Finite Element Analysis of Head and Brain During Frontal Impact Collisions

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ABSTRACT

Traumatic brain injury (TBI) is known as the most important reason for human fatalities in car accidents. Many studies have been performed to understand the structure and mechanisms of head and brain injuries. By the development of computer science in engineering, new numerical-based methods have been introduced to develop and analyze head and brain models and inquire a better explanation for head and brain traumas. Finite element method (FEM) opened a new gateway to perform easier and efficient numerical analysis. In order to obtain more realistic results, more sophisticated and comprehensive models have been also developed. To my knowledge, many numerical studies have been performed to find pressure or stress distribution on brain only. Therefore, the behavior of the parts in human head has not been revealed. The goal of this thesis was to develop a three dimensional (3D) FE model that contains scalp, cerebrospinal fluid (CSF), dura, pia, falx and tentorium, and brain of human head and then analyze pressure and Von-Misses stress distribution on these parts. The FE model contained 12 parts of the human head. The brain was modeled as viscoelastic material and the other parts were modeled as linear elastic materials. The aim was to investigate the impact effects of 5.95 kg object with the velocity of 9.49 m/s and the angle of 45° with the x axis. Therefore, Von-Misses stress and pressure distribution caused by the impact have been observed from the FE analysis. It was seen from the results that the maximum amount of pressure and stress were obtained on most head parts between 0.004 seconds and 0.00475 seconds after the impact. To my knowledge, the stress and pressure analyses on dura, pia, falx, tentorium, and scalp were performed for the first time in this thesis that can be considered as the novelty of the present work. In order

to validate the model, the obtained results from the pressure analysis of brain were compared with the published experimental results and a good correlation was witnessed between them.

Keywords: Brain, Head, Injury, Frontal Impact, Finite Element Analysis, Modeling, Biomechanics

ÖZ

Travmatik beyin hasarı, araba kazalarında insan ölümlerinin en önemli nedeni olarak bilinir. Birçok çalışma, kafa ve beyin yaralanmalarının yapısını ve mekanizmalarını anlamak için yapılmıştır.

Gelişmiş bilgisayar biliminin mühendislik alanında kullanılmasıyla, yeni nümerik yöntemler, baş ve beyin modelleri analizlerini ve beyin travmalarının çalışmalarını da geliştirmiştir.

Böylece, sonlu elemanlar yönteminin gelişmiş bilgisayarlarda kullanılması, daha kolay ve verimli nümerik analizler gerçekleştirmek için yeni bir ağ geçidi açmıştır. Daha gerçekçi sonuçlar elde etmek amacıyla da, daha gelişmiş ve kapsamlı modeller de geliştirilmiştir. Bildiğim kadarıyla, birçok nümerik çalışma sadece beyin üzerindeki baskı ya da stres dağılımını bulmak için yapılmıştır. Bu nedenle, insan kafatası içindeki diğer parçaların üzerindeki baskı ya da stres dağılımı bugüne kadar çalışılmamıştır.

Bu tezin amacı, baş derisi, beyin omurilik sıvısı, sert zar, ince zar, beyin orağı, beyin içi, beyin ve baş gibi bölümleri kapsayan üç boyutlu sonlu elemanlar modeli geliştirip daha sonra da bu bölümler üzerindeki stres ile Von-Misses stres dağılımınının analizini yapmaktır. Bu çalışmada kullanılan sonlu elemanlar modeli toplam oniki beyin ve baş bölümünden oluşmaktadır. Bu çalışmada yalnızca beyin viskoelastik malzeme olarak modellenmiş ve diğer parçalar da elastik malzeme olarak modellenmiştir. Yürütülen sonlu elemanlar analizinde, 9.49 m/s hızı, 5.95 kg ağırlığı ve yatay ile 45°' lik açısı olan bir nesnenin baş ve başın diğer bölümleri üzerindeki darbe etkileri incelenmiştir. Darbeden sonra elde edilen sonuçlar, baskı ve stresin 0.004 ile 0.00475 saniyelerinde maksimum değerlere ulaştığını göstermektedir. Böylece, darbenin baş ve başın diğer bölümlerine olan etkileri ilk kez bu nümerik calışmada bulunmuştur. Elde edilen sonuçların ve geliştirilen sonlu elemanlar modelinin geçerliliği daha önce yapılan çalışmalarla karşılaştırılarak onaylanmıştır.

Anahtar Kelimeler: Beyin, Baş, Yaralanma, Önden Darbe, Sonlu Eleman Analizi, Modelleme, Biyomekanik

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LIST OF SYMBOLS/ABBREVIATIONS

3D	Three Dimensional
AIS	Abbreviated Injury Scale
ATD	Anthropomorphic Test Devices
BET	Brain Extraction Tool
CFD	Computational Fluid Dynamics
CNS	Central Nervous System
COR	Center of Rotation
CPU	Central Processing Unit
CSF	Cerebrospinal Fluid
CT	Computed Tomography
FE	Finite Element
FEA	Finite Element Analysis
FEM	Finite Element Model/Finite Element Method
GAMBIT	Generalized Acceleration Model for Brain Injury Threshold
HIC	Head Injury Criterion
HIP	Head Impact Power

MRI	Magnetic Resonance Images
PDE	Partial Differential Equations
PNS	Peripheral Nervous System
TBI	Traumatic Brain Injury

Chapter 1

INTRODUCTION

1.1 Increasing Need of Biomechanical Modeling of Humans Head

Head and brain injuries are one of the most important causes of fatalities. There are over 50 million traumatic brain injury victims each year which causes 1.2 million deaths every year (1 every 4 minutes). The exact percentage of fatalities of head or brain trauma injuries are unknown, but about 5 to 10 percent of those who known hospitalized for head or brain injuries, die (Kleiven, 2002), and over half of the deaths occur within 2 hours of hospitalization. The main cause of brain injuries are motor vehicle accidents (51%), falls (21%), assaults and violence (12%), sports (10%) and other reasons (6%). Not necessarily an impact may cause the brain injury, but also a sudden linear or angular acceleration may lead to a brain injury, for example whiplash in a car crash or vigorous shaking of an infant or young child by the arms, legs, chest or shoulders can also cause head or brain injuries.

Growth of number of cars is followed by growth of traffic accidents which is the biggest cause of brain traumas. In order to treat brain traumas it's crucial to get a better understanding of human's body. Pathology of the organs should be studied accurately and for this, scientific developments have been made in medical and biomechanical fields. With use of Magnetic Resonance Images (MRI), hundreds of images are taken from different layers of the brain and by this tool, the structure of head, brain and even the trauma, can be discovered in details. By benefitting MR images a huge improvement has been achieved in medical and biomechanical field by giving a better understanding of internal structure of head, brain and the trauma. These images can be used to create three dimensional (3D) model of human's head and brain.

Many studies have been initiated in order to understand the structure of damages occur during an impact. These studies generally have been performed experimentally, analytically and with numerical methods. In experimental methods, cadaver and dummies are used to model the behavior of the subject to stimulus. The accuracy in this method is highly dependent to the quality of motion detectors and accelerometers, and as it has highly expensive procedure and extremely difficult process to perform, make a handful of obstacles on the way of performing the experiment. There are certain advantages in experimental method though. However, performing the cadaveric experiment with similar body mass distribution is advantageous. The body geometry may be the same as the geometry of alive human, although the characteristics of human's body start changing just after death, but it still has the closest properties to a live human body.

By development of computers in engineering science, new methods emerged that eased the way of researchers to perform their analysis. Finite Element Method (FEM) is the offspring of computer usage in engineering and is one of the mostly used tools for modeling the behavior of materials and bodies under the influence of any stimulus. FEM is a numerical method that converts Partial Differential Equations (PDE) into simpler linear equations by dividing a complex geometry into small elements and applies calculus variation methods and minimizes the errors by iterations and gives an approximate result. This method is especially useful for solving PDEs which are impossible to solve by analytical methods in complex geometries.

By improvements achieved in FEA method, better 3D models designed and more accurate results were achieved. Using MRI and CT scan images, softwares can extract 3D models of human head and brain. Computer programs like SFLView by university of Oxford (FSLView, 2013) by using images and Brain Extraction Tool (BET) has been developed for this issue.

For analyzing the 3D models, many software packages had been developed. Commercial software like ANSYS and ABAQUS are widely used. Besides commercial softwares, there are also open source programs written in C++ or FORTRAN.

The objective of the thesis is to develop a 3D FE model of human head and brain, analyze the effects of a frontal impact on the model, obtaining intracranial pressure and Von-Misses stress. Then it is aimed to evaluate the observed result by comparing them with the published data. The results from this thesis can then be used to predict the damage level of human head due to a frontal impact.

1.2 Anatomy of Human head

For a better understanding of the outcome of the research it crucial to have a good knowledge in the anatomy of human's head and brain.

Central Nervous System (CNS) and Peripheral Nervous System (PNS) are the two main parts of the human's nervous system. The primary objective of nervous system is to analyze and respond to the data received from sensors all around human's body (i.e. Eyes, Ears, Skin and Tongue). The brain tissue is soft and non-replaceable, so it should be protected by a strong solid and fortified tissue, such as the skull. The skull itself is covered by skin which is a protection layer as well. In the following subsections, parts of the head and brain are explained in detail.

1.2.1 Skin

Epidermis, Dermis and Hypodermis are the three layers of the skin with hairs, nails, sweat glands, oil glands and blood vessels, form Constitutive System. Epidermis is the outer layer and the first protective layer in the skin. This layer is waterproof and skin tone is provided by this layer. It mainly consists of dead cells. Hair follicles, sweat glands and a vascularized fibrous connective tissue are placed in Dermis and Hypodermis and consist of fat and connective tissue.

1.2.2 Skull

Occipital, Sphenoid, Ethmoid, Paired Frontal, and Paired Parietal form the skull base, connected together by Sutures. There are 22 bones constructing the structure of skull which have enough strength to protect the highly sensitive organ inside of it, the brain, and the frontal bones that form the shape of humans face.

1.2.3 Meninges

The purpose of meninges is to cover and protect the Central Nervous System and blood vessels. This protective layer is consisted of three sub-layers. Dura Mater is the outermost layer and Cerebrospinal Fluid and Brain are kept in, as they are by this layer. Pia mater is the innermost fluid impervious layer that emprise Cerebrospinal Fluid, and Arachnoid mater is placed in between Dura and Pia mater.

1.2.4 Cerebrospinal fluid (CSF)

CSF is a fluid that encloses the brain and helps nutriment of it. It flows in subarachnoid area and is one of the protective factors of the brain. Another function of this important material is to provide a dense environment to reduce the net weight of the brain. The structure of CFS is very similar to bloods plasma and flows through ventricles.

1.2.5 Brain (Cerebrum)

Brain is one of the most complicated and sensitive structures of humans' body. This organ consumes up to over 30% of the daily calories and it absorbs the most of it from carbohydrates. As this structure consists of a high volume of water, (about 80%) it is almost incompressible. Brain consumes Oxygen more than any other organ in human's body. This organ is the most important part of Central Nervous System (CNS) and in humans consists of different parts such as Thalamus, Hypothalamus, Pons and Brainstem.

1.2.6 Falx and Tentorium

Falx is a wrinkle of dura matter that is extended in between brain hemispheres and attached to another Dura mater fold, Tentorium. Tentorium protects the upper side of Cerebellum and occipital lobes are sustained by this structure.

1.3 Head and Brain Injury Mechanism

In a head trauma various parts may be involved. Scalp, skull, meninges and the brain itself are of the structures that can be affected and injured. It should be noted that in a severe brain injury it might not necessarily be distinguished in scalp or skin, and vice versa, an extensive damage to skull may be uphold without a significant damage to the brain.

Theoretically, a damage to head is caused by the momentum energy of a sudden change in acceleration of head or an impact between a stationary object and a moving head. Otherwise damage may occur with a stationary head and a moving object or both moving head and object (Moritz, 1943). Not only the velocity and direction, but the shape of the moving object is significantly effective in severity of trauma.

1.3.1 Focal and Diffuse Injuries

Medical researches indicate that brain traumas can be categorized into two groups: focal injuries and diffuse injuries. Focal injuries are of those injuries that can be observed by eyes. These can include scalp injuries and skull deformations. On the other hand, diffuse injuries referred to a kind of injuries that are difficult to detect and cannot be identified by eye witnessing. Brain dysfunctions caused by external stimulations, damages to blood vessels or traumas caused by meningitis can be counted in this group of head traumas.

Usually focal damages are caused by direct impacts to a certain place of head like strikes to the head in a car accident while diffuse injuries are mostly caused because of a sudden change in acceleration, direction or rotations. Regarding this point that although these are two different categories of head injuries, but they can happen individually or both at the same time with same stimulus. It means that, brain can suffering diffuse damages without occurrence of focal damage and vice versa.

1.3.2 Abbreviated Injury Scale (AIS)

Measurement of the extent of damage is performed by various criteria. Abbreviated Injury scale is used to measure and compare the severity of damage particularly in car accidents. For example, it can be used to compare the severity and predict the probability of survival between drivers and front seat passengers. The scale start from 0 to 6 and with the number growing the intensity of damage will be increased (Prasad P, 1985).

1.3.3 Head Injury Criterion (HIC)

The most popular method for scaling and prediction of the damage severity is Head Injury Criterion (HIC) which was introduced by Lissner et al., in 1960and it is the relevancy of acceleration and duration. For a person, maximum HIC value is defined 1000 and above this number, it is considered a threat for human's life (Kleiven, 2002). HIC limits have been suggested by Kleinberger et al (1998) for 36ms and by Eppinger et al in (2000) for 15ms (HIC_{36} and HIC_{15}). A summary of proposed values is given in Table 1.1.

Table 1.1: Suggested HIC limit values (Michael Kleinberger, 1998) & (Rolf Eppinger, 2000)

Dummy Type	*Mid-sizeed Male*	*Small female*	*6 years old child*	*3 years old child*	*12 months old child*
$\boldsymbol{HIC_{36}}$	1000	1000	1000	900	660
$\boldsymbol{HIC_{15}}$	700	700	700	570	390

1.3.4 Head Impact Power (HIP)

Newman et al (2000) presented a new criterion called head impact power (HIP) to measure brain damage that besides kinematics of the impact also considers variation rate of rotational and translational kinetic energy of the impact and presented coefficients to normalize the results for various directions with respect to the errors in certain directions. Rotational and translational kinetic energy rate of change can be defined as:

$$Power = P = \sum m\bar{a}.\bar{v} + \sum I\bar{\alpha}.\bar{\omega} \qquad (1)$$

Where:

$\bar{a}$ is linear acceleration, m is mass, v is velocity,I is mass moment of inertia,$\bar{\alpha}$ is rotational acceleration, and $\bar{\omega}$ is angular velocity.

Extending of the equation in three directions gives:

$$HIP = m\int a_x dt + m\int a_y dt + m\int a_z dt + I_{xx}\alpha_x \int \alpha_x dt + I_{yy}\alpha_y \int \alpha_y dt + I_{zz}\alpha_z \int \alpha_z dt \quad (2)$$

The value of HIP changes from 0 to a maximum value and then drops to 0 again. At the peak, the rate of kinematic energy change is maximum, and in this point, the probability of head injury will be the highest.

1.3.5 Generalized Acceleration Model for Brain Injury Threshold (GAMBIT)

Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) is another proposed criteria by Newman (1986) based on combination of rotational and translational kinematic thresholds. The following equation has been derived to evaluate *G(t)*:

$$G(t) = [\left(\frac{a(t)}{a_c}\right)^n + \left(\frac{\alpha(t)}{\alpha_c}\right)^m]^{\frac{1}{s}} \quad (3)$$

In this equation:

a(t) is instantaneous translational acceleration, α(t) is instantaneous rotational acceleration,a_c is the translational acceleration threshold, $\alpha_c$is the rotational acceleration threshold and n, m and s are experimental constants

Due to insufficient data required for deriving threshold and neglecting the important factor of direction, this criterion was never popular among car manufacturers and safety institutes administrations.

Organization of the Thesis

In this thesis the contents has been categorized into 5 chapters. In Chapter 1 the introduction is written. The major reason of initiating the investigation in the introduction is describing of human's head anatomy, overviewing of common head and brain traumas and the criteria used to evaluate and predict the damage. In Chapter 2, literature review is written and it is focused on the process of progress in biomechanical modeling of human head and brain, and the development of finite element models of human head. Also experimental analyses through years have been followed. Chapter 3 demonstrates all the steps of developing of a 3D human head and brain model, defining the materials and software analyses in FEA. Results have been presented in Chapter 4 and finally the discussion about the results, conclusion, and suggestions for future work are placed in chapter 5.

Chapter 2

LITERATURE REVIEW

A biomechanical model can be used wherever there is a need to examine musculoskeletal body, joint, and tissue behavior. As it is very difficult and expensive to obtain accurate data and results from experimental tests, biomechanical models are mainly used instead. Biomechanical models are utilized to understand stresses, impacts and forces of bodies. These can be analyzed by analytical methods and compared with experimental methods. In this chapter, biomechanical studies about the response of skull and brain to frontal impact collisions with various directions are explained.

2.1 Biomechanical Modeling of Human Head and Brain

A human head model consists of several parts. In order to achieve better and more realistic results in analysis, not only the design and geometry, but also quantification of materials and converting the raw model into a FEM in a proper way are essential. In order to acquire the ability of designing all the sections and layers of the human head and brain model in the most accurate state, it is crucial to know about the anatomy and physiology of human head and brain very well.

During the past 40 years biomechanical models of head had been gone through many changes and development. After vogue of widespread use of computers the more detailed models developed and the analysis of these models got so much easier and significantly accurate.

Physical models had been widely used for studying the traumas. Holburn (1943) reported as the first person who developed a physical model of human head with an idealized geometry using sagittal plane and No-slip condition. Gurdjian and Lessner (1961), and Thibault (1987) improved the former model. Aldman (1981), Margulies (1987), Viano (1997) and Ivarsson (2000) considered the Slip condition in their models. In the meantime, Thibault (1987) and Margulies (1990) published their results using coronal plane with exact geometry and No-slip condition and Meany and Thibault (1990), Meany (1991) performed the same researches with an idealized geometry and consideration of Slip condition in the coronal plane.

