DEVELOPMENT AND VALIDATION OF A NEW FINITE ELEMENT MODEL OF HUMAN HEAD

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ABSTRACT

Head injuries are one of the main causes of death or permanent invalidity in everyday life.
The main purpose of the present work is to build and validate a numerical model of human head in order to evaluate pressure and stress distributions in bones and brain tissues due to impact.
Geometrical characteristics for the finite element model have been extracted from CT and MRI scanner images, while material mechanical characteristics have been taken from literature. The model is validated by comparing the numerical results and the experimental results obtained by Nahum in 1977.
The proposed numerical model is promising even if some quantitative differences with the experimental results can be found due to the fact that all the inner organs are considered as a continuum (without sliding interfaces or fluid elements) and due to the geometrical difference between the head used in the experimental test and the head used as reference to build the numerical model.
The protecting action of the ventricles and of several membranes (dura mater, tentorium and falx) has been evaluated taking into account known injury mechanisms.

INTRODUCTION

Head injuries are one of the main causes of death or permanent invalidity in everyday life, especially among young people. Head injuries do not occur only on road accidents, but also during sport or work activities. During many years scientists have been trying to explain pathologies due to cerebral trauma searching for injury mechanisms, psychophysic consequences and possible treatments. In the last fifty years the consequences of head trauma have been studied also from a biomechanical point of view through the use of mathematical models.
Currently the parameter used in order to quantify the severity of a head damage as a consequence of a collision is the Head Injury Criterion (HIC). This parameter has been widely criticized. Its main limits are related to the fact that only linear accelerations are taken into account and that it should be used only when impacts against rigid surfaces are analyzed. Instead, several studies [1,2,3] concerning cerebral lesions demonstrates the influence and the importance of many other mechanical aspects as, for example, the angular accelerations and contacts responsible of stress and pressure distributions inside the cranium.
An effective way to predict several different head injuries (skull fracture, contusions, hemorrhage, diffuse axonal injury...) is the implementation and application of a finite element human head models validated by means of results obtained in experimental tests. Head models can be used to study the possibility of injury due to an external load (See figure 1).

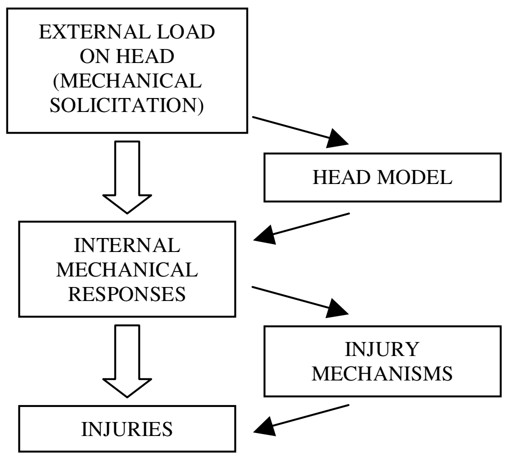

Figure 1. Block diagram representing injury sequence model.

Accelerations and forces are applied to the head model that tries to reproduce the behavior of a real human head in terms of internal mechanical responses. Injury mechanisms found by scientists (especially medicine doctors) for different biological tissues allow to estimate the possibility and severity of injuries due to the evaluated internal responses.
Different authors have proposed FEM models of human head during last years.
One of the first three dimensional model has been developed by Nahum, Smith and Ward in 1977 [4] in order to reproduce the experimental tests carried out on corpse heads. In this model the brain has

been modeled by means of 189 eight node brick elements while dura mater, falx and tentorium membranes have been modeled by means of 80 four node shell elements. A linear-elastic behavior has been adopted to model tissue mechanical properties.
A few FEM models of the human head have been proposed starting from this, each one characterized by several improvements:
- more realistic geometrical data due to the use of diagnostic medical instruments as Computer Tomography (CT) scans or Magnetic Resonance Images (MRI) ,
- introduction of different tissues and anatomical parts previously not present,
- more complex material models,
- higher number of elements due to the increased computational capabilities.

