DYNAMIC RESPONSE OF HEAD UNDER VEHICLE CRASH LOADING

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ABSTRACT

In this paper, a three-dimensional (3-D) nonlinear finite element (FE) method is used in association with the Articulated Total Body (ATB) biodynamics method, to study the human brain response under dynamic loading. The FE formulation includes the detailed model of the skull, brain, cerebral-spinal fluid (CSF), dura mater, pia mater, falx and tentorium membranes. The brain is modeled as viscoelastic material, whereas, a linear elastic material model is assumed for all other tissue components. Proper contact and compatibility conditions between different components are assumed. Instead of direct contact, inertial load resulting from the acceleration and deceleration of the head mass system is implemented. The ATB biodynamic package is used to simulate real vehicle impact scenarios, and to extract the six translation and rotation acceleration data at the center of the mass of the head component. These six-degrees of freedom (6-DOF) kinematic descriptions are used to represent the inflicted inertial loadings. The magnetic resonance imaging (MRI) outcomes, from two incidents with head impact, are compared with the biomechanical FE simulations to present the model capabilities. To examine and verify the material parameters used in FE formulations, experiments are conducted on a simulated brain material made from silicon dielectric gel. The results support that the combination of the FE deformation analysis and the ATB rigid body model is an effective method in head impact analysis and traumatic brain injury (TBI) identification.

Keywords: head impact, traumatic brain injury (TBI), Articulate Total Body (ATB) Model, dynamic analysis, three-dimensional (3-D) finite element (FE) model

INTRODUCTION

Every year vehicle crashes cause over a million fatalities and a hundred million injuries worldwide [1]. In the United States (U.S.), traffic accidents have been the leading cause of death for the age groups of 1 to 34, in recent years [2]. In European countries, 45,000 fatalities and 1.5 millions injuries were reported in 1995 [3]. The societal and economic annual cost of traffic accidents is estimated to be $200 million in the U.S. [4], and over $160 million in European countries [5].

Due to the devastating consequences of traumatic brain injury (TBI), crash analysis and head injury biomechanics are important fields in biomedical research. The head is routinely identified as the body part most frequently involved in life-threatening injuries in vehicular collisions [6]. In theU.S., approximately 2 million cases of TBI are recorded each year [7]. About one third of the hospitalized victims suffer from permanent disability [8]. Most of these victims undergo injury associated physical and psychological distress with a resulting high societal burden and cost. Although many injury protection devices, such as safety belts, airbags and helmets, have been developed and improved, traffic accidents are still responsible for most TBI cases [9]. Crash analysis of head injury biomechanics focuses on head impact and injury mechanisms that are very important in the development of effective TBI prevention and minimization strategies.

In an attempt to better understand head injury mechanisms, both clinical and laboratory studies have been conducted for decades. Mathematical models have been acknowledged as increasingly valuable tools in crash analysis. Sophisticated three-dimensional (3-D) finite element analysis (FEA) and rigid body biodynamic methods can be used to study impact injury events and the associated biomechanical response of the human head. The rigid body model is used to determine the gross dynamics and movements of a subject's head with the results introduced in finite elements (FEs) to determine local brain deformation. Complex geometry and constitutive models of multiple materials can also be employed under dynamic loading conditions. Combining these methods of analysis is time saving and improves effectiveness of the analysis [10].

The Articulated Total Body (ATB) Model, employed here for dynamic analysis of the body, is a validated 3-D rigid-body biodynamic model. This method has been successfully used by the Air Force Research Laboratory (AFRL) and other organizations for crash simulation and the prediction of gross human body response in crash dynamic environments. The ATB Model is quite general in nature and can be used to simulate the dynamic response of a wide range of physical problems approximated as a system of connected, or free, bodies and is not limited to crash dummy, or human, applications [11].

