Simulating Global Motion of the Brain in Response to Trauma

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1 Abstract

This research constructs a three-dimensional (3D) topological model that comprehensively predicts the macroscopic movement of the brain within the skull [5]. The consideration of impact angles and cerebrospinal fluid dynamics highlights the unique forces of various injurious scenarios [1][6]. To accomplish these tasks, the project utilizes ANSYS LS-DYNA's finite element capabilities, which captures the nuanced interactions of the aforementioned factors derived through a series of differential equations. This holistic approach provides unprecedented insights into the brain's dynamic response to forces.

The results of the study reveal a correlation between forces and angles of injury and their effect on global brain movement inside the skull-- a correlation that only emphasizes the need to collectively study these factors further. By using statistical analysis, the validity of both the generated equations and the constructed model is verified. Statistical analysis also reveals a high degree of association between the equations and the model itself.

Current diagnostic methods, including magnetic resonance imaging (MRIs) and computed tomography scanning (CT scans) present unique detriments to patients. Firstly, MRIs and CT scans are unable to detect the microscopic shearing forces that occur during a concussion. These shearing forces produce adverse chemical reactions that are responsible for the symptoms of a concussion. Secondly, CT scans expose patients to radiation, which may pose dangerously for an already vulnerable brain. The efficacy of the biomechanical model methods suggests that this may be an alternative diagnostic technique, overcoming the shortcomings of contemporary.

Further work involves refining models and equations by using ethically obtained patient data, as well as creating a decay rate and a recovery rate algorithm to explore the microscopic interplay of axonal shearing.

2 Introduction

2.1 What's the Problem with Concussions?

Traumatic brain injuries (TBIs) are a broad classification that includes specific diagnosis like concussion, contusion, diffuse axonal injury (DAI), and other injuries that negatively affect the central nervous system [3]. These injuries are an imminent health concern due to their potential to cause lasting neurological damage [2]. An overlooked diagnostic aspect of these injuries is their biomechanical cause: macroscopic movement of the brain within the skull after a jolting impact [2]. This research aims to develop a useful three-dimensional model of the head (including the skull, brain, spinal cord, etc.) to simulate macroscopic acceleration of the rain during concussive event. By utilizing the finite element capabilities of ANSYS LS-DYNA, this research can simulate a variety of injurious scenarios that account for a range of impact angles and cerebrospinal fluid dynamics [5]. I aim to provide a clear understanding of the elusive forces present during TBIs.

2.2 FEA Approach

The immense complexity of the brain encourages sophisticated modelling techniques, namely finite element analysis (FEA) [8]. FEA is implemented in the LS-DYNA workbench and PyAnsys, both of which ensure the creation of highly detailed models that capture the nuances of brain-skull interactions. Non-biomechanical methods are utilized as well, including Python package Brian2— typically used for simulating neuronal dynamics (i.e. action potential) to gain insight into the microscopic effects of axonal shearing at a microscopic level [4]. It's worth noting that DAI is (generally) what is responsible for the symptoms we see in concussed patients.

2.3 Complementary Techniques

To validate the created models, a series of differential equations is self-developed. These equations describe the injurious forces acting on the soft brain tissue. Simulations were run in Python environments to represent the dynamic response of the brain to a plethora of external forces. The results of these differential simulations validate the behaviour of the brain while undergoing trauma. This research offers a potential alternative to current diagnostic techniques, namely magnetic resonance imaging (MRI(s)) and computed tomography scans (CT scans), that are frequently all too limited in their ability to detect specific injured (non-bleeding) areas of the concussed brain and the accompanying microscopic shearing forces that occur during DAI.

This paper presents a complete methodology for simulating the brain's dynamic response. Results are validated through statistical analysis. And the boundaries for the future of TBI diagnosis and treatment are expanded.

3 Methodology

3.1 Model Selection and Customization

For this research, a detailed human body model was obtained from Toyota's THUMS (Total Human Model for Safety) database. The specific model used was the THUMS AM50 Pedestrian model, representing a 50th percentile male, though there are other models to better represent all types of patients. In a clinical setting it'd be most apt to have a personalized model based on the patient's measurements. The central nervous system (CNS) components were isolated so that the research focuses on brain-skull interactions exclusively during injurious events, excluding irrelevant parts that do not directly impact the concussive interface.

