Numerical and Experimental Modelling of Human Head under Impact - New Tolerance limits and experimental injury criteria

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Abstract: Head injuries are frequent and severe injuries in almost all types of traffic accidents with height societal and economic costs. Therefore, head injury reduction is a high priority for traffic safety improvement. After a review of the state of the art of human head modelling, the present paper presents original physical and numerical human head models followed by there modal and temporal validation against human head vibration analysis in vivo and cadaver impact tests. The human head finite element model developed by ULP at Strasbourg University presents two particularities : one at the brain-skull interface level were fluid-structure interaction is taken into account, the other at the skull modelling level by integrating the bone fracture simulation. Validation shows that the model correlated well with a number of experimental cadaver tests and predicted intra-cranial pressure accurately. However, for long duration impacts the model reaches its limits. The skull stiffness and fracture force were accurately predicted when compared with experimental values from the literature. In a second step a new dummy head prototype named Bimass 150 is presented. It has been constructed using a Hybrid III headform and comprises two masses : a skull and a mass to represent the brain attached to the skull with a damped spring system. The novel feature of this device is that it can simulate the brain - skull relative displacement at a frequency close to 150 Hz as recorded under vibration analysis in vivo. This numerical and physical improved human head surrogates have then be used for experimental and numerical real world accident reconstruction. Helmet damage from thirteen motorcycle accidents was replicated in drop tests in order to define the head's loading conditions. A total of twenty two well documented American football head trauma have been reconstructed as well as twenty eight pedestrian head impacts. By correlating head injury type and location with intra-cerebral mechanical field parameters, it was possible to derive new injury risk curves relative to specific injury mechanisms. As a summary, for the numerical ULP human head FE model following limits were drawn in this study.

INTRODUCTION

Head injuries are frequent and severe injuries in almost all types of traffic accidents with high societal and economic costs. Therefore, head injury reduction is a high priority for traffic safety improvement. Car safety standards rely upon criteria for human tolerance, which are based on biomedical research performed more than 30 years ago. Measures designed to improve head protection are typically evaluated against a measurement of the Head Injury Criterion (HIC). The predictive capacity of this criterion has been widely criticised because of its limited ability to predict a probability of brain injury. It has been suggested that specific deformation of skull material and brain tissue and a measure of the relative motion of the brain and skull would be much better means of assessing head protection.

The objective of the research performed at ULP since 1990 is to construct and validate a numerical model of the human head suited to reconstruct real world accidents and to derive improved tolerance limits to specific injury mechanisms.

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Finite Element Methods (FEM) were considered to be the best tool with which to investigate the response of the human head under impact condition. To date, more than ten different 3D human head models have been described but only Ruan's and Zhou's model [1,2] were validated and then used for accident reconstruction to investigate brain injury tolerance limits. Most of the essential head components were incorporated in this model, which was meshed with 37,040 elements. Recent modifications of this model [3] include the addition of a 3-layered skull and an improvement of the brain-skull interface to allow the brain to "move more freely" relative to the skull. The model was validated against the intra-cranial pressure data from impact tests onto the front of the head of cadavers and against brain-skull relative motion using data obtained from the high-speed X-ray experiments. An improved version of the WSU model has been published by Zang in 2001 [4]. A refined brain meshing was proposed (314 500 elements) and the validation procedure showed realistic results up to linear and rotational accelerations up to 200 g and 1200 rd/s^2 respectively. Bandak et al [5] also published a finely meshed human head FE model and proposed an in depth experimental and numerical analysis of the subarachnoïdal CSF layer influence on the brain response.

Zhou et al. [2] simulated a fully documented road accident with this model and the shear stresses predicted by the model agreed approximately with the location of axonal injury described by the medical report. More recently Newman et al. [6] presented a detailed methodology for the assessment of mild traumatic brain injury based on the reconstruction of American professional football accidents, using Zhou's human head FE model. The findings suggested that mild traumatic brain injury occurred with a Von Mises stress of 0.07 kPa and a pressure of 0.03 kPa. Main objective of the WSUBIM (WSU Brain Injury Model) was however to evaluate the correlation between Von Mises intra-cerebral shearing stresses with angular head acceleration on the one hand and intra-cerebral pressure with the head linear acceleration on the other hand. Correlation coefficient of respectively 0.86 and 0.82 was fund in this study as reported by Yang and King [7]. In 2001 Bandak et al. [8] presented a first version of a head injury assessment tool based on a very simplified head FE model and called SIMon. An attempt was made in this study to distinguish between brain injury mechanisms such as cumulative strain damage, dilatation damage and relative motion damage and to derive specific injury criteria by simulating existing experimental head impacts.

Willinger et al. developed the "ULP" human head FE model [9] and this is briefly described below. The first accident replicated was using an initial version of this model that involved a head impact caused by a motorcycle accident [10]. The results thereof showed that it was possible to compare intracranial field parameters with neuropathlogical brain injury details and hence leading to the conclusion that intra-cranial pressure did not correlate well with intra-cerebral haemorrhage but that Von Mises stress distribution correlated very well with lesions. The research described in this paper is a more extensive use of the ULP head FE model in real world accident simulation in order to investigate tolerance limits. The final objective of the present paper is to derive new head tolerance limits to specific injury mechanisms by using the ULP-FE head modelling the framework of accident reconstruction. Hereafter the model development and validation are reported before the presentation of the 64 reconstructed accidents. Injury risk curves for mild and severe neurological lesions, subdural haematoma as well as skull fracture are then presented

Although sophisticated FE models are now emerging, experimental analysis of head impacts remains an important and frequently used method both for research and Standards. Several attempts were made in the past to propose more biofidelic headforms. Brinn et al [27] proposed an aluminium shell covered by a vinyl layer and a second layer made of a frangible material in order to replicate skull fracture. Kenner et al. [28] suggested a water filled aluminium sphere to study wave propagation during an impact. Margulies et al. [29] described the use of a gel-model to correlate shearing force with diffuse axonal injuries. Today the current practice is to fit a single mass dummy head with an accelerometer array and to record linear and angular acceleration. Newman [22] used this approach, in an attempt to correlate a computation of the two acceleration components known as GAMBIT (Generalized Acceleration Model for Brain Injury Threshold) with injury.

One of the criticism of this studies was the use of a single mass to represent the head and brain. It is not possible with this approach to record the effect of shearing strain within the brain and the "brain -skull relative motion, which are well known accident injury mechanisms. Theoretical human head modelling is able to represent many more degrees of freedom. The first human head mathematical models were lumped mass models (Hodgson et al [30], Stalnaker et al [31], Willinger et al. [32]), validated mainly against experimental head impedance recordings. This type of model was of fundamental importance for understanding head injury mechanisms. It is well known and accepted that the head comprises several compliant components. Nevertheless, Standards that are used to certify protective systems continue to specify a rigid body for the headform. Thus, it is not possible to reproduce the various injury mechanisms that could occur during a head impact and injury potential can only be assessed against criteria such as HIC, widely criticised by the scientific community. Moreover, the force generated between the headform and the object struck is a function of the headform and this is very different to a human head. As an illustration of this problem it was recently shown that the optimisation of the helmet liner is a function of the head substitute (Willinger et al.[33])

The second step of the research presented in this paper describes a new dummy head prototype comprising two masses: a skull and a mass contained within the skull linked by a damped spring system. This headform is designed to be able to reproduce the motion of the brain within the skull and hence is known as the "Bimass”. It has been validated against the mechanical impedance recorded on the human head in vivo. As for the numerical model, 13 motorcycle accidents, for which all the details including head injuries were available, were then replicated to derive new experimental injury criteria to specific injury criteria.

THE ULP HUMAN HEAD FE MODEL

The geometry of the inner and outer surfaces of the skull was digitised from a human adult male skull. The data given in an anatomical atlas [11] was used to mesh the human head using the Hypermesh code. For this study, the option was chosen to retain a given realistic human adult anatomy rather then trying to find an average geometry, which may not exist. Figure 3 shows the 3D-skull surface obtained by digitising external and internal surfaces of the skull as well as the meshed model. Figure 4a 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, falx, tentorium, subarachnoid space, scalp, cerebrum, cerebellum, and the brain stem as well as the ventricles.