Besides physical models some experiments have been done on cadavers and animals as well. Nahum (1977) performed a test on cadavers which the data is based on intracranial pressure and most of the few analytical results which have been validated by an experimental data, referenced Nahum's research. Bradshaw (2001) believed that it is not the best interest to rely only in intracranial data and showed that low shear modulus of brain tissue can cause a significant shear respond to an impact.

2.2 Development of Head and Brain Finite Element Models

Finite Element Method (FEM) is a numerical approach that can solve Partial Differential Equations (PDE), which is almost impossible to solve by hand and by regular analytical methods. In this method a complicated physical model is divided into several small elements and these elements are connected to each other by nodes. As a result there is a relation between each element by these nodes that is defined by PDE. FEM by eliminating the spatial derivatives from the PDE finds approximate solutions for them. Many computer programs and softwares have been developed to solve problems and complicated equations obtained from various fields of studies such as structural tests in civil engineering, fluid mechanic analysis and biomechanics.

Finite element analysis on human's head performed earlier in 1973 by Hardy (1973) for the first time and it was then improved a year after by Nickell (1974). The human's head model consisted of a skull and brain only and other parts inside of the skull were neglected. The model was very simple and contained only 45 nodes. The data collected from their research was concluded that a 3500 lbs. load on front and 1400 lbs. on the sides of the head are the amount of load that triggers the first cracks on the skull. In 1975, Shugar improved Hardy's model and took the brain into the consideration as a fluid and faced a great deal of improvement in the results in comparison with Hardy and Marcals study. The details in the 3D model of head significantly improved by Horsey in 1981. Skull bones, spinal cord, brain and cerebrospinal fluid space (CSF) were combined and included in a single model. The exact geometry and morphological continuity of each part was set up carefully.

Trosseille in 1992 developed a test on cadavers in order to obtain the accurate and more realistic properties of human head. CSF pressure in various points, in different conditions and 3D dynamics of head were evaluated. This test became a reference for evaluating and comparing the other 3D models of head.

Ruan (1994) defined the brain material as inviscid continuum material and CSF as a thin viscous layer. He illustrated that intracranial pressure caused by an occipital impact on the frontal region which is higher than the pressure in occipital part caused by a frontal impact. This means that Traumatic Brain Injuries (TBI) are more likely to happen when the impact is from behind.

In 1995, three head models were proposed by "Laboratoire des Systemes Biomecanique" (LSBM) and Willinger (1995) commenced a research to calculate intracranial properties by applying stresses on each of these models. In these models head and brain were taken into consideration as a complex structure which includes non-deformable and deformable elements. The results concluded that besides intensity of loads, proximity of energy absorption to the natural frequency of head and brain is important as well. It means that if the aim is to damp the energy, the frequency of the energy should not be close to the natural frequency of heads elements.

In 1996, Kang et al. developed a model in University of Louis Pasteur (which also passed Trosseille test in frontal impacts) based on the properties calculated by Willinger, et al., and took the brain closed to a rigid body. Two impacts were applied to the model such as, a short duration and a long duration impacts. The results for the long duration impact illustrated a couple in the frontal and occipital intracranial pressure which did not happen in the experiments. As a result, reconsideration was needed for Willingers data.

Willinger, et al. (1999), conducted a new research to understand the reason of the mismatches in data collected by Kang, et al. After developing a 3D FEM of human head and applying several impact loads he concluded the modulus which is defined for CSF is not constant under different conditions. He described that, in order to model and analyze CSF, it either should be defined by nonlinear models or the model should be solved via a fluid-structure approach.

In 2002 Brands, et al., investigated the nonlinearity of brain tissue properties in rotational accelerations. First a physical model (cylindrical cup) was filled with silicone gel, it was subjected to rotational acceleration and the results were compared with the data collected from 3D FEA model with non-linear properties. The results illustrated up to 20% difference in strain between the linear and non-linear models and 11% difference in stresses respectively.

2.3 Experimental Studies for Testing Traumatic Skull-Brain Injuries

The validity of analytical data is determined by comparing it with experimental data. As well as, non-conformed simulation results might not be accurate and cannot be reliable. Therefore it is vital for a strong research to evaluate physical and experimental tests. Performing an experimental test needs lots of effort and investments, and in many cases it is very difficult to perform. In some cases it is even impossible to obtain experimental data. There are very few experimental data used to be compared with analytical experiments. To the best of my knowledge, the first test performed on human head is Nahum's test which was performed in 1977. This test is the most referenced test for validating analytical data.

After many years, Trosseille's test was performed in 1992. In this test, cadavers were subjected to apply impact to the head in several directions. In this particular study in order to obtain data, several accelerometer sensors were designed and implanted in the cadavers head. The results were more accurate and general that is used for evaluating 3D models of head and brain.

Another way to collect experimental data is to perform test on dummies. In most of the cases, the material model used as brain tissue is silicone gel. Bradshaw simulated a head impact on a dummy in 2001. Silicone gel was used as the brain material, CSF was modeled with paraffin. Medical images (MR images and CT scans) were used for designing 3D model.

Recently surrogates are widely used in simulating impacts in experiments and they are known as anthropomorphic test devices (ATD). In 2012, a test has been performed on an ATD (HYBRID III) by Adam Bartsch et al. (2012) to achieve better understanding of head-neck response to direct impacts from all three directions (Frontal, oblique frontal and lateral) were applied. The results of this test led to a better understanding of the response of a head-neck impact on HYPRID III ATD and improved head and neck testing standards.

There are so few of these models which have been validated by experimental data. The best known is FE model that was validated by experimental data was Ruan's (1994) model which has been compared to Nahum's experiment. Willinger et al., in 2002 performed a comprehensive research and simulated short duration impacts on FE model presented at University Luis Pasteur and validated his results with two experimental data (Nahum's and Trossielle's). This was the first time a FEM had been validated by two experimental data.

Kleiven, S. (2006) developed a FE model and studied the effect of impact loads in different directions (frontal, occipital and lateral), durations and validated his results from short duration impact with Nahum's published data. The results from long duration impacts were validated with Trossielle's set of data. In (2006) Zong et al., validated their FE model against two sets of experimental data and improved the results obtained by Willinger (2002).

The studies and investigations that preformed indicate the fact that each investigation has focused on a few parts and there were so few FE models that includes a comprehensive data in various layers of head and brain. These analyzes are even more limited when they get to the part that the comparison between analytical data and experiments should be done. Meanwhile, by increasing the processing capacity of computers, FE models should also get more accurate and comprehensive.

In this thesis, it is attempted to develop and modify a FE model that includes most of head and brain tissues and extract a comprehensive data from multiple sections of head and brain. Von-Mises stress and pressure analysis have been performed of CSF, dura, pia, scalp, falx and tentorium, and viscoelastic brain. In order to evaluate the model, the extracted results from pressure analysis of brain, have been compared with Nahum's experimental results. Comparing the results clearly illustrates a good correlation between them. To my knowledge, pressure and stress analysis have not been performed on several sections of head and brain, so some parts of extracted data can be considered as a newly acquired data and can be used in future researches.

Chapter 3

MATERIALS AND METHODS

Many studies on human head are mainly focused on the head influenced by stresses. The stimulus can be linear and rotational acceleration, impacts or loads, and the effects of these stresses are generally obtained by experimental, analytical and numerical methods. Dummies and cadaver are used in experimental methods but in many cases the experimental method is far too difficult or expensive to perform and the accuracy of the results are highly dependent on the experiment condition and the quality of the motion sensors. Another way to analyze the biomechanics of humans head is analytical methods but biomechanical analyses are generally way more complicated than that to be handled by analytical methods. Numerical methods, especially after introduction of computer, are used in many biomechanical issues. Finite Element Method (FEM) uses approximations to solve partial differential equations (PDE) by converting them to simple linear equations and in complicated geometries like simulation of humans head, it is very beneficial.

The sequence in Finite Element Analysis (FEA) is summarized in 3 steps. First, a 3D model should be constructed. In the next step the stimulus is applied, and finally, after FEA is done, the results should be compared and validated by experimental results.

3.1 Basic Principles

3.1.1 Conservation of Mass and Momentum

In almost every numerical simulation of impact modeling conservation of mass and conservation of momentum laws are used as the fundamental principles of the study.

The definition of conservation of mass in impact simulations is, when and impact happens between two objects, the mass of both objects will not change as given in equation (4).

$$\int \rho \, dV = const \tag{4}$$

In this equation, V is the volume of the body and ρ is mass per unit volume (density).

The conservation of momentum states that if in and isolated system, a collision occurs between two objects, total amount of momentum of the two objects will be the same, before and after the collision.

$$F = m\frac{dv}{dt} \tag{5}$$

So for a closed loop system that contains n number of masses, the conservation of momentum equation is given as:

$$\sum_{i=1}^{n} m_i v_i = const \tag{6}$$

The mass is presented by *m,* and *v* is velocity in this equation.

3.1.2 Impact Force

Calculation of the amount of force which is caused by the impact between the elastic bodies (impact force) is one of the challenges in FEA of any instance. This value can be calculated by the following equation:

$$F = K\alpha^{\frac{3}{2}} \tag{7}$$

Where:

F is Contact force

K is a constant which is varies in different material and geometry properties and it is called Hertzian contact constant, and the relative displacement between the bodies is shown by α (Nahum, 1977).

3.1.3 Von Mises Stress

It is vital to investigate about the stress distribution in an impact simulation. In material science, it is said that, a material starts yielding when it reaches to a certain amount of stress called yield strength. In order to predict when a material starts yielding, Von Mises Stress is calculated. This equation sums up stresses in all axes and the value is used for prediction on yielding in under stressed materials.

The following shows the calculation of Von Mises stress:

$$\sigma_v = \sqrt{\frac{1}{2}[(\sigma_{11} - \sigma_{22})^2 + (\sigma_{22} - \sigma_{33})^2 + (\sigma_{11} - \sigma_{33})^2 + 6(\sigma_{23}^2 + \sigma_{31}^2 + \sigma_{12}^2)} \tag{8}$$

3.1.4 Head Injury Criteria (HIC)

Head Injury Criterion (HIC) is a scale to measure how harmful is an impact can be on human head. It can be used to measure safety of motor vehicles and sports instruments. For a better perception of HIC, for an instance, researchers believe that HIC of over 1000 is absolutely dangerous and harmful. An 18% chance of intense traumatic brain injury (TBI) is predicted in HIC of 1000, also a 55% chance of serious damage and 90% chance of a moderate damage to an average adult human head is predicted (Maas AI, 2007).

HIC can be calculated by the following equation:

$$HIC = max_{t_1,t_2}\{(t_2 - t_1)\left(\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)\,dt\right)^{2.5}\} \tag{9}$$

Where:

t is the time duration which impact happens

a is acceleration during the impact. The values of the variables are t in seconds and a in units of gravity acceleration (g = 9.81 m/s^2).

The values of HIC and the predicted probability of injury scaled by maximum abbreviated injury scale are shown in Figure 3.1.

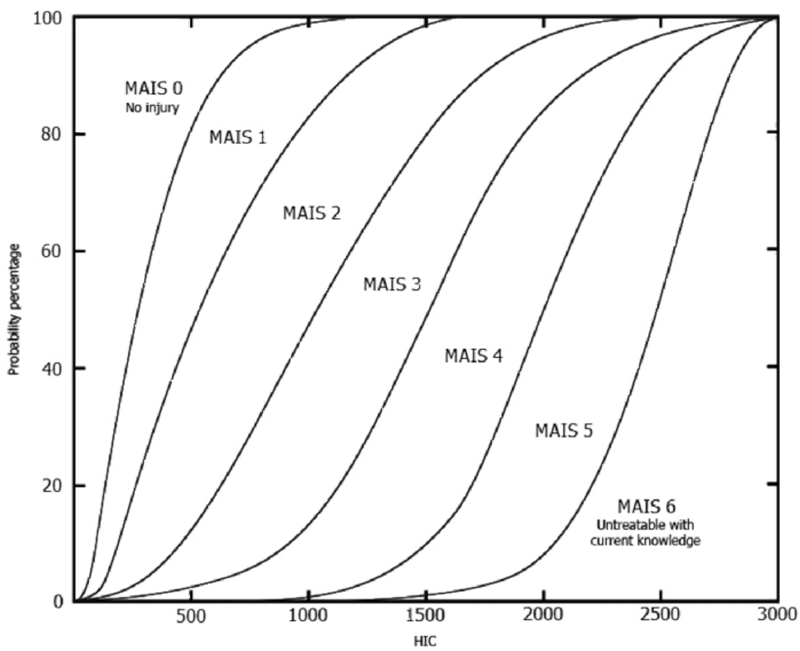

Figure 3.1: Values of HIC and the predicted probability of injury scaled by maximum abbreviated injury scale (Mackay, 2007)

3.2 Finite Element Analysis of Head and Brain with ABAQUS Software

Computer software has been developed during the past 35 years by increasing demand of using computer in engineering to speed up the calculations. In this thesis, ABAQUS software is used for FE analysis of 3D head. The fundamental principles of ABAQUS are able to compute linear and non-linear (for complex geometries and non-linear mechanical problems) FEA using continuum mechanics. This software can be used in wide range of mechanical problems. The ability of solving thermodynamic related problems in fluid mechanics like computational fluid dynamics (CFD) problems and dynamic and static analysis of solid materials in complex geometries are the eminent advantages of this software.

3.3 Construction of 3D Head Model

The primary form of the 3D head model was offered by Polytechnic University in Tehran, Iran. Changes had been made in order to make it suitable for the current research (Figure 3.2). In order to represent the impact body, a rectangular rigid body was constructed. The human head model shown in Figure 3.2 consists of 12 parts: Viscoelastic Brain, CSF, Cortical Bone, Trabecular Bone, Scalp, Dura, Tentorium, Pia, Arachnoid, Arachnoid Trabecular, Falx and Facial Bones.

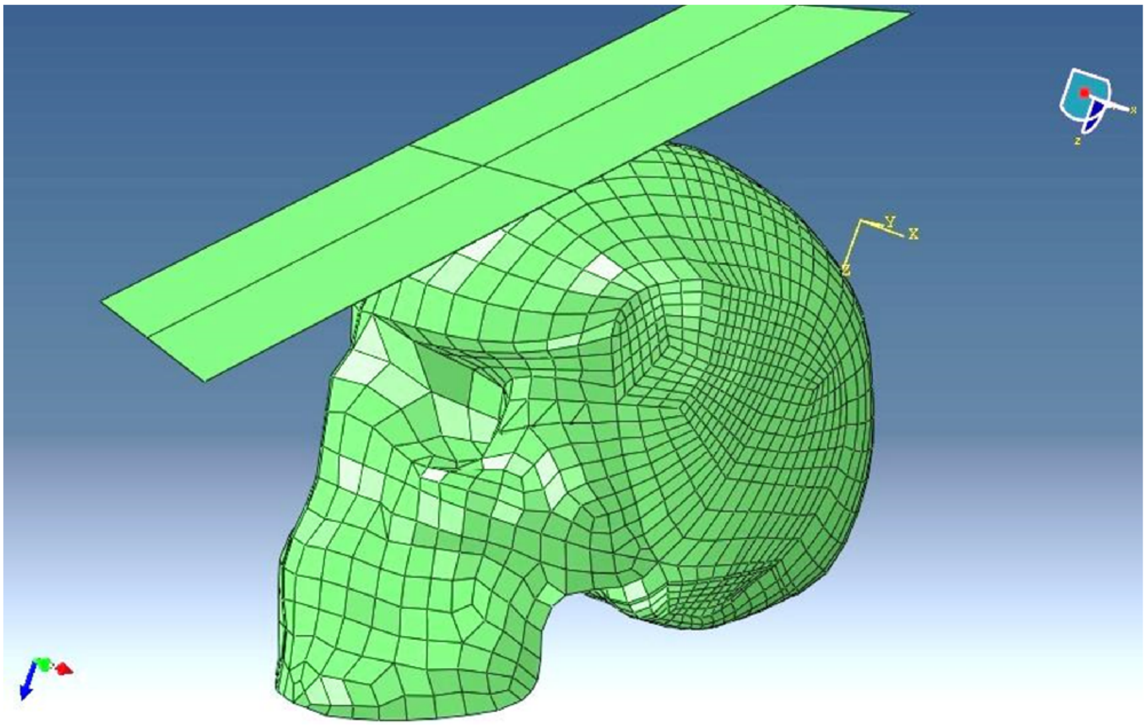

Figure 3.2: Finite Element model of human head

In order to validate this model, it has been modeled similar to Nahum's study (Nahum, 1977). The head is rotated 45° to the front and the equivalent forces that simulates the impact forces recorded in Nahum's experiment produced by a 5.95 kg impactor with 9.49 m/s has been applied to the forehead as the stimulus (test No. 42). Direction of the impact forces passed through the center of gravity of head to avoid rotations.

The impact forces have been applied in 1ms and duration of experiment is set to 1.5ms. The amount of impact forces recorded in Nahum's experiment caused by the impactor is drawn in Figure 3.3 with respect to time of impact. A rigid body has been defined in order to simulate the impactor.

