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# The Influence of Heterogeneous Meninges on the Brain Mechanics under Primary Blast Loading

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## Abstract

In the modeling of brain mechanics subjected to primary blast waves, there is currently no consensus on how many biological components to be used in the brain–meninges–skull complex, and what type of constitutive models to be adopted. The objective of this study is to determine the role of layered meninges in damping the dynamic response of the brain under primary blast loadings. A composite structures composed of eight solid relevant layers (including the pia, cerebrospinal fluid (CSF), dura mater) with different mechanical properties are constructed to mimic the heterogeneous human head. A hyper-viscoelastic material model is developed to better represent the mechanical response of the brain tissue over a large strain/high frequency range applicable for blast scenarios. The effect of meninges on the brain response is examined. Results show that heterogeneous composite structures of the head have a major influence on the intracranial pressure, maximum shear stress, and maximum principal strain in the brain, which is associated with traumatic brain injuries. The meninges serving as protective layers are revealed by mitigating the dynamic response of the brain. In addition, appreciable changes of the pressure and maximum shear stress are observed on the material interfaces between layers of tissues. This may be attributed to the alternation of shock wave speed caused by the impedance mismatch.

**Keywords:** layered structures, mechanical properties, interface/interphase, finite element analysis, blast wave

## 1. Introduction

Blast-related traumatic brain injury (TBI), including injuries caused from primary shock waves, penetration, impact, and fire/ toxic gases, is a common injury in the course of current military conflicts. A study on a combat brigade returning from Iraq showed that 22.8% of soldiers had at least one TBI confirmed by a clinician, and more importantly, 88% of those injuries were caused by exposure to blasts resulting from improvised explosive devices [1]. Understanding the mechanisms of TBI is necessary for developing more appropriate protective systems and diagnostic tools.

Finite element (FE) analysis has emerged as a powerful tool for investigating injuries of the human head under different loading conditions. The level of geometric complexity, constitutive equations and material properties determines the accuracy of blast–head interaction results. The brain floats within the skull surrounded by CSF and meninges layers that allows for relative motion between the brain and the skull. This movement caused the rupture of bridging veins, which account for the majority of TBI [2]. A previous study examined the role of CSF properties on the response of human brain under certain impact loadings [3]. To the best of our knowledge, no published data is available on the role of meninges in trans-

ferring blast impacts to the brain. There is no consensus on including meninges [4,5] or not [6,7]. There is also no discussion on the right biological components required to be included in the blast FE models. Different biological components within the head (i.e., skull, dura and arachnoid mater, CSF, pia mater, and brain) have different densities with many interfaces separating these components with varying magnitudes of acoustic impedances. Recent findings have shown that shock waves are reflected/transmitted/converted at heterogeneous interfaces and their ratios directly depend on the mismatch in acoustic impedance at the surface of separation [8]. Further, the frequency content of the sharply rising shock wave overpressure dictates the spatial resolution at which the impedance mismatch becomes pronounced. For example, a wave with a frequency in the MHz range will have three orders of higher spatial resolution compared to a frequency in the kHz range wherein the waves either reflect, transmit or split [9].

The aim of this study is to investigate the influence of the structural heterogeneities of the human head on damping the dynamic response of the brain under primary blast loading conditions. Two-dimensional plane strain FE models with detailed geometries of the human head, including meninges and CSF, have been developed using an explicit nonlinear dynamic code LS-DYNA (Livermore Software Technology Cor-

poration, Livermore, CA, USA). This model is used to carry out parametric studies in order to help understand the influence of material models of pia maters, dura maters, and CSF on the dynamic response of brain.

## 2. Material and methods

Anatomically, human brain is encased in the triple layers of skull (outer table, diploe and inner table) and is suspended and supported by a series of three fibrous tissue layers, dura mater, CSF and pia mater, known as the meninges, as shown in Figure 1. The FE model is composed of 6700 8-noded solid elements including Eulerian elements to represent the CSF. A summary of the material properties for the various tissue layers is listed in Table 1.

