Prevention of Head Injuries
Focusing Specifically on Oblique Impacts

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Preface and acknowledgements

This thesis was carried out at the Division of Neuronic Engineering, in the Department of Aeronautics from its inception until January 2003, and subsequently at the "Centre for Technology within Healthcare" - both situated at the Royal Institute of Technology (KTH) in Stockholm. All the work was done in close cooperation with my supervisors, Prof. Hans von Holst and Dr. Peter Halldin, and colleagues and friends Dr. Svein Kleiven and Dr. Karin Brolin. I would like to thank all of you for your support and inspiration. I would also like to thank everyone else at the Department of Aeronautical and Vehicle Engineering. Furthermore, I wish to express my gratitude to Länsförsäkringars forskningsfond, VINNOVA, NUTEK and Rikspolisstyrelsen for their financial support. I would also like to thank Peter Collopy for helping me with the proofreading of my English texts. Last but not least, I would like to thank my family for being there for me and of course you Therese.

Stockholm, November 2003.

Sammanfattning

Det stora antalet skador till följd av trafikolyckor i världen är ett ökande problem, speciellt i utvecklingsländerna. 1988 dödades mer än en miljon människor i trafiken, medan ungefär tio gånger så många skadades. Skador på det centrala nervsystemet och speciellt skador på huvudet är extra kritiska för den skadades levnadssituation. Föreliggande avhandling innehåller fem forskningsartiklar, som fokuserar på huvudskador och skydd för huvudet. Resultaten i dessa artiklar föreslår nya och mer effektiva sätt att skydda huvudet, speciellt vid trafikolyckor med motorcyklar och mopeder.

För att få en överblick över vilka skador, som är vanligast vid motorcykel och mopedolyckor, utfördes en statistisk studie från data på svenska olyckor (Artikel A). 27,100 personer vårdades och bodde kvar minst en natt på svenska sjukhus pga skador orsakade från motorcykel- och mopedolyckor mellan åren 1987 och 1999. Antalet skadade per år och det totala antalet vårddagar per år till följd av dessa skador minskade under den senare delen av denna tidsperiod. Män var åtta gånger så utsatta som kvinnor. Skador på huvudet var den vanligaste diagnosen följt av skador på underbenen. Hjärnskakning var den vanligaste huvudskadan. Resultaten visar på behovet av bättre system för att skydda huvudet.

Det vanligaste islaget mot huvudet i en motorcykel- eller mopedolycka är enligt statistiken ett sneda islag. Sneda islag genererar rotationer av huvudet, vilket är en vanlig orsak till de allvarligaste typerna av hjärnskador. En ny typ av testrigg har konstruerats för att kunna analysera dessa sneda islag (Artikel B). Den nya testriggen ger värdefulla testdata data i hastigheter upp till 50 km/h, med islagsvinklar varierande från rent tangentiella till rent radiella. Resultaten visar att den nya konstruktionen av testrigg förbättrar metoden för analys av accelerationer i ett Hybrid III huvud vid hjälmprovning för sneda islag.

Vid analys av prestanda hos nya motorcykelhjälmar uppkommer ofta avvikelser i testresultaten. För att avgöra vad, som orsakar dessa avvikelser, simulerades ett antal sneda islag med hjälp av Finita Element Metoden (FEM) (Artikel C). Av testade parametrar visade sig friktions-koefficienterna mellan islagsyta och hjälm och mellan hjälm och huvud ha störst inverkan på rotationsaccelerationerna. Resultaten visar att tunnare och därmed även vekare hjälmskal samt en mjukare liner (cellplast/energiabsorbent), gav bättre skydd mot de studerade islagen.

Idag saknas globalt accepterade toleransvärden för när en skada kan antas uppstå vid sneda islag. Därför har mer relevanta toleransvärden studerats och vilka tar hänsyn till både translationer och rotationer av huvudet (Artikel D). I studien användes FE modeller av (a) ett mänskligt huvud, (b) ett Hybrid III huvud och (c) en experimentell hjälm. Olika kriterier föreslogs för olika islag. Resultaten visar att såväl translations- som rotationseffekterna är viktiga när man skall förutsäga töjningsnivåerna i hjärnvävnaden vid sneda islag mot huvudet.

I syfte att reducera antalet huvudskador i samhället och för att få en bättre förståelse för islag mot hjälmar utifrån andra aspekter, studerades även ett ballistiskt islag (Artikel E). Metoden simulerade effekten av olika styvhet i hjälmskalet samt effekten av olika islagsvinklar. Vid analysen utnyttjades samma FE modell av det mänskliga huvudet som i Artikel D, men med den skillnaden att det använda huvudet skyddades av en modell av en ballistisk komposithjälm. Resultaten visar att hjälmskalet bör vara tillräckligt styvt för att förhindra att hjälmens insida kommer i kontakt med huvudet vid ett islag. Dessutom visar resultaten att töjningarna i hjärnvävnaden var högre för vissa typer av sneda islag än för rent radiella.

Sammanfattningsvis beskriver avhandlingen mönstret för motorcykel- och mopedolyckor i Sverige under åren 1987 och 1999. Avhandlingen visar också att antalet huvudskador till följd av dessa olyckor kan reduceras genom bättre testmetoder och bättre hjälmkonstruktioner. Med hjälp av FE metoden är det möjligt att på ett bättre sätt simulera realistiska islag mot huvudet.

Abstract

The massive number of injuries sustained in traffic accidents is a growing problem worldwide, especially in developing countries. In 1998, more than one million people were killed in traffic accidents worldwide, while about ten times as many people were injured. Injuries to the central nervous system and in particular to the head are especially critical to human life. This thesis contains five research papers looking at head injuries and head protection, proposing new and more efficient ways of protecting the head, especially in traffic accidents.

In order to define the national dimensions of the patterns of injuries incurred in motorcycle and moped accidents in Sweden, a statistical survey was performed on data spanning a 13-year period (Paper A). In Sweden, 27,100 individuals received in-patient care for motorcycle and moped accident injuries between 1987 and 1999. The motorcycle and moped injury rate reduced in the second half of the study period, so too were the total number of days of treatment per year. Males had eight times the incidence of injuries of females. Head injuries were the single most frequent diagnosis, followed by fractures of the lower limbs. Concussion was the most frequent head injury. These statistics clearly show the need for better head injury prevention systems.

According to the statistics, the most common type of impact to the head in motorcycle and moped accidents is an oblique impact. Oblique impacts generate rotations of the head, which are a common cause of the most severe head injuries. Therefore a new test rig was constructed to reproduce oblique impacts to a helmeted dummy head, simulating those occurring in real life accidents (Paper B). The new test rig was shown to provide useful data at speeds of up to 50 km/h and with impact angles varying from purely tangential to purely radial. This innovative test rig appears to provide an accurate method for measuring accelerations in oblique impacts to helmets.

When testing the performances of motorcycle helmets, discrepancies are usually seen in the test results. In order to evaluate these discrepancies, the finite element method (FEM) was used for simulations of a few oblique helmet impacts (Paper C). Among the parameters studied, the coefficients of friction between the impacting surface and the helmet and between the head and the helmet had the most significant influence on the rotational accelerations. Additionally, a thinner and consequently also weaker shell and a weaker liner, provided better protection for the impacts studied.

Since there are no generally accepted global injury thresholds for oblique impacts to the human head, a study was designed to propose new injury tolerances accounting for both translations and rotations of the head (Paper D). In that study, FE models of (a) a human head, (b) a Hybrid III dummy head, and (c) the experimental helmet were used. Different criteria were proposed for different impact scenarios. Both the translational and the rotational effects were found to be important when proposing a predictor equation for the strain levels experienced by the human brain in simulated impacts to the head.

