Full Length Research Paper

Deformation of skull bone as intracranial pressure changing

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Raised intracranial pressure (ICP), a serious and often fatal condition, is often not preventable. In the present study, the relationship was determined between cranial deformation and ICP change. To record the deformation of skull bone, strain foil was placed on the exterior surface of parietal skull. Prior to construction of finite-element model (FEM), an in vivo study was undertaken. Three anesthetized adult rats were subjected to baseline recording followed by either experimental raising ICP. By using the ‘Ansys’ finite element processor, a three-dimensional FEM of a hollow sphere was constructed for human skull. The model was used to calculate the deformation of human skull with the intracranial pressure changing. The skull is a layered sphere constructed in a specially designed form with a Tabula externa, Tabula interna, and a porous Diploe sandwiched in between. The stress and strain deformations were well-proportional on the exterior surface of human skull. The deformation scope of human skull was theoretically from 1.0 to 3.4 μm with the changing ICP from 1.5 to 5.0 kPa. The cranium could move and human skull - dura mater system was deformed as the ICP fluctuates.

Key words: Deformation, skull bone, intracranial pressure, finite-element model, rat.

INTRODUCTION

Dr. Sutherland (1939) firstly perceived a subtle palpable movement within the bones of cranium. Dr. Upledger (Retzlaff et al., 1973) discovered that the inherent rhythm-mic motion of cranial bones was caused by the fluctuation of cerebrospinal fluid (CSF).

The ‘Monro (1823) - Kellie (1824) doctrine’ states that an adult cranial compartment is incompressible, and the volume inside the cranium is a fixed volume thus creates a state of volume equilibrium, such that any increase of the volumes of one component (i.e. blood, CSF, or brain tissue) must be compensated by a decrease in the volume of another. If this cannot be achieved then pressure will rise and once the compliance of the intracranial space is exhausted then small changes in volume can lead to potentially lethal increases in intracranial pressure (ICP). The compensatory mechanism for intracranial space occupation obviously has limits. When the amount of CSF and venous blood that can be extruded from the skull has been exhausted, the ICP becomes unstable and waves of pressure develop (Lundberg, 1960). As the process of space occupation continues, the ICP can rise to very high levels and the brain can become displaced from its normal position. The objective of this study was to determine the relationship between cranial deformation and ICP change.

MATERIALS AND METHODS

Prior to construction of the finite-element model (FEM), an in vivo study was undertaken using the rats. The deformations of the human skull-dura mater system were measured using strain gauge. Under the same loading conditions, the displacements of skull-dura mater system were calculated using the FEM. Strain foil was placed on the exterior surface of parietal skull to record the strains of skull bone. ICP variation was recorded simultaneously via strain foils on the skull. Three anesthetized adult rats were subjected to baseline recording followed by either experimental raising ICP induced by middle cerebral artery occlusion (MCAO). MCAO was produced as the model of raising ICP, and a computer continuously recorded the strains of skull bone. ICP variation could be obtained through the automatic processing based on the deformations of skull bone.

For the purpose of our analysis, we adopted a model consisted of a hollow sphere. By using the ‘Ansys finite element processor, an
initial three-dimensional FEM of a human skull was constructed. The model was used to calculate the deformation of human skull with the ICP changing.

Animal experiments

In the previous analysis, the skull deformation can reflect the ICP change. Some studies monitored ICP in animal models for a prolonged time (Jallo et al., 1997; Kusaka et al., 2004; Miller et al., 1984).

An in vivo study was undertaken on three anesthetized adult rats. Three adult rats weighing 320, 400 and 430 g were anesthetized by intraperitoneal injection of alpha-chloralose (60 mg/kg) and urethane (40 mg/kg). First a midline linear skin incision was made on the parietal skull of rats. The area was shaved for about 11 mm² and cleansed with alcohol and betadine. After drying, an incision was made and the skin and periosteum was pulled back until the parietal skull was visible. The BX120-2AA strain foil was placed at the measuring point with glue (Figure 1) and pressed with fingers for 2 min. Then the skin and periosteum were put back and the interrupted sutures were operated. The leading-out wires of strain foils were connected with the CM-1J-20 strain gauge. The insulating resistance was kept more than 50 megohm.

