

Assessment of Bicyclist Head Injury Risk under Tangential Impact Conditions

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Abstract Although cycling may be attractive for both economic and environmental reasons, cyclists are extremely vulnerable road users and subjected to falls or collisions with cars. Therefore, wearing a helmet is of crucial and vital importance considering the high percentage of head injuries in cyclists. The present research attempts to understand the degree of protection offered by a commercial bicycle helmet under realistic oblique impact conditions. Experimental research was carried out on an existing helmet to perform 2 oblique impact situations at ambient temperature. In parallel, the helmet was digitized and meshed to produce a finite element model (FEM) that was implemented under the PamCrash® crash code. Coupled to a 50th percentile Hybrid III dummy head FEM, the experimental tests were numerically reproduced in order to provide a validation of the helmet model under tangential impact conditions. Then, coupled to a detailed human head model in order to assess the head injury risk under tangential impact, results showed that even if frontal oblique impacts are more critical than lateral ones, computed injury risks are acceptable for a realistic head impact condition as extracted from bicyclist accident analysis.

Keywords: bicycle helmet, brain injuries, head modeling, helmet modeling, oblique impacts

I. INTRODUCTION

Cycling represents one of the most popular recreational sports, but also one of the main transportation means all over the world and more specifically in urban areas. As reported by the United Nations in 2003 [1], road traffic injuries caused the death of more than one million humans in the year 2000, representing the ninth highest cause of death worldwide. Many previous studies have outlined the effectiveness of wearing bicycle helmets [2-3] and also the possibility of the improvement of currently used helmet design [3]. According to Belgian statistics published in a recent survey [6], 7.3% of bicyclists used the bike as their main method of transportation to work and 19.2% to school. In 2000, 9.8% of all road accidents and 9.1% of road traffic deaths on Belgian public roads were represented by pedal cyclists. Besides, according to a survey published in 2012 concerning road accidents in France [7], 65% of the 4567 bicycle accidents in 2011 were collisions with cars whereas only 4% were collisions with heavy trucks.

The head represents the most vulnerable part of the human body and one of the most important to be protected. Between 21 to 61% of the victims of bicycle accidents seeking medical care have a head injury as reported by Collins et al. [8] and Eilert-Petersson & Schelp [9]. The medical care required by this large number of victims also implies high costs both economic and societal. Bicycling dramatically leads to increased numbers of fatalities due to head injuries which represent the cause of death in 69–93% of fatal accidents according to Oström et al. [10]. Therefore, wearing a helmet is of crucial and vital importance considering this high percentage of head trauma in cyclists.

The bicycle helmet protects the head against impacts by reducing the impact energy that is transmitted to the head through energy absorption and dissipation through elastic and plastic deformation of its components (external shell and foam liner), and by increasing the area over which the impact is distributed. Nonetheless, the EN 1078 standard [11] considered in Europe to homologate bicycle helmets does not take into account any effects of a tangential component since only normal impacts are performed. More realistic would be to test helmets for impact similar to the most commonly observed impacts in real life bicycle accidents. Otte et al. [12] showed that oblique impacts, with a significant tangential force applied on the helmet, are more common than radial (normal) impacts in crashes. The authors also found that the frontal regions and the sides of the helmets

were the most frequently damaged regions. More recently, Bourdet et al. [13] reconstructed a total of 26 bicyclists' accident cases with head injuries and computed both linear and tangential head impact velocity. The results showed that the head is impacted more often on the top parietal zone and the mean impact velocity is 6.8 ± 2.7 m/s with 5.5 ± 3.0 m/s and 3.4 ± 2.1 m/s for normal and tangential components, respectively. The most frequently sustained brain injuries in motor vehicle accidents that result in either fatality or the need for long-term rehabilitation are subdural hematomas (SDH) and diffuse axonal injuries (DAI) [14]. The main causes of SDH are ruptured arteries or bridging veins. DAI is caused by the tearing of neuronal axons in the brain tissue. It has been demonstrated and well accepted that there are correlations between these injuries and rotational loading of the head [13-15].

