Final report of Working Group 3:
Impact Engineering

A COST Action TU1101 / HOPE collaboration

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COST is supported by the EU Framework Programme Horizon 2020

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1 Citation Information for this Report

Recommended citation of this report: Willinger R, Halldin P, Bogerd CP, Deck C, Fahlstedt M (2015). Final report of Working Group 3: Impact Engineering. COST Action TU1101 / HOPE, Brussels, Belgium.

Uniform Resource Locator (URL): http://publications.tno.nl/publication/34618465/RwBmO5
ISBN: 978-90-5986-468-9
Publishing date: September 2015

COST Action TU1101 / HOPE internet resources:

- www.bicycle-helmets.eu
- www.cost.eu/domains_actions/tud/Actions/TU1101

This report is summarized in the final report of COST Action TU1101: Bogerd CP, Annaheim S, Halldin P, Houtenbos M, Otte D, Shinar D, Walker I & Willinger R (2015). Helmet Optimisation in Europe: The final report of COST Action TU1101. COST Action TU1101 / HOPE, Brussels, Belgium.

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Acknowledgements

The authors are grateful to Niqui Seuntjens (Apestaart) for her expertise and support regarding the graphical aspects. This report is based upon work from COST Action TU1101, supported by COST (European Cooperation in Science and Technology).

The authors would like to thank Länsförsäkringarnas Cooperation Research fund for funding parts of this study, as well as Fondation MAIF (France) and French project ANR"SUFEHM13" for their support.

2 Introduction

The Working Group 3 within COST TU1101 action was aimed to integrate biomechanics investigation in the context of bicycle helmet optimisation in terms of head protection. The initial plans were to better understand the impact kinematics for bicyclists via real world accident simulation and to develop an advanced helmet test methods which includes realistic head impact conditions and biomechanical based pass fail criteria. A final task was to propose improvements of head protection by investigating new material and design.

The work has been very focused on the question on how to design a better test method for bike helmets. This work should be based on real accident situations. The reason why this is essential is that the current test methods for bicycle helmets are not based on real accident situations. The current test methods for bike helmets include a linear shock absorption test where the helmet is dropped vertically to a flat surface. This COST action identified this as a problem as earlier accident reports suggested that an angled impact is more common than a pure radial impact to the ground (Verschueren 2009, Bourdet et al. 2013).

If angled impacts are more common the impact force to the ground will be a combination of a normal force and a tangential force between the ground and the helmet. The tangential force is due to the coefficient of friction which for the bicycle helmet is relative high when falling to the road. A tangential force if high enough could cause the helmet and the head to rotate. It has in earlier studies been shown that a rotational motion to the head could cause both concussion and more severe brain injuries as subdural hematoma and diffuse axonal injury (Holborn 1943, Genarelli et al. 1982, 1983, Deck et al 2007 and Kleiven 2007). It is therefore believed that helmets shoudl be tested for angled impacts and to measure the rotational kinematics transfered through the helmet to the head.

Further focus was also on the pass fail criteria as currently no head injury criteria exist for complex head impact configurations. Based on advanced head FE modelling and a number of real world head trauma simulations, a first attempt of model based head injury criteria as well as its implementation into a novel helmet test method was evaluated.

Due to the initial project proposal for TU1101, WG3 has delivered what could have been expected. Then at the beginning of the project we were a bit too enthusiastic making a too ambitious project plan. WG3 planned to improve helmets by addressing new material and design. That objective was not achieved.

The present report first focusses on the kinematic analysis of bicyclists in case of real world accident. A review of accident analysis is reported and three accident cases are exposed in details. The synthesis of this accident analysis is introduced in chapter 3 where new helmet impact conditions are addressed in terms of initial velocity vector and in terms of head boundary conditions. This chapter end out with a proposal of a new helmet test conditions including tangential tests. Chapter 4 presents an extensive review of existing head injury criteria, based both on global kinetic parameters and FE head modelling. Also a synthesis of existing injury criteria evaluation is reported before concluding with a proposal for pass fail criteria. In the very last chapters a synthesis of a new helmet test method proposal is presented followed by a section on further research.

3 Bicycle accident reconstruction

3.1 Review of accident conditions

This review focused on the typical impact situation for a bicyclist. There are not many detailed reconstructions published that gives a clear view of a typical impact situation including the impact speed and the impact angle.

One can of course say that a bike accident can result in a million ways. The detailed reconstructions reports that included detailed information on the impact speed and impact angle for the head in a bicycle accident was Verschueren 2009 and Bourdet et al. 2013.

Bourdet et al. 2013, have reconstructed 13 accidents from GIDAS database plus an additional 11 cases from the French Accident Database (EDA, Études d'Détaillées Accident). A total of 24 accidents studied, where all was a collision with a car. The results from this study showed that the average speed at impact with the car was 6.8m/s and the impact angle was about 58 degrees. The same research group has also carried out an extensive parameter study (MADYMO) for single-vehicle accidents in which the different initial positions of the cyclist analyzed together with other parameters such as speed and the cause of the accident (Bourdet et al. 2012).

In another study of Verschueren (2009) presented reconstruction of 22 pieces of accidents. In this study, the simulation program MADYMO, Figure 1. Verschueren found that the average velocity was between 6.0-7.7m/s and the impact angle was 40-50 degrees. The variation of the speed and angle of impact can be derived that the accident either a single vehicle accident in which the person fell in the slope or the person collided with a car, see Figure 3 and table 1.

The impact speed and impact velocity is summarized in Table 1.

The number of cases are low why it is difficult to draw a statistical based conclusion on the typical impact speed and angle.

Richter et al. 2007 presented data from the GIDAS database including 4264 accidents. Most of these accidents are accidents with a car. The mean impact speed was 6.4m/s which is close to the cases presented by Bourdet and Verschueren.

Reference	Study	Number of cases/simulations	Accident type	Speed (m/s)	Angle α (Degrees)	Surface
Verschueren 2009	Accident reconstruction	11	Single	7.7	40	Road
Verschueren 2009	Accident reconstruction	11	Car	6.0	50	Car
Bourdet et al. 2013	Accident reconstruction	24	Car	6,8	60	Car
Bourdet et al. 2012	MADYMO Paramtric study (5.5m/s)	612	Single	6.7	55	Road
Bourdet et al. 2012	MADYMO Paramtric study (11.0m/s)	612	Single	10.2	33	Road
Ricter et al. 2007	Real accident data (GIDAS)	4264	Car/Single	6.4	NA	Car or road

Table 1: Impact speed and angle from detailed accident reconstruction studies.

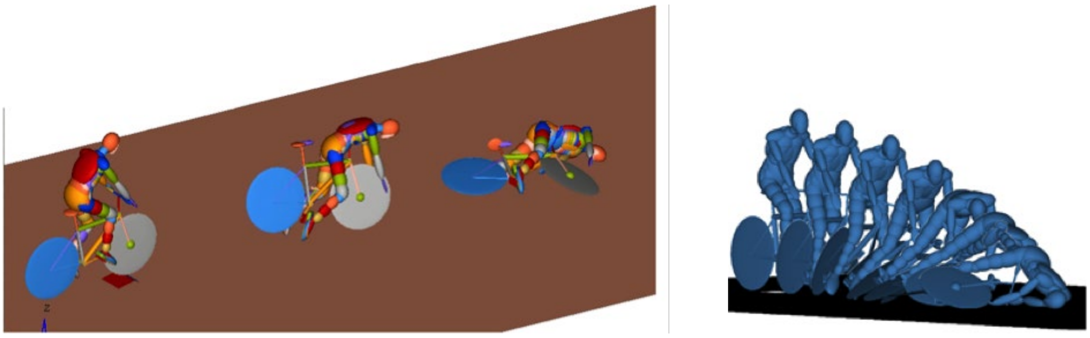


Figure 1: Reconstruction of a bicycle accident using the simulation program MADYMO

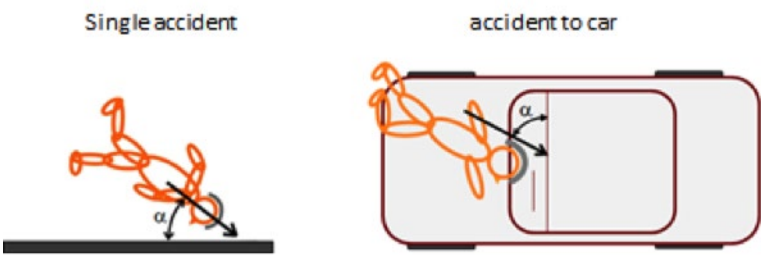
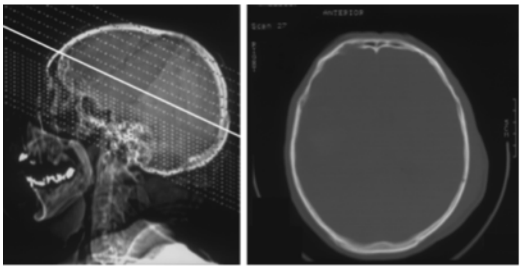


Figure 2: Schematic picture of a single accident (left) and a car accident (right).

3.2 Simulation of real world accident cases

The bicycle accident data used was collected by a research group at the KU Leuven in Belgium (Depreitere et al., 2004). They selected nineteen of the eighty-six collected accidents based on the availability of information about the accident that could be used in the accident reconstruction process with the multibody software MADYMO (Verschueren, 2009). Ten of the reconstructions were single accidents where only three cases had both medical images available and an indication of impact location on the head by scalp swelling. These three cases were therefore used in the study by Fahlstedt et al (2015a) to perform accident reconstructions with a detailed FE head model. None of the three cyclists wore a helmet. Therefore, also these three cases were used to study the protective effect of bicycle helmets.

The FE head model was positioned so that the initial impact point on the head corresponded to the maximum swelling of the scalp seen in the medical images. The initial velocity just before impact was taken from the MADYMO simulations performed by Vershueren (2009). An example of one of the cases is shown in Figure 3. In all three cases, the head impact was against concrete ground. Therefore, the ground was modeled with a rigid surface with a friction coefficient between the ground and scalp of 0.5.

a 

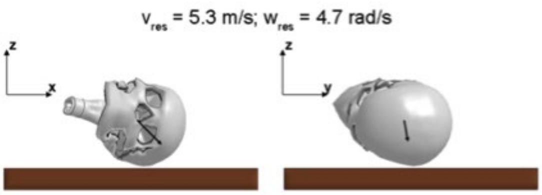
b 

Figure 3: Case 4; a) The swelling of the scalp b) the initial impact position and velocity. (Fahlstedt, 2015)

In all three cases the victim sustained skull fractures and brain injuries. More detailed information about the cases can be found in Fahlstedt et al. (2015). The reconstruction for these three accidents show that the FE model for the human head show high strains in the region of the documented injury in the medical images.

The three accident reconstructions of the head impact with and without a helmet showed substantial reductions of the brain tissue strain by between 33% and 43% (Figure 4) when including the helmet. Also an even larger reduction, from fracture level, was seen for the skull bone when including the helmet which suggest that the skull fracture could be avoided with a helmet. The strain level in Case 4 and Case 58 is close to the threshold for concussion (Kleiven 2007; Patton et al. 2013) which implies that the head injury severity is decreased or even not sustaining any head injury in the helmet case. In Case 15 the maximum strain was reduced 33% but still rather high in the when a helmet included. This accident was a severe accident in rather high velocity. The victim sustained skull fractures, large hemorrhagic contusion, intracerebral hematoma, subarachnoid hematoma, acute subdural hematoma and diffuse axonal injury. However, the brain tissue strain was decreased by 33% and the stress of the cortical skull bone went from fracture level of 80 MPa down to 15 MPa. Therefore, it is believed that the victim would not have sustained a skull fracture and the most severe brain injuries could have been avoided.

Also, the linear and angular acceleration as well as the angular velocity decreased when wearing the helmet. This is in line with previously published studies with rigid body simulations of bicycle accidents (McNally and Whitehead, 2014) and an experimental bicycle helmet study (McIntosh et al, 2013).

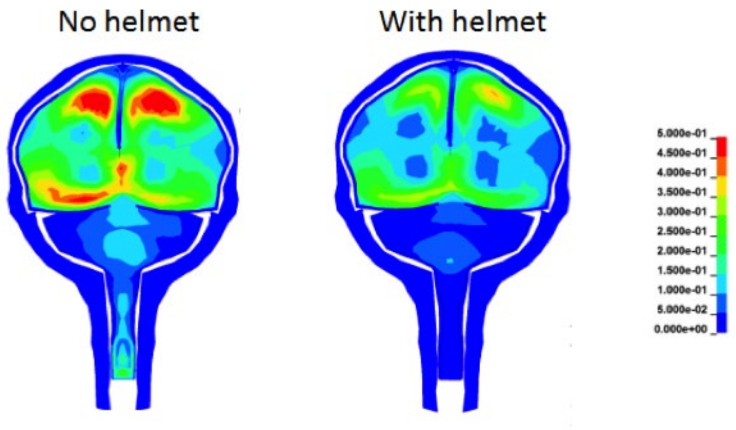


Figure 4: The 1st principal Green-Lagrange strain pattern for the brain for one of the cases.

3.3 CONCLUSION

The most common impact angle and impact angles in real accidents is difficult to evaluate exactly. Table 1 summarizes what is known today. 46 real accidents have been reconstructed and the impact angles are far from the pure radial 90 degree impact situation as in EN1078 and EN1080.

WG3 propose to keep the shock absorption test condition as defined in EN1078 and EN1080 but to complement this test with an angled (oblique) test using an impact angle of 45degrees at a speed of 6.5 m/s. These values should be seen as a proposal from the data that exists today (Table 1) and exposed in the literature review.

4 New helmet impact conditions

4.1 Test method design

This section will focus on the complementary angled impact as mentioned in Section 2.2

There are many ways to design a test method for an oblique/angled impact as shown in Fig. 6. At present, there is an existing test method as presented in (UN ECE reg. 22-05, Methods A) for motorcycle helmets. Test method A is designed to measure the tangential force between the helmet and the impacting plate, angled 15 degrees. The idea of dropping the helmet at an angle is tempting, as it is simple, with just one part moving the helmeted head. The simplicity of measuring the tangential and the normal force in the plate is interesting because it is a much less expensive alternative to having a number of accelerometers and/or rotational transducers. However, it has not been shown that the tangential force in the plate can measure the energy absorption in the helmet in the same way that accelerometers in the head form can. A possible improvement of the test used in ECE 22-05 would be to use a different head form and to install accelerometers or a combination of translational accelerometers and rotational transducers. Deck et al. 2012 presented a proposal for a new test method for bike helmets in which the helmet would be dropped onto a 45 degree angle. Deck proposed that the Hybrid III dummy head should be used eventually connected to the HIII neck. The advantages of the Hybrid III head compared to the ISO headform are that this headform has a much more realistic rotational inertia as shown in Table 2. The HIII head form can be instrumented with a 9-accelerometer-array as proposed by Padgoankar et al. 1975 or using three angular rate sensors plus a set of three linear accelerometers. It should also be mentioned that the HIII headform exists in different sizes as reported in Table 3. It is believed that two to three new HIII head sizes are needed to cope with all helmet sizes.

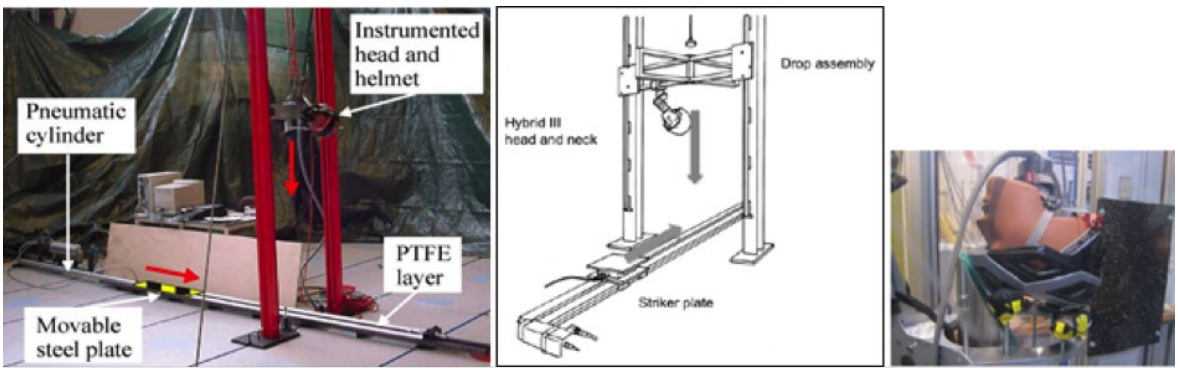


Figure 5: Examples of oblique helmet testing methods. a) Aare et al 2003 b) Pang et al. 2011 and c) photo of angled impact surface as proposed by Finan et al. 2008 and Deck et al 2012.

