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# UNISTRA-HEXR

**Expertise on Helmet Evaluation**

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## 1 Introduction

The objective of the present document is to proceed to a comparative analyze of the brain injury risk computed with HEXR helmeted with other helmets outcome under oblique head impact. In chapter 2, existing brain injury criteria based on linear acceleration are reported and limitations are expressed.

Chapter 3 gives a presentation of the Strasbourg University Finite Element Head Model (SUFEHM), in terms of mechanical properties, validation, as well as head injury criteria derived from extensive head trauma simulation.

This numerical head injury prediction tool has been used in order to assess the brain injury risk based on experimental headform acceleration data recorded under oblique impact. A total of three different helmet are considered for 3 oblique impacts according and leading to rotation around X, Y and Z axes. Each impact configuration is reproduced 3 times for reproducibility check. In chapter 4 the experimental versus numerical test is briefly reported and the results obtained with HEXR helmet are recalled. method and the results obtained with the HEXR helmet. Finally the comparative analysis of the HEXR helmet with other bicycle helmets is reported.

It is here gently recalled that the present report addresses the assessment of existing helmet performance compared to other helmets and is not a collaborative project on helmet design or joint helmet development. Also in this report no information regarding other helmets was foreseen. Therefore the comparative analysis will be limited to the brain response and brain injury risk obtained with the HEXR helmet compared to other helmet under three oblique impact conditions.

## 2 Existing Head injury criteria based on linear acceleration: presentation and critics

### 2.1 Introduction

Over the past forty years, a slant has been put by the biomechanical research on the understanding of 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 brain tissue level tolerance has very early been investigated, the global acceleration of the impacted head and the impact duration are usually being used as impact severity descriptors. This section gives an overview of head injury criteria expressed in terms of global linear head acceleration:

### 2.2 Maximum Resultant Head Acceleration

A head injury criterion 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 many 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, impact orientation, the type of injuries and the rotation...

A variation of this criterion is  $A_{3\text{ms}}$  value which refers to the maximum deceleration that lasts for 3ms. Even if a "kind" of time duration is taking into account, same limitations can be done for this criterion. The  $3_{\text{ms}}$  criterion is based on the WSTC.  $A_{3\text{ms}}$  should not exceed 80g (Got et al., 1978).

### 2.3 Wayne State tolerance curve

The Wayne State Tolerance Curve is considered to be the foundation of research on human head injury criteria. This curve evolved from the work of Lissner et al. (1960), Gurdjian et al. (1945, 1961) and Patrick et al. (1963), and gives the tolerable average acceleration in A-P direction (Anterior-Posterior) as function of the 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 used in the development of the curve which was obtained from different experiments with cadavers, animals and volunteers. That substantial acceleration causes injury over short durations, while smaller accelerations require longer duration to cause injury is an assumption fundamental to the curve formulation. The short duration part of the curve ( $2 < t < 6$  ms) was derived from cadaver tests in which skull fracture was chosen as injury criterion. 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 criterion 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 curve it was assumed that skull fracture and concussion correlate. Lissner et al.

maintained 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.

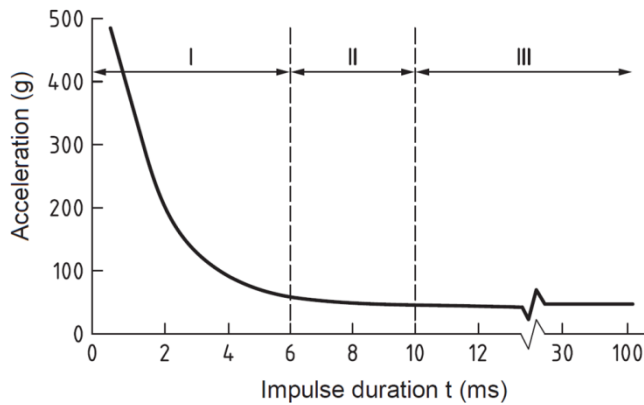


Figure 1. Wayne State Tolerance Curve The figure is divided into 3 parts:

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 (Beusenbergh 1991).

