

Finite Element Analysis of Helmeted Impacts and Head Injury Evaluation with a Commercial Road Helmet

Fábio A.O. Fernandes, Ricardo J. Alves de Sousa, Rémy Willinger, Caroline Deck

Abstract This research was conducted to assess the safety performance of a commercial motorcycle helmet, homologated by the most relevant current standards and currently available in the market. The evaluation was based on accurate reproduction of translational impacts specified by the ECE R22.05 standard to assess motorcycle helmets. The Finite Element numerical framework was validated against two different sets of experimental results. The first addressed the constitutive model of the expanded polystyrene, the material responsible for energy absorption during impact; the second related to the headform's centre of mass acceleration after the impacts defined in the ECE R22.05 standard. Both were successfully validated. Once a functioning and validated numerical motorcycle helmet model was created, the Strasbourg Finite Element Head Model was employed to evaluate in more detail the severity of sustained injuries after impacts in one of the four specified points of ECE R22.05 instead of employing a rigid headform. From this analysis, it was concluded that brain injuries such as concussion and diffuse axonal injury may occur even with a helmet that complied with the majority of existing motorcycle helmet standards. The need for improving current test procedures and head injury criteria is strongly recommended to reduce the exposure to these types of head injury.

Keywords biomechanics, finite element method, head injury, motorcycle helmet.

I. INTRODUCTION

Road accidents are one of the major causes of death in the world [1]. About 31 thousand people die and 1.6 million people are injured every year in the European Union as a direct result of road accidents [2]. Motorcyclists, only protected by the helmet, are less protected against road accidents than the users of 4-wheeled vehicles because car occupants are protected by safety belts, airbags and even by the car's body structure [3]. Motorcycle crash victims form a substantial proportion of all fatalities (21%) and also of the severely injured category (24.9%) [4-5]. Table 1 gives the number of all injury categories for powered two-wheeler (PTW) riders in Portugal.

TABLE I
NUMBER OF FATALITIES AND INJURIES SUFFERED BY PTW OCCUPANTS IN PORTUGAL IN 2010 AND 2011

Year	Total	Minor Injuries	Severe Injuries	Fatalities
2010	7603	6844	556	203
2011	7454	6703	564	187

Head injury is one of the most frequent and most severe injuries that results from motorcycle accidents as shown in [6] where head injuries occurred in 66.7% of COST database cases. For these reasons, head protection is fundamental.

Nowadays, motorcycle helmet standards are responsible for helmet quality and effectiveness by restricting market access only to the ones that are able to fulfill all the requirements. However, there are some issues regarding head protection criteria since current standards do not take into account rotational acceleration and its effects but rely only on the headform's peak translational acceleration (PLA) and head injury criterion (HIC). In [7] these same issues were highlighted, as well as the continued use of outmoded test methods that do not

properly reflect the real circumstances of accidents such as the biofidelity of the headform, the nature of the failure criteria and the impact kinematics, which influences the movement of a tested helmeted headform.

Motorcycle helmets have a hard outer shell that prevents penetration and distributes the impact force on a wider foam area, increasing the liner capacity to absorb energy and therefore reduce the load that reaches the head. The advance of CPU power allied with the inherent difficulties of experimental methods (time, cost, flexibility, repeatability) led to the development of virtual models, mainly using the Finite Element Method (FEM). Thus, modelling provides the capability to compute several variables such as the stresses and strains not only from the impacted helmet but also from the human head in order to assess the influence of the interaction between helmet and head. Virtual models have been used in many applications, for example, the study of motorcycle helmet materials [8-9]. Thus, with virtual testing it is possible to assess the influence of a large number of parameters in a way that would be extremely costly and less flexible for experimental testing.

The main goals of this work are: 1) the analysis of a commercially available helmet approved by the ECE R22.05 standard; 2) the assessment of its capacity to protect the user from head injuries that result from helmeted impacts through the use of a biomechanical FE head model; and 3) the creation of a FE motorcycle helmet model based on realistic geometric information and known material properties in order to further optimize it. To validate the numerical model, the simulated results are compared against experimental data from the energy-absorption tests prescribed by the ECE R22.05 standard. Once a functioning and validated numerical motorcycle helmet model is created, the time-acceleration curve at point P of the ECE R22.05 test measured at the headform's centre of gravity (COG) is induced into the Strasbourg University Finite Element Model (SUFEHM). Stress and strains are then compared against proposed injury thresholds, and conclusions about possible brain injuries are inferred. It can be concluded that an approved motorcycle helmet by the ECE R22.05 standard is not effective enough in protecting the user from head injuries.

II. METHODS

The motorcycle helmet used in this work is the CMS SUV Apribile modular motorcycle helmet manufactured by CMS Helmets and presented in Figure 1. This commercially available helmet fully meets ECE R22.05 regulation [10], the Brazilian Regulation NBR-7471 [11] and also the U.S. Regulation DOT [12].



Fig. 1. CMS SUV Apribile motorcycle helmet.

Motorcycle Helmet Modelling

The geometry of the helmet was provided by the manufacturer in a computer-aided design (CAD) file. The 3D CAD model of the different parts of the helmet was treated on CATIA V5R19 CAD system [13], improving the original CADs, in order to create the FE model. Only slight simplifications were made to the model, making the model easier to mesh but maintaining the overall geometry.