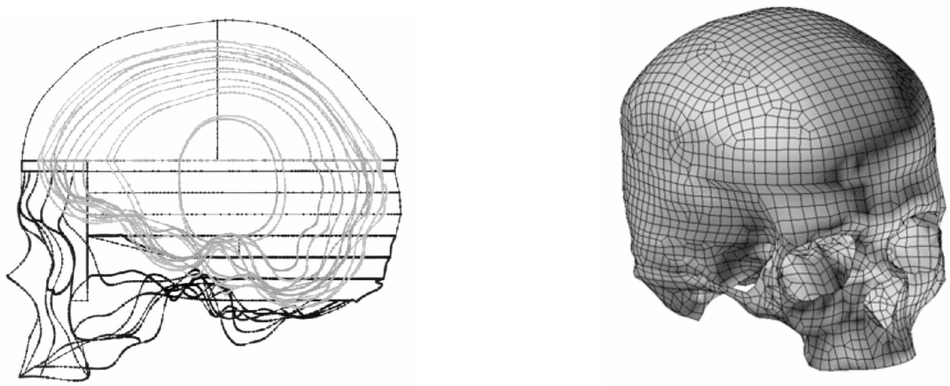


Figure 1 : 3D skull surfaces used for the ULP human head model construction and skull meshing.

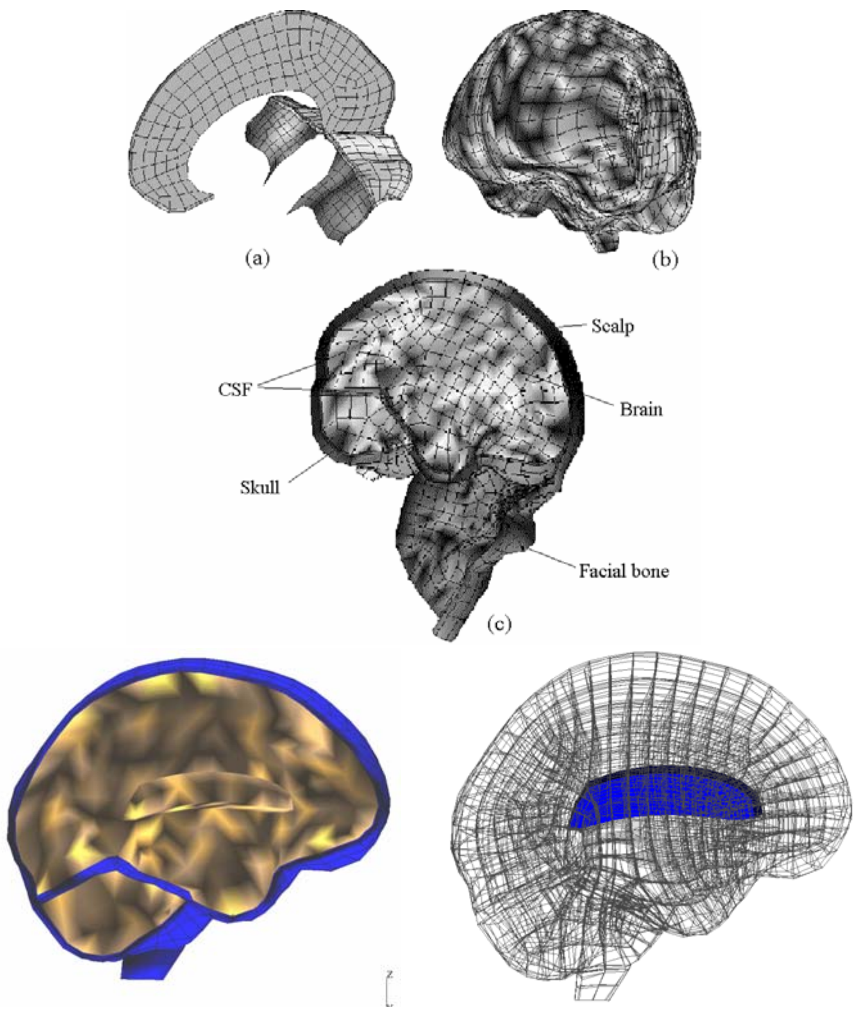


Figure 2. Meshing of the intra-cranial medium(falx and tentorium, brain representations, overview and Mid-sagittal view of brain and CSF including ventricules for the Arbitrary Lagrangie-Euler version of the model.

The finite element mesh is continuous and represents an adult human head. The falx and tentorium were simulated with a layer of shell elements, the skull comprised a three layered composite shell and the remaining features were modelled with brick elements. Of particular importance is the subarachnoïd space between the brain and the skull which was, as a first step, represented by one layer of brick elements to simulate the cerebral-spinal fluid (CSF). Lagrange formulation was therefore selected and the brain-skull liaison was modelled by an elastic material validated against the in-vivo vibration analysis [12]. In order to improve the simulation of the CSF for long

Duration impacts, a three layered brick element interface between brain and skull is proposed and an Arbitrary Lagrange Euler (ALE) formulation available in RADIOSS code was used in order to take into account the fluid-structure interaction as shown in figure 2. For this analysis the CSF brick elements were fixed to the skull elements at the exterior surface and to the brain or the membranes at the inner side. The tentorium separated the cerebrum and the cerebellum, and the falx separated the two hemispheres. Brick elements were used to simulate the CSF that surrounds these membranes in the same way as between brain and skull and ventricles were also integrated. A layer of brick elements also modelled the scalp, which surrounds the skull and facial bone. Overall, the Lagrange version of the head model consists of 13208 elements divided in 10395 brick elements (5376 for the brain, 2870 for the CSF and 1530 for the scalp) and 2813 shells elements (2424 for the skull and 389 for the membranes). Its total mass is 4.772 kg. The ALE version contains 6486 brick elements which describe the CSF layer and the ventricles instead of 2870 elements for the Lagrange version. The total number of elements for this latter head model is therefore 16824 elements.

Material characteristics are very important to the success of a finite element model and Table 1 lists the properties of the materials used. Material properties of the cerebral spinal fluid, scalp, facial bones, tentorium and falx are all isotropic and homogenous. The viscoelastic properties assigned to the brain were scaled from Khalil et al [13]. The behaviour in shear was defined by:

$$G(t) = G_\infty + (G_0 - G_\infty)\mathrm{Exp}(-\beta\, t) \tag{1}$$

with G_0: Short term shear modulus, G_∞: Long term shear modulus and β : Decay constant.

Two separate formulation are used for the CSF modelling, the Lagrangian and the Eulerian. Main objective was to evaluate under which condition CSF flow occurs and how this phenomenon influences the intra-cerebral mechanical response. For the Lagrangian version, the Young's modulus of the subarachnoid space was determined by Willinger et al [12] using modal analysis, based on the fact that a brain-skull decoupling occurs at the first natural frequency of the human head at around 100-150 Hz as reported in table 1. A large deformation formulation was used in order to have a realistic strain estimation in this layer of brick elements. The skull was modelled by a three layered composite shell representing the inner table, the diplöe and the external table of human cranial bone. In order to reproduce the overall compliance of cranial bone, a thickness in combination with an elastic brittle law were selected for each layer. To model the material discontinuity in the case of fracture, it was necessary to use values for the ultimate tensile and compressive stress (UTS and UCS in table 1) obtained from Piekarski [14] and integrated in the Tsaï-Wu criterion [15]. The material properties of the intra-cerebral membranes and the scalp are similar to those used by Zhou et al [2] and also reported in table 1. In the ALE version of the model the CSF is represented by an hydrodynamic fluid defined by a Bulk modulus of 21.9 GPa

A total of eight instrumented cadaver impacts were reconstructed with the objective to validate the ULP model under very different impact conditions. Currently head FE models are validated against Nahum's impact [16] and this was satisfactorily achieved with the first version of the model (with Lagrangian formulation), in a previous study [10]. The impact duration of Nahum's test was about 6 10-3 s . Strasbourg University devised a procedure to establish over what range the model was satisfactorily validated. The data used were taken from five highly dampened cadaver impacts with important angular components published by Troseille et al [17] and two extremely short cadaver impacts (to the front and to the vertex) inducing skull fracture, published by Yoganandan [18].

Table 2 summarizes the main characteristics of the three classes of impact, i.e. medium, long and short duration. The RADIOSS code developed by MECALOG was used for the finite element analysis and the method of one point integration was used for all analysis with an Hourglass energy below 0.1% of the total involved energy. This validation procedure is detailed in Willinger et al. [19].

