Towards advanced bicycle helmet test methods

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ABSTRACT

This paper exposes a critical analysis of current bicycle helmet standard tests and proposes an advanced test method. Different key aspects are considered consecutively, i.e. the bicyclist's head impact conditions in terms of velocity vector, head form boundary conditions, the head form itself with its instrumentation, the geometry of the impacted surface, the head impact location and finally the head injury criteria. Based on in deep analysis of bicycle accidents it has been shown that a significant component of the head velocity vector at the time of impact is the tangential velocity. On the other hand it has been shown that the head boundary condition at neck level does not play a significant role in the head response following to impact. Therefore it is suggested to consider tangential impacts against a helmeted Hybrid III dummy head, eventually connected to a Hybrid III neck. A critical analysis of the geometry of the impacted surface and the temperature at the time of accidents will be presented as well.

The current ISO head form presents a rigid contact surface with the helmet liner which is quite fare from human scalp-helmet interface. Therefore it is recommended in this proposal, to use a head form with a skin such as the Hybrid III dummy head. This improved head surrogate also presents the advantage to be fitted easily with rotational accelerometers, and to be connected to the Hybrid III neck without further modifications. Finally this head form presents mass and inertia properties much closer to the human head as ISO head form does.

The location of head impact in the current standard presents two key issues: The difficulty to guarantee an impact in line with the head form center of mass, and the recommended test line which excludes impacts to the temporal region which is often impacted in real world accidents. It is therefore suggested to prescript specific impact points in a similar way as for motorcycle helmets.

A final and acute issue with current bicycle helmet standard test is the head injury taken into consideration and related to results from the 1950's, as the threshold is still expressed in terms of acceleration amplitude and duration. In order to take into account the linear and rotational acceleration of the head after impact as well as improved model based head injury criteria, a coupled experimental versus numerical method is suggested in order to assess the head injury risk. In this method the helmeted head form is impacted in the previously descript impact conditions and the six head form acceleration versus time curves are implemented into a FE head model in order to compute the intra-cranial head response and to compare it to the model based head injury criteria. It is believed that the proposed approach would permit the evaluation and optimization of bicycle helmets against biomechanical criteria and under realistic impact conditions.

Keywords: bicycle helmet, test method, impact conditions, head injury criteria

1 INTRODUCTION

It is well known in the scientific community that head rotational acceleration is a critical head loading which can lead to brain injury. Concerning neurological injuries, Holbourn (1943) [1] suggested 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. In 1967, Ommaya *et al.* [2] 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 40 000 rad/s² will have a 99% probability of producing concussion which was expect to corresponds to an angular acceleration of 7 500 rad/s² for human.

Ommaya *et al.* (1968) [3] 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. In a study based on primates, Gennarelli *et al.* (1982) [4] proposed that a rotational acceleration exceeding 175 000 rad/s² would produce SDH in the rhesus monkey. With the objective to investigate the influence of the head rotational acceleration on the intra-cerebral mechanical parameters under accidental head impact, a total of 69 real world head trauma were simulated with and without considering the angular rotation by Deck *et al.* 2007 [5]. The numerical simulation of these head trauma by considering linear and rotational acceleration on the one hand and linear head acceleration only on the other hand permitted it to demonstrate and to express quantitatively the dramatic influence of the rotational acceleration on both intra-cerebral loading and brain-skull relative motion, supposed to lead respectively to neurological injuries and subdural haematoma respectively. In this study the effect of angular acceleration was found to increase the intra cerebral shearing stress for all accident cases considered of about 50% whatever the impact severity was. Kleiven *et al.* 2007 [6] as well as Zhang *et al.* 2001 [7] demonstrated that the angular kinematics of the head was the most important factor in determining the brain strain, based on numerical simulation of real world head trauma. More recently Takhounts et al [8], [9] used a head FE model in order to establish a head injury criteria for rotational acceleration called BRIC.

In parallel with the demonstration of the critical role of head angular acceleration in brain injury, a number of studies focussed on the head kinematics in real world accident in order to demonstrate that a tangential loading of the head does exist in addition to the normal impact velocity. Mills *et al.* (1996) [10] showed that oblique impacts are the most common situations in motorcycle crashes. More recently Bourdet *et al.* (2011, 2012) [11], [12] quantified the head rotational acceleration due to the tangential component of the head impact, by reconstructing real world and virtual motorcycle as well as bicycle accidents.