## 2.4 Head Injury Criteria, 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:

$$GSI = \int_T a(t)^{2.5} dt \quad \text{Where } T = \text{the total pulse duration, and } a(t) = \text{acceleration at the centre of mass of the head, as a function of time.}$$

This was described as the weighted impulse criteria for which a value of 1 000 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:

$$HIC \left[ \left( \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a_{res} dt \right)^{2.5} (t_2 - t_1) \right]_{\max} \quad \text{With: } t_1 \text{ and } t_2 [\text{ms}] \text{ any two points in time during any interval in the impact; } a = \text{resultant linear acceleration of the centre of mass of the head.}$$

After a number discussions and over years,  $t_1$  and  $t_2$  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 1 000 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 and currently head protection systems in the automotive domain consider injury curves issued by NHTSA and shown in figure 2.

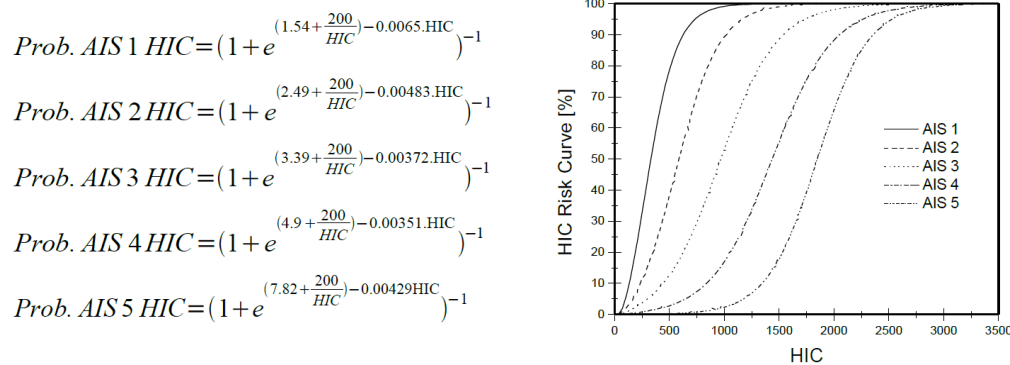


Figure 2. HIC injury risk curves according to AIS level

Critical aspect of HIC are that this metric is based on the linear accelerations of a one mass headform, HIC is based on skull fracture and not brain injury, HIC is not specific to direction of impact, and it also does not distinguish between injury mechanisms (skull failure, SDH, neurological injury). Further an essential limitation of HIC is that head rotational acceleration is not taken into account although rotation is known to be the primary cause for various types of traumatic brain injury, in particular acute subdural haematoma and diffuse brain injury [Adams *et al.*, 1983; Gennarelli *et al.*, 1987; Holbourn, 1943].

Moreover, Marjoux *et al.*, 2006 demonstrated that this criterion was poorly correlated with observed injuries. In framework of APROSYS SP5 project, the capability of HIC to predict injury was evaluated based on Strasbourg University 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 (SPSS software package), it appeared that HIC presents an acceptable correlation with severe neurological injury which means in most of the cases when victims died or in coma for a long time. Threshold parameter for a 50% injury risk obtained with this 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 is poor (Marjoux *et al.*, 2006).

## 2.5 Conclusion

Over the past years, several head injury assessment functions have suggested. Most of them are based on the Wayne State Tolerance curve. The most commonly acknowledged and widely applied head injury criterion is the HIC which is based on the assumption that the translational resultant acceleration of the head is a valid indicator of head injury. This criteria established in the 70s enabled vehicle safety to be improved. Nevertheless, it has shortcomings and does not take into account rotational acceleration and direction of impact. Furthermore, it is not clear how this injury criteria relates to skull or brain injury mechanisms.

Coming to the injury criteria (or pass/fail criteria) within standards, the choice of tolerance level depends on the headform, on the application, and on the level of accepted injury risk. The tolerance level for HIC in the motorcycle helmet standard ECE-R.22 is 2400 using a rigid

headform, whereas the tolerance level for HIC in the car crash standard FMVSS 208 [NHTSA, 1972] using a Hybrid III headform is 1000. Same maximum value (HIC=1000) is also required in Circular AC25.562.1b “Evaluation of Seat Restraint Systems and Occupant Protection on Transport Airplanes” calculated with a Hybrid II dummy. A number of standards also do not consider the impact duration and focus on a simple pass/fail criteria expressed in terms of maximum linear acceleration. This is the case for sport helmets (bicycle, ski, equestrian) in Europe with a limit at 250G and for the Motorcycle in US with a threshold of 350G.

