Concussion Simulation and Prevention

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***Abstract*—This document serves as a final report for a simulation project in which a human spine, skull, and brain are modeled to understand brain trauma. Accelerations and velocities of the brain during impact are determined, as well as the modeling methods used and data analysis. The purpose is to create a model that can be run in quick succession for preliminary results of concussion prevention methods.**

***Index Terms*—concussion, brain trauma, impact, helmet, skull model, discrete simulation**

I. Introduction

Approximately 1.6 to 3.8 million sports related concussion occur annually in the United States, which balloons even higher when including those from car accidents, bike accidents, or falls [1]. It's been estimated that 30 percent of high school athletes and 47 percent of collegiate athletes fail to self-report concussions, potentially leading to more severe symptoms. Serious or repeated concussions can result in permanent memory loss, depression, and neurodegenerative disorders. Even though concussions are so widespread, there are still many unknowns surrounding the condition, including which impacts have the highest risk of causing concussions [2]. To better understand and prevent mild traumatic brain injuries, further research on the kinematics of concussive impacts is necessary. This research can help identify contributing factors and lead to technologies that protect the brain to prevent long term cognitive diseases.

II. Project Objectives

The primary objective of this project was to develop a 2D model of the brain, surrounding cerebrospinal fluid, and skull that could quickly simulate impacts. A helmet was also added to simulate the effect of a protective layer with a lower elastic modulus. This model could be used to estimate stresses and strains within brain tissue, identify where a concussion may have locally occurred, and to optimize safety devices such as helmets designed to reduce rapid accelerations of the brain. Protective foams, shells, and fluids may be further integrated into the basic model to simulate the effectiveness of existing or proposed safety products at preventing concussions.The simulation was generated using beam elements for each body part connected by individual nodes. The head experienced acceleration until it contacted a perfectly rigid wall and rebounded. Simultaneously, the movement and deformation of the spine and brain was calculated using the Newton-Raphson method and beam deformation theory.

Various assumptions were made to simplify the model's design. This was necessary to develop a working model and could be compared to experimental results to drive further improvements to the simulation. Angular acceleration was not considered in this simulation, but has been shown to be a key cause of concussions. Therefore that could be added as different impacts are studied. The skull was modeled as a rigid body because it had a much higher elastic modulus than the brain and would therefore deform less. Additionally, the impact on the brain was the primary concern during an impact. The modified mass method was used to calculate the contact reaction between the head/helmet and the wall. The reaction of the brain on the skull was calculated using this approach as well, but using an applied reaction force. The brain was difficult to model due to a low elastic modulus, complex shape and many competing forces within the brain as well as from the neck and skull. The brain, having a low elastic modulus and complex shape, was difficult to model using only beam theory. Therefore the simulation also took into account the spring/damping forces within the brain, the brain's interaction with the neck, and the resulting impacts against the skull. Further improvements to the model could be made by increasing the accuracy of the brain's shape and material properties of different areas on the brain as it is a non-isotropic material. For each simulation, the model tracks the velocity and acceleration of the brain. Using realistic model parameters, the simulation was able model the actual mechanics of a head impact [1], [2], [4] and could be further expanded by varying these parameters to simulate various protection devices.

Moving forward, the existing simulation can also be modified to test various acceleration combinations as well as analyzing the forces on different areas of the brain during collisions. This can help identify areas of concern during a head injury and clarify which mechanics of the collision affect the brain the most. This knowledge could be used to narrow the focus of head injury prevention research to the most promising solutions.

III. Modeling Approach

This code simulated concussions by modeling a 2D skull as a rigid body, the brain as a series of interconnected elastic nodes, and the spine using discrete elastic beams. Each element incorporated accurate material and dimensional properties [1], [2], [4]. The brain was modified to include additional nodes and spring/damper interactions in order to better approximate its shape and deformation during impact. However, this model could be further improved by taking into account the various isotropic properties of an actual brain as well as better modeling its shape using additional nodes. For this simulation, the brain was modeled as a simple grid. Accelerations consistent with previous concussion studies were applied and the velocities and accelerations on the brain were computed. From here, the forces and stresses and strains could be estimated on the actual brain geometry. Angular accelerations have been shown to be particularly concerning for concussions so these could be added to future iterations, but were not practical to simulate in a 2D model [2]. Various collision studies could be compared with this model by comparing peak accelerations. Ultimately the goal of concussion prevention would be to decrease the peak acceleration (and forces) typically by increasing the time of impact. This results in an overall impulse decrease and better outcomes for the brain. Thus these comparisons can lead to additional development in the field and help to better understand concussion impact mechanics and which aspects should be modified to reduce head trauma.

