Development of Location Specific Finite Element Head Model for the Study of Damage Progression in Head Impact Injury

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Abstract

Background: Brain injuries are a primary health and a pecuniary issue all through the world. Numerical techniques like finite element (FE) methods may be used to investigate head injuries and optimize the safety, which can reduce the number of injuries. The FE head models were at first assessed for biofidelity by comparing with cadavers experiments. In any case, there are a few constraints in analyses as the body starts degrading after death. Human head FE models are mainly used for dynamic studies by creating scenarios of car crash, pedestrian & vehicle accidents in which different assumptions were considered.

Objective: The aim of the research is the development of realistic FE head model to study damage progression in head impact injury under external loading, sensitivity of the head impact injury to the elastic modulus and homogenized brain modeling in response to quasi-static loading at different locations.

Methodology: The FE model of the human head was used to study Von mises stress and displacement during location specific impact of the head. The human head FE model was divided into two models, one was simple and other was complex model. Both models consist of scalp, skull, spongiosa, cerebrospinal fluid (CSF), brain gray matter and white matter. These two models were then tested in three different cases with identical boundary conditions, forces and locations. Case I: Sensitivity of the injury to the elastic modulus of the brain layer by keeping all the other layers linear elastic with constant applied force to frontal region. Case II: Force Displacement Study i.e. by varying amount of force on homogenized brain model with frontal impact. Case III: Constant force was applied to homogenized brain model by varying locations of impact. Displacement and stress predicted from these models are then observed and analyzed.

Results: Preliminary outcomes of these simulations show that the brain injury may occur under applied conditions for simple model and is sensitive to the complexity of geometry. The stress and displacement profiles showed lower values for the complex model than simple model. Both the models showed linear relationship between force and displacement. By varying location, the maximum stress varied and was found maximum when the force was applied from the lateral side. So it was found that lateral impacts are more injurious and prone towards brain injury. Thus a complex model is more accurate and showed no injury in all cases in given amount of force while the simple model showed injury in two cases with the same applied conditions.

Conclusion: This study would be beneficial in order to better understand the brain response during head impact injury at different locations. It would serve as a guide line for non linear dynamic study of the head injury to the researchers in optimizing head protection and clinicians in terms of providing aid to the injured.

Keywords: Finite Element (FE) head model, head impact injury, quasi-static loading, sensitivity, elastic modulus, homogenized brain model.

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List of Abbreviations

CAD	Computer Aided Design
CSF	Cerebrospinal Fluid
CT	Computed Tomography
FE	Finite Element
FEA	Finite Element Analysis
FEM	Finite Element Method
ICP	Intracranial Pressure
MRI	Magnetic Resonance Imaging

CHAPTER 1: INTRODUCTION

Head injuries are life threatening and a major source of psychological & health disabilities and mortality. These injuries are the results of falls, collisions, crashes, sports and accidents, etc. The conditions associated with such injuries are critical and there is a need to investigate and understand the consequences of impact head injury and underlying phenomena's in terms of damage progression. Predictive modeling has emerged as an effective tool. Experimental studies are important but do have some limitations. One of the major is, few quantities cannot be measured directly especially in the structures that are located deep inside head. Also these studies are expensive and tedious. To handle these limitations and solving problems, FE method is used. The FE method is considered to be efficient approach to model these impact situations. This project seeks to develop a realistic FE human head model to study of damage progression in head impact injury under external loading. Sensitivity of the head impact injury to the elastic modulus has been assessed and modeling of homogenized brain model in response to quasi static loading has been studied.

1.1 Problem Statement

The mortality rate due to head injuries is escalating day by day. Head injuries are one of the nation's major health-care problems affecting many inhabitants. It is the main source of disability and mortality among youth. These injuries can be very expensive in terms of health care expenses, efficiency losses and lives lost. To gain better understanding of such incidents, it is possible to develop FE models which can be used as a prediction tool to find out the extent of damage inside the brain. FE techniques have advantages over other traditional approaches due to its robustness, reliability & efficiency. Therefore, computational modeling can be a better tool to study and investigate underlying effects of impact injury in the deep down structures of head. There has been a lot of work already available to-date for impact head injury, but fewer studies talk about sensitivity of the location to the impact injury. The aim of this study is thus, to develop FE head model and investigate the progression of damage in location specific impact head injury.

1.2 Significance of Study

This topic is selected so that effect of localized head impact injury can be studied on the human head model. There is a need to investigate localized injury on human head model which can be used to test different hypotheses and can help in the study of head impact injury and protection. The stress and displacement profiles are studied to determine damage progression in brain during head impact injury. The project will facilitate researchers as literature to-date do not provide an example of the linear analysis of head impact injury on a computer solver which is why it could be quick efficient and cost effective.

1.3 Objectives of study

a) Geometrical reconstruction (3D) and study of human head model for head impact injury using Finite Element Method (FEM).
b) To investigate location specific impact head injury using mechanically elastic linear approach in FEM under external loading.

1.4 Thesis layout

This dissertation has five chapters. Chapter 1 presents the research topic and discusses the problem statement, significance and objectives of the research. Chapter 2 gives an extensive literature review about anatomy and physiology of human head and brain, head impact injury, FE modeling of human head models, use of computational modeling , previous studies and their limitations. Chapter 3 discusses the adopted methodology to implement this research. The chapter sequentially elaborates the steps taken to develop this study. It covers the entire methodology starting from construction of Simple and complex model, meshing, material assignment, boundary conditions and doing finite element analysis of the developed model. The time taken for creating the realistic human head model, execution of simulation in terms o computational cost has also been elicited. Chapter 4 shows the results of the performed study and discusses the observations of the analysis. The brain injury was predicted by comparing the observed values with the literature. Chapter 5 concludes the study and provides improvement recommendations for the developed study. It also discusses the suggestion for future work.

CHAPTER 2: LITERATURE REVIEW

Head Injuries are a common phenomenon in today's times. The most common causes for these injuries are accidents, falls, physical assault, or traffic accidents. Adults undergo head injuries more often because of automobile crashes, falls, being struck or impacting by an assault or object than other age groups.

Principles of mechanics have a vital role in the investigation of head injuries and we cannot fully understand the mechanism without the help of these principles. Extensive research has been carried out since the early 20^{th} century. Most of these researches only revolved around the brain tissues and not the complete head layers. The other aspect is that the in-vivo and in-vitro characteristics of the brain tissues are quite different, and are still not quite accurately known till today. The experimental study is very critical to the analysis but it can be quite expensive and sluggish. Secondly, it is nearly impossible to measure some of the quantities directly therefore computational modeling which is an excellent approach to deal with all these problems on hand. In this approach the FE Models are widely used in the researches. 2D models were used in the past as computational powers of the computer were quite limited, but with the strides in computational power of the computers and advanced GPUs today we can use 3D models to study the brain injuries. The accuracy of these models is quite impressive and we are able to get much more realistic and definite results using these advanced models.

2.1 Human head anatomy

The human head has a very complicated geometry. To understand the biomechanics of head injury it is essential to know fundamentals of anatomy and physiology of human head. The head is a multilayered structure which majorly includes scalp, skull (compact bone), sandwiched layer spongiosa (spongy bone), cerebro-spinal fluid (CSF) and brain. The most complex organ in the head is brain, which is protected inside the bony skull. The brain has a network of interconnected neurons which controls all the major functions of the body including coordination, memory, intelligence, emotions, creativity and decision making. Every part of the brain performs the specific function, to perform more complex functions these regions are interconnected with each other. A significant disruption in the function of brain is observed in response to minor or mild injury. These injuries can bring drastic changes in the brain function and affect its performance.

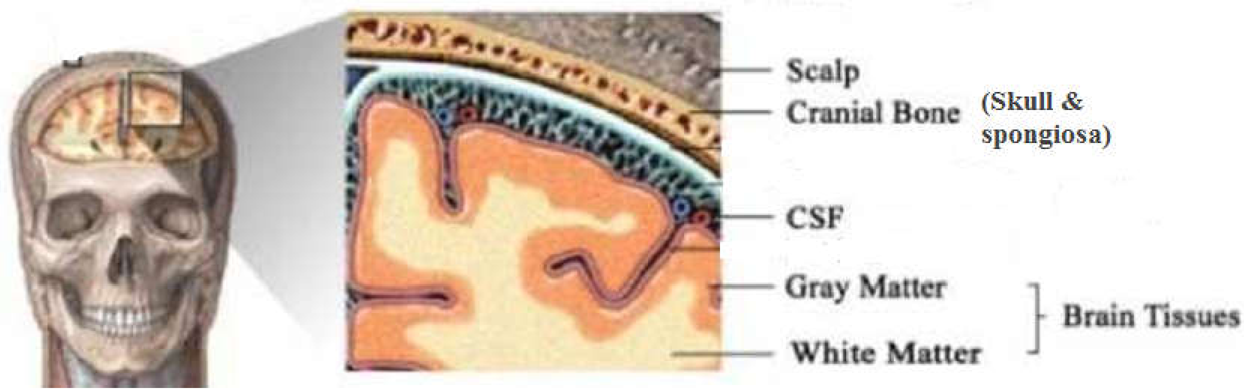


Figure 2.1: Human head and its important parts are illustrated (modified from [1])

2.1.1 Brain Anatomy and physiology

Physiologically, the brain is classified as left and right hemisphere. It has cerebrum, cerebellum and brain stem. The brain is divided into functional sections mainly as cerebellum, brain stem, temporal lobe, occipital lobe, frontal lobe and parietal lobe. The Table 2-1 provides the location and functions of these parts.

Table 2-1: Locations and functions of the different regions of brain.

Brain Lobe	Location	Function
Frontal	right under the forehead (Anterior side)	Attention, Self monitoring, Speaking(language expression), Personality, Awareness of abilities and limitations, Emotions, Mental flexibility, Problem solving, Motor planning & initiation, Judgment, Organization, Planning and anticipation, Inhibition of behavior, concentration.
Parietal	Located near the back and top of the head	Sense of touch, Spatial & Visual perception (i.e. identification and differentiation of shapes, size, and colors).
Temporal	Above ears, on the left and	Organization, Hearing, Understanding

	right sides of the head.	language (receptive language), Sequencing, Memory.
Occipital	Located at the back of the head (Posterior)	Vision.
Cerebellum	Situated beneath the cerebrum	Skilled motor activity, Balance, Coordination, Visual perception.
Brain stem (Includes the medulla, Pons & midbrain).	Anterior to the cerebellum. Located at the juncture of the spinal column & the cerebrum.	Attention and concentration , Arousal Breathing and consciousness, Heart rate, Sleep and wake cycles

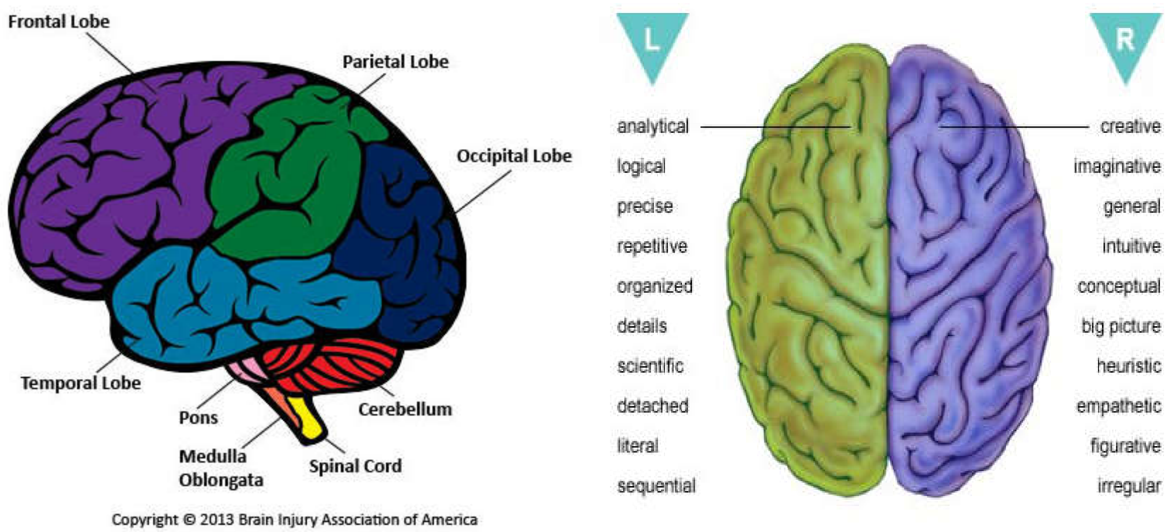


Figure 2.2: Brain regions, lobes and left and right hemispheres (L= left & R= right). (Source: http://www.biausa.org/living-with-brain-injury.htm.)

2.1.2 Head and Brain injury

In medical science, brain injury and head injury are interchangeably used[1-3]. According to literature head injury is a generic term which includes injuries to skull and brain [3, 4]. In a report[5], head injury is defined as “evidence of presumed brain involvement, i.e. concussion with loss of consciousness, post traumatic amnesia, neurologic signs of binary injury or even skull fracture.” The clinical findings and Goldsmith [6] confirmed that brain damage can occur with or without skull fracture and vice versa. Also brain injury is defined as "physical damage to, or functional impairment of cranial contents from acute mechanical exchange, excluding birth trauma"[7].

During impact, four major lobes (frontal, temporal, parietal, occipital) transmit forces to the brain which cause mild traumic brain injuries (MTBI). In investigating literature, it was found that of all injuries, MTBIs are most common and widespread. The reason of its prevalence is that stresses and strains are generated when the changes in intracranial pressure (ICPs) are happened[8, 9].

Table 2-2: Effects of brain injury on brain functions.

Sr. No	Injury at	Causes
1	The left side of the brain	Sequencing difficulties, Impaired logic, Decreased control over right-sided body movements. Anxiety, depression, Difficulties in speaking, Verbal memory deficits, Difficulties in understanding language.
2	The right side of the brain	Visual memory deficits, awareness of deficits decreased, Visual-spatial impairment, and control over left-sided body movements decreased, Altered creativity and music perception.
3	Diffuse Brain Injury (scattered injuries occurred throughout both sides of the brain)	Fatigue, Confusion, Reduced concentration and attention, in all areas impaired cognitive (thinking) skills, Reduced thinking speed.

2.2 Finite Element (FE) Modeling

A FE method is the technique in which an object from the real world is breakdown into many discrete finite elements. It is an efficient method that can handle irregular nonlinear geometries, nonlinearities& multiple compositions of material and complex boundary and loading conditions of the structures. Then mathematical tools are used to predict the response and behavior of these elements. Finally the computer computes and adds all the individual responses of these elements to give the response of the real object.

Finite Element Analysis (FEA) is quite a handy tool when it comes to the understanding of complex trauma mechanisms resulting from head impacts. Enhanced FE head models present the viewpoint of delivering precise, reconstruct-able experimental analysis of every types of head injury. The effectiveness of such models depends upon the accuracy of the geometry of the structures present inside the model. Most of the computational CAD based models are usually designed manually, but when it comes to the head and its complex internal parts, this approach tends to become error prone and can be subjected to human error. Thus incase of complex models it can provide a lot of challenges. A more appropriate approach that can minimize the inaccuracies related with modeling of extremely complex structure is well-known as 'image-based meshing'. It is an approach in which a volume scan data is converted from CT scan or MRI scan into a FE mesh directly by semi and completely automated processes through least possible user input [10]. Such approach is effective because it helps to achieve many distinguishable anatomical parts of complex structure with much needed accuracy.

2.2.1 Human head FE model

Since 1940s, many analytical solutions were provided for human head. Because of impediments and scientific challenges, these analytical and mathematical models elucidated by regular and simple geometries, homogeneous and isotropic material properties and simple boundary conditions. FE modeling is a versatile digital computational technique which has ability to handle the objects with non uniform shapes and inhomogeneous materials. It can provide solution by use of approximations to the analytical formulation for human head model efficiently.

Hardy et.al [11] were the first to design and model the performance of human head response by using FE modeling technique. They constructed a simple 2D FE head model which had the layer of skull only. This model was augmented afterwards by another researcher Shugar et al. [12]. In his model he came up with a fluid-filled, elastic layer of brain attached to the skull. Due to the lack of available image processing and computational resources in the 60s and 70s yet these 2D strain models were the only way to simulate deformations of the head in case of impact. The results of this model were quite impractical and improbable for large deformation simulation. 3D FE head models were represented using regular geometry. For example, Chan et al. [13]developed the human head model as an ellipsoid and spherical shell while Khalil et al. [14] employed an in viscid fluid core as the intracranial content and an ellipsoidal shell for skull. These models dealt with material features of only one of either skull or the brain. In early1980s, Hosey et al. [15] presented the first 3Dhomomorphic model with the basic anatomy of both brain and skull. This model also incorporated the inertial effects and material properties of the neck and head. It was considered the most comprehensive FE head and neck model in the 80s. At that time the potential of these earlier FE head models was not realized due to lack of computing power, but later on this model proved to be the basis of analysis of head injury biomechanics. Computing capability in 90s made it possible to develop more sophisticated and realistic 3D human FE head models based on real geometry of the human head. From the Wayne State University, Ruan et al.[16]modeled a detailed and well known FE humn head model, simulating the detailed and defined anatomy of the brain and skull.

In early 2000, Kleiven et al. [17] came up with KungligaTekniskaHögskolan (KTH) FE human head model. This model was unique in the sense that it was quite comprehensive and parameterized FE human head model of an adult. It included the scalp, brain, meninges, skull, cerebrospinal Fluid (CSF), simplified neck and eleven pairs of parasagittal bridging veins. It was modeled by using isotropic, homogeneous, nonlinear and viscoelastic material properties that were derived from previous work. The study revealed that in case of an impact, pressure response of the brain relied more on the interface type of skull-brain instead of constitutive parameters of the brain tissue. Upon consideration of effect of different interface conditions study also revealed that the tied interface provided the best correlation with existing literature of Nahum et al.[18]'s, Hardy et al.[19]', King et al.[20]'s experiments.