According to Aare et al. [17], there is currently a lack of good testing methods for evaluating the effects of oblique impacts. There is also a need for improved understanding of the effects on the human head subjected to oblique impacts. In current helmet standards, no rotational effects are measured in the headform, partly because there are no accepted global injury thresholds for a combination of rotation and translation. Rather than dropping the headform in an oblique direction onto a fixed horizontal surface, some experimental test rigs were constructed by researchers consisting of a freely-falling Hybrid III headform impacting a horizontally moving surface [18-20]. Aldman et al. [21] dropped a helmeted dummy onto a large rotating table moving at 8.3 m/s and observed values of peak angular head acceleration in the range of 7 to 15 krad/s². Similarly, Mills & Gilchrist showed in 1996 [22] that the peak linear acceleration was slightly reduced by the tangential velocities of 8 m/s. More recently in 2008 [23], they investigated the performance of bicycle helmets in oblique impacts with a simulated road surface, and both linear and rotational accelerations of a headform, fitted with a compliant scalp, were measured.

The present study aims at assessing a commercial bicycle helmet under tangential impact by means of both experimental and numerical methods. The literature reports a number of bicycle helmet models such as Willinger et al. in 1998 [24], Gilchrist & Mills in 2008 [25], Asiminei et al. in 2009 [26] and Milne et al. in 2012 [27]. Except for the last authors who assessed risks of both DAI and SDH under 90 normative impact cases, none of the reported bicycle helmet FE models were coupled to a human head FE model for a detailed head injury risk assessment against different injury mechanisms. In this context, the present paper reports the bicycle helmet modeling effort followed by a detailed validation against 2 experimental oblique impact tests. Once validated, the helmet model was coupled to an existing human head model (the Strasbourg University Finite Element model or SUFEHM) in order to compute intracranial field parameters such as the brain Von Mises stress and minimum pressure in cerebrospinal fluid (CSF). This step permitted an original head injury risk assessment and evaluation of the degree of protection offered by such a commercial bicycle helmet in case of realistic tangential impact.

II. METHODS

Experimental Approach

As a first step, experimental research was carried out to perform oblique impacts on a commercial bicycle helmet illustrated in Fig. 1. A moulding process allowed the external polycarbonate (PC) shell to be glued to an expanded polystyrene (EPS) foam liner of density 77 kg/m³. The helmet's shape is sharp with numerous vent holes and has a total mass of 0.235 kg.



Fig. 1. Lateral view of the helmet

To assess the dynamic energy absorption characteristics of such a helmet in case of tangential impact, automatic electro-pneumatic shock absorption test equipment with a falling speed up to 8 m/s was used. The helmet was positioned on an instrumented biofidelic 4.5 kg 50th percentile Hybrid III dummy head (Fig. 2) and the retention strap was tightly clamped to maintain the head-helmet system combined during the test. Three ENDEVCO 7264B/2000 piezoresistive accelerometers capturing accelerations up to 2000 g were placed at the centre of gravity of the headform to measure linear accelerations. One IES 3103-4800 3-axes gyro-sensors measuring angular velocities up to 4800 °/s was also mounted inside the headform. The sensors were linked to an ISO 6487 [28] certified data recorder equipped with a 250 kHz Sensor Input Module (SIM) card. The control software installed on the computer provided tools for performing data collection and display. An oblique anvil was manufactured in XC38 steel with an angle of 60°. As shown in Fig. 3, this anvil was covered by an 80 grain abrasive paper specified in BS 6658 [29] and ECE R-22.05 [30] standards to represent realistic road-helmet friction.

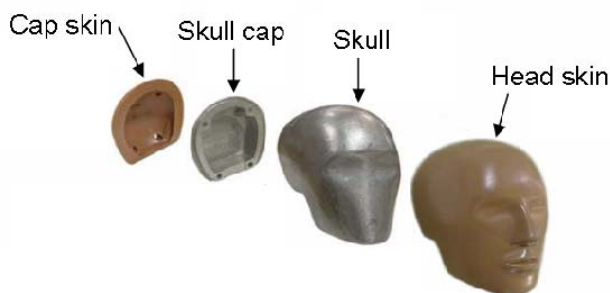


Fig. 2. Hybrid III dummy head

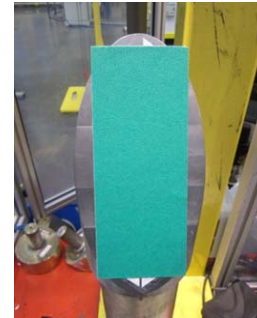


Fig. 3. Illustration of the 60° anvil covered by abrasive paper

In this test program, 10 bicycle helmets were impacted at velocity 5.42 m/s on both frontal and lateral regions on a 60° anvil and under ambient temperature. Finally, a total of 20 oblique impacts were performed experimentally. An example of positioning of the helmet against the 60° anvil is illustrated in Fig. 4a and Fig. 4b prior to frontal and lateral impact, respectively. For each oblique impact, linear accelerations along the three axes and rotational velocities were filtered respectively by CFC filters with 1 kHz and 180 Hz cutoff frequencies. Then, the angular velocity versus time curves were derived to get time evolution of angular accelerations.



