Human head tolerance limits to specific injury mechanisms inferred from real world accident numerical reconstruction

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ABSTRACT. *This study presents an original numerical human head model which is validated in both modal and temporal domain against vibration analysis in vivo and cadaver impact tests. The head finite element model developed presents two particularities : one at the interface between the brain and the skull were fluid structure interaction is taken into account, the other at the skull modelling level by integrating the bone fracture prediction. Validation shows that the model correlated well with a number of experimental cadaver tests. This improved numerical head surrogate is then used for numerical real world accident reconstruction. Helmet damage from thirteen motorcycle accidents is replicated in drop tests in order to define the head's loading conditions. A total of twenty two well documented American football players head trauma is reconstructed as well as twenty nine pedestrian head impacts on car windscreens. By correlating head injury type and location with calculated mechanical parameters, it is possible to derive new injury risk curves relative to specific injury mechanisms.*

RÉSUMÉ. *Cette étude présente une modélisation numérique originale de la tête humaine validée dans les domaines fréquentiel et temporel au regard d'une analyse vibratoire* in vivo *et de tests d'impacts de cadavres. Le modèle par éléments finis développé présente deux particularités : l'une à l'interface entre le cerveau et le crâne où l'interaction fluide structure est prise en compte, l'autre au niveau de la modélisation du crâne par la prédiction des fractures osseuses. La validation montre que la réponse du modèle est bien corrélée avec plusieurs tests expérimentaux sur cadavres. Ce substitut numérique amélioré de tête est ensuite utilisé dans le cadre de reconstructions numériques d'accidents réels. Les dommages sur le casque de treize motocyclistes sont reproduits lors de tests de chute libre pour définir les conditions de chargement de la tête. Un total de vingt-deux traumatismes crâniens bien documentés de footballeurs américains est reconstruit tout comme vingt-neuf impacts de têtes de piétons sur des pare-brises. En corrélant le type de lésions de la tête et leur localisation avec des paramètres mécaniques calculés, il est possible d'obtenir de nouvelles courbes de risques de lésions relatives à des mécanismes de lésion spécifiques.*

KEYWORDS: *head injury, finite element method, real world accident reconstruction, injury mechanism, tolerance limits.*

MOTS-CLÉS : *lésions à la tête, méthode des éléments finis, reconstruction d'accidents réels, mécanismes de lésion, limites de tolérance.*

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1. Introduction

Human 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 thirty 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 is criticised by (King *et al*., 2003) because of its limited ability to predict a probability of brain injury for example. It has been suggested that specific deformation of skull material and brain tissue and a measure of the relative motion between the brain and the skull would be much better means of assessing head protection.

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

Finite Element Methods (FEM) was considered to be the best tool with which to investigate the dynamic response of the head under impact condition. To date, more than ten different three dimensional (3D) head FEM have been described but only (Ruan *et al*., 1993) model and (Zhou *et al*., 1996) modifications of that model) was validated and then used for accident reconstruction to investigate brain injury tolerance limits. Most of the essential head components are incorporated in this model, which is meshed with 37 040 elements. Recent modifications of this model achieved by (Al-Bsharat *et al*., 1999) include the addition of a three layered skull and an improvement of the interface between the brain and the skull to allow the brain to move more freely relatively to the skull. The model is validated against the brain pressure data from impact tests onto the front of the head of cadavers and against relative motion between the brain and the skull using data obtained from the high speed X-ray experiments. An improved version of the Wayne State University model has been published by (Zang *et al*., 2001). A refined brain meshing was proposed (314 500 elements) and the validation procedure showed realistic results for linear and rotational accelerations up to 200 g and 1 200 rad/s^2 respectively. (Bandak *et al*., 1999) also published a finely meshed head FEM and proposed an in depth experimental and numerical analysis of the cerebrospinal fluid (CSF) layer influence on the brain response.

(Zhou *et al*., 1996) 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*., 2003) presented a detailed methodology for the assessment of mild traumatic brain injury based on the reconstruction of American professional football accidents, using Zhou's head FEM. These findings suggested that mild traumatic brain injury occurred with a brain Von Mises stress of 0.07 kPa and a brain pressure of 0.03 kPa. The main objective of the Wayne State University Brain Injury Model (WSUBIM) was however to evaluate the correlation between brain Von Mises stress with angular head acceleration on the one hand and brain 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, 2003). In 2001, Bandak *et al*., presented a first version of a head injury assessment tool based on a very simplified head FEM and called SIMon. An attempt was made in this study to distinguish between brain injuries 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*., 1999) developed the ULP FEM of the head at the University of Strasbourg. The first accident replicated was using an initial version of this model that involved a head impact caused by a motorcycle accident. This work has been achieved by (Kang *et al*., 1997). The results thereof showed that it was possible to compare intra cranial field parameters with neuropathological brain injury details and hence leading to the conclusion that brain pressure did not correlate well with brain haemorrhage but that brain Von Mises stress distribution correlated very well with this type of injuries. The research described in this paper is a more extensive use of the ULP FEM of the head in real world accident simulation in order to investigate injury mechanisms and tolerance limits.

Currently, real world accident analysis is used in an attempt to correlate a known head injury parameter with the Abbreviate Injury Scale (AIS) value sustained. An attempt by (Chinn *et al*., 1999) 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 cranial mechanical parameters calculated with the ULP FEM of the head, 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.

The final objective of the present work is to derive new head tolerance limits to specific injury mechanisms by using the ULP FEM of the head in the framework of accident reconstruction. Hereafter the model development and validation are reported before the presentation of the 64 reconstructed real world accidents. Injury risk curves for brain moderate and severe neurological lesions, subdural or subarachnoidal haematoma (SDH or SAH) as well as skull fracture are then presented before discussion and concluding remarks.