In order to reduce the level of head injuries in society and to better understand helmet impacts from different aspect, a ballistic impact was also studied (Paper E). The effects of different helmet shell stiffness and different angles of impacts were simulated. In this study, the same FE head model from Paper D was used, however here it was protected with a model of a composite ballistic helmet. It was concluded that the helmet shell should be stiff enough to prevent the inside of the shell from striking the skull, and that the strains arising in the brain tissue were higher for some oblique impacts than for purely radial ones.

In conclusion, this thesis describes the injury pattern of motorcycle and moped accidents in Sweden. This thesis shows that the injuries sustained from these accidents can be reduced. In order to study both translational as well as rotational impacts, a new laboratory test rig was designed. By using the finite element method, it is possible to simulate realistic impacts to the head and also to predict how severe head injuries may potentially be prevented.

Dissertation

Paper A
Injuries from Motorcycle- and Moped crashes in Sweden from 1987-1999
Magnus Aare and Hans von Holst
Published in Injury Control and Safety Promotion, Vol. 10, No. 1, pp. 131-138, 2003.

Paper B
A New Laboratory Rig for Evaluating Helmets subject to Oblique Impacts
Magnus Aare and Peter Halldin
Published in Traffic Injury Prevention, Vol. 4, Issue 3, pp. 240-248, 2003.

Paper C
Influence of Various Parameters on Rotational Acceleration inside a Helmeted Dummy Head during Oblique Impacts - a Parametric Study using FEM
Magnus Aare and Peter Halldin
Submitted to Traffic Injury Prevention

Paper D
Injury Tolerances for Oblique Impact Helmet Testing
Magnus Aare, Svein Kleiven and Peter Halldin
Presented and published (short version) at the 2003 International IRCOBI Conference on the Biomechanics of Impact, pp.349-350, Lisbon, Portugal, September 24-26, 2003.
Submitted to International Journal of Crashworthiness

Paper E
Evaluation of Head Response to Ballistic Helmet Impacts, using FEM
Magnus Aare and Svein Kleiven
Manuscript (Results presented in Proc. RTO Specialist Meeting, NATO, Koblenz, Germany, 2003).

Division of work between authors

Paper A
Injuries from Motorcycle- and Moped crashes in Sweden from 1987-1999.
Magnus Aare and Hans von Holst

M Aare was responsible for the outline of the project, the research work and the writing of the paper under the supervision of H von Holst.

Paper B
A New Laboratory Rig for Evaluating Helmets subject to Oblique Impacts.
Magnus Aare and Peter Halldin

P Halldin and M Aare developed the laboratory test rig. M Aare performed the laboratory tests and wrote the paper under the supervision of P Halldin.

Paper C
Influence of Various Parameters on Rotational Acceleration inside a Helmeted Dummy Head during Oblique Impacts - a Parametric Study using FEM
Magnus Aare and Peter Halldin

M Aare performed the numerical modelling and the simulations, the evaluation of data, and wrote the paper under the supervision of P Halldin.

Paper D
Injury Tolerances for Oblique Impact Helmet Testing.
Magnus Aare, Svein Kleiven and Peter Halldin

M Aare performed the numerical simulations together with S Kleiven. S Kleiven supplied the FE model of the human head, and M Aare developed the numerical model of the helmet. M Aare wrote the paper under the supervision of S Kleiven and P Halldin.

Paper E
Evaluation of Head Response to Ballistic Helmet Impacts, using FEM.
Magnus Aare, Svein Kleiven

M Aare performed the modelling of the helmet. S Kleiven supplied the FE model of the human head. M Aare performed the numerical simulations in close cooperation with S Kleiven. M Aare wrote the paper under the supervision of S Kleiven.

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1 Introduction

In 2000, road traffic injuries caused the death of an estimated 1,200,000 human beings, the ninth highest cause of death worldwide (United Nations, 2003a). By 2020, this figure is projected to nearly double unless something is done about it. Developing countries will bear most of the brunt of this predicted sharp rise. The injuries arising from traffic accidents in the world today cost societies enormous amounts of money. With an increase in the material standards in developing countries, a substantial increase in the number of motorcycle and moped accidents can be foreseen. However many of these accidents can be prevented. The most vulnerable part of the human body, and the most important to protect, is the central nervous system and in particular the head. Considering the expected increasing threat to the human head in traffic accidents, it is important to be able to provide relevant and adequate head protection.

The passive safety of motorcycle riders is an important issue even though motorcycle injuries represent a minor part of the total number of injuries sustained in traffic accidents. Analysing the outcome of the NASS (National Automotive Sampling System) and the GES (General Estimates System) databases published by the National Highway Traffic Safety Administration showed that in 1997, 2.1 million people were injured or killed in motor vehicle traffic crashes in the USA (Traffic Safety Facts, 1998). Some 55,000 of these were motorcyclists. However, comparing the occupant fatality and injury rates per 100 million vehicle miles travelled, it can be concluded that the risk of injury or death is more than 13 times higher travelling by motorcycle than by car, truck or bus. In Sweden, about 2000 people are injured in motorcycle accidents annually (Paper A).

The total annual rate of head injuries in Sweden over the last 14 years has been relatively constant (Kleiven et. al., 2003). Despite several national injury prevention strategies, these injuries have not decreased satisfactorily. The cost in terms of years of life lost resulting from road trauma is larger than from cancer (European Transport Safety Council, 1999). A significant number of road accidents affect the central nervous system in a devastating way by transferring high kinetic energy to the nervous tissue.

Previous studies have found that injured motorcyclists contribute to around 20% of all transport-related head injuries (Kleiven et. al., 2003). Motorcycle-related deaths contribute to around 12% of motor vehicle deaths (Sosin et. al., 1990). Head injury was the cause of more than half of these deaths. Investigations into the introduction of a compulsory helmet-wearing law for motorcycle riders has revealed findings similar to seat belt studies. The number of motorcycle accidents with severe head injuries has significantly decreased with compulsory helmet use (Sosin et. al., 1990).

A number of studies have suggested that the mean motorcycle speed is not very high in an accident, ranging from 30 to 45 km/h (Hurt H. H.Jr. et al., 1981; Otte D. et al., 1981; Whitaker J., 1980, Onser, 1983; Fuller P.M. and Snider J.N., 1987). Assuming therefore that the majority of motorcycle collisions take place at relatively low speeds, and that impacts at higher speeds cause more serious injuries, then a substantial proportion of serious injuries and fatalities occur at modest speeds where there is still some chance of providing protection. Otte et al. (1999) showed that the *oblique* impact condition, with a dominant *tangential* force to the head is the most common impact situation for the motorcyclists.

Considering the facts presented above, it is imperative to work toward providing better head protection. It should be possible to provide better head protection if one also considers the oblique impacts, which is the most commonly observed impact to the head in a motorcycle accident.

2 Objectives

The main objective of this thesis is to reduce the number of head injuries occurring in society, especially those caused by rotational force to the head.
This was pursued by focusing on the four following detailed objectives:

1. Studying epidemiological data to define the most frequent and severe types of head injury resulting from motorcycle and moped crashes in Sweden.
2. Designing a new test method for testing motorcycle helmets in a more realistic manner.
3. Improving helmet design to achieve better protection in crashes involving oblique impact to the head.
4. Proposing new injury tolerance criteria that take into consideration rotational effects transmitted to the head in a crash.

3 Anatomy of the head

3.1 The skull

The human skull consists of 22 bones, jointed together into one structural unit. 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.

3.2 The brain

The human brain is the upper and greatly expanded part of the central nervous system located inside the cranium. The brain can be divided into the cerebrum, the brainstem including the medulla, and the cerebellum (Figure 1). The *cerebrum* is the largest and most complex part of the brain. 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, and the occipital lobes. 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. The *cerebellum* is located in the posterior part of the head and includes two hemispheres.

