

Developing a Numerical Model for Human Brain under Blast Loading

Atacan Yucesoy¹, Thomas J. Pence¹, Ricardo M. Alvarez¹ and Adam M. Willis^{1,2}

¹Michigan State University, East Lansing, MI 48824.

²San Antonio Military Medical Center, Fort Sam Houston, TX 78234

Abstract

Blast-induced traumatic brain injury (bTBI) is one of the widespread causes of mortality and morbidity for military personnel. Exploring the mechanics of brain tissue is critical to predicting intracranial brain deformation and injury resulting from severe blast loading. This capability would help in obtaining a prognosis and choosing adequate neurosurgical procedures before a physical intervention takes place. In this conference paper, the aim is to build a numerical model using LS-DYNA[®] with the ability to capture the complex deformations induced by blast loading of the human brain. A coupling method of Load Blast Enhanced and Multi-Material Arbitrary Lagrange Eulerian are employed to generate the intracranial shock wave and to evaluate the interactions of brain layers, respectively. Relatively large displacement and velocity differences are observed between the skull and the gray matter. Complex interactions ensue when front-to-back moving coup waves meet back-to-front moving contrecoup waves. Shear stresses are highly localized at the interface of the gyri and cerebrospinal fluid and around the ventricles.

Introduction

Blast-induced traumatic brain injury (bTBI) is a leading cause of mortality and morbidity for personnel deployed in military conflict. bTBI occurs via a mechanical insult brought about by the explosion, possibly leading to temporary or permanent neurodegenerative diseases in the human brain. The mechanical insult can produce both localized and distributed forces across the brain [1]. The brain trauma can be in the form of pathological lesions, edema, contusion, and hemorrhage which are revealed by a variety of medical imaging technologies. Some physical and cognitive symptoms related to mild TBI are incoordination, nausea, vomiting, dizziness, headache, deficits in attention, loss of short-term memory and information processing [2,3].

The main objective of this work is to build a numerical model in order to investigate the intracranial shock wave generation in the human head for the purpose of determining injury mechanism. The recent advancements in computational tools can give us an opportunity to explore neuropathological damages occurring in human brain result from exogenous mechanical forces, such as blast-induced traumatic brain injuries. The numerical model is developed by using explicit nonlinear dynamic code LS-DYNA (Livermore Software Technology Corporation, Livermore, CA, USA). The human brain geometry used to construct the FE mesh is generated on the basis of Magnetic Resonance Imaging (MRI) scans. The FE model of the head consists of the skull, cerebrospinal fluid (CSF), and brain soft tissues; white and gray matter. The coupling method of Load Blast Enhanced and Multi-Material Arbitrary Lagrange Eulerian, available in LS-DYNA, scheme is employed to generate the blast wave and to treat the contact algorithm between the Eulerian and Lagrange elements, respectively.

Methods

The brain is encapsulated by distinct biological tissues such as the scalp, skull, and meninges. The primary functions of these layers are to protect the central nervous system against the mechanical shocks. The subarachnoid space, between the arachnoid membrane and pia matter, is filled with a watery fluid known as cerebrospinal fluid. Along with the physiological functions, CSF can mitigate the extracranial forces and

support the brain tissues against gravity forces by creating buoyancy [4]. The brain soft tissue is mainly found in two parts: cerebral white and gray matter. The surface of white matter is covered by gray matter.

The anatomical details and material properties play a significant role in an accurate assessment of injury mechanism under blast exposure. The human brain FE model is derived from an axial plane view MRI scan so as to obtain the anatomically correct distribution of skull, CSF, white and gray matter. The FE model of the human head is approximately formed by 266,000 hexahedral elements. The quality of mesh has a direct effect on the accuracy of numerical prediction, and computational time; therefore, the quality of mesh is evaluated in terms of smoothness, aspect ratio, skewness, and maximum and minimum ratios of the elements. Gray matter is found along the perimeter of the soft tissue, as illustrated in Fig. 1. CSF is found between the soft grey matter and the skull, the ventricles at the center are also filled with CSF. Other finer tissues aspect including falx, membrane, dura and pia mater are not resolved in this numerical study.