Brain injury is reported to correlate with intra-cerebral 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. Advanced computational techniques led to more accurate and more detailed numerical models of the human head. These models, when extensively used under real world head trauma simulation can lead to tissue level injury criteria that take into account the 6D time evolution of the head loading.

In the next section a state of the art Finite Element Head Model is presented with relevant tolerance limits, ready to be used as a head injury prediction tool.

### 3 Presentation of SUFEHM

As the evaluation of a head protection systems needs also a proper estimation of brain tolerance limits also called brain injury criteria, the present section deals with the model based brain injury criteria established at Strasbourg University. Strasbourg University Finite Element Head Model (SUFEHM) is a numerical model of the human head with realistic brain and skull material laws (*Deck et al., 2008, Sahoo et al. 2013, 2015 and 2016*) and which permits the computation of the mechanical brain response to an impact.

For this head model, the geometry of the inner and outer surfaces of the skull was digitized from a human adult male skull to ensure anatomical accuracy. The main anatomical features includes the brain, brainstem, skin and cerebrospinal fluid (CSF), represented by brick elements, and the skull, face and two membranes (the falx and the tentorium) modelled with shell elements. The SUFEHM presents a continuous mesh that is made up of 13,208 elements, including 1797 shell elements to the compose skull and 5320 brick elements for the brain. The total mass of the head model is 4.7 kg. Isotropic, homogeneous and elastic mechanical constitutive material models were applied to each of the SUFEHM parts except for the brain and skull.

A linear isotropic viscoelastic law is affected to the brain according to (Eq. (1))

$$G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t} \quad (1)$$

Where  $G_0$ ,  $G_{\infty}$  and  $\beta$  represent the short-time modulus, the long-time modulus and the decay constant, respectively.

Mechanical parameters were identified from the experimental data on human brain tissue proposed by *Shuck and Advani (1972)* as well as in vivo based values from Magnetic Resonance Elastography (MRE) published by *Kruse et al. (2007)*, with following values:  $G_0 = 49 * 10^3$  Pa,  $G_{\infty} = 1.62 * 10^4$ Pa,  $\beta = 145$  s<sup>-1</sup>.

Validation of this head model was proposed by *Deck and Willinger (2008, 2009)* against local brain motion data from *Hardy et al. (2001, 2007)*, and intracranial pressure data from *Nahum et al. (1977)* and *Trosseille et al. (1992)*.

The skull model considers a composite material model which incorporates fracture (*Sahoo et al. 2013*). The skull was modelled as a three-layered composite shell representing the inner table, diploe and outer table of the human cranial bone with a thickness of 3mm for the diploe layer and 2mm each for the two cortical layers. To demonstrate the robustness of the skull model, various parametric studies were conducted and reported in *Sahoo et al. (2015)*.

The skull and brain mechanical parameters implemented under LS-DYNA are represented in Tableau 1. A detailed presentation of different parts of the SUFEHM is shown in Figure 1.

| SKULL MECHANICAL PARAMETERS                            |               |             |
|--|---------------|-------------|
| Parameters   | Cortical bone | Diploe Bone |
| Mass density (Kg/m <sup>3</sup> )                      | 1900          | 1500        |
| Young's Modulus (MPa)                                  | 15000         | 4665        |
| Poisson's ratio  | 0.21          | 0.05        |
| Longitudinal and transverse compressive strength (MPa) | 132           | 24.8        |



| Longitudinal and transverse Tensile strength (MPa) | 90  | 34.8                 |
|--|---|----------------------|
| BRAIN MECHANICAL PARAMETERS                        |   |                      |
| Mass density (Kg/m <sup>3</sup> )                  | 1040  |                      |
| Viscoelasticity                                    | $G_0 = 49 * 10^3 Pa$<br>$G_\infty = 1.62 * 10^4 Pa$ | $\beta = 145 s^{-1}$ |

Tableau 1. Skull and brain mechanical parameters of the SUFEHM implemented under LS-DYNA (Sahoo et al., 2013, Deck et al., 2008).