Despite this widely recognized understanding of head rotational loading and effect of the induced rotational acceleration to the brain, no head protection standard is currently considering head rotational acceleration. Only ECE R2205 EU [13] motorcycle helmet standard considers a tangential impact condition but helmet evaluation is limited to the recording of the tangential force which informs about friction characteristics of the helmet but not about its protection capability against tangential impacts. One possible reason for the current situation beside the increased complexity and cost of the test device itself is that no accepted head rotation threshold has been established yet. A number of maximum head rotational accelerations have been proposed in the literature [10], [14]–[16] but none of them consider the time evolution of this parameter. Moreover, it is obvious that the maximum head rotational acceleration is a function of rotation axis and combined linear acceleration, two aspects which are not taken into account in existing proposals. To the author's opinion, the only way to integrate the complexity of brain geometry and brain material properties is to progress towards tissue level brain injury criteria as proposed in existing FE model based head injury criteria [6], [7], [17]**.** In 2004 Deck demonstrates that helmet optimization strongly depends on the head substitute and injury criteria taken into account. More recently first attempts were made (Tinard *et al.* 2012 [18]) in order to optimize new helmets against biomechanical criteria by coupling the human head model to a helmet FE model. Advanced model based head injury criteria have also been suggested in recent attempts to improve bicycle helmet test methods (Deck et al 2012) [19].

In the domain of bicycle helmet evaluation Milne et al 2012 and 2013 [20] suggested a new helmet assessment method using model based head injury criteria under both linear and tangential impact conditions, exactly as Hansen et al 2013 [21] in the context of the development of an advanced 'honeycomb" bicycle helmet. In a similar way, but in the context of hockey helmet evaluation Post et al. 2013 [22] suggested to impact a helmeted Hybrid III head neck system and to introduce the linear and rotational acceleration into an existing head FE model in order to assess the injury risk.

In order to progress in the field of helmet protection against tangential impacts a number of attempts were proposed in the literature. Aldman *et al.* 1970 [14] dropped a helmeted headform fixed to a dummy neck against a rotating steel disc. In 2001 Halldin *et al.* [15] designed a new oblique impact test for motorcycle helmets based on an instrumented free Hybrid III dummy head dropped vertically against a horizontally moving plate. More recently Pang *et al.* 2011 [16] published a novel laboratory test in order to investigate head and neck responses under oblique motorcycle helmet impacts using a mobile anvil. This proposal is based on a test rig considering a helmeted Hybrid III head fitted to the Hybrid III neck itself fixed to a 20 kg mass which drops against a sliding plate. This test permits the recording of the head kinematics as well as neck loading. Reported test conditions are characterized by a 5.4 to 7.7 m/s impact velocity and a plate speed adapted to provide a 45 to 90° impact angle. Impact directions are either frontal or lateral.

Concerning Hokey helmets, Gerberich *et al*. (1987) [23] and Flick *et al*. 2005 [24] investigated hokey head trauma and reconstructed experimentally typical impact conditions applied to the Hybrid III head and neck system. It was shown that both linear and rotational head accelerations are significant and can potentially lead to brain injury as long as head injury criteria proposed by Zhang *et al.* 2001 [7] are concerned. More recently, Rousseau et al. 2009 [25], [26] developed a hokey helmet test bench where the helmeted Hybrid III head was fixed to a Hybrid III neck and impacted frontally or laterally with an impactor. Linear and rotational head acceleration in the range of 100 to 120 g and 3 to 6 krad/s² were recorded respectively. This method was applied to helmet material evaluation and showed that specific helmet structure can have very different outcome in terms of linear versus rotational head acceleration. In addition, Rousseau et al. 2009 [27], investigated the influence of the impact point deviation in regards to the center of mass as well as the neck rigidity on the linear and rotational head acceleration. This study demonstrated that head rotational acceleration is very sensitive to impact position relative to the center of mass and that the increase of neck rigidity leads to a head rotation decreasing. Moreover, Walsh 2010 [28] investigated the effect of impact direction in a 5- 15 ° range on Hybrid III head kinematic, and it appeared that this parameter influences much more the rotational response as the linear one. In order to further investigate these aspects Walsh *et al*. 2009 [29], investigated helmeted Hybrid III head kinematic when fixed on Hybrid III neck and impacted at a number of points around the head and by considering several impact angles. First result is that even when directed along the center of mass, rotational acceleration can be as high as 10 krad/s². More generally, it was shown that impact angle influences significantly both linear and rotational acceleration. Finally Walsh *et al*. 2011 [30] consolidated these results with 20 further impacts and demonstrated that highest linear acceleration was obtained for radial directed impact for all impact points. In this study the authors plotted linear versus rotational acceleration for all impacts. Pure correlation was fund (R^2 = 0.4) demonstrating that both injury parameters must be recorded during test as one cannot be estimated means the other.