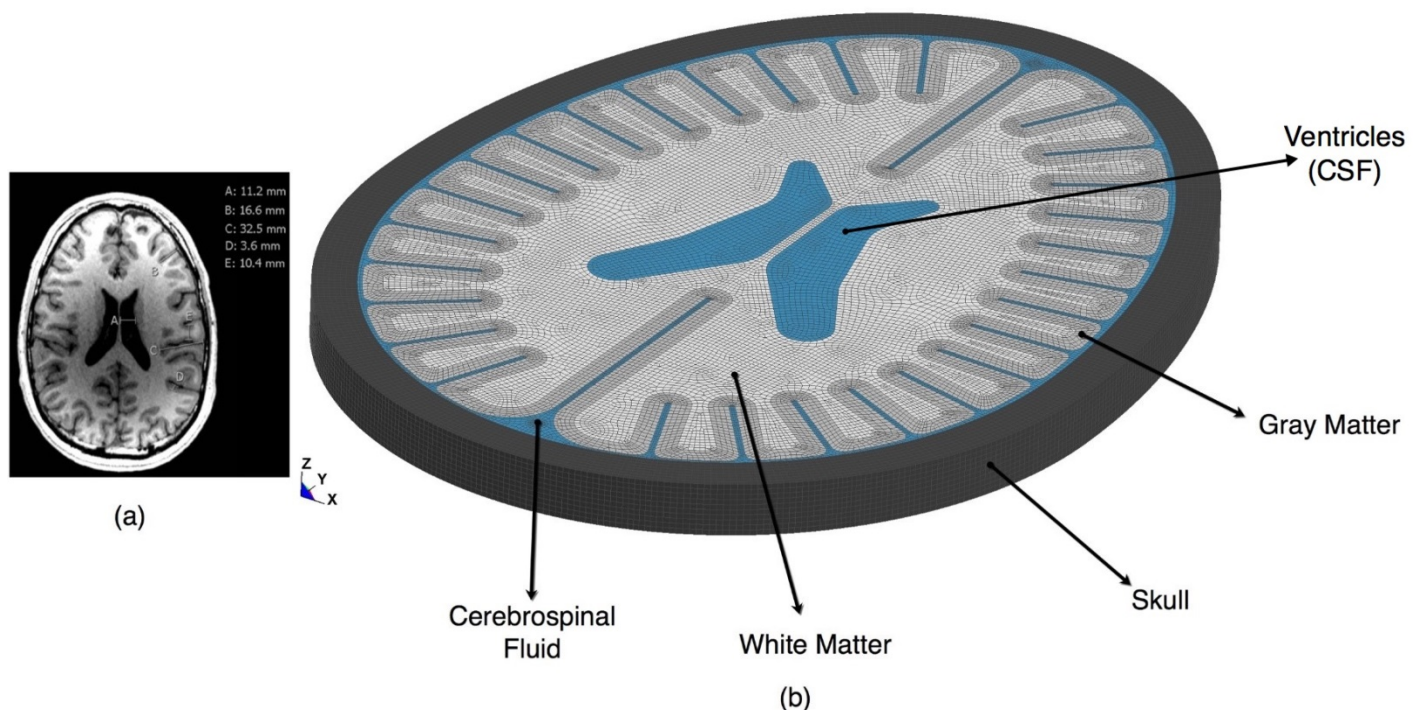


Figure 1. Magnetic Resonance Imaging (MRI) scan based FE model of the human brain. (a) Axial (traverse) plane of MRI scan with dimensions (b) FE model of human head

The skull material is simulated by MAT_013_ISOTROPIC_ELASTIC_PLASTIC_FAILURE material card because of its compressible, isotropic elastic-plastic material properties. In fact, for the simulations reported here, the failure criteria are never met so that the material always remain in the elastic regime. The material properties of the skull are listed in Table I.