In 1993 Ruan, Khalil and King [5] developed a model of human head with 6080 nodes and 7351 elements where the scalp, the cranium, the cerebro spinal fluid (CSF), the dura mater and the brain were represented. In this model a visco-elastic behavior was introduced for the brain tissue. This model is known as the first version of the WSUBIM (Wayne State University Brain Injury Model) and has been continuously improved. In 1995 Zhou, Khalil and King [6] built a model with 17656 nodes and 22995 elements representing: the scalp, the cranium, the grey matter, the white matter, the brainstem, the CSF, the ventricles, venous sinuses, the dura mater, the falx, the tentorium, the parasagittal bridging veins and the facial bones. In one of the last versions (WSUBIM 2001) proposed by Zhang, Hardy, Omori, Yang and King [7] the number of elements grew up until 245000.
In 1996 Willinger et al. [8] proposed a head model focusing his attention on CSF and made some in vivo experimental tests to find its mechanical properties. The same author in 1997 developed a model with Kang and Diaw [9,10] where a elastic-brittle constitutive law has been introduced to describe the mechanical behavior of the bone and to simulate fractures.
In 1997 Claessens et al. [11] developed a model of human head where the elements of the skull and the brain tissues have been separated by introducing a sliding interface.
In 2001 Kleivin and von Holst [12] proposed a parametric model to evaluate the influences of geometrical dimensions on impact response. They introduced particular formulations to model the brain tissue and to simulate the sliding interface between brain and skull.
Papers found in literature have been analyzed in collaboration with several doctors leading to the conclusion that the presence of some inner elements as the ventricles and the veins and of the differentiation between grey and white matter should be investigated with more attention in order to improve the knowledge of injury mechanisms.
The objective of the present work is the development and the validation of a finite element model of human head to be used to evaluate the intracranial pressure and stress distribution due to a frontal impact, with particular attention toward the protective effects of some inner organs as membranes and ventricles.

MODEL DESCRIPTION

The geometrical model of the head has been build by taking advantage of CT scan and MRI images. More than 160 CT scan images corresponding to sections at a 1.25mm distances of a 31 year old patient with a cranium trauma without serious cerebral consequences have been used to build the internal and external surfaces of the cranium and the facial bones. Surfaces describing the inner soft tissues, as the ventricles, have been taken from MRI images of another patient scaled and adapted

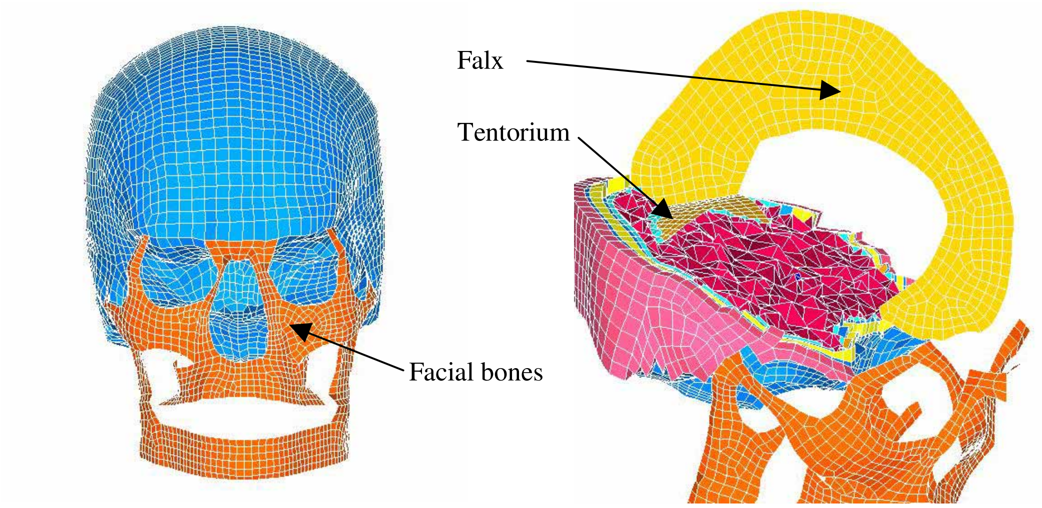

Figure 2,3. Frontal and perspective view of the head model.

to the surfaces obtained by the CT scan images.
CT scan images in the DYCOM format have been manipulated by using the software AMIRA. Triangulated surfaces in the STL format have been imported in PARASOLID and transformed in analytical surfaces for a better manipulation with the meshing code.

Finite Element Model

The finite element model (fig. 2-5) has been obtained by using Hypermesh 5.1. A continuous model has been adopted and contact elements between organs have not been defined.
The proposed numerical model is characterized by the following components:

- an external layer of brick elements with a 6mm thickness to represent the scalp,
- three layers of eight node brick element (two external layers of compact bone and one internal layer of cancellous bone) to represent the cranial bones,
- shell elements with only inertial contribution to describe the facial bones,
- four nodes shell elements to describe the dura mater, the falx and the tentorium membranes,
- eight node brick elements to describe the CSF,
- tetrahedral elements to model the brain tissues,
- tetrahedral elements to model the ventricles.

