Head Injury Prediction Tool for Protective Systems Optimisation

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Abstract: The objective of the present study is to synthesize and investigate using the same set of sixty-one real world accidents the human head injury prediction capability of the HIC and the injury mechanisms related criteria provided by the Louis Pasteur University (ULP) finite element head model. Each accident has been classified according to whether neurological injuries, subdural haematoma and skull fractures were reported. Furthermore, the accidents were reconstructed experimentally or numerically in order to provide loading conditions such as acceleration fields of the head or initial head impact conditions. Finally, thanks to this rather large statistical population of head trauma cases, injury risk curves were computed and the corresponding regression quality estimators permitted to check the correlation of the injury criteria with the injury occurrences. As different kinds of accidents were used, i.e. footballer, motorcyclist and pedestrian cases, the case-independency could also be checked. As a result FE head modelling provide essential information on the intracranial mechanical behaviour and, therefore, better injury criteria can be computed, especially for neurological injuries. Illustrations of how this new head injury prediction tool can participate to the head protection system optimisation is also provided.

1 INTRODUCTION

The head and more specifically the brain is among the most vital organs of the human body. From a mechanical point of view, the biological evolution of the head has lead to a number of integrated protection devices. The scalp and the skull but also to a certain extent the pressurized sub arachnoidal space and the dura matter are natural protections for the brain. However, these are not adapted to the dynamical loading conditions involved in modern accidents such as road and sport accidents. The consequences of these extreme loadings are often moderate to severe injuries. Preventing these head injuries is therefore a high priority.

Over the past forty years, a slant has been put by the biomechanical research on the understanding of the head injury mechanisms. One of the main difficulties of this research field is that a functional deficiency is not necessarily directly linked to a damaged tissue. Nevertheless, an injury is always a consequence of an exceeded tissue tolerance to a specific loading. Even if local tissue tolerance has very early been investigated, the global acceleration of the impacted head and the impact duration are usually being used as impact severity descriptors. The Wayne State University Tolerance Curve has therefore been proposed since the early Sixties thanks to several works by Lissner et al. (1960) [1] and Gurdjian et al. (1958, 1961) [2]. This curve shows the link between the impact of the head described by the head acceleration and the impact duration and, on the other hand the head injury risk. Hence, after the work of Gadd (1966) [3], the National Highway Traffic Safety Administration (NHTSA) proposed the Head Injury Criterion (HIC) in 1972. This is the tool used nowadays in safety standards for the head protection systems using headforms. Since it is based solely on the global linear resultant acceleration of a one mass head model, some limitations of this empiric criterion are well-known, such as the fact that it is not specific to direction of impact and that it neglects the angular accelerations. This is why Newman proposed the GAMBIT [4] and more recently the Head Impact Power (HIP) in the end of the nineties [5]. A methodology was described to assess brain injuries, based on multiple accident reconstructions of American football players' head collisions during recorded games.

However, in the computation of the HIC and the HIP criteria, the head is modelled as a rigid mass without any deformation. Since the finite element method exists and due to the improvement of computing capacities, the deformation of the skull and the internal components can now be simulated.

This method thereby leads to added useful mechanical observables which should be closer to the description of known injury mechanisms. Hence, new injury criteria can be proposed. In the last decades, more than ten different three dimensional finite element head models (FEHM) have been reported in the literature by Ward et al. (1980) [6], Shugar et al. (1977) [7], Hosey et al. (1980) [8], Di Masi et al. (1991) [9], Mendis et al. (1992) [10], Ruan et al. (1991) [11], Bandak et al. (1994) [12], Zhou et al. (1995) [13], Al-Bsharat et al. (1999) [14], Willinger et al. (1999) [15] and Zhang et al. (2001) [16]. Fully documented head impact cases can be simulated in order to compute the mechanical loadings sustained by the head tissues and to compare it to the real injuries described in the medical reports. It has for example been shown in Zhou et al. (1996) [17], Kang et al. (1997) [18] and more recently in King et al. (2003) [19] that the brain shear stress and strain rates predicted by their FEHM agree approximately with the location and the severity of the axonal injuries described in the medical report.

Since these finite element head models exist, new injury prediction tools based on the computed intracranial loadings should become available. The FEHM developed at the Wayne State University for instance has been used in Zhou et al. (1995) [13] to propose such tools. In the same way, thirteen motorcyclist accidents have been reconstructed by Willinger et al. (2001) [20] at Strasbourg Louis Pasteur University (ULP) using the FEHM presented in Willinger et al. (1999) [15] and described in the "Data sources" section of the present paper. This study established that the computed brain pressure was not correlated with the occurrence of brain haemorrhages, whereas brain Von Mises stress was. In order to undertake a statistical approach to injury mechanisms, more accident cases including footballers, motorcyclists and pedestrians were introduced in Willinger et al. (2003) [21] and a first attempt of injury criteria to specific mechanisms was proposed. Another FEHM presented in Takhounts et al (2003) [22] is very suitable for this kind of study due to the very short computing duration: the Simulated Injury Monitor or SIMon. A number of scaled animal model loading conditions lead the authors to propose as well injury mechanisms and related injury criteria based on animal experiments

In this context, the objective of the present study is therefore to synthesize and to investigate on a same set of real world accidents the injury prediction capability of the HIC criterion as well as the injury mechanisms related criteria provided by the ULP FEHM in order to illustrate of how this new head injury prediction tool (ULP model) can participate to the head protection system optimisation.