Though the selection of the material model is critical in the analysis of the response of shock wave, there is no published work that examines the selection of the right material model for the intracranial components. This requirement is exacerbated by the fact that the material model should be valid for the brain tissue over a large strain/high frequency range encountered in blast loading scenarios. In this paper, a hyper-viscoelastic material model for the brain is employed over a large strain/high frequency range. The model is formulated in terms of a large strain viscoelastic framework and considers linear viscous deformations in combination with non-linear hyperelastic behavior. This Cauchy stresses from both hyperelastic and viscoelastic frameworks are superimposed onto each other to describe the brain behavior.

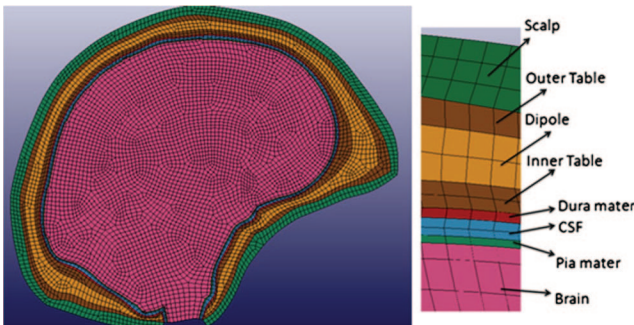
For the hyperelastic part of the material model, an Ogden hyperelastic strain energy function for incompressible material is adopted to describe the strain-dependent mechanical properties of brain tissue. The resulted Cauchy stress tensor  $\sigma$  is then calculated as:

$$\sigma = \frac{\partial W}{\partial \varepsilon} = \frac{\partial \left[ \sum_{i=1}^N \frac{\mu_i}{\alpha_i} (\lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3) \right]}{\partial \varepsilon} \quad (1)$$

where  $W$  is the Ogden strain energy function,  $\varepsilon$  is the Green strain tensor.  $\lambda_1, \lambda_2,$  and  $\lambda_3$  are the principal stretch ratios, and  $\mu_i$  and  $\alpha_i$  are constants to be determined experimentally for each value of  $i$ .

Four Ogden hyperelastic parameters were determined from the reported experimental data [10] as  $\mu_1 = -132.6$  kPa,  $\mu_2 = 0.481$  kPa,  $\alpha_1 = 0.00374$ , and  $\alpha_2 = 10.01$ . A comparison between the experimentally obtained stress-strain curve [11] and mathematically fitted one is shown in Figure 2. The agreement between numerical and experimental data appears to be quite good.

For the viscoelasticity part of the material model, the linear Maxwell is adopted and its associated Cauchy stress is computed through the following equation:



**Figure 1.** FE model arrangement with heterogeneous geometry and initial mesh of the model including triple layers of the skull (two cortical layers and middle diploe sponge-like layer), dura mater, cerebrospinal fluid (CSF), pia mater, and the brain.

$$\sigma_{ij} = J F_{ik}^T \cdot S_{km} \cdot F_{mj} \quad (2)$$

where  $\sigma_{ij}$  is the Cauchy stress component,  $F$  is the deformation gradient tensor, and  $J$  is the transformation Jacobian. The second Piola-Kirchhoff stress  $S_{ij}$  was estimated by a convolution integral of the form as:

$$S_{ij} = \int_0^t G_{ijkl}(t-\tau) \frac{\partial E_{kl}}{\partial \tau} d\tau \quad (3)$$

where  $E_{kl}$  is the Green's strain tensor, and  $G_{ijkl}(t-\tau)$  is the relaxation modulus function for the different stress measurements, which can be represented in terms of the Prony series:

$$G(t) = G_0 + \sum_{i=1}^n G_i e^{-\beta_i t} \quad (4)$$

where  $G_i$  is the relaxation modulus and  $\beta_i$  is the decay constant.

The relaxation moduli and decay constants are also derived from the published experimental data [11-17] at a wide frequency range between 0.01 MHz and 10 MHz. The fitted six term Prony series material parameters are:  $G_\infty = 2160$  Pa,  $G_1 = 156,488.3$  kPa,  $G_2 = 326,025.8$  kPa,  $G_3 = 0.0016$  kPa,  $G_4 = 1.2313$  kPa,  $G_5 = 17.583$  kPa,  $G_6 = 0.0254$  kPa,  $\beta_1 = 1.0763e + 9$  s<sup>-1</sup>,  $\beta_2 = 35.7999e + 6$  s<sup>-1</sup>,  $\beta_3 = 383.5146e + 3$  s<sup>-1</sup>,  $\beta_4 = 1e + 3$  s<sup>-1</sup>,  $\beta_5 = 10$  s<sup>-1</sup>, and  $\beta_6 = 3.6533$  s<sup>-1</sup>.

