Mild Traumatic Brain Injury-Mitigating Football Helmet Design Evaluation

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ABSTRACT

Concussion, as known as mild Traumatic Brain Injury (mTBI), is the most common sport-related head injury. Football is the most common sport with higher concussions in USA. Helmet is the equipment being used in mitigation of mTBI. There are numerous designs of helmets which meet the requirements of sport regulation committee. In this paper, a football helmet is evaluated using numerical methods. The brain and the tissues in human head are modelled using continuum Smoothed Particle Hydrodynamics (SPH). The brain tissues are generated by segmentation from human brain MRI data. The LSTC dummy is used to represent the football players. The brain tissue is fitted in the cavity of the dummy headform. Two different impact scenarios are simulated in this study. The results for these impact conditions are presented.

INTRODUCTION

Concussion, as known as mild Traumatic Brain Injury (mTBI), is the most common sport-related head injury. Centers for Disease Control (CDC) estimates reveal 1.6 million to 3.8 million concussions occur each year in USA. Football is the most common sport with higher concussion risk due to the nature of the sport. A professional football player will receive an estimated 900 to 1500 blows to the head during a season. Impact speed of a football player tackling a stationary player is around 25mph. The topic of concussions and the effect that they have on the human brain are being increasingly scrutinized by media, political personnel and research community. Tremendous amount of research studies have been conducted by medical, defense and academic communities in understanding the etiology of concussion, treatment protocols and mitigation strategies by designing better equipment.

Helmets are being developed and marketed in order to mitigate the mTBI. These helmets meet the requirements of the existing standards by National Operating Committee on Standards for Athletic Equipment (NOCSAE) [1]. Helmet testing to assess the impact performance is documented in Reference[2]. Impact test comparison of 20th and 21st century American football helmets is provides by Bartsch, et al[3]. The helmet designs vary widely. The designs are based upon simple inner foam padding, energy absorbing adapting head protection system, multi-directional protective systems, use of shear-thickening fluid layers,etc. The effects of pad composition, geometry and material stiffness were studied by Moss et al.[4]. Viano, et al studied the effect of mouthguards on head responses and mandible forces in football helmet impacts[5]. Externally applied foam is also used in football helmets for impact reduction [6].

The numerical analysis of the human head models using the finite element technique is being used extensively for the past few decades. A literature review of the finite element human head models which are used in the medical and engineering fields is given by Samaka et al. [7]. Patient specific finite element head models are generated by Johnson Ho [8] based on Magnetic Resonance Imaging (MRI) scans. A multiscale computational estimation of axonal damage under inertial loading of the head was investigated by Wright et al.[9] using two dimensional finite element models of the head constructed from detailed MRI and Diffusion Tensor Imaging (DTI).

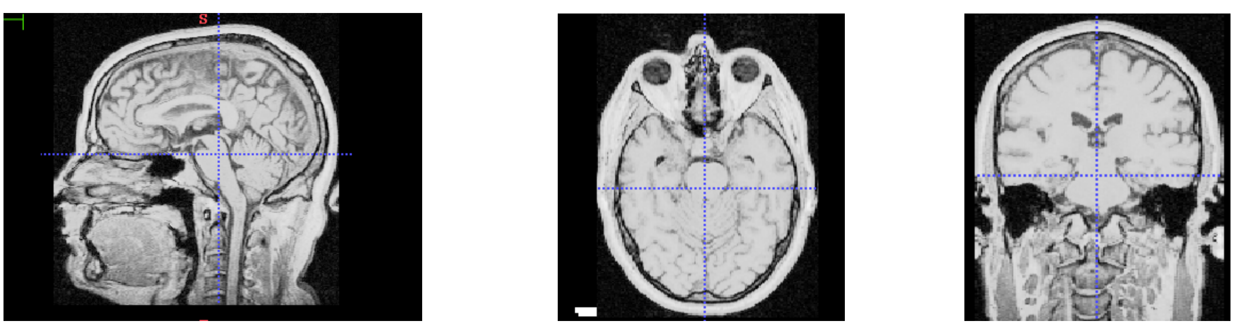
Full-scale anthropomorphic test devices (ATD) that simulates dimensions and weight of the human body are being used in automotive crash testing and occupant protection modeling. The finite element model of ATD for impact analysis is available from Livermore Software Technology Corporation (LSTC).

