Parametric Study of Effects of Brain-Skull Boundary Conditions and Brain Material Properties on Responses of Simplified Finite Element Brain Model under Angular Acceleration Impulse in Sagittal Plane*

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In the present study, the effects of brain-skull boundary conditions on the responses of a simplified three-dimensional finite element model of a thin sagittal slice of the human head were investigated. The model was excited using a time-dependent angular velocity with a maximum of 16 rad/s (maximum angular acceleration, around 5 000 rad/s²). The present study was conducted using the non-linear explicit finite element code LS-DYNA. Three methods for simulation of the brain-skull boundary conditions were investigated: 1) with brain rigidly attached to the skull; 2) using frictionless sliding contact allowing no separation between the brain and skull; and 3) direct simulation of cerebrospinal fluid CSF using a layer of eight-noded solid elements with fluid-like properties. Varying the method for simulation of the brain-skull boundary conditions appreciably affected the brain responses. However, varying parameters of a given method, such as viscosity of cerebrospinal fluid CSF and CSF-skull friction coefficient, exerted only minor effects on these responses. The present results suggest that accurate simulation of brain-skull boundary conditions requires direct representation of the subarachnoidal space/CSF as a fluid-like medium.

Key Words: Mechanics of Diffuse Axonal Injury (DAI), Brain-Skull Boundary Conditions, Modeling of Cerebrospinal Fluid, Finite Element Brain Model, Explicit Finite Element Modeling, Lagrangian Finite Element Formulation

1. Introduction

Brain diffuse axonal injury (DAI), i.e., widespread damage to axons in the white matter of the brain, typically occurs when a large angular acceleration is applied to the head as a result of a direct blow⁽¹⁾ (e.g., in traffic accidents resulting in contact between the occupant head and car interior) or inertial loading^{(1),(2)} (e.g., in traffic accidents with no contact between the occupant head and car interior). It is widely acknowledged that this type of injury results from shearing deformation of the brain tissue without

any damage to the skull⁽¹⁾. This implies that when attempting to understand the mechanics and predict the risk of diffuse axonal injury using mathematical modeling, such as finite element analysis, not only the brain material behavior but also the brain-skull boundary conditions should be appropriately simulated.

It is widely acknowledged that the brain tissue behaves as a nearly incompressible non-linear viscoelastic medium with the shear modulus of a few orders of magnitude lower than the bulk modulus⁽³⁾⁻⁽⁵⁾. However, in the field of impact biomechanics, the brain tissue has been often simplified using a linear viscoelastic material model^{(6),(7)} or even a linear elastic one⁽⁸⁾. Furthermore, a large variation in material constants of brain tissue is used in mathematical

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modeling studies. For instance, the shear modulus used by Ruan et al.⁽⁸⁾ is around two orders of magnitude larger than that used by Zhang et al.⁽⁷⁾

As with the data for modeling of the brain tissue, there is no consensus regarding the method for simulation of the brain-skull interface, i.e., subarachnoidal space and its interactions with the brain and skull. This lack of consensus is caused by the fact that the anatomy of the brain-skull interface is so complex that at the present stage of hardware/software development it is not feasible to adequately represent it by full-scale head models. This interface consists of the subarachnoidal space, optic nerves, and bridging veins (i.e., superior cerebral veins). The bridging veins connect the superficial cortical veins to the superior sagittal sinus, and mainly follow the sulci⁽⁹⁾. The subarachnoidal space is located between the brain and skull. It consists of arachnoidal trabeculae which form a net of canals for the cerebrospinal fluid (CSF). The subarachnoidal space is surrounded by two leptomeninges: 1) pia which forms the subarachnoidal space-brain boundary; and 2) arachnoid which is inferior to dura⁽⁹⁾. Various simplifications are needed when modeling the brain-skull interactions and subarachnoidal space. For instance, in the models by Khalil and Hubbard⁽¹⁰⁾ and Nishimoto and Murakami⁽¹¹⁾, the brain was rigidly attached to the skull, which is the simplest approach. The most advanced approach involves simulation of the cerebrospinal fluid (CSF) using a layer of brick elements interacting with the brain through a contact interface^{(6),(12)}.