Figure 2 shows the different parts of the FE helmet model developed, including the shell and the dual-density energy absorbing liner, composed of three different parts: the main padding that involves the entire cranium with the exception of the temporal region, the forehead padding insert and the lateral padding. The characteristics of the different meshes are shown in Figure 2 and detailed in Table III. The lateral liners are made

of Expanded Polystyrene (EPS) foam with density of 90 kg/m^3 and the remaining with 65 kg/m^3 density. Thickness values vary from 10 mm to 50 mm, being inversely proportional to the density. The outer shell made of Acrylonitrile-butadiene-styrene (ABS) has a thickness of 3 mm. In this study, the effects of the comfort liner, the retention system (chin strap) and the chin pad were not considered.

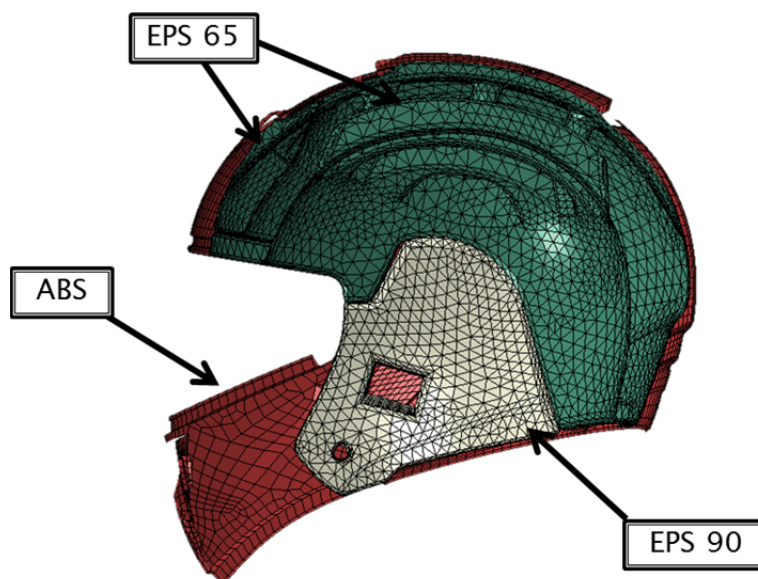


Fig. 2. Finite Element Helmet Model – A cut view from the sagittal plane.

The ECE R22.05 regulation used a 5.6 kg headform (M size). The headform model used in the simulations is shown in Figure 3 and mesh characteristics are given in Table II. After assembling all the helmet components, the headform was fitted in the helmet as shown in Figure 8.

TABLE II
HEADFORM MASS AND PRINCIPAL INERTIAL MOMENTS

Mass [kg]	$I_{xx} [\text{kg} \cdot \text{cm}^2]$	$I_{yy} [\text{kg} \cdot \text{cm}^2]$	$I_{zz} [\text{kg} \cdot \text{cm}^2]$
5.6	370	440	300

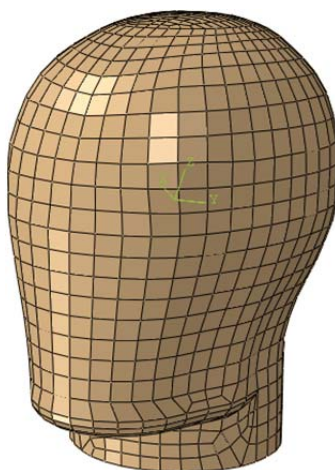


Fig. 3. Finite element model of the ECE 22.05 size M headform.

Finite Element Mesh

The summary of element types, integration schemes and mesh density is presented in Table III. Although ECE R22.05 requires more than one anvil, only the flat anvil was modelled because the most common object hit by the head in motorcycle crashes is the flat road surface [14-15], treated here as a rigid body.

The meshes of each part were created always avoiding distorted and warped elements and with special attention to the time increment, not having very small elements in order to have a reasonable computational

time but at the same time a mesh refined enough to obtain precise results.

TABLE III
CHARACTERISTICS OF MESHES USED TO MODEL THE DIFFERENT PARTS

Part	Element Type	Abaqus Element	Nº of Elements	Nº of Nodes
Main Covering Shell	Four-node, reduced integrated linear shell	S4R	9381	14073
Chin Guard Shell			2520	2723
Main Liner	Four-node linear tetrahedron	C3D4	65395	14073
Forehead Liner			4171	9587
Left Lateral Liner			12985	3317
Right Lateral Liner	Rigid quadrangular shell	R3D4	12497	2983
Headform			1346	1348
Anvil			1	4

Material Modelling

In order to simulate the helmeted impacts, it was necessary to choose suitable constitutive material models to simulate the mechanical behaviour of each material. Two different materials were modelled, the EPS and the ABS.

1) Modelling of EPS foam

EPS foam is a synthetic cellular material with closed cells widely used in energy absorption applications, such as protective gear. For this reason as well as its low cost, it is the most common liner material used in motorcycle helmets. EPS absorbs the energy during the impact through its ability to develop permanent deformation (by crushing). EPS foam uniaxial compression stress-strain behaviour can be divided into three regions as shown in Figures 6 and 7. The first region refers to linear elastic behaviour that arises from bending in the cell walls. The second region is often designated by stress plateau that arises from the plastic collapse of the cells in which strain increases at constant or nearly constant stress. The third region corresponds to the increased density of the foam in which the stress rises steeply, causing the cell walls to compress and the material to lose its capability to absorb more energy.