The brain that basically floats inside the skull is surrounded by cerebral spinal fluid. The head is subjected to linear acceleration and or a rotational acceleration during an impact or a blow. During the linear impact, the brain strikes the inner skull in acceleration and then hits the opposite side of the skull in deceleration and in the rotational acceleration scenario, the brain tissues are subjected to shear due to rotation of the head. In either case, the delicate neural pathways in the brain can become damaged, causing neurological disturbances.

In this study, a generic helmet with foam padding is used in design evaluation for mitigation of brain injury. The brain geometry is developed from MRI image using segmentation technique. The brain is modeled using continuum SPH technique. The player is represented by H3 LSTC 50th percentile dummy.

METHODS

BRAIN MODELING: A brain section is shown in Figure 1.



Sagittal Anterior Coronal

Figure 1. Brain Sectional Views

3D view of the brain tissue is obtained by segmentation of the MRI image using ITK-SNAP [10] program. The SPH model of the whole brain is shown in Fig. 2.

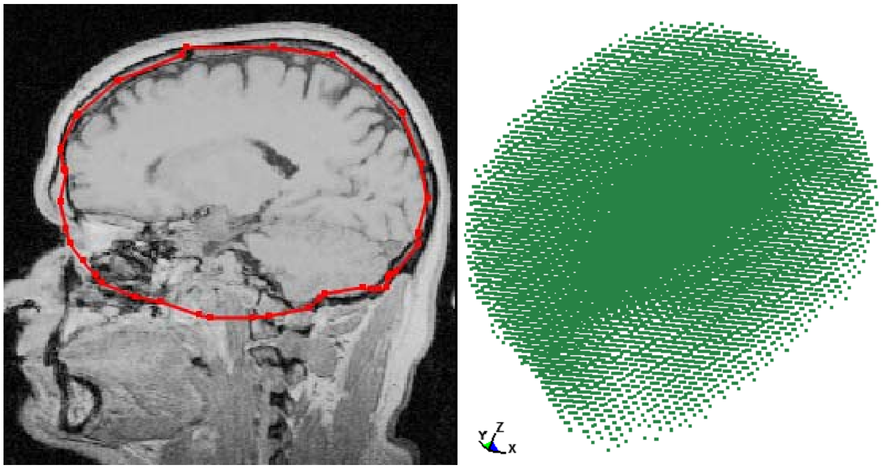


Fig.2 MRI image and approximated SPH mesh.

The material model for approximated brain is assumed as viscoelastic material and the constants used in the analyses are:-
Density = 1100 kg/m^3, Bulk Modulus = 500 MPa, G0 = 2.0MPa, Gi=1Mpa and Beta = 700/sec.

HELMET MODEL: A generic helmet model is shown in Fig. 3.

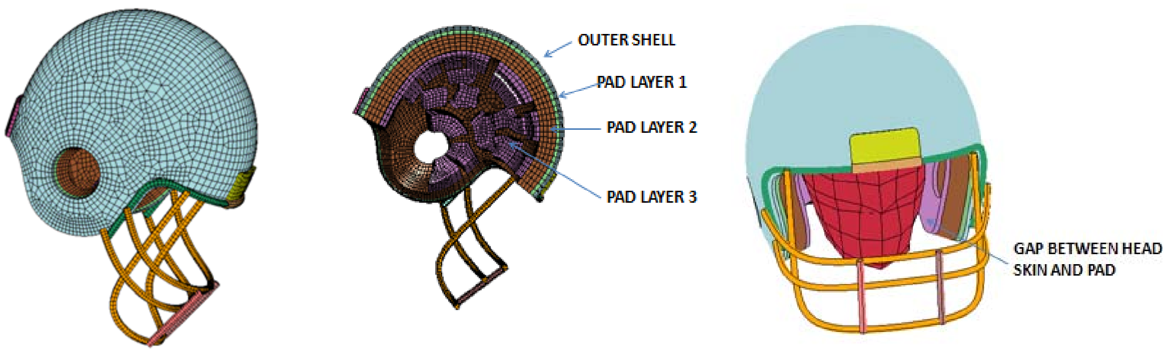


Fig. 3 Helmet Model and Energy Absorbing Pads.

The outer shell is modeled as plastic_kinematic material with Density=1200 kg/m^3, Young's modulus = 1.50GPa, Yield Stress = 80MPa and Tangent Modulus =1.5 MPa.

