

Development of numerical models for the investigation of motorcyclists accidents

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Summary:

The Finite Element (FE) method was employed to develop and enhance numerical models that can be used for simulating accidents involving motorcyclists. They are the FE models of a commercial helmet, the human head and the Hybrid III dummy. The composite shell and foam liner of the helmet, which are the most important components in terms of energy absorption, were generated using the Ls-Dyna preprocessing software. The FE model of the human head, which was developed in the Radioss environment, was converted to the Ls-Dyna format. In order to validate the head model with respect to the skull force-deflection response and intracranial pressures, two cadaveric tests reported in literature were simulated. The model of the head was coupled, through the neck, with the body of the Hybrid III dummy. This new dummy was capable of predicting the skull fracture as well as intracranial injuries such as the diffuse axonal injury. As an application of the models, the Hybrid III dummy and the new dummy were equipped with the helmet and dropped onto a flat anvil at the 7.5 m/s impact speed. The protective capability of the helmet was assessed with respect to kinematic injury predictors, such as the maximum linear acceleration of the head, and tissue level injury predictors, such as Von Mises stress in the brain.

Keywords:

Helmet, Human Head, Hybrid III Dummy, Drop Testing, Head Injury Predictors

1 Introduction

Motorcyclists are among the most vulnerable road users and their safety has become a crucial issue. The most severe type of injury in motorcycle accidents is the head injury, which may be mitigated by wearing a safety helmet. The protective capability of a helmet is assessed by dropping a rigid headform, equipped with the helmet, onto a steel anvil and measuring linear acceleration at the centre of gravity of the headform; the helmet is not approved if the peak of this acceleration and/or a function of the acceleration exceed(s) the relevant thresholds. Although this method is adopted by helmet standards, there have been some criticisms of its features; amongst them are the use of linear acceleration as a head injury predictor and using a detached headform (therefore, ignoring the rest of the body).

Linear and rotational acceleration of the head were first proposed as head injury mechanisms more than 60 years ago. Since then, extensive experimental research has been conducted to discover which of them is the dominant injury mechanism. Horgan [1] has reviewed the literature on linear and rotational injury theories. Over the past 10 years, the development of finite element models of the human head has allowed the modelling of detailed head injury mechanisms which can be assigned to tissue level injuries. For example, first principal strain and Von Mises stress in the brain are found to be the best predictors of concussion and diffuse axonal injury ([2] and [3]).

In real world motorcycle accidents the whole body is present and may affect the kinematics of the head under impact loading; however, there are few experimental studies on this effect. Aldman et al. [4-6] dropped a helmeted Ogle-Opat dummy onto a surface made of asphalt concrete at 4.4 m/s and 5.2 m/s impact speeds and measured linear and rotational accelerations of the head. Another study was carried out during the COST programme ([7]); a Hybrid III dummy was equipped with helmets and

dropped onto flat anvils at 4.4, 5.2 and 6 m/s impact speeds, and linear and rotational accelerations of the head were recorded. The conclusions of these investigations were restricted to the maximum impact speeds of 5.2 m/s and 6 m/s, while the helmets were certified with the European standard ECE22.05 [8], which requires an impact speed of 7.5m/s.

The present paper describes the FE models which have been developed in the Ls-Dyna environment ([9]) for the investigation of motorcyclists accidents. These FE models are specifically used to evaluate kinematic and tissue level head injury predictors in full-scale drop tests, i.e. tests including the whole body. To the best of our knowledge, this is the first time that FEM has been used for such a study.

2 FE Model of the Helmet

The geometry files of a full-face commercial helmet and ISO headforms were provided by Dainese s.p.a. The most important parts of the helmet in terms of energy absorption were considered for the FE modelling: the liner foam and the shell. The pertinent geometries were prepared for mesh generation using Hypermesh software. The liner and shell respectively, were discretized using single integration 4-node tetrahedral solid and single integration 4-node quadrilateral shell elements. The average size of the elements was chosen according to the guidelines suggested by Cernicchi et al. [10], who performed a convergence study on the element size of the liner foam and composite shell.

The liner foam was made of EPS (Expanded Poly-Styrene), and it was modelled using the *Crushable Foam* material model within Ls-Dyna. The shell was made of composite laminates, which had different lamination at the front-top-rear, sides and chin guard (Figure 1). The *Laminated Composite Fabric* material model was used to model the shell. For each lamina one through thickness integration point was defined and the relevant fibre orientation and material properties were assigned to this point. Material properties were obtained from conventional mechanical tests performed by Alessandro Cernicchi during his PhD at Imperial College London. He modelled a commercial helmet using FEA and validated it by simulating standard drop tests and comparing the results with the experimental data ([10]).



