Finite Element Modeling of Human Brain Response to

Football Helmet Impacts

by

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ABSTRACT

The football helmet is a device used to help mitigate the occurrence of impact-related traumatic (TBI) and minor traumatic brain injuries (mTBI) in the game of American football. The current design methodology of using a hard shell with an energy absorbing liner may be adequate for minimizing TBI, however it has had less effect in minimizing mTBI. The latest research in brain injury mechanisms has established that the current design methodology has produced a helmet to reduce linear acceleration of the head. However, angular accelerations also have an adverse effect on the brain response, and must be investigated as a contributor of brain injury.

To help better understand how the football helmet design features effect the brain response during impact, this research develops a validated football helmet model and couples it with a full LS-DYNA human body model developed by the Global Human Body Modeling Consortium (v4.1.1). The human body model is a conglomeration of several validated models of different sections of the body. Of particular interest for this research is the Wayne State University Head Injury Model for modeling the brain. These human body models were validated using a combination of cadaveric and animal studies. In this study, the football helmet was validated by laboratory testing using drop tests on the crown of the helmet. By coupling the two models into one finite element model, the brain response to impact loads caused by helmet design features can be investigated. In the present research, LS-DYNA is used to study a helmet crown impact with a rigid steel plate so as to obtain the strain-rate, strain, and stress experienced in the corpus callosum, midbrain, and brain stem as these anatomical regions are areas of concern with respect to mTBI.

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TABLE OF CONTENTS

Page

LIST OF TABLES....v

LIST OF FIGURES....vi

CHAPTER

1 INTRODUCTION....1

Literature Review of Modeling Techniques....6

Human Head Modeling....6

Football Helmet Modeling....11

Coupled Football Helmet Human Head Modeling....15

Thesis Objectives....16

Thesis Overview....16

2 MODEL DESCRIPTION....18

The Football Helmet....18

Football helmet model validation....24

Test Apparatus....25

Validation Results and Discussion....27

3 HUMAN BODY MODEL DESCRIPTION....37

Human Head Model....37

Coupling of Human Body and Helmet Model....41

4 FULL MODEL SIMULATION RESULTS....43

Wave Propagation....44

Impact Results....47

CHAPTER Page

5 CONCLUDING REMARKS..56

REFERENCES……………………………………………………………………...56

LIST OF TABLES

Table Page

1. Model Material Properties 21

2. Finite Element Properties 22

3. Mesh Convergence Data 30

4. Response Values for each Mesh Density 30

5. Convergence Data 31

6. Energy Checks 31

7. Low Frequency Acceleration Results 34

8. Linear viscoelastic Material Properties. 39

9. Elastic Material Properties. 40

10. Linear Elastic Plastic Material Properties. 40

11. Max Values for Crown Impact 54

12. Max Values for Frontal Impact 55

LIST OF FIGURES

Figure Page

1. Graph Comparing Horgans Results 11

2. Photographs of the First Football Helmets Used 12

3. Riddell Attack Revolution 18

4. FEM of the Helmet Headform System 20

5. Uniaxial Compressive Stress-Strain for EPP 22

6. NOCSAE Drop Test Apparatus 24

7. Test Apparatus 26

8. Helmet Assembly 27

9. Nodes Used in the Finite Element Model to Determine Model Convergence 28

10. h-Method Convergence Study for the Football Helmet Impactor 29

11. h-Method Convergence Study for the Football Helmet Components 30

12. Total System Energy 32

13. Experimental Validation of Y Acceleration in Impactor 33

14. FFT of Experiment 2 33

15. Filtered Experimental and Simulation data for Low Frequency Response 34

16. FE Validation Against NOCSAE Testing 36

17. The Human Head FE Model. 41

18. GHBMC Model Integration. 42

19. Location of Critical Anatomy to mTBI 44

20. Displacement at Time t = 15 ms 45

21. Displacement Propagation for the Crown Impact 46

Figure Page

22. Deformation of the Human-Helmet System During Crown Impact 47

23. Deformation of the Human-Helmet System During Frontal Oblique Impact 48

24. Crown Impact Relative Displacement 49

25. Frontal Impact Relative Displacement.. 49

26. Frontal Impact Max Strain 50

27. Frontal Impact Stress 50

28. Frontal Impact Max Strain rate 51

29. Crown Impact Max Strain.. 51

30. Crown Impact Max Stress.. 52

31. Crown Impact Max strain Rate 52

32. Acceleration Profiles of the Skull During Impact.. 55

Introduction

From 1869 to 1905, 18 deaths and 150 serious brain injuries were attributed to the sport of American football. In response to these troubling results the first helmet was created in 1893. The efficacy of such devices at onset was low but as the understanding of the biomechanics involved and the nature of the loading conditions in the sport became better known, the helmet design advanced to almost completely eliminate the occurrences seen in the sport. However, while helmets have had great success in reducing traumatic brain injury (TBI), a same level of was not observed when considering concussion and other minor traumatic brain injuries (mTBI). This is perhaps due to many factors but the most important one is that reduction in linear acceleration of the head leading to reduced TBI, did not minimize other head injuries considered as mTBI. This is because the relationship between each football helmet component and the brain response during impact is not well established. This research is designed to analyze a validated football helmet-human body finite element model in an attempt to better understand how the functional design properties of the helmet affect brain response during impact (Hoshizaki et al. 2004).

King et al. (King et al. 2003) studied the load conditions that caused head injury with a specific comparison of linear accelerations against rotational accelerations to the head and tried to determine if injury is a result of exclusively one or the other. It was determined that both acceleration types are significant in causing brain injury. However accelerations to the head are the wrong parameter to analyze when considering the effectiveness of a helmet. Instead of these loading parameters (linear and angular acceleration), the response of the brain with regards to strain and strain rate in the midbrain region of the brain was suggested. These parameters correlate with changes in brain matter integrity as a result of impact (McAllister et al. 2011) and the midbrain is selected because it relates to the highest levels of strain found in prior finite element analyses (King et al. 2003). Designing helmets that minimize brain response would be far more effective in minimizing injury because they would address the primary concern during impact, namely minimizing brain response. Brain injury is intimately related to the local response of the brain and not to the linear and angular accelerations of the head (King et al. 2003).

To understand the mechanical properties of the helmet that impact the brain response during impact, the ability to computationally predict human head response, the helmet response, and human head–helmet system response during impact must be obtained. The computational tool of choice for a study of this nature is the finite element method. A finite element model contains not only the geometry and the loading, but also the material properties for different components in the model. Clearly, for the brain, in-vivo validation of material properties during injury inducing loading conditions is very difficult. To circumvent this problem, cadaveric studies (Nahum et al. 1977; Trosseilee et al. 1992; Hardy and Mason 2007; Yoganandan et al. 1995; Hodgson et al. 1970) and animal studies (Prevost et al 2011; Rashid 2013) are used to define material properties in the brain. While not ideal, these experiments are good approximations and give insight into what should be expected for in-vivo material properties. Using this information, both academic and commercial institutions have published validated finite element models for use by the public. For example the following is a list of publicly available models:

- Total HUman Model for Safety (THUMS) developed by the Toyota Motor Corporation and Toyota Central R&D Labs (Total Human Model for Safety 2014)
- KTH FE Human Head model developed by the Royal Institute of Technology (KTH – School of Technology and Health 2008)
- TNO Head FE Model developed by the Netherlands Organization for Applied Scientific Research (TNO) (Gang et al. 2008)
- ULP FEM of the human head developed by the University of Louis Pasteur (Baumgartner et al. 2007)
- Politenico di Torino University finite element model of the human head (Belingardi et al. 2013)
- University College Dublin Brain Trauma Model (UCDBTM) (Horgan et al. 2003)
- Global Human Body Models Consortium (GHBMC) model with the human head model developed by Wayne State University (Mao et al. 2013)

Each model varies in the amount of validation data used (cadaveric and animal studies), meshing techniques, constitutive material models, and anatomical features included. By using these models, the brain response during impact can be studied and used to determine the severity of applied load conditions and their effect on brain tissue.

Finding and validating the response of the helmet is more straightforward because the tests do not involve invasive procedures. Studies have been carried out to show how the mechanical properties and design affect stress-distribution, energy attenuation, and accelerations of headforms wearing helmets (Tinard et al 2012; Moss and King 2011; Forero Reueda and Gilchrist 2012). The research shows that in general for head protection, a hard outer shell with an energy attenuating liner inside works best for minimizing the linear acceleration of the headform. The type of shell is typically a polycarbonate or composite, and the type of energy attenuating liner is typically a foam such as expanded polystyrene (EPS), expanded polypropylene (EPP), expanded polyethylene (EPE), vinyl nitrile, or cross-linked polyethelene. Since the game of football involves repeated impacts, the foams are restricted to EPP, EPE, vinyl nitrile, and cross-linked EPE (Deck and Willinger 2006). EPS is not suitable because it is characterized as a closed cell foam and undergoes cell rupture during impact.

From 2005 to 2010, a study monitoring 1,833 collegiate football players was done to determine if the type of football helmet affects the risk of concussion during play. The study compared two different helmet designs totaling 1,281,444 on-field head impacts from which 64 concussions were diagnosed (Rowson et al. 2014). The end result showed that one design type reduced the probability of concussion thereby validating the notion that there is connection between helmet design and mTBI (Rowson et al. 2014). However, before this study was completed, Post et al. (2013), using the UCDBTM finite element model for the human head, studied the brain response that occurs when three different football helmet designs are used. They constructed an impact simulating experiment and using each helmet, determined an acceleration profile to apply to the human head model. Each helmet type resulted in different acceleration profiles. However because they didn’t use a finite element model of the helmet in the study, only broad statements concluding that the total design impacts brain response can be made (Post et al. 2013). Similarly, a study by Deck and Willinger (2006) proved that optimization against headform response and human head response does not lead to the same results.

Transportation and military helmet applications studies have been carried out to explore how the helmet properties affect the brain response when the helmet and head are coupled in one finite element analysis. Pinnoji et al. (2007) studied the coupled impact of a general helmet using an EPS foam liner and a polycarbonate shell against a model with no helmet. In this study, the anterior of the frontal bone was impacted by a rigid surface and a decrease in the coup pressures in the head were substantially reduced by using the helmet. However it does not study specific anatomical structures in the brain and neglects the rotational contributions by excluding the neck anatomy in the model (Pinnoji et al. 2007). Jazi et al. (2014) studied the influence of helmet padding materials on the human brain under ballistic impact. They modeled the human head-neck system with the helmet and determined the sensitivity of pad stiffness showing that the stiffness did affect resultant brain pressures. This was a great exploratory exercise; however no model validation was done to bring confidence in the model validity (Jazi et al. 2014). Tinard et al. (2012), using the Strasbourg University Finite Element Head Model and a motor cycle helmet model coupled the helmet-head system and compared the results for risk of injury during motorcycle crash impacts. It was determined that for current motor cycle standard test conditions, the head is still susceptible to brain injury based on the determined brain response. The helmet was validated against experimental data and gives great insight into how to couple the human head and helmet in LS-DYNA. However, if the neck is ignored in a simulation involving a football helmet, the proper angular accelerations experienced by the head during impact is likely to be ignored and as a result, the brain response will be incorrectly predicted (Tinard et al 2012).

In the present research, it is proposed to use the most current and validated finite element models of the human body, i.e. the GHBMC full body model, and couple it with a fully validated finite element model of a football helmet. The full body model will be used so the effect of all the currently approximated anatomical features will be included in studying the brain response to head-helmet impact scenarios. The results will help isolate how the design features of the football helmet interact and contribute to the response of brain material.

1.1 Literature Review of Modeling Techniques

To identify the current state of knowledge in regards to football helmet impact on brain tissue, a review of current modeling techniques for football impacts is presented. Also to better understand the capabilities of the current GHBMC human model, a review of modeling techniques for the human head is presented.

1.1.1 Human Head Modeling

Human head models occupy a difficult space for research in that validation of such models must rely on experimental procedures that approximate the in-vivo material testing conditions. As a result the models have to find secondary approaches such as bench testing, cadaveric testing, and animal scalability testing to validate the material properties of the brain. The goal of these models is to predict injury and use them to prevent future occurrences. However in addition to material model difficulties, the variability in the geometry of brain anatomy across the population can also have a great impact on the injury mechanisms being modeled. These difficulties provide a challenging environment for research. However the last 50 years have produced a truly impressive advancement in such technology (Meany et al. 2014).

The first models of the human head were comprised of simplified geometries and material properties. One of the first was done by Advani and Owings (1975) and modeled the brain as an elastic core within an elastic spherical shell. They assumed the displacements at the skull brain interface were continuous, ignoring the now critical Cerebral Spinal Fluid (CSF) effects. Chan and Liu (1974) took the approach of viewing the brain as a fluid-filled spherical shell and applied both radial and tangential loads to the surface of the shell. They determined the transient response of the brain by a Laplace transformation in time and an eigenfunction expansion to result in a closed form analytical solution. Many other tests exploring different geometries such as ellipsoids and material properties such as purely viscoelasticity and completely rigid skull structures were explored in the same time period. However the limitation of all of these studies is that there was no validation data to support their claims [Khalil et al. 1974; Khalil and Viano 1982).

The first experimental procedures to be used for model validation were the tests done by Nahum et al. (1975). They ran a series of blunt head impacts on stationary un-embalmed human cadavers and were designed to simulate realistic fluid pressures within the CSF space and cerebral blood vessels. Human head models utilizing that data to validate the FE model followed quickly after. Ward et al. (1977) developed one of the first models validated by the pressure results in Nahum's tests. Ward's model included the cerebrum, cerebellum, brainstem, ventricles and the dural membrane. The load conditions for the model were determined experimentally with dropped rigid headform tests and then the resulting headform accelerations were applied to the brain model. The results of the model showed that high normal stresses and load conditions caused serious brain injury while combined tension and shear stress produced subarachnoid hemorrhage. However, the impact of skull deformation was lost because the experimental acceleration curves had to be applied to a rigid skull.

The rigid skull assumption is problematic because displacements of brain tissue are dependent on the way the skull deforms. To improve this effort, Ruan et al. (1994) developed a model that was validated by the cadaveric experiments and also included the elasticity of the skull. A parametric study was conducted to determine the effect of varying impact locations, changes in impact velocity, and mass of the impactor. It was found that the maximum shear stress occurred in the brainstem, that the viscoelasticity of the brain had an insignificant effect on pressure response in the brain, and that the velocity of impact had a greater effect on brain response than impactor mass. However, they did not include the cerebral membranes in this model and the effect that they have on brain response was not included. They suggested that the falx, cerebral tentorium and dura had to be investigated in future models.