2 METHODOLOGIE

A database of sixty-one real world accident cases is used in the present study in order to compare the injury prediction capability of the HIC and the ULP FEHM derived criteria. These footballer, motorcyclist and pedestrian accidents are described in the "Data sources" section hereafter. Each accident case is classified according to its medical report as follows:

- cases with neurological injuries are the cases where a concussion, unconsciousness, a coma or diffuse axonal injuries have been reported. Such injuries are wholly of brain origin and they stem especially from the neurological system of the brain matter rather than the vascular. For practical convenience, they are called moderate neurological injuries when the unconsciousness last less than twenty-four hours and severe neurological injuries when lasting more than twenty-four hours.
- cases with subdural haematoma (SDH) when vascular injuries with bleeding are observed between the brain and the skull.
- cases with skull fracture which can be linear or depressive. Among these cases, there is not any case where the only reported skull fracture is a basilar facture.

No special classification is used for subjects with injuries in more than one of the three categories.

Moreover, each accident provides loading condition of the head. These loading conditions can be described in terms of linear and angular acceleration curves of the head center of gravity or in terms of relative position and velocity between the head and the impacted surface at the time just prior to the impact. Although the ULP FEHM can be driven for both kinds of loading conditions, the HIC can only be computed using 3D acceleration fields. These accelerations are obtained from experimental or numerical accident replications using a Hybrid III dummy head. Since experimental replications had already been achieved for the footballer and motorcyclist accident cases, the 3D acceleration fields were already available. Thus, numerical accident replications using finite element models of the Hybrid III head and the windscreen were only necessary for the pedestrian cases. Finally, all the sixty-one cases could be considered for HIC computation and could be simulated with the ULP FEHM. This methodology synthesized in Figure 1 allows the computation of the HIC and ULP injury criteria for the whole set of accident data.

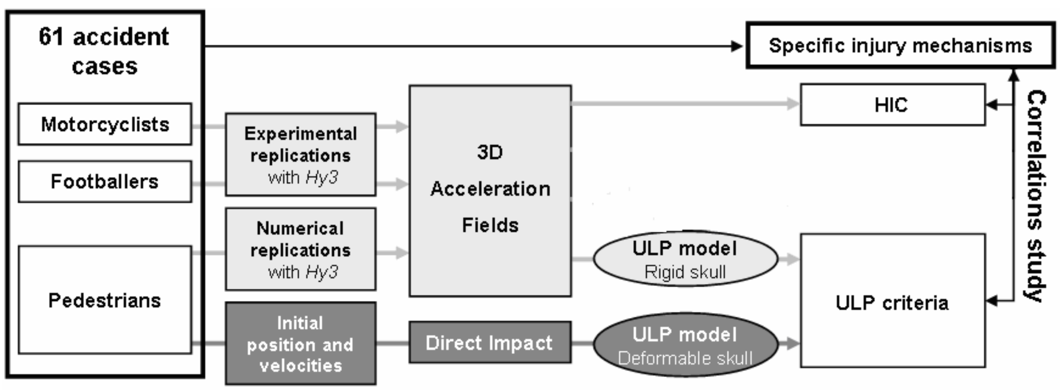

Figure 1 - Methodology permitting the computation of HIC and ULP criteria for all the 61 real world head trauma cases.

As a last step and in order to evaluate the injury prediction capability of the different criteria, an injury mechanism related approach was adapted. For each kind of injury, the correlation between the injury parameter values and the injury occurrences was reported and illustrated through histograms. Injury risk curves could then be computed for each injury mechanism following the method described in Nakahira et al. (2000) [23].

3 DATA SOURCES

This section describes the real world accidents used in the present study and details how the initial conditions are handled to drive the head model in order to compute the related injury criteria. Furthermore, head model and details on criteria computation are also synthesized in the present section.

4 THE REAL WORLD ACCIDENTS USED IN THIS STUDY

Twelve motorcyclist accidents, twenty-two footballer accidents and twenty-seven pedestrian accidents have been used in this study. The injuries sustained by the victims are summarized in Table 1.