Figure 1: FE model of a commercial full-face helmet

3 FE Model of the Human Head

The Strasbourg University Finite Element Head Model (SUFEHM) was employed in order to predict head injuries more bio-faithfully. This model, which includes almost all mechanically relevant parts of the head, e.g. the scalp, skull, brain, cerebrospinal fluid (CSF), tentorium and falx (Figure 2), was developed for the Radioss code by Willinger et al. [11], and validated against Nahum's, Trosseille's and Yoganandan's cadaveric experiments ([12], [13] and [14]). Through the replication of 61 real world accidents using this model, Willinger and Baumgartner [3] found that Von Mises stress in the brain, strain energy of the CSF and strain energy of the skull are the best predictors of diffuse axonal injury (DAI), subdural haematoma and skull fracture, respectively.

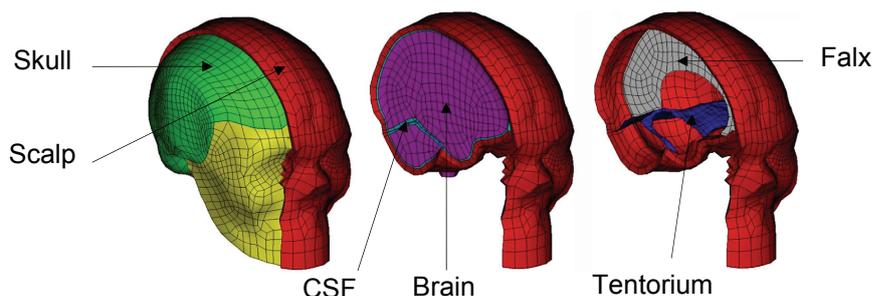


Figure 2: Strasbourg University human head FE model [15]

To make the head model consistent with the existing FE models of the helmet and dummy, it was converted from the Radioss to Ls-Dyna format. The parts meshes along with their material properties were remained intact in the conversion process with the exception of the skull. The facial bone, falx and tentorium were modelled using elastic single integration 4-node quadrilateral shell elements, and for the CSF and scalp elastic single integration linear pentahedron and hexahedral elements were applied. For the brain and brain stem single integration pentahedron and hexahedral elements were used, and their material properties were defined as linear visco-elastic using the *Brain Linear Viscoelastic* material model of Ls-Dyna. Table 1 presents the material properties of the different components of the head model.

Table 1 Material properties of the components of the SUFEHM

Parts	Density (kg/m^3)	Young's modulus (MPa)	Poisson's ratio
Facial bone	2500	5000	0.23
Falx	1140	31.5	0.45
Tentorium	1140	31.5	0.45
Cerebrospinal fluid	1040	0.012	0.49
Scalp	1000	16.7	0.42
Brain and Brain-Stem	1040	$K = 1125MPa$ $G_0 = 49kPa, G_\infty = 16.2kPa$ $\beta = 0.145ms^{-1}$	-

K : bulk modulus, G_0 : short term shear modulus, G_∞ : long term shear modulus,
 β : decay constant

The skull material definition is very important because the deformation of the skull under direct impact loading, for instance in motorcycle accidents, may affect the stress state inside the brain. After describing the material model selected for the skull, two cadaveric experiments will be introduced. The model in the Ls-Dyna format was validated by replicating these experiments to guarantee that the conversion did not reduce the capability of the model to reproduce experimental data.

3.1 Skull Material and Section Definitions

The skull bone is composed of three layers: inner and outer tables made of compact bone (cortical bone) and the middle spongy layer (diploe). Researchers have studied the mechanical behaviour of each layer through performing tension, compression or bending tests on specimens provided from cadavers. For instance, Wood [16] showed that, unlike the other bones of the body, the skull compact bone is isotropic in the direction tangent to the skull surface. Furthermore, the Young's modulus and ultimate tensile stress and strain of the cortical bone are sensitive to the rate of loading.

In the thickness direction, the skull bone can be modelled as a sandwich panel comprised of three layers. On account of the relatively small thickness of the skull, the skull was discretized using linear

quadrilateral shell elements and three through thickness integration points were defined to represent the inner table, diploe and outer table. These layers are transversely isotropic and brittle; hence, the *Laminated Composite Fabric* material model of Ls-Dyna was selected for modelling their behaviour.