Knowledge of the effects exerted by the brain-skull boundary conditions on the brain model responses is needed to determine a reliable method for simulation of the brain-skull interface when developing models for understanding the mechanics of DAI. So far, this knowledge has remained rather limited. Only recently, Darvish and Crandall⁽⁴⁾ conducted a systematic analysis of the effects of brain material properties and brain-skull boundary conditions on the brain model responses under transient loading. Although they took the role of the CSF carefully into account, they used a two-dimensional finite element model in which the brain and CSF were discretized using shell elements. Therefore, some of their algorithms, such as the LS-DYNA tied sliding contact interface and plane-strain shell element formulation, are available only for two-dimensional models, and cannot be fully utilized when developing three-dimensional head models.

Thus, in the present study, we investigated the effects of the brain-skull boundary conditions on the responses of a simplified three-dimensional model of a thin sagittal head slice, and preliminarily analyzed the

effects of brain material properties on these responses. When investigating the effects of the brain-skull boundary conditions, we focused on the subarachnoidal space whereas the role of the bridging veins was not investigated. Our rationale being that adequate simulation of these veins would require a sophisticated three-dimensional model of the brain and its vascular system.

2. Methods

2.1 Head model

A three-dimensional finite element model of a thin sagittal slice of the human head (Fig. 1) was developed and computed using the LS-DYNA 960 explicit finite element code (Livermore Software Technology Corporation, Livermore, CA, USA). The model geometry is based on the magnetic resonance images of the human sagittal sections publicly available from the BrainWeb website^{(13),(14)}. The section located 4 mm from the mid-sagittal plane was used.

The mesh used in the present model was constructed using the HyperMesh preprocessor (Altair Engineering, Troy, MI, USA). The brain was discretized using 3 487 eight-noded solid elements with a single integration point, and 210 elements of the same type were used to discretize the skull.

2.1.1 Assumptions and simplifications As the present study focused on understanding the effects of brain-skull boundary conditions rather than the development of a complete head model, the head/brain geometry was simplified to reduce the computation time. The foramen magnum was disregarded and the cranial cavity was closed. The basis for this simplification is that our study focused on duration phenomena of up to 20 ms. This seems too short for any significant real-time exchange of CSF between the cranial and spinal cavities. We applied only iner-

Fig. 1 Simplified model of thin sagittal slice of the human head developed in present study

tial loading, which implies that the skull deformation could be disregarded. Therefore, the skull was simplified as a rigid body.

The brain tissue was assumed to be homogenous and isotropic. Thus, no distinction was made between the white and gray matter, and the ventricles were not modeled. Although these simplifications may locally affect the brain responses, it seems unlikely that they could distort the relative changes that occur when varying the brain-skull boundary conditions and brain material properties.

Following the simplification often used in the field of impact biomechanics (see *Introduction* section), the brain tissue behavior was simulated using a linear viscoelastic material model⁽¹⁵⁾:

$$G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t}, \quad (1)$$

where G is the relaxation shear modulus, G_0 is the short-time shear modulus, G_{∞} is the long-time (i.e., infinite) shear modulus, β is the decay constant, and t is time. The brain tissue bulk modulus K was designated as that of water, i.e., $K = 2.2$ GPa.

2.1.2 External load The present model was excited using an idealized angular velocity impulse of up to 16 rad/s and duration of around 14 ms applied to the skull. The angular acceleration magnitude was up to around 5 000 rad/s² (Fig. 2), above the threshold of moderate DAI proposed by Margulies and Tibault⁽¹⁶⁾ for lateral rotation. The angular velocity was applied in the anterior direction about the point located near the gravity center.

