Correlation of an FE Model of the Human Head with Local Brain Motion – Consequences for Injury Prediction

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ABSTRACT – A parameterized, or scalable, finite element (FE) model of the human head was developed and validated against the available cadaver experiment data for three impact directions (frontal, occipital and lateral). The brain material properties were modeled using a hyperelastic and viscoelastic constitutive law. The interface between the skull and the brain was modeled in three different ways ranging from purely tied (no-slip) to sliding (free-slip). Two sliding contact definitions were compared with the tied condition. Also, three different stiffness parameters, encompassing the range of published brain tissue properties, were tested. The model using the tied contact definition correlated well with the experimental results for the coup and contrecoup pressures in a frontal impact while the sliding interface models did not. Relative motion between the skull and the brain in low-severity impacts appears to be relatively insensitive to the contact definitions. It is shown that a range of shear stiffness properties for the brain can be used to model the pressure experiments, while relative motion is a more complex measure that is highly sensitive to the brain tissue properties. Smaller relative motion between the brain and skull results from lateral impact than from a frontal or occipital blow for both the experiments and FE simulations. The material properties of brain tissue are important to the characteristics of relative brain-skull motion. The results suggest that significantly lower values of the shear properties of the human brain than currently used in most three-dimensional (3D) FE models today are needed to predict the localized brain response of an impact to the human head.

KEYWORDS – Finite element (FE) analysis; human head; brain displacement; intracranial pressure; brain material properties; brain-skull interface.

INTRODUCTION

A significant number of road accidents influence the central nervous system in a devastating way. Mechanical input to the nervous tissue initiates a cascade of biochemical processes, and often results in severe injuries with poor prognosis. There has been increased interest for the use of FE modeling for the human head during the last decade (Bandak and Eppinger, 1994; Zhou et al., 1995; Willinger et al., 1995; Ruan et al., 1997; Claessens et al., 1997; Zhang et al., 2001). However, the FE models of the human head today are often average models (such as 50th percentile male), and have fixed mesh density. The geometry of these models varies in detail. Of particular importance to a given model of the human head are the brain-skull interface characteristics, the constitutive properties used for various structures, and level of validation that has been achieved. Because motion between the brain and skull during head impact has been considered potentially important to head injury for more than fifty years (Pudenz and Shelden, 1946; Gurdjian et al., 1968), a primary concern in FE modeling of the human head has been the interface between the brain and skull. Modeling the cerebral spinal fluid (CSF) using linear elastic solid elements with low shear modulus (Ruan et al. 1997; Willinger et al.; 1995; Turquier et al., 1996; Zhou et al., 1995; Zhang et al., 2001) is likely to allow only small relative motion between the brain and skull (Al-Bsharat et al., 1999). Another approach to modeling the brain-skull interface applies contact algorithms between the brain and the dura mater. This contact has been defined in different ways ranging from completely fixed to frictionless sliding (Bandak and Eppinger, 1994; Cheng et al., 1990; Dimasi et al., 1991). Few models include the cranial
aspect of the spinal column together with the important foramen magnum interface.

Kuijpers et al. (1995) used a 2D FE model of a parasagittal cross section of the human head to simulate impacts to the frontal bone of human cadavers (Nahum et al., 1977). Good agreement was found between the simulations and experiments for the coup pressure when using sliding contact conditions, while the model with the coupled (no-slip) interface showed poor agreement. Claessens et al. (1995) developed a 3D FE model of the human head, also using sliding and coupled conditions for the brain-skull interface, and also compared the model results to the experiments of Nahum et al. (1977). In contrast with the work of Kuijpers et al. (1995), the no-slip model showed good agreement with the experiments for the coup pressures, while the sliding interface model did not. However, neither study was able to accurately simulate the experimental contrecoup pressures.

Miller et al. (1998) used a 2D coronal cross section to simulate rotational tests performed in an experimental model of severe diffuse axonal injury (DAI) in the miniature pig. In one version of the model the CSF was represented by linear elastic solid elements with low shear modulus. In a second version a sliding contact algorithm was specified between the dura mater and the brain with a coefficient of friction of 0.2. By comparing principal strain, von Mises stress, and pressure, it was shown that the sliding interface approach was better able to predict the location and distribution of axonal injury and cortical contusions. This study also found the effect of the coefficient of friction (varied between 0.001 and 0.4) in the sliding interface to be small.

Another important issue in modeling of the human head is the selection of material properties for various intracranial structures. The aforementioned three-dimensional (3D) models use linearly elastic or viscoelastic constitutive properties and conventional (displacement-based) finite element formulations that can create severe numerical instabilities when dealing with nearly incompressible materials. The choice of shear properties for the brain tissue is difficult since the span of published values varies several orders of magnitude. Donnelly (1998) reviewed and reported the average values of the shear relaxation modulus for brain tissue. According to this study, the average value of the instantaneous shear relaxation modulus for brain tissue is the order of 1 kPa. Most 3D FE modeling studies have included properties that are around 10-1000 times larger than the average published values. Bandak et al. (1995) used 68 MPa for the linearly elastic brain in a study of a procedure for generating a 3D FE model of the human head from CT images. However, no simulations were performed using this model. The only study using a instantaneous shear modulus near 1 kPa is a 2D study reported by Al-Bsharat et al. (1999). However, the only result reported was that a high level of distortion was produced in the model.

Perhaps of greatest importance to FE modeling of the human head is the level to which a given model has been validated. Most models have been validated against the pressure data of Nahum et al. (1977). However, Bradshaw and Morfey (2001) concluded that it is not acceptable to validate FE models for pressure and then use them for injury prediction. This is apparent since tissue level models (Bain and Meaney, 2000) have shown that diffuse axonal injury (DAI) is a function of strain not pressure. The more relevant parameter for validation of a FE model of the human head should therefore be strain. Such strain data do not exist, but relative displacement data between the brain and skull are available, and provide a means of model validation of localized brain motion that is more complete than pressure alone.

