COUPLING OF STRASBOURG UNIVERSITY HEAD MODEL TO THUMS HUMAN BODY FE MODEL: VALIDATION AND APPLICATION TO AUTOMOTIVE SAFETY

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ABSTRACT

Human body segments and whole body models are more and more used in automotive safety research. Detailed in deep validated segmental models exist and are used for the definition of improved injury criteria, transforming the models into injury prediction tools. The present collaborative work's objective is to couple and to validate the Strasbourg University Head FE Model (SUFEHM) with the THUMS human body model under Ls-Dyna code before applying the new tool under accidental environment.

In a first effort, Strasbourg University Head Model and related injury criteria developed in earlier studies under RADIOSS code had to be transferred under Ls-Dyna code, both at constitutive laws and injury criteria definition level. For this, a validation of the SUFEHM against Nahum and Yoganandan's experiments in order to validate brain and skull behavior respectively under Ls-Dyna has been done. After these validations the reconstruction of 59 real world head trauma has been conducted in order to propose head tolerance limits to specific injuries under Ls-Dyna code.

After this, the SUFEHM was coupled to the THUMS neck in order to create a hybrid "THUMS-Strasbourg head" model. At geometrical level the coupling was performed by creating interfaces at bone contact level and connecting ligaments and soft tissue elements to the head model. At mechanical level the coupled FEMs was validated under front, lateral and oblique impact regarding head-neck kinematics superimposed to experimental data.

This coupled model constitutes an original research tool for further investigation on the importance of human head boundary condition in case of head impact, whatever the accident condition are, car occupant, pedestrian or even motorcyclists.

INTRODUCTION

Since several years human body models became a useful tool for the simulation of the human body under impact conditions. Human body models are directly modelled like the human body. In comparison the common dummy simulation models are modelled like the anthropomorphic test device which they are representing.

Generally the results which can be obtained by a simulation model of the dummy are limited to the measurements which are delivered by the anthropomorphic test device.

The results which human body simulations deliver have of course also limitations. But compared to an anthropomorphic test device human body models are able to represent the kinematics of the human body in a more realistic way. A human body model can be loaded from different directions. For instance the same model is able to simulate the frontal and the side impact loading of an occupant. In fact the same model should be able to simulate a pedestrian accident situation if positioned in a standing posture.

All over the world several human body models are available. Additionally detailed models of body regions are also available. Body region models can be used for special impacts like a leg model impacting a front end model of a car. Regarding the body regions under the aspects of crash and impact the head is one of the most vulnerable body regions despite of having a strong bone structure. Therefore models of the head have been developed to simulate the impact and the possible injury caused by these impacts.

A typical human body model for the whole human body is the THUMS model used with Ls-Dyna code [Iwamoto2002, Oshita 2002]. Due to the fact, that the basic version of the THUMS model has a rigid skull modelling, the head impact of this whole body human model cannot be evaluated for possible head or brain injuries.

A typical state of the art head model is the Strasbourg University Head FE Model (SUFEHM), which is able to predict head injuries caused by impacts. This model is available under Radioss code. The aim of this study is first to transfer SUFEHM under Ls-Dyna code, validate it and propose some tolerance limits to specific injuries by reconstructing real world head trauma under Ls-dyna code, second to couple SUFEHM with the THUMS and validate the head-neck kinematics.

STRASBOURG UNIVERSITY FINITE ELEMENT HEAD MODEL UNDER L-DYNA CODE

Meshing presentation

Kang et al., in 1997, has developed the Strasbourg University Finite Element Head Model. The geometry of inner and outer surfaces of the skull was digitised from a human adult male skull. The main anatomical features modelled were the skull, falx, tentorium, subarachnoid space, scalp, cerebrum, cerebellum, and the brainstem. The finite element mesh is continuous and represents an adult human head. Globally, SUFEHM model consists of 13208 elements. Its total mass is 4.7 kg, a representation is given in Figure 1.

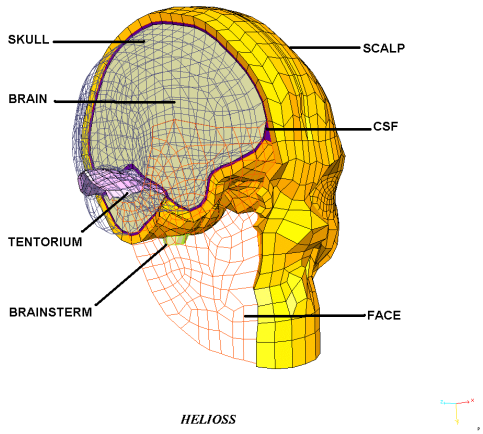


Figure 1. Section through the Strasbourg University Finite Element Head Model (SUFEHM).

Mechanical properties under Ls-dyna software

Introduction - The source model is available under Radioss code; the aim here is to implement mechanical properties under Ls-Dyna code before SUFEHM's validation. Material properties of the cerebral spinal fluid, scalp, facial bones, tentorium and falx are all isotropic, homogenous and elastic, with mechanical properties similar than those used under Radioss code (*MAT_ELASTIC law) (Willinger et al., 1995).

Brain material law choice – The brain is assumed to be visco-elastic. The visco-elastic law used under Ls-Dyna code is Material Type 6 (MAT_VISCOELASTIC). This model allows the modeling of visco-elastic behavior for beams, shells and solids. The shear relaxation behavior is described by:

$$G(t) = G_\infty + (G_0 - G_\infty)Exp(-\beta t)$$

With G_0 short-time shear modulus, G_∞ Long-time shear modulus and β Decay constant. Values of the parameters are the same than for Radioss code i.e. G_0=4.9E^{-02} MPa, G_∞=1.62E^{-02} MPa and β=145s^{-1}.

Skull material law choice - The skull was modelled by a three layered composite shell representing the inner table, the diplöe and the external table of human cranial bone. For this an INTEGRATION_SHELL card has been implemented in order to define the three skull layers (cortical bone and diploe) as layers' thicknesses. The material model 55, which is available under a single label "*mat_enhanced composite_damage*", in LS-DYNA was used to represent the mechanical behavior of the skull bones The material model 55 has three failure criteria expressions for four different types of in-plane damage mechanisms. Each of them predicts failure of one or more plies in a laminate. The expressions accommodate four in-plane failure modes: matrix cracking, matrix compression, fiber–matrix shearing and fiber breakage. Skull mechanical parameters are presented in Table 1.