A 3-D FE brain dynamic analysis under impact is also employed in this paper. In a previous publication [12] the researchers reported the suitability of material modeling under frontal head impact scenarios. In this paper, the combination of FE and the articulated rigid body (ARB) dynamics is used to simulate and examine brain behavior under direct impact to the occipital portion of the head.

FE MODELING

The geometrical data for the development of the FEs of the human head, modeled here, represents a modification of existing geometric data obtained, and previously published, by Horgan [13]. A 3-D simulation, with multiple material model and load conditions, is then created. Altair Hyperwork 7.0 (Altair Engineering, Troy, Michigan) was used for FE pre-processing modeling and data post-processing.

The ATB modeling facilities are used to perform biodynamic simulation and to extract the head kinematic data under vehicle impact. These combined six degrees of freedom (6-DOF) acceleration data are used in the FE model for head response simulation. The 3-D FE model takes into account the detailed structure of the human head anatomy including the brain, falx and tentorium, cerebral spinal fluid (CSF), dura mater, pia mater, skull and scalp. The brain, CSF and skull are modeled as first-order eight nodded brick elements. The falx, tentorium, dura, pia and scalp are modeled as four-nodded membrane, or shell elements, with uniform thickness. Figures 1 and 2 show the 3-D FE model of these components.

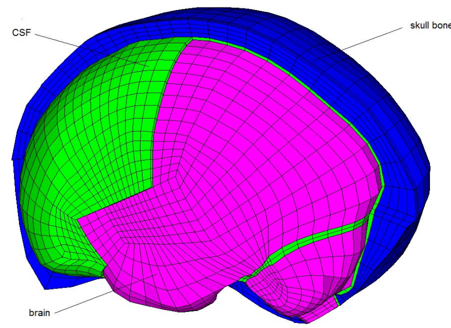

Figure 1. The right half model of brain CSF and skull bone.

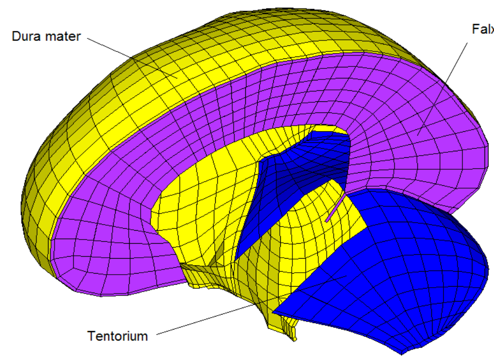

Figure 2. The right half model of dura mater, falx and tentorium.

The general-purpose 3-D nonlinear FE code LS-DYNA is used as the solver. The main solution methodology is based on explicit time integration using the central difference method differentiating scheme. The explicit method is computationally efficient due to the small time steps in this problem to assure the convergence and stability of the solutions. The entire duration of the crash analysis is typically 10-200 ms and small time steps are required, therefore, and are suitable for a converged and accurate solution procedure [14].

Using a Lagrangian formulation [15], the time-dependent finite deformation of continuum material can be expressed in terms of convected coordinates X_j, and time t:

$$x_i = \phi_i(X_j, t) \quad (1.)$$

where ϕ_i is the mapping function between the reference configuration and the current configuration.

The equation of motion (balance of momentum) describing continuum deformation states that:

$$\frac{\partial P_{ji}}{\partial X_j} + \rho_0 b_i = \rho_0 \ddot{u}_i \quad (2.)$$

where P_{ij} is the nominal stress, b_i is the body force density, ρ_0 is the density in the reference configuration, u_i is the displacement of a material point and $\ddot{u}_i$ is the acceleration. By integrating the equation of motion over the reference configuration, we have:

$$\int_{\Omega_0} \delta u_i \left(\frac{\partial P_{ji}}{\partial X_j} + \rho_0 b_i - \rho_0 \ddot{u}_i \right) d\Omega_0 = 0 \quad (3.)$$