Numerical simulations were performed to validate the constitutive law chosen to model EPS, comparing the results against experimental data obtained from compressive uniaxial tests, as shown in Figures 6 and 7. Figure 5 shows the numerical simulation setup, where the sample consists of a cylinder of diameter D and length L , and the EPS sample is placed between two rigid plates where the bottom plate is fixed and the top plate has one degree of freedom in the axial (vertical) direction, in order to replicate the experimental compressions as shown in Figure 4, performed in a Shimadzu testing machine. The samples were obtained directly from the helmet's liners by cutting it with a band saw. A total of seven samples for each density were tested, six of each at a compression rate of 10 mm/min and one sample of each at a rate of 1 mm/min in order to determine the foam Young's modulus. The compressive load P of 30kN was more than enough to exceed 90% average strain of the sample and as soon as this value was achieved, the test was terminated. The initial dimensions of the EPS samples and the properties obtained from the experimental tests are given in Table IV, where E is the Young's modulus, σ_{co} is the compressive yield stress and ν is the Poisson's ratio.

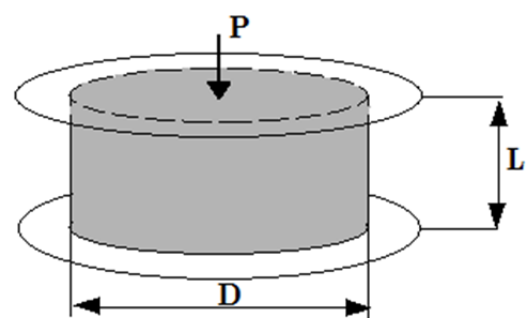
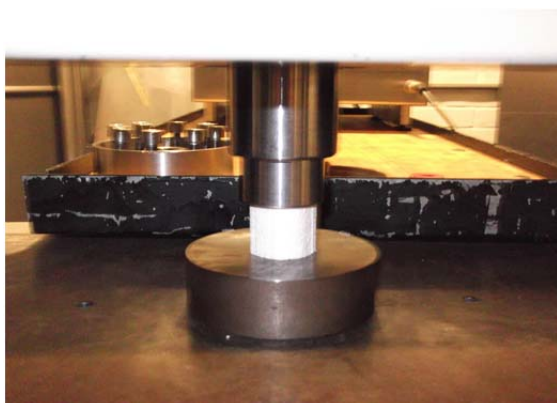


Fig. 4. Setup of the experimental simulation used for mechanical characterization of the EPS foam.

Fig. 5. Setup of the numerical procedure used for mechanical characterization of the EPS foam.

TABLE IV

INITIAL DIMENSIONS AND MECHANICAL PROPERTIES OF EPS FOAM SAMPLES USED TO CHARACTERIZE THE HELMET LINER MATERIAL

ρ [kg/m ³]	D [mm]	L [mm]	E [MPa]	ν	σ_{c0} [MPa]	σ_{c0}/p_{c0}	P_t/p_{c0}
65	33.61	25.75	7.51	0	0.31	1.5	300
90	31.87	19.65	8.64	0	0.61	0.1	4

The punch was numerically modelled as a rigid plate such as the bottom plate. The EPS samples were modelled as deformable solids with four-node tetrahedral elements as all the parts of the liner. To simulate the contact between the sample and the plates, a surface-to-surface type of contact was used. The Explicit version of commercial FE package Abaqus 6.10 [16] was used to perform the simulations.

According to the ECE R22.05 standard, the helmet-headform system is dropped against an anvil with a velocity of 7.5 m/s. Since the helmet's denser liner has an average thickness of 22.5 mm and the less dense liner (main liner) has an average thickness of 32.5 mm, velocities of 6.55 m/s and 5.94 m/s respectively were prescribed to the top plate, in order to guarantee the same strain rates of 333.3 s⁻¹ and 230.8 s⁻¹ respectively. However, these strain rates are much higher than the ones performed experimentally (quasi-static). Nevertheless, it was considered that the deviation from quasi-static to dynamic behaviour of EPS is negligible, following the conclusions of [17] that strain rate effects become pronounced only at rates above approximately 1000/s. Also, [18] performed quasi-static and dynamic tests on EPS and increased the strain rate magnitude several times up to high values and it was concluded that the use of characteristics measured through quasi-static tests does not lead to significant design errors. The EPS foam was modelled as elasto-plastic material, where the elastic behaviour of EPS is modelled with Hooke's law (eq. 1).