The numerical and experimental data are presented in Figure 3. More terms of Prony series can be used to obtain a better fit, however the six terms in the Prony series expansion is the limit in the commercial FE code LS-DYNA. In summary, the hyper-viscoelastic material model of the brain is depicted by seventeen material parameters in this work. This is the first material model for the brain that covers such wide range of frequencies.

The skull is modeled as a three-layered non-homogeneous material, including two cortical layers, i.e., outer table and inner table, and middle diploe sponge-like layer. Each layer is modeled as an isotropic material with properties listed in Table 1. A Gruneisen equation of state was used to mimic the behavior of CSF with a bulk modulus of 2.19 GPa. An equation of state (EOS) determines the hydrostatic behavior of the material by calculating pressure as a function of density, energy, and/or temperature and represented by Equation (5) in most generic form.

$$p = \frac{\rho_0 C^2 \mu \left[ 1 + \left( 1 - \frac{\gamma_0}{2} \right) \mu - \frac{a}{2} \mu^2 \right]}{\left[ 1 - (S_1 - 1) \mu - S_2 \frac{\mu^2}{\mu + 1} - S_3 \frac{\mu^3}{(\mu + 1)^2} \right]} + (\gamma_0 + a\mu) E \quad (5)$$

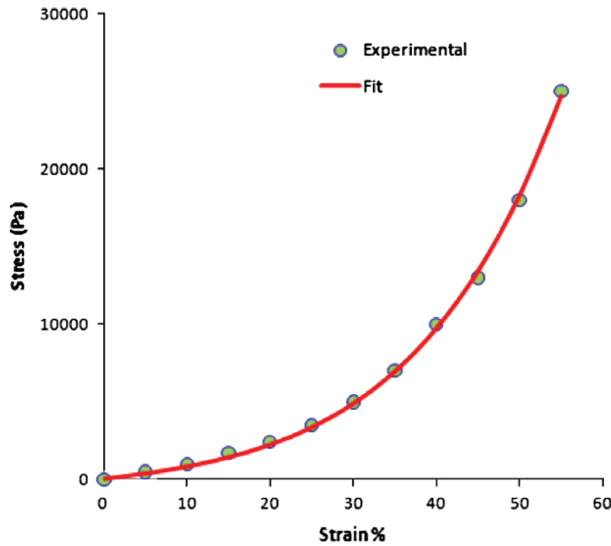
where  $\mu = \rho/\rho_0 - 1 = 1/V_0 - 1$ .  $C$  and  $S_1$  are parameters in the shock velocity ( $v_s$ ) and particle velocity ( $v_p$ ) according to the relation:  $v_s = C + S_1 v_p$ .  $C$  is the intercept of the  $v_s - v_p$  curve,  $S_1, S_2,$  and  $S_3$  are the coefficients of the slope of the  $v_s - v_p$  curve. Additionally,  $\gamma_0$  is the Gruneisen gamma,  $a$  is the first order volume correction to  $\gamma_0$  and  $E$  is the internal energy. In this work,  $S_1, S_2, S_3, \gamma_0$  and  $a$  are set to zero.

The mechanical properties of the meninges layers are not well established in the literature and there is a wide range the elastic moduli attributed to them [18,19]. For dura and pia maters, a second order Ogden hyperelastic model and two elastic models based on the published experimental work [18-20] have been employed to estimate the influence of the materials, as described in Table 2 and Figure 4. Jin et al. [19] performed uniaxial quasi-static and dynamic tensile experiments on pia mater at strain-rates of 0.05, 0.5, 5 and 100 s<sup>-1</sup>. Since high strain rate data is more suitable and applicable for blast loading scenarios, the data regarding 100 s<sup>-1</sup> strain rates was used to formulate the constitutive model in this study. The fitted material behaviors were added in Figure 3.