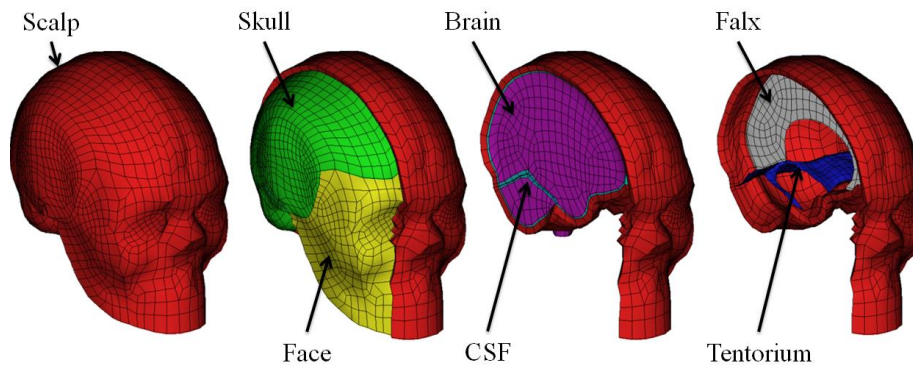


Figure 1. Illustration of the different parts of Strasbourg University Finite Element Head Model (SUFEHM), with 5320 brick elements of brain.

The proposed mechanical model of the head fulfilled typical requirements of state-of-the-art head models as long as stability and validations are concerned.

This model was used in order to derive tolerance limits to specific injury mechanisms. The main objective was to develop robust and accurate model based injury criteria to predict skull fracture and moderate diffuse axonal injuries (moderate DAI also called concussion). To do so, well-documented real-world head trauma cases collected from different existing accident databases and involving pedestrian, cyclists, motorsport, American football player and motorcycle accidents were simulated in order to compute the skull and brain mechanical response for the different head trauma. The correlation of these mechanical responses with the occurrence of a given injury permitted it to derive injury criteria for specific injury mechanisms.

A total of 85 well-documented head trauma cases were reconstructed numerically with the head model to develop a skull fracture injury risk curve. The proposed tolerance limit for 50% risk of skull fracture was associated with 453 mJ of skull internal energy calculated with the head model (Sahoo et al. 2016).

Further 109 real-world head trauma cases were simulated to develop a robust brain injury criterion in terms of intracerebral Von Mises stress to predict moderate DAI or short coma accurately. The head trauma modelling was performed in accordance with the victim's kinematic analysis. Based on an in-depth statistical analysis of different intra-cerebral parameters, it was shown that Von Mises stress was the most appropriate metric to predict moderate DAI. The proposed brain injury tolerance limit for a 50% risk of moderate DAI, which corresponds to a loss of consciousness (AIS2+) known to be reversible brain injury, has been established at 36 kPa.

Injury risk curves to predict probability of skull fracture by addressing skull strain energy and moderate brain injury by addressing brain Von Mises stress are illustrated in Figure 2

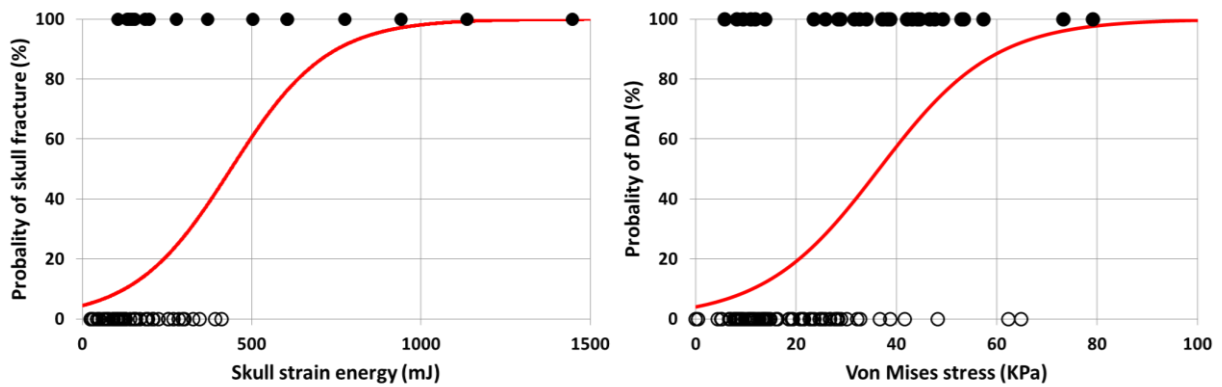


Figure 2 Injury risk curves to predict probability of skull fracture by addressing skull strain energy and moderate DAI (moderate diffuse axonal injuries which means loss of consciousness AIS2+, possible reversible brain injury) by addressing brain VonMises Stress.