$$\sigma = E\varepsilon, \quad (1)$$

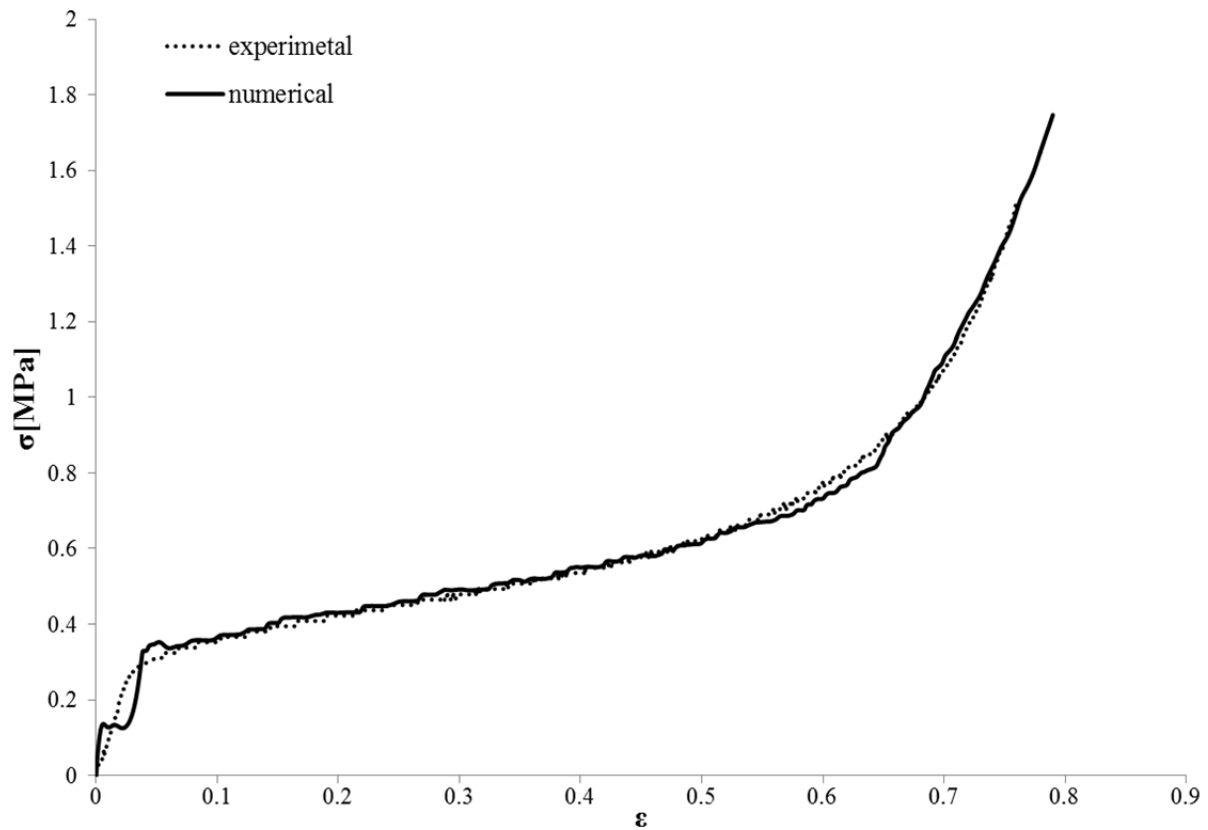
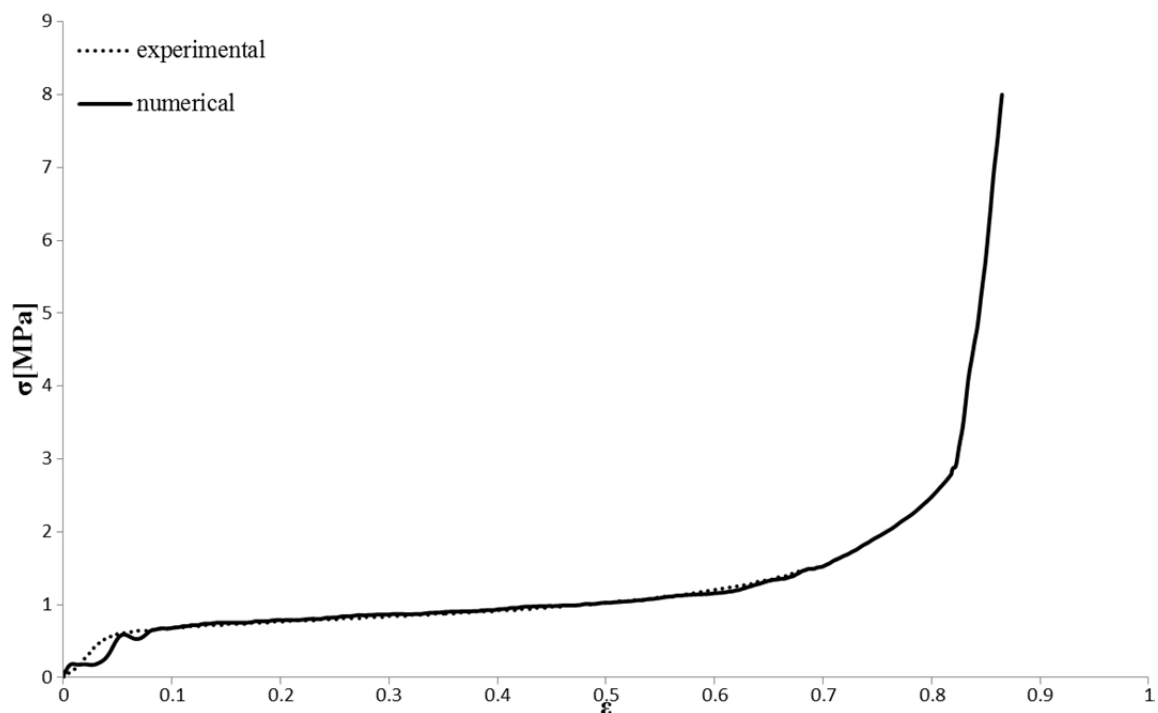
where σ is stress, E is Young's modulus and ε is engineering strain.