Later on many researchers like Zhang et al.[21], Belingardi et al.[22], Lashkari et al.[23], Tse et.al[24], Yang et al.[25], Jin et al[26],Song et al.[27] Cotton et.al[10] have developed their 2D & 3D models with different layers, material properties, boundary conditions and different impacts. These models were tested and validated against literature available experimental results [18, 28]. Table 2-3 provides a brief summary of recent development of 3D human head model including description and validation of each reported model.

Table 2-3: Summary of few developed 3D human head model.

Author	Year	Description	Validation
Cotton et al.[10]	2015	3D model developed from MRI data, in Simplewareand Abaqus are used softwares, tetrahedral mesh elements are used, total 3.72 Million volumetric elements of 33 layers are generated. This research developed a realistic model of complete neck and human head, used in TBI related study.	Nahum et al.[18],
Jin et al.[26]	2015	3D model, constructed from CT scan of a individual, 74,636 brick 24,391 shell elements were used. Scalp, facial bone, cerebrum, brainstem, cerebellum, two layered skull, CSF, falx, pia and dura matter. The major finding of the study was found that for anticipating the brain response and detailed analysis of injuries during impact to the human head, CSF with high bulk modulus was more befitting and suitable.	Nahum et al.[18] for translational impact, Hardy et al.[19] for skull/brain motion in rotational impact. Trosseille et al.[28], For Rotational impact.
Yang et.al[25]	2014	3D model, 1,173,039 tetrahedral elements white and gray matters, cerebellum, CSF the entire ventricular system of the brain, midbrain, and brainstem. the maximum Von Mises stress in the brain and the maximum principal stress in the skull were discussed through this model.	-
Tse et al.[24]	2014	Mimic & hyper mesh software are used . Model constructed by using CT scans Linear hex and 03,176, linear	For localized brain motion Hardy et al.[19],

		tetra1,337,903 mesh are used. Skull bone, brainstem ,CSF, cerebral peduncle, cerebellum, ventricles, cerebrum, white and gray matter, tooth , cartilages, soft tissues were modeled. The study revealed that simulated results were remarkably in accordance with experimental measured relative displacements and ICP. The Soft tissue modeled aided in damping out the oscillations of the models response.	For intracranial pressure for long duration impulse Trosseille et al.[28]. Pressure data, force, and intracranial acceleration from Nahum et al.[18] for impact for short duration impulse.
Kleiven et al.[17]	2002	Developed the Kungliga Tekniska Högskolan FE Head Model. This model has scalp, skull, meninges, brain,CSF, 11 pairs of veins and a simplified neck. It also included sliding condition between brain and skull. Different parametric researches on various skull-brain interface conditions revealed that the human head is responsive to the modeling of such interface condition during impact. The study also revealed that brain's pressure response in an impact relied more on the interface type of skull-brain instead of constitutive parameters of the brain tissue.	Nahum et al.[18], Hardy et al.[19] and King et al. [20] experiments.

2.3 Research Gaps

While going through the literature it was found that there is a need to do some basic level of study which should discuss the localized injury scenario during accidental mishaps and serves as a guideline for building a realistic human head and brain model for addressing head impact injury scenarios.

2.4 Chapter Summary

In this chapter, an in-depth review of literature has been presented. The anatomy and physiology of human head and brain, head impact injury and the role of computational modeling have been elaborated. The chapter also discusses about the research gaps in previous work and how to mitigate them.

CHAPTER 3: METHODOLOGY

The aim and first step of this research is to develop location specific, three dimensional FE human head model which helps the study of damage progression in impact head injury. The study was conducted in three main steps; Preprocessing, Analysis and Post-processing.

Preprocessing: The first step of finite element analysis (FEA) is the geometry construction of the structure which is to be analyzed. After geometry construction, model needs to be meshed. Meshing is a method to discretize and define the model into small elements. When the FE model is made, by applying appropriate interactions and constraints the material properties are assigned to each part. In the end, loads and boundary conditions are assigned.

Analysis: Subsequently, with in the predefined constraints, loads and boundary condition convergence of the model is checked by using numerical solver. In order to test physical conditions efficiently, different analysis for example time based analysis (static/quasi-static/dynamics) can be used. In static, input and output parameters are time independent; hence balancing of the load and boundary conditions for the system is solved. Quasi-static means that we may assume the static condition over given instant of time and with negligible inertial effects. In this way we can simplify the non linear system of equations to a linear system at small increment. In dynamic, input and output are time dependent; therefore the model shows variable behavior over the period of time. Example of dynamic case is impact loading in which mechanical response varies with the rate of loading. In this research we have assumed quasi-static conditions for all the analysis.

Post-processing: Last step is the post-processing, in which visualization of the results is performed. The outputs of the study give corresponding values for each node and element of different parameters for example displacements, stresses and strains etc.

The steps in the development of location specific, 3D human head FE model is shown in Figure 3.1

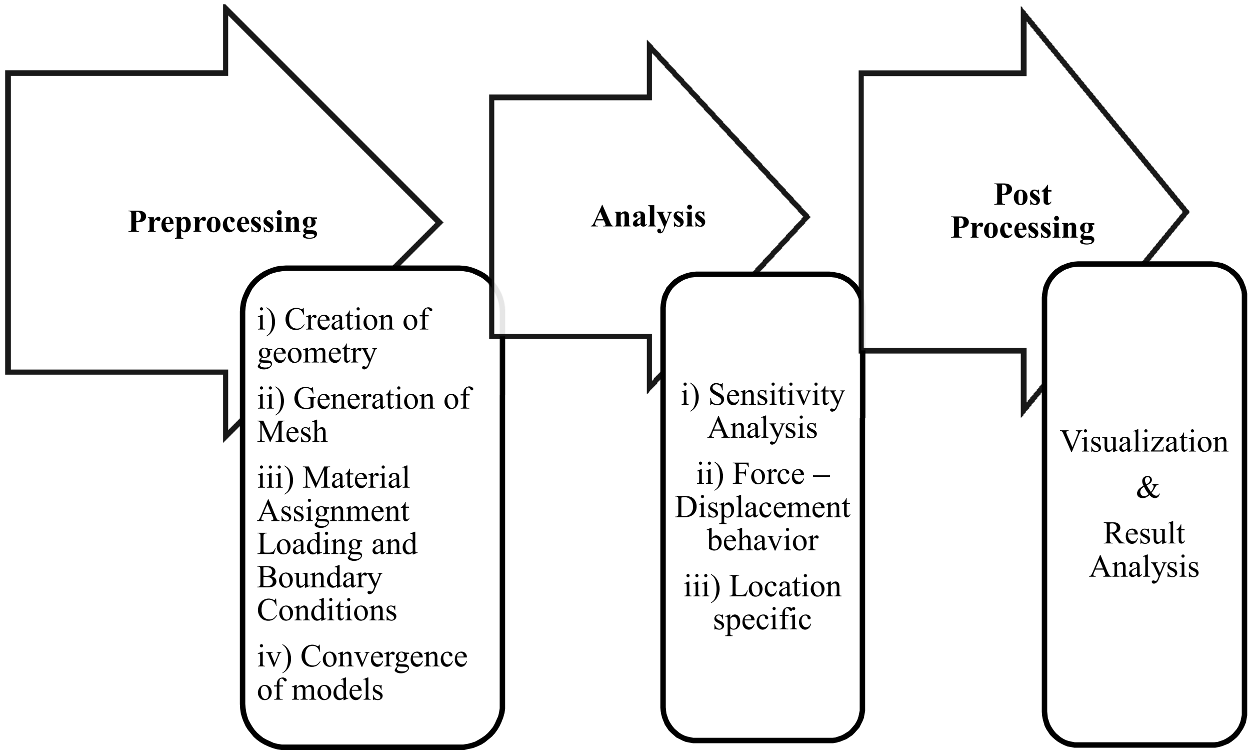


Figure 3.1: Steps of development of location specific, 3D human head FE model for the study of damage progression.

3.1 Development of Human head model

3.1.1 Geometry Construction:

In this study, two head models were used to understand the effects of quasi-static loading on human head during impact injury. The primary form of the model was developed and used for electrical analysis. This model was utilized using research conducted by Salman et.al[29]. The said model was modified and few changes were made so that it can be utilized for the current research. Model 1 is “the complex model” which was originally developed by Salman et.al[29]. Three dimensional realistic human head model was constructed by performing image processing tools on the MRI data sets in the module, “Scan IP” of the simple ware, shown in Figure 3.2. In this study both of the models had six layers namely scalp, skull, spongiosa, cerebrospinal fluid (CSF), white matter and gray matter. The scalp is the outer most layer while white matter is the inner most layer of the models.

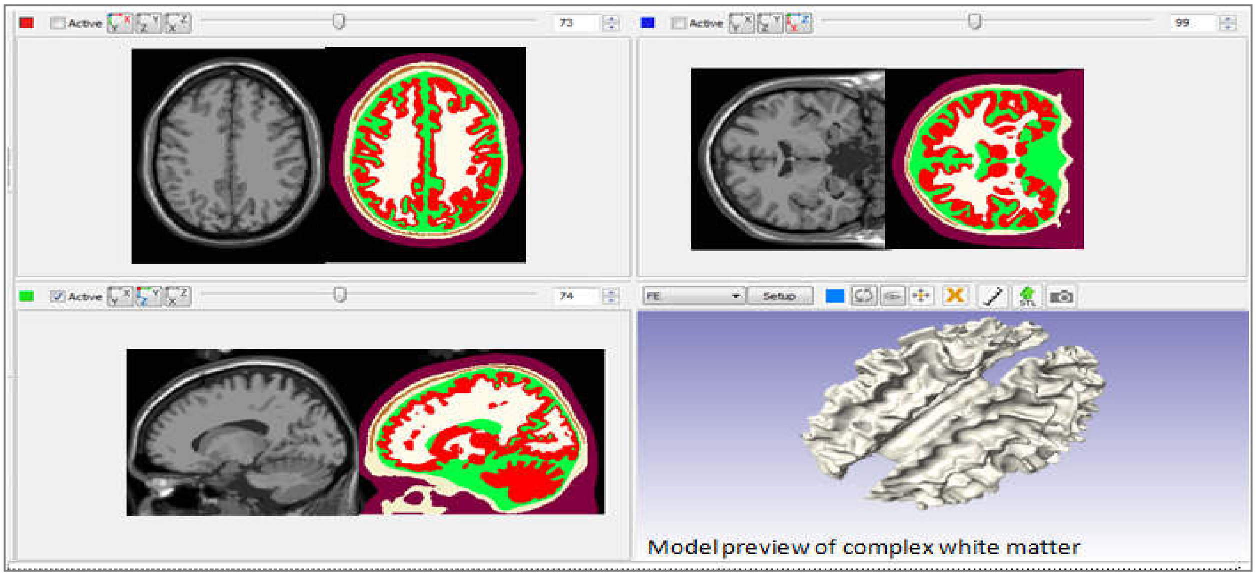


Figure 3.2: Segmented head layers of complex model (Model 1) in Scan IP.

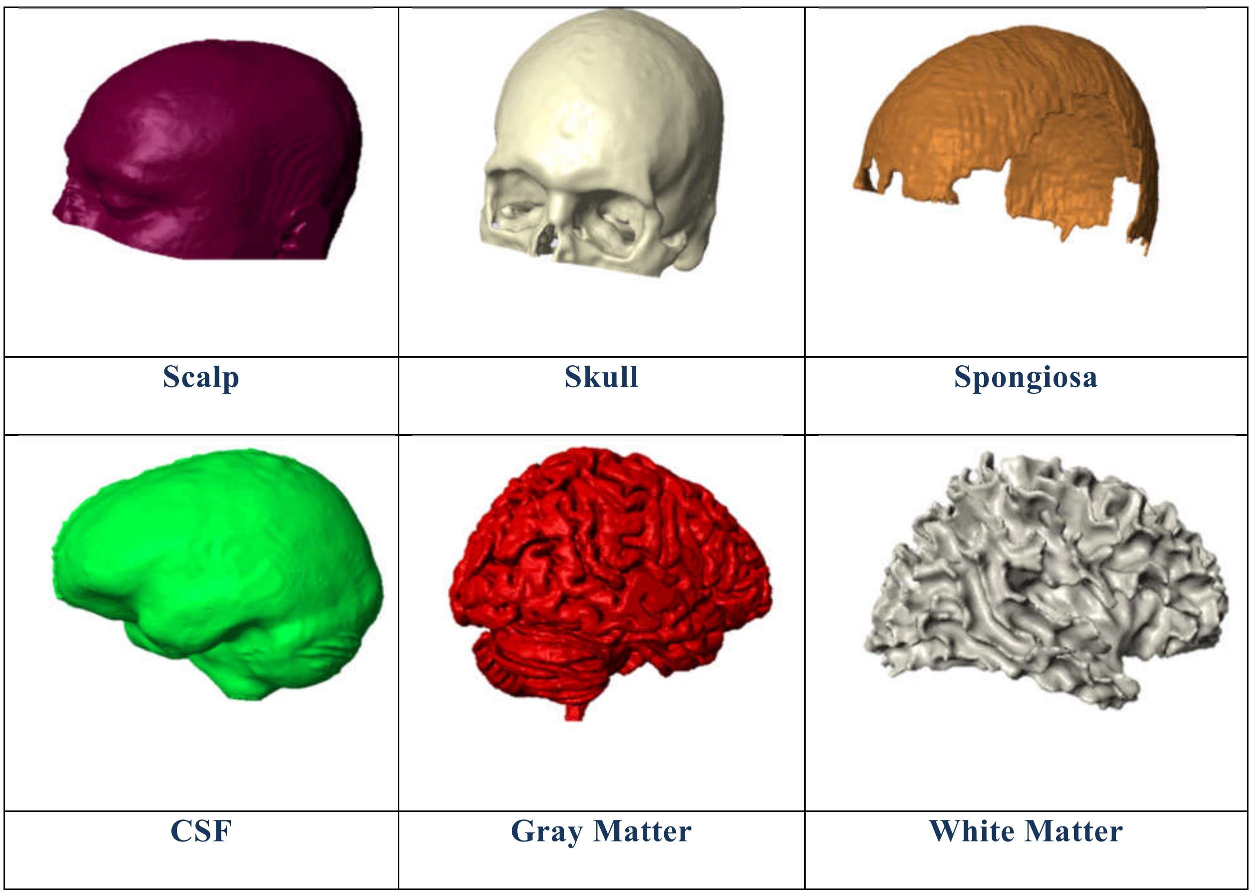


Figure 3.3: 3D Models of the segmented layers of complex model.

Model 2, “the simple model”, was derived and simplified from the realistic human head model of Salman et.al[29]. In the simple model the geometry of the brain layer was simplified by removing convolutions of gray and white matter.

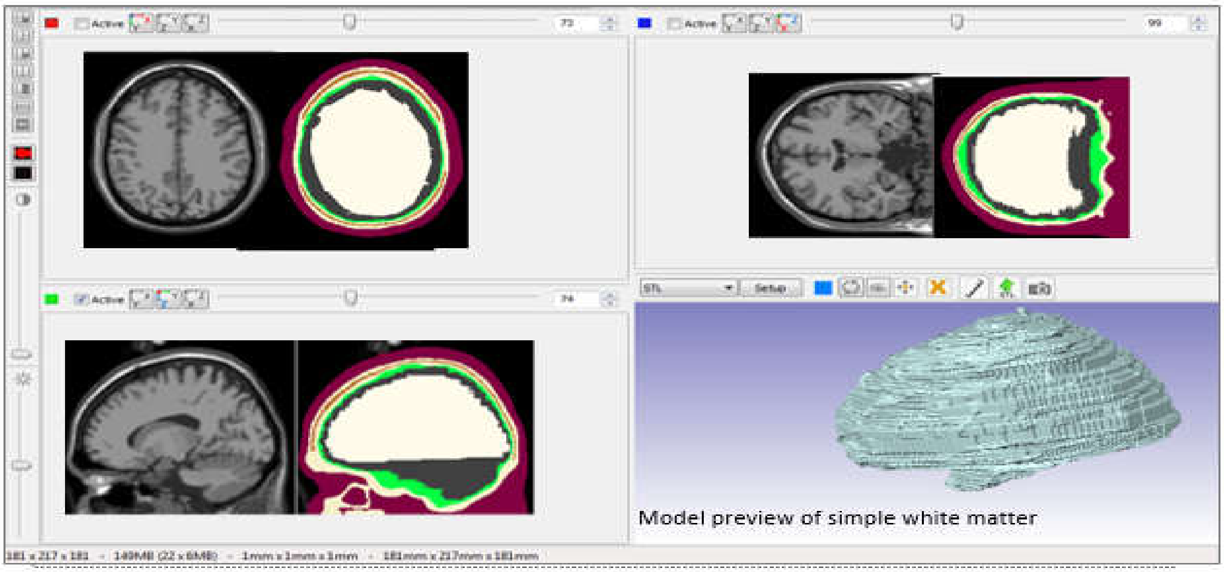


Figure 3.4: Segmented head layers of simple model (Model 2) in Scan IP.