Table 1. Skull mechanical parameters under Ls-Dyna code

	Cortical bone	*Diploe bone*
Mass density [Kg/m3]	1900	1500
Young modulus [MPa]	15000	4665
Poisson's ratio	0.21	0.05
Shear stress parameter	-0.5	-0.5
Longitudinal and transverse compressive strength [MPa]	145	24.8
Longitudinal and transverse tensile strength [MPa]	90	34.8

Validation

After Strasbourg University Head FE meshing transfer under Ls-Dyna code and after the identification of the material laws, the SUFEHM validation under this code for Nahum's impact (in order to validate brain response) and for Yoganandan's impact (in order to validate the skull behaviour and bones failure) is proposed.

Nahum's validation - The experimental data used in order to validate brain behaviour were published by Nahum et al.(1977) for a frontal blow to the head of a seated human cadaver. For this

impact configuration, a 5.6 kg rigid cylindrical impactor launched freely with an initial velocity of 6.3 ms-1 generates an interaction force and a head acceleration characterised by their peak values which are respectively 6900 N and 1900 ms-2 over a duration of 6 10-3 s. Intracranial pressures were also recorded in this test, at five well defined locations : behind the frontal bone, adjacent to the impact area, immediately posterior and superior to the coronal and squamosal suture, respectively in the parietal area, inferior to the lambdoidal suture in the occipital bone (one in each side), and at the posterior fossa in the occipital area.
Since the neck was not included in this model, a free boundary condition was used to simulate Nahum's impact. This hypothesis is based on the justification that the time duration of the impact is too short (6 ms) for the neck to influence the kinematics head response during pulse duration.
In order to reproduce the experimental impact conditions, the anatomical plane of the SUFEHM was inclined about 45°, as shown in Figure 2, like in the Nahum's experiment. For modelling a direct head impact, the model was frontally impacted by a 5.6 kg rigid cylindrical impactor (with an elastic padding, E= 13.6MPa, Poisson's ration=0.16) launched freely with an initial velocity of 6.3 m/s.

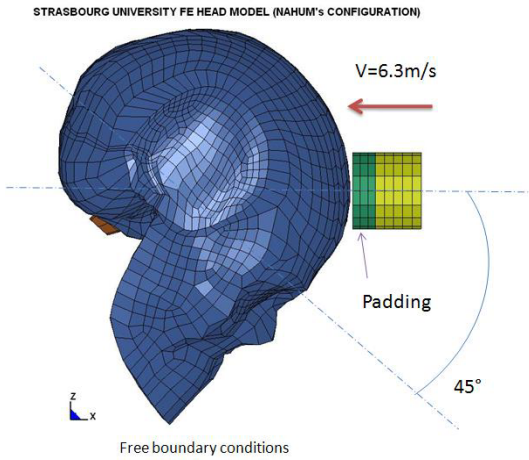


Figure 2. Nahum's configuration

<u>Yoganandan's validation</u> - Experimental tests carried out by Yoganandan et al. in 1994 has been used in order to validate the ability of the human head finite element model to predict a skull fracture. The impact configuration is shown in Figure 3. The surface of the impactor was modelled by a 96mm diameter rigid sphere. Initial conditions were similar to the experimental ones i.e. a mass of 1.213kg with an initial speed of 7.1 m/s. The base of the skull was embedded as in the experiment. For the model validation, the contact force and the deflection of the skull at the impact site, were calculated.

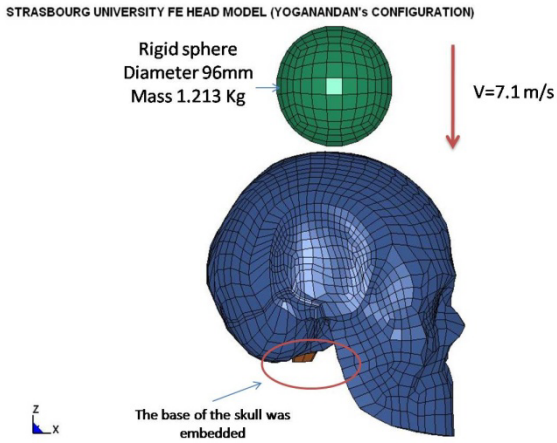


Figure 3. Yoganandan's configuration

Tolerance limits to specific injury mechanisms

SUFEHM tolerance limits to specific injury mechanisms are available under Radioss code and published by Deck et al. (2008). The objective here is to propose tolerance limits under Ls-dyna code. For this, 59 head impact conditions that occurred in motorcyclist, American football and pedestrian accidents were reconstructed with the SUFEHM under Ls-Dyna code. A summary of the type and number of accident reconstructions is given through Table 2.
The reconstructions involved applying the motion of the head from the accidents to the rigid skull of the SUFEHM. Same methodology (statistical analysis) than methodology used by Deck et al. (2008) has been undertaken.
For the statistical analysis the injuries for the accident data were categorised into the following types and levels based on the details of the medical report from each accident case:

- <u>Diffuse axonal injuries (DAI)</u>: DAI cases covered all incidences in which neurological injuries occurred and covered concussion, unconsciousness and coma. Incidences of DAI were broken down into mild and severe levels according to coma duration (<24H for moderate DAI and >24H for severe DAI)
- <u>Subdural Haematomas (SDH)</u>: This category of injuries covered all incidences in which vascular injuries with bleeding were observed between the brain and the skull of which there were six cases.

Table 2. Summary of the type and number of accident reconstructions

Accident Type	Number of cases
Motorcycle accidents	11
American football accidents	20
Pedestrian accidents	28

THUMS SUFEHM COUPLING

THUMS presentation

The Total Human Model for Safety (THUMS) was originally developed in the late nineties by the Biomechanics Laboratory from Toyota, with the help of a part of the model developed by the Wayne State University […]. The geometry of the model represents a 50th percentile American adult male body with 175 cm height and weighting 77 kg. This initial version of the model, composed of about 80,000 elements, is presented in Figure 4. Several projects performed at Daimler have led to an advanced model (165,127 elements) that is shown in Figure 5. This later model was used in the present study. Figure 6 gives an overview of the resulting global THUMS with the implemented SUFEHM (174,058 elements).