Tissues' Mechanical Property

It is very difficult to estimate the mechanical properties of the different tissues used in the model. They are very variable, depending on the experimental tests used to evaluate them. In this work literature data have been considered. For some parameters an average value of different literature values has been used, while other parameters have been adapted to the model.
All the tissues, except the brain, have been modeled with a linear-elastic behavior. The mechanical properties of the different components of the FEM model are summarized in table 1.
Mechanical properties used to model the scalp are ρ=1200 Kg/m^3, E=16.7 MPa and ν=0.42. This values have been used in WSUBIM [5], in the model by Zhou et al. [6] and in the model proposed by Willinger-Kang-Diaw [9,10].
A lot of experimental tests have been done by different authors in order to evaluate the bone mechanical properties. It has been chosen to adopt the same values used in Willinger et al. [9,10] to model the cranium bones. For the compact bone it has been used a Young modulus E=15000MPa, a Poisson ratio ν=0.21 and a material density ρ=1800Kg/m^3.

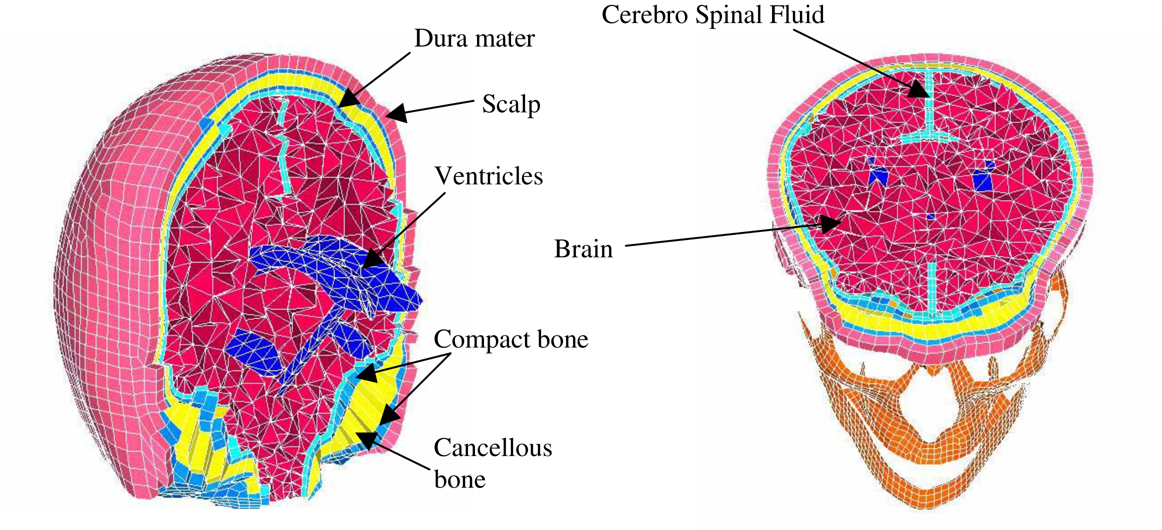

Figure 4,5. Coronal and axial section of the head model.

Dura mater has been obtained from the internal surface of the skull, while falx and tentorium have been built manually based on anatomical images. Cerebro spinal fluid has been obtained with a 2 mm offset from membrane surfaces. A layer of CSF surrounds all membranes and the brain. Internal surfaces of the CSF have been used as external surfaces for the brain volume.
The overall model is composed of 55264 elements and about 26000 nodes.

For the cancellous bone it has been chosen a Young modulus E=4500MPa, a Poisson ratio ν=0.01 and a material density ρ=1500Kg/m^3. Mechanical stiffness properties of the facial bones are not relevant for the proposed model and only inertial properties have been considered. The material density of the facial bones has been evaluated in order to keep a realistic mass.
For membranes, it as been found a general agreement in using the values (E=31.5MPa,

ν=0.21, ρ=1133kg/m^3) obtained by Nahum et al. and used in their FEM model [4].
CSF surrounding the membranes and filling the lateral ventricles has been modeled using a linear-elastic material with a 'fluid' option. In this case the element loses its ability to support shear stress and only compressive hydrostatic stress states are possible. For the fluid option the bulk modulus (K) has to be defined as the Young modulus and the Poisson ratio are ignored by the computational code. With the fluid option fluid-like behavior is obtained where the bulk modulus K is given by:

$$K = \frac{E}{3(1-2\nu)} \quad (1).$$

and the shear modulus is set to zero. The mechanical properties introduced in (1) are not well defined. The bulk modulus K varies in literature from 4.76MPa in Zhou et al. [6] to 2125 MPa in Willinger et al. [9,10]. Considering a value of the Young modulus E=0.012MPa and of the Poissons ratio ν=0.49 (nearly uncompressible material) a bulk modulus K equal to 0.2MPa has been obtained [8]. The CSF material density has been set to ρ=1040kg/m^3.
The brain tissue has been modeled by using a visco-elastic material model with shear relaxation behavior described by:

