Structural Engineering and Mechanics, *Vol. 48, No. 5 (2013) 661-679*
DOI: http://dx.doi.org/10.12989/sem.2013.48.5.661

Finite element analysis of helmeted oblique impacts and head injury evaluation with a commercial road helmet

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(Received May 22, 2013, Revised October 26, 2013, Accepted November 1, 2013)

Abstract. In this work, the safety performance of a commercial motorcycle helmet already placed on the market is assessed. The assessed motorcycle helmet is currently homologated by several relevant motorcycle standards. Impacts including translational and rotational motions are accurately simulated through a finite element numerical framework. The developed model was validated against experimental results: firstly, a validation concerning the constitutive model for the expanded polystyrene, the material responsible for energy absorption during impact; secondly, a validation regarding the acceleration measured at the headform's centre of gravity during the linear impacts defined in the ECE R22.05 standard. Both were successfully validated. After model validation, an oblique impact was simulated and the results were compared against head injury thresholds in order to predict the resultant head injuries. From this comparison, it was concluded that brain injuries such as concussion and diffuse axonal injury may occur even with a helmet certified by the majority of the motorcycle helmet standards. Unfortunately, these standards currently do not contemplate rotational components of acceleration. Conclusion points out to a strong recommendation on the necessity of including rotational motion in forthcoming motorcycle helmet standards and improving the current test procedures and head injury criteria used by the standards, to improve the safety between the motorcyclists.

Keywords: biomechanics, finite element method, head injury, motorcycle helmet, oblique impact

1. Introduction

Road accidents are one of the major causes of death in the world (WHO 2009). In the European Union, about 31 thousand people die and 1.6 million people are injured every year as a direct result of road accidents (ERSO 2012). Motorcyclists, protected by the motorcycle helmet, are less protected in road accidents than the users of 4-wheeled vehicles because the last ones are protected by safety belts, airbags and even by the car's body structure (Fernandes and Alves de Sousa 2013). Thus, motorcycle crash victims form a high proportion of those killed and injured in road accidents. For example, in Portugal, 21% of all road accident fatalities and 24.9% of all road accident severe injuries in 2011 were sustained by powered two-wheeler (PTW) occupants (ANSR 2010, 2011), as shown in Table 1.

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http://www.techno-press.org/?journal=sem&subpage=8 ISSN: 1225-4568 (Print), 1598-6217 (Online)

Table 1 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 a common result from motorcycle accidents, as shown in COST321 (2001), where head injuries occurred in 66.7% of COST database cases and the majority of these injuries were severe. For these reasons, head protection and safety are very important.

Nowadays, motorcycle helmet standards are responsible for helmets quality and effectiveness by restricting the market just for the ones that are able to fulfil all the requirements. However, there are some issues regarding for example the head injury criteria, once current standards do not take into account the rotational acceleration and its effects by relying only in headform's peak translational acceleration (*PLA*) and in head injury criterion (*HIC*). Newman (2005) highlighted these same issues, such as the lack of progress, and the current use of old fashioned test methods that do not properly reflect the real circumstances of accidents, like the biofidelity of the headform, the nature of the failure criteria, as well as the impact kinematics, which influences the movement of a tested helmeted-headform. The linear tests demanded by the standards are criticized due to not being quite equivalent to a real impact. Basically, the most frequently and severe injuries sustained in motorcycle accidents are head injuries, being many of these injuries caused by rotational forces (Aare *et al.* 2004, Gennarelli 1983), commonly generated as a result of oblique impacts found in motorcycle crashes (Otte *et al.* 1999). When these forces act on the head, the results are large shear strains in the brain, which have been proposed as a cause of traumatic brain injuries (TBI), such as diffuse axonal injury (DAI) and subdural haematoma (SDH) due to the tearing of neuronal axons in brain tissue and the rupturing of bridging veins respectively (Gennarelli 1983, Margulies and Thibault 1992).

Motorcycle helmets are composed by to main parts, having a hard outer shell that prevents penetration and distributes the impact force on a wider foam area, increasing the liner capacity to absorb energy, reducing the load that reaches the head. The advance of CPU power allied with the issues of experimental methods, such as time, cost, flexibility and repeatability, led to the development of virtual models, mainly using the Finite Element Method (FEM) to compute several variables such as the stresses and strains, allowing to assess the influence of the interaction between helmet and head, providing much more information about the helmet's impact. These types of models have been used in many applications, for example the study of motorcycle helmet materials (Alves de Sousa *et al.* 2012, Coelho *et al.* 2013). Thus, with virtual testing 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 in this work are the analysis of a commercially available helmet, approved by the ECE R22.05 standard, assessing its capacity to protect or not the user from head injuries that result from helmeted impacts, creating a FE motorcycle helmet model based on realistic geometric info and known material properties to perform such task. 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, an oblique impact is simulated, and the results are compared against proposed injury thresholds. It can be concluded that rotational motion should be considered in the

Fig. 1 CMS SUV Apribile motorcycle helmet

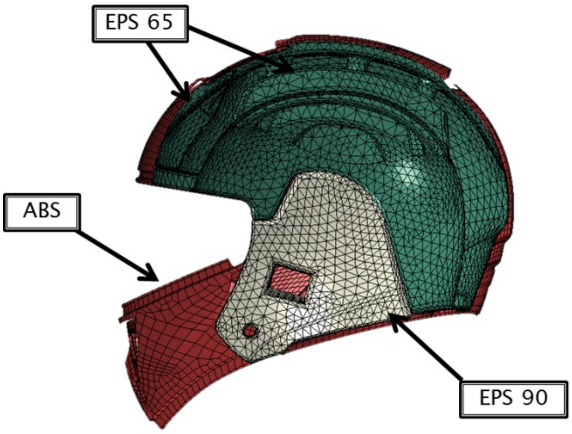


Fig. 2 Finite element helmet model - a cut view from the sagittal plane

standards tests and criteria, establishing a solid ground for further improvements on motorcycle helmets.