Figure 4. Basic THUMS model

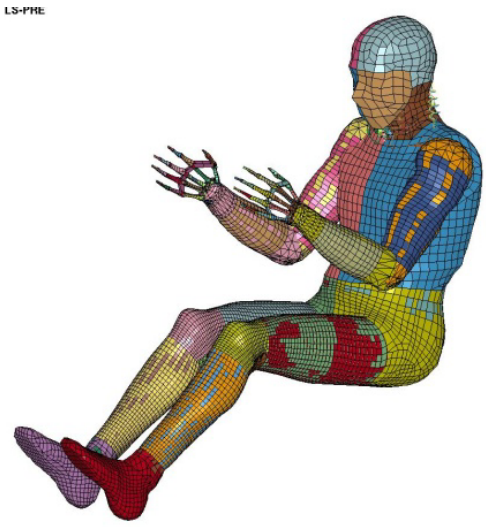


Figure 5. Advanced THUMS model

Figure 6. Advanced THUMS model with Strasbourg University Head FE Model (SUFEHM)

Coupling

The coupling of the SUFEHM head with the advanced THUMS model has been done manually by using a FE pre processor. The original head of the THUMS models has been replaced completely. The mesh of the THUMS neck has been changed in the connection area between the SUFEHM head and the THUMS neck. The connection methods are the same like in the original THUMS model. Between head and neck elements the same nodes are shared. The mesh is continuously without any tied contacts or boundary conditions. The Figure 7 shows the new generated head-neck complex.

Due to the changed head model it is necessary to verify the validation of the new generated head neck complex.

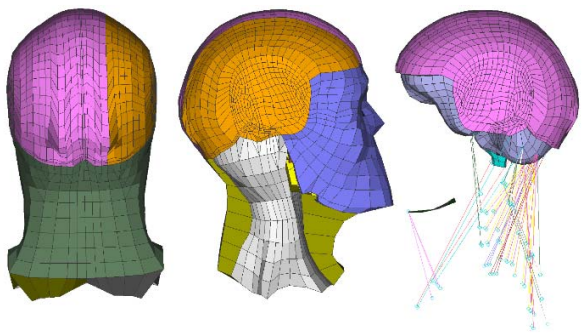


Figure 7. Head neck complex with SUFEHM head and modified THUMS neck.

Validation

After implementation of the SUFEHM on the THUMS model, a validation of the head-neck coupling was carried out according to data from studies on the human volunteer experiments conducted by Ewing et al. (1968, 1977) at the Naval Biodynamics Laboratory (NBDL).

In the experiments of Ewing et al. very healthy volunteers, from U.S. Army Air Defense Command, were put in seated position on a chair rigidly attached to a sled consisting of a horizontal accelerator that is shown in Figure 8. They were equipped with pelvic and torso restraint systems while allowing the head and neck to move freely as illustrated in Figure 9.

The subjects were then submitted to impact accelerations in frontal, lateral and oblique directions.

Two redundant systems were used to monitor the kinematics of the head-neck coupling. The first one is composed of transducers positioned over the posterior spinous process of the first thoracic vertebra (T1), over the posterior superior aspect of the head and at the mouth level as shown in Figure 10. The second system consists of high-speed cameras mounted to the sled.

Results of these tests give access to the dynamic responses (linear and angular accelerations and displacements) of the head with respect to the input accelerations, velocities and accelerations which have been recorded at T1.

Table 3 summarizes the conditions of the experiments for the different impact configurations.

Figure 8. Human volunteer positioned in the sled for a lateral impact (NBDL website)

Figure 9. Restraint systems (NBDL website)

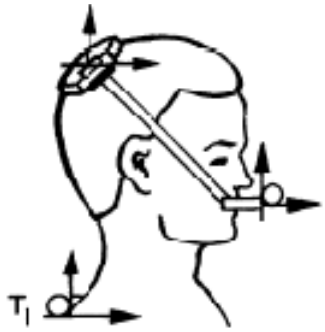


Figure 10. Anatomical locations for the transducer systems on volunteer head (Ewing et al. 1968)

Table 3. Summary of the set-up of the experiments for the three impact configurations (adapted from van der Vorst, PhD. Thesis, 2002)

Parameters	Frontal impact	Lateral Impact	Oblique Impact
Subject	Human volunteer	Human volunteer	Human volunteer
Total number of subjects	5	9	
Total number of tests	9	9	9
Average mass	68.6kg	76kg	
Average height	1.69m	1.77m	
Seat type	Rigid (90° to the floor)	Rigid (90° to the floor)	Rigid (90° to the floor)
Max sled pulse	15g	7g	>7g

In the validation procedure the THUMS model with the SUFEHM head model has to be loaded with the same accelerations like the volunteers in the tests of Ewing et al. To achieve the same loading conditions it is necessary to either model the same environment like in the test set up including the belt system or a simplification which allows validating the head-neck complex has to be done. A simplification of an existing deformable finite element model can be achieved by reducing the elements degrees of freedom for a part of the model. In this study the head-neck complex has been left deformable. The rest of the body has been defined as a rigid body. The rigid area includes the T1 bone (Figure 11).

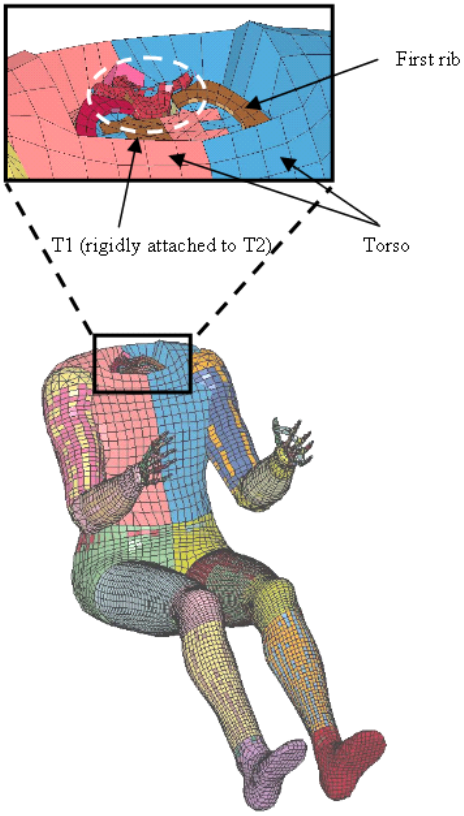


Figure 11. THUMS model rigid parts including T1 bone for validation procedure.

The velocities in the sled tests measured kinematics at the T1 bone can now be applied to the rigid part of the model. This approach saves computation time and allows validating the head-neck complex as an independent system in the first step.

From the tests of Ewing et al. average velocity curves are available for the three different loading directions. Figures 12 to 14 show the measured average speed.

For the validation of the model a change of any material property of the model has to be simulated and compared for all three loading directions (latatefrontal and oblique).