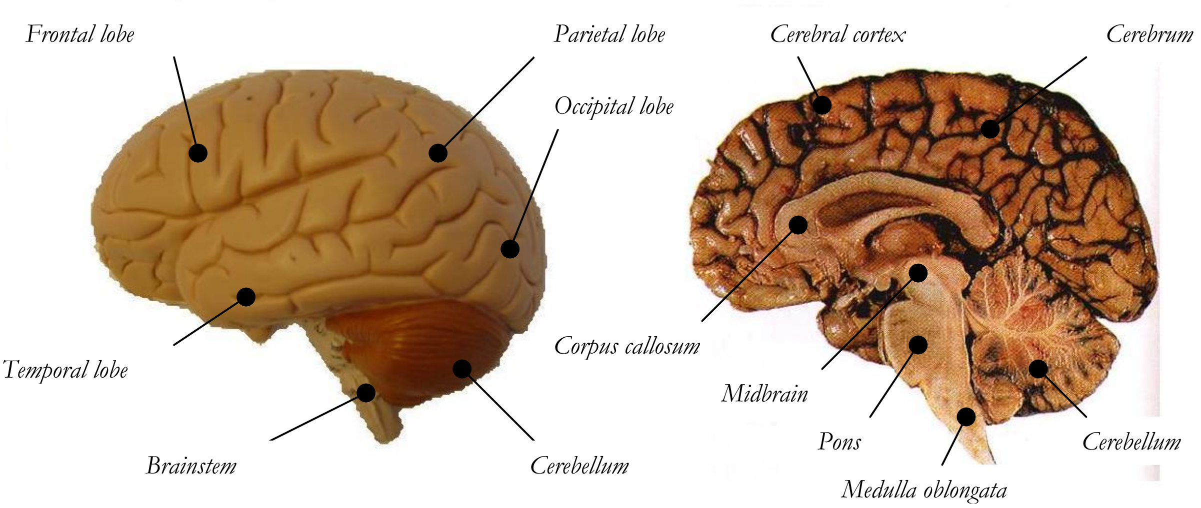


Figure 1 *The different parts of the human brain.*

4 Biomechanics of head injuries

In many accidents the human head is exposed to loads exceeding the loading capacities of its natural protection features. The consequences of severe head injuries are often fatal or at least long lasting and not fully recoverable. Falls and transport accidents are the main causes of head injuries (Kleiven et. al., 2003). Head injuries can be divided into primary and secondary injuries. The primary injuries, which are a direct consequence of the physical loading of the head, arise at the time of the accident. These injuries can result in a range of pathological changes, such as necroses, post-traumatic oedema, increased intracranial pressure, hypoxia, ischaemia, and intracranial hypertension. These changes can also give rise to subsequent injuries, called secondary injuries. These secondary injuries are related to the severity of the primary injuries and appear some minutes to days after the accident. For this reason, the efficiency of the medical intervention provided immediately after the accident is crucial for the overall outcome of the patient.

Mechanisms causing head injuries are still not fully understood. Generally, individual mechanisms of trauma produce very specific types of head injury (Youmans, 1996). Several factors must be considered in accurately determining the severity and type of injuries caused by accidents. The injury pattern depends not only on the primary mechanical damage but also on the complex interaction of the pathophysiological events that follow.

Primary head injuries are categorised into three distinct types: skull fractures, focal injuries, and diffuse brain injuries.

4.1 Skull fractures

Skull fractures do not necessarily cause neurological disability. However, when bone fragments penetrate blood vessels or brain tissue, the complications may be mild, moderate or severe. Skull fractures can be either open or closed. A closed fracture is a fracture of the bone 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.

4.2 Focal injuries

Focal injuries are characterized by macroscopic damage that is more or less limited to a local and well-defined area. Focal injuries may require surgery and approximately two thirds of the deaths associated with head injuries are attributable to these.

4.2.1 Epidural haematoma

An epidural haematoma (EDH) is the result of trauma to the skull or to the underlying meningeal vessels and is not due to injury to the actual brain (Figure 2). EDH are not as lethal as subdural haematomas.

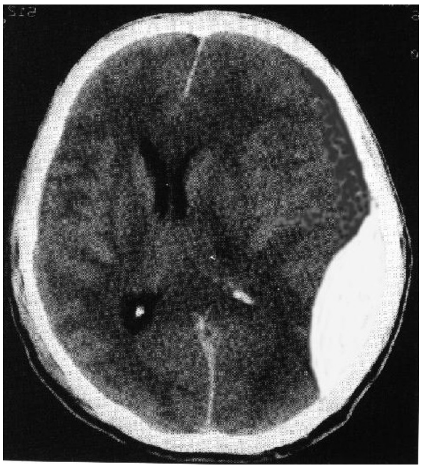


Figure 2 *Epidural haematoma, with bleeding visible on the lower right.*

4.2.2 Subdural haematoma

One of the most frequent injuries to the brain resulting from motor vehicle accidents that result in fatality or the need for long-term rehabilitation is subdural haematoma (SDH) (Gennarelli, 1983). A SDH is caused by a rupture to an artery or bridging veins (Figure 3). This injury arises from *tangential* force against the skull, and is directly related to *rotational* effects on the brain (Gennarelli, 1983). The most common type of SDH is due to disruption of 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 (Youmans, 1996).

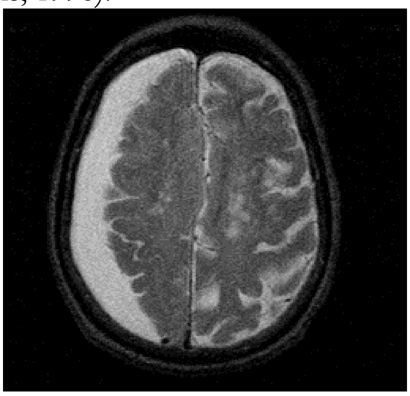


Figure 3 *Subdural haematoma with the bleeding visible on the left.*

4.2.3 Contusion

Contusion is the most frequent focal or localised brain injury. 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 appears at the site of impact while *contrecoup* appears at the opposite side in the head.

4.3 Diffuse injuries

Diffuse brain injuries are fundamentally different from focal injuries in that they are associated with widespread brain dysfunction, usually without macroscopic structural damage. Diffuse brain injuries account for approximately 40% of patients with severe brain injuries, and one third of deaths due to head injury (Whitaker J., 1980). These injuries are the most prevalent cause of persisting neurological disability in crash survivors.

Generally an impact to the head results in acceleration or deceleration of the head, which leads to inertial loading of the intracranial structures. Inertia—being the tendency for a body to maintain its initial velocity (even when it is zero)—results in the brain lagging behind the movement of the skull during acceleration of the head. This lag in motion can cause tearing of the brain tissue.

4.3.1 Concussion

Concussion is the most common head injury diagnosis resulting from motorcycle and moped accidents (Paper A). Concussion usually involves loss of consciousness. Recovery from this injury is usually good.

4.3.2 Diffuse axonal injury

Diffuse axonal injury (DAI) is a frequently occurring brain injury resulting from motor vehicle accidents, and often results in fatality or the need for long-term rehabilitation (Gennarelli, 1983). DAI is caused by the disruption of neuronal axons in the brain tissue. 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.

5 Criteria, tolerances and thresholds for head injury

This chapter presents an overview of the most commonly used criteria, tolerances and thresholds used for predicting head injury.

5.1 Head injury criterion

The most commonly used criterion for the prediction of injuries to the head is the head injury criterion (HIC) (NHTSA, 1972). The formula for calculating the HIC is expressed in equation 1 below:

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

where a is the resultant translational acceleration at the centre of gravity expressed in g; t_1 and t_2 are two points in time between which the acceleration acts, expressed in seconds. There are two expressions for HIC: the HIC15 and the HIC36. The numbers are representing the period of time between t_1 and t_2 that is used (that is either 15 or 36 ms). A HIC value of more than 1000 is considered to result in severe head injury. Hopes and Chinn (1990) indicated that there is an 8.5% probability of death at an HIC value of 1000, 31% at 2000 and 65% at 4000.