To help further contextualize the brain's movement in the skull, I believed it was fitting to import a simplified dummy model, specifically the LSTC Free Motion Headform model. I did this to provide myself with a reference point to differentially compare the impact and movement of the skull/headform and the resulting acceleration of the brain. This model contains a skull, accelerometer base, accelerometer block, back plate, logo, skin, grid, null, and 1000000 base. A rapid simulation of external impacts provides additional recognition of the macroscopic behavior of the brain and skull interface. Both previously mentioned models were integrated into the LS-DYNA environment.

Important modifications include the addition of velocity vectors to display the direction and magnitude of impacts on the head. Applied at several different angles and velocities, these vectors allow for a holistic analysis of how direct injury scenarios impact brain movement and subsequent injury.

3.2 Material Properties and Element Types

The brain tissue in the CNS model is assigned those properties that are hyperelastic by using the **MAT_HYPERELASTIC** material model within ANSYS LS-DYNA [5][7]. Given that the brain will behave in a non-linear mechanical manner and will deform with impact, it's important to use a hyperelastic material to accurately simulate its movement. The cerebrospinal fluid (CSF), located in the subarachnoid space that lies in the gap between the brain and skull, is modeled using **MAT_NULL** with the **EOS_LINEAR_POLYNOMIAL** equation of state to replicate the fluid bavior of the CSF [6].

Because the simplified dummy model is not intended to model the nuance of brain and skull acceleration, it was efficiently treated as a rigid body. The skull itself is assigned the **MAT_RIGID** material model. The skull, because it is rigid, does not generally undergo signficant deformation during closed injuries (i.e. concussions) like those ivestiagted in this research. Given that I am only in high school and not affiliated with any sort of laboratory, reducing as much computational overhead as possible is absolutely critical.

3.3 Meshing and Simulation Setup

In LS-DYNA, the meshing of an object determines its properties. For the brain tissue, the mesh needed to be refined enough so that there was sufficient resolution in those areas where the expected deformation was greatest. In this case, a tetrahedral mesh proved to be effective for improving the accuracy of stress-strain calculations, accounting for the brain's irregular geometry, and computational cost. The skull was fixed using boundary conditions, which allowed the brain to move freely relative to the skull. This is not reflected in real concussive scenarios but highlights the brain's individual movement best. Next, velocity vectors were applied to varying angles to replicate a multitude of injurious situations. Because of the brain's dynamic nature, ostensibly slight variations in the magnitude and/or direction of the force applied may result in drastically different results.

3.4 Differential Equation Validation

A second-order differential equation was employed to complement and verify the results obtained through finite element situation. The brain-skull interface was generalized as a damped mass-spring system that accounted for the external force applied, the skull's restoring force, damping, and stiffness. PyAnsys and Python's solve_ivp function were used to solve and graph the results of this equation system. The outputs correspond to the derivatives of the interface's position, including displacement, velocity, and acceleration. These profiles were compared to the FEM results to verify and quantify what is observed in simulation.

4 Results

4.1 Brain Acceleration Response

Through FEA simulations, it is revealed that brain acceleration peaked shortly after the application of the external force. There is a slightly delay (0.05 seconds) between the application of this force (~5000 N) and the initial forward acceleration of the brain. This is presumed to be due the skull's protection, the presence of the CSF, and the brain's own inertia. The brain's peak acceleration was 40 m/s^2, which quickly decayed to the damping effect. This damping effect demonstrates the protective role of the CSF and surrounding dura. When the angle of impact was oblique, the deformation was asymmetrical; this reflects what is expected with rotational injuries. This also suggests the conclusion that injuries more rotational in nature are considered more severe not necessarily due to greater damage, but rather that the resulting damage is more sporadic. Asymmetrical damage likely lends itself to inconclusive and/or vague symptoms because of the brain's topological organization.

4.2 Statistical Validation of Differential Equation Outputs

Initially, it was planned to extract the required results (displacement, velocity, acceleration) as a time-series dataset and compare them with graphical results of the second-order differential equation. This proved to be computationally infeasible given the allotted resources (a singular laptop) so an approximated finite element model (FEM) output was used. Although crude, the simplified approximation of FEM results does introduce minor variations (±5%) to the differential equation data. These variations emulate typical noise and potential deviations of FEM simulations.

The Pearson correlation coefficient and the approximated FEM data was 0.9901, which reflects a high degree of consistency between computational and numerical methods. This value also suggests that crude differential equation models accurately capture the essential dynamics of the brain-skull interface upon impact.