2. Finite element modelling

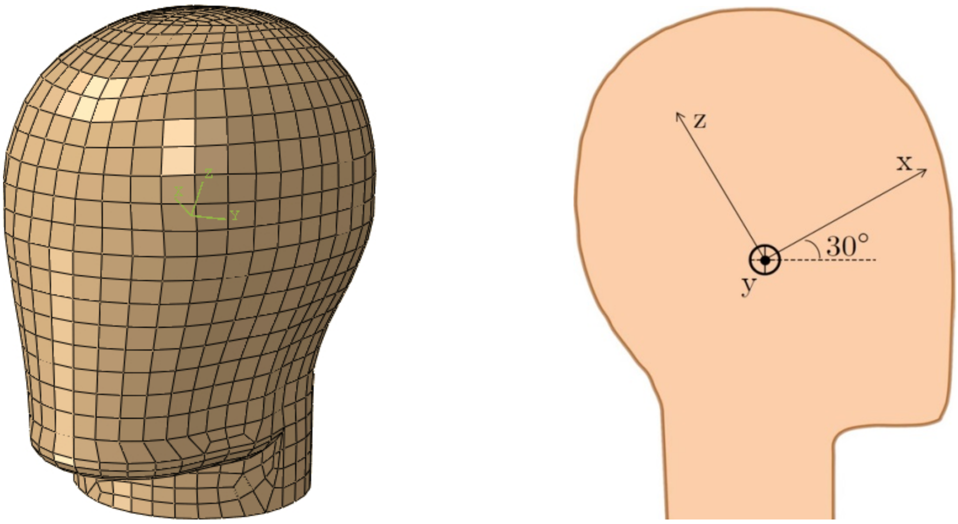
The motorcycle helmet used in this work is the CMS SUV Apribile modular motorcycle helmet manufactured by CMS Helmets and presented in Fig. 1. This commercially available helmet fully meets ECE R22.05 regulation (ECE R22.05 2002), the Brazilian Regulation NBR-7471 (ABNT 2001) and also the U.S. Regulation DOT (U.S. Department of Transportation 2012).

2.1 Motorcycle helmet modelling

The geometry of the helmet was provided by the manufacturer. The 3D CAD models of each helmet part were treated on CATIA V5R19 CAD system (CATIA 2008). Only slight simplifications were made to the models, making the model easier to mesh but maintaining the overall geometry.

Table 2 Headform mass and principal inertial moments

Mass [kg]	I_{xx} [kg.cm^2]	I_{yy} [kg.cm^2]	I_{zz} [kg.cm^2]
5.6	370	440	300



(a) Headform finite element model

(b) Illustration of the coordinate system used to apply the inertial moments

Fig. 3 ECE 22.05 size M headform

Fig. 2 shows the FE helmet model developed, which includes the shell and the dual-density energy absorbing liner composed by three different parts: the main padding that involves the entire cranium with the exception of temporal region, the forehead padding insert and the lateral padding (positioned in the temporal region). The different meshes characteristics are shown in the Fig. 2 and detailed in Table 3. 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.

According to ECE R22.05 regulation it is used a 5.6 kg headform (M size). The FE headform model used in the simulations is shown in Fig. 3a and its inertial characteristics are given in Table 2. In Fig. 3b, it is shown the coordinate system used to apply the principal inertial moments. After assembling all the helmet components, the headform was fitted in the helmet as shown in Fig. 7.

2.2 Finite element mesh

The FE motorcycle helmet model was created using four-node linear tetrahedral elements (Abaqus's C3D4 element) to mesh the foams, where the main liner was modelled using 65426 elements, the forehead insert was modelled using 17949 elements and the lateral liners were modelled with 14034 elements for the left one and 12463 to the right one. This type of element was used to model the foams mainly due to its complex geometry. On the other hand, the shell was

Table 3 Characteristics of meshes used to model the different helmet parts

Part	Element type	Abaqus's element	Nº of elements	Nº of nodes
Main covering shell	Four-node, reduced integrated linear shell	S4R	9385	9587
Chin guard shell			2569	2723
Main liner	Four-node linear tetrahedron	C3D4	65426	14073
Forehead liner			17949	4171
Left lateral liner			14034	3318
Right lateral liner			12463	2983
Headform	Rigid quadrangular shell	R3D4	1346	1348
Anvil			1	4

modelled with 9385 four-node linear shell elements with reduced integration (Abaqus's S4R element) with enhanced hourglass control for the main covering shell and 2569 to the chin guard, which makes a total of 11954 elements to the shell. The headform and the flat anvil were modelled with rigid quadrangular elements (Abaqus's R3D4 element), 1346 and 1, respectively. Between the anvils required by ECE R22.05, only the flat anvil was modelled because it is the most common object (flat road surface) hit by the head in motorcycle crashes (Shuaeib *et al.* 2002, Vallée *et al.* 1984), treated here as a rigid body.

The meshes were created always avoiding distorted and warped elements and with especial 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. A summary of the meshes is presented in the Table 3.

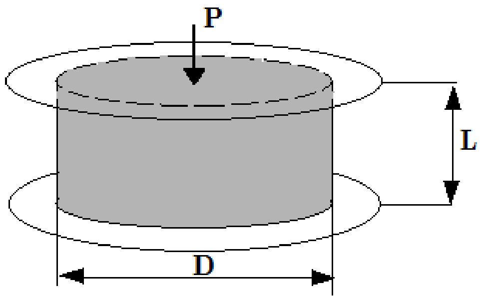
2.3 Material modelling

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.