The simulation of the intra-crânial behaviour was satisfactory where short impacts were concerned such as reported by Nahum et al. [16] (see figure 3). For more dampened impacts as reported by Troseille et al [17], the numerical and experimental skull rotational and linear acceleration were found to agree perfectly with the experimental result. Nevertheless, in high damped long time duration impact configurations as in Troseille's cases, pressures at different locations inside the brain are not satisfactorily reproduced by the model. This may be due to a CSF flow actually occurring in experiment and not taken into account in numerical simulation. In this case, the thorax of an instrumented cadaver has been impacted by a plate at 5 m/s. Moreover the CSF flow was represented by three layers of brick elements in ALE (Arbitrary Lagrangian Euler) formulation. This ALE formulation is specially indicated for fluid-solid coupling in FE modelling. The goal was to study the effect of ALE formulation for the CSF modelling and to compare the model response to the model response without ALE formulation. Both cases are confronted to Troseille's experimental data in terms of intra-cranial pressures in frontal and occipital regions as shown in figure 4. The main result is that at each location the pressure time history is better reproduced in the FE model including ALE formulation for CSF. There still remain some discrepancy with the experimental measurements probably since the intra-cranial material properties are not known accurately enough. In frontal lobes as well as in the occipital region, model response shows less oscillations when ALE formulation is used and the variations towards experimental pressure are notably reduced. To conclude, efforts have still to be made at the intra-cranial material properties formulation level in order to get closer to real life characteristics of the human head. Using ALE formulation for CSF seems to be a first step in that sense in the case of high damped long duration impacts. Finally such kind of impacts remains difficult to be modelled and this study points out the validation domain of existing models. At the skull response level, the numerical force-deflection curves are compared to the average dynamical response of experimental data. The dynamic model responses agree well with the experimental results, both the fracture force and the stiffness level. The model indicates multiple fracture located around the impact point which complies with pathological observations as reported in figure 5.

As a summary, the validation shows that the ULP FEM of the human head correlated well with a great variety of experimental cadaver tests and predicted intra cranial pressure accurately enough. Nevertheless, for long duration impacts the model reaches its limits. Moreover, the skull stiffness and fracture force were very accurately predicted when compared with values from the literature.

Table 1 : Material properties of the human head FE model from experience reported in the literature. Only CSF elastic modulus results from a numerical modal analysis.

Part	**Material property**	**Material parameter**	**Value**	**Element type**	**Shell thickness (mm)**
Face	Elastic	Density	2500 $Kg.m^{-3}$	Shell	10.0
		Young modulus	5.0E+03 MPa		
		Poisson's ratio	0.23		
Cranium (Cortical)	Elastic Plastic Orthotropic	Density	1900 $Kg.m^{-3}$	Shell	2.0
		Young modulus	1.5E+04 MPa		
		Poisson's ratio	0.21		
		Bulk modulus	6.2 E+03 MPa		
		UTS	90.0 MPa		
		UCS	145 MPa		
Cranium (Trabecular)	Elastic Plastic Orthotropic	Density	1500 $Kg.m^{-3}$	Shell	3.0
		Young modulus	4.6E+03 MPa		
		Poisson's ratio	0.05		
		Bulk modulus	2.3E+03 MPa		
		UTS	35.0 MPa		
		UCS	28.0 MPa		
Scalp	Elastic	Density	1.0E+03 $Kg.m^{-3}$	Solid	-----
		Young modulus	1.67E+01 MPa		
		Poisson's ratio	0.42		
Brain	Viscoelastic	Density	1040 $Kg.m^{-3}$	Solid	-----
		Bulk modulus	1.125E+03 MPa		
		Short time shear modulus	4.9E-02 MPa		
		Long time shear modulus	1.62E-02 MPa		
		Decay constant	145 s^{-1}		
CSF	Elastic	Density	1040 $Kg.m^{-3}$	Solid	-----
		Young modulus	0.12E-01 MPa		
		Poisson's ratio	0.49		
Falx	Elastic	Density	1140 $Kg.m^{-3}$	Shell	1.0
		Young modulus	3.15E+01 MPa		
		Poisson's ratio	0.45		
Tentorium	Elastic	Density	1140 $Kg.m^{-3}$	Shell	2.0
		Young modulus	3.15E+01 MPa		
		Poisson's ratio	0.45		

Table 2 : Main characteristics of experimental cadaver tests from the literature as used for validation

Test	Impact area	Impactor (kg)	Impactor velocity (m/s)	Force (N)	LA maxi (g)	RA maxi (rd/s^2)	Duration (ms)
Nahum 1977	front	cylinder (5.6) with padding	6.3	6900	198	-	6.5
Trosseille 1992 MS 428-2	face	steering wheel (23.4)	7	-	102	7602	15.8
Yoganandan 1994	front	rigid sphere (1.213)	7.1	10500	-	-	2

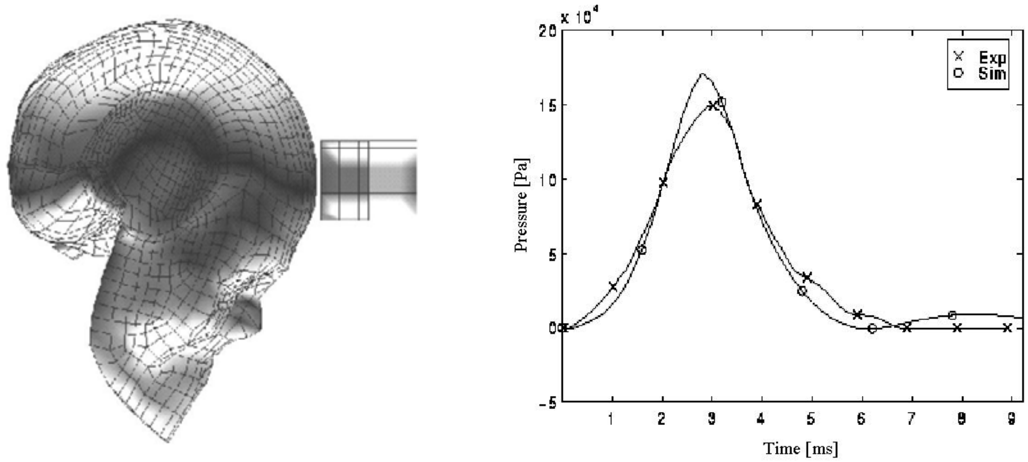


Figure 3 : Right : ULP FEM of the human head in frontal impact configuration. Left : measured by Nahum et al. [16] (x) and calculated (o) intra cranial frontal pressure.

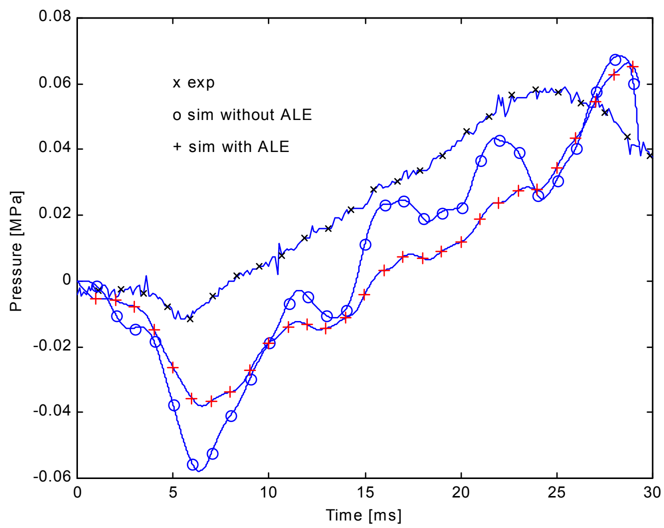


Figure 4 : Experimental (Troseille et al. [17]) and simulated (with and without) ALE formulation for CSF frontal pressure for a long duration impact with high angular component.

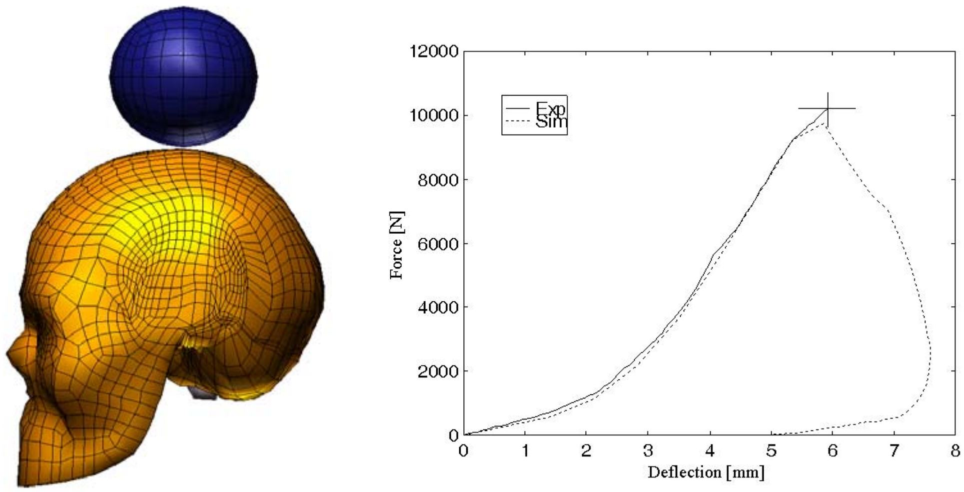


Figure 5 : Right : ULP FEM of the human head in vertex impact configuration. Left : measured by Yoganandan et al. [18] (full line) and calculated (doted line) to the head applied force versus deflection curve. + indicates the bone rupture point.

