Effect of Helmet in Preventing Human Head and Brain Trauma in Traffic Accidents

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ABSTRACT

To date, there have been number of studies to demonstrate the effectiveness of helmet during head impacts in car or motorcycle accidents, but these studies were focused on frontal impacts to the head only. There is still a need to study more on helmet to make it more effective, especially in lateral impacts. To my knowledge, previous studies did not focus on the effects of helmet during lateral impacts. This study focuses on lateral impacts and how the helmet prevents head injuries during such impacts. In this thesis, 3D models of the human head and helmet were created in order to build a parameterized Finite Element (FE) model of a protected human head. In the FE model, brain is assumed to be hyper-elastic, the helmet is considered as elasto-plastic and the skull is considered as a linear elastic material.

Two separate FE analyses were carried out in this study to find out how the helmet prevents lateral impacts to the head. In the first analysis, the head was not protected and it hits the impact object directly. In the second analysis the head model is protected with a one-layered helmet. In each of these simulations normal and Von Mises stresses are calculated and in the end compared with each other in order to understand the influence of helmet in reduction of such stresses on human head. Afterwards the acceleration of center mass of the head during two simulations are calculated and compared to each other. Helmet reduces head acceleration by 83%, which is a very important cause of Traumatic Brain Injury (TBI). It is clear that the helmet can be very crucial in avoiding brain trauma during car accidents under lateral impacts.

Through applying more sophisticated modeling in the FEA, the stress distribution at each layer of the helmet and at each part of the head could be illustrated. Therefore, this knowledge will provide insight into designing and developing helmet and safety products to reduce head and brain trauma.

Keywords: Traumatic Injury, Helmet, Lateral Impacts, Head, Biomechanics

ÖZ

Bu güne kadar yapılan bir çok araştırmada, araba ve motor kazalarında kask kullanımının etkisi incelenmiş fakat bu araştırmalar, kazalarda başın ön tarafına gelen darbeleri ele almıştır. Pek çok trafik kazası sonucunda baş yaralanmaları yan taraftan alınan darbelerden dolayı da olabilir. Fakat hala daha yan taraftan alınan darbelerin kask ile nasıl zararsız hale getirileceği incelenmemiştir. Yaptığım araştırmalara dayanarak, çok az bilimsel çalışmada başın aldığı yan darbeler ele alınmıştır. Bu araştırma tezi, başın aldığı yan darbelerde kask kullanımını ve yaralanmalardaki etkilerini incelemiştir. Bu tezde ilk olarak yapılan, üç boyutlu baş ve kask modellerinin yaratılmasıdır. Daha sonra bu modeler sonlu elemanlar yazılımına aktarılmış ve analize hazır duruma getirilmiştir. Sonlu elemanlarda kullanılan modellerde, kafatası doğrusal ve elastik, beyin, hiper elastik ve kask elasto-plastik olarak tanımlanmıştır.

Sonlu elemanlar kullanılarak iki tür analiz yapılmıştır. Bunlardan birincisinde kask kullanılmamış ve başa alınan yan darbelerde, başın gösterdiği mekanik davranış elde edilmiştir. Ikinci çalışmada, hem baş hemde kask modelleri kullanılmıştır. Her iki analizde de hem normal hemde Von Mises stress dağılımları elde edilmiş ve birbirleriyle karşılaştırılmıştır. Burada elde edilen sonuçlardan kask kullanımının başa gelen yan darbenin şiddetini düşürdüğü ve başı koruduğu anlaşılmıştır. Daha sonra ivme hesaplamaları iki tür analiz için de yapılmıştır. Böylece yan darbelerde başı korumak için kask kullanımının önemi ortaya konulmuştur.

Bu çalışma ileride daha sofistike kask ve baş modellerinin yaratılmasının analizlerde daha detaylı sonuçlar elde edilmesinde yararlı olacağını desteklemektedir. Bunların yanında elde edilen detaylı analiz sonuçları da beyin ve baş travmalarını önlemeye yarayan kask ve benzeri koruyucu ürünlerin tasarım ve geliştirilmesine katkıda bulunacaktır.

Anahtar Kelimeler: Travmatik Yaralanma, Kask, Yan Darbeler, Baş, Biyomekanik

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LIST OF ABBREVIATIONS

3D Three Dimensional

CSF Cerebrospinal Fluid

COR Coefficient of Restitution

CPU Central processing Unit

CT Computer Tomography

DAI Diffuse Axonal Injury

EDH Epidural Hematoma

EMC Expectation Maximization Classification

FE Finite Element

FEA Finite Element Analysis

FEM Finite Element Method

MRI Magnetic Resonance Imaging

Pa Pascal, a pressure unit

TBI Traumatic Brain Injury

Chapter 1

INTRODUCTION

1.1 Need for Head and Brain Modeling in Traffic Accident Analysis

Since the invention of car, car accidents have become a danger to human lives. Every day men, women and children from all over the world become victims of head injuries, which can range in intensity from mild concussions to coma and death. Nowadays, deaths over car accidents have become one of the leading causes of death worldwide. The estimated brain injury mortality rate is 22-25 per 100,000 people annually (Kleiven, 2002).

Introduction of the new imaging technologies such as magnetic resonance imaging (MRI) has improved our knowledge of the human head and as a result, medical treatments of head injuries have become more efficient. On the other hand, with the increasing rate of brain trauma related fatalities, prevention of such accidents has become more significant.

There are many theories on how the trauma happens to the brain but almost all the theories agree that the injury is related to sudden acceleration or deceleration of the head, even if the impact is not applied directly to the head. The brain is loosely situated inside the skull so it can be smashed against the interior walls of the skull upon impact.

Many studies have been conducted to simulate brain trauma that can occur from head impact to find the reasons why the trauma happens. Some experimental studies have been performed by using animals, cadavers and crash dummies to calculate the forces, accelerations and displacements (Nahum et al. 1977). Using cadavers have many advantages because they have the same geometry and mass distribution as an alive person, although experiments on cadavers related to the mechanisms of damage are not accurate. This is due to some differences in texture between a cadaver and a living body due to dehydration and decaying of the cadaver. On the other hand, they are not widely available to everybody due to their costs and ethical reasons.

Development of the Finite Element Method (FEM) brought many advantages to biomechanical researches. FEM is a numerical technique for finding approximate solutions to partial differential equations and it can be used to simulate impacts of the head and to find approximate results for the response of brain to impacts. Introduction of FEM encouraged researchers to develop more accurate 3D models. The early models were only simple geometries simulating head and brain, but as the technology improved more accurate models were created.

There are various ways to create skull and brain models either by using Magnetic Resonance Imaging (MRI) and Computer Tomography (CT) scan data and extracting the brain and the skull using the Brain Extraction Tools (BET) available in various software products (e.g. FSLView manufactured by Oxford University, Mimics and 3D doctor) or by using 3D printers to scan and model them inside Solidworks and other design software and even by creating the models from the scratch using Rhino 3D and

Solidworks. For the head and brain trauma analysis, there are many software packages available, including ANSYS, LS-DYNA and ABAQUS.

All the studies previously conducted on impacts to the head suffer from a big problem and that is the insufficient knowledge we have on human body. As a result researchers included different material properties for the body parts, which would affect the performed analysis.

On the other hand, due to lack of technologies and availability of powerful processors, researchers had to simplify models, and impact conditions and use lower quality meshes to reduce the processing time. So it is clear that there is still a need to model the human head tissue more accurately in order to be able to understand the conditions under which brain trauma happens under lateral impacts and hopefully try to find better ways of preventing these traumas.

1.2 Anatomy of the Human Head

Human head consists of many parts and each one of them has specific parameters. The size and shape of these parts vary between individuals. Before going into more details about the human head modeling, it is adequate to explain the main parts of the human head as follows:

1.2.1 Scalp

Scalp has 5 different layers which overall are between 5 to 7 mm thick. These layers are skin, connective tissue, epicranial aponeurosis, loose areolar connective tissue and pericranium. Although the scalp is a very thin part, due to its high elasticity, it helps to absorb impacts and thus reducing the impact forces that reach the brain.

1.2.2 Skull

Skull is the boney structure of the head, and its thickness varies between 4 to 7 mm. It has 8 cranial bones and 14 facial bones. All the bones of the skull are joined together so they have little movement, except the mandible, which can freely move and rotate. Skull protects the brain from impacts and shapes the head. The brain tissue is so soft and very vulnerable to impacts and can be easily damaged in impacts and the skull's role is to protect it from damaging.

1.2.3 Membrane

Membrane envelops the brain and its job is to protect the brain. It has 3 different layers. The outermost layer is dura mater. Dura mater is impermeable to fluid, which helps to enclose CSF. The second layer is called pia mater, which is the innermost layer of the membranes. The third part is arachnoid mater, which is placed between the other two layers.

1.2.4 Cerebrospinal Fluid (CSF)

CSF is a colorless and clear fluid that contains proteins and is produced in the choroid plexus of the brain. CSF protects the brain from bruising into the skull when the head is hit although it cannot protect the brain in severe hits.

1.2.5 Brain

Brain is the center of the nervous system and it consists of cerebellum, cerebrum and brain stem. The largest part of the brain is cerebrum and it has two hemispheres and each of them is divided into four lobes the outmost layer of the cerebrum is grey matter and underneath it is the white matter. Cerebellum is underneath the cerebral hemispheres at the bottom of the brain. Brain stem is the part where the brain and the spinal cord meet.

1.2.6 Cerebral Vasculature

Cerebral vasculature is the movement of blood through a network of blood vessels. Major veins in the cranial part are inferior sagittal sinus, the superior sagittal sinus, cavernous sinuses, straight sinus, transverse sinuses and sigmoid sinuses.

1.3 Head and Brain Trauma in Car/Motorcycle Accidents

The most common cause of death in traffic accidents is head injuries and that happens when the head is subjected to loads higher than the loading capacities of the head. Head injuries can be divided into two categories, primary and secondary injuries. If the consequences of the physical loading of the head happen at the time of injury, they are called primary injuries, which may result in post-traumatic oedema, hypoxia, increased intracranial pressures and more. Secondary injuries are the result of the severity of primary injuries, which may appear minutes from injury up to days after injury and quick medical treatment can prevent them from happening.

The following are the different variations of primary head injuries that can happen in traffic accidents:

1.3.1 Skull Fracture

The skull is made of eight bones that form the cranial portion of it. If a blunt force trauma occurs that breaks one of these bones a skull fracture happens. If the surrounding skin ruptures the skull fracture is considered as an open skull fracture. The skull fracture can absorb the excess energy applied to the head and transfer less energy to the brain; however, a significant blow can damage the brain and cause other trauma.

1.3.2 Epidural Hematoma

In traumatic brain injuries (TBI) the damage may cause bone fractures, which can rupture blood vessels and make the internal parts bleed. If the blood builds up between the skull and dura mater the condition is called epidural hematoma (EDH). Since the blood built up between the skull and dura mater can increase the intracranial pressures, cause brain shift or compress brain tissue, EDH is potentially fatal. According to the literature about 15 to 20% of EDH are fatal. EDH happens in about 1 to 3% of head injuries.

