Finite element modeling of the human head under baton impact

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ABSTRACT

Purpose: This research will try to predict damage probability and calculate the main stress resulted from baton impacts by finite element (FE) modeling of the human head considering skull texture, brain and cerebrospinal fluid.

Materials and Methods: A three dimensional FE model of the skull-brain complex was constructed for simulating the baton impact. The FE analysis was carried out using ANSYS program with a nonlinear transient dynamic procedure and the Euler-Lagrangian coupling method. The data used this study were taken from the literature, mentioned in Tables 2 and 3.

Results: Different results were carried out with different values of the bulk modulus and the short-term shear modulus (G₀) for the cerebrospinal fluid and brain material. Considering the values from Figure 8, it was found that the short term shear modulus of the neural tissue had the biggest effect on intracranial frontal pressure and on the model’s Von-Mises response. A comparison between different mesh densities showed that a coarsely meshed model is adequate for investigating the pressure response of the model, while a finer mesh is more appropriate for detailed investigations.

Conclusion: Because of the complexity of this phenomenon, in spite of its importance, there is a little understanding of how baton impact affects the human head. In this paper, the model was validated against a series of cadaveric impact tests. We can conclude that a well validated FE modeling is a powerful tool for investigating the physical process of simulating head trauma.

Keywords: impact force; traumatic brain injury; finite element analysis; Von Mises stress; viscoelastic material.

INTRODUCTION

Head injuries due to direct impact are major sources of death and disability as the result of transportation collisions, falls, assaults, military and sport accidents. Today, because of economic and political crises, there are several reports of people protesting in different countries which in many cases it leads to clashes of security forces with the protesters. Brain injuries due to baton impact are one of the most common medical emergencies that can cause death or health problems for the people in future. So in order to gain a better understanding of head injury mechanisms, it is useful to study the dynamic response of the head-brain complex as a mechanical system subject to impact loads (Figure 1).

Subdural hematomas and diffuse axonal injuries are more lethal than most other brain lesions.¹ This is of special interest in deriving injury criteria for them. Gennarelli and colleagues² have suggested that subdural hematoma is produced in short duration and high amplitude of angular accelerations, while diffuse axonal injury is produced in a longer duration and low amplitude of coronal accelerations. A threshold for diffuse axonal injuries has been proposed which accounts for rotational impulses in the coronal plane (Figure 2).³

Head Injury Criteria

Generally, the head injury criterion (HIC) is used when evaluating the consequences of an impact to the head.
HIC is exclusively based on the resultant, translational acceleration of the head. The basis underlying HIC was first introduced as a curve fit to the Wayne State Tolerance Curve (WSTC). The basic finding described by the WSTC was that high acceleration can be withstood for short durations, while lower accelerations can be tolerated for longer intervals. Moreover, Ueno and Melvin and DiMasi and colleagues found out that the use of either translation or rotation alone may underestimate the severity of an injury.

Recently, Zhang and colleagues and Ma and colleagues concluded that both linear and angular accelerations are significant causes of mild traumatic brain injuries. The generalized acceleration model for brain injury threshold (GAMBIT) was an early effort to combine thresholds for translational and rotational kinematics. Since no dependency of the impulse duration is included, the GAMBIT can be seen as a peak-acceleration criterion for a combined rotational and translational impulse. Lately, a new global kinematic-based head injury criterion, called the head impact power (HIP), was presented. The HIC and HIP are calculated according to the below formulas:

\[
HIC = \left( \frac{1}{(t_2 - t_1)} \right) \int_{t_1}^{t_2} a(t)^{2.5} dt 
\]

In this formula \(a(t)\) denotes the translational head acceleration in g’s as a function of time, and \(t_1\) and \(t_2\) represent the initial and final times of an interval that maximizes this function. In the 2000 revision, maximum critical time reduced 36 ms (HIC > 1000→serious brain injury) to 15 ms (HIC > 700→serious brain injury). Empirically determined relationships between HIC scores and the probability of head injury are widely used in the automotive industry to estimate the risk of injury. Some have compared abbreviated injury scale (AIS) scores for real life injuries to HIC scores or other indices of injury calculated from the reconstruction. HIC and tolerance levels have been explained in Table 1.