(a)



(b)

Fig. 4. Impact on a 60° anvil: a) Frontal, b) Lateral

Computational Modeling

The helmet under study was digitized to get a Computer-Aided Design (CAD) of its geometry. The developed helmet model consists of two parts, i.e. an external shell and a liner. The CAD was automatically meshed under the PamCrash® Mesh module. After mesh convergence tests performed to obtain optimal results, 4535 2D tria elements were created for the shell and 24380 3D tetra elements for the lining. Hence, the complete helmet FEM consists of 28915 elements. The whole model is presented in Fig. 5. Shell elements have a minimum and maximum size of 2.35 mm and 7.5 mm, respectively. Concerning tetra elements, the minimum element size is 2 mm. Finally, the meshing was generated to ensure a mesh continuity between the external shell and the liner.

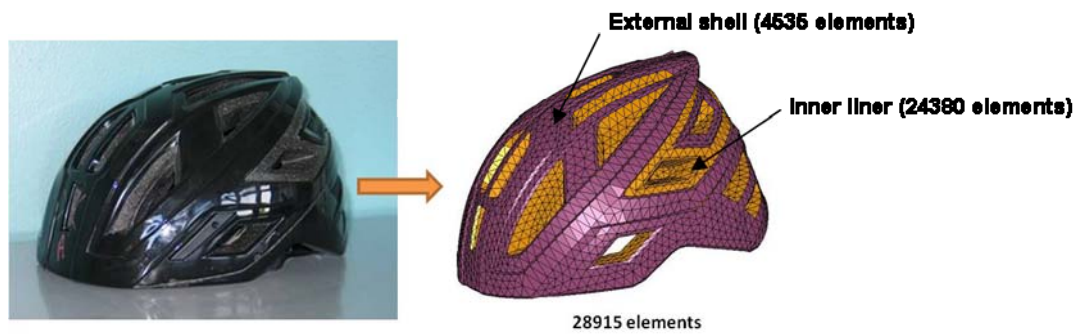


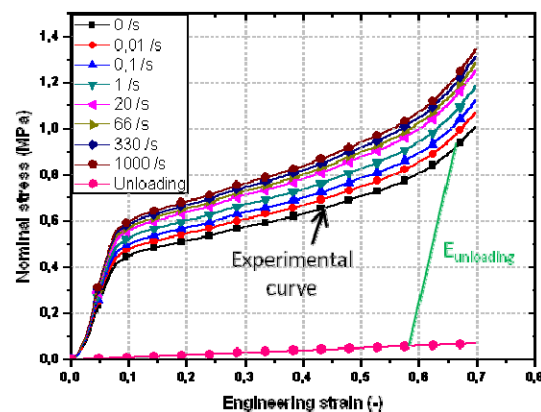
Fig. 5. General view of the bicycle helmet finite-element model

The mesh was imported under the PamCrash® explicit crash code and the material properties of the two constitutive parts of the bicycle helmet model were implemented. It was observed that under impact, the EPS foam liner is compressed up to large deformations. In order to simulate such deformations, the EPS behavior was modeled by a tabulated "General Nonlinear Rate Dependand Foam with Optional Energy Absorption" law material. The material properties implemented in the model are based on experimental EPS compression tests at quasistatic strain rate and summarized in Table 1. This quasistatic loading curve is supposed to represent the material behavior at strain rate "0 /s" according to the PamCrash® formalism for the considered law. Stress versus strain curves at higher strain rates were extrapolated from the first loading curve by applying scaling factors deduced from an experiment-based logarithmic interpolation. The behavior of the EPS is defined by a set of 8 stress versus strain loading curves, each one related to a strain rate value. Depending on the velocity, a linear extrapolation beyond the last curve and/or a linear interpolation in between two curves is operated by the unloading slope $E_{unloading}$.

