# Computational Mechanics of Traumatic Brain Injury under Impact Loads

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***Abstract*—A finite element (FE) study was performed to investigate the dynamic response of the brain under impact loading using computational mechanics to better understand the mechanisms of impact induced traumatic brain injury (iTBI). North Dakota State University Finite Element Head Model (NDSUFEHM) was used to investigate the pressure and stress responses of the brain under different impact conditions. The impacts were carried out at a 45 º-tilted orientation using two different impact velocity, 10 m/s and 13 m/s, which resulted in a total of two different impact scenarios. LS-Dyna nonlinear FE solver and LS-PrePost were employed to perform all simulations, record data and visualize results. Specifically, the intracranial pressure (ICP), maximum shear stress (MSS), were recorded and analyzed for two different impact velocities. These biomechanical responses were recorded at different locations on and inside the brain to starting from the impact site (coup) to the opposite site (countercoup). This was done to analyze the variations of ICP and MSS through the brain in order to understand the role of these parameters in injury mechanisms. The impact severity was shown to have more effect on the level of pressure response while its effect on peak MSS was not much. ICP variation was linear between coup and countercoup sites. It was observed that unlike pressure, shear stress traveled slower through the brain tissue. Our findings suggested that using only one biomechanical parameter can't justify the fidelity of the FE head models.**

***Index Terms*—computational mechanics, traumatic brain injury, shear stress, intracranial pressure, finite element analysis**

## I. Introduction

Impact induced TBI (iTBI) which accounts for 75% of the TBI cases, is a localized injury which results in from contact of the brain with the skull due to the relative motion of the brain with respect to skull upon acceleration/deceleration of the head upon impact. This type of brain injury, is visible and can be detected by routine imaging techniques such as Magnetic resonance imaging (MRI) or Computed Tomography (CT) scans. The iTBI can appear in the form of skull fracture, rupture of blood vessels inside and on the brain surface (intracerebral and subdural hematoma), and cerebral contusion [1]. The severity of iTBI varies by several parameters such as the anatomy of the heads, the intensity of the force applied on the head, and the location of impact [2]. The iTBI mainly involves two stages of brain injury: primary and secondary brain injuries. Primary injury refers to the damage of the brain tissue and blood vessels at the instant of the assault due to the structural displacement of the brain [3]. On the other hand, the second injury which is perceived as the indirect mechanism injury, involves the deterioration of the cellular activities such as blood-brain barriers damage and dysfunction of neurons. The common causes of iTBI are the car accidents, falls, and sport-related accidents such as football and hockey. Several in vivo animal studies, in vitro experiments, human cadaver and volunteer tests have been performed on the impact induced TBI [4,5]. As kinematic-based metrics rely on the linear and angular accelerations of the head, they neglect the material properties and behavior of different head components, especially the brain. Due to structural inhomogeneity of the human head and the intrinsic differences of head components in terms of shape, material, and tolerance, different parameters such as the impact velocity (intensity), location (directionality), the type of blunt impact (struck against or struck by) can affect the mechanical response of the head [6,7]. Accordingly, these variations would influence the stress and strain wave propagation, as well as the intracranial pressure (ICP) gradient distribution throughout the brain tissue, which would produce different levels of injury. As a common criterion in the literature, the head acceleration is utilized for defining

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various injury thresholds [8]. It is postulated, however, that the head trauma, such as diffuse brain injury, subdural hematoma, and contusion, is mainly associated with brain tissue parameters such as shear stress rather than head kinematics [9,10]. Impact loads which propagate inside the cranium and pass through different layers of head components also create stress waves. It is stated that the dynamically induced pressures at the coup and countercoup sites may generate stresses high enough to cause injuries in the brain [11]. Fig. 1 shows potential injury mechanisms for impact-induced TBI in a schematic manner. The dynamic loads in the form of dilatational and distortional stress waves have also been proposed as main contributors to TBIs [12,13]. However, due to moral and technical complexities in evaluation of tissue response of human head, the kinematical parameters of the head motion under assault are commonly used as the injury predictors.