Zou et al. (1995) continued Ruan's work on the three dimensional human head model. The brain was remeshed using smaller elements so that the difference between the white and gray matter could be modeled with different material properties. Also the ventricles were determined necessary to properly match regions of high shear stress to locations of diffuse axonal injury (DAI). However this added complexity and decrease in element size introduced a computation time problem. To reduce the computation time, the three-layered skull was simplified to one. The model included the scalp, skull, dura, falx, tentorium, pia, cerebral spinal fluid (CSF), venous sinuses, ventricles, cerebrum, cerebellum, brainstem, and the bridging veins (Zou et al. 1995). It was not until Al-Bsharat (1999) modified the model that an agreement with experimental results was achieved. A sliding interface between the skull and brain in the sub arachnoid space was introduced. This allowed for the inner surface of the CSF to slide relative to the outer surface of the pia matter. With these changes included in the model, the response matched Nahum's intracranial pressure data. In addition to the pressure data, the model was able to predict large displacements with respect to the skull at low speeds. The displacement data was gathered from experiments conducted at Wayne State University. One limitation of this model was that the results included a small amount of penetration in the sliding surfaces at impact energies that were higher than 100J (Al-Bsharat 1999)

As technology improved and the amount of computational power increased, the ability to study the effects of mesh density and mesh quality to the result of the model became more easily available. Horgan and Gilchrist (2003) performed a parametric study on the effects of several meshing patterns and the sensitivity of material properties to the model outputs. The results of the study showed that the short term shear modulus of the neural tissue has the biggest effect on intracranial frontal pressure and the model's von Mises response. Also the material properties defined by Zhou et al. (1995) resulted in the best agreement with cadaveric experiments by Nahum (1977), as seen in Figure 1. The parametric study added clarity to model decisions in future versions and allowed some clarity as to which material models provide the most reliable results (Horgan et al. 2003).

Currently the most robust model available for experimentation is the model by Mao et al. (2013). The geometry of the model was gathered from MRI and CT scan data. The model was validated against a more diverse set of experiments and was designed to be robust, capable of predicting a variety of injury mechanisms under many types of load conditions. Some examples of the new features included more attention to the connectome of the brain tissues. The connectome proved to have some significance in DAI and up until this point the details of how it should be modeled was relatively unexplored. This new model also investigated the optimal number of bridging veins to include in a finite element model. As was discussed previously, the anatomical variability in the general population can impact a models results and the number of bridging veins in a brain affects the overall stiffness of the organ. Research found that for the 50th percentile male, 11 bridging veins was a good approximation. Overall the model included the cerebrum, cerebellum, brainstem, corpus callosum, ventricles, thalamus, bridging veins, CSF, skull, facial bones, flesh, skin, and membranes – including falx, tentorium, pia, arachnoid, and dura. It was also validated against 35 different experimental procedures - more than any model before it (Mao et al. 2013).

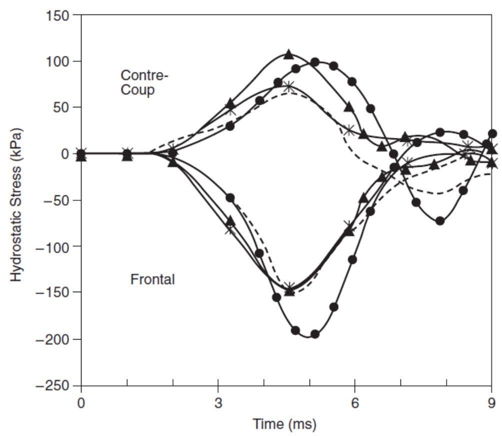

Figure 1: Graph comparing Horgans results when the material properties of Ruan (solid triangles), Zhou (crosses) and Willinger (solid circles) were used with the model, against the experimental results of Nahum (dashed line) (Horgan et al. 2003)

1.1.2 Football Helmet Modeling

In 1893 the first helmet was used in American football during an Army-Navy game. Figure 2 shows the replicas of those first helmets used. They were made exclusively with leather and in time started to implement metal alloys to allow for more support and protection. Moderate adjustments were made in efforts to decrease the number of spine and head injuries that were extremely common in the sport. However it wasn't until 1973 that helmet standards generated by the National Operation Committee on Standards for Athletic Equipment (NOCSAE) were developed to regulate the safety of football helmets. These standards were based on experimental procedures and the primary variable for measuring the performance of the helmet was the acceleration of a headform (a solid brain surrogate). This regulation encouraged football helmets to reduce the effects of linear acceleration by method of trial and error. However, the mechanisms of how each component in the helmet helps keep the head safe were unknown [Bennett 1977; Hoshizaki et al. 2004).

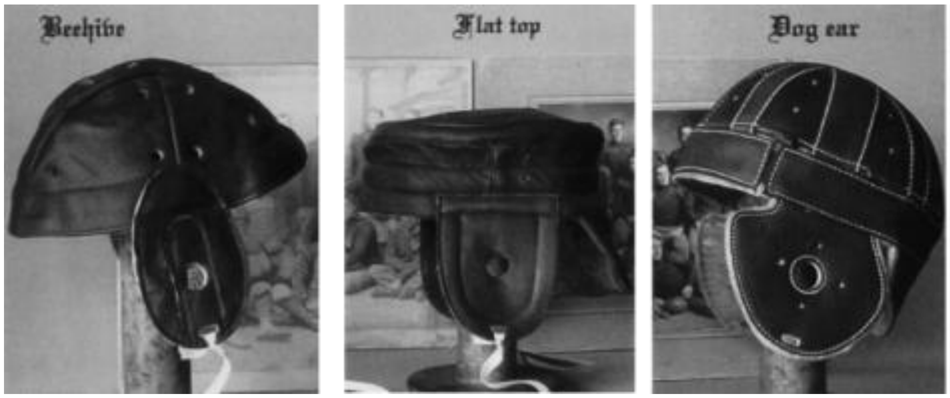

Figure 2 Photographs of the first football helmets used: “beehive”, “flat top”, and “dog ear” (Bennett 1977)

Kahlil (1974) wrote a pioneering piece on investigating the response of different material properties during impact loading scenarios and determined how they participate in affecting propagated pressure waves throughout the structure. By using the finite element method, a numerical model for concentric shells that housed either Styrofoam, water, polyurethane foam, and steel were created. The numerical model was then exposed to drop simulations and the resulting pressure waves throughout the cylindrical structure (assumed as a brain surrogate) was tested and compared against physical experiments that validated the results. This investigation showed that including a covering of any type helps to minimize the peak pressure amplitude experienced in the structure. Also it was illustrated that a cover composed of a hard outer shell and a soft, energy dissipating interior lining was significantly more effective than a helmet featuring only metallic protective surfaces. This illustrated how modeling an experimental procedure can elucidate mechanical mechanisms present in a larger structural response [38].

Vetter and Vanderby (1987) were one of the first to study the structural effectiveness in attenuating impact energy by a finite element parametric study. They generated a simplified shell and liner model. After validating a baseline model for the helmet, they adjusted parameters to determine the sensitivity of material properties and geometry on the output of the simulation. They showed numerically that the helmet shell accounts for only a small portion of the energy absorbed in impact. The constitutive material model used for the foam was a piecewise linear elastic. This is not in agreement with the non-linear material models that are currently used for foams today. As a result the sensitivity that foam thickness has on reducing linear acceleration was predicted incorrectly (Vetter and Vanderby 1987).

The effect of foams on open-cell foam impact in helmets was clarified in a study by Lawrence Livermore Labs (Moss et al. 2011) where they used experimental uniaxial compression tests to characterize foams for military applications and for football. The experimental curves were input directly into their finite element code and used for computing the response of the helmet in impact. By using this model for foam, a good agreement with experimental cases was accomplished and three significant results were found. First, performance of the pad depends on the range of impact velocities that the foam will experience. Second, softer pads are better for low velocity and hard pads were better for higher velocity. Lastly, thicker pads perform better at all velocities (Moss et al. 2011).

The emergence of practical models allowed for helmets to be optimized and improved for a variety of different industries. The motorcycle helmet industry had a unique model in that their loading conditions were higher than football and one time occurrences of impact were acceptable. This allowed for the use of damage in the foams to attenuate more of the impact energy [Mills et al. 2009; Ghajarj et al. 2009). In addition, Kostopoulos et al. (2002) generated a model to study the effects of composite shells in motorcycle helmets. It was determined that their greater stiffness leads to higher accelerations of the headform at lower impact energies, meaning that it would be a poor material for the application of football helmets.

The public availability of football helmet finite element models is scarce compared to the motorcycle, equestrian, and bicycle industry. However two significant works have been developed specifically for the football industry. Post et al. (2013) did a study that experimentally tested three different football helmets and monitored the effect of brain deformations. The helmets were tested by performing drop tests on the Hybrid III Headform (About headforms 2014). Accelerations of the headform were then applied to a finite element model of the brain. The resultant linear and angular acceleration profiles were applied to a rigid skull and the effect of each helmet was compared. The results of the brain simulation showed that the different helmet types resulted in different acceleration profiles and that the design features of the helmet can have an impact on the total safety of the helmet (Post et al. 2013). Also the typical indicator for safety of a helmet had been maximum acceleration or HIC (head injury criteria). The finite element model illustrated that peak strains and peak stress that occur in the brain from typical loading conditions did not occur at the peak acceleration but later in the impact scenario (Post et al. 2013). Further studies need to be done to characterize how the football model characteristics affect the brain deformation results to optimize the performance of protective equipment in football.

1.1.2 Coupled Football Helmet Human Head Modeling

Modeling the helmet-head coupling in one finite element model is a relatively new field of study. As presented before, impact data are traditionally gathered from experimental studies and then the load conditions are applied to the head for the study of brain response. The motor cycle helmet industry has seen some significant advances in this field. Deck and Willinger (2006) studied a validated finite element model of a motorcycle and coupled it with a human head model. They were able to monitor the Von-Mises stress in the brain due to 4 validated impact scenarios and also proved that using a rigid headform model in a simulation results in significantly different responses than the coupled system (Deck and Willinger 2006). Tinard et al (2012) expanded on that idea and optimized the design of the motorcycle helmet using modal methods with respect to the brain response metrics. These models were significant in that they used validated helmet models in the coupling of the two systems. Other exploratory works have been done in other industries such as the study by Pinnoji et al (2007) who looked at the coupled impact of a general helmet. In that study, the frontal loading condition was considered and the helmeted vs non-helmeted brain pressures from impact were compared. This is very useful as an exploratory study but the helmet model still needs to be validated in order to consider the helmet for design applications. Jazi et al (2014) studied the response of the coupled head-helmet system in response to ballistic impact. In this model they included the neck in the model and showed the sensitivity of helmet thickness to brain response data however it still lacked a validated helmet model for the study of design applications.

1.2.1 Thesis Objectives

The major objectives of this research are as follows.

(1) Develop a finite element model of a football helmet in LS-DYNA. The model will mimic the Riddell Revolution Attack football helmet which is made up of a bi-layer foam system with a polycarbonate shell. Also, for model validation a headform corresponding to the NOCSAE standards for rigid headforms will be integrated into the football helmet model.

(2) This helmet-headform system will then be validated by a laboratory experiment designed to mimic the NOCSAE standards for helmet safety. It will be designed to replicate the crown impact scenario for impact and use the impactor accelerations for model validation

(3) Integrate the football helmet model with GHBMC FBM v4.1.1 in LS-DYNA. With the integrated model conduct an impact on the crown of the helmet. The objectives are to observe how stress, strain, and strain rate occur in known brain tissues significant to concussion with respect to a fully validated human body model.

1.3.1 Thesis Overview

The model description for the helmet model is discussed in Chapter 2 and contains details of the general function of a football helmet and the standards that govern their design and safety. It also includes the geometry generation, meshing techniques, application of material properties and the experimental procedures used to validate the football helmet model. For the validation techniques, the results for the convergence analysis and experimental test are shown. Chapter 3 discusses the details of the human body model GHBMC FBM v4.1.1. The material properties of the brain materials, the model validation techniques, and the head injury mechanisms that are the target result of these models are presented. The integration of the helmet model with the human body model is also discussed. Chapter 4 presents the results of the coupled system and discusses the results in detail. Finally, in Chapter 5 the research findings are summarized and suggestions made for future study.

Model Description

2.1 The Football Helmet

In this study, a simplified football helmet model is created containing three components: the shell, the energy absorbing liner, and the comfort liner. As a geometry reference and tool for model validation, this research uses the Riddell Attack Revolution Youth helmet (L) (Riddell 2014) as the default model. Since the exact material properties were not readily available, initial material properties were acquired through literature review. The outer shell was determined polycarbonate-unfilled-low-viscosity, and the material properties were gathered from efunda.com (efunda 2014). For the comfort and energy absorbing liners the exact material is proprietary knowledge that Riddel did not share, however the material is a type of foam. As a result, uniaxial compression tests were performed by Moss et al. (2011) and the resultant stress-strain curves were input directly into LS-DYNA. The face mask and retention system are not included because they are not required in the NOCSAE standardized testing.

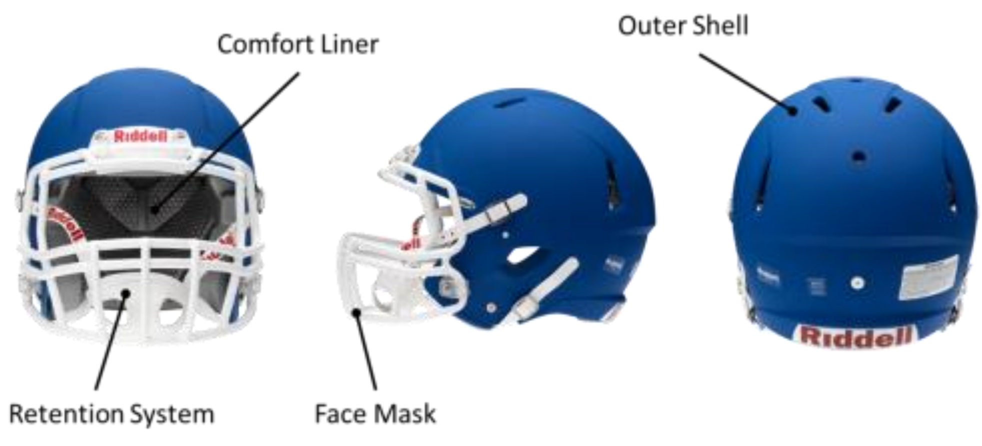

Figure 3 Riddell Attack Revolution (Riddell 2014)

The geometry of the helmet was created in a CAD software package, Solidworks (2014). Reference points were measured from the physical helmet and then images of the helmet were calibrated to the size of those reference point dimensions for use in the CAD software. Using the image, reference curves were traced to define surfaces for the geometry. The solids of the model were then extruded normal to these surfaces and complied with the measured thicknesses for each component of the helmet. Using the CAD software resulted in a geometry that does not match exactly with the helmet but is a reasonable approximation. This method was chosen over a 3D scanning method since one of the long term goals of the research is to have a parameterized model for helmet design optimization. The finite element mesh was created using Hypermesh (2011) and careful attention was paid to the mesh quality as deformations in foam analysis can be large leading to numerical instabilities and contact algorithm difficulties. LS-DYNA v971 (2011) was used to perform the explicit finite element analysis.