IV. Model Architecture

The model containing the spine, skull, and brain sufficiently simulates the mechanics of a head hitting a rigid wall. The following sections detail how each body part was modeled. Material and dimensional properties approximate average human values, making it a moderately realistic model. Additionally, elastic collision equations are incorporated to calculate the rebounding forces which are applied using the Newton-Raphson and Modified Mass Methods. The primary concern of this study was the movement and deformation of the brain and neck as the most flexible components. The skull was therefore treated as a rigid body as this would have minimal impact on the actual head trauma. The focus in the simulation was on the movement of the brain, elastic and damping affects within the brain, movement of the neck, dampening from the cerebral-spinal fluid and attachment of the brain to the neck. The brain therefore responded and deformed based on forward and backward collisions, internal and external damping/spring forces, and the positional/angular rotation of the spine.

A. Spine

The spine was modeled as a 2D discrete elastic beam with an adjustable number of nodes. The effect of muscles below the neck were neglected since the stiffness of the torso would be much higher than that of the neck. This is demonstrated by whiplash during a collision. The first two nodes of the neck were fixed and accelerations were applied to every other node to initiate movement. Mass was also applied to each node. The model was designed with virtual limits or "walls", which rebound the motion of the beam in the x-direction when the head or helmet contacts. The rebound is calculated using the Modified Mass Method on any node that passed beyond the wall assuming an elastic collision. The node's x coordinate index is switched to fixed, and that x value is set to the x value of the wall. The simulation continues running. Once the x value of that node is evaluated to be to the correct side of the wall, the fixed index switches back to free and the simulation could continue. This way, the beam was able to effectively bounce off the barrier while only considering its innate discrete elastic parameters. Fig. 1 shows the collision model containing a spine and skull impacting a wall, protected by a helmet. Further detail could be added to the wall in the future to simulate a collision against a softer surface such as the ground or an airbag.

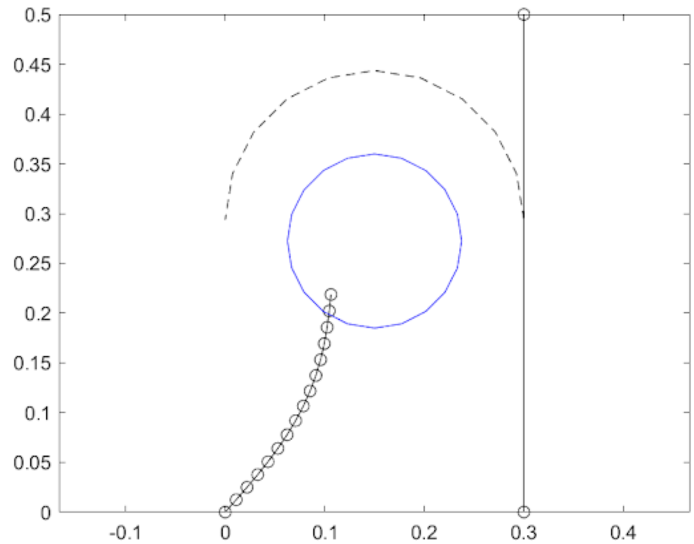

Fig. 1. Spine, skull, and helmet collision against wall.

B. Brain

Modeling the brain's complexity was one of the major challenges of this project. Initially, the brain was modeled as a simple beam attached in a circle that behaved according to the Newton-Raphson beam method. To improve the accuracy of the model, the brain was re-modeled as a four by four grid of nodes, connected by springs and dampers. Each square of nodes was connected as shown in Fig. 2. This model better approximates the complex interactions within an actual non-isotropic brain while still utilizing a simplified shape [4]. The code was able to accommodate a brain of any size by increasing the number of nodes and ultimately 16 nodes were used in the final iteration. The brain travels through viscous fluid contained in the skull and each connected node moves according to beam theory taking into account elastic stretching, bending as well as damping between nodes. It was critical to take into account how the spine impacted the brain in order to properly model the collisions using a modified mass method. Ultimately, the brain had a low modulus of elasticity and was very sensitive to small variations within the model. Therefore, forces derived from a momentum calculation were applied to the nodes during the collision to prevent small variations from distorting the brain's shape. Fig. 3 displays the brain in its undeformed state, and Fig. 4 displays the brain deformed under a force on the top right nodes.