## 4 Comparative analysis of results

### 4.1 Recall of HEXR helmets results

A total of three different HEXR helmets are considered for 3 oblique impacts at an initial speed of 6.0 m/s according to figure 4.1, and leading to rotation around X, Y and Z axes. Each impact configuration is reproduced three times as shown in Table 4.1

For each impact, the 6D headform kinematic has been recorded versus time and considered as initial conditions for the simulation of the brain response with SUFEHM model. This novel experimental versus numerical helmet test method is illustrated in figure 4.2. Finally table 4.2 reports the results in terms of mean brain Von Mises stress and injury risk, for the three impact conditions.

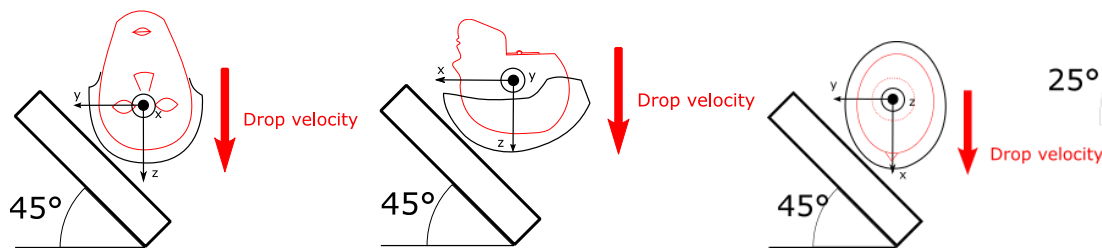


Figure 4.1 : Illustration of the three oblique impact conditions leading to rotation along X axis (XRor), Y axis (Yrot) and Z axis (ZRot)

Table 4.1 : Test matrix applied for the three available HEXR helmets.

| Helmet ID | First Impact | Second Impact | Third Impact |
|-----------|--------------|---------------|--------------|
| H1        | YRot         | XRor          | ZRot         |
| H2        | ZRot         | YRot          | XRor         |
| H3        | XRor         | ZRot          | YRot         |

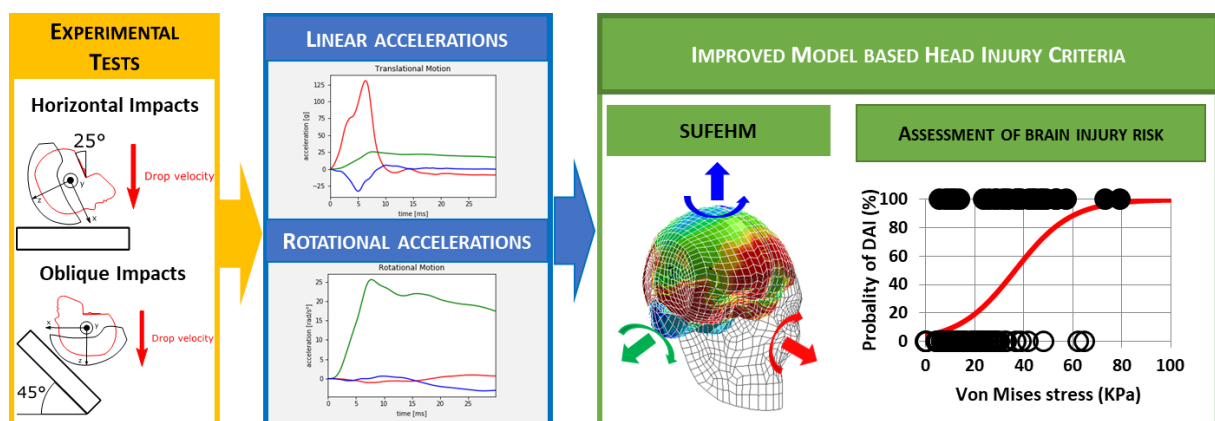


Figure 4.2 : Illustration of the experimental versus numerical test method. Experimental 6D headform response is introduced into the SUFEHM model in order to compute the brain Von Mises Stress and to assess the brain injury risk based on the injury risk curve.