To simulate the EPS plastic behaviour, the *Crushable Foam* material model available in Abaqus (suitable for rigid polymeric foams) was used. In addition to the properties presented in Table IV, where p_t/p_{c0} and σ_{c0}/p_{c0} are the ratios of the initial yield pressures in hydrostatic tension and compression respectively, the uniaxial compressive data from the experimental tests were used as input to this model. This model is based on the uniaxial-compressive response of low density closed-cell polymer foams given by:

$$\sigma_c = \sigma_{c0} + \frac{P_0 \varepsilon}{1 - \varepsilon - R}, \quad (2)$$

where σ_c is the engineering compressive stress, P_0 is the effective gas pressure in the cells and R is the foam relative density.

The results of simulations are presented in Figures 6 and 7. It is concluded that the constitutive material models used to simulate the compressive mechanical behaviour of EPS is reasonably adequate, despite the initial instability generated by the contact algorithm.

Fig. 6. Stress-strain curves of 65 kg/m³ density EPS.Fig. 7. Stress-strain curves of 90 kg/m³ density EPS.

2) Modelling of ABS

The outer shell is made from ABS, a widely used material on motorcycle helmets. The ABS is a stiff thermoplastic material very resistant to heat and penetration. The ABS material properties used to model the shell are listed in Table V. To simulate ABS mechanical behaviour, an isotropic linear-elastic material model was considered (Hooke's law). This choice is supported by the fact that during an impact the outer shell is mainly responsible for spreading out the impact's concentrated load and generally deforms only elastically, which is an acceptable simplification for a shell made from a thermoplastic like ABS.

TABLE V
MECHANICAL PROPERTIES OF ABS

$\rho[\text{kg/m}^3]$	$E[\text{MPa}]$	ν
1200	4000	0.37

Boundary Conditions

In order to simulate the interfaces between the headform and the liner and also the interactions between the anvil and the helmet's shell, a surface-to-surface type of contact with a friction coefficient of 0.4 was used. The same type of contact but with a higher friction coefficient of 0.5 was used to model the interaction between the shell and the liner. Also, a "tie" type contact was used to simulate the tie between the different parts of the liner (glued parts). This same type of contact was used to fully constrain the helmet's chin guard relative to the main shell.

According to the ECE R22.05 standard, the helmet-headform system is dropped, without any restriction, against an anvil with a velocity of 7.5 m/s. The simulated impacts were always against flat anvils only, which is enough for a first stage of the model validation. The flat anvil was fixed (fully constrained) and an impact velocity of 7.5 m/s was determined for the model. Figure 8 shows the impact configurations according to the ECE R22.05 standard, the B, P, R and X points. The explicit (dynamic) solver of Abaqus was used to simulate the impacts with durations of 20 ms, with the large deformation option activated. In order to reduce the computational time required, the helmet was placed very close to the anvil.