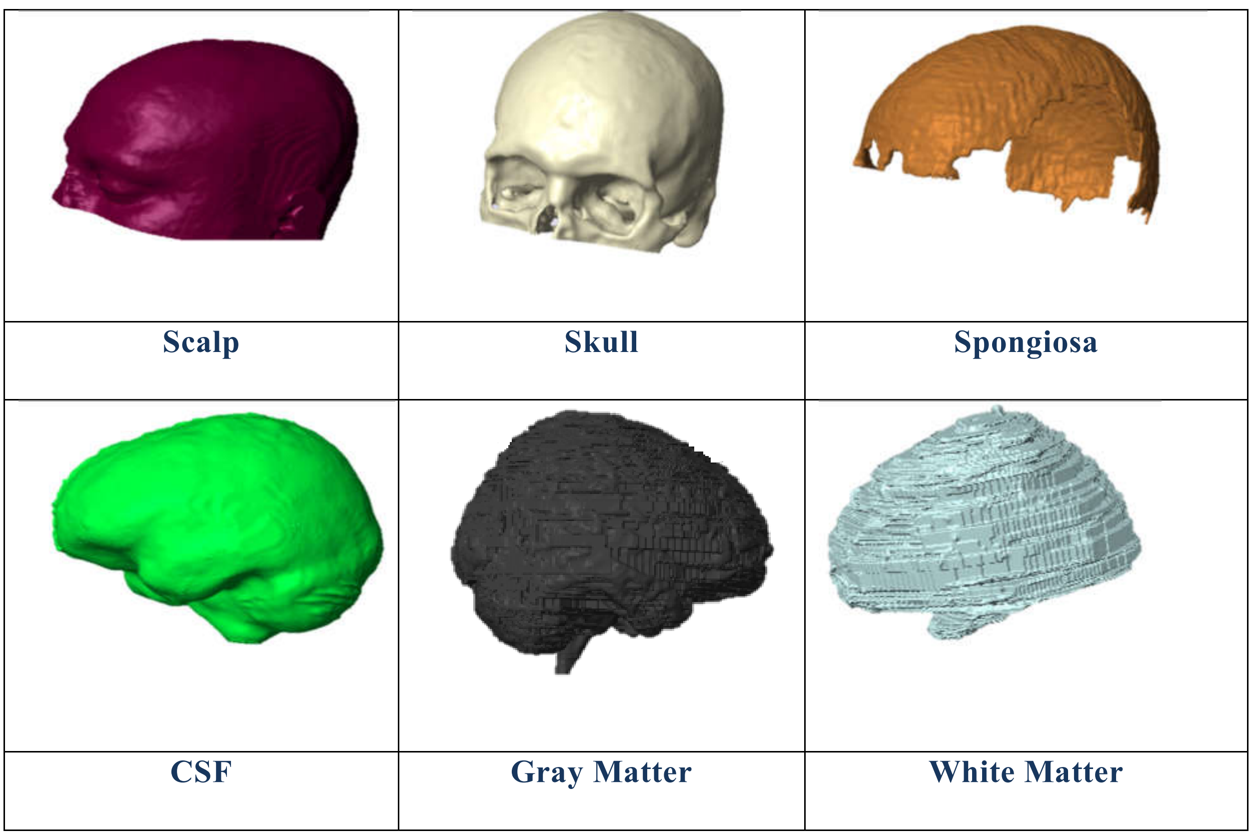


Figure 3.5: 3D models of segmented layers of simple model.

In Salman et.al[29] the modeling of electrodes was according to international 10-10 EEG electrode system and was modeled in the other module "ScanCAD" of the Simpleware. Each of the electrodes had a disk shape, having 6mm radius and 2mm thickness. For this study, the locations for loading were selected and derived from electrode locations. Figure 3.5 shows the locations selected for loading. Total six numbers of locations were selected. Table 3-1 provides the electrode locations translated to loading locations used in this research.

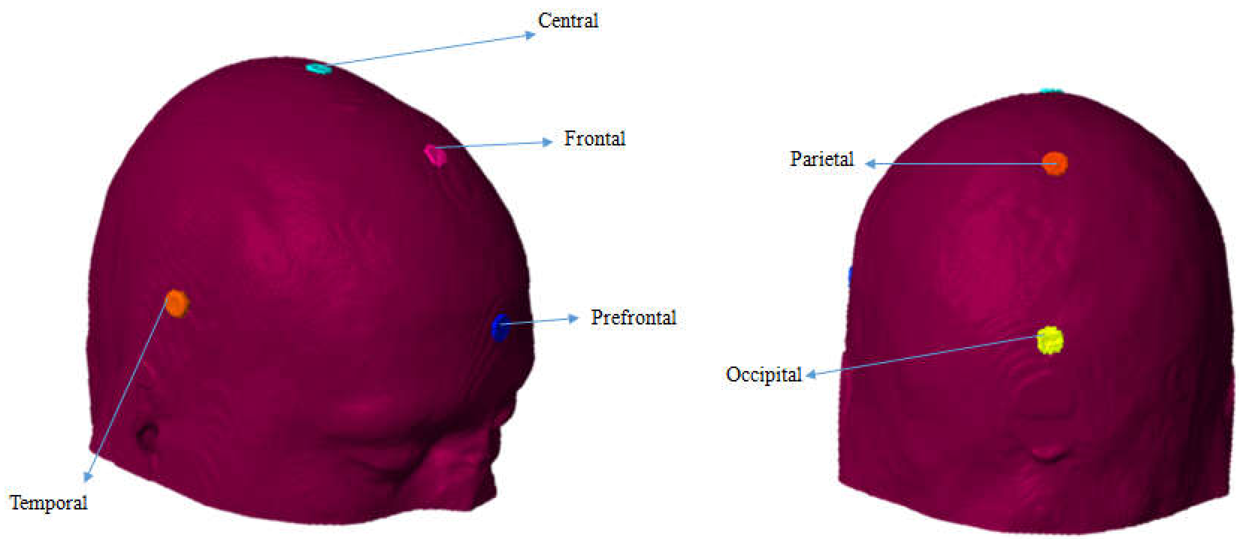


Figure 3.6: Loading locations selected for this study.

Table 3-1: Loading locations derived from 10-10 EEG electrode system.

Sr. No	Electrode Location	Loading Location
1	Fpz	Prefrontal
2	Fz	Frontal
3	Cz	Central
4	Pz	Parietal
5	Oz	Occipital
6	Tz	Temporal

3.1.2 Meshing procedure

The geometries created in Scan IP for both models were transferred to Scan FE module of the Simpleware in order to create the volumetric mesh.+FE Free method is used to generate volume meshing of the model. In this study to represent the discretized volumetric 3D model, all tetrahedral linear elements were selected. This type of meshing can efficiently represent the highly convoluted brain layer in best form. The default tetrahedral quality metric of ScanFE is the normalized in-out radius.

$$Q = 3 \times \frac{R_{in}}{R_{out}} \tag{3.1}$$

Where $R_{in}$is inscribed sphere radius and R_{out} is circumscribing sphere radius.

From coarser to denser four different levels of meshes were selected by adjusting compound coarseness slider. Table3-2 illustrates the different levels of mesh for both models, their corresponding number of elements and time to generate these meshes at each level.

Table 3-2: Different levels of mesh, number of elements and time to generate each mesh.

	Complex Model		**Simple Model**	
Level	Mesh density (element number)	Solution Time (minutes)	Mesh density (element number)	Solution Time (minutes)
1	2176895	40	1537867	30
2	2308601	45	1876045	38
3	2671721	55	2319666	45
4	3254665	80	2890309	60

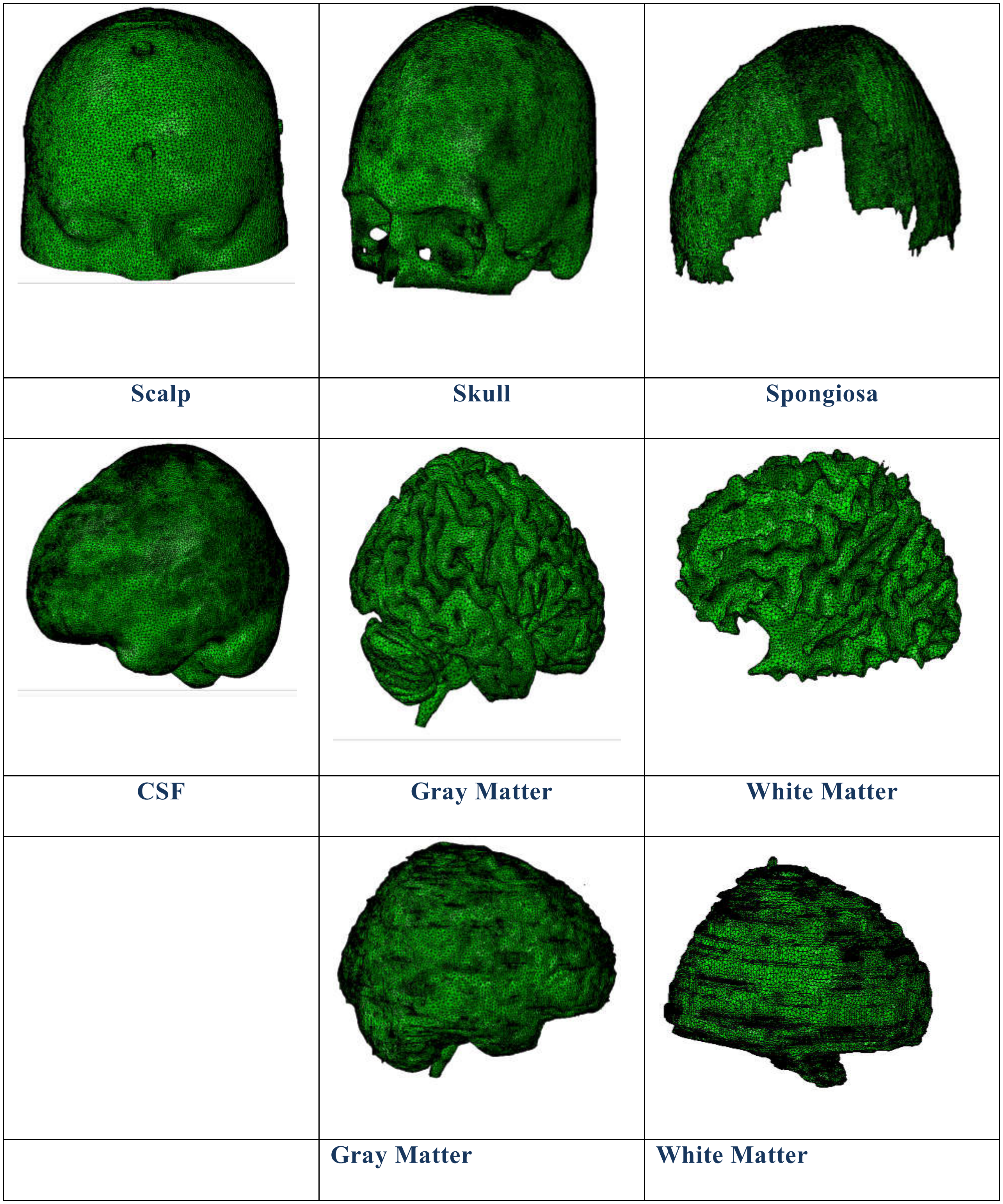


Figure 3.7: Meshes of the segmented layers, generated by Scan FE.

3.1.3 Material Assignment

Material properties assignment is the most important aspect of FE procedure as they determine the material behavior at the end when subjected to external loading. Many FE models considered different layers of head as isotropic, homogeneous, and linear/viscoelastic. The complex structures for example human skull, prior head models accepted the skull bone as a homogeneous and isotropic material. In few studies it can be seen that skull bone was treated as heterogeneous, indicated by layers and porosity (Cortical bone with 5-10% porosity, cancellous bone with 50-95% porosity) [12-14, 30]. Many researchers put their efforts to determine the constitutive properties of the human brain material. To study such properties, experimental testing on animal and cadaver brain was performed. In few earlier FE models the brain was modeled as linear elastic material [15, 18, 31, 32] while some had considered the brain as viscoelastic [13, 14, 30]. Even though the significance of both of the material models with respect to brain tissue response was controversial[33] but it's worth mentioning that the difference of responses between viscoelastic and linear model increased with the loading time [34]. In 2007,Gao et al.[35], used inhomogeneous brain model, as mainly consists of three types of tissues, white matter gray matter and CSF. In this study two types of material models were examined, linear model and homogenized model.

3.1.3.1 Linear Model

As the human head is complex in nature hence it is difficult to estimate material properties of human's head and brain. In this research, properties of the material were consulted and selected from different literature. The material properties of different layers of the head are selected as isotropic, homogenous and elastic. The model has total six layers, namely scalp, CSF, skull, spongiosa, gray matter and white matter. The values of elastic modulus, Poisson ratio used in this model have been listed in Table 3-3. The densities (tone/mm^3)of the extra cranial tissues (scalp, skull, spongiosa) used in the model were selected from[36, 37]. Density of scalp was 1.130E-09, skull was 1.5000E-09 and of spongiosa was 1.740E-09 tonne/mm^3 respectively.

Table 3-3: Material Properties of the linear model.

Sr No	Material	Model Type	Young Modulus (MPa)	Poisson Ratio	References
1	CSF	Linear Elastic	1.000E-05	0.499	[35]
2	Gray Matter	Linear Elastic	4.000E-03	0.495	[35]
3	White Matter	Linear Elastic	2.100E-03	0.495	[35, 38]
4	Scalp	Linear Elastic	1.670E+01	0.42	[39],[40],[37]
5	Skull	Linear Elastic	1.500E+04	0.21	[39], [36, 40, 41],[42]
6	Spongiosa	Linear Elastic	3.400E+03	0.22	[37]

3.1.3.2 Homogenized model

In this model, all the layers of the human head are considered as linear, elastic except the brain layer (combined gray and white matter). In this study, in order to address non linearity in the linear model, concept of homogenization is introduced. It is assumed that the brain tissue is a complex composite and is composed of filler and matrix. To calculate effective modulus or the bulk modulus of the brain, rule of mixture was used. It can be seen in literature that effective modulus was estimated by using effective medium theories that approximates the estimation by homogenizing the complex medium. The simplest approach of elastic modulus estimation for two phase material is the classical averaging schemes. It is believed that the Voigt estimation or the rule of mixture has upper bound of the effective Young's modulus of composites. However, it holds true where the Poisson effect is not significant. The effective modulus as calculated as proposed by Voigt et al.[43].

$$E = E_1 v_1 + E_2 v_2 \tag{3.2}$$

Where

E_1: Elastic modulus of Filler;

E_2: Elastic modulus of Matrix;

v_1: Volume fraction of Filler;

v_2: Volume fraction of Matrix.

$v_1 + v_2 = 1.$

This mixing rule is for the iso-strain state. This model assumes the constituents of a composite to be in parallel arrangement subjected to the same strain.

In the model of this study it was assumed that brain is filler and void/porous is the matrix with the elastic modulus of nearly equal to zero. Details of volume calculation and effective modulus are mentioned in the Appendix A. The calculated effective modulus used in this model was 6.743925×10^{-1} MPa.

3.2 Simulation Parameters

After material assignment, simulation parameters are assigned to the models in order to run simulations. A static general study was performed on the developed models. With selected amount of forces, tie constraints among the layers and fixed base, simulations were performed in ABAQUS CAE 6.12. The direction of applied forces, constraints & fixed base, used in this study can be seen in Figure 3.10.

3.3 Analysis

In this, analysis to be performed on Model 1 and Model 2 were designed. The developed human head models were then analyzed under three different studies and implemented on both, simple and complex model.

Case I: Sensitivity analysis, in this case the sensitivity of brain injury to elastic modulus was investigated. Linear and homogenized models of simple and complex human head models were considered and compared. After analysis, homogenized model was selected to perform Case II and Case III.

Case II: According to Oliver et al.[44] load and displacement (L-D) curve can be used determine hardness and elastic modulus of the sample. This L-D curve is shown in Figure 3.8. The important parameters of this curve are maximum load (P_{max}), maximum displacement (h_{max}) and final depth (h_f). In this study, Force-Displacement behavior of complex and simple model was studied at different amount of applied force. Four discrete levels of forces (Figure 3.9) were selected from literature [10, 18, 23, 25, 27] and were used to examine the response of both homogenized models at each level. After investigation, one force (causing head injury in the simple model) was selected and used as an input for Case III.

The gradient (slope) of the Force-Displacement curve can be used to determine the actual elastic modulus of the brain.

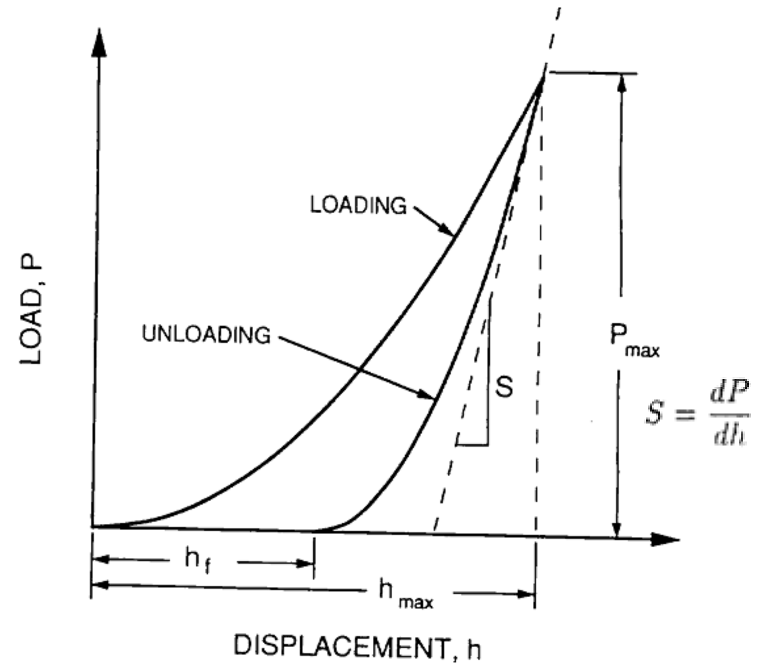


Figure 3.8: Typical Load-Displacement curve.[44]

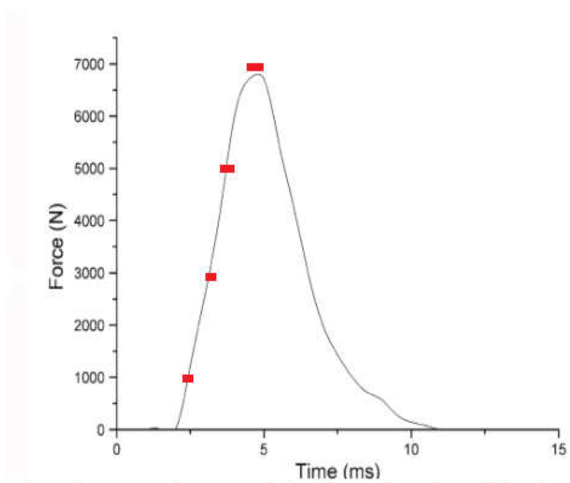
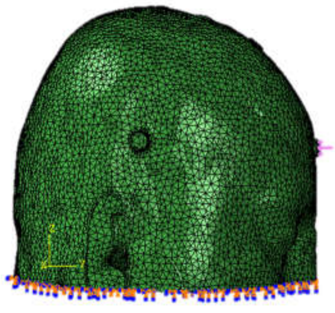
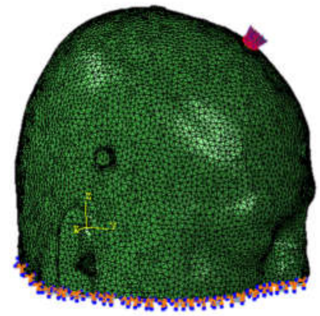


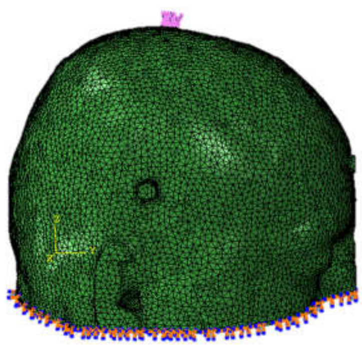
Figure 3.9: Levels of forces (1000N, 3000N, 5000N, 7000N), points highlighted in red, are selected from the rising curve of input force. [10, 17, 22, 24, 26].