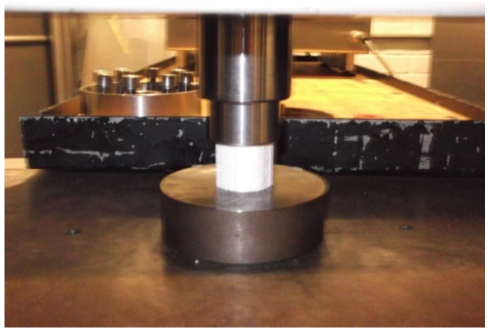
2.3.1 Modelling of EPS foam

EPS foam is a synthetic cellular material with closed cells and widely applied on energy absorption applications, such as protective gear, being the most common liner material used in motorcycle helmets, mainly due to its excellent shock absorption properties and low cost. 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 Figs. 5-6. 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 and the third corresponds to the densification of the foam in which the stress rises steeply, where cell walls are mostly compressed and the material loses 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 Figs. 5-6. Fig. 4(a) shows the numerical simulation setup, where the sample consists in a



(a) Setup of the numerical simulation used for mechanical characterization of the EPS foam



(b) Setup of the experimental procedure used for mechanical characterization of the EPS foam

Fig. 4 Mechanical characterization of the EPS foam

Table 4 Initial dimensions and mechanical properties of EPS foam samples used to characterize the helmet liner material

$\rho[kg/m^3]$	$D[mm]$	$L[mm]$	$E[MPa]$	v	$\sigma_{c0}[MPa]$	p_t/p_{c0}	σ_{c0}/p_{c0}
65	33.61	25.75	7.65	0	0.31	1.5	300
90	31.87	19.65	8.64	0	0.61	0.1	4

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 shown in Fig. 4(b), 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 on Table 4, where E is the Young's modulus, σ_{c0} is the compressive yield stress and v is the Poisson's ratio.

The punch was numerically modelled as a rigid plate. On the other hand, 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, it was used a surface-to-surface type of contact. The Explicit version of commercial FE package Abaqus 6.10 (Abaqus 6.10 2010) 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 denser liner has an average thickness of 22.5 mm and the less dense 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 rates). Nevertheless, it is considered that the deviation from quasi-static to dynamic behaviour of EPS is negligible, following the conclusions of Ouellet *et al.* (2006) that strain rate effects become pronounced only at rates approximately above 1000 s^{-1}. Also, Di Landro *et al.* (2002) performed quasi-static and dynamic tests on EPS and increased the strain rate magnitude several times up to high values and 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, function of Young's modulus E and Poisson's ratio v. To simulate the EPS plastic behaviour, the Crushable Foam material model available in Abaqus was employed. In addition to the properties presented in Table 4, 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 necessary as inputs. This model is based on the uniaxial-compressive response of low density closed-cell polymer foams given by

$$\sigma_c = \sigma_{c0} + \frac{P_0 \varepsilon}{1-\varepsilon-R} \tag{1}$$

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 (foam density divided by solid polymer density).

The results of simulations are presented in Figs. 5-6 and it is concluded that the constitutive material models used to simulate the compressive mechanical behaviour of EPS are reasonably adequate, despite the initial instability generated by the contact algorithm.

2.3.2 Modelling of ABS

The outer shell is made from ABS, a material commonly used on motorcycle helmets as shell material. 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 5. In order to simulate ABS mechanical behaviour, an isotropic linear-elastic material model was considered (Hooke's law). This choice is an acceptable simplification for a shell made from a thermoplastic like ABS, being 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.