Al-Bsharat et al. (1999) published the first data showing the relative motion between the brain and the skull in the human cadaver. These data were in the form of resultant magnitudes, and were not resolved into components with respect to an anatomical coordinate system. Therefore, the resulting FE model was validated against magnitudes of relative motion between the brain and skull for a few tests.

The first relative motion recorded during human cadaver head impacts in anatomical coordinate components was provided by Hardy et al. (2001). The generalized 3D kinematics of the head was compared to the relative brain-skull motion for translational and rotational input to the frontal and occipital regions within the sagittal plane. A high-speed bi-plane x-ray system was used to image columns of neutral density targets (NDTs) implanted within the cadaver brain. More recently, King et al. (2002) compared relative brain-skull displacements for rotation in the coronal plane to those in the sagittal plane. The results of the cadaver tests show that the local brain motions follow loop or figure eight patterns within the plane of rotation, with peak displacements on the order of ± 5 mm.

Comprehensive correlation between FE model output and relative motion between the human cadaver brain and skull in anatomical X, Y, and Z components has
not been demonstrated previously. Therefore, a goal of this study was to examine the effects of the brain-skull interface modeled by various FE contact definitions and brain-tissue constitutive parameters on the relative brain-skull displacement time histories. Another goal of this study was to compare model results with cadaver experiments conducted for three impact directions: Frontal and occipital (sagittal rotation), and lateral (coronal rotation).

METHODS

A parameterized Finite Element (FE) model of the adult human head was created including the scalp, skull, brain, meninges, cerebrospinal fluid (CSF), and eleven pairs of parasagittal bridging veins (Fig. 1). A simplified neck, including the extension of the brain stem to the spinal cord, dura mater, pia mater, vertebrae, and muscles, was modeled also. In order to better simulate the distribution of stress and strain, separate representations of gray and white matter were included, and distinct ventricles were implemented.

Material Properties

To cope with large elastic deformations, a Mooney-Rivlin hyperelastic constitutive law was used for the CNS tissues. Hyperelasticity or Green elasticity is path-independent and fully reversible, and the stress is derived from a strain energy potential. It can be shown (Malvern, 1969) that the stored strain energy for a hyperelastic material which is isotropic with respect to the initial, unstressed configuration can be written as a function of the principal invariants \( I_1, I_2, I_3 \) of the right Cauchy-Green deformation tensor, i.e.,

\[
W = W(I_1, I_2, I_3)
\]

Mooney and Rivlin showed that the simple form:

\[
W(I_1, I_2) = C_{10}(I_1 - 3) + C_{01}(I_2 - 3) \quad (1)
\]

closely matches results from large deformation experiments on incompressible rubber.

To model the brain tissue as an unconstrained material a hydrostatic work term, \( W_H(J) \), is included in the strain energy functional which is a function of the relative volume, \( J \) (Ogden, 1984):

\[
W(J_1, J_2, J) = C_{10}(J_1 - 3) + C_{01}(J_2 - 3) + W_H(J) \quad (2)
\]

\[
J_1 = I_1J^{-1/3}
\]

\[
J_2 = I_2J^{-2/3}
\]

The stress tensor corresponding to the strain energy density is derived using:

\[
S_{ij} = \frac{1}{2} \left( \frac{\partial W}{\partial E_{ij}} + \frac{\partial W}{\partial E_{ji}} \right) \quad (3)
\]

in terms of the second Piola-Kirchhoff stress, \( S_{ij} \), and Green’s strain tensor, \( E_{ij} \). In addition, rate effects are taken into account through linear viscoelasticity by a convolution integral of the form:

\[
S'_{ij} = \int_{0}^{t} G_{ijkl}(t - \tau) \frac{\partial E_{kl}}{\partial \tau} d\tau \quad (4)
\]

This stress is added to the stress tensor determined from the strain energy functional. The stress relaxation function, \( G_{ijkl} \), is represented by two terms in a prony series, given by:
Based on the previous work by Mendis et al. (1995) and Donnelly and Medige (1997), the stiffness parameters $C_{10}$, $C_{01}$, $G_1$, and $G_2$ were scaled, while the decay constants were not altered (Table 1).

### Table 1 – Mooney-Rivlin, and linear viscoelastic constants used in this study.

<table>
<thead>
<tr>
<th></th>
<th>$C_{10}$</th>
<th>$C_{01}$</th>
<th>$G_1$</th>
<th>$G_2$</th>
<th>$\beta_1$</th>
<th>$\beta_2$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stiff</td>
<td>620.5</td>
<td>689.4</td>
<td>8149</td>
<td>4657</td>
<td>125</td>
<td>6.7</td>
</tr>
<tr>
<td>Average</td>
<td>62</td>
<td>69</td>
<td>814</td>
<td>465</td>
<td>125</td>
<td>6.7</td>
</tr>
<tr>
<td>Compliant</td>
<td>31</td>
<td>35</td>
<td>407</td>
<td>233</td>
<td>125</td>
<td>6.7</td>
</tr>
</tbody>
</table>

Two additional, more compliant models were implemented. One model used properties slightly less than the most compliant data available on brain tissue (those of Prange et al. (2000) hereafter referred to as “compliant”). In an intermediate model (hereafter referred to as “average”), the properties were adjusted to levels near the average values reviewed by Donnelly (1998). These values fell between the compliant and stiff data reported by Mendis et al. (1995). The average relaxation function chosen in this study agrees very well with the stress relaxation results of Arbogast et al. (1997), and with the transformed results of the complex modulus by Arbogast and Margulies (1997) presented by Donnelly, (1998). The shear relaxation moduli can be seen in Fig. 2.

\[
g(t) = \sum_{i=1}^{N} G_i e^{-\beta_i t} \quad (5)
\]

This is effectively a Maxwell fluid that consists of dampers and springs in series. Where $G_i$ represents the shear moduli, and $\beta_i$ are the decay constants. Mendis et al. (1995) derived the rate dependent Mooney-Rivlin constants $C_{10}$ and $C_{01}$ and time decay constants $\beta_i$, using experiments published by Estes and McElhaney (1970) on white matter from the corona radiata region.

