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FINITE ELEMENT RECONSTRUCTION OF CRANIOCEREBRAL INJURY

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Summary: *The paper deals with reconstruction of a sport accident using detailed FE model of skull and brain is presented in the paper. The sport accident involved a 12-year old boy on whom a handball cage fell during school sport activity. FE model of human head was developed using series of CT scans obtained postsurgically. Rigid body model was used to assess initial conditions at the moment of the impact which were used in finite element modeling of the head impact. The detailed FE model was imposed to the initial conditions obtained just before the head impacted the playground.*
The pressure, shear stress response, von-Mises stress response and logarithmic strain values were evaluated in four regions: (i) frontal, (ii) parietal, (iii) occipital and (iv) midbrain region. For the head injury assessment a criterion based on the criterion proposed by Miller et al. in 1998 is applied. Results from the numerical analysis of the accident showed good agreement with clinically observed head injuries.

1 Introduction

Brain injury is the leading cause of death in those aged under 45 years in both Europe and the USA. One of the application of forensic biomechanics is reconstruction of sport, traffic or daily-activity accidents using numerical modeling. The dynamics of the impact can be described by an equation of motion, that is by a second order differential equation which can be solved using the Finite Element Method (FEM).

To study impact conditions during an 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 modeled 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].

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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. These models usually reflect only human skull and brain, both modeled 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 [6].

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. Geometrically detailed model of human head is built on the basis of series of CT scans obtained post-traumatically. The FE model of human skull is then filled with a simplified model of human brain. The simplification consists in considering the brain composed of one tissue only, neglecting the different material properties of white and gray matter. Resulting FE model is subjected to the same initial conditions as during the accident.

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 5 mm slices. For the surface reconstruction, a generalised Marching Cubes Algorithm [7] is used to identify the inner and outer surface describing the skull. The volume of the cranial region is filled with elements of high quality representing the brain.

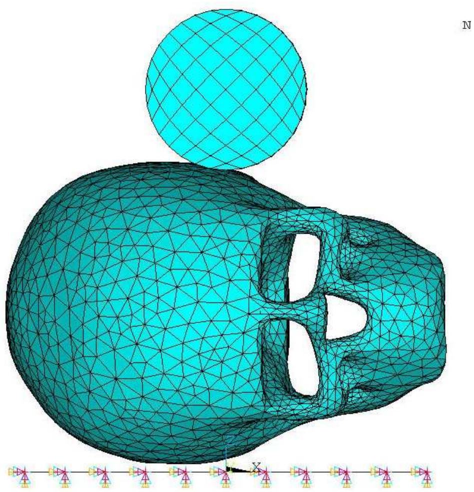

Figure 1: FE model of the skull showing the upper bar of the falling cage

The initial configuration of the head impacting the ground as well as the initial velocity of the steel cage were obtained from rigid-body modeling of the fall. For the rigid-body simulation of the fall MADYMO software package was used with the help of the 5% female pedestrian model. The small female model (1.52 m and 49.8 kg) was closest in weight and size to the boy injured (1.58 m and 40 kg). From the female model, the ellipsoids representing breasts were removed.

Because of the lack of knowledge of the initial conditions of the fall, two model cases were considered and resulting accelerations compared. Both cases were considered, however, for illustrative purposes only one case is depicted in which the body was subjected to gravitational acceleration. The initial configuration is depicted on the first image of the series presented in Fig. 2.

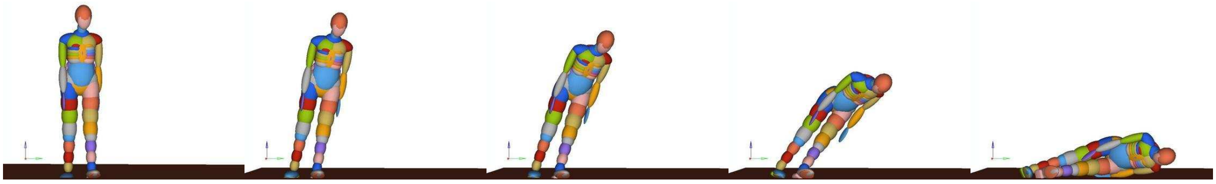

Figure 2: Rigid body reconstruction of the fall using the 5% pedestrian model

From the rigid body simulation of the fall several important values can be easily determined, e.g. acceleration of the center of gravity of the head. Using the acceleration history (see Fig. 3) we can calculate the head injury criterion as well as other important injury characteristics. In our case, the rigid body simulation was used to obtain the initial conditions of the head used in later FE modeling.