1.3.3 Subdural Hematoma

Subdural hematoma or subdural hemorrhage is a type of hematoma happening inside the skull and is usually the cause of traumatic brain injury. If the blood accumulates between dura mater and arachnoid mater the condition is called subdural hematoma. There are three kinds of subdural hematoma according to the speed of their onset: acute, sub-acute and chronic. The most fatal of all head injuries are the acute subdural hematomas and if not quickly treated with surgical decompression they have very high mortality rates.

1.3.4 Cerebral Contusion

Contusion is a type of hematoma and if happens to the brain's tissue, it is called cerebral contusion and happens to 20 to 30% of severe brain injuries, which often involves hemorrhage, edema and infraction in the site of the impact.

1.3.5 Diffuse Axonal Injury

Diffuse Axonal Injury (DAI) happens when the brain inside the skull is accelerated specially in cases where the rotation of the head happens. And the cause of DAI is the traumatic shearing forces that happen when the head is under rapid accelerations. Almost all the patients with severe DAI go to comma and 90% of them will never gain consciousness again and between those who wake up, often will get disabled.

1.4 Preventing Head and Brain Trauma by using helmet

Helmet is a protecting device that shields human head from injuries and has many different applications such as military, vehicular, sports, constructions and etc. What helmet basically does is to absorb the mechanical energy that is applied to the outer layer as the energy is transferred towards the head and it is clear that the protective capacity of the helmet is directly related to the application of the helmet and the impact energy it has to support.

A motorcycle helmet is a full-face helmet worn by motorcyclist to protect the head from head injuries. A review in 2008 has shown that motorcycle helmets reduce head injuries by 69% and death by 42%. There are many types of motorcycle helmets available in the market. A full-face helmet covers the head entirely and is the safest type of helmet for motorcyclists. Although avoided by some people because of the excess heat created inside the helmet and the problems with ventilations. Half-helmets only cover the upper parts of the skull and face and the chin is still at risk of injury, although there are some variations that cover the eyes as well.

1.4.1 Helmet shell and liner

Helmets are usually consisted of two major parts the outer shell and the inner protective layer called the liner. The most important part of the helmet in absorbing the impact energy is the liner and its thickness is usually 3 to 4 cm, and is made up of a kind of foam.

Hopes and Chinn (1989) investigated the protection power of the helmet with regards to the stiffness of the liner and the shell. Their result showed that the stiffer the shell and liners were suitable for higher impacts and accelerations but in lower impacts they could not handle the impacts efficiently and even at some point they had negative effects on reducing the head injuries where the degree of the severity of the impacts were low. Although their study showed that the shell and the liner must be designed stiffly so that they can protect the head in situations where the head injury is unavoidable.

1.4.2 Fiberglass versus plastic shells

Vallée's study in 1984 showed that the risk of injury will dramatically rise of the outer shell fractures and that fiber-based materials where more durable against fracture. However, Noel in 1979 showed that on the factor of aging, plastic shells were more resistant against fiber-based materials. But from a safety point of view, fiberglass shells are preferred to plastic shells.

Organization of the Thesis

This thesis has 6 chapters, the appendix and references. Chapter 1 consists of the introduction of brain traumas, the biomechanical conditions under which brain trauma happens and how they can be prevented. In Chapter 2 literature survey is conducted and the development of human head models to simulate car crashes is presented. Moreover, the creation and development of devices to prevent brain traumas are presented as well. In this thesis only the skull and the brain will be included in the FE analysis.

In Chapter 3, a FE model of the human head will be developed and later on that chapter a FE model of the helmet will be developed as well. The results are illustrated in Chapter 4 and discussed in Chapter 5. Finally, in chapter 6 conclusion and suggestion for future work are presented.

Chapter 2

LITERATURE REVIEW

2.1 Head Injury Prediction

Many experimental tests have been performed since early 90s to predict the damages to the head in different circumstances like car crash, falling and etc. One of the most commonly used methods is Head Injury Criterion (HIC), an improved method of Wayne State Tolerance Curve (WSTC). WSTC was first introduced by Lissner in 1960, which described the link between the impact of the head caused by acceleration, period of the impact and the risk of head injury. Gadd (1966) introduced a new method of estimating injury hazard. He developed a Severity Index (SI) number to improve the impact tests by defining how severe is a head trauma.

The National Highway Traffic Safety Administration (NHTSA) used Gadd's work to introduce the Head Injury Criterion (HIC) in 1972, which is still being used in head protection systems using head forms. There are some limitations for the HIC according to Newman (1986), the direction of the impact is not specified in HIC and angular accelerations are not considered as well. So the Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) was introduced by Newman, and in the late 1990s Head Impact Power (HIP) was introduced by him as well.

Although rotational accelerations have a massive effect on brain injuries, HIC only considers linear accelerations and that's why Willinger (2008) developed an improved head injury criteria based on his previously created FE model.

2.2 Development of Head, Neck and Torso Models for Simulating Car/Motorcycle Crashes

The development of the FE method brought many advantages to biomechanics researchers. The FE method is a numerical technique for finding approximate solutions to partial differential equations. In the past two decades various researchers have introduced many FE models of the head. The first geometrically similar to the head FE model, was introduced by Hardy and Marcal (1971) and it was used for static simulations. Nickell & Marcerl (1974) developed a model containing only the skull.

In 1977, Shugar presented a new model that was used for head impact simulations. Until now, many different FE models have been introduced which differ in complexity and accuracy. Mendis, (1992) considered rotational acceleration of the head and developed a FE model to analyze stress and strain of such accelerations.

Nahum et al., (1977) conducted an experimental test using human cadavers for frontal blow to the head. The blow to the head was done using a cylindrical impactor with padding. The final results showed the epidural pressures in different locations of the head.

Ruan et al., (1993) presented a model that was geometrically based on Shugar's model consisting of the scalp, skull, brain, dura matter and the falx cerebri. The simulation setup of the analysis was based on cadaver tests of Nahum. He considered the brain as a viscoelastic material in his study. Ruan also did a similar study but with different material properties of skull, brain and cerebrospinal fluid (CSF) which affected the intracranial pressure levels and stress and strain levels of the skull.

Kumaresan et al., (1995) used the geometry of a dummy head (Hubbard, 1974) and created a FE head model containing the skull, CSF and brain and he approximated the mesh with his own software. The final model had 293 shell elements for the skull and 555 brick elements for the CSF and brain.

During the same year, Krabbel and Müller introduced a new way of creating FE models of head using visible human data extracted from CT and MRI scans. The inner and outer layers of the skull were created from CT scan data and brain model was extracted using MRI data. They scaled down the pixel sizes from 0.9 mm to about 0.5 mm to be able to use fewer elements for the model to make the analysis easier. The final model had 9,732 brick elements for the brain and the skull. The study resulted in creating FE models in a much easier and more accurate way.

In 1995 Zhou et al. created a new more detailed FE model, which was based on the model developed by Ruan but in addition it contained grey and white matter, the ventricles and the bridging veins and also finer meshing was used for the brain. All the parts were considered to be linear elastic. The impact in the simulation was the same as Nahum's experimental impact but the model was subjected to angular acceleration pulse as well.

Ruan et al., (1997) conducted a new study that again Nahum's impact experiment was used in the simulation and the comparison of the results with HIC values verified the accuracy of the results.

In the same year, Eindhoven University of Technology created a CT and MRI scan based model consisting of the brain, skull and facial bone. (Claessens et al., 1997). They used the projection method to generate the mesh for the brain and by expanding the meshes of the brain they meshed the outer layer of the skull as well. Later, the CSF, falx, dura mater and tentorium were added to the model (Brands et al., 2002) and the model was validated against Nahum's experiments. (Nahum et al., 1977)

Wayne State University researchers developed a model of the head in 1993 called the Wayne State University Brain Injury Model (WSUBIS). Later, they upgraded the model and release several versions of the model. The first version of the model (1993-1997) includes the essential parts of the human head. The total mass of the model is 4.3 kg. The model can predict diffuse axonal injury (DAI) in porcine brain because the grey and the white matter have different material properties. The second version of the model (1998-1999) was an improved version of the first model.

In the new model the brain can slide inside the skull, which improves the accuracy of brain trauma results in direct, and indirect head trauma caused by accelerations and deceleration of the head.

In 1997, Kang developed a new FE model of the human head, which is known as Université Louis Pasteur (ULP) model. In 2004, Deck et al. developed a new modified version of the ULP model including a very detailed geometry of the skull, considering different thickness throughout the skull. And for the first time the model included the reinforced beams in the skull, which had a significant role in modeling the skull fracture. McGuffie et al (1998), studied golf related head trauma on children. In their results they showed the effects of the velocity of the ball on the trauma caused to the head.

In 2002, Kleiven and Hardy developed a new head model consisting of the scalp, skull, brain, meninges, cerebrospinal fluid and eleven pairs of parasagittal bridging veins based of the visible human data. And a simplified neck model was introduced as well with the spinal cord, dura matter, spine, muscle and skin. The model has 18400 elements.

In 2003, Horgan and Gilchrist created a model (The University Collage Dublin Brain Trauma Model (UCDBTM)) based on CT scan data, which was used to model the pedestrian accidents. Using interpolations and threshold schemes smoothed the skull to create a FE model. To mesh the CSF and brain, the inner surface of the skull was used. An improved version of the model was introduced a year after consisting of the grey and white matter in addition to other parts. The model then validated against localized brain motion experiments (Hardy et al., 2001 and cadaver impact tests (Horgan and Gilchrist, 2004).

Using willinger's model (1999), Zhang et al (2004) presented the injury threshold for mild brain injuries. Zong, et al (2004), analyzed impact power flow on a 3D FE head model. Gong, et al (2004), modeled a solid impact to human head model and analyzed the response of the head to the impact.

In 2006, Kimpara et al. developed a 3D FE model with a highly detailed brain and spinal cord geometry. The model was used to simulate the entire central nervous system's mechanical behavior in impacts to the head simultaneously

In the same year, Paul and Corey (2006) used CT scans of a healthy female head and segmented the skull, brain and CSF. Each axial plane had a high resolution but the resolution between the planes was low. The details of the head model was not mentioned nor validated against other studies. In their study they developed their own version of the head model to use on their future studies.

2.3 Modeling and Analysis of Head Trauma During Traffic Accidents

Vallée et al (1984) showed in their study that the risk of injury increases if the helmet breaks and in helmeted head impacts fiber materials were more durable than plastic material against fractures. However in Noél's study (1979) which was about aging of materials that showed plastic materials were more durable than fiber materials so Vallée's results were true only in safety point of view.

Köstner (1987) and Van Schalwijk (1993) described the early helmet models but none of them were validated. Helmet models are usually modeled as linearly elastic although in some researches the helmet was considered as rigid.