Using the relationship $G=2(C_{10}+C_{01})$ for the prony terms gives the following constants: $G_1=8149$ Pa, $\beta_1=125$ 1/s, $G_2=4657$ Pa, and $\beta_2=6.67$ 1/s. The law has been introduced for both white matter (Estes and McElhaney, 1970) and the gray matter (Prange et al., 2000). The Mooney-Rivlin constants for the brain stem were assumed to be 80 % higher than those for the gray matter in the cortex (Arbogast and Margulies, 1997). Although significantly less stiff than those used in previous 3D FE models of the human head, these parameters give stiffness for the brain tissue that is higher than the average published values (Donnelly, 1998). A curve fit of the data presented by Donnelly and Medige (1997) revealed almost identical decay factors and ratios between prony terms and long-term moduli, similar to the constitutive parameters proposed by Mendis et al. (1995).
A selectively reduced (S/R) integration scheme, which uses reduced integration for the volumetric (pressure) terms, and full integration for the deviatoric (shear) terms was used for the brain tissue. The S/R scheme was used to avoid hourglass instabilities associated with reduced integration, and locking phenomenon that is associated with full integration of lower order elements. A summary of the properties used in this study is given in Table 2.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Young's modulus [MPa]</th>
<th>Density [kg/dm³]</th>
<th>Poisson's ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Outer table/Face</td>
<td>15,000</td>
<td>2.00</td>
<td>0.22</td>
</tr>
<tr>
<td>Inner table</td>
<td>15,000</td>
<td>2.00</td>
<td>0.22</td>
</tr>
<tr>
<td>Diploe</td>
<td>1000</td>
<td>1.30</td>
<td>0.24</td>
</tr>
<tr>
<td>Neck bone</td>
<td>1000</td>
<td>1.30</td>
<td>0.24</td>
</tr>
<tr>
<td>Neck muscles</td>
<td>0.1</td>
<td>1.13</td>
<td>0.45</td>
</tr>
<tr>
<td>Brain</td>
<td>Hyperelastic/Viscoelastic</td>
<td>1.04 0.499999994-0.499999997</td>
<td></td>
</tr>
<tr>
<td>Cerebrospinal Fluid</td>
<td>$K=2.1$ GPa</td>
<td>1.00</td>
<td>0.5</td>
</tr>
<tr>
<td>Sinuses</td>
<td>$K=2.1$ GPa</td>
<td>1.00</td>
<td>0.5</td>
</tr>
<tr>
<td>Dura mater</td>
<td>31.5</td>
<td>1.13</td>
<td>0.45</td>
</tr>
<tr>
<td>Falx/Tentorium</td>
<td>31.5</td>
<td>1.13</td>
<td>0.45</td>
</tr>
<tr>
<td>Scalp</td>
<td>16.7</td>
<td>1.13</td>
<td>0.42</td>
</tr>
<tr>
<td>Bridging veins</td>
<td>$EA=1.9$ N</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

$K=$Bulk modulus, and $EA=$load/unit strain.

The Poisson’s ratio was varied (Table 2) to keep the bulk modulus of constant value of 2.1 GPa (Stalnaker, 1969, McElhaney et al., 1976), for all the chosen shear moduli.

This FE model is described in more detail by Kleiven and von Holst, (2001 and 2002a) and comprises nonlinear viscoelastic and incompressible material modeling, experimental validation and parametric studies. This model has been experimentally validated against pressure data, as well as relative motion magnitude data in previous studies.

**Interface Conditions**

Based on the anatomy and physiology of the brain-skull interface, the interface between the dura and the skull was modeled with a tied-node contact definition in LS-Dyna. Because of the presence of CSF between the meningeal membranes and the brain, sliding contact definitions were used for these interfaces. Different sliding contact definitions were compared with a tied contact definition. Three different strategies were tested: A tied interface including the CSF as fluid, a sliding interface that allowed separation, and a sliding interface that did not allow any separation. The sliding interface without separation consisted of an additional layer of pia mater, which was modeled within and close to the outer surface of the cortex and tied to the dura. This contact definition allowed sliding in the tangential direction and transfer of tension and compression in the radial direction. This was done in part because a fluid structure interface is likely to experience a vacuum when a pressure wave reflects at the contrecoup site, or when inertia forces create tension in brain regions opposite impact.

Average CSF thickness of roughly 2 mm was used, which corresponds to approximately 120 ml of subdural and subarachnoidal CSF. For all the sliding interfaces a coefficient of friction of 0.2 was used, as proposed by Miller et al. (1998).

**Intracranial Pressure**

Results from simulations with the FE model were compared with the intracranial pressure-time recordings from experiment No. 37 conducted by Nahum et al. (1977), where impacts to the forehead by a padded impactor were performed. The kinematics of this experiment was applied to the skull. The intracranial pressure-time characteristics in the frontal and occipital regions from this experiment were compared to the ability of the different contact definitions to transfer tensile and compressive loads in the direction normal to the surfaces, as well as the differences in pressure characteristics due to a change in shear stiffness of the brain tissue.