Table 1. Mechanical properties of the EPS foam liner implemented under PamCrash®

ρ (kg/m ³)	77
$E_{unloading}$ (MPa)	10

Tabulated loading curves at ambient temperature



A linear elastic behavior was assumed for the polycarbonate shell and modeled through the "Elastic-plastic thin shell material" law. The values of the four parameters had to characterize the PC material, in accordance with Mills & Gilchrist [25], and are reported in Table 2. A measured thickness value of 1.18 mm was attributed to all shell elements.

Table 2. Mechanical properties of the PC shell

ρ (kg/m ³)	1200
E (MPa)	2400
ν (-)	0.35
σ_{yield} (MPa)	50

To simulate the 2 oblique experimental tests, a 50th percentile Hybrid III dummy head FEM (Fig. 6a) was imported under the PamCrash® environment. This 4.5 kg headform was meshed with 1888 brick elements and 944 shell elements as shown in Fig. 6b. An accelerometer was defined at the centre of gravity of the head in order to compute rotational accelerations. Moreover, its inertia was in good agreement with that of the physical model as demonstrated in Table 3.

Table 3. Inertia of the 50th percentile Hybrid III dummy head

Model	$I_{xx} (x10^{-3} \text{ kg.m}^2)$	$I_{yy} (x10^{-3} \text{ kg.m}^2)$	$I_{zz} (x10^{-3} \text{ kg.m}^2)$
FEM	17.7	18.6	23.4
Physical	17.1	18.9	22.7

The head-helmet contact of the coupled system illustrated in Fig. 7 was defined by taking into account a “Symmetric node to segment with edge treatment” interface with a 0.5 friction coefficient.

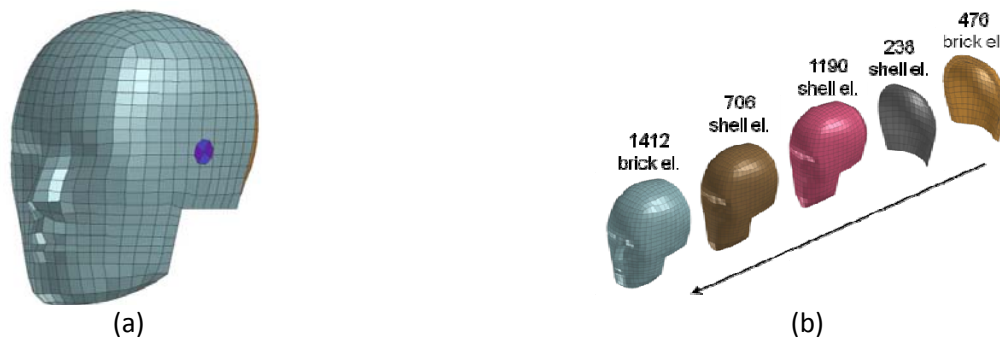


Fig. 6. Hybrid III dummy head: a) FE model, b) Main components

The geometry and the dimensions of the anvil complied with the EN 1078 prescriptions (minimum diameter of (125 ± 3) mm and 24 mm thick). The “Null material for shell elements” material law was implemented with parameter values mentioned in Table 4. A “Symmetric node to segment with edge treatment” interface with a friction coefficient of 0.5 was defined to compute the shocks on this inclined anvil.

Table 4. Mechanical properties of the anvil

$\rho \text{ (kg/m}^3\text{)}$	2700
$E \text{ (MPa)}$	200000
$\nu \text{ (-)}$	0.35



Fig. 7. Simulation of oblique impact tests at velocity 5.42 m/s on a 60° anvil: (a) Frontal, (b) Lateral

In order to assess the risk of head trauma in terms of DAI and SDH under tangential impacts, a coupling between the bicycle helmet FEM and the Strasbourg University Finite Element Head Model (SUFEHM) was performed under the PamCrash® software. This model developed by Kang et al. [33] and illustrated in Fig. 8 was meshed with 10395 brick elements and 2813 shell elements. Based on anatomical atlas, the head model contains the face, scalp, skull and a homogenous brain. Intracranial membranes and cerebrospinal fluid (CSF) are also represented in this model. The constitutive material laws of the different parts of the model have been defined by the material laws specified in appendix 1 according to Willinger et al. [34]. This model was validated under dynamic experimental tests reported by Nahum et al. [35], Trosseille et al. [36] and Yoganandan et al.