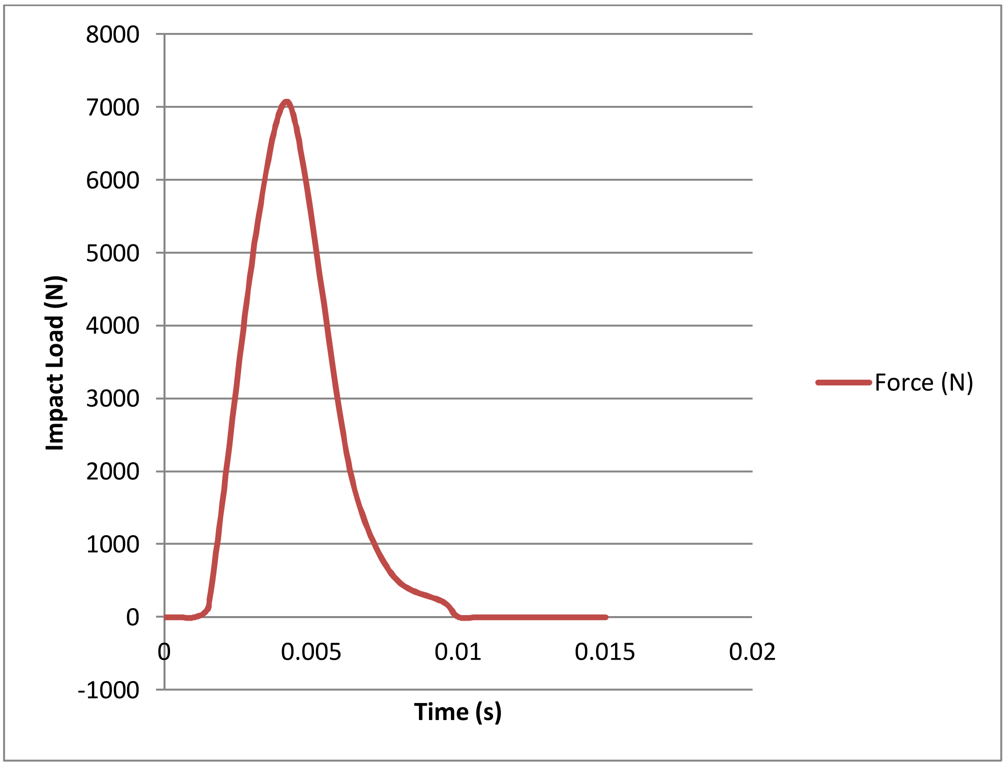

Figure 3.3: illustration of the applied impact force during 15ms

3.3.1 Materiel Properties of Tissues

Since human head has a complex geometry, estimating material properties of human's head and brain tissues, is so difficult. According to the experiments, (C´elia Maria Maganhotto de Souza Silva, 2005) results are highly dependent to material properties and when a high speed impact is involved, it makes the problem much more complicated because of great deal of parameters being involved in the analyzes. In this study, material properties have been retrieved from recent studies and literature. In some parts, the average amount of the properties in different literatures has been used. (Ruan J. K., 1993) (A. Charalambopoulos, 1999) (Kleiven, 2006)

All sections of the human head are defined with linear properties except brain which defined as a non-linear viscoelastic material. The viscoelastic frequency data has been set with respect to time. For the viscoelastic brain, a Young's modulus of 22.8 KPa and Poisson ratio of 0.4999982742 has been chosen from the literature (Ruan J. K., 1993). According to Tamer El Sayed (2008) yield stress of brain and other soft tissue materials is set to 20.0 KPa. Density has been set to 1040 Kg/m^3. In a viscoelastic material, the stress that is applies to the material, initially affects the elastic property of the material, and then due the time, stress starts to relax which is a result of the viscoelastic property of the material. In this case shear stress relaxation behavior of a viscoelastic material is defined by Prony series as:

$$G(t) = G_\infty + \sum_{i=1}^{N} G_i . e^{-t\beta} \tag{10}$$

where G_∞ is the infinite shear modulus, G_0 is the short-time shear modulus and β is the decay coefficient. This variable is time *t* in this equation. For the current research, shear relaxation behavior is described by the following equation:

$$G(t) = G_\infty + (G_0 - G_\infty). e^{-t\beta} \tag{11}$$

The values has been set according to (G. Belingardi, 2005) previously performed research. In the head model, besides brain, Arachnoid, Arachnoid Trabecular, Falx and Tentorium, Cortical bone, Trabecular bone and Pia were defined by linear viscoelastic properties, so the Prony series were used in order to simulate their behavior.

In literature, Young's modulus used for scalp varies from 6 MPa (Willinger, 2002) up to 16.7 MPa (Ruan J. K., 1993). For this case the Young's modulus used is 16.7 MPa with Poisson's ratio of 0.42, yields stress of 5 MPa (McPherson S. Beall III, 2003) and density of 1130 Kg/m^3 as a linear elastic material. Thickness of scalp varies from 5 to 7 cm, but in this study it is set to 5 cm.

Facial bone used in this thesis is not assigned to be analyzed and the only reason for updating the facial bone properties is to consider the effect of inertia caused by the mass of the facial bone. Facial bone's Young's modulus was taken from the study provided by Willinger (2002) and was set to 5000 MPa. Poisson ratio was set to 0.23 and density was 2500 Kg/m^3.

CSF has been defined as a linear elastic fluid, which concludes that as a fluid, there will be no support for shear stress, so the quantity which is considered as shear modulus is zero. The properties for CSF have been taken from Johnson Ho, 2007. A Poisson's ratio of 0.499989 set for CSF shows that it is considered as a very close to an incompressible material and density is set to 1000 Kg/m^3. As the Poisson's ratio and Young's modulus will be neglected by FE solver, Bulk modulus should be defined as followed:

$$K = \frac{E}{3(1 - 2\vartheta)} \tag{12}$$

A summary of properties used in this thesis is shown in table 3.1:

Table 3.1: Material properties of different tissues of human head in this research

Tissue	*Young's Modulus (MPa)*	*Density (Kg/m^3)*	*Poisson's Ratio*
Arachnoid	*19.32*	*1130*	*0.45*
Arachnoid Trabecular	*0.054810*	*1130*	*0.45*
Brain	*N/A*	*1040*	*0.4999982742*
Cerebrospinal fluid	*K=2.1 GPa*	*1000*	*0.499989*
Cortical bone	*8100*	*2000*	*0.22*
Trabecular bone	*880*	*1300*	*0.3*
Facial	*5000*	*2500*	*0.23*
Scalp	*16.7*	*1130*	*0.42*
Dura	*11.72*	*1140*	*0.23*
Falx	*11.72*	*1140*	*0.23*
Tentorium	*11.72*	*1140*	*0.23*
Pia	*19.32*	*1130*	*0.45*

3.3.2 Meshing

In order to achieve accurate results, it is crucial to choose the right meshing type for 3D models. A good mesh can reduce processing time and give us more realistic results in a shorter time. It is important to take meshing method, element type and element size into the consideration. An element with a big size can result inaccurate answers and an element with a very small size will increase CPU usage and processing time without resulting a big change in output data which reduces the efficiency of analysis. Other than the effect of different meshing methods, modifying the model affects the results as well. Omitting steep corners and smoothening the surface, provide better meshing, thus, more accurate and realistic results.

There are several types of elements as each of them has their specific characteristics and in each experiment, it is vital to choose the right type of element that is suitable with geometry of the model, material properties and the nature of experiment. So, in order to achieve satisfying results, the element types have to be chosen wisely.

Benzley (1995) concluded that for incompressible and close to incompressible fluids, linear hexahedral shown in Figure 3.4, or in other words, Brick elements, has the best efficiency compared to linear tetrahedral elements. Therefore, in this study these conditions have been met as much as possible. Because the current 3D FE model consists of a large number of different parts, using brick elements for in almost all of them results a very long processing time. So shell elements were used in some parts as an alternative. In Table 3.2 the information about the different parts of the model and element are given.

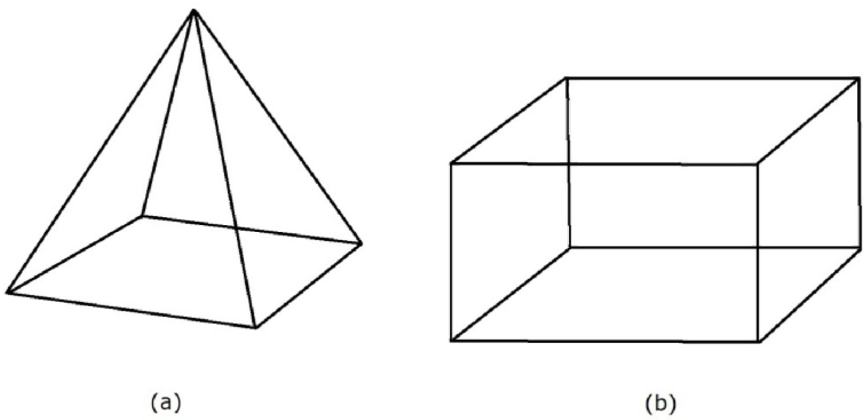

Figure 3.4: Element types: (a) linear tetrahedral and (b) linear hexahedral

In this thesis, 7 parts of the whole model were studied which are Scalp, Brain, Cerebrospinal fluid, Dura, Pia, Falx and Tentorium. All parts have been meshed by

Altair, HyperMesh 12.0 software. Detailed explanations of models with figures are given as follows.

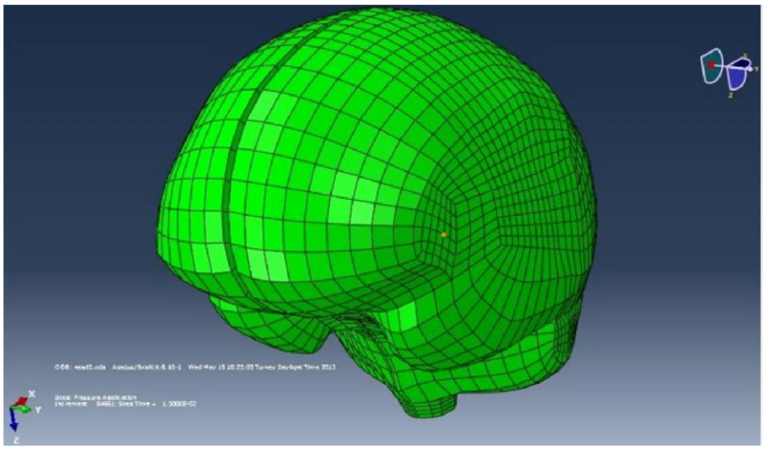

Figure 3.5: A 3D outlook of finite element model of brain

3.3.2.1 Brain

This part contains 7318 brick elements and modeled as solid and homogenous material. (Figure 3.6)

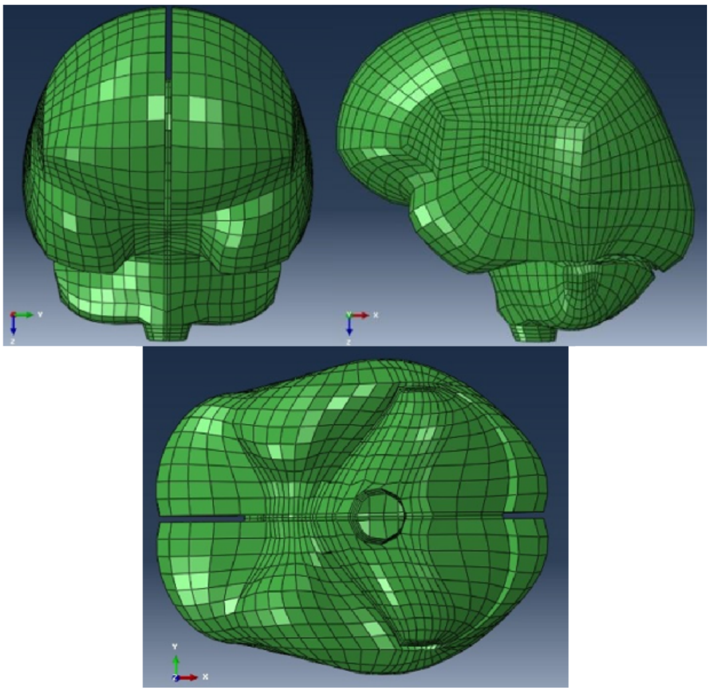

Figure 3.6: Illustration of brain finite element model from 3 views, X axis (left), Y axis (middle) and Z axis (right)

3.3.2.2 Cerebrospinal fluid (CSF)

Contains 2874 elements in two layers of homogenous brick elements. (Figure 3.7)

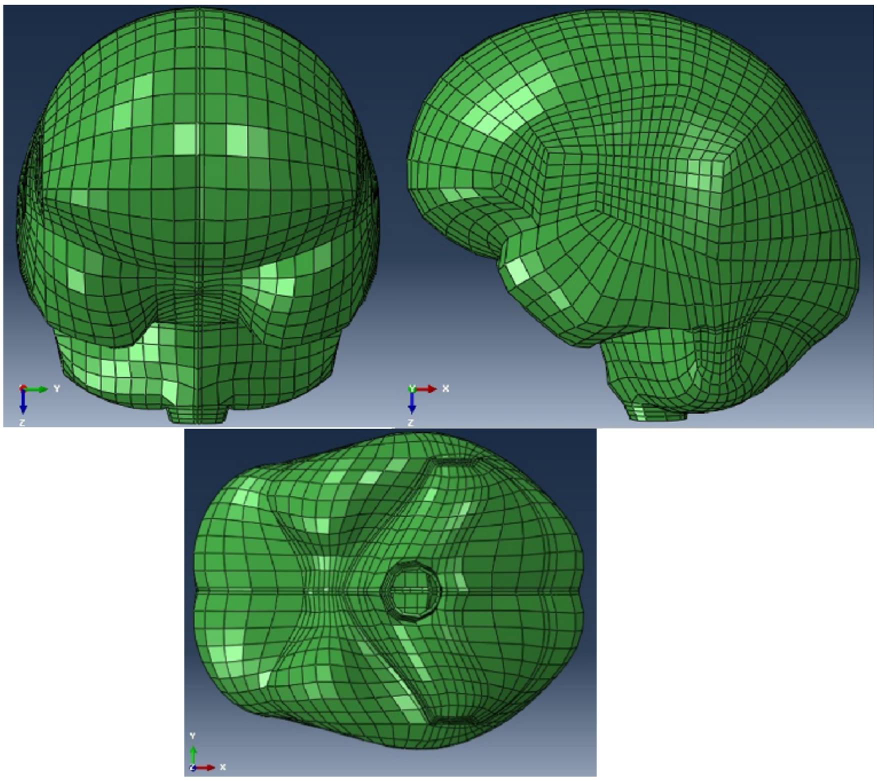

Figure 3.7: Cerebrospinal fluid shown from 3 directions, X (upper left), Y (upper right) and Z (below all)

3.3.2.3 Scalp

This tissue has been modeled by 2064 solid homogenous shell elements. Thickness has been set to 5 millimeters. (Figure 3.8)

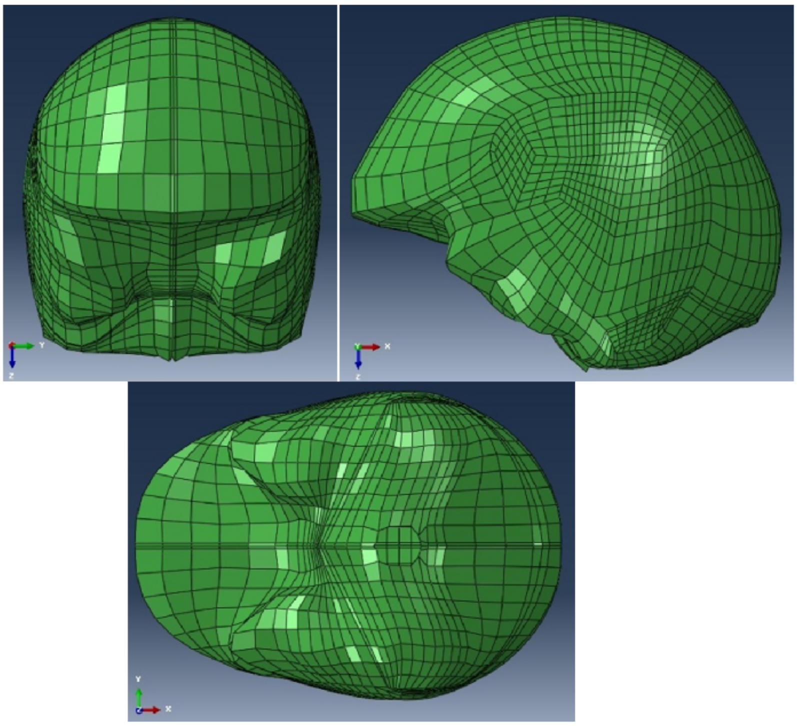

Figure 3.8: FE model of scalp from 3 viewpoints X, Y and Z

3.3.2.4 Dura

Contains 2157 shell homogenous elements. Thickness in this section has been set to 0.4 millimeters. (Figure 3.9)

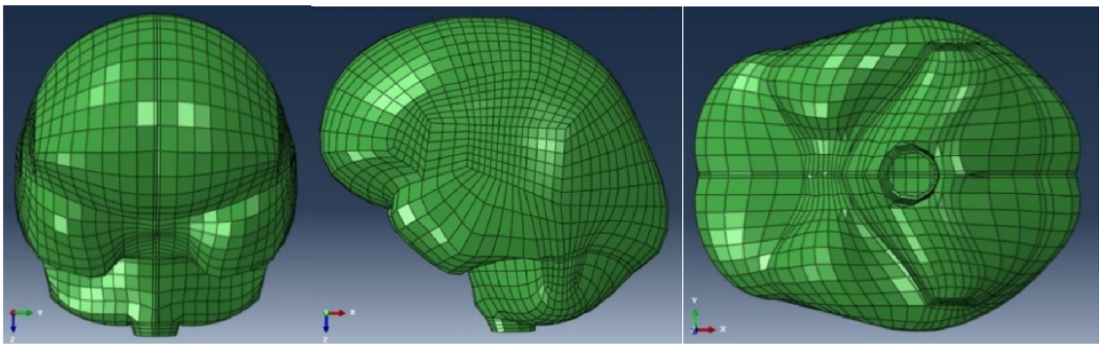

Figure 3.9: FE model of Dura

3.3.2.5 Pia

Contains 2876 shell homogenous elements and the thickness has been set to 0.15 millimeters. (Figure 3.10)

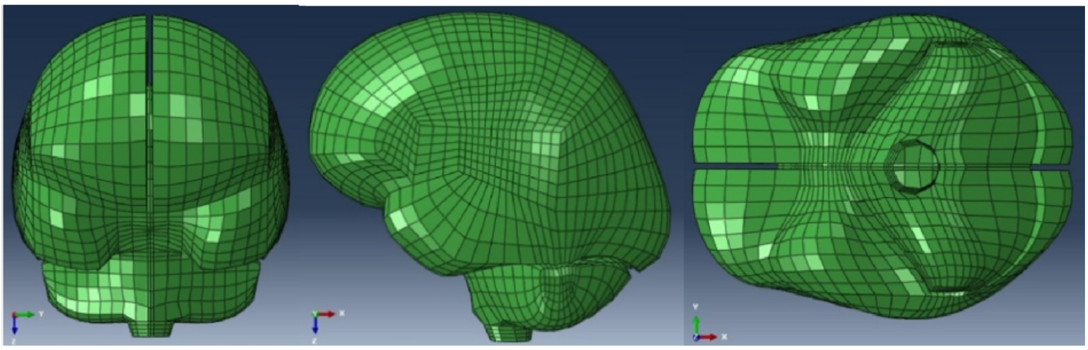

Figure 3.10: FEM of Pia

3.3.2.6 Falx and Tentorium

These sections are separated and falx contains of 235 and Tentorium contains of 230 shell homogenous elements and their thicknesses were set to 0.4 millimeters. (Figure 3.11)

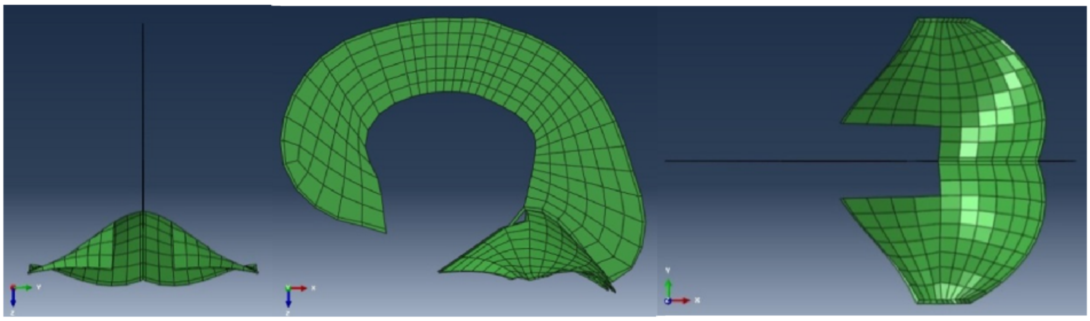

Figure 3.11: FEM figures of Falx and Tentorium shown attached together.