Applying the derivative product formula, and the divergence theorem:

$$\underbrace{\int_{\Omega_0} \left(\delta u_i \rho_0 \ddot{u}_i \right) d\Omega_0}_{\delta w^{kin}} + \underbrace{\int_{\Omega_0} \left(\delta \left(\frac{\partial u_i}{\partial X_j}\right) P_{ji} \right) d\Omega_0}_{\delta w^{\text{int}}} - \underbrace{\int \delta u_i \rho_0 b_i d\Omega_0 - \sum_{i=1}^{n_{SD}} \int_{\Gamma_{t_i}^0} \delta u_i \bar{t}_i^0 \, d\Gamma_0}_{\delta w^{ext}} = 0 \quad (4.)$$

where δw^{int} δw^{ext} and δw^{kin} are, respectively, virtual internal work, virtual external work and virtual inertial work. The surface tractions are denoted by $\bar{t}_i^0$ denotes the surface tractions and SD is the number of space dimensions. By discretizing the domain into a Lagrangian mesh of FEs, where the geometry and field variables are described in terms of shape functions $N_I(X)$, the stationary form of the equations is finally written in the simple form of the balances of inertia forces (mass matrix times acceleration) , internal and external forces as:

$$M_{ijIJ} \ddot{u}_{jJ} + f_{iI}^{\text{int}} - f_{iI}^{ext} = 0, \quad \text{with}$$

$$M_{ijIJ} = \delta_{ij} \int_{\Omega_0} \rho_0 N_I N_J d\Omega_0, \quad f_{iI}^{\text{int}} = \int_{\Omega_0} \frac{\partial N_I}{\partial X_j} P_{ji} d\Omega_0 \quad \text{and} \quad f_{iI}^{ext} = \int_{\Omega_0} N_I \rho_0 b_i d\Omega_0 + \int_{\Gamma_{t_i}^0} N_I \bar{t}_i^0 \, d\Gamma_0 \quad (5.)$$

where the upper-case indices stand for the node number and the lower-case indices stand for the directions number. The time rate equations part of the equation (5) can be solved using the central difference explicit time integration method. For this purpose, the time domain is divided into a sequence of time steps and the solution is sought with the marching in time.

MATERIAL CONSTITUTIVE MODEL

The FE analysis of head impact biomechanics is usually based on small deformations of elastic, or viscoelastic, material. The assumptions of linearity, homogeneity and isotropy are used for the head tissues in this work (see Table 1). The linear elastic material model is used for the skull, scalp, dura mater, pia mater, falx and tentorial membranes. A linear elastic model is also used for the CSF. Low shear modulus and high bulk modulus were used to simulate incompressibility. Fluid option is used for the CSF linear solid elements, in which the deviatoric stress is eliminated for fluid like behavior.

Table 1.
Tissue structure and finite element model

Tissue	Anatomical structure 50th perc. male	Constitutive model	Finite element model	# of Elements
Scalp	5-7mm thick	Linear elastic	6 mm thick shell element	2064
Skull	195mm length, 155mm breadth 225 mm height 4-7 mm thick	Linear elastic	Solid element	8256
Dura, falx, tentorium	1 mm thick	Linear elastic	1 mm thick membrane element	2622
Pia	1 mm thick	Linear elastic	1 mm thick membrane element	2786
CSF		Low shear modulus, high bulk modulus incompressible	1.3 mm thick solid element	2874
Brain	165 mm length 140mm transverse diameter	Homogeneous linear viscoelastic material	Solid element	7318

A linear viscoelastic material model is used for the brain tissue. The shear relaxation behavior is described by the Maxwell model as:

$$G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t} \quad (6.)$$

where G_0 is the short term shear modulus, G_{∞} is the long term shear modulus, and β is the decay factor. Table 2 shows the material properties used in the FE model.

Brain Substitute Material

Since it is not feasible to use actual brain tissue in collision simulation, it is necessary to find a suitable modeling material. The material should be viscoelastic and have a complex modulus similar to brain tissue when subjected to shear strain.