	Mass [kg]	I_{xx} [kg.m²]	I_{yy} [kg.m²]	I_{zz} [kg.m²]
ISO Pedestrian	4.5	11.10^{-3}	11.10^{-3}	$110.5.10^{-3}$
Hybrid III 50th	4.5	$17.088.10^{-3}$	$18.872.10^{-3}$	$22.685.10^{-3}$
Human Head	4.5	$17.996.10^{-3}$	$18.360.10^{-3}$	$21.902.10^{-3}$
ISO Helmet	5.7	Not controlled		

Table 2: Synthesis of headform inertial properties, and comparison with human head characteristics.

EN 960 headform size	Headcircumference [mm]	Dummy model	Head circumference [mm]
A	500	HIII 3 Year Old	508
B	510		
C	520	HIII 6 Year Old	520.7
D	530		
E	540	HIII 5% Female	538.5
F	550		
G	560		
J	570		
K	580	Hybrid III 95% Large Male	584
L	590		
M	600	HIII 50t% Male	597
N	610		
O	620		
P	630		
Q	640		

Table 3: Existing sized for Hybrid III headform compared to ISO EN 960 headform.

One benefit of a test method using a vertical drop onto an angled surface is that it can be installed in most test institutes with only minor changes, since the existing drop towers can be used.

Another method is to drop the helmet against a plate that is accelerated to a controlled speed, Halldin et al. 2001, Mills et al. 2008 and Pang et al. 2008. The main difference between impacting the movable plate and dropping the helmet onto an angled surface is the difference in the gravitation vector in relation to the normal force vector against the helmet. This could result in different outcomes for the two methods, even when testing identical helmets with the same impact speed and impact angle. The movable plate could therefore be more realistic in simulating a fall from a bike or a horse to the ground. However, the movable plate has drawbacks compared to the angled surface since it is more complex and because it can be difficult to maintain a constant speed of the plate during the impact.

A third potential test method is the NOCSAE (2006) pneumatic linear impactor, which was first developed by Binokinetics in Canada. The linear impactor is equipped with a curved plastic surface attached to a disc made of vinyl nitrile foam, to mimic a helmet-to-helmet hit (designed for

American football or ice hockey helmets). In this test, the head form is attached to a HIII neck and a sled moving horizontally. The test method specifies different impact locations on the helmet, all of which result in impacts to the centre of gravity in the dummy head (NOCSAE 2006). Rousseau et al. (2011) have proposed a modification of the test method by hitting the helmet at directions that are offset from the centre of gravity of the head in order to simulate real impacts as seen in football and ice hockey games. This test is however not considered as the impact result in little tangential force and is not very realistic for a bike accident.

Thus, two methods are identified to introduce tangential force to the helmet and to measure the energy absorption in the head. The method with the vertical drop to an angled impact surface is proposed due to simplicity and robustness.

Several tests using the versions of the vertical drop against an angled impact surface have been performed in different test labs in Europe (KTH (Stockholm, Sweden), UNISTRA (Strasbourg, France), OXYLANE (Lille, France) and SP (Swedish governmental test institute, Borås, Sweden). One example is seen in Figure 6 and Figure 7 where a test machine from CADEX has been rebuilt. The changes are:

- the impact surface (45 degree impact angle)
- the helmet basked to fixate the helmet during the vertical drop
- HIII head form equipped with a 9-accelerometer-array

The example shown here is from a benchmark test of 17 bicycle helmets performed at SP, Sweden (Folksam 2015). The test shown here is a test to the front of the helmet resulting in a rotation of the helmet and the head form around the Y-axis (ear-to-ear axis).

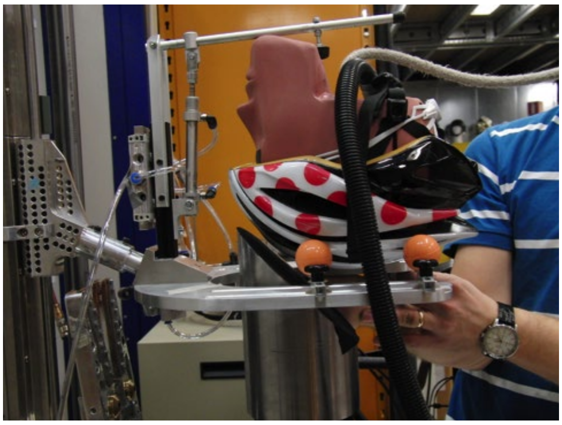


Figure 6: Showing the test rig at SP, Sweden. The impact shown here is a frontal impact resulting in a rotation around the Y-axis.

Figure 7: Shows pictures from the high speed camera.

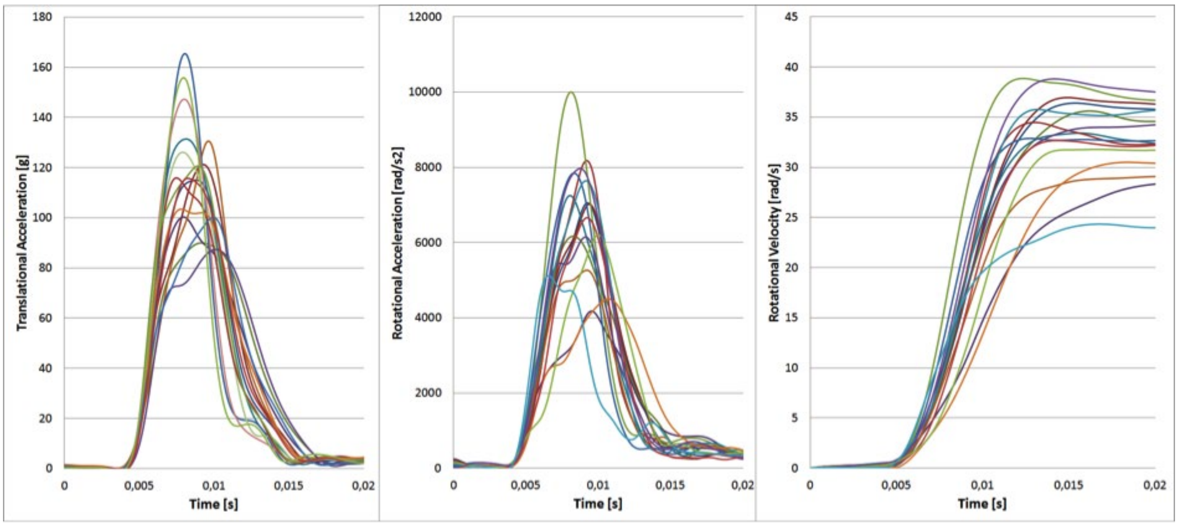


Figure 8: Benchmark test of 17 bicycle helmets from the Swedish market. The Figure shows the translational acceleration, rotational acceleration and the rotational velocity as function of time from a 6m/s impact and an impact angle of 45 degrees.

The results seen in Figure 8 show a G-level between 87-165G, Angular acceleration between 4200-10000rad/s2 and angular velocity between 24-39rad/s. All impacts where controlled with a high speed camera to make sure that the initial position of the helmet and head form were as specified. All helmets are certified and pass the EN1078 standard. Jet, the helmets differ a lot when tested in angled impacts. This test shows that there is a great potential for helmet manufacturers to improve the energy absorption if such a test is used in the design process.

4.2 Boundary condition for the head

In current test methods, the head either falls unrestrained onto the impact surface (European test standards) or is constrained to a monorail by means of a rigid arm attached to the head (US test standards). This can be said to represent the two extremes of the scale. Between these extremes is the normal situation, in which the head is constrained by the human neck. In order to design an oblique test method, questions remain as to whether the neck will affect the measured translational and angular accelerations in the dummy head. It is clear that the head is restrained by the neck and that it will, at some time, rotate around a point in the neck, or even lower down in the thoracic region. Earlier studies like the COST 327 study, have shown that the amplitude of the angular acceleration is affected by the neck COST 327 (2001). Helmeted full body Hybrid III dummies were dropped onto an angled surface and compared to free-falling helmeted head forms. The results showed that the angular acceleration differed in amplitude by about 20%. Beusenberg et al (2001) presented a numerical study on helmet-to-helmet impacts simulating an American football accident. It was concluded that the neck did indeed change the characteristics of the angular acceleration comparing impacts with and without a neck. In the study by Bausenberg, however, the impacts were close to a radial impact to the helmet, where the neck is the only cause for the rotation of the head, i.e. there was no or little tangential component in the impact. Ghajari et al. 2012 showed that the angular acceleration components could differ by as much as 40% comparing a helmet impact with the full body and the head only. In this study, Ghajari used the

THUMS finite element model and simulated an oblique impact to the lateral (temporal) portion of the helmet. Ghajari proposed changing the inertial properties of the head in order to compensate for the neck and the body if using only the head in an oblique impact test.

Forero 2009 reconstructed 12 jockey accidents using MADYMO. Two of these were studied in detail in simulations with and without the body in a helmet-to-racetrack turf impact. The angular acceleration was increased from 6462rad/s2 to 10104rad/s2 in one case and from 5141rad/s2 to 6444rad/s2 in the second case, comparing the simulation with a complete body and a simulation with the head only. Forero also mentioned that absence of the neck and the body might cause the direction of the acceleration to be altered. This study stated that the MADYMO human body model provides an unrealistic representation of the flexibility in the vertebral joint that could have resulted in this large discrepancy.

Verschueren et al. 2009 performed a reconstruction of 22 bike accidents using MADYMO. Nine of the accidents were simulated both with the head only and with the entire body. The results of this study showed that the correlation of the angular acceleration between the head-only simulation and the simulation with the complete body was good for four out of nine reconstructions. The correlation was defined as medium for three, while two out of nine were defined as poor, with a difference of about 30% for one of those examples defined as poor. Forero discussed the duration of impact pulse, pointing out that it is different in a jockey accident against racetrack turf (8-20ms) compared to bike accidents against a hard road (5-10ms). Therefore, if a test is to be designed with a surface mimicking racetrack turf for jockey helmets, a neck might prove necessary.

The conclusion that can be drawn here is that, in general, the neck affects the motion of the head. It can also be argued that a test method could be defined with impact angles where the effect of the neck is small during the short time (5-10ms) during which the helmet comes into contact with the impacting surface. Experimental tests on human cadavers show that the upper part of the human neck is flexible and could be seen as decoupled from the head for a certain amount of displacement or rotation. Motion in a human joint that does not result in a force or moment is defined as the neutral zone. The neutral zone in which the upper part of the neck allows the head to rotate without extensive load is in the range of 10 degrees, depending on the axis of rotation (Ivancic 2014, Camacho et al. 2007). Thus, when we do not take muscle activity into account, the head can rotate around 10 degrees without having any effect on the kinematics in horizontal loading of the head. Looking at an example in Appendix B, the free-falling head rotated 10 degrees during the first 10ms of impact. Based on this, one could argue that there would not be sufficient time for the neck to significantly effect the head in this specific impact direction. Neither Ivancic or Camacho et al. did however analyse a helmet impact situation with a vertical compression force to the neck.

In order to define the importance of the neck in a typical helmet to ground impact situation the partners in this COST action performed a study presented in detail in Appendix A. The conclusion was that the neck affect the kinematics of the head, but that it is dependent on the impact point and direction. The result showed in one of the studies also that helmet simulations with the HIII neck is less human like than simulations without the neck.

It can be seen that there are a multitude of issues relating to the use of the neck as the boundary condition for the head. Other aspects of the neck/no neck questions that would need to be taken into consideration when designing a new test method are:

- Assuming that the human neck does not affect the head during the first 10ms in most impact situations, is the result the same when the musculature is tensed to a theoretical maximum contraction?
- The HIII dummy neck (NHTSA) is designed and validated only for frontal car collisions at speeds of around 11m/s resulting in a flexion motion of the neck. Thus, the HIII dummy neck is not validated for compression loading, lateral bending or rotation around the vertical axis - as shown by Myers et al. 1989 and Disentis 1991.
- There are additional disadvantages to the use of a neck, like the cost involved and the need for calibration. However, the positive aspects include the fact that a neck could make the positioning of the helmet easier, as the neck keeps the head in position.

The conclusion, taken all aspects known today into account, is to propose a test method without the neck.

The other boundary condition that needs to be taken into account is the way the helmet is fastened to the head. McIntosh et al. [32] presented among other results in a study how hard the helmet was tightened by the chin straps to the head form, during oblique impact tests of bike helmets. There was no clear difference in the measured rotational components when tightening chin. The spread in the data was however lower when the helmet was tight compared to less tight.

Mills and Gilchrist [12] performed oblique tests on bicycle helmets using a HIII head equipped with an acrylic wig to mimic the hair and scalp. Aare and Halldin [1] also performed tests using an artificial scalp. These tests showed that artificial hair or scalp models did affect the angular acceleration measured. An experimental study was therefore performed in which equestrian helmets were tested in the test lab described by Aare and Halldin [1]. The test is described in full in Appendix B. The result showed that in comparison to having the HIII head form covered with stocking, testing helmets with a wig resulted in a reduction of the angular acceleration of 17% and a reduction in angular velocity of 4%. Thus, it could be argued that if the oblique test is performed using a headform covered with a wig, then it does not matter what is done as regards the fastening since the helmet slides on the wig in a typical oblique impact situation in any case. However, as presented in Appendix B, another test was conducted using a helmet with a different structural design (Helmet B). With Helmet B, the angular acceleration was reduced by 42% and the angular velocity by 34%. Therefore, it can be concluded that even if the helmet slides on a human head due to hair, it is still possible to design a helmet capable of absorbing more energy driven by more realistic test methods. The question then is whether or not the test method should be designed using a wig. The proposal is not to use a wig due to calibration problem of the wig.

The fixation of the helmet on the head is important and needs to be controlled. Today, helmets use either a fit system built of foam material called comfort foam, or using a head restraint system that can be adjusted using a screw or air pump system. A test standard would also need to define the degree of adjustment in any fastening.

4.3 Impact location on the helmet

The impact location on the helmet should, if possible, be selected based on accident statistics like the impact locations on the helmet presented in COST 327 [21], McIntosh et al. [33] and Bourdet et al. [34]. Figure 9 shows proposed impact points.

The impact location could be defined either by impact point or a region/area. Both approaches have benefits. However, the limitation with defining a point on the helmet is that it could be that the helmet performs well for that point only. Meanwhile, defining a region on the helmet can result in a large variation of the measured kinematics if changing the impact point within the region. Appendix C shows results where the impact point has been shifted 4-5cm. Ican be seen that the Angular acceleration and the angular velocity are sensitive to shifts in the X-direction but not in the Y-direction. Fahlstedt et al. (2014) performed a sensitivity study based on FE simulations for one helmet on the market with an irregular shape. For the five different side impacts the peak first principal strain of the brain tissue varied between 0.21 and 0.59 (Figure 10).

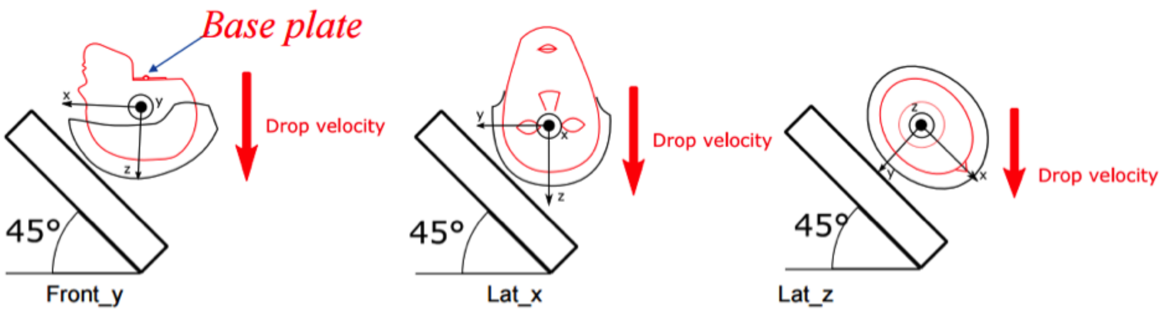


Figure 9: Proposal for impact points and impact directions.