2. The ULP FEM of the head

2.1. *Geometry and meshing*

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 from (Fener *et al.*, 1985), were used to mesh the 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 1 shows the 3D skull surface obtained by digitising external and internal surfaces of the skull as well as the meshed model. Figure 2 shows a cross section of the model and illustrates the anatomical features that are taken into account in that model. The main anatomical features modelled are the skull, the falx, the tentorium, the subarachnoidal space, the scalp, the cerebrum, the cerebellum, and the brain stem as well as the ventricles.

The finite element mesh is continuous and represents an adult head. The type of elements used is the following: 3 nodes (tria) and 4 nodes (quad) elements for the shell elements and 6 nodes (hexa) and 8 nodes (tetra) elements for the brick elements. The average size of the ridge of an element is about 5 mm. The meshing is assumed to be regular in terms of element dimension, angle and warpage. 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 subarachnoïdal space between the brain and the skull which was, as a first step, represented by one layer of brick elements to simulate the CSF. Lagrangian formulation was therefore selected and the link between the brain and the skull was modelled by an elastic material validated against the in vivo vibration analysis proposed by (Willinger *et al.*, 1995). In order to improve the simulation of the CSF for long duration impacts, a three layered brick element interface between the brain and the skull is proposed and an Arbitrary Lagrangian Eulerian (ALE) formulation available in RADIOSS © finite element 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 separates the cerebrum and the cerebellum, and the falx separates the two hemispheres. Brick elements were used to simulate the CSF that surrounds these membranes in the same way as between the brain and the skull. The ventricles were also integrated. A layer of brick elements also modelled the scalp, which surrounds the skull and the facial bone. Overall, the Lagrangian version of the head model consists of 13 208 elements divided in 10 395 brick elements (5 376 for the brain, 2 870 for the CSF and 1 530 for the scalp) and 2 813 shells elements (2 424 for the skull and 389 for the membranes). Its total mass is 4 772 g. The ALE version contains 6 486 brick elements which describe the CSF layer and the ventricles instead of 2 870 elements for the Lagrangian version. The total number of elements for this latter head model is therefore increased to 16 824 elements.

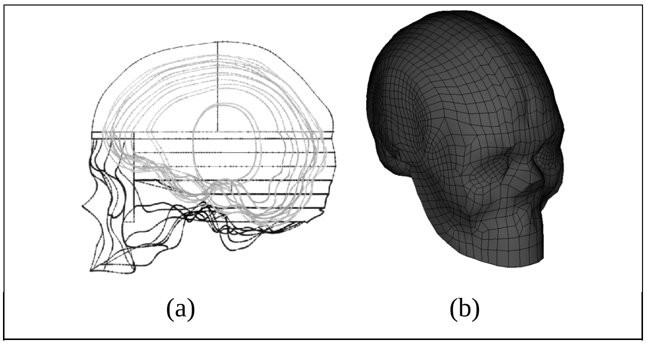


Figure 1. *ULP FEM of the head construction, (a) 3D skull surfaces, (b) skull meshing*

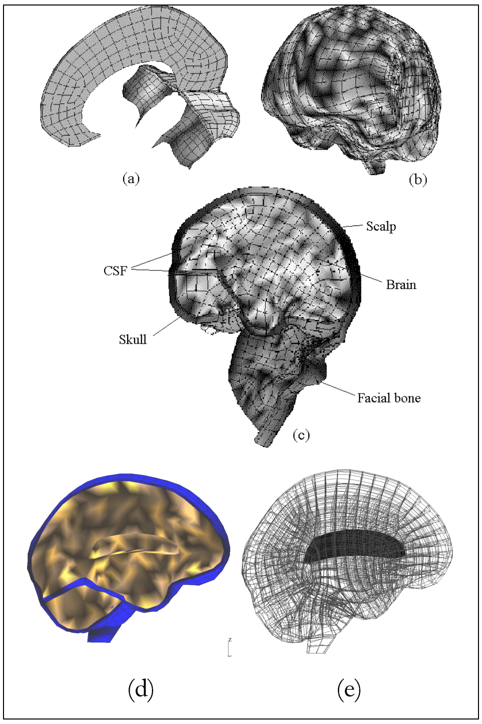


Figure 2. *Meshing of the intra cranial medium, (a) falx and tentorium, (b) brain, (c) overview of the head, (d) brain and CSF, (e) ventricules*

2.2. *Material properties*

Material characteristics are very important to the success of a finite element model and Table 1 lists the properties of the materials used for the ULP FEM of the head. Material properties of the CSF, the scalp, the facial bones, the tentorium and the falx are all elastic, isotropic and homogenous. The viscous elastic properties assigned to the brain were scaled from (Khalil *et al.*, 1977) data. The behaviour in shear was defined by a viscous elastic law.

Two separate formulations are used for the CSF modelling: the Lagrangian and the ALE. The main objective was to evaluate under which condition the CSF flow occurs and how this phenomenon influences the intra cranial mechanical response. For the Lagrangian version, the Young's modulus of the subarachnoidal space was determined by (Willinger *et al.*, 1995) using modal analysis, based on the fact that a brain skull decoupling occurs at the first natural frequency of the head at around 100 and 150 Hz as reported in Table 1. A large deformation formulation was used in order to have realistic strain estimation in this layer of brick elements. In that ALE version of the model the CSF is represented by a hydrodynamic fluid defined by a Bulk modulus of 21.9 GPa obtained through the study of (Willinger *et al.*, 1995).

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 the 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, 1970) and integrated in the (Tsaï-Wu, 1971) criterion.

The material properties of the intra cranial membranes and the scalp are similar to those used by (Zhou *et al.*, 1996) and also reported in Table 1.