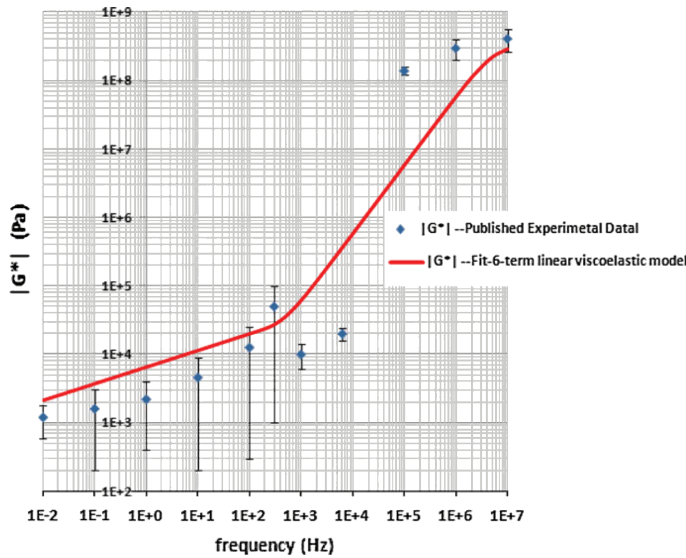
In this study, a blast scenario characterized by positive pulse duration and peak overpressure has been used with reference to the Bowen curves, which indicated that the unpro-

**Table 1.** Properties of the human head components [21,23,24].

Layer	Young's modulus $E$ (MPa)	Density (kg/m <sup>3</sup> )	Poisson's ratio
Scalp	16.7	1200	0.42
Outer table	15,000	2000	0.22
Diploe	1000	1300	0.24
Inner table	15,000	2000	0.22
Dura	Hyperelastic & elastic models	1130	0.4999
CSF	$K = 2.19$ GPa	1000	Incompressible
Pia	Hyperelastic & elastic models	1130	0.4999
Brain	Hyper-viscoelastic	1040	$\nu = 0.49999948$



**Figure 2.** Second order Ogden hyperelastic model of the brain tissue.

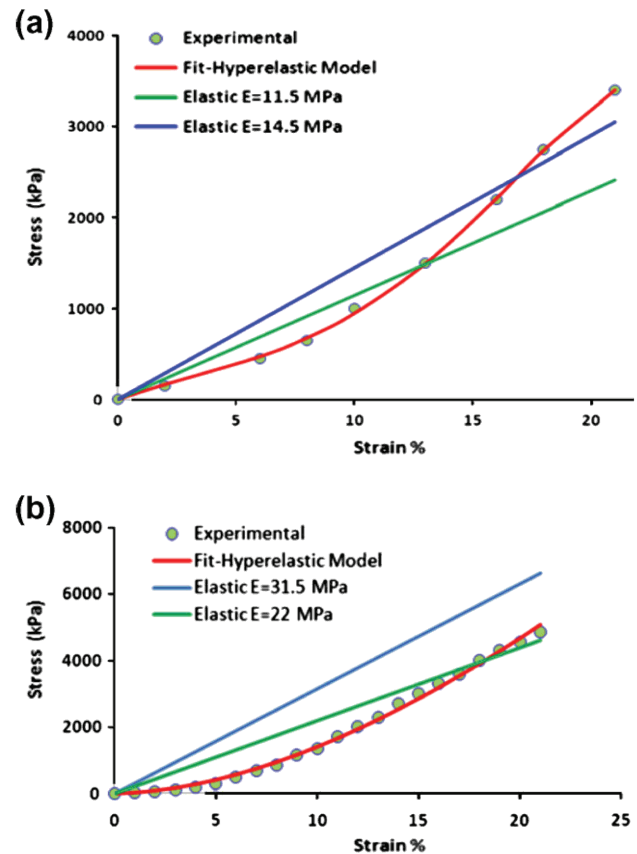


**Figure 3.** Complex shear modulus—experimental data and the fitted 6-term Maxwell viscoelastic model.

tected lung injury threshold is 5.4 atm peak pressure [21]. The simulated shock load, illustrated in Figure 5, is associated with free-air detonation (TNT weight of 0.5 lb and 2.5 feet as stand-off distance) and was previously validated by experimental data [22]. A tied contact algorithm (i.e. no tangential sliding and no separation) was suitable for the brain-membrane interfaces because it was capable of transferring loads in both compression and tension. This is necessary because the penalty algorithms are inadequate in representing the tensile re-