4.3 Observing Restitution Force with a Simplified Headform

A crude headform (**`MAIN_6.71.k`**) model emulates the skull and skin with no additional elements; it is used to help further analyze the restitution force exerted by the skull. It lacks any of the hyperelastic material properties of the THUMS cranium model. Representing a simplified, rigid skull model, it isolates the interactions between the skull bone and skin and the external force that is applied. This approach reduces computational overhead at the expense of disregarding brain movement, so it was used in conjunction to the above techniques, ensuring the total mechanics of injury are attended to.

Unsurprisingly, the generated **`D3PLOT`** indicates that the skull's restitution force parallels the external force profile. Only minor damping effects are attributable to the isolated interface. In the context of the THUMS model and differential equation data, the skull's rigidity is suggested to be a dominant protective force up to a certain threshold, whereas the CSF and other surrounding tissues cushion the brain's movement once this rigidity is overcome.

5 Relevant Figures

5.1 THUMS AM50 Model (Isolated Elements)

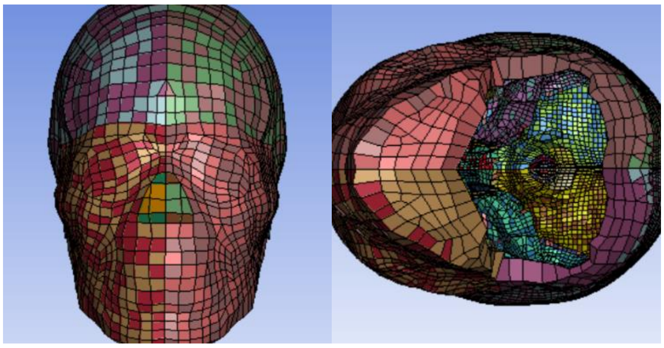

Fig 5.1: Front and inverted angles

5.2 THUMS AM50 Model (Isolated Elements) with Velocity Vectors

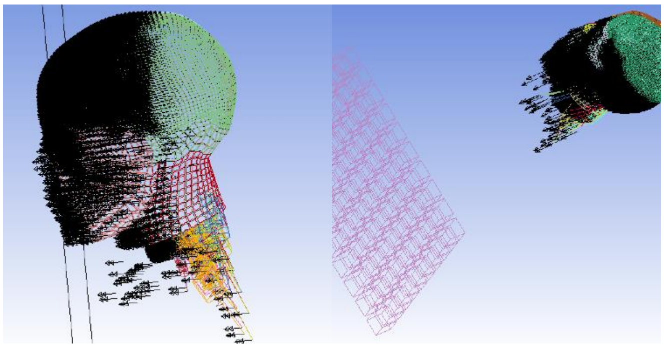

Fig 5.2: Modified THUMS Model and rigid wall; velocity vectors applied at varying angles

5.3 Simplified Headform

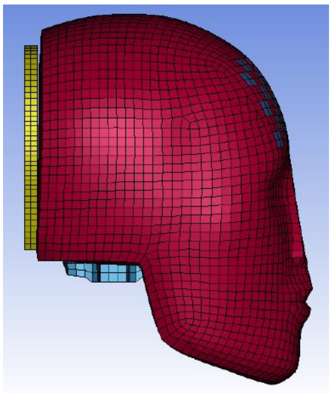

Fig 5.3: Simplified Headform Model of skull and skin only, including accelerometer