[37]. This 4.7 kg head model has been extensively used for the simulation of a number of well documented real world types of head trauma in order to correlate intra-cranial mechanical parameters with the occurrence of injury. This approach helped to define model-based head injury criteria reported by Deck et al. [38], summarized as follows:

- 50% risk of moderate neurological injury (DAI1+): Brain Von Mises shearing stress of 29 kPa,
- 50% risk of severe neurological injury (DAI2+): Brain Von Mises shearing stress of 53 kPa,
- 50% risk of subdural hematoma: minimum pressure of -315 kPa in CSF.

The interface between the energy-absorbing foam and the human head model (Fig. 9a) was computed by a “Symmetric node to segment with edge treatment” PamCrash® contact with a friction coefficient set to 0.5. An example of positioning of the head-helmet system for a frontal impact against the anvil is shown in Fig. 9b.

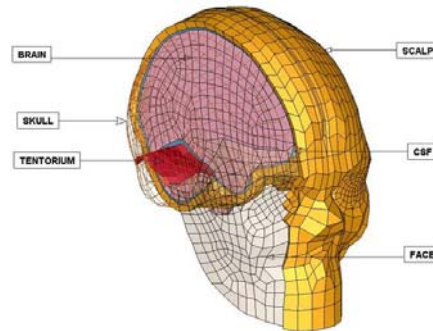


Fig. 8. Illustration of the Strasbourg University Finite Element Head Model



Fig. 9. Helmet-human head models coupling: (a) Illustration of the head-helmet interface, (b) Initial positioning of the coupled head-helmet system against the inclined anvil

III. RESULTS

During the validation process and for each impact location, numerical resultant linear acceleration and angular acceleration were computed at the centre of gravity of the Hybrid III headform in order to superimpose them with the respective experimental results. In the framework of this helmet validation effort under oblique impact configurations, results in terms of time-history acceleration curves were superimposed with mean experimental records and their associated standard deviation. Curves were numerically filtered by CFC filters with 1 kHz and 180 Hz cut-off frequencies, respectively. An illustration is given for a frontal impact at 5.42 m/s on a 60° anvil in Fig. 10 and Fig. 11 for resultant linear acceleration and angular acceleration around Y (and X) axis, respectively. A reasonable agreement is observed in terms of time durations and peak values demonstrating that the defined stress-strain curves correspond approximately to the energy-absorbing material in the considered loading case. Numerical peak g and peak krad/s² values are 67 g and 5.1 krad/s². Related experimental values are 60 g and 4.8 krad/s² leading to relative errors less than 10%.

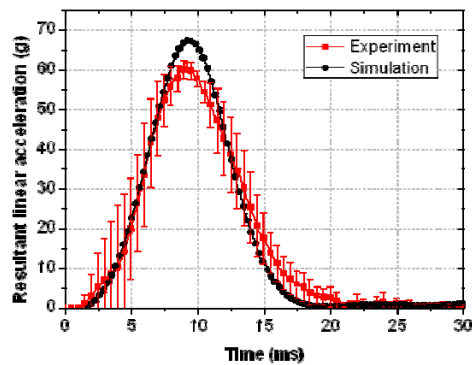


Fig. 10. Superimposition of time-history resultant linear acceleration curves for a frontal impact ($f_c = 1$ kHz)

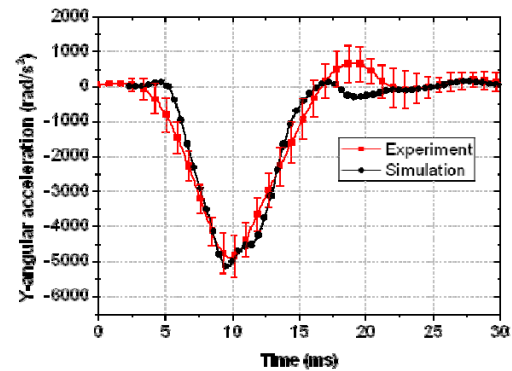


Fig. 11. Superimposition of time-history angular acceleration curves for a frontal impact ($f_c = 180$ Hz)

Time-history resultant linear and angular accelerations obtained for a lateral impact are shown in Fig. 12 and Fig. 13. By analyzing the time durations and peak values between numerical and experimental records, a good agreement is also observed. Numerical simulation of tangential impacts tests led to peak values of 70 g and 4.7 krad/s². Compared to experimental values, relative errors are 5.7% and 18%, respectively.

