motion by keeping the various measure constant while varying the impulse duration and it was found that the change in angular velocity mirrored the level of strain in the brain better than the HIP and the peak angular acceleration did. An almost constant level of strain was found for a constant change in angular velocity, while for both the HIP and the peak angular acceleration gave an increasing strain level for an increase in the impulse duration. This corresponds to Holbourn's hypothesis [37] that the strain (and the injury) is proportional to the change in angular velocity for rotational impulses of short durations. The same was done for translational motion and it was found that the HIC and HIP mirrored the level of strain in the brain better than the change in velocity (linear) did. An almost constant level of strain was found for a constant HIC and HIP, while a constant change in velocity gave a decreasing strain level for an increase in impulse duration. This supports the results presented by Newman *et al.* [226], where a good correlation was found between concussion and both the HIC and the HIP for predominantly translational impact data.

However, HIC and HIP do not seem to capture the level of intracranial response for different impulses (HIC and HIP for purely translational impulses of short duration and peak change in angular velocity for purely rotational impulses) [285]. This was expected since a single-mass model used by criteria such as the HIC or the HIP is not able to correctly model the intracranial mechanical behavior [322]. A zero HIC value is predicted for a pure rotational impulse while higher levels of stresses and strains are found compared to a corresponding translational impulse in the same direction. This underlines findings by previous investigators [152]. One possible explanation could be the as yet unexplored synergic effects of combined loadings. This is included naturally by the product of inertia terms for the angular components in the impact power formulation when using anatomical coordinates. Since the anatomical directions do not coincide with the principal directions of inertia, the product of inertia, I_{xz}, is non-zero. However, in the case of the human head, the power terms containing the products of inertia Ixy, Ixz and Iyz are insignificant compared to the moments of inertia I_{xx}, I_{yy} and I_{zz} [292, 293].

More recently, it was found that a simple linear combination of rotational velocity and HIC showed a high correlation with the maximum principal strain in the brain [363].

This criteria represent a step in the right direction, however improvements are still needed. Aare *et al.* [273] defended that a new injury criterion should include both rotational and translational parameters, such as the change in rotational velocity and HIC and should probably be dependent of direction.

Strain correlated with both accelerations

Aare *et al.* [464], after developed a test rig to perform oblique impacts, have tried to develop a criterion that correlate both translational and rotational acceleration with strains in brain tissue. In a previous work of this group, Kleiven and von Holst [171] found that the change in angular velocity correspond best with the intracranial strains found in the FE model. For translational impulses on the other hand, the HIC and the HIP have shown the best correlations with the strain levels found in the model [171].

Thus, as the strains in the brain tissue are proportional to the HIC value for pure translations, and also proportional to the change in rotational velocity for pure rotations of short impact durations [171], it was suggested by Aare *et al.* [464] that the output data can be fitted to the following formula:

$$\varepsilon = k_1 \Delta\omega + k_2 HIC \tag{2.8}$$

where ε is the maximum strain in the brain tissue, $\Delta\omega$ is the peak resultant change in rotational velocity, HIC is the head injury criterion and k_1 and k_2 are constants. These constants were obtained by regression analysis for each impact and are available in Aare *et al.* [464].

No threshold was proposed by Aare *et al.* [464], however the authors used in the same study the threshold proposed by Bain and Meaney [195],where a strain of 20% was shown to be critical to the brain tissue and the maximal principal strain in the brain tissue was chosen as a predictor of injuries, as it has been shown to correlate with DAI.

Nevertheless, other thresholds could be correlated with the result of this criterion. Some are reviewed in the following section 2.2.4.

Stress and Strain Based Injury Criteria

As seen in the previous sections, kinematic head injury predictors are not capable of perfectly distinguish injuries to head tissues, such as DAI and SDH. This drawback has formed a tendency among researchers to use head injury predictors that are based on the tissue level response of the head, rather than on its kinematics.

Brain injury is reported to correlate with stress, strain and strain rate [269, 300]. However, strains and strain rates inside the brain (during impact) are difficult to measure [36]. This can be achieved by using anatomical detailed and accurate Finite Element models of the human head, where the stresses and strains used to compute the injury parameters are calculated from Finite Element simulations, using a Finite Element model of the skull and intracranial contents.

Therefore, these models bring a detailed injury assessment closer to reality, since they enable stresses and strains to be examined and the behaviour of the brain can be predicted and improved injury criteria can be developed and implemented into safety standards. These FE models often contain a detailed geometrical description of the anatomical components but lack some accurate descriptions of the mechanical behaviour of the brain tissue and other features, which is currently an active area of research.

Therefore, the use of more delicate tools, such as FEHM together with local tissue strain thresholds seems to be the best way to evaluate the influence of rotational motion in the head structures [336]. For example, Kleiven [337] found that HIC predicted the strain level in the brain of a FEHM for purely translational impulses of short duration, while the peak change in angular velocity showed the best correlation with the strain levels in FEHM for purely rotational impulses.

DiMasi *et al.* [167] and Bandak [228, 229] developed three component level injury predictors representing the general types of brain injuries experienced in traffic accidents: the cumulative strain damage measure criteria (CSDM), Dilatation Damage Measure (DDM), Relative Motion Damage Measure (RMDM). Over the years, other predictors have been proposed, such as the applied brain pressure tolerance (ABPT) and the von Mises brain shear stress (VMSS)

It is possible that future methods used to assess head injury risk and protective head gear will rely on the predictions from numerical head models, which should hopefully provide more robust and accurate means of assessing head injury risk instead of HIC.

More recently, Takhounts *et al.* [277] proposed the SIMon FE model based on three injury metrics proposed by DiMasi *et al.* [167] and Bandak [228, 229], the CSDM, the DDM and the RMDM. The SIMon tool incorporates a numerical head model with purpose designed model based head injury criteria, and the head model possesses a rigid skull in order that headform motion data measured in physical tests can be used to load it.

Deck *et al.* [6] concluded that an optimisation based on biomechanical criteria is different than the optimisation with HIC criterion which is correlated with acceleration of headform's centre of mass and used for helmets homologation, showing the importance of development of good criteria based on stress and strain features of a FEHM.

Applied brain pressure tolerance (ABPT)

This threshold was first proposed by Nahum *et al.* [192] and then implemented by several researchers to investigate its validity. This criterion was based on the use of a skull-brain biomechanical head model. The use of computational head model is obvious as no any mechanical measurement could be made on the human brain during impact. Therefore, the more the realistic biomechanical head model, the more accurate are the results would be expected. This is in fact applicable to all the analytical approaches which are based on biomechanical modelling of the human head. The criterion proposed by Nahum *et al.* [192] states that a 172,3 kPa intracranial pressure will produce a moderate injury while a pressure in excess of 234,4 kPa will produce a sever to fatal injury. Ward and Chan [158] proposed tolerances for intracranial pressure, where less that 173 kPa is a minor or absent injury, while a greater pressure than 235 kPa means a serious injury. Between these two values are obviously the moderate injuries. This criterion has been implemented by Liu and Fan [224] to simulate helmeted-head impacts using a validated Nahum's skull-brain FE head model and concluded that ABPT has a better sensitivity for very short time impact than HIC. However, Kang *et al.* [220] and Miller *et al.* [223] had criticized this criterion for the prediction of brain injury particularly DAI brain injury type. Also Turquier *et al.* [222] had found that there was a shift between the experimental and the numerically predicted brain pressures. A brain pressure reaching 200 kPa is an indicator for brain contusions, oedema and haematoma proposed by Raul *et al.* [448].

Nevertheless, some values were proposed as tolerance. Ward *et al.* [303] proposed an intracranial pressure tolerance of 235 kPa for serious brain injuries. Baumgartner [304] proposed a brain pressure reaching 200 kPa as an indicator for brain contusions, oedema and haematoma. Yao *et al.* [302] proposed critical values for coup pressure of 180 kPa.

Willinger and Baumgartner [320] using the University Louis Pasteur (ULP) finite element head model, established that computed brain pressure is not correlated with the occurrence of brain hemorrhages, whereas brain Von Mises stress is.

Brain von Mises shear stress (VMSS)

This criterion assumes that the shear stress and not the pressure is the cause of brain damage. This criterion was developed by Kang *et al.* [220] using an ULP (University Louis Paster) head model. The model was validated with experimental tests and Nahum's computational model for the same impact conditions. The predicted ULP model injury was compared with real motorcycle accident autopsy and the results were found to be promising. The criterion sets an injury tolerance of range 11-16.5 kPa. Zhou *et al.* [449] reported a threshold of 11 kPa. It has been reported that there results were found to be in agreement with other published work using different biomechanical head models such as WSU (Wayne State University) brain injury model or simulating experimental ferret head impact with a 3D finite element head model of a ferret brain [220]. A similar criteria based on shear stress was used by Zhou *et al.* [225] who had proposed a higher value of 20 kPa for DAI to occur in piglets. Therefore, a more refinement work is clearly still required to converge to a closer range of Kang *et al.* [220] work and confirm the 20 kPa stated by Zhou *et al.* [225] for human models. Miller *et al.* [223] showed in a 2D FE study that the maximal von Mises stress predicts comparable patterns of axonal and macroscopic haemorrhagic cortical contusions in piglet. Anderson [366] proposed a limit of 27 kPa.

Later, Baumgartner *et al.* [266] performed a study that led to the consideration of intracranial von Mises stress as a good injury criterion for concussion or other mild traumatic brain injuries when reaching values above 15 kPa. These results are consistent with the conclusions of an injury tolerance limit study on motorcyclists accidents published by Willinger *et al.* [265] where the critical von Mises stress values for the brain resulting in concussion was estimated to be above 20 kPa, which is higher than the limit deduced from Baumgartner *et al.* [266]. Deck *et al.* [6] proposed a maximum value of 40 kPa for von Mises stress indicating that is a good indicator of concussion. Zhang *et al.* [263] proposed a shear stress of 7.8 kPa as the tolerance level for a 50% probability of sustaining a MTBI, in a study about injury criteria and threshold for MTBI, using an advanced FE head model. Yao *et al.*

[302] proposed critical values for von Mises and shear stress of 12 kPa and 6 kPa respectively.

Cumulative strain damage measure (CSDM)

This method was presented by Bandak and Eppinger [164] to evaluate the strain-related damage within the brain. It is based on the finite element human head modelling technique. The idea behind their hypothesis was that it is possible to quantify the mechanical damage to the axonal components of the brain once the responsible state of strain is characterized.

Therefore, a cumulative damage measure, based on the calculation of the cumulative volume fraction of the brain that has experienced a specific level of stretch (maximal principal strain), is used as a possible predictor for deformation-related brain injury such as DAI [322, 424, 439]. For instance, it was postulated that DAI is associated with the cumulative volume of the brain matter experiencing tensile strains over a critical level sometime during the impact. The measure was based on the maximum principal strain calculated from the objective strain tensor that was obtained by integration of the rate of deformation gradient with appropriate accounting for large rotations. This measure was used to evaluate the relative effects of rotational and translational accelerations in both the sagittal and coronal planes, on the development of strain damage in the brain. Bandak and Eppinger [164] used a 3D FE human head model developed by DiMasi [405] and subjected it to various pulses of sagittal and coronal plane centroidal and a noncentroidal angular velocities, as well as translational acceleration pulses in the anterior posterior direction. The authors concluded that translation impulses had little influence on CSDM. In this study, it was also found that the sagittal plane rotation produced a larger CSDM than the corresponding coronal, which contradicts the monkeys experimental results obtained by Gennarelli *et al.* [132, 157]. Vezin and Verriesr [278] found a high value of CSDM to correlate with high values of the longitudinal and vertical linear accelerations. This observation shows that a rotation of the head around the transverse axis associated with a high deceleration in the sagittal plane can lead to DAI.

The CSDM is based on the hypothesis that DAI is associated with the cumulative volume fraction (%) of the brain matter experiencing tensile strains over a critical level. At each time increment, the volume of all the elements that have experienced a principal strain above prescribed threshold values is calculated. The affected brain volume monotonically increases in time during conditions where the brain is undergoing tensile stretching deformations and remains constant for all other conditions (compression, unloading, etc.). Bandak *et al.* [230] found that a CSDM level 5 corresponds to mild DAI and a CSDM level of 22 corresponds to moderate DAI severity, which means that 5 and 22 % represent respectively the brain volume experienced strain in excess to the critical level of 15%, proposed [282]. Bain and Meaney [195] and Morrisson *et al.* [358] presented a local injury criterion for strains in the brain tissue, suggesting that strains in the brain tissue greater than 20% could cause injuries such as DAI. First principal strain of 0.35 proposed as the tolerance threshold for mild traumatic brain injury (MTBI) or concussion was used as strain damage threshold to assess the injury severity [342, 19, 343, 341]. Zhang *et al.* [341] concluded that strain magnitude increased as angular velocity increased and found that for AIS 1 concussion, peak strain in all brain regions was <0.30 while in AIS 2 concussion, large areas had strains >0.35. It is suggested that maximal principal strain higher 30-60% can cause vascular rupture [191, 384]. Kleiven [441] proposed strain brain tissue as a predictor of concussion and DAI suggesting 0.1 and 0.2 as thresholds, respectively.

The CSDM is the most promising stress and strain based injury criterion, since it is based on probably one of the most important parameters in brain injury (strain) and try to relates it with DAI brain damage which was reported to be one of the most expected and severe type of damage [229, 322]. Aare *et al.* [273] concluded that rotational violence to the head results in large shear strains in the brain, which has been proposed as a cause for traumatic brain injuries like DAI, showing the importance of brain strain prediction. Moreover, Kleiven [285] showed that the strain in the brain shows a large sensitivity to the shear properties utilized for the brain tissue. However, all the stress and strain based injury criteria discussed above, are based on a Finite Element model of the head that need to be validated fin order to obtain accurate results.

Dilatation Damage Measure (DDM)

The dilatation damage measure (DDM) is a pressure-based injury criterion and was proposed by Bandak [229]. This mechanical measure evaluates brain injury caused by large dilatational stresses. It is supposed to be a correlate with contusions [322, 424, 439]. The probability of contusion is correlated with the fraction of brain volume where negative pressures can produce damage [278]. DDM monitors the cumulative volume fraction of the brain experiencing specified negative pressure levels. However, no direct observational evidence has been reported on the relationship between pressure mechanisms and the production of axonal, vascular or other soft tissue injury. However, the authors themselves reported that there is no direct observational evidence on the relationship between intracranial pressure and head injury [229]. Therefore, this injury criterion may not be adequate.

Similar to the CSDM calculation, at each time step, the volume of all the elements experiencing a negative pressure level exceeding prescribed threshold values is calculated. Bandak *et al.* [230] suggested a DDM value of 5 at a threshold level of -101 kPa as an injury threshold, but also indicated that further research was necessary.

Bain and Meaney [195] have shown that DAI is a function of distortional strain, and not dilatation (pressure).

Recently, Yao *et al.* [302] proposed critical values for countercoup pressure of -130 kPa.

Relative Motion Damage Measure (RMDM)

The Relative Motion Damage Measure (RMDM) was proposed by Bandak [229] for the evaluation of injury related to brain movements relative to the interior surface of the cranium. RMDM monitors the tangential motion of the brain surface resulting from combined rotational and translational accelerations of the head. Such motions are suspected to be the cause of SDH associated with large-stretch ruptures of the bridging veins [322], due to the brain motion relative to the skull. The rupture tolerance levels of bridging veins are a combined function of both strain and strain rate as reported by Löwenhielm [244]. The bridging veins have been reported by Lee and Haut [269] to have an ultimate strain of about 0.5 in tension, while Löwenhielm [191] reported failure strain values ranging from 0.2 to about 1.0, depending on the strain rate.

The great majority of FEHM does not have bridging veins modelled. Moreover, RMDM does not require the modelling of the bridging veins but rather monitors the relative displacement of several node pairs. Each pair represents a bridging vein tethered between the skull and brain near the parasagittal sinus. The measure accounts for the large-stretch modes of rupture while leaving open the possibility of using other micro or macro rupture-modes associated with more complex vascular tethering states.

The RMDM heavily relies on a correct model of the interface between brain and skull. However, the full mechanical behaviour of the skull-brain interface is not yet known, but some research was already done regarding this matter [231, 232, 233, 143]. If the interface is modelled correctly, the RMDM is potentially a good injury criterion to predict SDH [322, 424].

SIMon Criterion

In both linear and angular loadings, the brain encounters deformations that can damage the neuronal or vascular structures. Numerical models of the head could be useful tools to improve the understanding of brain injury mechanisms. SIMon (Simulated Injury Monitor) proposed by Takhounts *et al.* [277], is one of the models, originally developed by DiMasi *et al.* [167] and later improved by Bandak and Eppinger [164] and Bandak *et al.* [381], that used the last three criteria explained previously, the CSDM, the DDM and the RMDM.

The SIMon FE head model consists of the rigid skull, the dura-CSF layer, the brain, the falx cerebri and the bridging veins, represented in figure 2.36. The brain is a simplified model. The region under the brain and the tentorium was modelled as a continuation of the dura-CSF layer and did not account for either the cerebellum or the midbrain. The skull was assumed to be rigid, whereas the rest of the structures were considered as deformable, linear viscoelastic, isotropic, and homogeneous. The brain was characterised as viscoelastic. Data from animal experiments were used to determine critical values for each injury metric. In order to apply these data, the linear and angular kinematics recorded for the animal's head were scaled in magnitude and time to what a human would experience. These responses were then applied to the rigid skull of the SIMon FEM. The injury metrics were computed from each test and logistic regression was used to establish the critical values.

SIMon proposes three new specific criterion for the three common types of brain injuries: DAI, contusions, and Acute SDH.

The CSDM is a correlation for DAI and assumes that DAI is associated with the cumulative volume of brain tissue submitted to critical tensile strain level. The tolerance limit proposed is 55%.

The DDM is used for injury resulting from dilatational stress conditions. The probability of contusion is correlated with the fraction of brain volume where negative pressures can produce damage.The tolerance limit proposed is 7.2%.

The RMDM is used for injuries related to brain motion relative to the skull, such as the rupture of the bridging veins and has a tolerance limit of 1.

The advantage of the SIMon model is that it assesses brain injury risks based on local rather than global data. Consequently, this tool could provide injury assessments for any impact direction without requiring an adjustment of the injury criteria [278].

These criteria are computed using the intracranial mechanical behaviour simulated by the finite element head model described in Bandak *et al.* [165] and Takhounts *et al.* [277]. The advantage of its simple geometry is of course the short computing duration, which makes the statistical approach simpler.

A limitation of this model is the skull, which is considered as rigid, and the FEM can only be driven by acceleration fields. Even if no skull fracture criterion can be calculated from a loading descriptor computed by the SIMon itself, a criterion named the skull fracture criterion (SFC) is available in the software developed by Takhounts *et al.* [277]. Moreover, the advantage of the relatively short computing duration has a cost, the results are not accurate enough.

Recently, to overcome the issues of this first model, Takhounts *et al.* [424] proposed a new FEHM that is comprised of several parts: cerebrum, cerebellum, falx, tentorium, combined pia-arachnoid complex with CSF, ventricles, brainstem, and parasagittal blood vessels, which are represented in figure 2.37. The model's topology was derived from human computer tomography (CT). It was investigated the potential for traumatic brain injuries (TBI) using this newly developed, geometrically detailed FEHM within the concept of a simulated injury monitor (SIMon), based on the injury criteria described in Takhounts *et al.* [277]. In conclusion, the new SIMon FEHM offers an advantage over the previous version because it is geometrically more representative of the human head, leading to more accurate results. This advantage, however, is made possible at the expense of additional computational time.

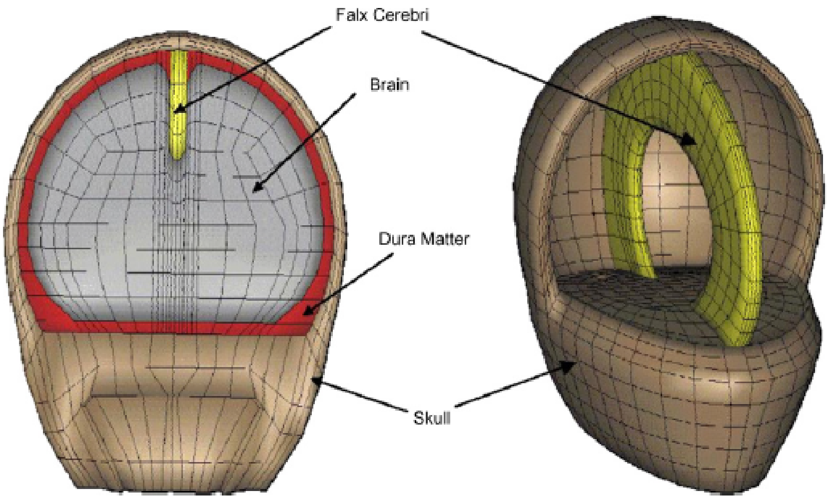


Figure 2.36: *SIMon finite element head model [277].*

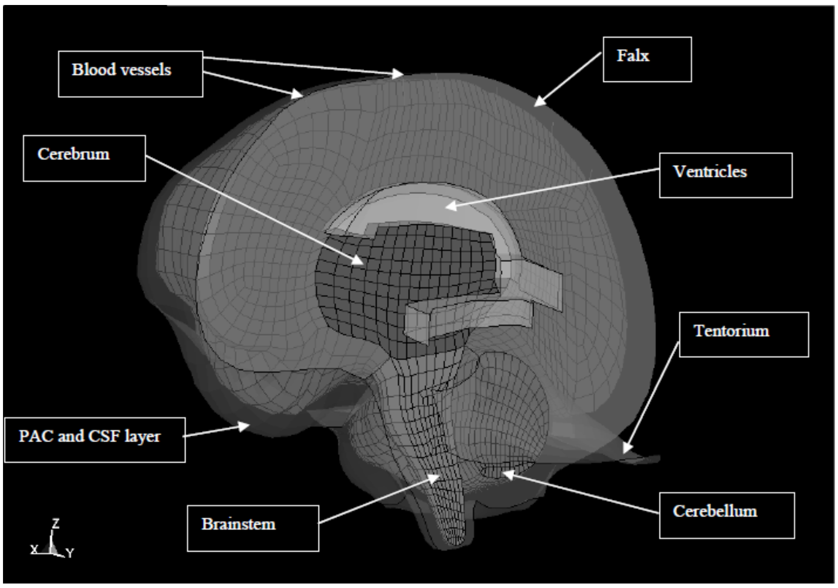


Figure 2.37: *New SIMon FEHM [424].*

ULP Criteria

Marjoux *et al.* [322] carried out a study with the objective of assess and compare the injury prediction capability of the HIC, HIP as well as the criteria provided by the SIMon FEHM and the ULP FEHM. To do that, the ULP three-dimensional FEHM represented in figure 2.38, proposed by Kang *et al.* [220] and detailed in Willinger *et al.* [321] was used.

As described in Willinger and Baumgartner [320], three injury criteria are computed with this model:

- The maximal von Mises stress value reached by a significant volume of at least 10 contiguous elements (representing about 3 cm^3 of brain volume) from the brain is proposed as a correlate to neurological injury occurrences. Von Mises stress was preferred to strain for empirical reasons: it was shown to be better correlated to neurological injuries in previous studies. Other authors like Anderson [366] made the same conclusions. Another reason is mathematical: von Mises stress is a frame independent scalar (such as pressure) whereas strain depends on the orientation of the frame. Marjoux *et al.* [322] for a moderate and severe neurological injury this von Mises stress value is about 27 kPa and 39 kPa respectively. More recently, Deck and Willinger [433] updated these tolerance limits to 28 kPa and 53 kPa, respectively;

- The maximum value reached by the global internal strain energy of the elements modelling the space between the brain and the skull is proposed as a correlate to subdural haematoma occurrences. This value represents the integral of the $\sigma \times \varepsilon$ product among the whole space between the brain and the skull. It is a way to quantify the energy absorbed by this space. Marjoux *et al.* [322] found the maximum value reached by the global strain energy of the subarachnoidal space is proposed as a correlate to subdural haematoma occurrences with a value about 4211 mJ. More recently, Deck and Willinger [433] updated this tolerance limit to 4950 mJ;

- The maximum value reached by the global internal strain energy of the deformable skull is proposed as a correlate to skull fracture occurrences. This criterion is only computed for the pedestrian cases where the deformable skull FEHM is driven with a direct impact. This value aims to quantify the energy absorbed by the skull. Marjoux *et al.* [322] found an internal energy of the skull of 833 mJ;

More details about the ULP/SUFEHM criteria are reported by Deck and Willinger [433]. From the results obtained by Marjoux *et al.* [322], the ULP FEHM based criteria seem to have the best prediction capability for each type of injury in comparison with SIMon head model [277]. This is particularly true concerning the neurological injuries since the injury criterion based on the peaks of von Mises stress keeps its accuracy even when predicting the moderate neurological injuries.

In the same study, the results obtained with SIMon based criteria were bad. Marjoux *et al.* [322] justified these results with the simplicity of this model (the first SIMon model), where the ULP model geometry seems closer to the real anatomy of the head and, therefore, the computed intracranial mechanical parameters could be more realistic. This was also suggested by Franklyn *et al.* [367] that computed CSDM with the WSU (Wayne State University) model that is much more accurate than the SIMon model.

Regarding skull fracture, Marjoux *et al.* [322] concluded that ULP criterion and the SFC of SIMon software, which is based on a single-mass head model, leads to comparable results.

More recently, Deck and Willinger [345, 433] proposed a rational approach in order to evaluate the ability of head models to predict brain pressures and strains by using a statistical approach, where it was proposed DAI criteria using SUFEHM head model (also known as ULP head model):

- Brain von Mises stress of 28 kPa for mild DAI and 53 kPa for severe DAI;

- Brain von Mises strain of 30% for mild DAI and 57% for severe DAI;

- Brain First principal strain of 33% for mild DAI and 67% for severe DAI.

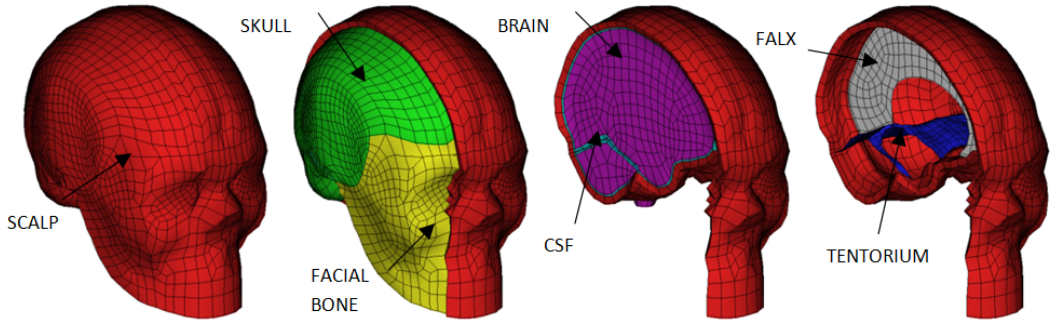


Figure 2.38: *Strasbourg University Human Head FE Model developed by Kang et al. [220].*

These predictors were used by Chatelin *et al.* [434] by integrating it in head trauma simulation by computing axonal elongation for each finite element of the brain model in a post-processing of classical simulation results. Axonal elongation was selected as computation endpoint for its strong potential as a parameter for DAI prediction and location.

2.2.5 Computational modelling - Headform and head modelling

Over the past few years, Finite Element Models of the human head have been powerful tools used to understand and to predict the head's response under impact conditions. This allowed a further accurate, computational-based prediction of brain injuries and relating it to medical investigations observed in autopsies of corpses involved in real accidents [220].

Actually, there are two different types of heads used in the impact simulation, the headform and the head model. Nowadays, with the huge development of the computational technology, the efforts of biomechanical researchers are concentrated in the development of feasible head models, that are capable of reproduce the injuries in an impact simulation. Also, in order to apply and to compute the methods described in the section 2.2.4, a advanced human head model is needed.

Therefore, headform modelling corresponds to the numerical model of the headform as described by either helmet testing standard headform or the numerical modelling of the physical head-form of the crash dummies such as hybrid II or III [376]. On the other hand, head modelling is devoted to the biomechanical head numerical modelling.

Shuaeib *et al.* [67] concluded that from helmet design point of view, knowledge of both models is required to serve certain purposes. For example, the headform model is required to simulate helmet standard drop tests, while biomechanical head model could be utilized to judge the validity of the drop test conditions, and also simulate real accident injury characteristics. However, from the biomechanical point of view the results obtained with a FEHM allows a further prediction of head injury.

Headform modelling

Helmet testing headform is relatively easy to model and detailed data that describes the headform geometry and material types is usually available in most of the helmet standards [8, 63]. However, standards usually does not provide all the required modelling details and some approximations are required. Nevertheless, these approximations must be within a certain limits.

However, physical headform modelling is more complicated, as the one developed by Delille *et al.* [373]. Some of these physical headforms are already modelled and available to use in some FE software such as LS-Dyna or Pam-Crash [368, 369], being very useful in helmet crash studies [370, 371]. An example of these dummies is the Hybrid III dummy head (HIII head) developed by Fredriksson [376].

Actually, there are several types of headforms. Basically, it is possible to divided them in single mass headforms and multi-mass headforms. The last ones are usually more advanced headforms that may provide better biofidelic data. One such headform, is the B150 headform that has been developed by Willinger *et al.* [382]. The single mass headform represented in figure 2.40 is usually used by the current standards, based on the ISO DIS 6220 standard [457].

The B150 headform consists of the following components: skin layer, aluminium skull shell, Hybrid III dummy headform mounting, steel sleeve, contact plug, brain mass and cylindrical cushion, as represented in figure 2.39. The skull shell, which is covered by a viscoelastic skin layer, is rigidly fixed to the Hybrid III dummy headform mounting. Secured to the top of the Hybrid III mounting is a steel sleeve, which fits around and secures the lower end of a flexible polymide contact plug. Fitted around the upper end of the contact plug is a steel block representing the brain mass. The flexible contact plug allows motion of the brain mass relatively to the skull during impacts or high accelerations of the outer skull shell, so representing the expected response of the real skull and brain system. Between the steel sleeve and the brain mass, a cylindrical cushion is fitted in order to damp the relative motion between the brain and skull. Accelerometers are fitted to both the brain and skull in order that the independent motions of these structures can be measured.

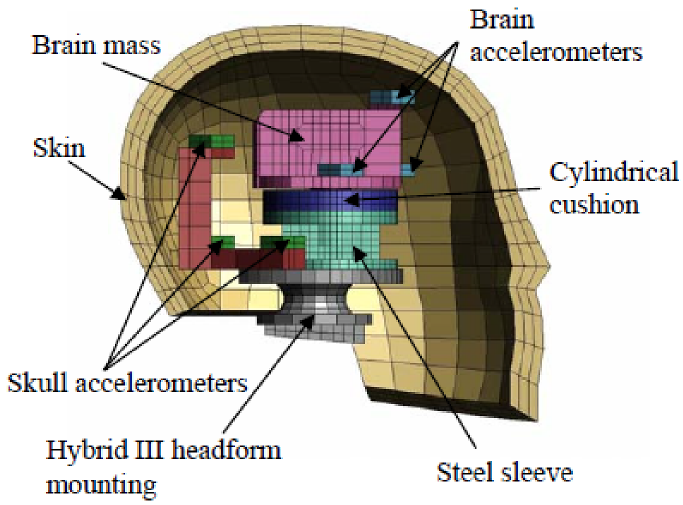


Figure 2.39: *Sagittal sections of B150 headform model [380].*

Neale *et al.* [380] carried out a study where this headform was compared to a single mass one and the results from this investigation suggest that rigid single mass headforms do not provide representative biofidelic responses under short duration impacts that justifies their used over the single mass headforms. Nevertheless, the B150 headform model demonstrated trends matching those predicted by a human head model.

Other models include deformable brain tissue mimicking materials such as water, oil or silicone gel. The currently used headforms may not properly mimic the response of the human head correctly in helmeted head impact because rigid headform does not model the flexibility of the human brain and skull, which may be inadequate to model the helmet-head interaction correctly. To overcome such deficiency, an anatomically more detailed, deformable headform was developed by van den Bosch [36] that allows more realistic helmet-head interaction. This

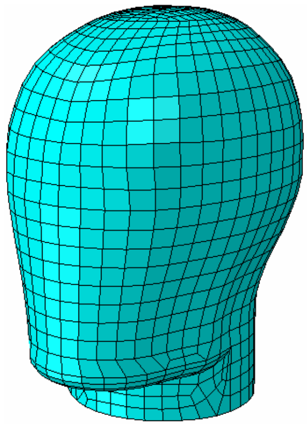


Figure 2.40: *Single mass headform model.*

anatomically detailed headform is represented in figure 2.41.

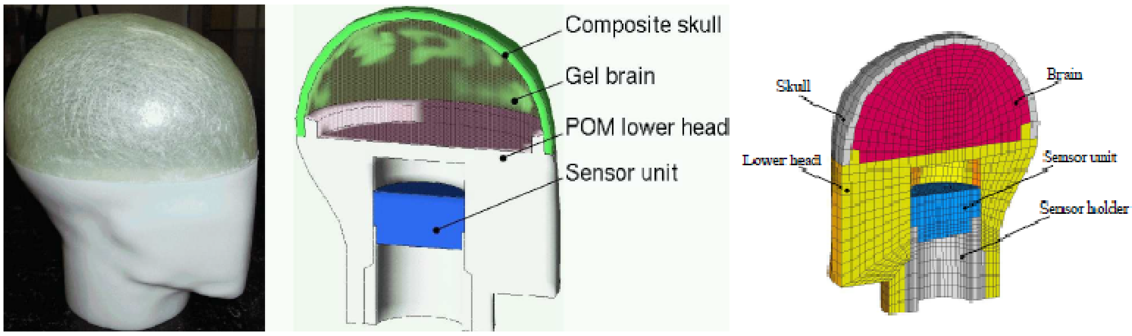


Figure 2.41: *Anatomically improved headform. Left: Real model. Middle: Schematic cross-section. Right: FE headform model [36].*

Finite Element Human Head modelling

As previously seen, it is relatively easy to model a standard test headform for impact simulation but the actual human head modelling for the same purpose is still under gradual progression. The head constituent materials are biological materials which commonly are anisotropic, inhomogeneous, nonlinear, viscoelastic and are different for each individual. In addition, there is a great variability between different individuals, which means different properties for each person.

During the energy transfer process, the head can deform under the influence of some force, it is not rigid as a headform, which lead many researchers to the head modelling and its respective components and materials. Moreover, optimisation against headform's response and human head's response does not lead to the same results as shown by Deck *et al.* [6] and also affirmed by Zhang *et al.* [379]. This was also indicated by Aare *et al.* [273], Kleiven [443] and Tinard *et al.* [7], where more realistic results were obtained with a FEHM over a headform. Thus, in general, researchers agree that helmets should be optimized according to biomechanical criteria using FEHM.

Also, one of the issues for head impact that has attracted the attention of researchers, was what type of impact loading is responsible for producing any of the previously mentioned brain injuries (section 2.2.2). Other advantage of these models is that they can be used for detailed reconstruction of injurious events and correlating reconstruction information to medical data, as done by Willinger *et al.* [279], King *et al.* [19] and Marjoux *et al.* [322].

FEHM review

Creating a FEM model starts by creating a topological description of the structure's geometric features, which can be in either 1D, 2D or 3D form, modelled by line, shape, or surface representation, respectively, although nowadays 3D models are predominantly used. Only 3D FE models of the human head are relevant for most impact and inertial load analysis. Due to the low shear resistance and large bulk modulus, material exchange between regions is likely to occur (CSF, foramen magnum etc.), when subjected to large deformations [121]. This can only be described from a 3D point of view. However, 2D models are useful for parametric studies of controlled planar motions.

Indeed there is great interest for the use of the FEM in head injury research. The development of the biomechanical head models had attracted the attention of the biomechanical researchers since a long time ago. At the beginning, most of the work was either experimentally modelled or approximated by a standard shapes such as spheres or spheroidals in order to be able to apply mathematical theories [452, 453, 386, 387, 388, 389, 454, 455]. Recently, with increasing computational technologies, research on head modelling had evolved tremendously. An example of the evolution of the first human head FE models can be observed in the work performed by Haug *et al.* [389] and a review of the latest models is presented by Raul *et al.* [414].

Over the last years, several models have been proposed. One of the first 3D models was developed by Ward and Thompson [413] in 1975 to reproduce the experimental tests carried out on cadaver heads . This model was constituted by the brain, dura mater, falx and tentorium membranes, rigid skull and CSF. The membranes and the foramen magnum were also modelled.

In 1977, Shugar and Kahona [403] developed a 3D model based on a previous 2D model developed by his work group in 1975 [159]. This model incorporated a thin layer to represent the subarachnoid space between the skull and the brain. In 1980, Hosey and Liu [404] developed a FEHM with the aim to show the cavitation phenomenon in the contrecoup area. In 1991, DiMasi *et al.* [405] created a 3D FEHM in order to simulate damped and undamped impacts with accelerations varying from 165 to 302 g. This model was a first step in the comprehension of brain injuries in car crashes [414]. In 1992, Mendis [406] developed a FEHM in order to analyse brain stresses and strains during a rotational acceleration of the head and to correlate the axonal injury intensity observed experimentally on primates by Gennarelli and Thibault [139] with strains predicted in the brain.

Ruan *et al.* [415] in 1992 illustrated the typical contrecoup phenomenon with a new FEM based on Shugar and Kahona's model [403]. Over the last few years, several versions of the Wayne State University head injury model (WSUHIM) were developed. WSUHIM version I (1993-1997) simulated essential anatomical compartments of the head [398, 449]. By differentiating the material properties of grey matter from white matter, the model was capable of predicting the location of DAI in the brain. It was used to predict the directional sensitivity of the brain to impacts from varying directions. The model has been continuously improved by Zhou *et al.* [169], where more head features were developed, including ventricles and gray and white matter were differentiate. Later, the model was revised and upgraded to WSUHIM version II (1998-1999) by introducing a sliding interface between the skull and brain surface. More recently, the model was used by Zhang *et al.* [281] in it's final form and it was suggested that the intra-cranial pressure is largely a function of the translational acceleration of the head, while the maximum shear stress is more sensitive to rotational acceleration, where the aim was to develop a model capable of simulating direct and indirect impacts over a wide range of impact severities. This model features fine anatomical details including the scalp, skull with an outer table, diploë and inner table, dura, falx cerebri, tentorium, pia, sagittal sinus, transverse sinus, CSF, hemispheres of the cerebrum with distinct white and gray matter, cerebellum, brainstem, lateral ventricles, third ventricles, and bridging veins. Moreover, this model included a facial model that consists of 14 facial bones, nasal cartilage, temporal mandibular joint, ligaments, soft tissue and skin [281]. Concerning the mechanical properties of this model, the brain behaviour was characterised as viscoelastic and an elastic-plastic material model was used for cortical and cancellous bones of the face. This FEHM was used to reconstruct 53 cases of sport accidents where there were 22 cases of concussion by King *et al.* [19] and it has been subjected to rigorous validation against available cadaveric intracranial and ventricular pressure data, the relative displacement data between the brain and the skull, and facial impact data [281, 343].

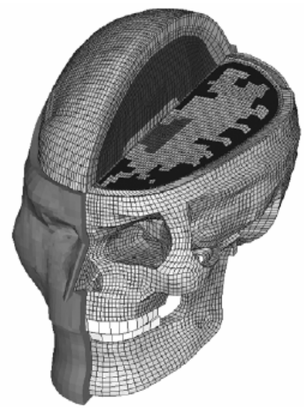


Figure 2.42: *Wayne State University Head Injury Model [277].*

In 1999, Al-Bsharat *et al.* [232] thereafter modified the model presented by Zhou *et al.* [169], by introducing a sliding contact definition between the CSF and the pia matter in the subdural space of the brain. In this way, the sliding of the brain within the skull could be simulated.

In 1997, Claessens *et al.* [221] developed a three dimensional head model using the Visible Human Project, resembling the study developed by Kuijpers *et al.* [456], although the latter was based in a two dimensional model. Two different approaches were used for the skull brain interface. A coupled interface, allowing no relative motion between the skull and the brain, and a sliding contact condition allowing for separation. This model was validated by being subjected to a frontal impact as in the experiments of Nahum *et al.* [192]. The results obtained by Claessens *et al.* [221] contrasted with those obtained by Kuijpers *et al.* [456]. The no-slip model showed good agreement with the experiments for the coup pressures, while the sliding interface model did not. On the other hand, these two studies agreed when it came to the contrecoup pressure, which not even the 3D study was able to mimic.

In 2002, the model developed by Claessens *et al.* [221] was completely transformed by Brands *et al.* [417, 418]. This model is shown in figure 2.43. The anatomical structures included in the current Eindhoven model are grouped into three components: the cranium, the meningeal layers and CSF and the brain tissue. All structures were assumed to be rigidly connected to each other. The objective of this model was to study the effect on non-linear material behaviour on the predicted brain response. Their conclusions were that the pressure response remained unaffected by application of non-linear behaviour; the pressure gradient being completely determined by the equilibrium of momentum and, thus, independent of the choice of the brain constitutive properties. This model was validated accordingly to cadaver experiment by Nahum *et al.* [192].