5.4 D3PLOT of Simplified Headform with External Force Applied

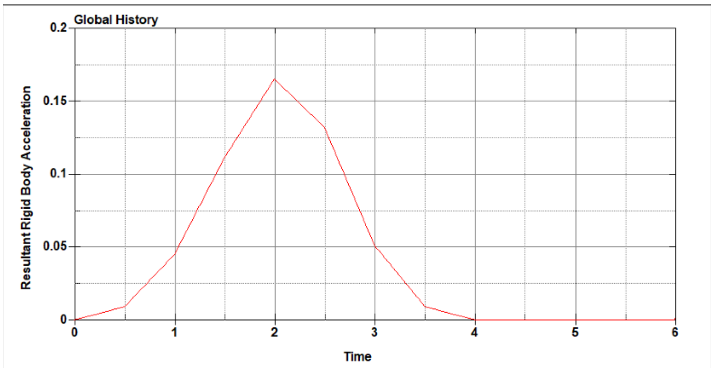

Fig 5.4: Acceleration of skull after application of external force

5.5 Graph of Differential Equation System of Internal Brain Acceleration

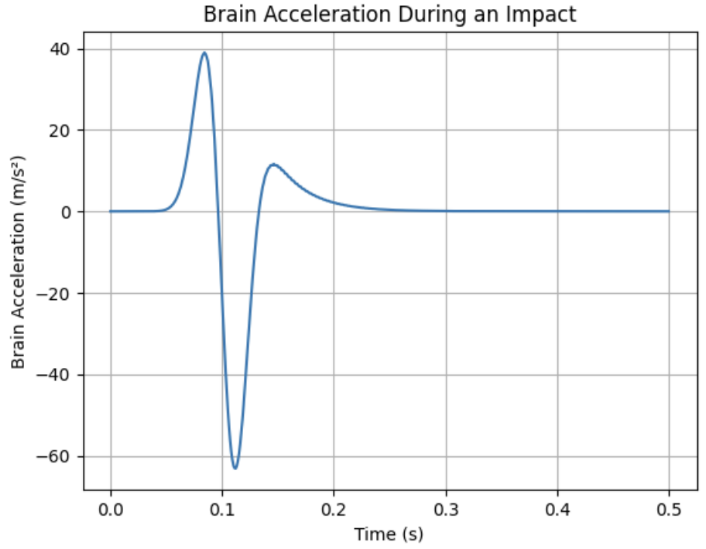

Fig 5.5: Graph of second-order differential equation (see below) representing the brain's acceleration in the skull upon application of external force

5.6 Graphical Comparison of Brain Acceleration: Differential Equation Model vs. Approximated FEM Results

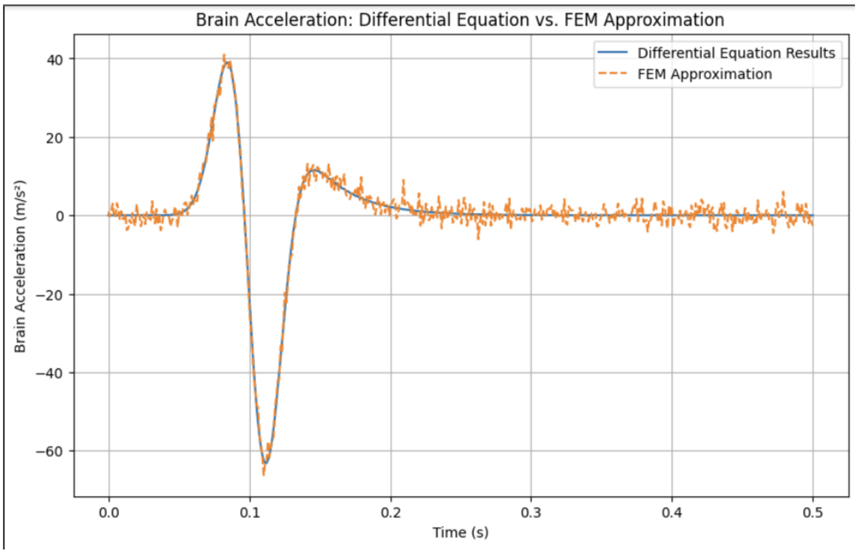

Fig 5.6: Comparison of brain acceleration from the differential equation model and approximated FEM results, highlighting consistency with minor variations.

6 Equations

6.1 Second-Order Differential Equation Modeling Brain Movement in Isolation

$$m\frac{d^2x}{dt^2} + c\frac{dx}{dt} + kx = F_{ext}(t) - F_{skull}(t)$$

Where:

- $m = 1.4\, kg$ is the effective mass of the brain
- $c = 1500\, N(s/m)$ is the damping coefficient
- $k = 50{,}000\, N/m$ is the spring constant
- $F_{ext}\,(t)$ is the external impact force(s)

- $F_{skull}\ (t)$ is the force exerted by the skull

7 Summary

This study used the THUMS AM50 model in simulations of the brain-skull interface upon the application of traumatic force. It focuses on the restoring effects of the skull and damping properties of the cerebrospinal fluid (CSF). A simplified headform was utilized to refer to the restitution dynamics of the skull itself. Validation of a self-generated second-order differential equation representing a mass-spring-damper model against approximated FEM data illustrates a high degree of correlation. With these findings, a foundation is provided to include further rotational dynamics and personalized modelling to better understand TBI diagnostics in a more pragmatic environment.

8 Relevant Literature

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