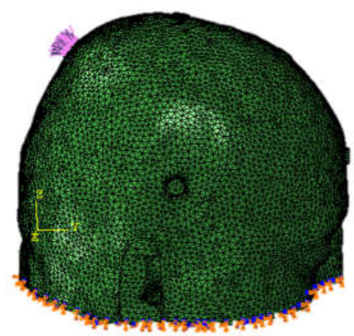
Case III: Location specific study was performed on homogenized model to determine the response of simple and complex model. Six different locations were studied and compared for both models. Figure 3.10 shows the locations, constraints and boundary conditions used in this study.



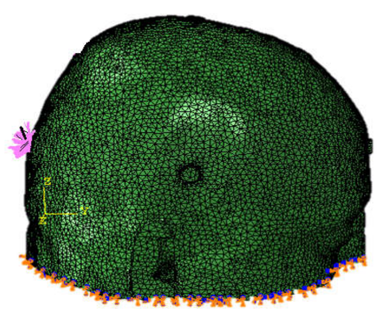


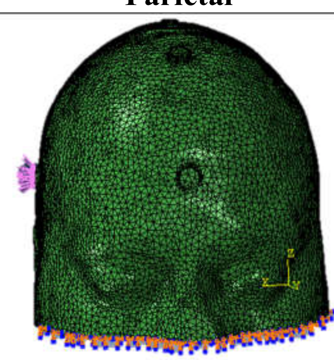
Prefrontal	Frontal





Central	Parietal





Occipital	Temporal

Figure 3.10: Model showing different loading locations (in pink) with tie constraints and fixed base (in blue and orange) of the model.

Following Table 3-4 gives the details of the designed analysis performed on both simple and complex model.

Table 3-4: Analysis performed on the developed models.

Cases	Variable	Constant
Case I	Brain Modulus of Elasticity	i. Location : Prefrontal Impact ii. Amount of force= 1000 N
Case II	Amount of Force (1000 N, 3000 N, 5000N, 7000N)	i. Homogenized model ii. Location: Prefrontal
Case III	Locations (Prefrontal, Frontal, Central, Parietal, Occipital, Temporal)	i. Homogenized model ii. Amount of force = 5000 N

3.4 Mesh Convergence Study

Once the model is built, it is important to perform convergence study to finalize the mesh size for any refinements. A refined mesh yield better results. To decide the suitable levels for this study, convergence analysis was performed using a human head model with both complex and simple geometry. In both types of the model (Model 1 & Model 2), linear tetrahedral elements are used to mesh the model. This type of meshing is mostly suitable for preserving small features while decimating the mesh elsewhere. This type of meshing is automatically generated in +FE Free algorithm of Simpleware. This approach considerably reduces number of elements and nodes for models.

The simple model has shown irregular, dense mesh in the periphery of the brain while coarsely meshed region lies in the center of the brain. In comparison to simple model, the complex model has uniform; regular and dense mesh in the convoluted structure of the brain while in brain stem bit coarser mesh can be seen in Figure 3.10. All of the models were assigned the material properties to each layer as mentioned in section 3.1.3.These models were then tested under the same interactions, constraints and loading conditions. For initial test of convergence, a force of 1000 N (least one selected from Figure 3.9) was applied to the prefrontal region of the head, i.e. the forehead with the fixed base of the model. Tie constraints were used between the adjacent scalp and surface of loading locations. Deformation was measured at the level of brain layer. Hence this displacement was compared among each simple and complex model so that ideal mesh size can be selected for both models. Table 3-5 shows the total number of elements and time to generate each model. Level 4 of the complex model, had number of elements which were beyond the computational capacity of the system.

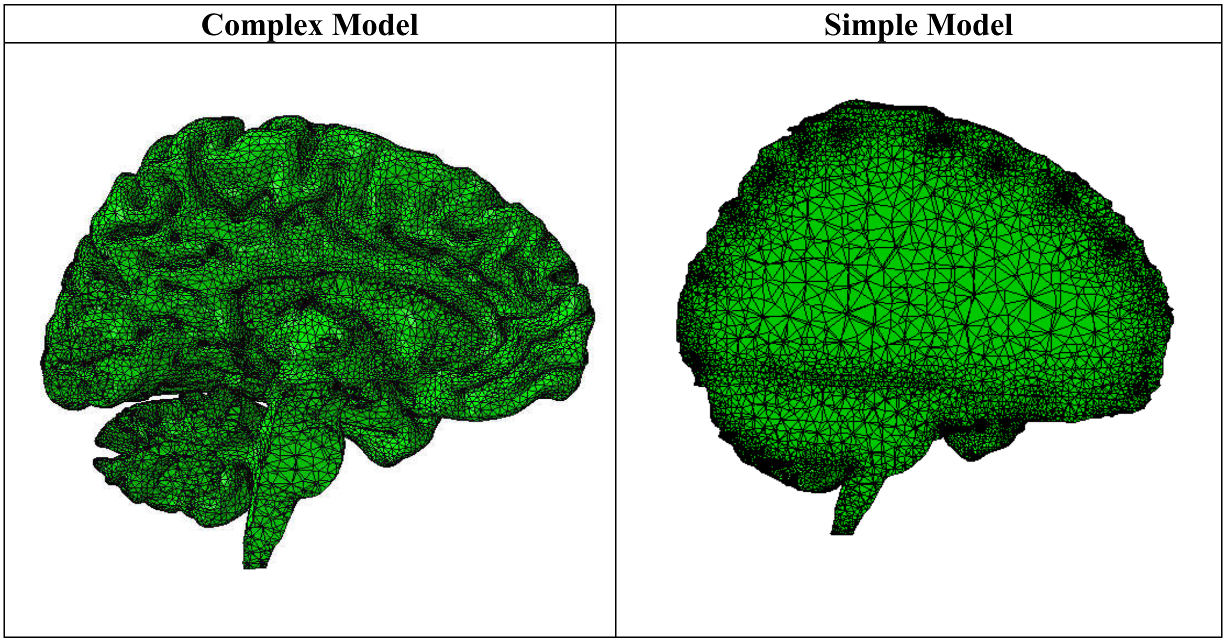


Figure 3.11: Difference between the mesh of complex and simple model.

Table 3-5: Number of elements for each levels of complex and simple model.

	Complex Model		Simple Model	
Level	Number of Elements	Simulation Time (minutes)	Number of Elements	Simulation Time (minutes)
1	2176895	11	1537867	5
2	2308601	13	1876045	7
3	2671721	20	2319666	13
4	3254665	Warning	2890309	25

In general practice it is observed that using a fewer number of elements would yield inaccurate solution in less run time [45]. With the use of greater number of elements in a model, yield results would take too much time without effective changes in them.

The Figure 3.11 shows the convergence of simple model. The similar behavior was observed for convergence of complex model (See appendix A). From our results of convergence it can be concluded that both of the models are converged and we can use any level of the model to do further analysis. In this research level 3, which is the optimum level to perform more studies is used.

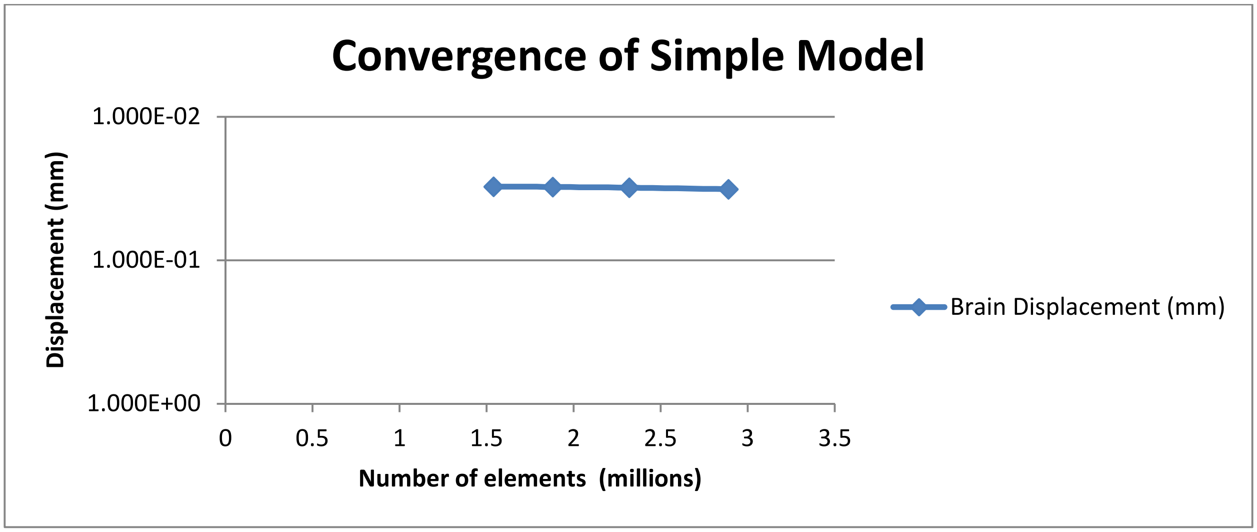


Figure 3.12: Convergence of simple model of FE human head with tetrahedral elements.

3.5 Computational Resources

The tasks performed in Modeling and Simulation is quite expensive computationally. Therefore, it is worth mentioning here the resources and software utilized in this project are given in Table 3-6.

Table 3-6: Computational Resources of the project.

Sr. No	Tasks for Each Model	Computational Resource
1.	Tissue segmentation (Simple ware)	Dell T5500 workstation 24 GB RAM 2.0 GHz Xenon processor.
2.	3D head model & tetrahedral mesh generation (Simple ware)	RCMS Supercomputer 24GB RAM node, 2.4 GHz processor.
3.	Simulation and Analysis (ABAQUS CAE 6.12)	HSL SMME Dell Optiplex 9020 Intel-core i7, 16GB RAM3.6 GHz processor.

3.6 Chapter Summary

In this chapter, the methodology for implementation and designing of the project has been explained. The methodology had three major steps and begun with the construction of 3D geometry, generation of mesh. Next setting of simulation parameters, designing of the analysis, and simulations were performed for both of the developed models. In next chapter results of these performed analyses will be shown and discussed.

CHAPTER 4: RESULTS AND DISCUSSION

In this study, the developed models (simple and complex) were analyzed by visualizing VonMises stress and displacement profiles in the brain. In section 4.1 and section 4.2, the results of simple and complex model under different cases have been illustrated and discussed. The purpose of this analysis is to study the behavior of brain in response to external loading. The results have been confirmed by using different values of Young's modulus and location specific quasi-static loads as well. All the analyses were run successfully using ABAQUS CAE 6.12 on workstation with 16 GB RAM except Level 4 of the complex model which was beyond the computational power of the said system. Both simple and complex analysis is performed for three cases:

1. Sensitivity analysis to Elastic Modulus
2. Force Analysis
3. Location Specific Analysis

4.1 Simple Model Analysis

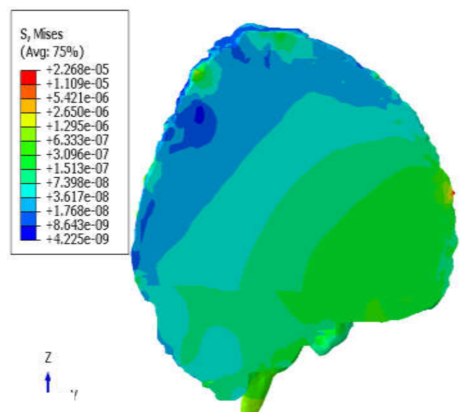
The simple finite element model was analyzed in response to external loading. The model consists of six layers namely scalp, skull, spongiosa, cerebro-spinal fluid (CSF), gray matter and white matter. This simple FE model was analyzed under sensitivity of elastic modulus, forces and locations respectively. In the following sections, the obtained results of each case are discussed.

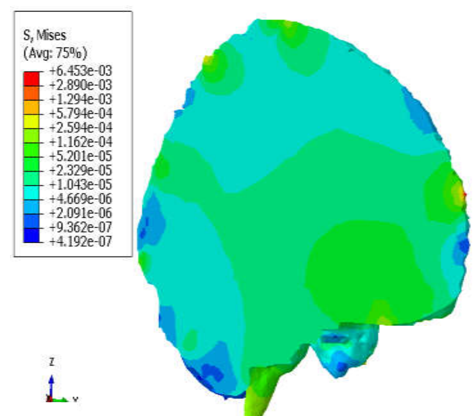
4.1.1 Case I: Sensitivity Analysis

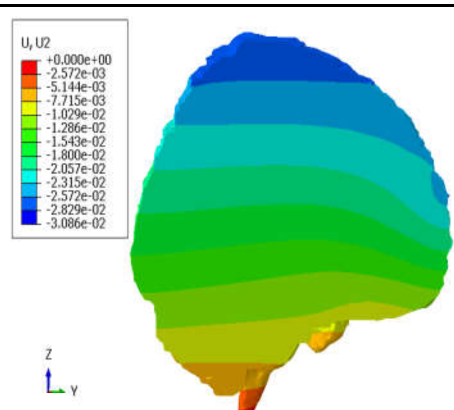
Head impact injuries are important to study because the mechanisms on the macro-scale are incomparable with micro-scale damage[46]. In this case i.e. Simple Analysis, sensitivity tests were performed using different values of elastic modulus at first. This is shown in Figure 4.1. The stress and displacement profile of the linear model is compared with that of homogenized model as discussed in section 3.1.3.2 of chapter 3. The maximum values for the Von Mises stress of both models (linear and homogenized) are observed at the points of loading (prefrontal region). The homogenized model shows the higher value and uniform distribution of the Von Mises stress than linear model (Figure 4.1 (a) and (b)).Higher values of stress are found in brain stem, frontal and temporal regions. However uniform pattern is observed in the central temporal area of the brain. In the peripheral region of the brain, random high and low values for the stresses can be seen.

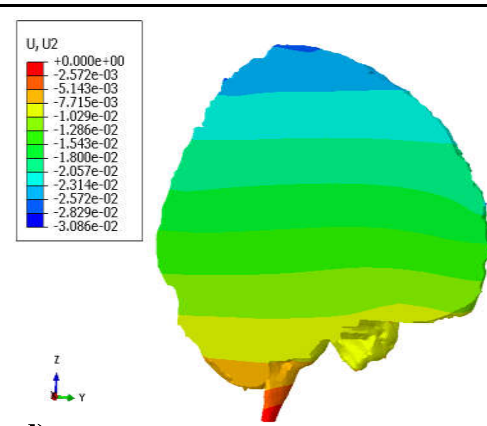
Under this analysis, displacements in y-direction (U2) are analyzed. This is because the displacements were calculated in the direction of the applied loads. The observed magnitude for both models is the same but the pattern of these distributions is different and visually can be distinguished. The maximum displacement occurs at the top of brain and minimum in the brain stem as shown in Figure 4.1 (c & d). In convention, the results displayed in the legends are in Mega Pascal (MPa) for Von Mises stress and in millimeter (mm) for displacement.

Figure 4.1: Sensitivity analysis of simple model. a-b) Stress profile. c-d) Displacement profile.