- The motorcyclist accidents are those described in Chinn et al. (1999) [24]. They were experimentally reconstructed in collaboration between ULP, the Transport Research Laboratory (TRL) and the Glasgow Southern General Hospital. The helmet of the victim was collected on the accident scene. The acceleration field sustained by the head during the impact was then inferred experimentally by using an instrumented Hybrid III dummy head which was fitted inside a new helmet similar to the one worn by the victim. Head and helmet were thrown at different velocities against different kinds of anvils in order to reproduce on the new helmet the same damages as those observed on the victim's helmet. In this study, the motorcyclist accidents are referenced with a letter "M".

- The footballer accidents are those described in [5]. In American football games, two cameras have been used in order to determine the relative position, orientation and velocities between the helmeted head of two players when colliding together. Then, the scene has been replicated experimentally thanks to two helmeted Hybrid III dummy heads. The validation of this method is based on the rebound of the full body dummies after the experimental replication compared to the filmed rebound of the football players' bodies. In this study, the footballer accidents are referenced with a letter "S".

- The pedestrian accidents are those reconstructed from the database of the Accident Research Unit of the Medical University of Hanover. These are all accidents with a main impact (i.e. the supposed injurious impact) consisting on a head hit by the middle of a windscreen. A great variety of parameters were collected on the accident scene and were used as the inputs of an analytical rigid body study in order to infer the kinematics of the pedestrian body until the impact of the head. For each case, the results of this simulation are compared to the damages observed on the car and to the wounds sustained by the victim. In order to obtain the acceleration curves undergone by the center of gravity of the head, a numerical replication using a finite element model of a windscreen and of a Hybrid III head was then performed. The windscreen model was previously described in Willinger et al (2001). It consists on three layers of composite shell elements with a mechanical behaviour based on the experimental data presented by Harward (1975) [25]. The finite element Hybrid III dummy head model was modelled with a viscoelastic skin and a rigid mass which inertias are close to those measured on a real dummy. In this study, the pedestrian accidents are referenced with a letter "P".

Table 1 - Injuries sustained by accident victims

Accident Origin	Skull fractures	SDH	Mod. Neuro. Inj.	Sev. Neuro. Inj.
Motorcyclists 12 cases	0	1	6	1
Footballers 22 cases	0	0	9	0
Pedestrians 27 cases	18	5	8	8
Total 61 cases	18	6	23	9

5 HEAD INJURY CRITERIA DESCRIPTION

5.1 HIC criterion

Proposed by the NHTSA in 1972, the head is seen as a one-mass structure. It is computed using the following formula:

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

where a [m.s-2] is the resultant linear acceleration measured at the center of gravity of the Hybrid III dummy head. t1 and t2 [ms] are chosen in order to maximize the HIC value.

5,2 ULP head model and criteria

The ULP 3D Finite Element Head Model used in this study has been developed by Kang et al. (1997) [18]. The present ULP model (Figure 2 in Appendix) includes the main anatomical features: skull, falx, tentorium, subarachnoid space, scalp, cerebrum, cerebellum, brain stem. Falx and tentorium have a layer of shell elements, skull is simulated by three layered composite shell and the others were constituted by brick elements. The finite element mesh is continuous and represents an adult human head. The subarachnoid space was modelled between the brain and the skull to simulate the cerebral-spinal fluid. This space is constituted by a layer of brick elements and surrounds entirely the brain. The tentorium separates the cerebrum and cerebellum and the falx separates two hemispheres. A layer of brick element simulating the cerebral-spinal fluid surrounds theses membranes. The scalp was modelled by a layer of brick elements and surrounds the skull and facial bone. Globally, the present human head model consists of 13208 elements. Its total mass is 4.5 kg.

Table 2 – Mechanical properties and element characteristics of the ULP human head model

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

Material properties assigned to the different parts are all isotropic, homogenous and elastic. The Young's modulus of the subarachnoid space was found by Willinger et al. (1995) [26] by modal analysis. The viscoelastic properties assigned to the brain were scaled from Khalil et al (1977) [27]. The behaviour in shear was defined by:

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

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

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. In order to model the material discontinuity in the case of fracture, it was necessary to use values for the limiting (ultimate) tensile and compressive stress obtained from Piekarski (1970) [28] and integrated in the Tsaï-Wu criterion. All mechanical properties and element characteristics of the ULP human head model are summarized in table 2.

A total of eight instrumented cadaver impacts were reconstructed with the objective of validating the ULP model under very different impact conditions. Currently head FE models are validated against Nahum's et al. impact (1977) [29] and have moreover been validated against other experimental data as those of Trosseille et al. (1992) [30] for high damped long impact durations, and those of Yoganandan (1994) [31] for very short impact durations including bone fracture. The ULP FEHM can be both driven by acceleration fields applied to a skull supposed to be rigid (motorcyclist and footballer cases) or throw a direct impact with a deformable skull and using the windscreen finite element model (pedestrian cases).

As described in Willinger et al. (2001) [20], three injury criteria are computed with this model:

- The maximal Von Mises stress value reached by a significant volume of at least ten contiguous elements from the brain is proposed as a correlate to neurological injury occurrences.