**Relative motion between the brain and skull**

New experimental data has been presented by Hardy et al. (2001) and King et al. (2002), which describes the relative displacement between the brain and skull of the human cadaver. The cadaver experiments
focused on measuring the relative brain-skull motion using a high-speed biplane x-ray system and neutral density targets. The NDTs are polystyrene cylinders that are 3.9 mm long and 2.3 mm in diameter. Centered within the polystyrene tubing are 1.9-mm tin granules. The density of these targets is roughly that of brain tissue. Typically, the NDTs were implanted in two vertical columns located in the occipitoparietal region, and in the temporoparietal region, with spacing between the centers of the NDTs of approximately 10 mm. The cadaver head was inverted and suspended in a fixture that allowed rotation and translation. Each specimen was mounted to the moving fixture at the level of T2. Frontal (Fig. 3, left), occipital, and lateral (Fig. 3, right) impacts toward an angled acrylic surface were conducted. The impact speed ranged from 2.5 to 3.5 m/s. The rigid body motion of the skull was eliminated from the NDT (brain) motion data, leaving the brain-skull relative displacements. The NDT’s numbered “a6” and “p6” were located in the brain toward the apex of the skull, while the “a1” and “p1” were located just above the base of the skull.

Figure 3 – Specimen and test configurations from the cadaver tests showing the NDT implant schemes for a frontal impact involving sagittal rotation (left) and a lateral impact involving coronal rotation (right). The markers used in the parametric studies of interface modeling and shear properties for the brain tissue are highlighted for the frontal impact condition.

Four experiments from the literature were simulated using the model. To increase accuracy, the actual geometry of the specimen (overall length, width and height) was reproduced. The kinematics and HIC values for the impacts are summarized in Table 3.

The first three tests were conducted using one cadaver, and involved rotation in the sagittal plane. Two of these impacts were to the frontal region, and one to the occipital region. The fourth experiment was a lateral impact involving a separate cadaver and coronal rotation. The linear and angular accelerations for a typical frontal impact can be seen in Fig. 4. The initial positions of the neutral density targets used in the experiments were used to locate the targets within the FE model. The targets were modeled as rigid beams. For each simulation, the full kinematics (all six degrees of freedom) of the experiments was applied to the skull. One of the frontal impacts (C383-T1) was used to evaluate the influence of shear properties of the brain tissue and the ability of the different contact definitions to describe the relative motion between the skull and the brain. Since the displacement characteristics in the experiments and simulations were similar for most locations, the
peak magnitudes as well as the temporal differences in peak values were used as an assessment of the correlation. For a few locations (e.g. the second peak in X-displacement of NZT a3, Fig. 6), there were differences in the response characteristics between the simulation and experiment. In such cases the model response values at the time of the experimental peaks were used for comparison, and the temporal differences for these peaks were not included in the analysis.

RESULTS

Intracranial Pressure

The predicted intracranial pressure responses from the FE model varied based on the interfacial condition, with some of the interfacial conditions producing responses that agreed well with previously published intracranial pressure recordings during impact. As seen from Fig. 5, the calculated curves for the frontal and occipital pressures gave magnitudes and characteristics similar to the experimental results for the tied interface, while the sliding interface (with separation) produced coup pressure magnitudes 90% higher than in the experiments. Also, the sliding interface with separation gave a positive pressure in the occipital area due to the lack of tensile resistance which made the inertia forces of the spinal cord and brain stem apply pressure on the cerebellum and occipital pole. The sliding interface without separation provided a way to allow sliding while giving tensile resistance in countercoup areas. This produced negative occipital pressure magnitudes that were 20% lower than the experiments, and that occurred about 0.5-1.0 ms later in time. The frontal pressure magnitudes for the sliding-only interface were 45% higher and occurred approximately 1.0 ms earlier than in the experiments.

Figure 4 – Linear and angular accelerations of a typical frontal impact.

Figure 5 - Simulation of the pressure results from the cadaver experiments by Nahum et al. (1977) as a function of interface properties using a FE model of the human head (left), and as a function of shear properties of the brain tissue (right).
A summary of the peak frontal and occipital pressures in the experiments and simulations is seen in Fig. 6. The magnitudes of the pressures show a low sensitivity for a variation of the stiffness properties of the brain tissue (Fig. 6A). The different contact formulations used for the brain-skull interface result in large variation in the magnitudes of the pressures (Fig. 6B).

**Relative motion between the brain and skull**

The results for the relative displacement of six locations in the occipitoparietal and temporoparietal regions for a 3 m/s frontal impact are shown in Fig. 7, and are used to assess the ability of different material properties of brain tissue to reproduce the experimental relative brain-skull motion. Each plot represents a given NDT location. The curves in the top half of the figure show relative displacement in the X direction, and the bottom curves show relative Z-direction displacement for the same locations. The motion of the markers is typically characterized by a maxima or minima occurring between 20-40 ms before rebounding through the initial position (zero) between 70-90 ms, and then reaching a minima or maxima between 90-105 ms.

It was found that the magnitude of the relative motion in the brain increases with decreasing stiffness of the brain tissue. When comparing the model having the stiffest shear properties for the brain with the experiments, the maxima and minima were underestimated by an average 69 percent for relative displacement in the X direction, and 67 percent in the Z direction. These values reflect the average discrepancy between the model and experimental results for all six target locations examined, and represent an average difference in the prediction of the displacement magnitudes of 68 percent for both the X and Z directions. The opposite was found when using the compliant properties, where the average overestimation of the maxima and minima was 104 percent for relative displacement in the X direction, and 147 percent in the Z direction. When using the average published values of shear properties for the brain tissue, an average difference of 59 percent (72 percent in the X direction and 46 percent in the Z direction) between the model and experimental displacement peaks was found.

