Improved helmet design and test methods to reduce rotational induced brain injuries

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ABSTRACT

Accidental impacts to the human head are often a combination of translational and rotational accelerations. The most frequent severe brain injuries from accidents are diffuse axonal injury (DAI) and subdural hematoma that both are reported to arise from rotational violence to the head. Most helmet standards used today do only take the translational accelerations into account. It is therefore suggested that an oblique impact test that measures both translational and rotational accelerations should be a complement to the helmet standards used today. This study investigates the potential to reduce the risk for DAI by improving the helmet design by use of an oblique helmet impact test rig. The method used is a detailed finite element (FE) model of the human head. The FE model is used to measure the maximum principal strain in the brain which is suggested as a measurement for the risk to get DAI. The results clearly show the importance of testing a helmet in oblique impacts. Comparing a pure vertical impact with a 45 degree oblique impact with the same initial impact energy shows that the strain in the central parts of the brain is increased with a factor of 6. It is therefore suggested that a future helmet impact standard should include a rotational component so that the helmet is designed for both radial and tangential forces. Such a test method, an oblique impact test, was used to compare two different helmet designs. One helmet was manufactured with the shell glued to the liner and one helmet was designed with a low friction layer between the shell and the liner (MIPS). It was shown that the strain in the FE model of the human head was reduced be 27% comparing the MIPS helmet to the glued helmet design.

Introduction

This paper is mainly focused on motorcycle safety. However, much of the results and discussion are directly applicable to helmets used for other activities and situations where a helmet is used as the primary head protection device.

Passive safety of motorcycle drivers is an important topic even if motorcycle injuries represent a minor part of the total number of injuries in automotive accidents. Analyzing the outcome from the NARS and the GES databases published by the National Highway Traffic Safety Administration showed that in 1997, 2.1 million people where injured or killed in motor vehicle traffic crashes in USA (Traffic safety facts [1998]). 55000 of these were motorcyclists. However, comparing the Occupant Fatality and Injury Rates per 100 Million Vehicle Miles Traveled, it can be concluded that it is more than 13 times higher risk to travel on a motorcycle than in a car, a truck or a bus.

Head injuries are the most common severe injuries from motorcycle accidents [Doyle and Sturrock 1996, Otte et al 1999, Aare et al. 2003], and the main protective device used to prevent these injuries is the safety helmet. Most safety helmets currently used are designed to meet the requirements from standard tests [ECE22.05, FMVSS218, BS6658]. In these tests, the helmet is typically dropped vertically onto a flat or curved rigid surface that is tangential to the helmet surface. During the drop tests, the translational acceleration of a head form is measured.

According to statistics there is no typical motorcycle accident. However, Otte et al. [1999] showed that the oblique impact condition, with a dominant tangential force to the head, is more common than a pure radial impact. The oblique impact condition will produce linear (translational) and angular accelerations to the head. There is no standard used today that measures the rotational acceleration in an oblique impact. The only helmet standard that includes an oblique impact is the British Standard BS6658, but there is no measurement of the rotational acceleration why it is difficult to correlate to injuries to the human brain. The oblique impact tests in BS 6658 ensure;

- that projecting visor mounts etc, shear off easily when there is an impact with a series of parallel bars; and
- that the tangential force on the helmet shell, when it impacts a rough flat surface, is no larger than that for typical shell materials used in 1985 (the year of introduction of the test).

This has resulted in helmets with good protection for radial impacts while their protection for oblique impacts is unknown. At an oblique impact to a helmeted head the rotational energy can be absorbed by friction energy between the helmet shell and the impacting surface, internal energy in the liner and comfort foam due to shear deformation, friction energy and relative motion between the liner and the head.

At the division of Neuronic Engineering (KTH) in Stockholm, Sweden and the School of Metallurgy in Birmingham UK, there is an ongoing project that tackles the helmet safety with focus on oblique impacts. This project has until now investigated:

1. A new oblique helmet impact test [Halldin et al, 2001 and Aare and Halldin 2003].
2. A new safety helmet design [Halldin et al, 2001 and Aare and Halldin 2003].