- The maximum value reached by the global strain energy of the subarachnoidal space is proposed as a correlate to subdural haematoma occurrences.

- The maximum value reached by the global strain energy of the deformable skull is proposed as a correlate to skull fracture occurrences. This criterion is only computed for the pedestrian cases where the deformable skull FEHM is driven with a direct impact.

The different previously described injury criteria which are candidates for each injury mechanism are summarized in the Table 3.

The SDH and neurological injuries prediction capability of these criteria is assessed using the whole set of accidents.

The skull fracture prediction capability is assessed using only the pedestrian cases. In these cases, HIC is computed with 3D acceleration fields obtained from the previously described numerical reconstructions whereas the internal deformation energy of the ULP FEHM deformable skull is computed throw direct impact simulation.

Table 3 – Proposed candidate criteria for each kind of injury

	Skull fracture	**SDH**	**Neurological injuries**
Linear accelerations	HIC	HIC	HIC
Intracranial field parameters computed with the ULP FEHM	Internal deformation energy of the skull	Internal deformation energy of the CSF space	Intra-cerebral Von Mises stress peak

6 RESULTS

The determination of the head injury risk curves for specific injury mechanisms is based on a correlation study between the values of the proposed candidate criteria and the injury occurrences. A histogram is built for each specific injury and the value taken by a given criterion for each case is plotted. These accident cases are sorted according to the injury classification as explained in the methodology section, i.e. moderate and severe neurological injuries, SDH and skull fractures. When the injury predictor candidate is adequate, a clear distinction is visible between the low values of the uninjured cases and the high values of the injured cases and a threshold can thereby be determined.

This threshold can accurately be calculated since it is the value leading to a 50% risk of an injury risk curve. In this work, the Modified Maximum Likelihood Method is chosen. It is a logistic regression method developed and described by Nakahira et al. (2000) [23] which shows better results than the classical Maximum Likelihood Method and the method described by Mertz et al. (1982) [32]. On the obtained curves the circles represent the victims with mention to their injury statement (uninjured = 0 and injured = 1) in y-coordinate and to their considered injury predictor candidate value in x-coordinate. The injury risk curve is a sigmoid with the following formula:

$$P(x) = \frac{1}{1+e^{-(a+bx)}}$$

where P is the probability of injury for the given value x of the injury predictor candidate. The a and b parameters are determined using maximum likelihood method to maximize the function's fit to the data. The estimator of the goodness of fit has been called *EB* by Nakahira *et al.* and is defined as equal to the log likelihood:

$$EB = \frac{1}{n} Log\left\{\prod_i P(x_i) \times \prod_j \left(1 - P(x_j)\right)\right\}$$

Where n is the total number of accident cases, x_i are the predictors of the injured cases and x_j the predictors of the uninjured cases. In addition, another estimator called *EA* by Nakahira *et al.* evaluates the assumption "the injury probability approaches zero when the injury related parameter approaches zero". A *EA* = 5% level has been used as proposed by Nakahira *et al.*

The quality of the regression is thereby given by the negative estimator *EB* which should be as close to zero as possible. Thus, the 95% confidence limits of the each injury risk curve has been calculated and plotted. It notably gives the 95% confidence intervals of the deducted thresholds for risks of 5%, 50% and 95%. These thresholds are indicated on the figures as well as the a and b regression parameters and *EA* and *EB* corresponding estimators.

For the four injury mechanisms and the two injury criteria results are reported as follows:

- Figure 3 (in Appendix) shows the results for the prediction of neurological injuries, both moderate and severe.
- Figure 4 (in Appendix) shows the results for the prediction of subdural haematoma.
- Figure 5 (in Appendix) shows the results for the prediction of skull fractures.

Finally, a synthesis of the prediction capability of each injury criterion in terms of *EB* value is reported for the different injury mechanisms in Figure 6 (in Appendix) and results are synthesized in table 4.

Table 4 - Summary of the main results of the injury risk curves

Injury type	Proposed injury criterion	EB	50% risk
Skull fracture	HIC	-1.0	667
	ULP Skull IE [mJ]	**-0.6**	833 mJ
SDH/SAH	HIC	-0.9	1429
	ULP CSF IE [mJ]	-0.9	4211 mJ
Moderate neurological injury	HIC	-1.3	533
	ULP VM [kPa]	**-0.6**	27kPa
Severe neurological injury	HIC	-0.8	1032
	ULP VM [kPa]	**-0.5**	39kPa

7 DISCUSSION

The logistic regression analysis has been made on a rather relevant statistical population of sixty-one accident cases when considering neurological injuries or SDH and of twenty-seven accident cases when considering skull fractures. The estimator EB of the logistic regression takes the quality of the statistical populations into account as well as the correlation between the proposed injury metric and the injury occurrences. It is also important to note that there are different kinds of accidents so that the injury mechanisms should not be case-dependants.

