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THE EFFECT OF SHOCK WAVE ON A HUMAN HEAD

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ABSTRACT

When a pressure wave of finite amplitude is generated in air by a rapid release of energy, such as high-pressure gas storage vessel or the blast from dynamite, there may be undetected brain injuries even though protective armors prevent the penetration of the projectile. To study brain tissue injury and design a better personnel head armor under blast wave, computational models of human head have been developed. Models with and without helmet are built to quantify the intracranial pressure and shear stresses of head subjected to blast wave. All the models are compared against injury thresholds for intracranial pressure and shear stresses. Overall pressure and shear stress level is highest in model without helmet and lowest in model with helmet having foam layer on inner side of helmet.

The results show that helmet reduces the pressure and shear stresses generated in the brain. However this reduction in pressure and shear stresses might not be sufficient to mitigate early time, blast induced, traumatic brain injury. The validated results will provide better understanding of the energy transfer characteristics of blast wave through helmet and the injury mechanism of human head.

INTRODUCTION

The yearly incidence of traumatic brain injuries (TBI) in the United States has been estimated at 1.4 million people, including 50,000 deaths and 235,000 hospitalizations [1]. The incidence of TBI injuries on the military personnel and civilians have been recently increased due to the tactics of asymmetric warfare, where enemy combatants detonate improvised explosive devices. Recent statistics from the conflict in Iraq shows that several thousand US soldiers have sustained TBI, 69% as a result of blasts [2,3]. The injuries from blast can be broadly divided into primary, secondary and tertiary injuries the details of which can be found elsewhere[4].

TBI has been often linked to traumatic axonal injury, most often referred to as diffuse axonal injury (DAI) [5-6]. The role of brain accelerations in development of diffuse axonal injury have been studied for many years [7-10]. Recently few researchers have shown the role of early time intracranial wave motion in generation of mild TBI [11,12]. They have shown that early time intracranial wave motion can generate significant intracranial pressure, volumetric tension (negative pressure) and shear stresses in the brain which can cause brain disturbance leading to TBI. The effectiveness of protective devices like helmet in reducing pressure, volumetric tension and shear stresses in the brain have been questioned by several researchers [11,13] .

The goal of present study is to understand the effectiveness of helmet in mitigating early time blast induced mild traumatic brain injury. Finite element method is used to characterize the effect of shock wave on human head.

METHODOLOGY

Two dimensional plane strain finite element models of helmet-head under shock loading are studied to compare effectiveness of helmet. Figure 1 describes three comparative models. Model 1 simulates the head (skull and brain) response subjected to the shock loading without protective helmet. Model 2 includes helmet which is in perfect contact with skull. In model 3 we have added layer of foam pad between helmet and skull as shown.

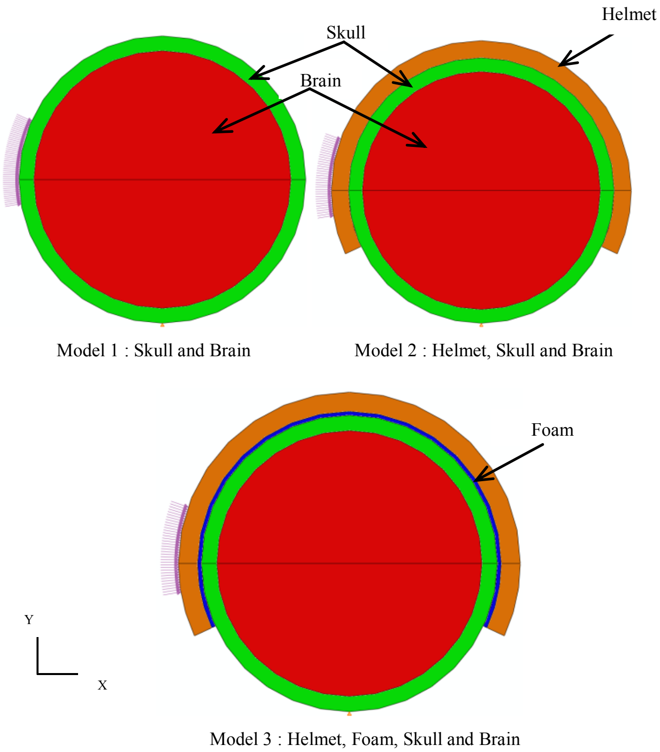


Figure 1. Plain strain models of with helmet and without helmet cases

The brain tissue is modeled as cylindrical core [14] of diameter 138 mm . The skull, helmet and foam are modeled as cylindrical shell. The shell thickness of skull, helmet and foam is 8 mm, 10 mm and 2 mm respectively. These simplified geometries were selected so that the analysis would not be overly complex and prohibitively expensive.