Studies from different species (human, porcine, bovine) and from different parts of the brain (white matter, cerebrum, brainstem) have consistently demonstrated that the viscoelastic properties of brain tissue fall into a predictable range [16-26]. Silicone dielectric gel (specifically Dow Corning Sylgard 527 A&B) has been demonstrated to have viscoelastic properties approaching those of actual brain tissue and has gained widespread acceptance as a physical substitute for brain tissue [16, 25-28]. It is worth noting, however, that the gel exhibits a lesser degree of dynamic deformation because the phase angel of the gel material increases at a greater rate with respect to frequency than does brain tissue at frequencies above 1 Hz. This means that the gel exhibits greater viscous damping than brain tissue at finite strains. In other words, the gel material provides an accurate estimate of the response of brain tissue to oscillatory shear strains below 1 Hz in frequency and a conservative estimate above 1 Hz. The gel is, therefore, an excellent alternative for benchmarking studies [16, 25].

The suitability of silicone dielectric gel is confirmed through the independent testing in this research using the accepted technique of measuring the complex modulus of brain tissue by applying an oscillating shear strain and measuring the resulting strain and the phase shift between input stress and output strain. Testing is conducted using an Advanced Rheometric Expansion System (ARES) Rheometer (LS714306), from TA Instruments, in the University of Minnesota's Rheological Measuring Laboratory (serial no. 199815770). Figure 3 shows the results of this testing, along with results published by Brands, et al. [16], which includes their tests of the same gel material, the results of their testing with several samples of porcine brain tissue and other findings in the literature, which include several human brain tissue samples.

Table 2.
The head tissue material parameters used in the finite element model

Tissue	**Young's mod.** (GPa)	**Density** (kg/mm^3)	**Poisson's ratio**
Skull	8	1.21×10^{-6}	0.22
Dura mater	0.0315	1.13×10^{-6}	0.45
Dura tentorium	0.0315	1.13×10^{-6}	0.45
Dura falx	0.0315	1.13×10^{-6}	0.45
Pia mater	0.0115	1.13×10^{-6}	0.45
Scalp	0.0167	1.0×10^{-6}	0.42

Tissue	**Young's mod.** (GPa)	**Bulk mod.** (GPa)	**Shear mod.** (GPa)	**Density** (kg/mm^3)	**Poisson's ratio**	**Static shear mod.** (GPa)	**Dynamic shear mod.** (GPa)	**Decay const.** (ms^{-1})
Brain	667×10^{-6}	2.19		1.04×10^{-6}	0.49999635	5.28×10^{-5}	1.68×10^{-5}	0.4
CSF	667×10^{-6}	2.19	5.0×10^{-7}	1.004×10^{-6}				

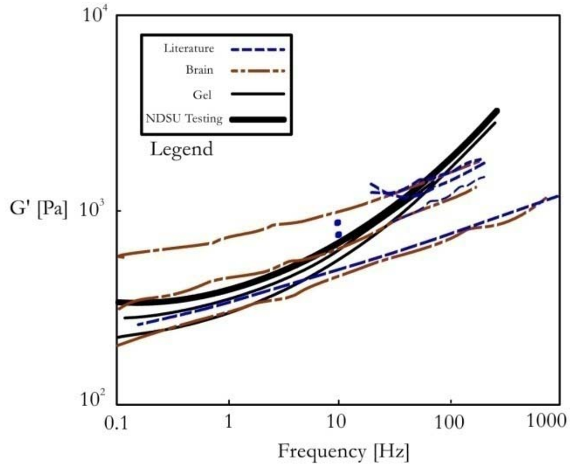

Figure 3. Our (NDSU) independent testing of storage modulus of silicone dielectric gel over a range of frequencies in addition to results for gel and several samples of brain tissue originally reported in Brands et al. [16]. Literature data includes porcine cerebrum [18, 20], porcine brainstem [21], calf cerebrum [19], human cerebrum [22], and human white matter [23].