The energy absorbing pads are modeled as low density EPP foam with varying densities, Distribution of the foam is unsymmetrical and symmetric as shown in Fig 3. The symmetric foam model has some gaps between the skin and the foam pad as shown in the figure 3.

PLAYERS MODEL 1: The players are modeled using LSTC ATD dummy as shown in Fig. 4

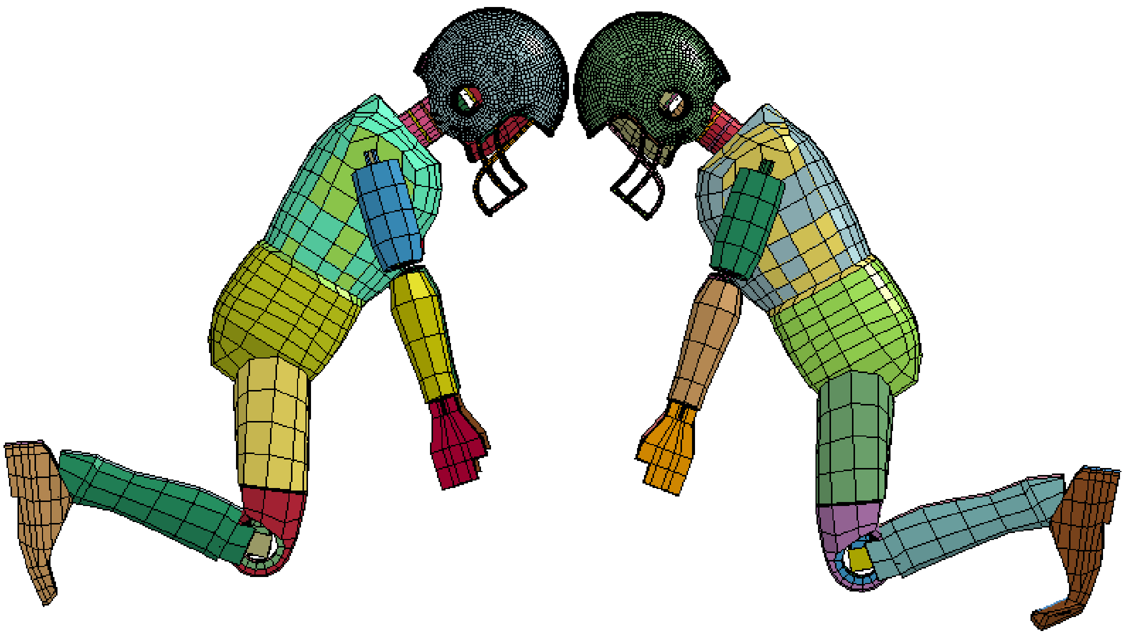


Figure 4. Two opposing players.

In this configuration, the opposing players are moving at 6.704 m/sec, a speed of 6.705m/sec.

PLAYERS MODEL2: Both striking and the struck players are modeled using LSTC ATD dummy as shown in Fig. 5

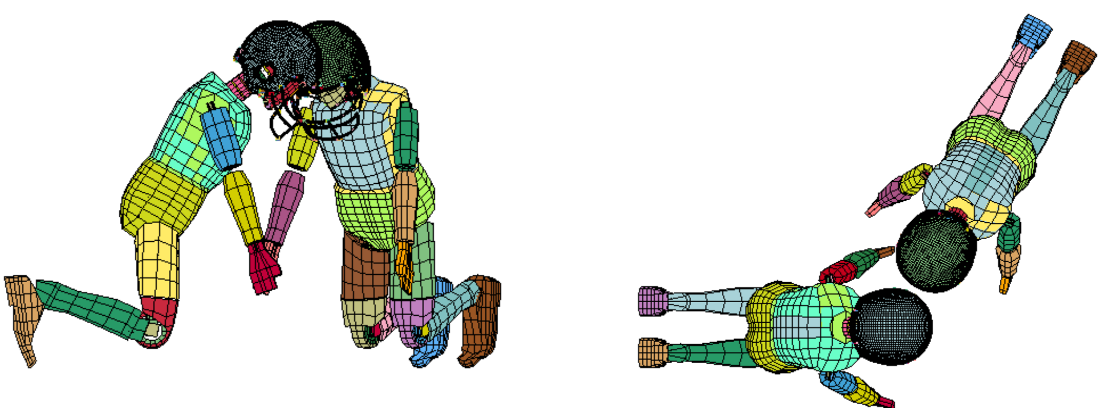


Fig. 5. Struck player (left) hit by striking player (right).