Brain is modeled as a single homogenous material with average brain property. Details of brain like white matter, grey matter, Cerebral Spinal Fluid (CSF), dura mater, pia mater and cerebrum will be added in the future. The brain tissue is modeled as linear, isotropic, viscoelastic material with properties adopted from Taylor et al.[11]. Standard Linear Solid (SLS) model is used to characterize shear response. These properties are within close proximity of material behavior of brain tissue as mechanical properties of brain tissue reported in literature varies over wide range and often subjected to controversy [15].

The skull and helmet are modeled as linear, elastic, isotropic materials based on data reported in the literature [16-23]. We have used average of all reported values for various material parameters like Young's modulus, Poisson's ratio and density. The foam is modeled as elastic, plastic polyurethane foam (crushable foam) with isotropic hardening as shown in Fig. 2 The properties of polyurethane foam are taken from Abaqus Benchmark Manual [24]. The detailed material properties are summarized in Table 1-3.

Table 1. Elastic Material Properties

	Density (kg/m^3)	Young's Modulus (MPa)	Bulk Modulus (MPa)	Poisson's Ratio
Brain	1040	0.123	2.37	0.499989
Skull	1710	5370	-	0.19
Helmet	1380	76000	-	0.3
Foam	60	7.5	-	-

Table 2. Viscoelastic Properties of brain tissue

	Instantaneous Shear Modulus (kPa)	Long-term Shear Modulus (kPa)	Relaxation time (sec)
Brain	41.0	7.8	0.00142857

Table 3. Material Parameters for isotropic hardening of foam

Yield Strength Ratio	1.0
Plastic Poisson's Ratio	0.0

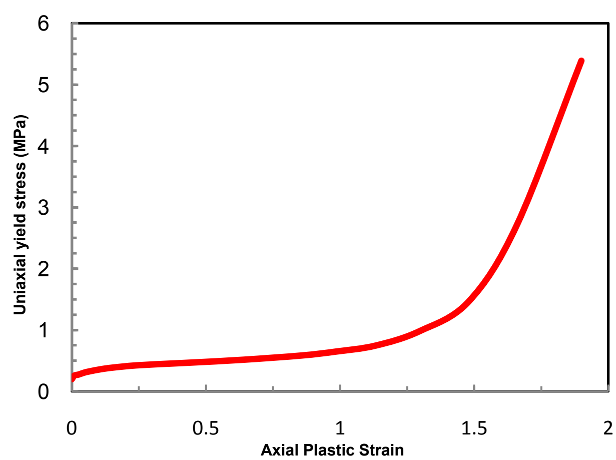


Figure 2. Stress-Strain curve for foam

Dynamic, explicit simulations are carried out in Abaqus 6.8. The skull is constrained at bottom in vertical degree of freedom as shown in Fig. 3.The total simulation time of 2 milliseconds (msec) is used. This time interval was selected by observing the fact that the majority of the intracranial wave mechanics had played out by that time and the associated stress peaks had been established. The pressure pulse with peak pressure of 4 MPa is applied over angle of 30 degrees for 0.05 msec as shown in Fig. 3. The magnitude and profile of pressure pulse is selected based on results of Taylor et al. [11]. They allowed the blast wave propagating in the air to envelope the head and monitored how pressure evolves with time at air skull interface. They reported sudden peak pressure of about 4 MPa at air skull interface.

