National Conference with International Participation
ENGINEERING MECHANICS 2006
Svratka, Czech Republic, May 15 – 18, 2006
paper no. 298

HEAD INJURY CRITERIA ASSESSMENT THROUGH FINITE ELEMENT MODELLING

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Summary: *Paper presents a detailed finite element model of human head used for head injury criteria assessment. Today most used HICs are based on the history of linear acceleration only. However, the rotational acceleration plays also important role, due to the rupture of bridging veins involved in the relative motion between the skull and the brain. Presented model allows for new approach to define the HIC based on either strain energy or on combination of linear and rotational acceleration history. The model is based on a series of CT and MRI scans of a cadaver head of high resolution and is composed of three different regions: (i) skull, (ii) brain and (iii) subarachnoidal space. Discussion on material properties of these four distinct regions is given in the paper.*

1 Introduction

To study impact conditions during a traffic accident a detailed, anatomically correct FE model of human skull and brain is needed. There exist a number of finite element models of human skull. One of the earliest FE models of human skull for investigation of human head response was developed by Hardy and Marcall [1]. However, these first three-dimensional models reflected only the skull, not the brain. With the advancement of more powerful meshing techniques first FE models containing the brain were built. Early models considered the brain material to be linear elastic, later it was modelled as an inviscid fluid [2]. Viscoelastic properties of human brain were considered few years later in a number of articles, e.g. in [3] or [4].

One of the first three dimensional model verified against experimental data was developed by Nahum, Smith and Ward in 1977 [5]. This model was built to reproduce the experiments carried out using cadaver heads. In this FE model the brain is 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. Material properties of all tissues are modeled using linear-elastic behavior.

More realistic three-dimensional models of human skull and brain are developed using CT data of high resolution [6], [7]. These models usually reflect only human skull and brain, both modelled as linear elastic materials. Another approach is to reflect all the structures presented in the skull (bone and brain, but also the scalp, cerebrum, cerebellum, spinal cord and other structures), but these models are geometrically very simplified [8]. Only few of them [9] were

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validated against experimental data. In the recent years, complex models containing not only the human skull and brain, but also other very important tissues, like dura, the subarachnoid space filled with the cerebrospinal fluid [10] and even containing the cervical spine and spinal cord [11] were built. These very detailed models are used in numerical modelling of human head impact with emphasis on new injury criteria assessment. Great emphasis is paid to the brain-skull interface where fluid-structure interaction is taken into account [12]. Validation of the model developed by Willinger and Baumgartner shows good correlation with a number of experimental cadaver tests. Predicted intra-cranial pressure corresponds with experimental data as well [12]. However, for long duration impacts their model reaches its limits.

In 1993 Ruan et al [13] developed a new FE model of human head consisting of the scalp, the cranium, the cerebro spinal fluid (CSF), the dura mater and the brain. The total number of elements in this model was 7,351. Visco-elastic behavior was introduced for the brain tissue. This model has become later known as the WSUBIM (Wayne State University Brain Injury Model) and since than has been continuously improved.

Most of the models are based on a simplified geometry of the skull, brain and other structures. In the paper, detailed FE model of human head built using CT and MRI scans of human head is presented. A different approach is used – geometrically detailed model of human head and brain is built on the basis of series of CT scans. The FE model of human skull including the brain, dura mater and subarachnoidal space is developed based on scans of high resolution. To validate the FE model acceleration curves were obtained from a numerical simulation of guided fall. The guided fall was simulated according to cadaver experiments done by Got et al [14]. The HIC values obtained from these simulations were compared with the experimentally assessed ones.

2 Materials and Methods

The FE model of human skull including the brain, dura mater and subarachnoidal space is constructed using data obtained from Computer Tomography scans. These scans were acquired in resolution of 512x512 pixels taken in 1mm slices. For the surface reconstruction, a generalised Marching Cubes Algorithm [15] is used to identify the inner and outer surface describing the skull.

The triangular surfaces defined by the MCA were converted to NURBs surfaces which were divided into regions suitable for construction of hexahedral mesh. The volume of the cranial region is filled with elements of high quality representing the brain.

A great effort was put in shape quality checking and optimisation of the final hexahedral mesh. From the nature of the explicit formulation using reduced integration, elements must be of the first order and therefore it is advisable to use hexagonal meshes only. Material properties of the cortical bone were assumed linear elastic and homogeneous with Young's modulus of elasticity 14 GPa and Poisson's ratio 0.23. The brain is modelled as linear viscoelastic. Subarachnoidal space, which is a 2–3mm layer filled with cerebrospinal fluid separating the arachnoid from the pia is modelled as a linear elastic material with almost incompressible behaviour.