BOUNDARY CONDITIONS

The free-boundary condition is defined as the junction of the neck and the head in the model. This means that there is no constraint effect at the head-neck joint, since for short duration impacts, such as 6 ms in Nahum's frontal impact [29], the neck does not influence the dynamic response of the head. Further, in inertial loading analysis, the 6-DOF kinematic description is sufficient and no extra boundary conditions are required.

According to the anatomical structure and physiology of the human head, the interfaces between the scalp and skull, the skull and dura, as well as the brain and pia, are modeled as a tied surface-to-surface contact definition. The interface between the dura, tentorium and falx is defined as a tied node-to-surface contact model as these components physically adhere to each other. A tied contact algorithm is preferred for the brain-membrane interfaces because it transfers loads in both compression and tension; only loads in compression, however, are transferred in a penalty contact algorithm, so a gap will be created in the contrecoup region where tensile loading is possible in frontal impact. [30]

Due to the presence of CSF, and the fact that the relative motion between the skull and brain during impact has been observed [30], the interfaces between the dura, pia, falx, tentorium and CSF are modeled as an automatic surface-to-surface sliding contact with a friction coefficient of 0.2, as previously reported [31]. This contact definition is also appropriate for the simulation of CSF fluid behavior by linear solid element methods.

HEAD RESPONSE ANALYSIS UNDER VEHICLE CRASH LOADINGS

Head impact against padded, or rigid, surfaces is a common and important source of loading to the human brain. In the modeling presented here, an ATB biodynamic package is used to reconstruct impact scenarios of a real vehicle. Six translation and rotation acceleration data are extracted at the center of gravity (CG) of the head. Since the combined acceleration data reflects the head kinematics, restraint system interaction and head/neck reaction forces during the impact events, these DOF kinematic descriptions can be used to replicate the angular and translational acceleration momentum and resulting inertial loads experienced by the head tissue system in the dynamic conditions of a vehicle crash [32]. The head kinematic data from the ATB is then applied to the FE model to replicate the head biodynamic response in car crash cases. Finally, the mechanical response outputs are compared with the magnetic resonance imaging (MRI) observations of brain tissue injury to validate the simulation methodology.

ATB Simulation

The ATB computer program is a 3-D, rigid-body dynamic crash simulator developed jointly by the National Highway Traffic Safety Administration (NHTSA) and Armstrong Aerospace Medical Research Laboratory at Wright-Patterson Air Force Base (AMRL/WPAFB) to predict human body dynamics during events such as automobile collisions, pilots' ejections and other hazardous events [33]. The Generator of Body Data (GEBOD) preprocessing program is used to generate the necessary input parameters for ATB, including geometric and mass properties of various body segments and locations and range of motion characteristics of joints [34]. This system can be used to predict both human and manikin body motion, as well as to provide injury assessment. ATB is used here to simulate the actual incidents and to determine the motion of the head for further FE simulation in cases for which MRI data is also available.

Case I: This case represents a rear end collision simulation. A tested impact scenario is based on a simulated human male subject weighing 79.3 kg (175 lbs) with a height of 1.8 meters (71 inches). The subject is positioned with a head separation of 7.5 cm (3 in) and a head rotation of 30º to the left at the time of impact, striking the headrest with a head angulation of 70º yaw, 0º pitch and 0º roll.

The impact scenario consists of a rearward acceleration resulting in a change in velocity of approximately 12.9 km/h (8mph). The translational and rotational accelerations, in three directions, are shown in Figures 3 and 4.