2.3. *Model validation*

A total of eight instrumented cadaver impacts were reconstructed with the objective to validate the ULP FEM of the head under very different impact conditions. Currently FEM of the head are validated against (Nahum *et al.*, 1997) impact and this was satisfactorily achieved with the first version of the model (with Lagrangian formulation), in a previous study realised by (Kang *et al.*, 1997). The impact duration of Nahum's test was about 6 ms. (Kang e*t al.*, 1997) 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 (Trosseille *et al.*, 1992) and two extremely short cadaver impacts (to the front and to the vertex) inducing skull fracture, published by (Yoganandan *et al.*, 1994).

Table 1. *Material properties of the ULP FEM of the head*

structure	ρ [kg/m3]	E [MPa]	ν	σ_t [MPa]	σ_c [MPa]	K [MPa]	G_0 [KPa]	G_{inf} [KPa]	β [m/s]
cortical bone	1800	15000	0,21	90	145				
trabecular bone	1500	4500	0	35	35				
subarachnoidal space	1040	0,012	0,49						
brain and cerebellum	1040					1125	49	16,7	0,14
scalp	1200	16,7	0,42						
falx and tentorium	1140	31,5	0,23						

Table 2. *Main characteristics of experimental cadaver tests from the literature as used for validation (LA: linear acceleration, RA: rotational acceleration, Force: peak value)*

Test	Impact point	Impactor [kg]	Impactor velocity [m/s]	Force [N]	LA max [g]	RA max [rad/s²]	Duration [ms]
Nahum et al. (1977)	front	Cylinder [5,6]	6,3	6900	198		6,5
Trosseille et al. (1992) (MS 428_2)	face	Steering wheel [23,4]	7		102	7602	15,8
Yogonandan et al. (1994)	vertex	Rigid sphere [1,213]	7,3	10500			2

Table 2 summarizes the main characteristics of the three classes of impact, *i.e*. medium, long and short duration impacts. The RADIOSS © code developed by MECALOG was used for the finite element analysis. The method of one point integration was used for all analysis with an hourglass energy below 5% of the total involved energy. The time step relies upon 10^{-3} ms which agrees with the propagation time in the involved materials. This validation procedure is detailed by (Willinger *et al*., 2000).

In the whole undergoing study the head is assumed to be free in its six degrees of degrees of freedom. In fact, the only boundary conditions that could influence the dynamical response of the head under impact are localized at the neck level. A complementary study shows that the neck does only influence the dynamical response of the head under impact 30 ms after the beginning of the impact. In the whole cases under study, shall it be in the validation procedure or in the accident reconstruction, the duration of the impact does not exceed 15 ms. Therefore the head is assumed to be free in terms of translation and rotation. Moreover, the gravitation acceleration is neglected in comparison to the high levels of acceleration which are applied to the head in each kind of impact. Eventually, the Frankfort plan of the head is considered as horizontal at the beginning of the impact is each simulation under study except for:

– The numerical replication of (Nahum *et al*., 1977) experimental tests. In that specific case, the Frankfort plan of the head is inclined of about 20° underneath the horizontal;

– The numerical replication of the pedestrian accidents where the initial relative position between the head and the windscreen is taken into account as detailed below.

The simulation of the intra cranial behaviour was satisfactory when short impacts were concerned such as reported by (Nahum *et al*., 1977) (see Figure 3).

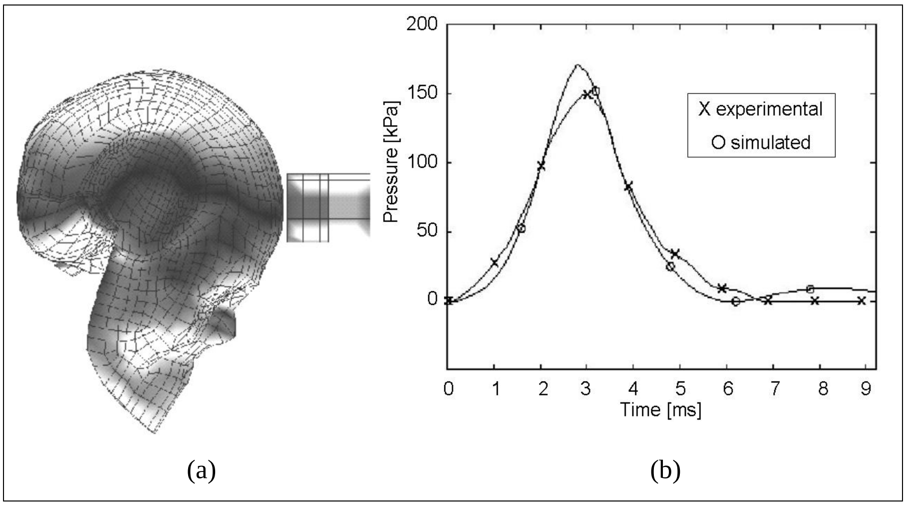


Figure 3. *(a) ULP FEM of the head in frontal impact configuration. (b) Measured by (Nahum* et al.*, 1977) (x) and calculated (o) brain frontal pressure*

For more dampened impacts as reported by (Trosseille *et al*., 1992), 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 (Trosseille *et al*., 1992) 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 a velocity 5 m/s. Moreover the CSF flow was represented by three layers of brick elements in ALE formulation. This ALE formulation is specially indicated for coupling between fluid and solid in finite element 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 (*i.e*. Lagrangian formulation). In that case, the method of eight integration points was used for the whole analysis with an hourglass energy below 5% of the total involved energy. The time step relies upon 10^{-3} ms which agrees with the propagation time in the involved materials. Both cases are confronted to (Trosseille *et al*., 1992) experimental data in terms of brain pressures in frontal and occipital regions as shown in Figure 4. The main result is that at each location the brain pressure time history is better reproduced in the ULP FEM of the head including ALE formulation for CSF. There still remains 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, the model response shows fewer oscillations when ALE formulation is used for the CSF and the variations towards experimental brain 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 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.

