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Numerical Model of the Human Head under Side Impact

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ABSTRACT

Head injury constitutes approximately 50 percent of all injuries sustained in transition and it is a common injury in sport and other human activities. Mathematical models provide powerful tool in the analysis of the mechanics of head impact. In particular the finite element method lends itself for the construction of a mathematical head model because of its capability to describe complex geometries.

In this study human head dynamic response to side impact is consider with finite element method. A two dimensional model of coronal section of human head has been designed using the actual human anatomy.

The reference model consisting of the three layer of skull (inner layer and outer layer are compact bone and mid layer is spongy bone), cerebral spinal fluid (CSF), brain membranes (falx cerebri and tentorium) and brain tissue. The model is loaded by a sinusoidal pulse with a peak pressure of 40 kPa.

Finite element analysis was conducted using Ansys software. Time-pressure history in the coup pressure region was studied.

The purpose of this study was to determine the effects of the membrane and effects of viscoelastic material properties for brain tissue on the dynamic response of the brain during side impact.

In the reference model –as opposed the reality-no relative motion at all is possible between brain and skull at their interface. Therefore the skull-brain interaction has been investigated in a parametric study using a contact algorithm.

KEYWORD: head injury, finite element, contact algorithm, viscoelastic, transient response.

1-INTRODUCTION

The human head is one of the most vulnerable parts of human body, when subjected to an impact loading. Every year many unfortunate victims suffer brain trauma. Head injury mechanisms have been proposed for many years and in spite of much research devoted to their verification, there are still many unanswered question. In particular, the mechanical factor causing brain dysfunction is still not cleared. It has been hypothesized that intracranial pressure during head impact produce brain contusions and that relative displacement between the brain and skull produce shear stresses in the brain causing cerebral bruising and hemorrhage. [1,2]

The study of human head under impact conditions may be split in two tasks. Firstly: the calculation of the deformation patterns of the skull and its contents as a result of an impact loading and secondly the identification of possible relationships between a particular deformation in a tissue and an injury in this tissue. For the first task, numerical model provide a powerful toll for the simulation of the deformations. They reduce the necessity of performing large number of experiments. Moreover they enable the calculation of tissue loads and deformations that cannot possibility be determined in experiment. For the second tasks, experiments need to be performed to determine the level of response at which the biological tissues fail to recover. The focus in this study will be on the first task, the simulation of the deformation of the human head using the numerical model**.** The objective of this study was to determine the affects of membranes on the brain protection, the material property of the brain tissue, free contact between the brain and scull in the transient load.

2-MATERIALS AND METHOD

2-1-Numerical model

Reference head model have been obtained using MRI scan and actual human head geometry [3,4]. The above data have been digitalized for different surfaces of the head consisting of 3 layers of skull, 3 surfaces of brain tissue, cerebral spinal fluid (CSF) and falx, tentorium membrane making up the boundary of the model. Fig.1 [5].

Affect of brain-skull contact has been modeled using the actual structure from the above data. In the reference model the brain-skull contact is in coupled form, while that of extended model the free contact between the brain and skull has been taken into account, providing a relative motion to the internal parts. (Figures.2, 3).

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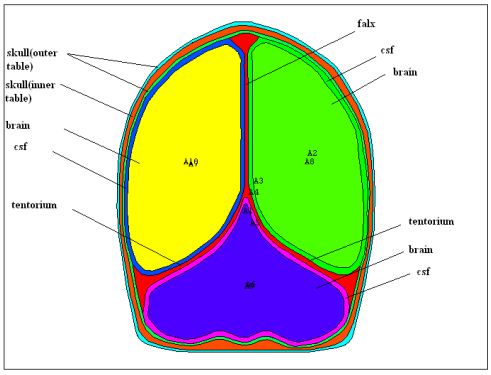

Fig.1 2.D model of head coronal section

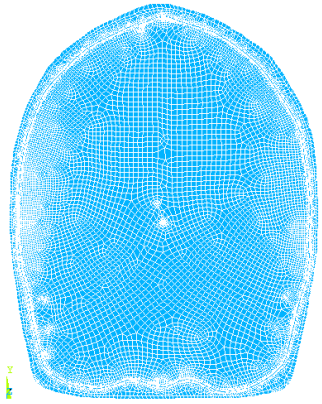

Fig.2 Meshing model of free contact

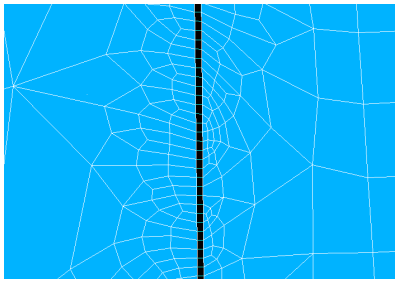

Fig.3 contact element between brain and skull

2-2- Constitutive equations

The deformation of a continuum is governed by a set of two equations, describing conservation of mass and balance of momentum. [6, 7]

$$\rho \frac{dv}{dv_0} = \rho_0 \qquad (1)$$

Where ρ is the density of the continuum and V the volume occupied by the material.