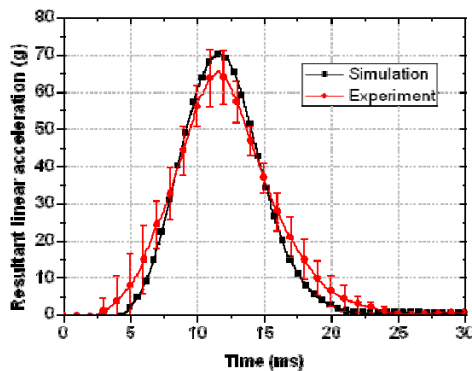


Fig. 12. Superimposition of time-history resultant linear acceleration curves for a lateral impact ($f_c = 1$ kHz)

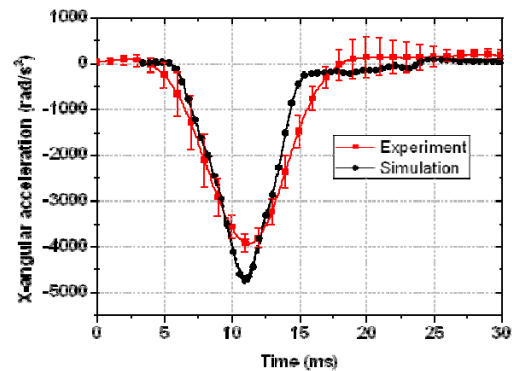


Fig. 13. Superimposition of time-history angular acceleration curves for a lateral impact ($f_c = 180$ Hz)

When coupled to the human head model, intracranial parameters were computed in order to assess the injury risks in terms of diffuse axonal injuries and subdural hematoma under tangential impacts, in accordance with the model-based injury criteria described in the Methods section. 10 oblique impact configurations performed with a helmeted head model are illustrated in Table 5. 3 head impact angles and 3 impact velocities were investigated to assess the influence of both normal and tangential components on head injury risks for a reasonable range of tangential impact conditions.

Table 5. Oblique impact loading cases simulated with helmeted SUFEHM

Impact	α (°)	V (m/s)	V_N (m/s)	V_T (m/s)
Frontal	35	6.5	5.3	3.7
	45	6.5	4.6	4.6
		5.42	2.7	4.7
	60	6.5	3.3	5.6
		8.0	4.0	6.9
Lateral	35	6.5	5.3	3.7
	45	6.5	4.6	4.6
		5.42	2.7	4.7
	60	6.5	3.3	5.6
		8.0	4.0	6.9

As for the numerical simulations with the Hybrid III head FE model, the quality of each simulation was verified by comparing hourglass to total energy of the system and making sure that hourglass energy did not exceed 10% of the total energy involved.

The peak intracerebral Von Mises stress computed under the 10 previously presented oblique impact conditions are shown in Fig. 14 in which 50% tolerance limits in terms of moderate and severe DAI are also indicated. Globally speaking, lateral impacts lead to lower intracranial shearing values than frontal impacts under given loading conditions. For a 60° head impact angle, the peak Von Mises stress increases from 29 kPa at a velocity of 5.42 m/s to nearly 55 kPa at 8.0 m/s. Moreover, for an impact velocity of 6.5 m/s, peak Von Mises stress reaches its higher value (45 kPa) for a 45° head impact angle. For most of the performed simulations, values higher than 29 kPa are calculated leading to a risk of moderate DAI of 50% at least.

Results obtained for the same 10 impact configurations in terms of minimum pressure in the CSF layer are presented in Fig. 15. Again, the 50% tolerance limit in terms of SDH is drawn. Peak values of -117 kPa and -195 kPa were obtained for a frontal impact on a 60° anvil at 5.42 m/s and 8.0 m/s, respectively.

Injury assessment in terms of percentage of risk to sustain a moderate or severe DAI for the helmeted head subjected to oblique impacts is presented in Fig. 16 and Fig. 17, respectively. By considering a realistic head impact condition reported by Bourdet et al. [13], i.e. $V=6.5$ m/s and $\alpha=35^\circ$, risks of moderate and severe DAI for frontal impact (respectively lateral) are 88% (77%) and 23% (15%). Risks for the helmeted user to suffer from a SDH in case of tangential impact conditions are very limited and detailed in Fig. 18. For the 10 impact configurations performed, risks of SDH never exceeded 10%.