$$G(t) = G_\infty + (G_0 - G_\infty)\cdot e^{-\beta t} \quad (2).$$

where:
G_∞ = long-time (infinite) shear modulus,
G_0 = short-time shear modulus,
β = decay coefficient,
t = time.
Considering the first [8] and the second [9,10] model proposed by Willinger the decay coefficient β varies from β=0.035ms^{-1} to β=0.145ms^{-1}, the short-time shear modulus G_0 from G_0=528 KPa to G_0=49KPa and the long-time (infinite) shear modulus G_∞ from G_∞=168KPa to G_∞=16.7KPa.
According to a previous model developed in our department [13] it has been chosen to set up the bulk modulus equal to 5.625MPa, the decay coefficient β equal to 0.145ms^{-1}, the short-time shear modulus G_0 equal to 490KPa and the long-time (infinite) shear modulus G_∞ equal to 167KPa.
A material density value of ρ=1140Kg/m^3 has been used for brain tissues. A total mass value of about 1.4Kg has been obtained: it is acceptable considering the cerebrum weight (1.2÷1.5Kg), the pons and medulla oblungata weight (50÷75g) and the cerebellum weight (about 150g).

Table 1.
Material characteristics.

Tissue	Material model	ρ (kg/m^3)	E (MPa)	ν
Compact bone	Linear elastic	1800	15000	0.21
Cancellous bone	Linear elastic	1500	4500	0.01
Facial bone	Linear elastic	4500	10000	0.3
Brain	Visco-elastic	1140		
CSF	Linear elastic	1040	0.012	0.49
Ventricles	Linear elastic	1040	0.012	0.49
Scalp	Linear elastic	1200	16.7	0.42
Dura mater	Linear elastic	1133	31.5	0.45
Tentorium	Linear elastic	1133	31.5	0.45
Faulx	Linear elastic	1133	31.5	0.45

Simulations have been solved using dynamic finite element code LS-DYNA.

Boundary Conditions

The model has been considered as free in correspondence of the neck because the impact phenomenon is too fast to be influenced by neck constraints.
Experimental tests carried out by Nahum in 1977 [4] have been taken into consideration in order to set up the loading condition and to validate the numerical model. In these tests pressurized corpse heads have been frontally hit by means of a metal impactor. In particular, test number 37 has been considered as reference due to the geometrical similarities between the impacted head used in the test and the proposed FEM model.
The impact force and the pressure distribution in correspondence of the frontal area of the skull and of the posterior-fossa subarachnoid space have been considered as reference parameters for the validation.
Experimental tests also consider the pressure distribution on the frontal, occipital and parietal lobes. These data have not been considered because of the difficulty to find the exact corresponding area on the FEM model, whose position depends on the unknown layout of the cranial sutures.

The impactor has been modeled by the finite element method. It is characterized by the same mass (5.6kg) as in the experiment 37 and has been covered with a layer of elements with an elasto-plastic behavior. Contact between impactor and scalp has been defined as surface/surface contact in LS-DYNA code by using a static and dynamic friction coefficient equal to f=0.2.
Several mechanical parameters have been maintained constant in all simulations while others have been changed, in a significative range, to find a better correlation with experimental tests.

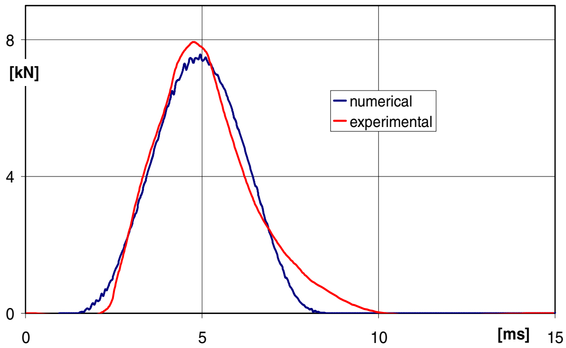

Figure 6. Impact force behavior.

The impact force behavior obtained by Nahum, related to the impact energy that has to be absorbed by head tissues, has been used as the first reference response to be reproduced with the numerical model.
Impactor mechanical properties influence the behavior of the impact force: the impactor speed influences mainly the peak value while the mechanical properties of the covering layer of the impactor control the time length and the shape of the force impulse.
In order to obtain the same peak of impact force, the impactor speed has been decreased from the real value of 9.6 m/s to a value of 7.0 m/s. This value is quite different from the experimental one used by Nahum in the analyzed experimental test (-27.5%) and it is probably due to differences in geometry, mass and stiffness distribution and to energy absorption mechanisms that are present in human tissues and have not been considered in the proposed numerical model. This difference became more relevant if kinetic energy of impactor is considered (-47%).
By setting up the impactor speed at V=7.0m/s, the best correspondence between numerical and experimental impact force behavior (see figure 6) has been obtained by using the mechanical properties of the impactor and of the covering layer shown on table 2.