The outer shell on the helmet was meshed using 4-noded thick shell elements with the full-integration element formulation (Type 16) and 5 integration points through the thickness. The foams were meshed using 8-noded hexahedral elements using 1-point integration (element formulation 1) in order to have a manageable number of elements, better control over the mesh quality, and prevent negative volumes in the elements due to the large deformations experienced in the analysis. Using this type of element formulation reduces the total computation time but requires special attention to prevent hourglass modes in the simulation. To eliminate such modes, LS-DYNA has an algorithm that uses an orthogonal set of hourglass shape vectors that resist the 12 modes of hourglass deformation (Hallquist 2006). In this model the type 2 hourglass control

(Flanagan-Belytschko) a viscosity based method, was used. The headform geometry was created by using the top half of the headform specified in NOCSAE standard (DOC(ND) 002-13). The elements used were rigid 4-noded thin shell elements (Type 16) with 1 integration point through the thickness. The headform shell was considered rigid because it was an approximation for the solid headform. Figure 4 (a) shows the football helmet with a cutout in the helmet model to show the cross section of the helmet features and the headform surface.

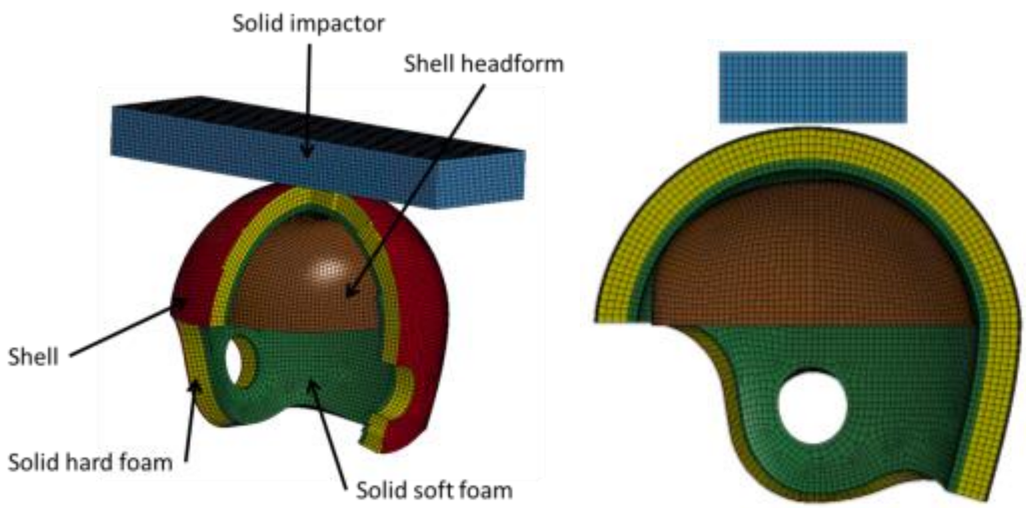

Figure 4 FEM of the helmet headform system (a) Section cut of the helmet headform system (b) Planar cross section of the helmet headform system

The helmet liners were modeled as open-celled foam (OCF) (McIntosh and McCrory 200). OCF's are suitable as a shock absorbing materials in football helmets because they are excellent at absorbing kinetic energy and are capable of full recovery after impact within a prescribed range of load conditions. Foams are a special material where principal engineering stresses are uncoupled, i.e. depend only upon the stretch ratio in the corresponding principal direction (Du Bois 2009). As a result, quasi-static compressive experiments can be performed and the resulting stress-strain curves can be directly input into LS-DYNA to define the material behavior. The keyword MAT_LOW_DENSITY_FOAM implements the constitutive model for this material (Hallquist 2006). In tension this material model assumes a linear elastic behavior until a particular defined tearing point. These foams are also highly strain rate dependent. However in the absence of experimental data only one strain rate data was used.

For the baseline model, the experiments done by Moss et al. (Mao et al. 2013) were used for reference and the densities and reference modulus of 2.8 $x10^{-7}$ kg/mm^3 and 0.08 MPa was used for the energy liner and $3.0x10^{-8}$ kg/mm^3 and 0.0448 MPa was used for the comfort liner (Hallquist 2006). The material properties were then tuned to match the experimental results. Figure 5 shows the stress strain curves of the two foam materials and the final adjusted material properties of the model. The final values are listed in Table 1.

Table 1: Model material properties

Component	**Material**	**Density (kg/mm^3)**	**E (MPa)**	**ν**
Energy foam	High density foam	$2.8x10^{-7}$	0.08	0
Comfort foam	Low density foam	$3.0x10^{-8}$	0.0448	0
Outer shell	Polycarbonate	$1.1996x10^{-6}$	2,415	0.329
Headform	Magnesium Alloy	$1.74x10^{-6}$	45,000	0.3
Impactor	6061 Aluminum Alloy	$2.7x10^{-6}$	206,000	0.3

Table 2 Finite element properties

Component	**Material**	**Element Type**	**Element Formulation**
Energy foam	MAT_LOW_DENSITY_FOAM	8-Noded Hexahedral	Reduced 1-ip
Comfort foam	MAT_LOW_DENSITY_FOAM	8-Noded Hexahedral	Reduced 1-ip
Outer shell	MAT_ELASTIC	4-Noded Quadrilateral	Full-integration, 5-ip through thickness
Headform	MAT_ELASTIC	4-Noded Quadrilateral	Full-integration, 1-ip through thickness
Impactor	MAT_ELASTIC	8-Noded Hexahedral	Reduced 1-ip

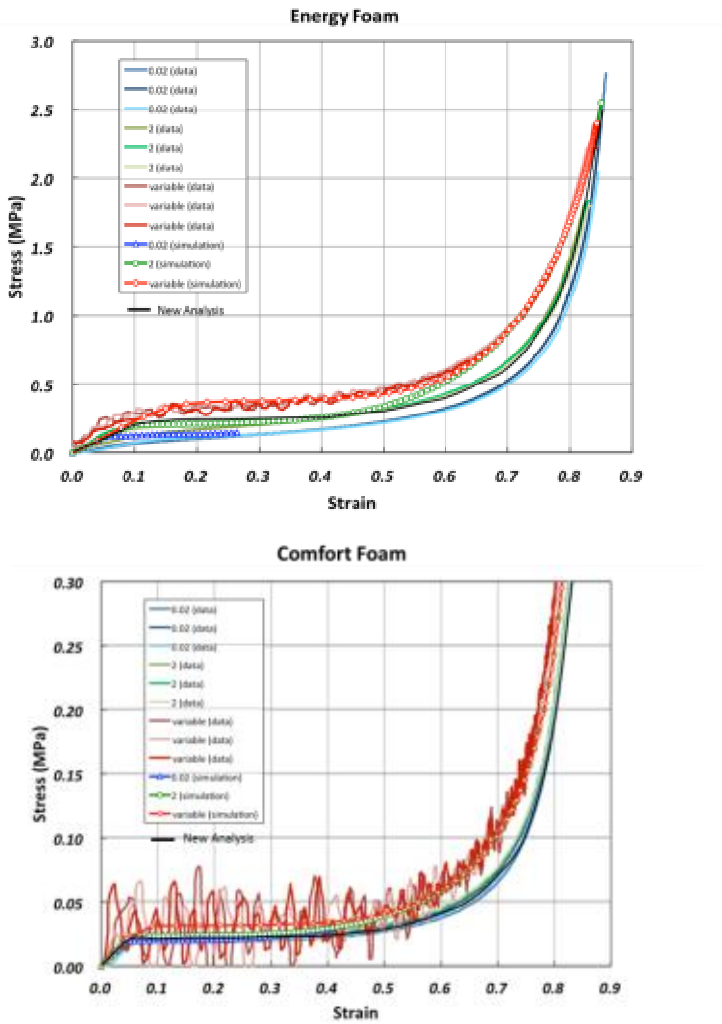

Figure 5 Uniaxial Compressive Stress-Strain for EPP (Moss et al. 2011)

The outer shell of the helmet was defined as polycarbonate (Zhang 2001). To model its behavior, a linear elastic material model was used, MAT_ELASTIC (Hallquist 2006). No plastic deformation in the shell was assumed because the loading conditions were designed to be below levels that induced damage to the helmet. To ensure this assumption is valid the max strain in the shell was monitored and for all simulations stayed below 10% elongation and 60 MPa of stress. While strain rate dependency may exist, due to lack of data and for the sake of simplicity of analysis, no strain rate dependence is used.

The material for the headform is defined in the NOCSAE standard. NOCSAE recommends magnesium alloys because of their light weight and strength. The material model for this component was MAT_ELASTIC. For simplicity of the model, numerical stability, and better control of the contact initiated between the headform and soft foam, 4-noded thin quadrilateral elements were used.

To define the interaction between the components contact details are used in the LS-DYNA model. The comfort foam and energy foam as well as the outer shell and the energy foam are coupled via tied contact. When defining automatic contact between two dissimilar materials such as the comfort foam and the magnesium headform, care has to be taken to see how the penalty function is implemented in contact. As a result the SOFT = 1 command is implemented in which causes the contact stiffness to be determined based on stability considerations, taking into account the time step and nodal masses.

2.2 Football helmet model validation

To ensure that the football helmet model is valid, the finite element model is run through an experimental validation and a convergence analysis. For the experimental set up a crown impact drop test was reproduced that has similar attributes to the NOCSAE standard requirements. The NOCSAE standard traditionally is designed to test a helmet's ability to minimize the Gadd severity index (Gadd 1966), which is a head injury criterion that analyzes the acceleration of a headform during the impact. To test this, a drop test machine is used and a Hybrid Headform III is the surrogate head to place the helmet on. The helmet headform is then dropped from prescribed heights to obtain certain impact velocities and the acceleration of the headform at the center of gravity is recorded. Figure 7 shows a sample schematic from the standard that illustrates the test set up (NOCSAE 2013).

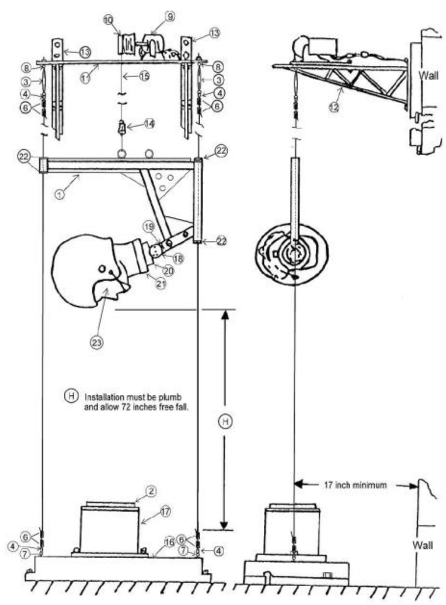

Figure 6 NOCSAE Drop Test Apparatus (NOCSAE 2013)

To replicate these standardized testing procedures a drop apparatus available at the Structures Testing Lab is used in the impactor test. The drop apparatus can be configured to run crown impacts from an impactor that has all degrees of freedom suppressed except for the vertical direction. This test scenario can be recreated in the finite element model of the helmet-headform system by fixing the headform in all degrees of freedom and allowing the impactor to drop onto the helmet at a prescribed velocity. The velocity chosen for this model is based off of the velocities found in the NOCSAE standard which range from 3.46 to 5.46 m/s. However, ensuring that no damage occurred to the helmet during testing the impact velocity was reduced to 3m/s (NOCSAE 2013).

2.2.1 Test Apparatus

The drop test system was designed to mimic the NOCSAE standard for football helmet quality, however materials such as the headform and official drop test apparatus were not available. As a result, a custom system that could repeat the crown impact test scenario was designed. The standard called out for a total drop mass (headform, helmet, and mounting system) of 5.93 kg (NOCSAE 2013). Since the Hybrid Headform III was not available, a sled was created that had approximately the same mass, 5.95 kg. The sled's degrees of freedom were restricted by two parallel shafts that allowed only vertical movement. Instead of dropping the headform-system onto an impact site, the impactor was dropped onto a fixed headform helmet system. The headform was designed by cutting five one inch thick pieces of 6061 aluminum alloy in a profile that was similar to the headorm specifications called out in the NOCSAE standard. For the areas critical for matching the dimensions of the human head, such as the crown of the headform, and to ensure a good fit, a rapid prototyped part using fuse deposition modeling with ABS material was formed to place over the aluminum structure (Headform cap). The experimental apparatus is shown in Figure 8.

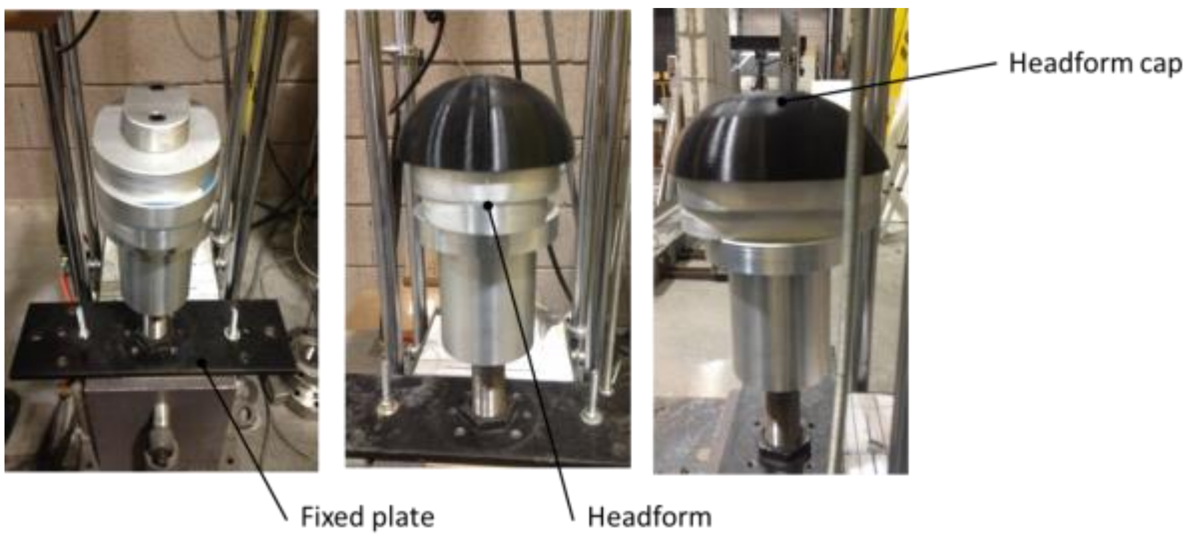

Figure 7 Test apparatus

To ensure a non-damaging loading condition, an impact velocity of 3 m/s was used. The sensor measuring the performance of the experiment is a uni-axial accelerometer placed on the impactor's top face (away from the contact surface). The goal was to measure the acceleration of the impactor in the vertical falling direction. The testing apparatus with the helmet included is shown in Figure 8.