An increased delay in the relative motion was seen when using the more compliant properties for the brain, which increased with decreasing stiffness. This was examined by comparing the temporal difference between the experimentally seen peaks in displacement and the results from the simulations using the various stiffness properties. It was found that the displacement peaks in the X direction were occurring (average for the six targets) 16.4 ms earlier in the simulation using the stiffest properties, while they were observed 13.6 ms earlier in the Z direction. When using the average properties, this lead effect was reduced to 7.3 ms for the peaks of the X displacements, and 5.9 ms for the Z displacements. For the model using the most compliant properties the opposite of was true, and a lag of (average for the six targets) 9.7 ms and 9.5 ms compared to the experiments was found for the timing of the peak X displacements and Z displacements, respectively.

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**Figure 6 -** A summary of the pressure magnitudes observed in the experiments and simulations. (A): Pressure magnitudes for a variation of the stiffness properties of the brain tissue. (B): Pressure magnitudes for the different approaches used for the brain-skull interface.
Figure 7 - Simulation of relative motion in the sagittal plane. Results for the model with tied contact interfaces between the brain (pia mater) and the CSF, and varying stiffness parameters of the brain. For marker locations and coordinate system directions, see Fig. 3, left.

The results for six locations in the occipitoparietal, and temporoparietal regions for a 3 m/s frontal impact are shown in Fig. 8, and are used to assess the ability of the different approaches used to model the brain-skull interface to reproduce the experimental relative brain-skull motion.

The comparison between various contact definitions shows a small sensitivity to the interface for relative motion, and insignificant differences were found both for the timing of the displacement peaks as well as for the magnitudes of the peaks in the X direction (Fig. 8). However, differences related to the various contact definitions are more pronounced for relative motion data in the Z direction (Fig. 8). When comparing the magnitudes of the displacements in the Z direction, it was found that the tied interface resulted in peak magnitudes that were an average of 46-percent higher than in the experiments for the six chosen targets. The corresponding value for the sliding interface was 66 percent, and the value for the sliding without separation interface was 65 percent.
A summary of the average values of the peak displacement magnitudes observed in the experiments and simulations is seen in Fig. 9. The difference shown is between the average magnitudes for all six NDT locations. The average magnitudes of the local brain motion show a high sensitivity for a variation of the stiffness properties of the brain tissue (Fig. 9A). The different contact formulations used for the brain-skull interface result in less variation in the average magnitudes of peak displacements (Fig. 9B).

For the second frontal impact (C383-T2) simulation using the average properties for the brain tissue and a tied interface, correlation between the experiment and the model similar to that for the first impact (C383-T1) was found (Fig. 10). The results of both the frontal impacts are in agreement, where an underestimation of the magnitudes of the X displacements and an over prediction of the Z displacement are found for most of the markers. This can be seen when looking at the magnitudes of the motion in the X direction for the posterior-superior markers (p3 and p6), which are smaller (20-100 percent) in the model than in the experiments, while the magnitudes for the motion in the Z-direction for the same markers are larger (30-100 percent) in the model than in the experiments.
Figure 9 - The average values of the peak displacement magnitudes observed in the experiments and simulations. (A): For variation of the stiffness properties of brain tissue. (B): For the different brain-skull interfaces used.

Figure 10 - Simulation of relative motion in the sagittal plane for a frontal impact (C383-T2). For marker locations and coordinate system directions, see Fig. 3, left.
Typical target motions in the sagittal plane for the C383-T2 (Frontal) impact can be seen in Fig. 11. The markers are plotted together with the falx and the tentorium to clarify target orientation and direction of motion, and the initial positions of the markers are superimposed in black. The motion of the superior markers (the lower markers in this figure) is shown to be opposite the direction of the inferior markers for both the experiments and simulations. This is especially apparent at 20 and 90 ms, where the magnitudes of the displacements are near their peak values.

For the occipital impact (C383-T4) simulation using the average properties for the brain tissue and the tied interface, a better correlation between the model and the experimental results is seen for the superior markers when compared to the frontal impacts (Fig. 12). There is 45-percent difference (average for both X and Z directions) in displacement magnitudes.

Figure 11 – Target motion in the sagittal plane for six markers in a frontal impact (C383-T2). The initial positions of the markers are shown in black, while the brain targets are shown in gray.
between the experiment and simulation for the two superior markers (a6 and p6) in the occipital impact, whereas the first frontal impact (C383-T1) exhibits 83-percent difference between the experiment and simulation for the same two target locations. On the other hand, the simulated Z-direction motion of the a1 and a3 markers deviates from the experiments more than 4 mm after 60 ms. Also, for the X-direction motion of the p1 and p3 markers, a 6-7 mm smaller magnitude of the second peak is found for the simulation, while the general shapes of the responses are similar for the simulation and experiment. The prediction of larger X-direction displacement for the anterior-superior target (a6), and the prediction of smaller motion for the inferior targets (a1 and p1) compared to the experiments are effects common to all the frontal and occipital impacts. When simulating the lateral impact, the characteristics similar to those seen in the experiment were found for Y and Z displacements (Fig. 13). The motion of the brain targets relative to the skull shows a characteristic peak around 10-15 ms, which subsequently decays. When comparing the experimental magnitudes to the simulation at this initial peak, it was found that for all the superior markers (aL5, pL5, and pR5), the differences were within 1.7 mm. While the model and experiment showed a discrepancy of 3.3 mm for the initial peak in Z displacement of the left anterior-inferior marker (aL2), the differences were within 0.9 mm for the Z displacement of the remaining targets.