Hopes and Chinn (1989) used different stiffness for the helmet shell and the liner in their study and the results showed that stiffer shell or liner was better for high accelerations.

Yettram (1994) developed a simplified FE model of the helmet. The helmet had only one layer and a wooden spherical head was used inside the helmet. The result showed that the density of the liner and the stiffness of the helmet had an extensive effect on the performance of the helmet and the lower the density, the higher the performance of the helmet.

Brands (1996) created a 3D FE model and used a dummy for the head and used elastic material for the shell and an elasto-plastic material for the liner, which was glued to the helmet. This model could not handle lateral impacts and it was used for frontal impacts only.

Liu (1998) simulated the head trauma using a FE model of helmet and again the model had only one layer. For the simulation process a biomechanical human head (the FE model of the Hybrid III dummy) including the neck was used. The results of that research were acceptable and were validated against other works.

Chang (1999) used a more detailed model for the helmet and included a rigid head model. The result was that higher stiffness of the helmet and a more dense liner was better for high-speed impacts and vice versa.

Chang (2000) conducted another simulation using chin bar and an energy absorbing liner which improved the safety of the helmet in facial impacts and showed that a less stiff helmet with the combined chin bar and energy absorbing liner could be much safer in crashes.

Willinger (2000) calculated the response of the head under normative impacts. The von mises stress and the pressure of the brain under the impacts was calculated and was validated against Nahum's cadaver head impacts.

Kotsiopoulos (2002) developed their helmet model using composite materials, which showed that composite shell helmets were absorbing more energy during impacts and were more reliable during crashes.

Aare and Halldin (2003) modeled a plastic shell helmet with elastic material and the liner was energy absorbing expanded polystyrene and was validated against oblique and radial impacts.

Pinnoji et al. (2007) combined Remy's head and brain model with a FE model of the helmet and calculated the force applied to the head during an impact with and without the helmet to determine the effects of the helmet in absorbing the force transferred to the head during frontal impact. The FE model used in the study had two layers and they monitored the results when the thickness and density of the liner were changed to find how it affects the impact forces and pressures absorbed by the helmet and transferred to the head.

Hamouda et al (2007) modeled the helmet and for the liner's material used expanded polypropylene and showed that using expanded polypropylene would increase the energy absorption of the liner.

Callahan (2011) analyzed a composite FE head model and for the head a Roma Plastilina #1 clay-filled rigid and stationary headform was used. The study showed the number of the layers of the helmet and the crated affected the results and predicted that the crater size in the clay have an important role in the head injury caused by the trauma to the helmet.

Through the literature survey, it is clear that there is still a need to work further on lateral impacts to the head and there is no need to mention that most of the previous projects were emphasizing the head only and the effects of the helmet were not analyzed. Very few FE helmet models are reported in the literature. The previous projects that were focused on helmet basically considered a rigid head form inside the helmet and only the effects of the helmet on reduction of the impact forces were analyzed, in other cases however, the helmet was simplified to a rigid body or a one part body.

Chapter 3

MATERIALS AND METHODS

3.1 Construction of 3D Models

In this thesis, Solidworks is used to create 3D models of brain, skull and helmet. All the parts are designed in Solidworks and the measurements are taken from available data. The methods, by which these parts are created, are explained as follows.

3.1.1 3D Model of the Helmet

To create the helmet, in a 2D sketch, a half circle with the diameter of 10 cm is created and with the Revolved Extrude feature a half sphere is created shown in Fig. 3.1.

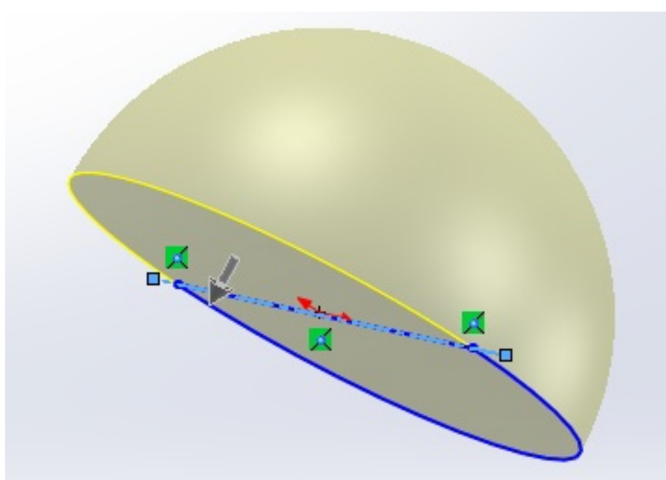


Figure 3.1: A Half Sphere Created as a Part of Helmet Model

In the next step, from the center of the sphere, a Boss Extrude with the same diameter of the half sphere is created at 15 degrees angle with the length of 5.99 cm (see Figure 3.2).

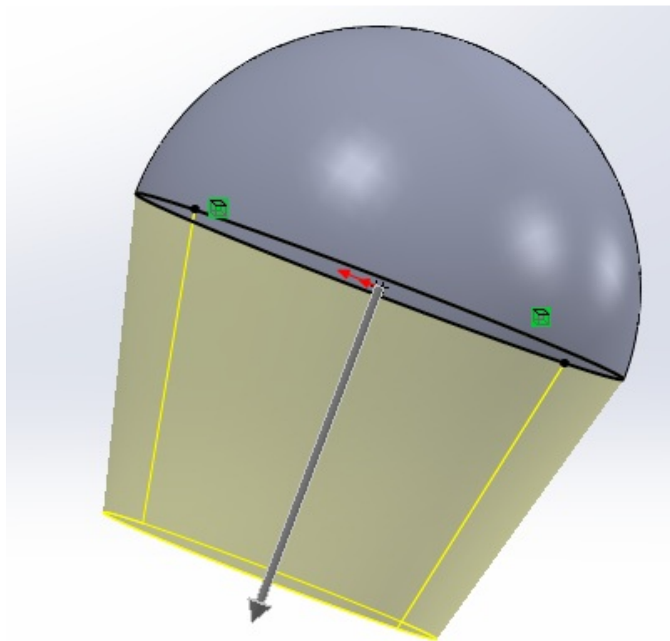


Figure 3.2: A boss extrude with 15 degrees angle is created from the center of the sphere

Once the second extrusion is completed, a fillet with a 4.5-cm diameter between the two extrusions is created and by using the Shell feature, the model is transformed from a solid body to a shell with the diameter of 2.2 cm, which is the thickness of the helmet according to the literature.

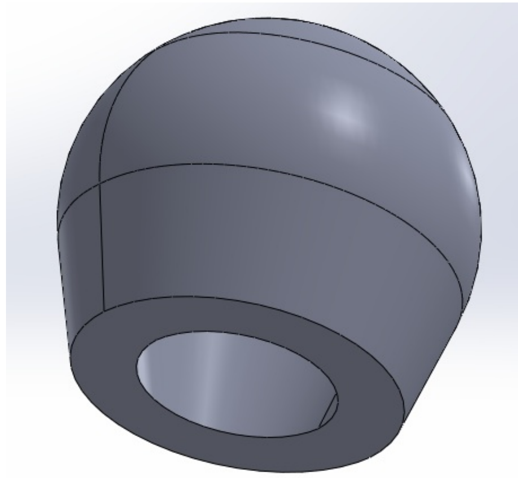


Figure 3.3: The helmet model is converted into a shell

In the final step, a surface cut is made in the center of the model to create the visor of the helmet. And in the edges of the visor a 1.27-cm radius fillets are created as well (see Fig. 3.4).

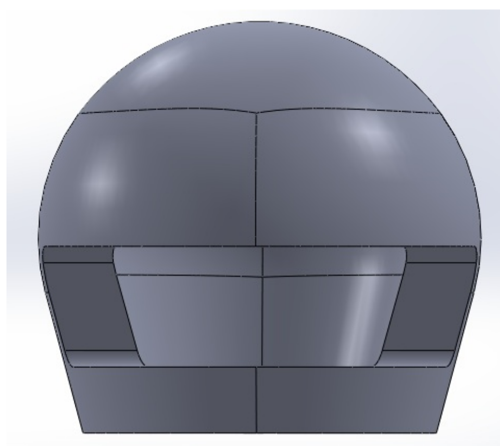


Figure 3.4: Final Geometry of the Helmet Model After a Surface Cut is Performed.

3.1.2 Brain and Head

For the scope of this thesis, accurate models of brain and head are not needed. So for the brain and the helmet, simple spheroids are used instead. Although for the skull, according to the literature thickness of 7 mm, which is the mean thickness of the skull, is used. Finally both brain and skull are assembled in Solidworks using the Mate feature. (See Fig. 3.5)

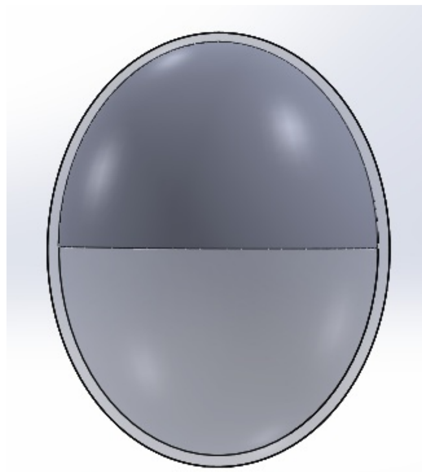


Figure 3.5: Assembly of the Skull and Brain Models

3.1.3 Final Assembly

For the assembly, all the parts are either scaled down or up in order to fit in the assembly. Figure 3.6 shows the final assembly created for the analysis. The impact object is situated very close to the helmet in order to reduce duration of the FE analysis.

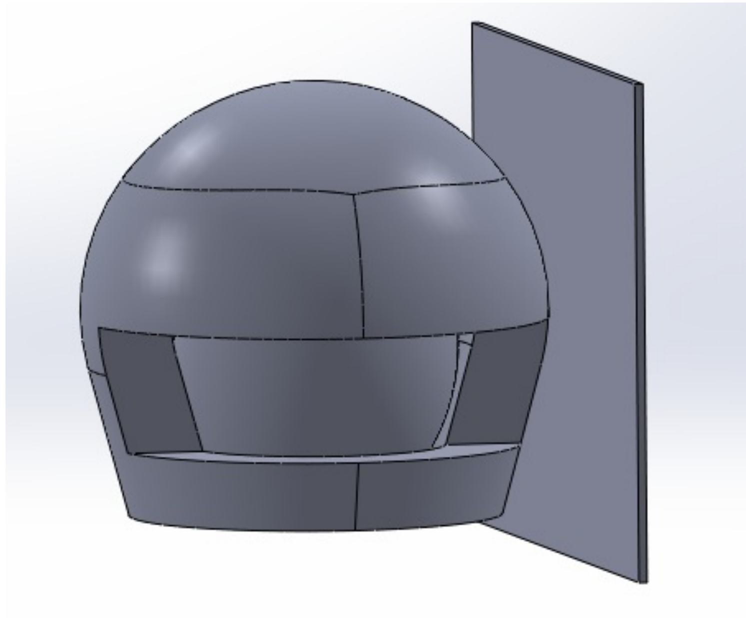


Figure 3.6: Final Assembly of the Protected Head Model.