REAL WORLD ACCIDENT RECONSTRUCTION

The previous presented human head model was then used for extensive real world accident reconstructions. A total of 64 head impacts were simulated, 35 protected and 29 non protected direct head impacts. Given the time duration of the impacts taken into account in the present study, lasting between 5 and 10 milliseconds, the Lagrangian version of the head model was used. The protected impacts came from helmeted victims (motorcyclists and American Football players) whereas the unprotected were pedestrians impacting a car's windscreen. Due to the very different impact conditions between protected and unprotected impacts, two separate methodology were designed in order to define the boundary conditions of the model itself.

Concerning the helmeted victims it was an experimental head impact reconstruction which permitted it to define the skull 3D kinematics for motorcyclists and the American football players. A total of thirteen motorcyclists cases were replicated with drop tests of a helmeted headform at Transport Research Laboratory London (TRL). The aim of this work was to replicate head impacts sustained during the accidents while measuring the dynamics of the head. In this experimental study [20], TRL replicated the helmet damage using a purpose-built helmet drop test facility. The method allowed impact parameters, including impact speed, angles and targets, to be controlled and quantified. By inspection of the helmet it was possible to modify the impact parameters until the desired damage was produced. Instrumentation was used to measure the dynamics of the impact and ultimately enable the accelerations, likely to have been experienced by the casualty, to be estimated. Analysis of the damage to the shell and liner was used to identify the kinematics of the impact. The accuracy of the replication was judged by comparing the replicated damage with the accident damage. The test helmet was an identical make and model to the accident helmet to ensure similar performance during the impact and up to five tests were sometimes necessary to obtain a satisfactory replication of the accident helmet damage. The American Football player cases were studied in collaboration with BIOKINETICS-Canada and detailed in by Shewchenko et al. [21]. When football players' helmeted heads collide with one another during games a two cameras device allows it to determine the relative position and velocity between the two involved heads at the time of impact. Both parameters are applied to two instrumented and helmeted dummy's heads that represent the heads of the football players in order to replicate experimentally the real world heads collision documented by Newmans et al. [22]. The validation of that method is based on the rebound of the dummies after the experimental replication compared to the filmed rebound of the football players (figure 6). Like for the motorcyclist cases this experimental impact test delivers the acceleration fields sustained by the heads of the victim during the impact.

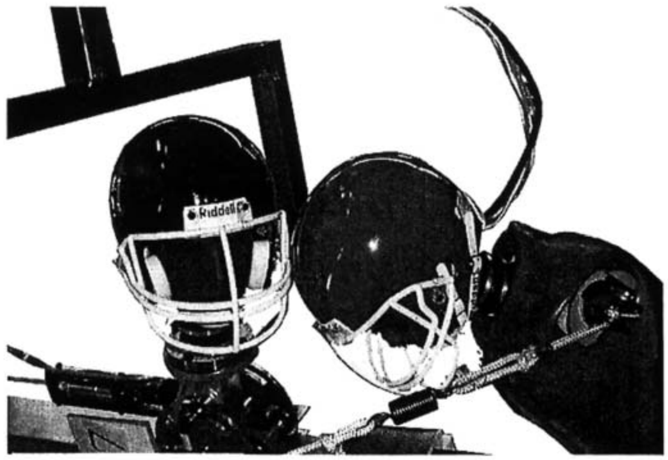


Figure 6 : Experimental protected head impact replication of American football player collision.

For the 35 accident cases involving helmeted heads and reconstructed experimentally, the reconstruction report was transferred to ULP-Strasbourg. In addition ULP was provided with an electronic copy of the results of the 3D linear and angular acceleration of the dummy head.

From the 3D-acceleration time histories provided, the velocity was calculated as a function of time at three points on the skull FE model supposed as rigid and this was used as the input to the FE accident simulation. Due to the duration of the impacts, the intra-cranial material properties used were the ones presented in table 1.

As mentioned previously, the non protected head impacts came from pedestrian which were impacted by a car . These pedestrians accidents reconstructions involve 29 cases stemming from the database of the Accident Research Unit of the Medical University Hanover (ARU-MUH) and precisely described by Baumgartner et al. [23]. When the head of pedestrians who are knocked down by a car strikes the windscreen, a great variety of parameters are collected on the accident scene. These parameters are used as inputs for an analytical model that simulates the kinematics of the pedestrian before the impact of his head on the windscreen. The aim of that analytical model is to establish the relative position and velocity between the head and the windscreen of the knocking down car. For each pedestrian accident case, the results of that analytical simulation are compared to the damages that are observed on the car and the wounds which are sustained by the victim. The ULP head FEM, this time with a three layered deformable skull model of the head, is then positioned towards the windscreen in respect to the calculated angular position just before the impact. The initial relative velocity between the head and the windscreen is set on the nodes of the windshield which has been modelled separately by a tree layered frangible structure as illustrated in figure 7 .

For the 64 head trauma retained for this study, the intra-cranial response was then computed with the RADIOSS FE code in order to calculate the intra-cerebral pressure and Von Mises shearing stresses as well as the global strain energy in the CSF as a function of time. The maximum values of the shearing stresses are then determined as well as the location where these maximum values are reached in order to be correlated with the sustained neurological lesions. The maximum shearing stress levels are distributed at different parts of the brain according to the considered case. For example, Figure 8a illustrates the calculated Von Mises stress field sustained by the motorcyclists G174 through both the field representation 9 ms after the beginning of the impact and the time history at the location where it reaches its maximum value. In case of direct head impact, skull response was computed in terms of deleted elements (figure 8b) but also interaction force and global strain energy in the skull.

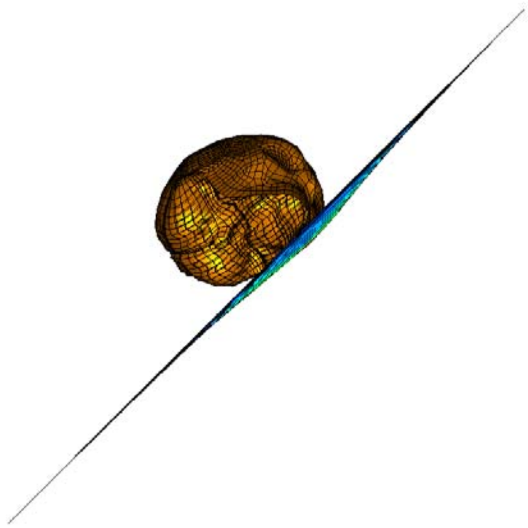


Figure 7: Lateral view of the pedestrian head impact on the windscreen for case H8362, 4 ms after the first contact.

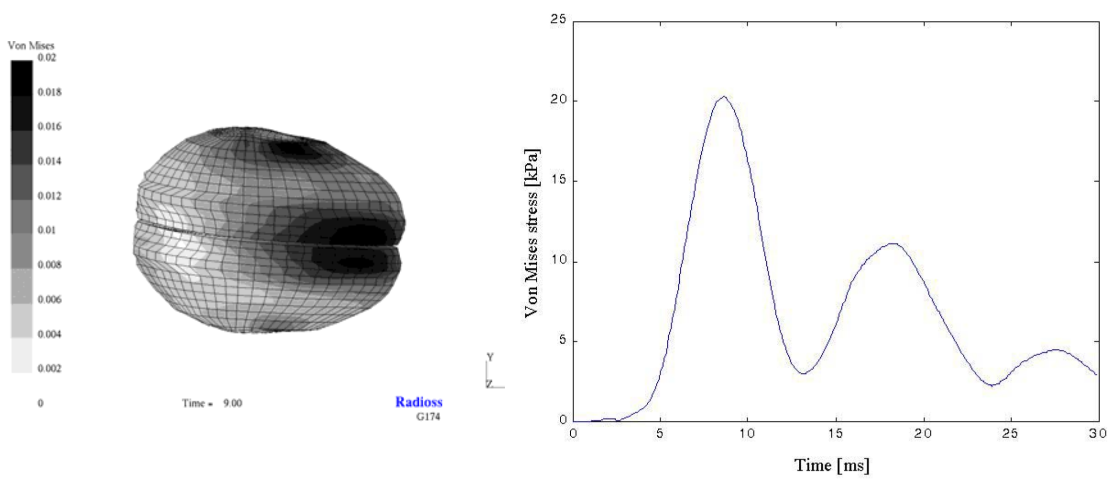


Figure 8 a : Brain Von Mises stress for the motorcyclist G174. Left : BrainVon Mises stress field 9 ms after the beginning of the impact. Right : Time history of the brain Von Mises stress at the location where it reaches its maximum value.