<table>
<tr><th>Model</th><th>Linear</th><th>Homogenized</th></tr>
<tr><td>Stress (MPa)</td><td>

a)</td><td>

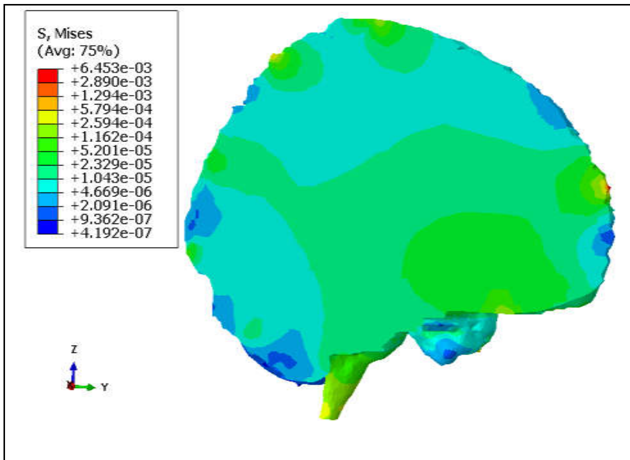
b)</td></tr>
<tr><td>Displacement (mm)</td><td>

c)</td><td>

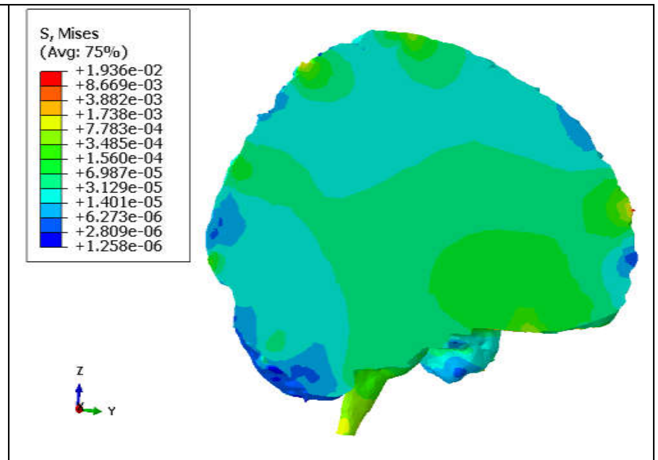
d)</td></tr>
</table>

4.1.2 Force Analysis

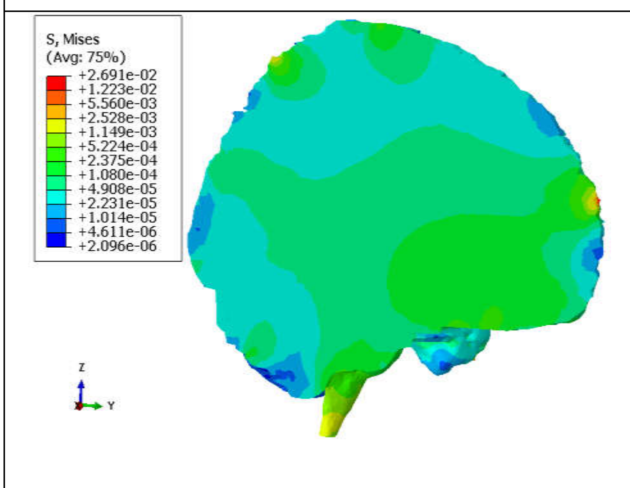
This case utilize varying amount of force to demonstrate head impact injury over time. In the force analysis, four different levels of forces are used. Again, the stress and displacement profiles were observed against each force. It was noticed that with an increase in amount of force, displacement and stress linearly increased. The highest Von Mises stress was found in the frontal region where the input force was applied (coup site).The maximum value of stress i.e 3.767 MPawas achieved at 7000 N applied force. The selection of the force was made using Nauhams's study[18]. The patterns of stress distribution in each amount of force were almost similar. The minimum amount of stress was observed in some peripheral regions of frontal and occipital lobes of the brain (Figure 4.2 (a-d)).The model showed a linear relationship of displacement with respect to applied force (Figure 4.3).



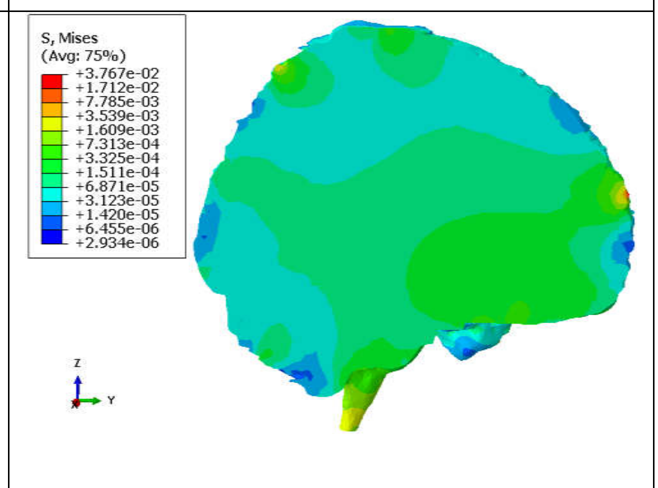
a) 1000N



b) 3000 N



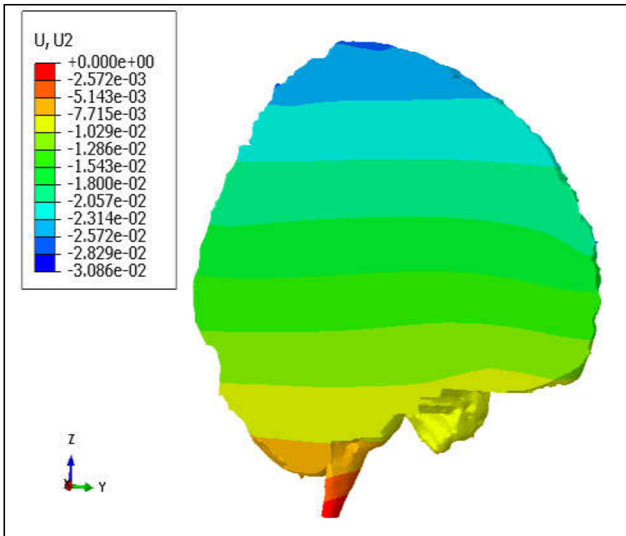
c) 5000N



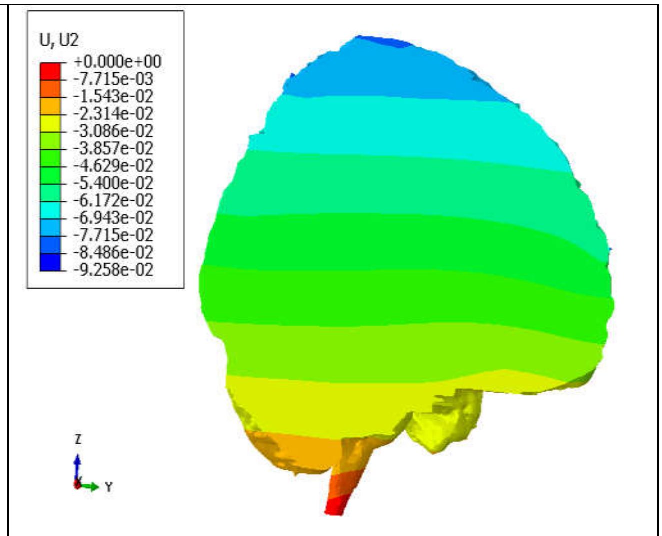
d) 7000N

Figure 4.2: Stress profile of force analysis on simple model. a), b), c), d) Stress at 1000 N, 3000N, 5000N, 7000N applied force respectively

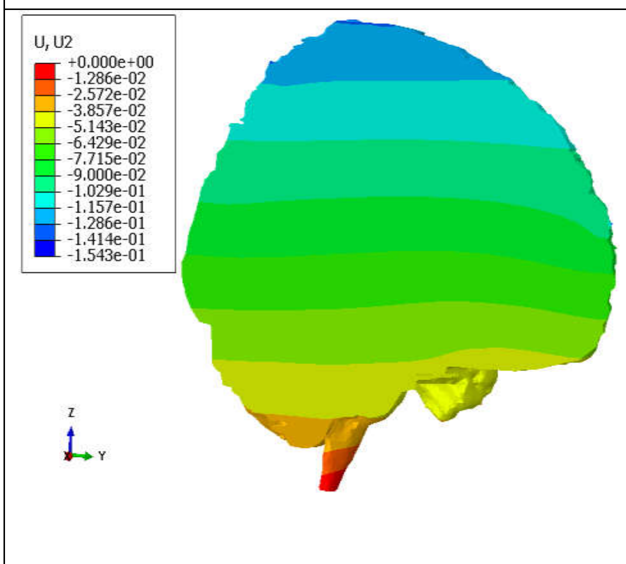
According to literature in case of dynamic studies which are mostly utilized for head impact problems, threshold values for stress causing brain injuries are ≥11 kPa, ≥15 kPa, ≥20 kPa in car, motorcycle and mild traumic injury respectively[47-49]. From the Figure 4.2 it can be seen that the stress at 5000 N and 7000N are above the threshold and hence indicating the brain injury at the loading location.



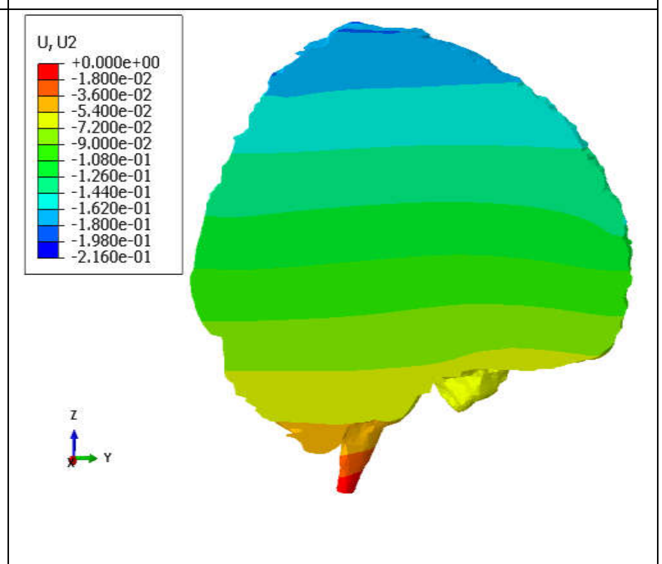
a) 1000 N



b) 3000 N



c) 5000 N



d) 7000 N

Figure 4.3: Displacement profile of force analysis on simple model. a), b), c), d) Displacement at 1000 N, 3000N, 5000N, 7000N applied force respectively.

The displacement profile for the force analysis is shown in Figure 4.3. The pattern observed in all four forces is similar but the magnitudes of the displacements are different. Maximum displacement can be seen on the top of the brain while it is negligible in the brain stem. The highest value of the displacement is achieved in this model is with 7000N force.

In the Figure 4.4, the relationship between force and displacement is shown. The graph shows that the changes are in linear fashion and indicating the brain as a stiff and an elastic material.

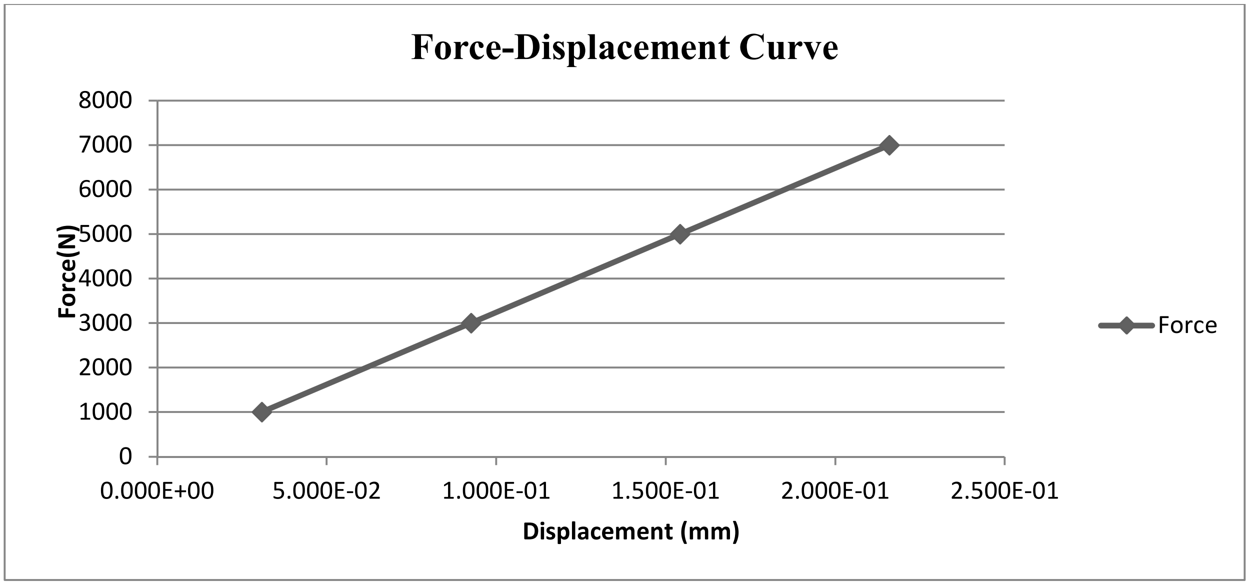


Figure 4.4: Force-Displacement curve for force analysis of simple model

4.1.3 Location Specific Analysis

In this, stress and displacement at multiple locations were analyzed. A 5000N force was applied to homogenized model at six different locations as to cover the brain from front, back, top and right side. From the obtained results it can be seen that in all cases maximum stress is at the location of loading i.e. coup site (Figure 4.5 (a-f)) except in Figure 4.5 b. The higher values can be observed in the lower temporal and occipital regions of the brain, while minimum is observed at coup site in frontal loading. This anomaly could be due to extra dense mesh developed because of interface between the two layers at coup site (Figure 3.8) variation in the trend was observed. A uniformly spread pattern is observed in the central region of the brain for all location. Figure 4.5 (a-f) shows the stress profile (cut plane sagittal view) of the brain for all loading locations. The maximum value of stress i.e 1.077 E-1 MPa (107.7 KPa) was observed during lateral or temporal loading and the minimum is observed during central loading i.e. 1.146 E-2 MPa (11.4 KPa). Assuming that the Von Mises stress ≥ 11 KPa (dynamic case) can cause brain injury, the predictions of this model suggest that more injury would happen in brain in temporal loading than in the central loading case.

For direction of displacement, displacements for all loading locations are observed. The y-displacements (U2) for prefrontal and occipital loading, z displacement (U3) for central, x-displacement (U1) for temporal and magnitude (U) for intermediate level of loading (frontal and parietal) are observed. These displacement profiles are illustrated in Figure 4.6 (a-f). In the study the levels of contours are adjusted for displacement plots only such that blue corresponds to maximum displacement and red to minimum value. From Figure 4.6 it can be seen that the minimum and lowest displacement is observed in the lower end of the brain stem. As the base of the model is fixed therefore almost zero displacement is observed at brainstem, in all locations. It should be noted that negative values in the contour shows the direction of displacement. The highest displacement is observed in temporal or lateral loading and different patterns are observed in different loading conditions. Although the negligible amount of displacement occurred in all cases, but it can be observed and analyzed with the help of FE analysis, which was nearly impossible to measure otherwise.

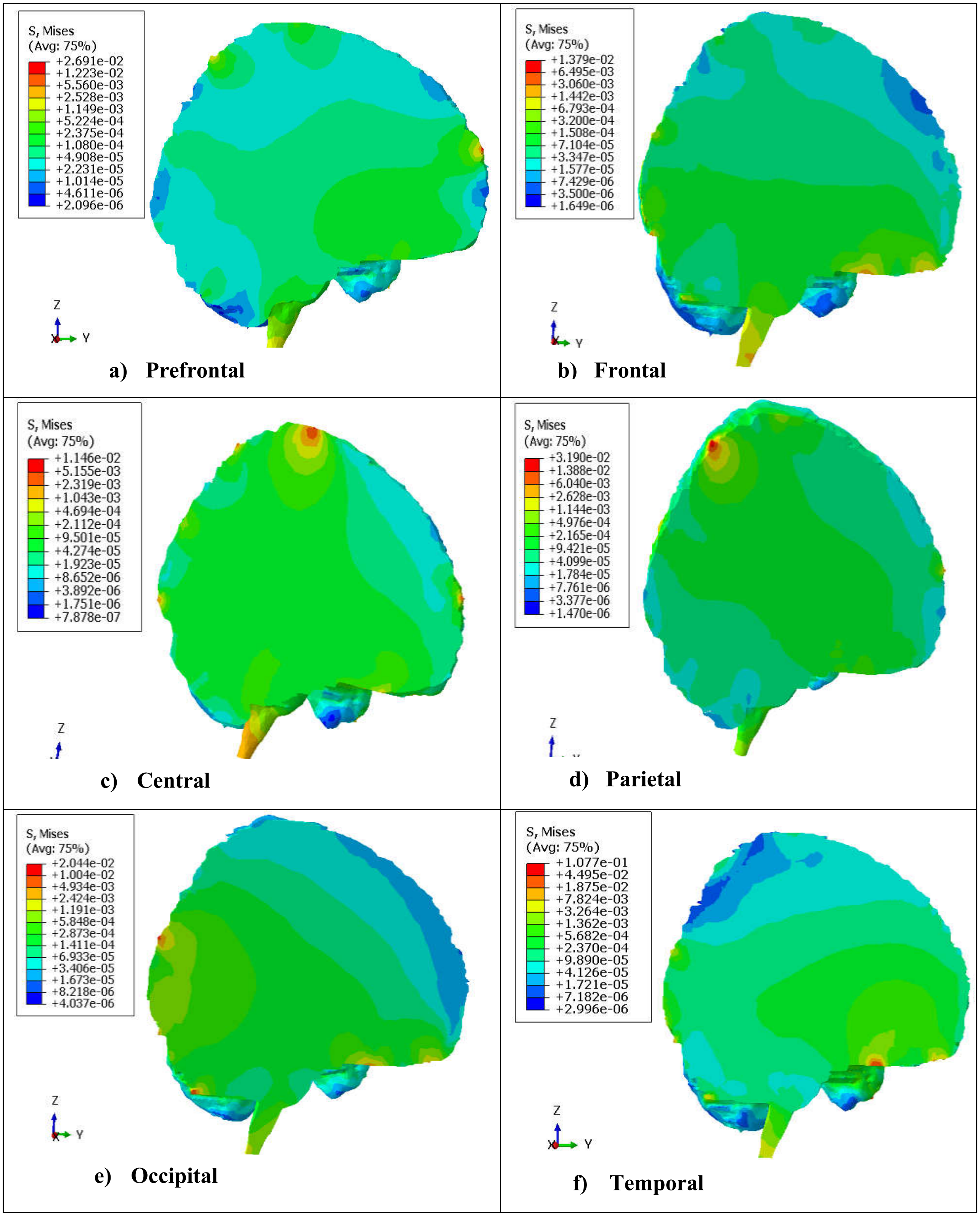


Figure 4.5: Stress profile for location specific loading in simple model.