**Table 2.** Three material models for pia and dura maters.

	Ogden model [19]	Elastic model [20]	Elastic model [18]
Pia mater	$\mu_1 = 106.33$ kPa $\mu_2 = -260.3$ kPa $\alpha_1 = -54.895$ $\alpha_2 = -47.472$	$E = 14.5$ MPa $\nu = 0.45$	$E = 11.5$ MPa $\nu = 0.45$
Dura mater	$\mu_1 = 19.073$ MPa $\mu_2 = 17.853$ MPa $\alpha_1 = -3.1478$ $\alpha_2 = 3.4246$	$E = 22$ MPa $\nu = 0.45$	$E = 31.5$ MPa $\nu = 0.45$



**Figure 4.** Mechanical behavior of (a) pia mater and (b) dura mater.

gions which is encountered in the counter-coup areas under blast loading conditions [23]. The hourglass energy was monitored and found to be negligible through the whole simulation process. This indicated that the simulations did not encounter any numerical instabilities. The free boundary condition is used for this head model. For the blast scenarios it is known that stress wave action is significant with negligible gross head motion [6,7], thus fixed vs. free boundary condition do not significantly affect the dynamic head response.

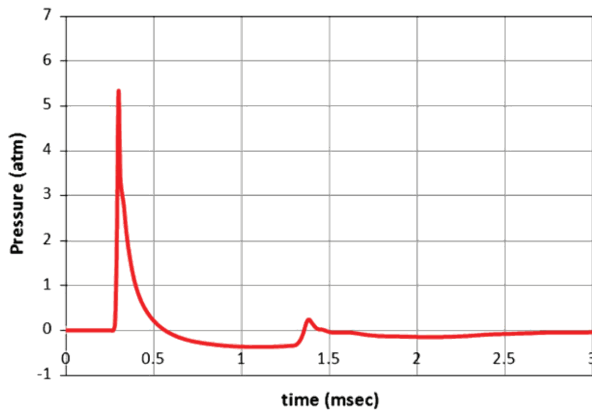


Figure 5. Blast shock wave load with 5.4 atm peak overpressure applied to the head.

3. Results and discussion

The focus of this study is to understand the influence of meninges on the dynamic response of the brain subjected to blast loadings. The intracranial pressure (ICP), maximum shear stresses, and principal strains are obtained to demonstrate the efficiency of material models and function of layered structure. It should be noted that the results presented in this work are based on two dimensional plane strain analysis. The actual magnitudes of ICP, maximum shear stresses, and principal strains for three dimensional head study could be significant different. The results here should be considered only in qualitative terms. Due to the comparative nature of this work, the obtained role of heterogeneities of the human head on brain mechanics was justified.

3.1. Role of meninges

Three case studies were developed to investigate the effect of meninges on dynamic response of brain tissues. Case study A refers to the full head model subjected to the blast loading. Case study B excluded the three layers of meninges from the full model. Case study C only included CSF layer to represent the meninges. A comprehensive comparison of ICPs and maximum shear stresses in the brain as a function of time for these three cases are presented in Figure 6. It shows that the head model without considering meninges (case study B) led to higher peak ICPs, maximum shear stress in the brain than that in case study A. The peak ICP reached 579 kPa at 0.27 ms and 292 kPa at 0.3 ms for cases B and A, respectively. It reveals that the existence of meninges including CSF can significantly reduce the amount of the peak ICP and also postpone its occurrence time. This is also valid for the maximum shear stress and principal strain induced in the brain. The peak maximum shear stress was 5.2 kPa at 0.4 ms and 3.5 kPa at 0.57 ms for cases B and A, respectively. The peak principal strain in case study B was approximately 2.5 times than those observed in case study A. This clearly demonstrated that meninges act as protective layers for the brain tissue subjected to blast loadings.