The striking player velocity is 13.41 m/sec resulting in a speed of 30 mph.

RESULTS

PLAYERS MODEL-1: The deformed configuration of the players and the neck deformation are shown in Fig. 6 and 7 for helmet model with gap between the skin and the foam pad.

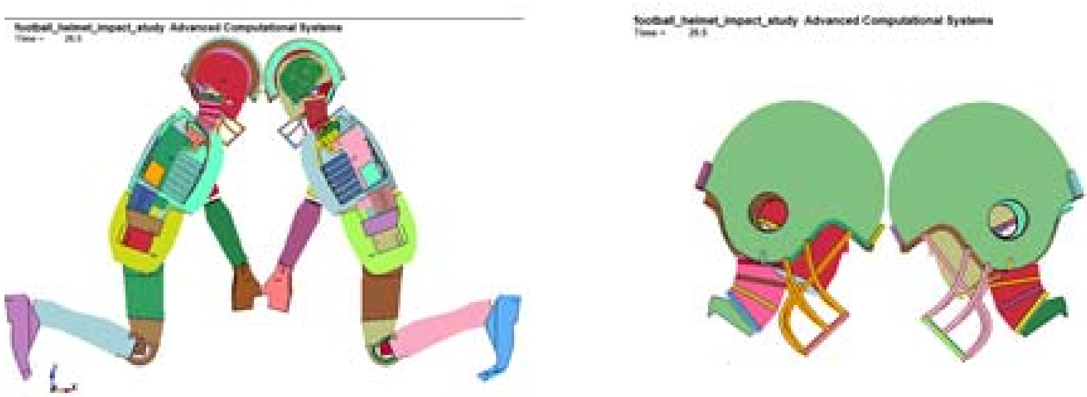
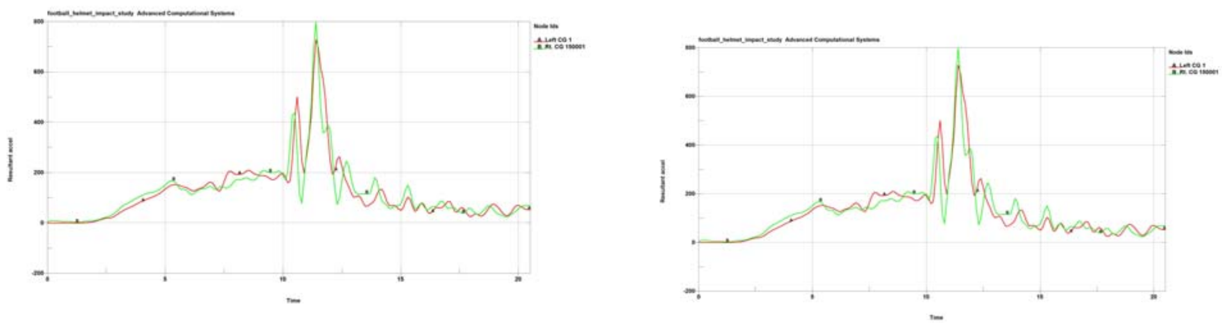


Fig. 6 & 7 Left: Deformed configuration at 20.5 msec. Right: Neck deformation



Figs. 8 &9 Linear and Rotational resultant acceleration at CG of heads.

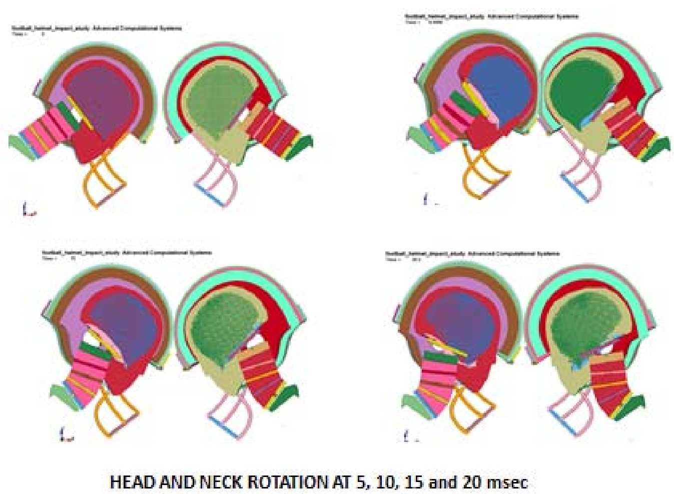


Fig. 10. Relative Rotation of the Head and Helmet.