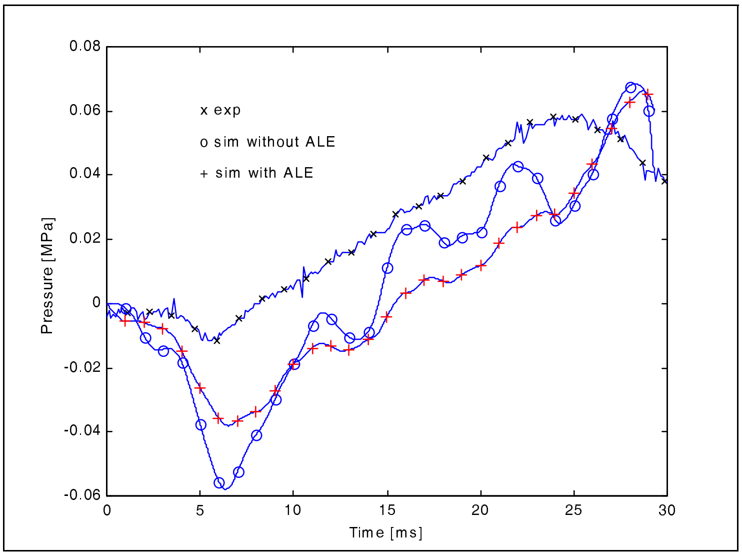


Figure 4. *Experimental (Trosseille* et al., *1992) and simulated (with and without) ALE formulation for CSF frontal brain pressure for a long duration damped impact with high angular component*

At the skull response level, the numerical force deflection curves are compared to the average dynamical response of experimental data obtained by (Yogonandan *et* *al.*, 1994). 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 head dynamical response correlated well with a great variety of experimental cadaver tests and predicted brain 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.

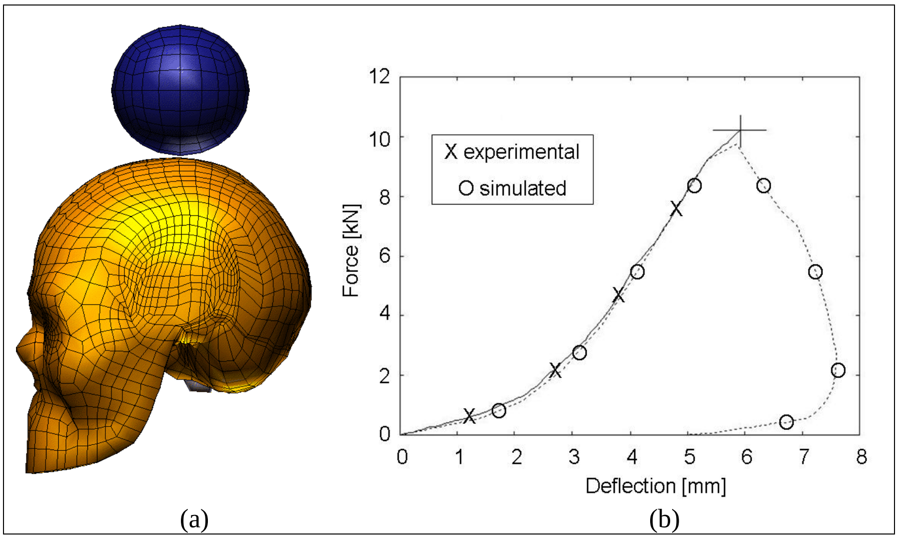


Figure 5. *(a) ULP FEM of the head in vertex impact configuration. (b) Measured by Yoganandan et al. (1994) (full line) and calculated (doted line) to the head applied force versus deflection curve. (+) indicates the bone rupture point*

3. Real world accident reconstruction

The ULP FEM of the head was then used for extensive real world accident reconstructions. A total of 64 head impacts were simulated: 35 protected and 29 unprotected direct head impacts. Given the time duration of the impacts taken into account in the present study, lasting between 5 and 15 milliseconds, the Lagrangian version of the 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 methodologies were designed in order to define the boundary conditions of the model itself.

3.1. *Reconstruction of the head 3D kinematics for helmeted victims*

Concerning the helmeted victims (*i.e.* the motorcyclists and the American football players) it was an experimental head impact reconstruction which allows the definition of the skull 3D kinematics. A total of thirteen motorcyclists cases were replicated with drop tests of a helmeted head form at the Transport Research Laboratory of 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 achieved by (Chinn *et al*., 1999), TRL replicated the helmet damages 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.

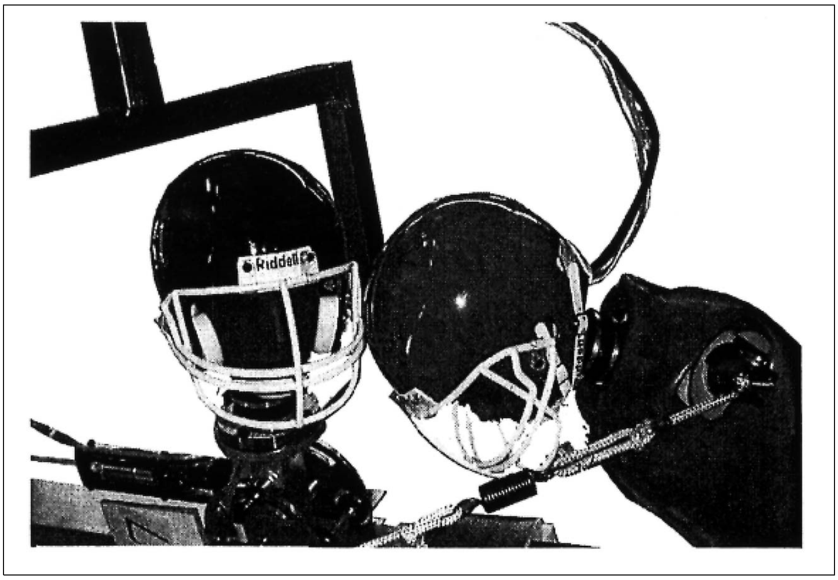


Figure 6. *Experimental protected head impact replication of American football player collision*

The American Football players cases were studied in collaboration with the Biokinetics Company in Canada and detailed in by (Shewchenko *et al.*, 2001). 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 (Newman *et al.*, 1986). 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.