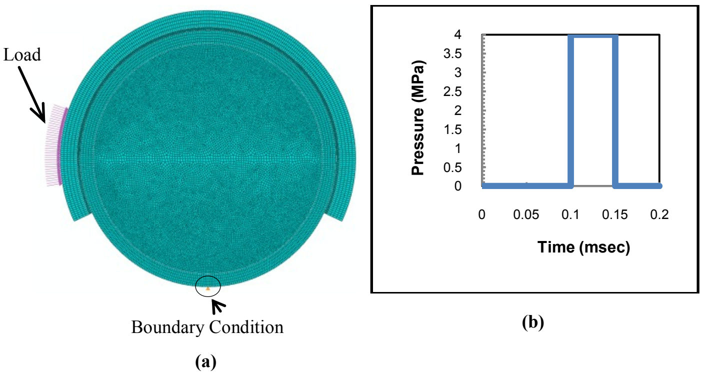


Figure 3. (a) Load and Boundary condition (b) Load applied as pressure pulse which varies with time

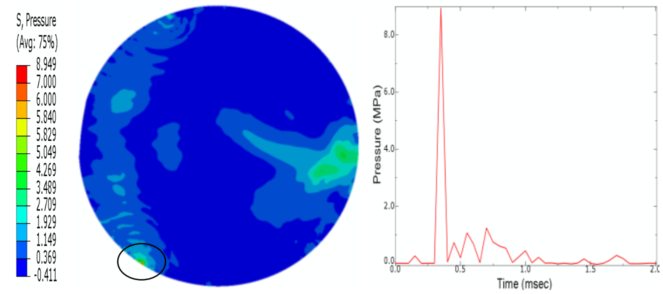
The surface to surface contact with kinematic contact method is used between brain, skull, foam and helmet. The friction coefficient of 0.1 is used between all these components. The model consist of 22,298 elements and 23,367 nodes.

RESULTS AND DISCUSSION

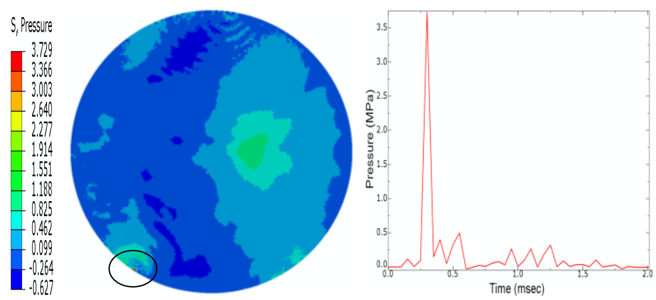
Each simulation data set has tracked calculations of pressure, volumetric tension (negative pressure) and deviatoric (shear) stress over span of 2 msec. In addition to these quantities simulations have also tracked various energies associated with each component. The output is written for every 0.1 msec. Time history output is created from the field output data for the elements where maximum values of pressure, volumetric tension and deviatoric stress have occurred.

Intracranial pressure distribution in the brain at different times for all three models is shown in Fig. 4. It shows that complex wave patterns are generated within the brain. This complex wave pattern is attributed to reflection of waves from finite boundaries , presence of helmet and skull which posses significant shear strength and wave mode conversion at material boundaries and interfaces. The intracranial pressure shows typical coup and countercoup pattern throughout the brain on early time scale (time = 0.15 and 0.25 msec). Once the early waves pass through the brain , a mixed intracranial pressure pattern develops at later time (time > 0.25 msec). This pressure distribution (qualitatively) is consistent with results observed in direct impact events [25]. This coup countercoup mechanism can cause contusion at early time and can widely spread throughout the brain at later time when mixed intracranial pressure patterns are dominant. We can also see from this figure that overall pressure level is highest in model 1 and lowest in model 3.

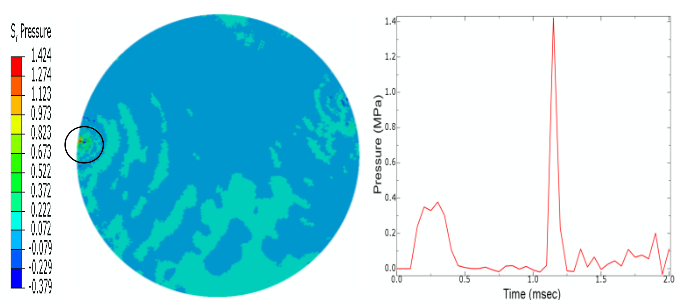
Figure 5 shows time history plots at locations of the brain for which maximum value of pressure is observed for model 1, model 2 and model 3. Difference in peak pressure timing and difference in location of peak pressure shows the complexity of wave motion inside the brain for all the three models. Maximum pressure of 8.949 MPa is observed in model 1 whereas model 2 and model 3 shows maximum pressure of 3.72 MPa and 1.424 MPa respectively. Reduction in maximum pressure from model 1 to model 2 is 58.43 % and that of model 2 to model 3 is 61.72 %.