As long as bicycle helmets are concerned no improvement of current standard tests has been proposed to the author's knowledge. Therefore the present paper' objective is to present a proposal on the improvement of the different aspects of bicycle helmet test method.

Presented is a global critical analysis of the existing EN 1078 bicycle helmet test method and the proposal of its evolution. A total of four separate aspects will be discussed, *i.e.* head impact conditions, head substitute, head impact location and head injury criteria.

2 HEAD IMPACT CONDITIONS

In the current standard test, impact conditions are characterized by a linear velocity of 5.42 m/s and 4.57 m/s, two temperatures (-20°C, +50°C) and one wet (+20°C) condition. The helmeted head form impacts two anvils, flat at 5.42 m/s and curbstone at 4.57 m/s and it is well known that the head is free at neck level for EN1078 standard.

If the head initial velocity seems to be reasonable in regard of real world accident situation, and fall alone simulations, the fact that this velocity has only a normal component is not acceptable. It has been demonstrated by (Bourdet et al 2012 [10]) that a significant tangential velocity exists which leads to head rotational acceleration in addition to the linear acceleration. Coming to the temperature it is important to mention that very few bicyclist's accidents occur at temperatures as low as -20 °C or at +50°C. Within COST action TU1101, it has been shown, based on 7180 real world bicycle accidents from GIDAS, that 92.6% of the accident occurred under dry weather conditions and that 96.8% happened at temperatures between 1 and 30 °C (figure 1). It is therefore no longer acceptable that helmet optimization includes material behavior at extreme low and high temperatures. In a similar way accident analysis shows that bicyclist heads only rarely impact curbstone and that the most frequent impacted surfaces are clearly flat rigid surfaces. Within COST T1101 and based on the GIDAS database only 4.4% of the cases revealed a head impact against an angular surface.

Following to the previous critical analysis of head impact condition, the hereafter proposal is made. First, just two temperatures (0°C and +30°C) and the wet condition at +20°C could be considered. In addition it is suggested to cancel the impacts against the curbstone and to maintain only impacts against flat anvils. No change is suggested for the linear impact velocity which could remain at 5.42 m/s. However, based on real world accident analysis and simulated accidents it is proposed to include three tangential impacts, characterized by a velocity of 6.5 m/s against a 45° inclined anvil. The impact points will be further defined in the following sections.

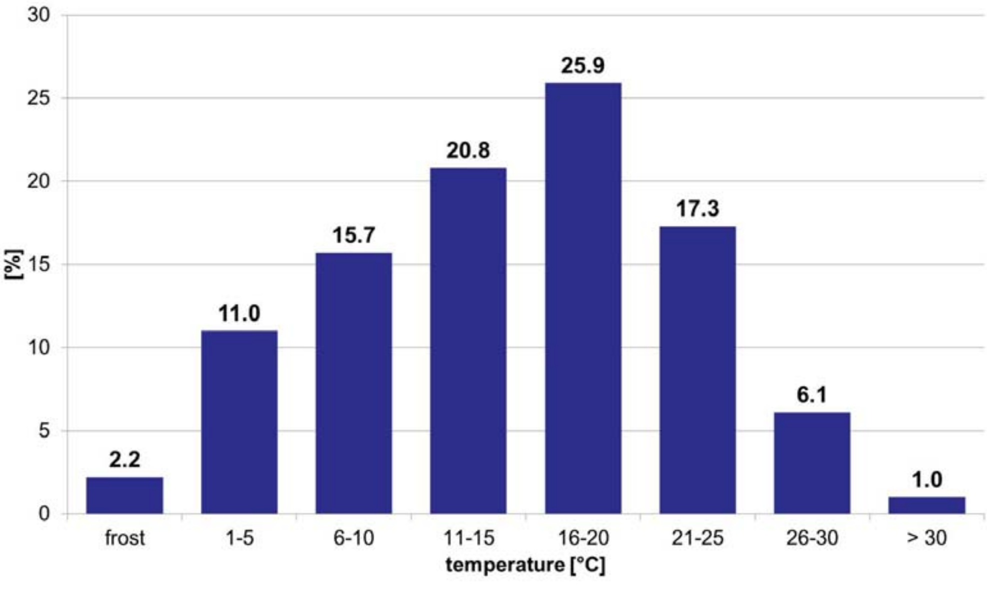


Figure 1 : Results based on GIDAS and illustrating that 96.8% of the accidents occur between 1 and 30 °C.