For the 35 accident cases involving helmeted heads and reconstructed experimentally, the reconstruction report was transferred to ULP. 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 supposed as rigid and this was used as the input to the numerical accident simulation. Due to the duration of the impacts, the intra cranial material properties used were the ones presented in Table 1.

3.2. *Reconstruction of the head 3D kinematics for pedestrian victims*

As mentioned previously, the non protected head impacts came from pedestrians 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. (2002). 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 FEM of the head, this time with a three layered deformable skull model, 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.

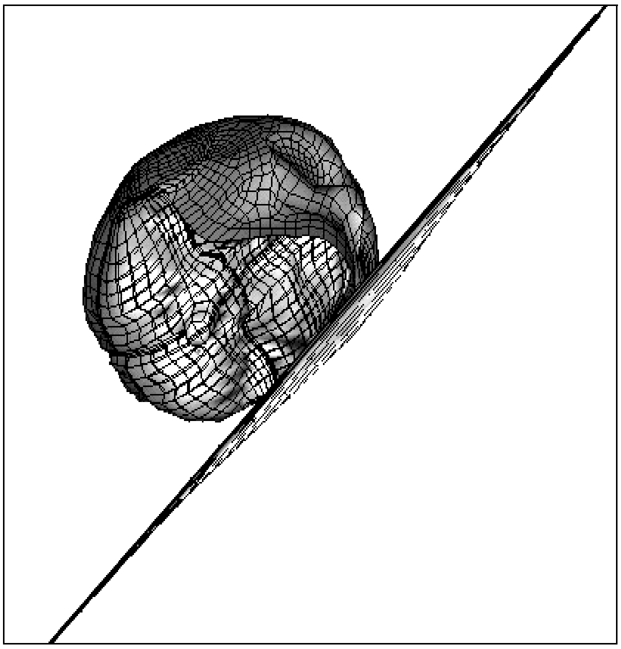
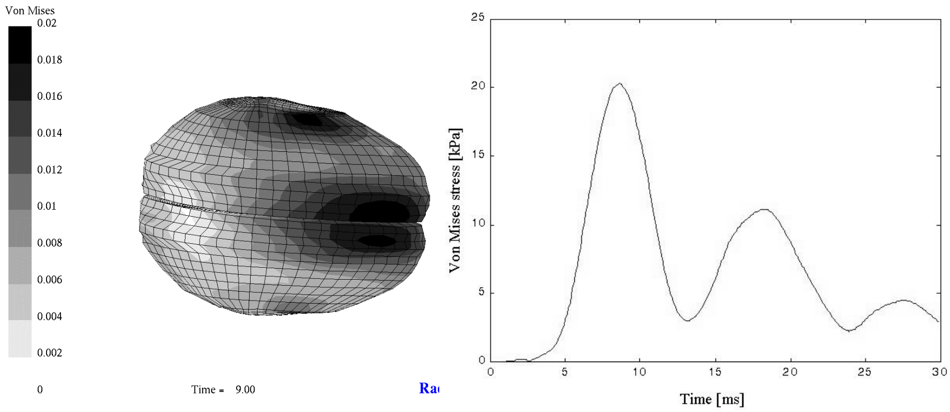


Figure 7. *Lateral view of the pedestrian head impact on the middle of the windscreen for case H8362, 4 ms after the first contact*

3.3. *Computation of the cranial and intra cranial response*

For the 64 head trauma retained for this study, the intra cranial response was then computed with the RADIOSS © code in order to calculate the brain pressure and Von Mises stresses as well as the global strain energy in the CSF as a function of time. The method of one point integration was used for all analysis with an hourglass energy remaining below 5% of the total involved energy. The time step relies upon 10^{-3} ms which agrees with the propagation time in the involved materials. 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 brain Von Mises stress levels are distributed at different parts of the brain according to the considered case. For example, Figures 8a and 8b illustrates the calculated brain 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, the skull response was computed in terms of deleted elements (Figure 8c) but also interaction force between the headand the windscreen and global strain energy in the skull.



(a) (b)

Figure 8 a et b. *Brain Von Mises stress for the motorcyclist G174. (a) Brain Von Mises stress field 9 ms after the beginning of the impact, (b) Time history of the brain Von Mises stress at the location where it reaches its maximum value*

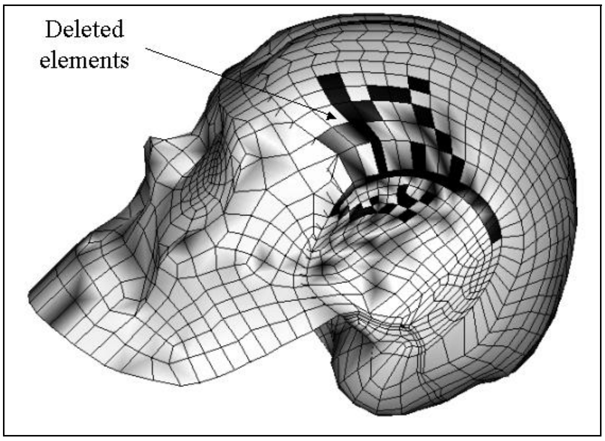


Figure 8c. *Deleted elements (colored in black) for H6351 unprotected pedestrian accident case*