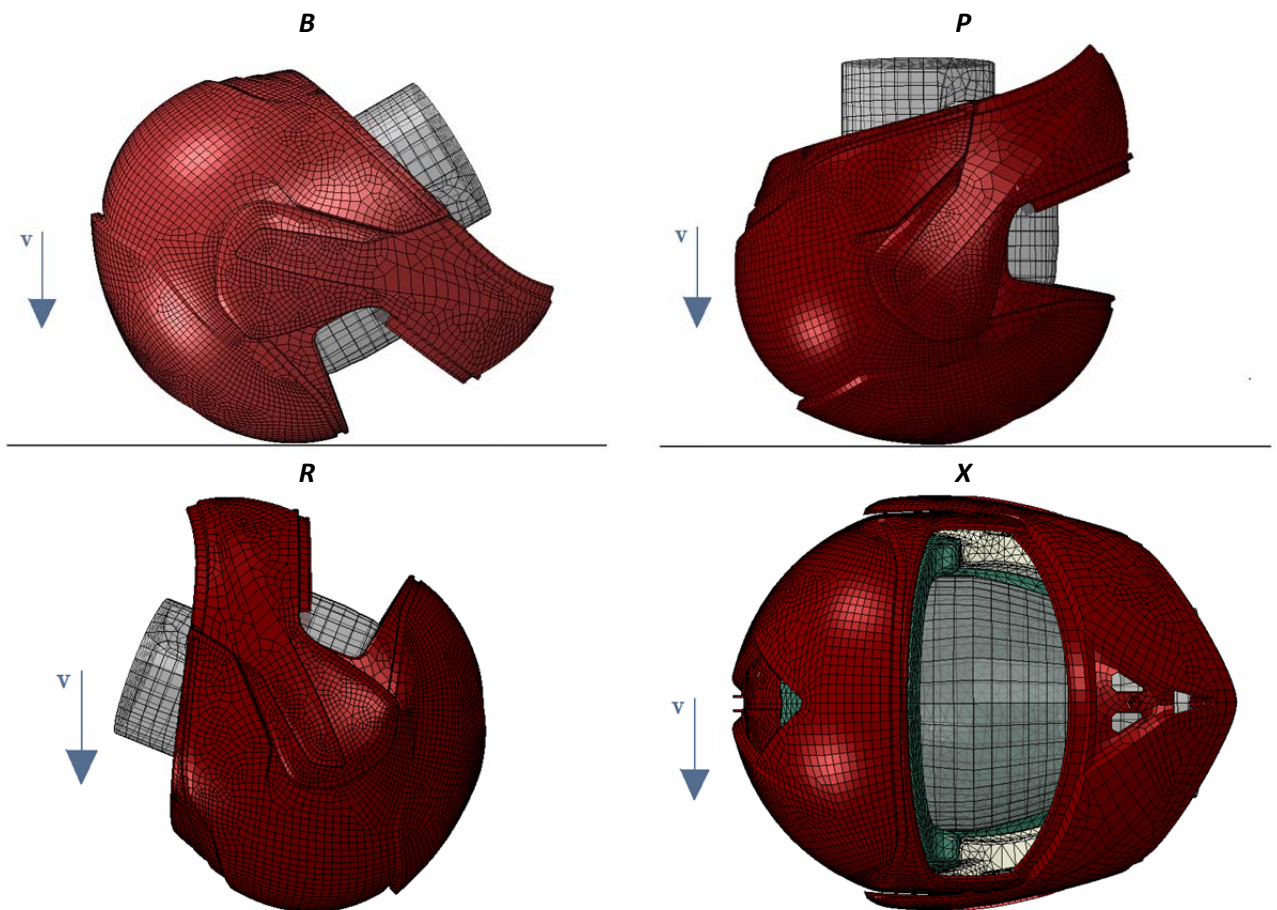


Fig. 8. ECE R22.05 impact configurations.

III. RESULTS

Motorcycle Helmet Model Validation

Numerical simulations of helmeted impacts were performed in order to validate the developed motorcycle helmet model, comparing its results against experimental data from energy-absorption tests required by ECE

R22.05 standard. This comparison based on the acceleration recorded at the headform COG is shown in Figure 9, and the PLA and HIC are assessed as well. The expression used to compute HIC is given by equation 3.

$$HIC = \left(\left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right]^{2.5} (t_2 - t_1) \right)_{max}, \quad (3)$$

where $a(t)$ is the resultant head acceleration (in g's), the interval $(t_2 - t_1)$ are the bounds of all possible time intervals defining the total duration of impact that must be less or equal to 36 ms, and t_1 and t_2 are any two time points of the acceleration pulse (in seconds). The PLA measured at the headform COG and the computed HIC values from numerical and experimental analyses are presented in Table VI.