3 HEADFORM CHARACTERISTICS

The current head forms are characterized by a non-deformable “head-shaped” masse. The value of the mass includes “some” neck effects and its inertia is not controlled. It should be mentioned that existing simple head forms such as the Pedestrian ISO head form presents inertia quite far from human head inertia as shown in table 1. Important improvement would be possible by replacing these ISO head forms by Hybrid III head. Main arguments to do so are:

- More realistic head mass
- More realistic head inertia, an essential aspect if rotation is considered
- Deformable skin, an essential aspect for helmet optimisation
- Easy link to Hybrid III neck
- Possibility to fix rotational transducers

A critical point which has to be managed is the sizing aspect, as currently a number of sizes are available for the ISO cyclist or motorcyclist helmet head form. This can be solved through the different version of the Hybrid III dummy heads as illustrated in table 2. As it is well known that sizes A, C, E, J, M and O represent 95% of sizes used in standard, only these sizes would show interest. It appears in table 2 that these sizes would adequately be covered by five sizes of the Hybrid III heads family.

	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 50^{th}	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 1. Synthesis of headform inertial properties and comparison with human head characteristics.

EN 960 headform size	Head circumference [mm]	Dummy model	Head circumference [mm]
A	500	Hybrid III 3 Year Old	508
B	510		
C	520	Hybrid III 6 Year Old	520.7
D	530		
E	540	H III 5th Female (or 10 years)	538.5
F	550	H III 5th Female (or 10 years)	538.5
G	560		
J	570	Hybrid III 95th Large Male	584
K	580		
L	590		
M	600	Hybrid III 50th Male	597
N	610		
O	620	Hybrid III 50th Male	597
P	630		
Q	640		

Table 2. Comparison between the EN 960 headforms circumferences and Hybrid III dummy heads. The A, C, E, J, M and O sizes represent 95% of size used in global standards and are covered by the Hybrid III heads family

A further question concerning the head form is related to its boundary condition at neck level. As for the tangential impact it is recommended to use the Hybrid III head a numerical simulation has been performed with three helmeted head form models, i.e. ISO head form, Hybrid III head and Hybrid III head connected to Hybrid III neck in framework of Cost TU0111. Impacts were simulated under occipital impact against a 45° inclined anvil. Results shown in figure 2 demonstrate that ISO head form has a very non-realistic response due to its non-controlled inertia. On the other hand both Hybrid III and coupled Hybrid III head and neck present similar results in terms of rotational acceleration during the 10 first milliseconds. Obviously the neck influences the head kinematic longer after the impact itself.