4. Tolerance limits and injury criteria

When the type of lesion, rather than the AIS, was used for comparison, then five distinct groups emerged from our accident data, *i.e.* uninjured (29 cases), brain moderate neurological lesions (24 cases), brain severe neurological lesions (11 cases), SDH or SAH (7 cases) and finally skull fractures (19 cases). 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 brain pressure, the maximum brain Von Mises stress, the maximum strain energy in the CSF layer and the maximum strain energy in the skull bone, calculated with the ULP FEM of the head 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 brain pressure, because of the wide variation within a victim group was not responsible for the brain neurological lesions. The maximum brain Von Mises stress, illustrated in Figure 9b, is of greater interest and show better correlation with brain neurological lesions. Uninjured victims sustained low values whereas brain moderate neurological lesions cases, sustained clearly higher values and cases with brain severe neurological lesions presented greater brain Von Mises stresses than those of the brain moderate neurological lesions group. The third histogram in Figure 9c is related to the maximum global strain energy in the CSF layer and shows that for the victims with SDH or SAH, 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 global strain energy. It appears that this parameter is a good candidate for a skull fracture prediction criteria given the high values computed in the cases where a fracture occurred.

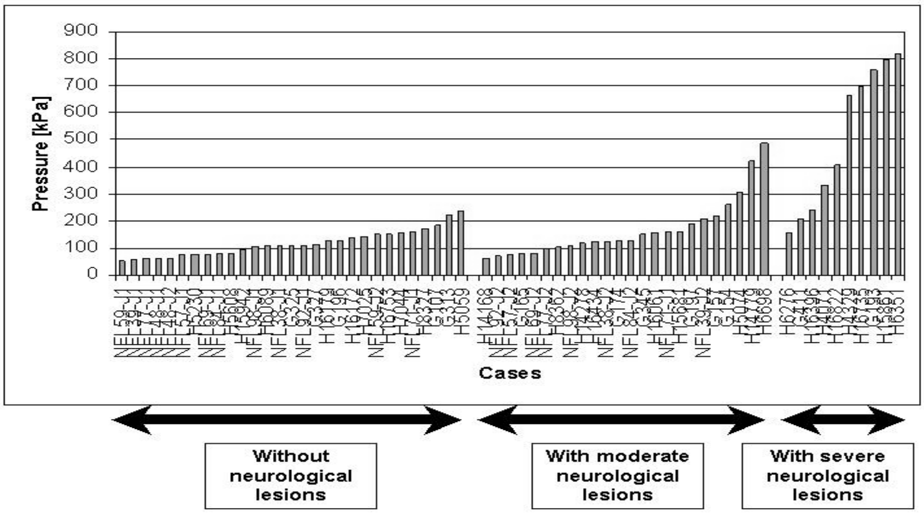


Figure 9a. *Histograms of brain pressure computed with the ULP FEM of the head for helmeted and non helmeted victims*

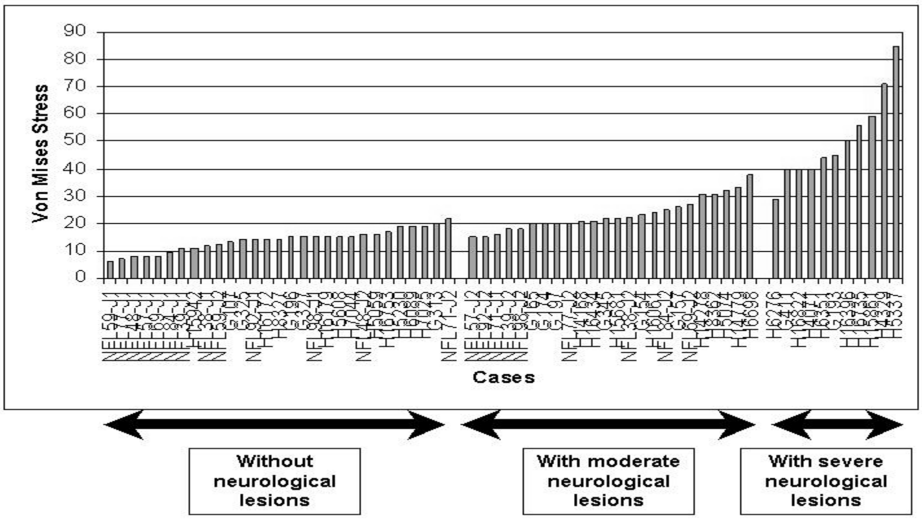


Figure 9b. *Histograms of brain Von Mises stress computed with the ULP FEM of the head for helmeted and non helmeted victims*

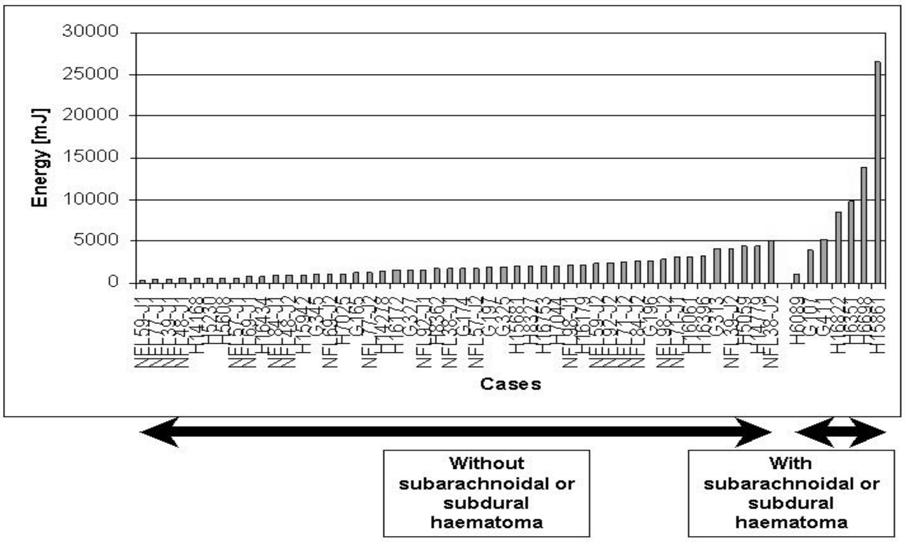


Figure 9c. *Histograms of global strain energy of the subarachnoidal space computed with the ULP FEM of the head for helmeted and non helmeted victims*