Table 2.
Mechanical characteristics of impactor.

Tissue	Material model	ρ (kg/m^3)	E (MPa)	ν
Impactor	Rigid	5304	210000	0.3
Covering layer	Elasto-plastic	1050	1500	0.3

RESULT ANALYSES

Impact force has a peak value of F=7.56kN, nearly the same obtained by Nahum (7.9kN), and also the general behavior is quite similar (see figure 6).
Once the correct impact force has been obtained, the mechanical responses of the proposed numerical model have been evaluated and compared to those obtained by Nahum in the experimental tests. In particular the influence of the value of some mechanical properties used to model the brain tissues and the CSF has been studied.

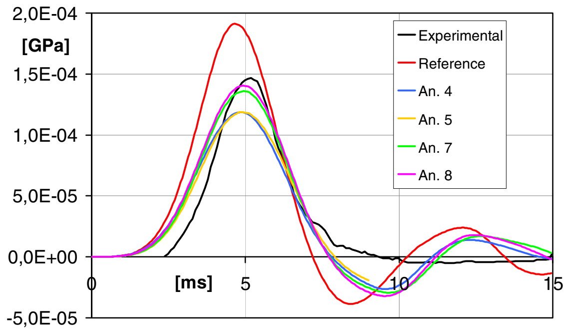

Figure 7. Frontal pressure behavior.

Different analyses have been carried out with different values of the bulk modulus for the CSF material and of the bulk modulus and of G_0 for brain material. The initial values chosen have been taken as reference and multiplied for a range of values varying between 0,1 and 10.

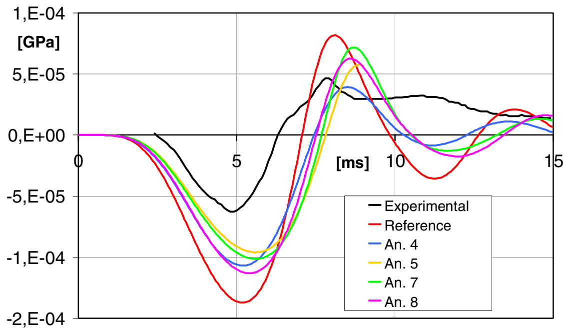

Figure 8. Posterior fossa pressure behavior.

Figure 7 and 8 show the numerical and the experimental pressure value evaluated in the frontal area of the cranium and in the posterior fossa for different values of the multiplying factors (summarized in Table 3). The pressure has been evaluated as an average value on four elements.

Table 3.
Multiplying factor for material characteristics in different analyses.

	CSF bulk	Enc. bulk	Enc. G_0
Reference	1	1	1
An. 4	1	0.2	2
An. 5	0.5	0.2	2
An. 7	0.5	0.3	2
An. 8	0.75	0.3	1.5

The general behavior is similar in both cases but there are differences concerning the peak values, especially for the posterior fossa pressure. Even with different values of the mechanical characteristics it has not been possible to obtain a significative improvement. This does not seems to be due to a wrong value in these mechanical parameters but to a lack of the model that seems to need the introduction of elements with a damping and/or retaining action for the brain tissues.
One of the main problems could be the modelization of the CSF. A structural analysis without fluid elements cannot correctly simulate the fluid damping and the fluid-dynamic migration of the CSF in different areas during impact although the very short duration of the phenomenon .
The activation of the fluid option for elastic elements improves the results by introducing a damping factor that reduces pressure oscillations but it is not sufficient to obtain accurate quantitative results.
Even if on the basis of some experimental tests [14] the bulk modulus has been evaluated equal to 2125MPa (nearly uncompressible) we have obtained better results (fig. 7,8) considering the brain compressible (bulk modulus less than 5.625Mpa). This is probably due to the fact that the compressibility assigned to the brain allows to take into account the mechanisms of movement of the CSF through the occipital foramen and of the blood flow of the inner vascular system that influence the history of the intracranial pressure during impacts.
Figures 9-17 shows the pressure distribution on a median sagittal section. A gradual transition from compression in the frontal zone to tension in the occipital zone can be observed. This is due to inertia forces that push the brain against the frontal portion of the skull and pull it from the occipital portion leading to large stresses in the connecting tissues between brain and bone. After the first