(a) Model 1



(b) Model 2



(c) Model 3

Figure 5. Time History plots at locations of brain for which maximum pressure is observed. The location where the maximum pressure is observed is circled.

Figure 6 shows time history plots at locations of the brain for which maximum value of volumetric tension (negative pressure) is observed for model 1, model 2 and model 3. Maximum negative pressure of 2.727 MPa is observed in model 1 whereas model 2 and model 3 shows maximum negative pressure of 1.084 MPa and 0.4170 MPa respectively. Reduction in maximum negative pressure from model 1 to model 2 is 60.25 % and that of model 2 to model 3 is 61.53 %.

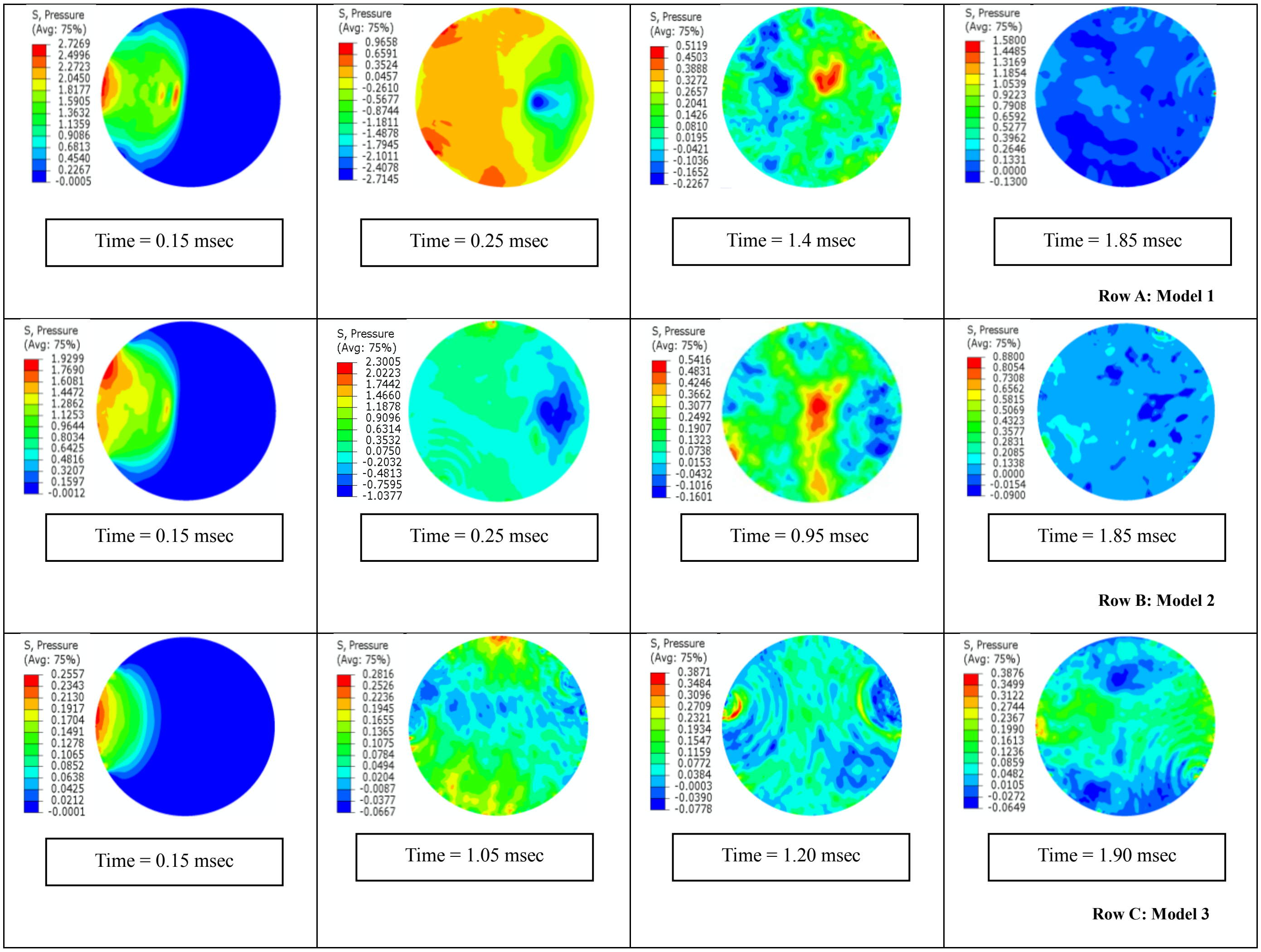
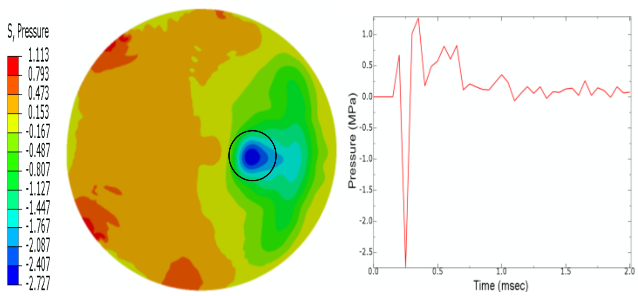
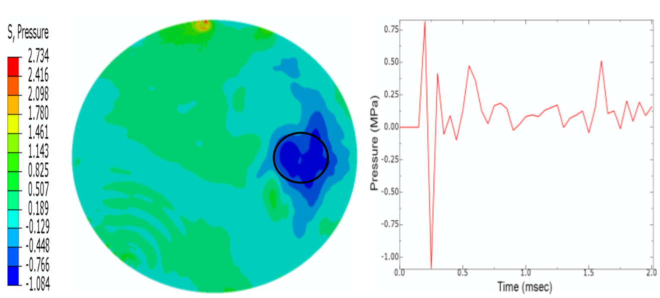