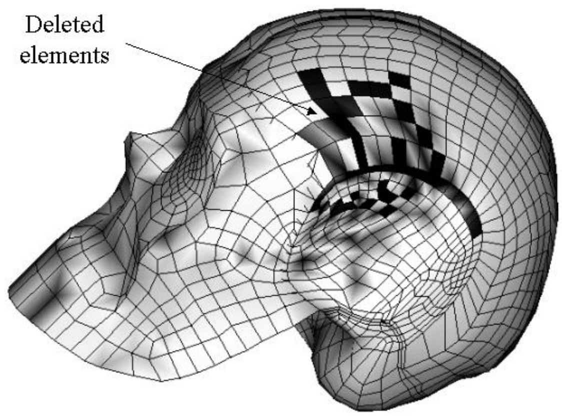


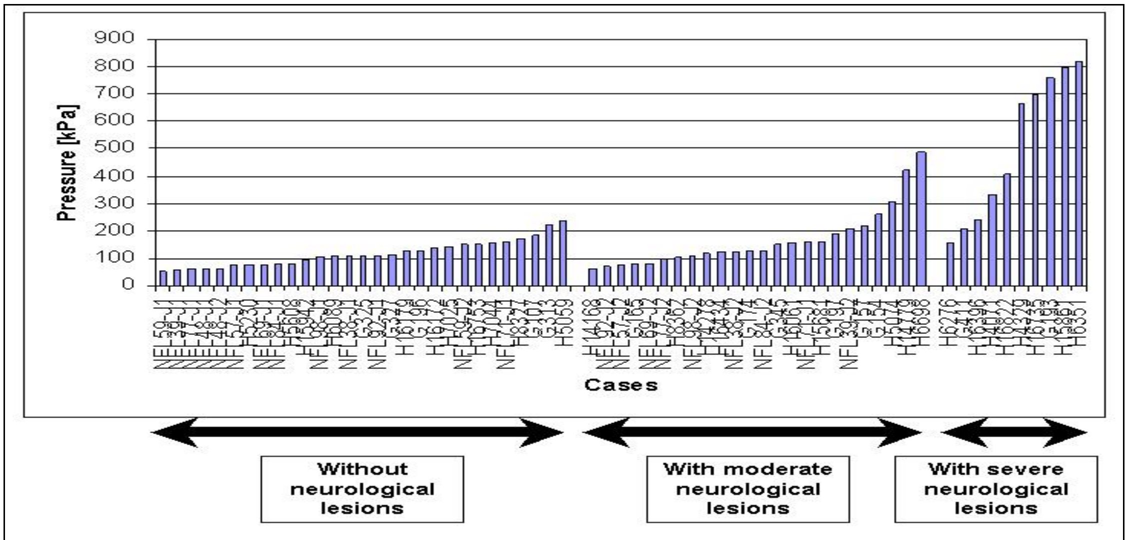
Figure 8b : Deleted elements (colored in black) for H6351 unprotected pedestrian accident case.

TOLERANCE LIMITS AND INJURY CRITERIA

Currently, real world accident analysis is used in an attempt to correlate a known head injury parameter with the AIS (Abbreviate Injury Scale) value sustained. An attempt by Chinn et al [20] to correlate initial head impact velocity, maximum linear and rotational acceleration, HIC value and GAMBIT versus AIS gave correlation coefficients of 0.3 to 0.6, which, in the authors opinion is not satisfactory. Even intra-cerebral mechanical parameters calculated with our head FE model, outputs were shown to give similar correlation with AIS which in our study ranged from 1 to 6. It is considered by the authors of this paper that a much better approach is to take into account the likely head injury mechanism. In fact, the main reason for the poor correlation between a given parameter and AIS is that the same AIS levels can be sustained from very different injury mechanisms. An original approach to derive injury tolerance to a specific injury mechanism would be to check the different head model output parameters and to correlate them with the type of the injuries sustained by the subject.

When the type of lesion, rather than AIS, was used for comparison, then five distinct groups emerged from our accident data, i. e. uninjured (n=29) , mild neurological injury (n=24), severe neurological injuries (n=11), sub-dural haematoma (n=7) and finally skull fractures (n=19). In order to go further in the analysis of the intra-cranial responses relative to the accidents under study, histograms which give for each case the maximum intra-cerebral pressure, the maximum Von Mises stress, the maximum strain energy in the CSF layer and the maximum strain energy in the skull bone, calculated with the ULP FE head model were successively plotted. After examination it was found that the value of some parameters for a specific group of accident victims was found to be valid as a means of estimating a tolerance limit for the injury sustained by that group. For example the histogram given in figure 9a shows that pressure, because of the wide variation within a victim group was not responsible for the neurological injuries. The maximum Von Mises stress, illustrated in figures 9b, is of greater interest and show better correlation. Uninjured, sustained low values whereas mild neurological lesions cases, sustained clearly higher values and cases with severe neurological lesions presented greater shearing stresses than those of the mild neurological injury group. The third histogram in figure 9c is related to maximum strain energy in the CSF layer and shows that for the victims with subdural haematoma, the values of this parameter was substantially greater than for the other groups. The last histogram in figure 9d concerns the skull response in terms of skull bone strain energy. It appears that this parameter is a good candidate for a skull fracture prediction criteria given the height values computed in the cases where a fracture occurred.

The above analysis, conduced injury mechanism by injury mechanism in each histogram leads to the following conclusions. The brain Von Mises stress is a good indicator for brain neurological lesions, shall they be moderate or severe. Moreover, this mechanical parameter allows to distinguish these lesions into two categories : moderate or severe. Global strain energy in the CSF layer and in the skull structure are reasonable indicators respectively for subdural haematoma and skull fracture. For the four thresholds defined thought the histograms (figures 9 a,b,c,d), a statistical regression analysis using the so called Modified Maximum Likelihood Method developed by Nakahira et al. [24] leads to the establishment of tolerance limits against specific injury mechanisms in terms of injury risk curves.



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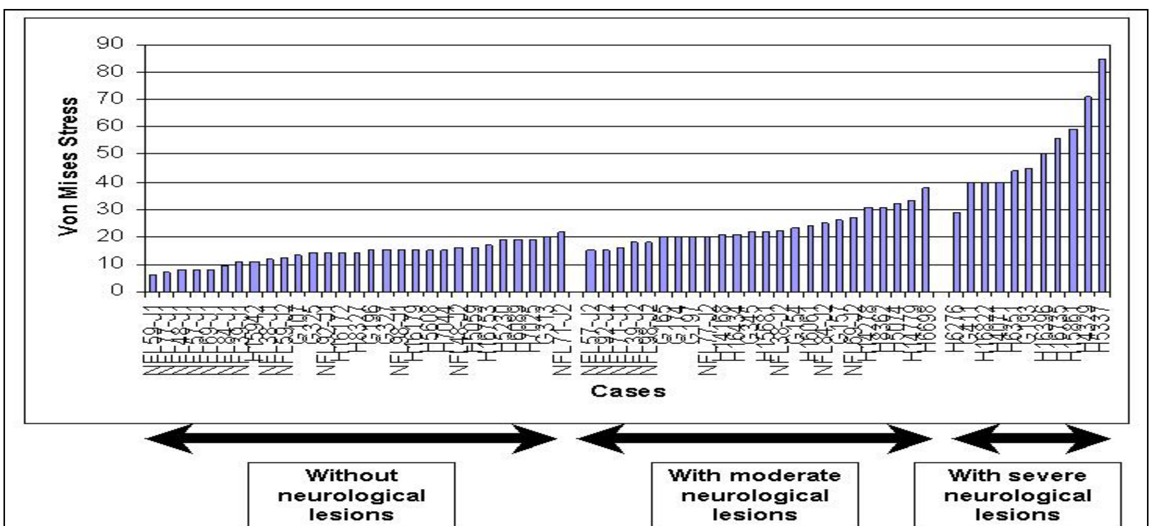