3.2 Developing a Finite Element Model of Head and Brain

3.2.1 Meshing

Meshing the model correctly is one of the most important parts in FE analysis. A bad mesh can affect the final results dramatically and on the other hand, it can increase the processing time, CPU usage and memory usage. Reducing the number of the elements in the mesh is a very good way to reduce the processing time of the analysis and there are many ways to do that, either by increasing the element size or by fixing the 3D CAD model, such as omitting the fillets or refining the sharp edges. In the meshing, there are two important factors that if chosen correctly can make the analysis much easier:

3.2.1.1 Element types

Choosing a correct element type is crucial to the accuracy of the final results in FE analysis. According to a couple of studies that were focused on how choosing element type woud affect the final results, usually use of the linear tetrahedral elements is avoided in biomechanical analysis because they can affect the results for nearly incompressible materials and the reason to that is that it causes volumetric locking and a stiffed bending responses. (Bonet et al., 2000)

Some studies have shown that linear hexahedral elements are much more accurate and have a much better performance in CPU time and accuracy than even quadratic tetrahedral elements. (Benzley *et al*. 1995)

Brain includes high amounts of water in itself and because of that is considered to be nearly incompressible (Poisson's ratio, $\upsilon \approx 0.5$) so using hexahedral elements for the brain is much more efficient in the analysis of brain injuries.

For the skull, using shell elements has better accuracy and lowers CPU time and studies have shown that using tetrahedral shell elements for the skull is more efficient in impact analysis.

Triangle meshes are avoided wherever possible and almost all the parts have quad type meshes.

3.2.1.2 Element quality

Errors in FE simulations are directly related to element quality. Large deformations in element will reduce its quality. The accuracy of the FE problems will reduce if the meshing of the part has large amounts of deformed elements.

Evaluating the quality of the element from its shape and dimension is possible using a variety of methods.

Jacobian value is a method to calculate the difference between an ideal element and the deviated element. The range of the Jacobian value is between 0 and 1 and usually values above 0.5 are considered as an acceptable deviation for the element. Another method for measuring the element's quality is measuring its aspect ratio. The ratio of the longest side to the shortest side of the elements must be less than 4:1 and in areas with quick changes in stress the ratio should be much smaller. Measuring the angle between the lines that join opposite sides is another method of determining the quality of the element. The aforementioned angle is called skew angle.

In the current thesis the meshes satisfy all the above methods. Figure 3.7 shows the meshed model in ANSYS. The meshes for the skull, brain and helmet are converted to hexahedrons by using the Hex Dominant Method.

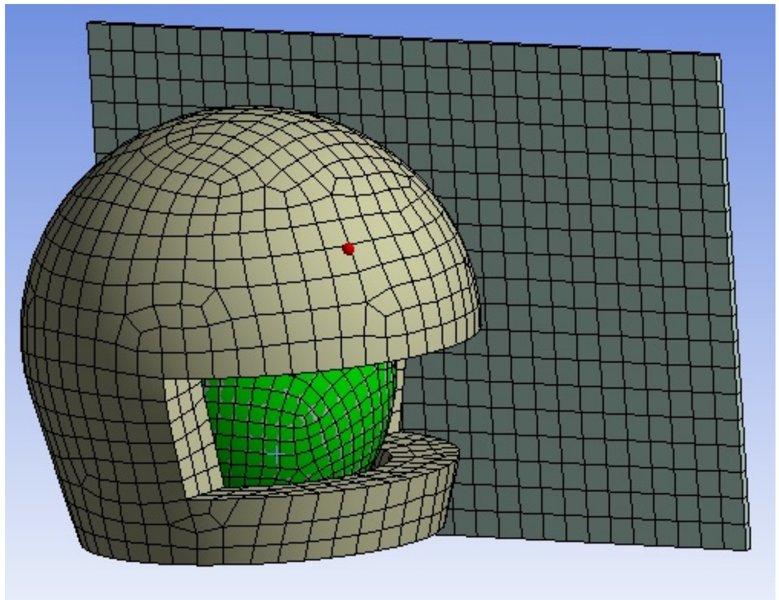


Figure 3.7: Hexahedral Elements for Meshed Model.

The model is meshed in ANSYS, and brain, skull and helmet have 493, 1367 and 2502 nodes and 2008, 4085 and 10404 elements respectively.

3.2.3 Material properties

Biological tissues are considered to be viscoelastic, nonlinear, anisotropic and inhomogeneous, but generally for analysis they are assumed to be linearly elastic. In this thesis, Material properties assigned to the different parts are all isotropic, homogenous and elastic.

3.2.3.1 Brain

With very large amount of water in the brain, it is assumed to be nearly incompressible with a poisson's ratio very close to 0.5. According to the literature the density of the brain is about 1.04 kg/dm3. The bulk modulus of the brain is close to water's bulk modulus and about 2.1 GPa (McElhaney et al., 1976; Stalnaker, 1969). Although in the brain, grey matter and white matter have different material properties but in the scope of this thesis they are neglected.

Although older studies considered the brain as a viscoelastic material in newer studies Mooney-Rivlin hyper elastic rubber like material model is one of the models that are used to describe brain's behavior (Hallquist, 2006). Marvern, 1969 has shown that the stored strain energy for a hyper elastic material and in our case the brain is a function of the principal invariants (I_1, I_2, I_3) of the right Cauchy-Green deformation tensor. That is W=W(I_1, I_2, I_3). Although this formula is implemented in ANSYS and only the constants in Table 3.1 are needed in order to use this formula.

$$W(J_1,J_2,J) = C_{10}(J_1 - 3) + C_{01}(J_2 - 3) + \frac{1}{2}K(J - 1)^2 \qquad (3.1)$$

$$J_1 = I_1 I_3^{-1/3} \qquad (3.2)$$

$$J_2 = I_2 I_3^{2/3} \qquad (3.3)$$

I_1, I_2, I_3 are the strain invariants of the right Cauchy-Green deformation tensor.

J is the Relative volume.

C_{10}, C_{01} are the Moony-Rivlin constants.

K is the Bulk modulus

W is strain energy density function and by inputting the constants to ANSYS equation 3.1 will be used to analyze the brain.

Since we cannot directly input the Poisson's ratio into ANSYS, the following formula is used to derive the incompressibility parameter from Poisson's ratio:

$$K = \frac{E}{3(1-2\nu)} \tag{3.4}$$

$$d = 2/K \tag{3.5}$$

So we have:

$$d = \frac{6(1-2\nu)}{E} \tag{3.6}$$

Where:

K is the Bulk modulus

E is the Young's modulus

v is the Poisson's ratio

In the analysis, the bulk modulus of the brain is considered to be 2.19 GPa. Poisson's ration of the brain is assumed to be 0.4996, which resembles a material nearly incompressible. Using the equation (3.4), the young's modulus of the brain is calculated 5.26 MPa and by putting the result in equation (3.6), we will find the incompressibility factor to be 9.125E-10.

Table 3.1 summarizes inputs that are used in ANSYS

Table 3.1: Summary of material properties of the brain model used in the FE analysis.

Tissue	Material properties	Density Kg/m^3	Poisson's ratio
Brain	Mooney-Rivlin rubber C_{10}=124 C01=138 d=9.125E-10	1040	0.4996

3.2.3.2 Skull

Generally, skull is assumed to be linearly elastic. Density of the skull is about 1.3 Kg/dm^3 and the Young modulus of the skull is about 15 GPa. Poisson's ratio of the skull is about 0.22 (McElhaney *et al*., 1970). Skull has different thicknesses in different parts and to make the analysis easier the mean thickness of the skull is assumed to be 7 mm. For the analysis, bulk modulus of the skull is needed, which is calculated from (3.4) equation and is 8.93 GPa. The total weight of a human head differs in different ages and genders and in adults it varies between 4 to 6 kg. The head model, which is used in this thesis, is weighed 5.4 kg.

3.2.4 Contacts between brain and skull

In the literature review (Sec. 2.2), 3 types of contacts are defined between the skull and the brain. The first method is to bond the skull and the brain together, so that there is no relative movement between the two.

The other method is to consider frictionless movement between brain and skull and the third method is to consider a frictional movement between the parts and considering that the brain is drowned in the CSF inside the skull this method is more realistic than the others.

In this thesis, the brain has a frictionless movement inside the skull and the reason to choose this method is that CSF is outside the scope of this thesis.

Aside from contacts, the interactions between all the bodies are considered to be frictionless as well.

3.3 Finite Element Model of Head and Brain with Helmet

In the second part of the analysis, effects of the helmet on reducing the stresses reached to the brain during the impact to the head will be analyzed. In this thesis, the helmet consists of only one part, the inner liner. According to the literature the most important part of the helmet in absorbing the impact forces is the liner. The helmet used in this study is a full-face helmet, covering all the parts of the head. The liner is made of expanded polystyrene foam. The thickness of the helmet is 40 mm.

The helmet model is meshed with quadrilateral elements, which brings acceptable results and reduces the processing time needed for the analysis. The total mass of the helmet is 0.3 Kg and its low weight is because the outer shell is not considered in this thesis. The head model used for this part of the thesis is the model introduced in section 3.1. Material properties of the helmet are shown in Table 3.2 and physical properties of the FE model are given in Table 3.3.

Table 3.2: Summary of material properties of the helmet used in the analysis

Material	Model	Young modulus [Gpa]	Poisson's ratio	Density [Kg/m^3]	Comment
Expanded polystyrene	Elasto-plastic	1.5e-3	0.05	0.25	Thickness 40 mm Yield stress = 0.35 MPa

Table 3.3: Summary of physical properties of all the models.

Object Name	*Impact Object*	*Brain*	*Skull*	*Helmet*
State	Meshed			
Definition				
Stiffness Behavior	Rigid	Flexible		
Properties				
Volume	1.2233e-003 m³	2.4658e-003 m³	1.2683e-003 m³	1.1785e-002 m³
Mass	3.0582e-002 kg	3.5902 kg	1.8467 kg	0.29462 kg
Centroid X	0.17963 m	5.4676e-003 m	5.9034e-003 m	5.0068e-003 m
Centroid Y	4.493e-002 m	6.789e-002 m	1.9108e-002 m	5.3139e-002 m
Centroid Z	1.495e-002 m	-1.6557e-002 m	-1.6297e-002 m	-3.1897e-002 m
Statistics				
Nodes	1160	493	1367	2502
Elements	532	2008	4085	10404

3.4 Impact conditions and Finite Element analysis of the model

A very important factor affecting the intensity of injuries during the impact is weather the head is able to move freely after it is hit or not. If the head cannot move freely after the impact the injuries are directly related to the location where the head hits an object and how the skull deforms. However in most cases the head impacts a stationary object and because of that the injuries are mostly due to the acceleration of the head. To simulate the impact of the head, a rigid object has been used to hit the head laterally. The velocity of the impact is assumed to be 4 m/s as specified by the standard. The duration of the analysis is 0.1 sec.