Because of the complexity of the model, meshing of all the parts demanded to operate refinement tool in software for several times. Steep parts of the model should have been softened in order to make the model suitable for a proper meshing. Defining and choosing between types of meshing in different parts of the model has been done according to the latest experiments. (Othman, 2009), (Brands, 1999), (Kleiven, 2002) (N.J. Mills, 2008), (V. Tinard, 2012)

3.4 Finite Element Analysis of Human Head

In the next step the effects of a frontal impact on human head and brain will be analyzed. The 3D model has been prepared and updated with the latest data for the material properties. A summary of material properties used in this thesis is given in Table 3.1 and a summary of the 3D model properties is given in Table 3.2

Table 3.2: Summary of 3D model of human head and brain

Part	**Number of Elements**	**Element Type**
Arachnoid	2030	Homogenous / Shell
Arachnoid Trabecular	3748	Beam
Brain	7318	Solid Homogenous / Brick
Cerebrospinal fluid	2874	Solid Homogenous / Brick
Cortical bone	4160	Solid Homogenous / Brick
Trabecular bone	4096	Solid Homogenous / Brick
Facial	406	Homogenous / Shell
Scalp	2064	Homogenous / Shell
Dura	2157	Homogenous / Shell
Falx	235	Homogenous / Shell
Tentorium	230	Homogenous / Shell
Pia	2786	Homogenous / Shell
Total	32104	

FEA of this model has been initiated with ANSYS 14.5. 3D body models have been prepared and imported into the software and after meshing and refinement, the analysis has been started. After a series of trial and error and illogical results, it was convincing that ANSYS is not suitable for this analysis and it is required to change the FEA software. Another well-known FE program which is used in vast areas of studies, experiments is ABAQUS. It was concluded that the next best choice after ANSYS is ABAQUS and the analyzes reinitiated in ABAQUS 6.1 with a new model adapted to this software.

As it was mentioned earlier, the conditions of impact forces has been set as 9.49 m/s impact of a 5.95 Kg impactor in 15ms and in a 45° angle. Von Mises stress and pressure distribution in brain, skull, cerebrospinal fluid, dura, falx, tentorium and pia have been calculated and the results are expected to match the results which have been obtained by (Nahum, 1977) with a good approximation.

Chapter 4

RESULTS

According to the literature survey, traumatic brain injuries are responsible for 52000 annual deaths only in United States (M. Faul, 2010). 23% of external reasons of fatalities belong to motor vehicle transportation. Among injuries causing death in motor vehicle accidents, 78% are head injuries. This indicates the vulnerability of humans head to impacts (Kleiven, 2006). In this thesis, modeling of human head is focused on frontal impacts to human's head and simulated a common impact that happens in motor vehicle accidents. Impact forces equal to a 5.95 Kg impactor with 9.49 m/s has been applied to the model during 15ms.

In Section 4.1 intracranial Von Mises stress has been visualized in 7 different layers of head: scalp, CSF, dura mater, falx, tentorium, pia and brain. In Section 4.2 the result for pressure analyzes has been shown and external and intracranial pressure variation during the impact time has been measured and illustrated for the same 7 parts of the head.

4.1 Von Mises Stress Analysis

As it is discussed in the introduction, Von Mises stress analysis has functional applications in engineering. Materials start yielding after Von Mises stress passes its critical point of yield strength, so it's a critically important factor to consider in analyzing the model. In the next sections, Van Mises stress distribution in 7 different parts of head is illustrated.

4.1.1 Stress Distribution

Von Mises stress distribution of different layers of head caused by the impact object in 0.05 seconds has been illustrated in the following Figure 4.1, Figure 4.2, Figure 4.3, Figure 4.4, Figure 4.5 and Figure 4.6.