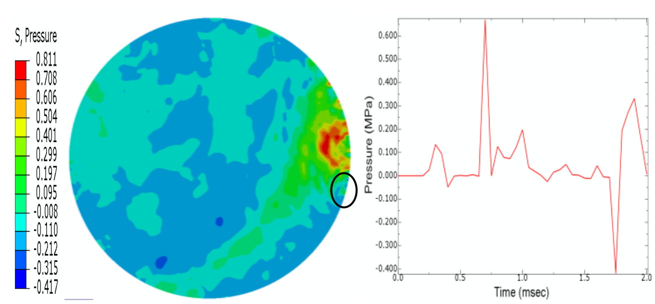
Figure 4. Intracranial pressure distribution in the brain. Coup and countercoup patterns are seen at early times (time < 0.25 msec). Mixed intracranial patterns are seen at later times (time > 0.25 msec)



(a) Model 1



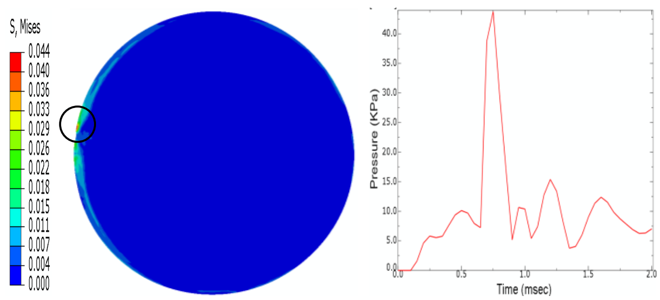
(b) Model 2



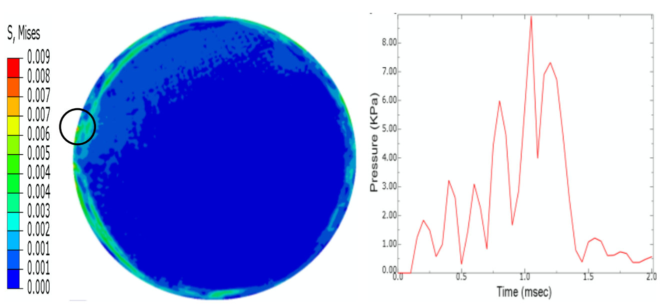
(c) Model 3

Figure 6. Time History plots at locations of brain for which maximum volumetric tension is observed. The location where the maximum volumetric tension is observed is circled.