The new oblique helmet impact test rig shown in Figure 1, is a new version of the test rig presented in Halldin et al. 2001. A Hybrid III dummy head is fixed in a helmet, which is placed on a frame. The frame is attached to two pillars and can travel almost without friction in the vertical direction. The helmet will strike a plate, which is moving horizontally on two PTFE-covered rails. The plate is accelerated by a pneumatic cylinder. Inside the head, a system of nine accelerometers was mounted. With this method it is possible to measure both linear accelerations in all directions and rotational accelerations around all axis. A more detailed description of the test rig is given in Aare and Halldin 2003.

It was believed that helmets could be improved by separating the shell and the liner by a low-friction layer. This idea mimics one of the safety systems found in nature, the human head. When the head is subjected to tangential impact or rotational violence, the brain has possibility to move/slide relative to the skull in the cerebro spinal fluid and thereby reducing the impact energy. This system has been copied to the helmet named MIPS (Multi-directional Impact Protection System) where the outer shell has possibility to move relative to the liner, Figure 2. The MIPS system ads an energy absorbing layer compared to the conventional helmet design, where the shell normally is glued to the liner.

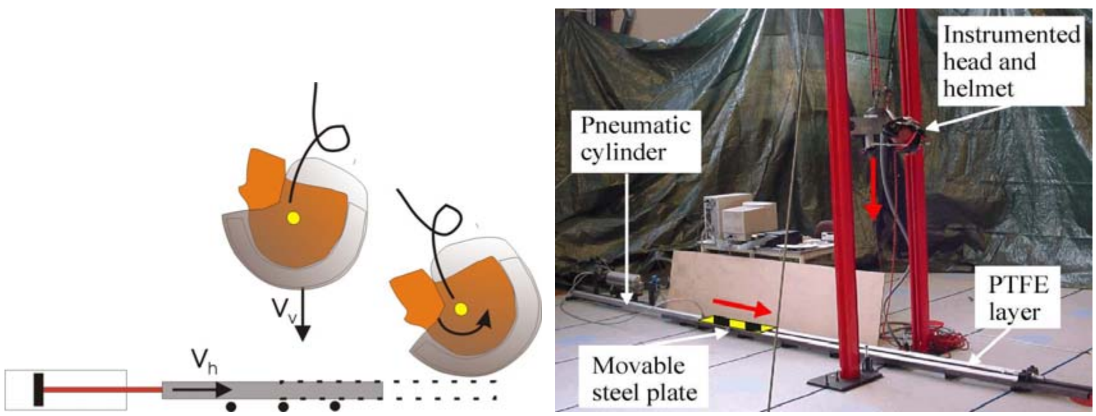


Figure 1. The oblique impact test.

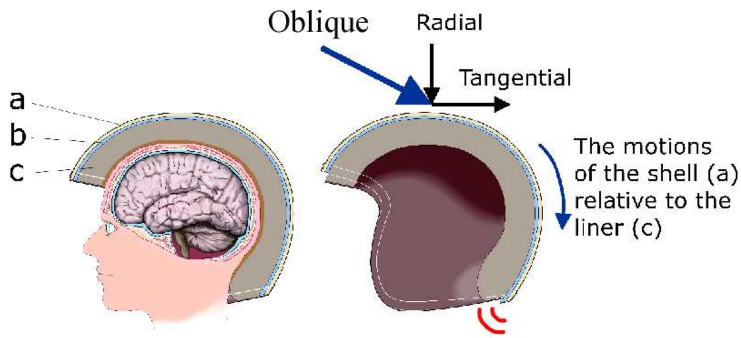


Figure 2. Schematic picture of the MIPS helmet, made of a hard shell (a) separated from the liner (c) by a low friction layer (b). At an oblique impact the shell can move relative to the liner.