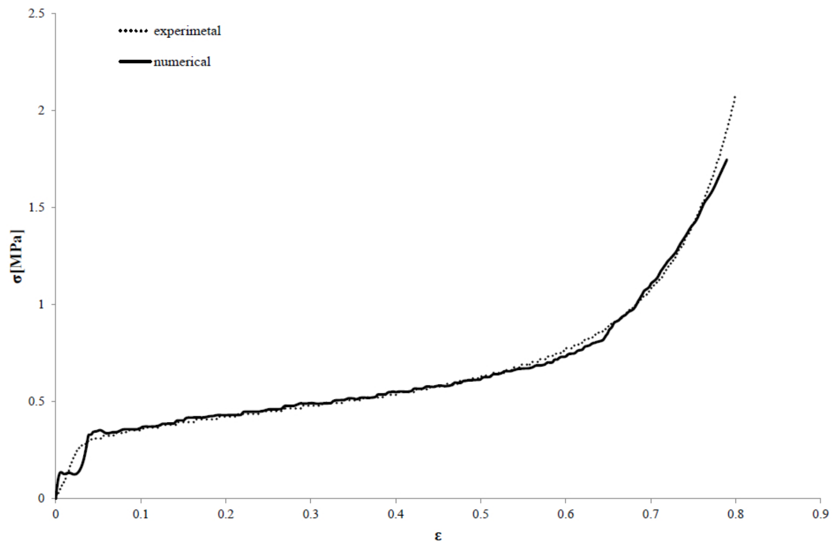


Fig. 5 Stress-strain curves of 65 kg/m^3 density EPS

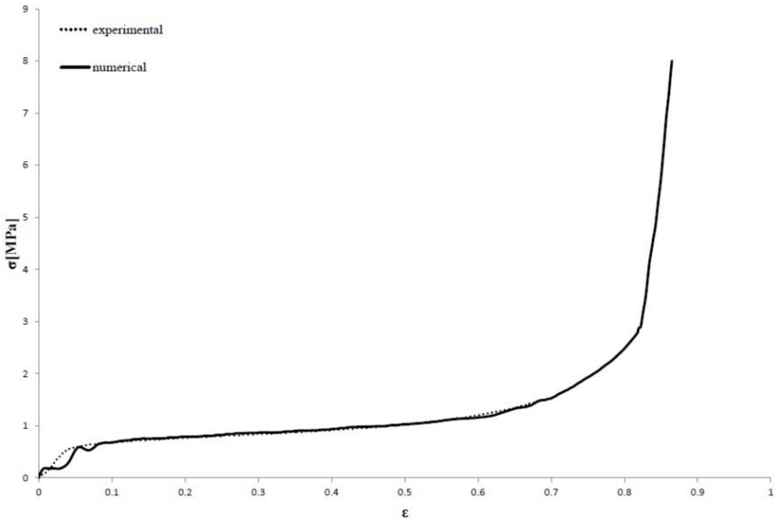


Fig. 6 Stress-strain curves of 90 kg/m^3 density EPS

Table 5 Mechanical properties of ABS

ρ[kg/m^3]	E[MPa]	ν
1200	4000	0.37

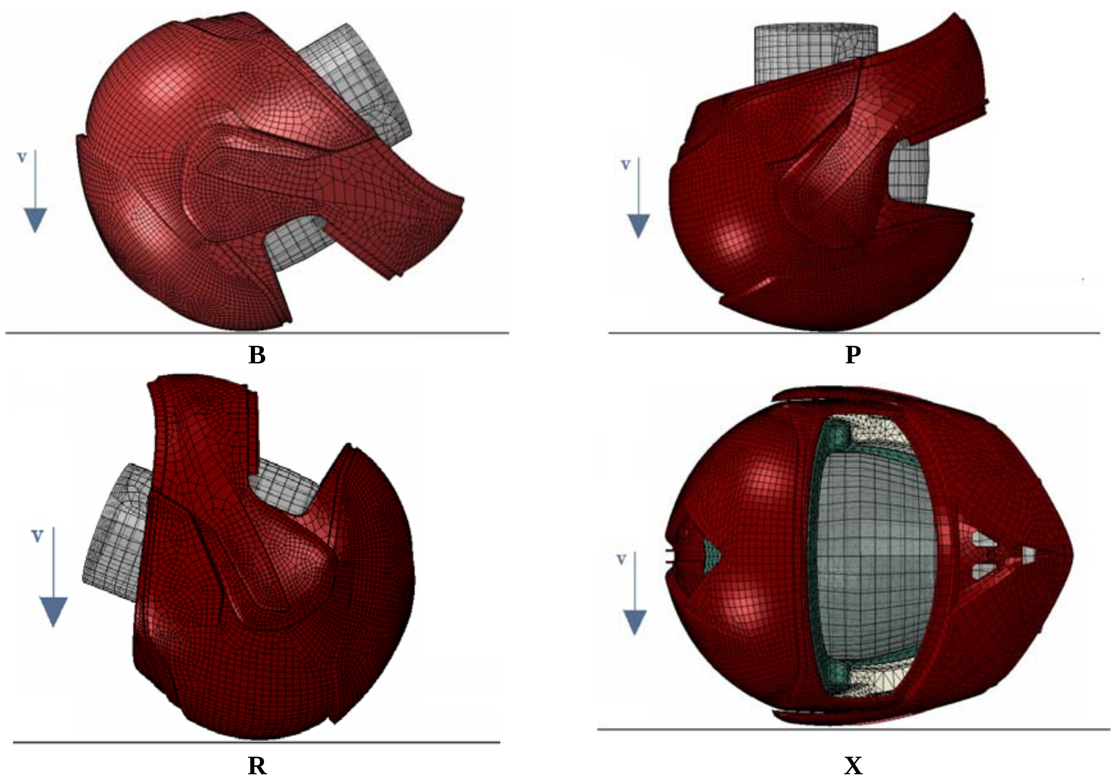


Fig. 7 ECE R22.05 impact configurations

2.4 Boundary conditions

In order to simulate the interfaces 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 friction coefficient of 0.5 was used to model the interaction between the shell and the liner and between the headform and the liner. Also, a "tie" type contact was used to simulate the tie between the different parts of the liner (glued parts). A "tie" was also used to fully constrain the helmet's chin guard relatively 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 prescribed to the model. Fig. 7 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 large deformation option activated. In order to reduce the computational time required, the helmet was placed very close to the anvil.

2.5 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 demanded by ECE R22.05 standard. This comparison based on the acceleration recorded at the headform's centre of gravity (COG) is shown in Fig. 8 and the *PLA* and *HIC* are assessed as well. The expression used to compute *HIC* is given by the Eq. (2).

$$HIC = \left(\left[\frac{1}{t_2-t_1}\int_{t_1}^{t_2} a(t)dt\right]^{2.5}(t_2-t_1)\right)_{max} \tag{2}$$

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 6.

Table 6 Headform maximum acceleration and HIC values calculated from the numerical and experimental studies

Impact point		a_{max} [g] (≤275)	*HIC* (≤2400)
Point B	Numerical	213.9	1876.8
	Experimental	208	1696
Point P	Numerical	228.5	2061.7
	Experimental	227	1903
Point R	Numerical	221.2	2296.2
	Experimental	234	2235
Point X	Numerical	235.6	2018.6
	Experimental	237	1714