In the *Laminated Composite Fabric* material model, the parameter that controls failure of a layer is maximum effective strain. When this maximum value is reached, the layer of the element is completely removed. However, this parameter was not used due to the uncertainty about the definition of the effective strain. Instead, another capability of the material model for simulating failure was employed, which is reducing the strength of a layer of an element to a scaled value of the given strength when the failure criterion is satisfied. There are two possibilities for the definition of the failure surface: quadratic, which is a combination of all stresses, and faceted, which treats each stress individually. The skull material is transversely isotropic and its failure is independent of the material direction tangential to its surface. Therefore, the quadratic failure surface was employed.

Table 2 shows the required inputs for this material model, their values for the head model and the pertinent formula, if applicable. Young's modulus and ultimate tensile strength of the cortical bone and diploe were measured by Wood [16] and Melvin et al. [17], respectively.

Table 2 Material properties of the skull layers used in Ls-Dyna input file

	Formula	Cortical bone	Diploe
Young's Modulus (MPa), E	-	12000	1000
Poisson's Ratio, ν	-	0.21	0.05
Density (kg/m^3), ρ	-	1900	1500
Ultimate Tensile Stress (MPa), S_{ut}	-	100	32.4
Ultimate Compressive Stress (MPa), S_{uc}	-	100	32.4
Ultimate Shear Stress (MPa), τ_u	$\min(S_{ut}, S_{uc})$	100	32.4
Thickness (mm)	-	2	3
Shear Modulus (MPa), G	$E/2(1 + \nu)$	4959	476
Ultimate Tensile Strain, ϵ_{ut}	S_{ut}/E	0.0083	0.0324
Ultimate Compressive Strain, ϵ_{uc}	S_{uc}/E	0.0083	0.0324
Ultimate Shear Strain, γ_u	τ_u/G	0.0202	0.068

3.2 Replication of Yoganandan's experiment (skull deformation and fracture)

Yoganandan et al. [14] conducted static and dynamic force-deflection tests on intact heads. The heads were embedded at the inferior site and loaded statically and dynamically at different locations. In dynamic tests, a metal sphere with a radius of 48 mm was dropped on the heads at 7.1- 8 m/s impact velocities, and the applied external force and the actuator displacement were recorded. Failure forces and displacements ranged from 8.8 kN to 14.1 kN and 3.4 mm to 9.8 mm, respectively. The accuracy of the element and material model used for the skull were studied by replicating this experiment. Figure 3 (a) shows the set-up of the simulation. The output parameters were the vertical reaction force on the impactor and its displacement. Data sampling in the FE simulation was about 33 kHz, which was four times higher than that in the tests (8000 Hz).

Force-displacement curves from FE simulation along with the experimental results are compared in Figure 3 (b). The experimental curve in this figure is the overall mean of the dynamic tests results disregarding the loading site. The FE failure forces and displacements are well in the range of the experimental data and close to the overall mean value, which shows that the FE model is valid. The influence of changing the elastic modulus and ultimate strength of the cortical bone layers on the force-deflection behaviour of the skull are shown in Figure 3 (b). The comparison between the light solid curve and the light dashed curve indicates that an increase in the elastic modulus of the inner and outer tables by 42% made the head stiffer, but this influence was not considerable. In addition, the ultimate strength of the cortical bone did not affect the stiffness of the skull (light and dark solid

curves). However, this parameter changed the failure force; an 11% increase in the ultimate strength of the cortical bone increased the fracture force of the skull by 10%.