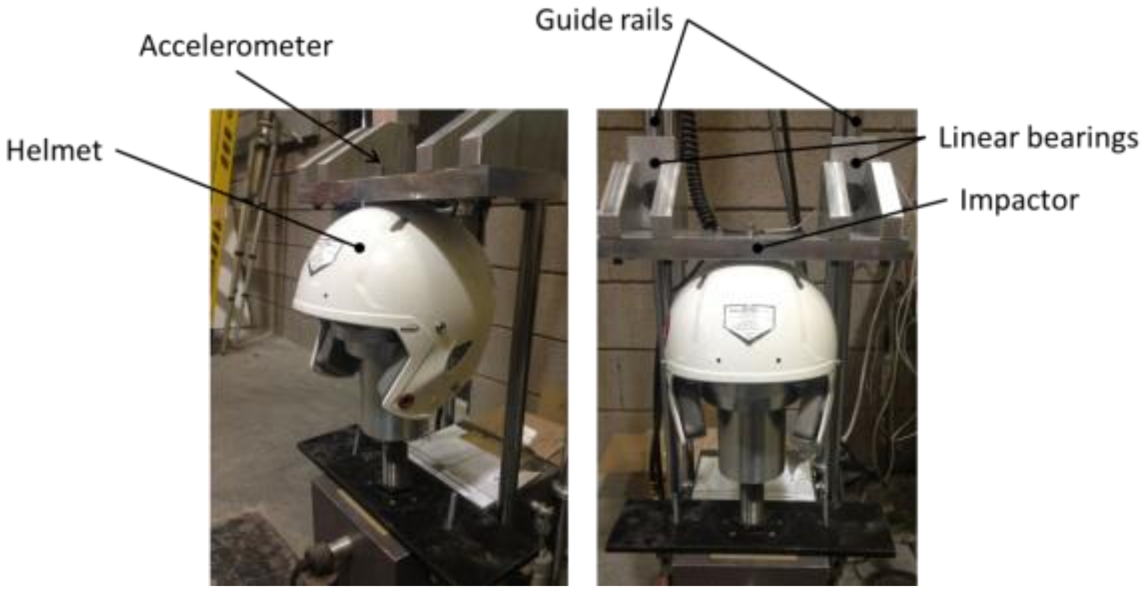

Figure 8 Helmet assembly

2.2.1 Validation Results and Discussion

To ensure accuracy of the finite element model, a convergence study is performed. Three simulations with varying mesh densities were completed to compute the order of convergence and grid convergence index (GCI) for the finite element model. Table 4 gives the mesh density details for each component, and Table 5 gives a summary of the convergence indices.

To obtain a stable converged helmet model, attention was given to the mass scaling parameter in LS-DYNA. LS-DYNA determines the time step of a simulation based on the stiffness of materials and the smallest element in the model. To increase the time step of the simulation virtual mass can be added to elements that are below a certain time step threshold that is defined by the user. This added virtual mass reduces the simulation time for the system. However since the penalty stiffness in contact is defined by the time-step and mass of the nodes, this mass scaling can introduce significant virtual energy to the system if it is applied to elements that are in contact. This contributes to a model that will not converge. As a result special attention was placed on the location of elements and the allowable threshold for the mass scaling to occur. For this simulation the mass scaling threshold is adjusted for each simulation and limited to allow for a maximum of 1 percent of the mass of the system to be virtual.

To monitor the convergence of the system the impactor and the helmet components were analyzed. For the impactor, the maximum acceleration of the center of gravity (COG) and the velocity of a corner node of the impactor were recorded. For the helmet, acceleration of the nodes at the front edge of the helmet was used. Figure 9 illustrates the location of the nodes used.

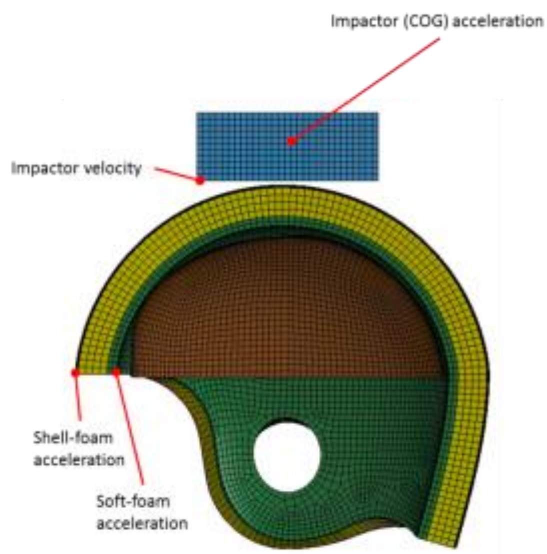

Figure 9 Nodes used in the finite element model to determine model convergence

The order of convergence and the corresponding GCI is listed on table 5. For the impactor, it is probable that the convergence difference is a result of the COG acceleration being computed by an inconsistent node location whereas the velocity is monitored at a consistent node location for each mesh density. The GCI for the soft foam was 0.8579 and 1.5895 for the shell-foam node, which shows that further mesh refinement is needed for a good confidence in the approximated solution.

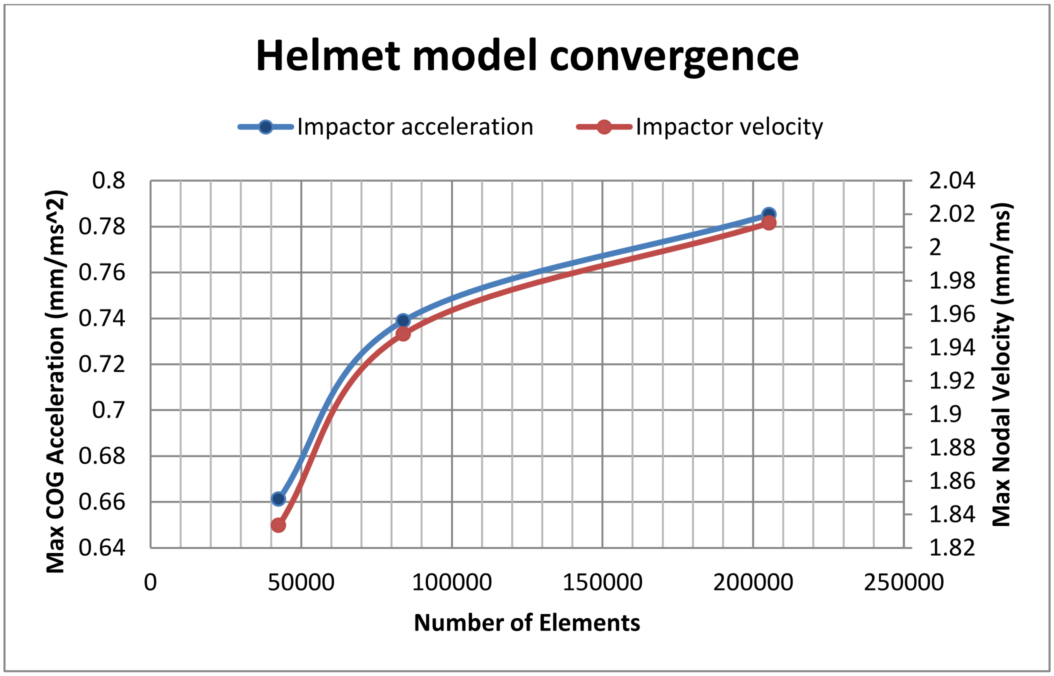

Figure 10 h-Method convergence study for the football helmet impactor

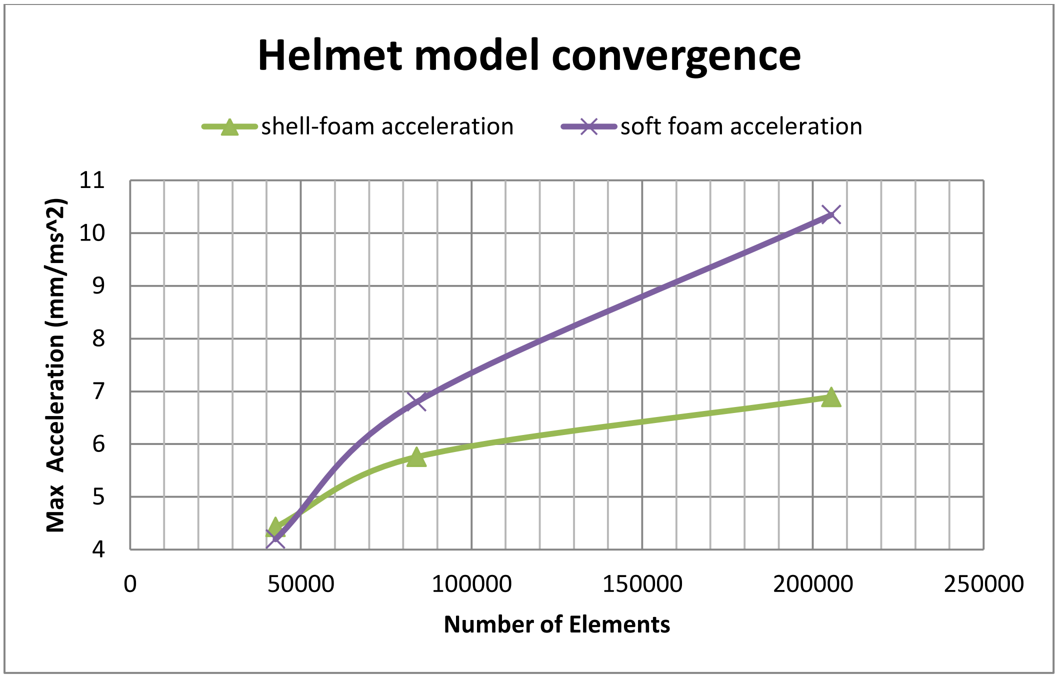

Figure 11 h-Method convergence study for the football helmet components

Table 3 Mesh convergence data

Mesh Density	Size of elements	Elements of soft foam	Elements of Hard foam	Elements of Shell	Elements of impactor	Elements of headform	Total Elements
Coarse	5mm	4788	14016	5736	18480	2319	45339
Medium	4mm	13772	27672	8569	30305	3690	84008
Fine	3mm	38559	74986	15898	69342	6617	205402

Table 4 Response values for each mesh density

Mesh Density	Impactor max vel. (mm/ms)	Impactor max accel. (mm/ms^2)	Shell foam max accel. (mm/ms^2)	Soft foam max accel.(mm/ms^2)
Coarse	1.8334	0.66118	4.4275	4.1945
Medium	1.94821	0.738961	5.7575	6.7997
Fine	2.0146	0.78517	6.8929	10.347

Table 5 Convergence data

	Impactor COG accel.	Impactor nodal vel.	Shell-foam accel.	Soft foam accel.
GCI	0.0559	0.0281	0.8579	1.5895
p	0.9778	1.0324	0.2825	0.3845

Table 6 Energy checks

Description	Acceptable Limit	Computed value
ER	> 0.9 and <1.1	Min = 0.99999; Max = 1.0024
Maximum SER (sliding energy/total energy)	< 0.1	0.040409894
Maximum KER (kinetic energy/ total energy)	< 1.0	0.997614841
Maximum IER (internal energy/ total energy)	< 1.0	0.918109541
Maximum HER (hourglass energy/total energy)	< 0.1	0.024739399

The energy of the system is also valuable for validating the quality of the finite element model. A list of energy checks are presented in Table 5 and the plot of the energies for the entire system are plotted against the time of the impact simulation. These results show that the energy levels are consistent with the current best practice energy values.

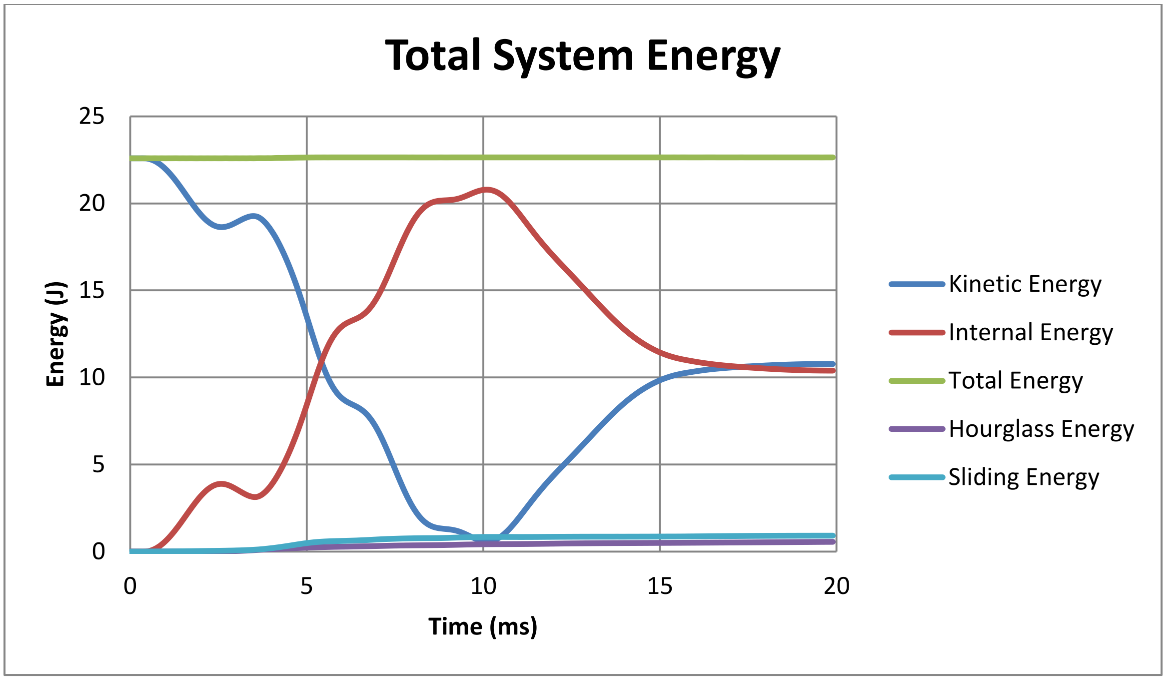

Figure 12 Total system energy comparing the kinetic, internal, hourglass, sliding, and total energy.

A total of five drop experiments were performed with the drop apparatus. The results of the experiments compared with the finite element model simulation are shown in Figure 11. The frequency content of the signal is determined by a fast fourier transform (FFT) of the experimental signal. The FFT shows that the majority of the signal amplitude comes from the low frequency response. As a result, the high frequency data is filtered with a moving average filter and the resulting low-frequency response is used for comparison.