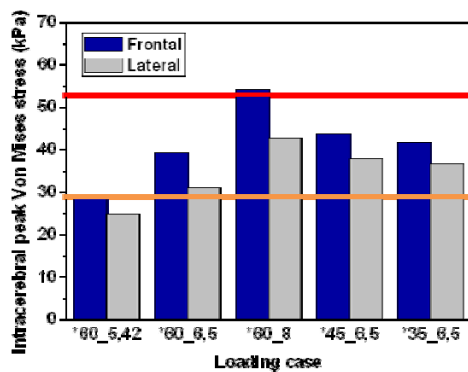


Fig. 14. Intracranial Von Mises stress computed for each of the 10 oblique impact configurations

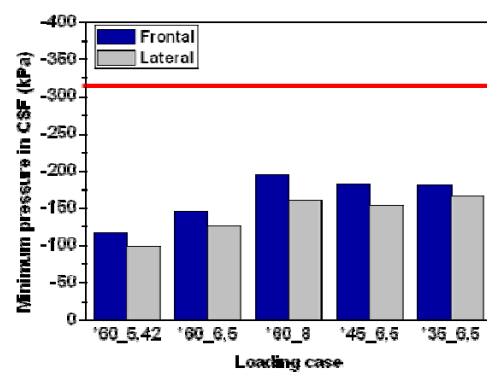


Fig. 15. Minimum pressure in CSF computed for each of the 10 oblique impact configurations

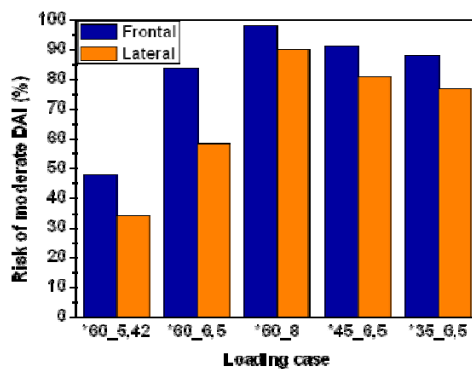


Fig. 16. Risk of moderate DAI for the 10 oblique impact configurations

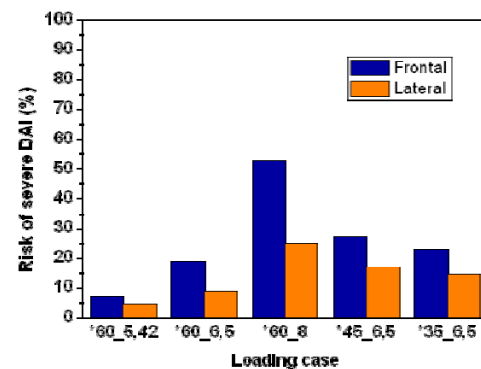


Fig. 17. Risk of severe DAI for the 10 oblique impact configurations

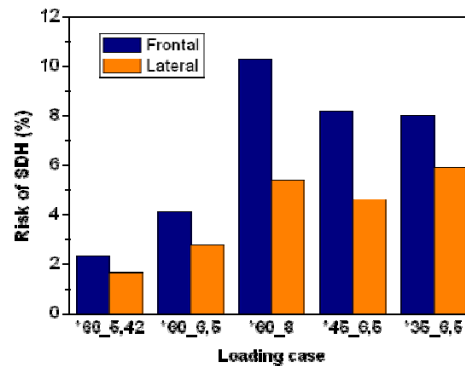


Fig. 18. Risk of SDH for the 10 oblique impact configurations

IV. DISCUSSION

This paper proposes an evaluation of a FE model of an existing commercial bicycle helmet under tangential impacts. After being subjected to experimental oblique tests, numerical simulations were performed under the PamCrash® crash code to validate the helmet model impacted at 5.42 m/s on 60° anvil and on both frontal and lateral regions. For the impact conditions considered in this study the rotational acceleration is mainly along one single axis only. However, in a further study it would be important to investigate more complex impact situations leading to combined rotational accelerations.

For the two simulations performed, a good agreement with acceptable relative errors has been shown between experimental and numerical results both in terms of time-history resultant linear acceleration and angular acceleration curves, computed at the centre of gravity of the Hybrid III dummy head FE model. For each impact location, numerical results are within the experimental corridors based on 10 impacts for friction coefficients of 0.5. This is also a limitation of the study and, in a further step, there is a need to conduct a parametric study on these parameters.