In order to ripen the injury thresholds inferred by the logistic regression, an important alternative point was to select as much non-extreme accidents as possible, i.e. neither too mild nor too violent. This selection should nevertheless explain the overlap in the histograms between some non-injured cases whose considered injury mechanism value is high and some injured-cases for which this value is low though.

Concerning the quality of the statistical population, a comment should be made about the proportion between injured and non-injured cases. Although this proportion is acceptable for neurological injuries and for skull fractures, it may be more arguable concerning the SDH since there are very few injured cases. However, this disproportion should explain the law quality of the regression as indicated by the EB regression quality estimator.

Since the injury criteria have been computed on the same set of accident cases, the comparison of their injury prediction capability is thereby possible. In terms of EB regression quality estimator as reported in table 4, the ULP FEHM based criteria seem to have the best prediction capability for each type of injury. This is particularly true concerning the neurological injuries since the injury criterion based on the peaks of Von Mises stress keeps its accuracy even when predicting the moderate neurological injuries. An injury mechanism based on the computed intracranial mechanical behaviour of the brain was obviously the main motivation for building a finite element model of the human head.

While the injury prediction capability is assessed using the EB estimator, the accuracy of the injury thresholds inferred by the regression analysis can be evaluated with confidence limits curves. In this study, like in most biomechanical studies, the number of data is limited. Thus, the data are usually censored since they are biased in one direction or another. The sign of the bias is known but not the magnitude. This explains the quite important width of the 95% confidence limits plotted on the figures. However, the slopes underlying risk function are steep and according to Di Domenico et al. (2003) [33], the steeper these slopes, the smaller the sample size that is needed to obtain "good" risk estimation are. Given the censored nature of the data, Consistence Threshold (CT) methods may be used in a future work as presented in Kent et al. (2004) [34] for instance.

Another limitation of this study is the hypothesis that there is no correlation between the different categories of injuries. For instance, the energy absorbed by a skull fracture could allow decreasing the loading of the brain and therefore prevent from neurological injuries. This is taken into account by the ULP FEHM with a deformable skull (pedestrian cases) but not by the HIC criterion. The loading of the brain might thereby be over-evaluated in cases with fractures and the resulting tolerance limit relative to brain injuries could be affected. Besides, skull fracture is often accompanied by extra-dural haematoma, but there is not any case with this kind of injury in the data-base used in this study. Tolerance limits of a second impact might also be affected after a first impact. This is not taken into account by any injury criterion and it is obviously a strong limitation.

The overall main limitation for such a study is the reliability of the replication of the accidents which are used. The authors must trust the reconstructions which have been made by specialists. The footballer cases are well known and have been used and discussed in several studies such as the one by Newman et al. (2000). The motorcyclist cases have been made by the TRL using reliable experimental techniques. The TRL evaluates the uncertainty on the acceleration field to about 10%. Finally, an uncertainty of about 20% on the resulting initial velocities is proposed by the Accident Research Unit of the Medical University of Hanover for the pedestrian cases.

8 CONCLUSION AND PERSPECTIVES

Sixty-one real world accident cases have been reconstructed in order to provide head acceleration fields and head initial impact conditions so that the HIC and the ULP criteria could be computed. New tolerance limits to specific injury mechanisms were deduced for the ULP head FE model and the relevance of their capability to predict injuries could therefore be investigated comparatively with HIC criterion, using histograms and injury risk curves. The advantage of this methodology is that this injury prediction capability is not deduced from ex-vivo or animal experiments but on real-world accidents. The main result of this study is the good capability in predicting skull failure and moderate and severe neurological injuries of criteria based on a finite element head model, the ULP human head model. This was expected since a single-mass model used by criteria such as the HIC is not able to correctly model the intracranial mechanical behaviour.

Although the quality and the accuracy of the accident replications and reconstructions are obviously arguable, the relevance of this study should be found in the high number of considered accidents. This statistical approach should decrease the consequences of possible errors. However the statistical population of cases with subdural haematoma must imperatively be consolidated.

Finally, after a demonstration that the HIC is not a good injury criterion, we can say that this new head injury predictive tool, which is ULP model, can participate to the head protection system evaluation and optimization (Figure 7 in Appendix).