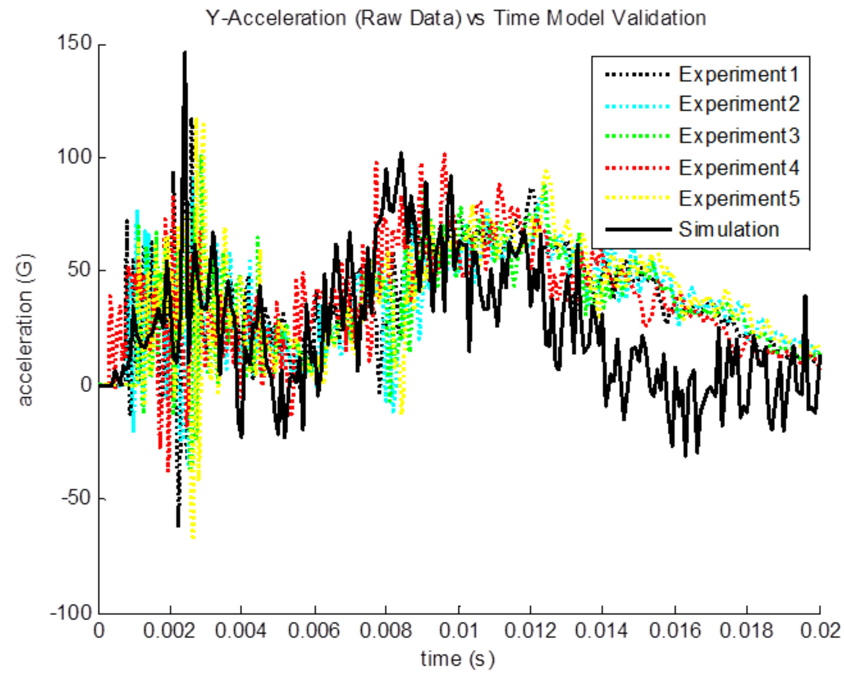

Figure 13 Experimental validation of Y acceleration in impactor

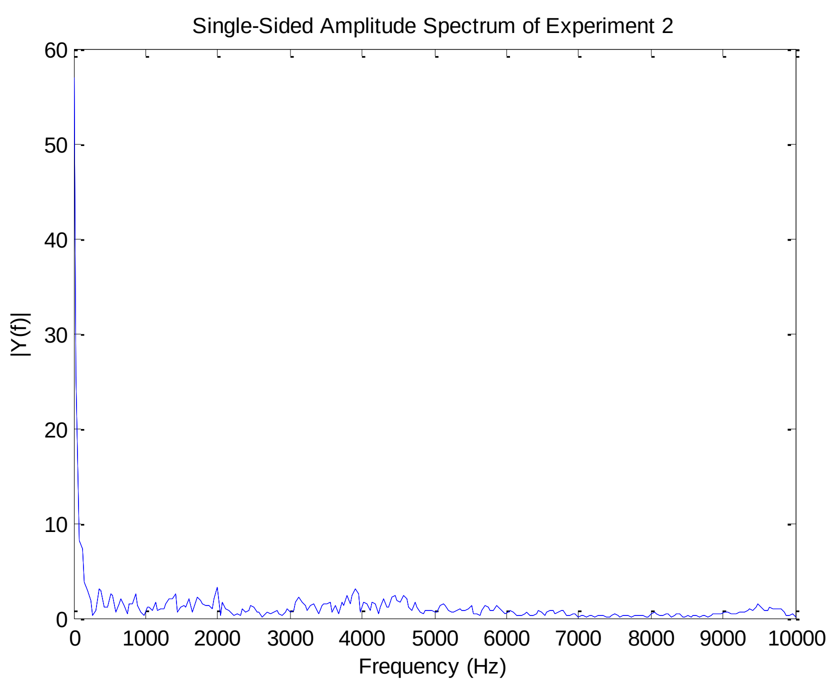

Figure 14 FFT of experiment 2

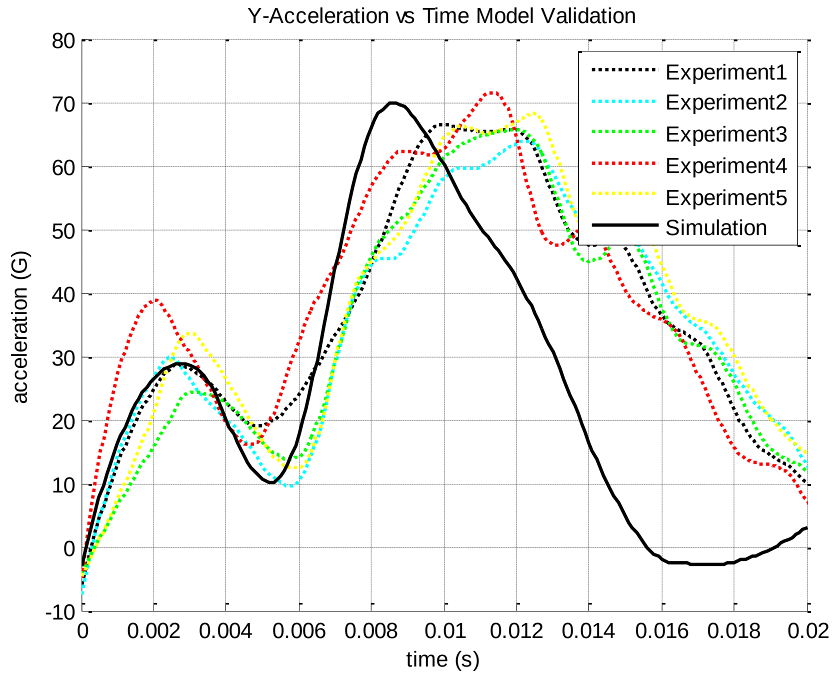

Figure 15 Filtered experimental and simulation data for low frequency response

Table 7 Low Frequency Acceleration results

Model	**Max Acceleration (G)**	**Max Acceleration Time (s)**
Simulation	69.8903	0.0086
Experiment 1	66.4897	0.0099
Experiment 2	63.9	0.0123
Experiment 3	65.66	0.012
Experiment 4	71.57	0.0124
Experiment 5	68.2208	0.0114
Average	67.1681	0.0116
Standard Deviation	2.603842187	0.000918695

The low frequency response shows that the model is able to recreate the salient features of the experimental acceleration profiles. Also the level of maximum acceleration was comparable for the simulation and the experimental simulation. The major discrepancy between the model and the experimental results is the time of max acceleration. The maximum acceleration in the simulation occurred at 0.0086 s and the experiments averaged to 0.0116 s for the max acceleration. See the tabulated data in Table 6 for reference. The reason for this discrepancy is due to an over-damping of the finite element model when compared to the experimental setup. For example, the simulation restricted all the degrees of freedom except for the dropping direction of the impactor. In the experimental apparatus the impactor is fixed at the location of the linear bearings.

As an additional layer of validation, the results from the model are compared against those obtained by Zhang (2001) who tested a football helmet model using the NOCSAE standard drop test equipment. It should be noted that Zhang used the Riddell VS4 helmet, and not the Riddell Revolution Attack helmet. The results are presented in Figure 11.

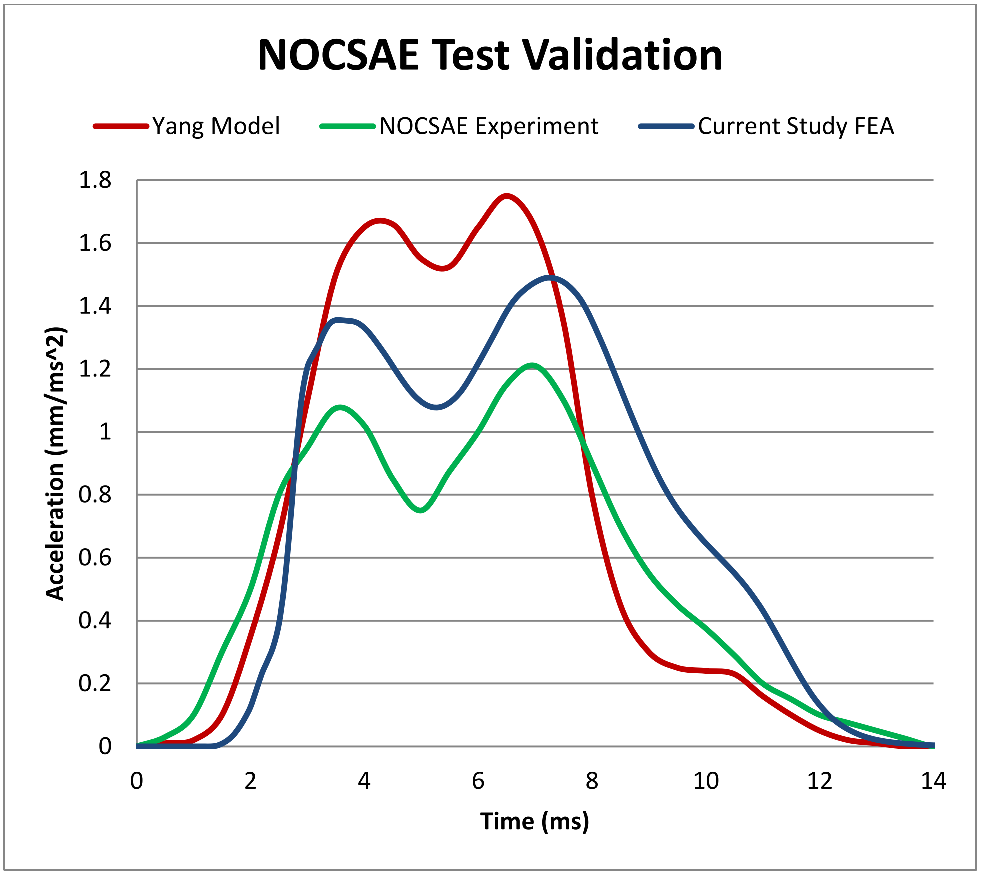

Figure 16 FE Validation against NOCSAE testing (Zhang 2001)

Both the current study and Zhang’s simulation exhibit a stiffer response than the experiment. This is probably due to the fact that the headform in the model was treated as a rigid body and the impactor is made of aluminum, whereas in the experiment there is a rubber pad at the site of impact and the headform is composite Hybrid headform III. However, in general the double peak behavior present in the experiments is well captured by the simulation models and the timing of the peak locations are much closer than in the prior experimentation.

Human Body Model Description

It is desirable to have as much of the human body model to study the dynamics of the system in impact scenarios. In this study the Global Human Body Models Consortium (GHBMC) is used. GHBMC is a consortium of seven automakers and one supplier who have consolidated their individual research and development activities in human body modeling into a single global effort to advance crash technology (Global Human Body Models Consortium 2014). The following list describes each component and where the model comes from:

- Head Model – Wayne State University, PI: Zhang L (Mao et al. 2013)
- Neck Model – University of Waterloo, PI: Cronin D (Dewit et al. 2012).
- Thorax Model – University of Virginia, PI: Kent R (Kent 2008).
- Abdomen Model – Virginia Tech and IFSTTAR, PI: Beillas P, and Hardy W (2011)
- Lower Extremity Model – University of Virginia, UAB and carhs, PI: Untaroiu C., Crandall J., and Eberhardt A. (Untaroiu 2005)
- Full Body Model – Wake Forest School of Medicine, PI: Stitzel J. (Vavalle 2012)

The total model includes 2,215,224 elements.

3.1 Human Head Model

The anatomical human head model was developed at Wayne State University by Mao et al (WSUHIM) (Mao et al 2013). The baseline head geometry was provided by Wake Forest University. They used supine magnetic resonance imaging (MRI) and CT scans of an average adult male head to gather the skin surface, skull and facial bones, sinuses, cerebrum, cerebellum, lateral ventricles, corpus callosum, thalamus, brainstem, and cerebral white matter. Where anatomical features could not be defined with the imaging techniques literature reviews were used to create the features, for example, differentiating the skull with outer cortical layer, middle cancellous layer, and inner cortical layer (Mao et al 2013).

Since large deformations are expected to occur in the brain simulations, hexahedral elements were primarily used in the model. The falx and tentorium were developed using four node shell elements. The shape of the features were adjusted to agree with the imaging data and to ensure good mesh quality with the cerebral spinal fluid (CSF). The thickness was defined as 1 mm which is consistent with previous head models done by Wayne State University (Zhou 1995). The CSF was meshed using tetrahedral elements to represent the complex geometry.

Available literature was used to mesh the bridging veins that are hard to identify in medical images (Mao et al 2013). Eleven pairs of bridging veins were included with a length that ranged from 6.63 to 17.91mm, an outer diameter of 2.76 mm and a wall thickness of 0.03 mm. The pia was developed by extracting brain surface nodes using 2D quadrilateral elements. The arachnoid used the surface of the CSF with 2D triangular elements. A total of 270,552 elements were used to create the model. This included 150,074 hexahedral elements, 352 pentahedral, 60,828 tetrahedral, 45,140 quadrilateral shell, 14,136 triangular shell, and 22 1D beam elements.

Biological materials display both elastic and viscous properties. Brain tissue is a hydrated soft tissue that consists of 78% water (Biltson 2011). As a result, the material properties of the head were determined by a large set of mechanical tests that have been performed on cadaver or animal specimens using compression, tension, shear, indentation, or magnetic resonance electrography methods. For a detailed review of how the material properties were obtained see review articles by Biltson (2011) and Chatelin et al. (2010). In summary, the gray matter and white matter were defined as linear viscoelastic materials with the white matter 25% stiffer than the gray. The cortical and cancellous bones were modeled with elastic-plastic materials, and the flesh was defined as viscoelastic material which provides the best simulation robustness. The skin membranes, falx, tentorium, dura, arachnoid, and pia were defined as elastic materials. A list of material properties for the model are listed in Tables 7, 8, and 9; and the finite element model of the head is broken into pieces for clarity in Figure 14.

Table 8 Linear viscoelastic material properties (Mao et al 2013).

Linear Viscoelastic					
Component	**Density (kg/m^3)**	**Bulk modulus (GPa)**	**Short-time shear modulus (kPa)**	**Long-time shear modulus (kPa)**	**Decay constant**
Cerebrum gray, cerebellum, thalamus, brainstem, basal ganglia	1060	2.19	6	1.2	80
CSF, 3rd Ventricle, Later ventricle	1040	2.19	0.5	0.1	80
Corpus callosum, Cerebrum white	1060	2.19	7.5	1.5	80
Facial Tissue	1100	0.005	0.00034	0.00014	0.00003
Scalp	1100	0.02	0.0017	0.00068	0.00003

Table 9 Elastic material properties (Mao et al 2013).

Elastic			
Component	**Density (kg/m^3)**	**Young's Modulus (GPa)**	**Poisson's Ratio**
Membrane	1100	0.0315	0.315
Skin	1100	0.01	0.45
Dura	1100	0.0315	0.35
Falx, Pia	1100	0.0125	0.35
Arachnoid	1100	0.012	0.35
Tentorium	1100	0.0315	0.3
Maxillary, Phenoida and Ethmoidal sinus	1000	0.001	0.3

Table 10 Linear elastic plastic material properties (Mao et al 2013).