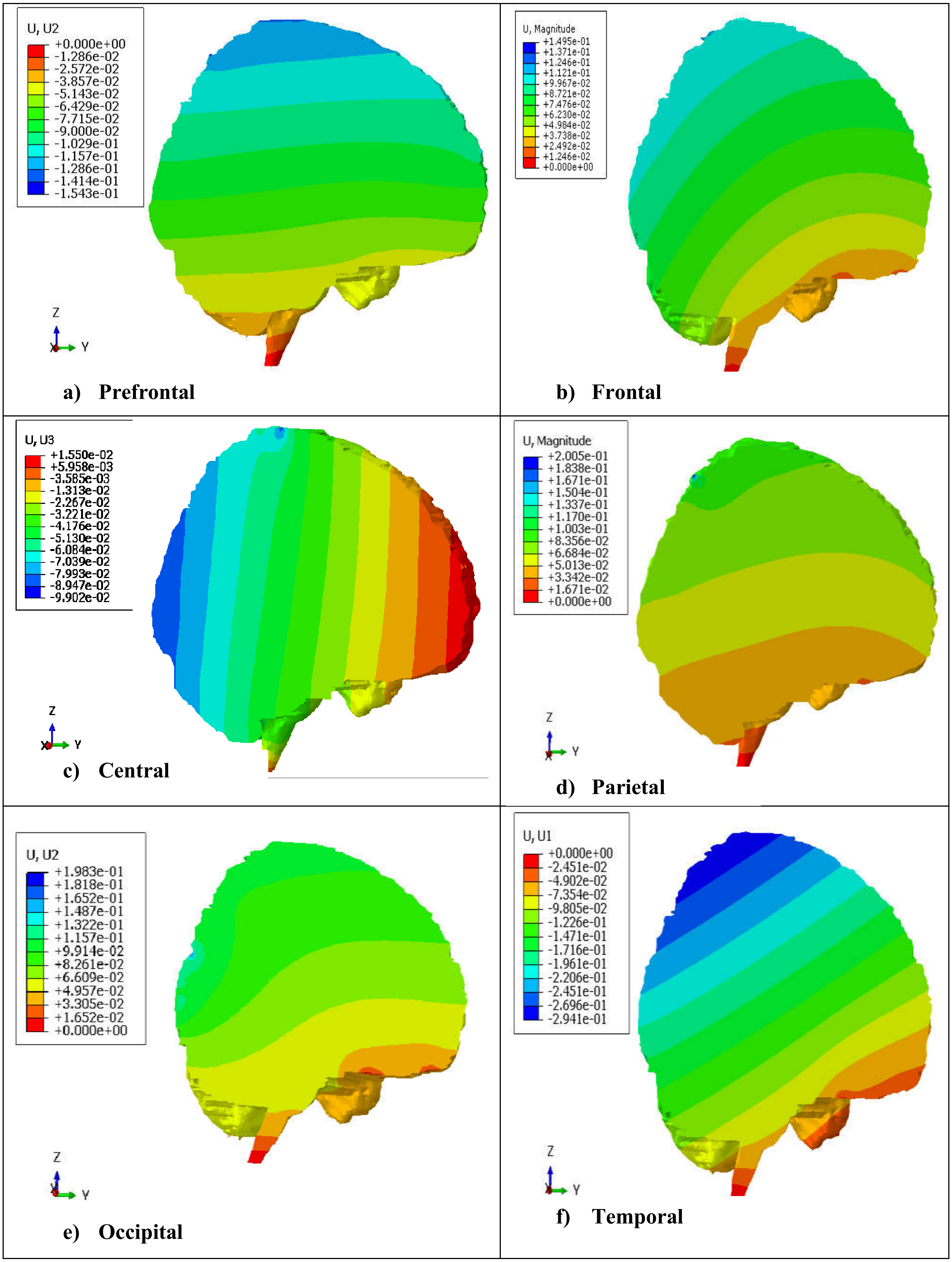


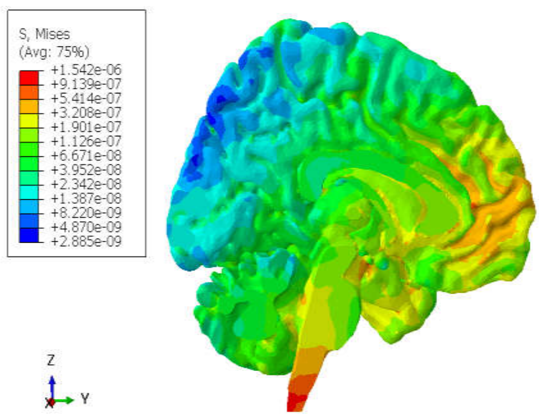
Figure 4.6: Stress profile for location specific loading in simple model.

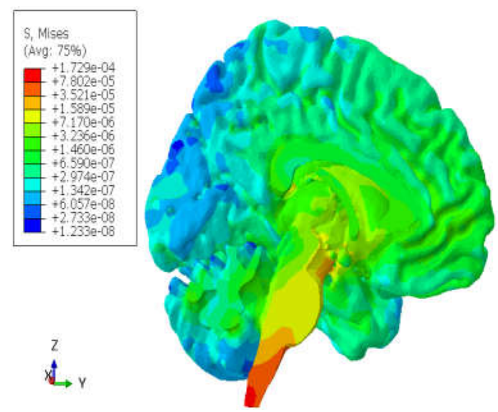
4.2 Complex Analysis

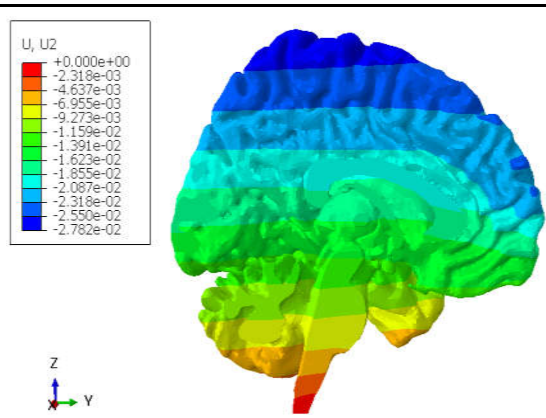
Like simple model, complex model was also performed for three cases. The complex model also had same six layers but with convulsions in the brain. In the following sections, the obtained results of each case can be seen.

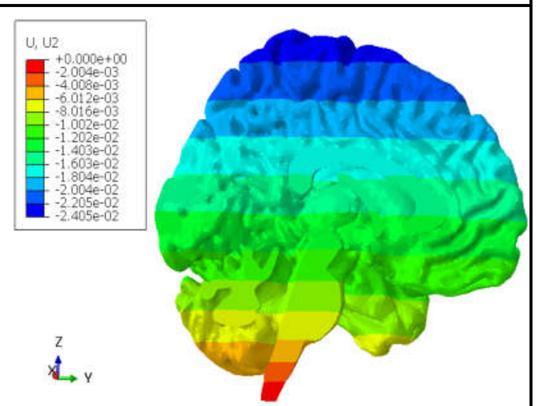
4.2.1 Case I: Sensitivity Analysis

In this case sensitivity tests to the complex model were performed using different values of elastic modulus at first. This is shown in Figure 4.7. The stress and displacement profile of the linear model is compared with that of homogenized model. Thirteen stress and displacement values of the color scale are used to measure the stress and displacement distributions quantitatively.

<table>
<tr><th>Model</th><th>Linear</th><th>Homogenized</th></tr>
<tr><td>Stress
(MPa)</td><td>

a)</td><td>

b)</td></tr>
<tr><td>Displacement
(mm)</td><td>

c)</td><td>

d)</td></tr>
</table>

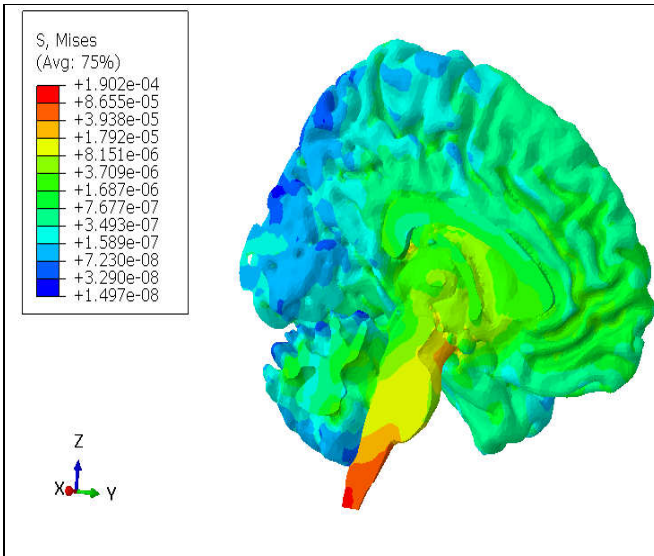
Figure 4.7: Sensitivity analysis of complex model. a-b) Stress profile, c-d) Displacement profile.

The maximum values for Von Mises stress of both models (linear and homogenized) are observed at the base of brain stem point. The homogenized model shows the highest value i.e 1.729 E-04 MPaand uniform distribution of the Von Mises stress (Figure 4.7 (a) and (b)). The higher values of stress are found in brain stem and mid brain region closer to the point of loading i.e. coup site.In linear model,however non-uniform and heterogeneous pattern is observed. Higher values in linear model can be seen in forebrain, mid brain and cerebellum.

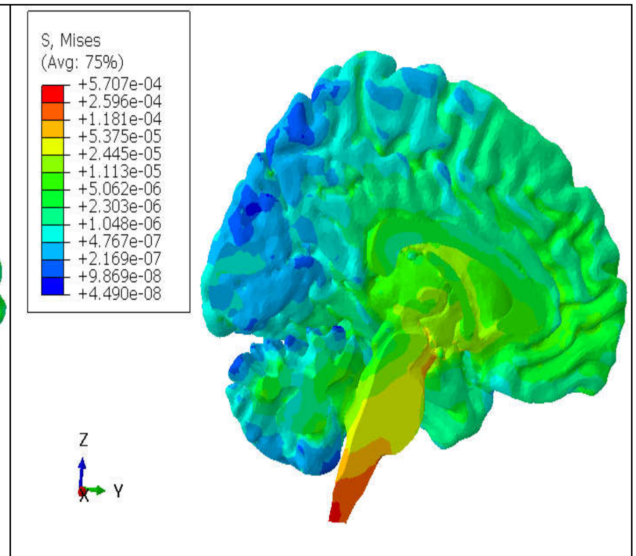
Under this analysis, displacements in y-direction (U2) are analyzed in the direction of loading. The observed magnitude for both models is bit different but the pattern of these distributions is similar. The maximum displacement occurs at the top of brain and minimum in the brain stem as shown in Figure 4.7 (c & d). In convention, the results displayed in the legends are in MPa for von mises stress and in mm for displacement. The negative sign shows the direction of the displacement. The blue color indicates the maximum displacement while red shows the minimum displacement in the brain.

4.2.2 Force Analysis

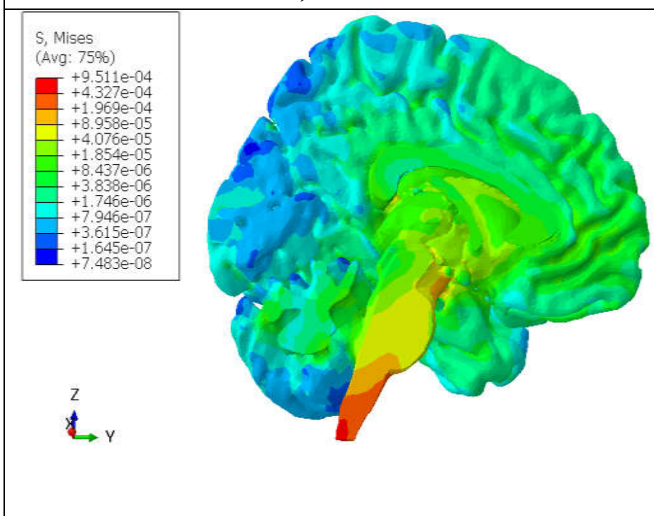
In the force analysis, four different levels of forces were used. The stress and displacement profiles were observed against each force. It was noticed that with an increase in amount of force, displacement and stress linearly increased. The model showed a linear relationship of displacement with respect to applied force (Figure 4.8). The highest Von Mises stress was found in the brain stem where the model was fixed. The maximum value of stressi.e 1.332E-3MPa was achieved at 7000 N applied force. The patterns of stress distribution in each amount of force were almost similar. The minimum amount of stress was observed in some peripheral regions of parietal and occipital lobes of the brain (Figure 4.8 (a-d)).



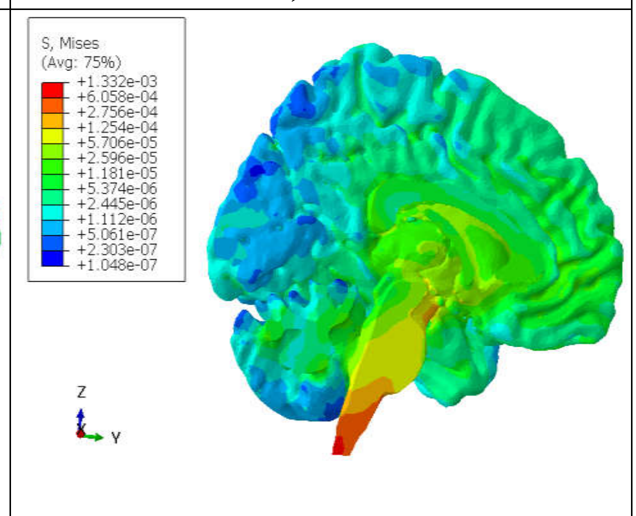
a) 1000 N



b) 3000 N



c) 5000 N



d) 7000 N

Figure 4.8: Stress profile of force analysis on simple model. a), b), c), d) Stress at 1000 N, 3000N, 5000N, 7000N applied force respectively.

When comparing these results with literature, it was found that the stress values achieved are significantly lesser than thresholds. The displacement profile for the force analysis is shown in Figure 4.9. The pattern observed in all four forces is similar but the magnitudes of the displacements are different. Maximum displacement can be seen on the top of the brain while it is negligible in the brain stem. Highest value of the displacement i.e. 1.724E-01 mm is achieved in the model with 7000N force. Negative sign in the contours indicates the direction of the displacement.

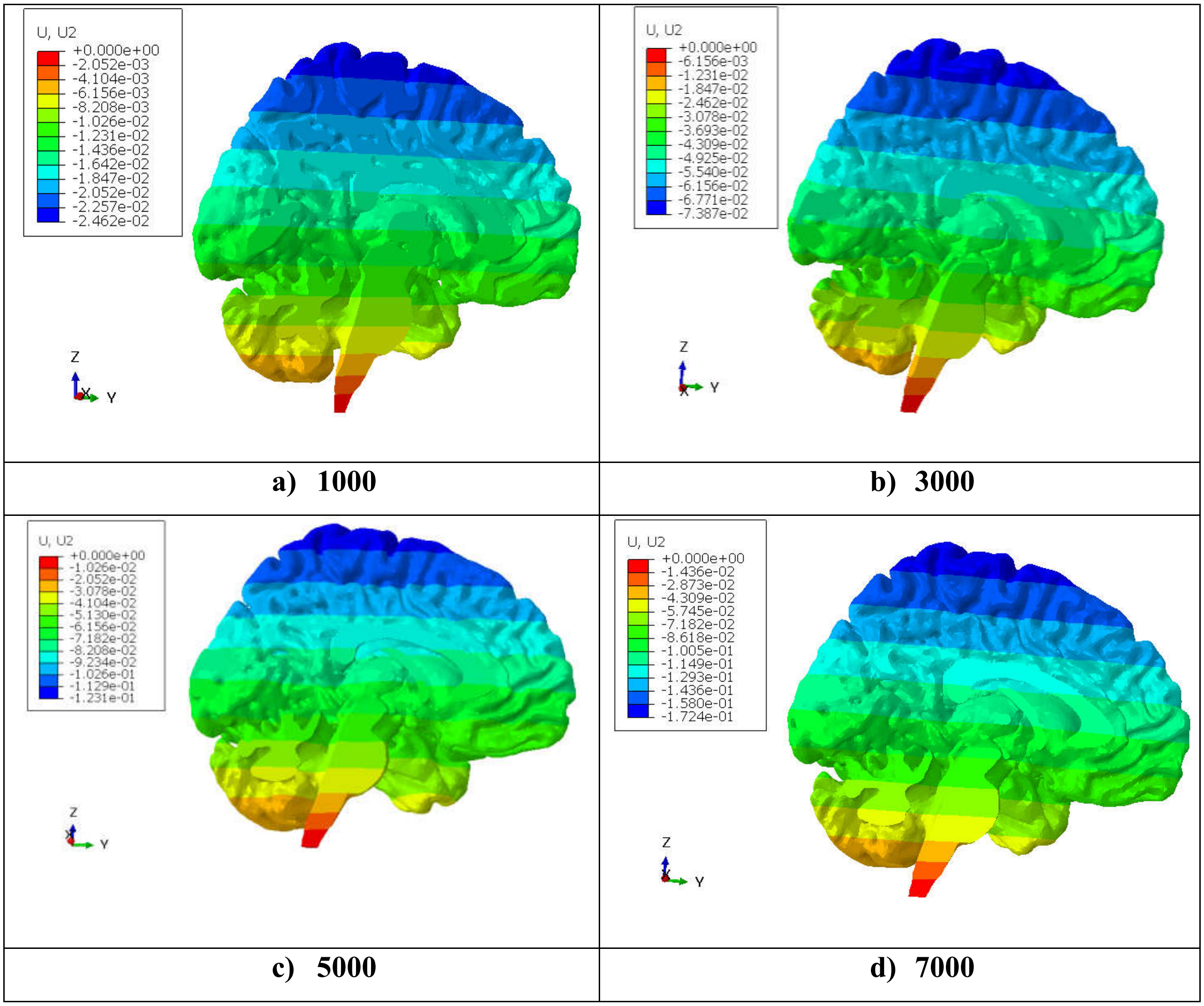


Figure 4.9: Displacement profile of force analysis on complex model. a), b), c), d) Displacement at 1000 N, 3000N, 5000N, 7000N applied force respectively.