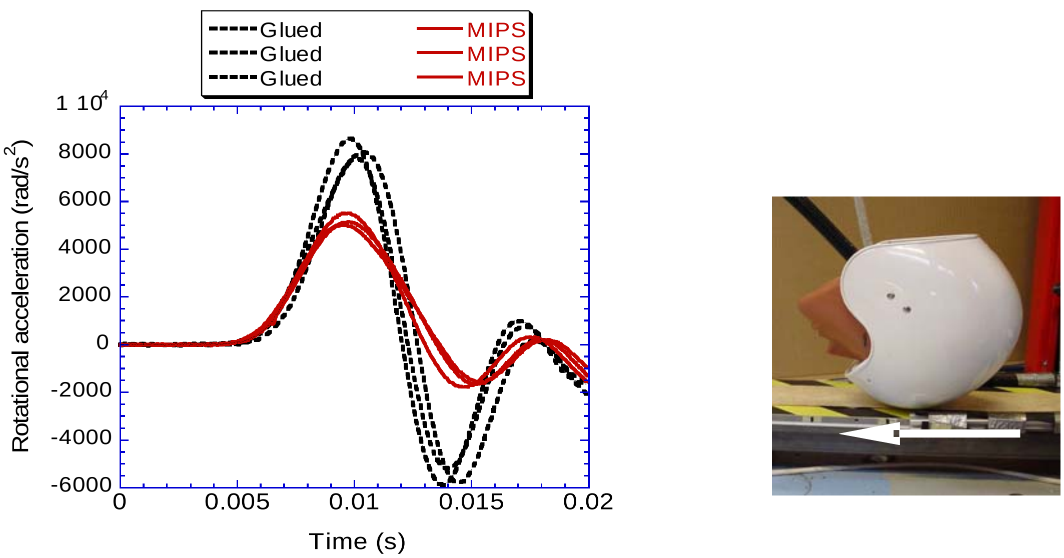


Figure 3 The rotational acceleration as function of time for two different helmet designs. The impact direction to the helmeted head is shown to the right. The arrow shows the direction of the plate.

By use of the oblique impact test it was shown, that the MIPS design reduces the rotational acceleration and the change of rotational velocity by 40% and 25%, respectively [Halldin et al. 2001], Figure 3. The research on how to test helmets in oblique impacts has been continued at Royal Institute of Technology Aare and Halldin [2003]. The results from that study also show that it is possible to improve the helmet protective properties in oblique impacts without changing the properties in vertical impacts. It as therefore suggested that the oblique impact test should be included in a future helmet safety standard. However, there are several questions and demands that need to be answered before such a standard test can be proposed and accepted:

1. The test must be robust and the variation between two similar tests must be controlled.
2. The injury thresholds for an oblique test must be proposed and accepted.

The first demand have been partly solved and discussed in Aare and Halldin 2003. The problem with an oblique test compared to a pure vertical test is that the fixation of the head form in the helmet is much more critical. It was concluded that it is possible to control the variation between similar test by a method where the rubber skin of the head form is pressurized against the comfort foam or the liner. The second demand is probably the most difficult part to solve. The injury thresholds or injury criteria used in an oblique helmet impact test should predict injuries frequently seen in traffic accidents. The most frequent brain injuries from motor vehicle accidents that result in fatality or the need for long-term rehabilitation are Subdural Hematoma (SDH) and Diffuse Axonal Injuries (DAI) (Gennarelli et al. [1983]). SDH is caused by rupture of an artery or a bridging vein. DAI is caused by the disruption of neuronal axons in the brain tissue. It has been shown that there are correlations between these injuries and the rotational acceleration and the change of rotational velocity of the brain (Melvin et al. [1993], Gennarelli et al. [1983]). The proposed onset for DAI is an angular acceleration of 10 krad/s^2 in combination with a change of the rotational velocity of 100 rad/s [Marguiles and Tibault 1992]. These thresholds are proposed pure angular motions. A set of thresholds used for a combination of angular and translational kinematics need to be corrected and the thresholds for the rotational acceleration and change of rotational velocity do probably need to be decreased as shown in the study by DiMasi et al. [1995].

One method that can be used as a complement to biological experiments on human cadavers in the struggle to find thresholds or criteria for an oblique impact to the helmeted head is the use of a detailed finite element model of the human head. It has been suggested that there is a correlation between the strain in the brain and DAI [Bain and Meaney 2000]. If the finite element model of the human head is well correlated to relevant experiments the strain computed in the model can be compared to accepted tissue base injury thresholds. There is a wide spectrum of proposed thresholds in the literature for the strain (5-50%) in brain tissue that has potential to cause DAI. Bain and Meaney [2000] proposed a threshold of 21% strain to cause DAI.

The objective of this paper is to investigate how the risk for rotational brain injuries like diffuse axonal injuries can be reduced by improving the helmet design and helmet test procedures.

Method

In order to investigate the possibilities and limitations with an oblique impact test method and the potential to improve the safety helmet design a new design scheme shown in Figure 4 has been used. The experimental oblique impact test of the helmeted head was modeled in a finite element (FE) program (LSDYNA). The FE model of the oblique impact test including a model of the Hybrid III dummy head and a model of the helmet was used to investigate which parameters that was of most importance to achieve a robust and repetitive experimental result. In this study a FE model of the human head has been used to investigate the impact response in the brain as function of impact impulse. Pure vertical impact was compared to a 45 degree oblique impact, with the same initial impact energy. The model was also used to compare two different helmet designs.