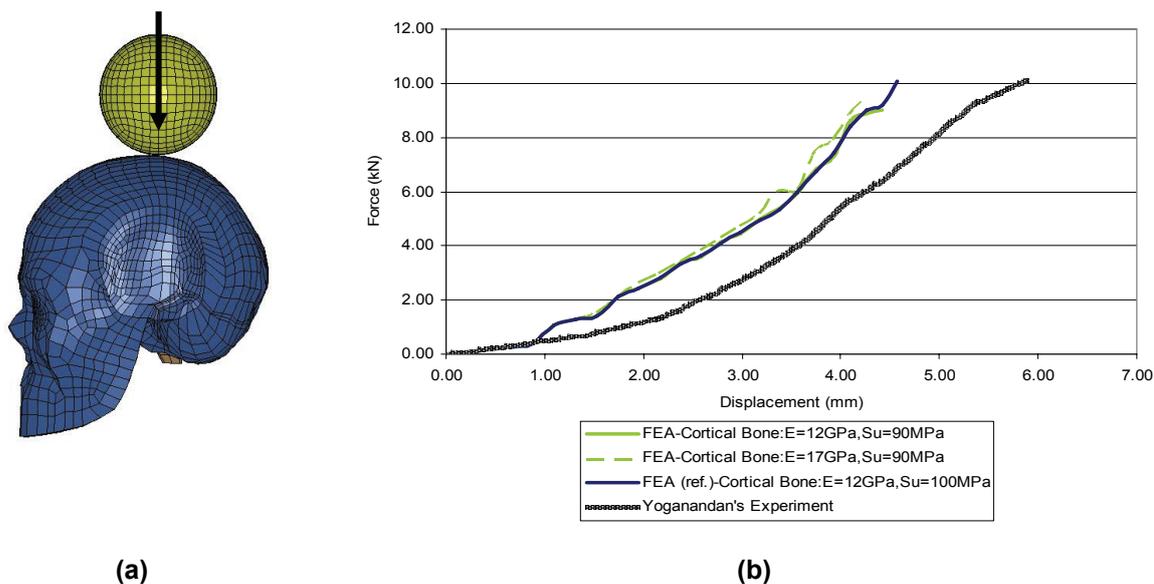


Figure 3: (a) Virtual test set-up for replicating the Yoganandan's experiment and (b) force-displacement curves from FE analysis and test

3.3 Replication of Nahum's experiment (Intracranial Pressures)

Nahum's experiment was comprised of a series of impacts on the frontal side of the head of stationary unembalmed (anatomical embalming is a process of using special fluids for long term preservation of cadavers) human cadavers ([12]). The head of the cadavers were inclined 45° forward with respect to the Frankfurt plane (which is a plane passing through the inferior margin of the left orbit and the upper margin of each ear canal) and impacted by a cylinder covered with a padding to reduce the force and spread it over time. Intracranial pressures were measured during impacts at five locations (Figure 4):

- 1) behind the frontal bone and adjacent to the impact area,
 - 2) immediately posterior and superior to the coronal and squamosal suture in the parietal area,
 - 3) and 4) inferior to the lamdboidal suture in the occipital bone (one in each side) and
 - 5) at the posterior fossa in the occipital area.
- These locations were called frontal, parietal, occipital 1 and 2 and posterior fossa, respectively.

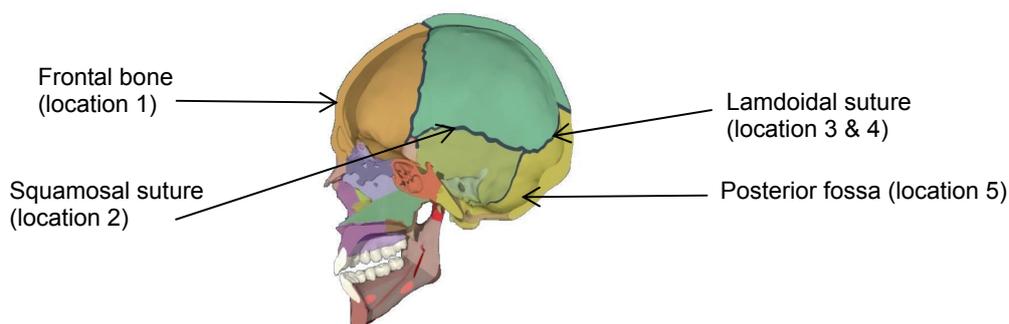


Figure 4: Approximate locations of pressure measurement ([18])

This experiment was replicated using the Ls-Dyna format of the head model in order to validate the model for short duration direct blows with respect to intracranial pressures. Figure 5 (a) illustrates the FE set-up of the test. Given a short duration for the impact (about 6 ms), no boundary conditions were applied to the head. The initial velocity of the impactor, which controls the amplitude of the force, and the elastic modulus of the padding, which controls the force pulse duration, were adjusted to obtain the same force-time curve as that reported by Nahum et al. [12] (Figure 5, b).