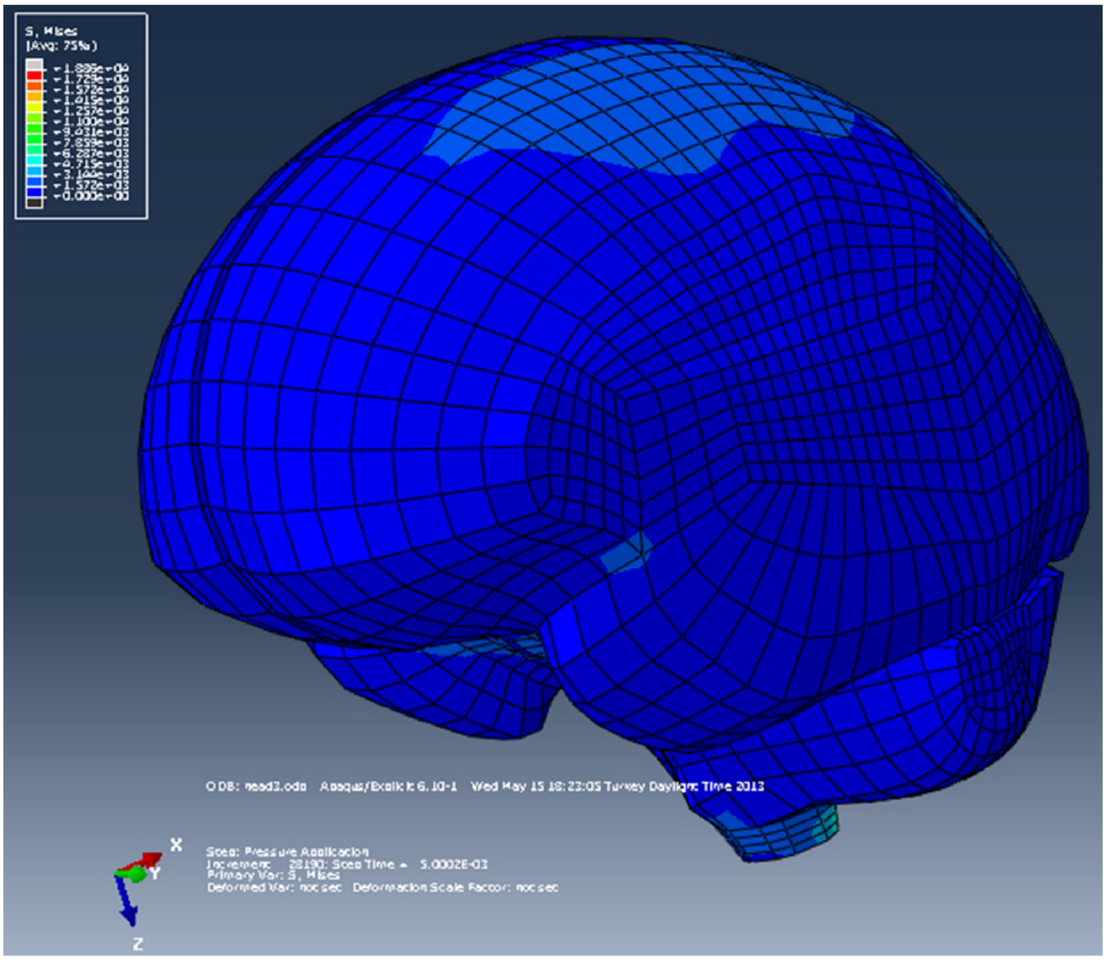

Figure 4.1: Von Mises stress distribution on viscoelastic brain

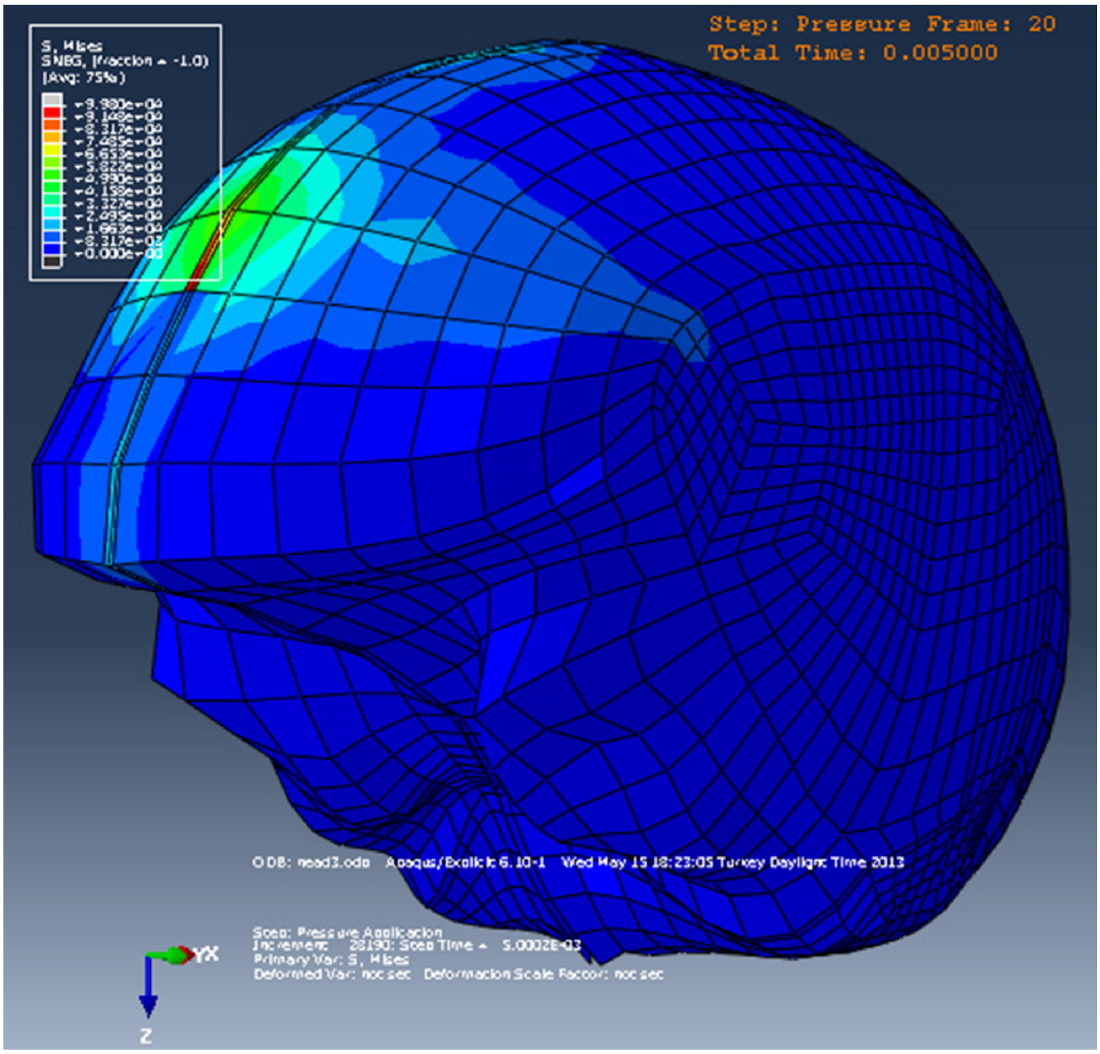

Figure 4.2: Von Mises stress distribution on Scalp

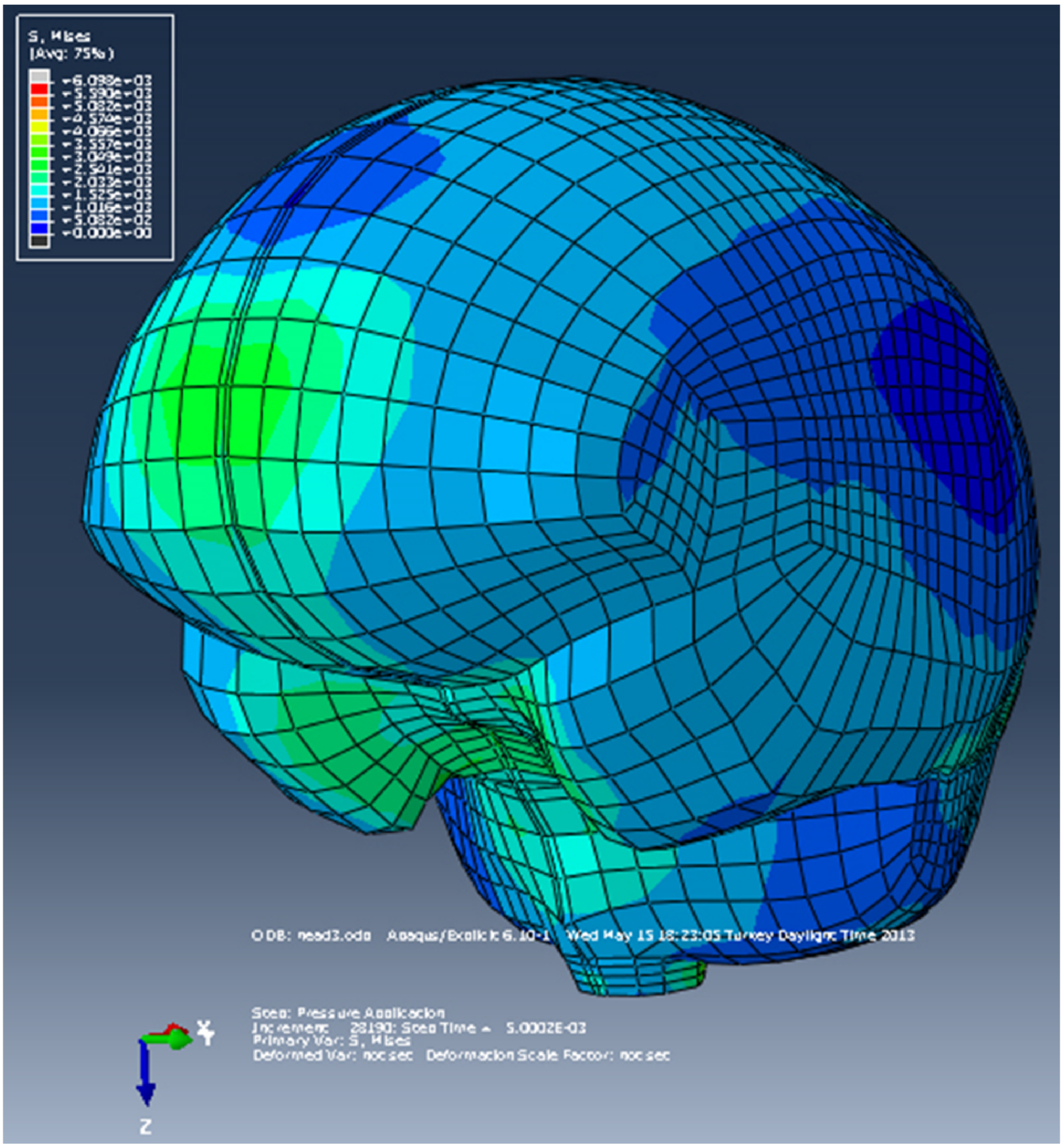

Figure 4.3: Von Mises stress distribution on CSF

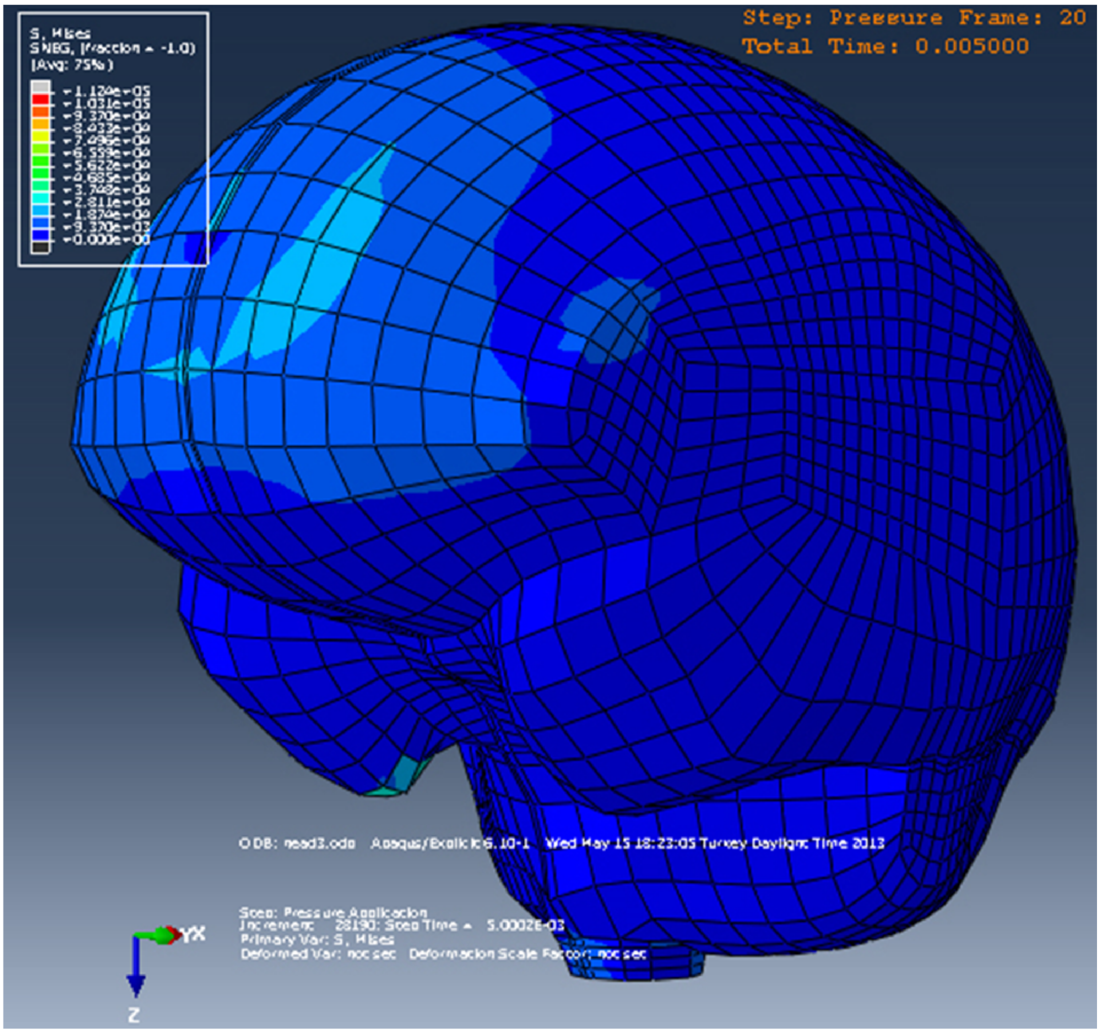

Figure 4.4: Von Mises stress distribution on Dura matter

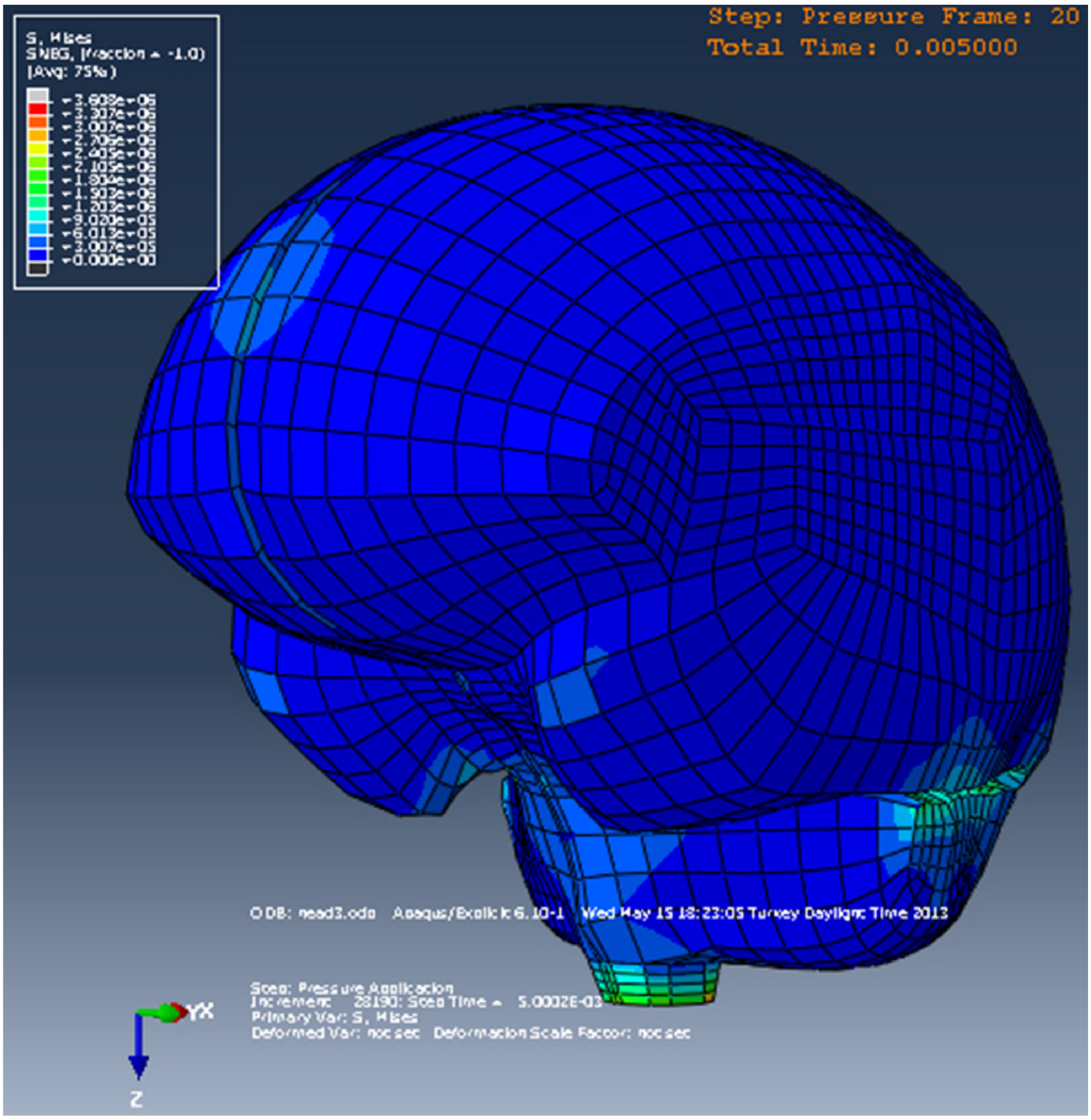

Figure 4.5: Von Mises stress distribution, Pia

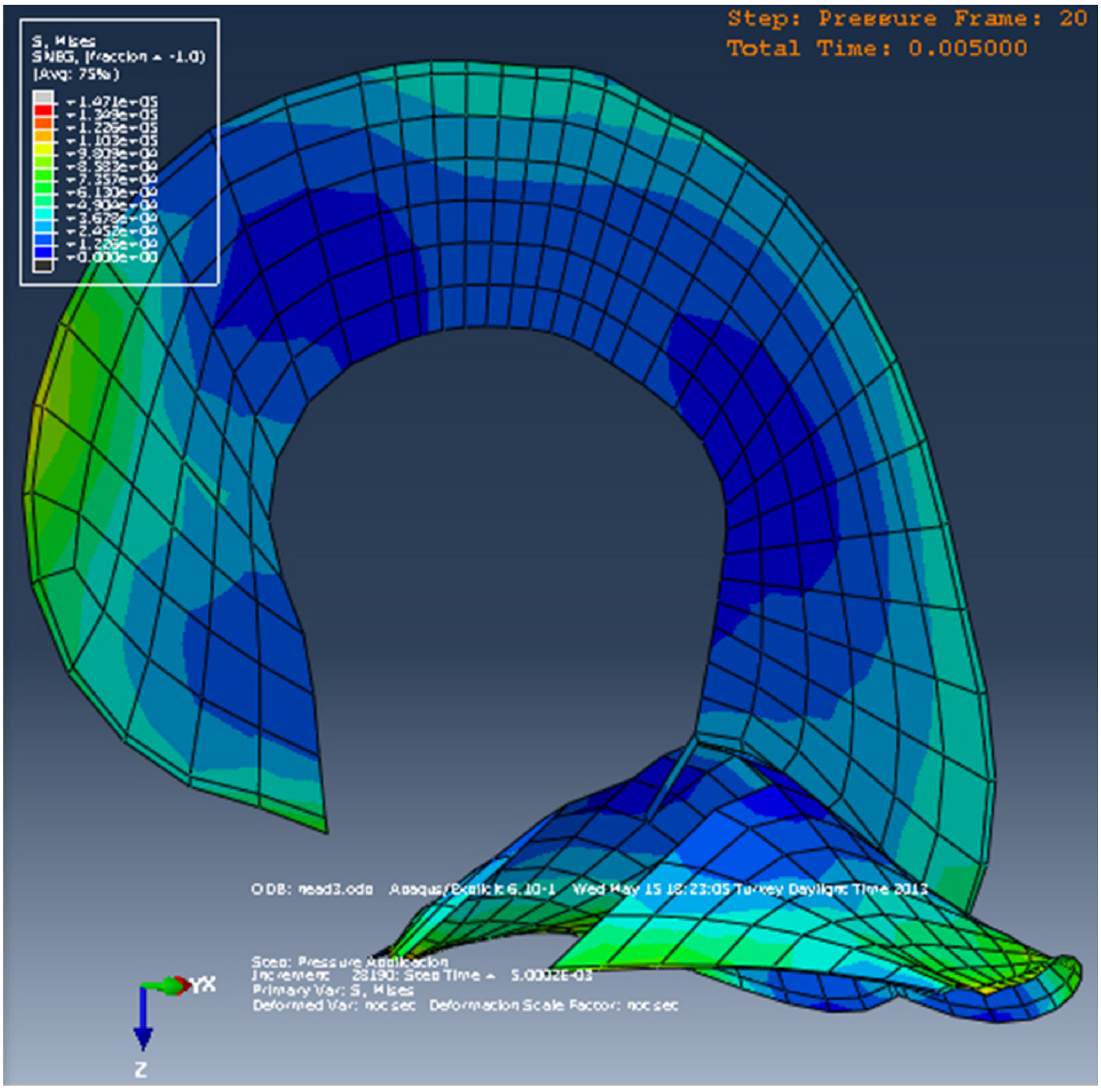

Figure 4.6: Von Mises stress distribution, Falx and Tentorium

Time histories of Von Mises stress is shown in the following figures to illustrate the changes during the impact and illustrate the maximum stress applied on different layers of the head model.

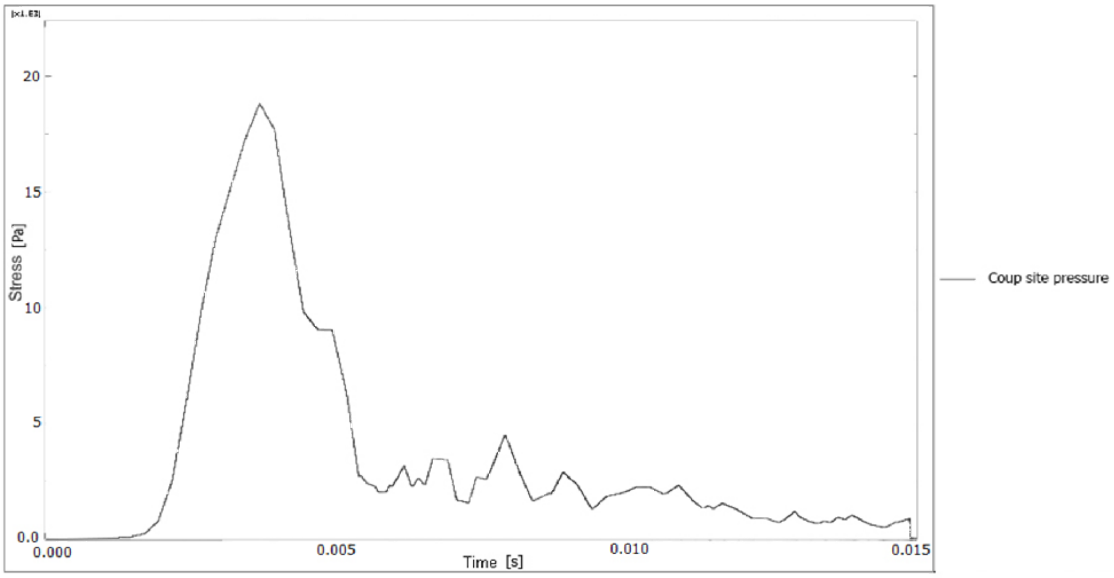

Figure 4.7: Von Mises stress time histories acquired from model in viscoelastic brain