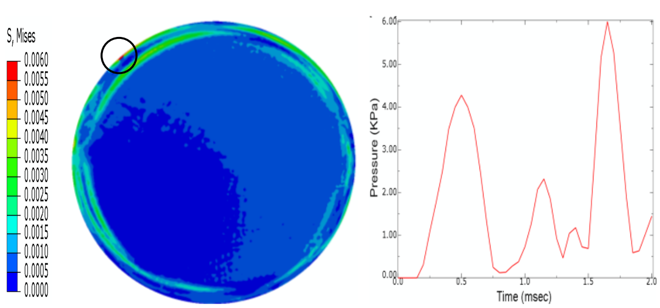
Figure 7 shows time history plots at locations of the brain for which maximum value of deviatoric (von mises) stress is observed for model 1, model 2 and model 3. Maximum deviatoric stress of 43.77 kPa is observed in model 1 whereas model 2 and model 3 shows maximum deviatoric stress of 8.938 kPa and 6.006 kPa respectively. Reduction in deviatoric stress from model 1 to model 2 is 79.57 % and that of model 2 to model 3 is 32.80 %.



(a) Model 1



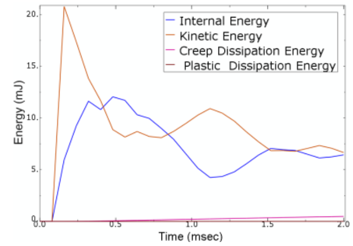
(b) Model 2



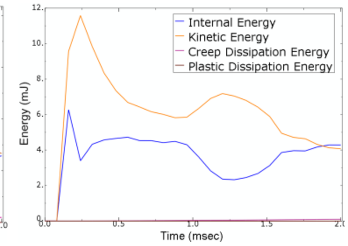
(c) Model 3

Figure 7. Time History plots at locations of brain for which maximum deviatoric stress is observed. The location where the maximum deviatoric stress is observed is circled.

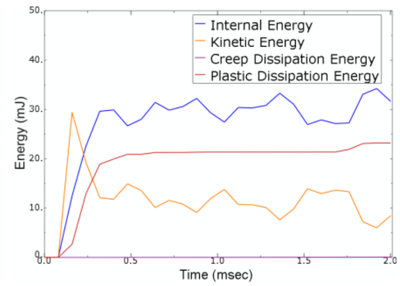
Figure 8 shows comparison of energies in all three models.



(a)



(b)



(c)

Figure 8. Comparison of Energies (a) Model 1 (b) Model 2 (c) Model 3.

As depicted in Fig. 5-7 the maximum values of pressure, volumetric tension and deviatoric stress in the brain reduces by 58.43 %, 60.25 % and 79.57 % respectively from model 1 to model 2. This can be attributed to volumetric and shear strength of the helmet and impedance mismatch between helmet and skull. Furthermore helmet delays the time over which load is applied on the skull. The reduction in maximum values of pressure, volumetric tension and deviatoric stress in the brain from model 2 to model 3 i.e. by adding foam layer of 2 mm thickness between helmet and skull is 61.72 % , 61.53 % and 32.80 % respectively. This can be attributed to energy dissipated in plastic deformation of foam (see Fig. 8) and impedance mismatch between helmet & foam and foam & skull. Foam being soft (low density, low Young's modulus) material reflects most of the incident pressure pulse allowing very little to transmit to skull. This in turn delays time over which load is applied on the skull and hence reducing the stresses incurred in the brain.

The effectiveness of helmet in mitigating mild TBI will be evaluated based on injury threshold criterions for intracranial pressure and shear stress.

We will use following two criterions to study intracranial pressure response:

1. Ward et al. [26] suggested an intracranial pressure injury index to access brain injury severity and the occurrence of the cerebral contusion. According to them, serious brain injury occurs when peak intracranial pressure exceeds 235 kPa and minor or no brain injury, for intracranial pressure below 173 kPa.

2. The bulk modulus of human brain is assumed to be 2.37 GPa which is very close to that of water (2.2 GPa) . This close property can be attributed to high water content of human brain. If this assumption is reasonable then water can be used to simulate the fluidic content of the brain tissue in a head injury model. Lubock and Goldsmith [27] used a spherical shell to resemble the head filled with water to access cavitation theories. They stated that water will form the cavities when exposed to negative pressure below100 kPa.