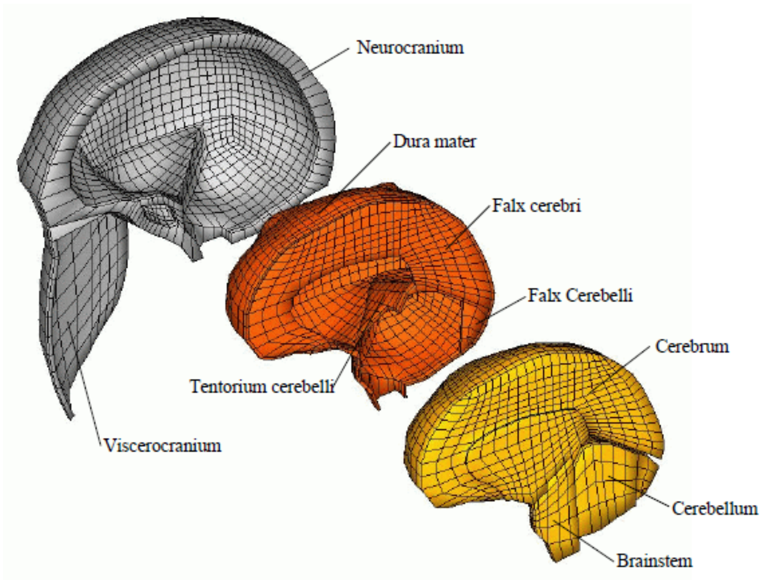


Figure 2.43: *FEHM developed by Claessens et al. [221] and transformed by Brands et al. [418].*

In 1997, Kang *et al.* [220] developed the Université Louis Pasteur (ULP) FE human head. The geometry of the inner and outer surfaces of the skull was digitised from a human adult male skull. This model includes anatomical features such as the scalp, a deformable skull, the face, the dura matter (including falx and tentorium), the subarachnoidal space, cerebral hemisphere,the brain, the cerebellum and brainstem as shown in figure 2.38. Recently, this FEHM started to be known as the Strasbourg University Finite Element Head Model (SUFEHM). For the CSF, a Lagrangian formulation was selected and the brain-skull interface was modelled by an elastic material validated against the in-vivo vibration analysis. Material properties of the CSF, scalp, facial bones, tentorium and falx are all isotropic and homogenous. The brain was assigned viscoelastic properties from Khalil and Viano [419]. This model continued to be developed by being validated against cadaveric experiments by Willinger *et al.* [321, 400] with regard to experimental tests [192, 409, 255, 233]. More details about the validation of this model were described by Deck and Willinger [433]. Also, more studies were done relating the model development and validation [279, 390, 280]. Moreover, tolerance limits were identified by Willinger and Baumgartner [280] and Marjoux *et al.* [322] through the replication of real world accidents. Their study established human head tolerance limits relative to DAI, SDH and skull fracture with a risk of occurrence of 50% and found that von Mises stress in the brain, the strain energy of the CSF and the strain energy of the skull are the best predictors of diffuse axonal injury, subdural haematoma and skull fracture respectively. Thus, this model constitutes a good basis for prediction of head injuries. FE head models have a good potential to predict DAI, since they describe local deformations within the brain [223, 351]. However, a well-defined correlation between mechanical loading and DAI using FE head models has not been achieved yet [348]. A possible contribution to this is that the gyri and sulci in the brain, which are not included in the actual FE head models, can play an important role in the local tissue deformations [410, 411, 412]. Ho and Kleiven [411] suggested that the inclusion of sulci should be considered in FE head models as it alters the strain and strain distribution in an FE model.

Later, Deck *et al.* [395] developed in detail the skull complex geometry including skull reinforced beams and thickness variation validated against existing head impacts involving skull fracture as well as a linear and a non-linear brain constitutive law which permit an accurate validation against brain deformation under impact. This new skull was incorporated in the ULP model, which was coupled with the cerebrum, cerebellum, falx, tentorium and the brain stem meshing taken from the ULP model [321] and is shown in figure 2.44. Concerning the brain mechanical properties, the authors improved two new laws based on original experimental tests by Nicolle *et al.* [420], focusing on high strain rates and non-linear behaviour in order to investigate the brain material properties' influence in the head model validation procedure against existing experimental brain deformation.

Nevertheless, more recently, the SUFEHM model was also validated through replicating two cadaveric experiments to guarantee that the conversion did not reduce the capability of the model to reproduce experimental data [89]. Recently, it was investigated the influence of the presence of the body in helmet oblique impacts by Ghajari *et al.* [352], where the SUFEHM was coupled to a body. It was found that the presence of the body considerably influences the intracranial response.

Actually, this model could be considered the state of the art of FE human heads. The potential of this model in several applications are tremendous. It could be applied to predict new injuries, to establish thresholds or even to optimize protective head gear, such as motorcycle helmets. This was recently done by Tinard *et al.* [7], by improving a motorcycle helmet regarding the injuries risks, using SUFEHM as head model.

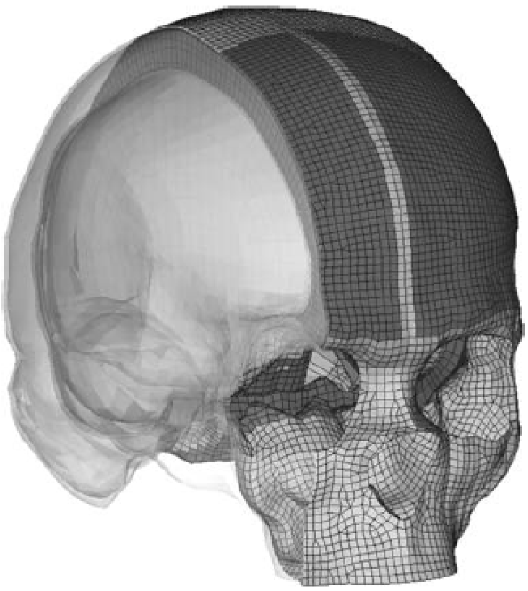


Figure 2.44: *ULP head model side view showing the sagittal reinforcement beam and the varying thickness of the skull developed by Deck et al. [395].*

The Kungliga Tekniska Högskolan (KTH) human head model represented in the figure 2.45, was developed by Kleiven [391, 121] and comprises nonlinear viscoelastic, incompressible material modeling and experimental validation. It is a detailed and parameterised FEM of the adult human head including the scalp, skull, brain, meninges, CSF and 11 pairs of parasagittal bridging veins. A simplified neck, including the extension of the brain stem to the spinal cord, dura mater, spine and muscle and skin, was also modelled. This homogeneous, isotropic, non-linear and viscoelastic constitutive model was based on the work by Mendis *et al.* [421]. In addition, dissipative effects are taken into account through linear viscoelasticity by introducing viscous stressee that are linearly related to the elastic stress. This model has been validated against experimental pressure data, as well as relative motion magnitude data [394, 143]. In 2007, Kleiven [363] compared various predictors for mild traumatic brain injuries, reconstructing real world accidents. Ho and Kleiven [438] studied the influence of the inclusion of the vasculature in the KTH model by modelling a set of blood vessels (the major veins and arteries) and concluded that it could be useful when studying acute SDH, since ruptures can be predicted by measuring the strain directly in the blood vessels. However, the vascular inclusion is not necessary in prediction of other brain injuries. Later, as already referred, Ho and Kleiven [411] studied and suggested that the inclusion of sulci should be considered in FE head models as it alters the strain and strain distribution in an FE model.

This model was experimentally validated against pressure data (intra-cranial pressure experiments) [196] as well as relative motion magnitude data [143]. Also, a comprehensive correlation between the FE model output and the relative motion between human cadaver brain and skull has been demonstrated for three impact directions [394]. More recently, it was also validated against intra-cerebral acceleration experiments [337] and skull fracture experiments [397].

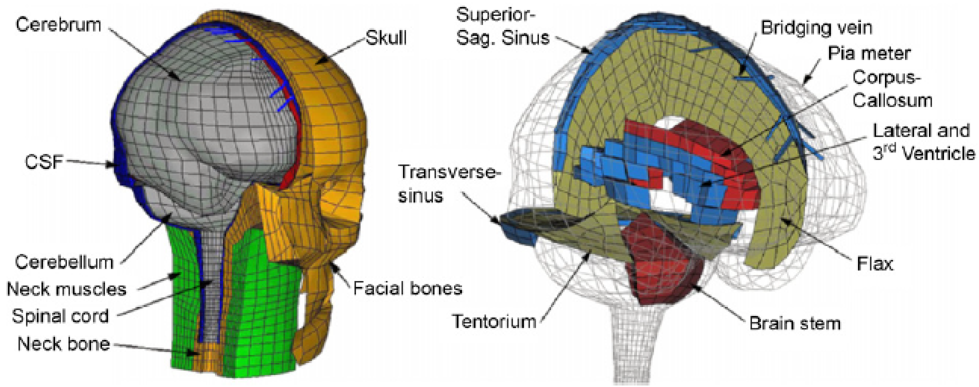


Figure 2.45: *Finite element head model developed by Kleiven [337].*

This detailed FEHM together with the SUFEHM and WSUHIM represent - at the point this work is written - the state of art for FE human head models.

Considering the increasing computational power that allowed the development of such models, other techniques revealed to be very useful. Examples are the Computer Tomography (CT), Magnetic Resonance Tomography (MRT) and sliced color photos. These were used by Horgan and Gilchrist [377] to develop a three-dimensional finite element, the University College Dublin Brain Trauma Model (UCDBTM), based on the same techniques. The same authors have improved this model [422]. Hereby, a realistic skull thickness variation was achieved in the different locations. This is an important feature as the variation in skull thickness is significant, extending from the thick and porous frontal bone to the thin temporal bones. A comparison between different mesh densities showed that a coarsely meshed model is adequate for investigating the pressure response of the model, while a finer mesh is more appropriate for detailed investigations [377]. The model was validated against intracranial pressure data from Nahum *et al.* [192] cadaver impact tests and brain motion against Hardy *et al.* [233] research. Further validations accomplished comparing real world brain injury events to the model reconstructions with good agreement [429].

In the same year, Takhounts *et al.* [277] proposed the simulated injury monitor (SIMon) FEHM based on the model originally developed by DiMasi *et al.* [167] and later improved by Bandak and Eppinger [164] and Bandak *et al.* [381]. Recently, Takhounts *et al.* [424] proposed a new geometrically detailed FEHM, more detailed than the first one, in order to obtain better results at the expense of additional computational time. Further details about this model were previous described (section 2.2.4).

Three dimensional models are predominantly used mostly in impact and inertial load analysis. However, two dimensional models are useful for parametric studies of controlled planar motions. Darvish *et al.* [374] proposed a two dimensional head model that included simplified representations of the skull, CSF, brain, a part of spinal cord and bridging veins as shown in figure 2.46. In this study, a simplified two-dimensional head was modelled, giving more emphasis to the material properties, suggesting the use of experimentally determined material properties rather than simplified material properties used for computational convenience in finite element models that can significantly alter the response of the brain.

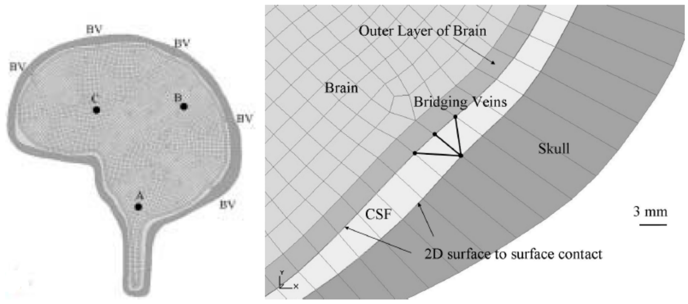


Figure 2.46: *Two dimensional Finite element head model developed by Darvish [374].*

A finite element model of the human head was developed by Cardamone [435], that includes the following the skull without facial bones, CSF, gray matter, white matter, cerebellum, corpus callosum, telencephalic nuclei, brain stem, ventricles, as shown in figure 2.47. A partial validation of this model has been carried out by El Sayed *et al.* [436] by comparing numerical predictions of intracranial pressure with the outcomes of an experiment on a human cadaver impacted by a rigid mass.

Kim *et al.* [425] developed a high-resolution human head/brain finite element model and used for the injury analysis. The model is reconstructed from computed tomography scans. The model consists mainly in skull, CSF and brain, showed in figure 2.48.

Belingardi *et al.* [451] developed a new FEHM. The geometrical characteristics were extracted from CT and MRI scanner images. The model was validated by comparing the numerical results and the experimental results obtained by Nahum *et al.* [192]. This model is composed by a brain, ventricles, dura mater, falx and tentorium membranes, CSF, skull and facial bones where compact bone and cancellous bone were modelled and scalp. However, the entire model was modelled with elastic properties, what could justify some differences from the experimental results.

In 2006, a three dimensional model of the head-neck complex has been developed by Kimpara *et al.* [423] with a detailed description of the brain and the spinal cord. The proposed model can be used to simulate the biomechanical behaviour of the entire central nervous system at the same time. For the authors, the brain-spinal cord model was useful to investigate the relationship between the restraint conditions and the central nervous system

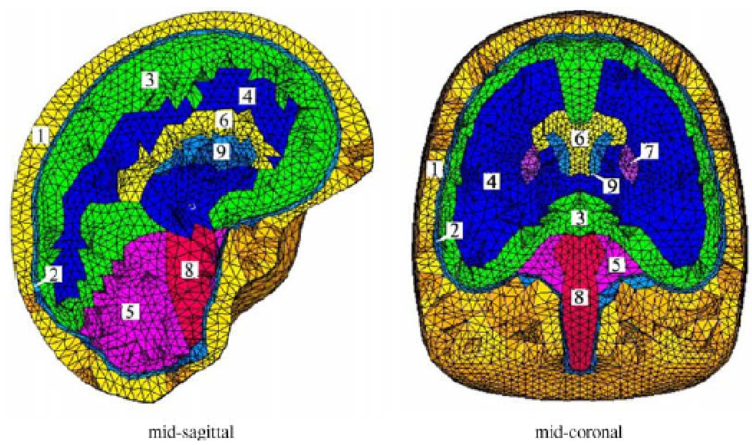


Figure 2.47: *Mid-sagittal and mid-coronal sections of the adopted head finite element model: (1) skull without facial bones; (2) CSF; (3) gray matter; (4) white matter; (5) cerebellum; (6) corpus callosum; (7) telencephalic nuclei; (8) brain stem and (9) ventricles. [436].*

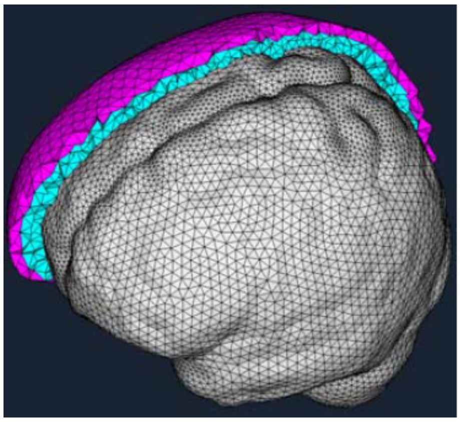


Figure 2.48: *FEHM developed by Kim et al. [425]*

injuries.

Yao *et al.* [302] presented a refined FE head model that includes the main anatomy structures of the head. To investigate the brain injury mechanisms, a head/brain model was developed based on the HUMOS2 head model which only consists of scalp, skull and a simplified brain as shown in figure 2.49. The refined head model includes more anatomy structures such as CSF, meninges, cerebral, cerebellum, brain stem, falx and tentorium. The FE head model was validated against Nahum's cadaver test [192].

Iwamoto *et al.* [375] proposed a finite element model of a mid-size adult. The skull model includes cortical bone modelled and spongy bone. The head also includes the brain model, the cerebrum, cerebellum, brainstem with distinct white and gray matter and CSF. Additionally, it was modeled sagittal sinus and the dura, pia, arachnoid, meninx, falx cerebri, and tentorium, as shown in figure 2.50. This head was developed to incorporate the Total Human Model for Safety (THUMS), a finite element model of the entire human body. The model was validated for head-neck motions in flexion-extension, lateral bending and rear end impact [431]. Nevertheless, THUMS was also tested with SUFHEM, where the results of the coupling were promising [432].

Motherway proposed a 3D FE model of the skull-brain complex includes scalp, a 3-layered

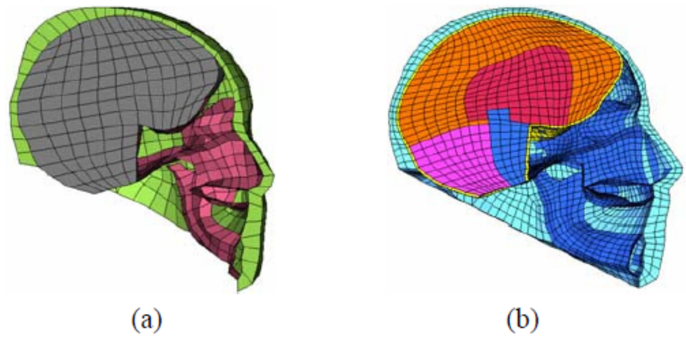


Figure 2.49: *Section view of (a) HUMOS2 head model and (b) refined head model (cut at the sagittal plane) [302].*

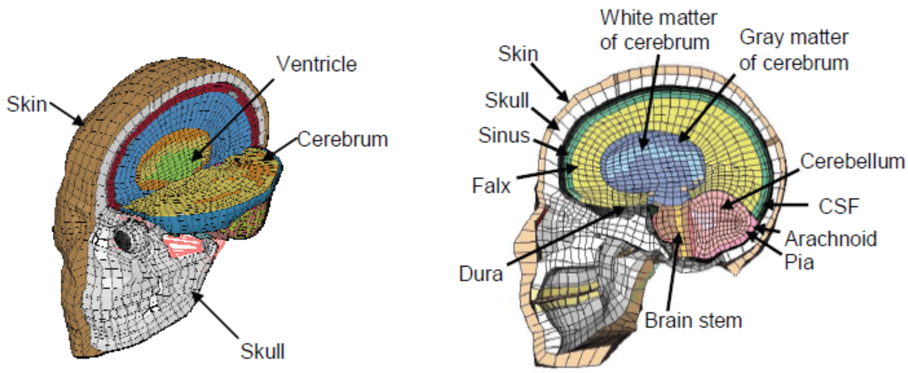


Figure 2.50: *THUMS model [430].*

skull (outer and inner tables, diploe), dura, CSF, pia, falx, tentorium, cerebral hemispheres, cerebellum and brain stem, represented in figure 2.51. The geometry of two human cadavers was determined by CT, MRI and sliced colour photographs. This model was used to assess the applicability of FEM in accident and forensic reconstruction, where the results showed that FEM could provide useful knowledge.

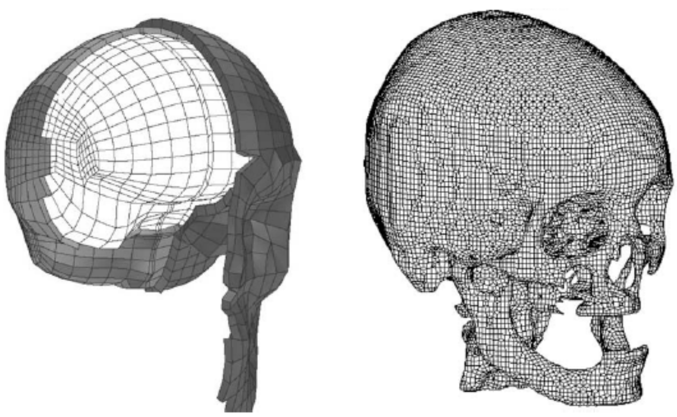


Figure 2.51: *Cutaway view (left) and surface view (right) of finite element head model [426].*

Recently, Dirisala *et al.* [444] developed a detailed FE head model, with almost all the components of a real human head. The geometrical data was based on the work of Horgan and Gilchrist [445] was employed in the development of the FE human head model. The model represented in the figure 2.52, has a skull with variable thickness, facial bone, all membranes such as dura mater, pia mater, falx and tentorium, the CSF and at last the brain. However, the realistic behavior of biological tissue is anisotropic, nonhomogenous, nonlinear, and viscoelastic while, all components were assumed to be isotropic, homogenous and linear. Moreover, the model is still not validated.

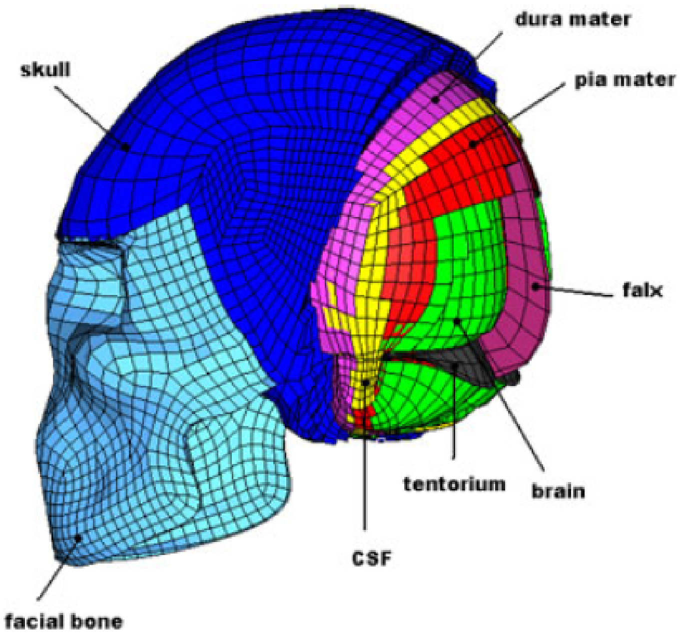


Figure 2.52: *Detailed FE human head model developed by Dirisala et al. [444].*

2.3 Helmet Safety Standards

Motorcycle helmet standards were created after the widespread introduction of the motorcycle and the first biomechanial studies suggesting that use of motorcycle helmets should be mandatory.

To evaluate the protective performance of helmets against head injuries, the helmet standards have been established in many countries. Some standards are regulated by governments such as European and American but in other countries they are issued by private organizations. Almost all standards are different from each other but are similar in their primary goal which is assessing the helmet impact energy absorbing capability. These standards prescribe a number of tests to ensure that the helmets satisfy the safety requirements.

Some standards also evaluate issues like comfort, ventilation, weight, fit, cost, appearance and availability. Because it is impossible to create a helmet for all impact conditions, designers have to create a helmet capable to resist to a higher number of situations possible.

Actually, all motorcycle helmets available in the market are designed, manufactured, and tested to meet standards. Therefore, the performance tests required by any standard induce the helmet design, the measured performance of the helmets in laboratory testing and therefore accident performance as well. Nevertheless, none of the standards are meant to precisely replicate the threats that a motorcyclist may see in a crash. This is justified by the need for reliability and repeatability in the testing environment.

Nowadays, it is well known that helmets substantially reduce head injury, being safer to a motorcyclist to wear a helmet rather than none. Nonetheless, today helmets are designed to reduce headform deceleration and not optimised to reduce head injury [6, 273, 443, 7].

The mass and size of the headforms specified by the standards are nearly the same. For instance, the dimensions of the ECE R22.05 and the latest version of Snell [63] headforms are based on the ISO-DIS-6220 standard [457] (figure 2.53), with their mass increasing with their size. Their dimensions are given by each standard. These standards test headforms comprises the entire head rather than the partial headform employed by DOT FMVSS-218 [459].

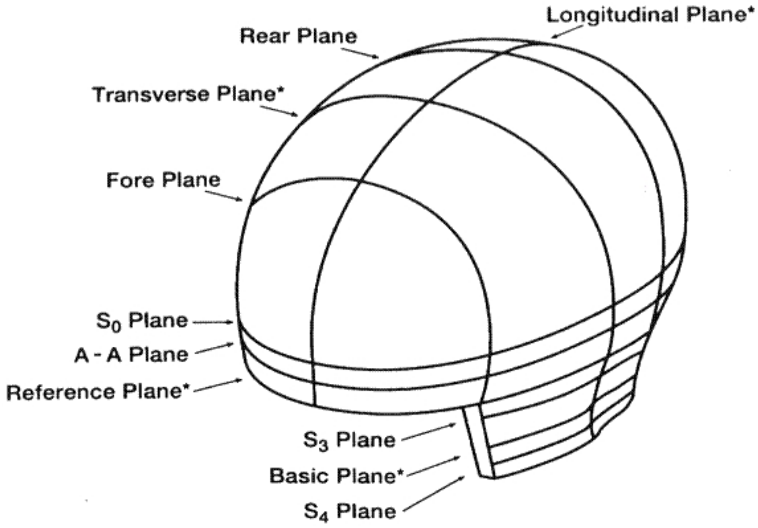


Figure 2.53: *ISO Head form - ISO DIS 6220-1983 [457].*

The head impact speed is an important variable in helmet impact study. Current standards impact speeds range up to 7,75 m/s although higher velocities are achieved riding a motorcycle. Nevertheless, the perpendicular impact speed of the helmet is usually not the same as the riding speed. When a motorcyclist falls, the impact is commonly oblique, which means that the impact speed is decomposed in two components, the perpendicular to the road surface and the tangential to the road surface. This range of impact speed used by the motorcycle helmet standards in their energy absorbing tests includes the velocities that are more common in real life [56]. It is also important refer that the tangential component is not assessed by current standards.

No helmet designed to a particular standard or standards can provide the maximum protection in all types of crashes and no helmet can protect the wearer against all impacts. There are many types of motorcycle helmets. Full-face helmets compared to the open-face types offer additional protection because covers the entire head. However, this additional protection comes with a weight penalty.

2.3.1 Common standard tests

Almost all the standards follow the same concepts in evaluating the effectiveness of the helmets during accidents, which are:

- the helmet has to be able to absorb enough impact energy;
- it has to remain on the head during the accident;

- it must resist to penetration.

The standard tests that a helmet needs to pass to fulfill the minimum performance requirements are described next.

Helmet Conditioning

Prior to all the tests, the helmet must be conditioned, by being exposed to these conditions:

- Ambient temperature and hygrometry conditioning;
- Low temperature conditioning;
- High temperature conditioning;
- Ultraviolet irradiation conditioning and moisture conditioning.

Other standards, such as Snell M2010 and DOT expose helmets to wet conditioning instead of UV radiation conditioning.

Shock absorption Test

The shock absorption test is designed to ensure that helmets retain structural integrity and attenuate impact energy during a variety of crash scenarios.

In all standards, tests are performed in a specially designed test rig. The helmets are dropped by gravity in a guided free fall accelerating the helmet until a required speed. By varying the drop height and the weight of the magnesium headform inside the helmet, the energy level of the test can be easily varied and precisely repeated. Headform should not get damaged and should not absorb energy (should be rigid) so that the test results are reproducible. During the test, the acceleration is measured and recorded thanks to a built-in triaxial accelerometer positioned at the centre of mass of the headform that precisely records the headform response.

After the helmet is dropped, it smash onto a fixed steel anvil. The helmeted headform fall is guided by either a steel track or a pair of steel cables. That guiding system adds friction to slow the fall slightly, so the test technician corrects by raising the initial drop height accordingly.

At the end, performance criteria, such as the PLA and HIC, dependent on the acceleration values are used to quantify the impact severity of the helmet.

Penetration Test

The penetration test simulates a head impact with a sharp object. This test is conducted by dropping a conical striker in guided free fall, with its axis aligned vertically, onto the outer surface of the complete helmet when mounted on a headform. This test evaluate the helmet's ability to resist an impact with objects which cause localised loads, possibly leading to penetration of the helmet and head.

This test has been criticized by Hume *et al.* [64] since the frequency of motorcycle accidents involving pointed objects is extremely small and this test causes the outer shell of the helmet to be excessively thick which results in a heavy weight helmet. Otte *et al.* [65] conducted statistical study and his findings supported the conclusions of Hume *et al.* [64].

The ECE 22.05 does only require such test to the visor, where a 3 kg hammer is dropped from a height of 1 m on a 0.3 punch striker. This striker must be stopped no less than 5 mm above the headform.

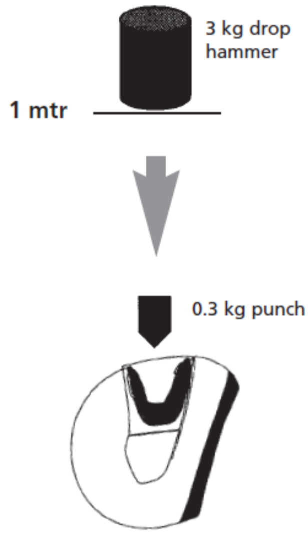


Figure 2.54: *ECE 22.05 penetration visor test (Adapted from [458]).*

Retention Test

The retention system test is done in order to ensure that the helmet remains securely fastened to the rider head. It is conducted by applying a tensile load to the retention assembly. There are two types of retention test, the dynamic test and the detaching test.

Dynamic test of the retention system In this test, the helmet is held by the shell at a point traversed by the vertical axis passing through the centre of gravity of the headform. The headform is equipped with a load-bearing device aligned with the vertical axis passing through the centre of gravity of the headform with an attached device to measure the vertical displacement on the loaded point, as shown in figure 2.55. A guide and arrest device for a falling mass shall then be attached below the headform. The mass of the headform must be 15 kg ± 0.5 kg, which pre-loads the retention system for determining the position from which the vertical displacement of the point of application of the force shall be measured. The falling mass of 10 ± 0.1 kg must be dropped in a guided free fall from a height of 750 ± 5 mm.

During the test, the dynamic displacement of the point of application of the force should not exceed 35 mm. After two minutes, the residual displacement of the point of application of the force, as measured under a mass of 15 ± 0.5 kg, cannot exceed 25 mm.

Retention (detaching) test A device to guide and release a falling mass of 3 ± 0.1 kg is attached on to the rear part of the shell in the median vertical plane of the helmet, as shown in figure 2.56. The falling mass of 10 ± 0.01 kg is then released and drops in a guided free fall from a height of 0.50 ± 0.01 m. The guiding devices shall be such as to ensure that the impact speed is not less than 95 % of the theoretical speed. After the test the angle between the reference line situated on the shell of the helmet and the reference plane of the headform cannot exceed 30°.

Rigidity Test

Rigidity tests involve application of a quasi-static, compressive force to evaluate the ability of a helmet to withstand compressive loads.

The helmet is placed between two parallel plates by means of which a known load can be applied along the longitudinal axis or the transverse axis, as shown in figure 2.57). The

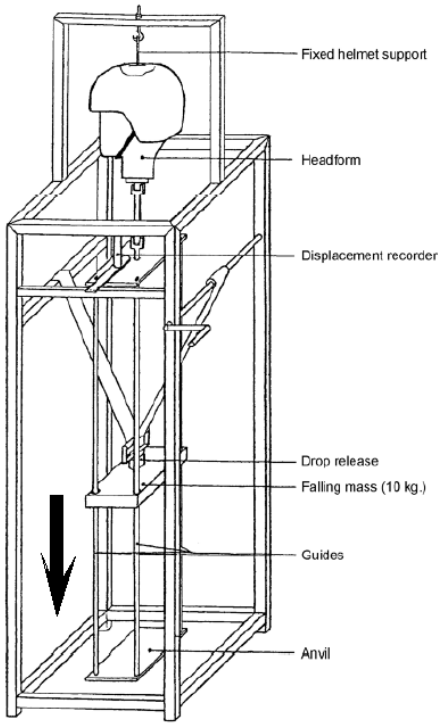


Figure 2.55: *Dynamic retention system test apparatus (Adapted from [8]).*

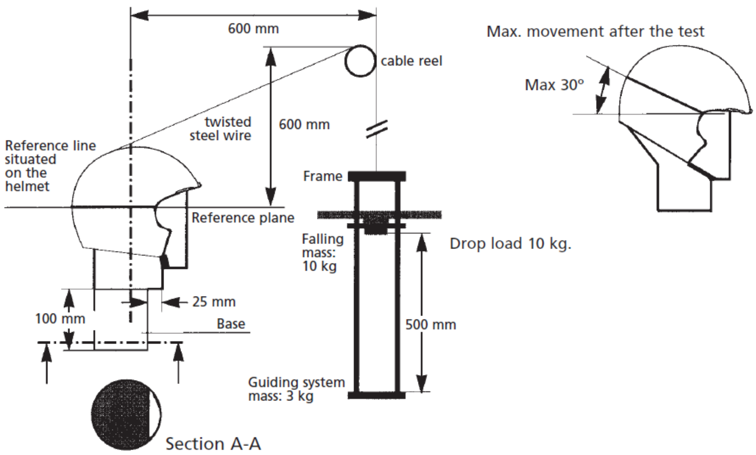


Figure 2.56: *Retention (detaching) test (Adapted from [8]).*

surface of the plates shall be large enough to contain a circle of at least 65 mm in diameter. An initial load of 30 N should be applied, at a minimum plates speed of 20 mm/min, and after two minutes the distance between the two plates shall be measured. The load shall then be increased by 100 N, at a minimum plates speed of 20 mm/min, and then wait for two minutes. This procedure must be repeated until the reaches a maximum load of 630 N. At this stage the deformation cannot be greater than 40 mm. At the end, the deformation measured must not exceed that measured under the initial 30 N load by more than 15 mm.

All these tests were presented regarding mainly the ECE 22.05. However, the procedure of these tests, as already referred is very similar in other standards.

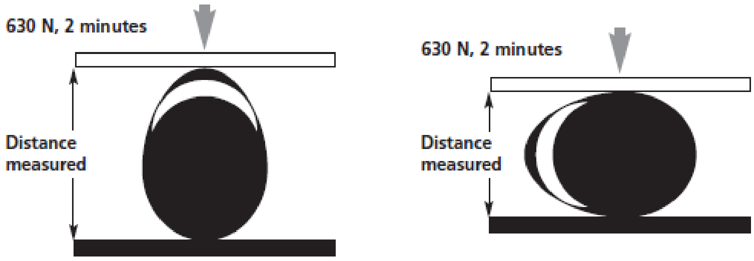


Figure 2.57: *Directions of rigidity test (Adapted from [458]).*

Types of anvil

The anvil is the impact surface during a drop-test. Every standard gives a detailed description of the anvil to be used. Some of these anvils are represented in figure 2.58

- **Hemispherical**:The Hemispherical Anvil is used by several standards.It has a spherical surface with a radius of 50 mm (± 1 mm).
- **Kerbstone**: The Kerbstone European Anvil has two sides forming an angle of (105°š 5°), each has a slope of (52.5°± 2.5°) towards the vertical and meeting along a striking edge with a radius of 15 mm ± 0.5 mm. The height is at least 50 mm and the length is not less than 125 mm. The orientation is 45°to the longitudinal vertical plane at points B, P, and R, and 45°to the base plane at point X (front low, back up).
- **Flat**: The Flat Anvil has a flat surface of a minimum 125 ± 3mm or 130 ± 3mm diameter circle and is at least 24 mm thick.
- **Edge**: Basically, it is a salient edge.

The most common type of object found in real crashes is flat and rigid [76, 79, 67], usually the road surface.

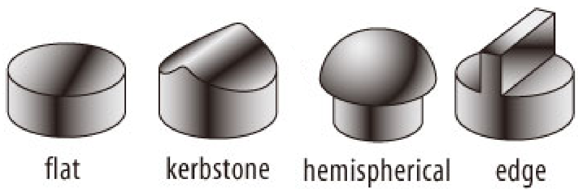


Figure 2.58: *Anvils (Adapted from [458]).*

2.3.2 ECE 22.05

This standard was created by the Economic Community of Europe (ECE) concerning the approval of protective helmets for riders and passengers of PTW. Globally, it is actually the most commonly used, being used by over 50 countries worldwide[8].

Compared to other standards, the ECE 22.05 has a big advantage which is that each manufacturer have to give for test two different samples in order that one will be re-tested for compliance, which is not done in the other standards where the manufacturer can produce the helmet forever and add some changes without being tested any more. Another advantage of the ECE 22.05 and maybe the most important is the requirement for mandatory batch testing of helmets even before they leave the factory, improving the helmet quality and certification is always ensured by a sample test for every production set that was produced.

The points of impact are defined for each helmet as represented in figure 2.59:

- B, in the frontal area, situated in the vertical longitudinal plane of symmetry of the helmet and at an angle of 20°measured from Z above the AA' plane,
- X, in either the left or right lateral area, situated in the central transverse vertical plane and 12.7 mm below the AA' plane,
- R, in the rear area, situated in the vertical longitudinal plane of symmetry of the helmet and at an angle of 20°measured from Z above the AA' plane,
- P, in the area with a radius of 50 mm and a centre at the intersection of the central vertical axis and the outer surface of the helmet shell,
- S, in the lower face cover area, situated within an area bounded by a sector of 20°divided symmetrically by the vertical longitudinal plane of symmetry of the helmet.

Impacts at points B, X and R should be within 10 mm radius of the defined point. After each impact the helmet shall be re-positioned correctly on the headform prior to the next impact, without interfering with the adjustment of the retention system.

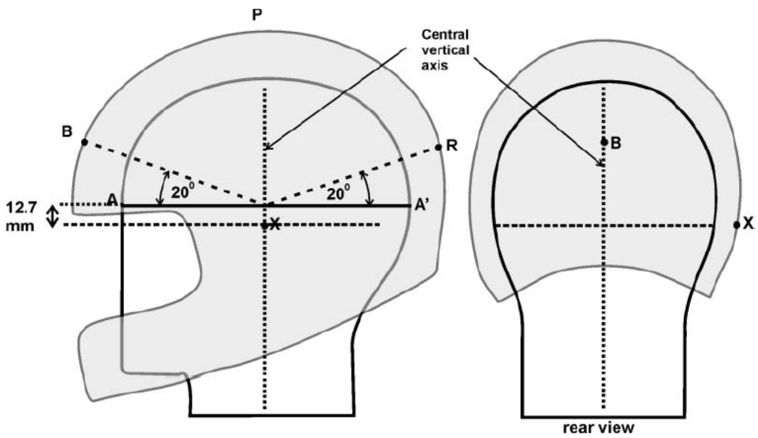


Figure 2.59: *ECE 22.05 impact points [465].*

The general characteristics of the headforms used in this standard are presened in the table 2.6. The headform used by ECE 22.05 is shown in figure 2.40. This headform differ from the others, having a short neck.

Table 2.6: General characteristics of ECE R22.05 test headforms

Symbols	Size [cm]	Mass [kg]
A	50	3.1 ± 0.10
E	54	4.1 ± 0.12
J	57	4.7 ± 0.14
M	60	5.6 ± 0.16
O	62	6.1 ± 0.18

The shock absorption test of this standard is made at the velocity of 7,5 m/s, for the points B, P, R and X and is made also an impact at the speed of 5,5 m/s for the point S. Nevertheless, only the points B, P, R and X are required. The anvils used for the test are the flat anvil and the kerbstone anvil, as shown in the figure 2.60.

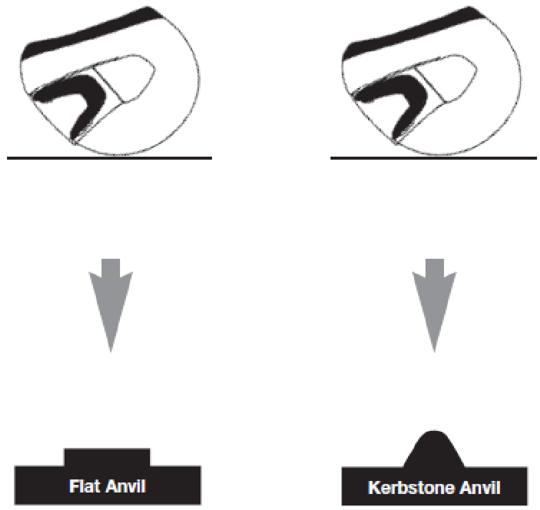


Figure 2.60: *ECE 22.05 shock absorption test (Adapted from [458]).*

This standard has a test that no other standard has, with the exception of BSI 6658 - 1985. This test is the friction test that can be considered an oblique test, where rotational motion is induced to the helmet. However, only the surface friction and force transmitted are assessed and unfortunately, nothing related to the rotational motion is evaluated.

Also, the shock absorption test is different between this standard and the others. The other standards such as DOT and Snell M2010 use a vertically-guided headform that cannot rotate during impact while the unrestrained headform method in ECE allows rotation in any direction as the headform responds to the test impact. However, this rotational motion and acceleration is not monitored in any way and important data is therefore lost.

2.3.3 DOT FMVSS-218

The US Department of Transportation (DOT), National Highway Traffic Safety Administration (NHTSA), proposed the North American motorcycle helmet standard, the Federal Motor Vehicle Safety Standard 218 (FMVSS 218) that is commonly referred as the DOT standard. To meet legal requirements for the states that have their own mandatory motorcycle helmet law, all motorcycle helmets must meet the DOT standard set by the Department of Transportation.

Beginning in 1974, motorcycle helmets were required to meet the minimum performance requirements established by FMVSS 218 standard criteria, where a helmet must fulfill those requirements to receive a DOT approval. In other words, DOT standard sets a level of protection that is targeted at most accidents but does not demand that helmets meet the most extreme impact threats [461]. Over the years, slight changes have been made to FMVSS 218, and since than the standard remains essentially unchanged from its original form until recently. At the beginning of the year 2012, this standard was updated [459]. However, again, few changes were made.

This standard is mandatory for every motorcycle helmet sold in the United States and Canada and implies a set of tests in impact protection, retention systems and how the helmet design affects the rider peripheral vision. If a helmet do not meet the DOT certification standard it cannot be sold as a motorcycle helmet. However, some U. S. states have eliminated the mandatory helmet use.

One problem with the DOT standard is that there is no entity that performed the experimental tests, relying only on the manufacturer's word.

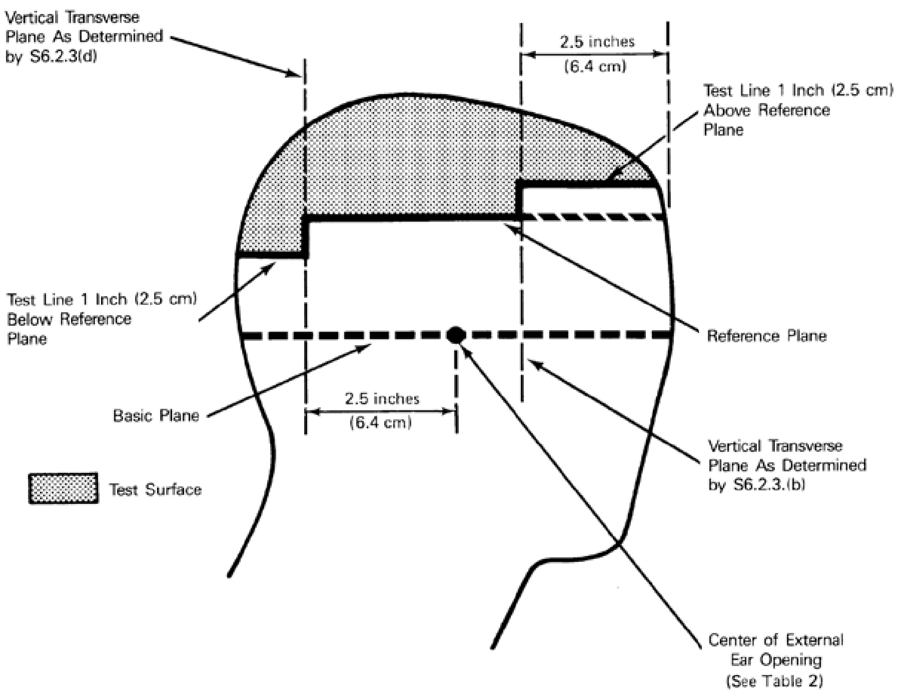


Figure 2.61: *Test area in DOT standard [459].*

Impact testing at each of the four sites, shall start at two minutes, and be completed by four minutes, after removal of the helmet from the conditioning environment. Each helmet is impacted at four sites with two successive identical impacts at each site. Two of these sites are impacted upon a flat steel anvil and two upon a hemispherical steel anvil. The impact sites are at any point on the area above the test line represented in figure 2.61, and separated by a distance not less than one-sixth of the maximum circumference of the helmet in the test area.