Figure 12 - Simulation of relative motion in the sagittal plane for an occipital impact (C383-T4). For marker locations and coordinate system directions, see Fig. 3, left.
A summary of the average values of the peak displacement magnitudes observed in the experiments and simulations for different impact directions (separated by test) is seen in Fig. 14. The prediction of larger Z-direction displacement and smaller X-direction displacement by the model has been described using the average error between peak displacement results obtained from the model and experiments. This phenomenon is further illustrated by the average values of the peak displacement magnitudes. This is evident for the first frontal impact and the occipital impact. For the second frontal impact there is essentially no average difference between the simulation and experiment in the Z direction. In the lateral-impact case the average values of both the peak Y- and Z-displacement magnitudes are larger in the model than in the experiments. The maximal principal strain (Green St. Venant) in the central parts of the brain showed variation for the different interface conditions and material properties used in this study (Fig. 15). The stiffness properties used for the brain tissue especially affected the strains in the brain. A result similar to that of the comparison with the relative motion experiments is seen. The stiff properties produced 60-80 percent lower peak values of maximal principal strains, which occurred 15-20 ms earlier than when using the average properties. The more compliant properties produced peaks roughly 100-percent higher, which occurred 5-10 ms later (Fig 15, left). Some smaller differences can also be seen when comparing the various contact interfaces, where the sliding interfaces produced stresses located centrally in the brain having peaks 20-30-percent higher in magnitude than the simulations produced using the tied interface (Fig 15, right).

Figure 13 - Simulation of relative motion in the coronal plane for a lateral impact (C291-T1). For marker locations and coordinate system directions, see Fig. 3, right.
Figure 14 - A summary of the average values of the peak displacement magnitudes observed in the experiments and simulations for the various impact directions.

**Comparison of Maximal Principal Strain in the Central parts of the Brain**

Figure 15 – Resulting maximal principal strain at the same element adjacent to the corpus callosum region of the brain for the various stiffness properties used for the brain tissue (left), and for the different contact interfaces between the brain and the skull (right).
DISCUSSION

The local motion of brain tissue described by Hardy et al. (2001) and King et al. (2002) has been simulated for three impact directions (Frontal, occipital and lateral), using a selection of material properties and interface conditions. The results of this effort show that simulation of local brain motion is highly sensitive to the shear properties of the brain tissue. The local brain motion in response to low severity impact seems to be relatively insensitive to the type of brain-skull interface used for simulation as well. The pressure response, on the other hand, seems to be more dependent upon the type of brain-skull interface than on the constitutive parameters chosen for the brain tissue.

Interface Conditions

Fluid elements should be used to simulate the CSF as proposed by Zhou et al. (1995), Bandak and Eppinger (1994), Miller et al. (1998), and Al-Bsharat et al. (1999), in order to adequately represent the effect of the ventricles, subarachnoid space, and brain-skull interface. However, adequate representation of fluid-structure interaction still remains a major limitation of most commercially available FE packages. Due to this problem, two different approaches that do not implement fluid elements for the CSF have been developed to model the brain-skull interface.

The first approach models the subarachnoid CSF using linear elastic solid elements with low shear modulus. This approximation has been used by several researchers (Ruan et al., 1997; Willinger et al., 1995; Turquier et al., 1996; Zhou et al., 1995). An alternate way of modeling the brain-skull interface includes contact algorithms between the brain and the dura mater. The contact has been defined in different ways ranging from completely fixed to frictionless sliding. Several parametric studies have been performed, where the effects of different interface conditions between the brain and skull have been studied (Cheng et al., 1990; Kuijpers et al., 1995; Claessens et al., 1997, and Miller et al., 1998). These studies concluded that the impact response of the human head is sensitive to the modeling of this interface condition. This is in keeping with the results from the present study, where the pressure response is sensitive to the interface conditions.

Localized brain motion during a low severity impact seems relatively insensitive to interface conditions. The sliding (with separation) contact algorithm used by Kuijpers et al. (1995) and Claessens et al. (1997), was found to be insufficient for the brain-membrane interfaces in the contrecoup region, and a gap was created in this region due to limited load transfer to compression only. In our study, a sliding-only contact algorithm, which transfers load in tension was also implemented. Because of this, large relative motion between the brain and skull was allowed and load was supported in tension at the contrecoup region. This resulted in comparable magnitude and shape of the pressure-time responses between the simulations and experiments even in the contrecoup region.

The sliding interface with separation resulted in positive pressure in the occipital area due to the lack of tensile resistance, which made the inertia forces of the spinal cord and brain stem apply pressure on the cerebellum and occipital pole. The looser connection between the brain and skull that is provided by sliding interfaces also makes it possible for the brain to slide forward due to the inertia forces, creating higher pressures frontally when simulating the pressure experiments. Further, this looser connection could explain the temporal differences of the maximal frontal and occipital pressures for the sliding interfaces when comparing the model to the experiments. For the tied interface the opposite is true since the connection does not allow any motion between components in the contact surface giving a more direct impulse transferred to the brain.

The hydrodynamics of a sealed, pressurized cranial cavity dictate that positive pressure cannot develop at the interface between the brain and skull at the contrecoup site. This effect is not reproduced in the model for the sliding interface simulations. In addition, since the model pressures were measured at locations at the surface of the brain as specified by Nahum et al. (1977), which are closer to the interface than the locations of some of the NDTs, it is reasonable that the pressure results should have a greater dependence on interface conditions than some of the NDT displacements.

The simulations showed larger differences in results between the interface conditions for the posterior markers during a frontal impact. An explanation might be that the frontal parts of the brain are naturally constrained by the skull base and anterior middle fossa during frontal impact so that the brain-skull interface would have less influence for the anterior markers. The posterior target located closest to the tentorium and the contrecoup area for a frontal impact (p1), had the greatest sensitivity to the interface conditions. This is likely due to the influence of both the looser connection for the sliding
interfaces, as well as the influence of the brain-tentorium interface. Relative motion between the brain and skull is a more complex response that requires unambiguous material characteristics. However, the relative motion response was found to be relatively insensitive to the interface conditions, since very small differences in magnitude and characteristics of the localized brain motion were seen (Fig. 6) for the different interfaces. This could be related to the relative motion primarily resulting from local distortions of brain tissue with little sliding occurring at the brain-skull interface.