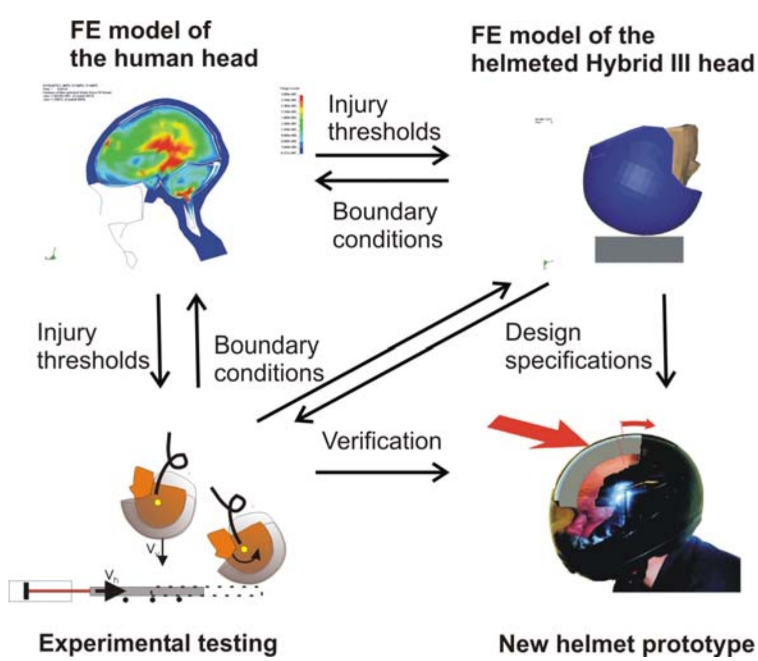


Figure 4 Design scheme used to improve a safety helmet for oblique impacts.

The Experimental Helmets

Two different types of open faced helmets were tested in Aare and Halldin [2003]. The two types of helmets had the same shell (ABS thermoplastic) and the same liner (expanded polystyrene, EPS, 40 kg/m3). These were the exact same type as Halldin (2001) used. In helmet type number one, called the GLUED helmet, the shell was glued to the liner. This is how most conventional helmets are manufactured. In helmet type number two, called the MIPS helmet (Multidirectional Impact Protection System), the shell was separated from the liner by a low friction layer (figure 2). This low friction layer was in this case Teflon, attached to the outside of the liner. This enables the shell to move relative to the liner. This motion can reach up to a maximum of two centimeter, when a tangential or oblique force is applied to the outer shell. In all these helmets, no comfort foam was used, which gives a high coefficient of friction (approximately 0.4) between the head and the liner.

FE model of the human head

The FE model of the human head was developed at the Department of Aeronautics is based on data from the Visual Human Database. Thus, the FE model is anatomically detailed. The model includes the scalp, the skull, the brain, the meninges, the cerebrospinal fluid (CSF) and eleven pairs of parasagittal bridging veins, Figure 6, Kleiven [2000], Kleiven and von Holst [2002a, 2002b]. This model has been experimentally validated against pressure data in a previous study (Kleiven and von Holst, 2002a) as well as relative motion magnitude data (Kleiven and von Holst, 2002b). Also, a comprehensive correlation between the FE model output and the relative motion between human cadaver brain and skull in anatomical X, Y, and Z components has been demonstrated for three impact directions (Kleiven and Hardy, 2002). The model correlates well to experiments by Nahum et al. (1977), Hardy et al. (2001), and King et al. (2002). The model has been validated with experiments performed using acceleration impulses of magnitudes and durations close to the ones in the present study.

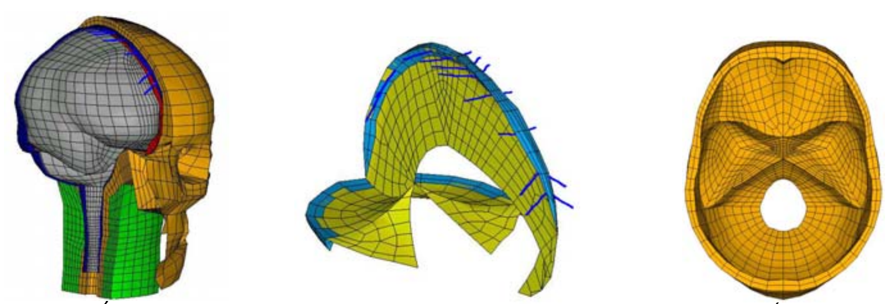


Figure 5: Finite element mesh of: a) The human head; b) falx and tentorium including transverse and superior sagittal sinuses with bridging veins; c) Skull bone.