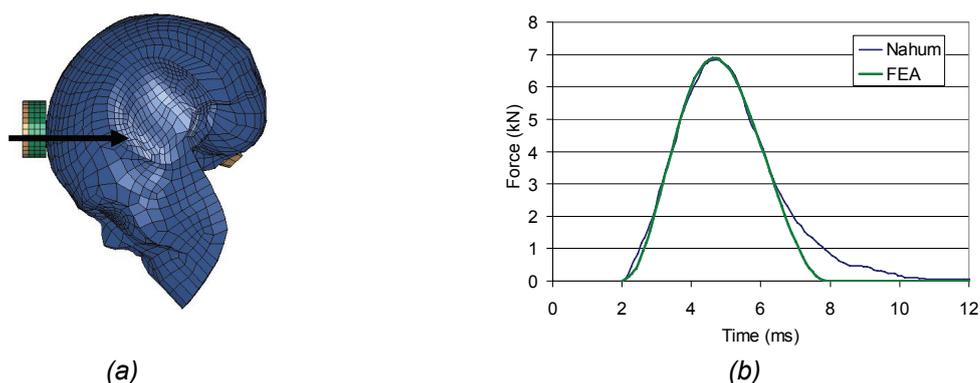


Figure 5: Nahum's experiment: (a) FE analysis set-up, (b) impact force versus time

The intracranial pressures measured in the tests are compared in Figure 6 to the pressures obtained from the FEA. For each location, the pressure at the integration point of an element was chosen for comparison with the experimental pressure. It can be concluded that the finite element model of the head predicts intracranial pressures with an acceptable accuracy, even though there are some discrepancies especially after the first pressure pulse. In fact, the important part of the curve is this pulse, which is accurately predicted by the FE model.

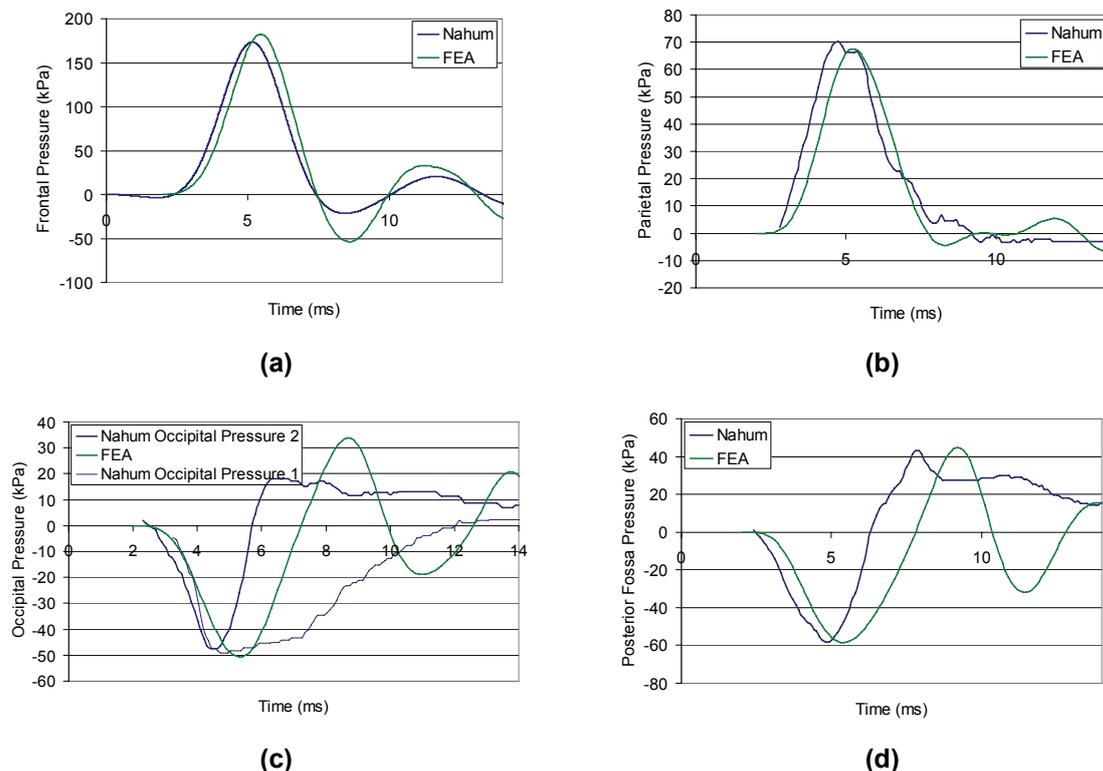


Figure 6: Experimental and FE results for the pressure response of the brain at test locations (a) 1, (b) 2, (c) 3 and 4 and (d) 5

4 Coupling the Head Model with the Hybrid III Dummy

In motorcycle accidents, the head is exposed to a direct load that may exacerbate the stress state inside the brain by deflecting the skull. However, running an FE model of the head with the six components of acceleration which are obtained from tests or simulations does not take into account the deflection of the skull. In addition, it has been shown that the presence of the body in drop tests increases the force on the skull compared to the same test using a detached headform ([7]). These effects were considered by replacing the head of Hybrid III dummy with the Strasbourg University head model.