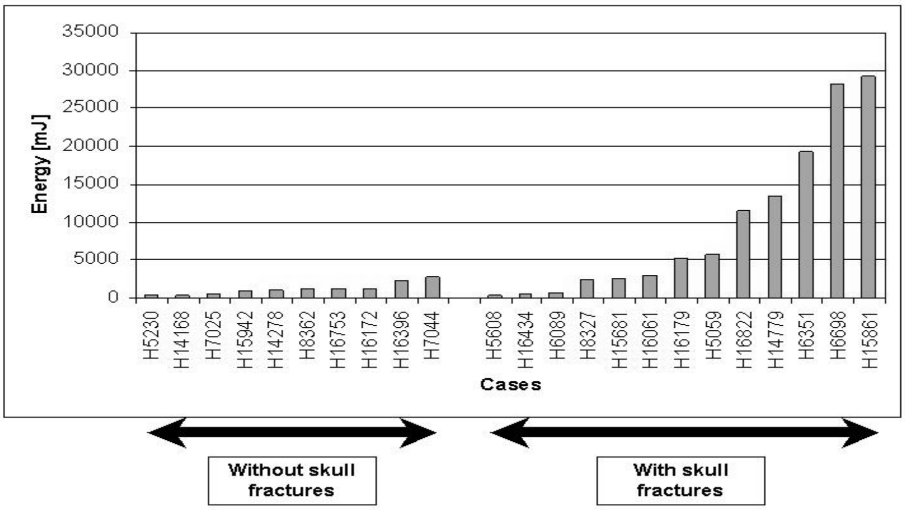


Figure 9d. *Histograms of global strain energy of the skull computed with the ULP FEM of the head for non helmeted victims*

The above analysis, conducted 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 is reasonable indicator respectively for SDH or SAH and skull fracture. For the four thresholds defined thought the histograms (Figures 9a, 9b, 9c and 9d), a statistical regression analysis using the so called Modified Maximum Likelihood Method developed by (Nakahira *et al.*, 2000) leads to the establishment of tolerance limits against specific injury mechanisms in terms of injury risk curves.

In Figures 10 the injury risk curves relative to the four criteria 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 Figures 10.

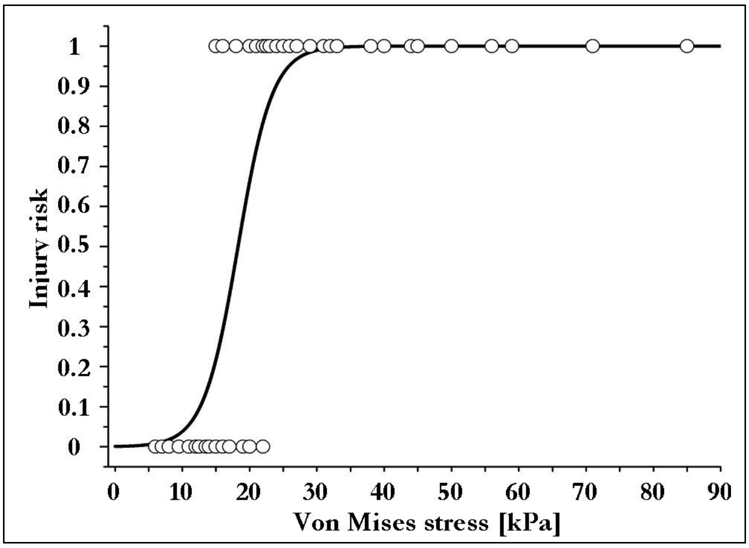


Figure 10a. *Injury risk curves to brain moderate neurological lesions defined with the ULP FEM of the head: brain Von Mises stress reaching 18 kPa for a 50% risk (EB= - 0.28)*

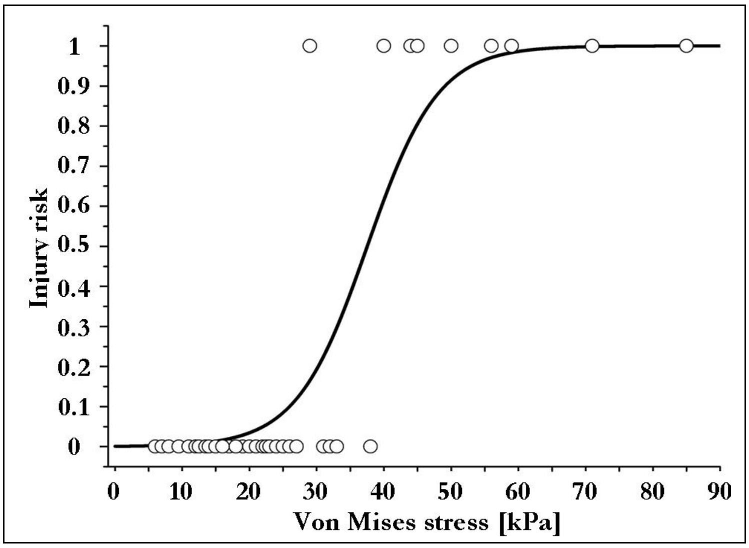


Figure 10b. *Injury risk curves to brain severe neurological lesions defined with the ULP FEM of the head: brain Von Mises stress reaching 38 kPa for a 50% risk (EB= - 0.11)*