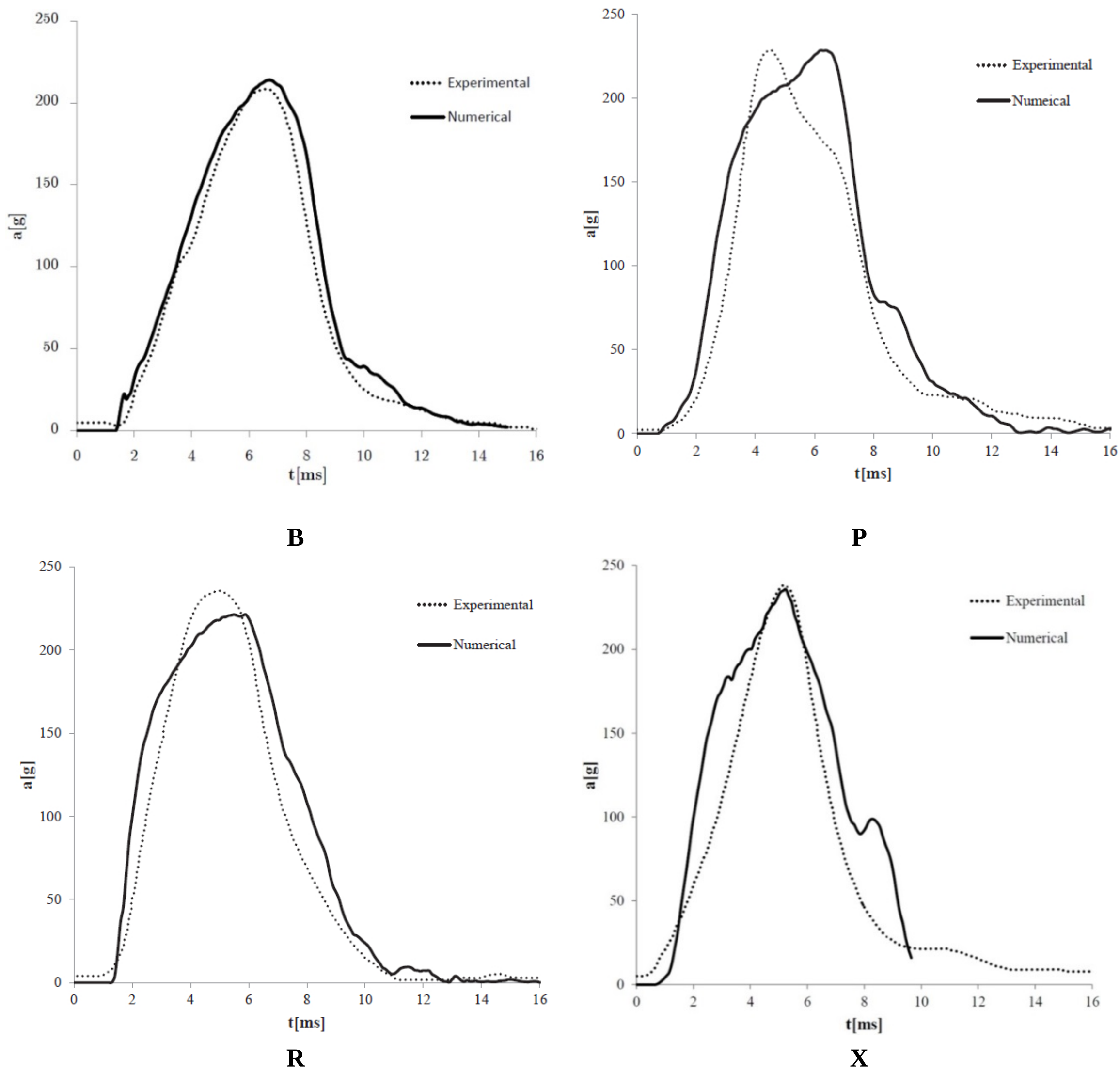


Fig. 8 Translational acceleration measured at the headform COG in simulations and experiments for each one of ECE R22.05 impact points

In overall, there is a good agreement between experiments and simulations for all four points (Fernandes *et al.* 2013). 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, since 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.

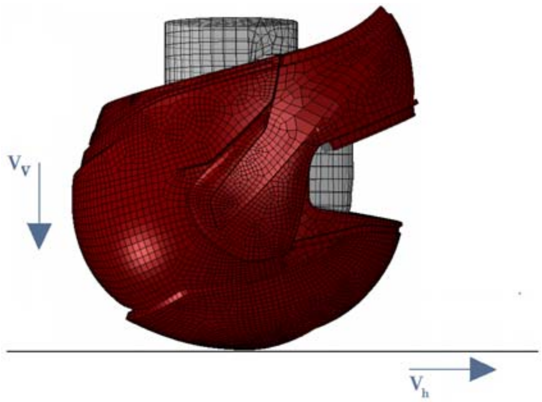


Fig. 9 Oblique impact - numerical setup

2.6 Oblique impact

After the full model validation carried out in the previous sections, rotational and translational parameters from a well-defined impact in real motorcycle accidents (Otte *et al.* 1999) were used to simulate the oblique impact. Results were compared against different thresholds proposed in several studies that will be presented in the results section.

In the simulated oblique impact, the vertex was defined as the impact location. In the present case, corresponds to the same region impacted previously to validate the helmet model (point P). Rotation was induced at the sagittal plane. This impact scenario was one of the most common impacts found in real motorcycle crashes (Otte *et al.* 1999). There is a great number of brain injury thresholds proposed for this type of impact with rotation on sagittal plane.

The oblique impact consists in a velocity of 7 m/s with an impact angle of 30°. The angle was induced by altering the speed of the helmet-headform system (vertical impact speed v_v) and the speed of flat road (horizontal impact speed v_h) to 3.50 m/s and 6.06 m/s respectively, as Aare *et al.* (2004) for the same configuration. This configuration is shown in Fig. 9. The interfaces between the model parts have the same properties except the interaction between the anvil and the helmet's shell, where the friction coefficient of 0.4 was updated to 0.55 in order to model the interaction between the shell and the road. These same contact properties were used by Mills *et al.* (2009) to perform oblique tests.