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APPENDIX:

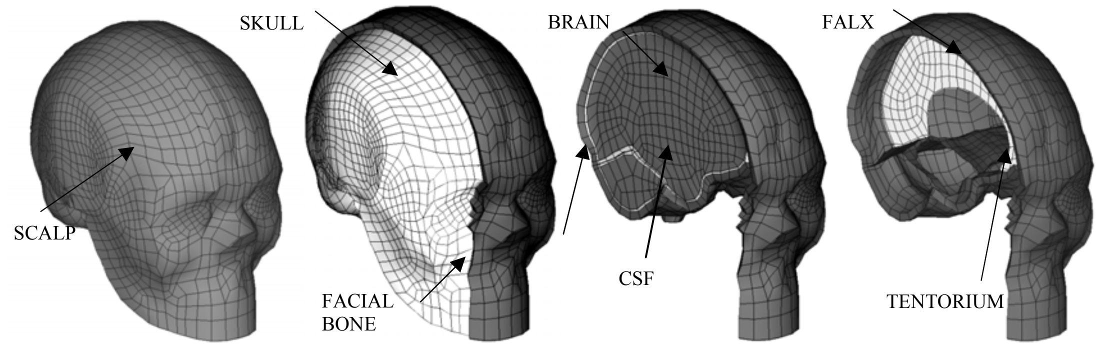

Figure 2 - Description of the finite element model of the human head (ULP model).

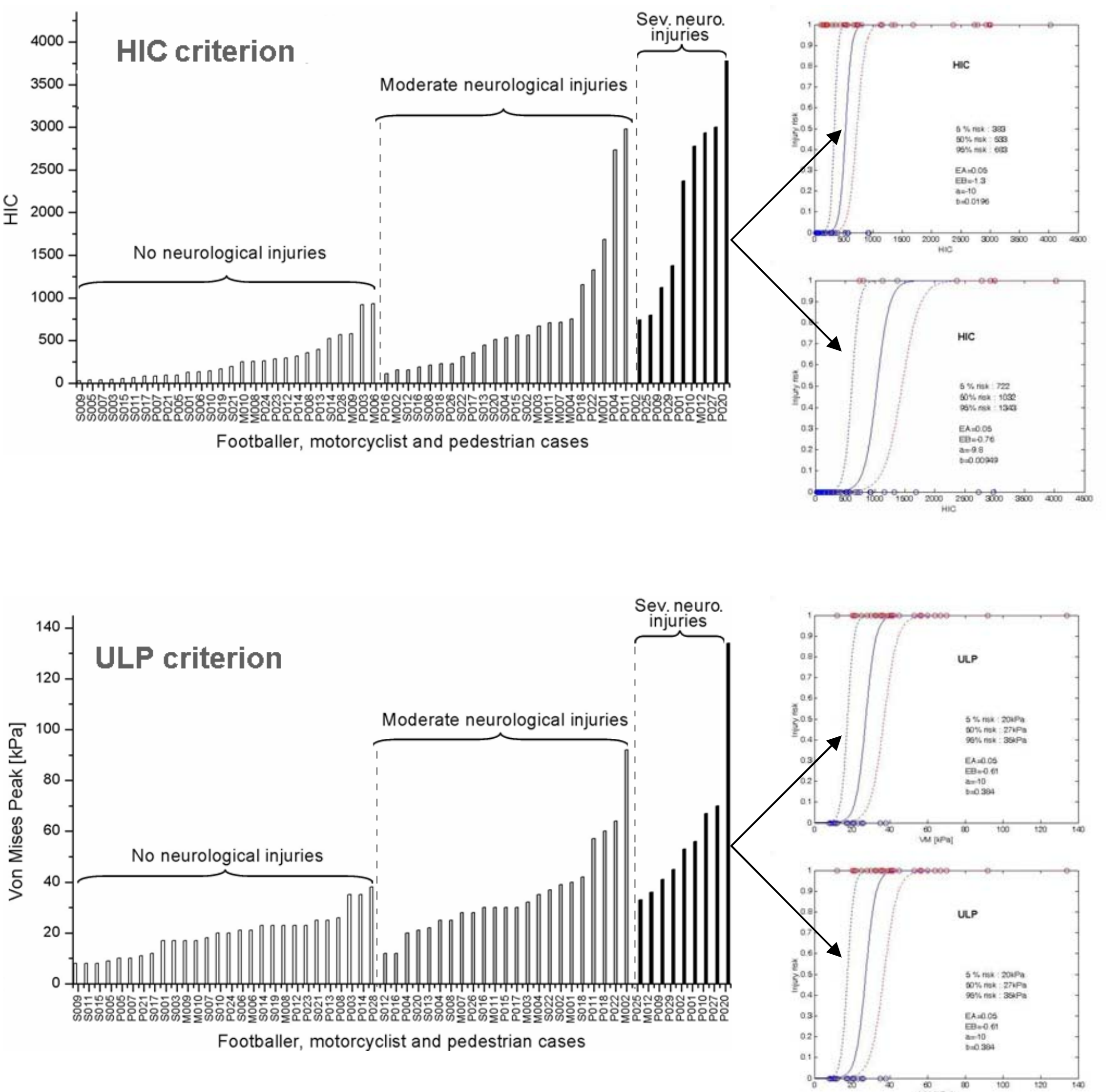

Figure 3 – Histograms and injury risk curves for the HIC and ULP injury criteria for moderate and severe neurological lesions.