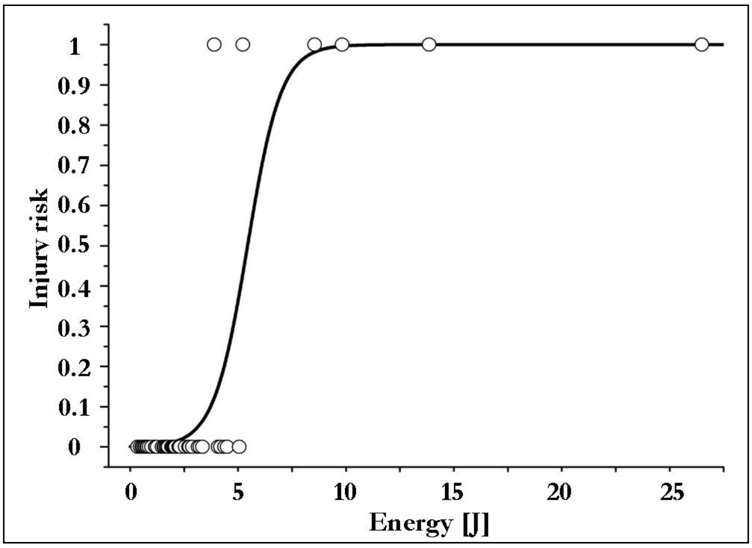


Figure 10c. *Injury risk curves to SDH or SAH defined with the ULP FEM of the head: Global strain energy in the CSF layer reaching 5.4 J for a 50% risk (EB= - 0.09)*

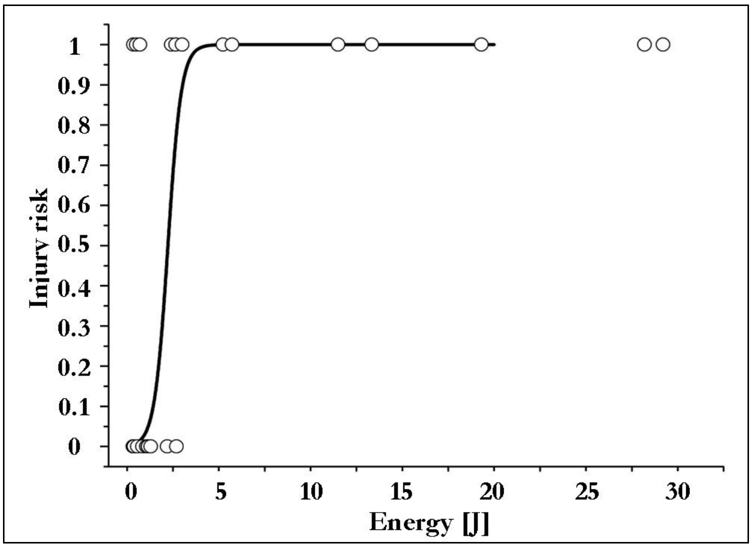


Figure 10d. *Injury risk curves to skull fracture defined with the ULP FEM of the head: Global strain energy in the skull reaching 2.2 J for a 50% risk (EB= - 0.15)*

5. Discussion

It is the first time that extensive real world accident reconstruction, based on living head data is used in order to derive tolerance limits to specific injury mechanisms according to the authors' knowledge. The tolerance limits to brain neurological moderate or severe injuries with a 50% risk are established at 18 kPa and 38 kPa brain Von Mises stress respectively. These values can be compared to previous reported attempts such as11 kPa by (Zhou *et al*., 1996) who reconstructed a car accident using the FEM of the head from (Ruan *et al*., 1993) or 15 kPa proposed by (Kang *et al*., 1997) who reconstructed a motorcycle accident using an initial version of the ULP FEM of the head and finally 27 kPa suggested by (Anderson, 2000) 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. Besides, (Bandak *et al*., 2001) suggested head angular acceleration as a good criterion for this kind of injury.

The tolerance limit to SDH or SAH established in our study rises to 5.5 J of global strain energy of the subarachnoidal space with a 50% risk of occurrence. In previous studies this injury is evaluated by parasagittal bridging veins elongation or elongation rate computed with the finite element modelling as suggested by (Bandak *et al*., 2001).

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, 1994) as well as in terms of strain energy (around 2 J) by (Gurdjian *et al*., 1958). In the present study a tolerance limit to skull fracture is established numerically at 2.2 J for the global strain energy of the skull 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 3 560 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.

6. Conclusion

The present paper presents an original numerical head model (the ULP FEM of the head) that is validated through modal and temporal analysis. An improved FEM of the head is also presented with two main originalities which are skull fracture modelling and the simulation of the interface between the brain and the skull by interaction between fluid and structure. Results show that it was possible to reconstruct the head kinematics of (Trosseille *et al*., 1992) experiments and to predict the brain pressure accurately at sites near to the impact location. However, the brain pressure predictions became less accurate as the distance from the impact location increased, especially for long duration impacts. The skull stiffness and fracture force were very accurately predicted when compared with values measured by (Yoganandan *et al*., 1994). This specifies the model's application domain to very short (2 ms) and very long (15 ms) impact duration.

The ULP FEM of the head is then used to simulate 64 real world accidents. For the protected head impacts, the head kinematics were obtained by experimental accident replication. For the non protected pedestrian head impacts, initial head velocity and position relatively to the windscreen were defined by pedestrian kinematics simulation. These data were transferred to Strasbourg University for the numerical accident simulation. The outputs from the ULP FEM of the head were compared with the head injuries recorded for each case. When head injury was examined the following five distinct groups emerged: uninjured, brain moderate neurological lesions, brain severe neurological lesions, SDH or SAH and skull fractures. 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 FEM of the head, following limits were drawn: (i) a brain Von Mises stress reaching 18 kPa for a 50% risk of brain moderate neurological lesions ; (ii) a brain Von Mises stress reaching 38 kPa for a 50% risk of brain severe neurological lesions ; (iii) a global strain energy in the CSF layer of 5.4 J for a 50% risk of SDH or SAH ; (iv) a global strain energy in the skull of 2.2 J for a 50% risk of a skull fracture .

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 or HIC values. 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.

7. References

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