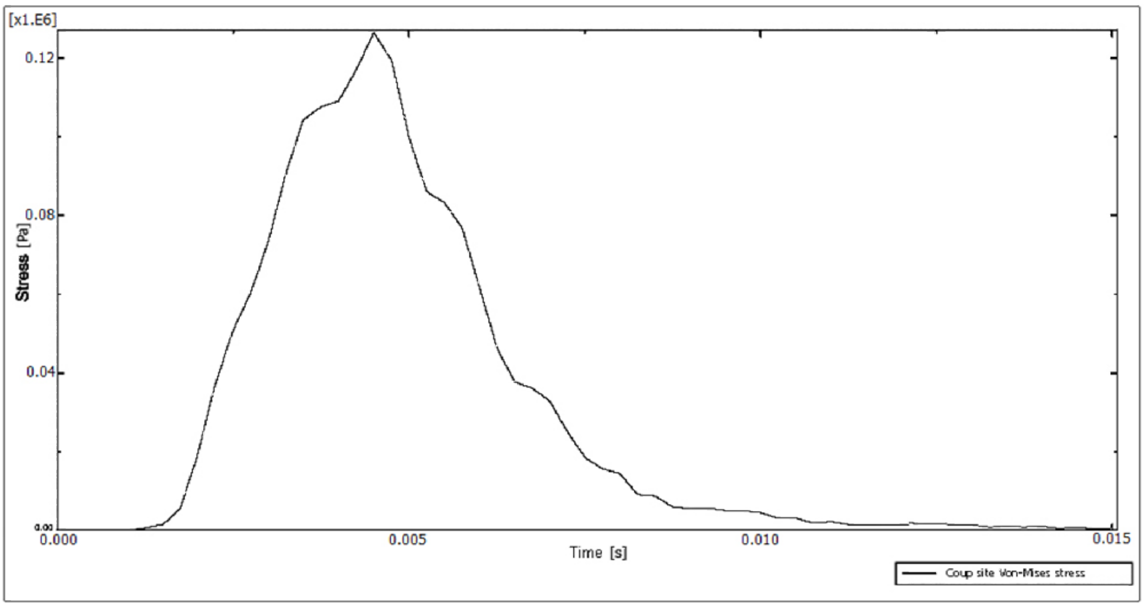

Figure 4.8: Von Mises stress time histories acquired from model in scalp

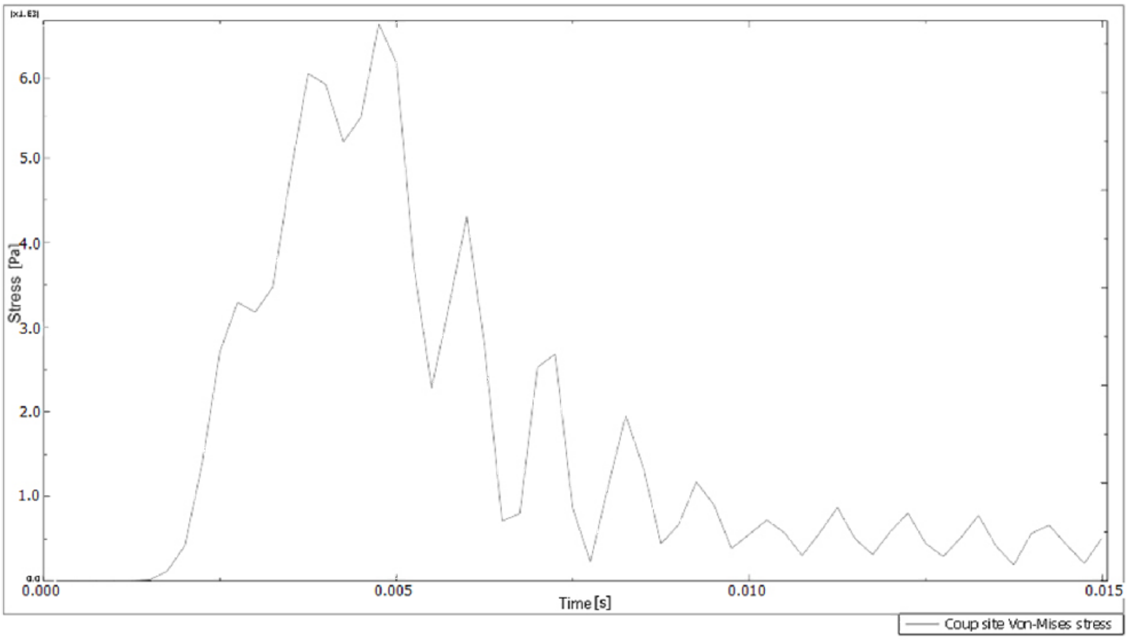

Figure 4.9: Von Mises stress time histories acquired from model in CSF

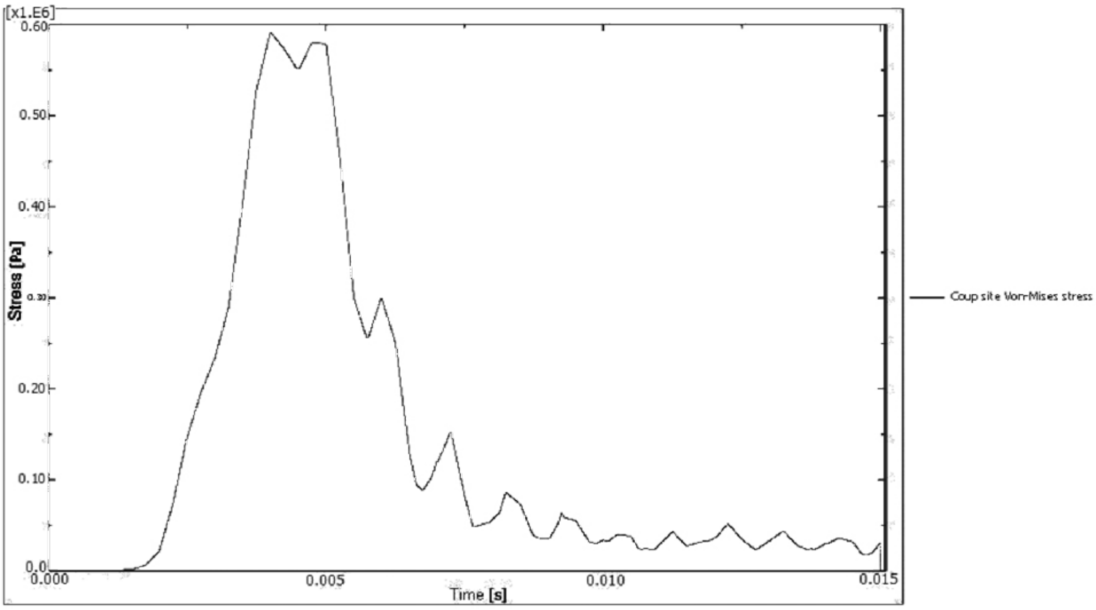

Figure 4.10: Von Mises stress time histories acquired from model in Dura mater

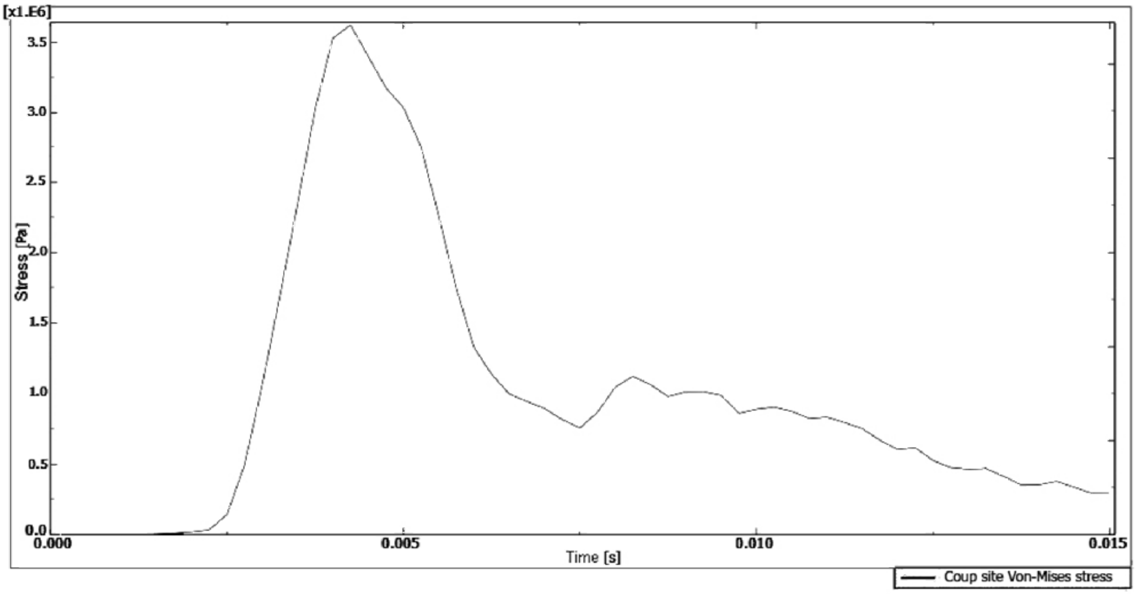

Figure 4.11: Von Mises stress time histories acquired from model in Pia

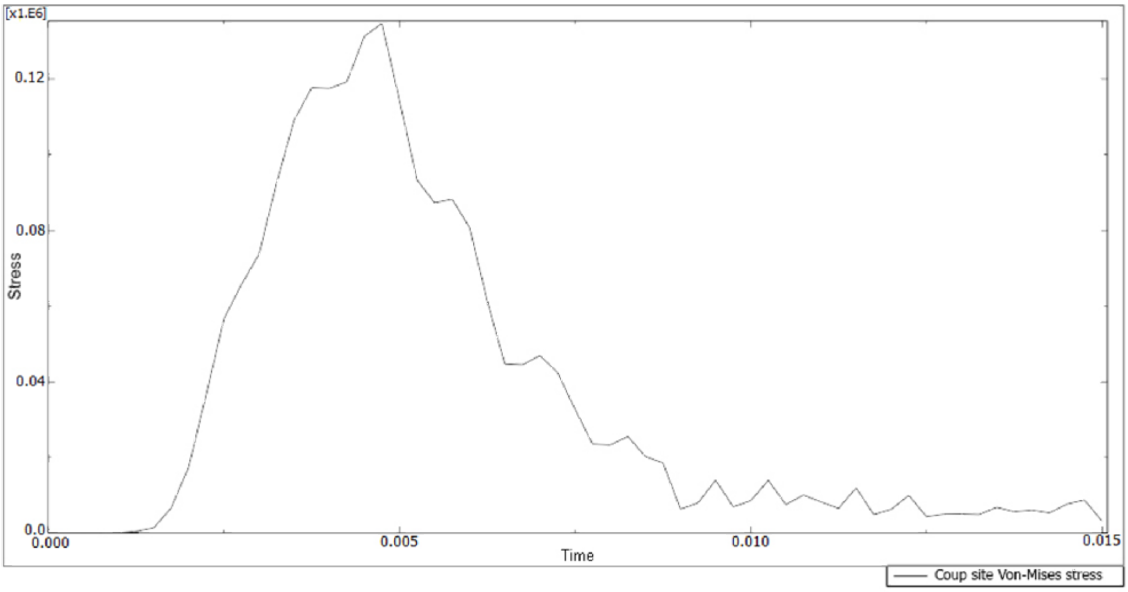

Figure 4.12: Von Mises stress time histories acquired from model in Falx

4.2 Pressure Analysis

The analysis of pressure has been performed during 15ms of impact. According to Nahum (1977) pressure analysis can be used to obtain the necessary data to predict impact collision damages. The results are explained in the following sections.

4.2.1 Pressure distribution

Depending on the geometry and material properties of layers, pressure analysis of human head develops the desired results. Subsequently, the retrieved results for pressure analysis of a frontal impact to the head model at 0.05s are presented with Figure 4.13, Figure 4.14, Figure 4.15, Figure 4.16, Figure 4.17 and Figure 4.18.