Tests with human volunteers deliver of course different results for each person despite of having similar body sizes and weights. Plotting the results in one diagram delivers usually a corridor of possible results. The aim in the validation is to achieve results for all three loading directions which are in the corridor.

The new created head-neck complex did not deliver simulation results which fit in the corridor of the test results with the original material properties of the THUMS neck model. It was necessary to change material properties for the neck in this validation procedure.

For realistic global kinematics of the head-neck complex it is necessary to define a contact between the head and the torso. Impact loadings of this severity can cause contacts of the head to the torso.

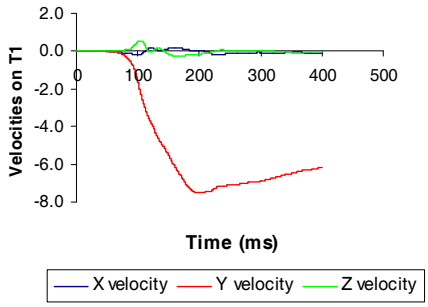


Figure 12 Velocities imposed on T1versus time for the lateral impact.

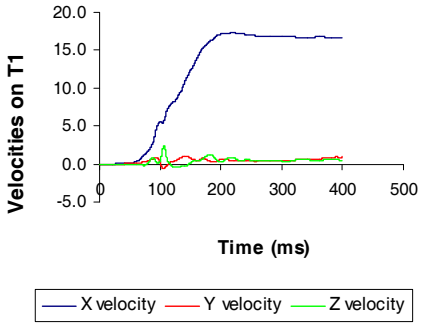


Figure 13. Velocities imposed on T1versus time for the frontal impact.

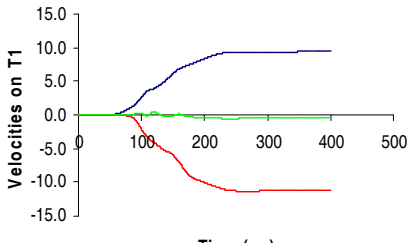


Figure 14. Velocities imposed on T1versus time for the oblique impact.

RESULTS

SUFEHM Results

Nahum's validation results – In order to validate brain mechanical properties under Ls-Dyna code, a Nahum's experiment has been numerically replicated. The comparison of numerical and experimental forces is shown in Figure 15a for the Nahum's impact. A good agreement for the impact force was found as the time duration of impact and the amplitudes were well respected. The comparison of pressure time histories between numerical and experimental data is presented in Figure 15b, c, d, e for the Nahum's impact simulation. As shown in these figures, five intracranial pressures from the model matched the experimental data very well. The maximum difference of pressure peak is under 10 %.

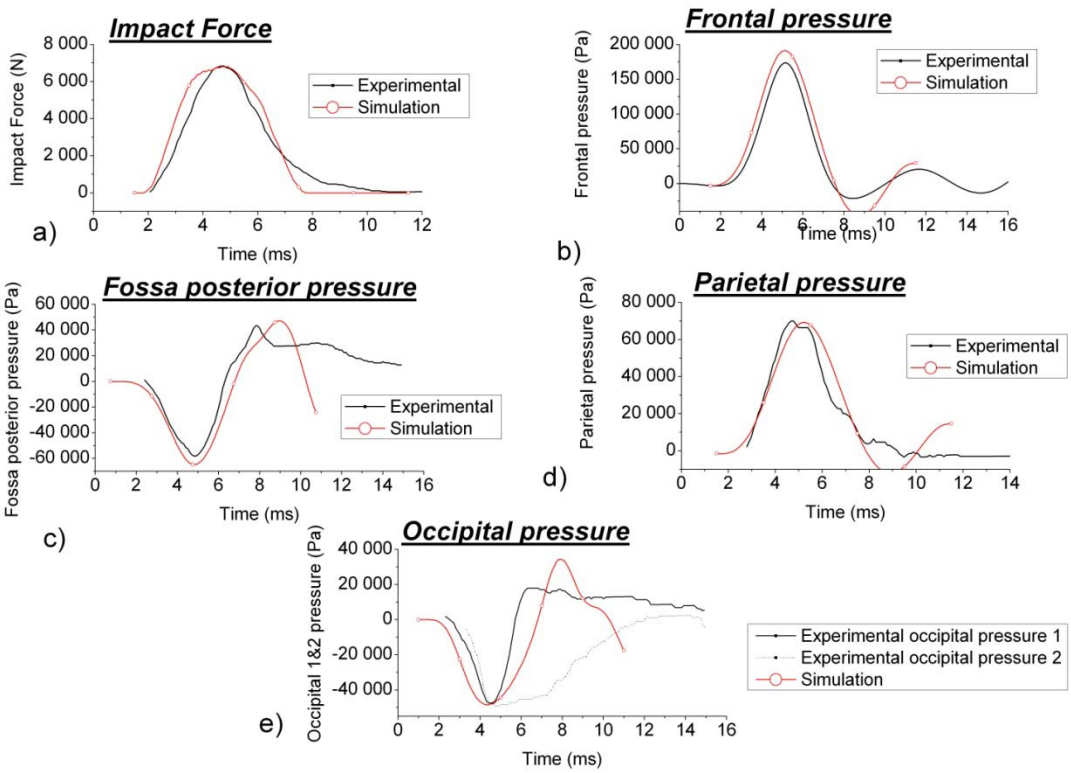


Figure 15. Experimental and numerical results comparison obtained for a Nahum's impact in terms of interaction force (a), frontal pressure (b), Fossa posterior pressure (c), parietal pressure (d) and occipital pressure (e) under Ls-Dyna code.

<u>Yoganandan's validation</u> - In order to validate material and section definition of the skull under Ls-Dyna software, Yoganandan's experiment was simulated. The numerical force-deflection curves are compared to the average dynamical response of experimental data (Figure 17). The dynamical model responses agree well with the experimental results, both the fracture force and the stiffness level.

When a layer fails, a parameter, called damage parameter, which is zero by default is set to one. Figure 16 illustrates damaged layer(s) in the simulation. The blue color indicates that at least one layer of the element failed. The model indicates fracture located around the impact point which complies with pathological observations.