The guided free fall drop height for the helmet and test headform combination onto the hemispherical anvil shall be such that the minimum impact speed is 5.2 m/s. The minimum drop height is 138,4 cm. The guided free fall drop height for the helmet and test headform combination onto the flat anvil shall be such that the minimum impact speed is 6.0 m/s. The minimum drop height is 182,9 cm. The drop height is adjusted upward from the minimum to the extent necessary to compensate for friction losses.

The flat anvil is constructed of steel with a 12.7 cm minimum diameter impact face and the hemispherical anvil is constructed of steel with a 4.8 cm radius impact face.

2.3.4 Snell M2010

After the dead of William "Pete" Snell, a race car driver that died in 1956 of massive head injuries sustained in a racing accident, Snively in 1957, created the Snell Memorial Foundation (SMF), which had a profound impact on modern helmet design and performance.

It is a not-for-profit organization with the goal of investigate and understand the mechanisms of head injuries in automotive sports and to encourage the development of protective helmets.

For the approval process, helmet manufacturers submit their products for certification. If their helmets pass the demanding series of performance tests, the manufacturers are invited to enter into a contract with the SMF. The contract entitles the manufacturer to use the Snell name and logo on their packaging and in their advertising. The manufacturer also purchases certification decals for use on their certified products. Under the contract with the SMF, the manufacturer is required to maintain their high standards for all of their certified production. Verification is achieved through a random sample test program. In this program, the SMF acquires helmets and tests them to certify the continuing quality of the products. The SMF makes an effort to ensure that these random sample helmets are drawn from the same supply as those sold in stores; so they are able to monitor the quality of the helmets sold directly to the consumer.

The Snell standard don't replace the DOT standard, it is not a mandatory standard. All motorcycle helmets sold in the U.S.A. must be DOT certified, but they are not required to be Snell certified.

To take advantage of technical and research development and to provide the highest degree of safety to consumers, Snell Standards are updated about every five years, which is an advantage of this standard relatively to others, being the most recent standard. The last form known of Snell is the Snell M2010 [63].

This last update of Snell standard approached it from ECE 22.05. The same was concluded from the last update of DOT standard. For example, the drop masses of the last version of the Snell, Snell M2010 are the same of the ECE 22.05.

The velocity for first impacts has been set to 7.75 m/s while the second impact velocities depend on headform circumference. M2010 seeks to demand all the protective performance reasonably possible for each different head size rather than select a single, uniform second impact velocity and, by doing so, limit helmet protection for all sizes to that achievable for the largest helmet sizes.

Table 2.7: M2010 Second impact velocities relatively to the headform size.

	Head Form					
	A	**C**	**E**	**J**	**M**	**O**
Size	50cm XXXS	52cm XXS	54cm XS, S	57cm M, L	60cm XL	62cm XXL
Criteria	275 G	275 G	275 G	275 G	264 G	243 G
Mass	3.1 Kg	3.6 K.g	4.1 Kg	4.7 Kg	5.6 Kg	6.1 Kg
1st Hit	7.75 m/s 3.06 m	7.75 m/s 3.06 m	7.75 m/s 3.06 m	7.75 m/s 3.06 m	7.75 m/s 3.06 m	7.75 m/s 3.06 m
2nd Hit	7.09 m/s 2.56 m	7.09 m/s 2.56 m	7.09 m/s 2.56 m	6.78 m/s 2.34 m	5.73 m/s 1.67 m	5.02 m/s 1.28 m

The Snell Memorial Foundation is the only that requires chin bar testing. The ECE 22.05 has this type of test but it is not required to pass the standard, a simple indication is done about this type of protection. The DOT standard simply does not consider the chin bar test. This is a problem for modular helmets, where just recently a motorcycle helmet pass this test required by the Snell M2010.

Although the heavier head forms imply lower peak accelerations for some impacts they also imply significantly higher impact energies suggesting problems in testing against the hemispherical anvil.

Since M2010 will call out head forms with a cubic relationship between circumference and drop mass just as do DOT and ECE 22.05, the impact attenuations demanded are much more similar. It is expected that manufacturers will be able to qualify their Snell certified models to DOT and to ECE 22.05 readily and demonstrate compliance with European and United States requirements.

The shock absorption test of this standard is made at the velocity indicated in the table 2.7 for first and second impact depending on the size of the headform used. The criteria depends too, on the size of the headform used. The anvils used for the test are the flat anvil, the hemispherical anvil and the edge anvil.

The impact site is at any point on the area above the test line represented in figure 2.62.

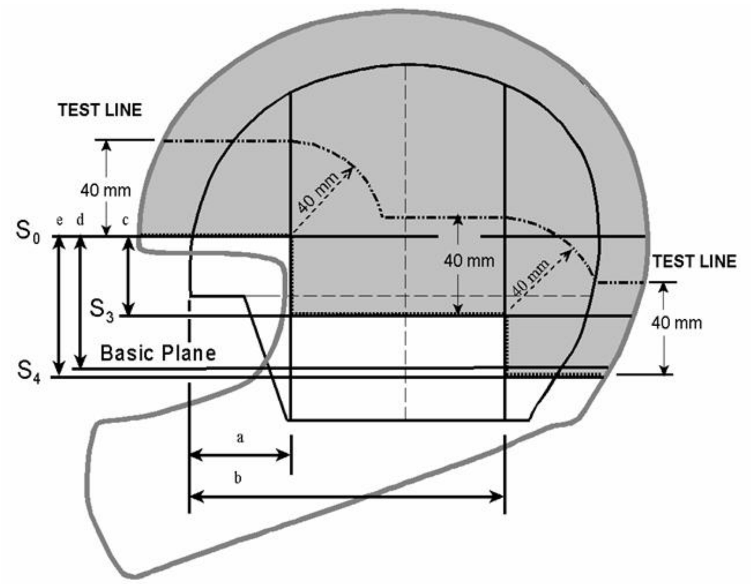


Figure 2.62: *Test area in Snell M2010 [63].*

2.3.5 Standards Comparison

European motorcyclists have to wear helmets that meet ECE 22.05 regulation even if in some cases this standard offer less protection than DOT or Snell.

In order to find a solution for this problem, Snell engineers developed the Snell M2010 standard tempting to approach the DOT and the ECE 22.05 requirements. Also, the last update of DOT was made in that sense.

Similarities between standards are well accepted, mainly from motorcycles manufactures, that could try to sell the same helmet in countries that have different standards, without changing or designing accordingly to each standard. However, differences are still visible and actually, a helmet that is approved from one of this standards, may not be approved from the others. An example, is the double impacts required by Snell M2010 and DOT against the single impact required by ECE 22.05. Double impacts are not typical of accident events, but the requirement is an acceptable procedure which provides a margin of safety for the consumer [460].

A short summary of the tests performed from each standard is presented in the table 2.8.

Table 2.8: Overview of motorcycle helmet standard tests.

Standard	ECE R22.05	Snell M2010	DOT FMVSS 218	BSI 6658
Impact	X	X	X	X
Penetration		X	X	
Retention	X	X	X	X
Roll off	X	X		X
Rigidity test	X			
Friction test	X			X

A comparison between the current standards from the impact point of view is summarized in table 2.9.

Table 2.9: Standards comparison

Standard	M2010 (Size J headform)	DOT	BSI 6658	ECE 22.05
Impact Criteria				
	Velocity:	Velocity:	Velocity: (flat or hemi anvil)	Velocity:
1^{st} impact	7.75 m/s	6.0 m/s	7.5 m/s or 7.0 m/s	7.5 m/s
2^{nd} impact	6.78 m/s	5.2 m/s	7.0 m/s or 5.0 m/s	-
Failure Criteria				
Peak	275 g	400 g	300 g	275 g
150 g	-	4 msec	-	-
200 g	-	2 msec	-	-
HIC	-	-	-	2400

Nevertheless, the HIC and the PLA remain as the only normative parameters used for helmet homologation in terms of protection against impacts. This means that no standard assess the rotational motion that a motorcyclist is subjected neither the local tissue thresholds. However, as already referred in section 2.2.4, the rotational acceleration is presented in all motorcyclists accidents and as a tremendous effect in brain injuries. Also, what is done today, is design helmets to pass the standards and no consideration is taken from the biomechanical point of view. So, an optimisation based on biomechanical criteria (for example strain and stress based head injury criteria) is different than the optimisation with HIC criterion which is correlated with acceleration of a rigid headform's centre of mass and used for helmets homologation.

Transmitted Force

In the figure 2.63 is compared the transmitted forces for a same helmet tested by ECE 22.05 and Snell M2010, where the dashed line is Snell M2010 and it follows the ECE 22.05 line for the smaller head form sizes but, for 60 cm and greater, the M and O head forms, it breaks to the lower Snell limit. Effectively, the M2010 criterion is the lower of ECE 22.05 for each helmet size.

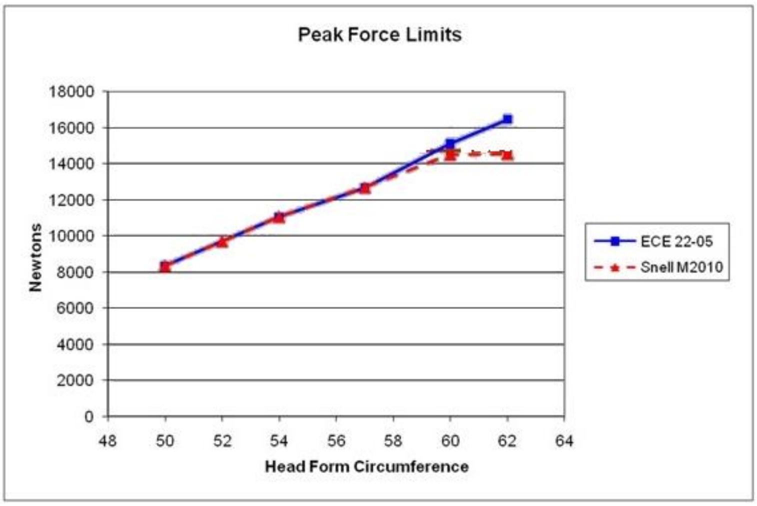


Figure 2.63: *Transmitted force (Adapted from [63]).*

Energy Management

DOT certification implies more energy management than ECE 22.05. Although ECE 22.05 calls out a higher impact velocity, there is only a single impact and the kerbstone anvil is much less aggressive than the hemispherical impact called out in DOT. However Snell M2010 implies substantially more impact energy management than either the DOT or ECE 22.05 standard. In the figure 2.64 is compared the energy management for these standards in an impact against a hemispherical anvil.

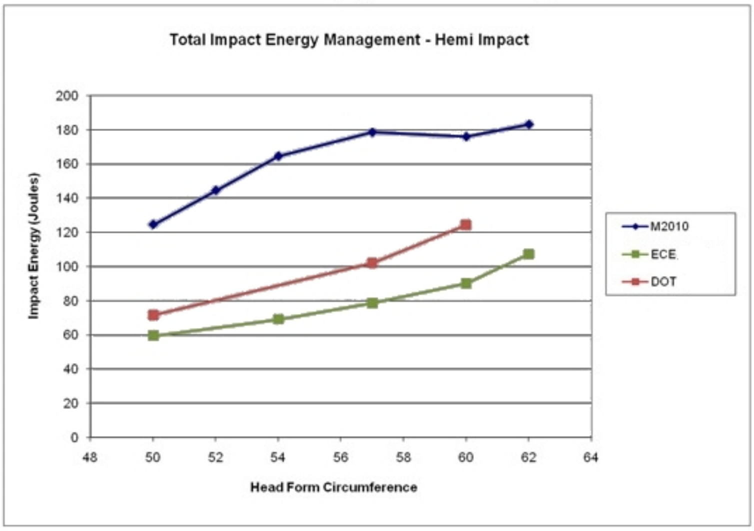


Figure 2.64: *Energy management [63].*

While there is always the temptation to increase the impact energy level with the expectation of providing greater protection, any change that is without support by research may adversely affect accident performance [460]. For example, if the impact energy of the standard test were increased, the typical design change would be an increase in liner density. These changes could provide the greater impact attenuation but may increase headform accelerations for impacts less than the standard test [50].

2.4 Oblique impact

The current criteria and shock absorption tests assessed and performed by the current helmet standards has a great limitation. The rotational motion is not considered, not taken into account the effect of this type of motion because current helmet standards only measure the headform linear acceleration in direct impacts.

One of the reasons for such miss, is that the criteria used, PLA and HIC, only assess the linear motion. The ECE R22.05 shock absorption test allows headform rotation during the impact, but unfortunately, rotational accelerations are not measured. There is other test that allows headform rotation. However, this is used only to assess external projections against the helmet's surface, to check that helmet projections do not cause excessive tangential forces. One of the reasons why there are no helmet standards measuring rotational effects is because there are no globally accepted injury tolerances for helmet impacts that include rotations [44, 464]. Several criteria were proposed over the years but no one was accepted as a globally well accepted injury criterion.

This ignores the fact that in reality almost always external load results in both translational and rotational head accelerations, and both determine the total deformation pattern of the brain. In addition, the effects of rotational acceleration are believed to be the main cause for specific types of traumatic brain injury, such as DAI and SDH, as seen in the section 2.2.3 where was reviewed this matter.

Halldin *et al.* [114] recognise rotational accelerations to be a major cause for head injury in motorcycle accidents, in particular SDH and DAI. Since oblique impacts, with a significant tangential force on the helmet, are more common than radial (normal) impacts in motorcycle crashes [106, 462], the authors developed an oblique test procedure to assess the helmet's ability to reduce rotational acceleration of the head during impact. In this test, a free falling helmeted headform impacts a horizontally moving steel plate covered with grit grinding paper as shown in figure 2.65, in order to be similar to an impact against the road surface.

The oblique impact test proposed by Halldin *et al.* [114] consists in a free falling headform that impacts a horizontally moving steel plate moved by a pneumatic cylinder of 1 m stroke. It is possible to perform a oblique impact at a desirable impact velocity by controlling the radial helmet velocity and the tangential velocity of the plate. A rough road surface was simulated by a grit grinding paper, bonded to a steel plate, which slides on flat PTFE (Polytetrafluoroethylene) bearings. In the headform centre of gravity an accelerometer capable of recording the linear and rotational acceleration components was positioned. Further developments were made by Aare [44].

In this study it was also concluded that higher angular accelerations are found in rougher surfaces. The basic idea behind this configuration shown in figure 2.65, was first presented by Harrison *et al.* [463].

More recently, Mills *et al.* [69] found that the peak headform rotational acceleration was shown to be a function of three main parameters, the impact velocity component normal to the road, the friction coefficient between the shell and road, and the impact site/direction. It was relatively insensitive to the tangential component of impact velocity. Several oblique impacts were performed with different friction coefficients between the headform and the inner liner and it was observed that raising the friction, the head angular acceleration raised too, proportionally.

The friction coefficient between the shell and the road was also identified by Finan *et al.* [479] as the parameter with more influence in oblique impacts, where reducing the friction between this two surfaces it was reduced the peak rotational acceleration, and vice-versa.

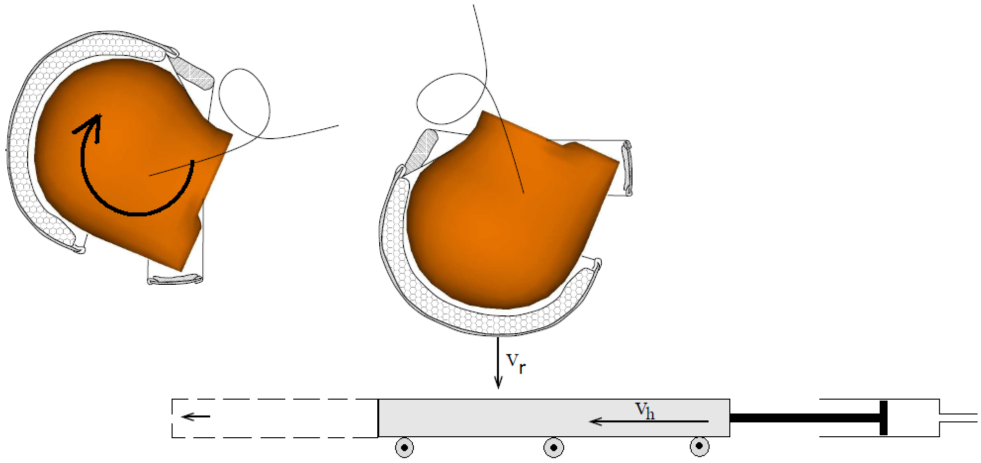


Figure 2.65: *Oblique impact (Adapted from [36]).*

2.4.1 Advanced Motorcycle helmets

In an oblique impact, there are three different types of slip that are important to address with regard to absorption of rotational energies during head impacts:

- the first is between the impacting surface and the outer helmet shell;
- the second is between the shell and the liner;
- the third is between the helmet and the human head.

Helmets are already very smooth on the outside for reduce the friction between the impacted object and the helmet. Nevertheless, since the helmet has to fit the human head well in order to avoid other injuries, the slip between the shell and the liner is the only place where a significant improvement is possible. Based on this, Halldin *et al.* [114] presented a novel helmet, the Multi-direction Impact Protection System (MIPS).

Other prototype helmet, the Phillips Head Protection System (PHPS) proposed by Phillips [115], aim to reduce the friction outside the helmet shell, by introducing easy-shear layer, contrary to the MIPS that introduces the easy-shear layer inside the helmet. The developers argued that this would reduce head rotational accelerations.

Multi-direction Impact Protection System (MIPS)

Besides the oblique impact test proposed, Halldin *et al.* [114] presented a novel helmet. In that study it was tested one helmet type with three different interfaces between outer shell and protective padding liner:

- The 'BONDED' helmet, where the outer shell and the protective padding liner were glued together;
- The 'FREE' helmet, the outer shell and the protective padding liner were joined by rubber strips at the bottom edge. No countermeasures for reducing the friction, between outer shell and protective padding, were taken;
- The MIPS helmet, which was designed to reduce the head's rotational acceleration. In this one, a low-friction Teflon film (a low-friction layer) is placed between the outer

shell and the protective padding liner which allows the shell to rotate relative to the liner in an oblique impact, as shown in figures 2.67 and 2.66. MIPS has also release mechanism that will make the helmet feel robust in normal handling but will release when a certain load is exceeded. Comparing to a conventional helmet, the weight is increased by less than 5%, while comfort and design will not change at all with the MIPS technology [471].

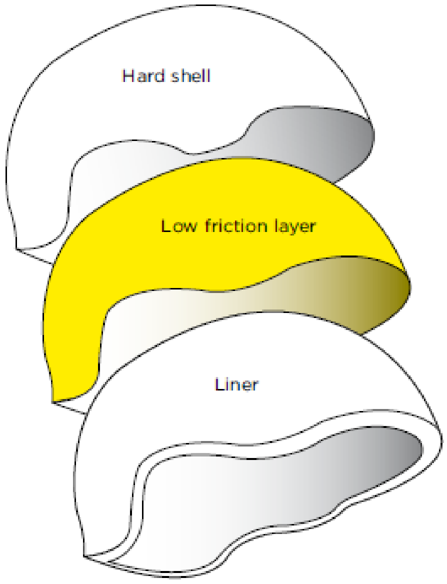


Figure 2.66: *Multi-direction Impact Protection System (MIPS) - construction [471].*

The MIPS helmet reduced the peak rotational acceleration by 28% and 39% compared with the FREE and the BONDED helmets, respectively [114]. It was also concluded that the comfort foam has influenced the results significantly, since one of the functions of the comfort foam is to provide a better fit, which is important in oblique impacts.

Basically, the MIPS mimics the brain's own protection system based on a sliding low friction layer between the head and helmet liner, brain injuries are significantly minimized in connection with angled impacts, as shown in figure 2.67. When the head is subjected to an impact, the brain slides along a membrane on the inner surface of the skull, which reduces the forces transmitted to the brain.

In numerical and experimental tests, the last version of MIPS has shown a dramatic reduction of the forces to the brain, as shown in the figures 2.68 and 2.69, respectively. The results showed that it was possible to reduce the forces to the brain by up to 40% at an impact angle of 45 degrees by adding the MIPS technology [471]. It has also been shown that helmets with MIPS technology perform well in the standard regulation test used today.

Already the MIPS system is globally employed by manufacturers of sports including equestrian, bike and ski/snowboarding helmets such as POC, SCOTT, RED Burton, Lazer and TSG. This technology results from years of biomechanical and neuroscientific research conducted by the Royal Institute of Technology and Karolinska Institute group.

The Phillips Head Protection System (PHPS)

The PHPS enhances traditional helmet design by adding a specially designed lubricated high-tech polymer membrane over the outside of the helmet [115]. Several materials were

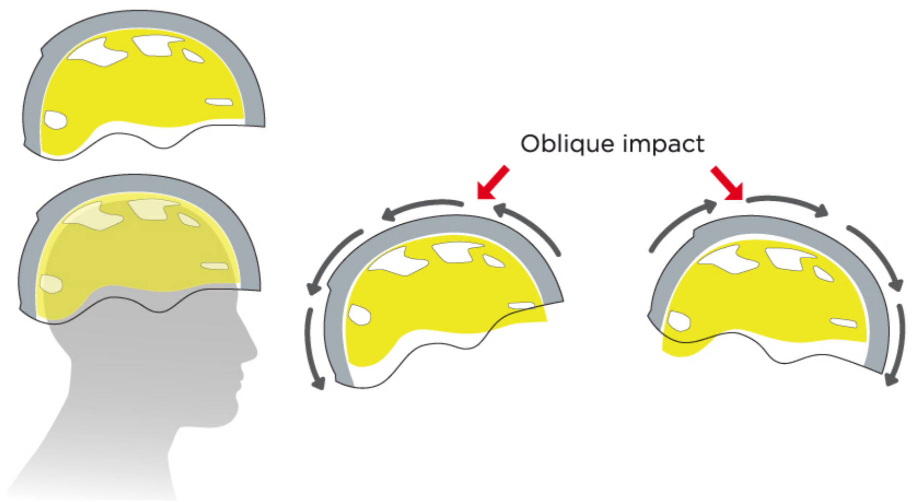


Figure 2.67: *Multi-direction Impact Protection System (MIPS) - function [471].*

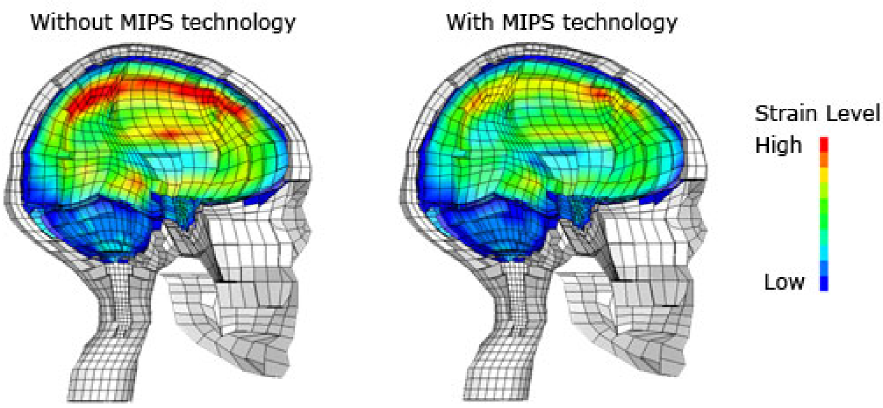


Figure 2.68: *Results of oblique impact simulation with KTH FEHM [471].*

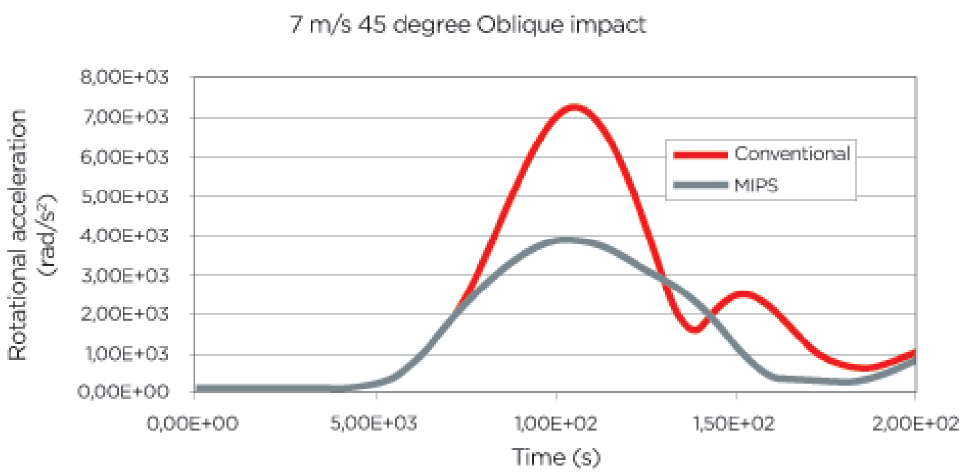


Figure 2.69: *Results of experimental oblique impact[471].*

used, such as closed cell plasticised PVC, high density PU foam and silicon foamed rubber [115]. The membrane is designed to slip in a controlled manner over the inner shell of the helmet. This concept mimics the human scalp, which is a natural protection to brain and skull. Thus, this is a layer on the outside of the helmet, which acts exactly like the scalp does in the human head, by sliding on the shell it limits rotation. An illustrative example is shown in figure 2.70.

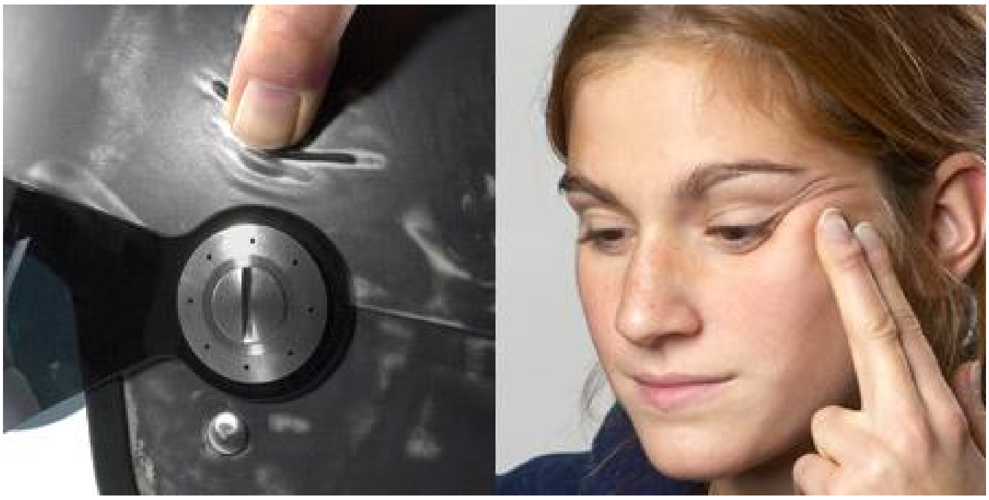


Figure 2.70: *Phillips Head Protection System (PHPS) [468].*

The lubricant and elastic quality of the PHPS membrane on a crash helmet decreases this rotational force and reduces its effect by over 60% in the critical milliseconds following a blow, significantly reducing the head trauma and reducing the risk of traumatic brain injury [115]. It decreases the friction of the helmet surface by moving and sliding over the hard shell. The PHPS high-tech polymer membrane was developed together with a specially designed lubricant. The membrane is designed to slip in a controlled manner over the inner shell of the helmet.

This technology is already commercialized by Lazer SuperSkin motorcycle helmet. Tests performed on the first commercial implementation of the PHPS by LAZER Helmets SA verified that upon head impact the LAZER SuperSkin helmet reduced the risk of intracerebral shearing by 67.5%, by reducing the mechanical effects of rotational acceleration by more than 50% [469]. These results were obtained by Remy Willinger through simulations performed with the FE head model of Strasbourg's University [471]. These tests show the rotational effect of the impact is significantly reduced with the addition of the PHPS membrane.

This concept was also proposed by Mellor and StClair [70] that tried to develop a advanced helmet,where it was tested several layers in several tests. The sacrificial layer in the exterior of the outer shell revealed good results.

Chapter 3

Finite element modeling - A framework for simulating helmeted impacts

This chapter describes the development of a three-dimensional numerical model of a helmet-head system, including different parts and suitable constitutive models for each material. Results show that the developed models can adequately reproduce the behaviour of EPS, in the context of a preliminary analysis. The referred helmet-headform is then submitted to impact on different points, as specified by the European Standard. The results from helmeted impacts are compared against experimental values provided by the helmet manufacturer. The HIC and PLA are also used to compare numerical results against the experimental data, in order to validate the helmet model. This chapter also presents the results obtained from the energy absorption tests and oblique impacts simulated in Abaqus and these results are compared with head injury thresholds, in order to predict the resultant head injuries. At the end, the impact point P of ECE R22.05 is performed with FEHM and again, the prediction of head injuries is made.

3.1 Introduction

Initially, the motorcycle helmet's design, impact behaviour investigation and optimization were based on the experimental investigation, where the results were restricted by varying few impact parameters such as the impact speed or the shape of the anvil. However, varying helmet parameters experimentally was an impossible task, due to testing sample manufacturing constraints, mainly the cost and time spend with such methodology.

This was overcome with the development of analytical models. The development of mathematical models is vital to a better understanding of the helmet impact and also head injury mechanisms. The exact manner in which helmets protect the head is still not fully understood. Over the years, several mathematical models have been proposed.

The earlier theoretical attempts to solve the helmeted-head impact problem were based on analogue techniques that the helmeted-head system could be approximated by an equivalent set of lumped masses, springs and dashpots [472, 475, 100, 79], that represented the helmet components and the headform. These lumped masses systems were then solved using basic dynamic and vibration theories such as modal analysis and dynamic compression [390]. The lumped mass models considered useful in parametric studies are usually simple model capable of provide a quicker and cheaper prediction than an empirical approach and also capable of describing deformation for one specified type of loading condition. However this solution, which is usually either one or two dimensional, had limited advantages due to the approximation degree involved and the incapacity of representing most of the essential impact features encountered in real accidents and - finally - with these models it is impossible to calculate the stiffness of the individual helmet parts from their shapes, dimensions and material properties. For example, the influence of the helmet fit on the headform is not taken into account by almost all these models, because it is difficult to model such interaction, mainly during impact. A few authors tried, such as Willinger *et al.* [390]. This means that the application of lumped mass models is very limited.

This allied with the advance of CPU power led to the development of detailed models by using the Finite Element Method, which led to more detailed results on stresses and strains not only from the impacted helmet but also from the human head. Finite Element Models do not only allow the modelling of the mechanical properties of the helmet components, but also include the geometry of the helmet. This allows the influence of the interaction between helmet and head to be investigated and provide much more information about the helmet's impact than a lumped mass model. The first attempt was reported by Khalil [481], performed with the concern about the biomechanical response of the head to the transient impact waves. More examples of the first simplified FE models, are the ones developed by Kostner and Stocker [473], van Schalkwijk [474], Yettram *et al.* [45]. Some of these models had some limitations, such as the not inclusion of a separate headform model and none of them took into account the effect of the soft comfort liner that provides the fit of the helmet on the head [83]. Also, the first models were not validated, but were used for trend studies only.

Few years later, more advanced FE motorcycle helmet models were developed by Liu *et al.* [370, 224, 480], Brands *et al.* [477], Scott [476], and Chang *et al.* [52]. In these models, helmet geometry was simplified, with either spherical or regular shapes adopted. Thus, these mathematical models differ from real-world situations.

More recently, more realistic models were developed, in order to study the helmet's material [62, 88, 74, 97, 94, 437, 91, 82], the optimization of head dummies [382, 36], the oblique helmeted impacts [44, 69, 352], the effect of impact velocities [39], the helmet's design optimization [73, 6, 7], the virtual modelling and simulation of impacts with motorcycle helmets [478, 75] and the biomechanics on helmeted impacts [279, 67, 380, 322] among many others available in the literature.

Therefore, once a functioning and validated numerical helmet model is created, a great variety of information can be obtained. Such a model may be a three dimensional Finite Element Model, to account for shell vibrations and to be able to use complex anisotropic material models.

Thus, to accomplish one of the objectives of this work, the FEM was chosen to develop and validate a reliable three dimensional helmet-head impact finite element model, based on realistic geometric features of a motorcycle helmet and known material properties to assess the protective performance of the helmet with respect to head injuries.

3.2 Explicit version of FEM

Finite Element simulations of helmet impacts are each time more used for a better understanding of impact kinematics and to validate new preliminary solutions for safety systems.

The time stepping methods are the centre of most structural dynamics problems. There are basically two time iteration methods outside of classical closed-form solutions available to analysts, the implicit formulation and the explicit formulation.

In an implicit scheme, the solution at any time t+Δt is obtained with a knowledge of the accelerations at the same time. Implicit methods are unconditionally stable, however, such stability is obtained at the expense of solving a set of equations at each time step. Generally, it may be said that the implicit integration method is more effective for static or low frequency problems while the explicit integration method is the best for high speed impacts [482]. Thus, helmet-head impacts are best considered by explicit dynamic finite element analysis (FEA) because load acts on the helmet-head system for a very short time interval [82]. Therefore, in order to simulate the different impacts in this study, the explicit method is chosen for such task.

The procedure for the discretized equation of motion is called explicit if the solution at some time t+Δt in the computational cycle is based on the knowledge of the equilibrium condition at time t.

The advantage of using the explicit direct time integration procedure in this transient analysis is that there is no need to calculate stiffness and mass matrices for the complete system, avoiding the need for matrix evaluation, assembly and decomposition at each time step as required by many implicit time-integration algorithms. Thus, the solution can be carried out on the element level and relatively little storage is required.

The drawback of the explicit method is that it is conditionally stable in time and the time step must be carefully chosen, the size of the time step must be sufficiently small to accurately treat the high-frequency modes that dominate the response in this type of problems and this accuracy is achieved at the cost of computational time.

Many finite element codes employ the explicit integration scheme to solve highly transient, non-linear problems. The most widely known is DYNA and its various commercial descendents, LS-DYNA [368], PAMCRASH [369] and MSC/DYNA [488]. Another code using this method that is not a DYNA derivative is ABAQUS/EXPLICIT [9] used in this study.

In order to create the finite element model and simulate the different impacts, the Explicit method of commercial FE package Abaqus 6.10 [9] is used. Abaqus automatically examines the finite element mesh and material properties in order to determine an appropriate time step size for numerical stability. This time step size is then automatically adjusted throughout the transient analysis to account for contact local material and geometric non-linearities.

Abaqus is capable of do the pre-processor, processor and post-processor, being used since the creation and model definition until the results assessment. The FEA of helmet-head impact required inputs consisting of geometry, initial and boundary conditions, interface conditions (contact problems), material properties, etc. The explicit module of Abaqus provides nonlinear, transient, dynamic response analysis of solids and structures using explicit time integration and is designed specifically to serve advanced, nonlinear continuum and structural analysis needs.

The main features of the computer codes , such as Abaqus, suitable for impact calculation have been reviewed by Hamouda and Hashmi [482] and it was concluded that it can give detailed understanding of physical processes and can be used to perform analytical experiments.

3.3 Material models

Before simulating a helmet impact, it is necessary to choose a suitable constitutive numerical model to simulate the mechanical behaviour of each material and set its parameters. Two different materials were modelled, the EPS foam for the energy absorbing liner and the ABS for the outer shell.

3.3.1 Material modelling of EPS foam

Expanded polystyrene (EPS) foam is a material commonly used on many applications, such as shock absorbing packaging of electronic goods or in protective gear. This low density foam has closed cells and is widely applied on energy absorption applications. EPS is the most common liner material used in motorcycle helmets, is a synthetic cellular material with excellent shock absorbing properties and low cost. EPS absorbs the energy during the impact of the helmet, through its ability to develop permanent deformation (by crushing), providing the required protection to the motorcyclist.

EPS foam uniaxial compression stress-strain behaviour can be divided into three regions, as shown in figures 3.5 and 3.6. The first region refers to linear elastic behaviour that arises from bending in the cell walls, the second region is often designated by stress plateau that arises from the plastic collapse of the cells in which strain increases at constant or nearly constant stress and the third corresponds to the densification of the foam in which the stress rises steeply, where cell walls are mostly compressed and the material loses its capability to absorb more energy.

Numerical simulations were performed to validate the energy absorbing liner material model chosen, against experimental data obtained from compressive uniaxial tests.

Figure 3.1 shows the setup of the numerical simulation, consisting of a cylinder with diameter D and length L, in order to replicate the experimental procedure shown in figure 3.2.

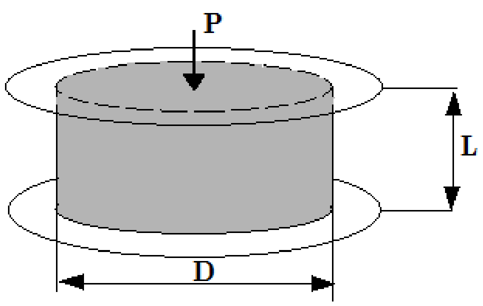


Figure 3.1: *Setup of the numerical simulation used for mechanical characterization of the EPS foam.*

In the figure 3.2 is possible to observe the experimental setup, where the EPS sample is placed between the fixed bottom plate and the movable compression punch.

To minimize possible errors from the experimental tests, a cylindrical punch was made and the contact face was machined in order to reduce any influence of the punch in the test

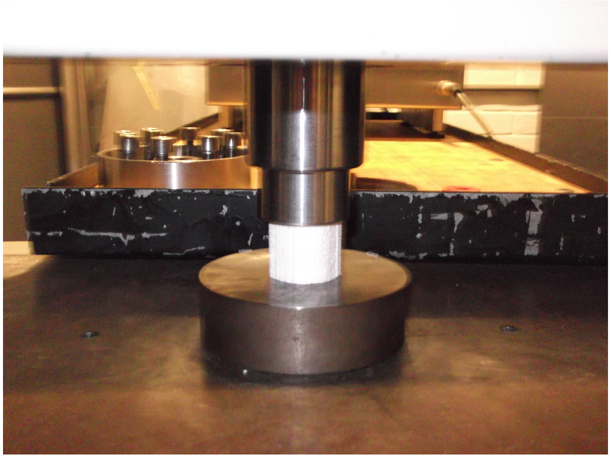


Figure 3.2: *Setup of the experimental procedure used for mechanical characterization of the EPS foam.*

results. This punch is shown in figure 3.3. The test equipment used to perform the tests was the Shimadzu shown in figure 3.4. The Punch has a diameter of 42 mm.

Figure 3.3: *Punch used in experimental tests.*

The energy absorbing liner of the helmet used in this study consists in a multi-density foam, with two different densities, 65 kg.m^{-3} and 90 kg.m^{-3}. In fact, the chosen helmet model includes the lower density EPS in the upper parts of the liner, the main liner and the forehead insert and the higher density is used in the lateral liners. The samples were obtained directly from the helmet's liners, by cutting it. A total of seven samples for each density were tested, six of each at a compression rate of 10 mm/min and one sample of each at a rate of 1 mm/min in order to determine the Young's modulus of the foam. The compressive load P was 30kN which was more than enough to exceed 90% average strain of the sample and as soon as this value was achieved the test was terminated. The initial dimensions of the EPS samples and the properties obtained from the experimental tests are given on table 3.1, where the E is the Young's modulus, the σ_{c0} is the compressive yield stress, the μ is the Poisson's ratio.

The punch was numerically modelled as a rigid plate such as the bottom plate. On the other hand the EPS samples were modelled as a deformable solid. To simulate the contact between the sample and the plates, a surface-to-surface contact using the kinematic contact

Figure 3.4: *Shimadzu testing machine.*

method with a friction coefficient of 0.75 was used. The kinematic contact method was used because previously it was observed numerical instabilities with the penalty method. The friction coefficient was the same used by Masso-Moreu and Mills [483] in a similiar experiment with contact between the steel plate and EPS foam as in this case.

The simulation was carried out at the rate of 7,5 m/s (top plate uniaxial velocity), which is equivalent to a strain rate of 291 /s and 382 /s for the less thick samples and for the thicker ones, respectively. This values are much higher than the ones performed experimentally (quasi static). Nevertheless, it is considered that the deviation from quasi static to dynamic behaviour of EPS is negligible, following the conclusions of Ouellet *et al.* [484] that strain rate effects become pronounced only at rates above approximately 1000/s. Also, Di Landro *et al.* [81] performed quasi static and dynamic tests on EPS and increased the strain rate magnitude several times up to high values and was concluded that the use of characteristics measured through static tests does not lead to significant design errors.

The samples were modelled with four-noded single tetrahedral elements, mainly due to the complex geometry of helmet's liners that only can be modelled with this type of element for geometric reasons. The simulations were terminated as soon as 90% average strain was reached.

The EPS foam was modelled as elasto-plastic material, where the elastic behaviour of EPS is modelled with Hooke's law. To simulate the EPS plastic behaviour, it was used the *crushable foam* material model available in Abaqus (suitable for rigid polymeric foams), that requires an input of the uniaxial compressive data, plus the ratios of the initial yield pressures in hydrostatic tension and compression, p_t/p_{c0} and σ_{c0}/p_{c0}, respectively. The uniaxial compressive data that was introduced in the model was the same data obtained experimentally and the ratios initially used were the same ones used by Mills *et al.* [69]. However, these ratios were tuned up to match the desired experimental curves from the compressive uniaxial tests and are given in the table 3.1. More details of how fit experimental

Table 3.1: Initial dimensions and mechanical properties of the EPS foam samples used on the material characterization models of the helmet liners.