Table I. Elastic material properties of skull [5].

| Part | Density (gr.cm ⁻³) | Shear Modulus (MPa) | Bulk Modulus (MPa) | Young Modulus (MPa) | Yield Stress (MPa) | Plastic Failure Strain (%) | Failure Stress (MPa) |
|-------|--------------------------------|---------------------|--------------------|---------------------|--------------------|----------------------------|----------------------|
| Skull | 1.21 | 3276 | 4762 | 8000 | 95 | 1.6 | 77.5 |

The cerebrospinal fluid is modeled as incompressible fluid due to its high bulk modulus and low shear modulus properties. In this numerical study CSF is characterized with MAT_ELASTIC_FLUID card without using any equation-of-state card. The material properties are shown in Table II.

Table II. The properties of MAT_ELASTIC_FLUID [5, 6].

| Parts | Density (gr.cm ⁻³) | Bulk Modulus (MPa) | P _{cut} (GPa) |
|-------|-----------------------------------|--------------------------|---------------------------|
| CSF | 0.9998 | 1960 | -1.0e-05 |

The mechanical behavior of human brain tissue exhibits viscoelastic and regional dependency under different loading condition [7]. Therefore, the brain tissues are modelled by using viscoelastic material models based on the rheological model in LS-DYNA material library MAT_006_VISCOELASTIC material card. Such a model effectively makes the pressure-volume relatively elastic and the shear response viscoelastic using a standard linear solid [8]. The time dependent shear modulus of the white and gray matter is defined by the equation as follows.

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

where t denotes time, G_0 is the short-term shear modulus, G_{∞} is the long-term modulus and β denotes decay constant. The mechanical properties of gray and white matter are given in Table III, as below.

Table III. Viscoelastic material properties of white and gray matter [5].

| Parts | Density (gr.cm ⁻³) | Bulk Modulus (MPa) | Short-term shear modulus G ₀ (kPa) | Long-term shear modulus G _∞ (kPa) | Decay constant β (s ⁻¹) |
|----------|-----------------------------------|--------------------------|--|---|--|
| White M. | 1.04 | 2371 | 41.0 | 7.8 | 40 |
| Gray M. | 1.04 | 2371 | 34.0 | 6.4 | 40 |

The skull, cerebral white and gray matter are discretized with a Lagrangian mesh, whereas the cerebrospinal fluid in ventricles and subarachnoid space is simulated as Eulerian part. The Fluid Structure Interaction (FSI) is modeled by CONSTRAINED_LAGRANGE_IN_SOLID card to define contact algorithm between CSF and skull and soft tissues interface. The blast wave loading can be generated by using ConWeb, MM-ALE and coupling of these two methods in LS-DYNA. In this study, the blast loading is directly applied to FE model of human brain by using LOAD_BLAST_ENHANCED card. This option is simplest way to generate the blast wave in LS-DYNA. In the MM-ALE and Hybrid method, the ambient around the head and explosive should be defined separately; however, modeling of these parts would require more computational resources. The standoff point is set at one-meter distance from the head model. As a boundary condition, the upper and lower surfaces of the layers under study (Fig.1) are constrained to against in the z-direction (normal) displacement. In the transverse directions, free boundary conditions are applied to the layer surface.