Nahum and Smith [14] performed cadaveric impact tests on 10 different seated cases. They used several impactors with masses ranging between 5.18 to 23.09 kg and constant impact velocities varying from 3.56 to 12.5 m/s. The head was inclined 45 degrees about its horizontal plane and a frontal impact was carried out on the frontal bone and brain ICP was recorded. The peak force delivered to the head was reported to be between 2.9 to 12 kN with a duration of 3-18 ms. They reported the peak positive pressure at the coup site and the peak negative pressure at the counter-coup site and reported a linear relationship between the linear head acceleration and the ICP. Stalnaker et al. [15] carried out cadaveric tests on 15 seated cadavers using a 10 kg impactor. The peak force was recorded as 4.2 and 14.6 kN with the durations of 3.2 and 10.6 ms. They reported a linear acceleration of 140 and 532 G's and reported the peak ICP as 140 kPa.

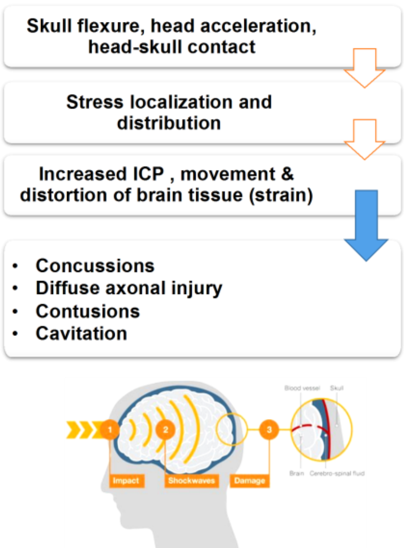


Figure 1. Mechanisms of impact-induced TBI and possible injuries

Despite such valuable studies, the cause of TBIs is too complex to be experimentally explained. Accordingly, mathematical modeling and computational algorithms have found a great attraction in the past few decades. One of the most effective and advanced computational mechanic methods for studying the mechanical response of the human brain to dynamic loads such as impact and blast, is the finite element analysis (FEA) [16-18]. FEA codes have been extensively used not only to evaluate the kinematics of the head, but also to assess the deformation of the soft tissue of the brain and different head components to provide estimates of the stresses and strains inside the brain.

Zhang et al. [5] reconstructed actual American football field incidents and used their kinematical data, as the input for their complex FE head model, to evaluate the mechanical response of the head. They recorded the shear stress and pressure response of the brain to several impacts and proposed some injury predictors and mTBI thresholds based on both kinematical and tissue-level parameters of the head. High shear stress levels were observed at the brainstem and thalamus locations. The linear head acceleration was found to have a greater effect on the ICP response of the brain. Finally, a threshold value of 7.8 kPa for the shear stress was asserted as the tolerance level for 50% probability of mTBI. El Sayed et al. [19] used a 3D FEHM, to carry out impact simulations for frontal and oblique impacts to evaluate the brain tissue response. The skull and the CSF were modeled by a hyperviscoelastic constitutive model. They reported that with respect to the frontal impact, the brain tissue predicted higher pressure responses at both coup and countercoup sites for the oblique impact, which was indicative of focal and diffuse damages in these sites. Sarvghad-Moghaddam et al. [2] investigated the effect of directionality on the response of the brain to impact loads. They used a 3D FE head model with neck attached. Using identical impact velocity of 2.2 m/s, they impacted the head model from front, back and side and reported that while the predicted tissue responses in the front and side impacts were quite close, the brain response to the back impact was significantly lower than those predicted in other scenarios, especially in terms of the shear stress. Their results confirmed directional dependence of the head response as they observed that for the back impact, the head was less prone to severe injuries, while front and side impacts predicted severe injury conditions.

The main focus of current study is to investigate the mechanisms of impact-induced traumatic brain injury using computational mechanics in terms of the biomechanical responses, ICPs and shear stresses, of the FE head to impact. Variations of ICPs and shear stresses in different parts of the brain were monitored and recorded while the head model was impacted with two different impact velocities.

## II. Computational Methodology and Matrials

### A. FE Head Model

The human head model used for this research was originally developed by Horgan and Gilchrist [20] from the Computed Tomography (CT) data provided by Visible Human Project, by 0.3 mm increment in the coronal plane. After stacking up the CT data, thresholding and interpolation techniques were performed to identify the voxels representation of the tissue. Smooth triangle surfaces of the head were then created by interpolation through the voxels. The CT data was employed to make a polygonal model of the head using VTK software. After smoothing the polygonal model, it was converted into IGES format and was imported in the MSC/Patran for meshing. The North Dakota State University Finite Element Head Model (NDSUFEHM), was developed in 2010 by adding the neck bone and skin, facial skin. The head model was discretized using HyperMesh using a total of 38,379 shell and brick elements. The anatomical components, as well as the discretization properties of all the head/neck components are presented in Table I. NDSUFEHM includes the major anatomical features of the human head and neck such as scalp, skull, dura mater, falx, tentorium, pia mater, CSF, brain, neck- bone, neck muscle, facial bone, and facial skin (Fig. 2).