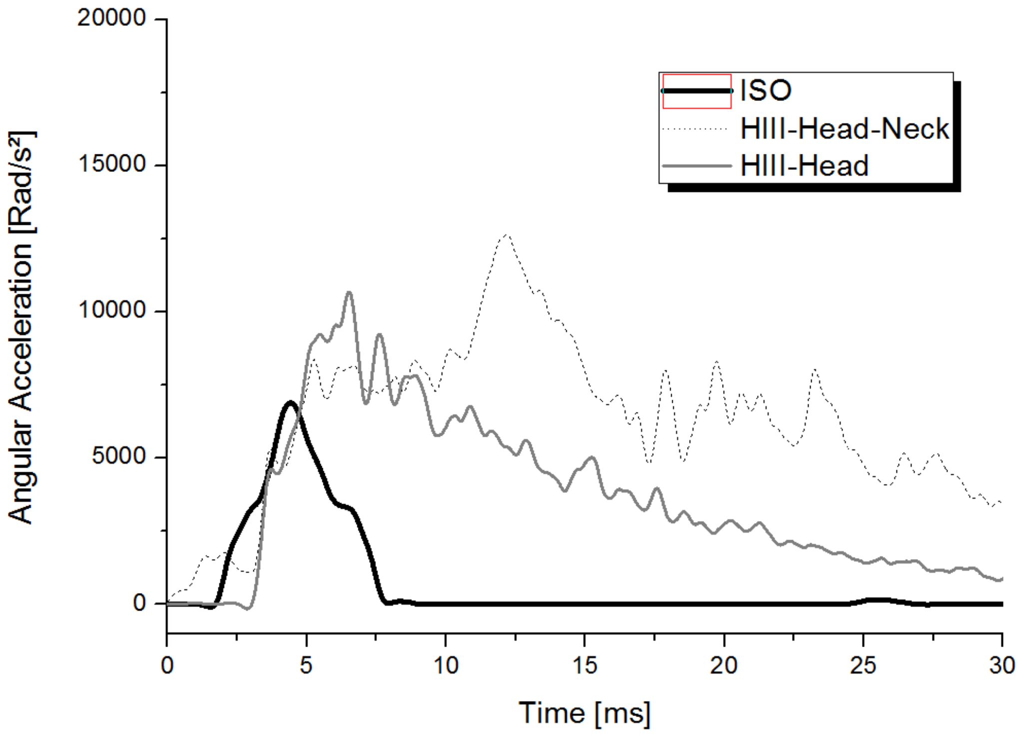


Figure 2 : Rotational acceleration computed with three different helmeted headforms and head boundary conditions under occipital impact against a 45° inclined anvil.

4 IMPACT POINTS

In the current standard impact points are chosen anywhere above the R-R' test line as illustrated in figure 3. Accident investigations (Bourdet et al 2012 [10]) clearly show that a critical area is the temporal zone and this aspect should be improved. Therefore the present proposal suggests to lower the test line in a similar way as suggested by Otte et al 2013 [31] in framework of Cost action TU0111 (figure 3).These linear impacts should be conducted with Hybrid III head alone and strictly in line with G (centre of mass of the helmeted head). Gyrometer should control this aspect and a very low accepted rotational acceleration should be imposed.

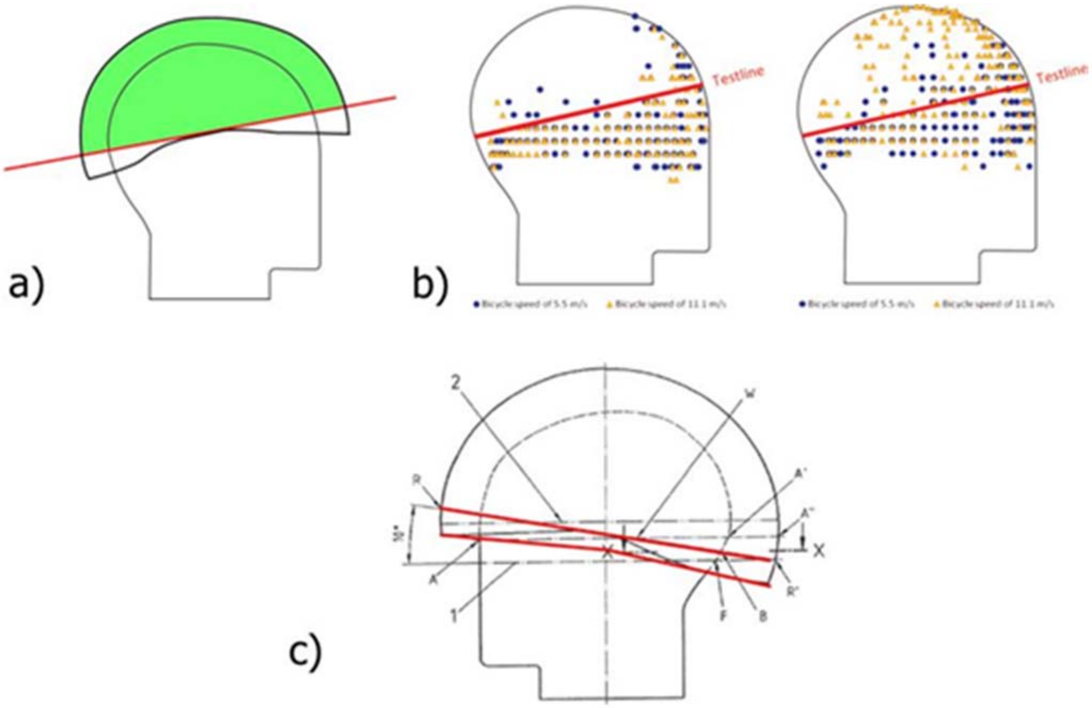


Figure 3 : Illustration of current test line a), impact points to the head (Bourdet et al 2012 b) and proposal of new test line as from Otte et al 2013 c)