The equation of motion (balance of momentum) states that the time rate of change of the total momentum of the continuum equals the vector sum of all the external forces acting on the continuum:

$$\nabla.\sigma + \rho b = \rho \ddot{u} \qquad (2)$$

That σ denotes the stress tensor, b the body force vector per unit mass and u the displacement vector.

The interactions between two or more continua that are in states of contact are mechanically accented for by applying a set of kinematics and dynamic conditions for each of the continua. These conditions are prescribed for those parts of the boundary that actually come into contact. This part of the boundary of a continuum A is denoting by Γ_e^A. For frictionless contact the contact conditions are:

1. No penetration may occur during the period of time the two continua are in state of contact. For two points P and Q (with material coordinates ζ^p and ζ^Q), belonging two body A and B, respectively, the requirement of no penetration can be formulated as

$$([x(\zeta^p)+u(\zeta^p,t)]-[x(\zeta^Q)+u(\zeta^Q,t)]).n^B(\zeta^Q,t)=0 \qquad (3)$$

Where X denote the initial position of a point and $n^A(\zeta^p,t)=-n^B(\zeta^Q,t)$ are be outward normal of the continua at the points of contact.

2. The contact forces in the point of contact are equal but opposite in sin for each of the bodies.

$$\sigma^A(\zeta^p,t).n^A(\zeta^p,t)=\sigma^B(\zeta^Q,t).n^B(\zeta^Q,t) \qquad (4)$$

3. Only compressive forces are transmitted between the two continua in the point of contact.

$$\sigma^A(\zeta^p,t).n^A(\zeta^p,t)\le 0 \qquad (5)$$

The contact conditions for contact without friction in Lagrange multiplier method are now restated as

$$\lambda \geq 0 \qquad g \geq 0 \qquad g\lambda = 0 \tag{6}$$

The set of equations are obtained using finite element method, by taking the $q(\lambda, g) = 0$ equation into account with free resistance assumption after substitution of contact forces ($f_c^p = \lambda.n(\zeta^p, t)$) in the momentum equation.

$$\underline{\underline{M}}\,\underset{\sim}{\ddot{u}} + \underline{\underline{K}}\,\underset{\sim}{u} + \underset{\sim c}{f} = \underset{\sim}{f} \tag{7}$$

$$\underset{\sim c}{\gamma} = \underset{\sim}{0} \quad , \quad \gamma_c^k = q(\lambda^k, g^k) \tag{8}$$

For to the displacement degrees of freedom u, a Lagrange multiplier λ^k is added to the column of unknown degrees of freedom for each node that comes into contact, where this extra unknown represents the normal contact force.

For every node that comes into contact an equation is added to the total system of equations. The advantage of this method is that the contact conditions exactly met.

2-3-Material properties

The skull is represented as a homogenous isotropic structure with linear elastic material behavior [8-13].

The inner table, outer table of skull has compact bone characteristics, and dipole with spongy bone characteristics. In the reference model the skull content with linier elastic characteristics was taken into account. Table .1 shows the material characteristics of head tissues.

Table.1 material characteristics of head tissues

Head Tissue	**E(pa)**	$\rho(\frac{Kg}{m^3})$	ν
Outer Table	12.2e9	3000	0.22
Diploe	5.66e9	1750	0.22
Inner Table	12.2e9	3000	0.22
CSF	1.48e6	1040	0.4887
Membrane	9.45e6	1113	0.45
Brain	66.7e6	1040	0.48

2-4-Impact load and boundary conditions

The boundary conditions and the load type for both reference and free contact models are the same.

Rotation of the head on the neck has been taken into account for the boundary conditions in the model, where some parts of the nods are taken as single hinge and some others as double support .Fig.4.

A side pressure with maximum amplitude of 40 Kpa in a sinusoidal pulse form was employed in the model with 10 ms duration. Fig 5.

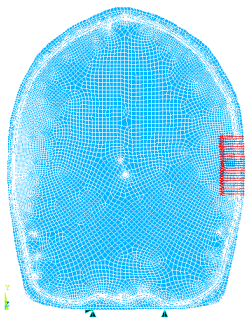

Fig.4 Schematic boundary conditions

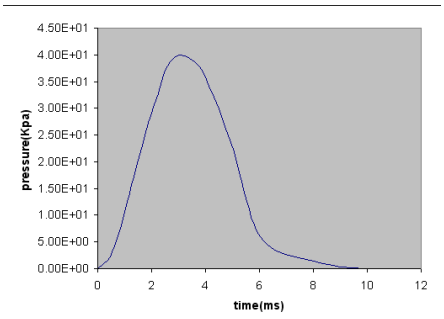

Fig.5 Sinusoidal pulse form was employed in the model

3-RESULT

3-1-Variation of linear viscoelastic brain properties

Simulation with the viscoelastic model for two different values of the time constant were compared with each other and with the elastic reference model. The short term behavior, governed by G_0 , was derived from the Young's modulus user for the reference model. The values used for the parametric variation are summarized in table.2 .The case with $\tau_2 = 10^{-4}$ represent a fast decays of the material stiffness whereas the model with $\tau_1 = 2 \times 10^{-2}$ lies between the former value and the elastic material model used in the reference model.