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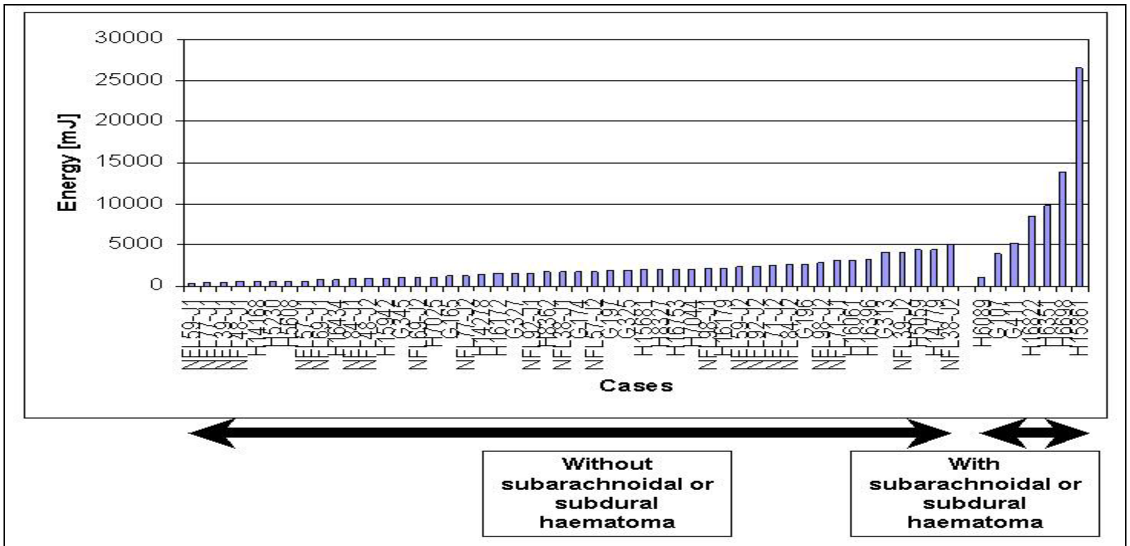
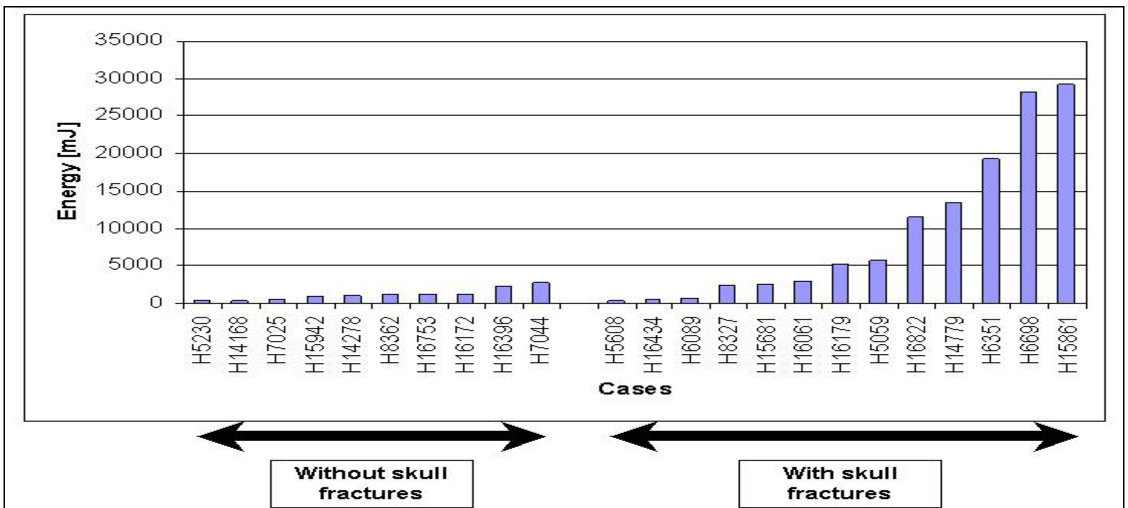


c)



d)

Figure 9 : Histograms of cranial and intra cerebral parameters computed with the ULP human head FE model for each accident case and for each injury group : a) Intra cerebral pressure, b) Intra cerebral Von Mises shearing stress, c) Global CSF strain energy, d) Local strain energy in the skull.

In figures 10 the injury risk curves relative to the four thresholds defined in this study are reported. Of particular importance, EB indicates the error committed through the regression between the logistic regression model and the observed cases. The regression is considered as appropriate for EB values ranging between -1 and 0. The more EB is close to 0, the better the regression model is. Main results can be summarized as presented in figure 10.

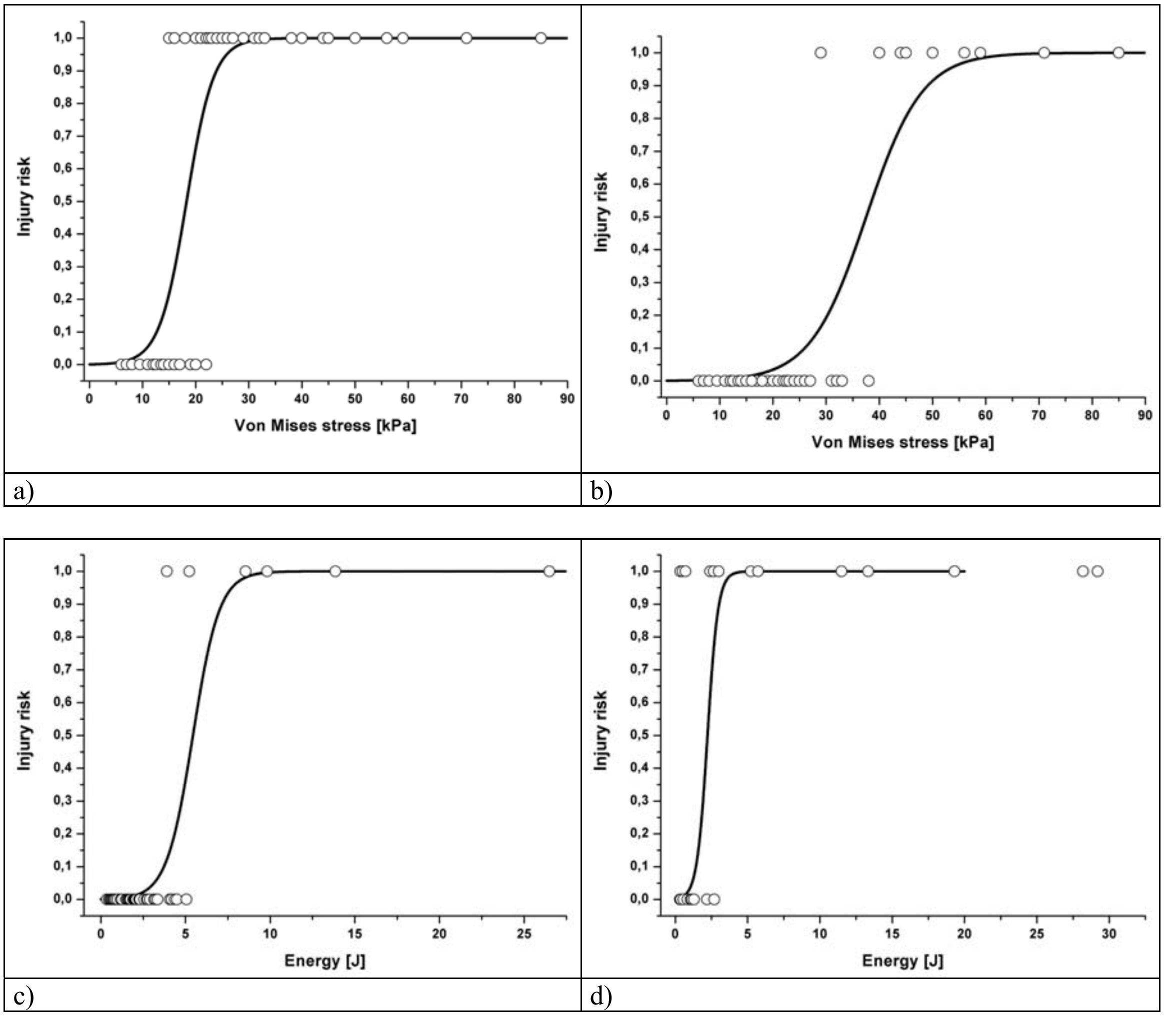


Figure 10 : Injury risk curves to specific injury mechanisms defined with the ULP human head FE model:
a) Von Mises stress reaching 18 kPa for a 50% risk of moderate neurological lesions (EB=-0.28)
b) Von Mises stress reaching 38 kPa for a 50% risk of severe neurological lesions (EB=-0.11)
c) Global strain energy in the CSF layer of 5.4 J for a 50% risk of a subdural heamatoma (EB=-0.09)
d) Local strain energy in the skull of 2.2 J for a 50% risk of a skull fracture (EB=-0.75)

DUMMY HEAD DEVELOPMENT AND EXPERIMENTAL INJURY CRITERIA

Frontal impedance recorded on the human head in vivo was used as a reference for validating a new headform prototype. The natural frequency, damping and de-coupled modal mass were used as reference parameters in the validation process. The objective was to prepare a physical model able to reproduce the same modal behaviour as the human head and thus, to describe the brain motion in the skull during an impact. This two-mass principle was fitted to a Hybrid III dummy head to demonstrate the applicability to a widely used device. In order to proceed rapidly with the design, a finite element model of the Bimass was constructed and used to establish design criteria. The HYPERMESH program was used for the geometry and explicit RADIOSS FE package for the numerical impact simulation. Theoretical modal analysis was obtained with the implicit ALGOR FE code. As a first step the standard Hybrid III head was modelled by four elements: the skull, the scalp, the base and the accelerometer amount. For the scalp, the material characteristics used were those of chlorure–vinyl, a polymer with a quasi incompressible viscoelastic behaviour. The typical Boltsmann model was used here with material constants kindly provided by the dummy manufacturer (First Technology Safety Systems) but not reported for confidentiality reasons. The last part of Hybrid III was a steel base screwed in the head, to represent a connection with the neck. This part was represented by a rigid body on which the skull accelerometer amount was attached. It was designed to record the three dimensional linear and angular rotation of the skull.