### Table 1. Levels of consciousness in relation to head injury criteria

<table>
<thead>
<tr>
<th>Head Injury Criteria</th>
<th>AIS Code</th>
<th>Level of Brain Concussion and Head Injury</th>
</tr>
</thead>
<tbody>
<tr>
<td>135 – 519</td>
<td>1</td>
<td>Headache or dizziness</td>
</tr>
<tr>
<td>520 – 899</td>
<td>2</td>
<td>Unconscious less than 1 hour – linear fracture</td>
</tr>
<tr>
<td>900 – 1254</td>
<td>3</td>
<td>Unconscious 1 – 6 hours – depressed fracture</td>
</tr>
<tr>
<td>1255 – 1574</td>
<td>4</td>
<td>Unconscious 6 – 24 hours – open fracture</td>
</tr>
<tr>
<td>1575 – 1859</td>
<td>5</td>
<td>Unconscious greater than 25 hours – large hematoma</td>
</tr>
<tr>
<td>&gt; 1860</td>
<td>6</td>
<td>Non-survivable</td>
</tr>
</tbody>
</table>

Key: AIS, abbreviated injury scale.
HIP = \int a_x dt + m a_y \int a_y dt + m a_z \int a_z dt + I_{xx} \alpha_x \int a_x dt
+ I_{yy} \alpha_y \int a_y dt + I_{zz} \alpha_z \int a_z dt
\int a_x dt + l_{xy} \alpha_y \int a_y dt + l_{xz} \alpha_z \int a_z dt

The x-axis was defined along the posterioranterior (PA) direction, the y-axis along the lateral-direction, and the z-axis in the inferiosuperior (IS) direction (Figure 3). The following values were calculated for the model: \( m \approx 4.37 \text{ kg}, I_{xx} \approx 0.0213 \text{ kgm}^2, I_{yy} \approx 0.0275 \text{ kgm}^2, I_{zz} \approx 0.0204 \text{ kgm}^2 \). These values are in the range of reported ones by Backer\(^1\) and Walker and colleagues.\(^2\)

\[ HIP = 4.50a_x \int a_x dt + 4.50a_y \int a_y dt + 4.50a_z \int a_z dt + 0.016a_x \int a_x dt + 0.024a_y \int a_y dt + 0.022a_z \int a_z dt \]

In this formula \( a_i \) is linear acceleration at the head’s center of gravity about anatomical coordinate axis \( i \) (i=x,y,z) (m/s\(^2\)) and \( \alpha_i \) is rotation acceleration about axis \( i \) (rad/s\(^2\)).

Hence, this research will try to predict damage probability and calculate the main stress resulted from baton impacts by finite element (FE) modeling of the human head considering skull texture, brain and cerebrospinal fluid.

**MATERIALS AND METHODS**

The human head consists of three main layers surrounding and protecting the brain (Figure 4). The dura and arachnoid mater are separated by a space called subdural space, while the arachnoid and pia mater are separated by another space called the subarachnoid space. In this subarachnoid space, there is string-like tissue called arachnoid trabeculae, which connects arachnoid to pia mater. Within the subarachnoid space, there is also water-like fluid called cerebrospinal fluid, which provides damping and cushions for the brain under impact situations.

**Table 2.** Elastic material properties of the skull and CSF layer.\(^2\)

<table>
<thead>
<tr>
<th>Tissue</th>
<th>( \rho ) (kg/m(^3))</th>
<th>( E ) (Pa)</th>
<th>( \nu )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skull</td>
<td>2070</td>
<td>6.5E + 9</td>
<td>0.20</td>
</tr>
<tr>
<td>CSF</td>
<td>1004</td>
<td>1.5E + 5</td>
<td>0.5</td>
</tr>
</tbody>
</table>

Keys: CSF, cerebrospinal fluid; Pa, Pascal; \( \rho \), density; \( E \), Young’s modulus; \( \nu \), Poisson’s ratio.
Various numbers of head components have been modeled by different authors. The head is modeled with Lagrangian elements and directly from the design (CAD 3D) drawings (Figure 5). The data used in this study are taken from the literature, mentioned in Tables 2 and 3.