Coming to the tangential impact tests, the ISO head forms has been replaced by the Hybrid III head, as this dummy head has much more realistic inertia properties. For these tests it is suggested to freely drop the helmeted head with a 6.5 m/s initial velocity against a 45° inclined anvil and to record rotational acceleration in addition to the linear acceleration.The first proposed impact is a tangential impact in the sagittal plan (point A) leading to rotation around the Y axis (lateral right to left) direction. The two next tangential impacts are located at parietal level (points B and C) and will be applied in the frontal plane, one introducing rotation around the X (postero-anterior) direction and one introducing a rotation around the Z (vertical ascendant) direction (figure 4).

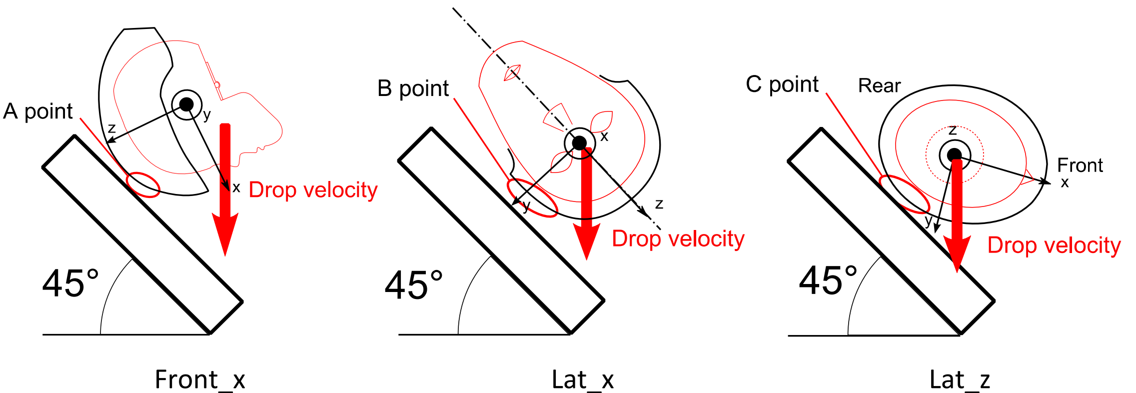
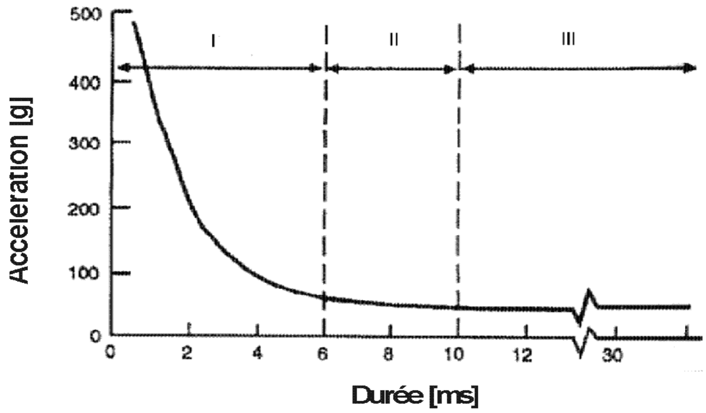


Figure 4 : Illustration of the three tangential impact conditions. From left to right, impact introducing angular acceleration around Y axes (called Front_x), X axes (called Lat_x) and finally Z axes (called Lat_z)

5 HEAD INJURY CRITERIA

Currently thresholds concerning helmet performance are set in terms of maximum headform acceleration (fixed at 250 G) according to the WSU tolerance curve reported in figure 5 and proposed in the 1950's. To protect the head in an automotive environment, HIC has been introduced in the 1970's as recalled in figure 5. 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. In addition it must be recalled that HIC does not integrate lateral direction or rotational acceleration and is unable to predict skull fracture. It is therefore a very limited head injury criterion.