Middle cerebral artery occlusion (MCAO) was adopted as the model of raising ICP. MCAO was produced by passing a nylon thread up through the internal carotid artery and piecing a hole on the arterial wall at the middle cerebral artery. The reproduction of the MCAO model used Brint’s method (Brint et al., 1988). By either experimental raising ICP induced by MCAO, rats were subjected to baseline recording followed. The rats were allowed to breathe room air spontaneously, and the femoral arterial blood pressure was continuously monitored. In all rats, baseline ICP was recorded for 1 h. During this period, the deformations of skull bone were tested by strain gauge in all rats and a computer continuously recorded the strains of skull bone.

Theoretical analysis

The craniospinal cavity may be considered as a balloon. For the purpose of our analysis, we adopted a model consisted of a hollow sphere. We presented the development and the validation of a 3D finite-element human skull model intended to better understand the deformation mechanism of human skull corresponding to the ICP change. Four different entities were distinguished in the complex anatomy of the cranial cavity: Tabula externa, Tabula interna, and porous Diploe sandwiched in between. Based on the established knowledge of cranial cavity importantly composed of skull and dura mater (Figure 2), a thin-walled structure was simulated by the composite shell elements of the finite-element software (Piekarshi et al., 1973). The important mechanical characteristic of cancellous bone and dura mater is viscoelasticity (Odgaard, 1997; Noort et al., 1981).

ICP is not a static state, but one that is influenced by several factors. It can rise sharply with coughing and sneezing, up to 50 or 60 mmHg to settle down to normal values in a short time. It also varies according to the activity the person is involved with. For these reasons single measurement of ICP is not a true representation. ICP needs to be measured over a period. The brain appears to be mild injury when ICP variation is about 2.5 kPa, moderate injury when ICP variation is about 3.5 kPa and severe injury when ICP variation is about or more than 5 kPa. Therefore, the following theoretical analysis was carried out within the ICP scope from 1.5 to 5 kPa.

In this work, the finite-element software ANSYS was applied to theoretically analyze the deformation of human skull with the changing ICP. The external diameter of cranial cavity is about 200 mm. The thickness of shell is the mean thickness of calvaria. The average thickness of adult’s calvaria is 6.0 mm, that of Tabula externa is 2.0 mm, diploe is 2.8 mm, Tabula interna is 1.2 mm and, dura mater in the parietal position is 0.4 mm.

The combined models made up of the primary elements are usually adopted to describe the viscoelastic performance of actual materials. The creep of linear viscoelastic solid can be simulated by the Kelvin model of three parameters or the generalized Kelvin model.

Viscoelastic model of human skull

Kelvin model of three parameters is shown in Figure 3a. Figure 3b shows the relaxation curves of human skull and Kelvin model of three parameters in the compressive experiment. Figure 3c is the creep curves of human skull and Kelvin model of three parameters. So the theoretical Kelvin model of three parameters could well simulate the mechanical properties of human skull in the tensile experiments. Thus the Kelvin model of three parameters was adopted to describe the viscoelasticity of human skull in this paper. For the Kelvin model of three parameters, the stress and strain of human skull are shown in equation (1),

\[
\begin{align*}
\varepsilon &= \varepsilon_0 + \varepsilon_1 \\
\sigma &= E_0\varepsilon_0 + \eta\dot{\varepsilon}_1 \\
\sigma &= E_1\varepsilon_1 + \eta\dot{\varepsilon}_1
\end{align*}
\]

After the calculation based on the equation (1), the elastic modulus...
Figure 3. Viscoelastic Kelvin model of human skull and dura mater. a. Three-parameters Kelvin model of human skull. b. Relaxation train-time curves of the experiment and three-parameters Kelvin theoretical model for human skull. c. Generalized Kelvin model of the human dura mater. d. Creep train-time curves under different actions for fresh human dura mater ($L_0 = 23$ mm, $\theta = 37^\circ$). e. Creep compliance curves of generalized Kelvin model for human dura mater.