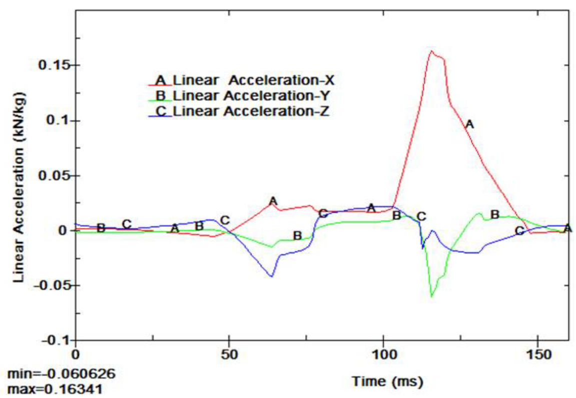

Figure 4. Translation accelerations in inertial coordinate

Case II: The second analysis is performed for a direct head impact in the occipital area. In this case, a female who was 1.5 meters (61 inches) tall and weighing between 54-59 kg, (120-130 lbs), loses her balance and strikes a rigidly attached wooden structure with the back of her head. At the time of impact, her head velocity is approximately 6.5 meters per second (4 mph). The translational and rotational accelerations, in three directions, are shown in Figures 5 and 6.

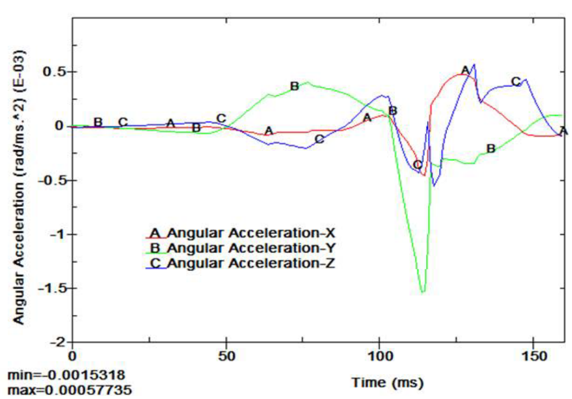

Figure 5. Rotational accelerations in inertial coordinate

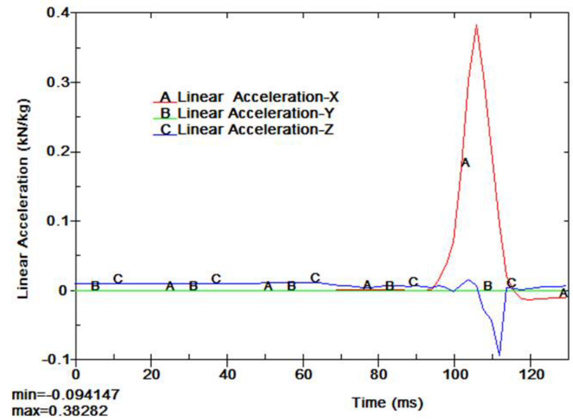

Figure 6. Translation accelerations in inertial coordinate

FEA of Inertial Loading Response

The FE head model and material model are used to analyze the inertial loading response of brain under these two scenarios. The translational accelerations from ATB simulation are applied at the center of mass of the head model, which is rigidly connected to the skull to introduce the loading to the entire system. The skull solid elements are defined as solid body to apply the rotational accelerations. Because head soft tissue injuries are known to occur without large deformations of the skull, the rigid skull assumption is reasonable for the analysis of soft tissue response under dynamic loading [32].

The Mechanical Response and Soft Tissue Injury

Brain soft tissue injuries result from the combination of many biomechanical factors such as the material nature of brain tissue, anatomic structure of head and brain tissue, kinematics and other constraints. Basically, the brain deforms when exposed to rapid momentum change due to direct impact forces, or non-contact forces, transferred through the neck as a result of the velocity differences between the head and human body.

Brain tissue is resistant to the dilatational deformation and hydrostatic stress due to the high bulk modulus. Due to the low shear modulus, however, the internal anatomical structure of the head-brain complex and the angular kinematic loading under impact conditions, brain tissue injury, such as, diffuse axonal injury (DAI), may be developed from shear deformation and shear stress [32].