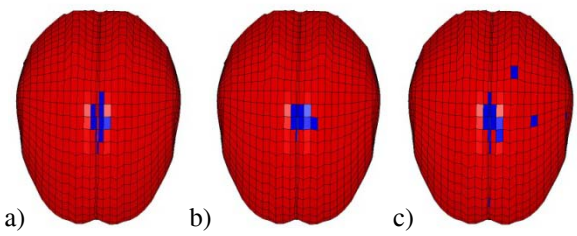


Figure 16. Skull failure description in terms of a) tensile fiber break, b) compressive fiber break and c) compressive matrix break

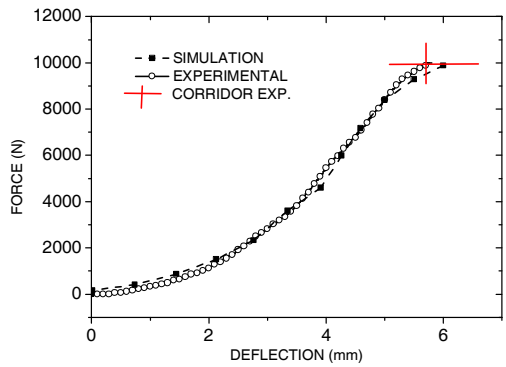


Figure 17. Experimental versus simulated force deflection curves until fracture (+ gives the corridor of Yoganandan's experimental results).

<u>SUFEHM Tolerance limits under Ls-dyna code</u>

Results computed with the SUFEHM under Ls-Dyna code are reported in terms of correlation coefficients (Nagelkerke R-Squared values) in order to express their injury prediction capability.

Based on SPSS method it appears that DAI are well correlated with intra-cerebral Von Mises

stress. Maximal principal strain as well as Von Mises strain presents also an acceptable correlation. Coming to maximum R^2 values, the maximum Von Mises stress conducts to 0.6 and 0.39 for respectively moderate and severe neurological injury.

The threshold for this parameter are of the order of 28 and 53 kPa respectively for moderate and severe neurological injuries as it appears in the injury risk curves reported in Figure 18.

Concerning the SDH injuries two mechanical parameters, i.e. CSF minimum pressure and CSF strain energy were considered.

With the SUFEHM it was shown (Table 4) that the best correlation with SDH was the maximum strain energy within the CSF, with a R^2 value of 0.465 and a threshold value of about 4950 mJ.

After the analysis of regression correlation method Figure 18 and figure 19 report the injury risk curves obtained with the SUFEHM for each of the injury types and the corresponding equations which permit to draw the S-curves. Finally Table 5 and Table 6 provide the tolerance limits for each injury mechanisms with an injury risk of 50%.

Table 4. Nagelkerke R-Squared value for the logistical regressions between the injury predictors computed with SUFEHM and the injury data.

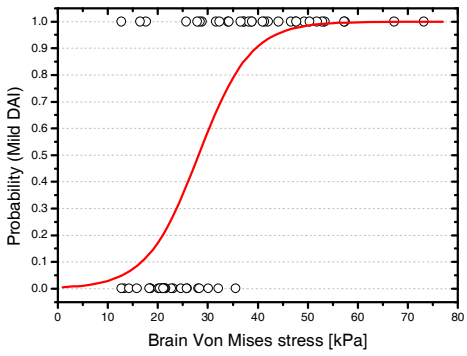
Injury Predictors	Mild DAI	Severe DAI	SDH
CSF minimum pressure			0.367
CSF strain energy			**0.465**
Peak brain Von Mises stress	**0.6**	**0.39**	
Peak brain first principal strain	0.43	0.355	
Peak brain Von Mises strain	0.43	0.35	

Table 5. Tolerance limits calculated for DAI injuries (mild and severe) with the ULP FE head model and LS-DYNA software.

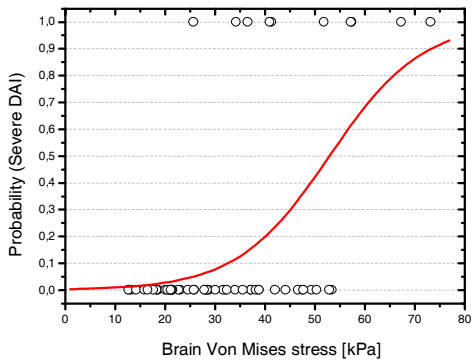
	Mild DAI	Severe DAI
Brain Von Mises stress [kPa]	28	53
Brain Von Mises strain [%]	30	57
Brain First principal strain [%]	33	67

Table 6. Tolerance limits calculated SDH injury with the ULP FE head model and LS-DYNA software.

	SDH
Minimum of CSF pressure [kPa]	290
CSF strain energy [mJ]	4950



Regression: log(p/(1-p)) = -5.438 + 0.193X
where X is the maximum brain Von Mises Stress



Regression: log(p/(1-p)) = -5.718 + 0.108X
where X is the maximum brain Von Mises Stress

Figure 18. Best fit regression models for DAI injury (moderate up and severe down) investigated for the SUFEHM considering brain Von Mises stress.

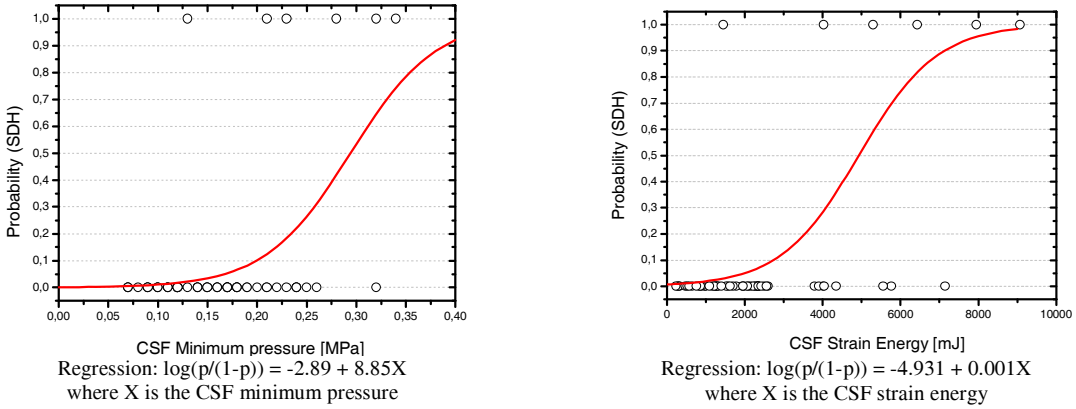


Figure 19. Best fit regression models for SDH/SAH injury investigated for the ULP head model considering CSF pressure (up) and CSF strain energy (down).

THUMS Validation Results

For the validation the simulation results can be compared for the head centre of gravity accelerations and displacements. In the tests of Ewing et al. the accelerations of the head centre of gravity have been calculated by considering the accelerometer positions which the occupants of the sled had fastened on their heads.