The test line on the bike helmet should as well be defined. Accident statistics show that the test line defined in EN1078 is too high. It is therefore proposed to lower the test line so that the helmet will cover more of the head as also expressed by Otte et al 2014 and Willinger et al 2014. The final test line must be chosen so that the helmet still will be attractive and accepted by the end consumer.

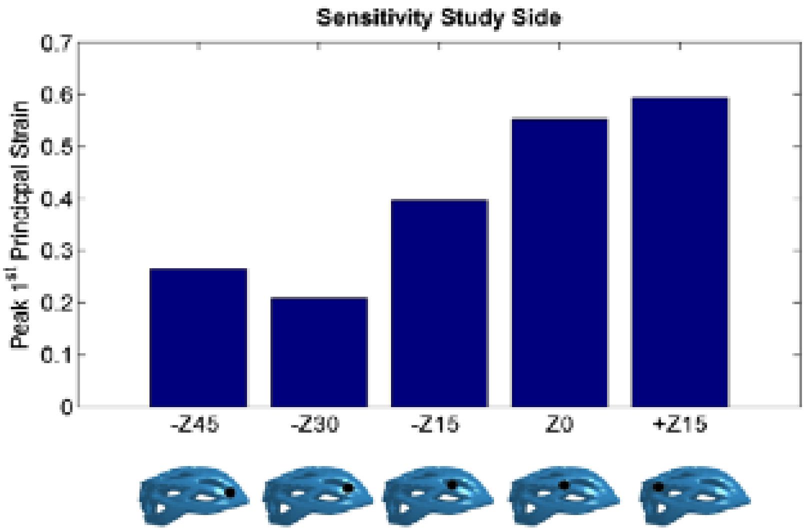


Figure 10: The peak first principal strain of the brain tissue in five different side impacts of a helmet available on the market with rather irregular shape.

4.4 Conclusion

It is proposed to maintain the linear impact conditions at 5.42 m/s as recommended in EN1078 but by using the Hybrid III head instead of the ISO head form. Main advantage at this level is to have a control of the rotational acceleration which may occur in case of non-controlled impact direction.

In addition an oblique test should be introduced including at least three tangential impacts with the helmeted head alone, at 6.5m/s against a 45°angled anvil, as illustrated in Figure 9.

5 Pass fail criteria

5.1 Introduction

Over the past forty years, a slant has been put by the biomechanical research community on the understanding of the head injury mechanisms. One of the main difficulties of this research field is that a functional deficiency is not necessarily directly linked to a damaged tissue. Nevertheless, an injury is always a consequence of an exceeded tissue tolerance to a specific loading. Even if local tissue tolerance has very early been investigated, the global acceleration of the impacted head and the impact duration are usually being used as impact severity indicator. Currently thresholds concerning helmet performance are set in terms of maximum headform acceleration (fixed at 250 or 275 G respectively for cyclists an motorcyclist) according to the WSU tolerance curve proposed in the 1950's. To protect the head in an automotive environment, HIC has been introduced in the 1970's as reported hereafter. This criteria, is based on the linear head acceleration evolution over time and has been set at around 1000 for linear frontal or occipital impact. For motorcycle helmets, this criterion has been set at HIC 2400 which has no sense in a biomechanical point of view. For bicycle helmets HIC is not considered. It must be mentioned here that maximum linear acceleration or HIC do not integrate lateral direction or rotational acceleration, when it has been demonstrated that the capability of the human head to support impact is strongly direction dependent *(Kleiven et al 2003)*. On the other hand it is well known since 1943 *(Holbourn et al 1943 and Ommaya et al 1968)* that rotational acceleration has a critical influence on intra-cerebral loading and in turn on DAI. These very simplified head injury criteria present therefore a number of limitations.

Hereafter a number of advanced head injury criteria proposed in the last decades are summarized and evaluated. This section is organized in 6 chapters dealing respectively with Head injury criteria based on linear acceleration, rotational acceleration, combined linear and rotational acceleration and also model based head injury criteria. A specific section is dedicated to the challenging task which is the objective evaluation of these head injury criteria, followed by a description of how to implement model based pass fail criteria into an advanced helmet test method.

5.2 Head Injury Criteria Based on Translational Acceleration

5.2.1 Maximum Resultant Head Acceleration

The Wayne State Tolerance Curve is considered to be the foundation of research on human head injury criteria (Figure 11). This curve derived from the research performed by *Lissner et al. (1960), Gurdjian et al. (1945, 1955, 1958 and 1961)* and *Patrick et al. (1963)*, and gives the tolerable average acceleration in A-P direction (Anterior-Posterior) as a function of pulse duration. The curve is given in Figure 1. Slight cerebral concussion without any permanent effects was considered to be within human tolerance. Only translational accelerations were considered in the development of the curve, which was obtained from different experiments with cadavers, animals and volunteers. The short duration part of the curve (2<t<6 ms) was derived from cadaver tests in which skull fracture was chosen as injury criteria. Cadaver and animal tests were used for the intermediate pulse durations (6<t<10 ms). For this part of the curve, intracranial pressure was used as the injury criteria in the cadaver tests and concussion was chosen as the injury criterion in the animal tests. The long duration part of the curve (t>10 ms) was obtained from volunteer tests. There was no head impact in these tests and no injuries were observed. By assembling all these tests in one single curve it was assumed that skull fracture and concussion correlate. Lissner et al. suggested that for a given duration, accelerations above the curve lead to injury (survival hazards), while accelerations below the curve are tolerable and cause, at most, cerebral concussion without permanent effects. Except for the long duration accelerations, the WST-curve has never been validated for living human beings.

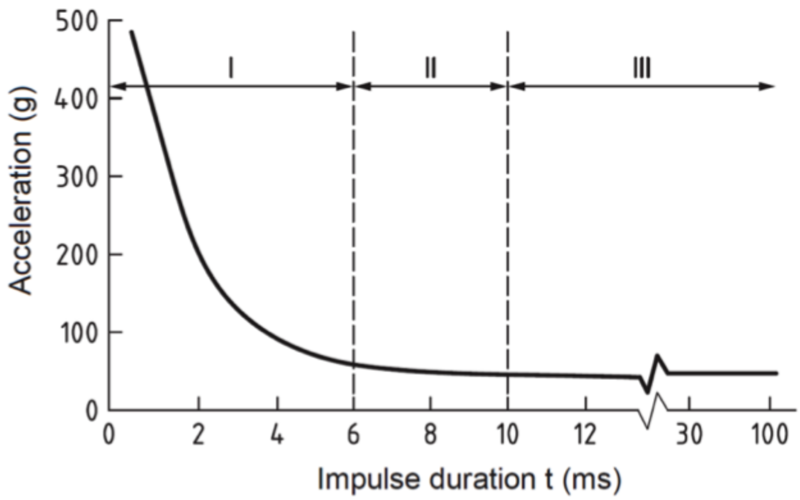


Figure 11: Wayne State Tolerance Curve The figure is divided into 3 parts:
1) Short duration area obtained from cadaver experiments;
2) Intermediate duration area, obtained from cadaver and animal experiments;
3) Long duration area, obtained from volunteer tests.
At a given duration, accelerations above the curve give injury, while accelerations below the curve do not lead to injury (Beusenberg 1991).

A very early head injury criteria which is often used because of its simplicity is the *maximum resultant head acceleration* (a_{max}). The threshold for a_{max} depends on its application, because of the time dependent nature of the resultant acceleration with respect to head injury. Maximum linear acceleration is used for many years and continues to be used in several helmet standards (Snell 1995, CSA 1985) $A_{max} < N$ with N a value which depends on the standard used. This criteria doesn't take into account the time duration of the impact even in some cases the maximum value is given for a maximum impact duration.

Therefore, a variation of this criteria is A_{3ms} value which refers to the maximum deceleration that lasts for 3ms. Even if a "kind" of time duration is taking into account, similar limitations can be done for this criterion. The A_{3ms} criteria is based on the WSTC.A_{3ms} should not exceed 80g *(Got et al.,* *1978)*. According to *Chin et al* (CEN TR16147), and based on *COST 327* reports, a head acceleration of 200 to 250 G and 250 to300G lead respectively to severe AIS4, respectively AIS5 head injury.

5.2.2 The Head injury criterion (HIC)

The Wayne State curve as described above led to the development of the Gadd Severity index (GSI), proposed by Gadd in 1966,which was expressed in the form:

$$GSI = \int_T a(t)^{2.5} dt \tag{1}$$

Where T = the total pulse duration, and a (t) = acceleration at the center of mass of the head, as a function of time.

This was described as the weighted impulse criteria for which a value of 1000 was considered unsafe. However, it can be shown that for irregular pulse shapes, there may exist within the pulse envelope which has a value greater than that for the whole pulse. The GSI has received significant scientific criticism, because it deviates considerably from WSUTC (Slattenschek & Tauffkirchen, 1970). Thus, it was decided that the maximum value within the pulse should be assumed to be the criterion for head injury. This became the Head Injury Criteria, HIC, which is given below:

$$HIC = \left[\left(\frac{1}{t2-t1}\int_{t1}^{t2} a_{res} dt\right)^{2.5} (t2-t1)\right]_{\max} \tag{2}$$

With: t_1 and t_2 [ms] any two points in time during any interval in the impact; a = resultant acceleration of the center of mass of the head.

After much discussion over many years, tl and t2 were defined to be any two times during the entire impact duration for which HIC is a maximum value. *Hodgson and Thomas (1975)* suggested that the critical HIC interval should be less than 15 ms, even if the HIC value exceeded the threshold of 1 000 over a longer interval. His finding was based on examination of events where the concussive outcomes were known or could be determined. The threshold of 1000 is still under discussion; because head injuries were found at HIC values of 500, while HIC values of 3 000 were sustained without major injury. The benefit of HIC over peak linear acceleration is that HIC is related to time and it is known that pulses with the same peak value but different duration can give a different injury outcome. Unfortunately, HIC and AIS values have never been satisfactorily correlated.

According to *Chin et al* (CEN TR16147), and based on COST 327 reports, a HIC1000, respectively HIC 2000 leads to 10-15% respectively 35-50% probability of death. Therefore it can be stated that HIC is an inaccurate criteria for extreme head injury which is not adapted for less severe brain injury.

5.2.3 The Skull Fracture Correlation (SFC)

SFC was developed by *Vander Vorst et al. (2003, 2004)* to predict skull fracture based on statistical analysis of PMHS test and FE simulation results. SFC was defined as the averaged acceleration over the HIC time interval.

$$SFC = \frac{\Delta V_{HIC}}{\Delta T_{HIC}} \quad (3)$$

Where ΔT_{HIC} is the time interval (t_1-t_2) that maximizes the integral in Eq 2. and ΔV_{HIC} is the change in velocity over the time interval.

Vander Vorst et al. (2003) studied the correlation between tensile skull strains, computed from an over simplified spherical FE head model with the SFC in frontal impact experiments. Vander *Vorst et al. (2004)* and *Chan et al. (2007)* extended the investigation to lateral impacts of skull and different shapes of the impactor for developing a generalized linear skull fracture criteria. However, due to lack of good correlation between strain and SFC for cylindrical impact surface, the study mainly focused on flat impact surface. The 50% Risk of skull fracture was proposed as SFC_{50} = 155g.

5.3 Head Injury Criteria based on Rotation Acceleration and velocity

Concerning neurological injuries, *Holbourn (1943)* suggests that the rotational acceleration induced by a given impact causes high shear strains in the brain, thus rupturing the tethering cerebral blood vessels, neo and subcortical tissue. This author was the first who suggests the importance of rotational acceleration in the appearance of cerebral concussion about 70 years ago!

5.3.1 Maximum rotational acceleration and velocity

There is no criterion related specifically to rotational acceleration. However, there has been research to determine what values of rotational acceleration are likely to cause injury and this is reviewed below. In 1967, *Ommaya et al*. 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. A chart of angular acceleration required to reproduce concussion in the rhesus monkey indicates that an acceleration of 40 krad/s^2 will have a 99% probability of producing concussion which corresponds to an angular acceleration of 7500 rad/s^2 for human.

Ommaya et al. (1968) studied the effect of whiplash injury on rhesus monkeys and showed that if the head was subjected to a rotational acceleration above a threshold value, subdural and subarachnoid injuries were obtained.

Unterharnscheidt (1971) 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). Based on animal experiments, he showed that a linear acceleration of 205g caused neither behavioral nor histological changes in the central nervous system. A linear acceleration of 280-400g produces commotions and considerable primary traumatic lesions are produced by impacts corresponding to more than 400g. In experiments concerning the effects of rotational acceleration on the brain, he showed that a rotational acceleration of 101-150 krad/s² lead to no injury. However for higher accelerations, up to 197 krad/s², he observed subdural hematomas combined with neurological injuries.

A series of head impact experiments was performed by Ono et al. (1980) using 63 live monkeys. In order to find a relationship between the impact and the observed lesions, several types of loading were used. The results indicated that the concussion and cerebral contusion depended on the translational and rotational acceleration impact. Brain contusions appeared at a rotational velocity of at least 300 rad/s, and 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.

In a primate study, *Gennarelli et al. (1982)* proposed that a rotational acceleration exceeding 175 krad/s² would produce SDH in the rhesus monkey.

Pincemaille et al. (1989) has conducted experimentations with volunteer boxers by equipping their head with an accelerometric system measurement in order to be able to record the kinematics of the head during fights. The angular acceleration limit recorded by the authors for a beginning concussion lied in the range of 13,6 krad/s² and 16 krad/s² which correspond to an angular velocity of 25 rad/s and 48 rad/s respectively. These values are higher than those proposed by *Ewing et al. (1975)* for the same type of analysis (1700 rad/s² corresponding to an angular velocity of 32 rad/s).

Pellman et al. 2003, generated injury risk curves for concussion from reconstructed NFL impacts using Hybrid III ATDs. In that study, the average concussive impact (n = 25) had a rotational acceleration of 6432 rad/s² and rotational velocity of 36.5 rad/s.

Rowson et al., 2012, proposed an estimate of rotational acceleration tolerance derived from direct acceleration measurements from instrumented human volunteers. The helmets of 335 football players were instrumented with accelerometer arrays that measured head acceleration following head impacts sustained during play, resulting in data for 300,977 sub concussive and 57 concussive head impacts. The authors developed an injury risk curve and proposed a nominal injury value of 6383 rad/s² associated with 28.3 rad/s represents 50% risk of concussion.

More recently *Patton et al 2012* reported maximum rotational acceleration respectively velocity of 4.5 krad/s² , respectively 33 rad/s² as a threshold for short or no loss of consciousness, based on a set of American football players head impact analysis.

According to *Chin et al* (CEN TR16147), and based on *COST 327* report and literature review, it could be shown that concussion AIS 1-2 can occur at 5 krad/s² and fatal injury AIS 5-6 can potentially occur at 10 krad/s². This correlates with data that indicate that there is a 35 % risk of a brain injury of AIS 3 - 6 at 10 000 rad/s².

5.3.2 Brain Injury Criteria, BrIC

In 2011, *Takhounts et al* proposed a new metric in order to define a head injury criterion. For this, the authors used a Finite element human head model (SIMon model, *Takhounts and Eppinger*, 2003). With this model, a total of 114 animal brain injury experiments were simulated in the development of the biomechanical injury metric – CSDM (cumulative strain damage measure). CSDM is based on the hypothesis that DAI is associated with the cumulative volume of brain tissue experiencing tensile strains over a predefined critical level.