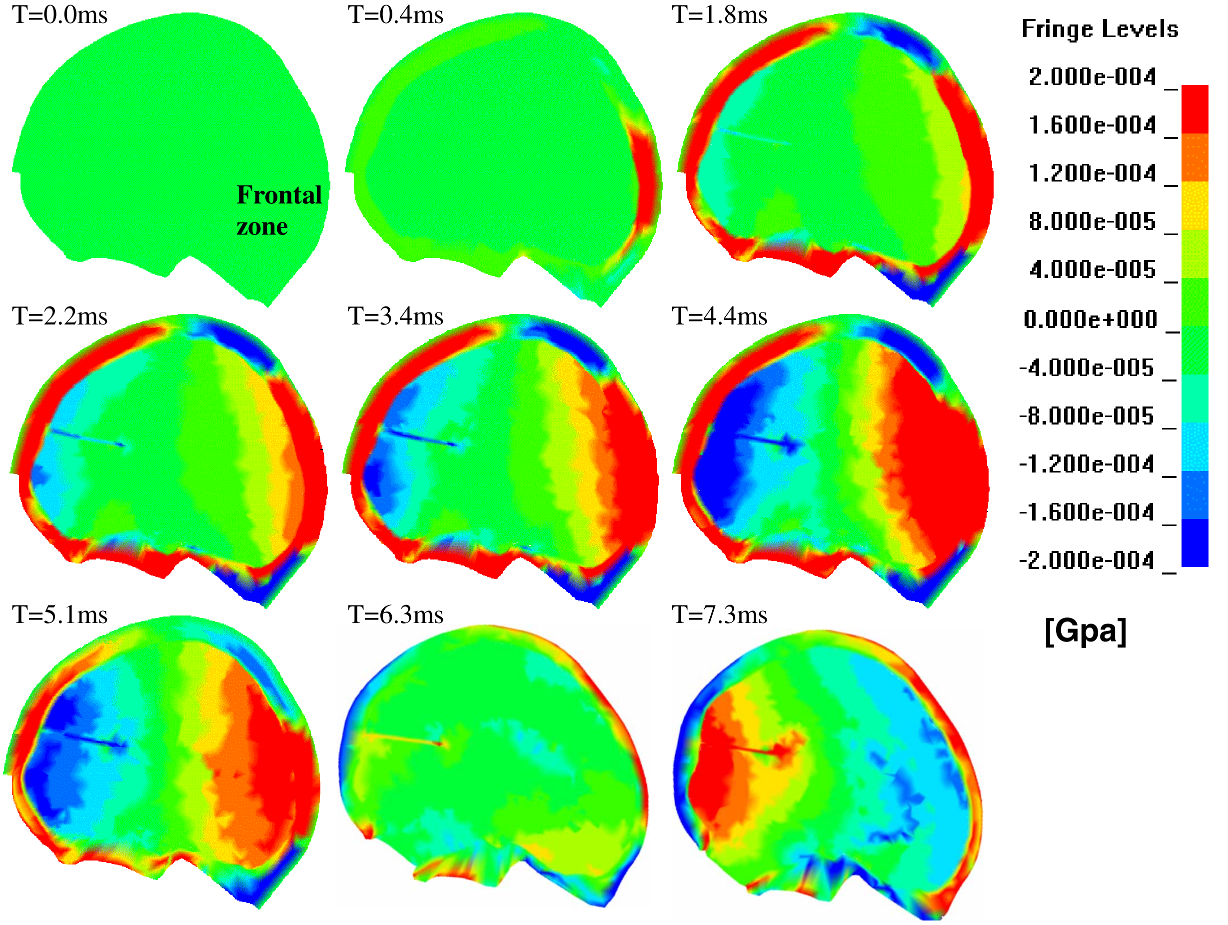

Figure 9-17. Pressure distribution on a sagittal section during impact.

bounce the brain return in the equilibrium position (about T=6.3ms) and the pressure distribution comes back to normality but the relative velocity between brain and skull, due to different inertial properties, creates the countercoulp effect when the brain is compressed towards the occipital zone (about T=7.3ms).
The analysis of the pressure distribution in different moments allows also to study the load transfer mechanism from the impacted area of the skull to the brain. In particular it is worth to notice the time delay of the mechanical responses in bone tissues and in the brain: high pressure values are reached in the bone about 1.8ms after impact while brain is still floating in the CSF and maximum values of pressure are reached after about 4.4ms in brain (figure 9-17.).

Influence of Ventricles and Membranes

Several medical studies have demonstrated the protecting effect of the ventricles and the membranes inside the cerebral mass. Some simulations have been done to investigate this behavior with the proposed model.

(about +300%) showing their protective effect in brain's central area.
Attention has then been focused on membranes. In this case elements corresponding to dura mater, falx and tentorium have been deleted. The absence of membranes leads to higher pressure peaks (+17% in the frontal zone and +18% in the posterior fossa, fig. 19) confirming also for these tissues an important protective effect.