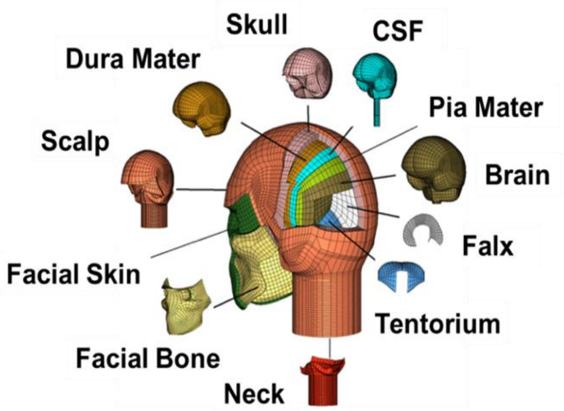


Figure 2. FE discretization of human head components for NDSUFEHM.

One of the challenges in developing exact FE models of the human head is the accurate modeling of the interfaces among different head components. To this end, it is very important that the CSF layer is properly modeled. As mentioned before, CSF is a water-like fluid layer which surrounds the cranial space and separates the skull from the brain. It acts as a natural shock absorber which provides the relative displacement of the brain with respect to the skull [21]. In our research, CSF was modeled using 8-noded brick elements with fluid-like properties, which allows for the relative skull/brain motion, agrees with impact experiments [22]. To comply with the anatomical characteristics of the head, while tied surface-to-surface contact algorithm was employed to replicate the interfaces between the scalp and skull, the skull and dura, and the brain and pia, tied node-to-surface was employed for defining the contacts among the dura-falx-tentorium components, which provides an application of both compressive and tensile loads on the brain [1].

### *B. Material Models*

Developing an accurate constitutive model for the brain tissue has always been a challenge and numerous studies have been performed to improve the modeling of head components in order to better predict the human head behavior under different loading conditions [23,24].

In the current study, linear elastic constitutive properties used for NDSUFEHM components such as the scalp, skull, pia and dura mater, tentorium, CSF, as well as the neck bone and muscle are adopted from the works of Zhang et al. [5] and Horgan and Gilchrist [20]. The mechanical properties of these components are described in Table II. However, a hyper-viscoelastic material model is utilized to better capture the nonlinear behavior of the brain tissue. This model, as also employed by Chafi et al. and Moghaddam et al. [1,22], develops effective constitutive properties in terms of the large deformations due to the dynamic blast loading. Intracranial pressure, defined as the pressure imposed by CSF, and blood on the brain is one of the major parameters in evaluation of the TBI as elevated ICP can induce severe neurological damages.

TABLE I. FINITE ELEMENT CHARACTERISTICS AND EMPLOYED MATERIAL MODELS FOR HEAD COMPONENTS

| Tissue | Constitutive model | FE model | # of Elements |
|---|---|---|---|
| Scalp | Linear elastic | 6 mm Solid element | 5938 |
| Skull | Linear elastic | Solid element | 8305 |
| Dura, falx, tentorium | Linear elastic | 1 mm thick shell element | 2590 |
| Pia | Linear elastic | 1 mm thick shell element | 2754 |
| CSF | Linear elastic & Viscoelastic | 1.3 mm thick solid element | 3354 |
| Spinal cord | Linear Elastic | Solid elements | 496 |
| Brain | Hyperviscoelast ic | Solid element | 7302 |
| Neck bone | Linear elastic | 6 mm thick solid element | 496 |
| Neck Muscle | Linear elastic | Solid element | 3772 |
| Facial Bone | Linear elastic | 2mm Solid element | 1124 |
| Facial Skin | Linear elastic | Solid element | 2248 |