Coefficient of Restitution (COR) for skull and brain are considered to be 1. The reason behind that is because in this thesis skull and brain are in the elastic zone so the collision of the objects is considered to be elastically. However, for the helmet, COR is considered to be 0.45.

The impact analysis of the head is performed twice. In the first simulation the head is not protected and it laterally hits the impact object. However, in the second simulation the head is protected with the helmet model and the same impact conditions are applied to the model. Normal and Von Mises stresses for both simulations are calculated and in the end compared with each other in order to find out about the effects of the helmet in reducing such stresses.

Chapter 4

RESULTS

The most common injury in car and motorcycle accidents is head injury (Otte et. al., 1999). Although in many countries using helmets on motorcycles are obligatory there is still a need to improve the helmet design to prevent head injury. In this thesis a FE analysis is conducted to simulate the lateral impact to head that may occur in car accidents then normal and equivalent stresses that are applied to the brain during such accidents have been analyzed. In this analysis the head is subjected to the impact twice. In the first analysis the head is not protected and the impact directly hits the skull at the speed of 4.5 m/s.

In the second analysis the head is protected with a helmet and again it is subjected to the same impact. These two analyses are performed separately and in Chapter 5 the results are compared to find out the effect of the helmet in absorbing external impacts to the head.

The results in the present chapter represent the stresses that are created during the impact of the head and the helmet to the impact object. Normal Stresses are shown in Section 4.1, which is divided into 3 subsections: Normal stresses on Helmet, skull (Protected with helmet and unprotected) and Brain (protected with helmet and unprotected) are given in Subsections 4.1.1, 4.1.2 and 4.1.3 respectively. Afterwards in Section 4.2 Von Mises stresses are calculated and given which is again divided into 3 Subsections as in 4.1.

4.1 Normal Stress Analysis

Normal stresses occur as a result of vertical internal and external forces. In this section normal stress distributions on different parts of the model under the lateral impact are illustrated. In all the figures green lines are Maximum stresses and red lines are minimum stresses. Minimum stresses happen due to internal forces rather than maximum stresses, which are the result of external forces. In the case of skull minimum stresses and in the case of brain maximum and minimum stresses are due to intracranial forces.

4.1.1 Helmet

Stresses on helmet during the impact of the impact object to the head at t=0.1 which is the end of the analysis are shown in Figure 4.1.

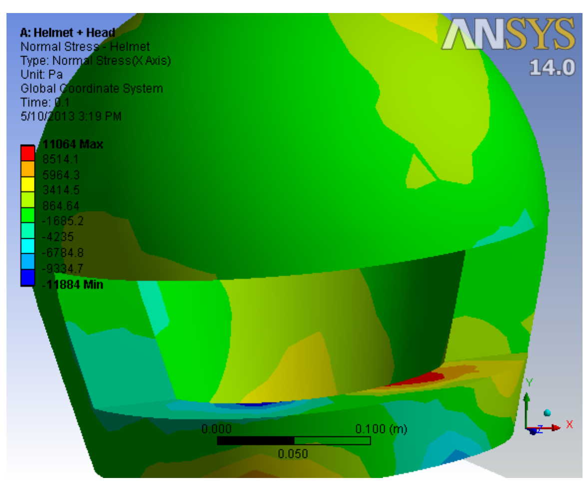


Figure 4.1: Normal Stress Distribution on the Helmet in the End of Simulation

Maximum and minimum stresses for the duration of the impact are shown in Table 4.1 and are illustrated in Figure 4.2.

Table 4.1: Maximum and Minimum normal stresses on helmet due to impact.

Time (s)	Minimum (Pa)	Maximum (Pa)
1.1755e-038	0.	0.
5.0009e-003	-29134	1620.2
1.e-002	-23835	2696.3
1.5e-002	-17385	8589.6
2.5e-002	-60138	4491.1
3.e-002	-60490	14693
3.5001e-002	-64673	9861.8
4.5e-002	-31637	4907.
5.0001e-002	-10348	4658.5
5.5e-002	-34763	10859
6.e-002	-66679	20529
6.5001e-002	-41939	18118
7.e-002	-42148	14733
7.5e-002	-36995	14017
8.0001e-002	-10853	12066
8.5001e-002	-10941	9911.6
9.e-002	-21676	11435
9.5001e-002	-10805	10786
0.1	-11884	11064

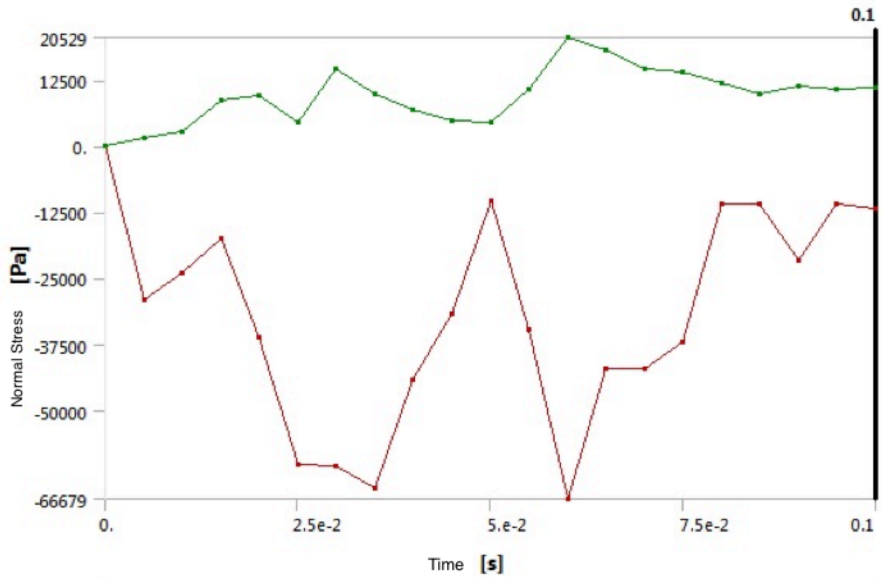


Figure 4.2: Visual Illustration of Table 4.1

4.1.2 Skull

4.1.2.1 Head protected with helmet

Normal stresses on skull during the impact when the head is protected with the helmet in the end of simulation (t=0.1) are shown in Figure 4.3.

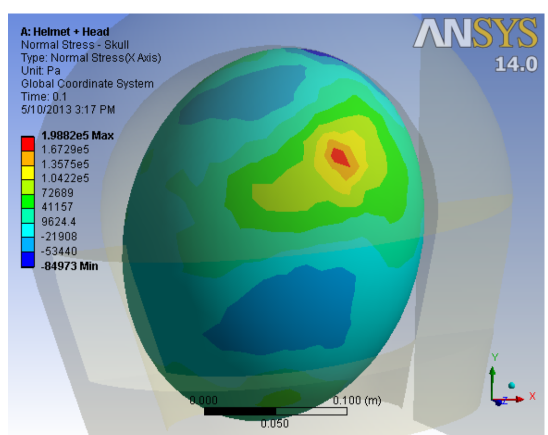


Figure 4.3: Normal Stress Distributions on Skull When Head is Protected at t=0.1 Sec in the End of Simulation.

Maximum and minimum normal stresses on the skull are given in Table 4.2 and are illustrated in Fig. 4.4.

Table 4.2: Maximum and Minimum normal stresses on protected skull.

Time [s]	Minimum [Pa]	Maximum [Pa]
1.1755e-038	0.	0.
5.0009e-003		
1.e-002		
1.5e-002	-1.1539e+005	1.7681e+005
2.0001e-002	-51624	33854
2.5e-002	-78161	1.5325e+005
3.e-002	-1.3959e+005	1.1142e+005
4.e-002	-1.0221e+005	47430
4.5e-002	-1.2137e+005	80595
5.0001e-002	-49656	52805
5.5e-002	-67289	81243
6.e-002	-84051	89508
6.5001e-002	-1.344e+005	1.0954e+005
7.e-002	-1.1095e+005	1.2741e+005
7.5e-002	-54918	1.0417e+005
8.0001e-002	-1.1229e+005	1.4438e+005
8.5001e-002	-1.6724e+005	1.8522e+005
9.e-002	-1.5342e+005	1.4723e+005
9.5001e-002	-1.2327e+005	1.2068e+005
0.1	-84973	1.9882e+005

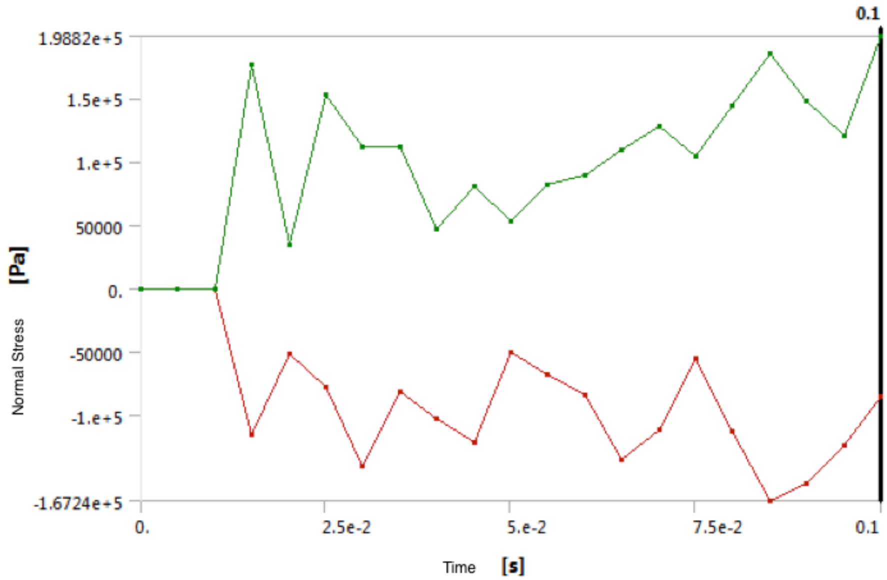


Figure 4.4: Visual illustration of Table 4.2

4.1.2.2 Unprotected head

Stresses in skull when the head directly hits the impact object without any protections in the end of simulation (t=0.1) are illustrated in Figure 4.5

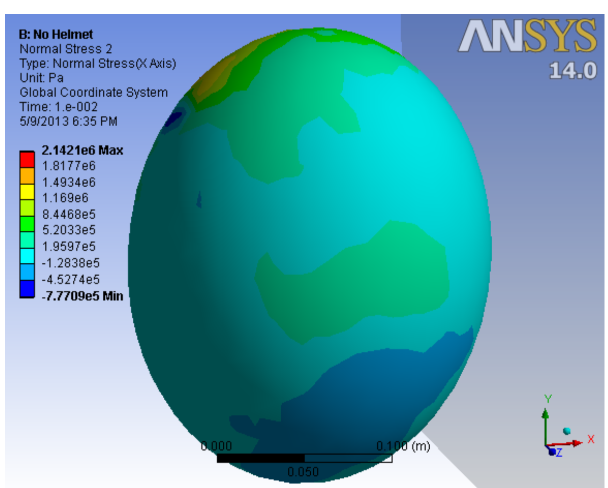


Figure 4.5: Normal Stress Distribution on Skull When Head is Unprotected at t=0.1 Seconds in the End of the Simulation.