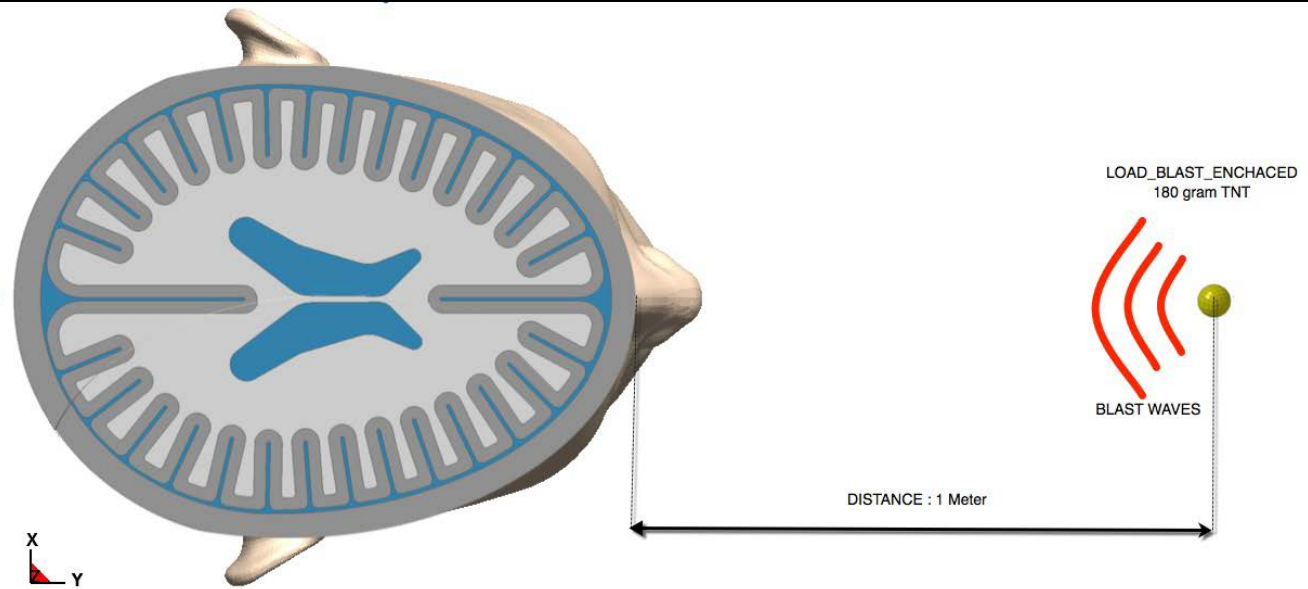


Figure 2. The top view of the general scheme of numerical simulations.

Results

The standoff point is located at one-meter away from the head. The blast wave is generated by using TNT explosive of 180 grams. As exhibited in Fig. 3, the initial interaction of the blast with the fore-head generates a high pressure (coup) wave that propagates through the tissue. The peak pressure is observed as 742 kPa at the 0.92 milliseconds, presented by deep red color in the figure. At the same time, a faster tensile wave propagates around the skull. When those waves collide at the back of the skull, it generates a negative pressure (contrecoup) wave that propagates through the brain in the opposite direction. The maximum observed negative (tensile) pressure in the contrecoup is 100 kPa, colored with deep blue at 1.10 milliseconds.