5.2 Head impact power

Newman (2000) presented a directionally-dependent injury criterion called head impact power (HIP), expressed in Equation 2 below:

$$HIP = ma_x \int a_x dt + ma_y \int a_y dt + ma_z \int a_z dt + I_{xx}\alpha_x \int \alpha_x dt + I_{yy}\alpha_y \int \alpha_y dt + I_{zz}\alpha_z \int \alpha_z dt \qquad (2)$$

where m is the mass, a is the acceleration, I is the inertia and α is the rotational acceleration. In the same study, it was proposed that coefficients should be used for the different directions to compensate for the different sensitivity. However, values of the coefficients for directional sensitivity were not presented. Head injury is assumed to correlate to the maximum HIP value.

5.3 Rotational/angular thresholds

A body has six degrees of freedom, consisting of translation in three directions and rotation around three different axes. 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. Except for skull fractures and EDH, virtually every known type of head injury can be produced by rotational force to the head. EDH is not related to head motion or acceleration, but to skull deformation. Due to the viscoelastic nature of biological tissue, the response of the tissue is dependent on the magnitude and rate of the acceleration or on the change of rotational velocity.

Several studies have been performed on animals to determine the thresholds of the rotational accelerations that might cause SDH. One of these was presented by Gennarelli and Thibault in 1982, where they performed experiments on primates (Figure 4). 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. The curves in the same figure show the proposed threshold values for SDH found by Bycroft (1973) and Hayashi T (1970).

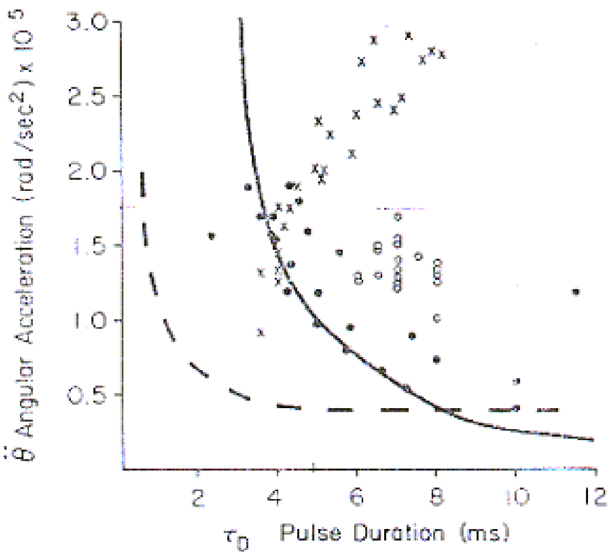


Figure 4 *Results presented by Gennarelli and Thibault (1982) where the x-axis shows the duration of the pulse in ms, the y-axis shows the rotational acceleration in rad/sec^2,* o *represents concussion,* • *represents diffuse injuries, and* × *represents acute SDH.*

Marguiles and Thibault (1992) proposed a criterion for predicting DAI based on animal (primates) tests scaled using an analytical model (Figure 5).

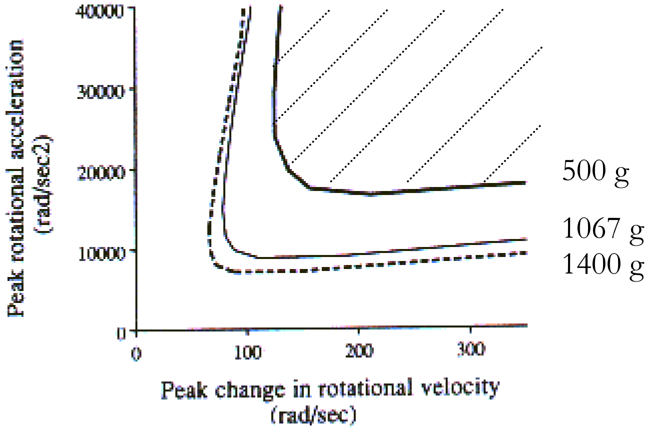


Figure 5 *Curves representing risk levels for DAI. The areas to the right and above the curves are areas of high risk and should be avoided. The different lines represent different mass of the brain. The thick line represents the threshold for a child (mass 500 g), the thin line represents the threshold for a smaller adult brain (mass 1067 g) and the dashed line represents the threshold for a larger adult brain (mass 1400 g).*

According to figure 5, a risk of DAI arises when the rotational acceleration is above 10 krad/s^2 in combination with a peak change in rotational velocity of more than 100 rad/s. Lighter brains appear to withstand higher levels of force. However this relationship cannot be assumed to hold generally, as for example the brains of children are not fully developed and are therefore more fragile than those of adults.

6 Motorcycle helmets - performance and current standards

There are many ways to design a helmet and still fulfil the legislated standards. This chapter describes some of the most important differences in helmet design, along with the most commonly used helmet standards and their differences.

6.1 Helmet design

6.1.1 Open versus full-face helmets

In recent years, the full-face helmet (integral helmet) has become more and more popular (COST 327, 1997), largely due to the belief that it provides better protection. From statistical studies, the full-face helmet appears to provide better protection to the head and face, but appears somewhat worse with respect to neck injuries. The slightly higher rate of neck injuries might be due to these helmets being somewhat heavier. There is also a slightly increased rate of fractures to the base of the skull for those wearing full-face helmets. Overall, a full-face helmet is preferable from a safety perspective.

6.1.2 Helmet shell and liner

Hopes and Chinn (1989) investigated the effect of helmet shell and liner stiffness on the ability of the helmet to protect the head. In this investigation it was concluded that a stiffer liner or shell, corresponded to a higher peak acceleration and HIC value from a given drop height. The standard helmet—in this case designed to comply with the British standard—was considered to be too stiff and too resilient. The standard helmet does not absorb energy efficiently enough in an impact where the degree of severity may be low enough to enable the wearer to survive. In other words, a less stiff or resilient helmet may result in less serious head injury for these survivors. Because of the penetration tests used in setting the helmet standards, the shell and the liner have to be designed stiffly and consequently the helmets absorb optimal energy in impacts where injury is probably unavoidable. Several other studies, such as those performed by Chang et. al. (1999), Gilchrist and Mills (1987), Kostoupulus et. al. (2002) and Yettram et. al. (1994) have shown the same results.

Many researchers in the area are convinced that some standards contribute to the construction of overly-stiff helmet shells and liners of too high a density.

6.1.3 Fibreglass versus plastic shells

Vallée et al (1984) studied the effects of fractures in the helmet shell and concluded that there is a much greater risk of injury when the helmet fractures. Fibre-based materials had a much lower rate of fracturing, whereas plastic shells fractured more often. However, in a study by Noél (1979) on the aging of materials, plastic shells showed superior performance than fibre-based materials. The rebound effect in the plastic-shell-fitted helmet is much higher than in the fibreglass-shell-fitted helmet, and is therefore less efficient. The fibreglass helmet is therefore recommended from a safety point of view.

6.2 Current helmet standards

There are a lot of different helmet standards in use around the world today. Different standards can be found for all types of helmets used in activities ranging from ice hockey and horse riding to mining and building construction. This thesis focuses on motorcycle helmets.

Depending on where a motorcycle helmet is sold, it has to comply with different standards. For example, there are different standards in place in Japan, Australia, Europe, Great Britain and North America. A helmet standard usually includes a series of tests addressing the following characteristics and features:

- shock absorption,
- penetration,
- chin strap,
- surface friction, and
- visor.

Only the shock absorption test has been addressed here.