The centre of gravity for the SUFEHM model has been calculated by pre processing software tools. For the comparison of the results the closest node of the calculated head centre of gravity has been used (Figure 20).

The validation results for the impact loading are strongly dependent from the contact between torso and head. Therefore both curves with and without head contact to the torso are plotted in the result plots (Figures 21, 23 and 25 for frontal, lateral and oblique impact respectively).

Motions for a frontal, lateral and oblique impact of the finite element model deprived of the contact definition are given in figures 22, 24 and 26 respectively.

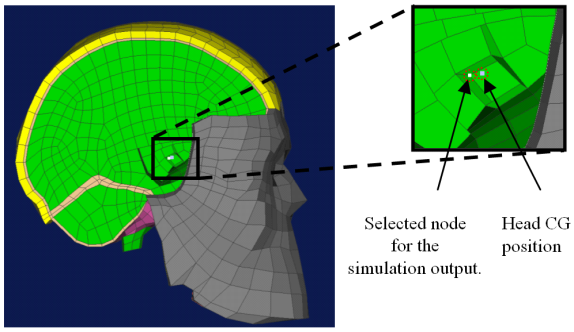


Figure 20. SUFEHM centre of gravity and closest node for comparison with simulation results.

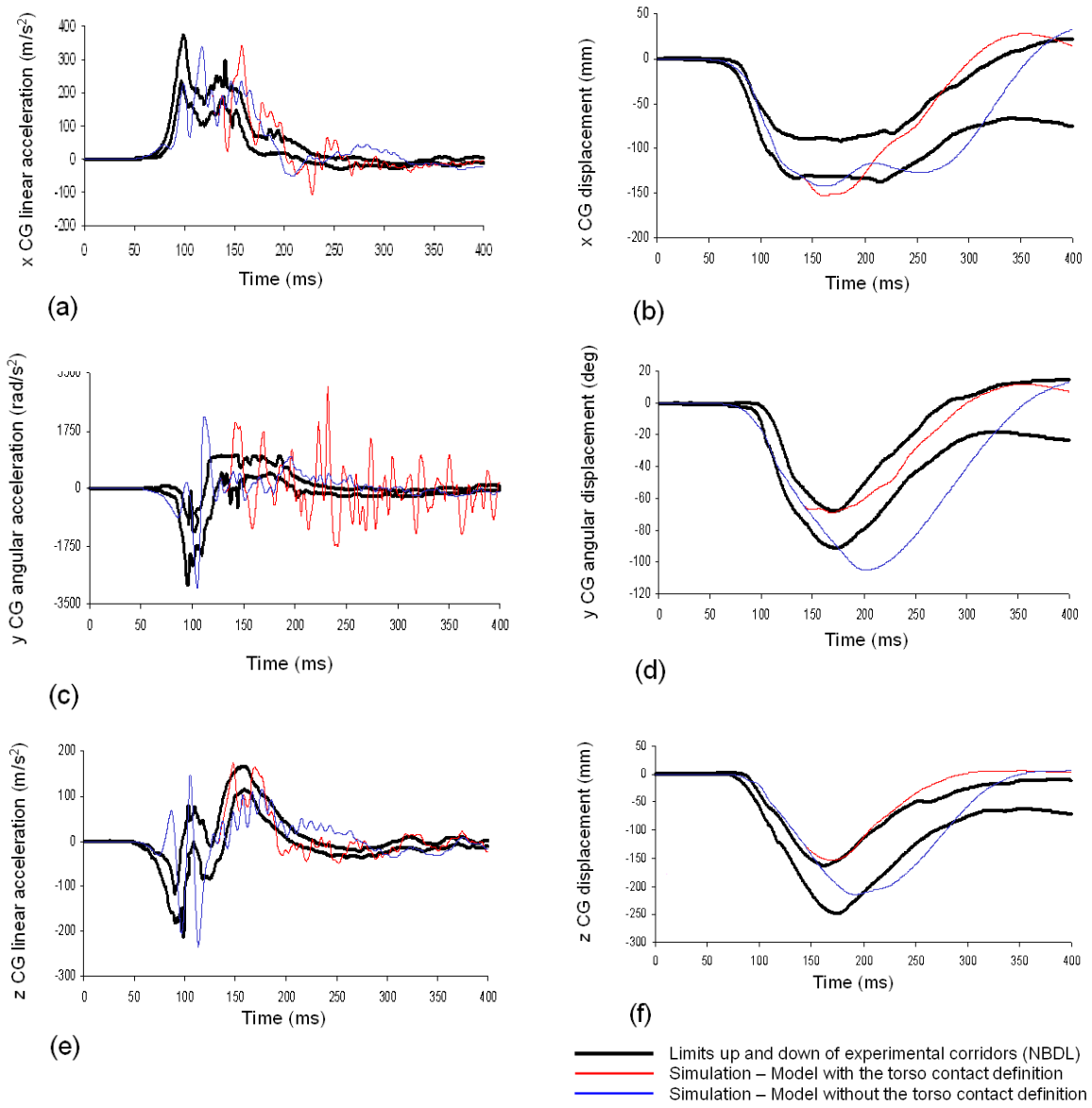


Figure 21. Simulation results of a frontal impact. Kinematics responses of the anatomical centre of the head: (a) x- angular acceleration; (b) y- linear displacement; (c) y- angular acceleration; (d) y- rotation; (e) z- linear acceleration; (f) z- linear displacement; Corridors and inputs for simulations follow experiments from Ewing et al. (1968, 1977)

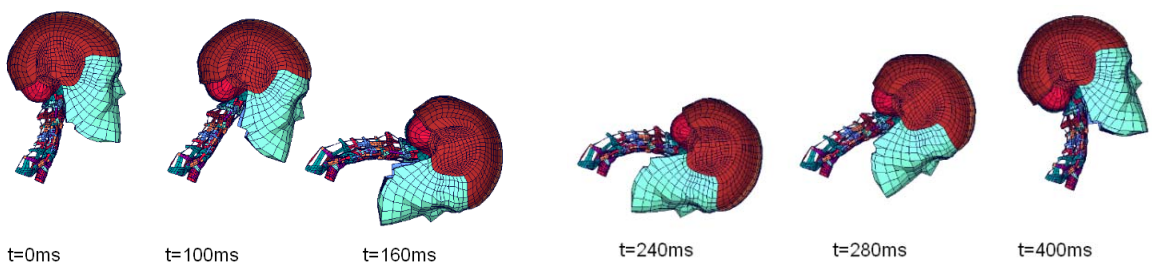


Figure 22. Motion for a frontal impact of the finite element model deprived of the contact definition (head and spine up to T1).