Intracranial pressure is defined as the hydrostatic pressure imposed on the brain mainly due to the fluid-like behavior of the tissue. The hydrostatic pressure, P, is the mean stress of three principal stresses $\sigma_1, \sigma_2$ *and* $\sigma_3$ [24]:

$$P = -\frac{\sigma_1+\sigma_2+\sigma_3}{3} \quad (1)$$

TABLE II. MECHANICAL PROPERTIES OF HEAD COMPONENTS

| Head Component | Density (g/cm3) | Elastic Modulus (GPa) | Poisson's Ratio |
|---|---|---|---|
| Scalp/ Skin | 1.2 | 0.0167 | 0.42 |
| Skull | 1.21 | 8.0 | 0.22 |
| Dura, falx, tentorium | 1.133 | 0.0315 | 0.45 |
| Pia mater | 1.133 | 0.0115 | 0.45 |
| Facial bone | 2.10 | 5.54 | 0.22 |
| Cervical Vertebrae | 2.5 | 0.354 | 0.3 |
| CSF | 1.004 | 2.19 Bulk Modulus | 0.499 |

In current material modeling, the Mooney–Rivlin formulation is adopted for describing the hyperelastic nonlinear behavior of the brain tissue while the Maxwell constitutive law is applied towards modeling the linear viscoelastic behavior of biological tissues. Accordingly, the resulting Cauchy stresses from both approaches are employed to describe the brain response under applied loadings. The material model used in this study correlated perfectly to the intrinsic behavior of the brain regarding its mostly incompressible behavior and sustainability under large deformations. The parameters for the Mooney-Rivlin constitutive equation are determined from the work of Mendis et al. [23].

The strain energy density function governing the Mooney-Rivlin constitutive approach for the incompressible, homogenous and isotropic materials is described by:

$$W = C_{10}(\bar{I}_1 - 3) + C_{01}(\bar{I}_2 - 3) + \frac{K}{2}(J_3 - 1)^2 \quad (2)$$

where $C_{10}$ and $C_{01}$ and K represent the material constants evaluated experimentally, $\bar{I}_1$ and $\bar{I}_2$ are the first and second invariants of the Cauchy tensor and $J_3$ is the elastic volume ratio. The Cauchy stress corresponding to the nonlinear hyperelastic model is obtained as:

$$\sigma = -\frac{\partial W}{\partial \varepsilon} \quad (3)$$

where $W$ is the strain energy potential of Mooney-Rivlin and $\varepsilon$ represents the Green's strain tensor. The parameters for the Mooney-Rivlin constitutive equation is determined from the work of Mendis et al. [23] and are shown for the brain tissue in Table III. On the other hand, applying the Maxwell model for the linear viscoelastic behavior of the tissue, one can find the resulting Cauchy stress tensor:

$$\sigma_{ij} = JF_{ik}^{T} \cdot S_{km} \cdot F_{mj} \quad (4)$$

where J depicts the transformation Jacobian, F indicates the gradient tensor of deformation and S is the second Piola–Kirchhoff stress tensor. The rate effects captured by this constitutive model are formulated using a convolution integral in terms of the second Piola–Kirchhoff stress tensor, $S_{ij}$ and the Green's strain tensor, $E_{kl}$ as follows:

$$S_{ij} = \int_0^t G_{ijkl}(t-\tau)\frac{\partial E_{kl}}{\partial \tau}d\tau \quad (5)$$

with $G_{ijkl}(t-\tau)$ indicating the relaxation functions at various stress levels.

The Prony series can be used to describe the relaxation functions which are employed to evaluate the Cauchy stress tensor:

$$G(t) = G_0 + \sum_{i=1}^{n} G_i e^{-\beta_i t} \quad (6)$$

where G is the shear modulus and $\beta$ is the decay parameter [14]. The data has been widely used in related literature and has shown to provide reasonable results [22, 25]. In order to prevent the rigid body motion and keep the stability of the solution, the inferior surface of the neck is constrained in all directions (Fig. 3).

TABLE III - MECHANICAL PROPERTIES OF HYPER-VISCOELASTIC BRAIN MATERIAL

| $C_{10}$ (Pa) | $C_{01}$ (Pa) | $G_1$ (kPa) | $G_2$ (kPa) | $\beta_1$ ($s^{-1}$) | $\beta_2$ ($s^{-1}$) | K (GPa) |
|---|---|---|---|---|---|---|
| 3102.5 | 3447.2 | 40.744 | 23.385 | 125 | 6.6667 | 2.19 |

## C. Impact Modeling

All the impact simulations were carried out using a rigid cylindrical impactor with a mass of 5.6 kg. Studies have shown that besides the severity of impact which clearly affects the brain response, the direction of impact would also change the brain tissue response [2,5]. The impacts were carried out at a 45 º-tilted orientation using two different impact velocity, 10 m/s and 13 m/s, which resulted in a total of two different impact scenarios.