Next, frontal impact tests with the Hybrid III dummy (43 NCAP tests - drivers and passengers – available from NHTSA database) were used to develop BRIC for frontal impact. To do so, first, based on criteria established previously with SIMon FEM, CSDM values were calculated for each test. Then optimization was carried out to obtain the best linear fit between CSDM and BRIC (in the form of the following equation) using critical values of angular velocity and acceleration ω_{cr} and α_{cr} as design variables and subjected to the constraint that BRIC=1 when CSDM =0.425 (30% probability of DAI/AIS4+).

$$BrIC = \frac{\omega_{max}}{\omega_{cr}} + \frac{\alpha_{max}}{\alpha_{cr}} \tag{4}$$

Where ω_{max} and α_{max} are maximum angular velocities and accelerations for each accident cases respectively. The linear relationship between CSDM and BRIC was then utilized to obtain risk curves for hybrid III dummy (Figure 12). The critical values of angular velocity and acceleration for the Hybrid III dummy were found to be ω_{cr}=46.41 rad/s and α_{cr}=39,774.87 rad/s².

After this, the authors used statistical artefact in order to propose head injury risk curves for different AIS levels for HIII dummy in frontal impact case. Same methodologies have been done for different dummies for different impact orientations and BrIC criteria have been proposed.

In 2013, Takhounts et al proposed a new definition of this BrIC criterion as follows:

$$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2} \tag{5}$$

where ω_x, ω_y, and ω_z are maximum angular velocities about X-, Y-, and Z-axes respectively, and ω_{xC}, ω_{yC}, and ω_{zC} are the critical angular velocities in their respective directions.

Even if the definition of BrIC value changed, the methodology to define head injury criterion was the same than previously described (based on SIMon model which is based on animal data).

The authors proposed some limitations of this criterion:

- "First, all the limitations that were applicable in the development and validation of SIMon finite element head model *(Takhounts et al, 2003, 2008)* are applicable" to this criterion. Main limitations of SIMon model are that this model is based on animals data
- "Second, only DAI type anatomic brain injuries in animals were investigated"
- BrIC is not an "ultimate" head injury criterion that captures all possible brain injuries and skull fractures"

The authors proposed to combine BrIC criterion with HIC value in order to "better capture head injury" and eventually to also take skull fracture risk into account.

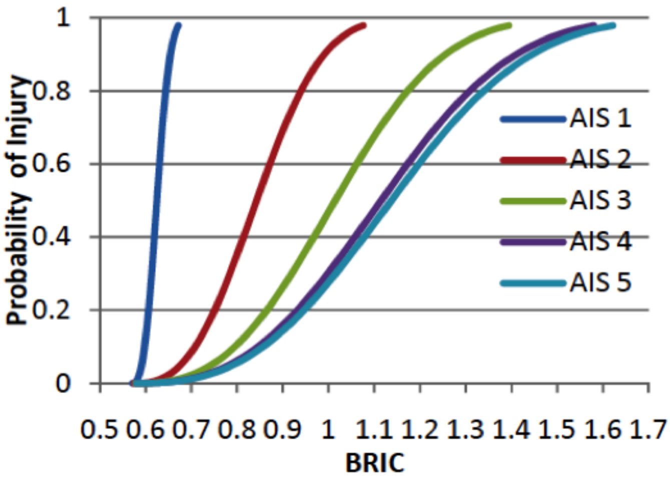


Figure 12: Risk of brain injuries as a function of BRIC for various AIS levels for Hybrid III (Frontal impact).

5.3.3 Injury criteria based on RIC

Kimpara et al. (2011) developed a new head injury criterion by taking into account the resultant rotational acceleration instead of resultant linear acceleration in similar way to HIC. This is called Rotational Injury Criterion (RIC), derived by substituting resultant angular acceleration of α (t) for in Eq.2. RIC is defined as

$$\mathrm{RIC} = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} \alpha(t)\,dt \right\}^{2.5} \right]_{\max} \quad (6)$$

The maximum integral time duration for RIC was set to 36 ms, which was the original time duration of HIC. RIC_{36} was correlated with CSDM computed from 31 impact events involving 58 American football players with strain thresholds of less than 15% (R>0.89). To predict mild TBI based on logistic regression (modified maximum likelihood method) the 50% risk value is RIC_{36} = 1.03x 10^7.

5.4 Head Injury Criteria based on Combined Rotational and Translational Accelerations

5.4.1 Generalized Acceleration Model for Brain Injury Tolerance, GAMBIT

In 1999 and 2000, *Newman et al.* proposed a new methodology to assess brain injuries, based on multiple accident reconstructions of American football players' head collisions during recorded 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. For the injury cases, the peak resultant linear and angular head acceleration varied from 48 to 138g and 2615 to 9678 rad/s² respectively. For the non-injury cases the peak value of head acceleration varied from 19 to 102 g and 1170 to 6613 rad/s² respectively.

In an attempt to combine translational and rotational acceleration, *Newman in 1986*, in contact with Transport Canada, introduces the concept of generalized GAMBIT (Generalized Acceleration Model for Brain Injury Tolerance). The model attempts to weight, in an analogous manner to the principal shear stress theory, the effects of the two forms of motion. G=1 is set to correspond to a 50% probability of MAIS 3. However, the GAMBIT was never extensively validated as an injury criterion. For example, the maximum time interval for a and m have never been set.

$$G(t) = \left[\left(\frac{a(t)}{a_c}\right)^m + \left(\frac{\alpha(t)}{\alpha_c}\right)^n\right]^{\frac{1}{s}} \quad (7)$$

Where a(t) and α(t) are the instantaneous values of translational and rotational acceleration respectively. α_c and a_c are limiting critical values and n, m and s are empirical constants (n = m = s = 2.5, a_c=250g, α_c = 25.000 rad/s²)

5.4.2 Head Impact Power, HIP

The Head Impact Power (HIP) was proposed by *Newman et al. (2000)*, the head is also seen as a one mass structure. It is computed using both linear and angular accelerations measured at the center of gravity of a *Hybrid III* dummy head as shown in the following formula:

$$HIP = \underbrace{C_1 a_x \int a_x dt + C_2 a_y \int a_y dt + C_3 a_z \int a_z dt}_{Linear\ contribution} + \underbrace{C_4 \alpha_x \int \alpha_x dt + C_5 \alpha_y \int \alpha_y dt + C_6 \alpha_z \int \alpha_z dt}_{Angular\ contribution} \quad (8)$$

The C_i coefficients are set as the mass and appropriate moments of inertia for the human head: $C_1 = C_2 = C_3 = 4.5$ kg, $C_4 = 0.016$ N.m.s², $C_5 = 0.024$ N.m.s², $C_6 = 0.022$ N.m.s².

- a_x, a_y and a_z [m.s^{-2}] are the linear acceleration components along the three axes of the inertial reference space attached to the dummy head.
- α_x, α_y and α_z [rad.s^{-2}] are the angular acceleration components around the three axes of the inertial reference space attached to the dummy head.

Since the HIP is a time-dependent function, the value taken as an injury predictor candidate is the maximum value reached by this function. 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 more severe brain injuries, since such experimental data is not yet available. 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 accelerations, HIP_{max} can also incorporate directional sensitivity, sensitivity for rotational accelerations and sensitivity for angular and translational velocities. However HIP was designed only for brain injury and not for SDH or skull fracture.

5.4.3 Injury criteria based on PRHIC

Kimpara et al. (2011) omitted the linear terms from Eq 8 to get the angular HIP-ang(t) which represented the rate of change in angular head kinematic energy as described:

$$HIP\_ang(t) = \sum I_{ii} \cdot \alpha_i \cdot \int \alpha_i dt \quad (9)$$

The plots of maximum value of HIP-ang(t) versus the time duration of HIP-ang formulate similar distribution as GSI or HIC definition. The resultant linear acceleration of Eq 2 was replaced by Hip-ang(t) and a new criteria called power rotational head injury criterion (PRHIC) was developed as described in Eq 10.

$$PRHIC = \left[\left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} HIP\_ang(t)\, dt \right\}^{2.5} (t_2 - t_1) \right]_{max} \quad (10)$$

The maximum integral time duration for PRHIC was set to 36 ms, which was the original time duration of HIC. $PRHIC_{36}$ was correlated with CSDM computed from 31 impact events involving 58 American football players with strain thresholds of less than 20% (R>0.90). To predict mild TBI based on logistic regression (modified maximum likelihood method) the 50% risk value is $PRHIC_{36} = 8.70x\ 10^5$.

5.4.4 Injury criteria based on Principal Component Score (PCS)

To quantify sensitivity of various biomechanical measures of head impact (linear acceleration, rotational acceleration, impact duration, impact location) to clinical diagnosis of concussion in American football players Greenwald et al. (2008) developed a novel measure of head impact severity which combines these measures into a single score called Principal component score (PCS) that better predicts the incidence of concussion. PCS is a weighted sum of translation and rotational accelerations, HIC, and SI with empirically determined weights, as shown below,

$$PCS=10 \cdot ((0.4718 \cdot sGSI+0.4742 \cdot sHIC+0.4336 \cdot sLIN+0.2164 \cdot sROT)+2) \quad (11)$$

Where: sX = (X – mean(X)) / (SD(X)), LIN = linear acceleration, ROT = rotational acceleration HIC = Head Impact Criteria, GSI = Gadd Severity Index.

On-field head impact data were collected from 449 football players at 13 organizations using in-helmet systems of six single axis accelerometers and Concussions were diagnosed by medical staff and later associated with impact data. When all impacts were considered, every biomechanical measure evaluated was statistically more predictive of concussion than guessing ($p < 0.005$). However, for the top 1% and 2% of impacts based on linear acceleration, a subset that consisted of 82% of all diagnosed concussions, only PCS was significantly more predictive of concussion than guessing ($p< 0.03$), and, when compared to each other, PCS was more predictive of concussion than classical measures for the top 1% and 2% of all data ($p < 0.04$).

5.5 Head Injury Criteria based on FE head modelling

It should be mentioned that the FE models mentioned in this section are models that have been used in studies where the output from the model has been correlated to a tissue level injury criteria. It is not therefore said that the models not described in detail have less potential to be used as an injury prediction tool. The models not described in detail are described in Hosey et al 1980, Ruan et al 1993,Claessens et al 1997, Iwamoro et al 2007, Colgan et al 2010.

Brain injury is reported to correlate with stress, strain and strain rate *[Lee & Haut, 1989; Viano & Lövsund, 1999]*. However, strains and strain rates inside the brain (during impact) are difficult to measure. Advancements in computational techniques have led to more accurate and more detailed numerical models of the human head. These models bring a detailed injury assessment closer to reality and at tissue level, since they enable stresses and strains to be examined. In the last decade tissue level brain injury criteria have been proposed in terms of:

- MPS Max Principal Strain
- SCC Strain in Corpus Callosum
- VM strain Max Von Mises strain
- SSR Strain*Strain rate
- Pmax Max Pressure
- VM stress Max Von Mises stress
- CSDM Cumulative Strain Damage Measure
- MAS Maximum Axonal Strain

The present synthesis focuses on the most recent results with special attention paid to the new generation of head FE models which enable it to compute axon elongation with advanced anisotropic brain models.

5.5.1 Simulated Injury Monitor, SIMon

Two head models have been proposed by the National Highway Traffic Safety Administration (NHTSA) in different generations. One is SIMon 2003 (Simulated Injury Monitoring) developed by *Takhounts et al. (2003)* and the other is SIMon 2008 by *Takhounts et al. (2008)*. Simon 2003, consists of a rigid skull, the dura-CSF layer, the brain, the falx cerebri, and the bridging veins. It represents the head of a 50^{th} percentile male head and has a mass of 4.7kg. The mass of the brain alone is 1.5 kg. It is built with 10475 nodes and 7852 elements (7776 hexagonal solid and 76 beam elements). 76 beam elements are used to represent the bridging veins. The brain is modeled with a linear isotropic viscoelastic material model. The other parts are modeled with elastic material except the skull which is considered as rigid. This 2003 version model is validated for brain motion for one test data from *Hardy et al., (2001)*.

The next detailed head model of SIMon was developed by *Takhounts et al., (2008)* and illustrated in Figure 13. The topology of the SIMon 2008 FEHM is based on CT scans of a single male individual with the head size close to that of 50th percentile male. Detailed surfaces of the cerebrum, cerebellum, and brain stem were generated. The SIMon FEHM consists of 42,500 nodes and 45,875 elements, of which 5153 are shell elements (3790 rigid), 14 are beam elements, and 40,708 are solid elements. This is a larger model compared to the previous (simpler) version of SIMon 2003. The mass of the new head model is 4.5kg including the brain mass of 1.5 kg. The brain was modeled with a linear isotropic viscoelastic material model. The PAC-CSF was also modeled with a viscoelastic model. The ventricle was modeled as an elastic fluid and the other part excluding skull (modeled as rigid) were modeled as elastic material.

This new model is validated against intracranial pressure data of *Nahum et al., (1977)* and *Trosseille et al., (1992)* experiments. Validation against 3 tests of *Hardy et al., (2001)* was done to study the local motion of brain.

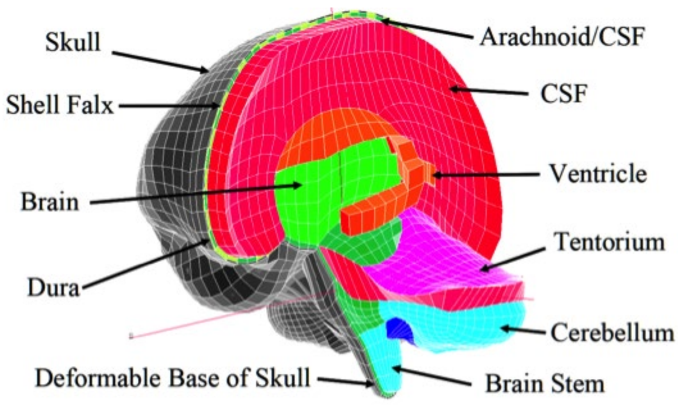


Figure 13. SIMon model version 2008 (Takhounts et al., 2008)

These two models were tested using available experimental animal injury data, including rhesus monkeys *(Abel et al., 1978; Gennarelli et al., 1982; Stalnaker et al., 1977; Nusholtz et al., 1984), baboons (Stalnaker et al., 1977)*, and miniature pigs *(Meaney et al., 1993)*. A total of 114 animal brain injury experiments were simulated in the development of the biomechanical injury metric - CSDM. The experimental kinematic loading conditions were scaled in amplitude and time to satisfy the equal stress/velocity scaling relationship.

Once correctly scaled, these loading conditions were applied to the SIMon model. It was assumed that the injury results from animal subjects were the same as that which would be observed from a human under the equivalent impact input. With this, authors developed a metric, the Cumulative Strain Damage Measure to predict brain injury. The CSDM is based on the hypothesis that diffuse axonal injury (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.

It was found that CSDM (0.25) and maximum principal strain correlated with DAI observed from animal tests.

5.5.2 Wayne State University Brain Injury Model (WSUBIM)

Over the last several years, several versions of the Wayne State University Brain Injury Model (WSUBIM) were developed. Here we will discuss the recent development done by *Zhang et al., (2001)* and King et al 2003 as illustrated in 15. This revised model had equivalent anatomical features of a 50th percentile male head including the scalp, skull with an outer table, diploë, and inner table, dura, falx cerebri, tentorium, pia, sagittal sinus, transverse sinus, cerebral spinal fluid (CSF), hemispheres of the cerebrum with distinct white and gray matter, cerebellum, brainstem, lateral ventricles, third ventricles, and bridging veins. It consisted of a total of over 281,800 nodes and 314,500 elements, with a mass of 4.5 kg. The brain of WSUBIM was modeled with a linear viscoelastic material model in PAM-CRASH Version 2000.

This model was validated against intracranial pressure data from *Nahum* and *Trosseille's* experiments as well as in terms of local brain motion *(Hardy et al., (2001))*. Facial bone of this model was validated against test data from *Nyquist et al., (1986)*. The nasal, frontal, zygomatic and maxillary bones were validated by comparing force-displacement curves from experiments with simulation results.

Based on the reconstruction of 58 American football impacts brain injury criteria have been proposed in terms of strain times strain rate (19 s^{-1} for 50% risk of mild TBI) by *King et al 2003*.