$\rho[kg/m^3]$	$D[mm]$	$L[mm]$	$E[MPa]$	ν	$\sigma_{c0}[MPa]$	p_t/p_{c0}	σ_{c0}/p_{c0}
65	33.61	25.75	7.51	0.1	0.25	5	1.5
90	31.87	19.65	8.64	0.1	0.51	1.9	0.8

data of compressive tests with the *crushable foam* model is given by Mills [57]. Also, to obtain similar results to the experimental tests, the Poisson's ratio was tuned up to -0.99. Only with this value it was possible to simulate the correct behaviour of the EPS and at the final of the simulated tests the diameter of the sample never was inferior to the initial diameter. There is no plausible explanation for only with this value it was possible to obtain results similar to the experimental. Nevertheless, the objective was to mimic the experimental results and such objective was accomplished.

One of the problems of the *crushable foam* material model is that the yield stress is preview to occur at 0.5% of deformation. However, in all of the samples tested, this occur at higher strains, which led to some fittings as already referred to better approximate the numerical results from the experimental results. This can also justify the sooner densification for the numerical curves observed in figures 3.5 and 3.6, where it is compared the numerical and experimental results for both EPS foam densities. Other explanation for this sooner densification can be the type of elements used. Although, tetrahedral elements are good for the creation of meshes for complex geometries, sometimes this elements lead to an excessive rigidity.

From the presented results one can conclude that the numerical model used to simulate the behaviour of EPS is fairly adequate in the content of a a preliminary study.

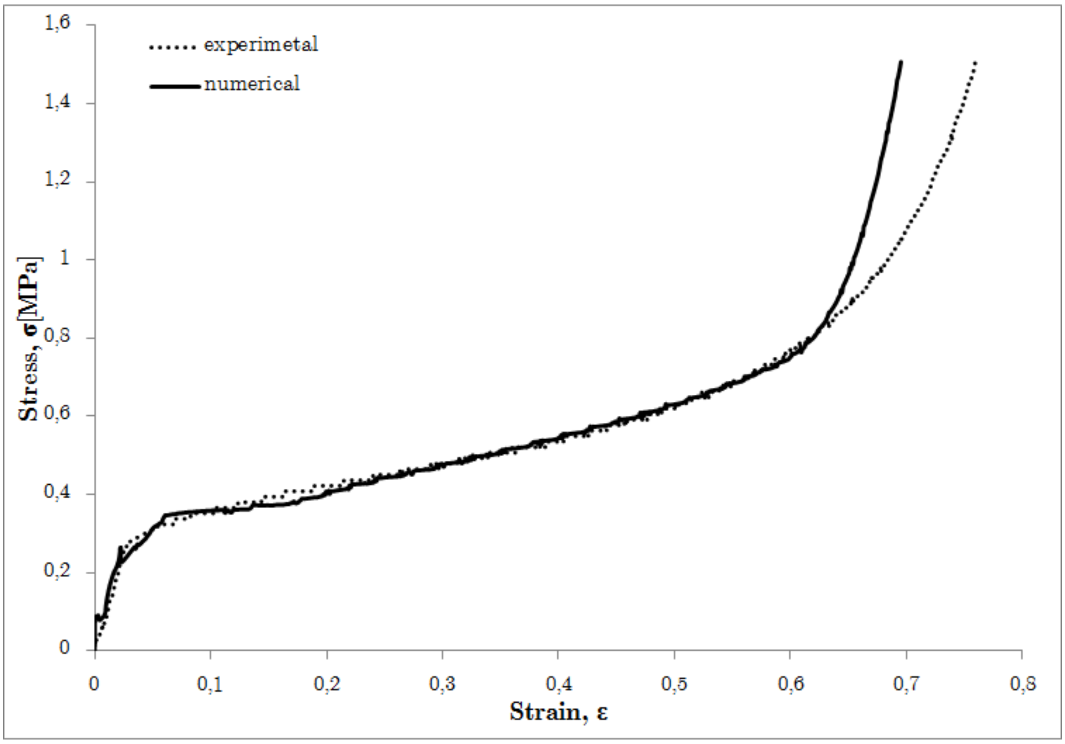


Figure 3.5: *Comparison of stress-strain curves of the EPS to a density 65.*

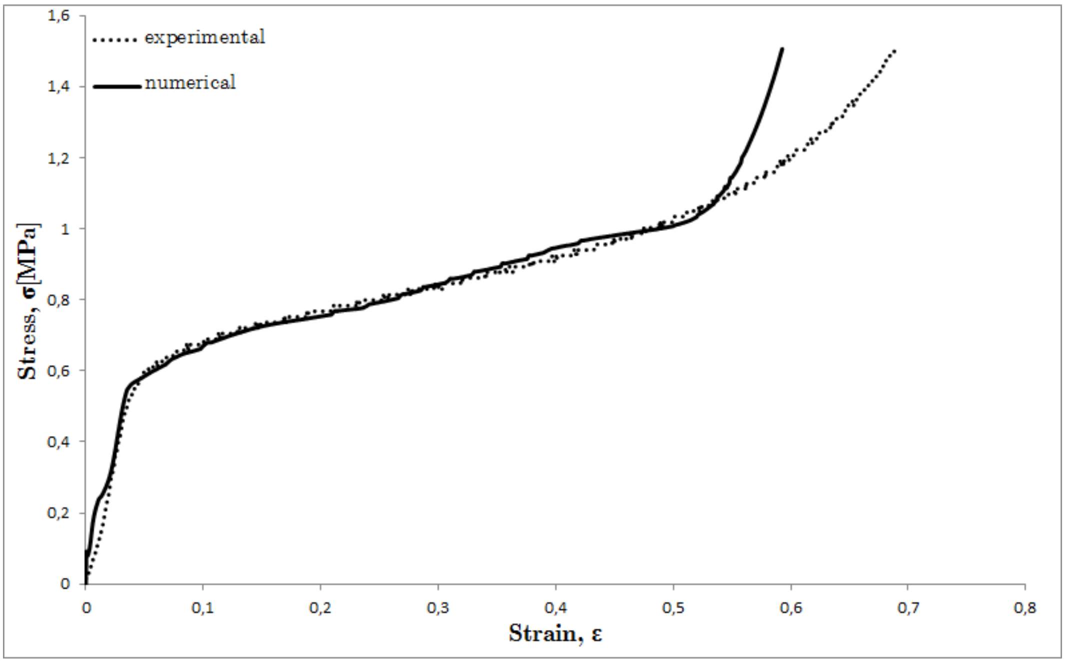


Figure 3.6: *Comparison of stress-strain curves of the EPS to a density 90.*

There are several material models proposed to model the foams behaviour such as EPS foams. Gibson and Ashby [96] proposed a simplified model, where it was determined the phenomena at each zone of EPS curve. Thus, each of the three zones has it equation. For the elastic zone, the material model was developed to allow strains until 5%, which is approximately the values found in the experimental data. At least, the yield stress for the EPS foam used in this study is better modelled with this material model than the crushable foam of Abaqus.

3.3.2 Material modelling of ABS

The outer shell of the developed model is made from Acrylonitrile-butadiene-styrene (ABS), a widely used material on motorcycle helmet as the outer shell material. The ABS is a stiff thermoplastic material very resistant to heat and to penetration. The outer shell material properties of ABS used on this work were based on the work of Pinnoji and Mahajan [86]. To simulate the material behaviour a isotropic linear-elastic material model was considered. This choice is supported by the fact that during an impact the outer shell is mainly responsible for spreading out the impact's concentrated load and generally deforms only elastically. Nevertheless, this simplification is only acceptable when a plastic shell, like ABS, is analysed [75]. Thus, it is only required the input of Young's modulus, Poisson's ratio and density. These material properties of ABS are listed in table 3.2.

Table 3.2: Mechanical properties used for ABS material.

$\rho[kg/m^3]$	$E[MPa]$	ν
1200	2000	0.37

However, the Young's modulus used in the final simulations was 10 GPa in order to match again the numerical data from the helmet's impacts with the experimental data. This change was based in the first simulations performed with the Young's modulus given in the table 3.2, where it was clear that the shell was easily deformed, leading to just local deformation of the liner (small liner area impacted). This was evident, by observing the shape of the impact curve that was similar to the experimental data, but the peak was much higher, and based on this, the increase of Young's modulus was made which led to a stiffer shell that increased the effective liner's energy absorption area, reducing the peak to real values as it will be observed in the results presented in the next sections.

3.4 Finite element motorcycle helmet modelling and simulation

The motorcycle helmet used in this work is the CMS SUV Apribile helmet. This is a modular motorcycle helmet manufactured by CMS Helmets that can also be used as a "Jet" styled open-face helmet. This commercially available helmet, shown on figure 3.7, fully meets European Regulation ECE R.22/05 [8], the Brazilian Regulation NBR-7471 [485] and also the U.S. Regulation DOT [459] and uses a dual-density energy absorbing liner, composed by three different parts: the main padding that involves the entire cranium with the exception of temporal region, the forehead padding insert and the lateral padding, as shown on figure 3.8.

Figure 3.7: *CMS SUV Apribile motorcycle helmet [105].*

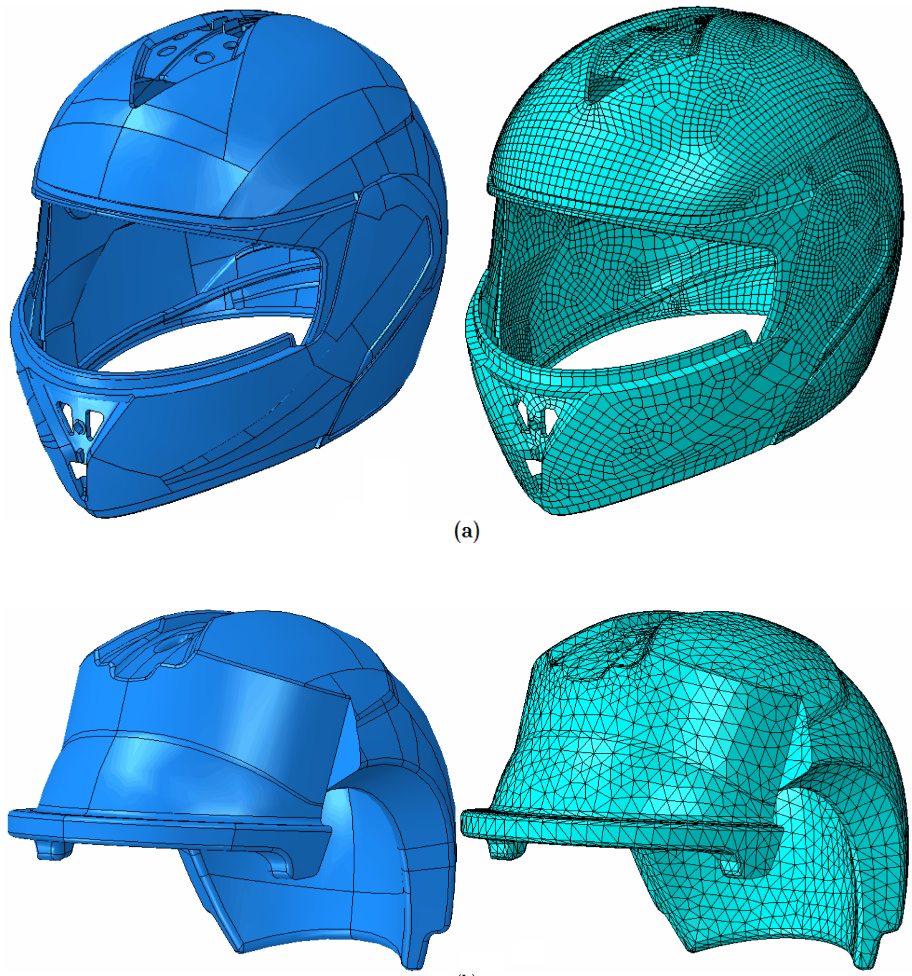
3.4.1 Helmet model

The geometry of the helmet outer shell and the foam was provided by the manufacturer in a computer-aided design (CAD) file. The three-dimensional CAD model of the helmet used to model the different parts of the helmet was treated on CATIA V5R19 CAD system [486], by improving the CADs provided by the manufacturer, in order to create the finite element model and simulate the different impacts, using the Explicit method of commercial

FE package Abaqus 6.10, as already referred. Only slight simplifications were made to the models, maintaining the overall geometry precisely just making the model easier to mesh.

Figure 3.8 shows the CADs and the different parts of the helmet's finite element model developed, including the outer shell, energy absorbing liners and the headform. To model the helmet as full-face helmet, the two shell's components were full constrained by constraining the degrees of freedom of the movable part (the chin guard) with a "tie" type of contact between both. The characteristics of the different meshes shown in the figure 3.8, are presented in the next section.

The lateral liners are made of EPS foam with density of 90 kg/m^3 and the remaining are made of EPS foam with density of 65 kg/m^3. This dual density can be justified by Mills *et al.* [69] that stated that the use of lighter foams in the top area compensates the excessive rigidity of the shell in the crown site caused by the double curvature and lack of free edges in proximity [51], resulting in a better protection of the head. Also, EPS parts with different thicknesses, since 1 mm to 50 mm, can be found in the various positions of the helmet (top, rear, sides) to ensure adequate comfort and protection to the user. Obviously, the thickness and the density are "connected", for example, the denser liner in this helmet is also the one with less thickness. The outer shell made of ABS has a thickness of 3 mm.



(a)

(b)

In this study, the effects of the comfort liner, the retention system (chin strap) and the chin pad were not considered. It is rare to found a study were this features were modelled or modelled with success. For example, none of this studies modelled the comfort foam

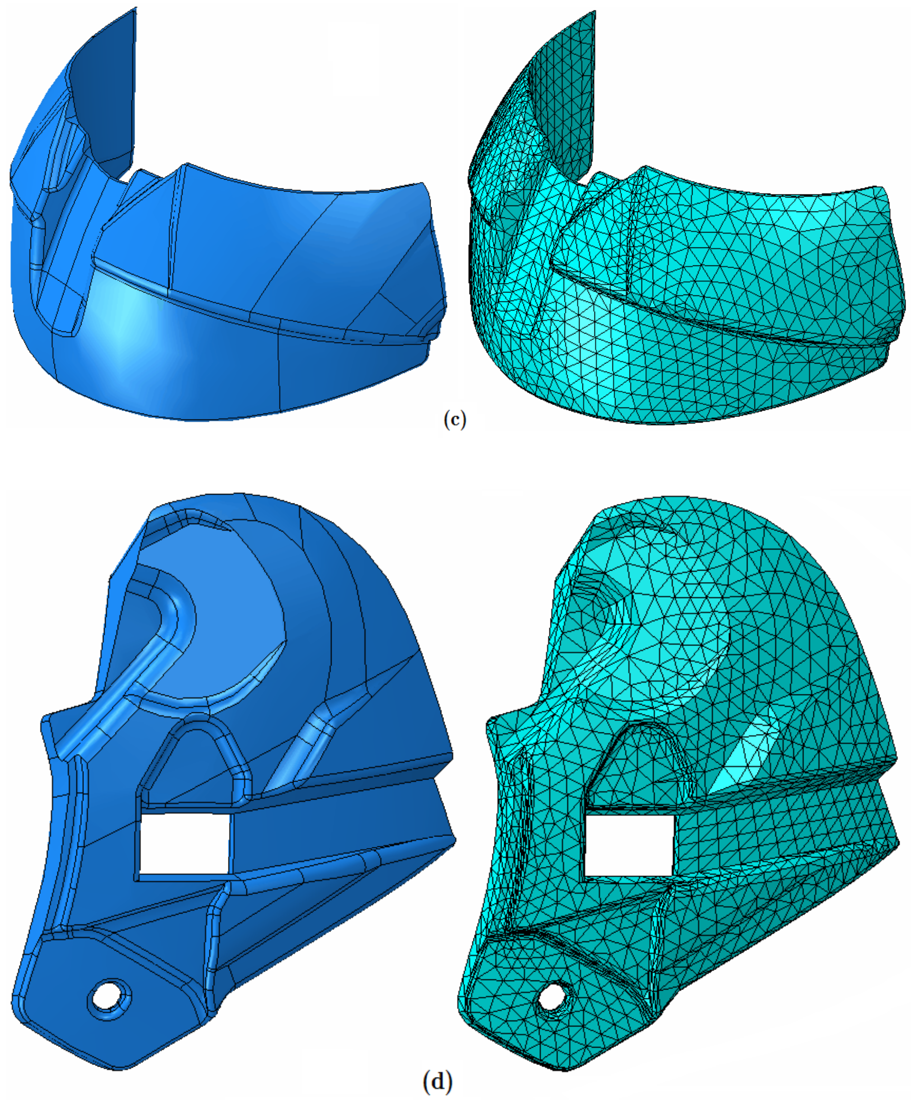


Figure 3.8: *CADs (at the left) and finite element models (at the right) of the helmet parts: (a) outer shell, (b) main liner, (c) forehead insert and (d) lateral liner.*

[481, 473, 474, 45, 370, 224, 480]. Brands [83] has modelled the comfort liner and concluded that it is significant for the shape of the headform acceleration peak, however due to the low stiffness of this liner in the model that causes problems with the numerical stability, it was removed. Also, Pinnoji and Mahajan [82] affirmed that this foam is often very soft and thin to absorb energy and is used only for fitting different head sizes, not having influence on headform response during an impact [74, 7]. Therefore, the majority of the cases doesn't model this foam because it is difficult to model and from a cost-benefit point of view, the improvements in the fitting of the helmet on the head do not worth this task.

At an initial stage, a plastic component responsible to control the air flux that ventilates the helmet was modelled. The geometry of this component was captured using a 3D scanner. This component was modelled mainly because it is positioned at the top of the helmet where is performed an impact. However, the helmet's shell was modelled with an elastic material model and the advantages of incorporate this component in the helmet model are minimum, it can even lead to some numerical problems. Thus, this component was not included in the helmet model.

This helmet is a size large (L) and is designed to fit a 58-60 cm circumference headform, an M size headform of 5.6 kg accordingly with the ECE R.22/05 regulation [8]. The headform used in the simulations is shown in the figure 3.9 and its characteristics are given in table 3.3.

Table 3.3: Headform mass and principal inertial moments.

Mass [kg]	$I_{xx}[kg.cm^2]$	$I_{yy}[kg.cm^2]$	$I_{zz}[kg.cm^2]$
5.6	316	329	176

This helmet is considered relatively light, taking into account the typical weight for a helmet with these characteristics. It weighs 1.625 kg, which makes it one of the lightest flip-up helmets available on the market. This helmet also has a the dual certification, which means that it meets DOT safety standards in North America and ECE 22.05 in Europe as both a full-face and Jet helmet.

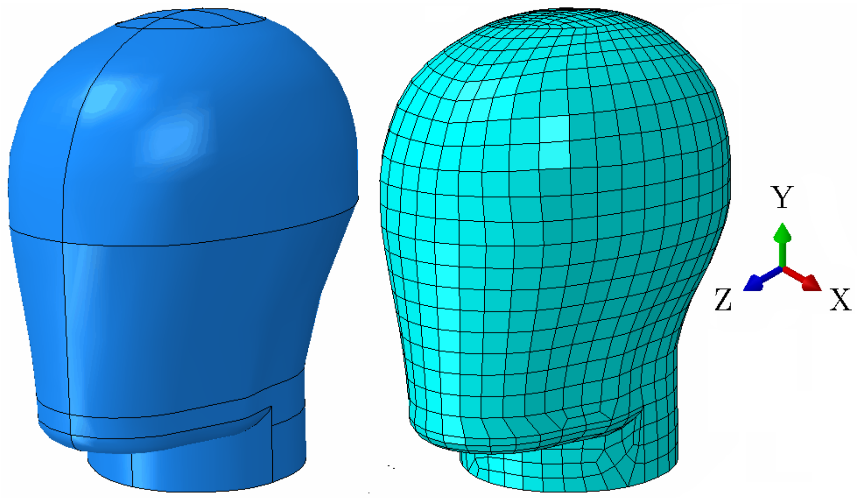


Figure 3.9: *The ECE 22.05 headform of size M: CAD (at the left) and finite element model of the headform (at the right).*

All these components modelled were assembled and the helmet was fitted on the headform. This is shown if figures 3.10, 3.11 and 3.12. From the sagittal cut view shown in figure 3.11 is possible to observe a good fitting between the helmet and the head. It is also possible to conclude that for the top and back of the head the thickness of the comfort liner is very small and considering its low stiffness, not model this foam was a good decision for these locations. However, observing the figure 3.12, it is observed a large gap between the headform and the lateral liners, where it should be the comfort liner. This could influence the results due to the thickness of the comfort liner at this site, especially in an impact at this region. Nevertheless, as shown behind in the literature, some researchers consider this influence negligible.

3.4.2 Finite element mesh

In order to create the finite element model of the helmet, the different parts were meshed. Four-node linear tetrahedral elements (Abaqus's C3D4 element) were used for the foams,

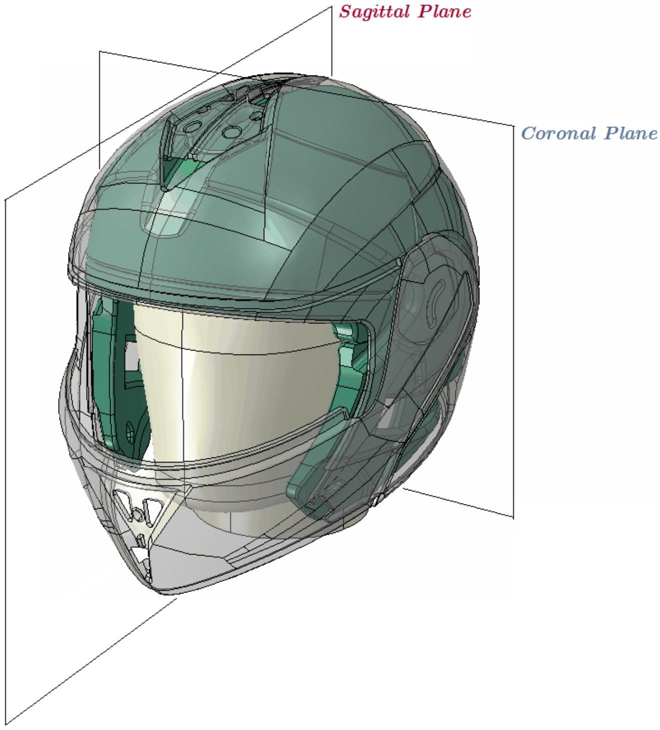


Figure 3.10: *A three dimensional view of the FE helmet-head model.*

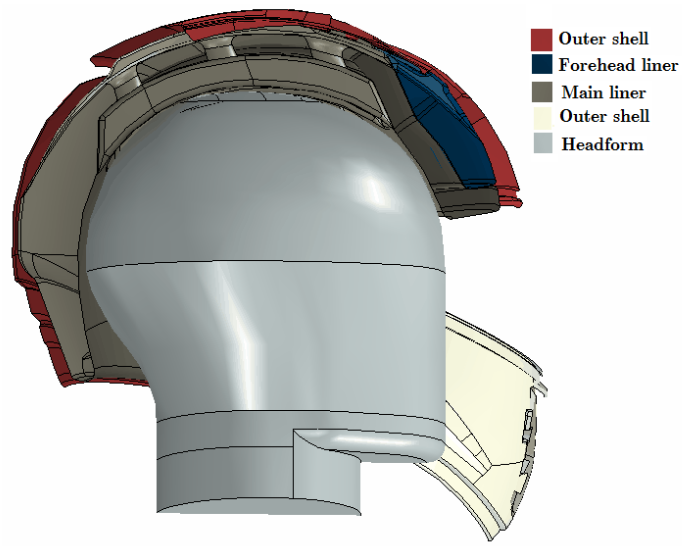


Figure 3.11: *A cut view from the sagittal plane.*

where the main liner was modelled using 65395 elements, the forehead insert was modelled using 17949 elements and the lateral liners 12985 elements for the left one and 12497 to the right one. This type of elements was used to modelled the foam mainly due to the complex geometry of the liners even known that this type of elements could lead to excessive rigidity.

Given the outer shell thickness of 3 mm (averaged value from the real outer shell), the shell was modelled using 9381 four-node linear shell with reduced integration elements (Abaqus's S4R element) with enhanced hourglass control for the main covering shell and 2520 to the chin guard, which makes a total of 11191 elements to the shell. The headform and

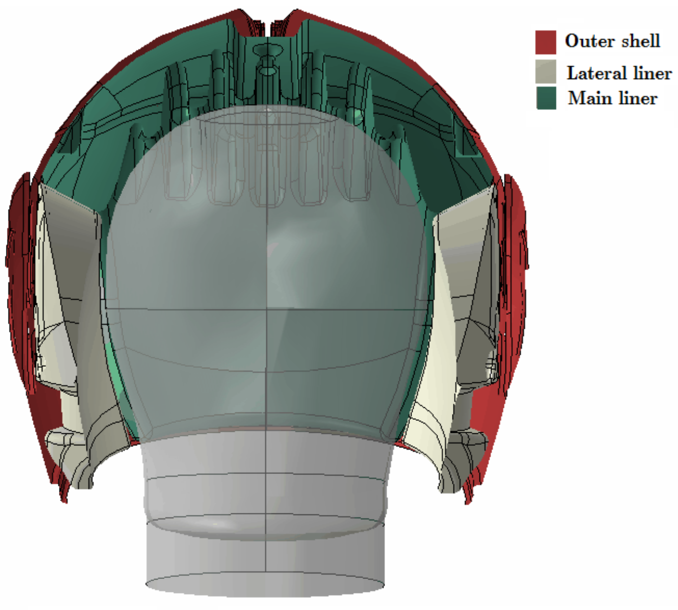


Figure 3.12: *A cut view from the coronal plane.*

the impact wall (equivalent to a flat anvil) were modelled with rigid quadrangular elements (Abaqus's R3D4 element), 1346 and 1, respectively.

Between the anvils required by ECE R22.05, only the flat anvil was modelled because the most common object hit by the head in motorcycle crashes is the flat road surface [41, 76], treated as a rigid body.

The meshes of each part were created always avoiding distorted and warped elements and with especial attention to the time increment, not having elements too small in order to have a reasonable computational time but at the same time a mesh refined enough to obtain good results. The meshes were also created in order to avoid hourglass issues and maintain a precise geometry of each part.

After a meshing of the helmet based on the CAD provided by the manufacturer, the mechanical properties of each component of the helmet (the outer shell and the foam) defined in the previously section 3.3 have been implemented under Abaqus FE code.

A summary of the different meshes used in the developed model are give in the table 3.4.

3.4.3 Contact and impact conditions

To simulate the interfaces between the different parts of the helmet, the head-liner interaction and the impact wall interaction with the helmet's shell was modelled using a surface-to-surface contact (Explicit) with a friction coefficient $\mu = 0.55$, as used by Mills *et al.* [69] to simulate such interaction. Moreover, a friction coefficient of $\mu = 0.5$ was used to model the interaction between the shell and the liner, also with surface-to-surface contact, based again in the work performed by Mills *et al.* [69]. The use of a high friction coefficient prevents the main padding and the crown from sliding out of the shell inner surface. In addition it also simulates the friction between the shell outer surface and the impact surface. Also, a "tie" type contact was used to simulate the real tie (glued parts) between the different parts of the helmet liner. A tie type of contact was also used to fully constrain the helmet's chin guard, and then simulate the impacts considering this helmet as a full face helmet. According to the ECE-R.22/05 standard, the helmet-head model is dropped, without any

Table 3.4: Characteristics of meshes used to model the different parts.

Part	**Element type**	**Abaqus's element**	**N° of of elements**	**N° of nodes**
Main covering shell	four-node linear shell	S4R	9381	14073
Chin guard shell	four-node linear shell	S4R	2520	2723
Main liner	Four-node linear tetrahedral	C3D4	65395	14073
Forehead liner	Four-node linear tetrahedral	C3D4	4171	9587
Left lateral liner	Four-node linear tetrahedral	C3D4	12985	3317
Right lateral liner	Four-node linear tetrahedral	C3D4	12497	2983
Headform	rigid quadrangular shell	R3D4	1346	1348
Anvil	rigid quadrangular shell shell	R3D4	1	4

restriction, against an anvil with a velocity v = 7.5 m/s. The simulated impacts were always against flat anvils only, which is enough for a first stage of the model validation. The flat anvil was fixed (fully constrained) and an impact velocity v = 7.5 m/s was prescribed to the model. The figures 3.13, 3.14, 3.15 and 3.16 show the impact configurations according to the ECE-R.22/05 standard, the B, P, R and X points respectively.

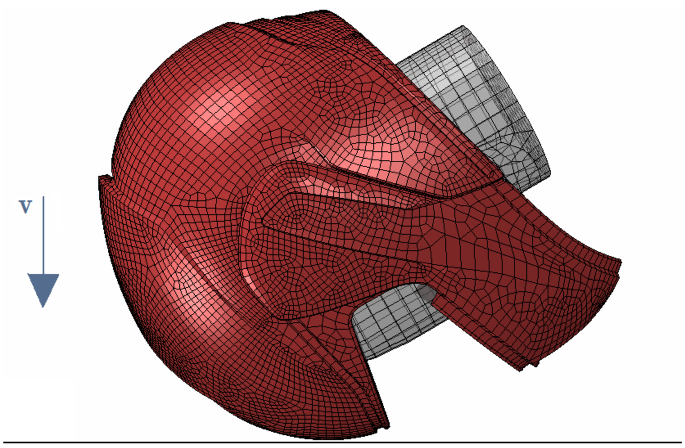


Figure 3.13: *Impact Point B.*

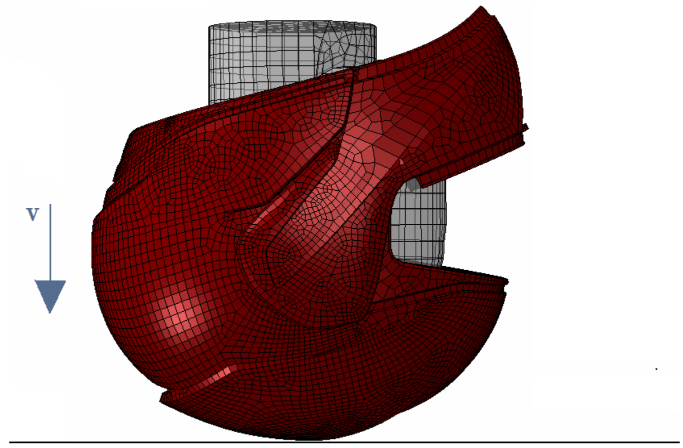


Figure 3.14: *Impact Point P.*

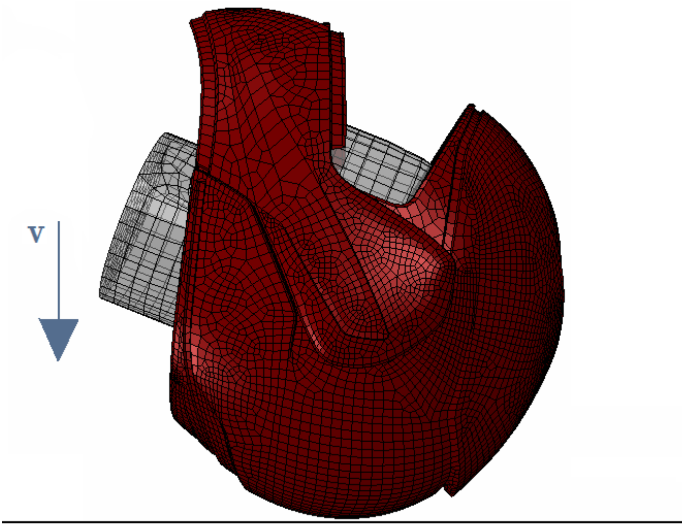


Figure 3.15: *Impact Point R.*

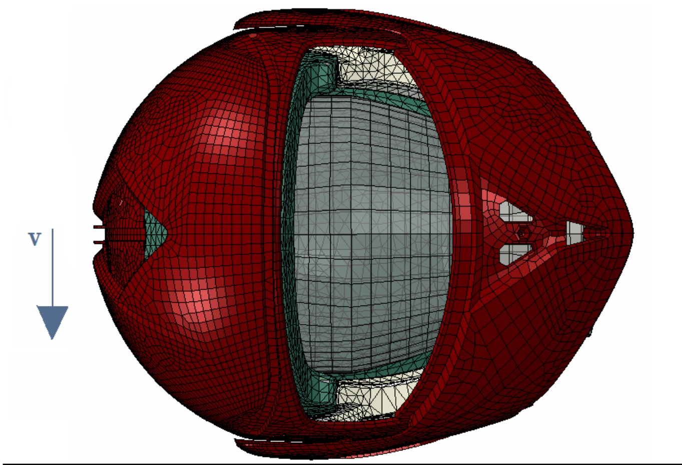


Figure 3.16: *Impact Point X.*

The explicit (dynamic) solver of Abaqus was used to simulate the impacts with durations $\Delta t = 20$ ms, with the large deformation option. To reduce the computational time required, the helmet was placed very close to the anvil and given an initial impact velocity instead of being dropped, as shown in figures 3.13, 3.14, 3.15 and 3.16.

3.5 Validation of the helmet model

As a first step to validate the model, numerical simulations of helmeted impacts were performed in order to validate the numerical simulation framework developed by comparing its results against experimental data provided by the helmet manufacturer. The experimental energy absorption tests were based on ECE R22.05 motorcycle helmet standard, which was already described. Only impacts against flat anvils were simulated and the head acceleration was recorded at the centre of gravity (COG). The comparison between the numerical and experimental results is shown in figures 3.17, 3.18, 3.19 and 3.20.

The PLA and the most disseminated injury criteria, HIC, presented previously in the sections 2.2.4 and 2.2.4, are assessed as well. The expression used to compute HIC is given by the equation 2.3.

According to the results in figures 3.17, 3.18, 3.19 and 3.20 there are noticeable differences between experimental and numerical behaviour for the different impacts. The maximum acceleration of the head and the HIC values calculated from the numerical and experimental analyses are shown in table 3.5.

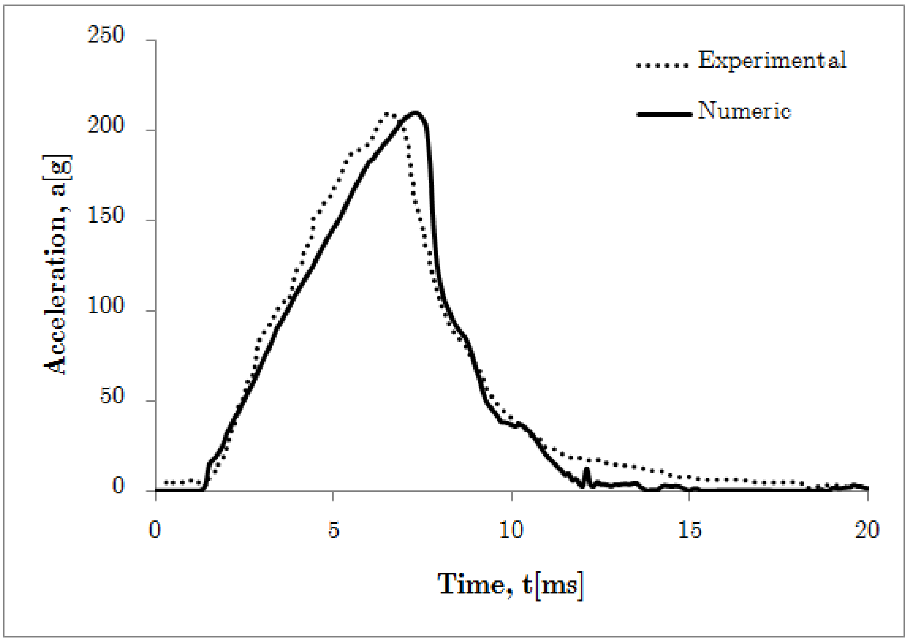


Figure 3.17: *Impact Point B.*

For points P, B and R the agreement between experiments and simulation is reasonable. However, for point X, there is a significant deviation. The differences between experimental and numerical results may be explained by the adoption of a simplified numerical model regarding to the number of helmet components modelled. Especially for point X, where the differences are more evident, it should be noted that in the real helmet the zone around the X point receives several parts, since metals and plastics, where is the visor locking

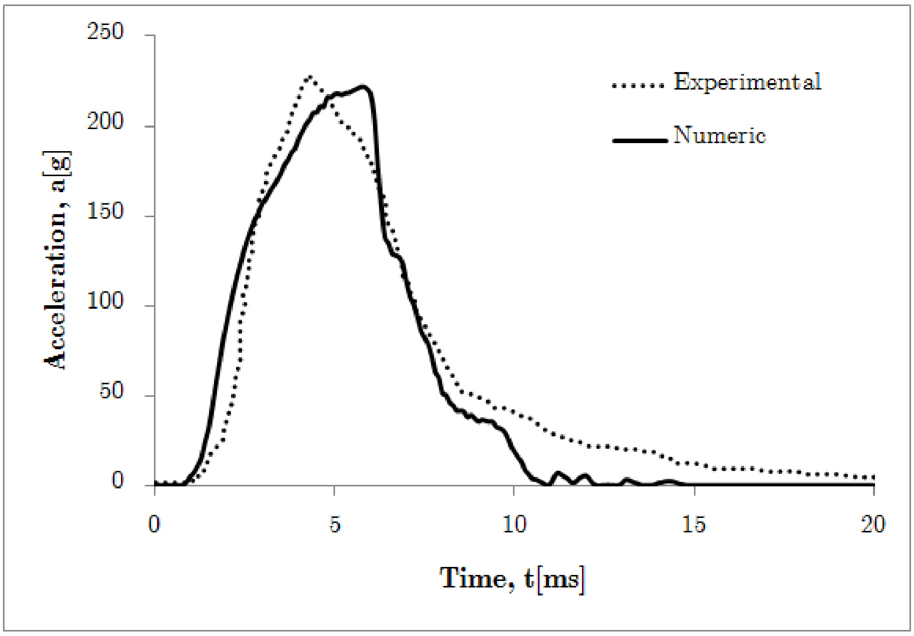


Figure 3.18: *Impact Point P.*

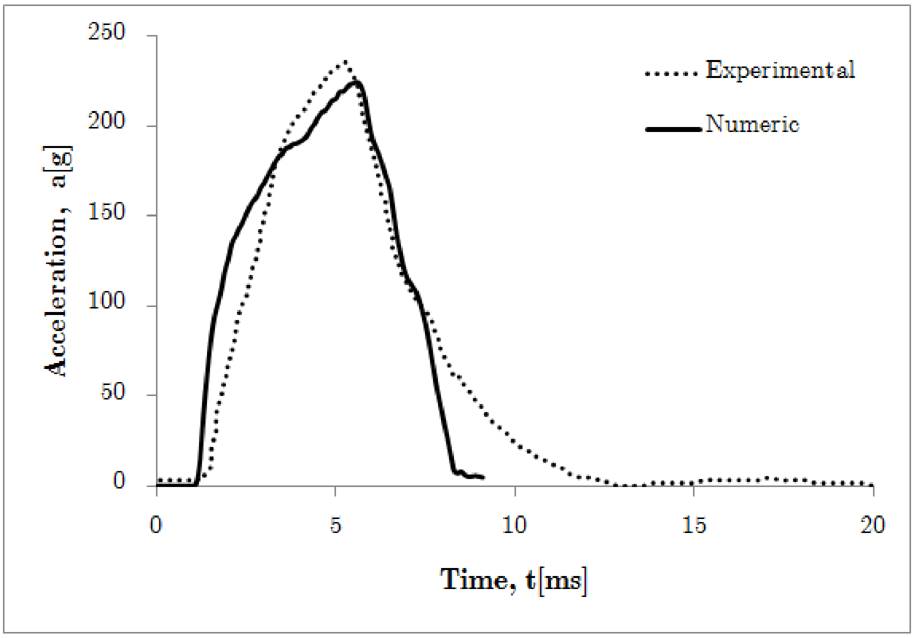


Figure 3.19: *Impact Point R.*

system and the chin strap fixation system and also the fixation system between the two shell parts, because in reality it is a modular helmet and not a full-face helmet. All of this contributes for a change in the energy absorption properties, as a less quantity of padding is included. Doing so simulation, considering just the liner and the shell in that area, provides a different response under impact. However, despite some differences between experimental and numerical impact results, the numerical model was considered adequate enough for a preliminary study on biomechanical issues.

Other explanation for the deviation of numerical results is the absence of comfort padding

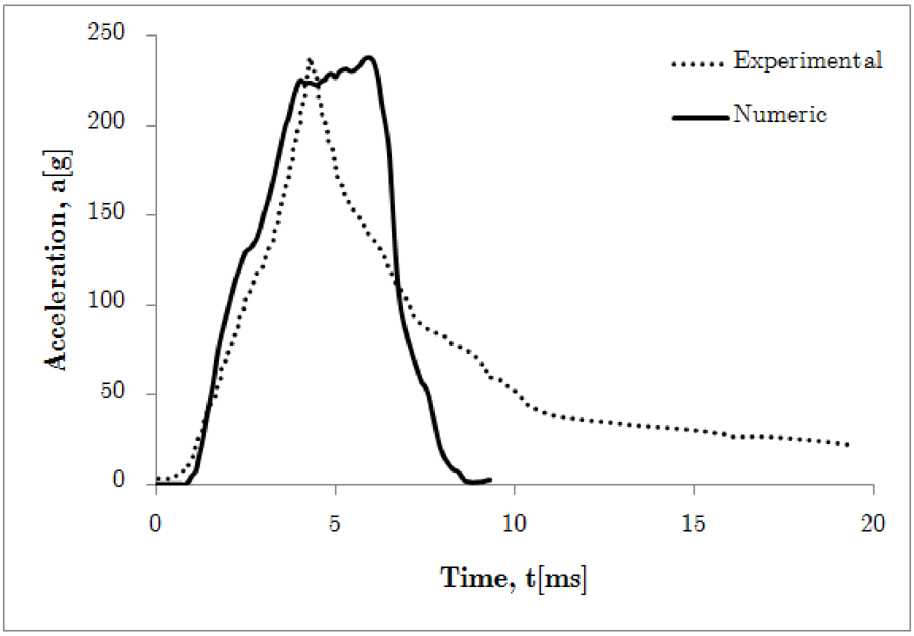


Figure 3.20: *Impact Point X.*

Table 3.5: Maximum acceleration of the head and HIC values calculated from the numerical and experimental studies.