In the Figure 4.10, the relationship between force and displacement is shown. The graph shows that the changes are in linear fashion and indicating the brain as a stiff and elastic material.

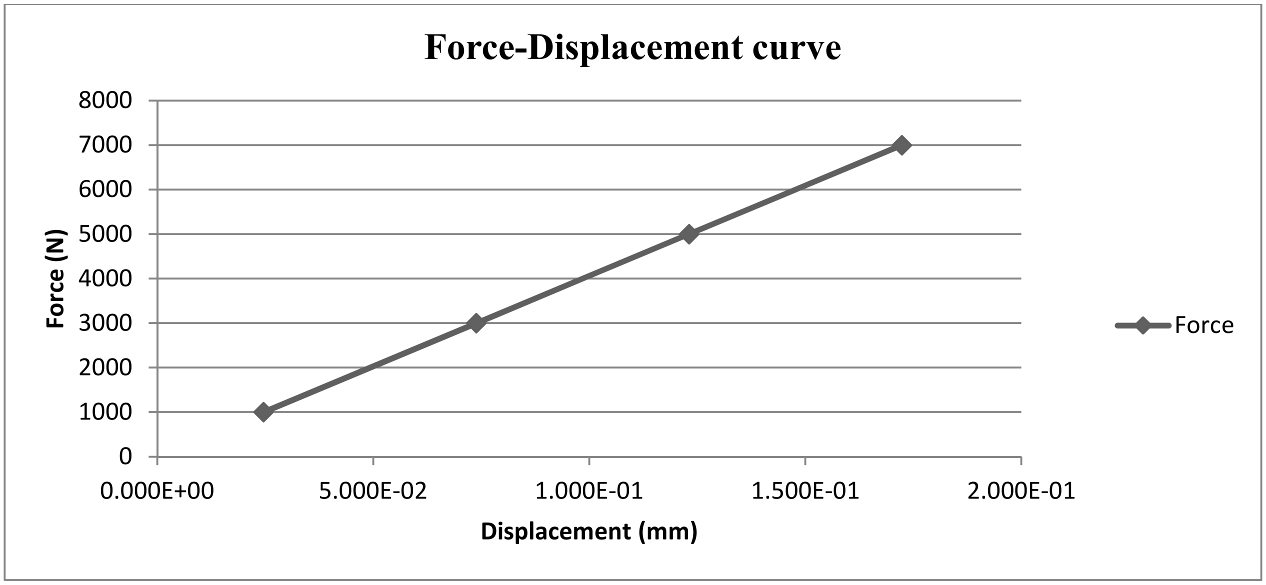


Figure 4.10: Force-Displacement curve for force analysis of simple model.

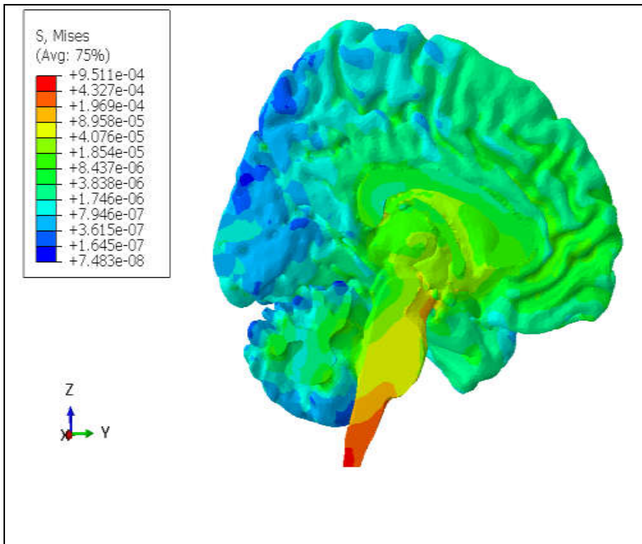
4.2.3 Location Specific Analysis

In this, stress and displacement at multiple locations were analyzed. From Figure 4.3(c and d) it was found that at 5000N and 7000N the stress values are above injury threshold 5000N. A force of 5000N was applied to homogenized model at six different locations i.e. covering the brain from front, back, top and right side. From the obtained results it can be seen that in all cases maximum stress is at the brain stem where the base of the model is fixed (Figure 4.11 (a-f)). The higher values for the stress can be seen in the mid brain and coup site (loading region) in each model.

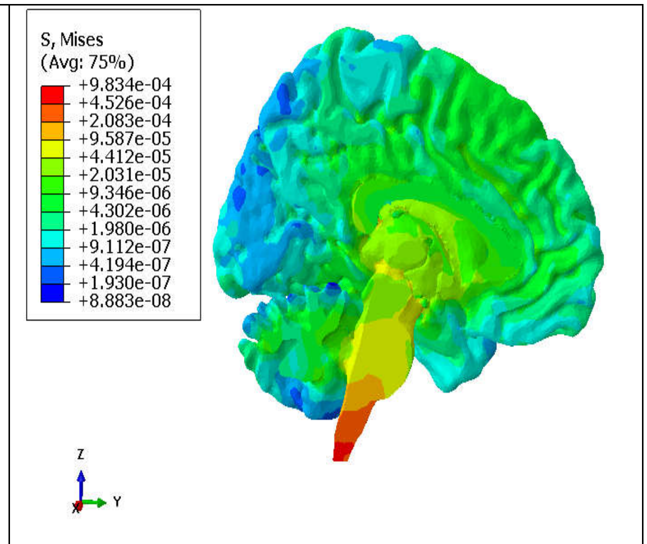
In prefrontal loading, higher stress is observed in forebrain (coup region and its nearby region) and mid brain and cerebellum of the hind brain. Minimum values can be seen in the parietal and occipital lobe. In frontal loading, trend is similar to prefrontal except that the minimum values are more confined in occipital region. In central loading, stress values are more in the central region and mid brain while minimum values can be seen in the frontal, temporal and occipital region. In parietal loading, uniform spread is observed in overall brain while only small region of the occipital lobe shows the minimum values. In occipital loading, the higher values are near coup site and mid brain, while lower values are observed in parietal and cerebellum region. In temporal loading, temporal lobe is the most affected area and shows the maximum stress in all cases, while least values can be seen in the peripheries of the other lobes.

Figure 4.11 (a-f) shows the cut plane sagittal view and stress profile of the brain for all loading locations. In this study, the complex model predicted higher and localized stresses in the brain stem and corpus callosum region for all loading locations (Figure 4.8). This may imply that these areas are are sensitive due to the head geometry, material properties and structural boundary conditions.

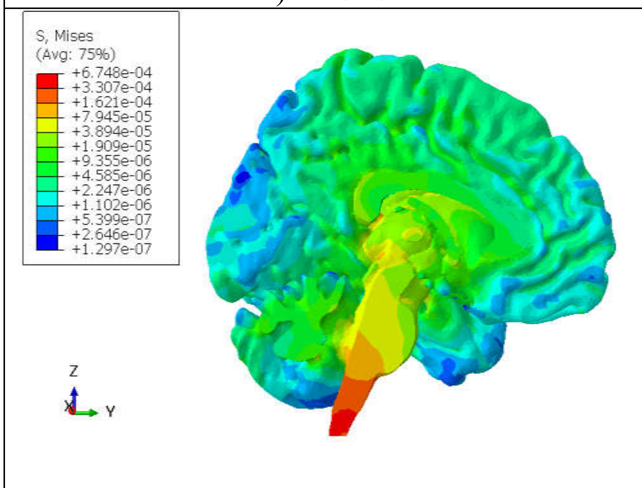
For direction of displacement, displacements for all loading locations are observed. The y-displacements (U2) for prefrontal and occipital loading, z displacement (U3) for central, x-displacement (U1) for temporal and magnitude (U) for intermediate level of loading (frontal and parietal) are observed. These displacement profiles are illustrated in Figure 4.12 (a-f). In the study the levels of contours are adjusted in such a way that blue corresponds to maximum displacement and red to minimum value. From Figure 4.12 (a, b, d, e, f) it can be seen that the maximum values are at the top of the brain and minimum or lowest displacement is observed in the lower end of the brain stem. In central loading (Figure 4.12 c), the observed maximum displacement occurred in forebrain region i.e. front side of the head. As the base of the model is fixed therefore almost zero displacement is observed at brainstem, in all locations. It should be noted that negative values in the contour shows the direction of displacement.



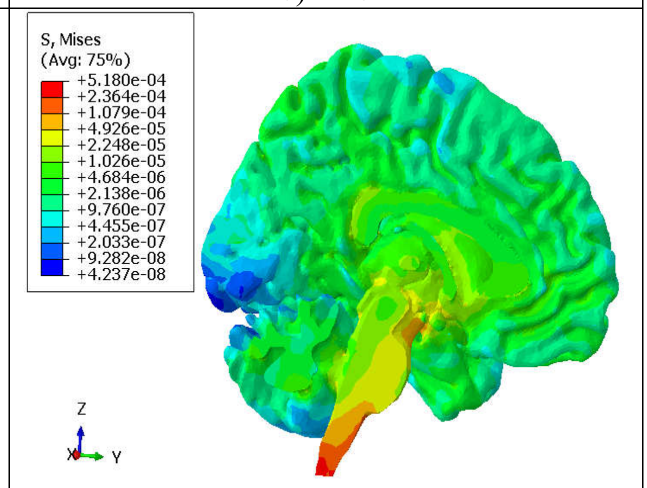
a) Prefrontal



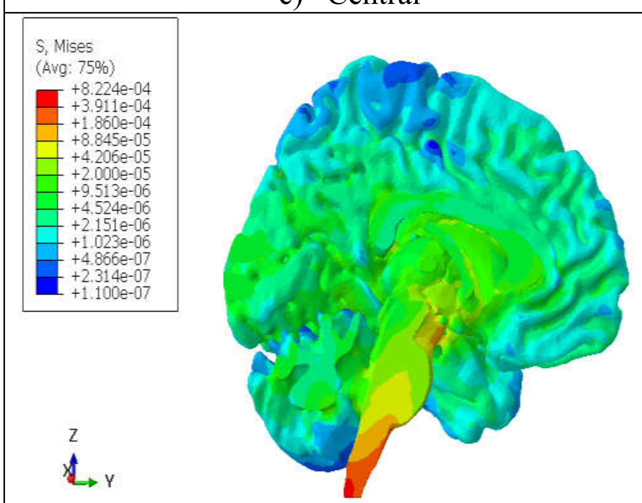
b) Frontal



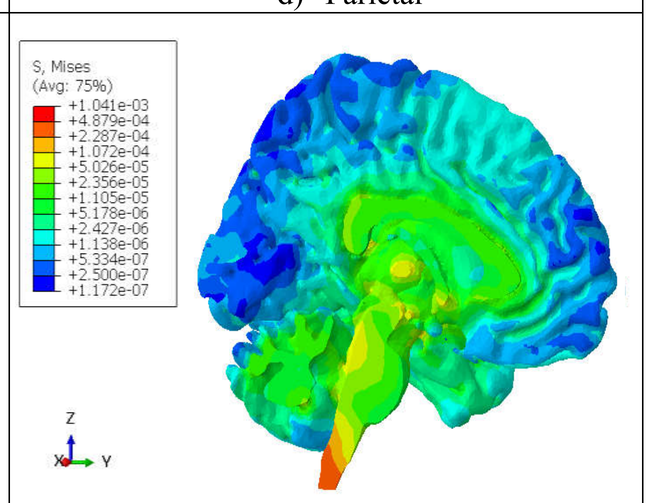
c) Central



d) Parietal

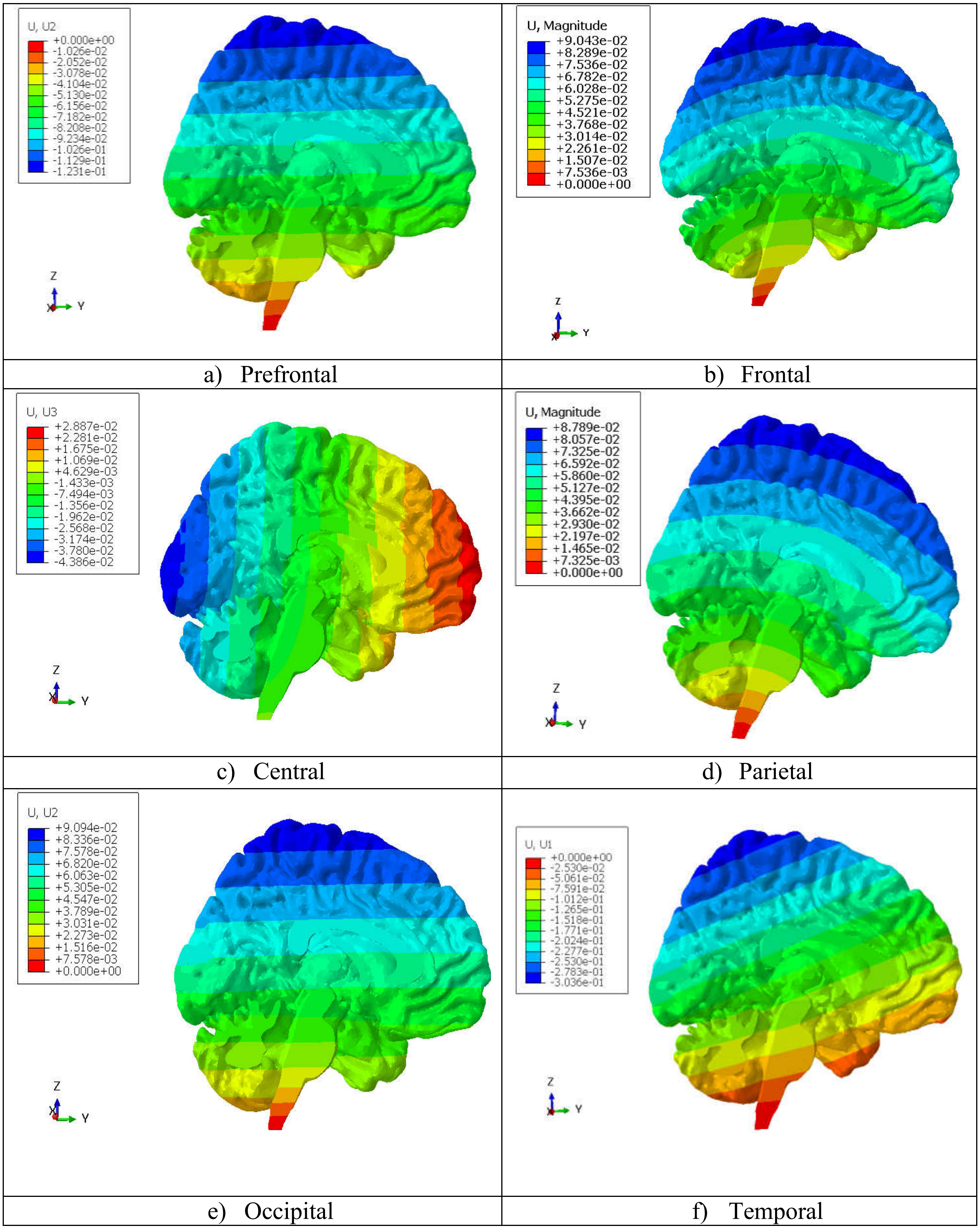


e) Occipital



f) Temporal

Figure 4.11: Stress profile for location specific loading in complex model.



a) Prefrontal
b) Frontal
c) Central
d) Parietal
e) Occipital
f) Temporal

Figure 4.12: Displacement profile for location specific loading in complex model.

4.3 Discussion:

A parametric designed human head model is developed and modeled in such a way that different stimulus impact in different locations are studied and analyzed. From literature it was found that neck constraint has not contributed considerably on the results in short impacts [16]. The model has head region only (without neck) with fixed base boundary condition, this contributes towards the reduction in performing calculations and processing time.The number of elements in meshes has been selected at certain level in order to get an efficient time computational solution. The model materials were considered to be isotropic, homogenous and linearly elastic in nature. To address non linearity in a linear model, concept of homogenized brain model has been introduced. Rule of mixture was used to calculate bulk modulus of the brain layer.

According to literature many researchers [10, 18, 23, 25, 27] validated their models with the pressure data and predict injury levels from Nahum et al.[18]. According to Bradshaw et al.[50], to predict injury strain is a definitive parameter than pressure. However, not that much data was found and in this thesis, the analysis is more confined on Von-Misses stress and displacement to be compared with different models.

From the above results it can be seen that both simple and complex model study showed sensitivity against elastic and effective modulus. Stress values are more sensitive to modulus than displacements. Higher the Young's modulus; higher the property of the material to with stand or resist the external load. In both models homogenized model is stiffer than linear model and showed and higher value of stresses and less or equal displacements as compared to linear model.

In force analysis,the stress values of simple model are much greater than complex model and linearly increasing. The complex model and simple model showed linear relationship for force displacement curve. The curve for complex model is steeper than simple model hence showing less displacement occurs in this model (Figure 4.13).

In our location specific analysis it was found that complex model has lower values of stress as compared to simple model because of refinement and detail in the meshes (Figure 4.14). The maximum stress was found during temporal or lateral loading. This is due to the geometrical complexity i.e. the temporal region has no spongiosa layer therefore the effect of the loading is maximum. Moreover, the oval shape of skull which was causing variation in observed brain damage. The directional dependence of computed brain response is consistent with experimental findings of decreased tolerance in a lateral impact [51]. Due to lack of any modeling and simulation literature of location specific analysis with such a small and sharp impactor and experimental results, models were verified by doing convergence test.