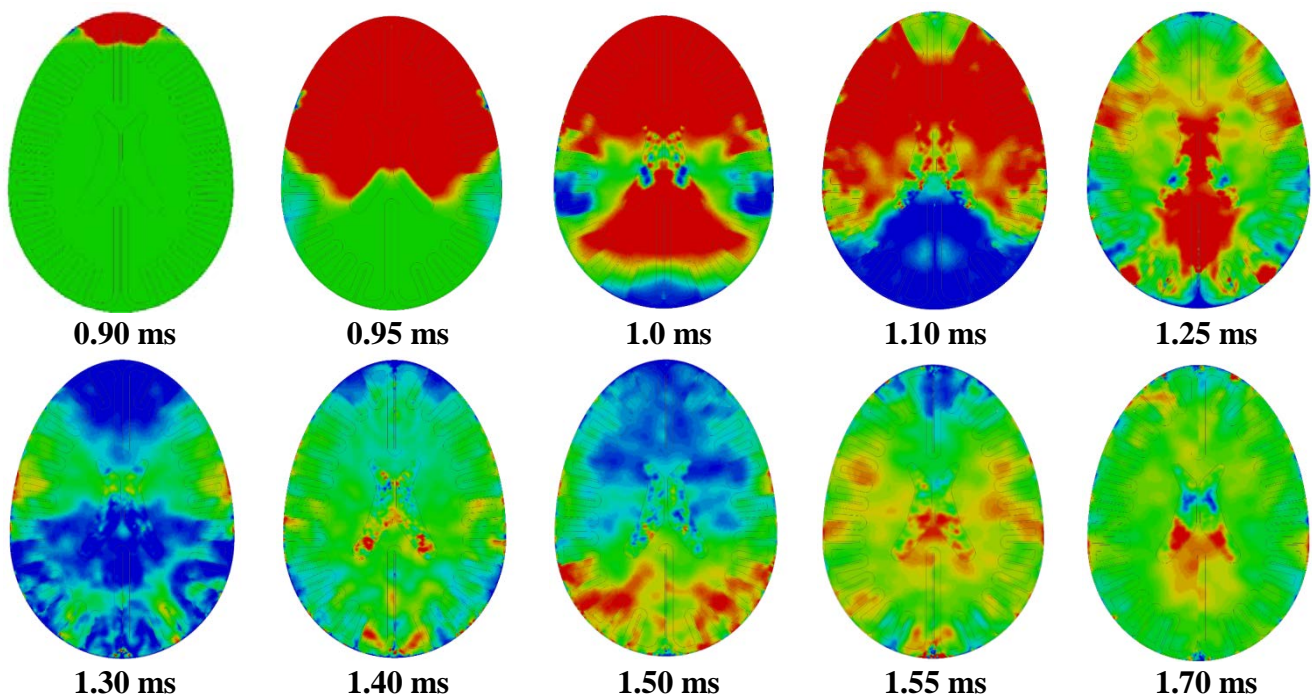


Figure 3. Time-lapse images of blast exposure showing the development of stress wave propagation inside the FE human head model.

As shown in Fig. 3, the initial coup stress gradually dissipates as it propagates through soft tissue in the general y-direction toward the back of the head model. As this process takes place, the tensile stress level increases gradually at the contrecoup. As evidenced in Fig. 4, a relatively obvious compressive spike of around 250 kPa is observed at the contrecoup around 1.3 ms after the initial blast interaction. In fact, the stress history at the contrecoup is of utmost importance given that this is a dominant focal region for injury.

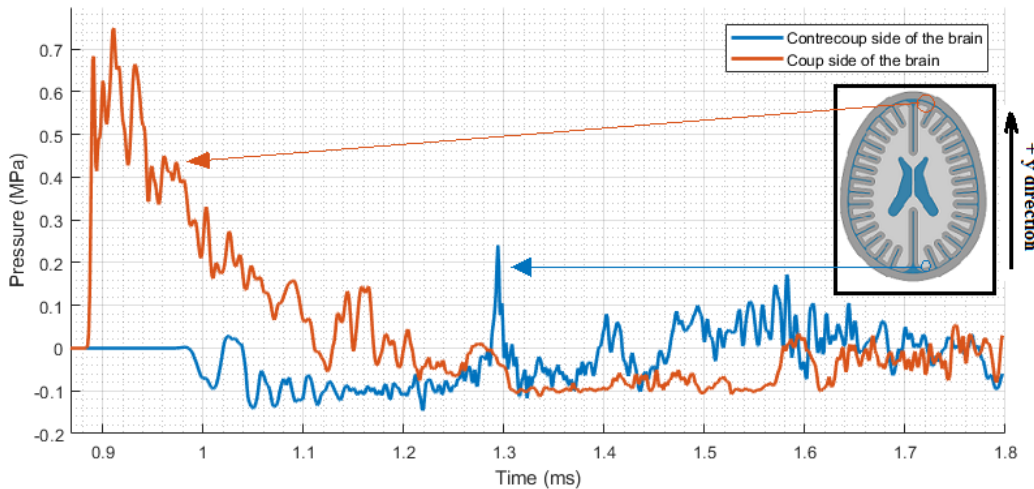


Figure 4. Pressure vs. Time histories at two different gray matter location in brain; contrecoup and coup.

Along with intracranial pressure waves, the blast exposure causes shear waves. These waves in turn propagate through skull contents relatively slower than the compressive waves. In particular, as displayed in Fig. 5a, the effective stress (second deviatoric stress invariant, commonly implemented in the Von-Mises criterion), tends to localize at material interfaces such as gray matter and cerebrospinal fluid, and gray matter and white matter. Since these interfaces are the boundaries between two materials with different shear stiffness, they become focal loci for the concentration of shear stresses. In particular, at around 1.22 ms after blast interaction, the deviatoric stress within the gyri at the contrecoup experience deviatoric stress levels of up to 15 kPa. As reported elsewhere, this deviatoric stress can cause deterioration in the neuronal cytoskeleton structure of the brain soft tissue on a microscopic level [9].