The regulations used today have resulted in helmets that provide good protection against radial impacts. As a rule, a safety helmet consists of a hard shell, attached more-or-less well to an energy-absorbing liner. Most of the research done to increase the level of safety provided by a helmet focuses on the characteristics of the shell and liner, and attempts to achieve the maximum absorption of energy in a radial impact. Some results (Hopes & Chinn, 1989, Yettram et al. 1994; Grandel & Schaper 1984) showed that the helmets made at that time were too stiff for the impacts that their wearers are normally exposed to.

Most safety helmets currently used are optimised to meet the requirements of standard tests such as the Federal Motor Vehicle Standard No. 218 (1997). In these tests, a helmet is dropped vertically from a height of 1.5-3 m onto a flat or curved, and rigid surface that is tangential to the helmet surface. During the drop tests, the translational acceleration of a model human head is measured. In Table 1, the shock absorption requirements stipulated in some helmet standards have been presented. The parameters used in these tests for evaluating helmet impact energy absorption are: resultant acceleration, cumulative duration, drop height and shape of the anvil.

The resultant peak acceleration must not exceed 300 g or 400 g, depending on the respective standard. In the ECE (United Nations, 2003b) and in the DOT (Federal Motor Vehicle Standard No 218, 1997) standards there are additional requirements for cumulative duration. Method A in Table 1 indicates a guided, freefalling model head fitted with a helmet subject to impact with different anvils, while Method B indicates a guided freefalling model head fitted with a helmet attached to a supporting arm, subject to impact with different anvils. The model head is made of a magnesium alloy, and varies between 3 and 6 kg.

Table 1 *Shock absorption requirements in helmet standards. Four of the most commonly used standards today are the ECE 22 (United Nations, 2003b), the Snell M95-M2000 (Snell Memorial Foundation, 2003), the DOT (Federal Motor Vehicle Standard No 218, 1997), and the BS6658 (BSI, 1999). The ECE 22 is the European standard, the Snell and DOT are used in North America, and the BS6658 is the name of the British standard.*

Standard	Anvil Shape	Drop Height (m) 1st Impact	2nd Impact	Cumulative duration (g)	Peak (g)	Method
ECE R22-05-2000	flat	2.50	none	150 / 5ms	300	A
	hemisphere	none	1.83	150 / 5ms	300	A
BS 6658:1985	flat	2.87	1.43	-	300	B
	hemisphere	2.5	1.27	-	300	B
	flat	2.15	1.08	-	300	B
	hemisphere	1.83	0.94	-	300	B
Snell - M95 and M2000	flat	2.35(2)*	1.72(2)*	-	300	B
	hemisphere	2.35(2)*	1.72(2)*	-	300	B
	edge	2.35	none	-	300	B
DOT 218	hemisphere	1.38	1.38	200/2ms and 150/4ms	400	A
	flat	1.83	1.83	200/2ms and 150/4ms	400	A

(2)* - Two impacts at each site

Although oblique impacts are the most frequently seen impacts in real life accidents, they are not considered in the ECE 22, the Snell M95-M2000 (Snell Memorial Foundation, 2003) and the DOT.

7 Review of numerical simulations on helmets

This chapter describes some of those numerical simulations performed on motorcycle and ballistic helmets over the last 10 years, that are considered to be of the most importance to this thesis. All literature presented below used the finite element method (FEM) for the simulations.

7.1 Motorcycle helmets

Yettram and co-workers (1994) constructed an FE model of a motorcycle crash helmet with a simplified geometry. The energy absorbing liner used in this model had only one structural element making up its entire thickness. As a surrogate for the human head, they used a spherically-shaped wooden head. The influences of shell stiffness and liner density were studied. The performance of the helmet was judged in terms of maximum translational acceleration of the model head. They came to the conclusion that the stiffness of the shell and the density of the liner had major effects on the behaviour of the helmet. According to their results, the stiffness of the helmet shell and the density of the liner ought to be kept low in order to increase the safety performance of the helmet. The limited space available for deformation ought be used to its maximum capacity. Another interesting conclusion concerned the expanded polystyrene foam, which was that the strain rate dependency was negligible for the velocity range considered in helmet testing.

Liu and co-workers carried out crash simulations on helmets using FEM, modelling the energy absorbing liner using only one element throughout its entire thickness (Liu et.al., 1997, 1998; Liu and Fan, 1998). Their main concerns were to develop a methodology and carry out a validation process rather than performing parametric studies. The simulations that were carried out used both simple and biomechanical head models. Some of the important features of their studies were the inclusion of the neck in the simulations, and the use of more accurate human head geometry with the Hybrid III dummy and biomechanical head models.

Chang and co-researchers used a relatively detailed FE model of a helmet and a rigid head-shaped model to conclude that a stiffer shell and a denser liner were preferable in high velocity impacts, whereas a less stiffer shell and a softer liner were preferable at low velocity impacts (Chang et. al., 1999).

Subsequently, Chang and co-researchers used FEM to evaluate the protective performance of chin bars to reduce head injuries from facial impacts (Chang et.al., 2000). Their results showed that the shell stiffness was important in determining the protective effect of the chin bar. However, the chin bar consisting of simply an outer shell and comfort foam provided inadequate protection. The inclusion of an energy absorbing liner was seen as essential for increasing the protective effect of the chin bar, and it was found that the liner density ought to be denser than that used in the cranial portion of the helmet. Results from the same studies showed that combining a chin bar with an energy absorbing liner and a shell design that is less stiff would provide better protection.

Kostopoulos and co-researchers conducted a parametric study on the effects of different composite shell stiffnesses on the dynamic response of a model head fitted with a helmet (Kostopoulos et. al., 2002). Much of that work focused on the modelling the mechanical behaviour of the composite materials, including damage and delamination progression. The dynamic response of the head model was measured in terms of the maximum translational acceleration at the centre of gravity. It was shown that composite shell systems exhibiting lower shear performance provided additional absorbing mechanisms and resulted in better helmet behaviour in crashes.

All these studies have focused on purely radial impacts and translational acceleration measurements, and have not taken the important rotational effects into account. This is a great disadvantage since the rotationally-induced injuries are important. Only the studies by Liu and co-researchers used an FE model of the human head; though they did not analyse the effects in the brain tissue during an impact, which are important for improving our knowledge of the injury tolerances for the human brain.

7.2 Ballistic helmets

The response of woven composite helmet materials to ballistic impact was studied experimentally and numerically by van Hoof et al., (1999). Ballistic impacts were performed on flat panels consisting of the same material as ballistic helmets, using 1.1 g fragments. A numerical model incorporating post-failure response was implemented in the LS-DYNA3D finite element code to predict the penetration and the backplane response of composite materials under ballistic impacts. It was concluded that it is possible to model the impact response when the rate dependency in the material is taken into account.

Later, van Hoof and co-researchers continued their studies by modelling a helmet and a head (van Hoof, 2001). It was then found that the helmet interior exhibited higher deformations than previously observed in flat panels fabricated from the same material. It was also found that the impact effects were localised and that the global motion of the helmet was negligible. These simulations showed that the helmet deformation could exceed the distance between the head and the helmet, resulting in an impact of the helmet interior onto the skull.

Baumgartner and Willinger (2003) studied the rear effect caused by a ballistic projectile launched at high velocity towards the helmet of military personnel. FE models were developed of the human head including its principal anatomical components, and a model of the ballistic helmet was made from an aluminium plate subject to impact by a steel bullet. Pressure in the brain, von Mises stress in the brain tissue, the force applied to the human head as well as the global strain energy of the skull were all calculated. A linear fracture of the skull was predicted in the case of the rear effect impact configuration, whereas it appeared that the tissue tolerance limit for sustaining neurological lesions in the brain was not reached.