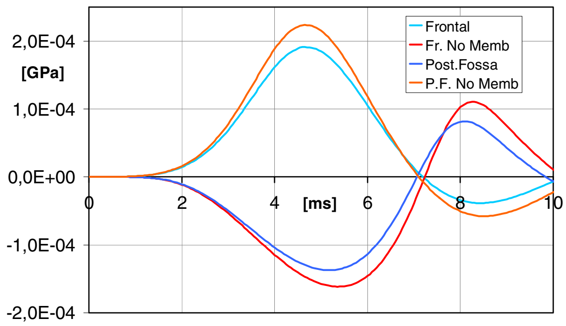

Figure 19. Pressure behavior with and without membranes.

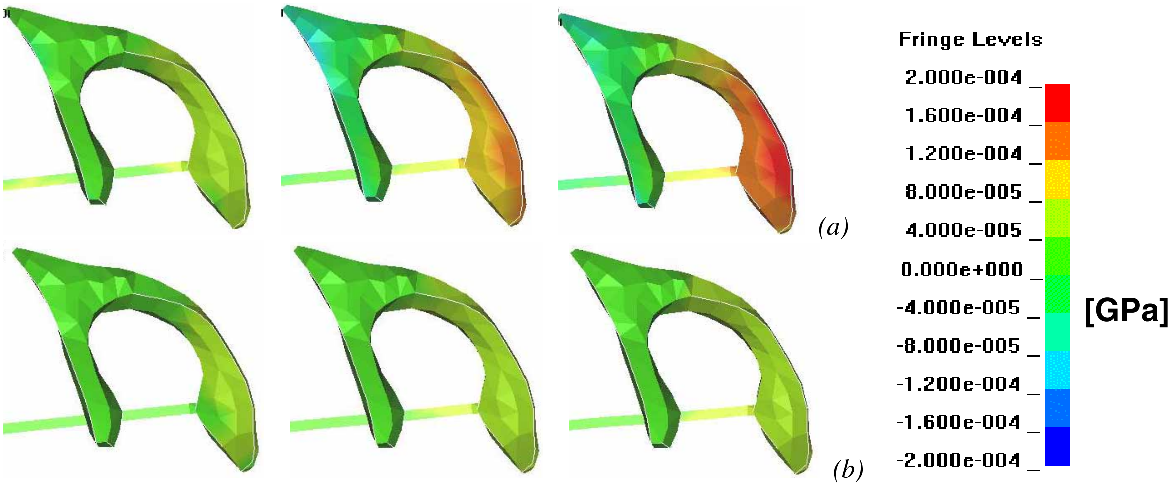

Figure 18. Pressure distribution in correspondence of ventricles after 3, 5 e 7ms (a) if they are modeled with brain material and (b) if they modeled with CSF material.

At first, attention has been focused on ventricles. The intracranial pressure distribution obtained by using the proposed model has been compared with that obtained by eliminating the ventricles. Ventricles elimination has been obtained by assigning to the corresponding elements the same mechanical properties of the surrounding brain tissues.
Pressure distribution in the frontal area of the skull and in the posterior fossa is not significantly different from that obtained by using the complete model. Otherwise relevant differences can be found in areas corresponding to ventricles surfaces (fig. 18) where peak pressure are strongly increased

The mechanical effect of membranes has been taken into consideration also by Claessens [11]. Also in his paper an increasing value for pressure distribution can be found eliminating membrane tissues.

Brain Injuries

Shear stress distribution has been analyzed on a median sagittal section and on a coronal section of the brain. Injuries concerning the brainstem and diffuse axonal injury (DAI) are usually related to the presence of large shear stresses.

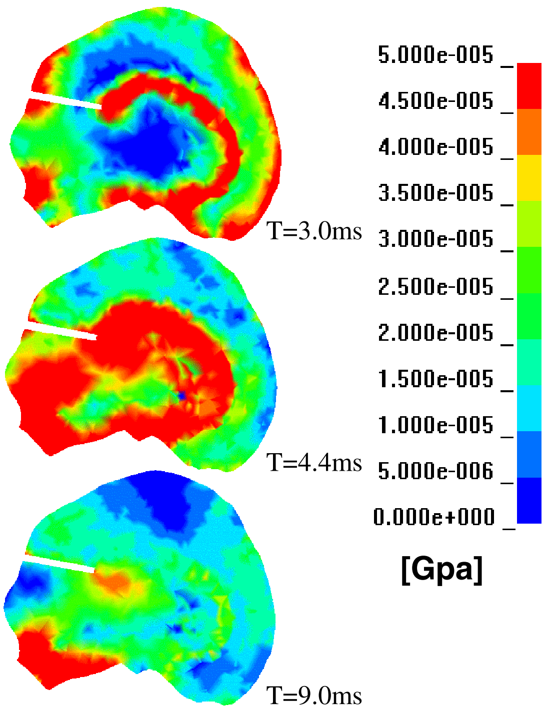

Figure 20-22. Shear stress distribution on sagittal section.