In addition, the bicycle helmet model was validated against experimental angular acceleration filtered at 180 Hz. A more pertinent validation could have been proposed if no filtering had been applied on these experimental records. It should be mentioned that this helmet model has been validated under linear impact conditions in an earlier study [27].

Even if current bicycle helmets are designed to reduce headform acceleration, they are still not optimized to reduce human head injuries under tangential impact. In order to progress towards this final objective, head injury risk assessment under tangential impact conditions was performed in the present study. It appears, for the two validated impact configurations, that the risk of severe DAI is less than 7%. A 2% risk of SDH has also been determined. It should be mentioned that the risk of skull fracture was not considered in the present study.

According to the realistic impact conditions reported by Bourdet et al. [13], head injury risks have been assessed for a helmeted head impacting the ground at a velocity of 6.5 m/s and 35°. Risks of severe DAI are 23% and 15%, respectively, for impacts performed at the front and the side, values that are acceptable regarding levels generally accepted in standards.

V. CONCLUSIONS

In this study, a previously published bicycle helmet FE model was implemented and validated under tangential impact condition under the PamCrash® explicit crash code.

The numerical simulation of two experimental oblique impact tests at ambient temperature was performed by coupling the helmet FEM to the Hybrid III dummy head FE model. Results in terms of resultant linear acceleration and angular acceleration versus time curves between experiment and simulation are in good agreement for both impact points.

Once validated, this helmet FE model was coupled to the Strasbourg University FE Head Model in order to assess the head injury risks of both DAI and SDH injuries. Results show that frontal oblique impacts are more critical than lateral ones. Results also show that the computed injury risk is acceptable for a realistic head impact condition as extracted from bicyclist accident analysis. This work is therefore a step towards helmet optimisation against biomechanical head injury criteria and the consideration of model-based head injury

criteria in future helmet standards.

VI. ACKNOWLEDGEMENT

The experimental approach has been carried out within the ANR PREDIT Projet "BICYTETE" led by the French Department of Transport.

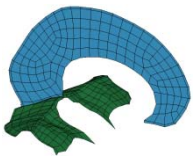
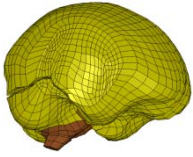
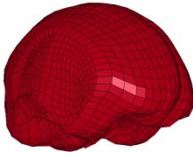
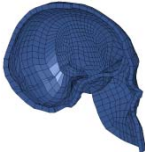
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VIII. APPENDIX

APPENDIX 1. MECHANICAL PROPERTIES OF THE SUFEHM CONSTITUTIVE PARTS

Segment	Illustration	Mechanical behavior	Mechanical parameters	Mechanical parameters
Falx and Tent		Linear elastic	$e = 1 \text{ mm}$ $\rho = 1140 \text{ kg/m}^3$ $E = 31.5 \text{ MPa}$ $\nu = 0.45$	/
Brain-Skull interface		Linear elastic	$\rho = 1040 \text{ kg/m}^3$ $E = 0.012 \text{ MPa}$ $\nu = 0.49$	/
Brain and Cerebellum		Viscous elastic	$\rho = 1040 \text{ kg/m}^3$ $K = 1125 \text{ MPa}$ $G_0 = 0.049 \text{ MPa}$ $G_{inf} = 0.0167$ $\beta = 145 \text{ s}^{-1}$	/
Skull		Fragile elastic-plastic	Cortical $e = 2 \text{ mm}$ $\rho = 1900 \text{ kg/m}^3$ $E = 15000 \text{ MPa}$ $\nu = 0.21$ $K = 6200 \text{ MPa}$ $UTS = 90 \text{ MPa}$ $UTC = 145 \text{ MPa}$	Spongy $e = 3 \text{ mm}$ $\rho = 1500 \text{ kg/m}^3$ $E = 4600 \text{ MPa}$ $\nu = 0.05$ $K = 2300 \text{ MPa}$ $UTS = 35 \text{ MPa}$ $UTC = 28 \text{ MPa}$
Face		Linear elastic	$e = 10 \text{ mm}$ $\rho = 2500 \text{ kg/m}^3$ $E = 5000 \text{ MPa}$ $\nu = 0.23$	/
Skin		Linear elastic	$\rho = 1000 \text{ kg/m}^3$ $E = 16.7 \text{ MPa}$ $\nu = 0.42$	/