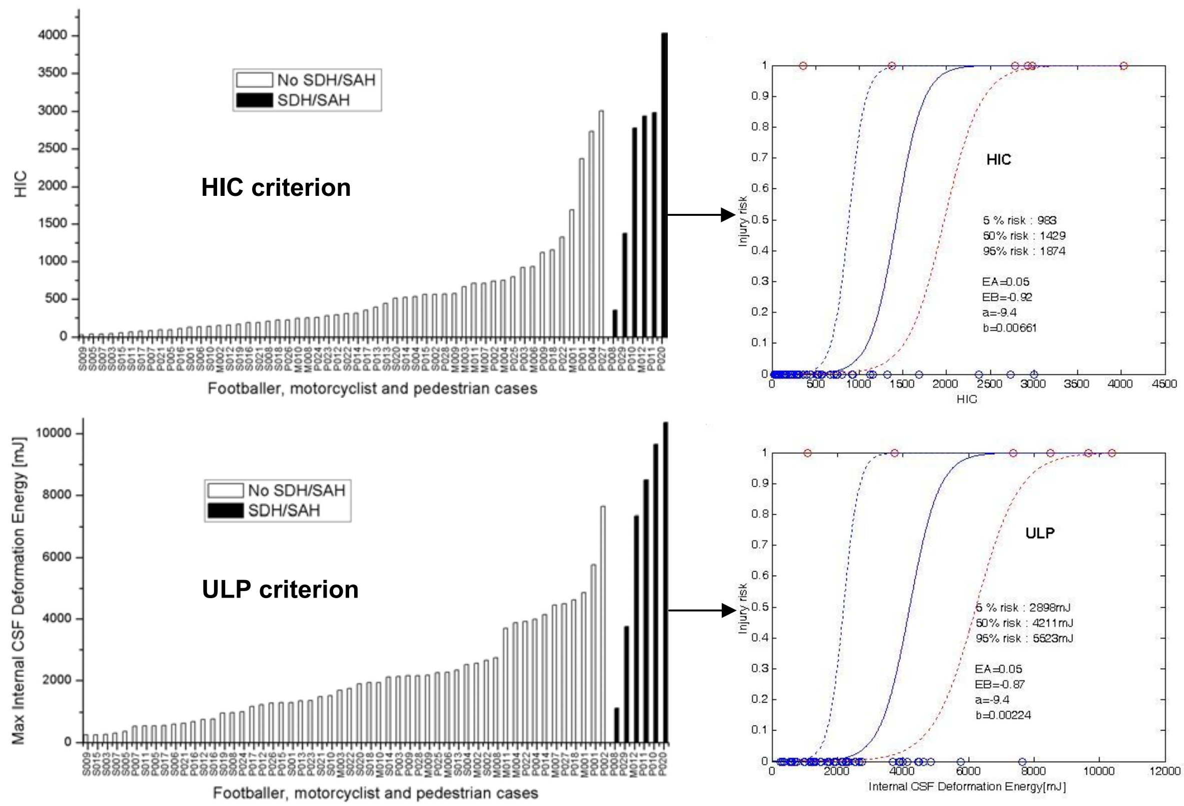

Figure 4 - Histograms and injury risk curves for the HIC and ULP injury criteria for subdural haematoma.