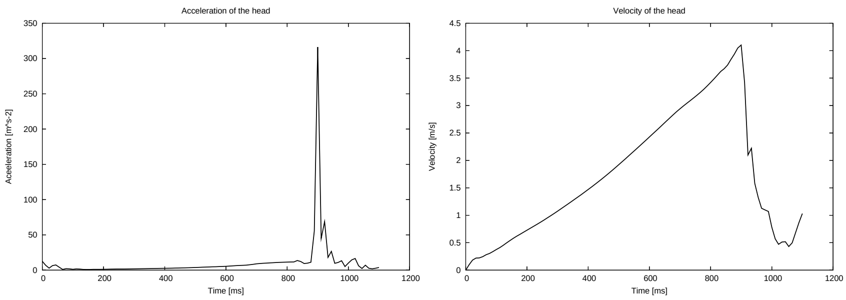

Figure 3: Velocity and acceleration history in the center of gravity of the head

To determine the velocity and angular acceleration of the impacting handball cage a simple rigid model was used. The cage was supposed to fall from its indifferent equilibrium position. Differential equation of motion was used to determine velocity and angular acceleration of the upper bar of the cage at the moment it hit the boy's head. Solution of the differential equation is not trivial, it leads to an elliptic integral. Therefore, the equation was linearized and solved incrementally for the angle to be sufficiently small (increment of 5 degree was assumed). The components of velocity and acceleration determined from the equation of motion for the angle 85 degrees (upper bar touching the skull) were used as initial conditions for the movement of the cage modeled as deformable bar in the FE analysis of the accident.

As it was mentioned earlier, the geometry of the model was prepared based on the post-operative images. The surface of the brain was reconstructed based on the same series of CT images because MRI data were not available. The outlines of the brain in the whole set of CT images were therefore defined by shrinking the inner outline of the skull and thus representing reduced intracranial region.

Within the FE model of the head four different regions were defined:

- skull, represented by volumetric tetrahedral elements for the spongional bone covered by inner and outer layer of shell elements for cortical bone of uniform thickness of 1 mm. Spongional bone was modeled as elasto-plastic, with following material properties: Young's modulus of elasticity E_{spon}=2.200 MPa, Poisson ratio ν=0.01, material density ρ=1500$\frac{kg}{m^3}$, ultimate strength in compression σ^-_{ult}=32 MPa, ultimate strength in tension σ^+_{ult}=30 MPa. Cortical bone was considered elasto-plastic as well with following material properties: E=12.000 MPa, ρ=1850 $\frac{kg}{m^3}$, ν=0.21, σ^-_{ult}=80 MPa, σ^+_{ult}=140 MPa

- brain, modeled as viscoelastic material, no differentiation between material properties of white and gray matter was considered. Bulk modulus was set to K=2200 MPa, density of the brain tissue is close to that of water ρ=1000$\frac{kg}{m^3}$, instantaneous shear modulus G_0=1.036 kPa, and the shear modulus at infinity G_∞=0.0185 kPa, reciprocal decay coefficient $\frac{1}{\beta}$=0.0165$\frac{m}{s}$.

- subarachnoidal space which is filled with the cerebrospinal fluid (CSF) is the main shock absorber and is composed mainly from water (99%). In this study, the subarachnoidal space is modeled with bulk modulus K=0.105 MPa, shear modulus G=1.086 MPa, density ρ=1130$\frac{kg}{m^3}$ and Poisson ratio ν=0.495.

The situation modeled was set according the accident: the boy's head touching the ground and the cage cross-bar falling on the left part of the head, see Fig. 1. The playground is covered with the CONIPUR material, which is an impact absorbing, permeable layer of polyurethane, often used for playground surfaces and for these purposes it is approved by Swiss Sport Institute and International Knowledge of Sport Surfaces Association. The surface was modeled using three layers of solid elements with elasto–plastic material properties with following constants: Young's modulus 4209 MPa, yield strength 132 MPa, density ρ=1050$\frac{kg}{m^3}$, Poisson's ratio ν=0.41.

The cage was modeled as a bar of the same cross-sectional properties as the real one, but only the upper bar was modeled. The density of the bar was scaled as to represent the overall force exerted to the head in the moment of contact. The cage is made of zinc-coated steel with following material properties (elasto–plastic material): Young's modulus 195 GPa, yield strength 230 MPa, density ρ=8030 $\frac{kg}{m^3}$, Poisson's ratio ν=0.3.