So from the above study it was found that the developed models when compared with each other, complex model showed less stress distribution and displacements. The complex model predicted no injury in any case while the simple model had shown in some cases. Hence it can be concluded that the complexity of the structure, geometry and the mesh density can bring significant changes and different responses under similar conditions. When these models are compared with the existing literature, they are found in a good agreement and consistent [3, 10, 21, 23, 27]. Quasi-statically the FE model exhibited the response as expected.

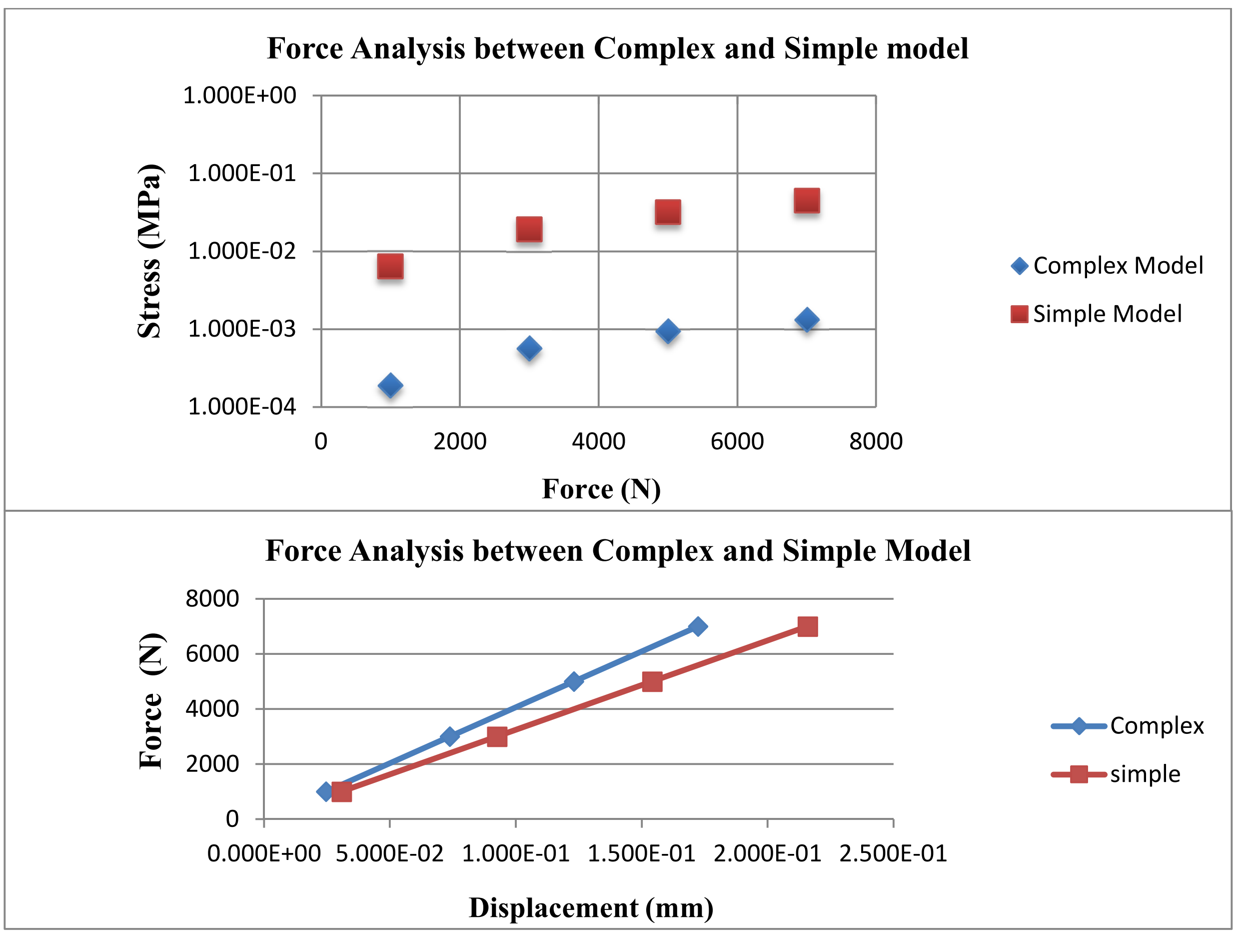


Figure 4.13: Comparison of Force analysis between complex and simple model.

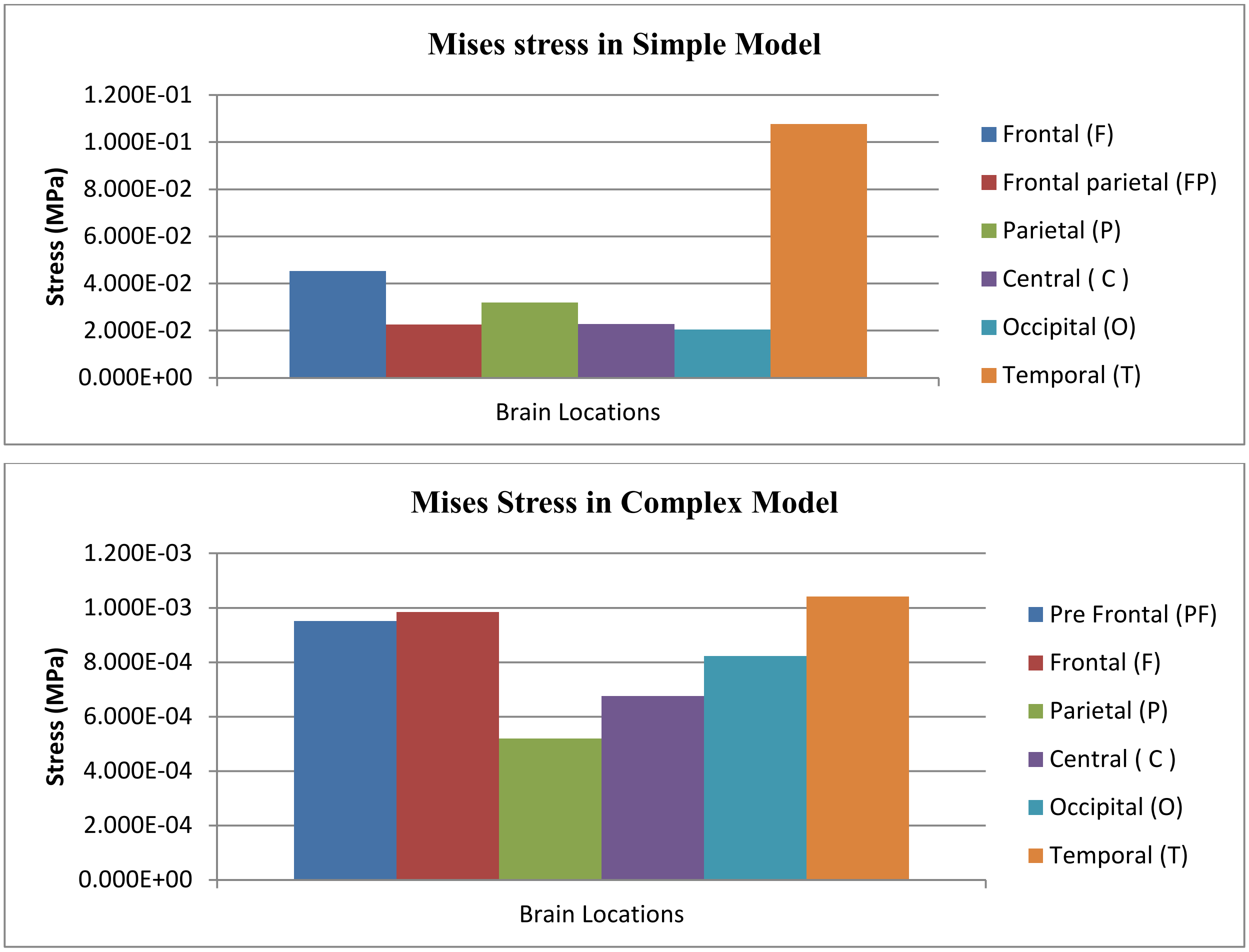


Figure 4.14: Comparison of Location specific analysis between complex and simple model.

4.4 Chapter Summary

In this chapter, the results have been presented. There is a noticeable change in Von Misses stress distribution and displacements when the change in modulus of elasticity of brain layer, geometry complexity and different locations of impact are introduced. Also, it has been concluded that the stimulation results depend predominantly on the geometry complexity, mesh density and the boundary conditions.

CHAPTER 5: CONCLUSION AND FUTURE PROSPECTS

The above mentioned approach is quite different from the usual manual creation of FE Models. Image processing has seen a lot of advancement in recent times which has allowed us to develop numerically reasonable head models which are immensely flexible. This approach not only allows us to tailor the head models according to our needs for specific scenarios but also allows us to capture the miniscule details and remarkable definition of the complex head and brain geometry.

Image-based modeling seems to have a promising future and with the persistent improvement of 3D-imaging methods, in future it will allow to us to construct extremely complex models, empowered by rapid advances in computing power. In future it will enable to investigate the use of principal component analysis for the generation of a population of head models which will facilitate and encourage the analysts to learn and study the changes in anatomy across the population according to impact response. The real challenge left to deal with is the synthesis of appropriate models for soft and hard tissue structures, as the constancy of the numerically predicted responses is based on realistic material models which are quite arduous to determine experimentally for high strain and amplitudes; strain rates in biological tissues.

5.1 Conclusion

3D simple and complex models of human head are developed. It can be concluded from this project, that the complexity in terms of geometry, material properties, loading and points of loading can greatly affect the overall behavior of the material. The accuracy of the predicted region of injury can be controlled by using effective modeling and simulation approaches.

Our study shows that elastic finite element analyses may greatly underestimate expected maximum stress values in location specific loading even when incorporating a certain degree of complexity in the geometry and material properties. From applications perspective, this model and investigative study can be useful for future research in more realistic modeling of human head injury. Human head image based modeling for research purposes for the first time is done by the Human Systems Lab, SMME, NUST. As a proud member of Human Systems Lab, SMME, NUST, my study will serve as a forward matter for hospitals, neural rehabilitation centers, and research and development organizations in Pakistan in the field of Neuro-modulation and neural rehabilitation.

5.2 Future Prospects

Following are some of the way forwards of this research (see figure 5.1).

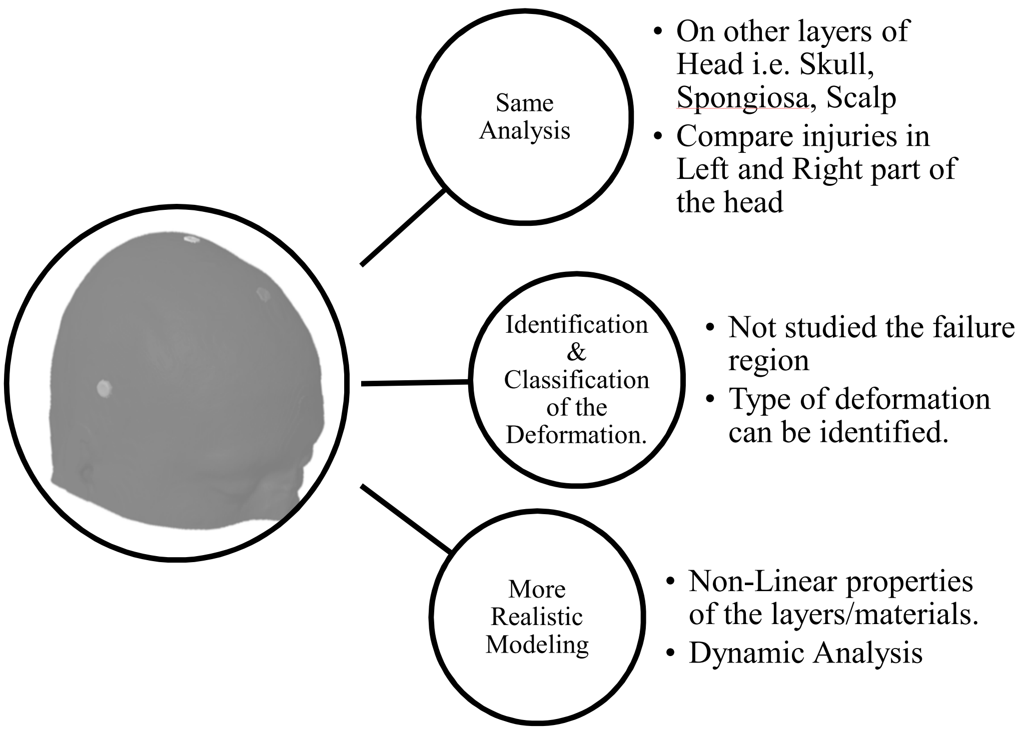


Figure 5.1: Future prospects of the developed study.

- This research has been based on effects of parameters on brain injury. The same model used in this study can be adapted for investigation of responses of other head layers e.g skull, CSF etc. Also this study has not considered the failure region, so identification and classification of the deformation can be done. The comparison or the left and right lateral impacts can be done in order to investigate the level of brain damage.
- The present study has been performed on static conditions. In future, time based dynamic analysis with non linear properties of brain layers can also be employed in the same models.
- The cost of computation (time and resources) remains a major challenge in computational modeling. The refinement in algorithms to reduce this cost is essential for making Image Guided Intervention a constructive tool for clinicians.

Appendix A

Convergence of complex model

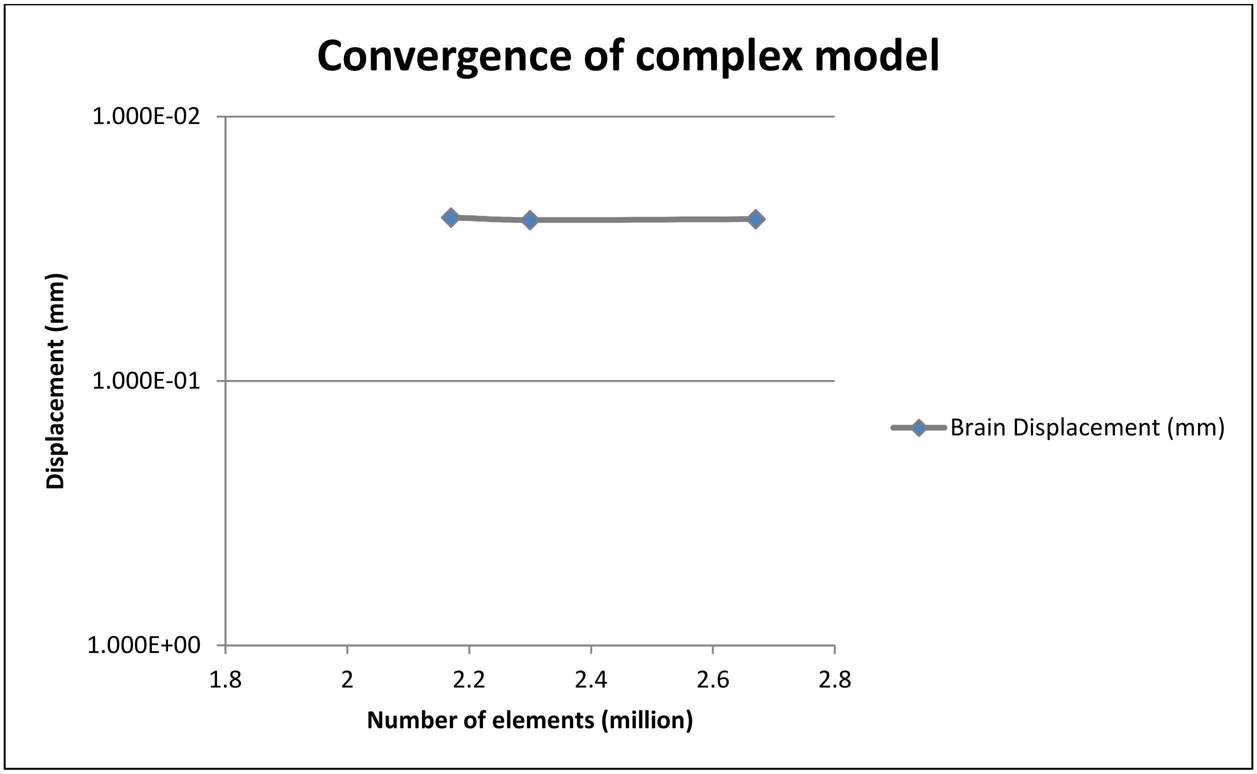


Figure A.1: Convergence of complex model of FE human head with tetrahedral elements.

Calculation of Effective Modulus for homogenized model.

The basic information required for the evaluation of the effective moduli is the volume fractions, and elastic moduli, of each phase.Voigt, proposed the effective elastic modulus of the composite as

$$E = E_1 v_1 + E_2 v_2$$

Where

E_1: Elastic modulus of Filler;

E_2: Elastic modulus of Matrix;

v_1: Volume fraction of Filler;

v_2: Volume fraction of Matrix.

&

$v_1 + v_2 = 1$.

To calculate volume fractions of the filler and matrix, volume brain and its layer should be known. These volumes are measured in Mimics suite.

Volume calculation

Total volume occupied by brain= $V_T = 1707188.46 mm^3$

Volume of White Matter layer= $V_1 = 1079851.08 mm^3$

Volume of Gray matter layer = $V_2 = 625819.28 mm^3$

Volume of Layers= $V_1 + V_2 = 1705670.36\ mm^3$

Volume occupied by voids= $V_{void} = (V_1 + V_2) - V_T = 1518.10 mm^3$

Volume fraction calculation

v_1 =volume fraction of filler or Brain = $\frac{\text{Volume of Layers}}{\text{Total volume occupied by brain}} = 0.9991$

According to literature: $v_1 + v_2 = 1$

$v_2 =$ volume fraction of matrix (Voids) $= 1 - v_1 = 0.8892 \times 10^{-3}$

Effective Elastic modulus

E_1= Linear Elastic modulus of brain= 6.75×10^{-1} MPa

$E = E_1 v_1 = 6.75 \times 10^{-1} \times 0.9991 = 6.743925 \times 10^{-1}$ MPa

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