The experimental data do not indicate whether there is, or there is not, sliding between the brain and skull. The model, however, showed a small relative motion (less than 1 mm) at the skull brain interface for the relatively low severity impacts used for the relative motion experiments. It has been shown (Kleiven and von Holst, 2001, 2002b), that when applying higher magnitudes of rotational accelerations, the motion close to the skull increases to several mm in magnitude. A larger difference should therefore be expected between the various contact definitions ability to describe localized brain motion for rotational inputs of higher magnitude.

**Brain Shear Stiffness Dependency**

A range of brain material properties can be used to effectively model the intracranial pressure responses obtained from cadaver experiments. It is possible to predict the pressure data of Nahum et al. (1977) using material properties for the brain 2000 times stiffer than the average published values (in keeping with levels used by many other modeling efforts). From the results (Fig. 5, right) it can be seen that the pressure responses for the models using stiff and compliant stiffness parameters for the brain tissue are virtually indistinguishable, although the parameters were varied more than one order of magnitude. This is in contrast to the aforementioned sensitivity of the pressure response to the type of contact interface, where the tied interface showed better correlation with the pressure experiments as compared to the sliding interfaces.

Relative motion between the brain and skull is very sensitive to the choice of stiffness for the brain tissue. When using the parameters reported by Mendis et al. (1995), significantly smaller relative motion than that found in the experiments was observed. Since this constitutive law is more compliant than what is used in most other 3D FE head models, it is tenuous to suggest that significantly less stiff parameters should be used in future FE modeling efforts. Two proxy terms with a fast decay and a slow decay in the viscoelastic terms of the model were used, and are probably necessary to simulate these relative motion experiments. This is apparent since a large-magnitude impulse (a couple of ms in duration) is followed by a long duration motion (150ms). It can only be conjectured as to what extent additional terms might improve the correlation.

A selectively reduced (S/R) integration scheme was used to avoid hourglass instabilities associated with reduced integration. This was necessary to stabilize the elements representing the brain when using the compliant brain tissue properties since the large-magnitude initial rotational impulse followed by long duration motion resulted in unacceptably high artificial (hourglass) energies when utilizing reduced integration. To the authors’ knowledge, brain properties roughly half the stiffness of the average published values (around those of Prange et al., 2000) have never been successfully implemented in a simulation before. However, by using such compliant properties a significant increase in relative motion as well as the strain in the brain appeared. The characteristics of the response changed as well, producing a “delay” in the local brain tissue motion.

**Brain Motion Considerations**

There is symmetry in the motion of the superior and inferior markers for both the model and the experiments for both sagittal, and coronal rotation. This might be explained by the nearly incompressible properties of brain tissue and skull geometry, and the associated response to rotation with little response to linear input. The model shows larger relative motion in the X-direction for superior markers, while the opposite is true for the inferior markers. For the central markers (between 1 and 6), the similar magnitudes are observed, while the characteristics are slightly different for markers located in the anterior column. All markers tend to go back to the initial position for both simulations and experiments. The relative influences of the characteristics of the superior regions of the model on the motions in the inferior regions are not known. However, discrepancies between the simulations and the experiments in one region will likely result in discrepancies in the other.

Both the model and the experiments show similar magnitudes of relative motion between the skull and the brain for the occipital and frontal impact scenarios. The lateral impact resulted in smaller relative motion than the frontal impact of similar severity. This is true for the cadaver experiments and
the simulations. This is mainly thought to be due to the supporting structure of the falx cerebri. Gennarelli et al. (1982, 1987) produced traumatic coma in monkeys by accelerating the head without impact in various non-centroidal rotation scenarios. It was found that the majority of the animals that were subjected to coronal rotation suffered more prolonged coma. Also, all the laterally impacted animals had a degree of DAI in the corpus callosum and superior colliculus similar to that found in severe human head injury. The present study supports the findings of Gennarelli et al. (1982, 1987): Smaller relative motion between the brain and skull suggests the influence of the falk, which may impinge upon adjacent structures such as the corpus callosum, potentially causing injury. This is also supported by the findings of higher shear stresses in the corpus callosum for coronal rotation compared to sagittal rotation when imposing a sinusoidal acceleration pulse corresponding to the same head impact power (HIP), which has been reported previously by Kleiven and von Holst (2002b).

The maximal principal strain (Green St. Venant) in the central parts of the brain varies depending upon the material properties used in this study (Fig. 15). The stiffness properties for the brain tissue especially affect the strains in the brain. The stiff properties produced 60-80-percent lower peak values of maximal principal strain, which occurred 15-20 ms earlier in time than when using the average properties. The compliant properties produced approximately 100-percent higher peaks, which occurred 5-10 ms later in time. It can also be noted that when using the stiff properties the peaks in maximal principal strains occurred close to the peaks in the acceleration pulses (Fig. 4). When using the average and compliant properties, the strain peaks occurred 20-30 ms, and 30-40 ms after the acceleration peaks, respectively. These differences are crucial for a FE model of the human head designed to predict Traumatic Brain Injury (TBI), and reinforces the need for comparison with the relative motion experiments. The importance of intracranial motion differences due to shear properties of the brain becomes apparent when looking at published tissue thresholds for DAI. Bain and Meane (2000) estimated a tissue threshold for axonal damage to a Lagrangian principal strain of about 0.2 in experiments on optic nerves of guinea pigs. Using this value, the potential for DAI is predicted for the relative motion experiments using the compliant and average properties, while the stiff properties predict values far below 0.2. When using the average properties, principal strain just above this level is found for the central parts of the brain.