3 Results

Finite element reconstruction of a sport accident using a geometrically accurate model of head is presented in the paper. The FE model was subjected to initial conditions assessed using a rigid model of the boy falling freely to the surface of the playground and FE model of the upper bar of the falling cage. As an illustrative example of the results the fields von Mises stress in the brain tissue (Fig. 4) is presented. Remarkable observation in the study is the fact, that the skull was fractured at the side of the impact with the cage only, whereas on the opposite side, at the contact with ground it remained intact. Also brain injuries on the side opposite to cage impact were not so severe as on the opposite side. Obvious explanation of this phenomena is that the playground was covered with 16 mm layer of cushion material (CONIPUR 2S) absorbing much of the deformation energy of the reverse side.

Prediction of skull fracture was based on Yoganandan et al. [8] experimental results where the force necessary to fracture cadaver skulls ranged between 8.8 kN and 14.1 kN, with an average of 11.9 kN. The authors also concluded, that for the fixed head the force–deflection

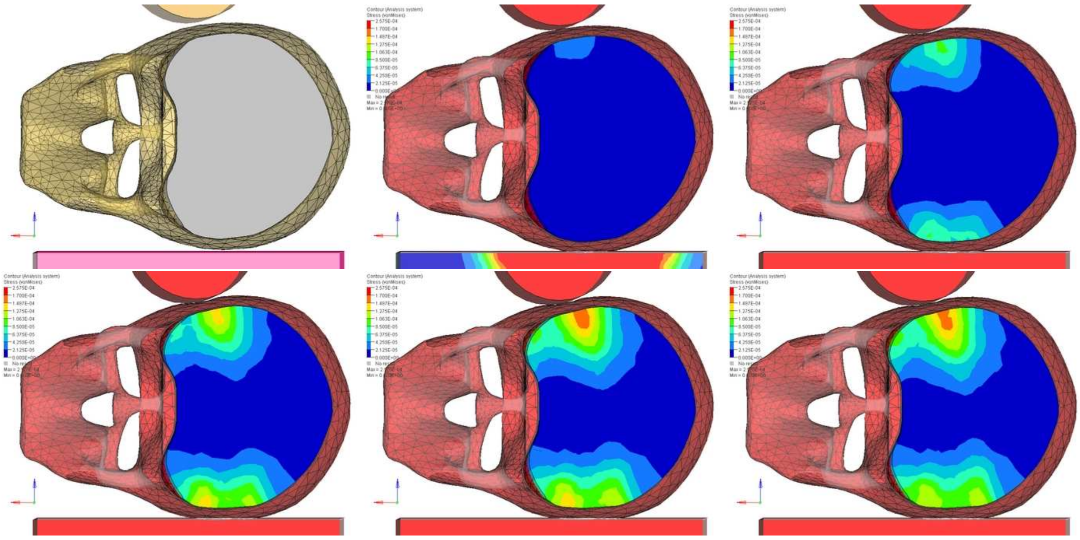

Figure 4: von Mises stress in the brain tissue in 0, 2, 4, 6, 10 ms time intervals

curve was found to be insensitive to impact location. The peak force from the MADYMO simulation was 7.4 kN indicating no skull fracture. In this case, the falling cage is not modeled and the peak force is obtained from the free fall only. This indicates that in case of free fall no skull fracture would occur. On the other hand, results from the finite element modeling show peak force more than double of that value clearly predicting skull fracture.

Results from the FE modeling were used for tissue thresholds. Most of the thresholds are used for axonal injury prediction rather than whether a particular type of injury would occur ([9]). In recent years few works with injury thresholds based on von Mises stress appeared, particularly the work of Willinger et al. [10] and Baumgartner [11].

In this work the injury limits were set according to recent work of Baumgartner and Willinger [12]. The thresholds were derived from a FE modeling of 64 accident involving helmeted motorcyclists, American footballers and pedestrians. The limits were set to 20 kPa for concussion, and 40 kPa for severe brain neurological lesions. The limit for subdural and subarachnoidal haematoma sets the global strain energy of the subarachnoidal space to 5 J. A global strain energy of the skull of 2 J leads to skull fractures.

Using presented FE model von Mises stress was evaluated at the side of impacting cage, at the opposite side (temporal regions) as well as at the occipital and parietal and midbrain regions. In the temporal regions the peak values of von Mises stress were 47 kPa and 23 kPa clearly predicting brain lesions and haematomas on both sides. Overall, the results from the numerical analysis were encouraging and showed good ability of the FE model to represent the impact situation studied and to investigate the brain injury mechanisms.

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