In this study, a material model was used which assumes linear viscoelastic, isotropic behavior for both grey and white matter. The standard linear solid model was applied to characterize the shear behavior, and the shear relaxation modulus was described by the following formula:

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

In this formula, \( G_0 \) is the short-term shear modulus, \( G_\infty \) is the long-term shear modulus, and \( \beta \) is decay constant. The material parameters used were the same as those in Zhang and colleagues.\(^7\) The interface between all these tissue types were modeled as tied contact. The bottom of the neck was constrained in all six degrees of freedom to avoid rigid body motion.

To verify the FE model, the numerical results were compared with those results of the cadaver experiment by Nahum and colleagues.\(^19\) The impact direction was along the specimen’s mid-sagittal plane, and the head

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**Table 3. Viscoelastic material properties of the brain.\(^20\)**

<table>
<thead>
<tr>
<th>Tissue</th>
<th>( \rho ) (kg/m(^3))</th>
<th>( K ) (Pa)</th>
<th>( G_0 ) (Pa)</th>
<th>( G_\infty ) (Pa)</th>
<th>( \beta ) (s(^{-1}))</th>
</tr>
</thead>
<tbody>
<tr>
<td>Grey matter</td>
<td>power</td>
<td>2.19E+9</td>
<td>3.4E+4</td>
<td>6.4E+3</td>
<td>400</td>
</tr>
<tr>
<td>White matter</td>
<td>1040</td>
<td>2.19E+9</td>
<td>4.1E+4</td>
<td>7.8E+3</td>
<td>400</td>
</tr>
</tbody>
</table>

Keys: \( \rho \), density; \( K \), bulk modulus; \( G_0 \), short-term shear modulus; \( G_\infty \), long-term shear modulus; Pa, Pascal; \( \beta \), decay constant.

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**Figure 5.** Finite Element (FE) model of the human head.

**Figure 6.** Experiment by Nahum and colleagues.\(^19\) (a) The 5 kg iron impactor impacted the frontal region of the head at 6 m/s, and intracranial pressures were measured in the frontal (point A), parietal (point B) and occipital (point C) region of the head. (b) The input force curve obtained from the experiment.
was rotated forward such that the Frankfort anatomical plane was inclined 45° from the horizontal plane. The outline of the experiment is shown in Figure 6. In the experiment, a 5 kg iron impactor was impacted to the head at 6 m/s.

RESULTS

In Figure 6 a marked difference in the shear stress can be observed between two boundary conditions. The model with fixed boundary at the head/neck junction predicts larger shear stress in the brain throughout the duration. This difference in shear stress magnitude was attributed to the fact that fixed boundary causes rotational motion of the head model while the free boundary leads to nearly translational motion.

Figure 7 shows the pressure and Von Mises stress envelopes for the head impact simulation. In the impact simulation, the maximum pressure of 20.98 MPa was experienced by the skull and gray matter, and the muscle, cerebrospinal fluid, skin/fat and air sinuses experienced peak pressures of 20.66 MPa, 16.22 MPa, 15.37 MPa and 12.4 MPa respectively, all at 8.07 ms.

A parametric study was performed to investigate the effect of different mesh densities on the models, the use of a composite shell element skull and the influence of material properties.

Different results have been carried out with different values of the bulk modulus and the short term shear modulus (G₀) for the cerebrospinal fluid and brain material. Considering the values from Figure 8, it was found that the short term shear modulus of the neural tissue had the biggest effect on intracranial frontal pressure and on the model’s Von-Mises response. A comparison between different mesh densities showed that a coarsely meshed model is adequate for investigating the pressure response of the model, while a finer mesh is more appropriate for detailed investigations.

Figure 9 illustrates the linearly distributed pressures in the skull at t = 2.5, 5, 7.5 and 10 ms for the models with fixed boundary conditions. The linear pressure gradients along the impact direction may be considered as a result of a quasi-static event rather than a dynamic one. However, the maximum speed of shear wave propagation in the brain was around 6.3ms, which was three orders of magnitude less than the speed of pressure waves, so that the minimum transit time of shear waves across the
brain was approximately 22 ms. When the shear wave transit time was compared to the impact pulse lasting 9 ms with a rise time of 2.5 ms, the wave effects became important.