2.2 Brain-skull boundary conditions

The present analysis of the effects of brain-skull boundary conditions on the brain model responses started using the simplest approach in which the brain was rigidly attached to the skull (simulation S1), i.e., with the skull and brain meshes merged. Then, the sliding-only contact (i.e., contact type 1 in the LS-DYNA code) was introduced between the brain and skull (simulation S2). This contact implies frictionless sliding with no separation between the brain and

Fig. 2 Angular velocity applied to the skull of the present head model and angular acceleration corresponding to this velocity

skull. Thus, the sliding-only contact may exert relatively large tensile forces on the brain. However, in the real world, such large tensile forces in the normal direction seem rather unlikely to occur as a result of interactions between the brain and subarachnoid space. Therefore, a layer of eight-noded solid elements (single integration point) was introduced between the brain and skull to simulate the CSF (simulations S3 to S17) (Figs. 1 and 3).

The CSF behavior was determined using the Mie-Gruneisen equation of state^{(15),(17)}:

$$p = \frac{\rho_0 a^2 c \left[1 + \left(1 - \frac{\gamma_0}{2} \right) c - \frac{a}{2} c^2 \right]}{\left[1 - (S_1 - 1)c - S_2 \frac{c^2}{c+1} - S_3 \frac{c^3}{(c+1)^2} \right]^2} + (\gamma_0 + ac)E \text{ for compression}, \quad (2)$$

$$p = \rho_0 a^2 c + (\gamma_0 + ac)E \text{ for expansion } (c < 0), \quad (3)$$

$$c = \frac{V_0}{V} - 1 = \frac{\rho}{\rho_0} - 1, \quad (4)$$

where p is the pressure, c is the compression (see Eq. 4), V is the instantaneous volume, V_0 is the initial volume, ρ_0 is the initial density, ρ is the instantaneous density, a is the bulk sound speed, E is the internal energy per initial volume, γ_0 is the Gruneisen gamma parameter, and a is the first order volume correction to γ_0 . S_1 , S_2 , and S_3 are the coefficients of the shock and particle velocity relation. In the present study, the parameters E , γ_0 , a , S_1 , S_2 , and S_3 were designated as zero, which simplifies Eq. (2) to the following formula:

$$p = -\frac{\Delta V}{V_0} K, \quad (5)$$

where ΔV is decrease in volume.

Thus, the present approach, in which the Mie-

Fig. 3 Model with sliding contact (surface-to-surface type) between the CSF and skull used in simulations S7 to S17. The contact was defined between the arachnoid membrane and dura. The arachnoid membrane was rigidly attached to the CSF, and the dura was rigidly attached to the skull. In this figure, some of the CSF and skull elements were removed to show the arachnoid membrane and dura

Gruneisen equation of state was applied to simulate CSF, implies that the CSF deviatoric stress results solely from the CSF viscosity μ , and the pressure is treated as a variable dependent only on the volume change. This approach would seem to more accurately represent CSF properties than the previous attempts by Trosseille et al.⁽¹⁸⁾, Turquier et al.⁽¹⁹⁾ and Canaple et al.⁽¹²⁾, in which elastic/hyperelastic material models were applied.

Two versions of the model with the CSF simulated using a layer of solid elements were used in the present study: 1) with the CSF rigidly attached to the brain and skull, i.e., CSF mesh merged with meshes of brain and skull (simulations S3 to S6); 2) with sliding contact (surface-to-surface type) between the arachnoid membrane discretized using elastic shell elements and dura (simulations S7 to 17) (Fig. 3). The model second version is referred to as the one with sliding between the CSF and skull.