3. Results

In order to predict any resultant injury, the results from oblique impact are assessed by comparing it against head injury thresholds based on linear motion, on rotational motion and others based on both motions. Several thresholds are used, such as the peak of rotational acceleration, the change in rotational velocity and its peak, the combination of these with *PLA* and *HIC* and other methods based on same concept of rotational and translational combined effects, such as Head Injury Power (HIP), Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) and even one proposed by Aare *et al.* (2004) that predicts the maximum strain in the brain tissue. The results obtained from the oblique impact simulation are given in the Table 7.

Table 7 Results computed from the oblique impact simulation

a_{max}	α_{max} *[rad/s²]*	$\Delta\omega$ *[rad/s]*	*HIC*	*HIP[W]*	*GAMBIT*	$\varepsilon_{brain\ tissue}$
111.50	6175.13	33.12	430.57	23417.00	0.7525	0.2090

In Fig. 10, it is shown the translational acceleration, the rotational acceleration and the rotational velocity measured during the impact simulation. The conclusions of Ueno and Melvin (1995) are verified, in the sense that linear results of oblique impact, such as *PLA* and *HIC* were considerably lower than the ones find in the linear impacts due to the introduction of an angular component. Also, the angular component induced in this impact is much greater due to the angle of impact chosen. In other words, the greater velocity of the anvil relatively to the velocity of the helmet-headform system, the higher is the rotational component.

3.1 Peak translational acceleration, peak rotational acceleration and change of rotational velocity

The results given in Table 7 and shown in Fig. 10 were compared with established thresholds found in the literature and a summary of the limits that were exceeded is presented:

• Gennarelli *et al.* (2003) hypothesized a magnitude of rotational acceleration of 2877.8 rad/s^2 required to induce mild cerebral concussion at an angular velocity of 25 rad/s. Observing Fig. 10, it is possible to conclude that higher values of rotational acceleration are observed at an angular velocity of 25 rad/s.

• Ommaya *et al.* (1967) proposed a 50% probability of cerebral concussion for a maximum rotational acceleration of 1800 rad/s^2 during a period of time inferior to 20 ms. This is much lower than the maximum determined in this study for the same period. A value of 1800 rad/s^2 was also predicted by Kleiven (2007a) as a threshold for cerebral concussion. More thresholds for cerebral concussion were proposed by Fijalkowski *et al.* (2006b) and Newman *et al.* (2000b), agreeing that 6200 rad/s^2 is a good predictor, where the last one is indicated with a 50% probability of concussion. This last threshold was not exceeded but the values are somewhat close.

• Löwenhielm (1974) predicted rupture of bridging veins for rotational acceleration values higher than 4500 rad/s^2. In Löwenhielm (1974) several brain injuries were predicted for the same threshold, such as brain injuries classified as AIS 5, which corresponds to a set of severe injuries like EDH, SDH and also DAI.

• Advani *et al.* (1982) predicted brain superficial tissue shearing for rotational accelerations higher than 2000 - 3000 rad/s^2, which was also exceeded in this study.

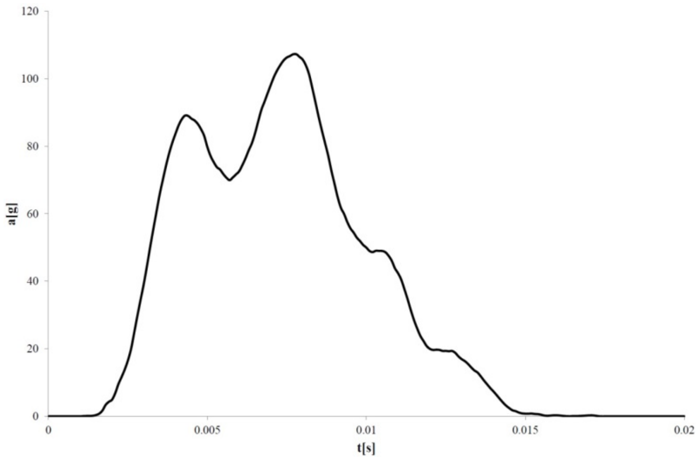
• Thomson et al. (2001) predicted the occurrence of brain injuries when rotational accelerations higher than 5000 rad/s^2 are found, as in this case.

• Davidsson *et al.* (2009) proposed a 50% probability of mild traumatic brain injury (MTBI) for a maximum rotational acceleration of 5900 rad/s^2, which was exceeded.

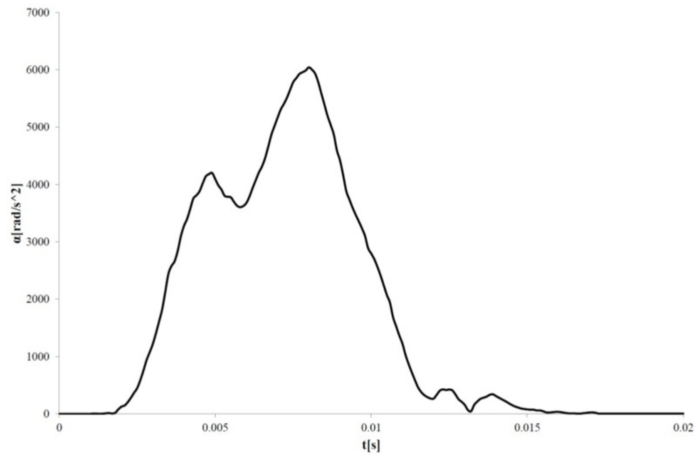
• King *et al.* (2003) suggested that a rotational velocity change of 19 rad/s is a good threshold for DAI in an exposition to sagittal plane rotation. A higher change was verified in this study. The authors also proposed a threshold of 5757 rad/s^2 with 50% of probability.

• Zhang *et al.* (2004) proposed rotational acceleration as a threshold for MTBI defining a probability of 50% for MTBI occurrence regarding values higher than 5757 rad/s^2.