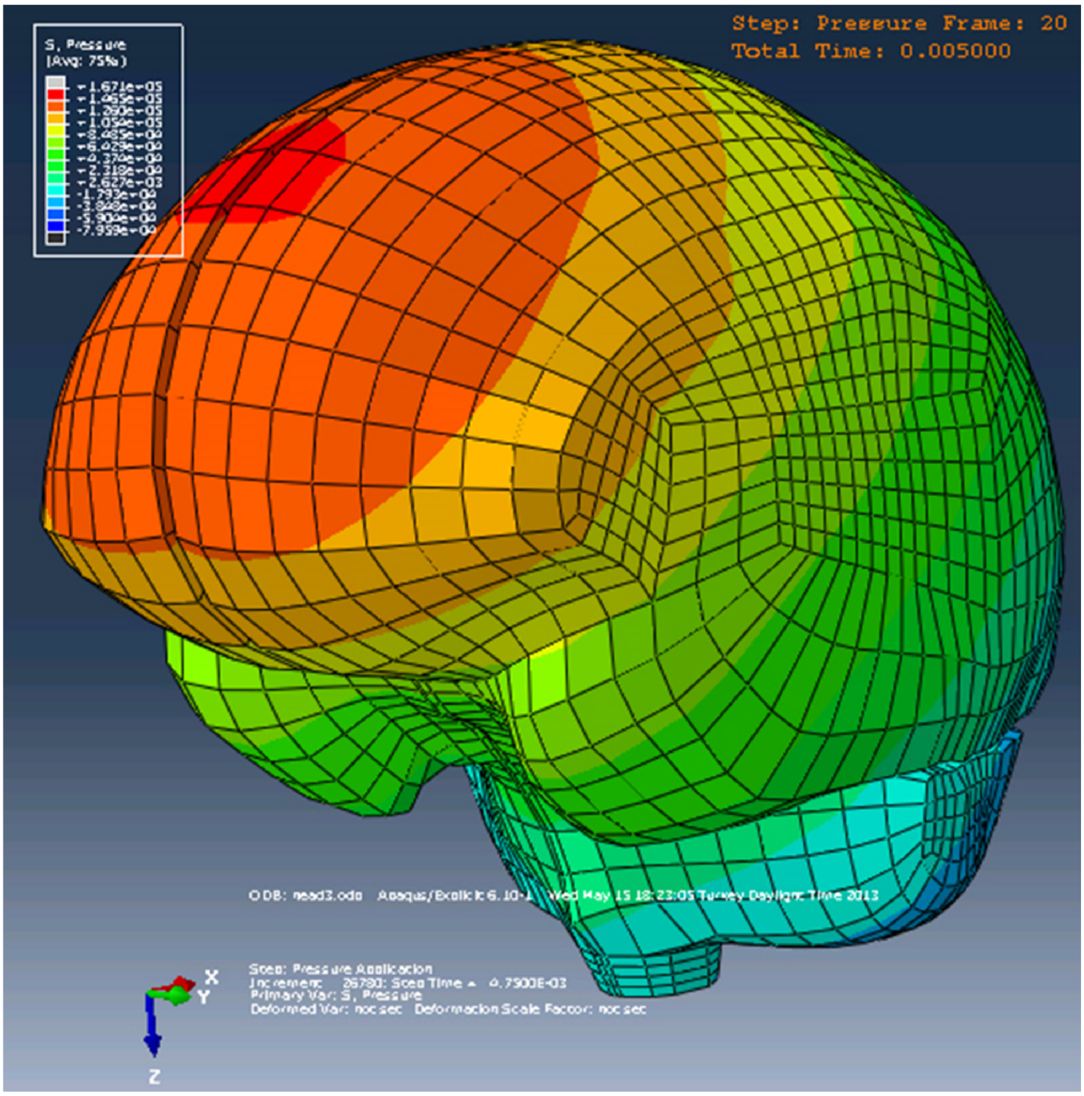

Figure 4.13: illustration of pressure analysis results at 0.05s on Brain

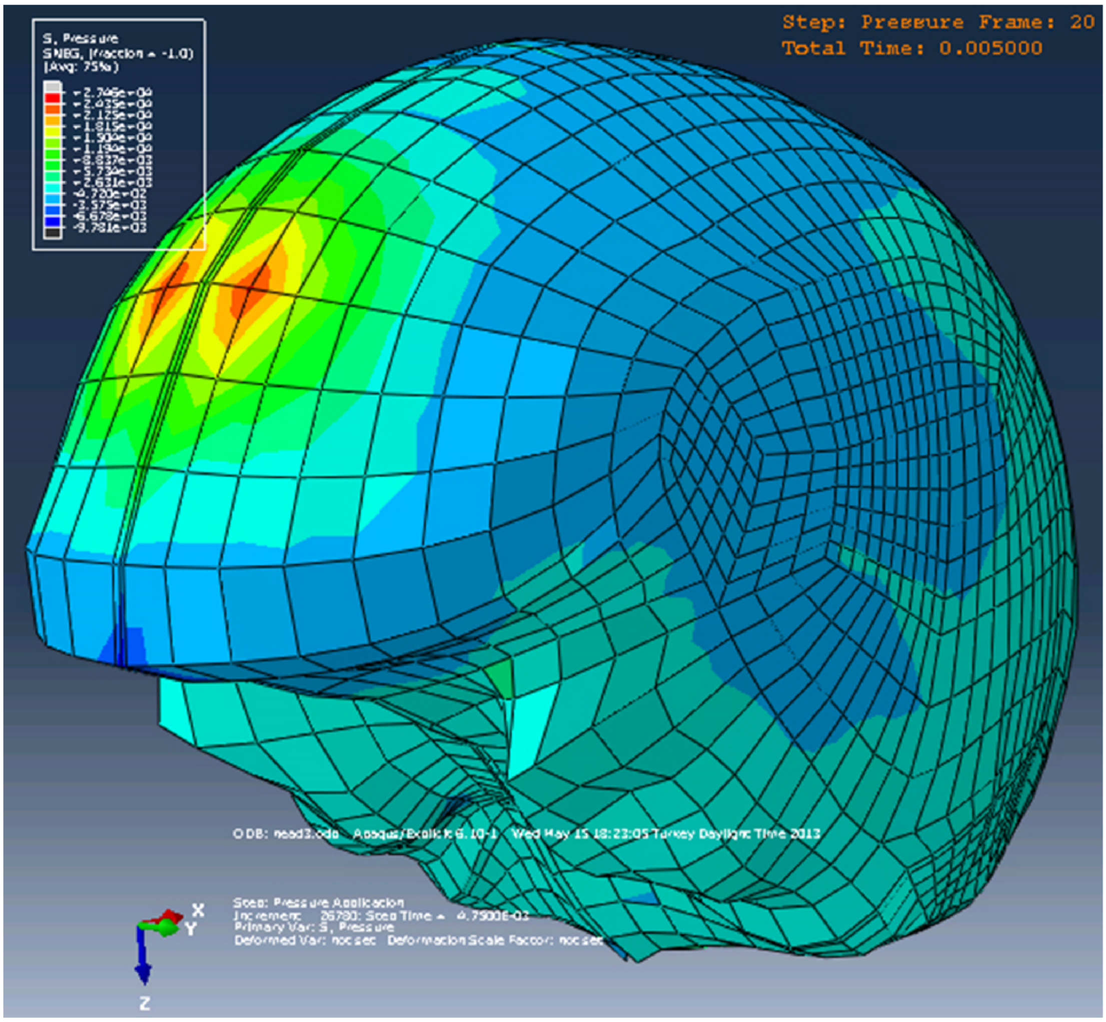

Figure 4.14: illustration of pressure analysis results at 0.05s on Scalp

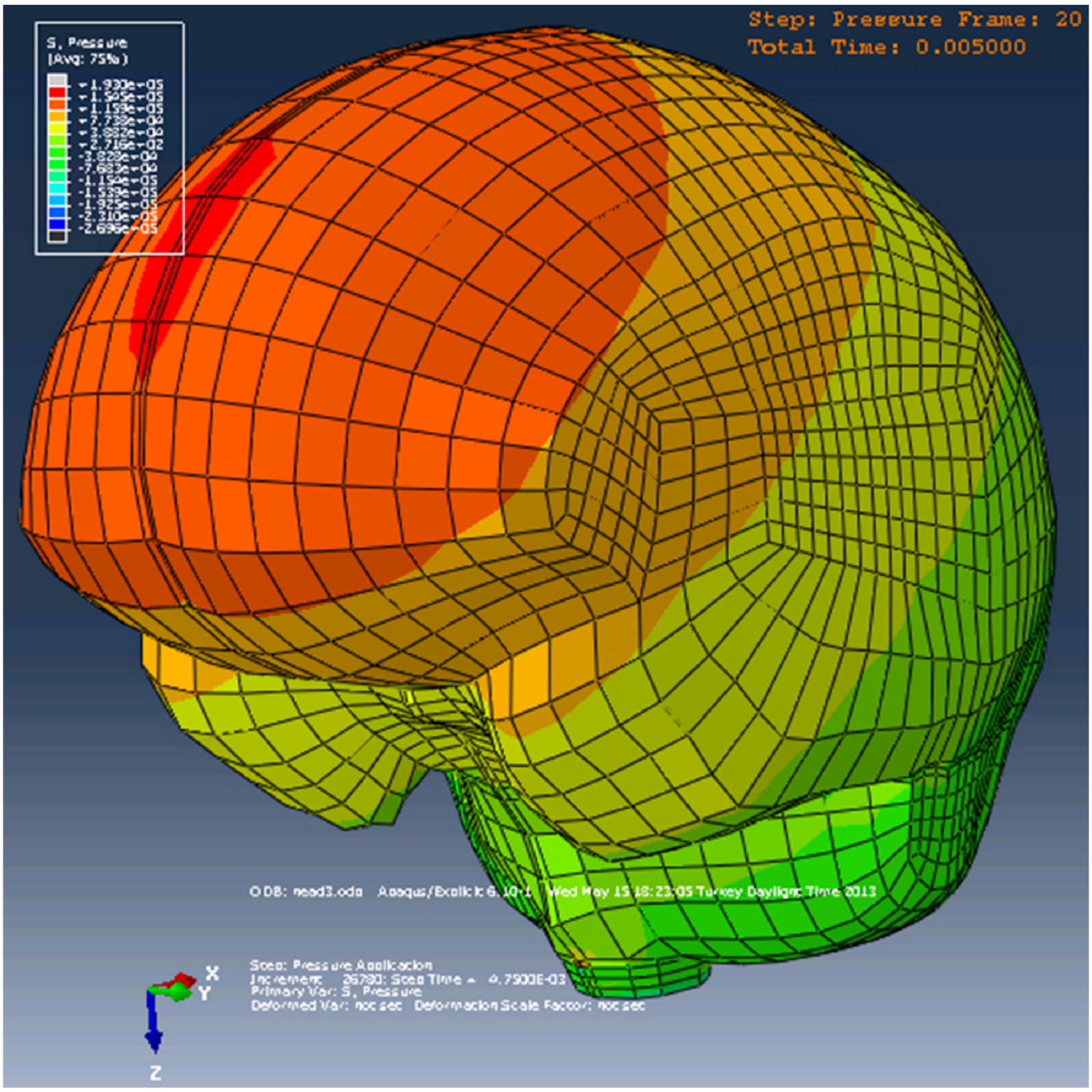

Figure 4.15: illustration of pressure analysis results at 0.05s on CSF

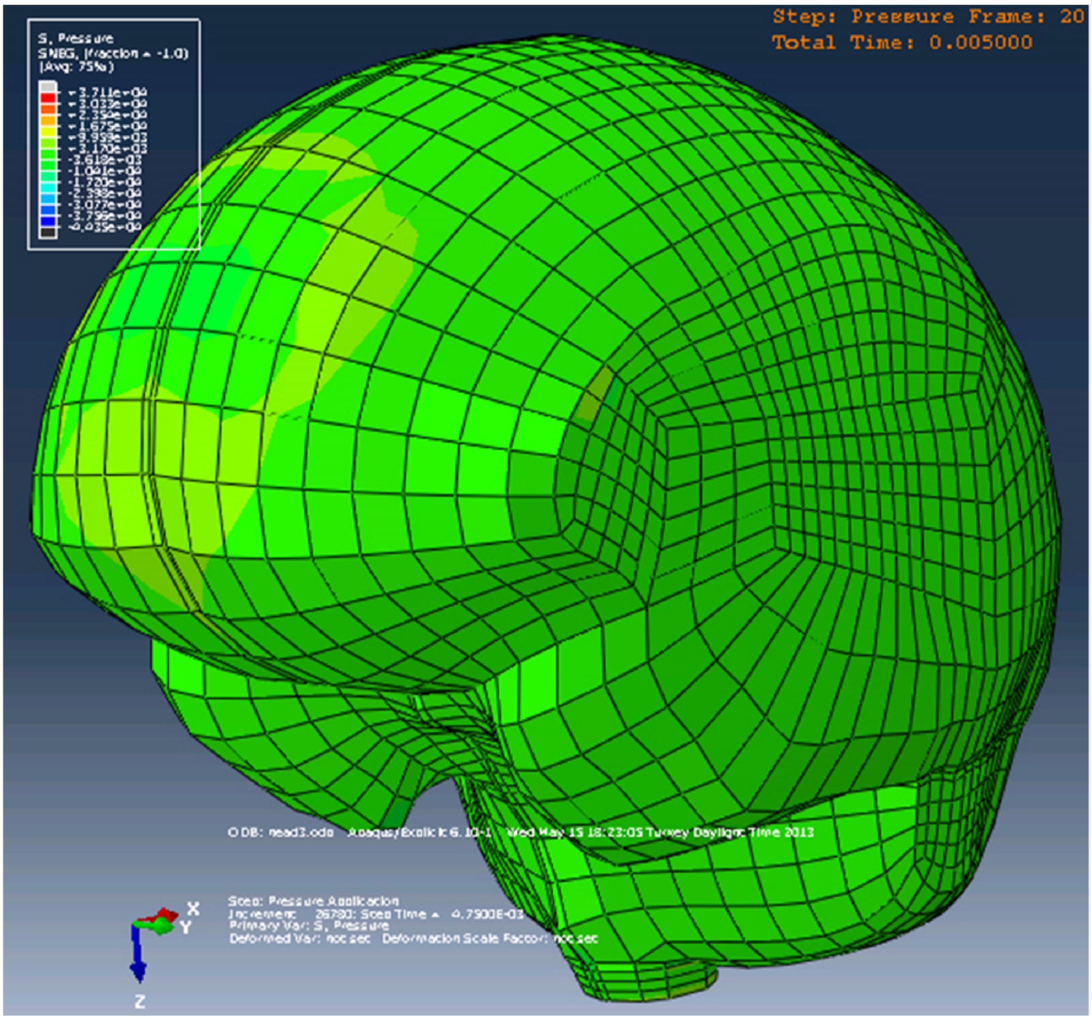

Figure 4.16: illustration of pressure analysis results at 0.05s on Dura

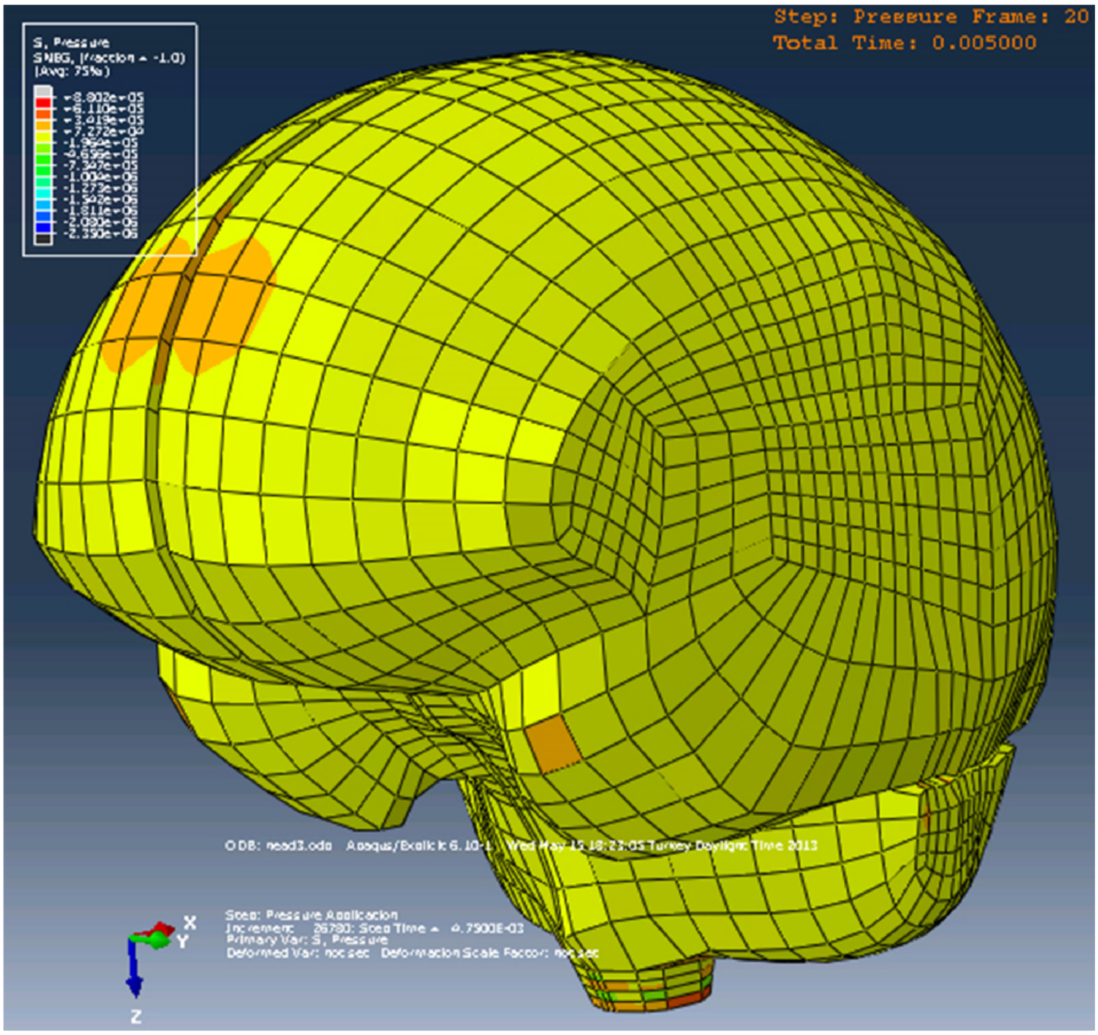

Figure 4.17: illustration of pressure analysis results at 0.05s on Pia

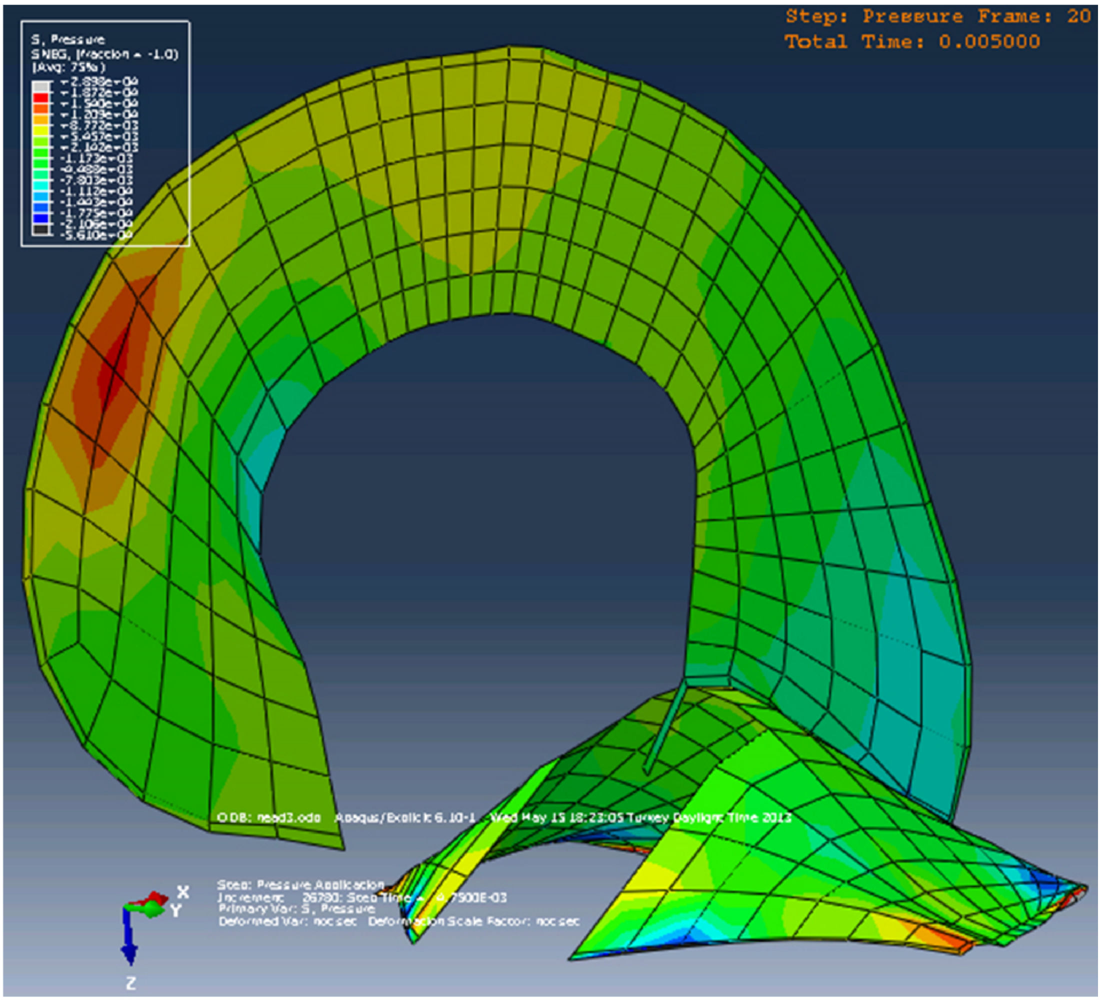

Figure 4.18: Illustration of pressure analysis results at 0.05s on Falx and Tentorium

Time histories of maximum and minimum pressure are drawn in the following Figure 19, Figure 20, Figure 21, Figure 22, Figure 23 and Figure 24 to illustrate the changes during the impact and draw the maximum stress applied on different layers of the model. These diagrams have been plotted for 2 points named coup and countercoup point. Coup point is coincident on the impact point and the countercoup point is located on the opposite side of the head behind the center of mass. The locations of these points are the same as the points that Nahum (1977) used in his experiment and thus, can be used to compare the acquired results with experimental results.

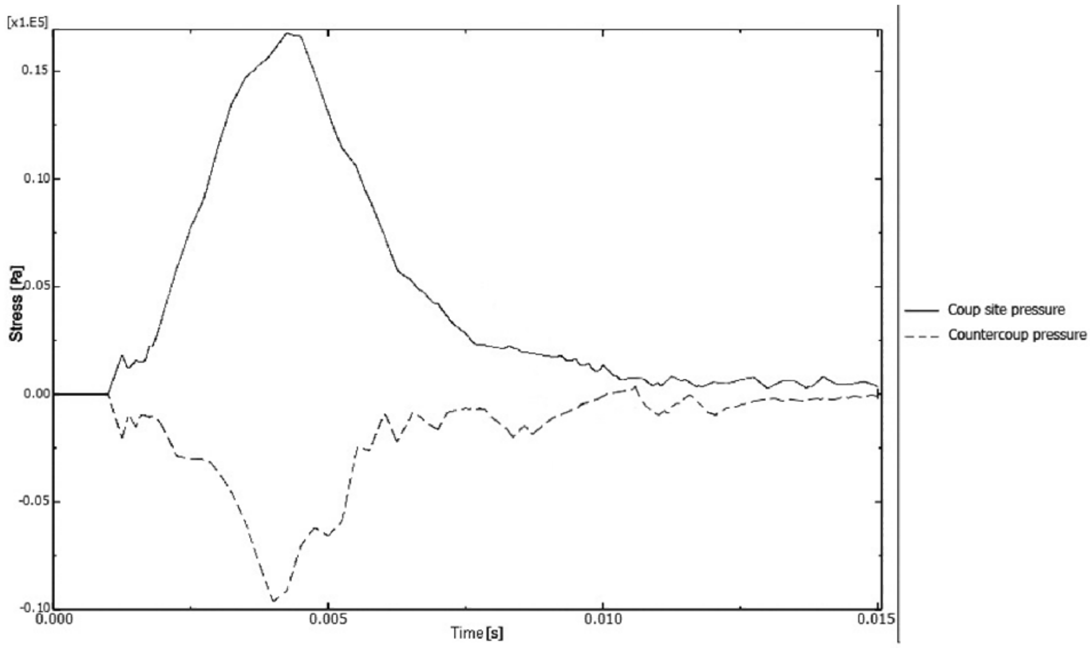

Figure 4.19: Maximum and minimum pressure time histories acquired from model in Brain

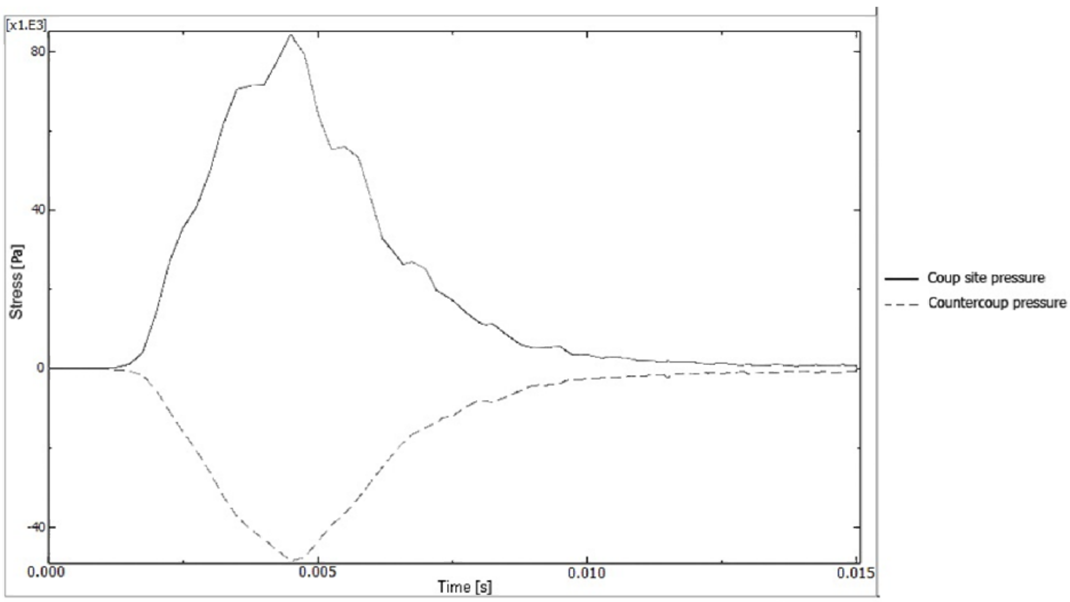

Figure 4.20: Maximum and minimum pressure time histories acquired from model in Scalp

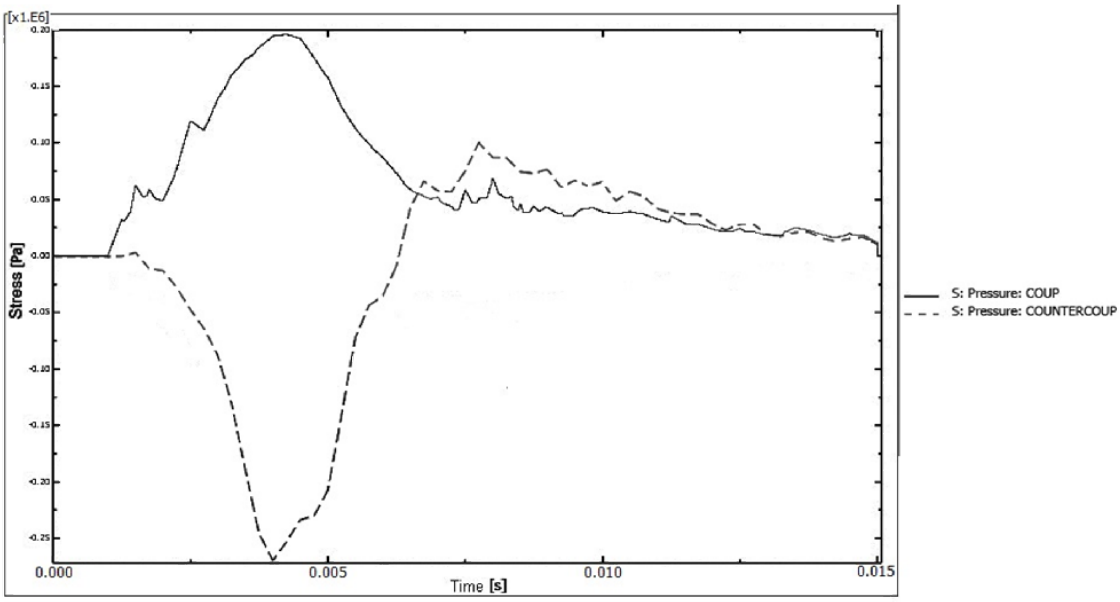

Figure 4.21: Maximum and minimum pressure time histories acquired from model in CSF