3.2. Effect of material models of meninges

Pia and dura mater used in the published FE models for head response simulations are mostly modeled as linear elastic model with young's modulus  $E = 11.5$  MPa and  $E = 31.5$  MPa and Poisson's ratio of 0.45, respectively [18]. However, the experimental data reported in the literature does not support these numerical values [19,20]. In order to observe the sensitivity of model outcomes to the material properties, three material models are employed here for both pia

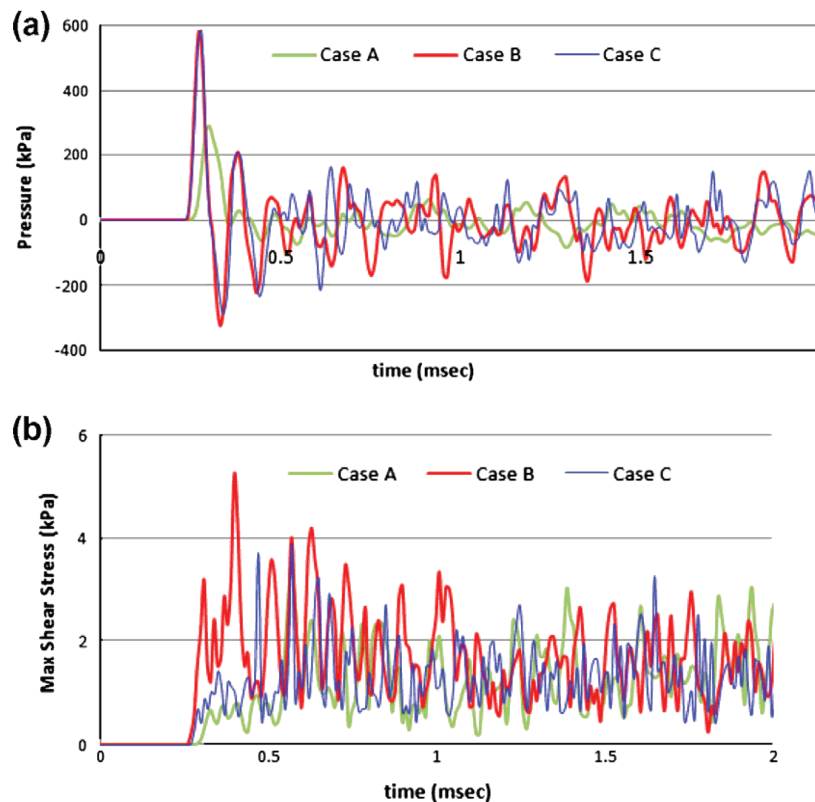


Figure 6. The influence of meninges on the dynamic behavior of brain tissue under blast loading conditions.