The model in which the CSF was rigidly attached to the brain and skull was used when investigating the effects of CSF viscosity on the brain model responses. Four values for the CSF viscosity were used: $\mu=0.0$ (i.e., CSF treated as inviscid fluid), $\mu=0.723 \cdot 10^{-3}$ Pa·s (water viscosity), $\mu=1.440 \cdot 10^{-1}$ Pa·s (approximate viscosity of glycerine), and $\mu=7.230$ Pa·s (around 70% of liquid honey viscosity). The reason for selecting a viscosity greater than $0.723 \cdot 10^{-3}$ Pa·s was that although the CSF is 99% water, one cannot rule out the possibility that the effective viscosity of the sub-arachnoidal tissue may exceed that of water due to possible interactions between the CSF and arachnoidal trabeculae. The viscosity of 7.230 Pa·s was used for comparative purposes only, and the authors do not intend to suggest that the actual viscosity of the subarachnoidal tissue can be that large.

The model with surface-to-surface sliding contact between the CSF and skull was applied to investigate the effects of CSF-skull friction, because although the friction coefficient f between the arachnoid membrane and dura seems rather low, its exact value has not been determined yet. Two coefficients were used here: 0.1, and 0.5.

The test matrix for the present investigation of brain-skull boundary conditions is summarized in Table 1.

2.3 Brain material properties

When investigating the effects of brain-skull boundary conditions (simulations S1 to S8), the following values were used for the parameters of Eq. (1): $G_0=1.036$ kPa, $G_\infty=0.185$ kPa, and $\beta=16.5$ s⁻¹. Thus, the present values for G_0 and G_∞ were below the lower limits of the data previously used in mathematical modeling studies reported in the literature⁽⁶⁾⁻⁽⁸⁾.

Table 1 Test matrix when investigating effects of brain-skull boundary conditions on brain model responses

Simulation	Description
CSF disregarded. Direct interactions between brain and skull.	
S1	Brain rigidly attached to skull.
S2	LS-DYNA sliding-only contact between brain and skull.
CSF simplified using layer of solid elements between brain and skull. CSF rigidly attached to brain and skull. CSF viscosity μ varied.	
S3	$\mu=0.0$ (i.e., CSF as inviscid fluid)
S4	$\mu=0.723 \cdot 10^{-3}$ Pa·s (i.e., water viscosity)
S5	$\mu=1.446 \cdot 10^{-1}$ Pa·s (i.e., viscosity similar to glycerin)
S6	$\mu=7.230$ Pa·s
CSF simplified using layer of solid elements between brain and skull. Surface-to-surface sliding contact between CSF and skull. CSF viscosity of $0.723 \cdot 10^{-3}$ Pa·s. CSF-skull friction coefficient f varied.	
S7	$f=0.1$
S8	$f=0.5$

Table 2 Test matrix when evaluating effects of brain material parameters

Simulation	Description
$G_0=1.0$ kPa, $G_\infty=0.2$ kPa (parameters similar to those used in the present study when investigating the effects of brain skull boundary conditions)	
S9	$\beta=8$ s ⁻¹ (based on Darvish ⁽²¹⁾)
S10	$\beta=80$ s ⁻¹ (from Zhang et al. ⁽⁷⁾)
S11	$\beta=800$ s ⁻¹ (based on Al-Bsharat ⁽⁶⁾)
$G_0=2.0$ kPa, $G_\infty=0.4$ kPa	
S12	$\beta=8$ s ⁻¹
S13	$\beta=80$ s ⁻¹
S14	$\beta=800$ s ⁻¹
$G_0=10.0$ kPa, $G_\infty=2.0$ kPa (parameters used by Zhang et al. ⁽⁷⁾ when modeling cerebrum)	
S15	$\beta=8$ s ⁻¹
S16	$\beta=80$ s ⁻¹
S17	$\beta=800$ s ⁻¹

Fig. 4 Elements for output of maximum shear strain-time and pressure-time histories. The numbers in parentheses denote the coordinates (in millimeters) of Gauss points of elements A, B and C in X-Y coordinate system. Element D is adjacent to element A, and element E is adjacent to element C. The origin of X-Y coordinate system is located at the center of prescribed rotation of the skull

To evaluate to what extent the results of the present investigation of the effects of brain-skull boundary conditions could be affected by selection of

the brain tissue parameters, these parameters were varied as summarized in Table 2 (simulations S9 to S17). The parameters were selected with the idea of representing the range of values used in the relevant experimental and mathematical modeling studies reported in the literature^{(3),(4),(6)-(8),(20)-(22)}.