Figure 7 compares the FE solutions for maximum shear stress distribution with the MRI observations of brain tissue injury for case I at 114 ms, the peak time of the impact for the case. In Figures 8 and 9, the FE solutions for the variation of maximum shear stress and maximum shear strain in the brain, from 114 ms to 116 ms, are shown.

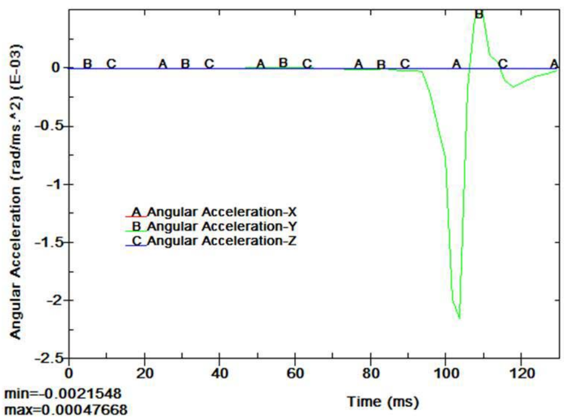

Figure 7. Rotational accelerations in inertial coordinate

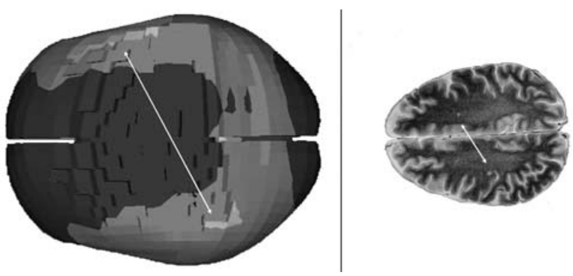

Figure 8. Case I brain shear injury MPI and maximum shear stress at 114ms (top-horizontal view)

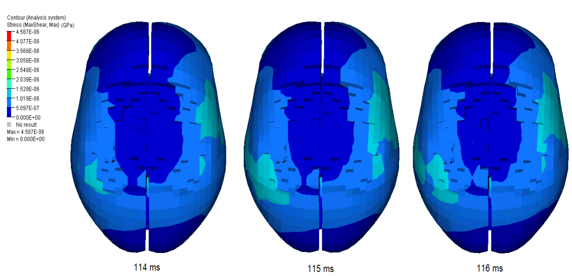

Figure 9. **Case I brain maximum shear stress (top-horizontal view)**

The MRI data for the cases are from patients referred for clinical evaluation and were obtained following informed consent under IRB approval. The red arrows in the Figures indicate the sites of maximum shear stress and observed shear injury of brain tissue. In Figures 10- 12, the maximum shear stress distribution is compared with the MRI brain tissue injury for case II at the peak impact time of 105ms from different view directions. In Figures 13, 14 and 15, a maximum shear stress variation of the brain from 104 *ms* to 106ms is shown from different view directions for this case.

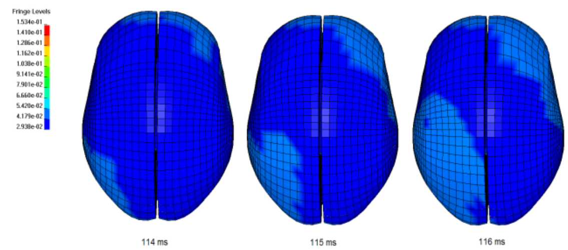

Figure 10. Case I brain maximum shear strain (top horizontal view)

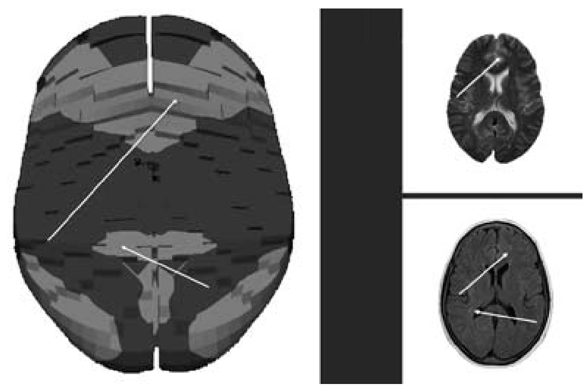

Figure 11. Case II brain shear injury MPI and maximum shear stress at 105ms (top-horizontal view)