On the sagittal section maximum values of shear stress can be found at first in correspondence of corpus callosum, while later in correspondence of brainstem (fig. 20-22). Medical studies indicate this two tissues as the most affected by DAI. Some shear stress concentration can also be found in correspondence of the falx border, but this could be due mainly to the numerical model that does not allow sliding between tissues.

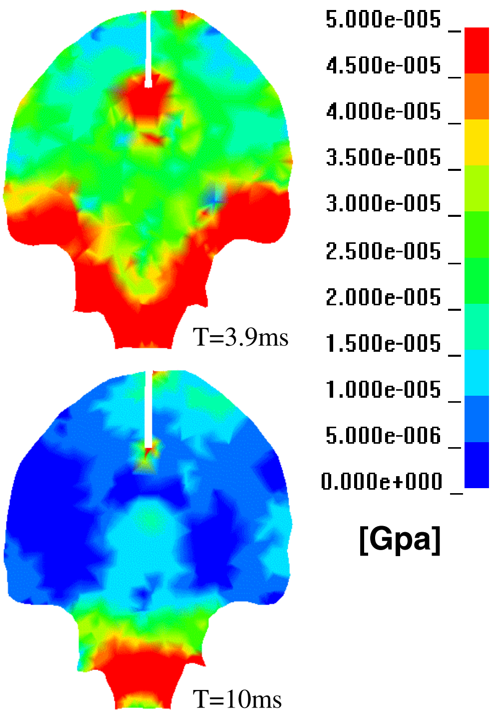

Figure 23-24. Shear stress distribution on coronal section.

A high value of shear stress can also be seen in the coronal section in correspondence of the brainstem. This area keeps being stressed for the greatest part of the impact phenomenon, also when all other tissues are almost relaxed (fig. 23-24). This behavior confirms the hypothesis that the brainstem is like a pivot for brain movements and may be seriously damaged by shear stresses.

CONCLUSIONS

A FEM model of human head has been built to study injury mechanism due to impacts.
The use of images obtained by TC or MRI scanners revealed fundamental to obtain a realistic geometry to be used as a starting point for the numerical model. Unfortunately it is not always possible to obtain the necessary TC or MRI images of the same head to build all the surfaces needed for the numerical model. In fact, in most cases, these medical analyses are focused on a particular pathology and there are some tissues that are put in evidence and other that are not visible.
The mechanical properties of several tissues to be used are not well known. In this work literature data have been considered to define the mechanical properties. For some parameters an average of different literature values has been used, while other parameters have been modified during the validation phase.
Some mechanical parameters have been kept constant in different simulations while others have been changed, in a significative range, to find a better correlation with experimental tests. Impact force intensity obtained by Nahum has been used as reference value to be obtained with the numerical model. Impactor speed has however been varied from the real value of 9.6 m/s to a value of 7.0 m/s to obtain the same peak of impact force. This difference is probably due to differences in geometry and in mass and stiffness distribution between head used in Nahum experiments and our model.
Good results have been obtained for the impact force and the pressure distribution behavior while there have been difficulties in simulating pressure behavior in posterior fossa. The same problem has been encountered also by others authors and is probably due to the material model adopted to simulate the cerebrospinal liquid. Using a continuous mesh and an elastic material with a low stiffness value (with fluid option), it is possible to simulate the “floating” effect of brain inside the cranium but not the motion of fluid in the subarachnoid space and the ventricles. The solutions, as already proposed by Claessens [11], could be a coupled analysis with fluid-solid

interaction or a contact interface between brain and dura mater to allow tissue sliding, as proposed by Kleiven and Von Holst [12].
Importance of some tissues introduced in this model for injury prevention has been investigated. In particular tentorium and falx structural stiffening function with respect to soft tissues has been pointed out. An important pressure absorbing capability of the ventricles has also been put in evidence.
The highest values of shear stress have been found in area where DAI lesions are usually found. They seem to be also responsible for injuries to brainstem and corpus callosum.
The model could be improved from an anatomical point of view, for example by introducing the bridge veins or the brain tissue differentiation between white matter, grey matter, cerebellum and brainstem behavior. Improvements could be reached by introducing more complex material models like, as an example, the real fluid behavior of the CSF or the fracture criterion for skull bones.
These considerations agree with some medical studies and more qualitative conclusions could be drawn with more experimental or clinical data. A close collaboration with doctors is considered as fundamental to obtain clinical data and information necessary to build more accurate models, to validate them and for a better comprehension of injury mechanisms.

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