The linear resultant acceleration and rotational acceleration are shown in Fig. 8 and 9 of the head for both left and right players. The peak resultant acceleration is of the order 800g. The rotational acceleration about Y axis is very high. Figure 10 shows the relative rotation of the head and the helmet.

The peak pressure is 12MPa at maximum acceleration occurring time 13msec.

PLAYERS MODEL -1 _Updated Foam Geometry: The padding geometry is updated so that there is a good fit of the helmet with the head skin. The same impact scenario as the previous case is used in this study. The deformed configuration at 18msec is shown in Fig. 11. The head acceleration plot is shown in Fig. 12. The peak acceleration is 350g.

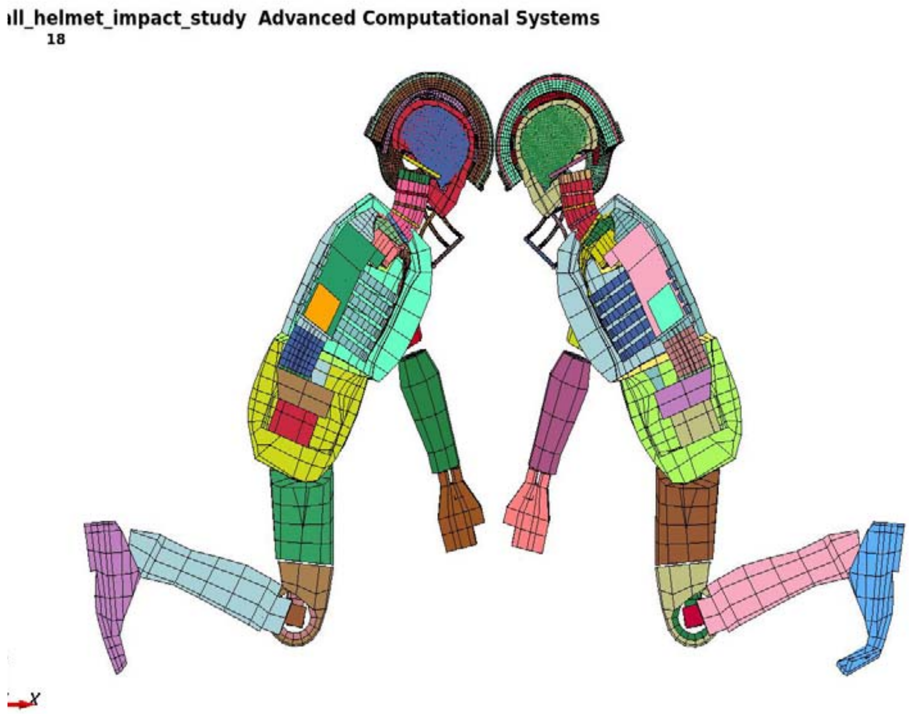


Fig 11. The deformed configuration shows upward movements of the helmets.