Though van Hoof and co-researchers constructed a very precise model of a ballistic helmet shell, they did not however have accesses to an FE model of the human head. On the other hand, Baumgartner and co-researchers used an FE model of the human head but used a very rough model of a helmet, simulating the helmet shell with an aluminium plate model. The combination of a detailed FE model of the human head and a good model of the helmet was therefore needed. None of the numerical simulations on the ballistic helmets mentioned above have taken the rotational aspects into account, which are important for predicting head injury.

8 Materials and methods

8.1 Statistics

The statistical study included here (Paper A) is a national survey of medical data on traumatic head injuries occurring in Sweden between 1987 and 1999 (Aare and von Holst, 2003). The size of the Swedish population is 8.8 million, and ranged between 8.4 and 8.8 million during this 13-year period. The statistics contained in the paper have been defined below by descriptive analysis.

Some data have been obtained from the Swedish National Road Administration database, namely where police have reported that a motorcycle or moped rider or passenger was injured or killed. The rest of the data were taken from the Swedish Hospital Discharge Register (HDR). From 1987, HDR includes all public, in-patient care in Sweden. Today, there are 93 emergency hospitals reporting to HDR. Information for the HDR is sent once a year to the Epidemiological Center (EpC) at the National Board for Health and Welfare in Sweden, from each of the 26 county councils in Sweden. There are four different types of information contained in the HDR: data on the patient, data on the hospital, administrative data, and medical data. Patient data include personal identification number (PIN), gender, age and place of residence. Administrative data include dates of admission and discharge, and length of stay. Medical data include diagnosis, E-code, surgery, medical treatment, and decease/discharge outcome.

Data from the HDR used in this study collected between 1987 and 1996 were based on the ninth revision of the International Classification of Diagnosis (ICD 9), and from 1997 to 1999 on the 10^{th} revision (ICD 10). However, a few counties still report in ICD 9.

8.2 Experiments

The laboratory experiments (Paper B) consist of a number of different helmet impact testing procedures. The basic objective of the test rig described here was to introduce tangential force into helmet testing. An oblique force is the resultant of both radial and tangential force components. The tangential component induces rotational acceleration in the test head. Figure 6 presents a photographical overview of the test rig in which a helmeted dummy head can be dropped onto a moving plate. The basic idea behind this configuration was first presented by Harrison et. al. (1996). Speeds of up to 10 m/s (36 km/h) can be achieved in both horizontal and vertical directions. This means that impact speeds of up to 14 m/s (50 km/h) can be reached. Sensors are mounted on the rails to measure the velocity of the plate, both before and after helmet impact. Sensors are also mounted on the pillars to measure the vertical speed of the falling frame.

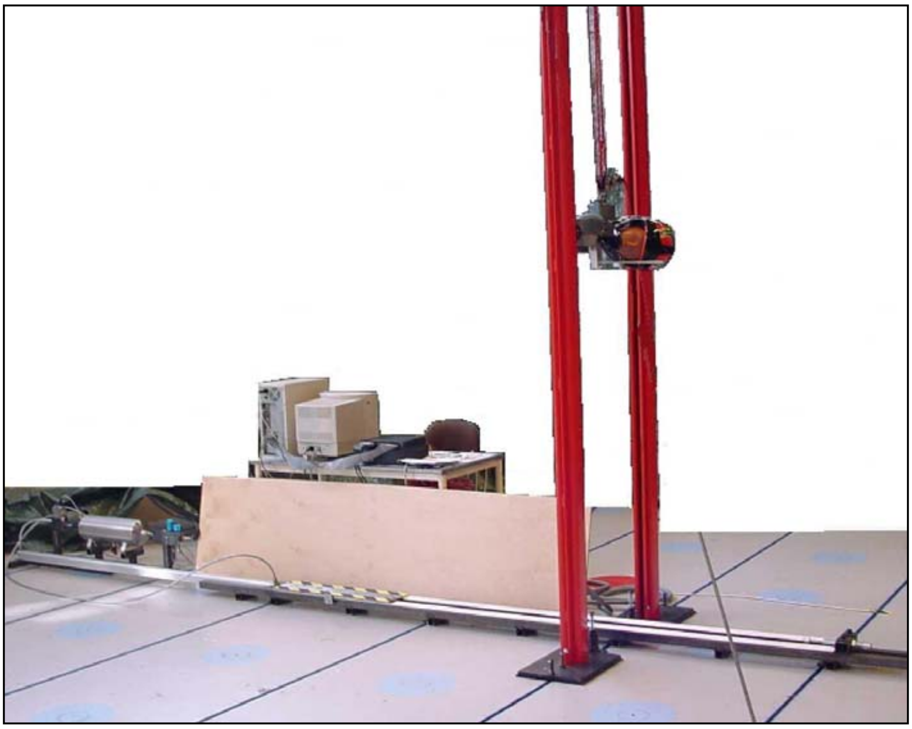


Figure 6 *Overview of the test rig.*

A helmeted Hybrid III dummy head was fitted with sensors to measure accelerations in translation as well as rotation during impacts. All the data were collected in one computer, which also manoeuvred the system. Two different types of helmets were tested in a number of impact scenarios. In helmet Type 1, the shell was glued to the energy absorbing liner. This is how most of the conventional helmets are manufactured. In helmet Type 2, the shell was separated from the liner by a low friction layer. This configuration enables the shell to move about 1 cm relative to the liner when a force is applied to the outer shell. The helmets were sized to fit the dummy head perfectly. An artificial scalp was also introduced to simulate the movement of the human scalp relative to the cranium. This artificial scalp was constructed from two rubber bathing caps with a section of medicine ball sandwiched between them. A thin layer of oil was introduced between the inner cap and the section of medicine ball in order to reduce friction. The original artificial skin on the dummy head was removed before attaching the artificial scalp to the dummy head.

8.3 Numerical simulations

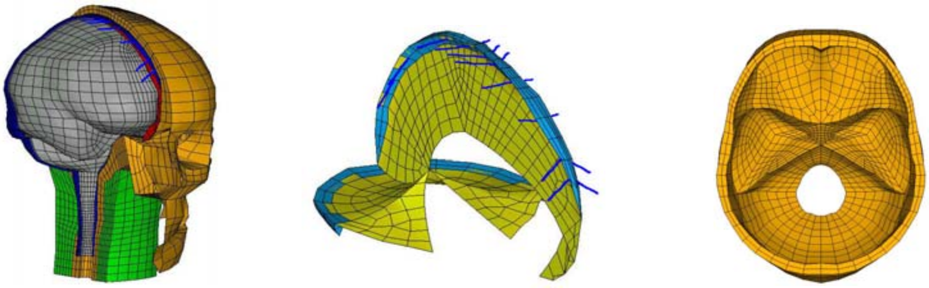
For the numerical analysis (Papers C, D and E) the finite element method (FEM) was used, implementing the non-linear, explicit and dynamic code LS-DYNA3D (Hallquist, 1998).

In this thesis, the following four finite element models were used:

- the human head (Kleiven, 2003),
- a Hybrid III dummy head (Fredriksson, 1996),
- an experimental motorcycle helmet, and
- a ballistic helmet.

8.3.1 The FE model of the human head

The FE model of the human head (Figure 7) developed at the Royal Institute of Technology (KTH) Stockholm, Sweden, is based on data from the Visual Human Database (Visible Human Project, 2003). The FE model is anatomically detailed and includes the scalp, the skull, the brain, the meninges, cerebrospinal fluid (CSF) and eleven pairs of parasagittal bridging veins (Figure 7b) (Kleiven and von Holst 2001, 2002a, 2003).