The simulations of this study were compared to previously reported experimental local brain motion data. The tests showed that the local brain motions followed loop or figure eight patterns within the plane of rotation. The peak relative displacements between the brain and skull were on the order of ± 5 mm. The general path of the loop or figure eight patterns were described as having a major displacement axis (MDA) within the plane of rotation, similar to the description for an ellipse. The perpendicular bisectors from each MDA were said to intersect at an average instant center (AIC) of rotation, which is a common point about which the movements of the brain at each target location for a given column of targets seem to be organized. The MDA orientations also rotated from inferior to superior. Motion for sagittal and coronal plane rotation was remarkably similar, as was the angular speed of each test. The brain motions exhibited interesting anterior-posterior and right-left symmetry. In general, the motion of the brain seemed to lag that of the skull for rotational input to the skull, while the brain moved little (less than 1 mm) for linear input to the skull. These phenomena are in keeping with the near incompressible nature of the intracranial contents. The model developed for this study exhibited many of the same trends of the experiments, as shown in the component-wise comparisons between the simulations and experiments (Fig. 6 through Fig. 11).

Limitations

The intracranial pressure measurements of Nahum et al. (1977) have been simulated by Kleiven and von Holst (2002). In that study the impact was simulated by modeling impactor padding, using elastic models for the skull and the scalp, and imposing the given impactor velocity. In this study, the kinematics of the head was imposed to avoid errors in the transferred impulse. However, to impose the kinematics of an impulse, the skull must be assumed rigid. In this case, the experimental impacts were padded so deflection of the skull should not have a great influence. Nevertheless, the initial pressure response is smaller and smoother in the experiments, which possibly could be explained by the presence of the padding and small local skull deflection. There are obvious differences between the experimental pressure data and the simulations, especially in the occipital region. However, since the differences between the right and left pressures (Occipital #1, and Occipital #2) reported in Nahum et al. (1977) are greater than the differences between experiments and simulations (Fig. 16), the differences between the experiments and simulations are considered insignificant.
In the model, the effect of baseline pressurization was not included. Baseline pressure results from atmospheric pressure and repressurization. When gage pressure is observed under such conditions, the contrecoup pressure will decrease from zero as compression of the brain decreases during impact. This occurs irrespective of the interface type. Some discrepancies between the simulation and experimental pressure results could result from the absence of baseline pressure in the model. However, it should be noted that the baseline pressure used in the relative motion experiments was small, being roughly 10 kPa.

In the cadaver experiments, the greatest specimen-preparation concerns were related to evacuation of gasses from the intracranial space to maintain coupling between the brain and the skull. If small air bubbles were trapped beneath the tentorium, the larger motion in the caudal cerebrum seen in the experiments as compared to the simulations may be explained in part. However, the high-speed x-ray images provided no indication of air within the intracranial contents. Parametric analyses conducted by introducing small levels of air (amounts that should be visible on x-ray) within the model showed minor differences in output as compared to the air-free simulations. These two factors combine to suggest that intracranial air had small influence on the experimental results, and therefore is a minimal factor in the discrepancies between the simulations and experiments.

Other limitations of this numerical study are the assumptions of isotropy, homogeneous properties, and lack of details in the boundaries between the ventricles and brain tissue. Also, the kinematics is applied to the skull, and the simplified neck is forced to follow this motion without accounting for the relative motion of the vertebrae in the neck. This should, however, have a limited effect on the localized motion of brain tissue since the targets are located on a distance from the spinal cord.

Bradshaw and Morfey (2001) showed that the intracranial pressure response in the experiments by Nahum et al. (1977) is a hydrostatic problem, controlled only by the density of the cerebrum. The impulse used in the intracranial pressure experiments by Nahum et al. (1977) is long (6-10 ms) compared to the transit time of dilatational waves within the brain (approximately 0.1 ms), so static equilibrium prevails (only inertia forces are acting) and wave effects are of no consequence (Graff, 1975). Therefore, intracranial pressure should be independent of the bulk modulus and shear modulus. This is supported by this study and the results of Thomas et al. (1967). In the presence of local skull deformation, the bulk modulus would play a larger role. However, the local skull deformation was insignificant in the relative motion experiments, and was not simulated in the model. High accuracy is required for relative displacement validation since strain is always a function of the spatial derivatives of the displacements. Since Bain and Meaney (2000) suggest that axonal injury is related to distortional
strain, validation against local displacement of brain tissue is particularly relevant for an FE model of the human head that is developed for traumatic brain injury prediction.

CONCLUSION

A parameterized finite element (FE) model of the human head has been developed. The parameterized nature of the model allows the user to adjust the geometry of the model to fit that of a particular specimen, reducing some of the concerns associated with scaling or the use of generic models.

It is possible to simulate both the magnitude and characteristics (shape) of complex three-dimensional localized brain tissue motion. This motion can be simulated for impacts of the head in multiple directions (frontal, occipital, lateral) and for motion in different planes (sagittal, coronal). Regarding relative brain-skel skull motion, and intracranial pressure, this study shows:

1. The pressure response seems to be more dependent upon the type of brain-skull interface than on the constitutive parameters chosen for the brain tissue, and the tied interface provided the best correlation with the experiments.

2. Smaller relative motion between the brain and skull results from a lateral impact than from a frontal or occipital blow for both the experiments and the FE model.

3. There is symmetry in the motion of the superior and inferior markers for both the simulations and the experiments during both sagittal, and coronal rotation.

4. Simulation of local brain motion is highly sensitive to the shear properties of the brain tissue, and average published values provided the best correlation with the experiments.

5. The local brain motion in response to low-severity impact is relatively insensitive to the type of brain-skull interface used for the simulation.

6. The results suggest that significantly lower values of shear properties than are currently used in most 3D FE models of the human brain should be used to predict local brain response during impact to the human cadaver head.

7. Further investigation into the effects of brain tissue shear properties for prediction of localized brain motion is needed.

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