Linear Elastic Plasticity						
Component	**Density (kg/m^3)**	**Young's Modulus (GPa)**	**Poisson's ratio**	**Yield Stress (GPa)**	**Tangent Modulus (GPa)**	**Plastic strain failure**
Bridging vein	1130	0.03	0.48	0.00413	0.0122	0.25

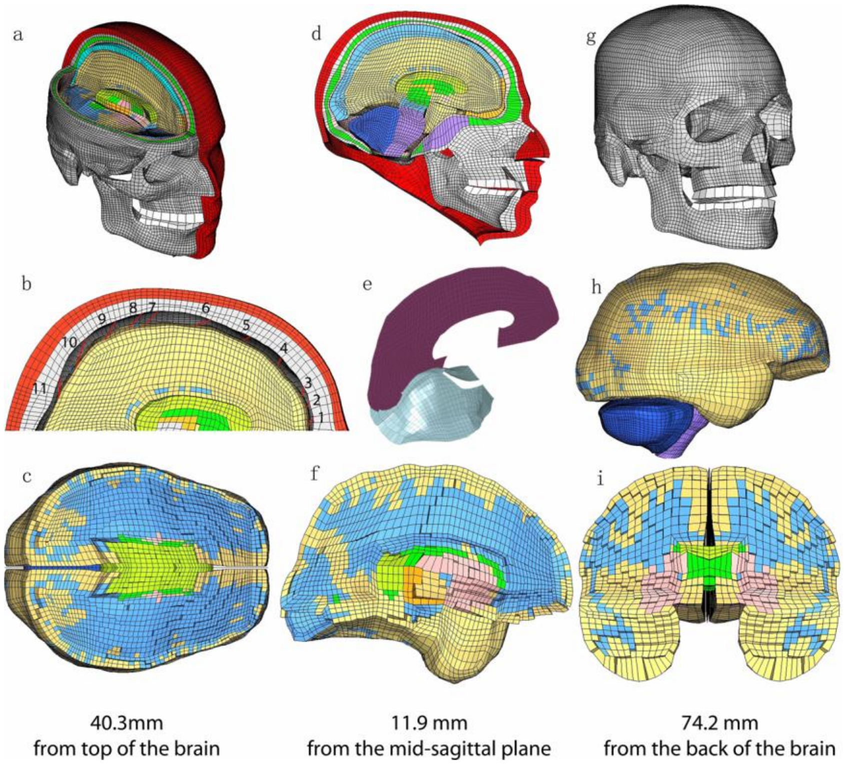

Figure 17 The human head FE Model. (a) Head model with brain exposed; (d) medium sagittal view of the head model; (g)skull and facial bones; (b) 11 bridging veins; (e) falx and tentorium; (h) brain; (c), (f), (i) brain sectional views in three directions (horizontal, sagittal, and coronal) (Mao et al 2013).

3.2 Coupling of Human Body and Helmet Model

The human body model and helmet interaction requires contact definition. The specific bodies that are identified for this contact are the skin of the human head and the comfort foam layer in the helmet. The dynamic coefficient of friction between the human head and the comfort foam is taken as 0.01. See figure 15 for a representation of the coupling of the two models.

Figure 18 GHBMC Model Integration: (a) full GHBMC model integrated with the football helmet model and impactor. (b) Cross section of the GHBMC to note anatomical detail.

There were two load conditions created using the coupled system. The first analyzed a crown impact that involved a plate impacting the crown of the helmet at a velocity of 5 m/s and data was taken for a duration of 15 ms. The second case was a frontal impact (as illustrated in figure 18) and it impacted the helmet at 5 m/s for a duration of 30 ms. The extended simulation time for the frontal impact is due to maximum response values appearing later in the impact simulation.

Full Model Simulation Results

In this chapter, the results of the coupled helmet-human body system in an impact on the crown of the head and an oblique impact to the frontal bone are presented and discussed. The GHMBC model computes the brain response in specific anatomical structures and since mTBI is a primary concern for football helmets the anatomy related to that category of injury is presented. McAllister et al. (2012) showed that the maximum strain and strain rate fields in brain tissue during impact agree with the changes in white matter integrity in the area of the corpus callosum after concussive events in the game of football. This validates that monitoring the strain and strain rate field accurately can help properly predict brain injury and the effectiveness of a football helmet. As a result the brain response data for the current study focuses on the midbrain, corpus callosum, and the brain stem, see Figure 16 for an illustration of the location of these components in the human head for this model. Pfister et al. (2003) showed that axonal injury and neural cell death occurs when applying strains from 20-70% and strain rates in the range of 0.020-0.090/ms to create mild to severe axonal injury. The strain and strain rate values obtained from the two load conditions were compared to these thresholds to determine if the helmet was effective in minimizing the probability of injury. The Gadd severity index (GSI) of the skull is also computed as a comparison to the brain response metrics.

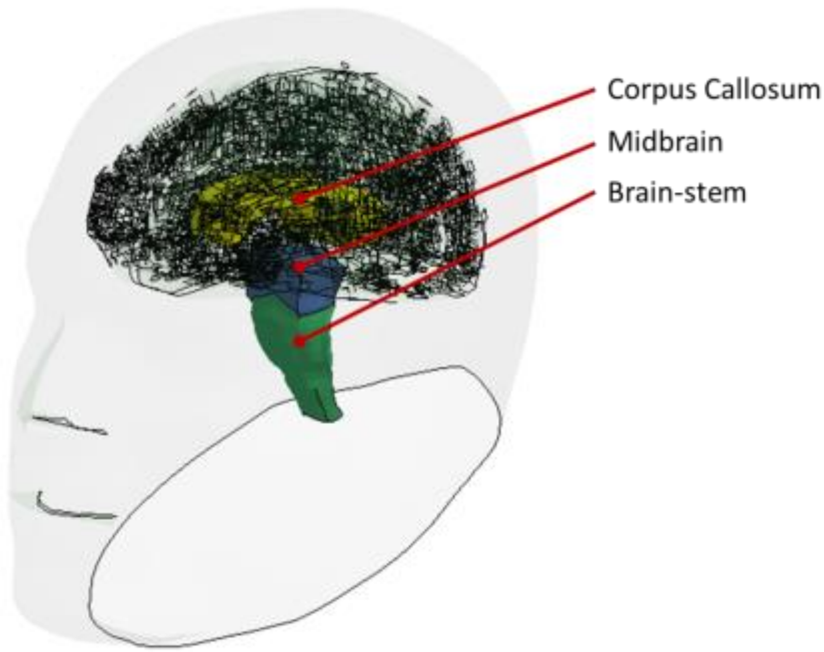

Figure 19 Location of critical anatomy to mTBI

The integration of a full human body model allows for an elimination of unnecessary assumptions regarding the heads relationship to the neck and the neck's relationship to the rest of the body. Figure 17 and 18 give the progression of deformation during the crown and frontal impact scenario, respectively. As a result in both cases the neck experiences severe compression during the crown impact. The GHBMC has an option to toggle the muscles on or off and for this case the muscles in the neck were considered off. A sensitivity analysis of the effect of muscle engagement should be done in the future.

4.1 Wave Propagation

To determine how much of the full body model is needed for future analysis a study of the wave propagation throughout the body during impact was performed. Figure 20 shows the final displacement through the body during the 15 ms crown impact. Then, by monitoring specific nodes placed evenly throughout the body the wave of displacement throughout the body is shown. Figure 21 shows the displacement response for the duration of the impact. This shows that the different materials lead to different wave speeds. For example, the spinal cord propagates the displacement field at a much faster rate than the surrounding soft tissue. As a result, the backbone experiences the greatest deformation and the displacement wave reaches down to about the pelvis area of the full human body model. This shows that for a 15 ms simulation including the arms and legs in the simulation do not have a significant impact on the response of the brain.

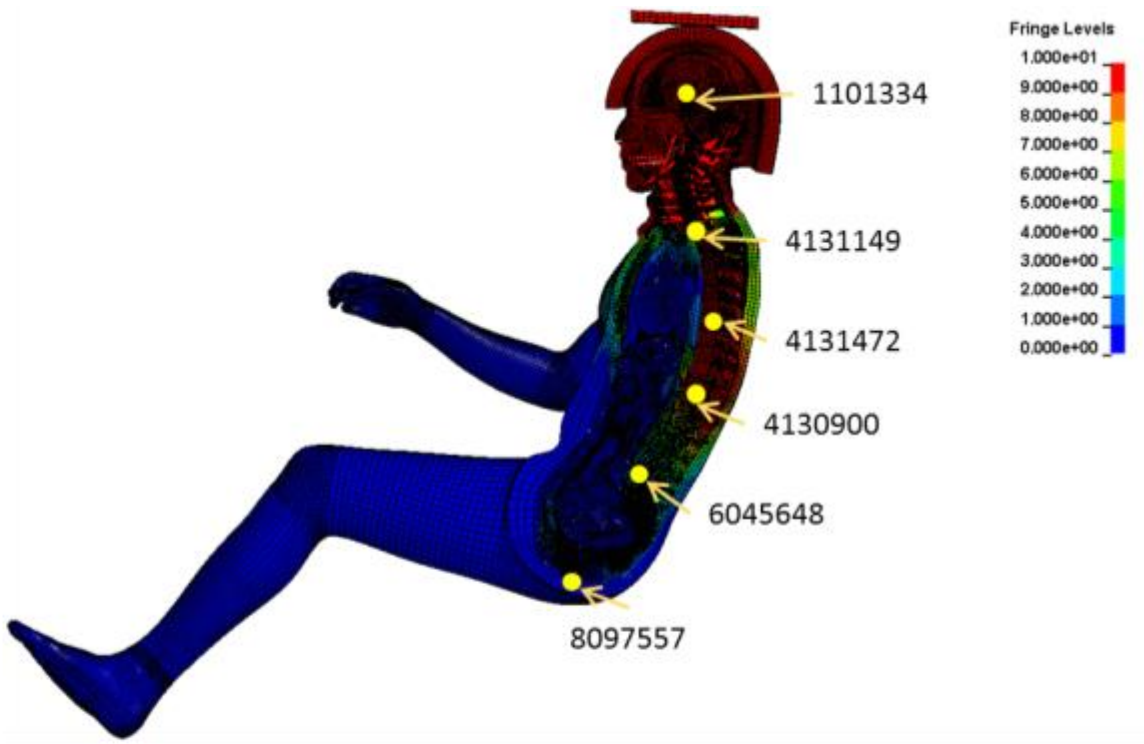

Figure 20 Displacement at time t = 15 ms

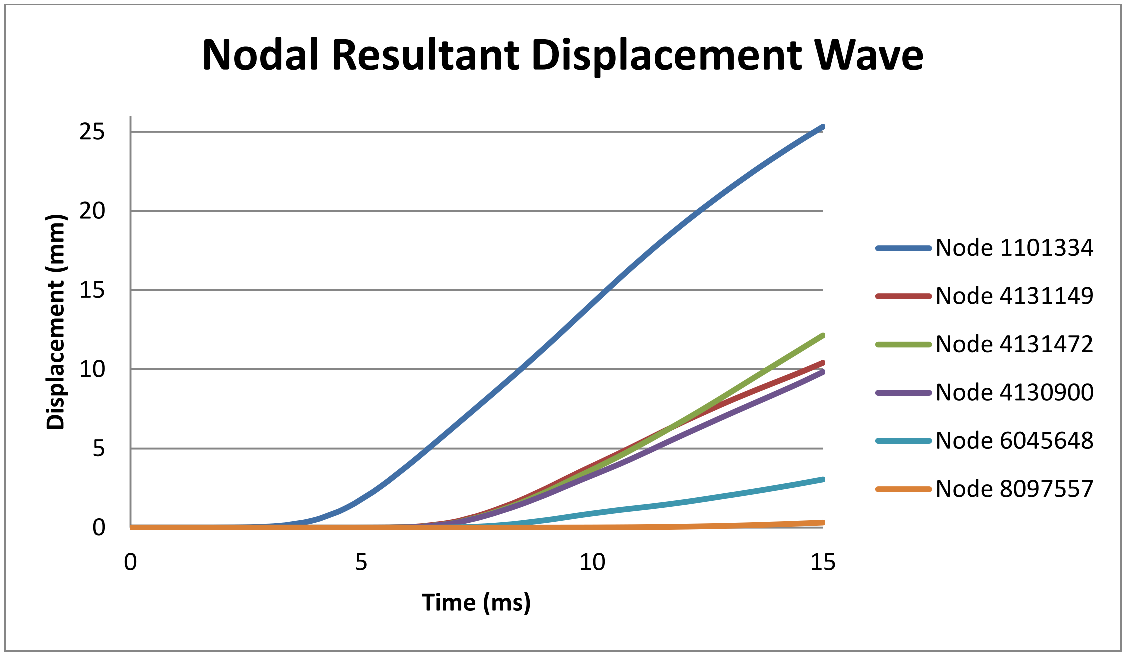

Figure 21 Displacement propagation for the crown impact

4.2 Impact Results

For the frontal loading condition the head is pushed back and sever deformation is noted near the chin of the human model. These plots give insight into the participation of the neck and chest for brain response data.

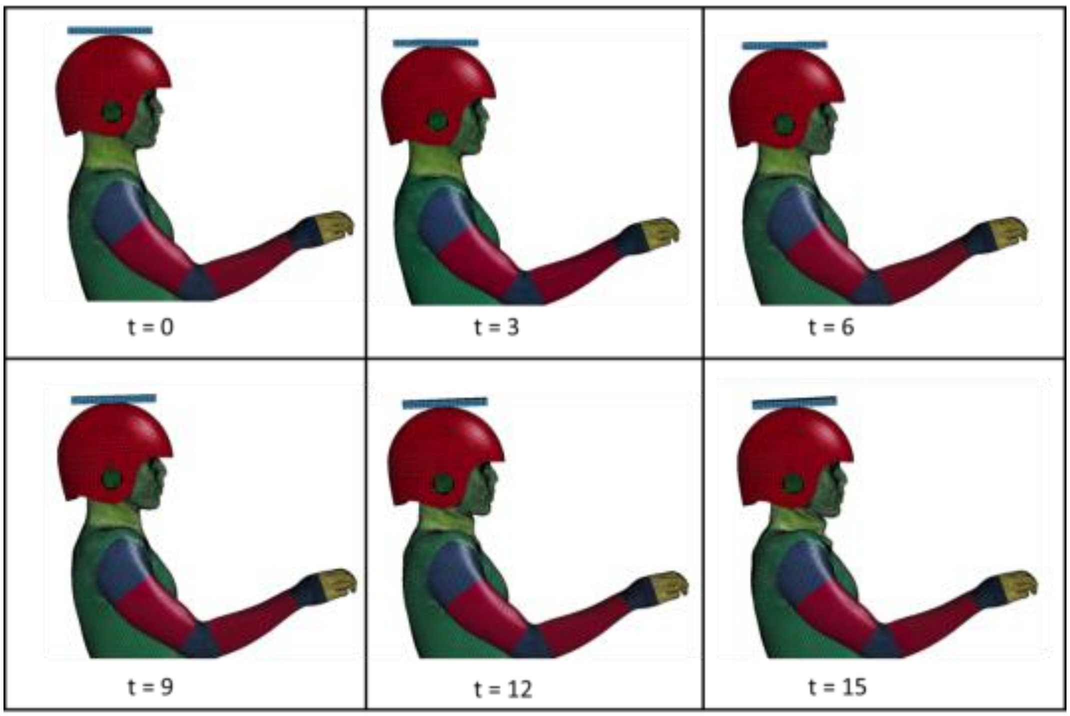

Figure 22 Deformation of the human-helmet system during crown impact

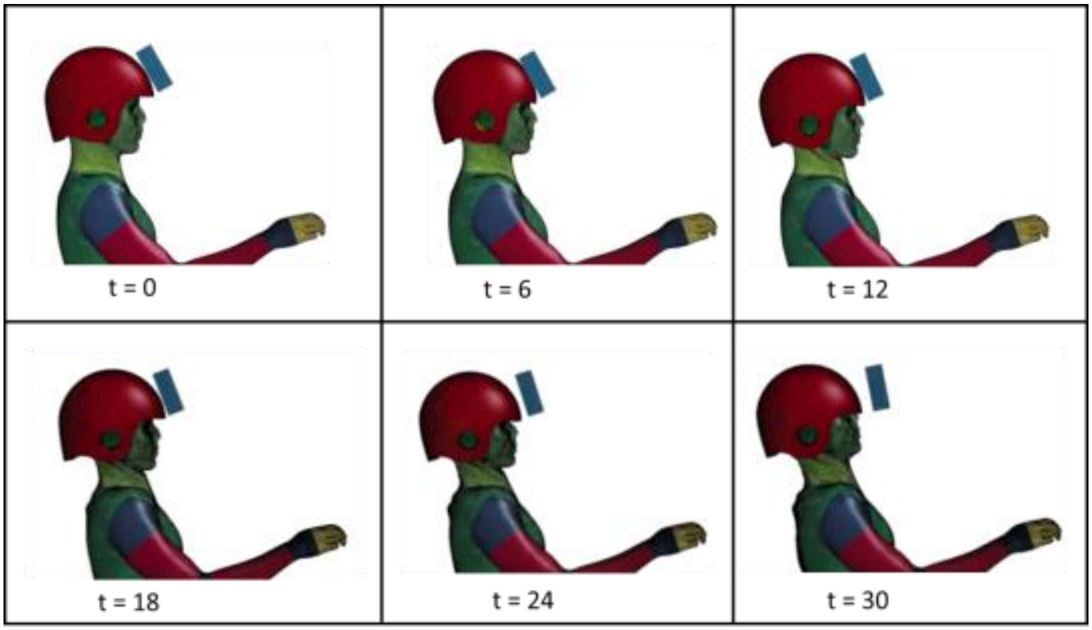

Figure 23 Deformation of the human-helmet system during frontal oblique impact

To identify critical time points during impact a deformation plots tracking the corpus callosum, midbrain, and brainstem with respect to the skull displacement for both the crown and frontal loading conditions are shown in Figure 24 and 25. The crown impact results in an oscillation of the brain features with respect to the skull. A different phenomenon occurs with frontal loading. The Corpus Callosum is pushed toward the top of the skull whereas the midbrain and brainstem components are pulled toward the neck. This equates to an impact condition that experiences a larger level of overall max strain however the magnitudes of the strain values still stay under the prescribed threshold conditions for injury. This is demonstrated through the max strain, strain rate, and stress seen in Figures 24-31 for the components of interest.