Maximum and minimum stresses on the skull are given in Table 4.3 for the duration of the impact and are illustrated in Fig. 4.6.

Table 4.3: Maximum and Minimum normal stresses on unprotected skull.

Time [s]	Minimum [Pa]	Maximum [Pa]
1.1755e-038	0.	0.
5.0031e-004		
1.0001e-003	-3.2453e+006	3.6802e+006
1.5003e-003	-7.173e+006	6.7372e+006
2.0001e-003	-2.4408e+006	7.4586e+006
2.5003e-003	-1.9674e+006	8.0699e+006
3.5e-003	-1.8227e+006	6.306e+006
4.0003e-003	-1.5552e+006	5.902e+006
4.5001e-003	-1.0347e+006	3.6825e+006
5.0003e-003	-1.2074e+006	3.5357e+006
5.5002e-003	-8.7079e+005	2.9168e+006
6.e-003	-7.5275e+005	1.8943e+006
6.5003e-003	-1.343e+006	1.5483e+006
7.0001e-003	-9.7209e+005	1.8506e+006
7.5003e-003	-1.6092e+006	3.3211e+006
8.0001e-003	-1.2411e+006	4.0006e+006
8.5003e-003	-1.4162e+006	3.9278e+006
9.0002e-003	-1.038e+006	3.775e+006
9.5e-003	-1.1279e+006	3.105e+006
1.e-002	-7.7709e+005	2.1421e+006

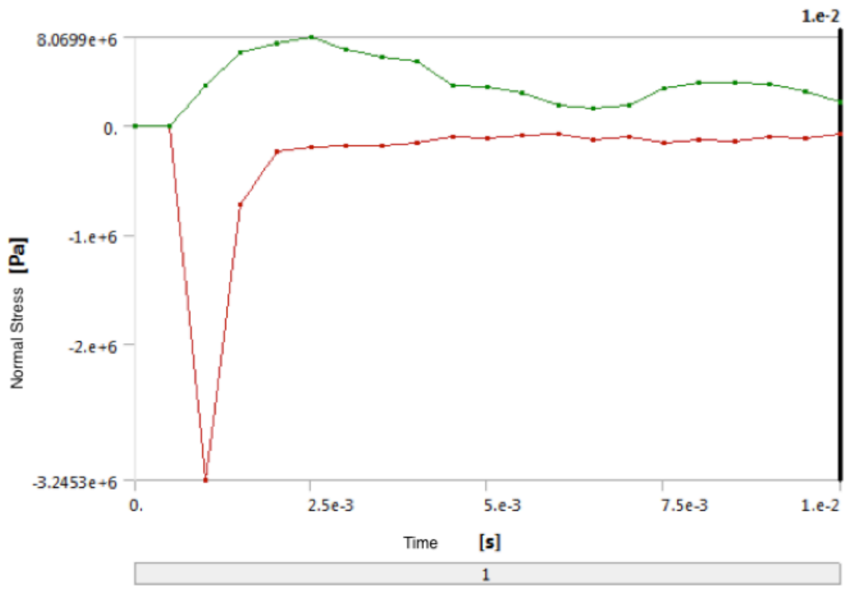


Figure 4.6: Visual Illustration of Table 4.3

4.1.3 Brain

4.1.3.1 Head protected with helmet

Stress distribution on brain in the end of simulation (t=0.1) when head is protected with a helmet and hits the impact object are shown in Figure 4.7.

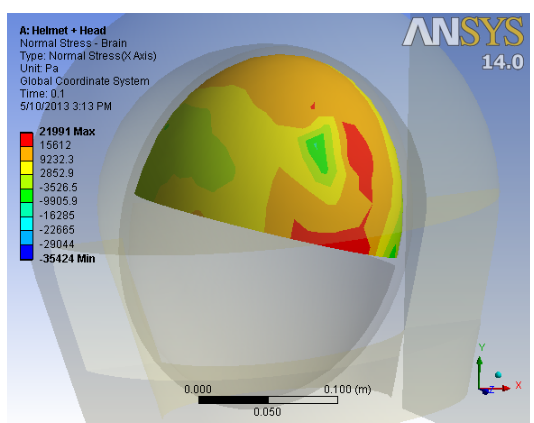


Figure 4.7: Normal Stress Distributions on Brain When the Head is Protected at t=0.1 Seconds.

Maximum and minimum normal stresses are given in Table 4.4 and are graphed in Fig. 4.8.

Table 4.4: Maximum and Minimum Normal Stresses on Protected Brain.

Time [s]	Minimum [Pa]	Maximum [Pa]
1.1755e-038	0.	0.
5.0009e-003		
1.e-002		
1.5e-002	-87815	3276.4
2.0001e-002	-6875.1	3572.2
2.5e-002	-63388	6910.7
3.e-002	-50352	5745.4
3.5001e-002	-22422	14584
4.e-002	-34967	9167.1
4.5e-002	-29911	20096
5.5e-002	-23865	17002
6.e-002	-50338	27652
6.5001e-002	-57520	67737
7.e-002	-41018	17844
7.5e-002	-76205	10132
8.0001e-002	-32773	18551
8.5001e-002	-74761	12230
9.e-002	-93927	25857
9.5001e-002	-46124	19427
0.1	-35424	21991

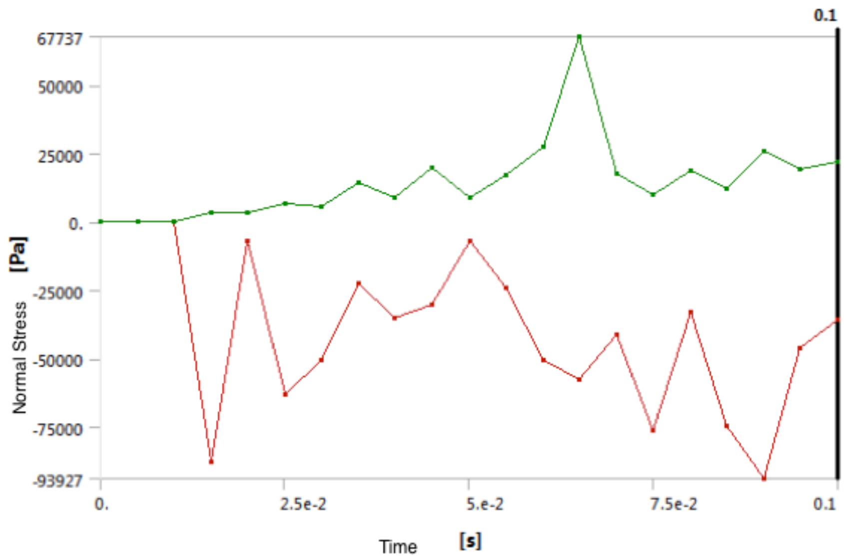


Figure 4.8: Visual Illustration of Table 4.4

4.1.3.2 Unprotected head

Stress distributions on brain when head is unprotected and it directly hits the impact object in the end of simulation (t=0.1) are shown in Figure 4.9.

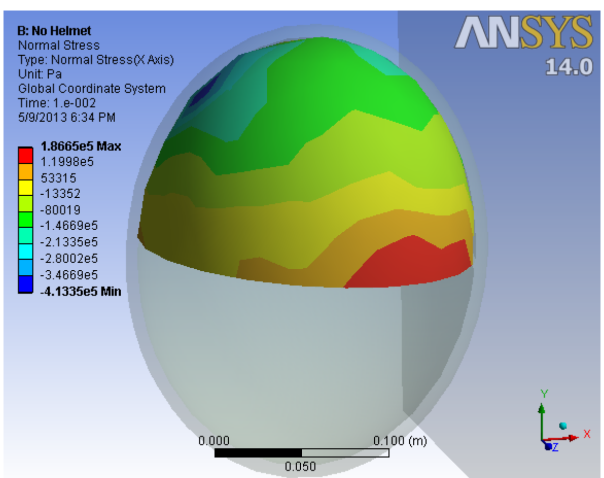


Figure 4.9: Normal Stress Distributions on Brain When Head is Unprotected at T=0.1 Sec.

Maximum and minimum stresses are given in Table 4.5 and are graphed in Fig. 4.10.

Table 4.5: Maximum and Minimum Normal Stresses on Unprotected Brain.

Time [s]	Minimum [Pa]	Maximum [Pa]
1.1755e-038	0.	0.
5.0031e-004		
1.0001e-003	-2.8358e+006	21124
1.5003e-003	-4.4991e+006	1.5009e+005
2.0001e-003	-4.6591e+006	1.242e+005
2.5003e-003	-4.9683e+006	1.6106e+005
3.0002e-003	-4.4624e+006	1.3493e+005
4.0003e-003	-3.5792e+006	83043
4.5001e-003	-5.6506e+005	5.0375e+005
5.0003e-003	-5.5702e+005	53281
5.5002e-003	-4.8186e+005	2.4012e+005
6.e-003	-4.2315e+005	1.3993e+005
6.5003e-003	-5.0887e+005	2.4279e+005
7.0001e-003	-4.8586e+005	3.6007e+005
7.5003e-003	-6.533e+005	1.1338e+005
8.0001e-003	-1.0526e+006	2.581e+005
8.5003e-003	-8.1239e+005	1.0483e+005
9.0002e-003	-6.1444e+005	2.36e+005
9.5e-003	-5.1395e+005	54944
1.e-002	-4.1335e+005	1.8665e+005

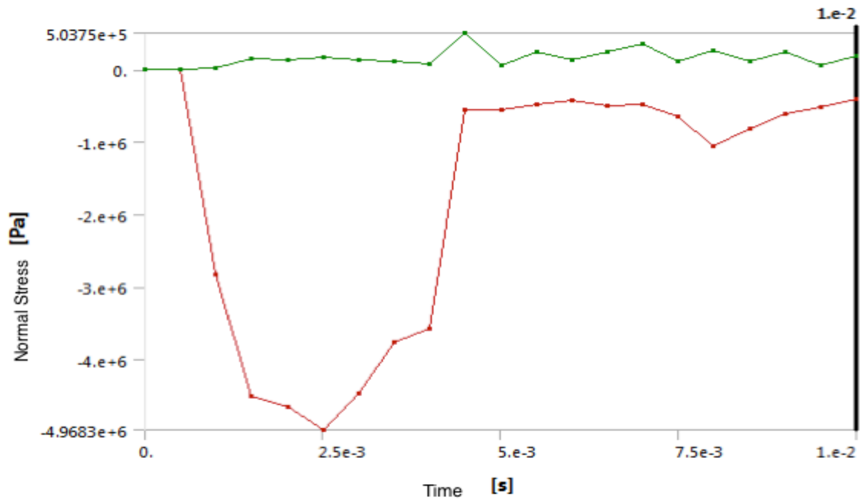


Figure 4.10: Visual Illustration of Table 4.5

4.2 Von Mises Equivalent Stress

In this section Von Mises equivalent stress distributions, which according to Willinger, R, 2000, are considered as good indicators of concussion, are illustrated for the duration of the impact.