C. Connections

The spine and brain are connected by fixing the leftmost column of brain nodes along the spine as a brain stem. This keeps each body part connected, but allows forces to transfer

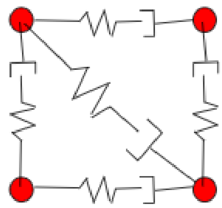

Fig. 2. Brain model incorporating a series of springs and dampers.

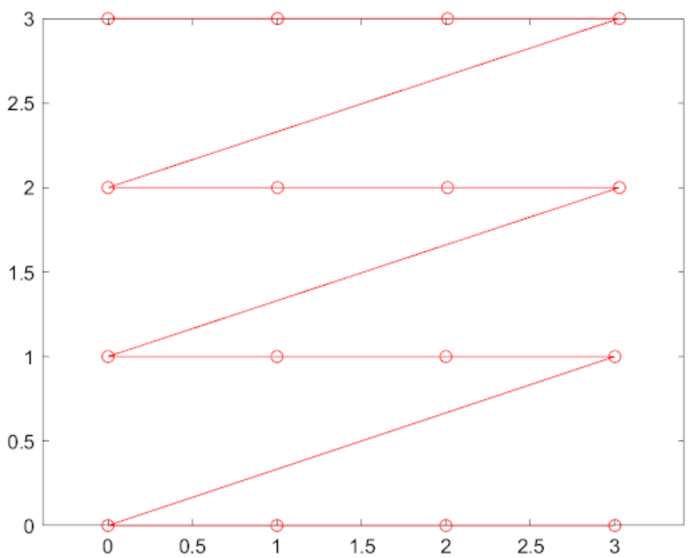

Fig. 3. Undeformed brain model.

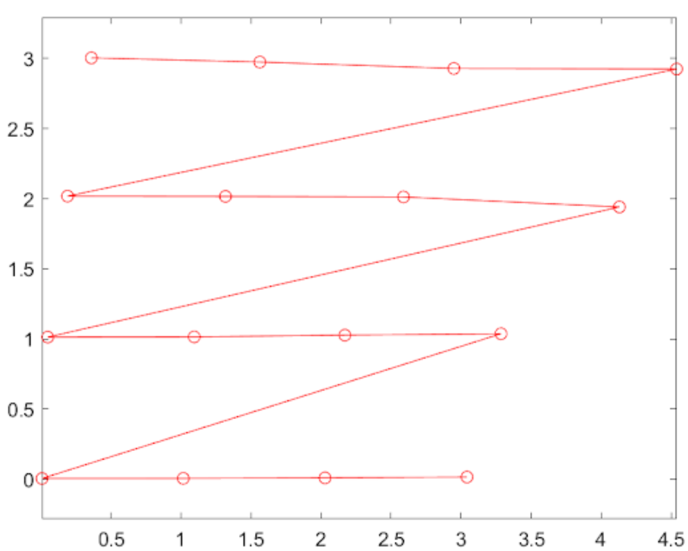

Fig. 4. Deformed brain model under force on top right nodes.

along the kinetic chain. The "fixing" is accomplished by calculating the x-y offset of the matched points after each iteration of Newton-Raphson method.The offset is then applied to ensure that the points move together while also taking into account deformation from the externally applied forces during a collision. The calculation also takes into account the angle of the spine/brain stem and adjusts the nodes accordingly each iteration. For interacting components such as the brain contacting the edge of the skull, a modified mass method was applied. The locations of each node were analyzed using for loops until contact was detected. Once the nodes intersect, a basic elastic calculation was performed and the forces for the current loop were updated. Fig. 5 shows the collision model continuing the spine, skull and brain. Ultimately this model was able to take into account the multiple interactions the brain has with both the skull and neck while still allowing elastic deformation.