For the time interval analyses (10 ms) it is expected that for the time constant τ_2 full relaxation of the material will have taken place, before any significant pressure build up occurs.
The pressure – time history for the side region in the brain are presented in figure 7.

Table.2 values used for the parametric studies

description	G_0 (N/m^2)	G_∞ (N/m^2)	τ (s)	ρ (Kg/m^3)	ν (_)
τ_1	5.28e5	1.68e5	$\tau_1 = 2e-2$	1.04e3	0.499
τ_2	5.28e5	1.68e5	$\tau_2 = 1e-4$	1.04e3	0.499

The decay in material stiffness becomes apparent after about 2 ms, and is best seen in the result for the pressure for the τ_2 value. The trend seen was that the larger time constant led to higher pressure levels than the smaller time constant.

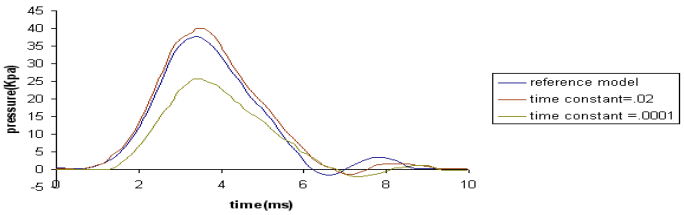

Fig.7 pressure – time history for various time constant

3-2-relative motion between skull and brain

A new model was constructed by decoupling the brain from the skull in the reference model. With this arrangement number of elements in reference model was increased from 4767 to 31441 in the free contact model, of whom 4033 are of contact elements.

Between the brain and the skull the corrected contact algorithm, presented in above was applied. The result for the pressure in the coup is present in fig 8. From the result it may be directly concluded that the free skull-brain interface has a large influence on the result. The pressure time histories for the coup region show much lower amplitude than the calculated pressures with the reference model. Dou to the application of contact algorithm only compressive forces can be transferred across the interface.

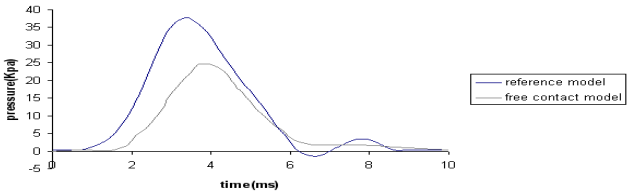

Fig. 8 coup pressure for the reference model and free contact model

3-3-Anatomical detail of the model

The falx cerebri and tentorium have a important function in the human head. The hypothesis regarding their mechanical function is that when the head is impacted, they support the cerebrum. The modified model

is consisting of all the reference model parts except for the falx and tentorium membranes, and connection between all structures are coupled. Different parts of the model are shown in figure 9.

The intracranial contents were attached to the skull, and all additional substructures were also rigidly connected to each other. No relation motion was therefore possible between brain and skull and skull and between separate structures.

All the structure represented as homogeneous linear elastic material.

Figure.10 depicted the calculated pressure in the coup region of the brain in comparison with the results calculated using the reference model. The coup pressure for the new model show a increase of maximum attained pressure level in comparison with the reference model. Decrease in the pressure in the reference model indicated the protective nature for cerebrum tissue.

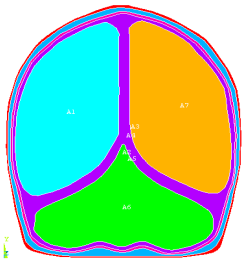

Fig. 8 Modified model without membrane

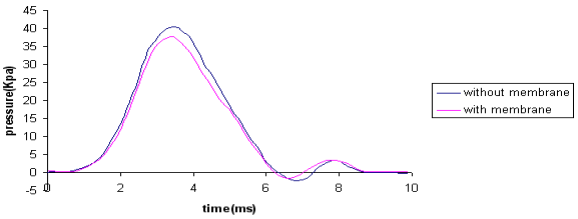

Fig. 9 Comparison of coup pressure in models with and without membrane

4-conclusion

The material properties and method of modeling the skull-brain interface showed to significantly change the result for pressures for coup region in the brain. For brain tissue viscoelastic materials the trend seen was that the larger time constant led to higher pressure levels than the smaller time constant.

In the contact model the pressure-time histories for the coup region show much lower amplitude than the calculated pressures with the reference model.

A geometrically realistic 2D finite element model of the human head was developed based on CT and MRI data.

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