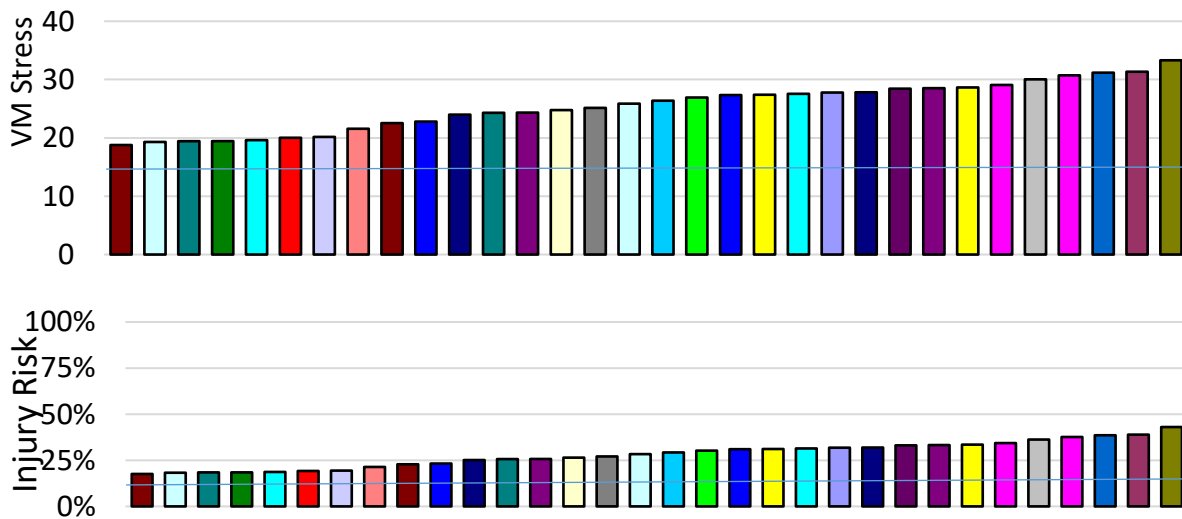
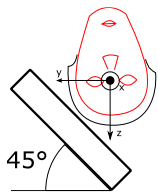
Table 4.2 : Summary of results obtained with the HEXR helmet in terms of mean brain Von Mises Stress (in KPa) and percentage of injury risk for the three impact condition.

| Impact | VMS  | R % |
|--------|------|-----|
| XRot   | 17.1 | 16  |
| YRot   | 28.6 | 34  |
| ZRot   | 23.3 | 24  |

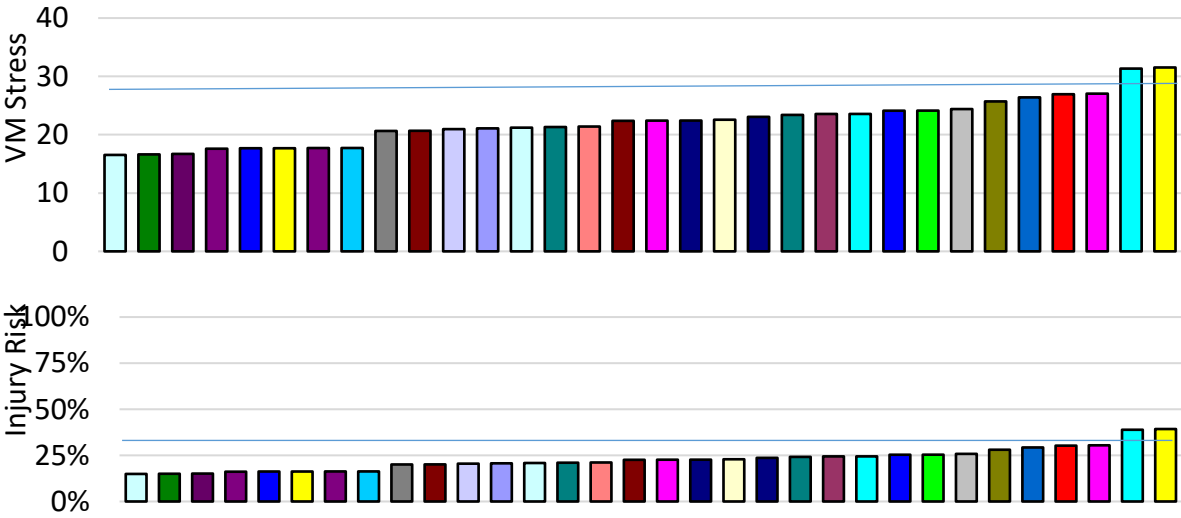
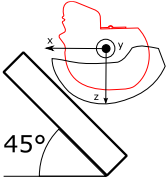
### 4.2 Comparative analysis

Results obtained with the HEXR helmet are compared to results obtained with a set of 32 existing bicycle helmets respectively under XRot, YRot and ZRot impact conditions. Comparison is expressed hereafter in terms of brain response i.e. mean Von Mises stress as well as in terms of risk of moderate AIS2+ brain injury. The values obtained with the HEXR helmet are indicated with the blue line.