4.2.1 Helmet

Von Mises stresses on the helmet during the impact are depicted in Figure 4.11.

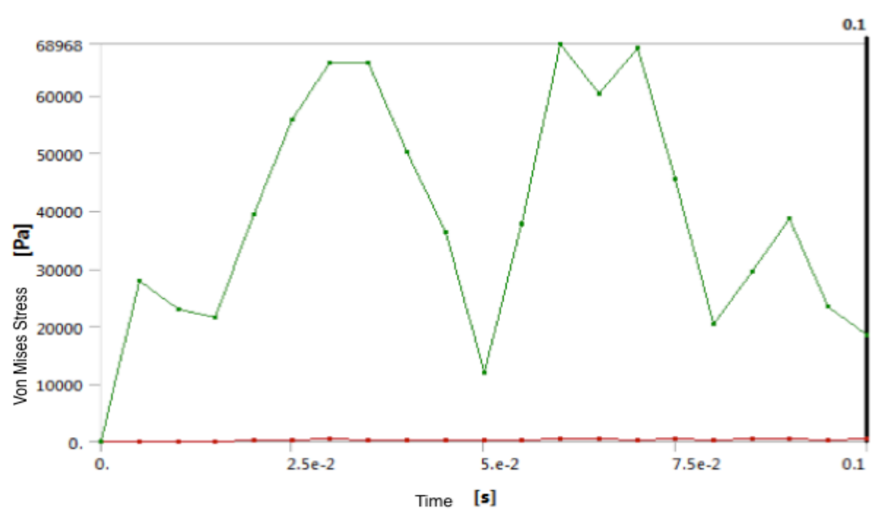


Figure 4.11: Visual Illustrations of Von Mises Stresses on the Helmet Model

4.2.2 Skull

4.2.2.1 Head is protected with helmet

Von Mises stress distributions on skull during the impact when head is protected with helmet are illustrated in Figure 4.12.

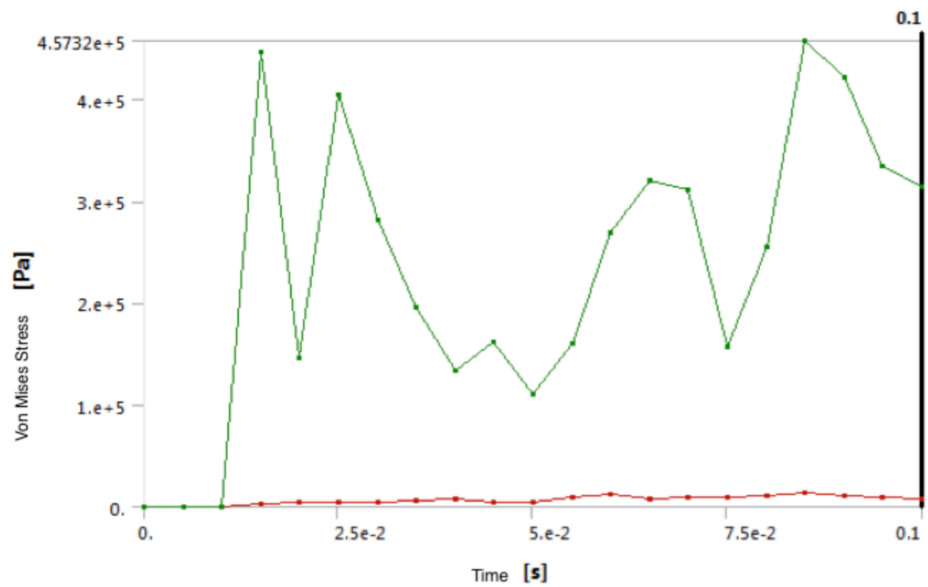


Figure 4.12: Visual Illustrations of Von Mises Stresses on Protected Skull

4.2.2.2 Unprotected head

Von Mises stress distributions on skull during the impact when the head is unprotected and it directly hit the impact object are illustrated in Figure 4.13

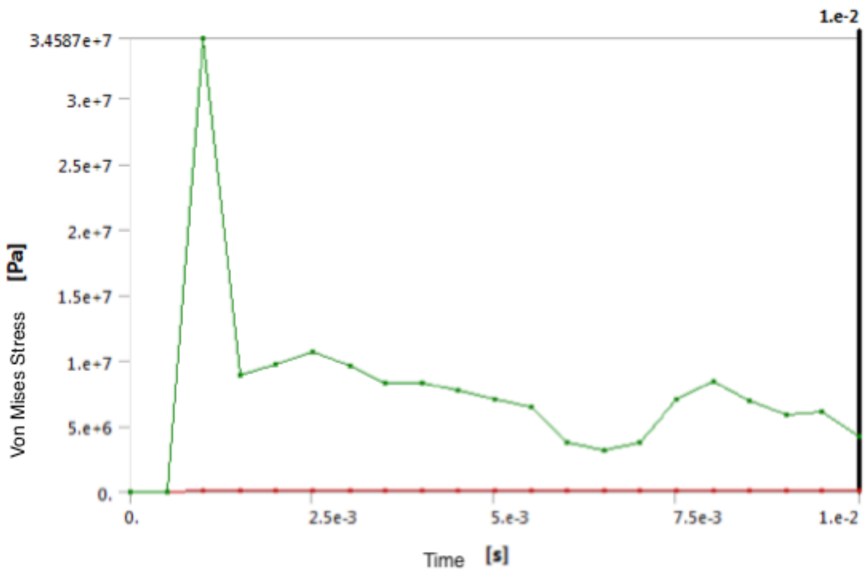


Figure 4.13: Visual Illustration of Von Mises Stresses on Unprotected Skull

4.2.3 Brain

4.2.3.1 Head is protected with helmet

Maximum and minimum Von Mises stress distribution on the brain when head is protected with helmet for the duration of the impact are illustrated in Figure 4.14.

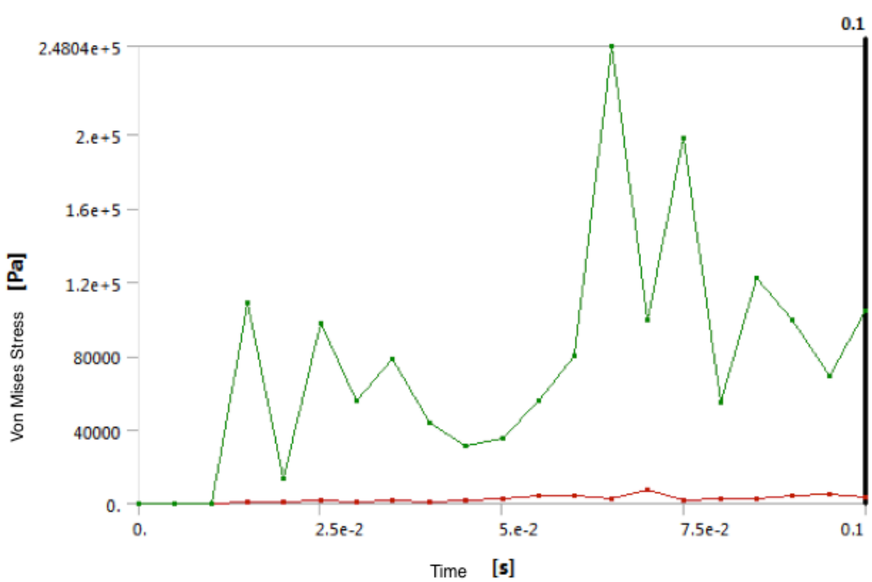


Figure 4.14: Visual Illustration of Von Mises Stresses on Protected Brain.

4.2.3.2 Unprotected head

Von Mises stress distribution on brain when the head is unprotected and it directly hits the impact objects are shown in Figure 4.15.

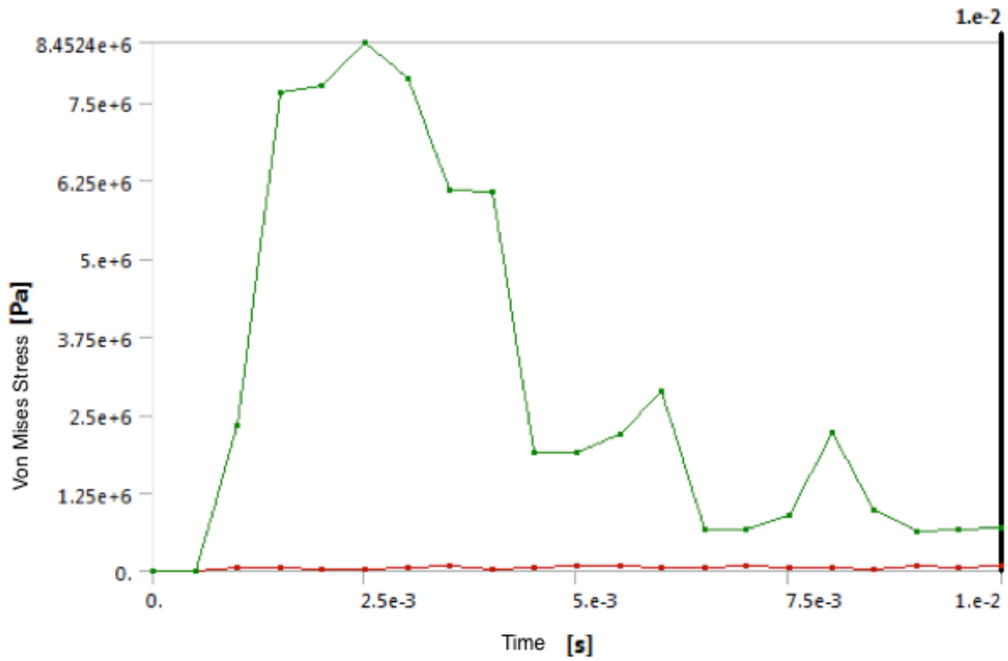
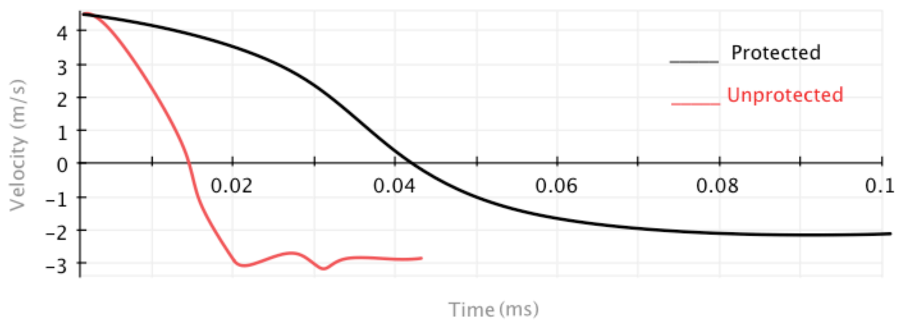


Figure 4.15: Visual Illustration of Von Mises Stresses on rain When Head is Unprotected.