(a) (b) (c)

Figure 7 *Finite element mesh of: (a) the human head, (b) falx and tentorium including transverse and superior sagittal sinuses with bridging veins, and (c) skull bone.*

8.3.2 The FE model of the Hybrid III dummy head

The dummy head used was a detailed model of a Hybrid III crash test dummy head (Fredriksson, 1996), (Figure 8). The FE model of the Hybrid III dummy head is very similar to the real Hybrid III dummy, with respect to geometry, weight, inertia, and material properties. A Hybrid III dummy head consists of an aluminium skull and a rubber skin. The aluminium skull model was rigid and the rubber skin was modelled as a viscoelastic material.

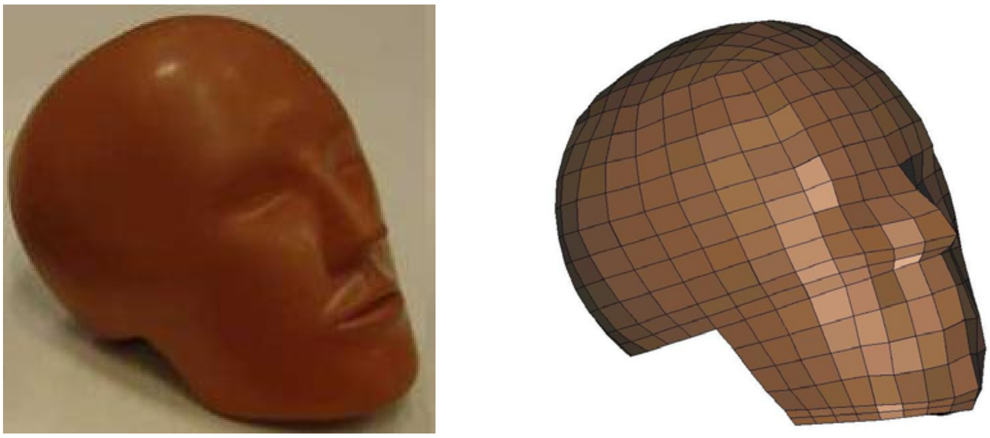


Figure 8 *The real Hybrid III dummy head (left) and the FE model of the Hybrid III dummy head (right).*

8.3.3 The FE model of the experimental motorcycle helmet

The geometry of the helmet model was similar to the one used by Aare and Halldin (2003), (Figure 9). This helmet was open-faced with a plastic shell (ABS thermoplastic), and was modelled using an elastic material model. Young's modulus was determined from tensile testing. The helmet liner was made of an energy-absorbing expanded polystyrene (EPS) foam (40 kg/m^3), modelled using a crushable foam material model (Hallquist, 1998). The material characteristics were determined using hydrostatic compression tests. There were no comfort foam inserts placed inside of the helmet. The complete model of the helmet was validated against both radial and oblique helmet impact tests (Aare and Halldin, 2003). This model had no chinstrap, as it was not deemed necessary considering the short duration of the impacts.

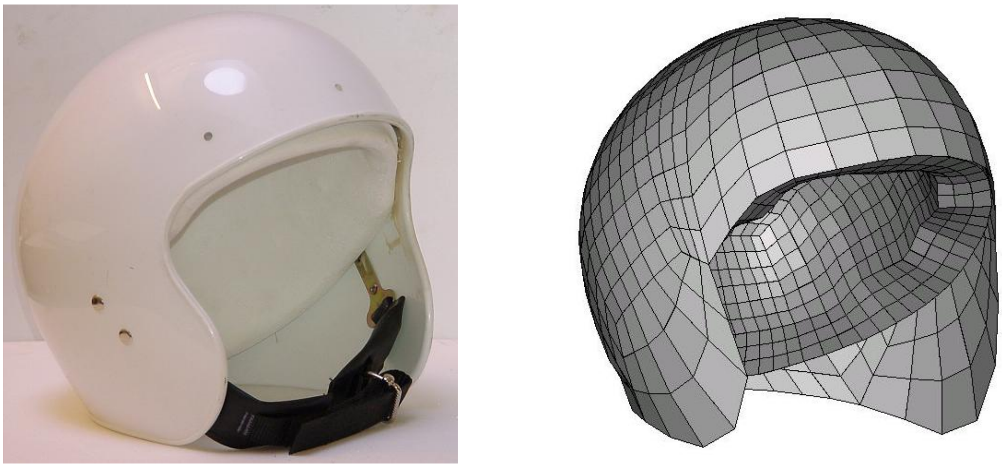


Figure 9 *The experimental helmet (left) and the FE model of the experimental helmet (right).*

8.3.4 The FE model of the ballistic helmet

An FE model of a ballistic helmet was made with a similar geometry to the PASGT, NATO standard (Figure 10). The protection level was good for stopping an 8 g, 9.0-calibre bullet, with a velocity of 360 m/s. The investigation focused on trying to obtain a realistic response similar to that of the human head during impact. This means that the weight of the different parts and the interaction between them were essential. The material properties of the woven-fabric-reinforced aramid laminates were taken from the literature (van Hoof et al., 1999, 2001). The material used for the helmet shell was modelled using a Chang-Chang Composite Failure model (Hallquist, 1998). The shape of the deformed helmet shell agreed well with ballistics tests. Cracking and penetration of the helmet shell was not modelled, as this only happens to the outer fibre layers and has little influence on the overall behaviour. The delamination of the composite shell was not modelled either. The total weight of the helmet in this model was 1.5 kg, which is comparable to most composite helmets on the market. The helmet shell model was validated against ballistics tests performed at Beschussamt Mellrichstadt. In these tests, the helmet was cleared from all interior fittings and the shell itself was attached to a fixture. To measure the dynamic deflection of the inside of the helmet shell, a piece of clay was applied on the fixture at the spot where the bullet was expected to hit. The deflection of the helmet shell varied from 23–35 mm.

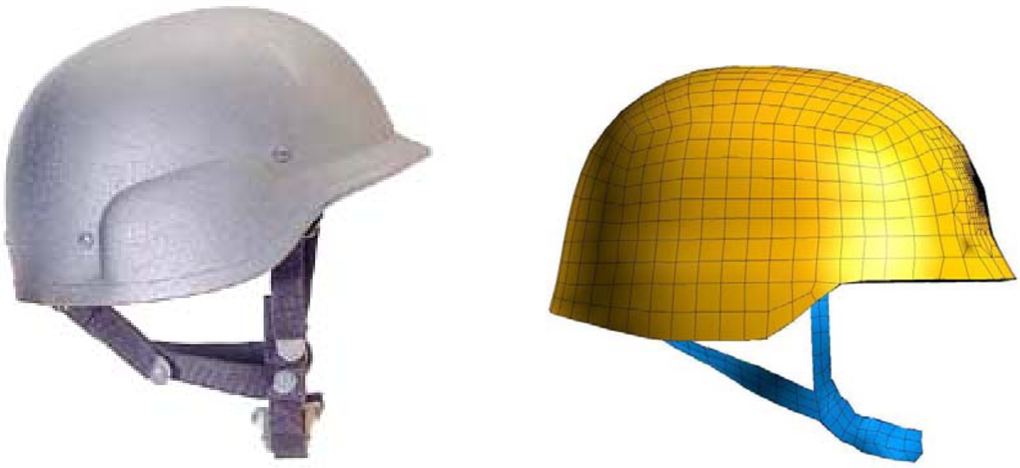


Figure 10 *A real ballistic helmet (left) and an FE model of a ballistic helmet (right).*

9 Results and discussion

Head injury is the most common severe injury resulting from motorcycle accidents (Otte et. al., 1999), (Paper A). In Sweden, 23% of all motorcycle-related injuries are head injuries. By far the most common type of head injury is the intracranial injury, occurring in 85% of all head injuries sustained in motorcycle and moped accidents. The annual costs of in-patient care relating to motorcycle and moped accidents in Sweden, excluding rehabilitation is approximately SEK 18 million (Paper A).