2.4 Analyzed variables

Since a large deformation of the CSF and brain was observed in the present study, we examined the total, internal and hourglass energy-time histories to detect any possible problems with numerical stability. When investigating the effects of the brain-skull boundary conditions on the brain model responses, we analyzed the contour plots of the maximum shear strain and pressure as well as the time histories of the maximum shear strain and pressure in the selected elements of the brain (elements A, B, C) and CSF (elements D and E) (Fig. 4). When investigating the effects of brain material properties, we focused on the peak values of the maximum shear strain-time histories in elements A and C.

3. Results

3.1 Effects of brain-skull boundary conditions

3.1.1 Models without CSF As expected, the model with the brain rigidly attached to the skull

(simulation S1) predicted a much larger shear deformation of the brain than the remaining models in which the CSF effects were either indirectly simulated using frictionless sliding contact allowing no separation between the brain and skull (simulation S2) or the CSF was directly represented using a layer of solid elements (e.g., simulations S4 and S7) (Fig. 5). Since in the present study the simulation time was only 20 ms, the differences in the shear deformation between simulation S1 and the remaining ones were concentrated in the outer layers of the brain elements. However, in simulation S1, large shear deformation quickly propagated towards the brain interior, which resulted in contour plots of the brain maximum shear strain qualitatively different from those observed in CSF simulations (Fig. 6). Therefore, disregarding CSF would be a gross oversimplification when simulating the brain responses under angular acceleration impulse in sagittal plane. Any model in which the brain is rigidly attached to the skull is likely to vastly overestimate the brain shear strain.

As with the maximum shear strain, the model with sliding-only contact (i.e., no friction and no separation) between the brain and skull (simulation S2) exhibited brain pressure-time histories appreciably different from those obtained under the assump-

Fig. 5 Maximum shear strain-time histories when varying method for simulation of brain-skull boundary conditions. S1: brain rigidly attached to skull; S2: frictionless contact allowing no separation between brain and skull; S4: CSF simulated using a layer of solid elements rigidly attached to brain and skull (CSF viscosity of $0.723 \cdot 10^{-3}$ Pa·s); and S7: CSF as in simulation S4 but with sliding contact (surface-to-surface type) between CSF and skull. a) Element A; b) Element B; c) Element C

Fig. 6 Maximum shear strain contour plots at 20 ms. a) Simulation S1 (i.e., brain rigidly attached to skull). b) Simulation S7 (i.e., CSF simulated using layer of solid elements with surface-to-surface sliding contact between CSF and skull)

Fig. 7 Pressure-time histories when varying method for simulation of brain-skull boundary conditions. S1: brain rigidly attached to skull; S2: frictionless contact allowing no separation between brain and skull; S4: CSF simulated using layer of solid elements rigidly attached to brain and skull (CSF viscosity of $0.723 \cdot 10^{-3} \text{ Pa} \cdot \text{s}$); and S7: CSF as in simulation S4 but with surface-to-surface sliding contact between CSF and skull. a) Element A; b) Element B; c) Element C

tion that the brain is rigidly attached to the skull (simulation S1). Addition of this sliding-only contact reduced the magnitude of negative pressure and resulted in oscillations that dominated the pressure response of elements A and C (Fig. 7). One possible explanation for these oscillations involves the finite element discretization of the skull and brain. The surfaces of finite elements were flat, which implies that small volumetric changes of the brain elements may be induced when the skull rotates in relation to the brain (Fig. 8).