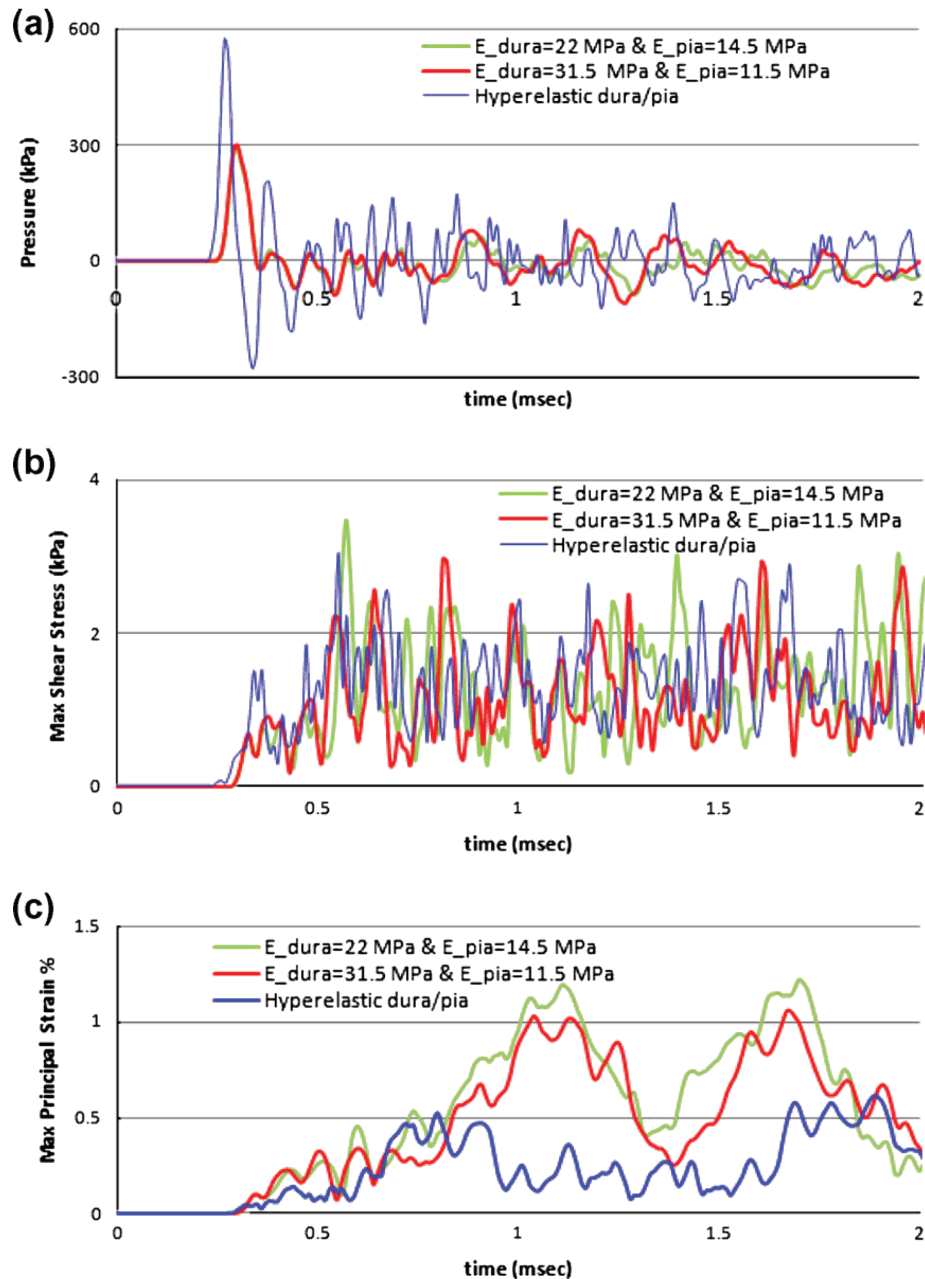


Figure 7. The influences of meninges material models on brain's dynamic responses.

and dura mater, as depicted in Figure 4 and Table 2. The resulted brain responses were shown in Figure 7. It is clear that the hyperelastic material models led to larger ICP and less strains in the brain tissue than the use of elastic model. It is also observed that the brain responses based on the hyperelastic model of meninges were very similar to the model without considering meninges (Case B and C in Figure 6), especially in ICP curves.

We then compared two elastic models for meninges. One adopted the Young's modulus of  $E_{dura} = 22$  MPa and  $E_{pia} = 14.5$  MPa from published experimental data, referred to as  $E1$ , the other is from published simulation work as  $E_{dura} = 31.5$  MPa and  $E_{pia} = 11.5$  MPa, referred to as  $E2$ . The obtained peak ICP, maximum shear stress and principal strain of the brain in the case  $E1$  were found to be 2%, 17% and 13% higher than those in case  $E2$ . This indicates that selection of material model affect the outcome of FE models.

### 3.3. Role of impedance mismatch on maximum shear stress responses

When a wave confronts a boundary between two media with mismatched impedances, part of the wave is reflected from the boundary while part of the wave is transmitted across it. The degree of transmission and reflection at the interface not only depends on the material properties of the two media, but also the ratio of impedance mismatch. This phenomenon can result in appreciable changes to the pressure and energy transferred at that interface [24]. The maximum shear stresses in the interfaces of the different constituents obtained with the full model (case study A) are revealed in Figure 8. Appreciable changes of maximum shear stress were observed at the brain/pia, pia/CSF, CSF/dura, and dura/skull interfaces. The mechanical impedance mismatch existing between different material layers plays an important role in the