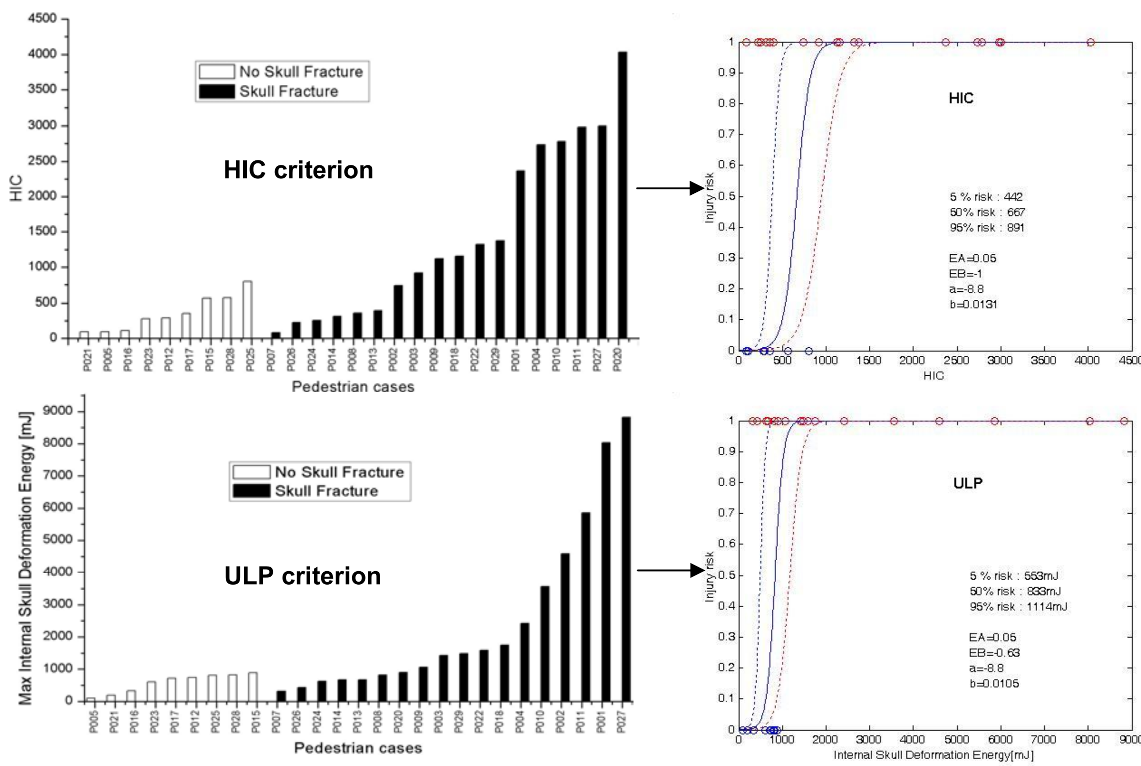

Figure 5 - Histograms and injury risk curves for the HIC and ULP injury criteria for skull fracture

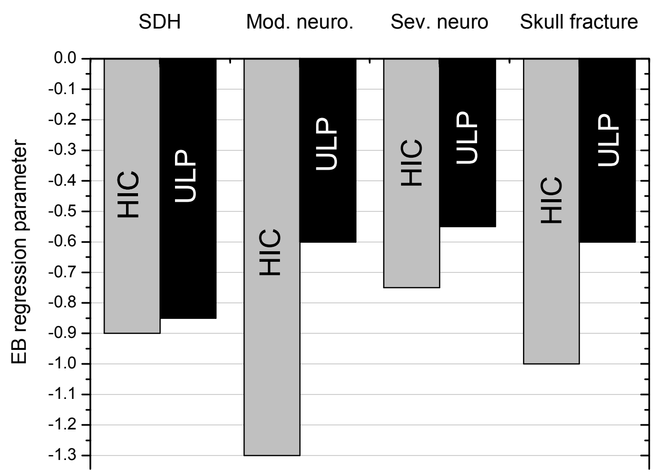

Figure 6 - EB regression quality estimator for each injury type with the associated injury criteria. The closest to zero this negative parameter is, the best is the quality of the regression

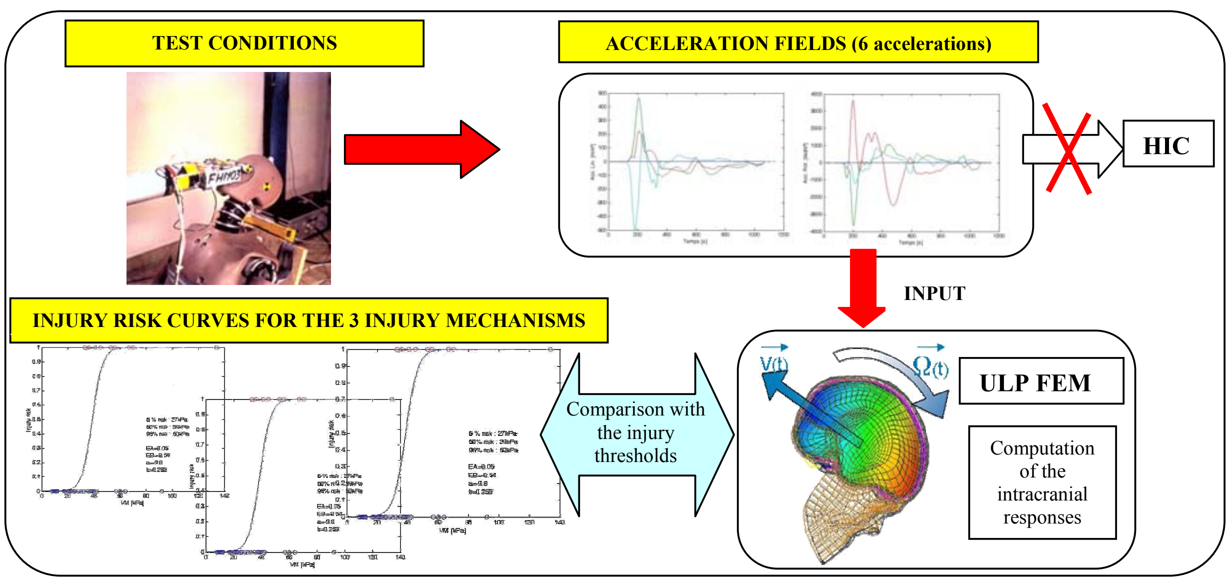

Figure 7 – Head injury predictive tool: Methodology