Numerical simulation of an oblique impact to the human head

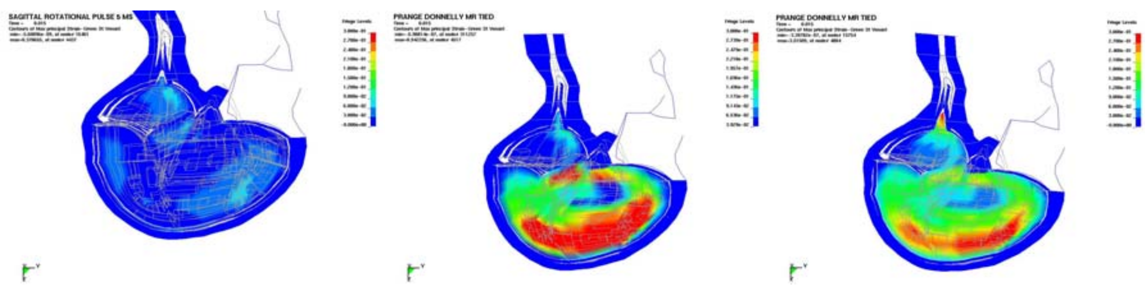
Tissue-level threshold for DAI is a function of distortional strain [Bain and Meaney 2000]. The numerical model of the human head was used to investigate the risk for DAI at different impacts. The skull of the human FE model was modeled as a rigid body and the boundary conditions from the experimental tests presented in Figure 3 were applied directly to the skull of the FE model. The boundary conditions include the translational acceleration in the X and Z direction and the rotational acceleration around the Y-axes. The coordinate system is defined with its origin in the centre of gravity for the head. The X-, Y- and Z-axis are defined in frontal, lateral and vertical directions, respectively. Three different sets of boundary conditions were investigated as presented in Table 1. The risk for DAI was analyzed by measuring the maximum principal strain in the central parts of the brain including the thalamus and the corpus calosum.

Table 1. The different impact impulses applied to the FE model of the human head.

	Helmet	Impact speed (m/s)	Impact angle (degree)	Z-acc (m/s^2)	X-acc (m/s^2)	Y-rotacc ($krad/s^2$)
Pure vertical impact	Glued	7	90	250	0	0
Oblique	Glued	7	45	175	45	9
Oblique	MIPS	7	45	175	38	5

Results

Figure 6 shows the strain in the xy-plane 5mm offset the sagittal plane. Figure 7 shows the strain as function of time for the different impact situations. The strain was measured in the element in the central part of the brain that showed the highest strain. Comparing the numerical results from the vertical impact and the oblique impact shows that the vertical impact gives 83% lower strain in the central parts of the brain. Comparing the results for the Glued helmet with the MIPS helmet shows that the MIPS helmet reduces the strain by 27% in an oblique impact.



Vertical impact (Glued helmet) Oblique Impact (Glued helmet) Oblique Impact (MIPS helmet)

Figure 6. The pictures show the strain in the xy-plane 5mm offset the sagittal plane. The red colors show the regions where the strain is close to the injury thresholds for DAI.