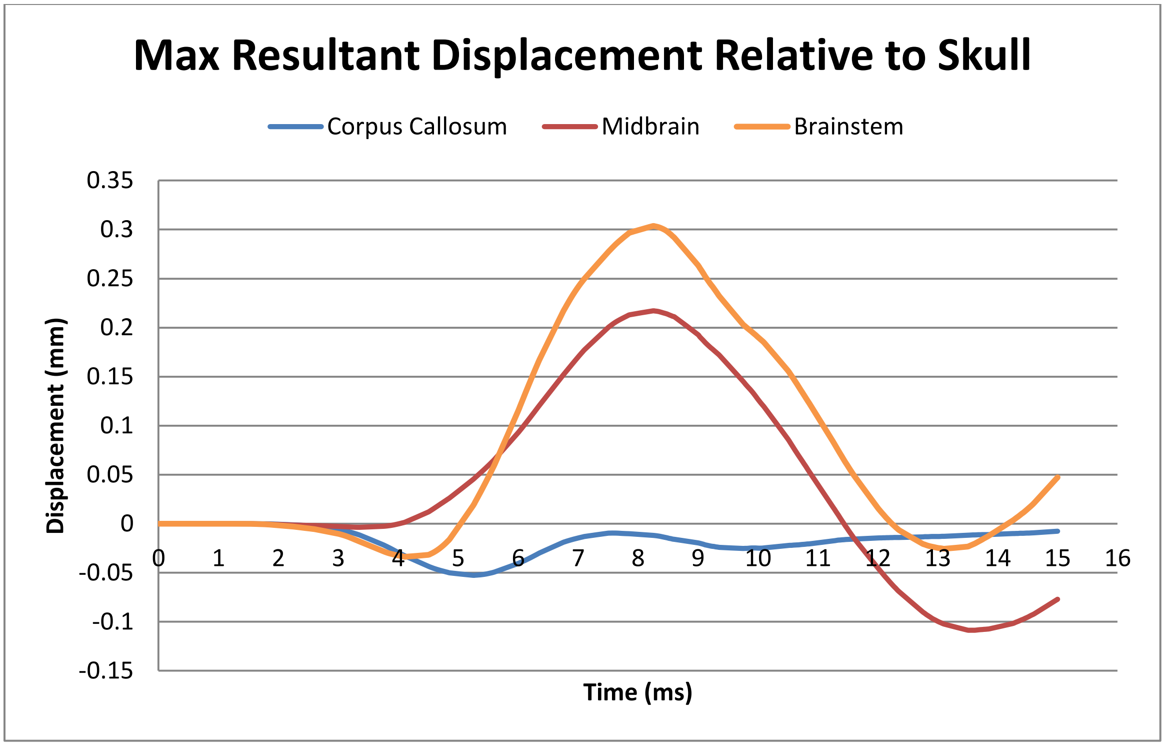

Figure 24 Crown impact relative displacement of corpus callosum, midbrain, and brainstem components.

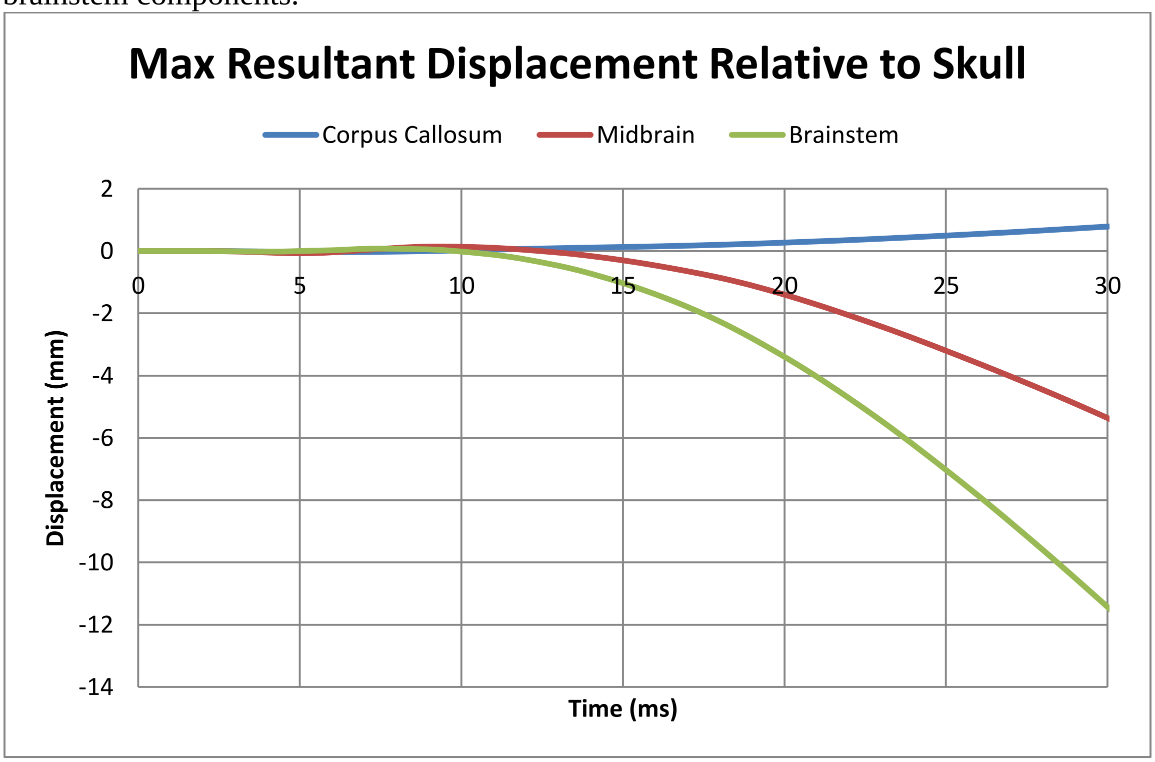

Figure 25 Frontal impact relative displacement of corpus callosum, midbrain, and brainstem components.

