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EXPLICIT FINITE ELEMENT ANALYSIS OF HUMAN HEAD IMPACTING A RIGID SURFACE

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Summary: *Article deals with explicit finite element modelling of contact conditions of human head impacting a flat rigid surface. For this purpose, detailed 3-D model of human skull and brain was developed based on a series of computer tomography scans of resolution 512x512 pixels taken at 1 mm distances. Different mesh generation techniques were studied to obtain the final FE model. Simple FE analyses of stress state under simple static loading were compared to several models of different resolution to obtain the final, optimised one, which was used in the case study.*
The FE model of human skull was filled with simplified model of human brain. The geometry of the brain was based on a series of MRI scans of the same individual. The brain is modelled as a homogeneous viscoelastic material. A case study of low-velocity impact to different regions of the human head (e.g. frontal impact and impact in the occipital region) is also modelled. After thorough validation these models will be used in conjunction with different designs of protective helmets to study contact stresses between different helmets and the skull as well as for evaluation of the influence of protective helmet design.

1 Introduction

To study impact conditions during a traffic accident a detailed, anatomically correct FE model of human skull and brain is needed. The geometry of both skull and brain were constructed using data obtained from Computer Tomography scans. These scans were acquired in resolution of 512x512 pixels taken in 1mm slices. Fully automated direct generation of the volumetric tetrahedral mesh based on the Marching Cubes Algorithm, Laplacian smoothing and Delaunay tetrahedralization was used to develop the geometry of both the human skull and the brain. For more detailed information about the whole procedure see e.g. [1].

For the purpose of explicit dynamic modelling material properties of the cortical bone are assumed linear elastic and homogeneous with Young's modulus of elasticity 14 GPa and Poisson's ratio 0.23 . The volume of the cranial region is filled with tetrahedral elements of high quality representing the brain. The brain is modelled as linear viscoelastic. Results obtained using this detailed FE model are compared to experimental results from a standard drop test. These experiments use a standard metal head form of variable size. After thorough validation of the FE model, it will be used in more complex dynamic simulations as well as for the Head Injury Criterion (HIC) assessment.

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2 Development of the FE model

The skull geometry was constructed on basis of a series of CT scans acquired at the resolution of 512x512 pixels taken in 1mm slices. For selection of the bone tissue, fully–automated procedure based on thresholding combined with repetitive application of Gaussian smoothing and consequent removal of islands (removing regions of tissue smaller then specified value) was used.

For the surface reconstruction, a generalised marching cubes algorithm [2] is used to identify the inner and outer surface describing the skull. This algorithm utilise the "divide and conquer" technique to determine the surface of an organ. Since this algorithm produces very large number of triangles (in our case of detailed CT data the total number of triangles was over million) describing the surface a relatively simple decimating algorithm [3] was used. The triangular surface mesh was smoothed using Laplacian approach and finally, the Delaunay tetrahedralization [4] was used to fill the volume with tetrahedral finite elements.

The finite element model of the brain was constructed using the same CT data. It is well known fact, that soft tissues are better distinguishable from MRI scans, however, for our purposes it was necessary to use the same imaging data. In this case, it was more complicated to detect the surface of the brain and more sophisticated contour-tracking approach was used. Some manual intervention was also necessary. Resulting FE meshes of the skull and brain are presented in Fig. 1.