Impact point		a_{max} **[g]** ($\leq$ 275)	**HIC** ($\leq$ 2400)
Point B	numerical	209.7	1461
	experimental	208	1696
Point P	numerical	222.0	1973
	experimental	227	1903
Point R	numerical	224.6	2201
	experimental	234	2235
Point X	numerical	238.3	2130
	experimental	237	1714

in the numerical model. This foam is responsible for the fitting between the helmet and the head and may have relevant importance on the impact kinematics. Although in the majority of the studies developed in this field it is not modelled, the comfort foam was found here that it is very important. Of course, this importance is different between the several helmet models and for this model the comfort foam can be neglected to the points B, P and R (little differences between experimental and numerical results, good agreement between the result for this three impact points) but for the impact X it is very important mainly due to the considerable foam thickness in that region. This feature should be modelled in the future, to make the model even more reliable and accurate. Observing figure 3.12 is possible to observe that the large gap between the headform and the lateral liner where it should be modelled the comfort foam, this gap increase from the top of the head to neck of the headform due to the helmet curvature. Thus, without the comfort foam, only a small area (at the top side of the head) absorbs impact energy at the beginning of an impact. During the impact, the area increases, putting more liner to work while if the comfort foam was present, the lateral area would absorb more energy at the beginning, due to the increase effective area. This fact explain the rapid increase of stress in the numerical simulation and the longer peak duration, shown for impact X in figure 3.20. In figure 3.21 it is shown the initially small area of the foam that effectively absorbs the energy highlighted with a circle while the ellipse highlight the gap where the comfort liner should be modelled to increase the EPS foam effective area.

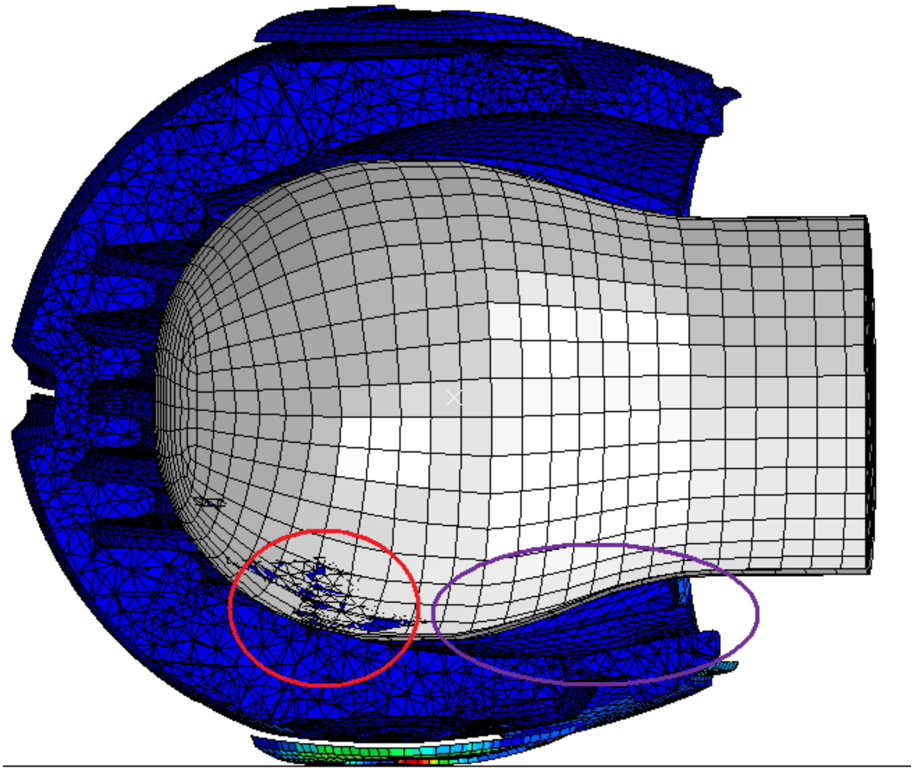


Figure 3.21: *Impact Point X - drawback of the comfort foam absence - cut view of coronal plane.*

3.6 Oblique impact

The most frequently sustained severe injuries in motorcycle crashes are injuries to the head, and many of these are caused by rotational forces [464]. Many of these injuries are caused by rotational forces [128] that are most commonly generated as a result of oblique impacts found in motorcycle crashes [462].

As seen in section 2.2.3, rotational force to the head results in large shear strains arising in the brain, which has been proposed as a cause of traumatic brain injuries like DAI by the tearing of neuronal axons in the brain tissue and SDH by rupturing bridging veins [128, 243]. Thresholds were proposed in these studies and many others reviewed in section 2.2.4. However, many of these thresholds was proposed based on pure rotational motion and DiMasi *et al.* [167] and Ueno and Melvin [168] concluded that these thresholds probably have to be decreased, if an angular motion is combined with a translational motion, which is typical in oblique impacts.

In the current helmet standards tests, no rotational effects are measured in the headform, partly because there are no accepted global injury thresholds for a combination of rotations and translations and there are no realistic test capable of reproduce impacts similar to the most commonly observed impacts in real life motorcycle accidents. Nevertheless, Aare and Halldin [487] recently proposed a new method to test helmets for oblique impacts, but they did not propose any injury tolerances for such a test. Later, in a following work, Aare *et al.* [464] proposed a threshold based on both types of motion, where the authors concluded that this threshold reviewed in section 2.2.3, make good predictions of brain injuries.

All of this considerations and indicated studies influenced this work, relatively to the oblique impact simulated with the previously validated helmet. In this study, rotational and translational parameters from a well-defined and commonly observed impact in real motorcycle accidents [462] was simulated, in order to compare these parameters against different thresholds proposed by several studies and thus, assess an approved ECE R22.05 helmet from the biomechanical point of view.

The impact location is defined to the top of the head, the same region impacted previously to validate the model. However, now is induced rotation at the sagittal plane to simulate the oblique impact. This impact scenario was one of the most common impacts found in real motorcycle crashes by Otte *et al.* [462]. However, the most frequent impact was found at lateral side of the head, in temporal region. This one was not simulated due to some differences between numerical and experimental results of the impact in this region previously simulated to validate the helmet model. On the other hand, as was seen, the numerical results of an impact to the top of the head are similar to the ones obtained experimentally. Also, the majority of the brain injury thresholds were proposed based on this type of impact, with rotation on this plane.

The oblique impact velocity of 7 m/s with a impact angle of 30° is used for the oblique impact simulation, where the impact angle of 30° was induced by altering the speed of the head and the speed of the impactor. The vertical impact speed (v_v)(of the head) and the horizontal impact speed (v_h)(of the anvil) as proposed by Aare *et al.* [464] for the same configuration are 3.50 m/s and 6.06 m/s, respectively. This configuration is presented in figure 3.22.

To perform the oblique impact simulation, only these changes were made to the simulation setup used in the previously helmet validation.

3.7 Results and discussion

The results of interest in this work are presented and discussed in the following paragraphs. Despite the motorcycle helmet model has already been validated, the results used to validate the model are assessed from the biomechanical point of view, by comparing these results against proposed head injury thresholds, based of course in linear motion. At a second stage, the results from the oblique impact are assessed in the same way. However, not only head injury thresholds based on linear motion are assessed. In addition, head injury thresholds based on rotational motion and others based on both motions are used to predict resulting injuries from the oblique impact.

For the impacts performed to validate the model, only PLA and HIC are used as thresholds, due to the fact of only translational acceleration was recorded as in a typical energy absorption test of ECE R22.05 standard.

In the case of oblique impact, several thresholds are used, such as the peak of rotational acceleration, the change in rotational velocity and its peak, the combination of these with PLA and HIC and other methods based on same concept of rotational and translational combined effects, such as HIP, GAMBIT and even the one proposed by Aare [464] that pre-

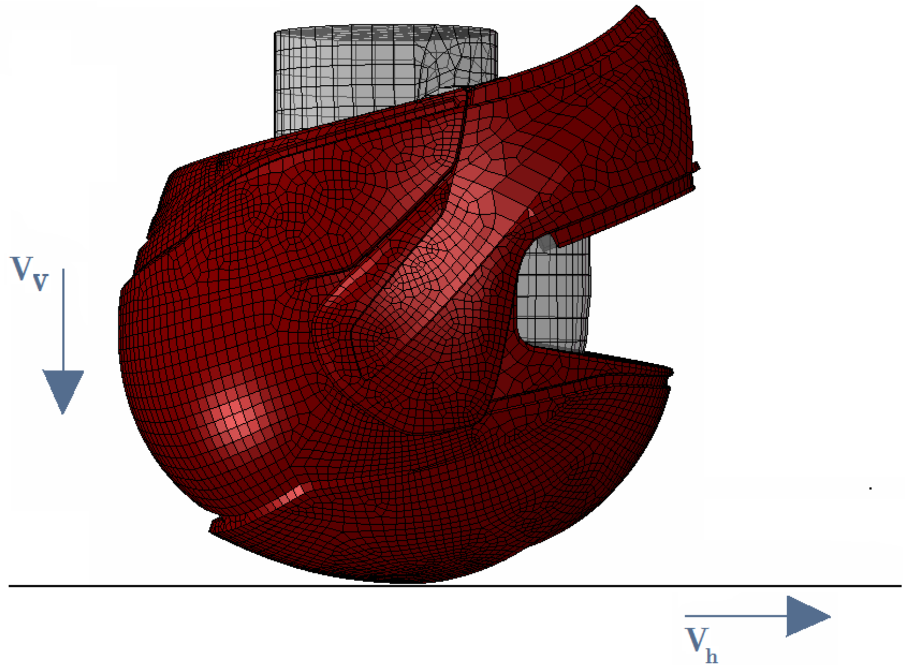


Figure 3.22: *Oblique impact.*

dicts the maximum strain in the brain tissue. Figure 3.23 shows the magnitude of resultant linear velocity, at the middle of the oblique impact, where is possible to observe the presence of rotational motion, because comparing the position of the helmet with the initial position observed in figure 3.22 is concluded that rotated. Also, observing the resultant linear velocity, it is clear that the red regions that corresponds to higher values are present at the points more far from the centre of rotation. Thus, from the distribution is possible to conclude that both, translational and rotational acceleration are present.

3.7.1 Energy absorption tests according to ECE R22.05 standard

These impacts were previously validated in the section 3.5, where the results are given in the table 3.5 and the translational acceleration curves are shown in the figures 3.17, 3.18, 3.19 and 3.20.

Comparing these results against the translational acceleration based injury criteria reviewed in the section 2.2.4, some injuries were predicted. The limit of 80 g for a duration that shall not pass 3 ms [209, 242, 261] to not occur any type of head injury was exceeded for all impacts. Mertz *et al.* [241] estimates a 5% risk of skull fractures for a peak acceleration of about 180g, and for all of the impacts this value was exceed. More recently, King *et al.* [19] estimated a MTBI tolerance for head linear acceleration of 98,4 g and for HIC_{15} of 333 to a probability of 75% of MTBI occurrence and again these were exceeded for all the impacts.

Hopes and Chinn [59] indicated that there is an 8.5% probability of death at an HIC value of 1000 and 31% at 2000. Thus, for impacts B and P, exist a minimum probability of 8.5% of death, while for the impact point R, exists a minimum probability of 31%. For the impact point X, numerical results showed a prediction above the 31%, however experimental results indicate that the probability is somewhere between 8.5% and 31%. This difference was already justified previously during the helmet model validation.

The HIC value is greater than 1000 for all the impacts and comparing this with the

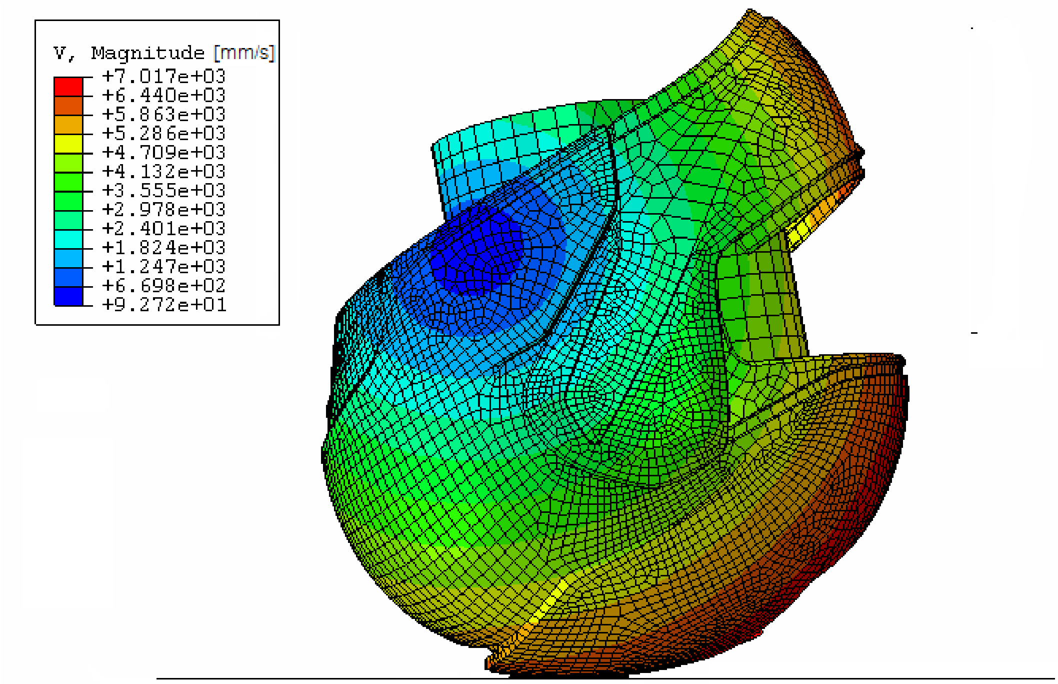


Figure 3.23: *Results from the oblique impact.*

conclusions of Horgan [442] there is a minimum probability of 16% of life threatening injuries occurrence.

Nevertheless, these tests are made by current standards to evaluate the helmet's protective capacity and do not reproduce an impact in a real accident, mainly because it is a pure translational impact, which is almost impossible or unlikely to occur in a real crash. This leads to high values of translational components, since there is no angular component as concluded by Ueno and Melvin [168] and DiMasi *et al.* [167]. Nevertheless, this section is important to conclude this same idea, by comparing next against the translational components present in the oblique impact. Also, it was not found any study in the literature that correlates brain injuries such as DAI with only translational thresholds contrary to what is found in the next section.

3.7.2 Oblique impact

The results obtained from the oblique impact simulation and used in the comparison against proposed head injury thresholds are given in the table 3.6.

Table 3.6: Head injury thresholds obtained and computed from the oblique impact results.

$a_{max}[g]$	$\alpha_{max}[rad/s^2]$	$\Delta\omega[rad/s]$	**HIC**	**HIP [W]**	**GAMBIT**	$\varepsilon_{brain\ tissue}$
88.47	7299.20	34.28	351.31	24098.63	0.81	0.2151

In the figure 3.24 is shown the translational acceleration, the rotational acceleration and the rotational velocity found during the impact. This figure is useful to the following injury assessment with thresholds that have time dependence (typically a pulse duration).

The conclusions of Ueno and Melvin [168] and DiMasi *et al.* [167] were verified. The linear results of this impact, such as PLA and HIC were considerably lower than the ones

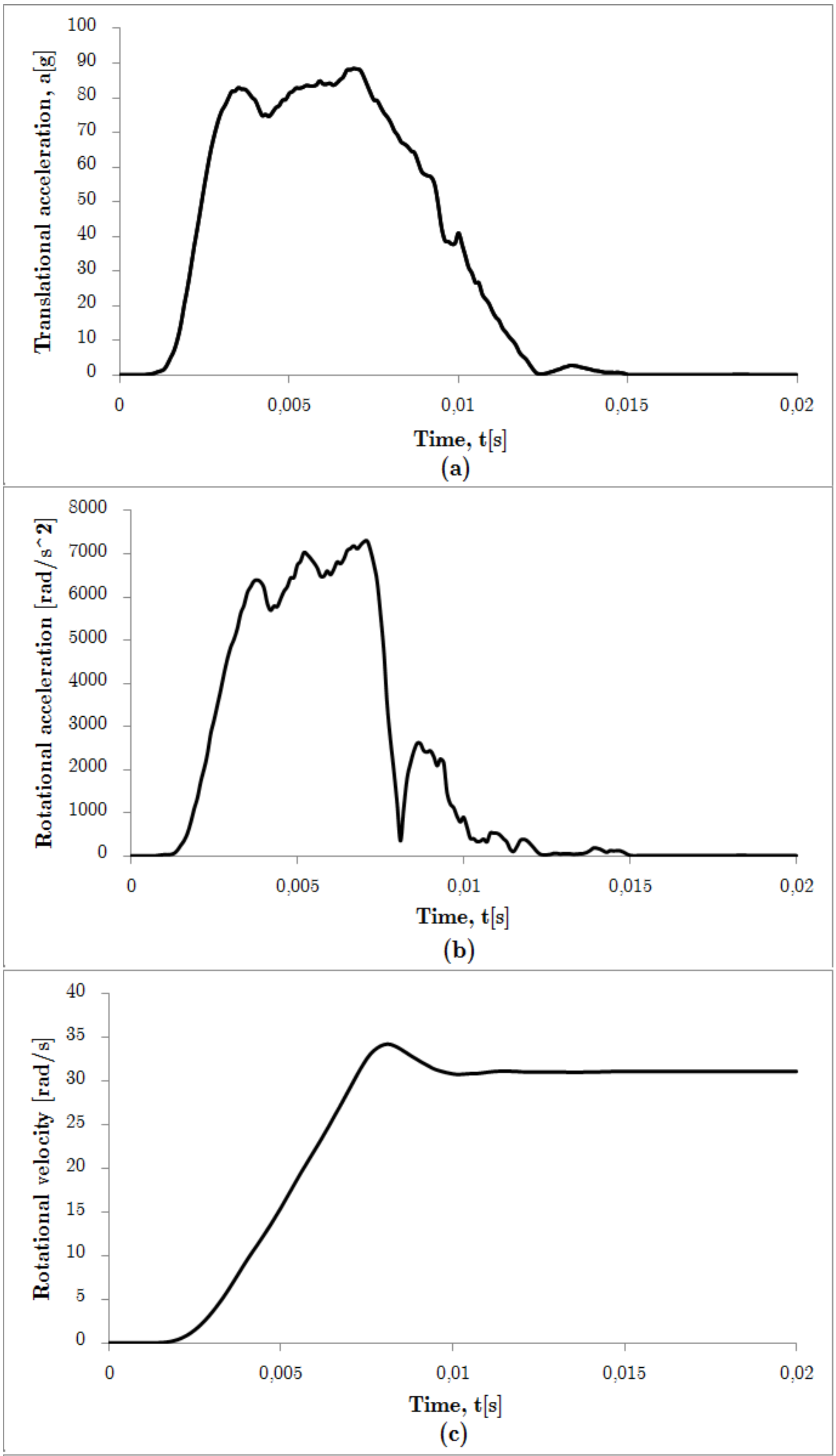


Figure 3.24: *Results of the oblique impact simulation: (a) translational acceleration, (b) rotational acceleration and (c)rotational velocity.*

find in the linear impacts. This is justified by the introduction of an angular component. Also, the angular component induced in this impact is much greater due to the angle of impact chosen, or in other words, the greater velocity of the anvil relatively to the velocity of the helmet-head system.

Peak translational acceleration, peak rotational acceleration and change of rotational velocity
The results given in the table 3.6 and shown in the figure 3.24 were compared with the thresholds proposed in the section 2.2.4 and a summary of the limits that were exceeded is presented:

- Accordingly to the work performed by COST-327 Motorcyclist's helmet working group [14], it was found that a 50% probability of head injury with AIS 3 results from 10 krad/s^2 and 0 g linear acceleration down to 0 krad/s^2 and 190 g. When compared to the results of 88.47 g and 7299.20 rad/s^2 found in this study, it is verified that such values fits in those intervals, ans such prediction is then applicable to this study.

- Gennarelli *et al.* [317] hypothesized a magnitude of rotational acceleration of 2877.8 rad/s^2 required to induce mild cerebral concussion at an angular velocity of 25 rad/s and both limits were exceeded in this study.

- Ommaya *et al.* [246] proposed a 50% probability of cerebral concussion for a maximum rotational acceleration of 1800 rad/s^2 during a period of time inferior to 20 ms.

- Löwenhielm [244, 364] predicted rupture of bridging veins when values of rotational acceleration or velocity higher than 4500 rad/s^2 or 50 - 70 rad/s respectively, are found. Here, it was found higher values for rotational acceleration.

- Advani *et al.* [257] predicted brain superficial tissue shearing for rotational accelerations higher than 2000 - 3000 rad/s^2, which was exceeded here too.

- Ommaya [365] predicted several brain injuries for rotational velocities higher than 30 rad/s where the rotational acceleration of 4500 rad/s^2 is exceeded too. Such brain injuries were classified as AIS 5, which corresponds to a set of severe injuries like EDH, SDH and also DAI.

- Newman [226] proposed a probability of 50% for cerebral concussion for rotational accelerations higher than 6200 rad/s^2, which was excceded in this study.

- Thomson *et al.* [267] predicted the occurrence of brain injuries when rotational accelerations higher than 5000 rad/s^2 are found. This was found in this study.

- King *et al.* [19] proposed several thresholds for MTBI. HIC and rotational acceleration were presented as thresholds where a probability of MTBI occurrence of 75% was proposed when values higher than 333 and 7130 rad/s^2 respectively, are found. In this work, both were exceeded.

- Zhang *et al.* [263] proposed a 50% probability of MTBI for a maximum rotational acceleration of 5900 rad/s^2, again exceeded in this study.

- Viano *et al.* [343] presented a limit for the magnitude of rotational acceleration of 6400 rad/s^2 required to induce cerebral concussion at angular velocities higher than 35 rad/s. The peak for roational acceleration proposed was clearly exceeded, however the rotational velocity found in this study is 34.28 rad/s, which is slightly inferior to the limit proposed by Viano *et al.* [343]. Nevertheless, the difference is minimal, and at least a mild cerebral concussion can be considered.

- More thresholds proposed for cerebral concussion were exceeded. Fijalkowski *et al.* [316] and Kleiven [441] predicted cerebral concussions for values of rotational acceleration higher than 6200 rad/s^2 and 1800 rad/s^2, respectively. Once again, the limits for cerebral concussion were exceeded.

- Recently, Davidsson *et al.* [347] suggested that the threshold for DAI is a rotational velocity change of 19 rad/s, in an exposition to sagittal plane rotation, which was exceeded.

The great majority of these head injury thresholds were determined in studies where the rotational motion was induced to the sagittal plane, exactly as is done in this study, which makes these comparisons even more valid. From this comparison, it is possible to conclude that these thresholds can be a good basis to predict concussion. From all the literature reviewed, several types of injuries were predicted. However, there was not a good agreement for some injuries. For example, it was predicted bridging vein ruptures for more than one threshold but SDH that almost always is associated with bridging veins was not predicted. DAI was also not predicted with such thresholds always higher than the values found in the simulation results. However, for the thresholds given in the literature, MTBI was almost always predicted and concussion was too, several times. Besides the differences between the studies, some were performed experimentally on animals or cadavers, others through numerical simulations, several studies proposed thresholds for cerebral concussion and all of them were exceeded. Thus, it can be said that at least cerebral concussion is predicted for the oblique impact simulated in this work with a motorcycle helmet homologated by the European standard, ECE R22.05. Moreover, concussion was considered the most common head injury diagnosis resulting from motorcycle and moped accidents [133]. However, this is not a very severe injury.

GAMBIT

The head injury assessment function GAMBIT, introduced by Newman [218] was computed. This criterion takes into account both translational and rotational acceleration, which makes it a useful tool in oblique impacts.

GAMBIT is obtained by using equation 2.6. By replacing the output accelerations of the simulation in the equation 2.6, a GAMBIT value of 0.81 was calculated. The constants used in the expression 2.6 are the more recent ones indicated in section 2.2.4.

More recently, Newman [226] reported a 50% probability of concussion for a GAMBIT value superior to 0.4, which is half than the value calculated in this study. This means that there is a minimum probability for concussion occurrence of 50%. GAMBIT has recently been employed in some studies involving protective headgear [227, 14], where it was highlighted the good capability of this method for predicting cerebral concussion.

Again, concussion was predicted using a different criterion that takes into account both motions.

HIP

Another criterion computed in this study that considers both translational and rotational motion is the HIP, proposed by Newman *et al.* [226]. Newman *et al.* [226] reasoned that the rate of change of translational and rotational kinetic energy, i.e. power, could be a viable biomechanical function for the assessment of head injury. HIP is computed by using the equation 2.7, where the constants (mass and principal moments of inertia) are substituted by the values given in the table 3.3 and the remaining variables of each principal direction are substituted by the output obtained in the simulations. Thus, the HIP value calculated is 24098.63 W, which is twice the value proposed by Newman *et al.* [226] for 50% probability of concussion at a maximum Head Impact Power (HIP_{max}) of 12.8 kW. However, this threshold was proposed with the suggestion of the utilization of average values for a 50th percentile adult male while in this study a size M headform is used and not the medium size J. This clearly affects the results because the mass and accelerations are higher and greater values of HIP are expected. Thus, there is no validity in establishing a comparison with this limit.

Brain tissue strain

Other criterion computed is the criterion developed by Aare *et al.* [464] which correlates both translational and rotational acceleration with strains in brain tissue, by considering that strains in the brain tissue are proportional to the HIC value for pure translations and also proportional to the change in rotational velocity for pure rotations of short impact durations. To calculate the maximum strain value (ε), it is used the equation 2.8, where HIC is computed by using the equation 2.3, the change in rotational velocity is given in the table 3.6 and the constants k_1 and k_2 are equal to 6.14×10^{-3} and 1.32×10^{-5}, respectively. These constants were suggested by Aare *et al.* [464] in an attempt to simulate the critical scenarios and were obtained by regression analysis for the same oblique impact simulated in this study with a regression coefficient (R^2) of 0.93 which represents a good match.

Thus, the strain in brain tissue is 21.51%, which is slightly greater than the limit of 20% proposed by Aare *et al.* [464] and predicted by Bain and Meaney [195] as critical to the brain tissue that could cause injuries such as DAI. This same threshold is presented by Morrisson *et al.* [358]. A lower value of 15% was proposed by Thibault *et al.* [282] as a critical level to brain strain. Also, Kleiven [441] proposed strain brain tissue as a predictor of concussion and DAI suggesting 0.1 and 0.2 as thresholds, respectively. All these thresholds were exceeded in this study, predicting the occurrence of brain injuries. In a similar study performed by Aare *et al.* [464], it was found that the change in rotational acceleration was the parameter that correspond to a better correlation with the intracranial strains. This was also concluded by Kleiven and von Holst [171].

Thus, it can be said that damage to the brain tissue really occurs in this impact. The limit proposed between the different studies is very similar, and some agree that it can cause concussion and a more severe injury, DAI. There is a high confidence in these thresholds that present similar results and conclusions, and each one of them resulted from different methodologies, since experimental tests, to real world accidents investigations, FE analysis with FE head models, etc.

Therefore, it can be concluded that in real accidents, a motorcyclist wearing an approved helmet can suffer brain injuries such as cerebral concussion and brain damage (in the basis of DAI, brain tissue strain exceeded), from an impact like the one simulated.

3.8 Reproduction of ECE R22.05 impact P with SUFEHM

An additional task is performed in this work. In this section is presented numerical results obtained with an advanced biomechanical FE head model to simulate an impact closer to the real conditions instead of a rigid headform. The anatomical head finite element model used for this study is the SUFEHM developed by Kang *et al.* [220]. This biomechanical head model will allow a further accurate computational-based prediction of brain injuries. More details about the development of this FEHM are described in section 2.2.5. In the next sections is presented the SUFEHM model, its characteristics and validation, the head injury criteria related to this model and finally, the results. This procedure was done in cooperation with the biomechanical group of Strasbourg University, since this FEHM was developed by them.

3.8.1 SUFEHM model

Figure 2.38 shows a cross section of the model and illustrates the anatomical features of the skull and the brain. The main anatomical features modelled were the skull, face, falx, tentorium, subarachnoid space, scalp, cerebrum, cerebellum, and the brainstem.

The geometry of the inner and outer surfaces of the skull was digitised from a human adult male skull. The data given in an anatomical atlas by Ferner *et al.* [489] was used to mesh the human head using the Hypermesh code. For this model, the option was chosen to retain a given realistic human adult anatomy rather then trying to find an overage geometry, which may not exist. Figure 3.25 shows the 3D skull surface obtained by digitising external and internal surfaces of the skull as well as the meshed model.

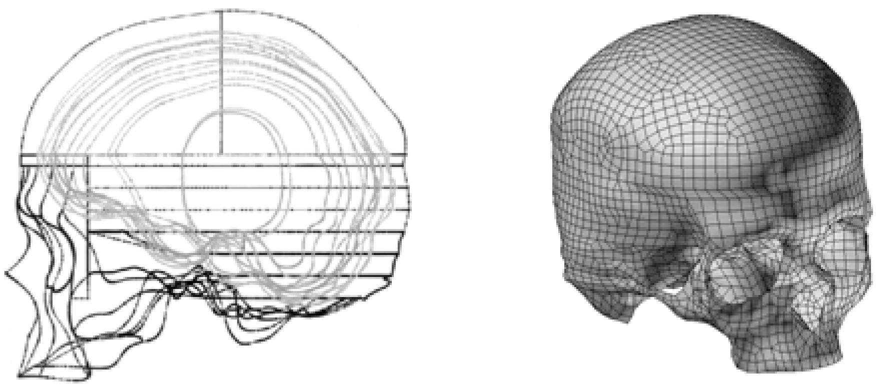


Figure 3.25: *3D skull surfaces used for the model construction and skull meshing.*

The finite element mesh is continuous and represents an adult human head. The falx and tentorium were simulated with a layer of shell elements, the skull comprised a three layered composite shell and the remaining features were modelled with brick elements. Of particular importance is the subarachnoid space between the brain and the skull which was represented by one layer of brick elements to simulate the CSF. Lagrange formulation was therefore selected and the brain-skull liaison was modelled by an elastic material validated against the in-vivo vibration analysis. The scalp was modelled by a layer of brick elements and surrounds the skull and facial bone (shell elements). Globally, the SUFEHM model consists of 13208 elements. More details about the mesh of each head component are given in the table 3.7.

Mechanical properties

Table 3.7 lists the properties of the materials used, where e is the thickness, ρ is the density, E is the Young's modulus, ν is the Poisson's coefficient, G_0 is the short term shear modulus, G_∞ is the long term shear modulus, β is the decay constant, K is the bulk modulus, UTS is the ultimate tensile strength and UTC is the compressive one. Material properties of the

CSF, scalp, facial bones, tentorium and falx are all isotropic, homogeneous and elastic. The Young's modulus of the subarachnoid space was found by Willinger *et al.* [393] by modal analysis. The viscoelastic properties assigned to the brain were scaled from Khalil *et al.* [419] and correspond to recent works published by Kruse *et al.* [490] on in vivo human brain mechanical properties. The behaviour in shear was defined by the following expression:

$$G(t) = G_\infty + (G_0 - G_\infty)\exp(-\beta t) \tag{3.1}$$

with G_0 the short term shear modulus, G_∞ the long term shear modulus and β the decay constant.

The skull was modelled by a three layered composite shell representing the inner table, the diplöe and the external table of cranial bone. In order to reproduce the overall compliance of cranial bone, a thickness in combination with an elastic brittle law were selected for each layer. In order to model the material discontinuity in the case of fracture, it was necessary to use values for the limiting (ultimate) tensile and compressive stress obtained by Piekarski [491] and integrated in the Tsaï-Wu criterion. The total mass of the SUFEHM model is 4.5 Kg.

Table 3.7: 3D skull surfaces used for the model construction and skull meshing.

Part	Mesh	Mechanical law	Mechanical parameters	Mechanical parameters
Membranes (falx and tentorium)	471 shell elements	Linear elastic	e=1 mm ρ=1140 kg/m^3 E=31.5 MPa ν=0.45	- -
CSF	2591 brick elements	Linear elastic	ρ=1040 kg/m^3 E=12 kPa ν=0.49	-
Brain (brainstem, cerebellum, cerebrum)	5508 brick elements	Viscoelastic	ρ=1040 kg/m^3 K=1125 MPa G_0=49 kPa G_{inf}=167 kPa β=145 s^{-1}	-
Skull	1813 shell elements	Composite law with failure criteria	<u>Cortical</u>: e=2 mm ρ=1900 kg/m^3 E=15 GPa ν=0.21 K=6.2 GPa UTS = 90 MPa UTC = 132 MPa	<u>Diplöe</u>: e=3mm ρ=1500 kg/m^3 E=4.6 GPa ν=0.05 K=2.3 GPa UTS=35 MPa UTC=25 MPa
Face	529 shell elements	Linear elastic	e = 10 mm ρ=2500 kg/m^3 E=5000 MPa ν=0.23	-
Scalp	2296 brick elements	Linear elastic	ρ=1000 kg/m^3 E = 16.7 MPa ν=0.42	-

SUFEHM validation

The SUFEHM model has been validated over the years, since it was developed until now. In the section 2.2.5, it was already done a short summary of different validation stages of this model. Even so, it is presented here more details SUFEHM validation, the FEHM used in this study.

A total of eight instrumented cadaver impacts were reconstructed with the objective of validating the SUFEHM's model under very different impact conditions SUFEHM's model was validated in terms of intracerebral pressure field by reproducing Nahum *et al.* [192] and Trosseille *et al.* [409] experiments, and in terms of deformation and rupture of the skull by reproducing Yoganandan *et al.* [255] tests.

Nahum's validation

The experimental data used in order to validate brain behaviour were published by Nahum *et al.* [192] for a frontal blow to the head of a seated human cadaver. For this impact configuration, a 5.6 kg rigid cylindrical impactor launched freely with an initial velocity of 6.3 m/s generates an interaction force and a head acceleration characterised by their peak values which are respectively 6900 N and 1900 m/s^2 over a duration of 6 ms. Intracranial pressures were also recorded in this test, at five well defined locations: behind the frontal bone, adjacent to the impact area, immediately posterior and superior to the coronal and squamosal suture, respectively in the parietal area, inferior to the lambdoidal suture in the occipital bone (one in each side), and at the posterior fossa in the occipital area. In order to reproduce the experimental impact conditions, the anatomical plane of the SUFEHM was inclined about 45°, as shown in Figure 3.26, like in the Nahum's experiment. For modelling a direct head impact, the SUFEHM was frontally impacted by a 5.6 kg rigid cylindrical impactor with an elastic padding launched freely with an initial velocity of 6.3 m/s.

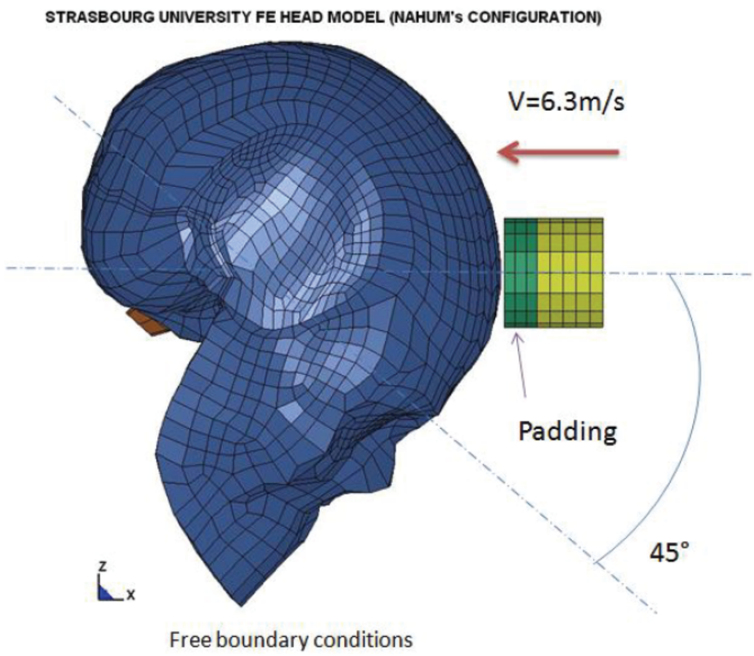


Figure 3.26: *Nahum's configuration.*

The comparison of numerical and experimental forces is shown in figure 3.27 for the Nahum's impact. A good agreement for the impact force was found as the time duration of impact and the amplitudes were well respected. The comparison of pressure time histories between numerical and experimental data is presented in figure 3.27b, c, d and e for the Nahum's impact simulation. As shown in these figures, five intracranial pressures from the model matched the experimental data very well. The maximum difference of pressure peak is under 10%.

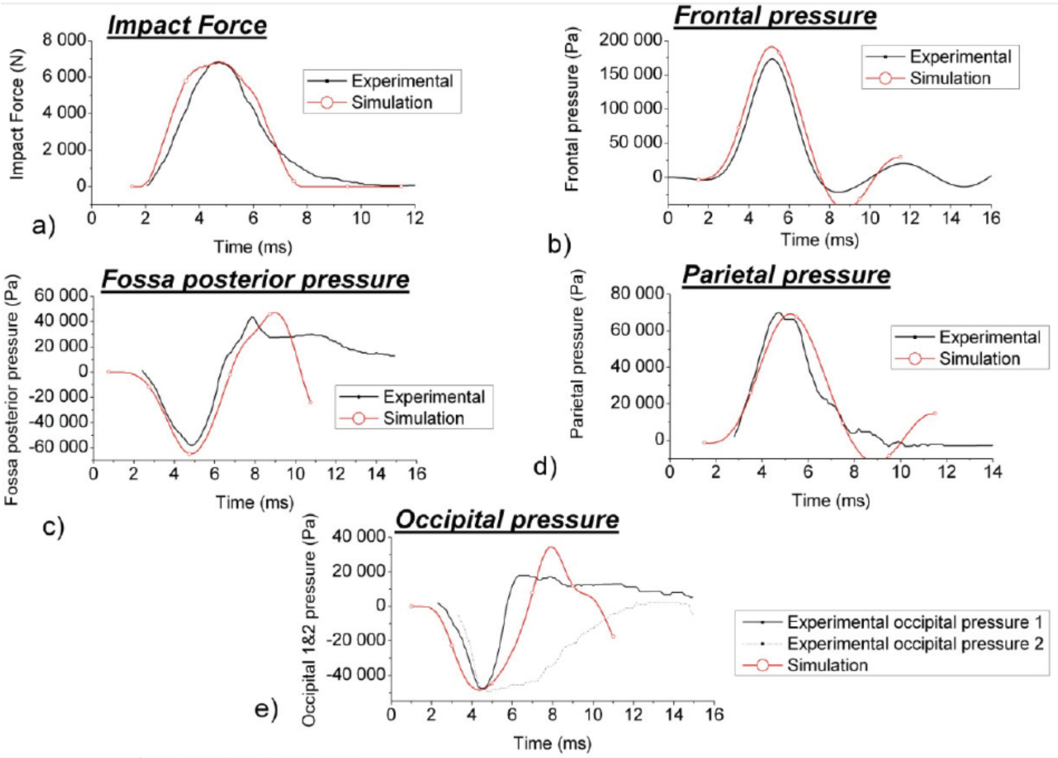


Figure 3.27: *Experimental and numerical results comparison obtained for a Nahum's impact in terms of interaction force (a), frontal pressure (b), Fossa posterior pressure (c), parietal pressure (d) and occipital pressure (e).*

Trosseille's valdiation

Simulations were based on the MS428-2 case involving an impact to the face by a 23.4kg steering wheel with an initial velocity of 7m/s. Because of the long duration of the impact and the influence that the neck would have on the head motion the head impact conditions were simulated by applying measured head kinematics from test case MS428-2 to rigid modelled skull in the model. Validation has been done in terms of:

- Frontal intra-cranial acceleration;
- Lenticular intra-cranial acceleration;
- Occipital intra-cranial acceleration;
- Frontal pressure;
- Lateral pressure;
- Occipital pressure;
- Temporal pressure;
- Ventricular pressure.

Yoganandan's validation

Experimental tests carried out by Yoganandan *et al.* [255] has been used in order to validate the ability of the human head finite element model to predict a skull fracture. One of the impact configurations is shown in Figure 3.28. The surface of the impactor was modelled by a 96 mm diameter rigid sphere. Initial conditions were similar to the experimental ones i.e. a mass of 1.213 kg with an initial speed of 7.1 m/s. The base of the skull was embedded as in the experiment. For the model validation, the contact force and the deflection of the skull at the impact site, were calculated. The numerical force-deflection curves are compared to the average dynamical response of experimental data, which is shown in figure 3.29. The dynamical model responses agree well with the experimental results, both the fracture force and the stiffness level.

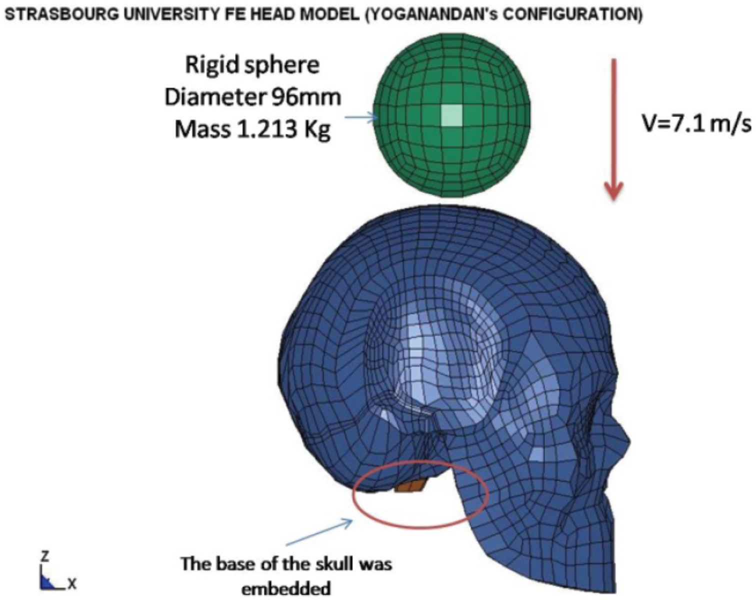


Figure 3.28: *Yoganandan's configuration.*

3.8.2 SUFEHM related head injury criteria

In order to establish some human head tolerance limits to specific injury mechanisms, some real world accident cases were reconstructed numerically. Altogether 68 head impact conditions that occurred in motor sport, motorcyclist, American football and pedestrian accidents were reconstructed with the SUFEHM model. A summary of the type and number of accident reconstruction is given through table 3.8.