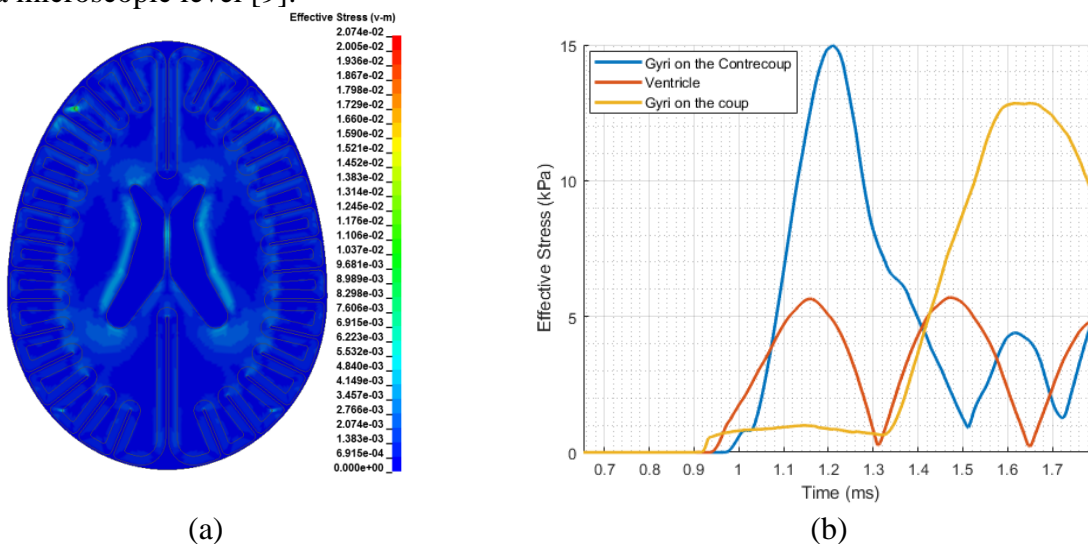


Figure 5. (a) The distribution of effective stress within brain at 1.74 milliseconds (b) Comparison of effective stress for the locations the contrecoup through the coup.