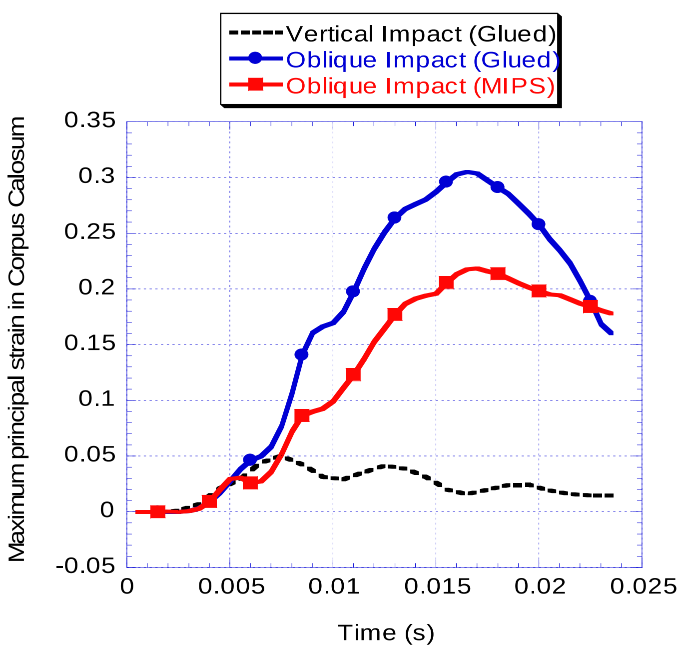


Figure 7. Shows the maximum principal strain in the central parts of the brain (Corpus calosum and thalamus).

Discussion

This paper has presented a method used to investigate an oblique helmet impact test and tried to answer the question if there is a need for an oblique impact test as a complement to the test standards used today? This study has tackled the question by use of a FE model of the human head. A FE model of the human head was used to compute the maximum principal strain in the brain and thereby analyze the risk for DAI. There are also other tissue level measurements that can be potential injury predictors like the strain rate, the strain*strain rate, the Von-Mises stress or the strain energy. In this study the strain has been chosen to analyze the risk for DAI, as this measurement has been experimentally verified [Bain et al. 1996, Bain and Meaney 2002].

A limitation to such a method is that the FE model needs to be well correlated to relevant experimental studies on the human brain. However, the model used in this study has gone through a long validation procedure and has shown good correlation to experiments found in the literature. Of special interest when looking at rotational injuries correlated to the strain in the brain is to compare the model to relative motion experiments like the one published by Hardy et al. 1995 and King et al. 2001. The model used in this paper has shown good correlation to these kinds of experiments. Another limitation to this study is that the loading is applied to the rigid skull. A more realistic method would be to include a FE model of the helmet and thereby introduce the forces to the head in a more realistic way.

It was shown that the strain in the brain at an oblique impact is significantly higher than that for a pure vertical impact, with the same impact energy. The oblique impact will result in rotational motions of the head and brain that have been shown to correlate to SDH and DAI. It is therefore our belief that the helmet test regulations should be complemented with an oblique impact.

This study also compares two different helmet designs. The major reason for this was to show that an oblique test can be an effective tool in the helmet design process. It has earlier been shown in experimental oblique helmet tests that it is possible to reduce the rotational acceleration and the change of rotational velocity with a new helmet design [Halldin et al. 2001]. The impact pulse from the experimental tests of the Glued and the MIPS helmet was applied to the FE model of the head. The result shows a reduction of maximum principal strain in the head of 27%. It was seen that the reduction of strain in the brain model is in the same order as the reduction of the change of rotational velocity (25%) but lower than the reduction of rotational acceleration (40%). There have been suggestions that the strain in the central parts of the brain do not correlate to the rotational acceleration for impact pulses with short duration [Holbourn 1947]. This study is too small to draw any conclusions, but the results indicate that the hypothesis by Holbourn is correct. However, there is a need to investigate this further in a study with a wider spectrum of impact energies and different impact directions.

There are as mentioned above specific difficulties related to an oblique impact test compared to a pure vertical impact. One problem that needs special attention is how the head form should be fixated in the helmet in order to measure the helmets performance and not how well the helmet is fixated on the head form. A solution to this problem is to pressurize the head against the helmet as discussed in Aare and Halldin [2003]. Another problem that needs special attention is to propose the injury thresholds under which helmet should pass. There is a need to better understand the injury mechanisms for rotational induced brain injuries. There are single parameter based thresholds like rotational accelerations or the change of rotational velocity. The proposed onset for DAI (100 krad/s2 and 100 rad/s) by Marguiles and Thibault could be a start value. However, these values need to be reduced when adding the translational acceleration to the impact pulse, [DiMasi et al. 1995]. It is also likely that the thresholds will need to be different for different impact directions. One potential injury criteria that takes the impact direction into account is the Head Impact Power (HIP) [Newman et al. 2000]. Regardless if the single parameter threshold or a criteria like the HIP criteria is chosen more studies need to be done in order to propose a helmet test criteria. One way to answer these questions are to use a FE model of the human head as a complement to experimental tests on human and dummy heads.

Acknowledgement

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