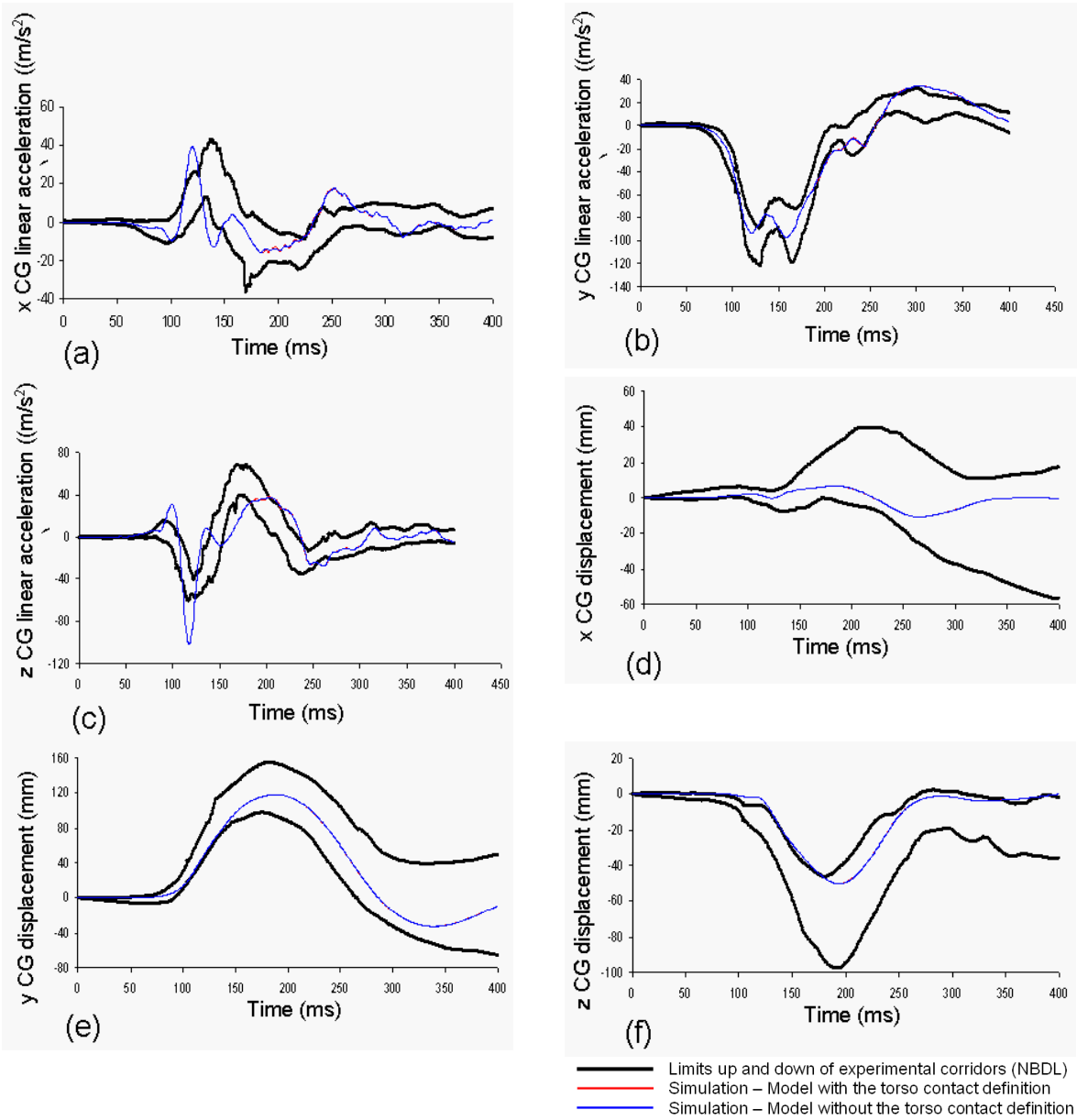


Figure 23. Simulation results of a lateral impact. Kinematics responses of the anatomical centre of the head: (a) x- (b) y- and (c) z- linear accelerations; (d) x- (e) y- and (f) z- linear displacements; (g) Motion of the finite element model deprived of the contact definition (head and spine up to T1). Corridors and inputs for simulations follow experiments from Ewing et al. (1968, 1977)

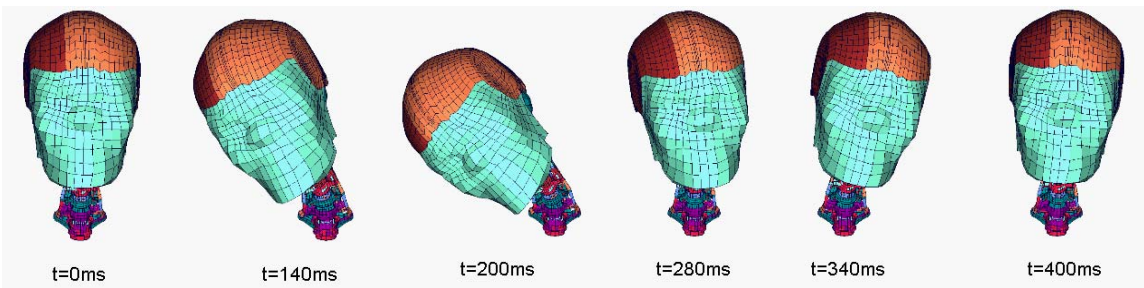


Figure 24. Motion for a lateral impact of the finite element model deprived of the contact definition (head and spine up to T1).