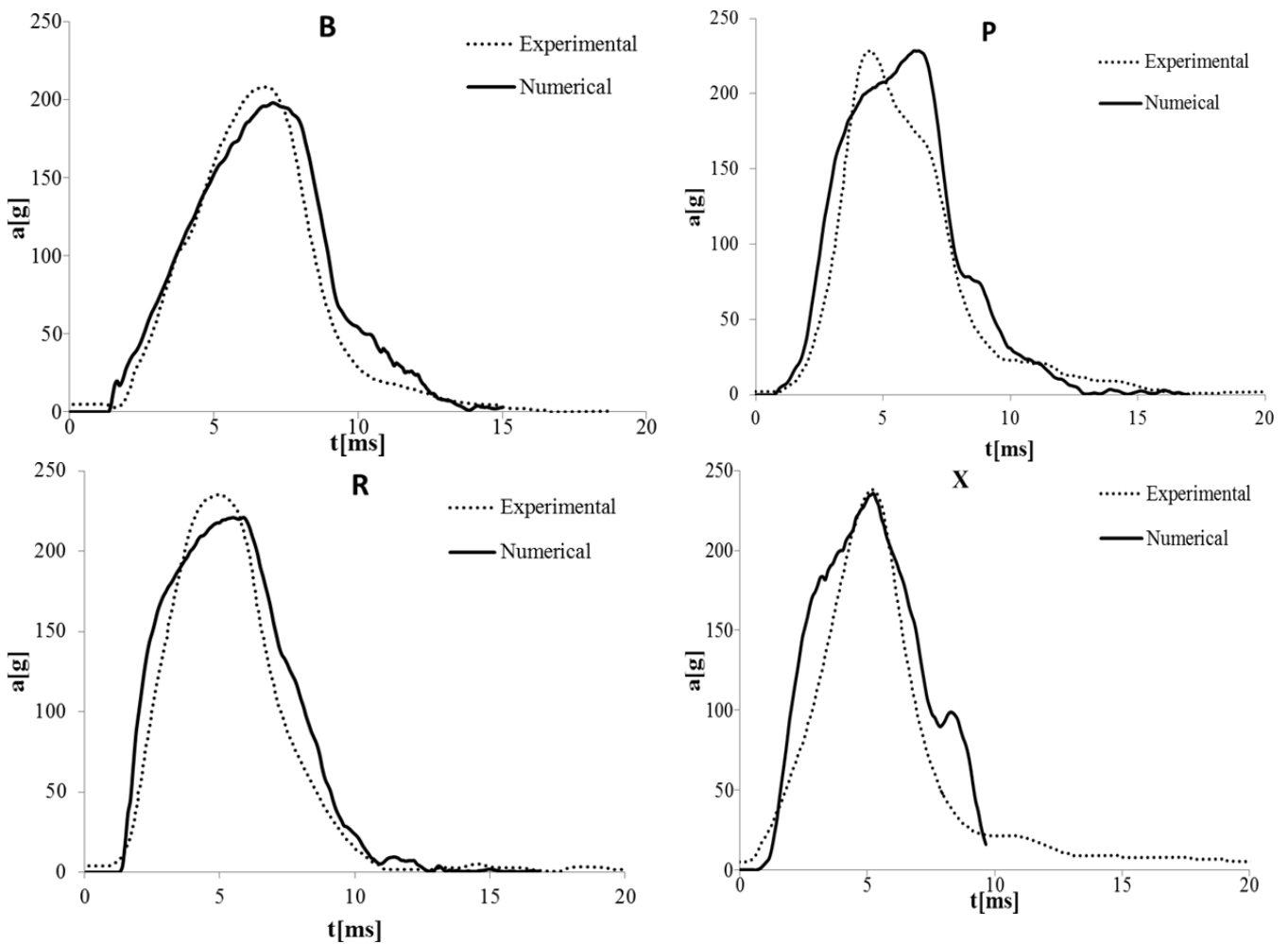


Fig. 9. ECE-R.22/05 impact results.

Overall, there is a good agreement between the experiments and simulations for all four points. The little differences between experimental and numerical results may be explained by the adoption of a simplified numerical model regarding the number of helmet components modelled. For example, the helmet impacted area at point X is a zone that has several parts that were not modelled such as the visor locking system, the chin strap, the fixation system between the two parts of the shell and the comfort padding that has a considerable thickness at this region. Despite some differences between experimental and numerical impact results, the numerical model was considered adequate enough for a preliminary study on linear and oblique impacts with motorcycle helmets.

TABLE VI
MAXIMUM ACCELERATION OF THE HEAD AND HIC VALUES CALCULATED FROM THE NUMERICAL AND EXPERIMENTAL STUDIES.

Impact Point		a_{max} [g] (≤ 275)	HIC (≤ 2400)
Point B	Experimental	208	1696
	Numerical	198.3	1659.1
Point P	Experimental	227	1903
	Numerical	228.5	2061.7
Point R	Experimental	234	2235
	Numerical	221.2	2296.2
Point X	Experimental	237	1714
	Numerical	235.6	2018.6

ECE R22.05 Vertex Impact with SUFEHM

The modelling of biomechanical human head models using FEM provides a strong basis for helmet design improvements over the conventional headforms used in helmet standards, allowing a further accurate computational-based prediction of brain injuries. As a first step, one of the impacts prescribed by the ECE R22.05 standard (Point P) with an advanced biomechanical FE head model was performed in order to assess the helmet from a biomechanical point of view. The FE head model used in this study is the Strasbourg University Finite Element Head Model (SUFEHM) developed in [19]. The linear resultant head acceleration measured in the experimental tests for impact point P is induced at SUFEHM's skull in order to drive SUFEHM's model. Figure 10 shows a cross section of the model and illustrates the anatomical features of the skull and the brain. The main anatomical features modelled were the skull, face, falx, tentorium, subarachnoid space, scalp, cerebrum, cerebellum and brainstem. The SUFEHM model has been validated over the years [20-22] and a short summary of its different validation stages is contained in [23]. The head injury criteria used are based on the SUFEHM head injury predictive tool presented in [24], predicting a 50% risk of subdural hematoma (SDH) for a cerebrospinal fluid (CSF) strain energy of 4950 mJ and a 50% risk of a moderate and severe diffuse axonal injury (DAI) for brain von Mises stress values of 28 and 53 kPa respectively.

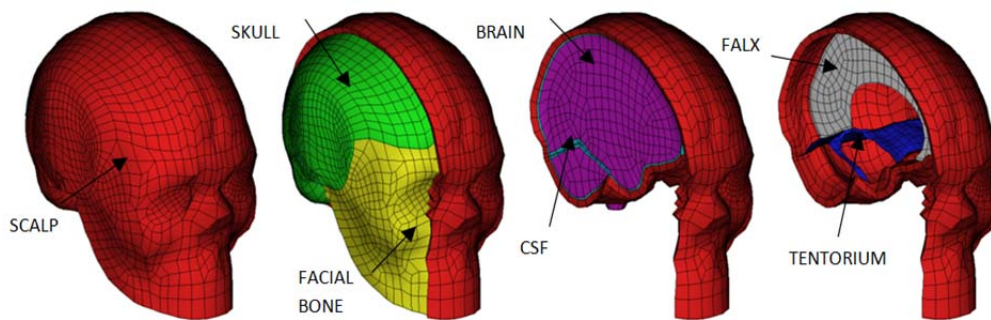


Fig. 10. Strasbourg University Finite Element Head Model.

The results obtained with the SUFEHM in terms of intracerebral field pressure and von Mises stress are illustrated in Figures 11 and 12 respectively. The maximum von Mises stress is located between the brain and the cerebellum and between the cerebellum and the brainstem while the maximum pressure is located in the vertex area. In Table VII, the results in terms of brain von Mises stress value (maximum) and CSF strain energy are summarized. The assessment of head injury risk is expressed in terms of percentage, concluding that for a vertex impact, there is a high risk of moderate neurological injuries (90%) and a 30% risk of subdural hematoma (SDH). Thus, considering that there is a considerable risk of severe DAI and SDH, and that moderate DAI is an almost certain injury (risk of 90.2%), it is clear that there are defects in the current standards, more specifically in their criteria for injury assessment, because an approved helmet does not protect the user in the tests in which it was validated.