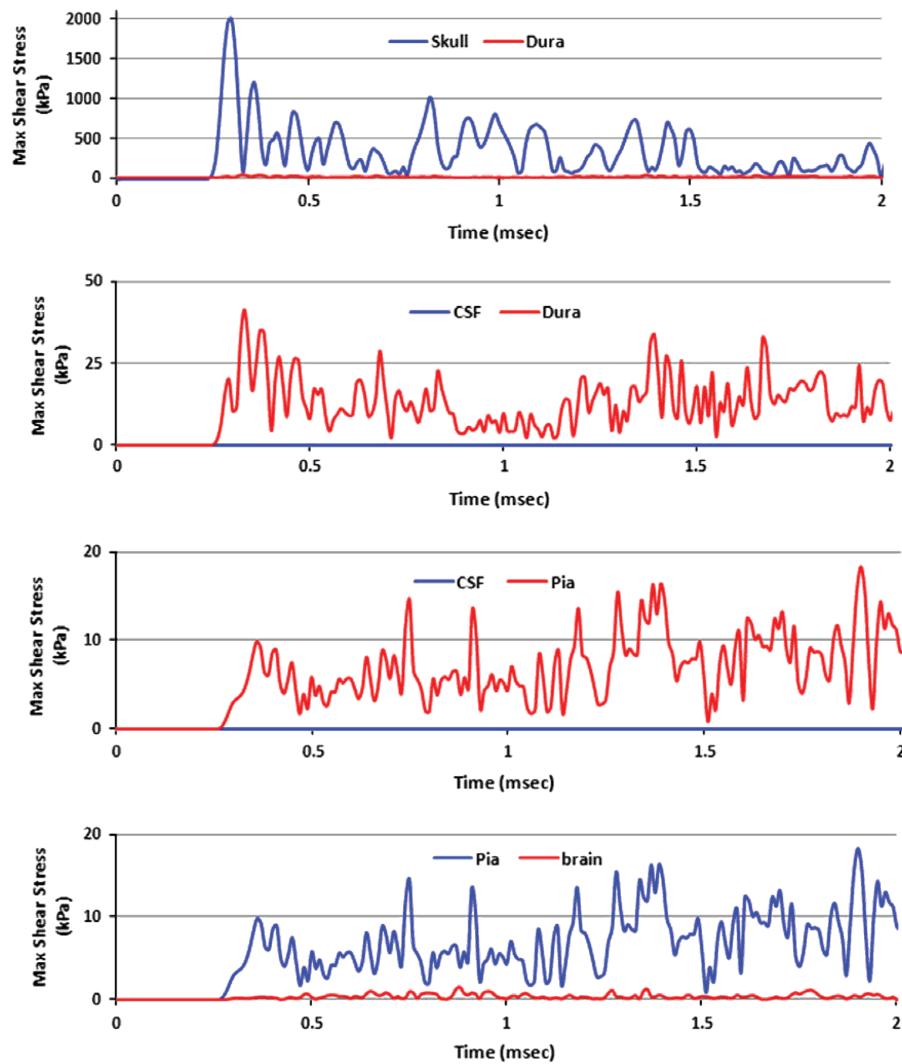


Figure 8. The influences of mismatched impedances on shear stress responses at different interfaces.

development of traumatic brain injuries. It is clear that each layer served to damp the dynamic response of the brain under primary blast loadings. The skull experiences higher stresses due to its more rigid material properties. Again, the results indicate that the meninges including CSF has an important influence on the brain responses.

#### 4. Conclusions

Computational modeling is emerging as a viable tool in understanding the effect of primary blast injuries and designing better protective devices. The influences of the heterogeneities of the human head in damping the dynamic responses of the brain tissue under blast loading conditions are investigated in this work and summarized as the following:

- The heterogeneous composite structures of the head have a major influence on the intracranial pressure, maximum shear stress, and maximum principal strain in the brain, which indicate possible brain injuries.
- A hyper-viscoelastic material model is developed to better represent the mechanical response of the brain tissue over a large strain/high frequency range applicable for blast scenarios.

- The meninges mitigate the dynamic response of the brain tissue subjected to blast loadings, and serves as protective layers of the brain. The material models of the pia and dura maters impact the brain response, and therefore the understanding of brain injuries.
- Appreciable changes of the pressure and maximum shear stress have been observed on the material interfaces between layers of tissues. This may be attributed to the alternation of shock wave speed caused by the impedance mismatch.

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