• COST327 motorcyclist's helmet working group (COST327 2001) performed a work where it



(a)



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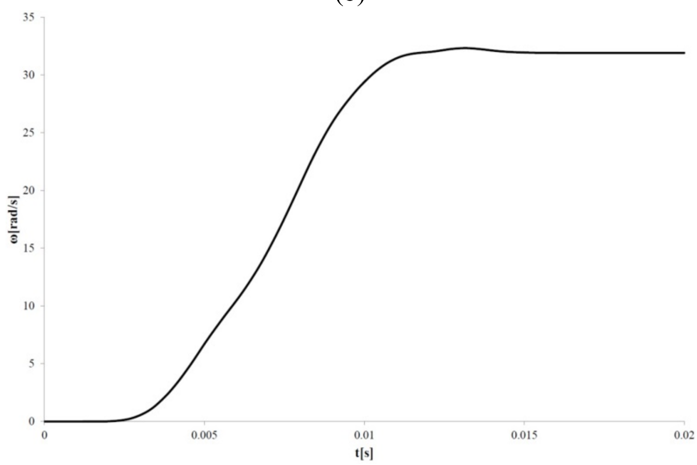

(c)

Fig. 10 Results of oblique impact simulation: (a) translational acceleration, (b) rotational acceleration and (c) rotational velocity was found a 10% probability of head injury with AIS 2-5 given from rotational acceleration values varying between 5000 and 6000 rad/s^2. In the present study, higher rotational accelerations were obtained.

• Ommaya (1985) predicted brain injuries classified as AIS 5 for rotational velocities and accelerations higher than 30 rad/s and 4500 rad/s^2 respectively.

The majority of these head injury thresholds were determined in studies where the rotational motion was induced to the sagittal plane, exactly as replicated in this study, which makes these comparisons more valid. From all the literature reviewed several types of head injuries were predicted, with more insight concerning concussion, as a higher numbers of thresholds were presented for this particular injury. Several of these were exceeded in this analysis, predicting concussion as an almost certain scenario. Thus, it can be said that at least cerebral concussion is predicted for the simulated oblique impact with the ECE R22.05 homologated motorcycle helmet under study. Moreover, concussion can be considered the most common head injury diagnosis resulting from motorcycle and moped accidents (Aare and von Holst 2003).

3.2 GAMBIT

Newman (1986) introduced a head injury assessment function that takes into account both translational and rotational accelerations, in an attempt to combine both into one injury criterion. In fact, they independently cause severe stresses in the brain, resulting in brain injury. On the assumption that translational and rotational acceleration equally and independently contribute to head injury, the GAMBIT expression is

$$G(t) = \left[\left(\frac{a(t)}{a_c}\right)^n + \left(\frac{\alpha(t)}{\alpha_c}\right)^m\right]^{1/s} \tag{3}$$

where *a(t)* and *α(t)* are the instantaneous values of translational (expressed in [g]) and rotational acceleration respectively (expressed in [rad/s^2]); n, m and s are empirical constants selected to fit available data; a_c and α_c represent critical tolerance levels for those accelerations.

Here, the GAMBIT function was computed by replacing the simulated output accelerations and the constants used by Newman (1986) ($n = m = s = 2$, $a_c = 250$ g and $\alpha_c = 25000$ rad/s^2) in the expression 3. Nevertheless, the value used for α_c was 10000 rad/s^2 as proposed by COST327 (2001) and used by Mellor and StClair (2005). A GAMBIT value of 0.7525 was computed. Newman *et al.* (2000b) reported a 50% probability of concussion for a GAMBIT value superior to 0.4, which is almost half of the value calculated in this study. This means that there is a high probability of concussion. GAMBIT has recently been employed in some studies involving protective headgear (Newman *et al.* 2000a, COST327 2001), where it was highlighted the good capability of this method for predicting cerebral concussion. Again, concussion was predicted using a different criterion that takes into account both motions.

3.3 HIP

The HIP criterion was also used to predict resultant injuries from the simulated oblique impact. Newman *et al.* (2000b) proposed HIP to scale the impact power for different directions, depending on the tolerance level for the actual direction. More details about this and the other criteria used in this study can be found in Fernandes and Alves de Sousa (2013).

The HIP is expressed by an empirical expression that relates a measure of power to head injury

$$HIP = Aa_x \int a_x\, dt + Ba_y \int a_y\, dt + Ca_z \int a_z\, dt + D\alpha_x \int \alpha_x\, dt + E\alpha_y \int \alpha_y\, dt + F\alpha_z \int \alpha_z\, dt \tag{4}$$

The coefficients A, B and C represent the mass of the human head, m [kg] and D, E and F represent the appropriate moments of inertia for the human head I_{xx}, I_{yy} and I_{ZZ} [kg.m^2] respectively, which denote the injury sensitivity for each of the six degrees of freedom of the head. The headform inertial properties used are the same indicated in Table 3(a). Newman *et al*. (2000b) estimated a 50% probability of concussion for a HIP of 12.8 kW, and a *HIC* value of 240 for a 50th percentile head.

The HIP value computed in the present study is 23417 W, from equation 4 where the constants (mass and principal moments of inertia) are substituted by the values given in Table 3(a). The remaining variables of each principal direction are substituted by the output obtained in the simulations. The HIP value computed is almost twice the value proposed by Newman *et al.* (2000b) for 50% probability of concussion at a maximum HIP of 12.8 kW. However, this threshold was proposed based on the use of average values for a 50th percentile adult male, while in this study an M size headform is used and not the medium size J. This clearly affects the results since mass and accelerations are higher, with greater HIP values expected. The computed value is slight below the limit of 24 kW predicted by Marjoux *et al.* (2008) for moderate neurological injury.