The incidence of motorcycle and moped crashes in Sweden has declined over the last few years, though the numbers of injuries are still unacceptably high. Most of these accidents involve young people—especially males—between 15 and 25 years of age, accounting for about 60% of all those injured (Paper A). While fractures to the lower limbs are a frequent occurrence in motorcycle and moped crashes, head injuries are by far the most important. The vast majority of these head injuries are intracranial. Concussion is the most common injury, followed by focal and diffuse lesions, while intracranial haematoma was less prominent. Head injuries are responsible for a substantial economic burden on the public health system. In addition, the human suffering that head injury cause the victims, their relatives, and society cannot be calculated in simply economic terms. Thus the current national data in Sweden shows that improvement in existing injury prevention strategies is clearly necessary.

Although legislation making helmet use obligatory prevents many head injuries, there is an important need for better tests in the development of improved helmet design. In this thesis, a new way of testing helmets, and a new helmet concept was studied (Paper B). This study investigated different interfaces between the shell and the liner in an experimental simplified helmet, at impacts with a dominant tangential component. A conventional helmet—generalized here with the shell fixed to the liner—was compared to a helmet with the shell and liner separated by a low-friction interface made of Teflon. The experimental results showed that the separated helmet had a substantial potential to reduce the energy transmitted to the head 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, and the third is between the helmet and the human head. Helmets are already very smooth on the outside for reducing the friction between the impacted object and the helmet. 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.

An artificial scalp was also introduced to better resemble the human head in an oblique impact situation (Paper B). However reconstructing the test scalp identically before each test session can be problematic. If a mechanical scalp is going to serve as a standard in the future, it is crucial that it can be easily validated to ensure identical characteristics for each test session. It is also essential to ensure that the head is fixed to the helmet in the same way during all tests. When it comes to measuring rotations, good head-helmet fit is also crucial.

Helmets manufactured today are optimised for different test procedures, according to the standard in the country where they will be sold. Helmet manufacturers claim that there are too many different standards currently in use around the world. In fact, if these standards are to be optimal for human safety, they ought to be the same. Furthermore, many researchers around the world are even suggesting that some of the tests are actually irrelevant and should be replaced by more realistic tests. The most important of some of the new suggestions is the oblique impact test. In this test the helmet is exposed to a more realistic impact situation. There have been attempts to introduce this new impact test, but the problem to date has been how to obtain accurate repeatability. In order to be able to determine whether one helmet is better than another when it comes to absorbing rotational energies, test results from two identical helmets should not differ too greatly. The results from the purely radial tests in today's standards vary less than 7% for two identical helmets. This might seem a bit high, but it is impossible to cope with all the uncertainties involved (Paper C). Much of the efforts devoted to future research should aim at trying to minimize the sources of errors in oblique impact testing.

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. In the struggle to find such tolerances, a large number of numerical simulations on both an FE model of a dummy head and an FE model of the human head were performed in the preparation of this thesis (Paper D). The maximal principal strain in the brain of the FE human head model was used as a measurement to predict brain injury. The assumption that the strain in the brain tissue is directly coupled to DAI or to brain injury is open to discussion. There are also other tissue-level measurements that can be used as injury predictors, such as strain rate, the product of strain and strain rate, von-Mises stress, and strain energy. However in this study, strain has been chosen to enable analysis of the risk for DAI, as this measurement has been verified experimentally (Bain et al, 1996; Bain and Meaney, 2000). An FE model needs to be well correlated to relevant experimental studies on the human brain. The FE model used in this study has been carefully validated and shown to correlate well with other experiments in the literature looking at how rotational injuries correlate to strain in the brain, as well as relative motion experiments (Hardy et al., 2001; King et al., 2002).

For rotational impulses of short duration, the change in angular velocity has been shown to correspond best with the intracranial strains found in the FE model (Kleiven and von Holst, 2003), which is in agreement with Holbourn's hypothesis (1943). However, it was seen in Paper C that the peak change of angular velocity is strongly coupled to the rotational acceleration for helmet impacts. For translational impulses on the other hand, the HIC and the HIP (Newman et al, 2000) have shown the best correlations with the strain levels found in the model (Kleiven and von Holst, 2003).

The fact that the head only is dealt with throughout this thesis might be seen as a limitation. However this is also the case in all known helmet tests to date. A more realistic simulation would probably involve the whole body, or at least the head, neck and torso. This would change the dynamics of the impact, as the boundary conditions for the head would change. Ruan et. al. (1991) showed that a single hinge coupling between the head and the neck had a limited effect on the intracranial pressure during impacts of short duration. Beusenberg et. al. (2001) simulated the influence of different neck models using data from head impacts recorded from the National Football League in the US. In that study, it was shown that in oblique impacts to the front of the head, the coupling between the head and the neck is important only for sagital plane rotation. Hering and Derler (2000) performed both radial and oblique helmet impact tests on both a detached and a complete Hybrid III dummy. It could be concluded from their study, that the influence of the neck and body on the rotational effects seen in the head was different for different impact locations. They also considered as a problem the fact that the Hybrid III neck is much stiffer that the human neck. The most interesting and important part of the impact with respect to injury levels is the interval of time spanning the first few milliseconds. During this short period of time, the influence of the human neck is limited. One other reason why the neck and torso are not used in this thesis, and in fact in any standards today, is the difficulty experienced when trying to standardize and repeat such a test. The lack of good surrogates for the neck and torso is also an important issue. For example, the Hybrid III dummy neck is designed for frontal collisions in cars at speeds well in excess of the speeds in helmet impact testing. It is however suggested that the influence of the neck should be investigated in future studies.

The impacts simulated and performed experimentally throughout this thesis were not only chosen to cover the most commonly observed impacts in motorcycle accidents, but also to cover rotations around all three axes.

In order to improve our knowledge of other impact injuries to the head, which could help in the designing of better crash helmets, this thesis also discusses a more unusual type of head impact, namely ballistic impact to a protected head (Paper E). Armed forces and police face ever increasing threats of serious injury as the access to weapons amongst criminals increases. The need for head protection in policing situations involving exposure to weapons is obvious. In the study involving ballistic helmets presented in this thesis, a parametric study was performed to find the optimum ballistic helmet shell stiffness. Effort was also put into investigating the effects on the human head when the impacting projectile hits the helmet from different angles. From the parametric study on ballistic helmet impacts, it could be seen that the deflection of the shell had a substantial influence on the load levels to the human head. It should be remembered that this is just a parametric study conducted to compare different helmet shells and different impact angles. There is little knowledge about short duration pulses in general, and about short duration rotational accelerations pulses in particular. The rotational accelerations do not exceed known thresholds, but that is not a guarantee that these are not dangerous. In these impacts, there are combinations of translational and rotational accelerations, which makes injuries even harder to predict.

10 Conclusions

It has been concluded that injuries to the head are the most common injury in motorcycle and moped accidents requiring in-patient treatment. Injuries to the head are by far the most severe, requiring the longest rehabilitation periods. They are therefore the most expensive to society. Many of these injuries are caused by rotational forces to the head.

A new type of helmet testing equipment has been designed, introducing a tangential force resulting in a rotational motion of the head. A new concept for developing and constructing helmets with better rotational energy absorption has also been proposed.

In the efforts to predict injuries in non-radial helmet impacts, new injury tolerances for three specific oblique impacts have been proposed.

Our knowledge in the field of ballistic head injuries has been improved. It was concluded that the most crucial task when designing a ballistic helmet is to produce a stiff enough shell. The results also predicted that projectiles impacting on the helmet in an oblique angle could be more injurious than radial impacts.

11 Future work

Measurements of the rotational energy absorption ought to be included in helmet standard testing. The results of this thesis present concrete progress in that direction. However, some difficulties remain and much of the effort devoted to future research should aim at trying to minimize the sources of errors in the oblique impact test. Future work should be aimed at trying to make the test as repetitive and robust as possible, and facilitate standardization.

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