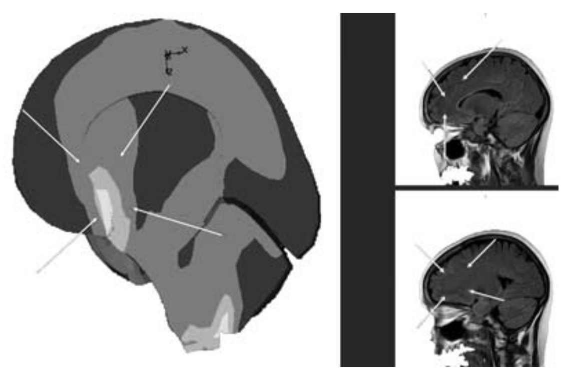

Figure 12. Case II brain shear injury MPI and maximum shear stress at 105ms (mid-sagittal view)

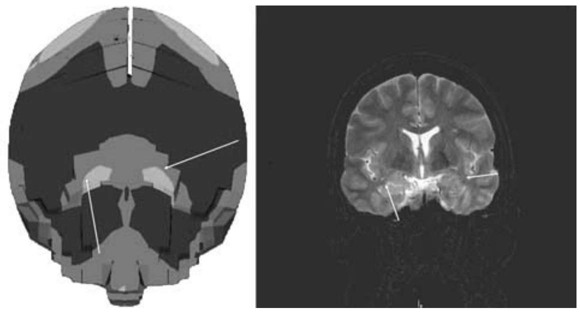

Figure 13. Case II brain shear injury MPI and maximum shear stress at 105 ms (coronal view)

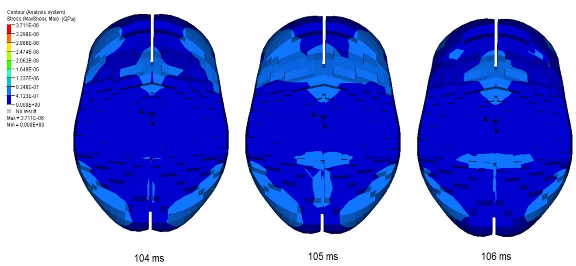

Figure 14. Case II brain maximum shear stress (top-horizontal view)

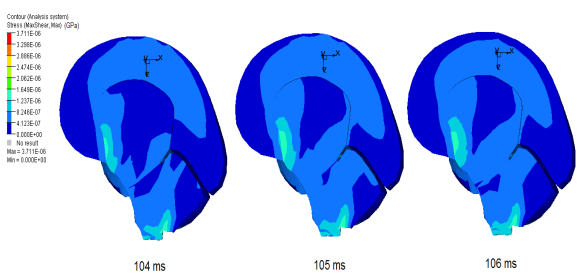

Figure 15. Case II brain maximum shear stress (mid-sagittal view)

Discussion

A good correlation between the internal injury sites and high shear stress regions is demonstrated. The FE head model accurately identifies and predicts locations of internal brain injury associated with blunt trauma as validated here. The maximum angular acceleration experienced by the head is 1532 rad/s^2 in case I, and 2155 rad/s^2 in case II.

These accelerations are in the range of the published values known to cause TBI in the human brain [35-36]. The type, magnitude, duration and direction of acceleration loads all play important roles in brain injury mechanisms.

CONCLUSIONS

The combination of FE deformation analysis and an ATB rigid body model is an effective method in head impact analysis and TBI identification. More real accident simulations can be done to test the accuracy and the validity range of the head model. Parametric analysis of crash simulations can be done to study the brain injury mechanism.

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