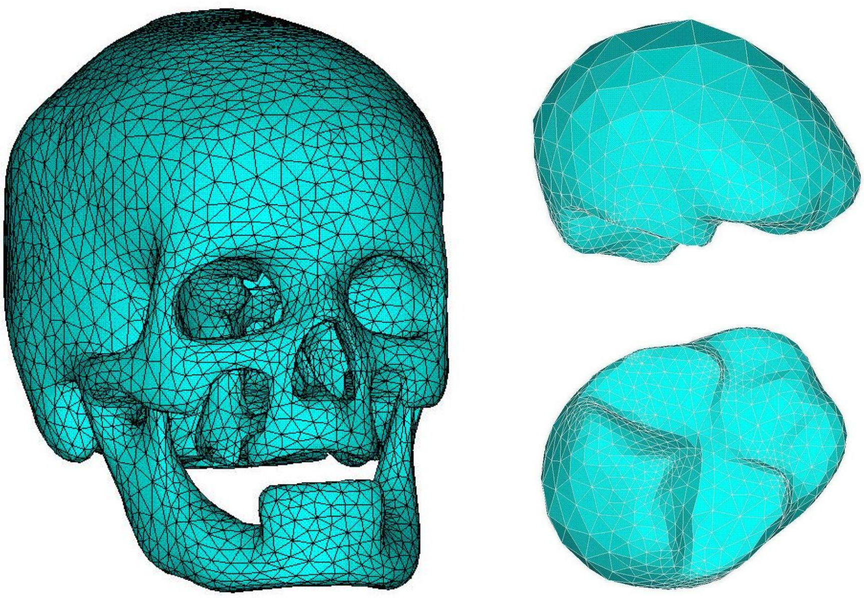

Figure 1: FE model of the human skull and brain

Five models of different resolution were constructed using the same input data. These models differed in the total number of nodes. To select the most suitable model for the time consuming drop test simulation a simple static analysis was computed using all these models. Results from this simple analysis were compared and the most convenient model (with sufficient number of nodes) was chosen.

3 Drop–test experiments

Protective capacity of helmets is usually investigated using drop tests. The test consists in a controlled impact where a helmet is positioned on a metal head form and then dropped in a guided fall. As the target surface, various steel test anvils (flat, edge, hemisphere or a horseshoe type) can be used to simulate different impact surfaces. The head form is instrumented with triaxial accelerometer to measure acceleration components and enabling thus to compute peak G force.

In our experiments, we use standard head forms given by the Czech National Standard CSN EN 960 change A1 832140. These head forms are of different sizes, made from an alloy with inertial properties corresponding to those of appropriate human head.

Figure 2: Drop test configuration

To categorise the possible injuries of human head and brain complex, several approaches has been used. One of the first head injury criteria taking into account the acceleration time history is the Head Severity Index (HSI) or Gadd Severity Index (GSI):

$$HSI = \int [a(t)]^{2.5} \, dt \tag{1}$$

Currently, the Head Injury Criterion (HIC) is 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. From the resultant acceleration–time curve average value of acceleration over the time interval is given by:

$$\overline{a} = \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \tag{2}$$

and the Head Injury Criterion is calculated using following formula:

$$HIC = \begin{matrix} max \\ t_1, t_2 \end{matrix} \left\{ (t_2 - t_1) \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} \right\} \tag{3}$$

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. [5].

4 Explicit dynamics

High quality FE models of both the skull and brain were imported into explicit dynamic code (Ansys LS-DYNA). The explicit method of solution is advantageous for short-time, large deformation dynamics or quasi-static problems with large deformations. In this case, the explicit code was chosen mainly due to the complex contact nature of the problem (contact between the soft brain and relatively stiff skull which is also in contact with rigid target representing a steel plate).

From the nature of the explicit formulation, elements used must be of the first order and therefore it is advisable to use hexagonal meshes. However, this is nearly impossible for domains of such complicated shape as for the human skull or brain. A great effort was therefore put in shape quality checking and optimisation of the final tetrahedral mesh.

Reduced (one–point) integration plus viscous hourglass control was used for faster element formulation. The use of one–point integration is advantageous due to savings on computer time and in case of large deformations of the brain (high–speed impacts). Contact between the surface of the brain and the inner surface of the dura was accomplished using the surface–to–surface contact elements with friction. The coefficient of friction is assumed to be dependent on the relative velocity, v_{rel} of the contacting surfaces and on the static and dynamic coefficients of frictions:

$$\mu_c = \mu_d + (\mu_s - \mu_d)e^{-v_{rel}} \tag{4}$$

Central difference time integration is used to solve for accelerations instead of displacements. Accelerations at time t are computed as follows:

$$\left\{a\left(t\right) = \left[M\right]^{-1} \left(\left\{F^{ext}\left(t\right)\right\} - \left\{F^{int}\left(t\right)\right\}\right)\right\} \tag{5}$$

The vector of internal forces $\{F^{int}\left(t\right)\}$ is given by:

$$F^{int}\left(t\right) = \sum \left(\int_{\Omega} B^T \sigma_n d\Omega + F^{hourglass} + F^{contact} \right) \tag{6}$$