The modelling of the brain and its link to the base was the main part of the dummy head development. The adopted principal used a plug which was designed so that a brain – skull decoupling occurred at 150 Hz for the three rotational degrees of freedom and two for translation in the horizontal plane. The steel brain comprised two symmetrical parts, which could be separated in order to fix them around the upper part of the contact plug. The position of the four three-axis accelerometers of this “brain”constituted a normal frame of 36 mm visible in figure 11. The geometry of the contact plug is illustrated in figure 8 and was constrained by inner geometry of the Hybrid III head. The brain-skull decoupling at 150 Hz was achieved through the choice of the plug materials, which were Polyamide (ERTALON) with Young's modules ranging from 1500 to 5200 MPa. In order to define the correct variety of Ertalon, the brain connection system was numerically modelled and analysed with a fixed boundary condition at the base. Reasonable decoupling frequencies were observed around 150 Hz for a Young modulus of 3100 MPa which indicated that Ertalon 6 SA material should be chosen.

A steel cylinder made of two symmetrical parts to be attached to the lower part of the contact plug and to connect the whole system to the base of the Hybrid III head was then designed. Finally a padding cushion was fixed on the top of this cylinder in order to dampen the brain motion under extreme acceleration. The padding cushion resembled a large washer made of soft polyurethane with a low rebound coefficient (Eladip 500). Both elements (steel cylinder and cushion) are shown in figure 7 and 8. The whole system, illustrated on figure 11, was finally introduced in the Hybrid III. In this approach of the human head model, the focus was the brain-skull decoupling. Thus, the brain, the steel cylinder, the base, the accelerometer amount and the skull were intended to be rigid and were therefore defined as rigid bodies in the FE model. Only three elements were considered as continuous deformable components: the scalp (properties not given for confidentiality reasons), the contact plug (E=3982 MPa; ν=0.397), and the padding cushion(G_0=5.34 MPa;G_{00}=520MPa;β=15/s).

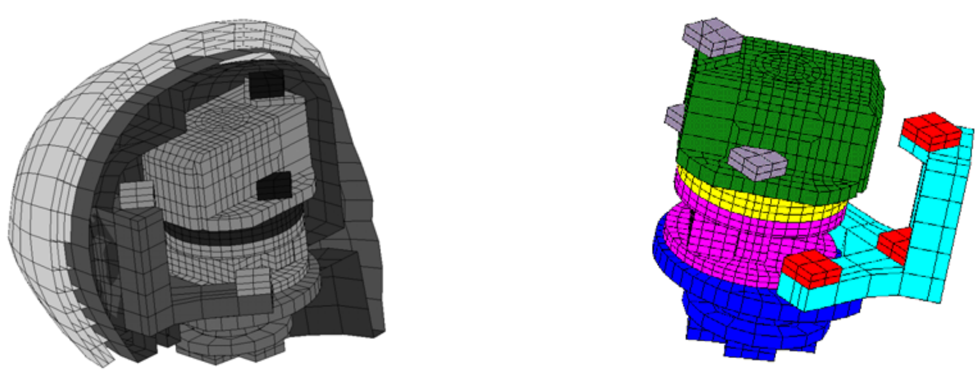


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Figure 11: General view of the main Bimass head components introduced in the Hybrid III dummy head (the contact plug is not visible but can be observed in figure 8).

The prototype (figure 12) was constructed using a Hybrid III dummy head for which the two tin masses of 300 g were cast and on which were fitted the accelerometer amount machined in steel, then

the Bimass system was integrated. The total mass of the Bimass prototype is 5.61 kg. Its general inertia properties are :
$I_{xx} = 1.585\ 10^{-2}\ kg.m^2$; $I_{yy} = 2.040\ 10^{-2}\ kg.m^2$; $I_{zz} = 1.767\ 10^{-2}\ kg.m^2$
This values are very close to the standard Hybrid III inertia properties (similar to the human head inertia properties) which are :
$I_{xx} = 1.590\ 10^{-2}\ kg.m^2$; $I_{yy} = 2.400\ 10^{-2}\ kg.m^2$; $I_{zz} = 2.200\ 10^{-2}\ kg.m^2$

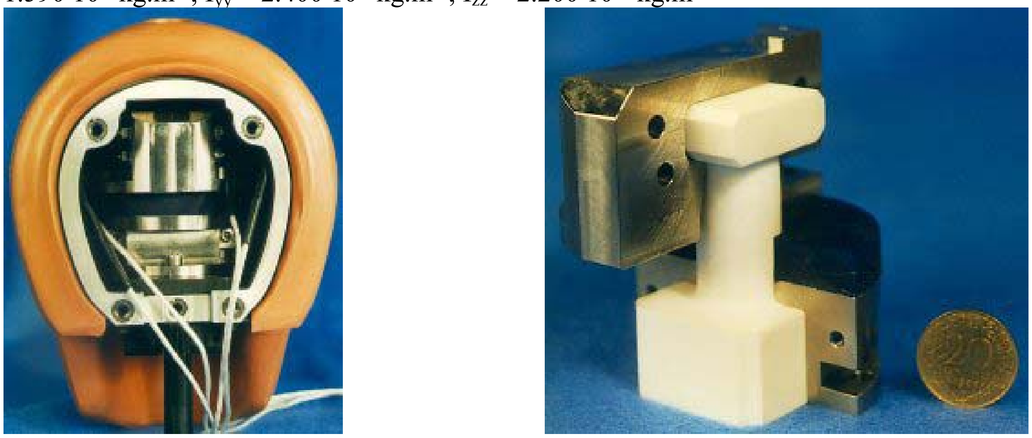


Figure 12 : Rear view of the Bimass dummy head and detail of the brain connection system (left half brain, contact plug, and anterior half of the steel cylinder and padding cushion).

The Bimass headform was validation in order to check if the brain-skull decoupling occurred close to the 150 Hz frequency. A de coupling of a mass of about 1 kg occurred at the natural frequency of 140 -150 Hz, with a dampening of 10%. This validated the Bimass against the human head in the 10 - 500 Hz frequency range. In the temporal domain, protected and unprotected head impacts were performed in order to check the dummy head’s reproducibility as well as the conformity of the prototype FE model’s response with the prototype itself. Dummy head related outputs are expressed by the time history of the brain and skull linear acceleration as well as the differential linear acceleration in figure 13. Rotational acceleration are even available but not shown in figures.