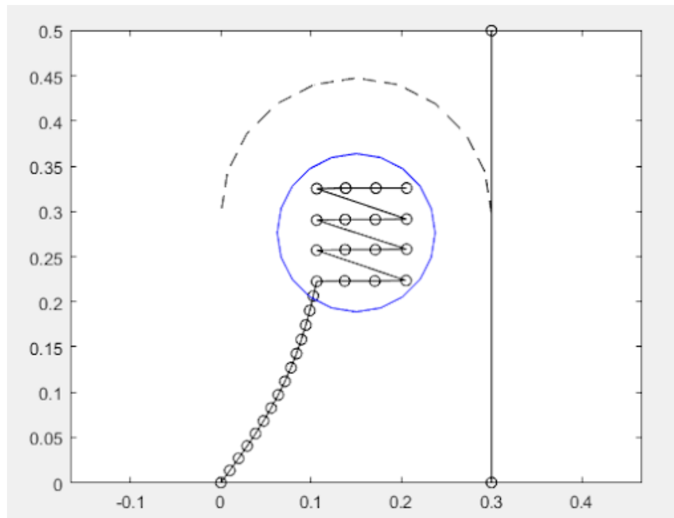

Fig. 5. Spine, skull, brain, and helmet collision with wall.

V. Data Capture

The simulation captures the brain's velocity and acceleration as shown in Fig. 6 and Fig. 7 which could be easily converted into force and stress values. However, this would likely require a more accurate geometrical interpretation of the brain in order to accurately approximate the area and moment of inertia. Therefore, for this simulation acceleration is the most critical measurement of a protective device's effectiveness. By taking a differential measurement, various collisions can be compared even if the actual stress in the brain is not known. As the brain model is improved, stress could feasibly be calculated. Furthermore elasticity values for the helmet can be altered to simulate the deformation and viscosity of the cerebrospinal fluid could be varied to model a Q-Collar's effectiveness [3].

VI. Challenges and Lessons Learned

The primary challenge in implementing this simulation was developing the code for the brain's movement once it was

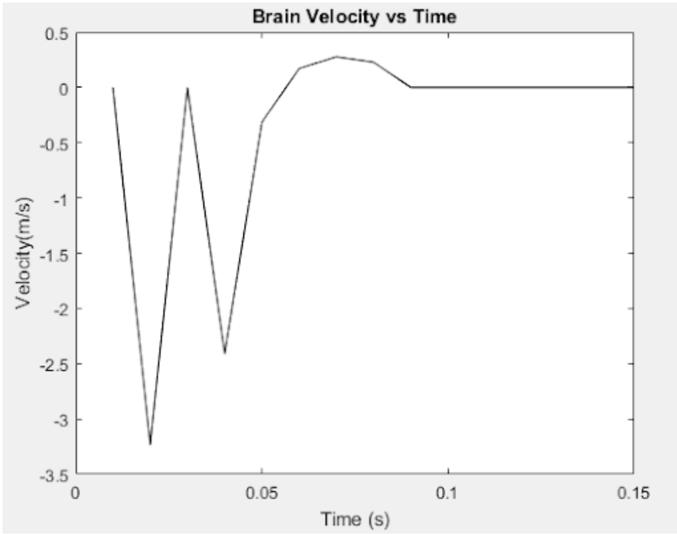

Fig. 6. Brain velocity vs time captured in the simulation.

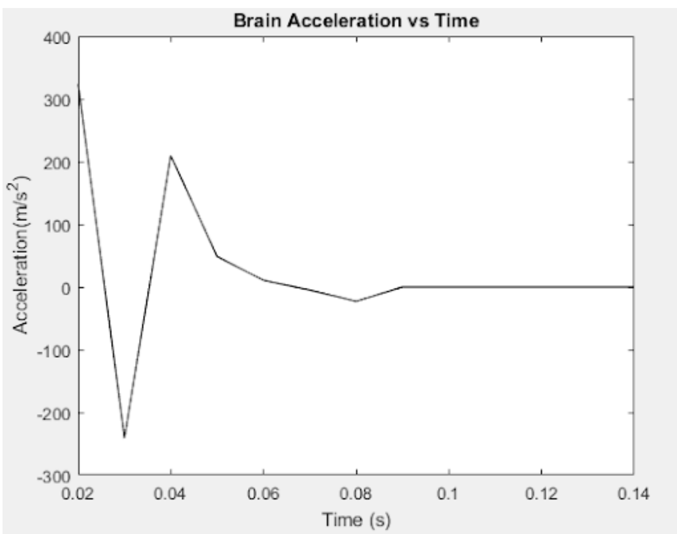

Fig. 7. Brain acceleration vs time captured in the simulation.

modeled as a series of springs and dampers. The brain had a low elastic modulus and was consequently very sensitive to small variations in the Newton-Raphson method or unevenness during a collision. The extremely low viscosity of the cerebrospinal fluid only exaggerated these difficulties. Despite previous studies promoting this model [4], this simulation suggests that this might not be the best model for numerically modeling a brain. Nonetheless, the code was able to control the brain such that it was sensitive to the neck movement, neck rotation and collisions. However, future research into alternative modeling approaches for the brain should be investigated in order to improve the simulation.