Furthermore, in simulations S1 and S2, substantial negative pressure was calculated in brain elements A, B and C. One possible source of this pres-

Fig. 8 Hypothetical mechanism inducing volume changes in brain elements due to skull rotation in relation to brain when brain-skull interactions are simplified using frictionless contact allowing no separation (i.e., LS-DYNA sliding-only contact).

sure may be tensile forces between the brain and skull. In simulation S2, these forces resulted from the behavior of the LS-DYNA sliding-only contact, which prevents separation between the brain and skull. However, in the actual brain, the only anatomical structures that can exert tensile forces on the brain boundary are bridging veins and intracranial nerves. Thus, there is no anatomical basis to assume that the effects of these forces will be as strong as those in simulations S1 and S2. Therefore, it can be concluded that the present results do not support application of frictionless sliding contact allowing no separation between the brain and skull as a method for simulation of brain-skull boundary conditions.

3.1.2 Models with CSF In the model in which the CSF mesh was rigidly attached to the skull, the negative pressure was still observed, but its magnitude was appreciably smaller than in simulations S1 and S2. Since the tensile forces acting on brain in this model were determined by the properties of the CSF, the largest negative pressure was observed in simulation S6 in which an extremely large CSF viscosity was assumed (Fig. 9). When the surface-to-surface sliding contact was added between the CSF and skull (simulations S7 and S8), no negative pressure in brain elements A, B and C was observed (Fig. 10), which can be related to the fact that this type of contact does not transfer tensile forces in the normal direction. Furthermore, the brain pressure oscillations in simulations S3 to S8 were clearly smaller than those in

simulation S2. One possible explanation for this phenomenon is that a layer of solids we applied to simulate the CSF prevented the CSF volume changes induced by the finite element discretization of the CSF and skull boundaries (compare Fig. 8) from directly affecting the brain pressure. Therefore, it can be concluded that addition of a layer of eight-noded solid elements to simulate the CSF seems to improve the pressure responses of the present brain model.

Increase in CSF viscosity from 0 (i.e., inviscid fluid) to a level 200 times greater than the viscosity of water exerted virtually negligible effects on the pressure and maximum shear strain in the analyzed elements (Figs. 9 and 11). One possible explanation for this phenomenon can be that the viscous forces generated in the CSF were too small to appreciably affect the entire brain responses. Based on the results of simulation S6, it can be concluded that for the CSF strain rates observed in the present study, the CSF viscosity should be several orders of magnitude larger than that of water to induce significant shear forces.

As with CSF viscosity, when the coefficient of friction between the CSF and skull was varied within a range of 0.1 to 0.5, only minor effects on the calculated pressure and maximum shear strain-time histories were observed (Figs. 10 and 12). One possible explanation for this phenomenon may be that the normal forces between the arachnoid membrane and dura were too small to result in a friction force of appreciable magnitude. Furthermore, friction was not

Fig. 9 Pressure-time histories when varying CSF viscosity. S3: inviscid fluid; S4: $\mu=0.723 \cdot 10^{-3} \text{ Pa}\cdot\text{s}$; S5: $\mu=1.446 \cdot 10^{-1} \text{ Pa}\cdot\text{s}$; S6: $\mu=7.23 \text{ Pa}\cdot\text{s}$. CSF simulated using layer of solid elements rigidly attached to brain and skull. The results obtained in simulations S3, S4 and S5 are visually undistinguishable. a) Element A; b) Element B; c) Element C

the only mechanisms for transmitting tangential forces that could result in brain/CSF shear deformation. Such forces were also transmitted as a result of

finite element discretization. As already explained in section 3.1.1 *Models without CSF*, the contact surfaces of the CSF and skull elements were flat. When the

Fig. 10 Pressure-time histories when varying CSF-skull friction coefficient f . S7: $f=0.1$; S8: $f=0.5$. CSF simulated using layer of solid elements. Surface-to-surface sliding contact between CSF and skull. CSF viscosity of $0.723 \cdot 10^{-3} \text{ Pa}\cdot\text{s}$. a) Element A; b) Element B; c) Element C