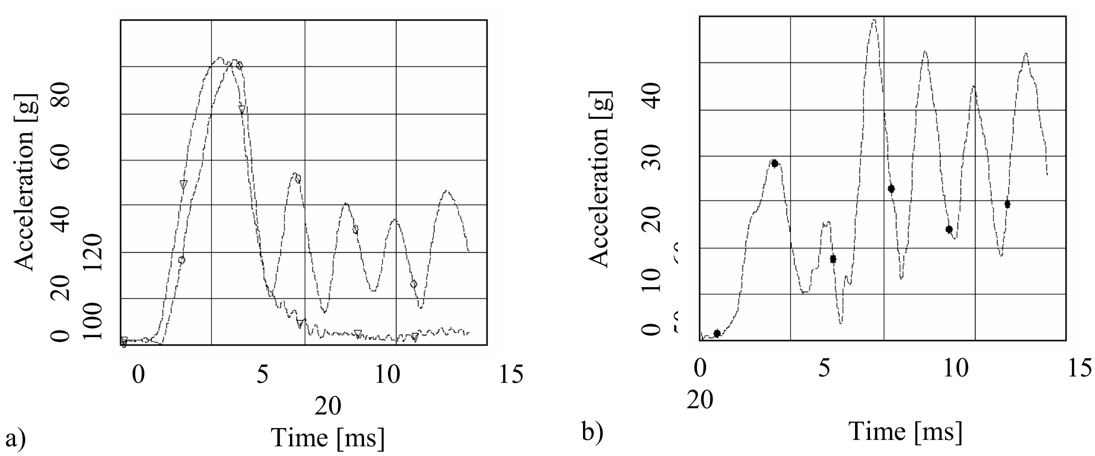
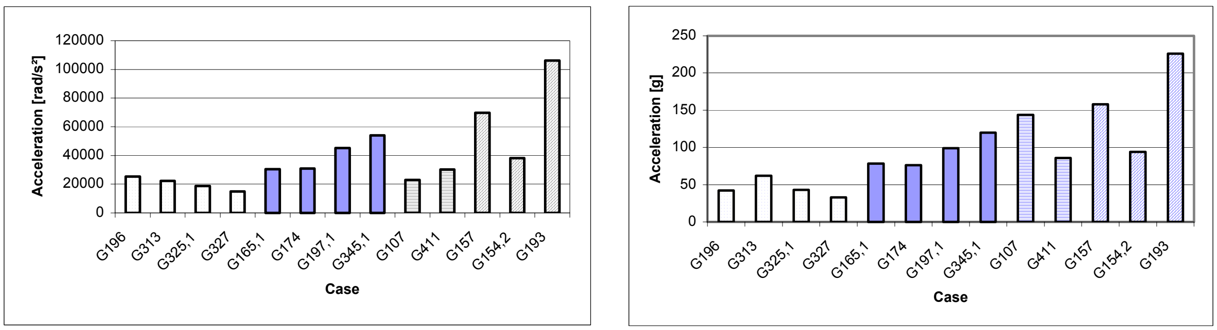


Figure 13 : Time evolution of Bimass response for accident replication case G174 :
a) brain(circle) and skull acceleration b) brain-skull relative acceleration.

This dummy head has been used at the experimental level for the reconstruction of the 13 motorcycle accidents. The outputs of the proposed physical head model are the skull acceleration, the brain acceleration and the brain-skull differential acceleration. Each one of these parameters can be related to a specific head injury mechanism. Skull acceleration may, in the future, related to skull deformation and lesions such as extra dural haematoma or skull fracture. The brain-skull differential acceleration or relative motion is an injury parameter to indicate subdural haematoma or focal cerebral contusions. Brain acceleration remains the parameter to indicate diffuse axonal injury and intra-cerebral contusions or haematoma. An attempt is given in figure 14 where the histograms of the brain maximum rotational acceleration are given for each case (Figure 14a) as well as the maximum brain-skull relative linear acceleration (Figure 14b). The histograms show a higher brain acceleration for the concussed cases (3 to 4 krad/s²) and high values for brain-skull relative acceleration when a SDH occurred (close to 130 g).

a) b)



on-injured oncussion DH Fracture

Figure 14a : Histogram of brain peak rotational acceleration a) and brain-skull relative linear acceleration b), given by the reconstruction of the13 accidents with Bimass dummy head.

An attempt for Bimass related injury criteria can also be made as follows. The histograms in figure13 show a higher brain acceleration for the concussed cases (3 to 4 krad/s²) and high values for brain-skull relative acceleration when a SDH occurred (130 g). The main limits of this study are the very limited number of SDH cases involved and obviously the fact that the Bimass 150 is not yet able to incorporate skull fracture in the analysis.

DISCUSSION

It is the first time that extensive real world accident reconstruction, based on living human head data is used in order to derive tolerance limits to specific injury mechanisms according to the authors' knowledge. The tolerance limits to neurological moderate or severe injuries with a 50% risk are established at 18 kPa and 38 kPa respectively. These values can be compared to previous reported attempts such as11 kPa by Zhou et al. [2] who reconstructed a car accident using the FEM of the head from Ruan et al. [1] or 15 kPa proposed by Kang et al. [10] who reconstructed a motorcycle accident using an initial version of the ULP-FEM of the head and finally 27 kPa suggested by Anderson et al. [25] who reconstructed 16 experimental head impacts on living sheep. These values are in the same range as the ones obtained in the present study by using the same method. Bandak et al. [8] suggested head angular acceleration as a limit for this kind of injury. The tolerance limit to subdural haematoma established in this study rises to 5.5 J with a 50% risk of occurrence. In previous studies this injury is evaluated by parasagittal bridging veins elongation or elongation rate computed with the FE model as suggested by Bandak et al [8]. Tolerance limits for skull fracture have been reported in the literature from experimental data in terms of impact force (4 to 14 kN) by Yoganandan et al. [18] as well as in terms of strain energy (around 2 J) by Gurdjian et al. [26]. In the present study a tolerance limit to skull fracture is established numerically at 2.2 J with a 50% risk. A tolerance limit in terms of the head applied force is also established in the framework of the present study but not illustrated because of the less confident error values obtained from the logistic regression (EB = -0.5958). Nevertheless, the tolerance limit derived (with 50 % risk) relies upon 3560 N. That value is in the same range as the currently proposed limits more especially as the whole skull bones are involved in the presented accident reconstructions and considered as isolated, and that of this fact the obtained tolerance limit of skull bones fractures reflects a mean value for skull bone, thus not a specific tolerance limit relative to an isolated bone.

As a global consequence of the presented results it can be stated that numerical head surrogates have been designed which enable it to optimise numerically or experimentally head protection systems against specific injury mechanisms.

CONCLUSIONS

The present paper presents original numerical human head models and ITS validation through modal and temporal analysis. An improved FE model of the human head is presented with two main originalities which are skull fracture modelling and the simulation of the brain-skull interface by fluid-structure interaction. Results show that it was possible to reconstruct the head kinematics of Troseille's experiments and to predict the intra-cranial pressure accurately at sites near to the impact location. However, the pressure predictions became less accurate as the distance from the impact location increased, specially for long duration impacts. The skull stiffness and fracture force were very accurately predicted when compared with values measured by Yoganandan [18]. This specify the model’s application domain to very short (2 ms) and very long (15 ms) impact duration.

This head model is then used to simulate 64 real world accidents. For the helmeted head impacts the head kinematics was obtained by experimental accident replication. For the direct pedestrian head impacts, initial head velocity and position were defined by pedestrian kinematics simulation. These data were transferred to Strasbourg University for the numerical accident simulation. The outputs from the model were compared with the head injuries recorded for each case. It was concluded that AIS does not correlate well either with the conventional test criteria such as acceleration, HIC and GAMBIT or with intra- cerebral parameters delivered by the improved models. However, when head injury was examined the following five distinct groups emerged: uninjured, moderate neurological injuries, severe neurological injuries, subdural haematoma and skull fracture. Histograms of several intra-cranial mechanical parameters were then correlated with injury types in order to derive tolerance limit for specific injury mechanisms.

For the numerical ULP human head FE model following limits were drawn: (i) a brain Von Mises stress reaching 18 kPa for a 50% risk of moderate neurological lesions; (ii) a brain Von Mises stress reaching 38 kPa for a 50% risk of severe neurological lesions; (iii) a global strain energy in the CSF layer of 5.4 J for a 50% risk of a subdural haematoma; and (iv) a local strain energy in the skull of 2.2 J for a 50% risk of a skull fracture.

For the Bimass 150 dummy head prototype two experimental criteria were derived. A angular brain rotation of 9240 rd/s^2 for a 50% risk of neurological injury. A linear brain acceleration of 273 g for a 50% risk of subdural haematoma.

At the experimental level we present the development and validation of a new dummy head prototype named Bimass. It has been constructed using a Hybrid III headform and comprises two masses : a skull and a mass to represent the brain attached to the skull with a damped spring system. The novel feature of this device is that it can simulate the brain - skull relative displacement at a frequency close to 150 Hz. This is the frequency recorded in tests in vivo. The measurements from the physical device obtained in impacts onto a flat anvil : rotational and linear brain and skull acceleration, rotational and linear brain-skull relative acceleration and brain-skull relative displacement closely matched the outputs from Bimass FE model. Furthermore, The modal response of the prototype physical headform in the frequency domain is in agreement with expected results in terms of natural frequencies as well as in mode shapes. The Bimass headform has been shown to be robust in typical impact tests. The eight tri-axial accelerometers connected to the brain and the skull as well as the different elements of the prototype were not in any way damaged during high energy impacts. In the temporal domain the headform was shown to be repeatable. Very first attempts for experimental injury criteria are proposed.

However, given the low number of cases involved in each injury group, this accident analysis must be continued. The first results presented in this paper demonstrate the interest of the proposed approach, and the need to analyse sustained injury by injury mechanisms and not simply by AIS value. This study shows that the final target, which is the definition of tolerance limits for a given head injury mechanism, can be reached. In the near future it will therefore be possible to optimise head protection systems against biomechanical criteria.

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