The FE model of the 50th percentile male Hybrid III dummy, released on 30 October 2008, was downloaded from the LSTC website, where there was the calibration report for this dummy; the model was calibrated by simulating neck extension and flexion and thorax impact tests as per FMVSS 572. To attach the human head model to the dummy, the head of the dummy was removed; some elements of the SUFEHM's skull around the foramen magnum were defined as rigid (Figure 7) and constrained as a slave part to the part that contained inertia properties of the dummy head. This new dummy, which is called H3-HeadFEM, is capable of predicting skull fracture as well as intracranial injuries such as diffuse axonal injury and provides more realistic head boundary conditions.

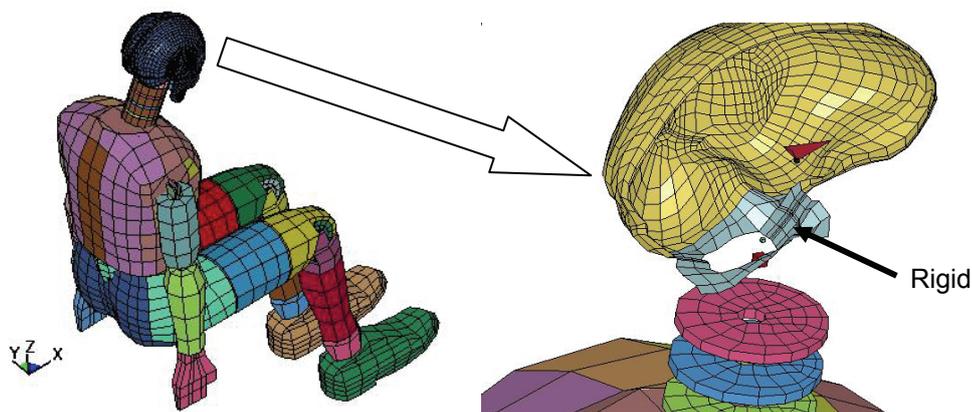


Figure 7: The Strasbourg University head model coupled with the body of the Hybrid dummy

In order to verify the accuracy of the head-dummy coupling, HIII and H3-HeadFEM dummies were equipped with the FE model of the helmet and dropped onto a flat anvil at the 6m/s impact speed. The body axis was parallel to the surface of the anvil and the blow occurred in the mid-sagittal plane at the front site (Figure 8). Without losing generality, the scalp and skull of the SUFEHM were defined as rigid. The resultant linear acceleration at the centre of gravity of the head of the HIII dummy and the SUFEHM along with their rotational acceleration in the sagittal plane are compared in Figure 9. Although there are some discrepancies, the agreement between the results implies that the coupling is correct. The discrepancies are the result of the different inertial properties of the heads (Table 3) and the curvature of their outer surfaces. In addition, the non-rigid components of the Strasbourg University head, e.g. the brain, slightly affect the accelerations.