The viscoelastic model of human skull is equation (2),

$$E = \left( \frac{E_0 E_1}{E_0 + E_1} \right) + \left( \frac{E_0^2}{E_0 + E_1} \right) \frac{t}{\eta}$$

Here, $\sigma$ = direct stress acted on elastic spring or impact stress acted on viscopot; $\epsilon$ = direct strain of elastic spring; $E$ = elastic modulus of tensile compression; $\eta$ = viscosity coefficient of viscopot; $\xi$ = strain ratio; and

$$P_1 = \frac{\eta}{E_0 + E_1}$$

Viscoelastic model of human dura mater

For the viscoelastic model of human dura mater composed of three Kelvin-unit chains and a spring, the stress and strain of human dura mater were shown in equation (3),

$$\begin{align*}
\sigma &= E_1 \epsilon_1 + \eta_1 \dot{\epsilon}_1 = E_2 \epsilon_2 + \eta_2 \dot{\epsilon}_2 = E_3 \epsilon_3 + \eta_3 \dot{\epsilon}_3 \\
\epsilon_0 &= \frac{\sigma E_0}{E_0} \\
\epsilon &= \epsilon_1 + \epsilon_2 + \epsilon_3
\end{align*}$$

After the calculation based on the equation (3), the creep com-
Table 1. Coefficients for the viscoelastic properties of human skull.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Elastic modulus (GPa)</th>
<th>Viscosity (GPa/s)</th>
<th>Delay time (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$E_0$</td>
<td>$E_1$</td>
<td>$\eta$</td>
</tr>
<tr>
<td>Compression</td>
<td>5.69±0.26</td>
<td>42.24±2.09</td>
<td>96840±5400</td>
</tr>
<tr>
<td>Tension</td>
<td>13.64±0.59</td>
<td>51.45±2.54</td>
<td>206100±15360</td>
</tr>
</tbody>
</table>

Table 2. Creep coefficients for the viscoelastic properties of fresh human dura mater.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Elastic modulus (MPa)</th>
<th>Delay time (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$E_0$</td>
<td>$E_1$</td>
</tr>
<tr>
<td>Dura mater</td>
<td>16.67</td>
<td>125.0</td>
</tr>
</tbody>
</table>

The compliance of human dura mater is equation (4),

$$J(t) = E_0^{-1} + E(1-e^{-\tau_1/\tau_1}) + E_2^{-1}(1-e^{-\tau_2/\tau_2}) + E_3^{-1}(1-e^{-\tau_3/\tau_3})  \quad 4$$

Then the elastic modulus of human dura mater is equation (5),

$$E = \left[E_0^{-1} + E_1^{-1}(1-e^{-\tau_1/\tau_1}) + E_2^{-1}(1-e^{-\tau_2/\tau_2}) + E_3^{-1}(1-e^{-\tau_3/\tau_3})\right]^{-1}  \quad 5$$

Here, $\sigma$, $\varepsilon$, $E$, $\eta$, $\dot{\varepsilon}$ = Ditto mark; $\tau_1$, $\tau_2$, $\tau_3$ = Lag time, that is $\tau_1 = \eta_1 / E_1$, $\tau_2 = \eta_2 / E_2$, $\tau_3 = \eta_3 / E_3$.

In the finite-element software ANSYS, there are three kinds of models to describe the viscoelasticity of actual materials, in which the Maxwell model is the general designation for the combined Kelvin and Maxwell models. Considering the mechanical properties of human skull and dura mater, the finite-element Maxwell model was adopted to simulate the viscoelasticity of human skull-dura mater system. The viscoelastic parameters of human skull and dura mater are respectively listed in Tables 1 and 2.

Human skull is the viscoelastic material (Charalambopoulos et al., 1998). Considering the viscoelasticity of human skull and dura mater, we used the viscoelastic option of the ANSYS finite-element program to analyze the strains on the exterior surface of human skull as ICP changing. According to the symmetry of 3D model of human skull, the preprocessor of the ANSYS finite-element program was used to construct a 1/8 FEM of human skull and dura mater consisting of 25224 nodes and 24150 three-dimensional 8-node isoparametric solid elements, shown in Figure 4f.