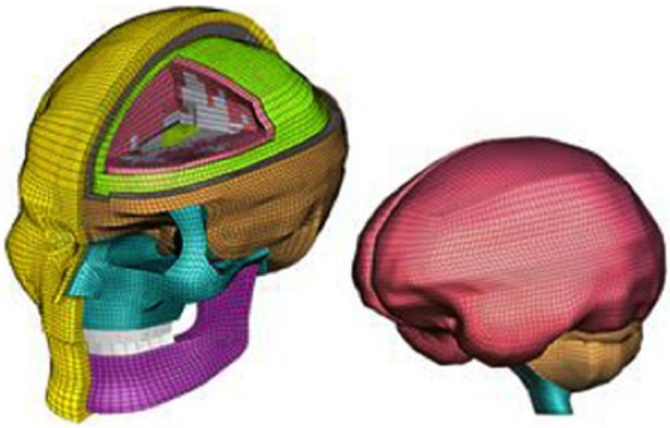


Figure 14: The WSU Brain Injury Model

5.5.3 KTH model and injury criteria

The KTH model presented in *Kleiven et al 2002* and shown in Figure 15-a. This model consists of scalp, skull, brain, meninges, cerebrospinal fluid (CSF), and eleven pairs of parasagittal bridging veins. A simplified neck, including the extension of the brain stem to the spinal cord, dura mater, pia mater, vertebrae, and muscles, was modeled also.

The model was comprised of 19350 nodes, 11454 eight-node brick elements, 6940 four node shell and membrane elements, and 22 two-node truss elements. The total mass of the head is 4.5 kg.

Mooney- Rivlin hyperelastic constitutive law was used for the brain model with addition of second order Proney series to account viscosity. The skull is modeled with elastic material with failure and the other parts are modeled with elastic material model.

The KTH head model was validated against intracranial pressure data from Nahum experiments and in terms of local brain motion *(Hardy et al. 2001, 2007)*.

This model was also used for the simulation of 58 American football impacts by *Kleiven et al 2007*. With this isotropic brain model a critical value for the First Principal Strain was established at 21% for corpus callosum and 26% for the white matter. Most recently the model was used to reconstruct bike accidents as presented in Section 2.2. Also the KTH head model was used in the Folksam Benchmark study of Bike helmets mentioned in Section 3.1. For this set of tests the strain computed in the FE model were between 16% and 30%. The computed maximum principal strain correlated best with the angular velocity and least good with the translational acceleration.

More recently *Giordano et al. 2014* implemented main axon bundle direction into this brain model and demonstrated that axonal strain is the best metric for brain injury description. Based on the reconstruction of 58 American football impacts a threshold of 7% to 15% axonal strain was proposed.

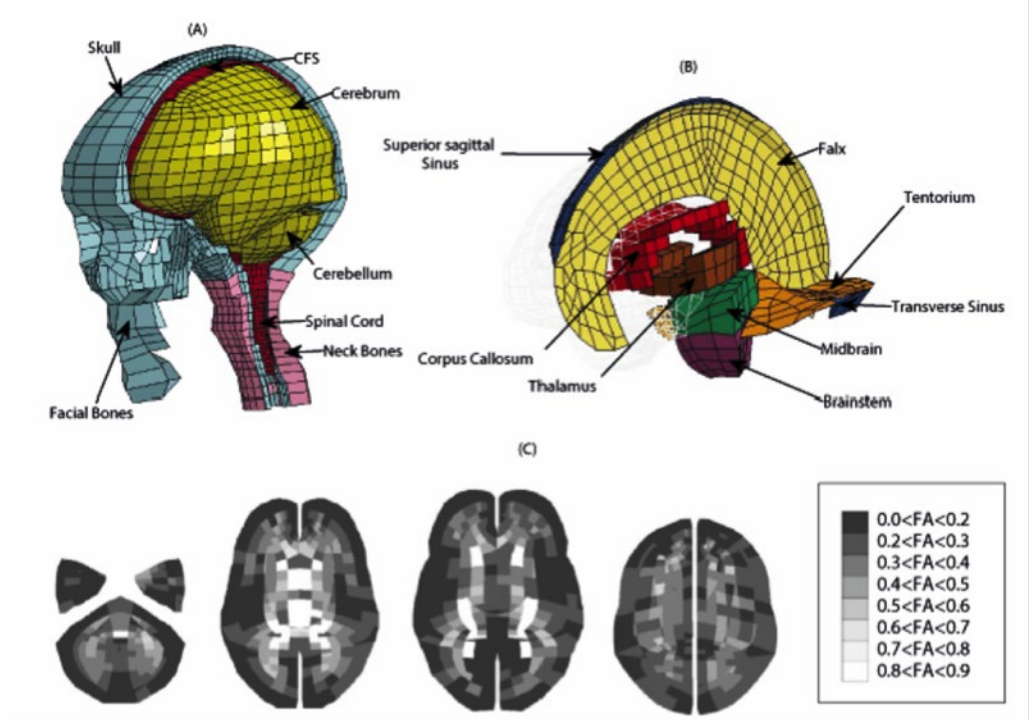


Figure 15: a) The isotropic KTH head model; b) internal view of the head model; c) Brain redefinition based on DTIs.

5.5.4 Strasbourg University head injury criteria (SUFEHM)

Based on head FE model developed by *Kang et al. in 1997, Deck and Willinger (2008)* developed head injury criteria based on the reconstruction of 68 accident cases. The proposed tolerance limits for 50% injury risk for different injury mechanisms are reported Deck et al 2008. The proposed head geometry is based on a human skull, which has been digitized externally and internally. Membranes such as falx and tentorium are based on anatomic atlas and a brain-skull interface of two millimeters thickness has been considered in order to represent the CSF. Brain, CSF and scalp are modeled with brick elements.

As a function of application two approaches exist for the skull model, i.e. a rigid skull, or a deformable and frangible skull, modeled by a three-layered composite structure with constant. Top of Figure 16 illustrates the main anatomical features taken into account like skull, the CSF, the membranes and the brain structure. Concerning the cerebral structure, the 3D directions of the main axon fibers have been implemented into the brain model, based on MRI medical imaging, and more precisely on Diffuse Tensor Imaging (DTI) by *Chatelin at al 2013* as shown in Figure 16.

This advanced model has then be validated at skull and brain behavior level by *Sahoo et al 2013, 2014, 2015 a,b*. It is important to consider these axon directions in order to enable the brain model to compute the axon elongation in case of impact, as it is well known that this phenomenon leads to neurological injuries. The present mechanical model of the human head permits it to compute skull deformation, brain skull relative motion as well as the strain of the main axon fibers, a physical parameter directly linked to the occurrence of DAI (Diffuse Axon Injury) known to be the cause of coma and death.

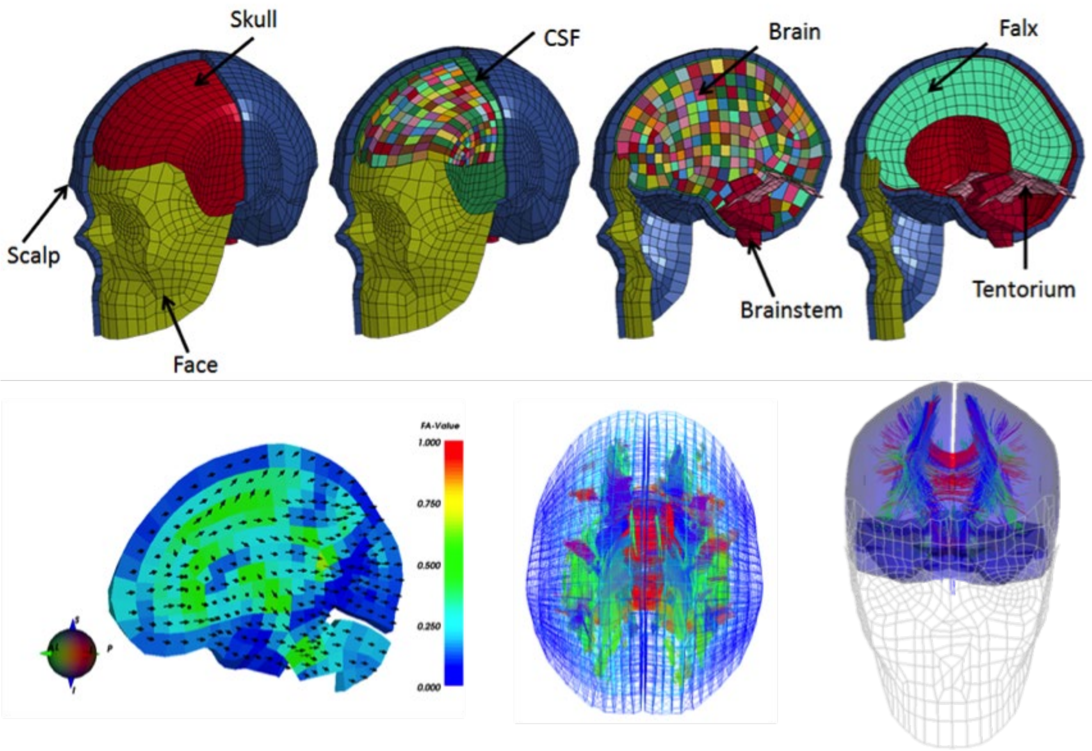


Figure 16: Illustration of the SUFEHM and the main axon fiber bundles implemented into the anisotropic brain model.

In order to establish human head tolerance limits (or head injury criteria), a total of 125 well-documented real world head trauma have been simulated with the above head model in collaboration with a number of international partners as reported by *Sahoo et al. 2015*. A detailed description of the head trauma database used is reported in *(Deck et al. 2008a&b, Peng et al. 2013, Munsch et al. 2009, Bourdet et al. 2013, Sahoo et al. 2014)*. Finally the methodology applied for the accident reconstruction and the simulation of the head trauma is illustrated in Figure 17. Several cranial and intra-cranial mechanical parameters have been computed for each case and correlated with the occurrence of skull fracture and neurological injuries. Concerning the neurological injuries threshold was set at a reversible brain injury classified as AIS2 injuries, corresponding to short-term coma.

Concerning the tolerance limit to skull fracture the statistical analysis has shown that the key parameter is the strain energy within the skull. Figure 18 reports the histogram with the strain energy in the skull for the injured and the non-injured cases. The regression analysis showed that the critical value (for a 50% risk of skull fracture) is 0.450 J. More precisely the injury risk curve for skull fracture is also shown in Figure 18. The robustness of this curve is characterized by a Nagelkerke parameter of 0.60.

Coming to the brain injury tolerance limit, the statistical analysis has shown that the key parameter is the computed axon strain. Figure 19 reports the histogram showing the axonal strain computed for the injured and the non-injured cases. The regression analysis demonstrated that the critical value (for a 50% risk of short term coma) is an axonal strain of 15%. More precisely the injury risk curve for brain AIS2+ injury is also shown in Figure 19. The robustness of this curve is characterized by a Nagelkerke parameter of 0.87. It should be mentioned that specific post-processors which compute the injury risk automatically also exist in order to ensure a non-user dependent assessment and to permit no end-users with limited skills in FE simulation to use the head injury prediction tool. The present head injury prediction tool has been used in similar consumer tests mentioned in section 3.1 published in Germany (Stftung Warentest) and in France (60 Millions de Consomateurs in August 2015.

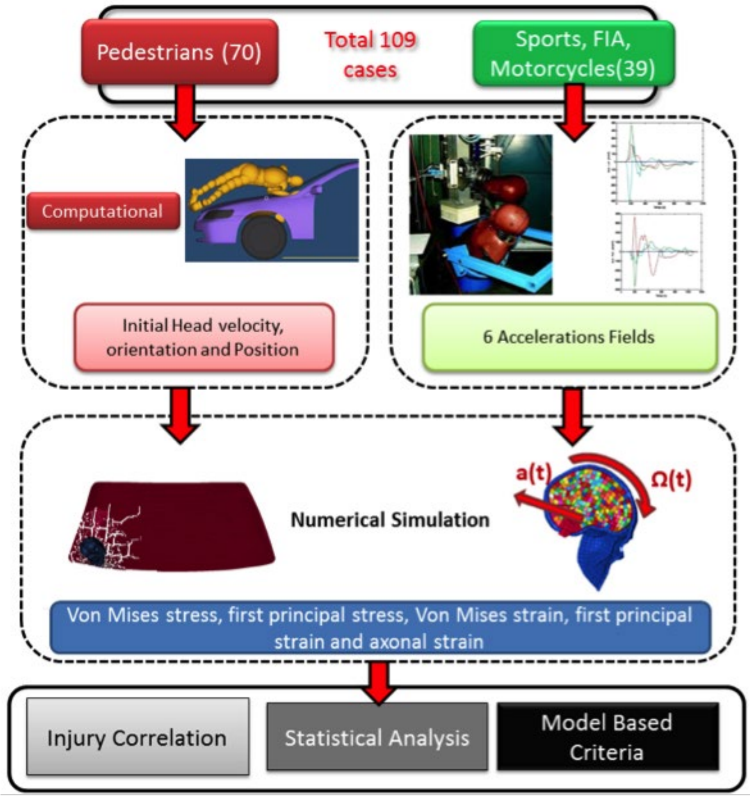


Figure 17: Illustration of the methodology implemented for the accident reconstruction and the simulation of the head trauma

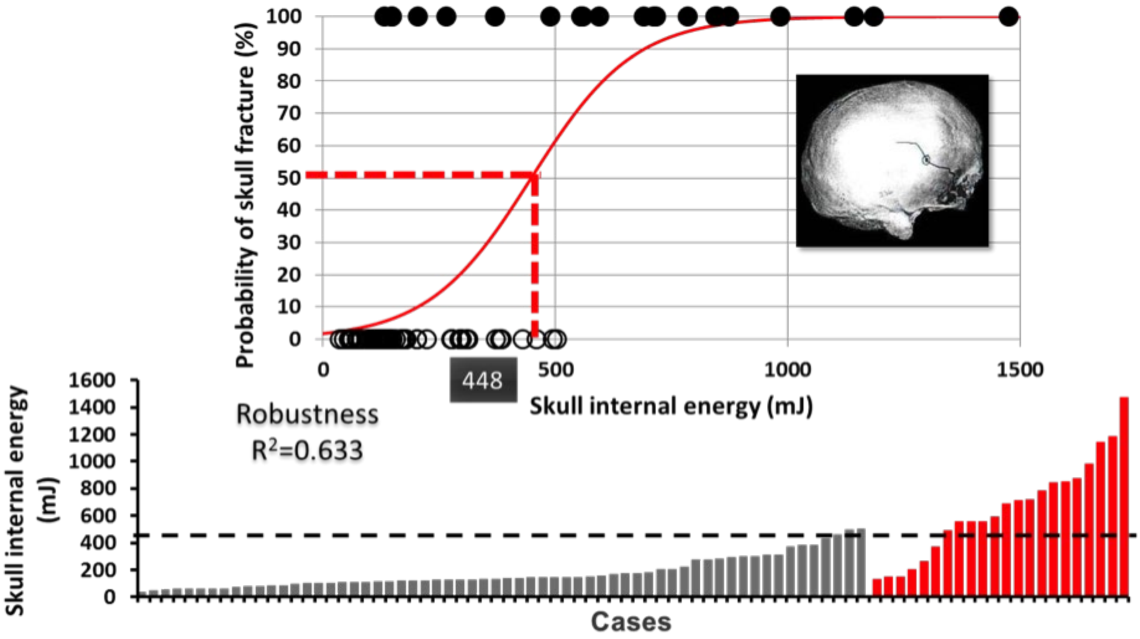


Figure 18: Head injury criteria in terms of skull fracture: Bottom - Histogram showing the computed skull strain energy for all head trauma cases, Top- skull fracture injury risk curve.