If we use above criterions to assess our models then all three models exceeds the injury threshold limit. Table 4 summarizes injury threshold limits and maximum values obtained from our models.

Table 4. Intracranial pressure injury threshold and comparison of our models based on this injury threshold

	Model 1	Model 2	Model 3
Maximum pressure, p (MPa)	8.949	3.72	1.424
Maximum volumetric tension (negative pressure), p (MPa)	-2.727	-1.084	-0.4170
Concussion injury Threshold [26]	p > 235 kPa injury p < 173 kPa no injury		
Cavitation injury Threshold [27]	p < -100 kPa		

The deviatoric stresses generated in the brain suggest another injury mechanism for brain. Deviatoric stresses tend to shear material. This deviatoric stresses would result in tearing of cellular membrane. The brain is principally composed of neurons which conduct electrical impulses along their outer membrane, a tear in this membrane would be synonymous with the loss of electrical conductivity and hence functionality. The injury threshold reported by various researchers for deviatoric stresses varies over certain range. Anderson et al. [28] reported that shear stress in the range of 8 to 16 kPa can cause widespread axonal injuries. Their FEM results showed good correlation with site of injury observed in sheep experiments. Kang et al [29], by series of FEM motorcyclist accident simulations, observed that predicted areas of shear stress concentration are correlated to the brain injury locations. They suggested that the shear stress in the range of 11 to 16.5 kPa can be regarded as the brain injury limit. Zhang et al. [30] investigated football collisions and reported that shear stress levels of 3.1 to 6.4 kPa can cause concussive injury and mild TBI. Table 5 summarizes injury threshold limits and maximum shear stress values obtained from our models.

Table 5. Shear stress injury threshold and comparison of our models based on this injury threshold

	Model 1	Model 2	Model 3
Deviatoric (shear) stress (kPa)	43.77	8.932	6.006
Injury threshold Anderson et al. [28]	8-16 kPa		
Injury threshold Kang et al. [29]	11-16.5 kPa		
Injury threshold Zhang et al. [30]	3.1-6.4 kPa		

If we use Anderson's and Kang's threshold limits then maximum shear stress incurred in models 1 and 2 is above threshold whereas shear stress incurred in model 3 is below threshold. However if we use Zhang's threshold limit all three models show shear stress exceeds threshold limit.

Our result shows that helmet with foam layer on inner side of helmet is most beneficial of all the models in regards to intracranial pressure , volumetric tension and shear stresses incurred in the brain. Results reported here are based on two dimensional models hence the conclusions drawn from these results might not be realistic. However these results can be helpful in developing mitigation strategies. The qualitative comparison can give idea about reduction in stresses of the brain in with helmet and without helmet cases.

We have also monitored maximum pressure and maximum deviatoric stress generated in skull. The maximum value of pressure is around 16 MPa whereas maximum value of deviatoric stress is around 20 MPa. The fracture stress reported by Taylor et al [11] for skull is 77.5 MPa. Maximum values of pressure and maximum deviatoric stress for skull are well below this fracture stress hence skull fracture will not occur under present blast load conditions.

CONCLUSION

In this study, we have evaluated effectiveness of helmet in mitigating blast induced traumatic brain injury (TBI). The blast conditions simulated here are equivalent to those predicted for a location 2 to 3 meters distant from a detonated explosive device constructed from 3 kg charge of Octol explosive. The results show that helmet can reduce the pressure and shear stresses generated in the brain. However this reduction in pressure and stresses might not be sufficient to mitigate early time, blast induced, traumatic brain injury. Addition of foam pad of 2 mm thickness between helmet and skull further reduces the stresses generated in the brain. Hence mitigation strategies should focus on materials which can deform plastically hence dissipating energy in plastic deformation. In actual combat scenarios, soldiers can frequently undergo such blast loading. In this regard, layer of material like foam on inner side of the helmet is better, which can undergo large plastic deformations before failure and also has low transmission coefficient. The simulations based on 3D modeling of head and helmet will be better in order to understand mitigation offered by the helmet, which is our future work.

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