NDSUFEHM was impacted at 45 º from Frankfort plane to generate a contact force similar to the one in the study of Nahum et al. [14] with the maximum force of around 8 kN on the skull, and the brain responses under the applied force were monitored. Ward et al. [26] proposed an injury criterion based on the ICP threshold of 173 to 235 kPa (i.e., moderate to severe injury corresponding to AIS 3–4 and AIS 5–6, respectively) for TBI. According to this criteria, the predicted tissue responses for the loading employed in this study approached the moderate TBI threshold. The second impact scenario was simulated by impacting the head model in the same orientation but at a higher velocity of 13 m/s. The aim was to create contact force of around 10.1 kN on the skull in these two impacts in order to create injury levels of severe TBI, or fatality. The waveforms of the brain tissue responses in terms of ICP and shear stress were recorded at different locations on and inside the brain and compared. In addition, the peak pressure and shear stress values were monitored at seven equally spaced locations in the brain across a line on the sagittal plane, connecting the coup and the countercoup sites (Fig. 3). The results could help gain a better understanding of the mechanism that happens inside the brain as a result of blunt impacts.

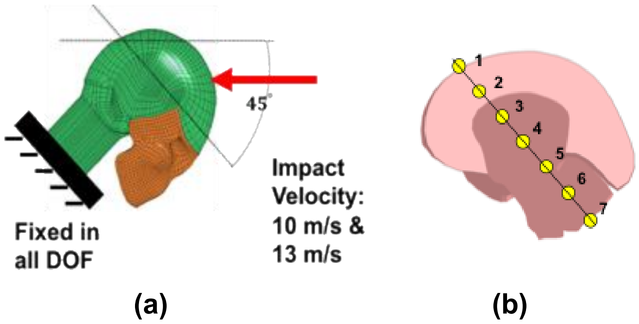


Figure 3. (a) Boundary conditions and head orientation at 45 °from Frankfurt plane; (b) sagittal plane of the brain and the schematic view of the locations that the biomechanical parameters are extracted.

## III. Results and Discussion

### A. ICP Response of the Brain

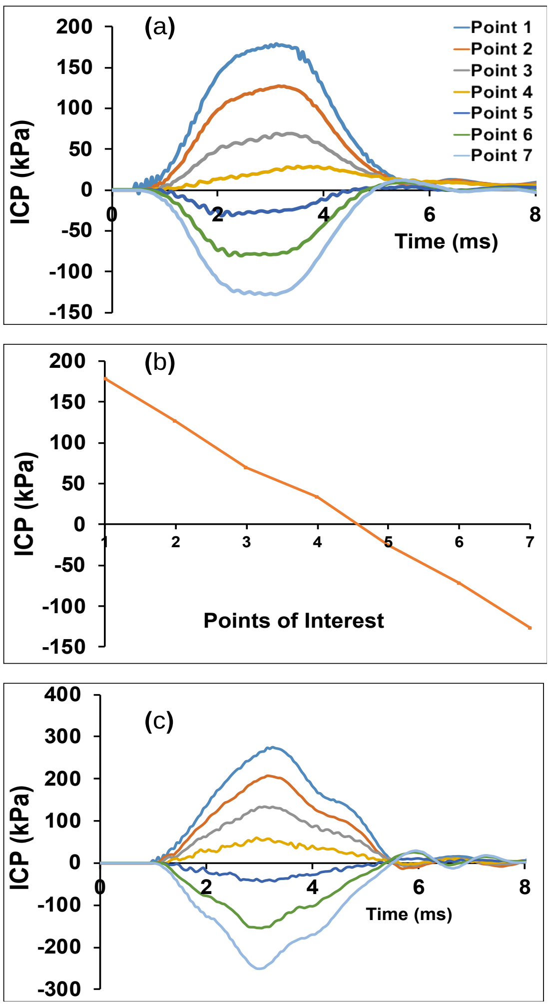


Figure 4. (a) ICP variations at different locations inside the brain on the connecting line of coup-countercoup sites when the impact velocity is 10 m/s; (b) Maximum pressure at different locations of the brain from coup (point 1) to countercoup site (point 7); (c) ICP variations at different locations for 13 m/s impact velocity