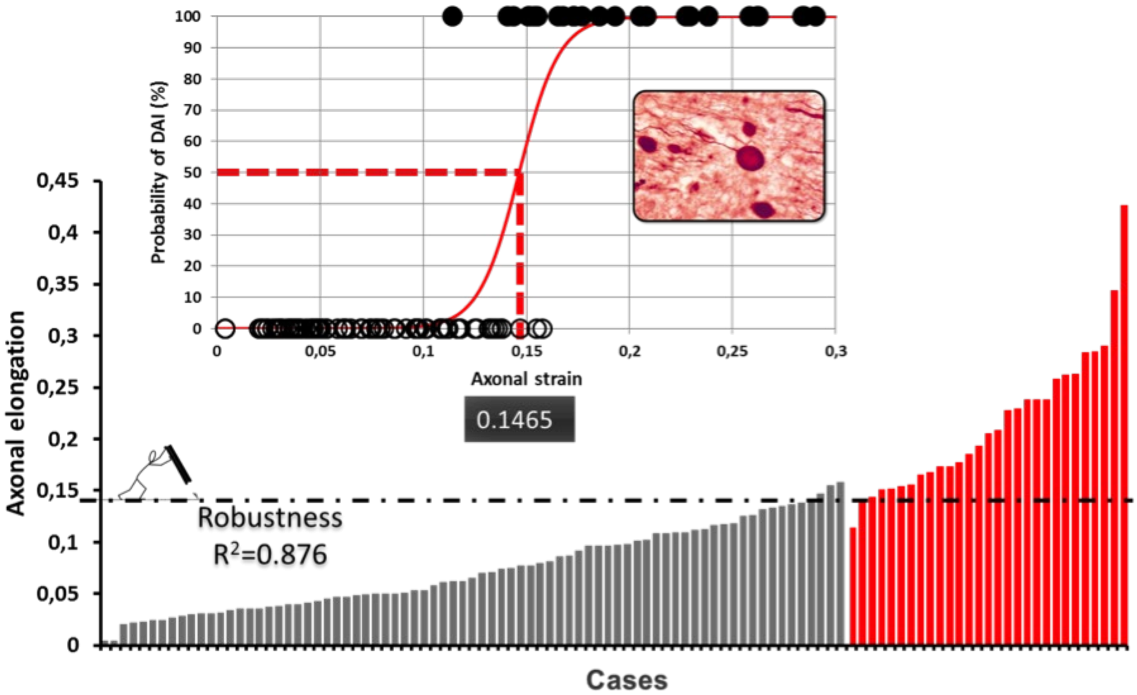


Figure 19: Head injury criterion in terms of brain injury: Bottom - Histogram showing the computed axonal strain for all head trauma cases, Top- brain injury risk curve.

5.6 Evaluation of head injury criteria

5.6.1 Introduction

The evaluation of a head injury criteria or the capability of a given criteria to predict injury is expressed via different statistical methods like binary logistic regression as illustrated in Figure 20 and reported by (Hynd et al 2004) or via the Hosmer–Lemeshow (HL) test (Hosmer et al 1980). These methods establish the correlation of a given parameter with the occurrence or not of a given injury. Typically these statistical methods establish an injury risk curve but also an objective statistic parameter which gives the quality of the statistic regression, such as for example the Nagelkerke R^2 parameter.

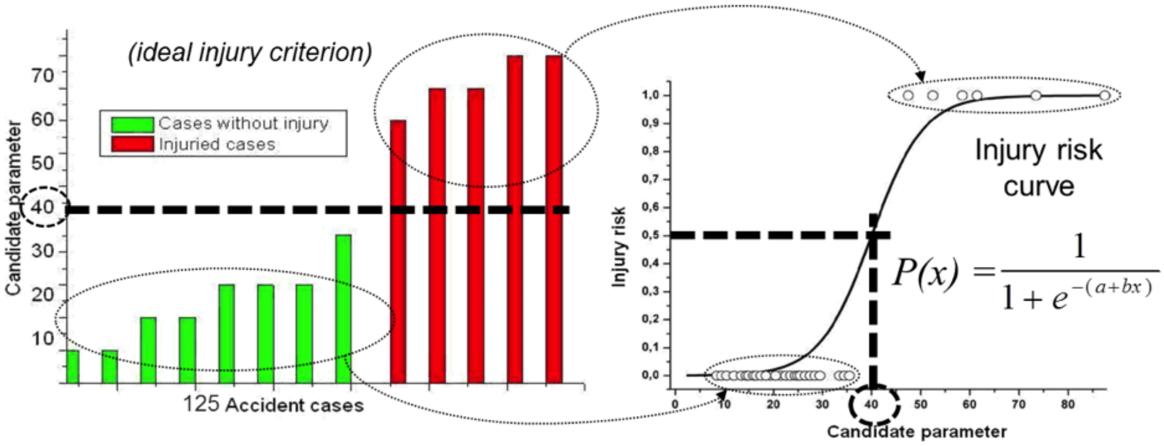


Figure 20: Illustration of the binary logistic regression method for the definition of a criterion.

It appears therefore that the evaluation of the quality of an injury criterion needs the definition of an adequate head trauma database. In the literature the following databases are reported:

Yoganandan et al 2004	86 Skull fracture experiences
Newman et al 2000	58 American Football players (25 concussions)
Willinger et al 2000	22 American Football
Takhounts et al 2008	114 animal tests
Kleiven et al 2007	58 American football players
Deck et al 2008	68 head trauma
Giordano et al., 2014	58 American Football players
Sahoo et al 2014	125 head trauma

The present document does not conduct any criteria evaluation. Its objective is just to synthetize briefly the evaluations reported in the literature and to try a conclusion. It should be mentioned here that some studies (ISO WG6) report a correlation study between some combinations of several kinematic parameters with some FE computed head responses. In no case this can be considered as an evaluation of head injury criteria.

5.6.2 Overview of existing evaluations

In 2006 *Marjoux et al. (2006)* proposed to investigate the human head injury prediction capability of the HIC and the HIP based criterion as well as the injury mechanisms related criteria provided by the SIMon, compared to shearing stress computed with the SUFEHM head model. Sixty-one real world accident cases have been reconstructed in order to provide head acceleration fields and head initial impact conditions so that the HIC, the HIP, the SIMon and the SUFEHM criteria can be computed. The advantage of this methodology is that this injury prediction capability is not deduced from *ex-vivo* or animal experiments but on real-world head trauma. The main result of this study was the poor capability of HIC, HIP and SIMon model to predict head injuries. As an illustration, Figure 21 reports comparative R^2 of some of the injury criteria.

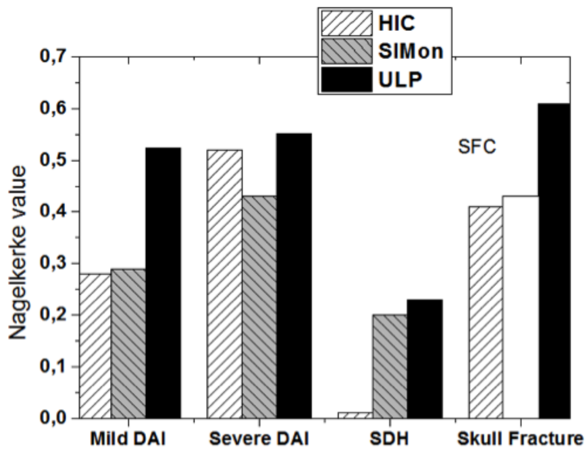


Figure 21: Comparative values of Nagelkerke parameters calculated for HIC, SIMon and ULP (former UNISTRA) by Marjoux et al 2006.

Kleiven 2007 reported an evaluation of injury criteria based on the simulation of 58 American football players. Nagelkerke parameters ranging from 0.3 to 0.6 were found for global kinematic parameters as well as for KTH model based injury criteria.

In the framework of EU project APROSYS-SP5, HIC criterion was evaluated based on Strasbourg accident database (i.e. 68 real world accidents). In a very first step global (input) parameters as well as HIC value have been considered in order to evaluate the correlation of these parameters with the occurrence of head injury. When the binary logistical regression method is used (using SPSS software package), it appeared that HIC presents an acceptable correlation with severe neurological injury (R^2=0.58 for HIC15) which means in most of the cases when victims sustained fatal injuries or suffered coma for a long time. Threshold parameter for a 50% injury risk obtained computed with the present set of accident is respectively 150 G for maximum acceleration and 1500 for HIC. However, correlation of HIC with moderate neurological injury as well as with SDH was poor (R^2=0 for SDH, R^2<0.3 for moderate DAI. Further this study showed that peak rotational acceleration present a poor correlation with observed injuries (R^2<0.33) especially for severe DAI and for SDH with a R^2<0.22. In this study SIMon and SKF were also evaluated and some of the Nagelkerke parameters are compared in Figure 22. Further details on this study are reported in *Deck et al. 2009*.

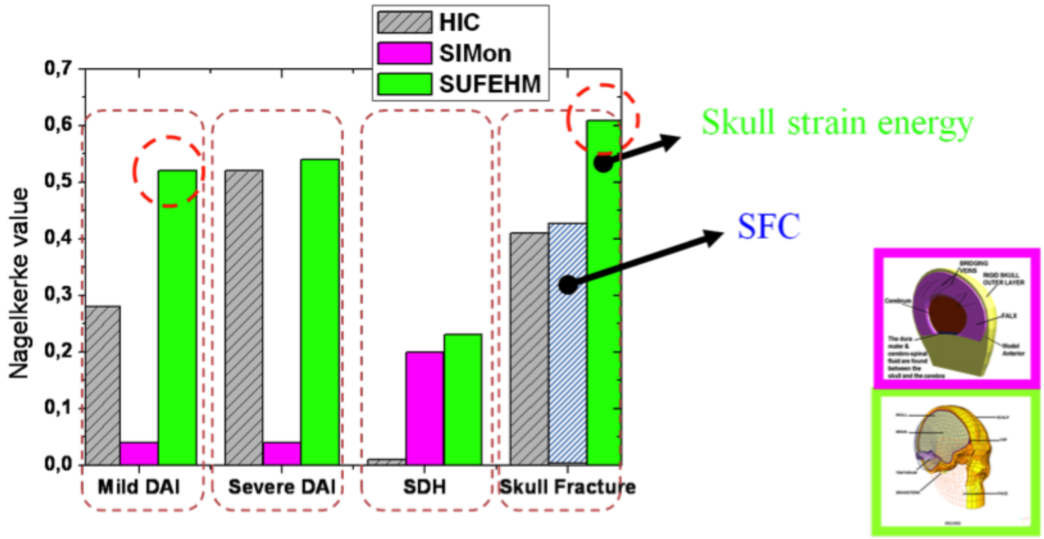


Figure 22: Comparative values of Nagelkerke parameters calculated for HIC, SIMon and SUFEHM by Deck et al 2009.

In *Giordano et al 2014* a total of 58 American football players head impact cases were computed and the logistic regression plots related to HIC, BrIC, MPS, MAS, AESM and CSDM was computed. This study showed that MAS presents the highest correlation with injury compared to MPS, CSDM, BrIC and HIC.

More recently *Sahoo et al, 2015* evaluated several head injury criteria, both based on global head kinematic and skull and brain tissue level injury criteria. The head trauma database involved in this study includes 125 cases. The evaluation of the different injury criteria is reported in Figure 23 and demonstrates that the axonal strain is the most relevant brain injury metric.

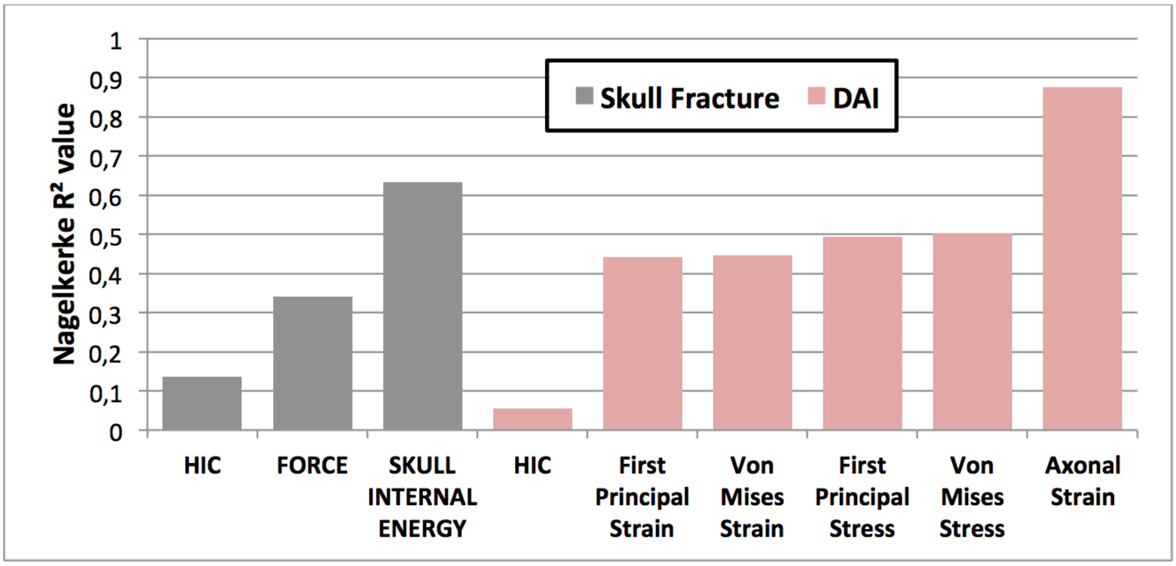


Figure 23: Comparison of regression parameters in terms of Nagelkerke parameter reported by *Sahoo et al 2015* and based on the simulations of 125 real world head traumas with the SUFEHM *(Sahoo et al 2013, 2015a and b.)*

5.6.3 Conclusion

It appears in this attempt to evaluate several injury criteria that one difficult aspect of injury criteria evaluation is the definition of the head trauma database. For example the database involving only American football players permits only the evaluation of criteria linked to mild concussion. On the other side databases considering scaled animal tests are also questionable. Finally real world head trauma coming from road accidents are built here and there but not yet harmonized. Despite the encountered difficulties it can be concluded and it has often been published that HIC has a very limited injury prediction capability and that FE model based and specifically axon strain based injury criteria permits a much more accurate brain injury risk assessment

5.7 WG 3 proposal for pass fail criteria

The pass/fail criteria for a helmet test for bicycle helmets need to be simple, robust and easy to use by the notified body and the helmet manufacturers. The pass/fail criteria should as close as possible be designed to measure the risk for a skull fracture and also a moderate to severe brain injury.

It could be argued that a pass/fail criteria only should focus on the brain injury as no risk for a skull fracture is seen in accident statistics and has again been demonstrated in section 3.1 of this report, when the bicycle rider is equipped with a helmet. However, the both test linear and tangential should use a global injury criteria like maximum amplitude criteria (250G) or HIC to keep the current protection level for skull fracture.

The pass/fail criteria should then in addition measure the risk for a brain injury. There are several candidates. The candidates could be divided into global kinematic criteria that are calculated directly from the 6DOF acceleration pulses, or tissue level injury criteria such as local strain or axon strain in the brain, calculated with the 6DOF acceleration pules and using a detailed FE model of the human head as illustrated in Figure 24.

The SUFEHM exposed in section 4 is one potential injury risk assessment tool. It is obvious that other head FE models could be used with the same method as presented in Figure 24. Further studies are under progress and will be needed in order to harmonize these results.

In order to go beyond a simple maximum acceleration, or HIC or even combination of several tentative kinematic criteria such as BrIC, RIC or PHRIC, for which the correlation with the occurrence of injury has not been demonstrated, but also in order to take into account the complex 6D loading for the assessment of the brain injury risk, the proposal made in the present project is to implement a coupled experimental versus numerical method as illustrated in Figure 13. In this approach the experimental linear and rotational head accelerations will constitute the inputs which will drive the head FE model, in charge of the latter to compute the brain strain or axon strain related to neurological injury. Post-processing tool that exist for SUFEHM and that will be developed shortly for other FE models, will extract the maximum brain strain from the FE simulation file and calculate the brain injury risk according to the injury risk curve relevant to the head model. This methodology can be applied for linear impact as well as for tangential impacts as suggested in *Willinger et al 2014* and under discussion within TC158 WG 11 working group, as it is the only which permits to take into account the multidirectional brain loading in one single brain injury criteria.

So, the conclusion is that there exist two candidates methodologies that can be used to measure the protective properties in a helmet in a shock absorption test, one based on global kinematic parameters, the other on tissue level brain loading. Both still need consolidation and international harmonization.