Table 3.8: Summary of the type and number of accident reconstructions.

Accident Type	Number of cases
Motorsport accidents	6
Motorcycle accidents	11
American football accidents	22
Pedestrian accidents	29

The head trauma database considered involved no less than 68 real world head trauma coming from:

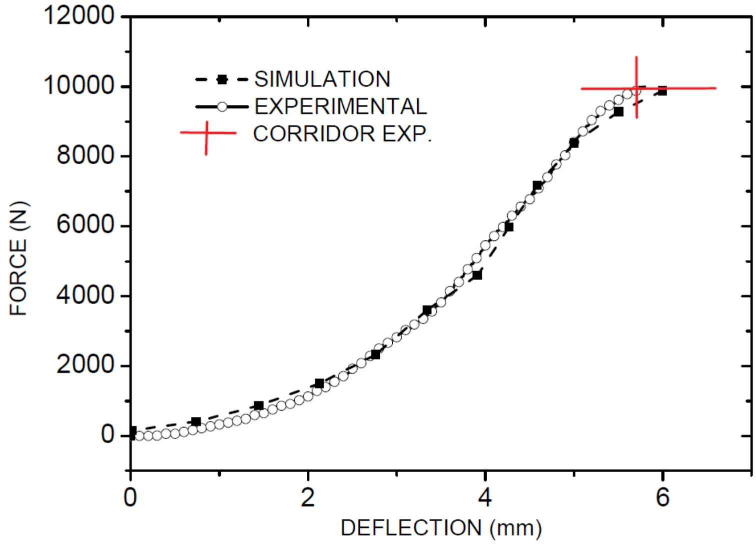


Figure 3.29: *Experimental versus simulated force deflection curves until fracture (+ gives the corridor of Yoganandan's experimental results).*

- COST 327 project (motorcyclists). The motorcyclist accidents are those described in Chinn *et al.* [147]. They were experimentally reconstructed in collaboration between Strasbourg University, the Transport Research Laboratory (TRL) and the Glasgow Southern General Hospital. The acceleration field sustained by the head during the impact was then inferred experimentally by using an instrumented Hybrid III dummy head which was fitted inside a new helmet similar to the one worn by the victim. Head and helmet were thrown at different velocities against different kinds of anvils in order to reproduce on the new helmet the same damages as those observed on the victim's helmet.

- BIOKINETICS (American football players). The football player accidents are those described in Newman *et al.* [218, 226]. In American football games, two cameras have been used in order to determine the relative position, orientation and velocities between the helmeted head of two players when colliding together. Then, the scene has been replicated experimentally thanks to two helmeted Hybrid III dummy heads. The validation of this method is based on the rebound of the full body dummies after the experimental replication compared to the filmed rebound of the football players' bodies.

- EU - APROSYS project (pedestrians). The pedestrian accidents are those reconstructed from the database of the Accident Research Unit of the Medical University of Hanover. These are all accidents with a main impact (i.e. the supposed injurious impact) consisting on a head hit by the middle of a windscreen. A great variety of parameters were collected on the accident scene and were used as the inputs of an analytical rigid body study in order to infer the kinematics of the pedestrian body until the impact of the head. For each case, the results of this simulation are compared to the damages observed on the car and to the wounds sustained by the victim. In order to obtain the head acceleration field, a numerical replication using a finite element model of a windscreen and of a Hybrid III head was then performed.

The reconstructions involved applying the motion of the head from the accidents to the rigid skull of the SUFEHM. For the statistical analysis the injuries for the accident data were categorised into the following types and levels based on the details of the medical report from each accident case:

- DAI: DAI cases covered all incidences in which neurological injuries occurred and covered concussion, unconsciousness and coma. Incidences of DAI were broken down into mild and severe levels according to coma duration (<24H for moderate DAI and >24H for severe DAI)
- SDH: This category of injuries covered all incidences in which vascular injuries with bleeding were observed between the brain and the skull.

Methodology used to reconstruct numerically accidents is presented in figure 3.30.

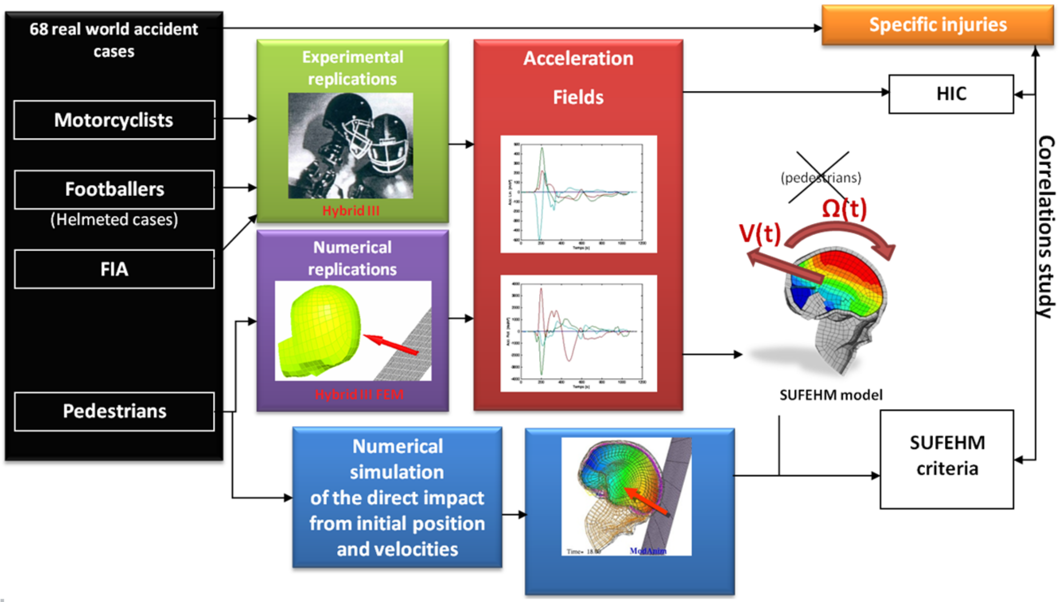
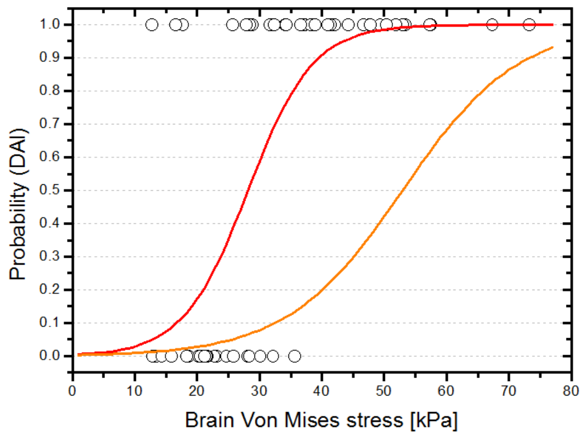


Figure 3.30: *Accident reconstructions methodology.*

Results computed with the SUFEHM's model were reported in terms of correlation coefficients (Nagelkerke R-Squared values) in order to express their injury prediction capability. Based on SPSS method it appears that DAI are well correlated with intra-cerebral von Mises stress. Coming to maximum R^2 values, the maximum von Mises stress conducts to 0.6 and 0.39 for respectively moderate and severe neurological injury. The thresholds for this parameter are of the order of 28 and 53 kPa respectively for moderate and severe neurological injuries as it appears in the injury risk curves reported in table 3.31. Concerning the SDH injuries CSF strain energy were considered. With the SUFEHM it was shown that the best correlation with SDH was the maximum strain energy within the CSF, with a R^2 value of 0.465 and a threshold value of about 4950 mJ. After the analysis of regression correlation method table 3.31 and 3.32 report the tolerance limits and the injury risk curves obtained with the SUFEHM for each of the injury types with an injury risk of 50%.

3.8.3 Results from the vertex impact.

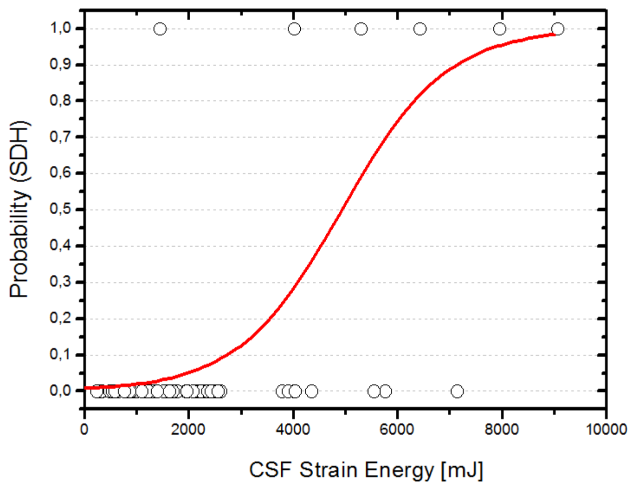
In this section, it is presented numerical results obtained with the SUFEHM model in terms of brain von Mises stress and CSF strain energy by introducing experimental headform



	Mild DAI	Severe DAI
Brain Von Mises stress [kPa]	28	53

Tolerance limit at 50% risk of DAI

Figure 3.31: *Tolerance limits calculated for DAI injuries (mild and severe) with the SUFEHM's model under LS-DYNA software and Best fit regression models for DAI injury investigated for the SUFEHM considering brain von Mises stress.*



	SDH
CSF strain energy [mJ]	4950

Tolerance limit at 50% risk of SDH

Figure 3.32: *TTolerance limits calculated SDH injury with the SUFEHM's model and LS-DYNA software and best fit regression models for SDH injury investigated for the SUFEHM's model considering CSF strain energy.*

acceleration field at the center of gravity of the SUFEHM. An illustration is proposed through figure 3.33. The linear resultant head acceleration has been used in order to drive SUFEHM's model. Results obtained with the SUFEHM's model in terms of intracerebral field's pressure and von Mises stress are illustrated in figure 3.34 and figure 3.35. The maximum von Mises stress is located between the brain and the cerebellum and between the cerebellum and the brainstem while the maximum pressure is located in the vertex area. Table 3.9 summarizes results obtained with the SUFEHM's model based on experimental input. Results are given in terms of brain von Mises stress value (maximum) and CSF strain energy. The assessment of head injury risk is expressed in terms of percentage. We can conclude for a vertex impact that head injury risks exist with a high risk of moderate neurological injuries (90%) and 30% risks to obtain a SDH.

From the results it is clearly the high risk of brain injury occurrence, which reveals again, defects in the current standards, more specifically in their tests specifications and criteria to

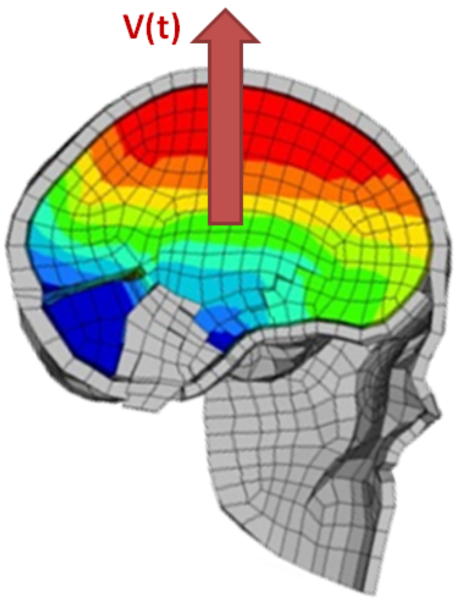


Figure 3.33: *Illustration of the imposed acceleration in the SUFEHM model with acceleration curve coming from experimental test.*

Table 3.9: Results obtained with SUFEHM in terms of percentage risk (Based on experimental acceleration recorded).

Impact Point	**Brain Von Mises Stress [kPa]**	**% Risk of moderate DAI**	**% Risk of Severe DAI**	**CSF strain energy [mJ]**	**% Risk of SDH**
P	39.7	90.2	19.3	4097	30.3

injury assessment. There is a considerable risk for occurrence of severe DAI and SDH, and an almost certain injury is moderate DAI with a risk of 90.2%.

In other works, other limit values are proposed based on the same criteria, von Mises Stress and CSF strain energy. Baumgartner *et al.* [266] considered intra-cranial von Mises stress a good injury criterion for concussion or other mild traumatic brain injuries when reaching values above 15 kPa. Willinger *et al.* [265]indicated von Mises stress values for the brain resulting in concussion was estimated to be above 20 kPa, which is higher than the limit deduced from Baumgartner *et al.* [266]. Deck *et al.* [6] proposed a maximum value of 40 kPa for von Mises stress indicating that is a good indicator of concussion. Again, concussion arises as a possible scenario.

Thus, even reproducing the same impact that is assessed by the ECE R22.05, it is evident that a certified helmet can't protect the user from brain injuries, mainly DAI and possibly concussion. It is also observe a considerable risk for SDH. This shows the incapacity of a good injury assessment from the utilisation of a rigid headform. Mainly due to this issue, FE head models, such as SUFEHM acquired great prominence. There are also solutions to the experimental procedures such as anatomical headforms as the one developed by Bosch [36].

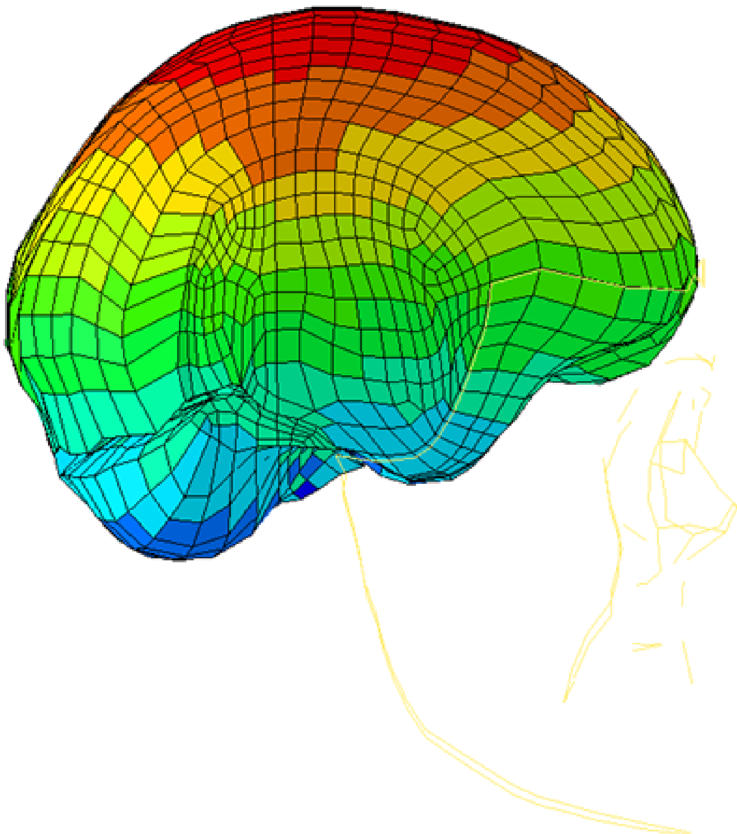


Figure 3.34: *Illustration of the Brain pressure field for an impact in P location.*

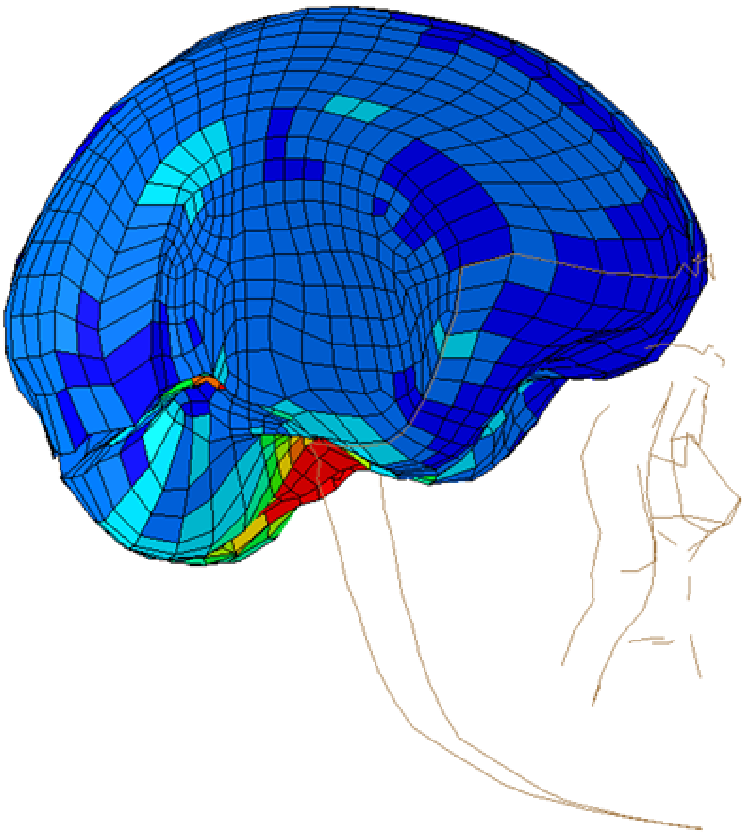


Figure 3.35: *Illustration of the Brain von Mises stress field for an impact in P location.*

Chapter 4

Conclusions

This chapter presents general conclusions and discuss the results obtained in this work. It also has some suggestions and ideas to implement in future works related to this study.

4.1 Conclusions

The main objective of this work was to assess the head injuries that possibly result in a motorcycle crash.

At a first stage, it was developed a FE motorcycle helmet model in order to be possible to perform FE analysis of impacts with helmet-head systems. A realistic and detailed FE model of a commercial motorcycle helmet was developed. Besides the geometric modelling, it was also modelled the constitutive models for the multi-density liner and the outer shell. A simple approach was made to the shell material modelling, where this was modelled as an elastic material. On the other hand, EPS foam was modelled with a more complex model in order to reproduce accurately its uniaxial compressive behaviour. Experimental tests (uniaxial compression tests) were performed and the results were compared against the ones obtained from numerical simulations of the uniaxial compression tests, where it was concluded that the material model *Crushable Foam* is suitable to reproduce the real mechanical behaviour of the EPS. Only a slight difference was found between the experimental results and the numerical ones. The EPS foam densify "faster" (for lower strains) than the experimental. This could be explained by some inconsistencies between the material model adopted and the real mechanical behaviour of this foam. Other explanation for this sooner densification can be the type of elements used, because tetrahedral elements are good for the creation of meshes in complex geometries, but sometimes this elements lead to an excessive rigidity. In overall, the material models were found suitable for its application in the helmet FE model.

After all the modelling has been made, the four impacts of the energy absorption tests against flat anvils were simulated accordingly to ECE R22.05 standard and the head acceleration recorded at the centre of gravity (COG) was compared against the experimental data provided by the manufacture. It was also compared the PLA and HIC. Good correlation was found for all the impacts, except one, the impact point X.

The differences between experimental and numerical results at this point may be explained by the adoption of a simplified numerical model regarding to the number of components modelled, because the retention system, the comfort padding and fixations devices were not modelled. Especially for point X, where the differences are more evident, it should be noted that in the real helmet the zone around the X point receives several parts, since metals and plastics, where is the visor locking system and the chin strap fixation system and also the fixation system between the two shell parts, because in reality it is a modular helmet and not a full-face helmet. However, despite some differences between experimental and numerical impact results, the numerical model was considered adequate enough for a preliminary study on biomechanical issues resulting from oblique impacts.

Other explanation for the deviation of numerical results is the absence of comfort padding in the numerical model. This foam is responsible for the fitting between the helmet and the head and may have relevant importance on the impact kinematics. For this model the comfort foam can be neglected to the points B, P and R (little differences between experimental and numerical results, good agreement between the result for this three impact points) but for the impact X it is very important mainly due to the considerable foam thickness in that region and due to the shell geometry in that region.

Another feature that can also be important is the chin strap, specially in the next step that was done in this work, the oblique impact, due to the considerable head rotational motion that could makes the helmet to fall off the head. This impact was performed due to its similarities with the real crashes and its setup in the simulations was based on records of real observed crashes. Thus, the results assessed, can provide information of what really happens in a motorcycle accidents regarding helmet impact. All of the considerations made to this impact were based on the findings of others authors.

A immediately conclusion was made by comparing the results from the impacts perform to validate the helmet model with the ones obtained from the oblique impact. The linear results of the oblique impact, such as PLA and HIC were considerably lower than the ones find in the linear impacts. This is justified by the introduction of an angular component. Also, the angular component induced in this impact is much greater due to the angle of impact chosen, or in other words, the greater velocity of the anvil relatively to the velocity of the helmet-head system. Thus, the conclusions of Ueno and Melvin [168] and DiMasi *et al.* [167] were verified. Moreover, these tests are made by current standards to evaluate the helmet's protective capacity and do not reproduce an impact in a real accident, mainly because it is a pure translational impact, which is almost impossible or unlikely to occur in a real crash. This led to other conclusion that motorcycle helmets are designed to perform for higher levels than the ones verified in real accidents, which could lead for example to stiffer helmets that can induce head injuries.

At a final stage, the results from the impacts were assessed from a biomechanical point of view, specially the ones from oblique impact, where it is compared against head injury thresholds proposed in the literature. The more conclusive results were found to the oblique impact, where it was found a good correlation with thresholds that predict a not too severe brain injury, cerebral concussion and a severe brain injury induced by critical strains in the brain tissue. This more severe brain injury is DAI. Therefore, it can be concluded that from this oblique impact with a helmet approved by the three current standards, in which are included the European ECE R22.05 and the US DOT FMVSS 218, results brain injuries such as cerebral concussion and brain damage (in the basis of DAI) (brain tissue strain exceeded) or at least a high risk of occurrence.

This means that a lot of work is steel needed by the standards to improve the motorcycle helmet safety. The rotational component in head kinematics should some how be assessed, not alone, but combined with the translational motion, assessing what real happens in a real accident. To do this, two important things are needed, well accepted established head injury thresholds or criteria and a standard test rig that can be the reference to replicate impacts with helmets, that typically occurs in the reality, such as the oblique impact performed in this study.

The results obtained in this study reinforce the idea already defended by some researchers in others studies, that rotational acceleration must be assessed in real world accidents, consisting in a step in the right direction.

Other issue was identified in the motorcycle helmet standards test procedures, which is the rigid test headform. In order to confirm the suspicions, the ECE R22.05 energy absorbing test at point P was performed with the validated finite finite element head model of Strasbourg University, SUFEHM, based on experimental acceleration previously recorded. Only this impact was performed at an initial stage, but was sufficient to prove the expectations. The results were truly revealing, where a high risk of moderate DAI was predicted and again, concussion arises as a possible scenario. For a severe DAI and SDH a minor risk but still, with a considerable probability of occurrence was observed.

Thus, from the results it is clear the high risk of brain injury occurrence, which reveals again, defects in the current standards, more specifically in their tests specifications and criteria to injury assessment. Therefore, it is evident that a certified helmet cannot protect the user from brain injuries, mainly DAI and possibly concussion. This shows the incapacity of a good injury assessment from the utilisation of a rigid headform. Mainly due to this issue, FE head models, such as SUFEHM acquired great prominence, being intensively developed in last decade, since a real geometry to the real properties of head components and also being validated from real experiments and reproducing real impacts to the head that occurs in accidents. There are also solutions to the experimental procedures such as anatomical headforms.

All these conclusions reveal that there is still a long way to go by the standards and head's safety gear, mainly in the motorcycle case.

4.2 Future work

In this work it was developed a numerical model of a motorcycle helmet, where this one was assessed from the biomechanical point of view. It was concluded that a motorcycle helmet approved by the standards, fail in task of protecting the head from injuries such as concussion or even DAI. This conclusion was done by simply assess the rotational motion together with linear motion. and also by assessing a linear impact of the ECE R22.05 standard with a FEHM. In the oblique case, there are not well accepted thresholds nor a test rig that could be the reference between the community. The same happens for the test headforms. This work is a small step in research in this area. The following future work is suggested:

- Study of the same oblique impact with a validated FE head model to verify the conclusions of this work for the oblique impact.
- Study of the same energy absorption tests used to validate the helmet model whit the coupling of a validated FE head model to verify the further validate this model, beyond the point P.
- Modelling of the chin strap and the comfort foam, at least at the lateral sides of the helmet, to improve the numerical impact performance at that region, and than validate the model for the impact point X.
- Study the impact performance of the helmet has a modular helmet at the lateral sides and in the chin region. To study the impact performance of the helmet as a modular

helmet at the lateral sides, all the fixation systems should be modelled.

- Propose a new criterion or verify (by assessing) the existing ones for brain injuries, such as DAI and concussion with results obtained after the study with a validated FE head model.

- Testing new material solutions for helmet components based on the same predictions, for example new liner materials or configurations to improve the helmet energy absorption capacity.

References

[1] WHO, Global status report on road safety: time for action, The World Health Organization, Geneva, 2009. http://www.who.int/violence_ injury_ prevention/publications/road_ traffic/world_ report/en/index.html 15/03/12

[2] European Road Safety Observatory, 2008. Annual statistical report 2010 - based on data from CARE/EC. http://ec.europa.eu/transport/road_ safety/pdf/observatory/historical_ evol.pdf 15/03/12

[3] Peden, M., The world report on road traffic injury prevention, World Health Organization, Geneva, 2004.

[4] Koornstra, M., Broughton, J., Esberger, R., Glansdorp, C., Koppel, W., Taylor, F., Cauzard, J.P., Evans, A., Hantula, L., Piers, M., Vanlaar, W., Transport safety performance in the EU: a statistical overview, European Transport Safety Council, Brussels, Belgium, p. 32, 2003.

[5] Lin, M., Kraus, J., Methodological issues in motorcycle injury epidemiology, *Accident Analysis and Prevention* **40**:653-1660, 2008.

[6] Deck, C., Baumgartner, B., Willinger, R., Helmet Optimisation on head-helmet modelling, *Struct. Mater.* **13**:319-328, 2003.

[7] Tinard V., Deck C., Willinger R., New methodology for improvement of helmet performance during impacts with regards to biomechanical criteria, *Materials and Design* **37**:79-88, 2012.

[8] ECE Regulation 22.05, Uniform provision concerning the approval of protective helmets and their visors for driver and passengers of motor cycles and mopeds, United Nations, 2002. Internet: http://www.unece.org/fileadmin/DAM/trans/main/wp29/wp29regs/r022r4e.pdf

[9] ABAQUS 6.10, Hibbitt, Karlsson & Sorensen, Inc., 2010. Internet: https://www.sharcnet.ca/Software/Abaqus610/Documentation/docs/v6.10/index.html

[10] Observatório de Segurança Rodoviária, Autoridade Nacional Segurança rodoviária (ANSR), Sinistralidade Rodoviária, Lisboa, 2010.

[11] ACEM, ACEM's view on PTW fatality statistics in Europe, The motorcycle industry in Europe, Belgium, 2006.

[12] Subramanian, R., NCSA, Traffic safety facts, bodily injury locations in fatally injured motorcycle riders, NHTSA's national center for statistics and analysis, USA, 2007.

[13] DaCoTA, Traffic Safety Basic Facts 2010 - Motorcycles & Mopeds, European Road Safety Observatory, 2011. Internet: http://ec.europa.eu/transport/road_safety/pdf/statistics/dacota/bfs2010_dacota-kfv-1-3-main_figures.pdf (accessed 15/03/12)

[14] COST327, Motorcycle safety helmets, final report of the action, European Communities, Belgium, 2001.

[15] NHTSA, Determining Estimates of Lives and Costs Saved by Motorcycle Helmets, DOT HS 811 433, Washington, DC: National Highway Traffic Safety Administration, 2011. http://www-nrd.nhtsa.dot.gov/Pubs/811433.pdf 15/03/2012

[16] NHTSA, Traffic Safety Facts, Data: Motorcycles. DOT HS 811 159, Washington, DC: National Highway Traffic Safety Administration, 2008. http://www-nrd.nhtsa.dot.gov/Pubs/811159.pdf

[17] Deutermann,W., Motorcycle Helmet Effectiveness Revisited. US Department of Transportation, National Highway Traffic Safety Administration, DOT HS 809 715, 2004.

[18] Brown, C., Hejl, K., Bui, E., Tips, G., Coopwood, B., Risk factors for riding and crashing a motorcycle unhelmeted, *The Journal of Emergency Medicine* **41**:441-446, 2011.

[19] King, A., Yang, K., Zhang, L., Hardy, W., Viano, D., Is Head Injury Caused by linear or angular acceleration?, Proceedings of IRCOBI Conference, pp. 1-10, Lisbon, September, 2003.

[20] MAIDS, In depth investigations of accidents involving powered two wheelers, Final Report 1.2, Brussels, 2004. http://ec.europa.eu/transport/roadsafety_library/publications/maids_report_1_2_september_2004.pdf 15/03/2012

[21] Liu, B., Ivers, R., Norton, R., Boufous, S., Blows, S., Lo., S.K., *Helmets for preventing injury in motorcycle riders*, Cochrane Database Syst Rev, 2008.

[22] Abbas, A., Hefny, A., Abu-Zidan, F., Does wearing helmets reduce motorcycle-related death? A global evaluation, *Accident Analysis and Prevention*, In Press, Corrected Proof, 2011.

[23] Servadei, F., Begliomini, C., Gardini, E., Giustini, M., Taggi, F., Kraus, J., Effect of Italy's motorcycle helmet law on traumatic brain injuries, *Inj. Prev.*, **9**:257-260, 2003.

[24] Forman, J., Lopez-Valdes, F., Pollack, K., Heredero-Ordoyo, R., Molinero, A., Mansilla, A., Fildes, B., Segui-Gomez, M., Injuries among powered two-wheeler users in eight European countries: a descriptive analysis of hospital discharge data, *Accident Analysis and Prevention*, In Press, Corrected Proof, 2011.

[25] The Greek Gold. http://www.greekgold.com/items/485/504.jpg 16/03/2012

[26] Newman, J., The biomechanics of head trauma and the development of the modern helmet. How far have we really come?, Proceedings of the IRCOBI Conference, Prague, September, 2005.

[27] P & K Military Antiques. http://www.pandkmilitaryantiques.com 15/12/2011

[28] Turner, A., Havey, J., Helmet, US patent 1953. http://ip.com/pdf/patent/US2634415.pdf 15/12/2011

[29] Roth, H., Lombard, C., Crash helmet, US patent, 1953. http://ip.com/pdf/patent/US2625683.pdf 15/12/2011

[30] Bell helmets, http://www.bellracing.info/page_contenu/history.php 15/12/2011

[31] Snively, G.C., Skull Busting for Safety, Sports Car Illustrated, 1957.

[32] Cairns H., Head Injuries in Motor-Cyclists: The Importance of the Crash Helmet, *British Medical Journal*, No. 4213, pp. 465-471, 1941.

[33] Cairns H., Crash Helmets, *British Medical Journal*, No. 4470, pp. 322, 1946.

[34] Nemirovsky, N., van Rooij, L., A a new methodology for biofidelic head-neck postural control, Proceedings of IRCOBI Conference, pp:71-84, Hanover, Germany, September, 2010.

[35] Fraga, F., van Rooij, L., Happee, R., Wismans, J., Symeonidis, I., Peldschus, S., Development of a motorcycle rider model with focus on head and neck biofidelity, recurring to line element muscle models and feedback control, ESV 09-0244, 2009.

[36] van den Bosch, H.L.A., Crash Helmet Testing and Design Specifications, PhD thesis, Technische Universiteit Eindhoven, 2006.

[37] Holbourn, A.H.S., Mechanics of head injury, *British Medical Bulletin*, **2**:438-441, 1943.

[38] Holbourn, A.H.S., The mechanics of brain injuries, *British Medical Bulletin*, **3**:147-149, 1945.

[39] Chang, L.T., Chang, G.L., Huang, J.Z., Huang, S.C., Liu D.S., Chang, C.H., Finite element analysis of the effect of motorcycle helmet materials against impact velocity, *Journal of the Chinese Institute of Engineers*, **26**:835-843, 2003.

[40] Liu, D.S., Chang, C.Y., Fan, C.M., Hsu, S.L., Influence of environmental factors on energy absorption degradation of polystyrene foam in protective helmets, *Engineering Failure Analysis*, **10**:581-591, 2003.

[41] Shuaeib, F.M., Hamouda, A.M.S, Hamdan, M.M., Radin Umar, R.S., Hashmi, M.S.J., Motorcycle helmet: Part II, Materials and design issues, *Journal of Materials Processing Technology*, **123**:422-431, 2002.

[42] Brain Injury Association of America. http://www.biausa.org/mild-brain-injury.htm. 21/11/2011

[43] Arai helmet. http://www.araihelmeteurope.com/2010/wpcontent/uploads/factory20.jpg 15/12/2011

[44] Aare, M., Prevention of head injuries focusing specifically on oblique impacts, Doctoral Thesis, Technical Report 2003-26, School of Technology and Health, Royal Institute of Technology (KTH), Stockholm, Sweden, 2003.

[45] Yettram, AL, Godfrey, NP, Chinn, BP, Materials for motorcycle crash helmets - a finite element parametric study, *Plastics and Rubber Processing and Applications*, **22**:215-221, 1994.

[46] Huang, S. C., Numerical Simulation of Human Head-Neck Dynamics, *Bio-Medical Materials and Engineering,* **9**:66-71, 1999.

[47] Huston, R. L., Sears, J., Effect of Protective Helmet Mass on Head/Neck Dynamics, *Journal of Biomechanical Engineering*, **103**:18-23, 1981.

[48] Kim, G. H., Lee, O. S., Yoo, S. S., Experimental Study on impact absorbing performance of motorcycle Helmets, *KSME International Journal* **11**:292-299, 1997.

[49] Shuaeib, F.M., Hamouda, A.M.S., Wong, S.V., Radin Umar, R.S., Megat Ahmed, M.M.H., A new motorcycle helmet liner material: the finite element simulation and design of experiment optimization, *Materials and Design*, **28**:182-195, 2007.

[50] Mills, N. J., Gilchrist, A., The Effectiveness of Foams in Bicycle and Motorcycle Helmets, *Accident Analysis and Prevention*, **23**:153-163, 1991.

[51] Gilchrist, A., Mills, N. J., Impact Deformation of ABS and GRP Motorcycle Helmet Shells, *Plastic Rubber Composites Processes Appl*, **21**:141-150, 1994.

[52] Chang, C. H., Chang, L. T., Chang, G. L., Huang, S. C., Wang, C. H., Head Injury in Facial Impact: A Finite Element Analysis of Helmet Chin Bar Performance, *Journal of Biomechanical Engineering*, **21**:640-646, 2000.

[53] Mills, N.J., Plastics, Microstructure and Engineering Application, Arnold, Paris, 1993.

[54] Mlyajlma, I., Kitahara, H., Helmet, IBM: Intellectual Property Network, US Patent No.5,943,706, 1999.

[55] Mills, N.J., Impact response, N.C. Halyard, A. Cunningham (Eds.), Low Density Cellular Plastics, Physical Basis of Behavior, 1994.

[56] Richter, M., Otte, D., Lehmann, U., Chinn, B., Schuller, E., Doyle, D., Sturrock, K., Krettek, C., Head Injury Mechanisms in Helmet-Protected Motorcyclists: Prospective Multicenter Study, *Journal of Trauma*, **51**:949-958, 2001.

[57] Mills, N.J., Polymer Foams Handbook- Engineering and biomechanics applications and design guide, Butterworth Heinemann, 2007.

[58] Bourdet, N., Deck, C., Tinard, V., Willinger, R., Behaviour of helmets during head impact in real accident cases of motorcyclists, *International Journal of Crashworthiness*, **17**:51-61, 2012.

[59] Hopes, P. D., Chinn, B. P., Helmets: a new look at design and possible protection, Proceedings of IRCOBI conference, Stockholm, pp. 39-54, 1989.

[60] Chang, L. T., Chang, C. H., Huang, S. C., Chang, G. L., A dynamic analysis of motorcycle helmet by finite element methods, Proceedings of IRCOBI Conference, Sitges, Spain, pp. 371-382, 1999.

[61] Gilchrist, A., Mills, N. J., Improvements in the design and the performance of motorcycle helmets, Proceedings of IRCOBI Conference, Birmingham, UK, pp. 19-32, 1987.

[62] Kostopoulos, V., Markopoulos, Y. P. ,Giannopoulos, G., Vlachos, D. E., Crashworthiness Study of Composite Motorcycle Safety Helmet, *Composites Part B* **33**:99-107, 2002.

[63] Snell M2010: Standard for Protective Headgear, Snell Memorial Foundation, North Highlands, CA, 2010. http://www.smf.org 20/02/2012

[64] Hume, A., Mills, N. J., Gilchrist, A., Industrial head injuries and the performance of the helmets, Proceedings of IRCOBI Conference, Switzerland, 1995.

[65] Otte, D., Chinn, B., Doyle, D., Strurrock, K., Schuller, E., Accident description and analysis of motorcycle safety helmets, Cost 327 Interim Reports, Accident Research Unit, Medical University Hanover, Germany, 1997.

[66] Zellmer, H., Investigation of the Performance of Motorcycle Helmets Under Impact Condition, Proceedings of 37^{th} Stapp Car Crash Conference , SAE P-269, p. 63., 1993.

[67] Shuaeib, F.M., Hamouda, A.M.S., Radin Umar, R.S., Hamdan, M.M., Hasmi, M.S.J., Motorcycle helmet part I. Biomechanics and computational issues, *Journal of Materials Processing Technology*, **123**:406-421, 2002.

[68] Motorcycle Safety Foundation, What you should know about motorcycle helmets, Cycle Safety Information, California, USA, 2002. http://www.msf-usa.org/downloads/helmet_ CSI.pdf 5/11/2011.

[69] Mills, N.J., Wilkes, S., Derler, S., Flisch, A., FEA of oblique impact tests on a motorcycle helmet, *International Journal of Impact Engineering*, **36**:913-925, 2009.

[70] Mellor, A., StClair, V., Advanced motorcycle helmets, Proceedings of the 19^{th} International Technical Conference of the Enhanced Safety of Vehicles, Washington, DC, pp. 1-13, 2005.

[71] BS 6658:1985, British Standard Institution, Protective Helmet for Vehicles Users, British Standard 6658, London, 1985. http://www.bsigroup.com/

[72] DeMarco, A.L., Chimich, D.D., Gardiner, J.C., Nightingale, R.W., Siegmund, G.P., The impact response of motorcycle helmets at different impact severities, *Accident Analysis and Prevention*, **42**:1778-1784, 2010.

[73] Mills, N.J., Gilchrist, A., Motorcycle helmet shell optimisation, Proceedings Association for Advancement of Automotive Medicine conference, pp. 149-162, Portland, Oregon, 1992.

[74] Tinard, V., Deck, C., Willinger, R., Modelling and validation of motorcyclist helmet with composite shell, *International Journal of Crashworthiness*, **17**(2):209-215, 2012.

[75] Cernicchi, A., Galvanetto, U., Iannucci, L., Virtual modelling of safety helmets: practical problems, *International Journal of Crashworthiness*, **13**(4):451-467, 2008.

[76] Vallée, H., Hartemann, F., Thomas, C., Tarriére, C., Patel, A., Got, C., The fracturing of helmet shells, Proceedings of IRCOBI Conference, Delft, pp. 99-109, 1984.

[77] Beusenberg, M.C., Happee, R., An experimental evaluation of crash helmet design and effectiveness in standard impact tests, Proceedings of IRCOBI Conference, Eindhoven, The Netherlands, 1993.

[78] Mills, N.J., Role of components of motorcycle helmets, Report-B15 2TT, School of Metallurgy and Materials, University of Birmingham, UK, 1995.

[79] Gilchrist, A., Mills, N.J., Modeling of the impact response of motorcycle helmets, *International Journal of Impact Engineering*, **15**(3):201-218, 1994.

[80] Ghajari, M., Galvanetto, U., Iannucci, L., Influence of the body on kinematic and tissue level head injury predictors in motorcyclists accidents, Proceedings of IRCOBI Conference, York, UK, pp. 9-11, 2009

[81] Di Landro, L., Sala, G., Olivieri, D., Deformation mechanisms and energy absorption of polystyrene foams for protective helmets, *Polymer Testing*, **21**:217-228, 2002.

[82] Pinnoji, P.K., Mahajan, P., Analysis of impact-induced damage and delamination in the composite shell of a helmet, *Materials and Design*, **31**:3716-3723, 2010.

[83] Brands, D.W.A., Development and validation of a finite element model of a motorcycle helmet, Master's Thesis, Eindhoven University of Technology, The Netherlands, 1996.

[84] Mellor, A., Dixon, P., Improved motorcycle helmet design: the relationship between shell design and injury potential, Vehicle Engineering Resource Centre, TRL, Project Report-PR/VE/263/97, 1997.

[85] Markopoulos, Y.P., Vlachos, D.E., Kostopoulos, V., Impact response of a polycarbonate motorcycle safety helmet, Third National Congress on Computational Mechanics, Volos, Greece, pp. 609-616, 1999.

[86] Pinnoji, P.K., Mahajan, P., Impact analysis of helmets for improved ventilation with deformable head model, Proceedings of IRCOBI conference, pp. 159-170, Madrid, September, 2006.

[87] Pinnoji, P.K., Haider, Z., Mahajan, P., Design of motorcycle helmets: computational fluid and impact dynamics studies, *International Journal of Crashworthiness*, **13**(3):265-278, 2008.

[88] Tinard, V., Deck, C., Bourdet, N., Willinger, R., Motorcyclist helmet composite outer shell characterisation and modelling, *Materials and Design*, **32**:3112-3119, 2011.