**DISCUSSION**

Many FE head models with various degrees of simplification have been developed in the past several decades. There are, however, two common problems with
the existing models. Firstly, the mesh generation method is often time-consuming, and the generated mesh is unable to represent the important geometric characteristics of the complex human head. Secondly, the existing models are validated using either intracranial pressure or deformation measured by cadaver experiments. However, the extent to which the experimental results may be applied to living human brains is uncertain due to discrepancies between material properties of in vivo and cadaver brains. Simulating identical impact scenarios with a range of different FE models, it has made possible to investigate the influence of model geometries. We can conclude that careful modeling of the cerebrospinal fluid (depth/volume) and skull thickness is necessary if the correct intracranial pressure distribution is to be predicted. In this paper, the model was validated against the cadaver impacts performed by Nahum and colleagues. The model was also scaled and warped in order to represent the same-sized head as that used in their experiment. To obtain a reliable prediction of a mechanical response from any FE model, the constitutive models used for the involved materials have to be chosen carefully.

The patterns of Von Mises stress and Von Mises strain predictions with the FE head model using different constitutive models for brain tissue are compared in Figure 8. Stress and strain predictions obtained with the different constitutive models show similar patterns. However, the magnitudes of stress and strain concentrations differ. During simulations with translational acceleration, the highest deformations in all cases were caused by the brain being obstructed by relatively stiff parts of the dura mater, the falx cerebri or the tentorium cerebelli. For all models, during posterior anterior linear acceleration (Figure 3), the maximum deformations were observed in the posterior and inferior side of the region corresponding to the occipital lobe of the brain at approximately 5 ms. For all models during posterior anterior angular acceleration (Figure 3), the maximum deformations were observed in the anterior, superior and inferior side of the region corresponding to the frontal lobe and anterior side of the region corresponding to the temporal lobe of the brain at approximately 9 ms. For lateral rotation, the maximum deformations were observed in the posterior side of the corpus callosum of the brain which is caused by the falx cerebri at 10-13 ms. The second major deformation area was found to be the inferior side of the region corresponding to the frontal lobe at the same time. For simplicity, only maximum values obtained during the first 15 ms of the simulation are shown in Figure 7.

In contrary, the variation in the Von Mises stress and Von Mises strain response between the simplified and the non-linear version of the constitutive model was found to accurately describe the non-linear response of brain tissue. In both cases the shear and compression were found to be only up to 17% and 40%, respectively. These differences were not dependent on the applied acceleration level in both translation and rotation. Therefore, the simplified version of the recently developed model could be used instead of the non-linear model to obtain reliable injury predictions with FE simulations, since the response can easily be scaled according to the constitutive model used.

Choosing a different constitutive model for brain tissue to be used in a FE model can have large consequences, depending on the presence of nonlinearities in the model. However, in the case of a simplified and non-linear version of a model that has been shown to match the non-linear response of brain tissue, the response predicted with a numerical head model for different conditions (i.e. severity and type of loading) varies consistently with the constitutive behavior used. Therefore, still a reliable assessment of injury can be made with the less accurate simplified constitutive model by using a model-specific injury criterion that is not a true threshold for injury of brain tissue.

Regarding the influence of inertial forces on all the degrees of freedom of the human head, this study shows the following:

1. The results obtained by the FE model correlated with previous clinical and animal studies.
2. HIC was unable to predict consequences of a pure rotational impulse while HIP needs individual scaling coefficients for the different terms to account for difference in load direction.
3. Three additional components, implemented as Heaviside’s step functions, were added to the original HIP in order to address the differences in response between the opposite directions.
4. When using the proposed scaling procedure, a better prediction of subdural hematoma was obtained.

**CONCLUSION**

From an engineering point of view, the final objective of brain injury research is to provide a predictive tool such as an FE model that can aid in injury diagnosis and design of protective devices. A well validated FE model can be such a powerful tool, especially when brain injury experiments are difficult and expensive to carry out. It should be emphasized that the scaling procedures and coefficients are proposed estimations using a parameterized and detailed FE model of the
human head. Although the results give insight into directional sensitivity of impacts to the human head, further experimental validation of intracranial responses for the model in response to higher rotational loads need to be performed before new HIC can be suggested.

CONFLICTS OF INTEREST
None Declared.

REFERENCES