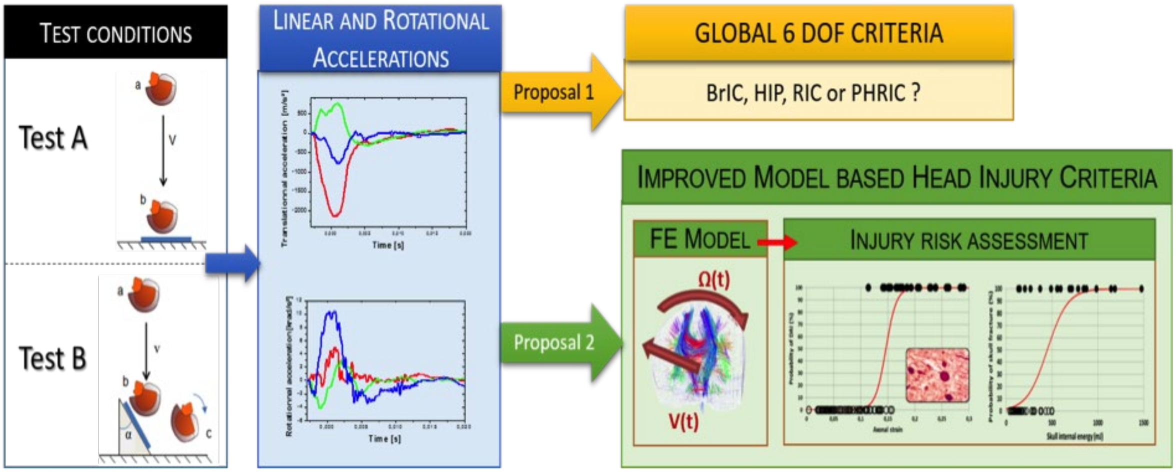


Figure 24: Illustration of the coupled experimental versus numerical head impact test method based on head FE modelling

6 WG3 conclusion for a new helmet test method

WG3 propose to change the way helmets are certified. The proposed changes to the current regulation test EN1078 and EN1080 is based on real accident data and new biomechanical knowledge. Single accidents are the most common followed by accidents to a car. In both accident types pure radial hits to the helmet and the head are rare. It is therefore suggested to complement the existing radial chock absorption test in EN1078 and EN1080 (called Linear impact Test) with an angled impact (called tangential impact test). The proposed impact speed should be 6.5m/s and the impact angle 45degrees for the tangential test.

The proposal is also based on the ongoing work within CEN TC158 – Working Group 11. The ongoing work in TC158 aims to define a test method to measure the energy absorption in tangential impacts with the demands to be robust, simple and cost effective. The test method proposed is therefore designed to use existing test machines for helmet shock absorption tests. The existing test machines are just slightly modified.

WG3 propose the chock absorption test to include both a vertical drop test from 1.5m to a flat anvil (Linear impact test) and a vertical drop from 2.2m to a 45degrees angled anvil (Tangential impact test). Both tests should use a free falling Hybrid III head form without a neck. Existing drop towers from CADEX in Canada or Ad Engineering in Italy can be used. A proposal for three impact sites inducing rotation along the three reference axis is presented in Figure 9.

The HIII family needs to be extended with size 56cm and 62cm in circumference as shown in Figure 5. The head forms should be instrumented to measure both the three linear accelerations and also the three angular accelerations around all axis, both for the linear and the tangential impact test.

It is believed from WG3 that both a global injury criteria and an FE based injury assessment tool could be used to measure the brain injury risk. Important is that the value defining the pass/fail level is based on robust injury risk.

It is believed that there is a need for two separate injury metrics for the linear impact. One for skull fracture and one for brain injury. For skull fracture the levels used today using the linear acceleration (250G) could still be used or HIC should be introduced. For brain injury it is believed that the pass/fail criteria used for bicycle helmets should measure the risk for a moderate or severe brain injury. For brain injury a kinematic criteria structured as BrIC, RIC or PHRIC could be used. Though more work is needed to set the exact limit. Also, there might be necessary to define one pass/fail limit per impact direction. In order to go beyond a kinematic criteria such as BrIC, RIC or PHRIC, the proposal made in the present project is to implement a coupled experimental versus numerical method as illustrated in 25. For this method the 6D accelerations curves are introduced into a head FE model and the brain injury risk is assessed automatically via a head FE model, by a post processor tool. It is considered that this methodology already applied in very recent bicycle helmet consumer tests in Sweden, Germany and France is the only which permits to consider one injury criteria under a complex 6DOF head loading.

The present proposal should not be seen as a standard but as an advanced scientific helmet test method proposal based on real world accident and biomechanical investigation which should be considered within the different standard bodies.

7 Future studies

WG11 within CEN TC158 are working to finalize the specification for a new test method. There are still details to be solved as to define a calibration tests of the instrumented HIII head forms. The spread and the variation between helmets and different test labs needs to be understood as well as the pass/fail criteria. Benchmark studies between different FE head models are under progress and will be needed in order to specify head FE models and to harmonize model based head injury criteria. Further research in the field will also be organized within EU projects (HEADS, MOTORIST, SmartHelmets, Safe2Wheelers) and national projects in Swede, Germany and France.

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Appendix A

FE analysis of the importance of a neck in angled helmet impacts.

Background

In order to design an oblique test method there are questions if the neck will affect the measured translational and rotational accelerations in the dummy head. It is clear that the head is restraint by the neck and at some time will rotate around a point in the neck or even lower down in the thoracic region.

Earlier studies like the COST 327 study has shown that the amplitude of the rotational acceleration is affected by the neck. Helmeted full body Hybrid III dummies was dropped to an angled surface and compared to free falling helmeted head forms. The result showed that the rotational acceleration differed in amplitude by about 20%.

It seems like the neck and the rest of the body do affect the amplitude of the rotational acceleration. It is however a complex procedure to test helmets using complete crash test dummies. The question is if a test can be designed without a representation of the cervical spine in order to improve the helmet safety.

In order to understand how the neck will affect the results during impact, a numerical study was therefore performed.

The aim of this study is to evaluate how the translational and rotational acceleration measured in a Hybrid III dummy head is effected by the inertial properties of a neck.

Method

The Finite Element (FE) method was used to analyze the oblique impact between a helmeted Hybrid III head and an angled impact surface. The vertical velocity was 6.5 m/s and the impact angle was 45 degrees.

All simulations were made using the Hybrid III head (Fredriksson 1996). One configuration was without neck. This was compared with simulations where the Hybrid III neck was attached to the head and compared with a model of the human cervical spine (Halldin et al 2001, Brolin et al. 2005). In the case using the human neck the skull base from the neck model was merged to the aluminum part of the HIII head modeled as rigid. The total added weight of the HIII and the human neck was 5kg each.

Helmet

The helmet is a conventional designed motocross helmet. The main difference in the helmet design compared to a conventional full face motorcycle helmet is the design of the cheek part. The shell was modeled as a glass fiber reinforce shell (*MAT_COMPOSITE_FAILURE_SHELL_MODEL in LSDYNA). The liner consists of three different parts and was modeled as EPS liner with densities 35, 50 and 70 kg/m^3 (*MAT_BILKHU/DUBOIS_FOAM in LSDYNA).

Simulation set up

Five different model configurations were compared, two using the HIII neck, two using the KTH neck and one without neck. The base of the neck (representing T1 vertebrae) was constrained from rotation and restricted to translational motion along the Z-axis, see Figure A1.Three different impact directions were studied, lateral, pitched and backward, Figure 1. The impact directions was chosen to generate rotations around all axis (X, Y, Z) in impact situations that is believed to be relevant. Impact to the crown is rare in reality why the lateral impact is questionable. However, it will evaluate the helmet in a lateral impact direction with rotation around the X-axis. The coefficient of friction between the helmet and the plate was set to 0.5.

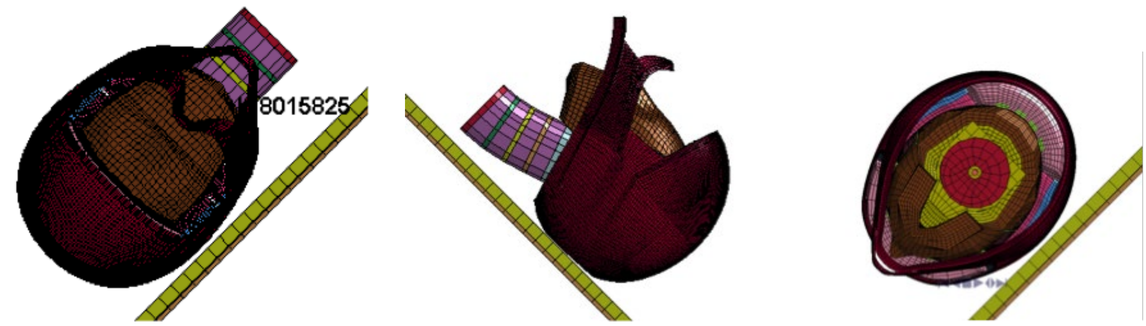

Figure A1: Left: side impact resulting in rotation around the X-axis. Middle: Back impact resulting in rotation around the Y-axis. Right : Back impact, resulting in rotation around the Z-axis.

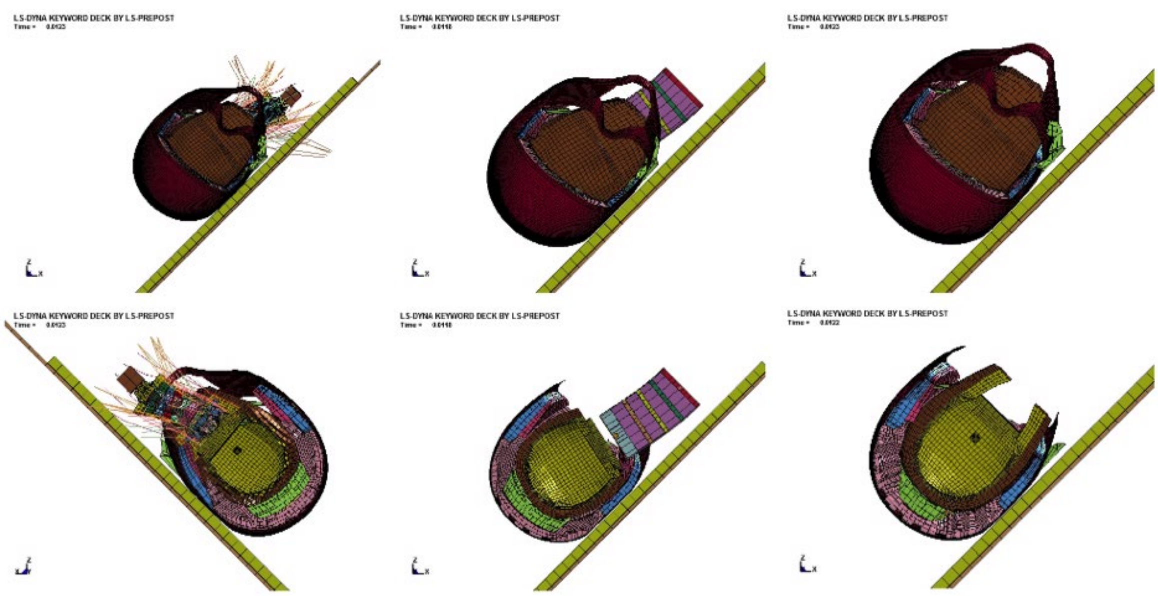

Figure A2: Left : Simulation with human neck model, Middle : HIII neck model and Right : No neck. Simulation of a side impact resulting in rotation around the X-axis.

Results

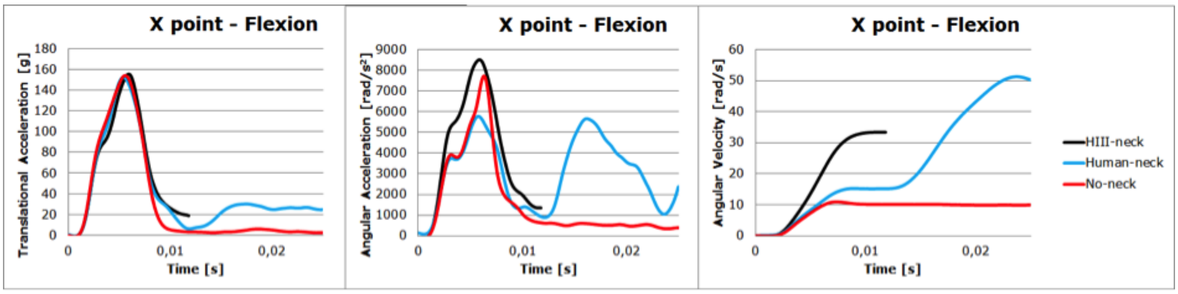

Figure A3: Translational acceleration, rotational acceleration and rotational velocity for X-rotation impacts. The results for the HIII neck is shown in black, Human neck model is shown in blue and the results for the No-neck configuration is shown in red.

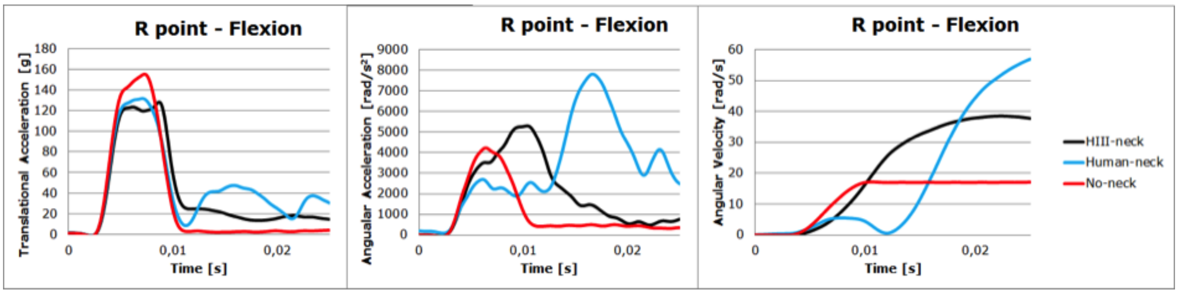

Figure A4: Translational acceleration, rotational acceleration and rotational velocity for Y-rotation impacts. The results for the HIII neck is shown in black, Human neck model is shown in blue and the results for the No-neck configuration is shown in red.

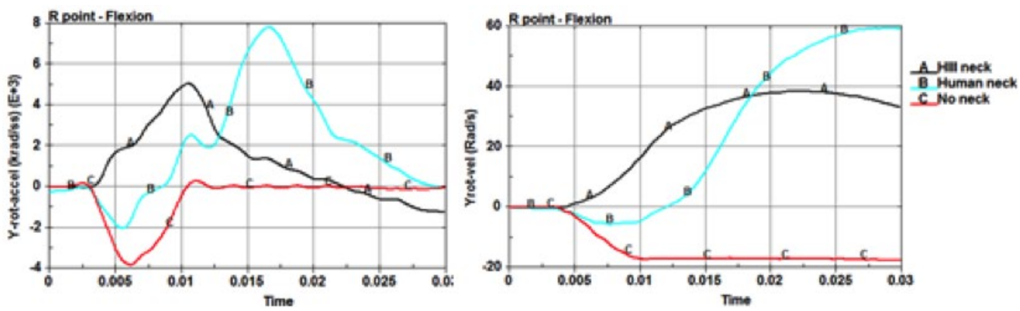

Figure A5: Y-component rotational acceleration and rotational velocity for the Y-rotation test.

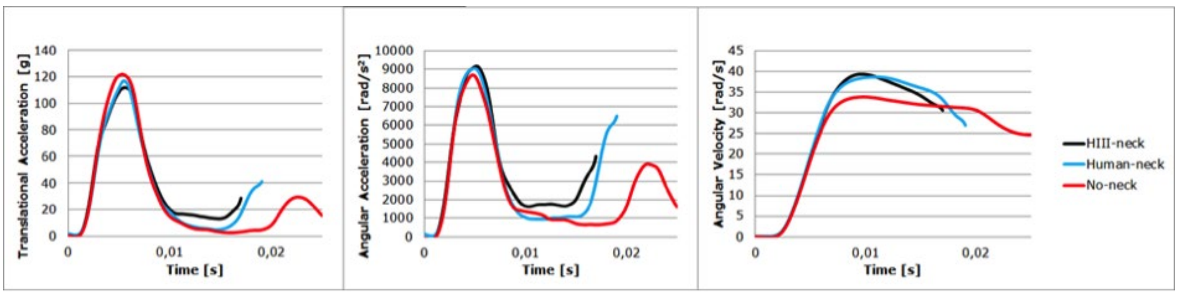

Figure A6: Translational acceleration, rotational acceleration and rotational velocity for Z-rotation impacts. The results for the HIII neck is shown in black, Human neck model is shown in blue and the results for the No-neck configuration is shown in red.