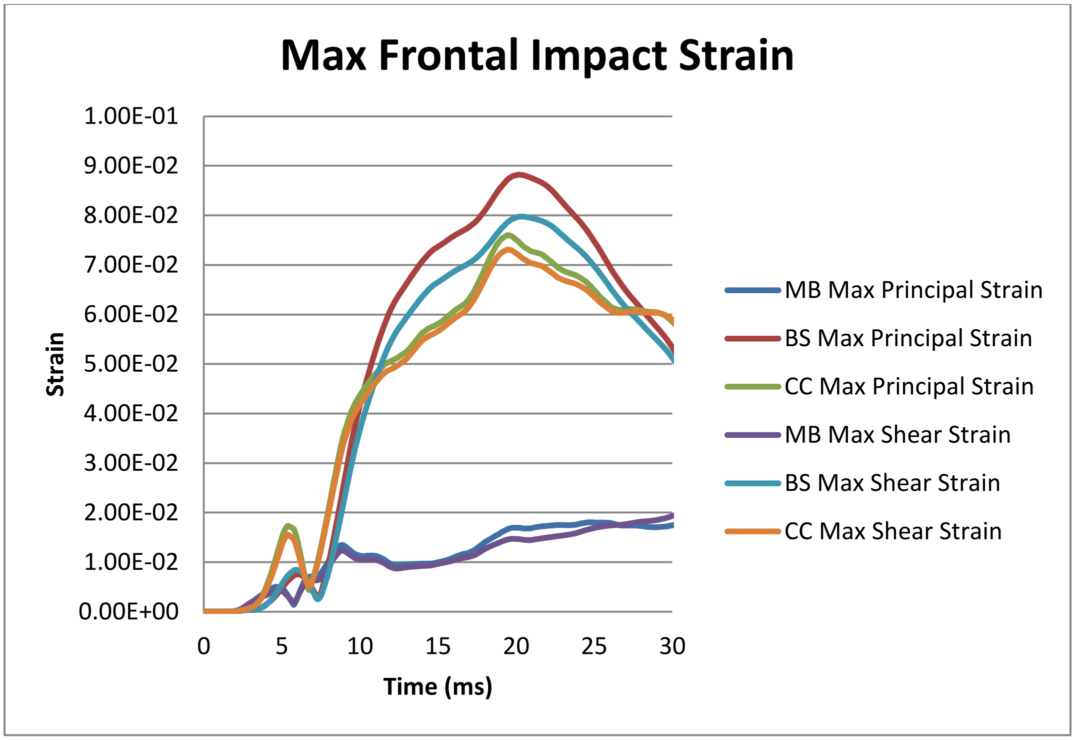

Figure 26 Frontal impact max strain

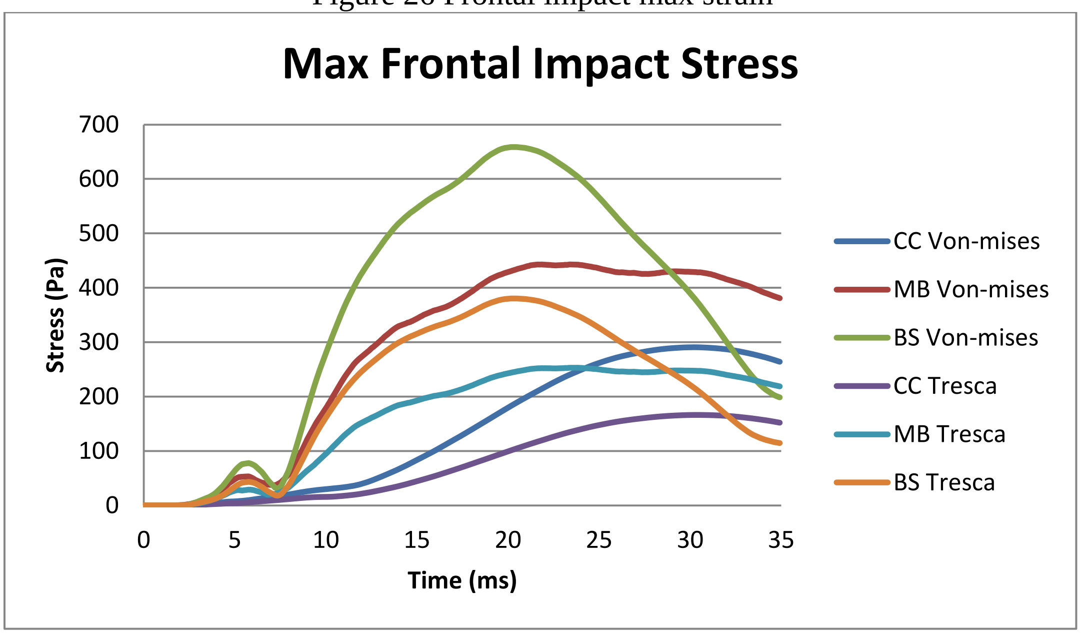

Figure 27 Frontal impact stress

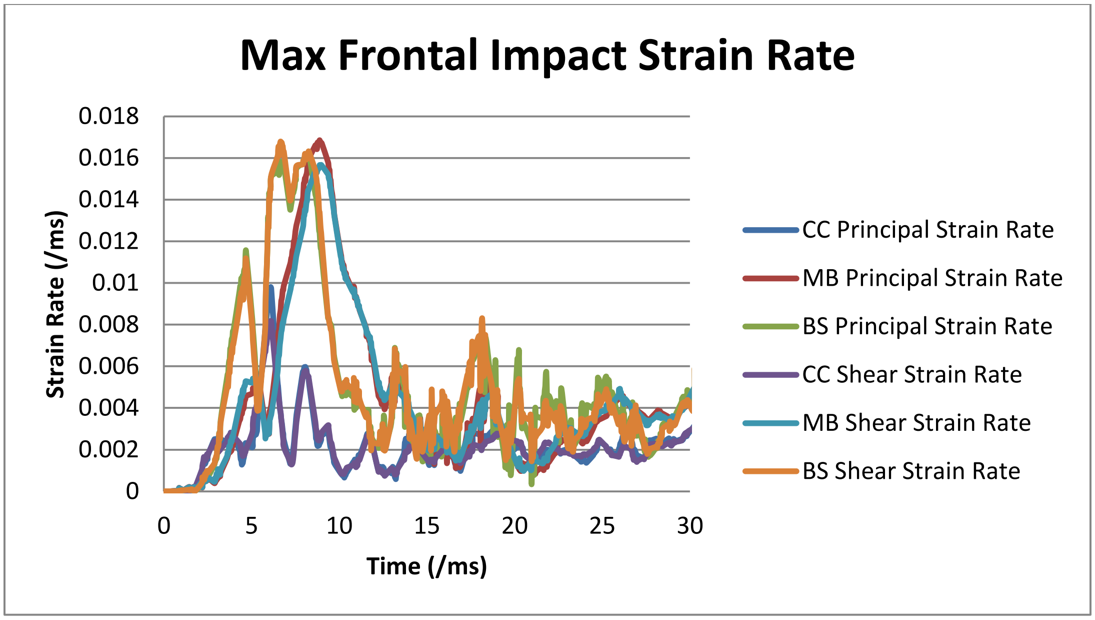

Figure 28 Frontal impact max strain rate

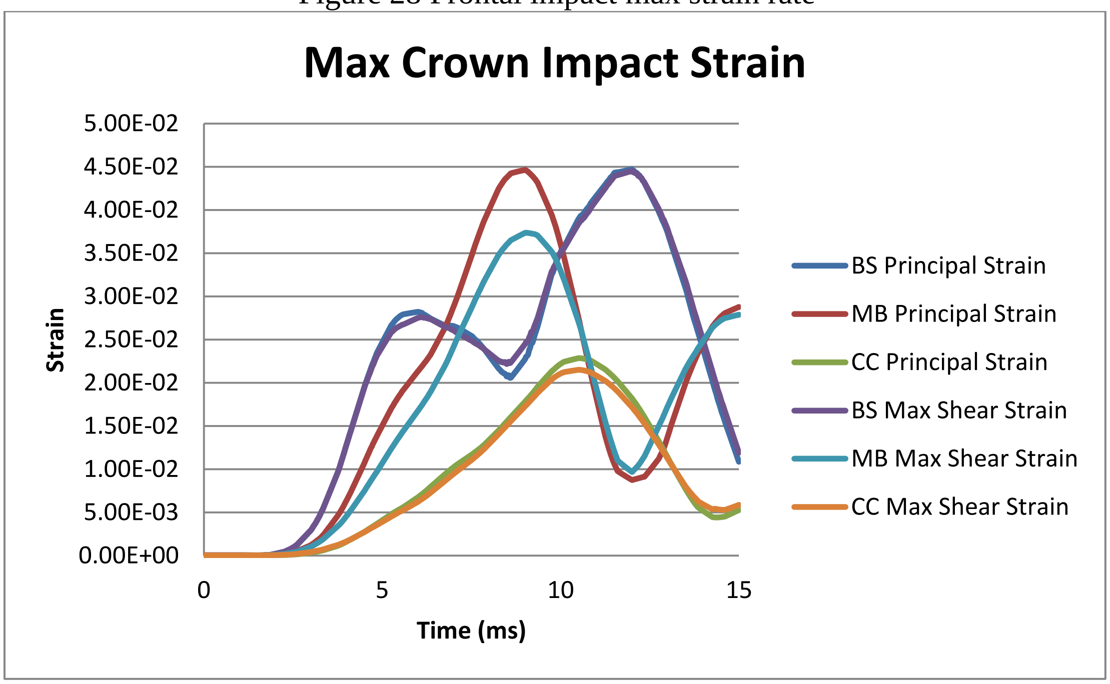

Figure 29 Crown impact max strain

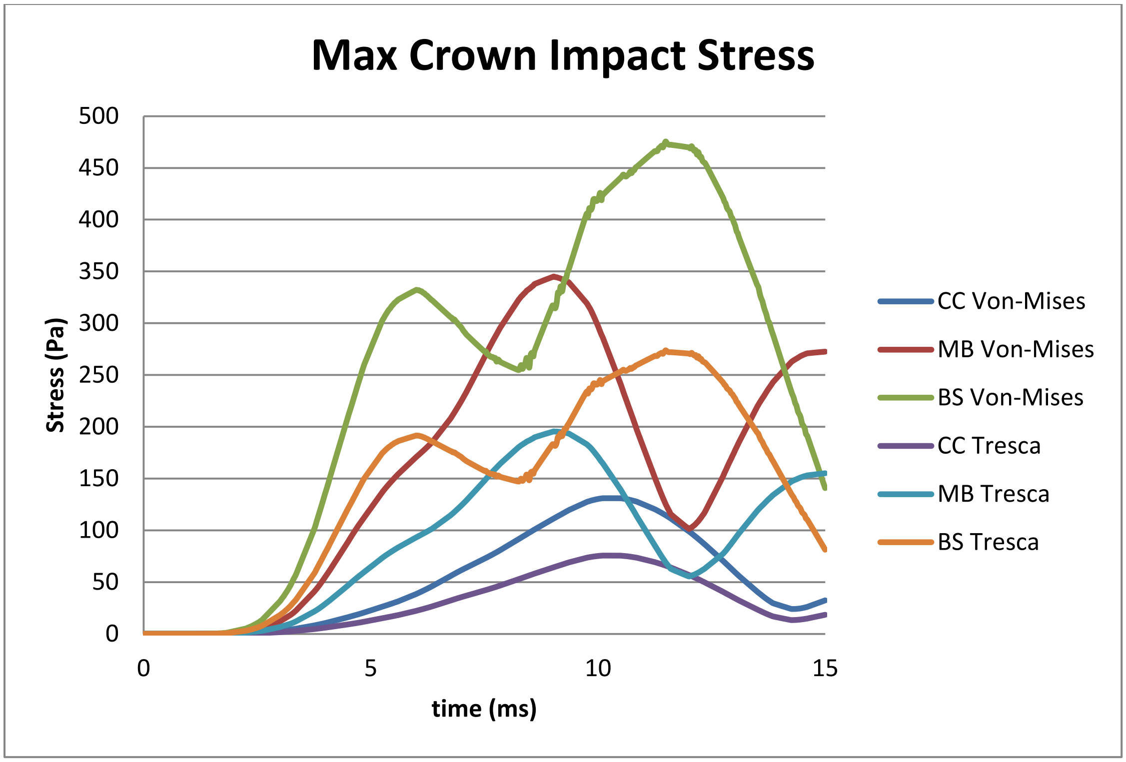

Figure 30 Crown impact max stress

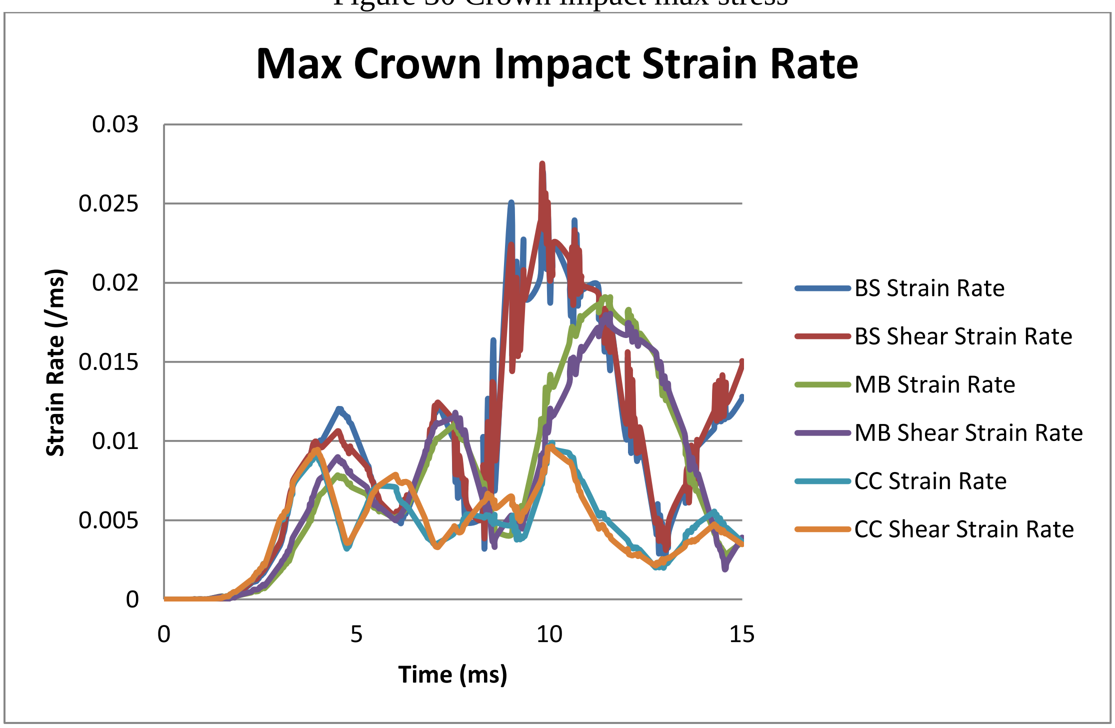

Figure 31 Crown impact max strain rate

Comparing Figure 26 and 29, the maximum principal strain obtained by the frontal impact condition was 8.88% and for the crown impact it was 4.47% both occurring in the brain-stem. While the strains doubled for the frontal impact condition the strains were still lower than the defined thresholds by Pfister et al. (2003) for brain injury. Comparing Figure 27 and 30, the maximum von-mises stress for the crown impact was 476 Pa and the frontal impact was 659 Pa both occurring in the brain-stem which is a similar story as the principal strain values however specific threshold values for the stress in the brain is unknown for this study. Comparing Figures 28 and 31, the maximum principal strain rate for the crown impact was 0.027 /ms and for the front impact it was 0.016 /ms both occurring in the brain-stem. These results when compared against the thresholds for injury defined earlier show that the crown impact generates a potential injury inducing loading condition as the strain rate obtains a value larger than 0.020 /ms. See Tables 10 and 11 for the list of maximum values.

The maximum values of stress, principal strain, and principal strain rate in the brainstem of the brain illustrate an agreement with the clinical evaluations seen when studying patients who have experienced an mTBI. The brainstem can be associated with regulating consciousness of the patient and so damage or high strains in that region would cause the patient to lose consciousness which occurs in some cases for mTBI. Using this model the crown impact would be characterized as a potentially damaging loading condition due to the strain rates experienced in the brainstem.

The NOCSAE standard uses the Gadd severity index (GSI) to determine if a helmet is adequate in preventing injury. In the physical test environment, this value is computed by gathering the acceleration of the center of gravity for the headform, and the duration of impact. The same procedure was performed using the skull of the finite element model and the center of gravity acceleration was gathered to compute the Gadd severity index. Figure 32 shows the acceleration profile for the two loading conditions. Using this data the GSI for the crown impact was calculated to be 1477.8 and the frontal impact was 1214. The NOCSAE standard defines a requirement for helmets to have a GSI lower than 1200. This means that for both impact scenarios the helmet would have failed. Also the crown impact should be more dangerous than the frontal impact. This is in agreement with what was found in the brain response simulation as the strain rates in the brainstem reached magnitudes that were within the threshold for brain tissue damage.

This model shows that by using a validated football helmet model and a robust human body model, great insight to the effect of potential loading conditions for the sport of football can be explored. This information can be used to improve helmet design and reduce the risk of mTBI currently found in football.

Table 11 Max values for crown impact

	Brainstem	**Midbrain**	**Corpus Callosum**
Max Principal Strain	0.044686	0.044627	0.022845
Max Shear Strain	0.044497	0.03736	0.021496
Strain Rate (/ms)	0.027129	0.019097	0.009888
Shear Strain Rate (/ms)	0.027465	0.018049	0.009662
Von-Mises (Pa)	476	345	131
Tresca (Pa)	274	196	75.6

Table 12 Max values for frontal impact

	Brainstem	Midbrain	Corpus Callosum
Max Principal Strain	0.0882	0.0294	0.0760
Max Shear Strain	0.0798	0.0283	0.0731
Strain Rate (/ms)	0.0162	0.0169	0.0098
Shear Strain Rate (/ms)	0.0168	0.0157	0.0082
Von-Mises (Pa)	658.755	442.893	290.679
Tresca (Pa)	380.307	253.003	166.217

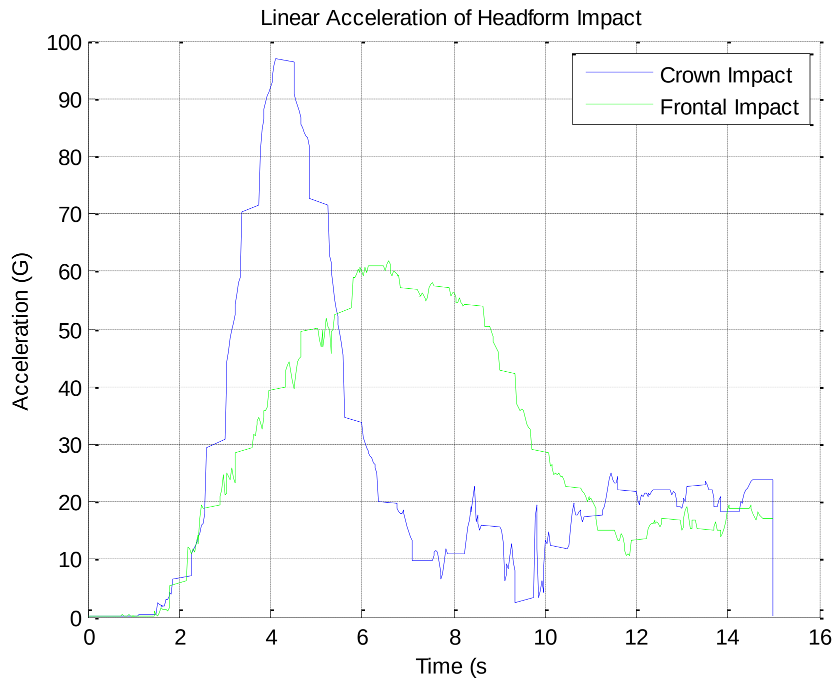

Figure 32 Acceleration profiles **of** the skull during impact

Concluding Remarks

An integrated human body-football helmet finite element model was created and analyzed using LS-DYNA. The GHBMC human body model was used. The Riddle Revolution Attack helmet was modeled with particular attention paid to its polycarbonate shell and bi-layer foam system, and was placed over the skull portion of the GHBMC model. The integrated model was then subjected to centric and non-centric impact scenarios.

The general conclusions from this research are as follows.

- It is necessary to recreate the geometry seen in modern football helmets in order to obtain reasonable simulation results.
- When meshing the foam and plastic layers in the helmet, care has to be taken in keeping a relatively good mesh quality with a majority of the elements being hexahedral elements. The large deformations experienced by the foam require adequate mesh density to allow the contact algorithms to work and provide numerical stability in the computations.
- Verification and validation of the material models is possible using publicly available data and experimental results. More sophisticated regression analyses can be used to minimize the error between the experimental results and numerical simulations.
- Creating a headform out of 6061 aluminum slabs and an ABS plastic cap is a valid method to producing a rigid headform for model validation. While this setup does not give the same biomimetic properties of the Hybrid Headform III, it is an

order of magnitude less expensive and more modular, i.e. the size can be adjusted easily to meet the needs of the experiment.

- The validated football helmet and human body model from the GHBMC can be integrated into one model in LS-DYNA. Load conditions can be applied to this model and anatomy of particular interest can be isolated to study the total system effects. For example in this analysis, a crown impact was performed and the strain-rate, strain, and stress values experienced in the corpus callosum, midbrain, and brain stem was determined.

The following future studies are suggested.

- A more robust modeling technique can be used to characterize the helmet foams. For example, the keyword MAT_FU_CHANG_FOAM allows for multiple load curves to be input into the material model to define the strain-rate dependent responses of the material [43]. Hysteresis behavior present in these low density foams can also be included in the model.
- Setup and analyze cadaveric impacts and compare them against the finite element model results.
- Improve the characterization of brain trauma using response parameters available from a typical finite element analysis.
- Optimize the helmet design via sizing, shape, topology and material optimization techniques so as to minimize the risk of mTBI.

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