Fig. 11 Maximum shear strain-time histories when varying CSF viscosity. S3: inviscid fluid; S4: $\mu=0.723 \cdot 10^{-3} \text{ Pa}\cdot\text{s}$; S5: $\mu=1.446 \cdot 10^{-1} \text{ Pa}\cdot\text{s}$; S6: $\mu=7.23 \text{ Pa}\cdot\text{s}$. CSF simulated using layer of solid elements rigidly attached to brain and skull. The results obtained in simulations S3 and S4 are visually undistinguishable. a) Element A; b) Element B; c) Element C

skull rotated in relation to the brain/CSF, the contact surfaces of the skull and CSF elements were no longer parallel to each other (compare Fig. 8). According to Al-Bsharat⁽⁶⁾, this could lead to tangential forces between the CSF and skull even when the CSF-skull contact interface is frictionless.

In all but S6 simulations in which the CSF was modeled using a layer of solid elements, the CSF exhibited very large shear deformation—the maximum shear strain was close to 1.0 (Figs. 12(d) and (e)). Despite such large deformation, no numerical instabilities were detected, and the ratio of hourglass to total energy at 10 ms was less than 0.20.

3.2 Effects of brain material properties

In all but S15 and S16 simulations, the maximum shear strain-time histories of element B did not reach its peak value till 20 ms. Therefore, only the results

obtained from elements A and C are analyzed here.

Increase in G_0 and G_∞ reduced the peak maximum shear strain in elements A and C, which is an intuitive result. On the other hand, increase in the decay constant β increased the peak maximum shear strain in these elements as summarized in Fig. 13. This is because large β implies a fast decrease in the shear modulus from G_0 to G_∞ , which increases the brain shear deformation (compare Eq. (1)). It should be also noted that only a small difference in the peak maximum shear strain was observed when β was increased from 8 s^{-1} to 80 s^{-1} . One possible explanation for this phenomenon may be that time constants $\tau = \frac{1}{\beta}$ associated with $\beta = 8 \text{ s}^{-1}$ and $\beta = 80 \text{ s}^{-1}$ (i.e., $\tau = \tau_{\beta=8} = 125 \text{ ms}$ and $\tau_{\beta=80} = 12.5 \text{ ms}$) were relatively long in comparison to the present simulations (i.e., 20 ms).

Fig. 12 Maximum shear strain-time histories when varying CSF-skull friction coefficient f . S7: $f=0.1$; S8: $f=0.5$. CSF simulated using layer of solid elements. CSF viscosity of $0.723 \cdot 10^{-3} \text{ Pa}\cdot\text{s}$. Surface-to-surface sliding contact between CSF and skull. The time histories of elements B, C, D and E obtained in simulations S7 and S8 are visually undistinguishable. a) Element A; b) Element B; c) Element C; d) Element D; e) Element E

Fig. 13 Peak maximum shear strain when varying brain material properties (simulations S9 to S17). $G_0 = 5 \cdot G_\infty$ in all simulations. CSF simulated using layer of solid elements. Surface-to-surface sliding contact between CSF and skull. CSF viscosity of $0.723 \cdot 10^{-3} \text{ Pa} \cdot \text{s}$. a) Element A; b) Element C

On the other hand, $\beta = 800 \text{ s}^{-1}$ corresponds to a very short time constant $\tau_{\beta=800}$ of 1.25 ms.

4. Discussion

4.1 Effects of method for simulation of brain-skull boundary conditions

The present study has indicated that the method used for the simulation of brain-skull boundary conditions appreciably affects the responses of the entire brain model. For instance, when the brain was either rigidly attached to the skull or frictionless contact allowing no separation was made between the brain and skull, a large negative pressure was observed in all the analyzed brain elements. No negative pressure in the analyzed elements was calculated when the CSF was simulated by means of a layer of solids with water-like properties and the CSF-skull interactions

were represented using surface-to-surface sliding contact (Fig. 7). As the surface-to-surface contact applied does not transfer tensile forces in the normal direction, it could underestimate the brain-skull coupling. One possible solution to improve the model biofidelity when using this contact may be direct simulation of bridging veins.