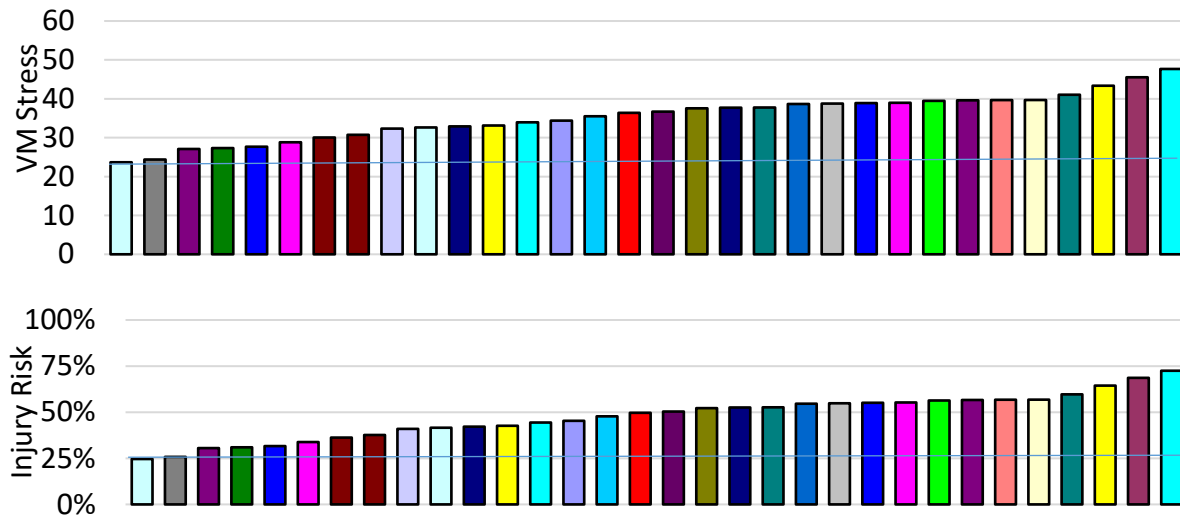
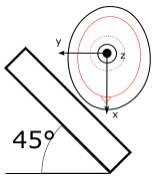
- i) Oblique impact leading to Rotation around X axis (XRot)



ii) Oblique impact leading to Rotation around Y axis (YRot)



## iii) Oblique impact leading to Rotation around Z axis (ZRot)



## 5 Conclusion

The head protection capability of the HEXR helmet under oblique impact conditions has been evaluated via a coupled experimental versus numerical helmet test method. The present report gives a synthesis of the novel method which considers the 6D headform kinematic for the numerical computation of the brain response using SUFEHM (Strasbourg University FE Head Model). The assessment of brain injury risk is conducted with a dedicated brain injury risk for moderate AIS2+ brain injury.

The oblique impact results were on average in:

- XRot-orientation 111G, 3.8krad/s<sup>2</sup> and 11krad/s in linear acceleration, rotational acceleration and rotational velocity.
- YTot-orientation 175G, 5.7krad/s<sup>2</sup> and 20krad/s in linear acceleration, rotational acceleration and rotational velocity.
- ZRot-orientation 117G, 6.6krad/s<sup>2</sup> and 11krad/s in linear acceleration, rotational acceleration and rotational velocity.

Under oblique impact leading to rotation around X axis and around Z axis the HEXR helmet presents exceptional protection capability with brain injury risk of respectively 16% and 24% when other helmets lead to risk typically over 25% (for XRot) respectively 50% (for YRot). This demonstrates that HEXR helmet is among the best helmets when considering the 32 shown in the present report.

Under oblique impact leading to rotation around Y axis (YRot), the HEXR helmet behave less good. Brain injury risk is about 34% when other helmets lead to a risk that is typically under 25%.

As a general conclusion, it appears that when evaluating the HEXR helmet under oblique impact and versus advanced model based brain injury criteria that consider moderate and reversible brain injury, also called commotion, the overall risk is 25%. It is recalled here that the minimum requirement suggested for future helmet standard is 80% risk of moderate brain injury.

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