The other key lesson learned was that proper geometry for the brain was necessary to approximate stresses in various areas of the brain. Brains have non uniform material properties and an unconventional geometry that could be incorporated into the model to get more refinement of critical areas of the brain during a collision. Nonetheless, the approach taken in this simulation of simplifying the collision of the skull and helmet and focusing on the forces impacting the brain appeared to be effective.

VII. Future Work

The current state of the model works under standard parameters, but can run into issues as parameters or collision situations are varied. This is due to the many factors affecting the brain's movement as well as the brain's low elastic modulus. The inclusion of many discrete, connected systems made the calculation of transferred forces difficult, and behavior at times unpredictable. In the future, the behavior of the brain specifically would have to be altered so that forces from the neck are more accurately transferred to the brain and the brain more realistically impacts with the interior of the skull. This would be accomplished by improving the geometry or potentially using an alternative brain model that the zigzag interconnected nodes suggested by previous literature [4]. This geometry was very difficult to model numerically so a new approach would be suggested. One possible alternative is replacing the current brain mesh calculations with ones closer to those presented in Chapter 8 of the course reader [11]. The mesh could be constrained along the z-axis, effectively making it 2D. An attempt was made at this, but the resulting mesh was not yet able to be effectively integrated with the rest of the components in time. The mesh would likely have more stability and could even be updated to include angular acceleration forces on the brain. Fig. 8 shows an example of the mesh being developed for this simulation.

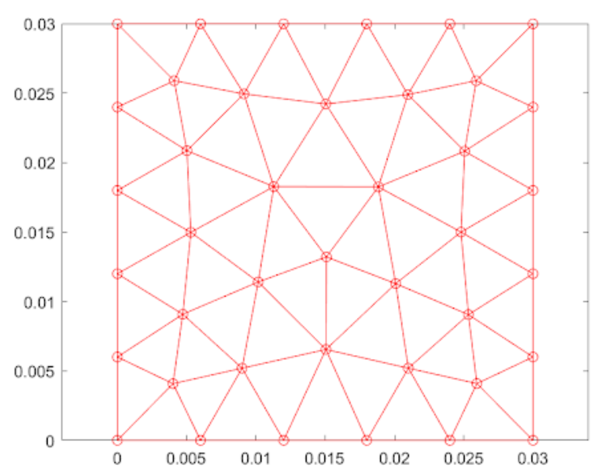

Fig. 8. Mesh simulation under development for the skull and brain.

Additionally, when the empirically found modulus of elasticity of the brain is used, the resulting deformation is greater than expected for a typical brain implying an additional factor in brain stress/strain. This could further suggest that additional geometry besides the spring/damper model used would be preferred. Further research into a better brain model could improve the simulation while ideally simplifying the model compared to an FEA simulation.The bio-mechanical characteristics of the brain are laid out in multiple referenced papers, and could be used to fine tune the simulation to a satisfactory state [1], [2], [4]. Once the model has been tuned, various forces, impacts, and protective devices could be simulated, and data could be used to determine the risk of concussion and the effectiveness of safety technologies.

A. Evaluating Technology

The properties of the helmet or viscosity parameters could be altered in this simulation to evaluate different techniques to minimize brain accelerations. The collision equations capture the change in acceleration experienced by the brain and could be compared to a bare head collision. In addition, the viscosity of the cerebrospinal fluid could be increased to simulate the effect of Q collars, which increase pressure in the skull. By changing these parameters we could determine the effectiveness of concussion mitigation strategies even if the model could be refined with an improved brain geometry.

VIII. Conclusion

This model of the spine, brain, and skull was simple enough to run quickly, yet accurate enough to evaluate accelerations and velocities in the brain. These accelerations could predict concussions and evaluate the effectiveness of certain concussion mitigation devices. While other methods do exist, like finite element analysis, this model could be run many times in succession for quick preliminary results. The biggest takeaway from this simulation was that previously developed modeling techniques for the brains involving interconnected mass/spring damper systems did not ideally mimic the actual behavior of the brain. Thus this simulation could be further improved by refining the brain model, while using the same simulation framework, to find a more effective modeling technique for a collision. These results could then be compared to existing experimental data to verify which brain model details are critical and which could be neglected to develop a fast simulation program.

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