The mass matrix is calculated as the lumped matrix enabling simple inversion. Then the system equations:

$$\left[M\right]\left\{\ddot{r}\right\} + \left[C\right]\left\{\dot{r}\right\} + \left[K\right]\left\{r\right\} = \left\{f\right\} \tag{7}$$

become uncoupled and can be solved explicitly (directly). All nonlinearities (in our case including contact) are included only in the vector of internal forces. The nodal velocities and displacements are computed using the central difference time integration:

$$\left\{v_{t+\Delta t/2}\right\} = \left\{v_{t-\Delta t/2}\right\} + \{a_t\}\,\Delta t_t \qquad \{u_{t+\Delta t}\} = \{u_t\} + \left\{v_{t+\Delta t/2}\right\}\Delta t_{t+\Delta t/2} \tag{8}$$

Updated geometry of the domain is found with the help of the displacement increments from the initial geometry:

$$\{x_{t+\Delta t}\} = \{x_0\} + \{u_{t+\Delta t}\} \tag{9}$$

The solution of an explicit problem is stable only if the time increment Δt is smaller then the critical one. The critical time increment can be calculated based on the Courant-Friedrichs-Levy criterion and for stability reasons is lowered by a scale factor of 0.9 to $\Delta t = 0.9\frac{l}{c}$, where l is the characteristic length of an element and c is the propagation velocity.

5 Results

Various impact conditions (according to the drop tests used in HIC evaluation) were studied. As an example analysis, simple drop test is chosen to demonstrate the possibilities of explicit solution. The material model is linear viscoelastic for the brain tissue defined by:

$$G(t) = G_\infty + (G_0 - G_\infty)\, e^{-\beta t} \tag{10}$$

with following material properties considered: density ρ=1000$\frac{kg}{m^3}$, bulk modulus K=2.2 GPa, G_0=1.036 kPa, G_∞=185 Pa, β=0.0165 ms^{-1}. The material of the human skull is modelled as linear elastic. Only the cortical tissue was considered with Young's modulus of elasticity E=17 GPa and Poisson's ratio ν=0.23.

As an example of the results obtained a drop test of the human skull from 2 m height onto a rigid plate is presented. To save computational time, the drop height was lowered and correspondent initial velocity supplied. As an example of these results, third principal stresses are presented in Fig. 3.

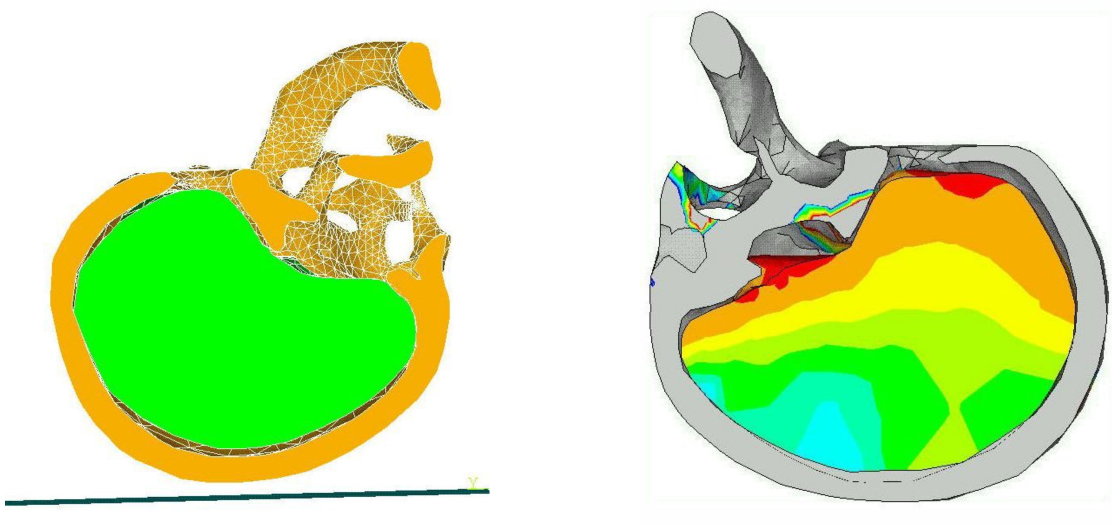

Figure 3: FE model and sample results (third principal stresses σ_3 in a simple drop test)

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