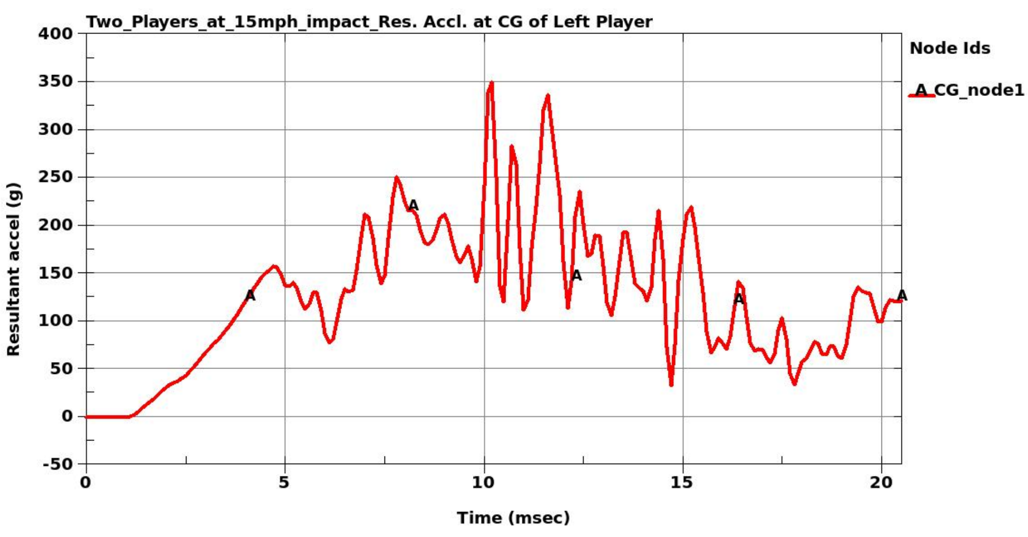


Figure 12. The resultant head acceleration

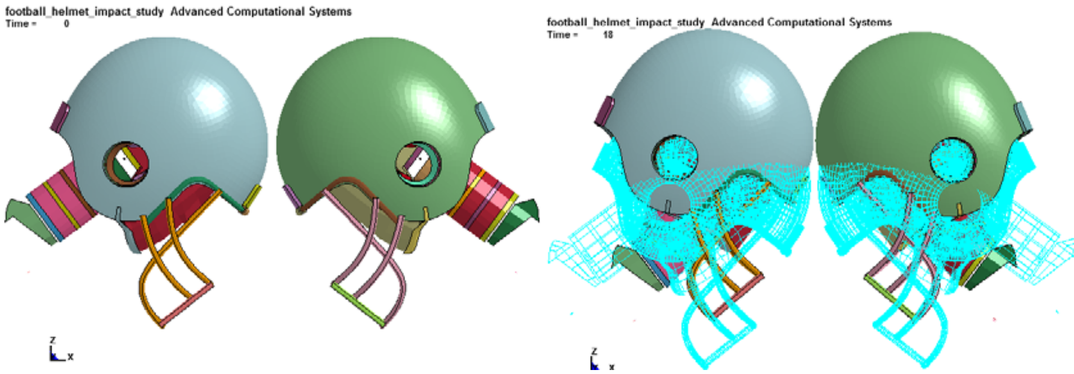


Figure 13. Vertical displacement of the head and helmet

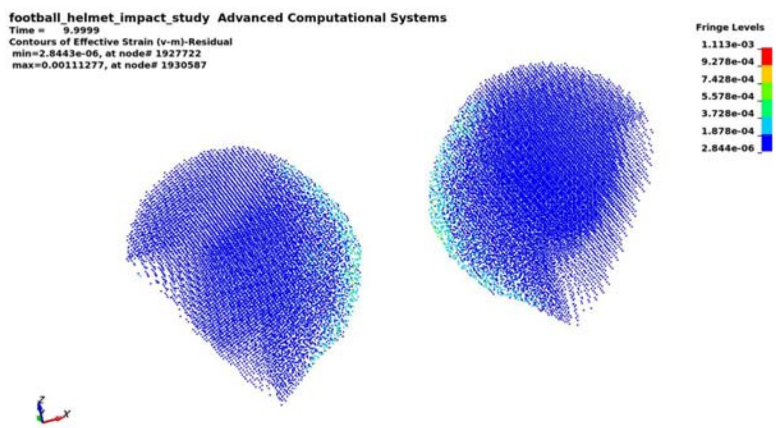


Fig. 14 Contours of effective strain distribution at 10msec

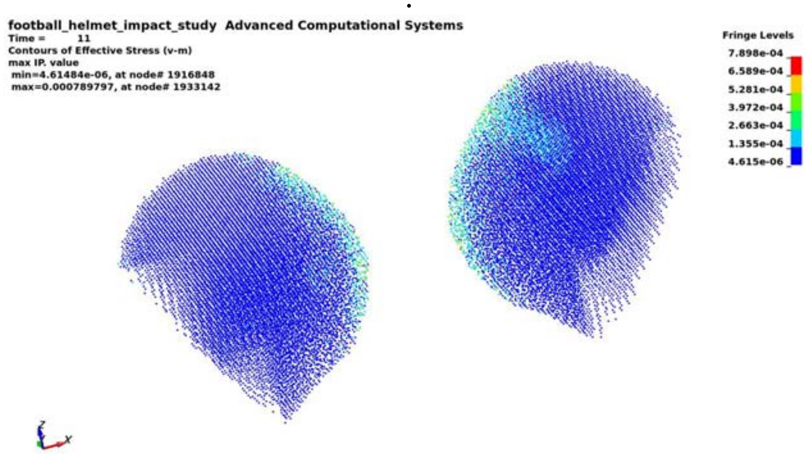


Fig. 15. Contours of effective stress at 11msec.