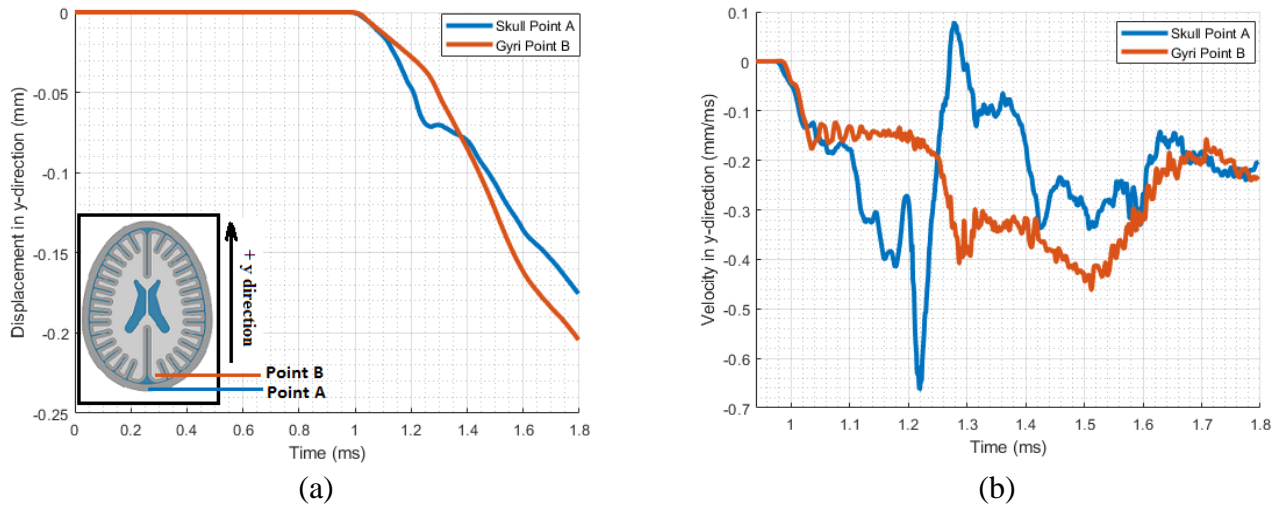


Figure 6. Comparison of the displacement (a) and velocity (b) profiles of the points chosen from the back side of the skull and the gray matter located on the contrecoup portion of the head.

Figure 6 shows a comparison of displacement and velocity histories collected from the skull and gray matter. The total displacement difference in $-y$ direction predicts as 0.03 mm at the end of the simulations. These differences can produce a localized force around the gray matter as a consequence of the brain's movement against the skull. Two instantaneous velocity increment is observed in the gray matter at 1.02 and 1.28 milliseconds, respectively.

Conclusion

The purpose of this work is to understand the effects of intracranial stress waves induced by a blast exposure. The numerical approach proposed herein includes a simplified model of the skull, CSF, and brain tissues to explore the underlying mechanisms for blast-induced TBI. A wealth of unsteady stress-strain events is observed after the blast-head interaction. For instance, right after the initial interaction, the coup experiences a relatively high compressive stress level, closely followed by a tensile stress at the contrecoup. Of interest, while the contrecoup experiences predominantly tensile stresses during the first 1.45ms, there are two isolated spikes in compressive stress (at 1.3 and 1.4 ms) presumably resulting from the complex interaction of stress waves inside the skull contents. Differences in velocity and displacement of skull and gray matter can potentially induce interactions between these two layers of the head, generating focal places of stress. Finally, it is also observed that the deviatoric stress is generally localized at material interfaces, ostensibly due to their difference in shear stiffness.

Acknowledgements

This work supported in part by NPRP grant No. 8-2424-1-477 from the Qatar National Research Fund (a member of the Qatar Foundation).

References

- [1] Hemphill, M.A., Dauth, S., Yu, C.J., Dabiri, B.E., Parker, K.K., (2015). Traumatic Brain Injury and the Neuronal Microenvironment: A Potential Role for Neuropathological Mechano-transduction. *Neuron*, 85(6), 1177-1192.
- [2] Kushner, D., (1998). Mild Traumatic Brain Injury: Toward Understanding Manifestations and Treatment. *Archives of Internal Medicine*, 158(15), 1617-1624.
- [3] McAllister, T. W., (2011). Neurobiological consequences of traumatic brain injury. *Dialogues in Clinical Neuroscience*, 13(3), 287–300.
- [4] Sotudeh, M.C., Dirisala, V., Karami, G., Ziejewski M., (2009). A finite element method parametric study of the dynamic response of the human brain with different cerebrospinal fluid constitutive properties, *Proc Inst Mech Eng*, 223(8), 1003–1019.
- [5] Taylor, P. A., Ludwigsen, J. S., & Ford, C. C., (2014). Investigation of blast-induced traumatic brain injury. *Brain Injury*, 28(7), 879–895.
- [6] Klug, C., Sinz, W., Brenn, G., Feist, F. Experimental sphere-in-sphere testing for the validation of a numerical cerebrospinal fluid model. 2013 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury. 483-496.
- [7] Budday, S., Sommer, G., Birkl, C., Langkammer, C., Haybaeck, J., Kohnert, J., Bauer, M., Paulsen F., Steinmann, P., Kuhl, E., Holzapfel, G.A., (2017). Mechanical characterization of human brain tissue. *Acta Biomaterialia*, 48, 319-340.
- [8] Wineman, A.S. and Rajagopal, K.R., (2000). *Mechanical Response of Polymers: An Introduction*. Cambridge University Press.
- [9] Strich, S.J., (1956). Diffuse degeneration of the cerebral white matter in severe dementia following head injury. *Journal of Neurology, Neurosurgery, and Psychiatry*, (19), 163–185.