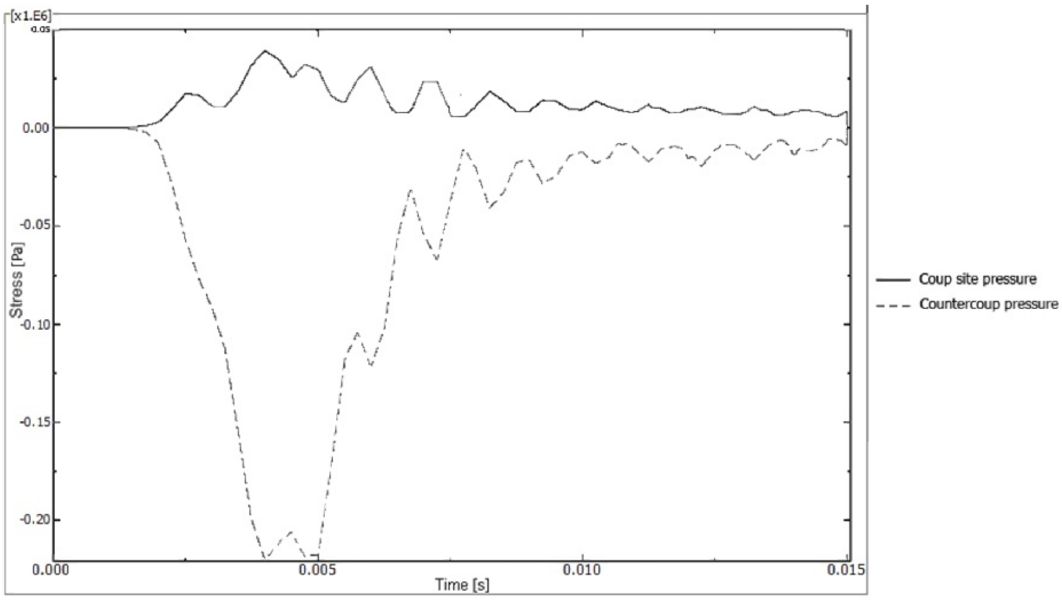

Figure 4.22: Maximum and minimum pressure time histories acquired from model in Dura

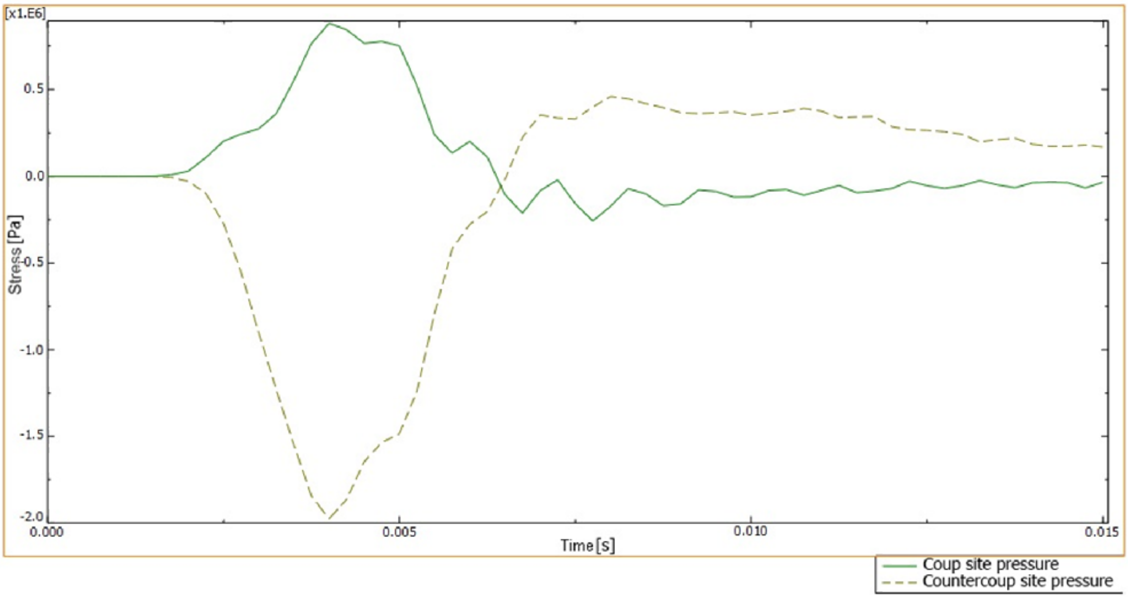

Figure 4.23: Maximum and minimum pressure time histories acquired from model in Pia

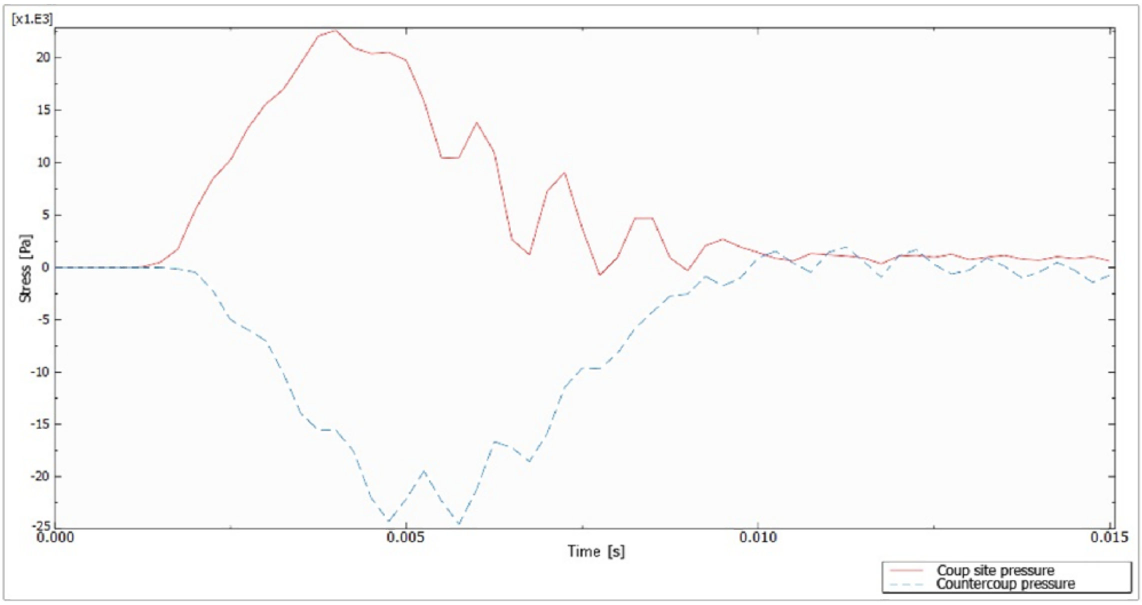

Figure 4.24: Maximum and minimum pressure time histories acquired from model in Falx

A summary of maximum and minimum values of Von-Mises stress and Pressure analysis are given in Table 4.1 and Table 4.2.

Table 4.1: Summary of Maximum and minimum Von-Mises stress data

Parts of head	**Maximum Von-Mises Stress for coup site (Pa)**
Viscoelastic brain	1.886×e4
CSF	6.098×e3
Scalp	9.980×e4
Dura	1.124×e5
Pia	3.612×e6
Falx	1.335×e5
Tentorium	1.471×e5

Table 4.2: Summary of maximum and minimum pressure data at coup and countercoup sites

Parts of head	Maximum pressure for coup site (Pa)	Maximum pressure for countercoup site (pa)
Viscoelastic brain	1.671×e5	9.153×e4
CSF	1.930×e5	2.741×e5
Scalp	8.244×e4	4.788×e4
Dura	3.711×e4	2.249×e5
Pia	8.802×e5	2.348×e6
Falx	1.872×e4	2.471×e4

Time histories of pressure changes in coup and countercoup sites and the results from Nahum's experiment are drawn in Figure 4.25 and Figure 4.26 to illustrate the differences between experimental and analytically analyzed results.

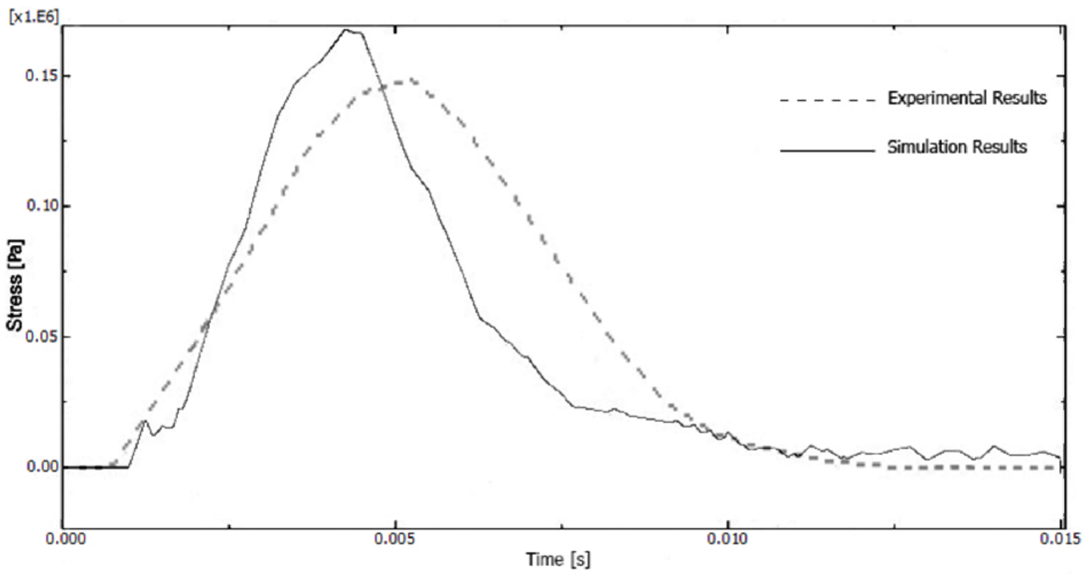

Figure 4.25: Comparison of Nahum's experimental results and currents studies analytical results for intracranial pressure at coup site

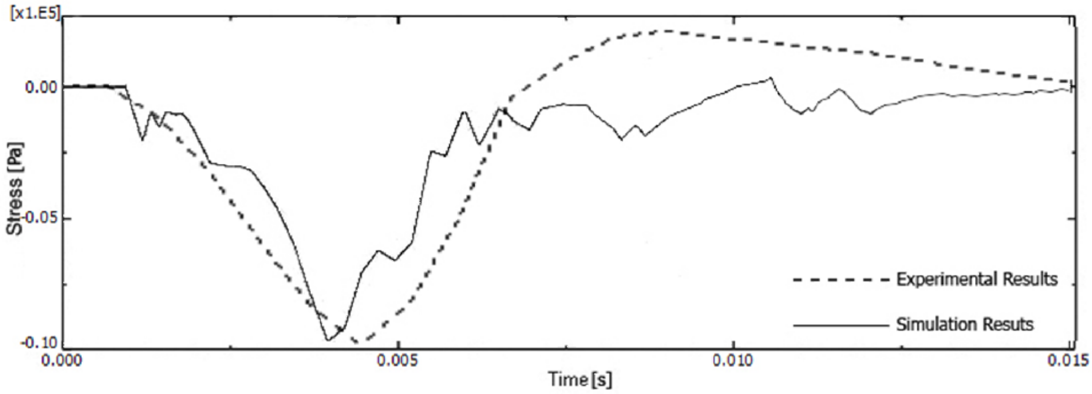

Figure 4.26: Comparison of Nahum's experimental results and currents studies analytical results for intracranial pressure at countercoup site

In the next chapter, the obtained results will be discussed and the FE model will be evaluated by comparing to experimentally analyzed data.

Chapter 5

DISCUSSION AND CONCLUSION

5.1 Discussion

From the literature review, it is understood that the number of head injuries, brain traumas and fatalities are much above an acceptable rate. Therefore, it is necessary to take action to prevent these injuries. The first step can be finding a better understanding of the brain and head trauma by performing experimental and analytical tests. Designing FE model can predict the head and brain trauma due to various impacts.

In this thesis, a modified 3D model from Polytechnic University of Tehran (M. Ghaffari, 2012), model has been used. FEM makes it possible to anticipate the stress distribution and the likelihood of injury in diverse intensities and impact angles by comparing the amount of internal pressure and stress enforced by the impact to injury criteria.

To simplify the model and avoid unnecessary complexities, some constraints have been imposed. These include the following cases:

- According to Ruan (1994) in short impacts, neck constraint has negligible effects on the results. Therefore the effect of neck was neglected in the current analysis and an open boundary condition has been assumed for the model to reduce the volume of calculations and reduce the processing time.

- The number of meshes has been limited to a certain level to prevent a long and inefficient processing time.
- The current FE model was not modeled parametrically. Since the human head size differs in a huge range, a parametrically designed head model will give enough flexibility to analyze a large range of size and mass. Heads in different sizes and weights, with different stimulus impacts in different directions can be modeled easily in a parametrically designed model.
- In this model, materials were mostly assumed to be continuum isotropic, homogeneous, linear viscoelastic, linear elastic or were designed as shell elements.
- The pressure data from the literature (Nahum, 1977) were used to validate FE models to predict injury levels. But according to Bradshaw (2001), pressure is not a suitable parameter to predict injury and strain is a more reliable parameter to measure and predict injury. However, such data does not exist, and in the present thesis, the analysis is more focused on pressure and Von-Mises stress to be compared with Nahum's experiment.

Von Mises stress

According to the Figure 4.7, the maximum total Von Mises stress occurred on the brain at 0.00375 seconds after the impact applied. Therefore, at this time, it was more likely that the brain started yielding. However, maximum total Von-Mises stress value (1.886×E4) remained below yielding point.

Figure 4.8 shows the maximum total Von-Mises stress of scalp which was 1.235×E5 Pa occurred at 0.0045 seconds. It can be seen from Figure 4.10 and Figure 4.11 that the peak point of stress for dura mater and pia was 5.886×E5 Pa and 3.612×E6 Pa in 0.00425 seconds. This value for falx was obtained at 0.00475 seconds for the amount of 1.378×E5 Pa (Figure 4.12). The obtained values indicate that the deformations occurred on the head showed that materials are all in elastic range and maximum total stresses were all below yielding point. In Figure 4.9, time histories of total Von-Mises stress were shown. However, plastic deformation/yielding is not defined for fluids.

Pressure

Definition on pressure is the ratio of the force over surface area. According to the Figures 4.12, 4.13, 4.14, 4.15, 4.16, 4.16, 4.17 and 4.18, the concentration of impact force was mostly coincided on impact point. From Figure 4.19, it was be understood that the maximum pressure in coup site occurred at 0.0045 seconds .It can also be seen that the peak point of pressure diagram for countercoup site observed at 0.004 seconds. Therefore, it can be said that, brain traumas are likely to happen at 0.0045 seconds on coup, and at 0.004 seconds on countercoup site. The maximum pressure on coup site of brain was about 1.671×E5 Pa and on the countercoup site was around -9.153×E4 Pa.

According to the Figure 4.20, the maximum and minimum pressure for scalp was obtained at 0.0045 seconds. The amount of pressure was 8.446×E4 Pa for coup and -4.840× Pa for countercoup site.

In CSF, as it is shown in Figure 4.21, the maximum pressure coup site was 1.955×E4 at 0.00425 seconds, and the minimum pressure of countercoup site was -2.696×E5 at 0.004 seconds.

CSF, dura mater and pia had almost identical behavior in their time-pressure diagram shiwn in Figure 4.20, 4.21 and 4.22. Dura had a maximum pressure of 3.895×E4 at 0.004 seconds and the minimum pressure of -2.196×e^4 at 0.004 seconds. Pia had the maximum pressure of 8.802×E5 at coup site at 0.004 seconds and the minimum pressure of -2.350×E6 Pa at countercoup point at 0.00425 seconds.

According to the Figure 4.24 the maximum pressure in coup site was 2.202×E4 and minimum pressure applied to countercoup site was -2.469×E4 at 0.00575 seconds.

Summary of obtained data for Von-Mises stress and pressure are given in table 4.25 and 4.26.

Literature review indicated that in 1977, Nahum performed an experimental study on the effects of impact collision on human head intracranial pressure. In this study, the stimulus properties were identical to Nahum's experiment number 42, thus, the results were validated with the available data from Nahum's experiment.

From the Figure 5.1, there was a very good correlation between experimental and analytical results. Maximum pressure in the analytical method was higher and occurred at sooner than experimental method in the coup site. This can be the result of the material. In the FE model, the materials were modeled as isotropic linear elastic and linear viscoelastic materials. Moreover, in FE modeling, terms of impacts were performed in perfect circumstances, but in practice, performing experiments like this are always coupled by errors. Another reason for this is neglecting the presence of neck and its effects.

The results from the Nahum's experiment and the FE model in countercoup site were compared in Figure 4.26 and the harmony between them was clearly observed. Just like coup site, there was a slight prematurity in analytical results. This can also be a result of excluding the effects of neck in the analysis, as well as, considering the materials as isotropic linear elastic and linear.

5.2 Conclusion and future work

In this study, a 3D FE human head model was developed to examine the effects of an accelerated stimulus on human's head. The FE model has been imported in ABAQUS software and intracranial pressure and Von-Mises stress for various parts of human's head were extracted and compared.

From observations of Von-Mises stress, it was seen that the peak points were below the critical values. Therefore, it was concluded that none of examined parts cross critical values and all deformations ware in elastic range. From the extracted results of pressure analysis, it was understood that in almost all sections maximum pressure occurred between 0.004 and 0.00475 seconds which leads us to conclude that in this case study, brain traumas are most likely to happen between 0.004 and 0.00475 seconds. The comparison of pressure between experimental and FE based results illustrated a good correlation in the range and behavior of pressure time historian diagrams (Figure 4.25 and Figure 4.26). This evaluation emphasizes the good accuracy of the current FE model.

By getting benefit from current findings, designers can develop new and more sophisticated systems in automotive to prevent TBI in accidents. As discussed before, TBI are more likely to happen between 0.004 and 0.00475 seconds after start of the impact which is a very short time. By designing an electro-mechanical system that identifies the threat and triggers the safety mechanism of the vehicles before the impact, can reduce the probability of TBI significantly.

It was shown that in all of the layers of the head, the pressure distribution is centralized on coup site. New designs for helmets should be able to diffuse the concentration of pressure and reduce the amount of applied impact force on the coup site.

For the future work, is strongly advised to perform the similar analysis with an additional helmet on the model and compare the results with experiments. With current knowledge and computer technology, it is challenging to create and model a more detailed model which considers the real properties of human's head and nonlinearity of materials. It would be advantageous to add a new section to represent neck in the model as well.

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