3.4 Brain tissue strain

Aare *et al.* (2004), after developing a test rig to perform oblique impacts, have tried to develop a criterion that correlate both translational and rotational acceleration with strains in brain tissue. In a previous work of this group, Kleiven and von Holst (2003) found that the change in angular velocity corresponds best with the intracranial strains found in a FE model. For translational impulses on the other hand, the *HIC* and the *HIP* have shown the best correlations with the strain levels found in the model (Kleiven and von Holst 2003). Thus, as the strains in the brain tissue are proportional to the HIC value for pure translations, and also proportional to the change in rotational velocity for pure rotations of short impact durations (Kleiven and von Holst 2003), it was suggested by Aare *et al.* (2004) that the output data can be fitted to the following formula

$$\varepsilon = k_1 \Delta\omega + k_2 HIC \tag{5}$$

where ε is the maximum strain component at the brain tissue, $\Delta\omega$ is the peak resultant change in rotational velocity, k_1 and k_2 are constants. These constants were obtained by regression analysis for each type impact and are available in Aare *et al.* (2004). For the impact simulated in this work, the constants $k_1 = 6.14\times10^{-3}$ and $k_2 = 1.32\times10^{-5}$ were used. No threshold was proposed by Aare *et al.* (2004), however the authors used the thresholds proposed by Bain and Meaney (2000), where a strain of 20% was shown to be critical to the brain tissue, and the maximal principal strain in the brain tissue was chosen as a predictor of injuries, once it has been shown to correlate with DAI.

The brain tissue strain computed using Eq. (5) is 20.90%, which is slightly greater than the limit of 20% proposed by Aare *et al.* (2004) and predicted by Bain and Meaney (2000) as cause of brain injuries such as DAI. This same threshold was presented by Morrison III *et al.* (2003) to predict DAI. A lower value of 15% was proposed by Thibault *et al.* (1990) as a critical level to brain strain, in order to predict DAI. Also, Kleiven (2007a) proposed strain brain tissue as a predictor of concussion and DAI suggesting 0.1 and 0.2 as thresholds, respectively. Shreiber *et al.* (1997) derived a threshold of 19% in principal strain in the cortex for a 50% risk of cerebral contusions, a value that was herein exceeded. Margulies and Thibault (1992) proposed 0.05 to 0.10 of strain, in order to produce moderate-to-severe DAI. Kleiven (2007b) also used first principal strain as a criterion to predict DAI, proposing a 50% probability of DAI for values of 0.21 for first principal strain in the corpus callosum. Recently, Wright and Ramesh (2012) defined an axonal injury threshold value of 18% of strain. All these thresholds were exceeded in this study, predicting the occurrence of brain injuries. In a similar study performed by Aare *et al.* (2004), it was found that the change in rotational acceleration was the parameter that corresponds to a better correlation with the intracranial strains. This was also concluded by Kleiven and von Holst (2003).

The limit proposed between the different studies can be considered very similar in the sense that all point to DAI occurrence, although each one of them resulted from different methodologies like experimental tests, real world accidents' investigations and FE analysis. Therefore, it can be concluded that in real accidents, a motorcyclist wearing a certified helmet still can suffer brain injuries such as cerebral concussion and DAI, based on the thresholds that were exceeded for this type of impact.

4. Conclusions

A certified motorcycle helmet was assessed by predicting head injuries that possibly will result during a motorcycle accident. A FE motorcycle helmet model was developed in order to accurately simulate an oblique impact. This type of impacts can be much more similar to a real impact in a motorcycle accident than the energy absorption tests required by motorcycle helmet standards, once rotational motion is considered, unlike the standards. In order to rely on the results, the helmet materials and the helmet-headform system was validated against experimental data from four energy absorption tests demanded by ECE R22.05 standard. Good correlation was found and the numerical model was considered adequate enough for a preliminary study on biomechanical issues resulting from oblique impacts.

The setup in the oblique impact simulations was based on records of real observed crashes available in the literature. Thus, the results assessed are reliable regarding what really happens in motorcycle accidents concerning helmet impact. The thresholds that only consider translation, such as PLA and HIC were considerably lower for oblique impact than the ones found in linear impacts. This is justified by the introduction of an angular component. This angular component is greater as the angle of impact increases. This led to other conclusion, that motorcycle helmets are designed to perform for higher levels of translational accelerations than the average ones verified in real accidents, which could lead for example to stiffer helmets that can induce head injuries.

The results from the oblique impact, including rotational and translational components were assessed from a biomechanical point of view, comparing it against head injury thresholds proposed in the literature. The more conclusive correlations were found with thresholds that predict cerebral concussion and severe brain injuries induced by critical strains in the brain tissue, such as DAI. Therefore, it could be concluded that an oblique impact with a helmet approved by established motorcycle standards (such as ECE R22.05 and US DOT FMVSS 218), leads to brain injuries such as cerebral concussion and DAI or at least a high risk of occurrence. Summing up, considerable improvements are still necessary for standards to raise motorcycle helmets safety.

The rotational component in head kinematics must somehow be assessed, and always combined with the translational motion.

To carry this goal on, two important things are needed: i) well accepted head injury thresholds or criteria and ii) a standard test rig that can be the reference to replicate impacts with helmets that typically occurs in the reality, such as the oblique impact performed in this study. The results here obtained reinforce this idea: rotational acceleration must be assessed. There is still a long way to go regarding motorcycle helmet safety, where standards demands greatly affects the pace of design evolution.

Acknowledgments

The authors are grateful to the Portuguese Foundation for Science and Technology (FCT) who financially supported this work, through grant PTDC/EME-TME/109856/2009, COMPETE Program and the scholarship SFRH/BD/91292/2012, and also to CMS helmets, Portugal.

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