As with the brain pressure, the maximum shear-strain time histories were also greatly affected by the method used for the simulation of brain-skull boundary conditions. When the brain was rigidly attached to the skull, the calculated maximum shear strain was clearly larger than in the other simulations (Figs. 5 (a) and 6).

4.2 Effects of CSF viscosity and CSF-skull friction coefficient

For a given method of simulation of brain-skull boundary conditions, varying the method parameters, such as CSF viscosity and CSF-skull friction, exerted only minor effects on the calculated pressure and maximum shear-strain time histories in the analyzed brain elements (Figs. 9 to 12). The present results regarding the effects of CSF viscosity on the brain model responses are consistent with those obtained by Darvish and Crandall⁽⁴⁾ who used two-dimensional head model.

4.3 Effects of brain material properties

The present study indicated that varying the parameters determining the brain tissue behavior appreciably affects the brain model responses. Increase in the decay constant β increased the peak maximum shear strain in the analyzed brain elements, whereas an increase in the shear moduli G_0 and G_∞ reduced these peak values (Fig. 13). These results are consistent with the behavior of Eq.(1). Therefore, although the G_0 and G_∞ used here were below the values conventionally applied in the mathematical modeling studies, it seems unlikely that their increase would appreciably affect the conclusions derived from the results of the present investigation of the effects of brain-skull boundary conditions on the brain model responses.

4.4 Study limitations and recommendations for further studies

The limitations of the present study are related to three factors: 1) assumptions made when developing the present head model; 2) verification of the calculated brain responses; and 3) computational algorithms.

As explained in the *Methods* section, the key assumptions in developing the present head model were that the brain is isotropic and homogenous. However, in the actual brain, the brainstem region is stiffer than cerebrum⁽²⁰⁾, and the material properties

of brain tissue tend to depend on load direction. Furthermore, in the present study, the brain behavior was determined using a linear viscoelastic model whereas several experimental studies^{(21),(22)} have indicated that the actual brain behavior is nonlinear at large strain. It seems unlikely that the present assumptions regarding homogenous, isotropic and linear properties of brain tissue could distort the relative changes when varying the brain-skull boundary conditions and brain material properties. However, they may locally affect the brain responses.

It should be noted that direct verification of the brain responses calculated in the present study could not be done as no experimental data regarding the pressure and strain-time histories in the human brain subjected to large angular acceleration in sagittal plane are available. However, lack of such validation seems unlikely to compromise our findings as we focused on understanding the relative changes in the brain model responses rather than development of a finite element model that accurately represents all the features of the actual brain responses under transient loads.

In the present study, Lagrangian finite element formulation was used due to its low computational cost. Yet, when the mesh deformation is large, the accuracy of this formulation becomes poor. Therefore, it was necessary to limit the present investigation to an impulse load of short duration. Further studies focused on brain/CSF deformation larger than that calculated in the present study would presumably require application of either Eulerian finite element formulation or mesh-free discretization to simulate the CSF.

5. Conclusions

The present results confirm that the way in which the brain-skull boundary conditions are simulated appreciably affects the responses of the brain model subjected to angular acceleration in the sagittal plane.

Very large shear deformation and large negative pressure in the analyzed brain elements were obtained for a brain rigidly attached to the skull, clearly indicating that such a simulation results in gross oversimplification. When frictionless sliding contact allowing no separation (i.e., LS-DYNA sliding-only contact) was made between the brain and skull, the brain pressure-time histories exhibited not only substantial negative values but also strong oscillations that cannot be explained based on the brain anatomy. Thus, our results do not support application of this type of contact as a reliable method for simulation of brain-skull boundary conditions. The present study therefore suggests that direct representation of the

subarachnoidal space/cerebrospinal fluid as a water-like medium is needed in order to accurately simulate these conditions. This suggestion is consistent with the findings of Darvish and Crandall⁽⁴⁾ who indicated that frictional or elastic representation of the CSF tends to overestimate the brain shear stress.

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