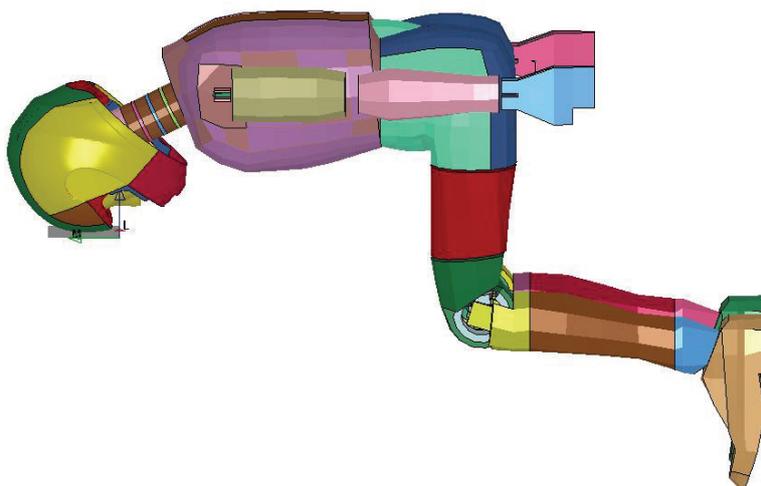


Figure 8: Full-scale frontal impact

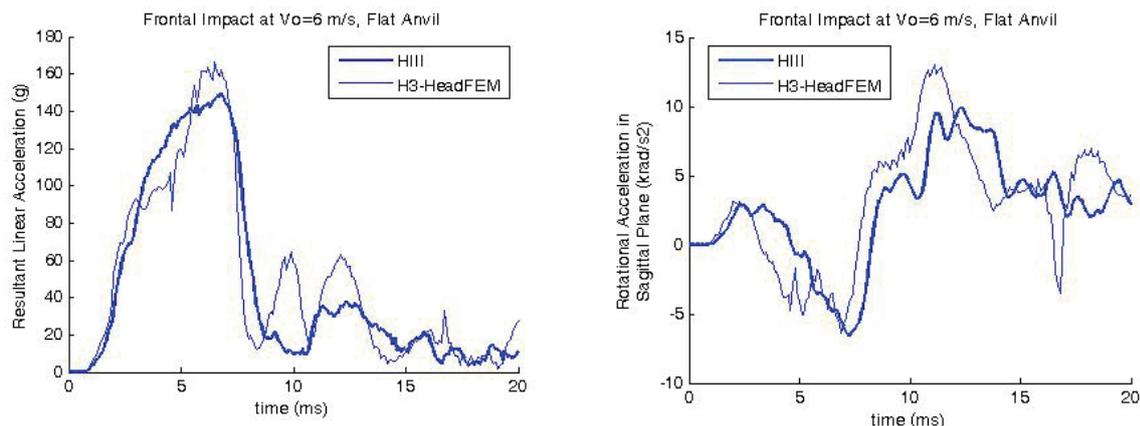


Figure 9: Comparison between the drop test results of the helmeted HIII and H3-HeadFEM dummies

Table 3 Inertial properties of the Strasbourg University head and the HIII head

	SUFEHM	HIII head	diff %
M	4.7	4.8	1.0
Ixx	2.004E+04	2.185E+04	9.0
Iyy	2.315E+04	2.628E+04	13.5
Izz	1.814E+04	1.971E+04	8.6
Ixz	0.253E+04	0.330E+04	30.5

M: mass (kg), I: moment of inertia ($kg \cdot mm^4$)

5 Full-scale drop test

The FE models described were used to simulate a full-scale drop test, i.e. the whole body is present rather than a detached headform, in order to investigate the performance of the helmet in more realistic impact conditions. The Hybrid III and H3-HeadFEM dummies were equipped with the helmet and dropped onto a flat anvil at an impact speed of 7.5 m/s. This is the speed adopted by the energy absorption test of the European standard ECE 22.05, and the helmet had been certified as per this standard. The body axis was parallel to the surface of the anvil and the blow occurred in the mid-sagittal plane at the front site (Figure 8).

The protective capability of the helmet was assessed in terms of two kinematic and three tissue level injury predictors as shown in Table 4, along with their injury limits. The limits of *PLA* and *HIC* are those set in the ECE 22.05 standard. The *PLA* threshold is in the range of 260g-280g for serious head injuries (AIS3); however, the limit of *HIC* is very high compared to the threshold of 1500 for AIS3 [7]. The limits of tissue level injury predictors have been found by Marjoux et al. [15] using the Radioss format of the SUFEHM. If different softwares are used to solve an identical FE model, they generate slightly different magnitudes for the same mechanical output; hence, the limits assigned to *VMB*, *IECSF* and *IESK* in Table 4 are not the precise injury thresholds for the Ls-Dyna format of the SUFEHM. Nonetheless, these injury thresholds were used to predict the probability of the injuries because they were the best available limits. The maximum Von Mises stress in the brain at each sampling time is the mean of the Von Mises stresses of 10 elements that have the largest value among all. This method was applied to avoid obtaining spurious results because of the poor aspect ratios of the elements and coarse meshes.

According to the data given in Table 4, the *PLA* exceeds the injury threshold, but *HIC* is much less than its limit. The tissue level injury predictors indicate that in these impact conditions, diffuse axonal injury, subdural haematoma and skull fracture are probable. Further investigation of the helmet revealed that the helmet liner was excessively compressed under impact loading such that it entered the densification region of its stress-strain characteristic curve at the crush zone. In these conditions, the foam cannot absorb more energy and a small deflection results in a very large normal force. All these phenomena, i.e. higher normal force and more compression of the foam, are the consequences of the body inertia.