Figure 13 shows the vertical movement of the heads due to impact. The vertical displacement is 80-90mm and the neck rotation is high. The resultant head rotational acceleration is 30000 rad/sec^2. Figure 14 shows the strain distribution in brain tissue and is 0.0011. Figure 15 shows the contours of effective stress at 11msec and the peak value is 0.78MPa.

PLAYERS MODEL2: The neck deformation is shown in figure 16. The head acceleration of the struck player is shown in Fig. 17. The peak acceleration is 240g. The effective strain in the brain tissue at 12msec is shown in Fig. 18. The strain is 0.000924.

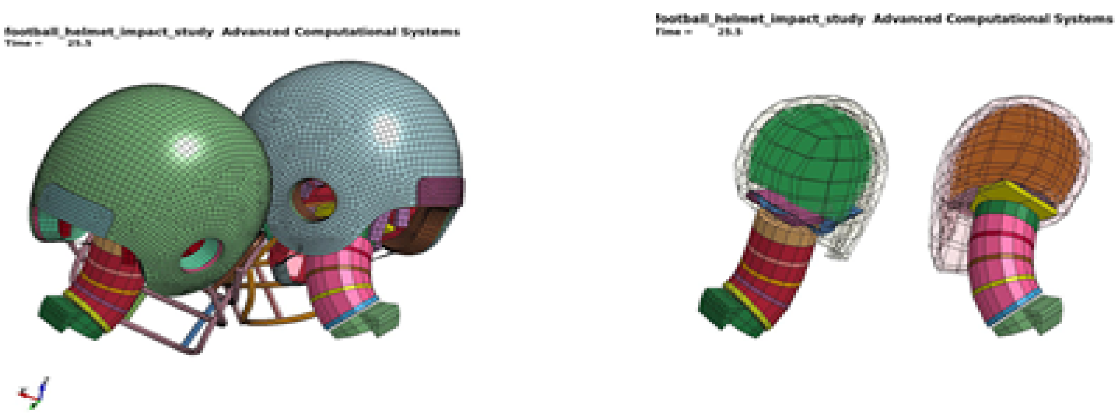


Figure 16. The Head and Neck Deformation at 25 msec.

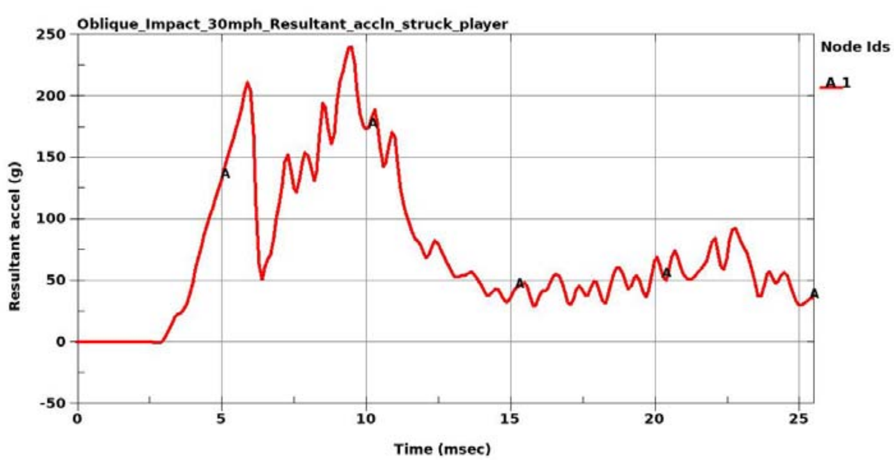


Figure 17. The head acceleration plot of struck player.

The shear stress distribution in the brain tissue at 12msec is shown in figure 19. The maximum shear stress is 0.48MPa.