[89] Ghajari, M., Deck, C., Galvanetto, U., Iannucci, L., Willinger, R., Development of numerical models for the investigation of motorcyclists accidents, 7^{th} European LS-DYNA Conference, Salzburg, Austria, 2009.

[90] Pinnoji, K. P., Mahajan, P., Two Wheeler Helmets with Ventilation and Metal Foam, *Defence Science Journal*, **58**:304-311, 2008.

[91] Pinnoji, P.K., Bourdet, N., Mahajan, P., Willinger, R., New motorcycle helmets with metal foam shell, Proceedings of IRCOBI Conference, pp. 449-452, Bern, Switzerland, September, 2008.

[92] Shuaeib, F.M., Hamouda, A.M.S., Radin Umar, R.S., Hamdan, M.M., Hashmi, M.S.J., Motorcycle helmet, Part III, *Journal of Materials Processing Technology*, **123**:432-439, 2002.

[93] Shuaeib, F.M., Hamouda, A.M.S., Wong, S.V., Radin Umar, R.S., Megat Ahmed, M.M.H., A new motorcycle helmet liner material: the finite element simulation and design of experiment optimization, *Materials and Design*, **28**:182-195, 2007.

[94] Alves de Sousa, R.J., Gonçalves, D.F.S., Coelho, R.M., Teixeira-Dias, F.M.V.H., Assessing the effectiveness of the use of a natural cellular material as safety padding in motorcycle helmets, *Simulation: Transactions of the Society for Modeling and Simulation International*, **88**(5):579-590, 2012.

[95] Pedder, J.B., Characteristics of serious and fatal motorcycle accidents, PhD thesis, University of Birmingham, 1993.

[96] Gibson, L.J., Ashby, M.F., Cellular Solids: Structure and Properties, 2nd ed., Cambridge University Press, Cambridge, 2001.

[97] Caserta, G.D., Iannucci, l., Galvanetto, U., Shock absorption performance of a motorbike helmet with honeycomb reinforced liner, *Composite Structures*, **93**:2748-2759, 2011.

[98] Blanco, DH, Cernicchi, A, Galvanetto, U, FE Modeling of Innovative Helmet Liners, 11^{th} International LS-DYNA Users Conference, **9**:1-11, 2010.

[99] Morgan, D.E., Cone-head technology, http://www.coneheadhelmets.com.au/ 17/12/2011.

[100] Gilchrist, A., Mills, N.J., Deformation analysis for motorcycle helmets, proceedings of IRCOBI conference, Eindhoven, The Netherlands, pp. 269Ũ-281, 1993.

[101] Gilchrist, A., Mills, N. J., Khan, T., Survey of Head, Helmet and Headform Sizes Related to Motorcycle Helmet Design, *Ergonomics*, **31**:1395-1412, 1988.

[102] van den Bosch, H.L.A., Modelling and specifications for an improved helmet design, Master thesis, Technische Universiteit Eindhoven, 1998.

[103] Chang, L.T., Chang, C.H., Chang, G.L., Fit effect of motorcycle helmet - a finite element modelling, *JSME International Journal*, **44**:185-192, 2001.

[104] Helmet Boys. http://www.helmetboys.com/images/motorcycle-helmets/ARAI/RAM3/arai-ram3-helmet-overall-airflow.jpg 15/12/2011

[105] CMS helmets, http://www.cms-helmets.com 7/10/2011

[106] Otte, D., Technical demands on safety in the design of crash helmets, 35^{th} Stapp Car Crash Conf., SAE paper 912911, pp. 335-148, Warrendale, PA, 1991.

[107] Chang, L., Chang, C., Chang, G., Experimental evaluation of chin bar on head injury in facial impact, *JSME International Journal*, **42**:294-300, 1999.

[108] Mills, N.J., Accident investigation of motorcycle helmets, *Impact*, **5**:46-51, 1996.

[109] Hurt, H.H., Ouellet, J.V., Thom, D.R., Motorcycle accident cause factors and identification of countermeasures, Vol. 1, Final Report, Technical Report, Traffic Safety Center, University of Southern California, CA, 1981.

[110] Web Bike World, http://www.webbikeworld.com/r2/zeus-helmet/zeus-zs-3000/ 16/05/2012

[111] Hitosugi, M., Shigeta, A., Takatsu, A., Yokoyama, T., Tokudome, S., Analysis of fatal injuries to motorcyclists by helmet type, *American Journal of Forensic Medicine and Pathology*, **25**:125-128, 2004.

[112] Yu, W.Y., Chen, C.Y., Chiu, W.T., Lin, M.R., Effectiveness of different types of motorcycle helmets and effects of their improper use on head injuries, *International Journal of Epidemiology*, **40**:794-803, 2011.

[113] Smith Family Powersports, http://www.sfpstore.com/servlet/the-578/XXL-2011-GMax-GM55S/Detail 20/11/2011

[114] Halldin, P., Gilchrist, A., Mills, N.J., A new oblique impact test for motorcycle helmets, *Int. J. Crashworthiness* **6**:53-64, 2001.

[115] Phillips, K.D., Protective headgear and protective armour and a method of modifying protective headgear and protective armour, European Patent EP1404189, 2004.

[116] Galford, J.E., McElhaney, J.H., A viscoelastic study of scalp, brain, and dura, *J. Biomech.*, **3**:211-221, 1970.

[117] Allsop, D., Kennett, K., Skull and facial bone trauma, Accidental Injury - Biomechanics and Prevention (Eds. Nahum, Melvin) Springer Verlag, New York, 2002.

[118] Pike, J.A., Automotive Safety: Anatomy, Injury, Testing and Regulation, SAE, 1990.

[119] Schmitt, K., Niederer, P., Muser, M., Walz, F., Head Injuries - Trauma Biomechanics, Springer Berlin Heidelberg, pp.55-81, 2007.

[120] UCSF, University of California, San Francisco. http://missinglink.ucsf.edu/lm/ids_104_cns_injury/Response%20_to_Injury/Injury_Images/MeningesBlum.jpg 09/12/2011

[121] Kleiven, S., Finite Element Modeling of the Human Head, Doctoral Thesis, Technical Report, School of Technology an Health, Royal Institute of Technology, Stockholm, Sweden, 2002.

[122] Chinn, B., Hynd, D., Technical response to the Unpublished "Critical Evaluation of the HARP Motorcycle Helmet Response" by Mills, N.J., Published Project Report PPR452, TRL, 2009.

[123] Mayfield Clinic http://www.mayfieldclinic.com/Images/PE-AnatBrainFig1.jpg 21/02/2012

[124] ProProfs http://www.proprofs.com/flashcards/upload/a5020641.jpg 21/02/2012

[125] Gennarelli, T.A., Mechanistic approach to head injuries: clinical and experimental studies of the important types of injury, Ed. Ommaya, A.K., in Head and neck injury criteria: a consensus workshop. U.S. dept. of transportation, national highway traffic safety administration, Washington DC, pp. 20-25, 1981.

[126] Kleiven, S., Peloso, P., von Holst, H., The Epidemiology of Head Injuries in Sweden From 1987 to 2000, *Journal of Injury Control and Safety Promotion* **10**:173?180, 2003.

[127] Youmans, J., Neurosurgery, W. B. Sounders Co., 1996.

[128] Gennarelli, T.A, Head Injury in Man and Experimental Animals: Clinical Aspects, *Acta Neurochirurgica, Suppl.* **32**:1-13, 1983.

[129] Whitaker, J., A survey of motorcycle accidents, TRRL Laboratory Report LR 913, Crowthorne, TRRL, 1980.

[130] Melvin, J., Lighthall, J., Ueno, K., Brain Injury Biomechanics, Accidental Injury, Nahum, A.M. and Melvin, J.W. (eds), Springer-Verlag, New York, pp. 269-290, 1993.

[131] Cooper, P.R., Post-traumatic intracranial mass lesions, Head injury, Williams and Wilkins, Baltimore/London, pp. 185-232, 1982.

[132] Gennarelli, T.A., Thibault, L.E., Adams, J.H., Graham, D.I., Thompson, C.J., Marcincin, R.P., Diffuse Axonal Injury and Traumatic Coma in the Primate, *Ann. Neurol.* **12**:564-574, 1982.

[133] Aare, M., von Holst, H., Injuries from Motorcycle and Moped crashes in Sweden from 1987-1999, *Published in Injury Control and Safety Promotion* **10**:131-138, 2003.

[134] Melvin, J., Lighthall, J., Brain injury biomechanics, Accidental Injury - Biomechanics and Prevention (Eds. Nahum, Melvin), Springer Verlag, New York, 2002.

[135] Gennarelli, T.A., The state of the art of head injury biomechanics - A review, 29^{th} Conference of the Association for the Advancement of Automotive Medicine, pp. 47-63, 1985.

[136] Prasad, P., Review of biomechanical impact response and injury in the automotive environment, chap . 1 - Head, report no. DOT HS 807 042, U.S. Department of transport - NHTSA, Washington D.C., USA, 1985.

[137] Wismans, J., Injury biomechanics, no. 4552, Eindhoven University of Technology, The Netherlands, second printing edn, 1994.

[138] Moore, K., Clinically Oriented Anatomy, Williams and Wilkins, Second edition, Baltimore, USA, 1985.

[139] Gennarelli T, Thibault L, Biomechanics of acute subdural hematoma, *J Trauma*, **22**:680-686, 1982.

[140] Voo, L., Pintar, F.A., Yoganandan, N., Sances Jr., A., Ewing, C.L., Thomas, D.J., Synder, R.G., Biomechanical analysis of tractor-induced head injury, *SAE Transaction Paper*, no. 941725, pp. 178-183, 1994.

[141] Viano, D., Biomechanics of head injury - toward a theory linking head dynamic motion, brain tissue deformation and neural trauma, *SAE Transaction Paper*, no. 881706, pp. 1070-1089, 1988.

[142] Kraus, J., McArthur, D., Epidemiologic Aspects of Brain Injury, *Neurological Clinic*, **14**(2):435-450, 1996.

[143] Kleiven, S., von Holst, H., Consequences of brain size following impact in prediction of subdural hematoma evaluated with numerical techniques, Proceedings of IRCOBI Conference, Isle of Man, UK, pp. 161-172, October 2001.

[144] Bullock, R., Graham, D.I., in: Graham, J.C., Dudley, H.A.F., Gann, D.S., Little, R.A., Maynard (Eds.), Non-penetrating Injuries of the Head, Section 2 of the Scientific Foundation of Trauma, Butterworths/Heinemann, London, 1997.

[145] Motherway, J.A., Verschueren, P., van der Perre, G., van der Sloten, J., Gilchrist, M.D., The mechanical properties cranial bone, Proceedings of IRCOBI Conference, Hanover, Germany, pp. 131-134, September, 2010.

[146] The Ohio State University Medical Center. http://medicalcenter.osu.edu/greystone/images/es_ 2705.gif 13/12/2011

[147] Chinn, B.P., Doyle, D., Otte, D., Schuller, E., Motorcyclists head injuries: mechanisms identified from accident reconstruction and helmet damage replication, Proceedings of IRCOBI Conference, pp. 53-72, 1999.

[148] Bandak, F.A., Biomechanics of impact traumatic brain injury, in: Ambrosio, J.A.C., Seabra Pereira, M.F.O., Silva, P.F. (Eds.), Crash-worthiness of Transportation Systems: Structural Impact and Occupant Protection, Kluwer Academic Publishers, Dordrecht, pp. 53-93, 1997.

[149] Unterharnscheidt, F., Higgins, L.S., Traumatic lesions of brain and spinal cord due to non-deforming angular acceleration of the head, *Texas Reports on Biology and Medicine*, **27**(1):127-166, 1969.

[150] Ommaya, A.K., Hirsch, A.E., Tolerances for Cerebral Concussion from Head Impact and Whiplash in Primates, *J. Biomechanics*, **4**:13-31, 1971.

[151] Gennarelli, T.A., Ommaya A.K., Thibault, L.E., Comparison of translational and rotational head motions in experimental cerebral concussion, Proceedings of 15^{th} Stapp Car Crash Conference, SAE P-39: 797-803, 1971.

[152] Gennarelli, T.A., Thibault, L.E., Ommaya, A.K., Pathophysiological Responses to Rotational and Translational Accelerations of the Head, SAE Paper No. 720970, Proceedings of 16^{th} Stapp Car Crash Conference, Society of Automotive Engineers, pp. 296-308, 1972.

[153] Abel, J.M., Gennarelli, T.A., Segawa, H., Incidence and severity of cerebral concussion in the rhesus monkey following sagittal plane angular acceleration, Proceedings of 22^{nd} Stapp Car Crash Conference, Society of Automotive Engineers, SAE Paper No. 780886, 1978.

[154] Hodgson, V.R., Thomas, L.M., Acceleration induced shear strains in a monkey brain hemisection, SAE Paper No. 791023, Proceedings of 23^{rd} Stapp Car Crash Conference, Society of Automotive Engineers, 1979.

[155] Hodgson, V.R., Thomas, L.M., Khalil, T.B., The role of impact location in reversible cerebral concussion, SAE Paper No. 831618, Proceedings of 27^{th} Stapp Car Crash Conference, Society of Automotive Engineers, 225-240, 1983.

[156] Ono, K., Kikuchi, A., Nakamura, M., Kobayashi, H., Nakamura, N., Human Head Tolerance to Sagittal Impact: Reliable Estimation deduced from Experimental Head Injury using Subhuman Primates and Human Cadaver Skulls, Proceedings of 24^{th} Stapp Car Crash Conference, SAE Paper No. 801302, 1980.

[157] Gennarelli, T.A., Thibault, L.E., Tomei, G., Directional dependence of axonal brain injury due to centroidal and non-centroidal acceleration, SAE Paper No. 872197, Proceedings of 31^{st} Stapp Car Crash Conference, Society of Automotive Engineers, pp. 49-53, 1987.

[158] Ward, C., Chan, M., Rotation generated shear strains in the brain, Proceedings of 8^{th} annual Int. Workshop on Human Subjects for Biomechanical Research, Troy, MI, USA, 1980.

[159] Shugar, T.A., Kahona, M.C., Development of finite element head injury model, *Journal of Structural Engineering - ASCE* **101**:223-239, 1975.

[160] Shugar, T.A., Kahona, M.C., A finite element head injury model, Vol. I: Theory, development, and results, U.S. Dept. Of Transportation Report no. DOT-HS-289-3-550-IA, 1977.

[161] Lee, M.C., Melvin, J.W., Ueno, K., Finite element analysis of traumatic subdural hematoma, Proceedings of 31^{st} Stapp Car Crash Conference, SAE Paper No. 872201, Society of Automotive Engineers, 1987.

[162] Huang, H.M., Three-dimensional finite element analysis of subdural hematoma, *The Journal of Trauma: Injury, Infection, and Critical Care*, **47**(3):538-544, 1999.

[163] Huang, H.M., Finite element analysis of brain contusion: an indirect impact study, *Medical & Biological Engineering & Computing*, **38**(3):253-259, 2000.

[164] Bandak, F.A., Eppinger, R.H., A three-dimensional FE analysis of the human brain under combined rotational and translational accelerations, Proceedings of 38^{th} Stapp Car Crash Conference, Society of Automotive Engineers, pp. 145-163, 1994.

[165] Bandak, F.A., van DerVorst, M.J., Stuhmiller, L.M., Mlakar, P.F., Chilton, W.E., Stuhmiller, J.H., An imaging based computational and experimental study of skull fracture: finite element model development, Proceedings of the Head Injury Symposium, Washington DC, 1994.

[166] DiMasi, Simulated head impacts with upper interior structures using rigid and anatomic brain models, Auto and trafic safety., no. 1, Ed. R. Strombotne, National Highway Traffic Safety Publication, Vol. 1, 1991.

[167] DiMasi, F., Eppinger, R.H., Bandak, F.A., Computational analysis of head impact response under car crash loadings, Proceedings of 39^{th} Stapp Car Crash Conference, Society of Automotive Engineers, SAE Paper No. 952718, Society of Automotive Engineers, Warrendale, PA, 425-438, 1995.

[168] Ueno, K., Melvin, J.W., Finite element model study of head impact based on hybrid III head acceleration: The effects of rotational and translational acceleration, *J. Biomechanical Engineering*, **117**(3):319-328, 1995.

[169] Zhou, C., Khalil, T.B., King, A.I., A new model comparing impact responses of the homogeneous and inhomogeneous human brain, Proceedings of 39^{th} Stapp Car Crash Conference, Society of Automotive Engineers, 121-137, 1995.

[170] Zhang, L., Yang, K.H., King, A.I., Comparison of brain responses between frontal and lateral impacts by finite element modelling, *J. Neurotrauma*, **18**(1):21-30, 2001.

[171] Kleiven, S., von Holst, H., Review and evaluation of head injury criteria, Proceedings RTO Specialist Meeting, the NATO, Koblenz, Germany, 2003.

[172] The Ohio State University Medical Center. http://medicalcenter.osu.edu/greystone/images/es_ 2704.gif 13/12/2011

[173] Ommaya, A.K., Mechanisms and preventive Management of head injuries, a paradigm for injury control, Proceedings of 32^{nd} AAAM Conference, 1988.

[174] Gurdjian, E .S., The mechanism of skull fracture, *Radiology*, **54**:313-339, 1950.

[175] Thomas, D .R., Hodgson, V .R., Skull fracture and management of open head injury, Neurological surgery (Edited by Youmans, J .R.), pp. 969-977, Philadelphia, Saunders, 1973.

[176] Adams, J. H., Brain damage in fatal non-missile head injury, *Journal of Clinical Pathology*, **33**:1132-1145, 1980.

[177] Chapon, A., Dedoyan, A .J., Verriest, J .P, Clinical study of head injuries in patients with traumatic unconsciousness, Proceedings of IRCOBI Conference, pp. 33-54, 1985.

[178] Gurdjian, E.S., Webster, J.E., Linear acceleration causing shear in the brain stem in trauma of the central nervous system, *Mental Advances in Disease*, pp. 24-28, 1945.

[179] Gurdjian, E.S., Webster, J.E., Lissner, H.R., Observations on the mechanism of brain concussion, contusion and laceration, *Surgery, Gynecology and Obstetrics*, **101**:680-690, 1955.

[180] Gurdjian, E.S., Lissner, H.R., Evans, F.G., Intracranial Pressure and Acceleration Accompanying Head Impacts in Human Cadavers, *Surgery, Gynecology, and Obstetrics*, **112**:185-190, 1961.

[181] Gurdjian, E.S., Lissner H.R., Patrick, L.M., Concussion-mechanism and pathology, Proceedings of 7^{th} Stapp Car Crash Conference, pp. 470-482, 1963.

[182] Ommaya, A.K., Hirsch, A.E., Martinez, J.L., The role of whiplash in cerebral concussion, Proceedings of 10^{th} Stapp Car Crash Conference, SAE Paper No. 660804, 1966.

[183] Unterharnscheidt, F.J., Translational versus rotational acceleration: animal experiments with measured input, Proceedings of 15^{th} Stapp Car Crash Conference, SAE Paper No. 710880, 1971.

[184] Lowenheilm, P., Mathematical simulation of gliding contusions, *J. Biomechanics*, **8**:351-356, 1975.

[185] Gennarelli, T.A., Adams, J.H., Graham, D.I., Acceleration induced head injury in the monkey I: The model, its mechanistic and physiological correlates, *Acta Neuropathol. Suppl.*, **7**:23-25, 1981.

[186] Thibault, L.E., Gennarelli, T.A., Biomechanics of diffuse brain injuries, Proceedings of 29^{th} Stapp Car Crash Conference, SAE Paper No. 856022, 1985.

[187] Fruin, A.H., Juhl, G.L., Taylon, C., Interhemispheric subdural hematoma, *J. Neurosurg.*, **60**:1300-1302, 1981.

[188] Hirakawa, K., Statistical analysis of chronic subdural hematoma in 309 adult cases *Neurol. Med. Chir.*, **12**:71-83, 1972.

[189] Jamieson, K.G., Yelland, J.D.N., Surgically treated traumatic subdural hematomas, *J. Neurosurgery*, **37**(2):137-149, 1972.

[190] Adams, J.H., Gliding contusions in non-missile head injury in humans, *Archives of Pathology & Laboratory Medicine*, **110**(6):485-508, 1986.

[191] Lowenhielm, C.G.P., Dynamic properties of the parasagittal bridging veins, *Zeitschrift fur Rechtsmedizin*, **74**:55-62, 1974.

[192] Nahum, A.M., Smith, R., Ward, C.C., Intracranial pressure dynamics during head impact, Proceedings of 21^{st} Stapp Car Crash Conference, pp. 339-366, 1977.

[193] McLean, A.J., Brain injury without head impact?, *J. of Neurotrauma*, **12**:621-625, 1995.

[194] COST 327, Motorcycle safety helmets : A literature review, Transport Research Laboratory, Crowthorne, Berkshire, England, 1997.

[195] Bain, B.C., Meaney, D.F., Tissue-level thresholds for axonal damage in an experimental model of central nervous system white matter injury, *Journal of Biomechanical Engineering*, **16**:615-622, 2000.

[196] Kleiven, S., von Holst, H., Consequences of head size following trauma to the human head, *Journal of Biomechanics*, **35**:153-160, 2002.

[197] Viano, D., King, A.I., Injury Mechanisms and biofidelity of dummies, Crashworthiness of Transportation Systems: Structural Impact and Occupant Protection, Kluwer Academic Publishers, Dordrecht, 1997.

[198] Glaister, D., Acceleration injury, Graham, J.C., Dudley, H.A.F., Gann, D.S., Little, R.A., Maynard (Eds.), Scientific Foundation of Trauma, Butterworths/Heinemann, London, pp. 314-323, 1997.

[199] Supprian, T., Bendszus, M., Hofmann, E., Becker, T., Retz, W., Rosler, M., The role of diffuse axonal injury in medical assessment of brain injured patients, *Fortschritre der Neurologie Psychiatrie*, **68**(3):121-128, 2000.

[200] Meaney, D.F., Smith, D.H., Shreiber, D.I., Bain, A.C., Miller, R.T., Ross, D.T., Gennarelli, T.A., Biomechanical analysis of experimental diffuse axonal injury, *J. Neurotrauma*, **12**(4):689-694, 1995.

[201] Nishimoto, T., Murakami, S., Relation between diffuse axonal injury and internal head structure on blunt impact, *J. Biomech. Eng.* **120**(1):140-147, 1998.

[202] Nishimoto, T., Murakami, S., Direct impact simulation of diffuse axonal injury by axial head model, *JSAE Rev.*, **21**(1):117-123, 2000.

[203] Nahum, A. M., Melvin, J. W., Accidental Injury-Biomechanics and Prevention, pp. 292-300, 1993.

[204] Gurdjian, E.S., Lissner, H.R., Latimerr, F.R., Haddad, B.F., Webster, J.E., Quantitative determination of acceleration and intracranial pressure in experimental head injury, *Neurology*, **3**:417-423, 1953.

[205] Lissner, H.R., Lebow, M., Evans, F.G., Experimental Studies on the Relation Between Acceleration and Intracranial Pressure Changes in Man, *Surgery, Gynecology, and Obstetrics*, **3**:329-338, 1960.

[206] Nokes, L.D.M., Roberts, A.M., Knight, B.H., Use of Gadd severity index in forensic medicine: a case study, *Forensic Sci. Int.*, **76**(2):85-90, 1995.

[207] King, A., Fundamentals of impact biomechanics: Part I-Biomechanics of the Head, Neck, and Thorax, *Annu. Rev. Biomed. Eng.*, **2**:55-81, 2000.

[208] Gadd, C.W., Use of a weighted-impulse criterion for estimating injury hazard, Proceedings of the 10^{th} Stapp Car Crash Conference, New York, pp. 164-174, 1966.

[209] Versace, J., A review of the severity index, Proceedings of 15^{th} Stapp Car Crash Conference, New York, USA, Paper no. 710881, pp. 771-796, Warrendale, Soc. Automot. Eng., 1971.

[210] Newman, J.A., Head injury criteria in automotive crash testing, Society Automotive Engineers, SAE 801317, 1980.

[211] Onusic, H., HIC and SI of impacts with different pulse shapes, *Int. J. Vehicle Des.*, **16**:194-202, 1995.

[212] Hutchinson, J., Kaiser, M.J., Lankarani, H.M., The head injury criterion (HIC) functional, *Appl. Math. Comput.*, **96**(1):1-16, 1998.

[213] Ruan, J.S., Prasad, P., Coupling of a finite element human head model with lumped parameter hybrid III dummy model: preliminary results, *J. Neurotrauma*, **12**(4):725-734, 1995.

[214] King, A.I., Ruan, J.S., Zhou, C., Hardy, W.N., Khalil, T.B., Recent advances in biomechanics of brain injury research: a review, *J. Neurotrauma*, **12**(4):651-658, 1995.

[215] Naick, P.A., Carnago, K.M., Theoretical relationship of HIC_d to crush space for FMVSS-201 head impact tests, and the effective use of FEA in predicting HIC for design, verification and optimization of countermeasures, SAE Transactions Paper No. 982349, pp. 2465-2473, 1998.

[216] Fenner Jr., H., Thomas, D.J., Gennarelli, T., Pintar, F.A., Becker, E.B., Newman, J.A., Yoganandan, N., HIC Workshop: Final report of workshop on criteria for head injury and helmet standards, Department of Neurosurgery, Medical College of Wisconsin, Milwaukee, WI, 2005.

[217] Lowenhielm, P., Tolerance levels for bridging vein disruption calculated with a mathematical model, *Journal of Bioscience and Bioengineering*, **2**:501-507, 1978.

[218] Newman, J.A., A generalized model for brain injury threshold (GAMBIT), Proceedings of IRCOBI Conference, Zurich, Switzerland, September 2-4, 1986.

[219] Shewchenko, N., Withnall, C., Keown, M., Gittens, R., Dvorak, J., Heading in Football. Part 1: Development of Biomechanical Methods to Investigate Head Response, *British Journal of Sports Medicine*, **39**:10-25, 2005.

[220] Kang, H., Willinger, R., Diaw, B.M., Chinn, B., Validation of a 3D anatomic human head model and replication of head impact in motorcycle accident by finite element modelling, SAE Transactions Paper No. 973339, pp. 849-858, 1997.

[221] Claessens, M., Sauren, F., Wismans, J., Modeling of the human head under impact conditions: a parametric study, SAE Transactions Paper No. 973338, pp. 3829-3848, 1997.

[222] Turquier, F., Kang, H., Trosseille, X., Willinger, R., Lavaste, F., Tarriere, C., Domont, A., Validation study of a 3D finite element head model against experimental data, SAE Transactions Paper No. 962431, pp. 1912-1923, 1996.

[223] Miller, R.T., Margulies, S.S., Leoni, L., Nonaka, M., Chen, X., Smith, D. Meany, D., Finite element modeling approaches for predicting injury in an experimental model of severe diffuse axonal injury, SAE Transactions Paper No. 983154, pp. 2798-2810, 1998.

[224] Liu, D.S., Fan, C.M., Applied pressure tolerance to evaluate motorcycle helmet design, Proceedings of International Crashworthiness Conference, Dearborn, Michigan, USA, 9-11 September 1998.

[225] Zhou, C., King, A., Khalil, T., Yang, K., Meany, D., Finite element modeling of experimental diffuse axonal injury: a parametric study, Crashworthiness, Occupant Protection and Biomechanics in Transportation Systems, AMD-Vol. 225/BED-Vol. 38, ASME, New York, 1997.

[226] Newman, J.A., Shewchenko, N., Welbourne, E., A New Biomechanical Head Injury Assessment Function: The Maximum Power Index, *Stapp Car Crash J.*, **44**:215-47, 2000.

[227] Newman, J., Barr, C., Beusenberg, M., Fournier, E., Shewchenko, N., Welbourne, E., Withnall, C., A New Biomechanical Assessment of Mild Traumatic Brain Injury Part 2 - Results and Conclusions, Proceedings of IRCOBI Conference, Montpellier (France), 2000.

[228] Bandak, F.A., On the mechanics of impact neurotrauma: A Review and critical synthesis, *Journal of Neurotrauma*, **12**(4):635-649, 1995.

[229] Bandak, F.A., Impact traumatic brain injuries: a mechanical perspective, in Neurotraumatology: Biomechanic aspects, Cytologic and Molecular Mechanisms, eds. M. Oehmichen and H.G. Konig, Schmid-Romhild, pp. 58-83, 1997.

[230] Bandak, F.A., Zhang, A.X., Tannous, R.E., DiMasi, F., Masiello, P., R. Eppinger, SIMon: a simulated injury monitor: application to head injury assessment, Proceedings of the 17^{th} Int. technical conference on the Enhanced Safety of Vehicles, Amsterdam, The Netherlands, 2001.

[231] Hardy, W.N., Foster, C.D., King, A.I., Tashman, S, Update on the study of head injury kinematics, Syposium Proceedings, Centers for Disease Control, pp. 177-184, 1998.

[232] Al-Bsharat, A.S., Hardy, W.N., Yang, K.H., Khalil, T.B., Tashman, S., King, A.I., Brain/Skull Relative Displacement Magnitude Due to Blunt Head Imapact: New Experimental Data and Model, Proceedings of the 43^{rd} Stapp Car Crash Conference, pp. 321-Ű332, Paper No. 99SC22, 1999.

[233] Hardy, W.N., Foster, C.D., Mason, M.J., King, K.H., King, A.I., Tashman, S, Investigation of head injury mechanisms using neutral density technology and high-speed biplanar X-Ray, Stapp Car Crash Journal, Proceedings of the 45^{th} Stapp Car Crash Conference, pp. 337-368, 2001.

[234] Gadd, C.W., Tolerable Severity Index in whole-head nonmechanical impact, Proceedings of the 15^{th} Stapp Car Crash Conference, pp. 809Ũ-816, 1971.

[235] NHTSA, Occupant crash protection - Head Injury Criterion, Report No. FMVSS 208, Docket Number 69-7, Notice 17, US Departement of Transportation, 1972.

[236] Adams, J.H., Graham, D.I., Gennarelli, T.A., Head injury in man and experimental animals: neuropathology, *Acta Neurochirurgica*, **32**:15-Ũ30, 1983.

[237] Eiband, A.M., Human Tolerance to Rapidly Applied Accelerations: A Summary of the Literature, NASA Memo, 1959.

[238] Melvin, J.W., Injury assessment reference values for the CRABI 6-Month Infant Dummy in a Rear-Facing Infant Restraint with Air Bag Deployment, SAE International Congress and Exposition, SAE Paper No. 950872, Society of Automotive Engineers, 1995.

[239] Kleinberger, M., Development of improved injury criteria for the assessment of advanced automotive restraint systems, National Highway Traffic Safety Administration, 1998.

[240] Eppinger, R., Development of improved injury criteria for the assessment of advanced automotive restraint systems - II, National Highway Traffic Safety Administration, 2000.

[241] Mertz, H., Prasad, P., Irwin, A., Injury risk curves for children and adults in frontal and rear collisions, SAE Technical Paper 973318, Society of Automotive Engineers, pp. 13-30, 1997.

[242] Stalnaker, RL, McElhaney, JH, Roberts, VL, MSC tolerance curve for human heads to impact, ASME paper No. 71-WA/BHF-10, pp. 1-9, 1971.

[243] Margulies, S.S., Thibault, L.E., A proposed tolerance criterion for diffuse axonal injury in man, *J. Biomech.*, **25**:917-923, 1992.

[244] Löwenhielm, P., Strain tolerance of the Vv. Cerebri Sup. (bridging veins) calculated from head-on collision tests with cadavers, *Z. Rechtsmedizin*, **75**(2):131-144, 1974.

[245] Voigt, G.E., Lange, W., Simulation of head-on collision with unrestrained front seat passengers and different instrument panels, Proceedings of 15^{th} Stapp Car Crash Conference, SAE Technical paper No. 710863, pp. 466-488, 1971.

[246] Ommaya, A.K., Hirsch, A.E., Harris, E., and Yarnell, P.: Scaling of experimental data in cerebral concussion in sub-human primates to concussive threshold for man, Proceedings of the 11^{th} Stapp Car Crash Conference, Society of Automotive Engineers, pp. 47-52, New York, October, 1967.

[247] Pincemaille, Y., Trosseille, X., Mack, P., Tarrière, C., Some new data related to human tolerance obtained from volunteer boxers, SAE Technical Paper 892435, 1989.

[248] Bycroft, G.N., Mathematical model of a head subjected to an angular acceleration, *Journal of Biomechanics*, **6**:487-495, 1973.

[249] Hayashi, T., Brain shear theory of head injury due to rotational impact, *Journal of the Faculty of Engineering*, University of Tokyo, No 4, 1970.

[250] Nahum, A., Gatts, J., Gadd, C., Danforth, J., Impact tolerance of the skull and face, Proceedings of 12^{nd} Stapp Car Crash Conference, SAE 680785, 1968.

[251] Hodgson, V., Thomas, L., Breaking strength of the human skull vs impact surface curvature, Wayne State University School of Medicine, Dept. of Neurosurgery, Report, 1971.

[252] Schneider, D., Nahum A., Impact studies of facial bones and skull, Proceeding of 16^{th} Stapp Car Crash Conference, SAE 720965, pp. 186-203,1972.

[253] Advani, S., Powell, W., Huston, J., Ojala, S., Human head impact response experimental data and analytical simulations, Proceedings of International Conference on Biomechanics of Serious Trauma, pp. 153-162, 1975.

[254] Allsop, D., Warner, C., Wille, M., Schneider, D., Nahum, A., Facial impact response - a comparasion of the Hybrid III dummy and the human cadaver, Proceeding of 32^{nd} Stapp Car Crash Conference, SAE 881719, 1988.

[255] Yoganandan, N., Sances, A., Pintar, F.A., Walsh, P.R., Ewing, C.L., Thomas, D.J., Snyder, R.G., Reinartz, J., Droese, K., Biomechanical tolerance of the cranium, SAE Transactions Paper No. 94172, pp. 184-188, 1994.

[256] Allsop, D., Perl, T., Warner, C., Force/deflection and fracture characteristics of the temporoparietal of the human head, Proceedings of 35^{th} Stapp Car Crash Conference, SAE 912907, pp. 139-155, 1991.

[257] Advani, S., Ommaya, A., Yang, W., Head injury mechanisms, in Human Body Dynamics, Ed. Ghista, Oxford Univ. Press, 1982.

[258] Association of Advancement of Automotive Medicine, The injury scale, AIS, Eds. Gennarelli T and Wodzin E, 2005.

[259] Gurdjian, E., Robert, V., Thomas, L., Tolerance curves of acceleration and intercranial pressure and protective index in experimental head injury, *J Trauma*, **6**(5):600-604, 1966.

[260] Tarriere, C., Relationship between experimental measuring techniques and real world accidents, Head Injury Symposium, New Orleans, AAAM report, 1987.

[261] Got, C., Patel, A., Fayon, A., Tarriere, C., Walfisxh, G., Results of experimental head impacts o cadavers: the various data obtained and their relation to some measured physical parameters, SAE Technical Paper 780887, pp. 57-99, 1978.

[262] Kramer, F., Passive Sicherheit von Kraftfahrzeugen, Vieweg Verlag, Braunschweig, Germany, 1998.

[263] Zhang, L., Yang, K., King, A., A proposed injury threshold for mild traumatic brain injury, *J Biomech Engineering*, **16**:226-236, 2004.

[264] Ommaya, A., Goldsmith, W., Thibault, L., Biomechanics and neuropathology of adult and paediatric head injury, *British J. Neurosurg.*, **16**(3):220-242, 2002.

[265] Willinger, R., Baumgartner, D., Chinn, B., Neale, M., Head tolerance limits derived from numerical replication of real world accidents, Proceedings of IRCOBI Conference, pp. 209-222, Isle of Man, UK, 2000.

[266] Baumgartner, D., Willinger, R., Shewchenko, N., Beusenberg, M., Tolerance limits for mild traumatic brain injury derived from numerical head impact replication, Proceedings of IRCOBI Conference, Isle of Man, UK, 2001.

[267] Thomson, R., Lövsund, P., Norin, H., Jakobsson, L., Boström, O., Håland, Y., Brain injuries in real world accidents - a multidisciplinary investigation, Proceedings of IRCOBI Conference, pp. 173-188, Isle of Man, UK, 2001.

[268] Ewing, C., Thomas, D., Lustick, L., Becker, E., Becker, G., Willems, G., Muzzy, W., The Effect of the Initial Position of the Head and Neck to Gx. Impact Acceleration, Proceedings of the 19^{th} Stapp Car Crash Conference, SAE, Warrendale, Pennsylvania, USA, 1975.

[269] Lee, M.C., Haut, R.C., Insensitivity of tensile failure properties of human bridging veins to strain rate: Implications in biomechanics of subdural hematoma, *J. Biomechanics*, **22**:537-542, 1989.

[270] Department of Transportation (DOT), National Highway Traffic Safety Administration(2000): CFR 49 FMVSS 571.201 Occupant Protection in Interior Impact, 2000.

[271] Eichberger, A., Steiner, J., Breitenhuber, W., Wernig, J., The effect of the fmvss201u free motion headform legislation on head injuries: a discussion based on numerical simulations, Proceedings of IRCOBI Conference, Munich (Germany), 2002.

[272] Deck, C., Willinger, R., Baumgartner, D., Meyer, F., Helmet optimisation against biomechanical criteria, Proceedings of IRCOBI Conference, Lisbon (Portugal), pp. 351-352, 2003.

[273] Aare, M., Kleiven S., Halldin, P., Injury criteria for oblique helmet impacts, Proceedings of IRCOBI Conference, Lisbon (Portugal), pp. 349-350, 2003.

[274] Bellora, A., Krauss, R., Van Poolen, L., Meeting Interior Head Impact Requirements: A Basic Scientific Approach, SAE 2001-01-0469, 2001.

[275] Kleiven, S., Influence of Impact Direction to the Human Head in Prediction of Subdural Hematoma, *Journal of Neurotrauma*, **20**(4):365-379, 2003.

[276] Gennarelli, T.I., Thibault, L.E., Clinical rationale for a head injury angular acceleration criterion, pp. 5-8 in Head Injury Mechanisms, Washington, 1989.

[277] Tahkounts, G.E., Eppinger, R.H., Campbell, J.Q., Tannous, E.R., Power, E.D., Shook, L.S., On the development of the SIMon finite element head model human, *Stapp Car Crash J.*, **47**:107-133, 2003.

[278] Vezin, P., Verriest, J.P., Evaluation of the simulated response of the human brain subjected to different accelerations during a frontal impact, Proceedings of IRCOBI Conference, Graz (Austria), pp. 319-320, 2004.

[279] Willinger, R., Baumgartner, D., Chinn, B., Neale, M., Head tolerance limits derived from numerical replication of real world accidents, Proceedings of the IRCOBI Conference, pp. 209-222, 2000.

[280] Willinger, R., Baumgartner, D., Human head tolerance limits to specific injury mechanisms, *International journal of Crashworthiness*, **8**(6):605-617, 2003

[281] Zhang, L., King, K.H., Dwarampudi, R., Omori, K., Li, T.B., Hardy, W., Khalil, T., King, A.I., Recent advances in brain injury research: A new human head model development and validation, *Stapp Car Crash Journal*, **45**:369-393, 2001.

[282] Thibault, L.E., Gennarelli, T.A., Margulies, S.S., Marcus, J., Eppinger R., The strain dependant pathophysiological consequences of inertial loading on central nervous system tissue, Proceedings of IRCOBI Conference, pp. 191-202, 1990.

[283] Stalnaker, R.L., Mechanical properties of the head, PhD Dissertation Thesis, West Virginia University, Morgantown, WV, USA, 1969.

[284] McElhaney, J.H., Roberts, V.L., Hilyard, J.F., Handbook of Human Tolerance, Japan Automobile Research Institut, Inc., pp. 143, 1976.

[285] Kleiven, S., Influence of direction and duration of impacts to the human head evaluated using the finite element method, Proceedings of IRCOBI Conference, Prague (Czech Republic), pp. 41-57, September 2005.

[286] Yang, K., Head Injury Biomechanics, ICB 118 - Injury Biomechanics Short Course, Warsaw (Poland), September 2011.

[287] Zhou, C., Khalil, T.B., King, A.I., Visoelastic response of the human brain to sagittal and lateral rotational acceleration by finite element analysis, Proceedings of IRCOBI Conference, Dublin, Ireland, 1996.

[288] Tarriere, C., Depressurized cadaver head impacts - some findings, Biomechanics Symposium on Head Injury Prevention Past and Present Research, Wayne State University, pp. 93-144, 1985.

[289] Ivarsson, J., Viano, D.C., Lövsund, P., Aldman, B., Strain relief from the cerebral ventricles during head impact: experimental studies on natural protection of the brain, *J. Biomechanics*, **33**(2):181-189, 2000.

[290] Strich, S.J., Diffuse degeneration of the cerebral white matter in severe dementia following head injury, *J. Neurol. Neurosurg. Psychiat.*, **19**:163-185, 1956.

[291] Dirnhofer, R., Walz, F., Sigrist, T., Zur Mechanischen Belastbarkeit des Tentorium Cerebelli, *Z. Rechtsmed.*, **82**:305-311, 1979.

[292] Becker, E.B., Measurement of Mass distribution Parameters of Anatomical Segments, Proceedings of 16^{th} Stapp Car Crash Conference, SAE Techn. Paper No. 720964, pp. 160-185, 1972.

[293] Walker, L.B., Harris, E.H., Pontius, U.R., Mass, Volume, Center of Mass, and Mass Moment of inertia of Head and Head and Neck of Human Body, Proceedings of 17^{th} Stapp Car Crash Conference, SAE Technical paper No. 730985, pps. 525-537, 1973.

[294] Bain, B.C., Billiar, K.L., Shreiber, D.I., McIntosh, T.K., Meaney, D.F., In vivo mechanical thresholds for traumatic axonal damage. Proceedings of AGARD AMP Specialists, Meeting, Mescalero, New mexico, USA, 7-9 November 1996, published in CP-597, 1997.

[295] Galbraith, J.A., Thibault, L.E., Matteson, D.R., Mechanical and electrical responses of the squid giant axon to simple elongationm *J. Biomech. Engineering*, **115**:13-22, 1993.