The coup-countercoup ICPs were recorded for both simulations. As illustrated in Fig. 4(a), the responses of the brain were monitored at 7 equally distant locations on the coup-countercoup connecting line, with the first point at the site of impact (coup), and the 7th point located at the opposite site of the impact (countercoup). The highest positive and negative pressure values were observed at the coup and countercoup regions, respectively, and started to decrease towards the mid-brain region. The peak ICP values, recorded at different locations of the brain across the so-called connecting line, are compared in Fig. 4(b). The pressure response showed a linear variation inside the brain from the site of impact (point 1) to the opposite site of impact (point 7) and approached zero somewhere in the middle of the brain (between point 4 and 5), which was indicative of the interaction of positive and negative pressures. The maximum inflicted pressures always happen on the outer surface of the brain at these two sites and as far as the ICP thresholds are concerned, the amount of pressures inside the brain is, therefore, of no interest.

In the second impact scenario, the simulation was replicated in the same orientation, but the velocity of the impactor was increased by 30% to about 13 m/s. Fig. 4(c) illustrates the variation of pressure at the same locations across the coup-countercoup connecting line for the high-velocity impact at 45 °from Frankfort plane.

### B. Shear Stresss Response of the Brain

Fig. 5(a) shows shear stress variations inside the brain at the aforementioned locations. A phase lag was observed in the generation of maximum shear stress waves, due to the low shear modulus of the brain tissue, with respect to the ICP pattern. To better understand the effect of impact severity on the tissue response of the brain, the impact velocity was increased to 13 m/s in the previous scenario, while other conditions were kept the same. Fig. 5(b) represents the shear response for the high-velocity impact.

The shear response followed a pattern similar to those for the low-velocity impact in Fig. 5(a). The maximum shear stresses were 1.6 and 2.5 kPa in the low and high velocity impacts, respectively.

Fig. 6 illustrates the distribution of ICP and shear stress in different regions of the brain. This Fig. clearly shows the band like layout of the pressure distribution all over the brain such that different regions are made. However, for shear stress, the distribution is not uniform and the maximum occurred at the brainstem.

The contact force on the head induced by a blunt impact leads to the sudden head motion and, as a result, the relative normal displacement of the brain with respect to the skull [7]. It was previously suggested that the relative brain/skull motion could contribute to the development of ICP variations inside the brain [7, 27]. Our ICP results showed that the variation pattern of pressure from coup to countercoup site is linear. The increase in the velocity of the impactor by 30% to about 13 m/s, at which serious injury or death occurs [26], contributed to similar results by the head models, emphasizing the independency of ICP variation pattern on the severity of the impact load. Figs. 4(a) and 4(c) illustrate how fast the pressures created at both coup and countercoup sites can travel throughout the brain. The rise in the respective positive and negative pressures at points 1 and 7 was observed to occur almost simultaneously. It was concluded that the positive and negative pressures at both sides of the brain, generated due to the movement of the whole brain inside the skull, traveled quickly toward the middle of the brain in the form of stress waves [28], as shown in 4(a) and 4(c). Due to the large bulk modulus of brain tissue compared to its shear modulus, the pressure responses within the cranium reached hydrostatic balance almost instantaneously [21]. Therefore, the pressure responses at different locations inside the brain reached their peak values almost at the same time.

While the determination of ICP by numerical methods appears relative straightforward, the detection, evaluation, and validation of the shear stress are quite complicated. Shear stress has been widely investigated in many studies and several injury thresholds have been defined based on that [21]. Fig. 5 shows how shear waves propagated throughout the brain tissue. Due to the low shear modulus of the brain tissue (compared with its high bulk modulus), the shear waves propagated slowly, when compared to the pressure wave propagation [28]. Similar to the pressure behavior, at the instant of the impact, shear stresses were also generated at the coup and countercoup sites of the brain, and propagated from both sides toward the middle of the brain. However, unlike the attenuating pattern of the pressure wave, shear stresses were notably amplified inside the brain. It was concluded, therefore, that the determination of shear stresses inside the brain needed a thorough validation of the head models against the wave propagation.