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

Figure 5 : Illustration of the WSU head tolerance curve (1950's) and expression of HIC criterion (1970's)

In the literature, rotational acceleration limits are proposed at 8 to 10 krad/s2. It must however be recalled that head tolerance limit to rotation is strongly time and direction dependant so that today there is no single limit known.

Within EU project APROSYS SP5 and French PREDIT projects PROTEUS and BIOCASQ, improved head injury criteria to specific injury mechanisms have been defined taking into consideration the time evolution of both linear and rotational head acceleration as recalled hereafter.

Today, state of the art FE head models exist and have been used for the definition of injury criteria to specific injury mechanisms. These models became much more powerful injury prediction tools that HIC, so the present proposal is to implement improved, model based head injury criteria into a new helmet impact test procedure. Based on the simulation of nearly 100 well documented head trauma, tolerance limits have been identified with respect to moderate and severe neurological injury. Human head tolerance limits relative to neurological injuries with a risk of occurrence of 50 % were established as follows:

- A brain Von Mises strain reaching 20 % for moderate neurological injury.
- A brain Von Mises s strain reaching 35 % for severe neurological injury.
- Finally a global strain energy of the sub-arachnoid space exceeding 4.2 J generates subdural and sub-arachnoid haematoma, and a local strain energy in the skull of 439 mJ for a 50% risk of a skull fracture.

In the proposed approach the experimental head linear and rotational head acceleration will constitute the inputs which will drive the head FE model, in charge of the latter to compute the injury parameters related to, subdural haematoma and neurological injury. By this methodology it will be possible to predict head injury risk means a coupled experimental versus virtual testing procedure as illustrated in figure 6.

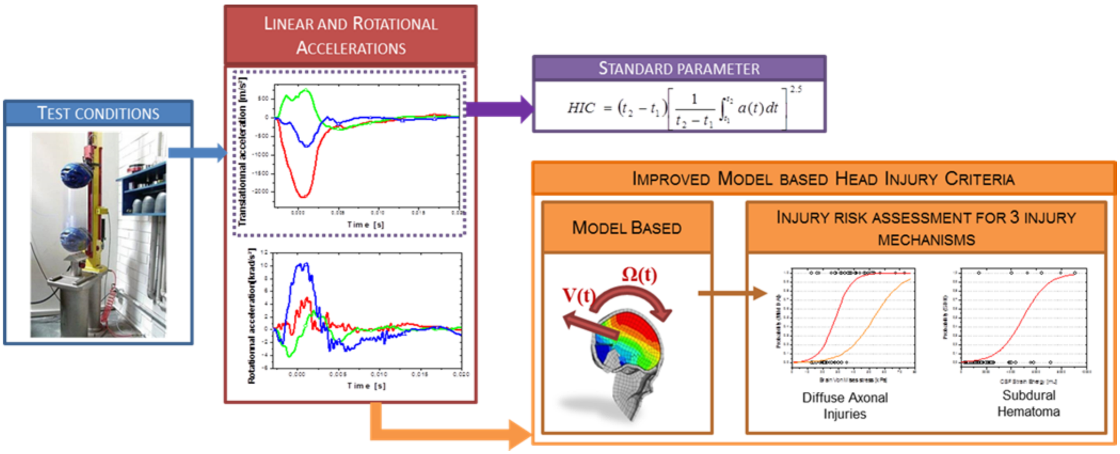


Figure 6 : Illustration of the coupled experimental versus numerical head impact test method based on novel model based head injury criteria.

6 CONCLUSION

This paper presents a proposal for a possible evolution of current EN 1078 bicycle helmet standard. A total of four key aspects have been reviewed in a critical way, i.e. head impact conditions, head surrogate, head impact location and head injury criteria. For each of these issues a concrete improvement proposal has been made in order to open a discussion on further research needed. At head impact conditions level it is proposed to implement at tangential head impact tests with a helmeted Hybrid III head. This improved head has also the advantage of more realistic inertial properties and interface characteristics between head form and helmet. It is also suggested to remove tests under extreme temperature or against the curbstone anvil. Further improvement concerns the impact location as the current test line excludes any impact to the temporal region which has been shown to be a critical area. A final key evolution which is proposed concerns the assessment of the head injury risk for which a coupled experimental versus numeric method is proposed in order to introduce model based head injury criteria. It is expected that the evolution of helmet standard test method will enable advanced helmet evaluation and optimization against biomechanical criteria.

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