[296] Gennarelli, T.A., Thibault, L.E., Tipperman, R., Axonal Injury in the Optic Nerve: A model of Diffuse Axonal Injury in the brain, *J. Neurosurg.*, **71**:244-253, 1989.

[297] Shreiber, D.I., Bain, A.C., Meaney D.F., In vivo thresholds for mechanical injury to the blood-brain barrier, SAE Paper No. 973335, Proceedings of 41^{th} Stapp Car Crash Conference, Society of Automotive Engineers, pp. 177-190, 1997.

[298] Anderson, R.W.G., Brown, C.J., Blumbergs, P.C., Scott, G., Finney, J.W., Jones, N.R., McLean, A.J., Mechanics of axonal injury: An experimental and numerical study of a sheep model of head impact, Proceedings of IRCOBI Conference, Sitges, Spain, pp. 107-120, 1999.

[299] Goldstein, D.M., Mazuchowski, E.L., Gdula, W., Thibault, L.E., In vitro and mathematical models of axonal injury in CNS tissue, Prevention Through Biomechanics, Symposium Proceedings, Wayne State University, pp. 207-215, 1997.

[300] Viano, D.C., Lövsund, P., Biomechanics of brain and spinal-cord injury: analysis of neuropathologic and neurophysiologic experiments, *J. Crash Prevention and Injury Control*, **1**:35-43, 1999.

[301] Verschueren, P., Delye, H., Berckmans, D., Verpoest, I., Goffin, J., Vander Sloten, J., van der Perre, J., Analysis of fracture characteristics of cranial bone for FE modelling, Proceedings of IRCOBI Conference - Madrid (Spain), pp.357-360, September 2006.

[302] Yao, J., Yang, J., Otte, J., Investigation of brain injuries by reconstructions of real world adult pedestrian accidents, Proceedings of IRCOBI Conference, Madrid, Spain, pp. 241-252, September 2006.

[303] Ward, C.C., Chan, M., Nahum, A.M., Intracranial Pressure: a Brain Injury Criterion, Proceedings of $24^t h$ Stapp Car Crash Conference, SAE 801304, 1980.

[304] Baumgartner, D., Mécanismes de lésion et limites de tolérance au choc de la tête humaine - Reconstruction numérique et expérimentale de traumatismes crâniens, Ph.D. dissertation, Université Louis Pasteur Strasbourg, 2001.

[305] Yoganandan, N., Pintar, F.A, Maiman, D., Head linear and rotational accelerations and craniocervical loads in lateral impact, Proceedings of IRCOBI Conference, Madrid, Spain, pp. 127-141, September 2006.

[306] Margulies, S. S., Thibault, L. E., Gennarelli, T. A., Physical model simulations of brain injury in the primate, *Journal of Biomechanics* **23**(8):823-836, 1990.

[307] Ivarsson, J., Viano, D. C., Lovsund, P., Influence of the lateral ventricles and irregular skull base on brain kinematics due to sagittal plane head rotation, *Journal of Biomechanical Engineering*, **124**:422-431, 2002.

[308] Gennarelli, T.A., Thibault, L.E., Biomechanics of head injury, Neurosurgery. Eds. R. H. Wilkins and S. S. Rengachary. 2nd ed. New York: McGraw-Hill Book Company, 1531-1536, 1995.

[309] Goldsmith, W., Plunkett, J. A Biomechanical Analysis of the Causes of Traumatic Brain Injury in Infants and Children, *J Forensic Med Pathology*, **25**(2):89-100, 2004.

[310] Fijalkowski, R.J., Stemper, B.D., Ellingson, B.M., Yoganandan, N., Pintar, F.A., Gennarelli, T.A., Inducing mild traumatic brain injury in the rodent through coronal plane angular acceleration, Proceedings of IRCOBI Conference, Madrid, Spain, pp. 115-125, September 2006.

[311] Douglass, J.M., Nahum, A.M., Roberts, S.B., Applications of Experimental Head Injury Research, Stapp Car Crash Conf, pp. 317-37, 1968.

[312] Higgins, L.S., Schmall, R.A., A Device for the Investigation of Head Injury Effected by Non-Deforming Head Accelerations, Proceedings of 11^{th} Stapp Car Crash Conference, pp. 57-72, 1967.

[313] Shatsky, S.A., Alter, W.A., Evans, D.E., Traumatic Distortions of the Primate Head and Chest: Correlation of Biomechanical, Radiological and Pathological Data, Proceedings of 18^{th} Stapp Car Crash Conference, pp. 351-81, 1974.

[314] Stalnaker, R.L., Robers, V.L., McElhaney, J.H., Side Impact Tolerance to Blunt Tauma, Proceedings of 17^{th} Stapp Car Crash Conference, pp. 377-408, 1973.

[315] Ommaya, A.K., Rockoff, S.D., Baldwin, M., Experimental Concussion: A First Report, *J Neurosurg*, **21**:249-67, 1964.

[316] Fijalkowski, R.J., Ellingson, B.M., Stemper, B.D., Interface Parameters of Impact Induced Mild Traumatic Brain Injury, *Biomed Sci Instrum*, **42**: 108-13, 2006.

[317] Gennarelli, T.A., Pintar, F.A., Yoganandan, N., Biomechanical Tolerances for Diffuse Brain Injury and a Hypothesis for Genotypic Variability in Response to Trauma, 47^{th} Annual Proceedings of the Association for the Advancement of Automotive Medicine, pp. 624-628, 2003.

[318] Margulies, S.S., Gennarelli, T, A Study of Scaling and Head Injury Criteria Using Physical Model Experiments, Proceedings of IRCOBI Conference, pp. 223-234, Göteborg, Sweden, 24-26 June 1985.

[319] Pellman, E.J., Viano, D.C., Tucker, A.M., Concussion in Professional Football: Reconstruction of Game Impacts and Injuries, *Neurosurgery*, **53**(4):799 - 814, 2003

[320] Willinger, R., Baumgartner, D., Numerical and physical modelling of the human head under impact - toward new injury criterion, *Int. J. Vehicle Des.*, **32**(1-2): 94-115, 2001.

[321] Willinger, R., Kang, H.S., Diaw, B.M., Three-dimensional human head finite element model validation against two experimental impacts, *Ann. Biomed. Eng.* **27**(3): 403-410, 1999.

[322] Marjoux, D., Baumgartner, D., Deck, C., Willinger, R., Head injury prediction capability of the HIC, HIP, SIMon and ULP criteria, *Accident Analysis and Prevention*, **40**:1135-1148, 2008.

[323] Margulies S.S., Thibault, L.E., An analytical model of traumatic diffuse brain injury, *Journal of Biomechanical Engineering*, **111**(3):241-249, 1989.

[324] Depreitere B., Van Lierde C., Vander Sloten J., Van Audekercke R., Van der Perre G., Plets C., Goffin J, Mechanics of acute subdural hematomas resulting from bridging vein rupture, *Journal of Neurosurgery*, **104**(6):950-956, 2006.

[325] Verschueren, P., Delye, H., Haex, B., Berckmans, D., Verpoest, I., Goffin, J., van der Sloten, J., van der Perre, G., Implementation of head injury criteria in cyclist-car accidents, Proceedings of IRCOBI Conference, Maastricht, The Netherlands, pp. 405-408, 2007.

[326] Ommaya A.K., Gennarelli T.A., Cerebral concussion and traumatic unconsciousness, Correlation of experimental and clinical observations of blunt head injuries, *Brain*, **97**:633-654, 1974.

[327] Maxeiner, H., Detection of ruptured cerebral bridging veins at autopsy, *Forensic Sci Int*, **89**:103-110, 1997.

[328] Yoganandan, N., Humm, J., Pintar, F.A., Gennarelli, T.A., Three-dimensional head physical properties on the accuracy of injury metrics, Proceedings of IRCOBI Conference, York, UK, pp. 323-326, 2009.

[329] Pang, T.Y., Thai, K.T., McIntosh, A.S., Head and neck dynamics in helmeted hybrid iii impacts, Proceedings of IRCOBI Conference, York, UK, pp. 315-318, 2009.

[330] Ommaya, A.K., Fass, F., Yarnell, P., Whiplash injury and brain damage, *Journal of the American Medical Association*, **204**:285-289, 1968.

[331] Newman, J., Beusenberg, M., Fournier, E., Shewchenko, N., Withnall, C., King, A., Yang, K., Zhang, L., McElhaney, J., Thibault, L., McGinnis, G., A new biomechanical assessment of mild traumatic brain injury - Part 1 - Methodology, Proceedings of the IRCOBI Conference, pp. 17-36, 1999.

[332] Deck, C., Baumgartner, D., Willinger, R., Influence of rotational acceleration on intracranial mechanical parameters under accidental circumstances, Proceedings of IRCOBI Conference, Maastricht, The Netherlands, pp. 185-197, 2007.

[333] Li, J., Zhang, J., Yoganandan, N., Pintar, F.A., Gennarelli, T.A., Role of rotational acceleration and deceleration pulses on brain strains in lateral impact, Proceedings of IRCOBI Conference, Maastricht, The Netherlands, pp. 173-183, 2007.

[334] Fijalkowski, R.J., Stemper, B.D., Pintar, F.A., Yoganandan, N., Gennarelli, T.A., Influence of angular acceleration duration on functional outcomes following mild diffuse brain injury, Proceedings of IRCOBI Conference, Maastricht, The Netherlands, pp. 161-171, 2007.

[335] Hirsch, A.E., Ommaya, A.K., Mahone, R.M., Tolerance of Subhuman Primate Brain to Cerebral Concussion, Washington DC: Dept Navy Naval Ship Res and Developmental Center, 1968.

[336] Mordaka, J., Kleiven, S.,van Schijndel de Nooij, M., de Lange, R., Casanova, L.J.G., Carter, E.L., von Holst, H., The importance of rotational kinematics in pedestrian head to windshield impacts, Proceedings of IRCOBI Conference, Maastricht, The Netherlands, pp. 83-94, 2007.

[337] Kleiven, S., Evaluation of head injury criteria using an FE model validated against experiments on localized brain motion, intra-cerebral acceleration, and intra-cranial pressure, *International Journal of Crashworthiness*, **11**(1):65-79, 2006.

[338] McElhaney, J.H., Fogle, J.H., Melvin, J.W., Haynes, R.R., Roberts, V.L., Alem, N.B., Mechanical properties of cranial bone, *J. Biomechanics*, **3**:495-511, 1970.

[339] Melvin, J.W., McElhaney, J.H., Roberts, V.L., Development of a Mechanical Model of the Human Head - Determination of Tissue Properties and Synthetic Substitute Materials. Proceedings of 14^{th} Stapp Car Crash Conf, Society of Automotive Engineers, SAE Paper No. 700903, 1970.

[340] Robbins, D.H., Wood, J.L., Determination of mechanical properties of the bones of the skull, *Experimental Mechanics*, **9**(5):236-240, 1969.

[341] Zhang, L., Yang, K., Gennarelli, T.A., Mathematical modeling of cerebral concussion: correlations of regional brain strain with clinical symptoms, Proceedings of IRCOBI Conference, Bern, Switzerland, pp. 123-132, 2008.

[342] Zhang, L., Yang, K.H., King, A.I., Viano, D.C., A new biomechanical predicator for mild traumatic brain injury - A preliminary finding. ASME Bioengineering Conference Proceedings, Key Biscayne, FL, USA, 2003.

[343] Viano, D.C., Casson, I.R., Pellman, E.J., Zhang, L., Yang, K.H., King, A.I., Concussion in professional football: Brain responses by finite element analysis - Part 9, *Neurosurgery*. **57**:891-916, 2005.

[344] Tamura, A., Nagayama, K., Matsumoto, T., Hayashi, S., Variation in nerve fiber strain in brain tissue subjected to uniaxial stretch, Proceedings of 51^{th} Stapp Car Crash Conference, pp. 139-154, 2007.

[345] Deck, C., Willinger, R., Improved Head injury criteria based on head FE model, *International Journal of Crashworthiness*, **13**(6):667-679, 2008.

[346] Munsch, M., Anderson, R., Deck, C., Ludes, B., Willinger, R., Influence of head boundary conditions in pedestrian real world head trauma simulations, Proceedings of IRCOBI Conference, York, UK, pp. 287-300, 2009.

[347] Davidsson, J., Angeria, M., Risling, M.G., Injury threshold for sagittal plane rotational induced diffuse axonal injuries, Proceedings of IRCOBI Conference, York, UK, pp. 43-56, 2009.

[348] Cloots, R.J.H., van Dommelen, J.A.W., Kleiven, S., Geers, M.G.D., Traumatic Brain Injury at Multiple Length Scales: Relating Diffuse Axonal Injury to Discrete Axonal Impairment, Proceedings of IRCOBI Conference, Hanover Germany, pp. 119-130, 2010.

[349] Gennarelli, T. A., Spielman, G. M., Langfitt, T. W., Gildenberg, P. L., Harrington, T., Jane, J. A., Marshall, L. F.,Miller, J. D., Pitts, L. H., Influence of the type of intracranial lesion on outcome from severe head injury, *J Neurosurg*, **56**:26-32, 1982.

[350] Gentleman, S.M., Roberts, G.W., Gennarelli, T. A.,Maxwell,W. L., Adams, J. H., Kerr, S., Graham, D. I., Axonal injury: a universal consequence of fatal closed head injury *Acta Neuropathol*, **89**:537-543, 1995.

[351] Smith, D. H., Meaney, D. F., Shull, W. H., Diffuse axonal injury in head trauma, *J Head Trauma Rehabil*, **18**:307-316, 2003.

[352] Ghajari, M., Galvanetto, L., Iannucci, R. Willinger, Intracranial Response in Helmet Oblique Impacts, Proceedings of Ircobi Conference, Warsaw, pp. 90-94, September, 2011.

[353] Asiminei, A.G., Baeck, K., Verbeken, E., Sloten, J.V., Goffin, J., Investigation on strain rate dependency in the biomechanical behaviour of the Superior Sagittal Sinus - Bridging Vein Complex, Proceedings of IRCOBI Conference, Warsaw, pp. 36-39, September 2011.

[354] Graham, D.I., Neuropathology of head injury, Neurotrauma, ed. Narayan RK, Wilberger JE, Povlishock JT, McGraw-Hill, New York, NY, pp. 43-60, 1996.

[355] Douglass, J.M., Nahum, A.M., Roberts, S.B., Applications of Experimental Head Injury Research, Stapp Car Crash Conf, pp. 317-37, 1968.

[356] Meaney, D.F., Smith, D., Ross, D.T., Diffuse Axonal Injury in the Miniature Pig: Biomechanical Development and Injury Threshold, *Crashworthiness and Occupant Protection in Transportation Systems*, **25**:169-175, 1993.

[357] Smith, D.H., Nonaka, M., Miller, R.T., Immediate Coma Following Inertial Brain Injury Dependent on Axonal Damage in the Brainstem, *J Neurosurg*, **93**:315-322, 2000.

[358] Morrison III, B., Cater, H.L., Wang, C.C.B., Thomas, F.C., Hung, C.T., Ateshian, G.A., Sundström, L.E., A tissue level tolerance criterion for living brain developed in an in vitro model of traumatic mechanical loading, Proceedings of 47^{th} Stapp Car Crash Conference, SAE Paper No. 2003-22-0006, 2003.

[359] Thibault, L.E., Brain injury from the macro to the micro level and back again: What have we learned to date?, Proceedings of IRCOBI Conference, Eindhoven, The Netherlands, pp. 3-25, 1993.

[360] Adams, J. H., Doyle, D., Ford, I., Diffuse axonal injury in head injury: definition, diagnosis and grading, *Histopathology*, **15**:49-59, 1989.

[361] Strich, S. J., Shearing of Nerve Fibres as a Cause of Brain Damage due to Head Injury, *Lancet*, **2**:443, 1961.

[362] Povlishock, J. T., Traumatically induced axonal injury: pathogenesis and pathobiological implications, *Brain Pathol*, **2**(1):1-12, 1992.

[363] Kleiven, S., Predictors for Traumatic Brain Injuries Evaluated through Accident Reconstructions. Proceedings of the 51^{th} Stapp Car Crash Conference, 07S-27, 2007.

[364] Löwenhielm P., Mathematical simulation of gliding contusions, *J Biomech*, **8**:351, 1975.

[365] Ommaya, Biomechanics of head injury, in biomechanics of Trauma, Eds. Nahum, Melvis, Appleton-Century-Crofts, Norwalk, 1984.

[366] Anderson, R., A study of the biomechanics of axonal injury, PhD thesis, University of Adelaide, 2000.

[367] Franklyn, M., Fildes, B., Dwarampudi, R., Zhang, L., Yang, K., Sparke, L., Eppinger, R., Analysis of computer models for head injury investigation, Proceedings of the 18^{th} International Technical Conference on Enhanced Safety Vehicles, 2003.

[368] Hallquist, J.O., User Manual for LS-Dyna, LSTC Lab., USA, 1997.

[369] Pam-Crash Software, ESI Group, France.

[370] Liu, D.S., Fan, C.M., Lee, M.C., Yen, C.Y., A dynamic finite element simulation of the impact of motorcycle helmets, *Crashworthiness, Occupant Protection and Biomechanics in Transportation Systems*, **38**:199-207, 1997.

[371] Moss, S., Huang, Y., Keer, T., Shah, B., Development of an advanced finite element model database of the hybrid III crash dummy family, SAE Transactions Paper No. 971042, pp. 1899-1917, 1997.

[372] Willinger, R., Diaw, B.M., Kang, H.S., Finite element modelling of skull fractures caused by direct impact, *Int. Journal of Crashworthiness*, **5**(3):249-258, 2000.

[373] Delille C., Drazetic P., Masson C., Markiewicz E., Canaple B., Cesari D., A new head physical model, Proceedings of IRCOBI Conference, Isle of Man, UK, October 2001.

[374] Darvish, K.K., Crandall, J.R., Influence of brain material properties and boundary conditions on brain response during dynamic loading, Proceedings of IRCOBI Conference, Munich, Germany, 2002.

[375] Iwamoto, M., Yoshikatsu, K., Watanabe, I., Furusu, K., Miki, K., Hasegawa, J., Development of a finite element model of the total human model for safety (thums) and application to injury reconstruction, Proceedings of IRCOBI Conference, Munich, Germany, 2002.

[376] Fredriksson, L.A., A Finite Element Data Base for Occupant Substitutes, Doctoral Thesis, Dissertations No. 447, Division of Solid Mechanics, Dept. of Mech. Engineering, Linköping University, Sverige, 1996.

[377] Horgan, T.J., Gilchrist, M.D., Investigations into finite element modelling aspects of the human head, Proceedings of IRCOBI Conference, Lisbon, Portugal, pp. 347-348, September 2003.

[378] Yoganandan, N., Zhang, J., Pintar, F.A., Gennarelli, T.A., Kuppa, S., Eppinger, R.H., Biomechanics of lateral skull fracture, Proceedings of IRCOBI Conference, Lisbon, Portugal, pp. 69-78, September 2003.

[379] Zhang, L., Ramesh, D., Yang, K.H., King, A.I., Effectiveness of the football helmet assessed by finite element modeling and impact testing, Proceedings of IRCOBI Conference, Lisbon, Portugal, pp. 27-38, September 2003.

[380] Neale, M., McGrath, M., Baumgartner, D., Willinger, R., Comparative study between finite element human head and dummy head model responses under impact, Proceedings of IRCOBI Conference, Graz, Austria, pp. 69-78, September 2004.

[381] Bandak, F., Zhang, A., DiMasi, F., Masiello, P., Eppinger, R. SIMon: A simulated injury monitor; application to head injury assessment, Proceedings of the 17^{th} International Technical Conference on the Enhanced Safety of Vehicles (ESV), Amsterdam, The Netherlands, 4-7 June, 2001.

[382] Willinger, R., Baumgartner, D., Chinn, B., Schuller, E., New dummy head prototype: development, validation and injury criteria, *International Journal of Crashworthiness*, **6**:281-293, 2001.

[383] Fredriksson, R., Boström, O., Zhang, L., Yang, K., Influence of pop-up hood systems on brain injuries for vulnerable road users, Proceedings of IRCOBI Conference, Madrid ,Spain, pp. 253-264, September 2006.

[384] Monson, K.L., Goldsmith W., Barbaro N.M., Manley G.T., Axial mechanical properties of fresh human cerebral blood vessels, *J Biomech Eng*, **125**(2): 288-294, 2003.

[385] Committee on medical aspects of automotive safety: rating the severity of tissue damage, the abbreviated scale, *JAMA*, **215**:277-280, 1971.

[386] Kumareasan, F.A., Radhakrishnan, S., Ganesan, N., Generation of geometry of closed human head and discretization for finite element analysis, *Med. Biol. Eng. Comput.*, *33*:349-353, 1995.

[387] Voo, L., Kumareasan, F.A., Pintar, F.A., Yoganandan, N., Sances, A.S., Finite element models of the human head, *Med. Biol. Eng. Comput.*, *34*:375-381, 1996.

[388] Ruan, J.S., Khalil, T., King, A.I., Human head dynamic response to side impact by finite element modeling, *J. Biomed. Eng. Trans.*, ASME **113**:276-283, 1991.

[389] Haug, E., Biomechanical models in vehicle accident simulation, in: J.A.C. Ambrosio, M.F.O. Seabra Pereira, P.F. Silva (Eds.), Crash-worthiness of Transportation Systems: Structural Impact and Occupant Protection, Kluwer Academic Publishers, Dordrecht, pp. 237-259, 1997.

[390] Willinger, R., Baumgartner, D., Guimberteau, T., Dynamic characterization of motorcycle helmets: modelling and coupling with the human head, *Journal of Sound and Vibration*, **235**:611-625, 2000.

[391] Kleiven, S., Finite Element Modeling of the Human Head, Licentiate Thesis, Technical Report 2000-9, Department of Aeronautics, Royal Institute of Technology, Stockholm, Sweden, 2000.

[392] Ruan, J.S., Impact head injury analysis using an explicit Finite Element Human Head Model, *J. Traffic Medicine*, **25**:33-40, 1997.

[393] Willinger, R., Taleb, L., Pradoura, P., From the finite element model to the physical model, Proceedings of IRCOBI Conference, Brunnen, pp. 245-260, 1995.

[394] Kleiven, S., Hardy, W.N., Correlation of an FE Model of the Human Head with Experiments on Localized Motion of the Brain: Consequences for Injury Prediction, SAE Paper No. 02S-76, Society of Automotive Engineers, Proceedings 45^{th} Stapp Car Crash Journal, 2002.

[395] Deck, C., Nicolle, S., Willinger, R., Human head fe modelling : improvement of skull geometry and brain constitutive laws, Proceedings of IRCOBI Conference, Graz, Austria, pp. 79-92, September 2004.

[396] Franklyn, M., Fildes, B., Zhang, L., Yang, K.H., Sparke, L., Analysis of finite element models for head injury investigation: Reconstruction of four real-world impacts, *Stapp Car Crash Journal*, *49*:1-32, 2005.

[397] Kleiven, S., Biomechanics as a forensic science tool - Reconstruction of a traumatic head injury using the finite element method, *Scand J Forens Sci.*, **2**:73-78, 2006.

[398] Ruan, J.S., Khalil, T.B., King, A.I., Finite Element Modeling of Direct Head Impact, Proceedings of 37^{th} Stapp Car Conference, SAE Paper No.933114, 1993.

[399] Ruan, J.S., Khalil, T., King, A.I., Dynamic response of the human head to impact by three-dimensional finite element analysis, *Journal of Biomechanics*, **116**:44-50, 1994.

[400] Willinger, R., Diaw, B.M., Kang, H.S., Finite element modeling of skull fractures caused by direct impact, *Int. Journal of Crashworthiness*, **5**(3):249-258, 2000.

[401] Horgan, T.J., Gilchrist, M.D., Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations, *Int. J. of Crashworthiness*, **9**(4):401-418, 2004.

[402] Zong, Z., Lee, H.P., Lu, C., A three-dimensional human head finite element model and power flow in a human head subjected to impact loading, *J. of Biomechanics*, **39**(2):284-292, 2006.

[403] Shugar, T.A., A finite element head injury model, Report nř DOT HS 289-3-550-TA, **1**, 1977.

[404] Hosey, R.R., Liu, Y.K., A homeomorphic finite element model of impact head and neck injury, *I. C. P. of Finite Elements in Biomechanics*, **2**:379-401, 1980.

[405] DiMasi, F., Marcus, J., Eppinger R., 3D anatomic brain model for relating cortical strains to automobile crash loading, Proceedings of the International Technical Conference on Experimental Safety Vehicles, NHTSA, **2**:916-923, 1991.

[406] Mendis, K., Finite element modelling of the brain to establish diffuse axonal injury criteria, PhD Dissert., Ohio State University, 1992.

[407] Bandak, F.A., van der Vorst, M.J., Stuhmiller, L.M., Mlakar, P.F., Chilton, W.E., Stuhmiller, J.H., An imaging based computational and experimental study of skull fracture: finite element model development, Proceedings of the Head Injury Symposium, Washington DC, 1994.

[408] Chu, C.S., Lin, M.S., Huang H.M., Lee M.C., Finite element analysis of cerebral contusion, *J Biomech*, **27**:187-194, 1994.

[409] Trosseille X., Tarrière C., Development of a FEM of the human head according to specific test protocol, Proceedings of the 36^{th} Stapp Car Crash Conference, pp. 235-253, 1992.

[410] Cloots, R. J. H., Gervaise, H.M. T., van Dommelen, J. A.W., Geers, M. G. D., Biomechanics of traumatic brain injury: influences of the morphologic heterogeneities of the cerebral cortex, *Ann Biomed Eng*, **36**:1203-1215, 2008.

[411] Ho, J., Kleiven, S., Can sulci protect the brain from traumatic injury?, *J Biomech*, **42**:2074-2080, 2009.

[412] Lauret, C., Hrapko, M., van Dommelen, J. A. W., Peters, G. W. M., Wismans, J. S. H. M., Optical characterization of acceleration-induced strain fields in inhomogeneous brain slices, *Med Eng Phys*, **31**:392-399, 2009.

[413] Ward, C.C., Thompson, R.B., The development of a detailed finite element brain model, Proceedings 19^{th} Stapp Car Crash Conference, SAE Paper, vol 751163, SAE, New York, pp 641-674, 1975.

[414] Raul, J.S., Deck, C., Willinger, R., Ludes, B., Finite-element models of the human head and their applications in forensic practice, *Int J Legal Med*, **122**:359-366, 2008.

[415] Ruan, J.S., Khalil, T., King, A.I., Dynamic response of the human head to impact by three-dimensional finite element analysis, *J Biomech Eng*, **116**(1):44-50, 1994.

[416] Zhou, C., Kahlil, T.B., Dragovic, L.J., Head injury assessment of a real world crash by finite element modelling, Proceedings of the Advisory Group for Aerospace Research and Development, AGARD-Conference Proceedings, New Mexico, pp 81-87, November 1996.

[417] Brands, D.W., Bovendeerd, P.H., Wismans, J.S.H.M., On the potential importance of non-linear viscoelastic material modelling for numerical prediction of brain tissue response, Proceedings 46^{th} Stapp Car Crash Conference, SAE paper vol 2002-22-0006, pp. 103-121, 2002.

[418] Brands, D.W., Predicting brain mechanics during closed head impact:Numerical and constitutive aspects, Ph.D. dissertation, University of Eindhoven, The Netherlands, 2002.

[419] Khalil, T.B., Viano, D.C., Critical issues in finite element modelling of head impact, Proceedings 26^{th} Stapp Car Crash Conference, SAE paper, vol 821150, pp. 87-102, 1982.

[420] Nicolle, S., Lounis, M., Willinger, R., Shear properties of brain tissue over a frequency range relevant for automotive impact situations: New experimental results, *Stapp Car Crash J*, **48**:1-20, 2004.

[421] Mendis, K.K., Stalnaker, R.L., Advani, S.H., A constitutive relationship for large deformation finite element modelling of brain tissue, *J Biomech Eng*, **117**:279-285, 1995.

[422] Horgan, T.J., Gilchrist, M., Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations, *Int J Crashworthiness*, **9**:401-418, 2004.

[423] Kimpara, H., Nakahira, Y., Iwamoto, M., Miki, K., Ichihara, K., Kawano Taguchi, T., Investigation of anteroposterior head-neck responses during severe frontal impacts using a brain-spinal cord complex FE model, Proceedings 50^{th} Stapp Car Crash Conference, pp. 509-544, 2006.

[424] Takhounts, E.G., Ridella, S.A., Hasija, V., Tannous, R.E., Campbell, J.Q., Malone, D., Danelson, K., Stitzel, J., Rowson, S., Duma, S., Investigation of Traumatic Brain Injuries Using the Next Generation of Simulated Injury Monitor (SIMon) Finite Element Head Model, *Stapp Car Crash Journal*, **52**, November 2008.

[425] Kim, J.E., Kim, Y.H., Li, Z., Eberhardt, A.W., Soni, B.K., Evaluation of Traumatic Brain Injury using Multi-Body and Finite Element Models, 17^{th} IMACS World Congress, Scientific Computation, Applied Mathematics and Simulation, Paris, France, July 11 - 15, 2005.

[426] Motherway, J., Doorly, M.C., Curtis, M., Gilchrist, M.D., Head impact biomechanics simulations: A forensic tool for reconstructing head injury?, *Legal Medicine*, **11**:S220-S222, 2009.

[427] Post, A., Hoshizaki, B., Gilchrist, M.D., Finite element analysis of the effect of loading curve shape on brain injury predictors, *Journal of Biomechanics*, **45**:679-683, 2012.

[428] Yogananda, N., Li,J., Zhang,J., Pintar,F.A., Gennarelli,T.A., Influence of angular acceleration deceleration pulse shapes on regional brain strains, *Journal of Biomechanics*, **41**:2253-2262, 2008.

[429] Doorly, M.C.,Gilchrist, M.D., The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls, *Computer Methods in Biomechanics and Biomedical Engineering*, **9**(6):371-377, 2006.

[430] Iwamoto, M., Nakahira, Y., Tamura, A., Kimpara, H., Watanabe, I., Miki, K., Development of advanced human models in thums, 6^{th} European LS-DYNA Users Conference, pp. 47-56, 2007.

[431] Iwamoto, M., Recent Advances in THUMS: development of the detailed head-neck and internal organs, and THUMS family, LS-DYNA & JMAG User Conference, Japan, 2003.

[432] Ipek, H., Mayer, C., Deck, C., Luce, H., de Gueselle, p., Willinger, R., Coupling of Strasbourg University head model to thums human body fe model: validation and application to automotive safety, paper number 09-0384, pp. 1-13, 2009.

[433] Deck, C., Willinger, R., Head injury prediction tool for predictive systems optimization, Proceedings of 7^{th} European LS-DYNA Conference, 2009.

[434] Chatelin, S., Deck, C., Renard, F., Kremer, S., Heinrich, C., Armspach, J.P., Willinger, R., Computation of axonal elongation in head trauma finite element simulation, *Journal of the mechanical behaviour of biomedical materials*, **4**:1905-1919, 2011.

[435] Cardamone, L., Analisi numerica del trauma cranico da impatto, Technical report, Bioengineering Laboratory, University of Salerno, Italy, 2005.

[436] El Sayed, T., Mota, A., Fraternali, F., Ortiz, M., Biomechanics of traumatic brain injury, *Computer Methods in Applied Mechanics and Engineering*, **197**:4692-4701, 2008.

[437] Pinnoji, P.K., Mahajan, P., Bourdet, N., Deck, C., Willinger, R., Impact dynamics of metal foam shells for motorcycle helmets: Experiments & numerical modelling, *International Journal of Impact Engineering*, **37**274-284, 2010.

[438] Ho, J, Kleiven, S., Dynamic response of the brain with vasculature: A three-dimensional computational study, *Journal of Biomechanics*, **40**:3006-3012, 2007.

[439] Zhang, J., Yoganandan, N., Pintar, F.A., Guan, Y., Gennarelli, T.A., Biomechanical differences between contact and non-contact head impacts in vehicle crash tests, Department of Neurosurgery, Medical College of Wisconsin, United States, Paper Number 07-0352, 2007.

[440] Krave, U., Al-Olama, M., Hansson, H.A., Rotational Acceleration Closed Head Flexion Trauma Generates More Extensive Diffuse Brain Injury than Extension Trauma, *Journal of neurotrauma*. **28**:57-70, 2011.

[441] Kleiven, S., Head Injury Biomechanics and criteria, Biomechanics and Neuronics, course litterature, KTH, 2007.

[442] Horgan, T.J., A finite element model of the human head for use in the study of pedestrian accidents, University College Dublin, Dublin, 2005.

[443] Kleiven, S., A parametric study of energy absorbing foams for head injury prevention, Royal Institute of Technology, Sweden, Paper Number 07-0385, 2007.

[444] Dirisala, V., Karami, G., Ziejewski, M., Effects of neck damping properties on brain response under impact loading, *International Journal for Numerical Methods in Biomedical Engineering*, **28**(4):472-494, 2011.

[445] Horgan, T.J., Gilchrist, M.D., The creation of three-dimensional finite element models for simulating head impact biomechanics, *International Journal of Crashworthiness*, **8**(4):353-366, 2003.

[446] Zhang, L., Yang, K.H., King, A.I., Viano, D.C., A new biomechanical predictor for mild traumatic brain injury - a preliminary finding, Proceedings of Summer Bioengineering Conference, Sonesta Beach Resort in Key Biscayne, Florida, June 25-29, 2003.

[447] McLean, A.J., Anderson, R.W.G, Biomechanics of closed head injury, Head Injury Edited by Peter Reilly and Ross Bullock, Published in 1997 by Chapman & Hall, London, ISBN 0 412 58540 5.

[448] Raul, J.S., Baumgartner, D., Willinger, R., Ludes, B., Finite element modelling of human head injuries caused by a fall, *International Journal of Legal Medicine*, **120**:212-218, 2006.

[449] Zhou, C., Kahlil, T.B., Dragovic, L.J., Head injury assessment of a real world crash by finite element modelling, Proceedings of the AGARD Conference, New Mexico, USA, pp 81-87, November 1996.

[450] Newman, J., On the Use of the Head Injury Criterion (HIC) in Protective Headgear Evaluation, Proceedings of 9^{th} stapp car crash conference, pp. 17-19, San Diego, California, 1975.

[451] Belingardi, G., Chiandussi, G., Gaviglio, I., Development and validation of a new finite element model of human head, Politecnico di Torino, Dipartimento di Meccanica, Italy, Paper Number 05-0441, 2005.

[452] Anzelius, A., The effect of an impact on a spherical liquid mass, *Acta. Pathol. Microbiol. Scand.*, **48**:153-159, 1943.

[453] Chan, H.S., Mathematical model for closed head impact, Proceedings of 18^{th} Stapp Car Crash Conference, Society of Automotive Engineers, pages 557-579, 1974.

[454] Kenner, V.H., Goldsmith, W., Dynamic loading of a fluid filled spherical shell, *International Journal of Mechanical Sciences*, **1**:557-568, 1972.

[455] Khalil, T.B., Hubbard, R.P., Parametric study of head response by finite element modelling, *J Biomech.*, **10**:119-132, 1977.

[456] Kuijpers, A.H., Claessens, M.H., Sauren, A.A., The influence of different boundary conditions on the response of the head to impact: A two-dimensional finite element study, *J. Neurotrauma*, **12**(4):715-724, 1995.

[457] ISO-DIS-6220: Headforms for Use in the Testing of Protective Helmets, International Standards Organisation, Geneva, Switzerland, 1983.

[458] Arai, Helmet testing standards, Netherlands, 2006. helmet http://www.araihelmeteurope.com 4/10/2011

[459] U.S. Department of Transportation, Federal Motor Carrier Safety Administration, Standard No. 218, Motorcycle helmets, Regulations current to 29/02/2012. (http://www.fmcsa.dot.gov/rules-regulations/administration/fmcsr/fmcsrruletext.aspx?reg=571.218 consulted in 26/03/2012).

[460] Thom, D.R., Hurt, H.H., Smith, T.A., Motorcycle helmet headform and test apparatus comparison, Head protection research laboratory, United States, Paper Number 98-S10-P-29, 1998.

[461] Thom, D.R., Comparison Tests of Motorcycle Helmets Qualified to International Standards, Proceedings of the 2006 International Motorcycle Safety Conference, Motorcycle Safety Foundation, 2006.

[462] Otte, D., Chinn, B., Doyl,e D., Mäkitupa, S., Sturrock, K., Schuller E., Contribution to Final Report of COST 327 Project, University of Hannover, 1999.

[463] Harrison, T. I., Mills, N. J., Turner, M. S., Jockeys head injuries and skull cap performance, Proccedings of International IRCOBI Conference, Sept 11-13, Dublin, pp. 49-62, 1996.

[464] Aare, M., Kleiven, S., Halldin, P., Injury tolerances for oblique impact helmet testing, *International Journal of Crashworthiness*, **9**(1):15-23, 2004.

[465] Mills, N. J., Critical evaluation of the SHARP motorcycle helmet rating, *International Journal of Crashworthiness*, **15**(3):313-342, 2010.

[466] Yoganandan, N., Pintar, F.A., Zhang, J., Gennarelli, T.A., Beuse, N., Biomechanical aspects of blunt and penetrating head injuries, Dublin: IUTAM Symposium Biomechanics of Impacts, pp. 173-184, 2005

[467] Aldman, B., Lundell, B., Thorngren, L., Oblique impacts - a parametric study in crash helmets, Proceedings of IRCOBI Conference, Lyon, pp. 129-141, 1978.

[468] Ask Nature. http://www.asknature.org/product/52b0b0cd3c9afbf7d6cbb8ed83aac6af#menuPopup 26/02/2012

[469] Lazer helmets. http://www.lazerhelmets.com/innovations/superskin/ 26/02/2012

[470] Phillips Head Protection System. http://www.phillipshelmets.com/HOME.htm 26/02/2012

[471] MIPS helmet. http://mipshelmet.com 26/02/2012

[472] Mills, N.J., Gilchrist, A., Mathematical modelling of the effectiveness of helmets in head protection, Proceedings of IRCOBI conference, pp. 215-226, Bergisch Gladbach, W. Germany, 1988.

[473] Kostner, H., Stocker, U.W., Mathematische Analyse der Sto absorption im Schutzhelmmaterial, *VOIbericht*, **657**:211-244, 1987.

[474] van Schalkwijk, R., Helmet shock simulation with MARC using a hypo-elastic foam model. MTR-9304, MARC Analysis Research Corporation, The Netherlands, 1993.

[475] Wilson, K.H., Carr, D.J., Head protection the application of mathematical modeling, Proceedings of IRCOBI Conference, 8-10 September, Eindhoven Technical University, Eindhoven, The Netherlands, 1993.

[476] Scott, R.S., A dynamic FE simulation of the impact of motorcycle helmets, Crashworthiness, occupant protection and biomechanics in transportation systems, AMD vol. 225, New York: ASME, 1997.

[477] Brands, D.W.A., Thunnissen, J.G.M., Wismans, J.S.H.M., Modelling head injury countermeasures: a 3D helmet model, AGARD Meeting on Impact Head Injury, New Mexico, November 7-9 1996, AGARD conference proceedings 597, 1997.

[478] Aiello, M., Galvanetto, U., Iannucci, L., Numerical simulations of motorcycle helmet impact tests, *International Journal of Crashworthiness*, **12**:1-7, 2007.

[479] Finan, J.D., Nightingale, R.W., Myers, B.S., The Influence of Reduced Friction on Head Injury Metrics in Helmeted Head Impacts, *Traffic Injury Prevention*, **9**(5):483-488, 2008.

[480] Liu, D.S., Fan, C.M., Lee, M.C., Yen, C.Y., Development and Application of a 3D Finite Element Simulation Model of Impact of Motorcycle Helmet IJCRASH'98 Conference, Dearborn, MI, 1998.

[481] Khalil, T.B., Goldsmith, W., Sackman, J.L., Impact on a model head-helmet system, *International Journal of Mechanical Sciences*, **16**:609-625, 1974.

[482] Hamouda, A.M.S., Hashmi, M.S.J., Modelling the impact and penetration events of modern engineering materials: characteristics of computer codes and material models, *Journal of Materials Processing Technology*, **56**:847-862, 1996.

[483] Masso-Moreu, Y., Mills, N.J., Impact compression of polystyrene foam pyramids, *International Journal of Impact Engineering*, **28**:653-676, 2003.

[484] Ouellet, S., Cronin, D., Worswick, M., Compressive response of polymeric foams under quasi-static, medium and high strain rate conditions, *Polymer Testing*, **25**(6):731-743, 2006.

[485] ABNT, Associação Brasileira de Normas Técnicas. Capacetes de motocicletas e similares. Technical Report NBR 7471:2001, ABNT - Associação Brasileira de Normas Técnicas, Av. Treze de Maio 13, 28 Andar, Caixa Postal 1680, Rio de Janeiro, 2001.

[486] Catia V5, User Manual, Dassault Systems, 2008.

[487] Aare, M., Halldin, P., A New Laboratory Rig for Evaluating Helmets subject to Oblique Impacts, *Traffic Injury Prevention*, **4**(3):240-248, 2003.

[488] MSC/DYNA User's Manual: MSC/DYNA Version 1, MacNeal-Schwendler Corporation, 1990.

[489] Ferner, H., Staubesand, J., Atlas d'anatomie humaine-tête, cou, membre thoracique, Edition Médicales internationales Paris, 1985.

[490] Kruse, S.A., Rose, G.H., Glaser, K.J., Manduca, A., Felmlee, J.P., Jack, Jr. C.R., and Ehman, R., Magnetic Resonance Elastography of the brain, NeuroImage 39, 231-237, 2007.

[491] Piekasrski, K., Fracture of Bone, *J. appl. Physics*, **41**, No. 1, 1970.

[492] Gentleman, S.M., Roberts, G.W., Gennarelli, T.A., Maxwell, W.L., Adams, J.H., Kerr, S., Axonal injury: a universal consequence of fatal closed head injury?, *Acta Neuropathol (Berl)*, **89**:537-543, 1995.

[493] Willinger, R., Césari, D., Evidence of cerebral movement at impact through mechanical impedance methods, Proceedings of IRCOBI Conference, Bron, France, pp. 203-213, 1990.