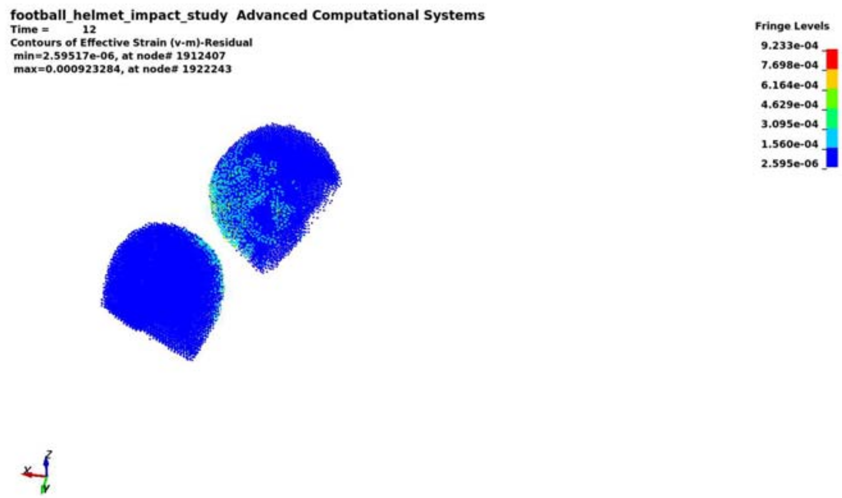


Figure 18. Effective strain distribution at 12msec in the brain tissue.

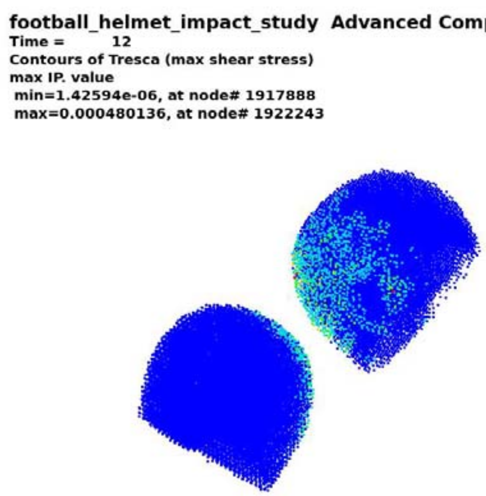


Figure 19. The shear stress distribution in the brain tissue at 12msec.

DISCUSSION AND CONCLUDING REMARKS

The human brain model was developed using MRI images. The brain tissues and surrounding soft tissues were lumped as one material. The finite element modeling of these brain tissues were approximated as structural SPH particles. The players were represented by using existing ATD model from LSTC dummy library. A generic helmet was modeled in this study.

The impact scenarios of two football players colliding with each other were presented. The first one was head on collision of two players at a velocity of 6.704 m/sec. The second one was a striking player running with a velocity of 13.4 m/sec impacting the standing stuck player. Two padding designs were studied. The first one was with the foam padding fully covered but not completely fitting with the head skin. In the second, the foam paddings were fitted so that they are in contact with the headform outer layer skin.

In the first configuration of head on collision, the results showed high peak accelerations. Since the impact energy was high, the neck rotations were also high. The heads also moved upward in the process of dissipating the energy and this resulted in higher neck rotation. The peak acceleration for the second foam design tightly in contact with the head showed 350g compared to the first configuration where in the pads were not in tight fit. It is better to keep tight fit of the pads so that the energy absorption starts as soon as the impact takes place.

In the second configuration, the striking player was moving at a high speed of 13.4 m/sec and the struck player was stationary. The striking player was hitting at an angle of 60 degrees. The peak acceleration induced was high and neck rotation was also high.

All these simulations were run using the Linux 64 bit smp d R7.0.1 LS-DYNA version. The LS-PrePost version 4.1 was used in meshing and the post processing the results.

A number of assumptions and approximations were made in these analyses. Mathematical modeling of biological tissues such as brain is complex because of wide variation in the geometry and unknown tissue properties. Some of these assumptions and approximations may not be exactly representing the actual structure or the product, however, the mathematical simulation provides insight in understanding the mechanics.

Coupling the advancement in medical imaging technology with powerful computational tools such as LS-DYNA help evaluate and understand the brain injury mechanism in finding solutions for mitigation of mild traumatic brain injury.

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