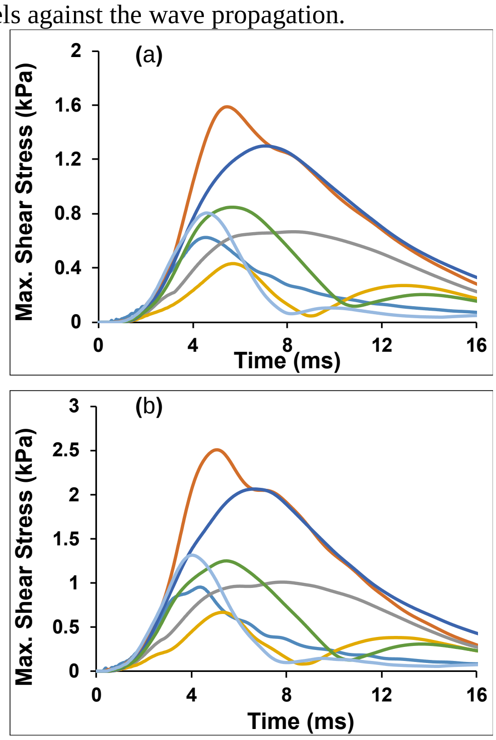


Figure 5. Maximum shear stress responses of the NDSUFEHM for (a) low-velocity impact; (b) High-velocity impact

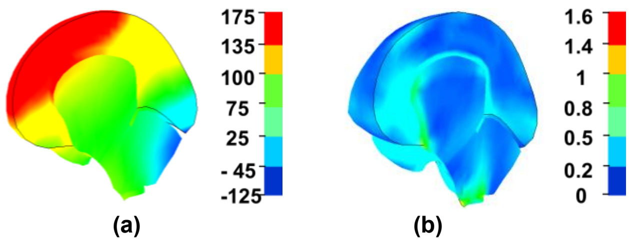


Figure 6. (a) ICP and (b) shear stress distribution in the brain upon impact at 10 m/s velocity

It has been postulated that the development of pressure gradients inside the brain can give rise to shear stresses deep inside the brain [29]. In other words, an accurate determination of maximum shear stresses (proportional to the difference between the first and third principal stresses) requires a precise evaluation of pressure waves. As the shear modulus of the brain (on the order of kPa) is far smaller than its bulk modulus (on the order of GPa), a slight deviation of principal stress values (or pressures) inside the brain may result in a greater deviation in the prediction of maximum shear stresses. Another consideration in affecting brain maximum shear stress could be the element size [28] as stress concentration may occur adjacent to large elements.

In terms of the risk of injury, it is seen that the higher speed impacts expose the brain to higher risks of concussive injury. Zhang et al. [5] found 235 kPa to be threshold for mild TBI. While the peak coup ICP was 175 kPa for the 10 m/s, it was increased to about 290 kPa for the 13 m/s impact which put the brain at a serious risk of concussive injuries. Maximum shear stress is used as the criterion for DAI. Zhang et al. [5] found the threshold for disuse injury to be shear stresses exceeding 7.8 kPa. For both the MSS was under 3 kPa in both impacts so risks of DAI was assumed to very low.

Moreover, it was observed that there was no correlation between the impact intensity and the ICP and MSS peak values. This could be due to the difference in shape, material, and functions of different parts of the brain.

## IV. Conclusion & Future Works

NDSUFE head model was used to study the mechanism of impact-induced TBI as it was impacted by an impactor at two 10 m/s and 13 m/s velocities. The maximum and minimum pressures were found on the brain surface and were shown to vanish toward the center of the brain. The ICP varied linearly from the coup to the counter coup sites inside the brain very quickly such that the whole brain tolerated the pressure simultaneously. Shear stress, on the other hand, showed a very different pattern compared to that of the ICP. While the pressure vanished quickly after a few milliseconds, the shear stress showed a phase lag and travelled inside the brain much slower due to the brain viscous behavior. Additionally, several milliseconds after the external loads and pressures disappeared, the brain experienced significant shear stress peaks at several locations.

Therefore, a head model only validated against ICP cannot be considered as sufficiently validated against other mechanical parameters relevant to brain injury. While our research addresses a very significant challenge, the authors believe some future works can further boosts this study. We plan to incorporate other FE head models and compare our results with them to investigate the accuracy of different FE head models. Furthermore, it would be beneficial to include impacts from other directions and it higher intensities to investigate the effects of impact velocity and direction. Finally, reconstruction of real impact events such as those happening in football or hockey can expand the scope of this study.

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