4.3 Acceleration on Center of mass of the head

Differences between changes in velocity of mass center of the head during the two impacts are shown in Figure 4.16



Acceleration of protected head 180 g

Acceleration of Unprotected head 1064 g

Figure 4.16: Change in Velocity of Center of Mass of the Head and the Accelerations Caused by That.

These results are compared in Chapter 5 in order to understand how helmet helps to protect the head from lateral impacts.

Chapter 5

DISCUSSION

In this chapter the results that have been obtained from analyses are discussed. The results of this study imply than helmet is very effective in reducing effects of external impact forces during lateral head trauma, which have a direct effect on stress distributions on the head.

In this study several limitations are included in order to reduce the processing time and to prevent using extra variations. These limitations include:

- The at which the head model hits the impact object was kept low at 4.5 m/s to eliminate the unwanted oscillations of brain inside the skull and in order to omit extra factors and achieve more accurate results.
- The models of the skull and brain were simplified to reduce the meshes and increase the quality of the meshes that were created.
- Helmet straps are neglected although it is very unlikely that straps have considerable effects on the results of this study.
- The helmet model only consists of the liner as other studies have shown liner is the most effective part of the helmet during impacts.

- The skin was excluded from this study as it has a minimal effect in reducing the impacts because of its low thickness around the skull.
- The interactions between brain and skull were considered frictionless to reduce the complexity of the analysis.

5.1 Discussion of the Results

Simply by comparing the results in two cases (protected and unprotected), it is apparent that helmet will effectively reduce the impact forces, which results in reduction of stress in the brain (Figures 4.8, 4.10) and the skull (Figures 4.4, 4.6). According to the results, the maximum normal stress on brain in an unprotected head is about 4.97 MPa which is as a result of intracranial pressures, comparing to result of the protected case, which is 93.9 KPa. So there was a 98% reduction in maximum stress on the brain when the helmet was used. Moreover, in the skull maximum stresses on the unprotected case and protected case were about 8 MPa and 199 KPa respectively, which has reduced maximum normal stress on the head by 97.5%.

Although the skull was not damaged in neither of the cases but the stresses on the brain in the unprotected case could be lethal according to the literature where the critical stresses on the brain were considered as 100 KPa. So the protected head is in the safe zone and it is highly unlikely that the impact would cause brain trauma.

We can see that minimum stresses, which are caused by internal forces that are created inside the head have higher absolute magnitudes so in normal stress they are used instead of maximum stresses which are as a result of external forces.

On the other hand, the helmet did not fail during the impact because according to the literature the failure of the helmet will probably have negative effects in protecting the head from trauma.

In this study Von Mises stresses were calculated as well. The reason behind that is according to the definition of Von Mises Criterion although stresses on the X, Y and Z direction may not cause failure in an object but the combination of these stresses may cause failure. So Von Mises stresses were also calculated to assure that combination of the stresses would not cause trauma in the protected head. According to the results maximum Von Mises stresses on the brain in unprotected and protected cases were 8.45 MPa and 248 KPa respectively which shows a 97% reduction in the maximum Von Mises stress on the brain. For the skull, the results were 34.6 MPa and 457 KPa respectively, which reduced the maximum Von Mises stress on the skull by 98%.

It should be mentioned that since absolute values of Minimum Von Mises stresses were considerably lower, maximum Von Mises stresses were used instead. Table 5.1 and 5.2 summarize these comparisons.

Table 5.1: Comparison of Maximum normal stresses in two cases.

Parts of the head	**Maximum normal stress on protected head with helmet (KPa)**	**Maximum normal stress on unprotected head (KPa)**	**Percentage of stress reduction (%)**
Brain	93.9	4970	98.1
Skull	199	8000	97.5

Table 5.2: Comparison of Maximum Von Mises stresses in two cases.

Parts of the head	**Von Mises Stress on protected head (MPa)**	**Von Mises Stress on unprotected head (MPa)**	**Percentage of stress reduction**
Brain	0.248	8.45	97
Skull	0.457	34.6	98

By comparing the all the figures related to normal and Von Mises stresses for unprotected and protected cases, it is clear that in protected cases although the maximum levels of normal and Von Mises stresses were lower but there were more fluctuations compared to unprotected cases. The reason behind this is because of the elasticity of the helmet liner, which has caused these fluctuations, and by considering the outer shell for the helmet these fluctuations would be either prevented or reduced dramatically.

Another important factor in two cases is the change in velocity of the mass center of the head after the impact (Section 4.3). According to figure 4.16 the total change in velocity of the protected head is nearly 80% of the unprotected head. Although this change is not significant but the duration under which this change happens is much higher than that of the unprotected head. This will result in lower acceleration of mass center of the head. According to Hutchinson. J., 1998, accelerations higher than 1000 g, increase the risk of brain trauma. Maximum value of mass center of the head acceleration is 180 g for protected head and 1064 g for unprotected head. So a considerable amount of the energy is absorbed in the protected head.

The results in both sections (4.1 and 4.2) were close to the literature (Afshari A. et al., 2008) with less that 15% difference, which is because of the simplified geometries, different material properties and different direction of the impact to the head (Lateral impact instead of frontal impact).

5.2 Key Research Accomplishments

- A comprehensive literature survey on head and helmet impacts
- Application of FEM for analyzing head trauma in two different cases (Head protected with a helmet and unprotected head).
- Determination of the importance of the helmet in reducing the normal and Von Mises stresses on brain and skull.
- Identification of the limitations of the study to clarify the areas that could be improved in future studies.

- Considering lateral impacts instead of frontal impacts on the helmet unlike previous studies.
- Considering the most current material properties for the brain, skull and helmet.

Chapter 6

CONCLUSION AND FUTURE WORK

In this study the 3D models of the brain, skull and helmet were created using Solidworks and two analyses were conducted in ANSYS to find out how the helmet will protect the head from lateral impacts under conditions of standard tests. Normal and Von Mises Stresses on different parts of the head and helmet were calculated and compared. The results showed a dramatic change in the intensity of the stresses on the brain and the skull between the first (the protected head) and the second analysis (unprotected head). Helmet has reduced stresses significantly on brain and more than 90% of the stresses were reduced on protected head.

Due to lack availability of high capacity CPU, the simplified versions of the models were designed and the element numbers were reduced. The acceptable range of the results show that the FE head model was developed properly.

It is advised that a more comprehensive model may be built that include all the layers of the head complex to find more realistic results and since the accuracy of the model would be higher a finer meshing will be applied to the models to increase the accuracy of the final result in the FEA software. The proposed model may include scalp, 3-layered skull (outer and inner tables, diploe), Dura, CSF, Pia, Falx, Tentorium, Cerebral

Hemispheres, Cerebellum and Brain Stem. The helmet will be improved as well to include the outer shell and the straps.

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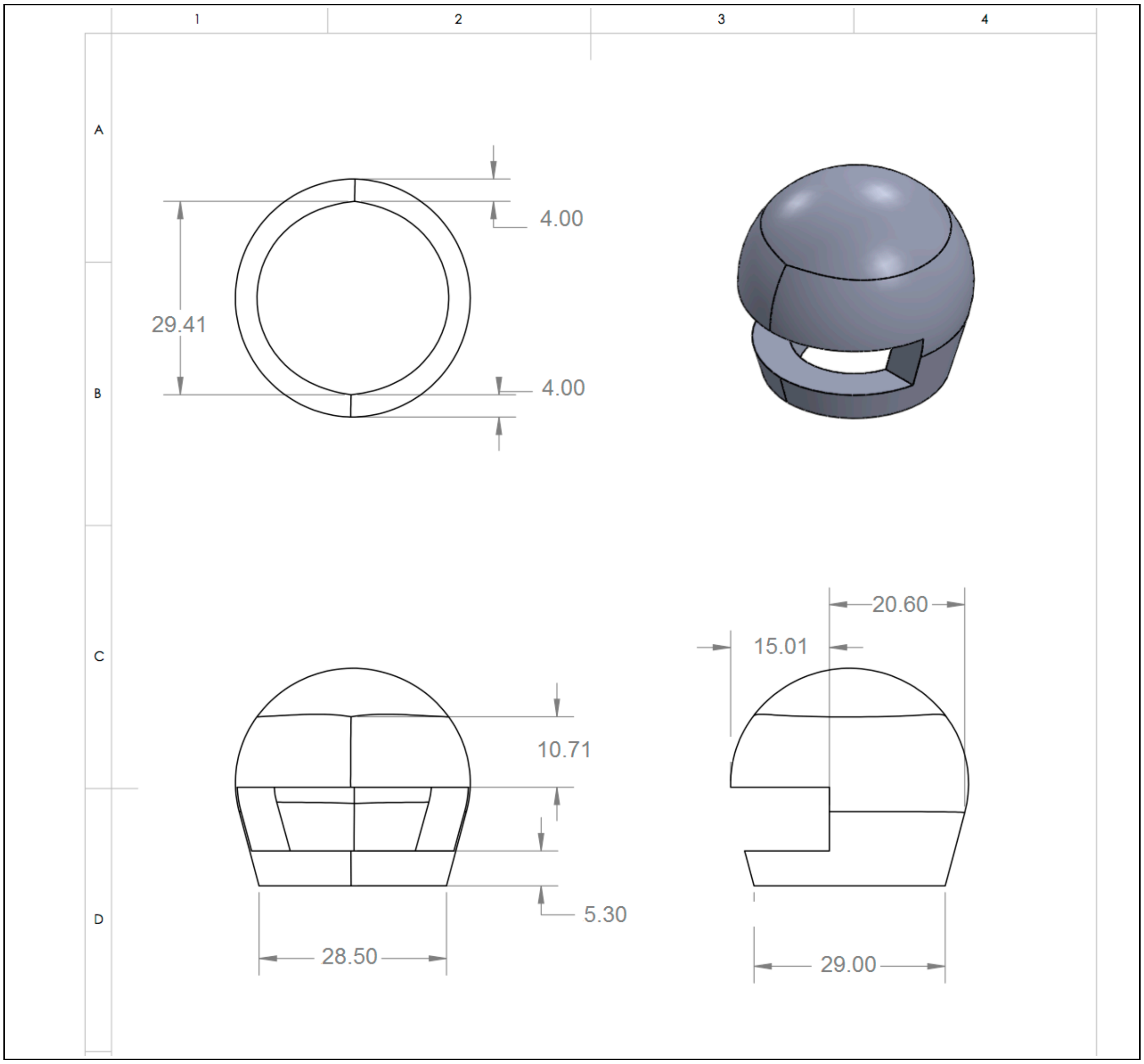
APPENDIX

Appendix 1



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