3 Results

Three different regions in the FE models were identified: (i) skull (ii) brain (iii) subarachnoidal space. Four material models were defined. The human skull was modelled as a sandwich

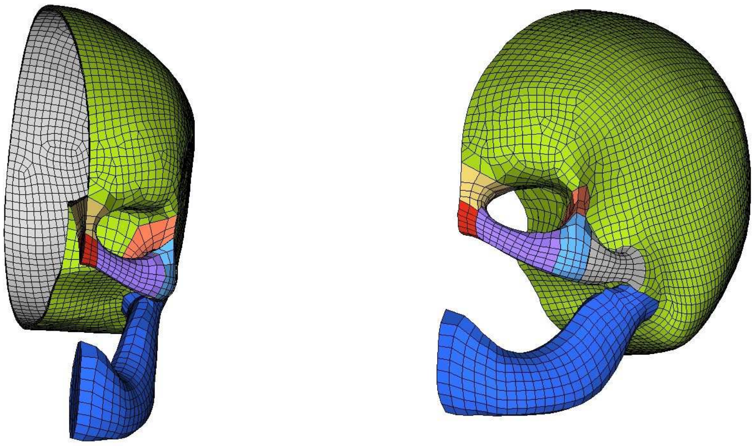

Figure 1: FE model of the skull showing the hexahedral mesh of high quality

construction, composed by a thin shell of cortical bone filled with elements representing the spongional bone. The difference between material properties of gray and white matter was neglected and the entire brain was modelled as viscoelastic material. Subarachnoidal space was identified by means of a Boolean operation.

The skull was modelled consisting of the inner and outer layer of shell elements representing the cortical layer. The thickness of this layer was assumed to be constant and was set to 1 mm. Material properties were assumed linear elasto-plastic with Young's modulus of elasticity $E_{cortical}$=12.000 MPa, density ρ=1850 $\frac{kg}{m^3}$ and Poisson number ν=0.21. The ultimate strength in compression and tension were considered σ^-_{ult}=80 MPa and σ^+_{ult}=140 MPa respectively. The space between these layers was represented by elements with material properties according to trabecular bone. The trabecular bone was also modelled elasto-plastic with the following material properties: E_{spongy}=2.200 MPa, ν=0.01, ρ=1500$\frac{kg}{m^3}$, σ^-_{ult}=32 MPa, σ^+_{ult}=30 MPa.

Material model used for brain was viscous elastic (both the white and gray matter was modelled using the same material properties) with following material properties: bulk modulus K=2200 MPa, density ρ=1000$\frac{kg}{m^3}$, value of the shear modulus at zero time G_0=1.036 kPa, shear modulus at infinity G_∞=0.00185 kPa, reciprocal value of the decay parameter $\frac{1}{\beta}$=0.0165$\frac{m}{s}$. The linear viscoelastic material model for the brain is defined by the time-dependant shear modulus according to the formula:

$$G(t) = G_\infty + (G_0 - G_\infty)\, e^{-\beta t}$$

The space between the brain and inner surface of the skull was filled with elements representing the subarachnoidal space. Material model for the subarachnoidal space was chosen linear elastic with Young's modulus of elasticity E=0.012 MPa, density ρ=1050$\frac{kg}{m^3}$ and almost incompressible (represented by Poisson's ratio ν=0.495).

To categorise the possible injuries of human head and brain complex, several approaches has been used. Nowadays, the Head Injury Criterion (HIC) is the most widely accepted criterion used to assess head injury potential in automobile crash test dummies. HIC is computed from the acceleration versus time curves using critical time span. It is based on the average value of the resultant translational acceleration over the most critical part of the deceleration. Head Injury Criterion is calculated using following formula:

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

According to this formula, HIC is the maximum value over the critical time period t_1 to t_2 for the expression in $\{\}$. The critical time period $\langle t_1, t_2 \rangle$ is chosen so as to maximise the resulting HIC. For more information about the severity index assessment, see e.g. [16].

Accelerations obtained from the FE simulation were in good correlation with those from experimental measurements. However, the presented model is relatively simple from the material point of view. The model does not account for the different structures presented in the brain, e.g. it does not make difference between the grey and white matter. Several existing FE models of human head account not only for white and grey matter, but also reflect other structures in the head that may play important role when investigating the strains and stresses in the tissues. Presented model is suitable for different HIC assessment, e.g. based on the rotational acceleration which is known to be important cause of severe damage to the brain. Other criteria, e.g. using shear strain or deformation energy of the brain tissue are possible as well.

Acknowledgement

The research has been sponsored by the Grant Agency of the Czech republic (grant No. 103/05/1(Financial support provided by the Grant Agency of the Academy of Sciences of the Czech Republic of the Czech Republic (grant No. AV 0Z 20710524) is fully acknowledged, too.

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