TABLE VII
RESULTS OBTAINED WITH SUFEHM IN TERMS OF PERCENTAGE RISK

Impact Point	Brain von Mises stress [kPa]	Risk of moderate DAI (%)	Risk of severe DAI (%)	CSF strain energy [mJ]	Risk of SDH (%)
<i>P</i>	39.7	90.2	19.3	4097	30.3

It has been shown that intracranial von Mises stress is considered a good injury criterion for concussion or other mild traumatic brain injuries (MTBI) for values above 15 kPa [25], while values higher than 20 kPa [26] and 40 kPa [27] were estimated to be a good predictor for concussion.

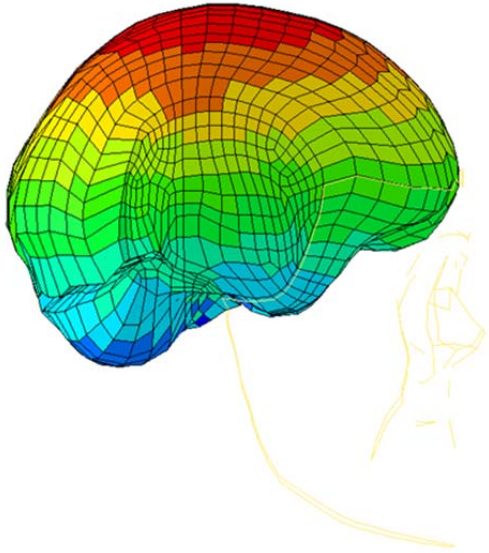


Fig. 11. Illustration of the brain pressure field for an impact in P location.

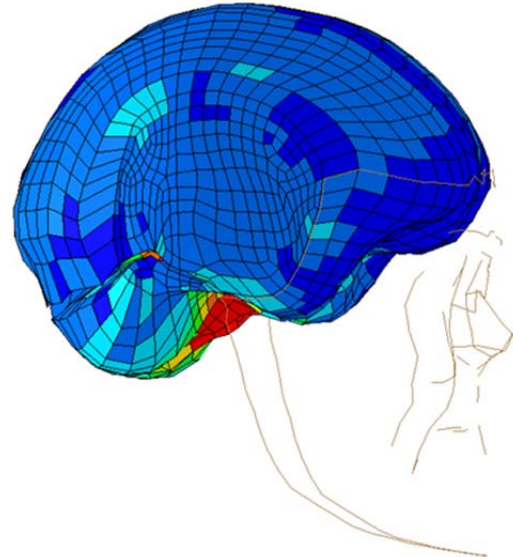


Fig. 12. Illustration of the brain von Mises stress field for an impact in P location.

Thus, by reproducing the same impact that is assessed by the ECE R22.05 standard, it is evident that a certified helmet cannot protect the user from brain injuries such as DAI and possibly concussion, the latter being the most common head injury diagnosis resulting from motorcycle and moped accidents [28]. This shows that limits for the PLA and HIC are not sufficient to assess the protection afforded by motorcycle helmets. Therefore, it can be concluded that a motorcyclist wearing an approved helmet can suffer brain injuries such as cerebral concussion and DAI, based on the analysis of the results from the same impact that was used to certify the motorcycle helmet.

IV. CONCLUSIONS

The main objective of this work was to assess a certified motorcycle helmet, predicting head injuries that can possibly occur in a motorcycle accident. In order to perform such an analysis, a FE motorcycle helmet model was developed. To ensure reliable results, the helmet materials and the helmet-headform system were validated against experimental data of the four energy-absorption tests specified by ECE R22.05. Despite some differences between experimental and numerical impact results, good correlation was found and the numerical model was considered adequate enough for a preliminary study on biomechanical issues.

The acceleration measured in an ECE R22.05 energy-absorption test at point P was induced at SUFEHM's skull. The results were revealing in that a high risk of moderate DAI was predicted and concussion emerged as a real possibility. Thus, these results reveal defects in current motorcycle standards in that a certified helmet cannot protect the user from sustaining brain injuries. Considerable work is still needed to improve helmet standards, including ECE R22.05 and the US DOT FMVSS 218, to ensure that helmets meeting these standards do protect the user from brain injuries such as cerebral concussion and moderate DAI. Some limitations of the standards are underscored from what was seen with SUFEHM where a replication of an impact test that is used to assess the helmets revealed a high risk of moderate brain injury from an impact that was used to validate the same helmet. This supports the premise that the current criteria for motorcycle helmet standards are not optimal. Further, the biofidelity of headforms used in these tests arguably does not result in adequately

reproducing the human head. Another issue is the fact that the current standards do not take into account rotational acceleration and its effects but rather relies only on the headform's peak translational acceleration (PLA) and head injury criterion (HIC). A new standard should take into account tangential and rotational components of the head in both injury criteria and testing procedures. All these conclusions reveal that there is still a long way to go regarding motorcycle helmet safety regulation that greatly affect this type of protective headgear design.

V. ACKNOWLEDGEMENT

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