Table 4 Results of full-scale drop test

Injury Predictors	Kinematic		Tissue Level		
	PLA (g)	HIC	VMB (kPa)	IECSF (J)	IESK (J)
Indicator of	Head injury	Head injury	Severe neurological injury (DAI)	Subdural haematoma (SDH)	Skull fracture
Helmeted HIII	293	1872	-	-	-
Helmeted H3-HeadFEM	-	-	51	6.5	3.5
Injury Threshold	275	2400	39	4.2	0.833

PLA: Peak Linear Acceleration

VMB: Maximum Von Mises Stress in the Brain

IECSF: Maximum Internal Energy in CSF

IESK: Maximum Internal Energy in the Skull

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

$a(t)$ is the resultant linear acceleration, in g unit and

t_1 and t_2 are, respectively, any starting and ending time in impact pulse duration

6 DISCUSSION

The head injury criterion (*HIC*) is based on the Wayne State Tolerance Curve (WSTC), which was obtained from animal concussion tests and the linear skull fracture of cadavers [19]. This curve suggests that the tolerance of the human head to linear acceleration decreases when the pulse duration is longer. The limits of 1000 and 3000 for *HIC* were defined as 16% and 99% probability of life threatening injuries, respectively [1]. Using *HIC* in helmet standards has been always a matter of debate [20]. Furthermore, researchers believe that its 2400 limit is more than the tolerance limit of the human being. Other kinematic injury predictors also have their own shortcomings. For instance, the maximum resultant linear acceleration does not account for the direction of loading, and it excludes the effect of rotational acceleration, which is believed to be an important injury mechanism.

Some of the drawbacks of the kinematic injury predictors are the consequences of the fact that these predictors are founded on the kinematics of the head disregarding its components. According to the levels of biological organisations, the next lower level is the tissue level. Finite element models of the head try to establish injury predictors at this scale; for instance, SUFEHM suggests that Von Mises stress inside the brain is the mechanism of neurological injuries. Tissue level injury predictors can predict head injuries more precisely compared to the kinematic predictors and provide more information about the type of injury; however, they cannot be measured directly during tests in contrast to kinematic predictors.

Although the helmet satisfied the requirements of the energy absorption test of the ECE 22.05 standard, it did not protect the head when the whole body was present in the drop test. According to the ECE 22.05 standard, the protective capability of a helmet is assessed by dropping a rigid headform, equipped with the helmet, onto a steel anvil and measuring linear acceleration at the centre of gravity of the headform. The helmet is certified when the peak of this acceleration (*PLA*) as well as *HIC* do not exceed the relevant thresholds given in Table 4. When the effect of the body was considered by using the dummy instead of the headform, the kinematic injury predictors indicated that head injury was probable. Moreover, the tissue level injury predictors specified the type of injuries which were diffuse axonal injury, subdural haematoma and skull fracture.

The Hybrid III dummy was developed for studying car rear impacts. Researchers believe that this dummy is not suitable for direct impact investigations, such as motorcycle accidents, because its neck is too stiff. A field of future research would be to employ a more realistic FE model for the human neck and to investigate its effect on injury predictors.

7 Conclusions

The finite element model of a commercial helmet was developed within the Ls-Dyna environment. The Strasbourg University FE model of the human head was converted from the Radioss to the Ls-Dyna format and validated against two cadaveric experiments, which are usually used to investigate the accuracy of head FE models in predicting skull force-deflection and intracranial pressures under

impact loading. The head model was coupled with the body of the Hybrid III dummy and equipped with the FE model of the helmet. This combination and the helmeted Hybrid III dummy were virtually drop tested onto a flat anvil. It was shown that the consequence of the presence of the body was more compression of the helmet liner foam such that at an impact speed of 7.5 m/s, which is the speed adopted by the ECE 22.05 standard, the foam bottomed out, and head injuries including the diffuse axonal injury, subdural haematoma and skull fracture were probable.

Acknowledgements:

The work presented in this paper was completed within the research training network MYMOSA funded by the Marie Curie fellowship of the 6th framework programme of the EU under contract no. MRTN-CT-2006-035965. The authors would like to thank Dainese s.p.a. for providing the geometry files of the helmet parts and Mr A. Cernicchi for providing the material properties of them in the Ls-Dyna format.

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