RESULTS

Figures 4 a to e are the analytic graphs of stress and strain with finite-element software ANSYS when ICP variation is raised up to 2.5 kPa. The stress and strain distributions were well-proportioned on the exterior surface of human skull. Human skulls were deformed with the ICP change. As the ICP changing from 1.5 to 5 kPa, the strains of cranial cavity are shown in Figure 5 with the finite-element software ANSYS. The strains of cranial cavity were coincident with ICP variation.

In three rats in the MACO group, the blood vessels were obstructed within 30 min after passing a nylon
Figure 5. Strain curves of human skulls of the theoretical FEA and simulative experiment with the changing ICP. The deformation strains of mild injury, moderate injury and severe injury are separately 1.6, 2.9 and 4.0 με or so.

Figure 6. Strain changes of rats' skull bone with the changing ICP. **No.1 rat**: The strain change was totally corresponding to the symptom. After placed strain foils on the exterior surface of skull bone and searched the internal carotid artery for about 1.9 h, strains became stable. A silicone rubber cylinder attached to a nylon surgical thread was inserted through the internal carotid artery in rats and the blood vessel was obstructed for about half an hour from 2.82 h. The strains of skull bone were reduced with the decreasing ICP during the period of cerebral ischemia from 2.82 to 3.3 h. The thread was suddenly pulled out of the carotid artery while 3.3 h or so. After the autoregulation and recirculation of blood in carotid artery for about 40 min, the strains went up and reached the stability after nearly 3.83 h. **No.2 rat**: The wound in carotid artery bled twice when the time was about 2.5 and 2.8 h. Then ICP was decreased for the hemorrhage from the carotid artery. Correspondingly, the strains curve appeared to be twice-sharp declines at 2.5 and 2.8 h respectively, and reached the stable state at 3.9 h or so. **No.3 rat**: The position to place strain foil wasn't level on the exterior surface of skull bone and damp-proof measures were hardly taken for the strain foil. The strain foil wasn't firmly placed on the skull, so the strains couldn't reflect the ICP change.
The relationship between stress and strain is 

\[ \sigma = E \varepsilon \]

where \( \sigma \) is stress, \( \varepsilon \) is strain, \( E \) is the elastic modulus. Based on the strain-electrometric mechanism and after the automatic conversion of strain gauge, ICP variation of rats’ could be obtained by the strains of skull bone with the measuring times (Figure 7). The skull strains were consistent with ICP change. All recordings were completed within 4 h. No attempt was made for prolonged recordings.

**DISCUSSION**

The deformation scope of human skull was theoretically from 0.9 to 3.4 \( \mu \) as the ICP changing from 1.5 to 5.0 kPa. In No.1 rat, the ICP responded immediately the strains of skull bone. The records of No.1 and No.2 rats showed the exact same variations and patterns. Similar recordings of ICP tendency for No.1 and No.2 rats were observed after the experiment. In No.1 and No.2 rats, the skull strains responded immediately after arterial puncture and ascended sharply. In a few seconds after pulling the thread out of the carotid artery, ICP reached the valley pressure and then increased rapidly. ICP variation showed a similar pattern with skull strains. Among all of the recordings, lower peak values from No.2 rat were observed, which were about 40 - 50% lower than the values from No.1 rat. We have noticed some leaking of blood from the wound in carotid artery, which might contribute to the ICP reduction. Finally, the recording of stable ICP was raised due to the obstruction. The curve of No.3 rat indicated that it was important to paste firmly strain foil on the skull bone, or else the strain couldn’t reflect the ICP change.

**Conclusion**

According to the mechanism of mechanical deformation, the cranium can move and human skull - dura mater system is deformed as the ICP fluctuates. The timing of ICP to reach the peak value, and the slope of curve to decay to the plateau is almost identical between the skull strains and the ICP change.

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