The Results in X-rotation (Figure A3) show that:

- The translational acceleration was similar for the three configurations (HIII-neck, Human neck and No neck).
- The resultant angular acceleration showed similar results for the HIII neck and the no/ neck configuration.
- The angular velocity showed that the No-neck configuration was closer to the human neck than the HIII neck.
- The simulation with the HIII neck stopped after 120ms due to contact problems in the simulation.

The results in Y-rotation (Figure A4) show that:

- The translational acceleration was similar for the HIII-neck and Human neck
- The angular acceleration around the Y-axis (ear-to-ear) and the angular velocity show large variations between the test configurations. The reason is that the HIII neck force the head to rotate in a non-human like way due to its stiffness as shown in Figure A5.

The results in Z-rotation (Figure A6) show that:

- The translational acceleration was similar for the three configurations
- The resultant angular acceleration and the angular velocity showed similar characteristics for the first 10 ms.

Conclusion

It could be seen that there are significant differences between the HIII neck and a human neck model for two of three impact directions. The results showed that for some impacts that the simulations without a neck could be more human like than impacts with the HIII neck. However, it should be stressed on that neither the FE model of the HIII neck nor the human neck model have been validated for these specific impact conditions. Further studies are needed to understand all aspects of the neck as a boundary for the head.

Appendix B

Six 'Back-on-Track'-brand equestrian helmets, size L were tested in the test rig at KTH, Sweden [1]. The helmets were dropped at vertical speed of 3.8m/s and a horizontal speed of the movable plate of 6.5m/s, giving a resultant speed of 7.1m/s and an impact angle of 30 degrees. The helmets are normally equipped with a low-friction layer between the EPS liner and the plastic outer shell. In three helmets, the low-friction layer was removed and the outer shell glued to the EPS liner (Helmet A). The helmet including the low-friction layer was named B. The mass of the helmet with and without the low-friction layer was controlled to be the same for Helmet A and Helmet B. The helmet was placed on a 50% HIII headform equipped with 3-2-2-2 accelerometer array [1]. Three tests were performed for each helmet A and B. First, the helmets were tested by simply attaching the helmet to the HIII headform. Second, the two helmets were tested by covering the HIII headform with ladies' stockings. Lastly, the HIII headform was covered with a wig.

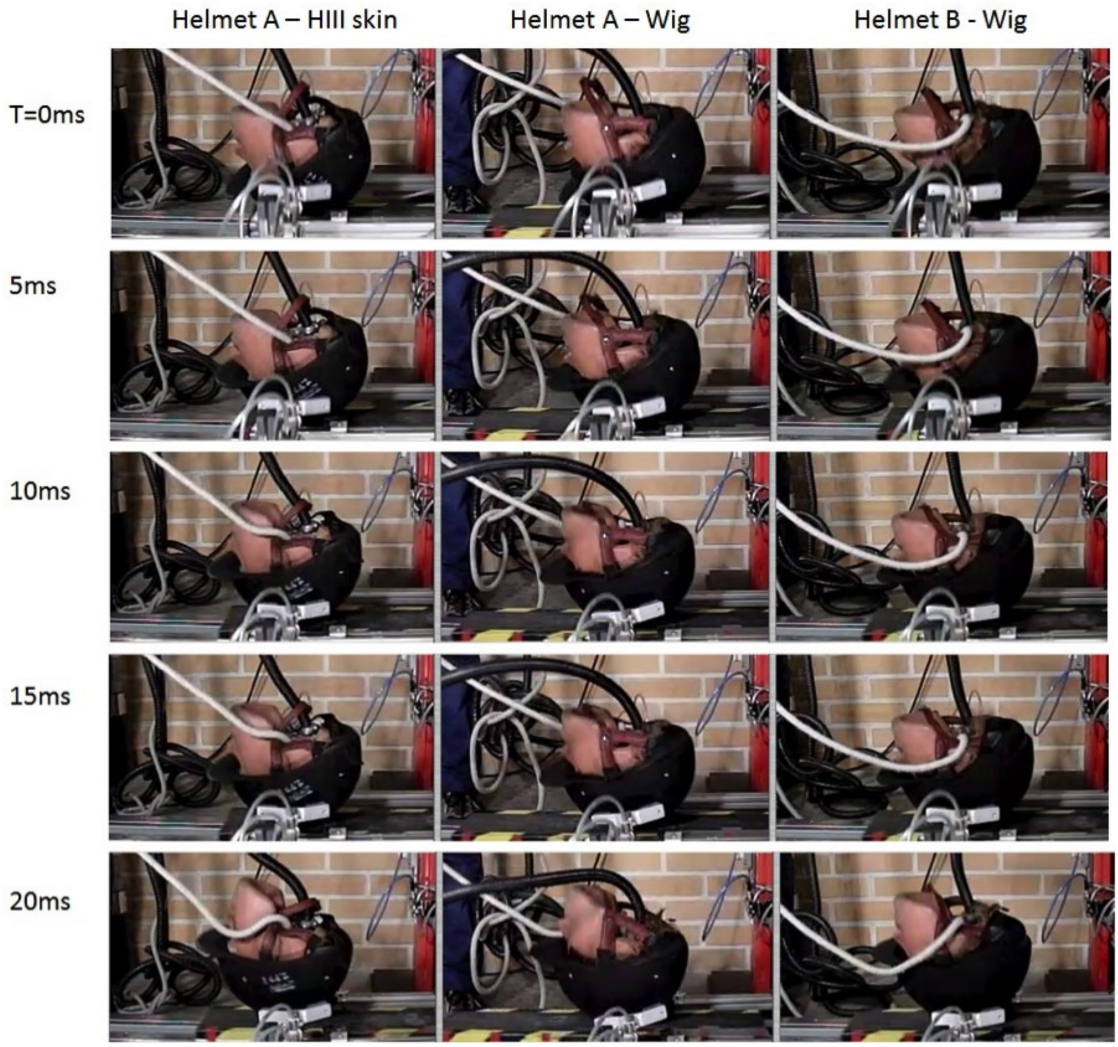

Fig B1: Shows images (600fps high speed camera) from the experimental tests of equestrian helmets. The column to the left shows the test of Helmet A without stocking or wig. Middle column shows helmet A with wig and the right column shows Helmet B with Wig.

The result showed that in comparison to having the HIII head form covered with stocking, testing helmets with a wig resulted in a reduction of the angular acceleration of 17% and a reduction in angular velocity of 4%. It could then be argued that if the oblique test is done covering the headform with a wig, then it does not matter what is done with the helmet since it slides on the wig in a typical oblique impact situation in any case. However, as presented in Figure B2, another test was conducted using a helmet with a different structural design (Helmet B). With helmet B, the angular acceleration was reduced by 42% and the angular velocity by 34%. Therefore, it can be concluded that even if the helmet slides on a human head due to hair, it is still possible to design helmet capable of absorbing more energy driven by more realistic test methods.

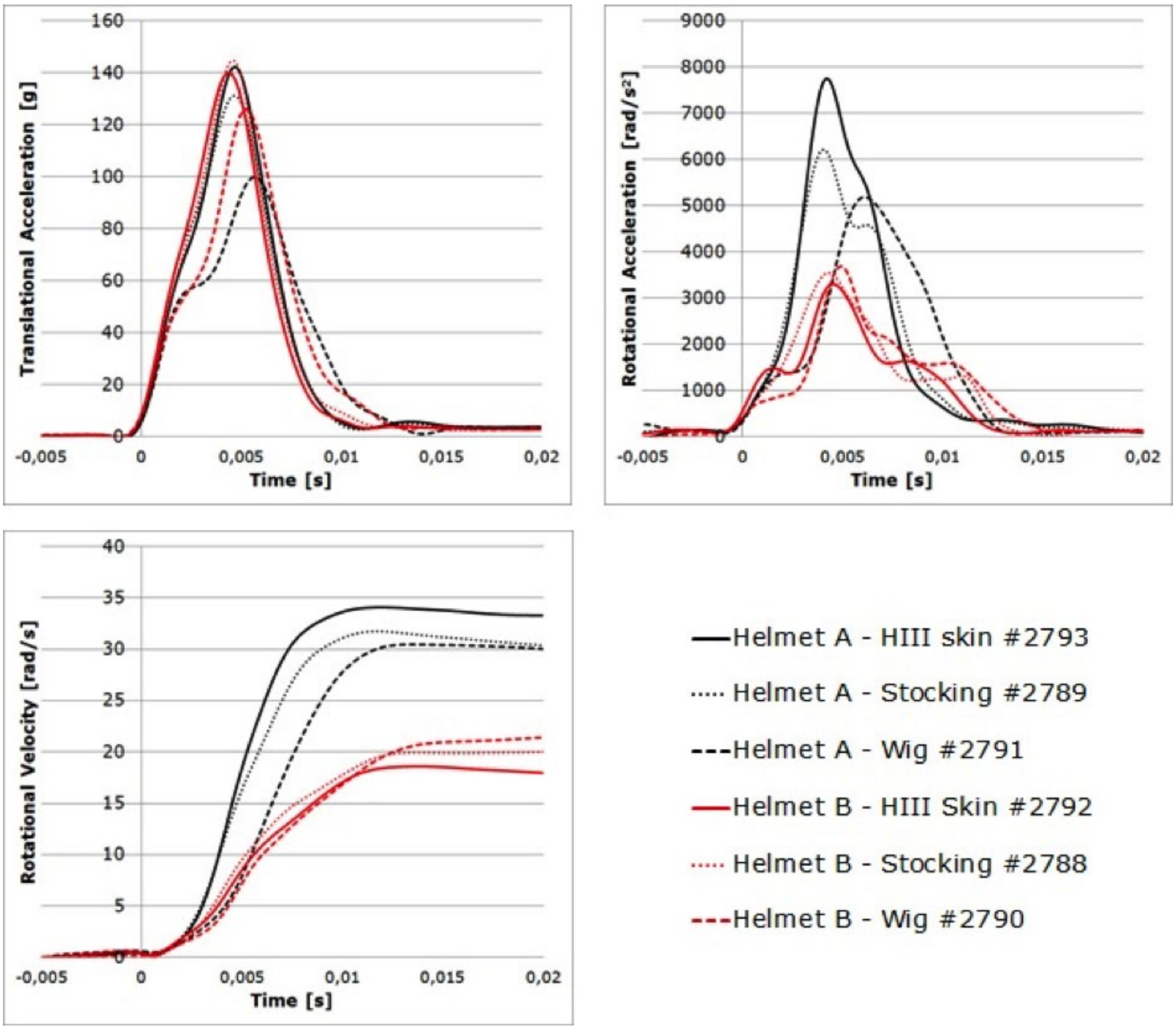

Fig B2: The resultant translational acceleration, resultant angular acceleration and the angular velocity as function of time is presented for the oblique impact tests on the equestrian helmet with and without the wig.

Appendix C

Another experimental study was performed where 15 helmets (Biltema Skate helmet bought in Sweden) were dropped in the KTH oblique test rig (Vertical drop to a movable plate). Vertical drop speed Vv=3.8m/s and the horizontal speed of the plate Vh=6.5m/s. The impact direction is shown in Figure 4. Five different impact points where as shown in Figure 5 (3 helmets per impact point). The results from the test shown in Figure 6 show that the measured kinematics is more sensitive for shifts in the X-direction than in the Y-direction.

The large difference in measured kinematics will have implications on the design of a test method. In the current test standard specification it is stated that the test engineer can choose the impact point on the helmet within specified boundaries. This result in that the test engineer can choose the impact location from his or her experience. If the test point then needs to be specified as in the motor cycle test standard ECE 22.05 then the helmet might be optimized only for these specific test points. More work is needed to solve this potential problem due to test point sensitivity.

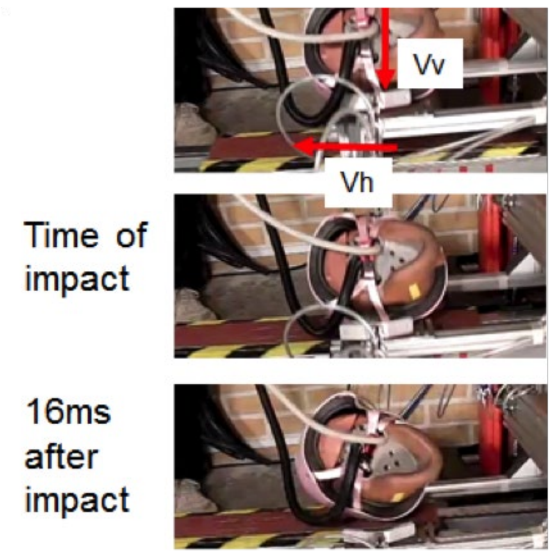

Figure C1: Pictures showing the helmet impacting the movable plate before, at time of impact and after impact..

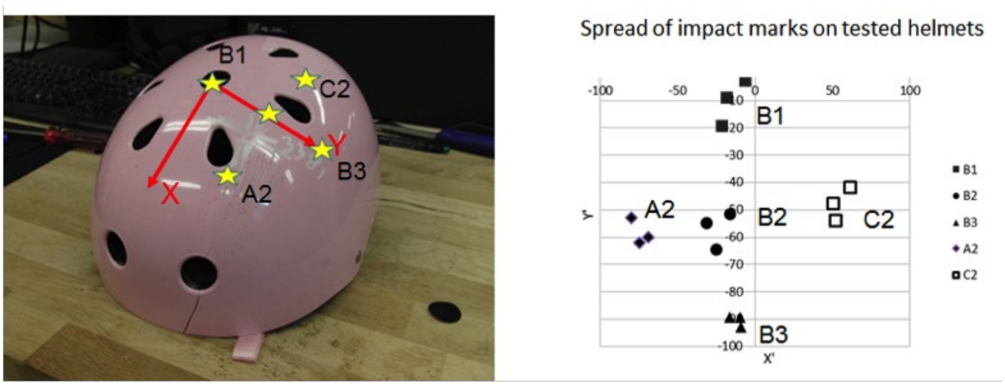

Figure C2: Impact points on the helmet.

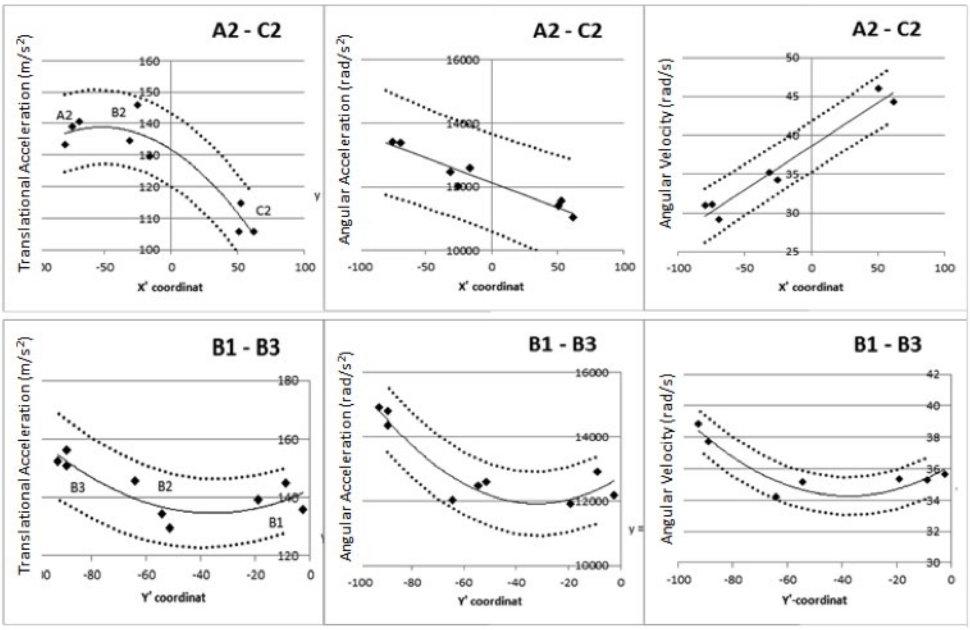

Figure C3: Showing the change in amplitude of measured kinematics.