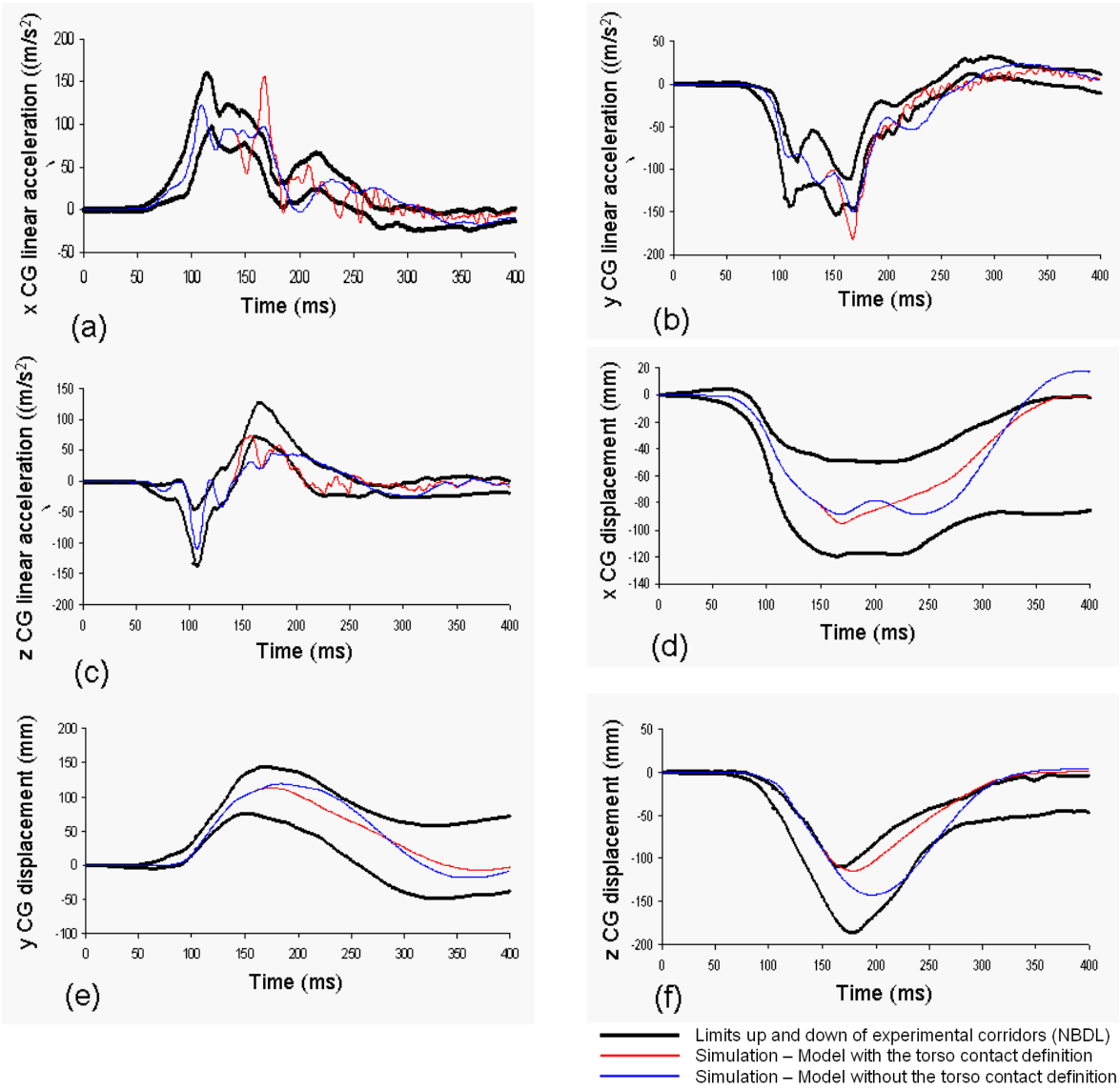


Figure 25. Simulation results of an oblique impact. Kinematics responses of the anatomical centre of the head: (a) x- (b) y- and (c) z- linear accelerations, (d) x- (e) y- and (f) z- displacements; Corridors and inputs for simulations follow experiments from Ewing et al. (1968, 1977)

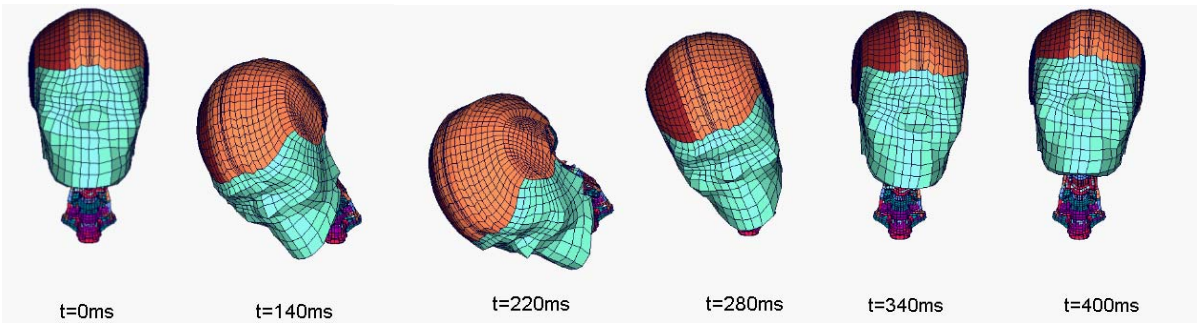


Figure 26. Motion for an oblique impact of the finite element model deprived of the contact definition (head and spine up to T1)

CONCLUSIONS AND DISCUSSION

In this study the Strasbourg University Finite Element Head Model (SUFEHM) has been transferred under LS-DYNA code, and mechanical properties have been implemented. Two experimental impacts have been replicated numerically, a Nahum's impact in order to validate the brain behaviour and a Yoganandan's shock to validate the skull stiffness and fracture.

In an attempt to develop improved head injury criteria under Ls-Dyna code, 59 real world head trauma that occurred in motorcyclist, American football and pedestrian accidents were reconstructed with SUFEHM. Statistical analysis was then carried out on intra cerebral parameters computed in order to determine which of the investigated metrics provided the most accurate predictor of the head injuries sustained in the accidents. Two tolerance limits to specific injury (for a 50% risk of injuries) have been computed:

- A maximum Von Mises stress value: 28 kPa for moderate DAI and 53 kPa for severe DAI.
- A maximum CSF strain energy: 4950 mJ for SDH

The original head model of the THUMS model has been replaced by the Strasbourg University Finite Element Head Model (SUFEHM). The Validation of the new head-neck complex has been done against the volunteer tests of Ewing et al. The results show a good fit for the linear accelerations and linear displacements of the head centre of gravity for all three impact directions.

The coupled model can now be used for further studies in which the whole body kinematics is important before the head suffers any impact. Compared with the real accident situations the whole body simulation is more realistic than a head impact model alone. The complete deformable finite element model can be used directly for simulations with other finite element models, without any pre simulations like with multi body models.

In car accidents the new model can be used for the analysis of pedestrian accidents and for occupants the oblique or multi directional loading can be simulated.

Another interesting study would be the accident analysis of motor cyclists. Especially the whole body kinematics with a helmet model can deliver new information.

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