A COMPUTATIONAL INVESTIGATION FOR POTENTIAL IMPROVEMENTS

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A COMPUTATIONAL INVESTIGATION FOR POTENTIAL IMPROVEMENTS IN SHOCK MITIGATION EFFICACY OF POLYUREA AUGMENTATIONS TO THE CURRENT ADVANCED COMBAT HELMET

A Thesis
Presented to
the Graduate School of
Clemson University

In Partial Fulfillment
of the Requirements for the Degree
Master of Science
Bioengineering

by
Angela Grujicic
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Accepted by:
Dr. Martine LaBerge, Committee Chair
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Dr. Larry Bowman
ABSTRACT

Mild traumatic brain injury (mTBI), colloquially known as a concussion, is the most common injury in modern wars. This domination of mTBI is hypothesized to be due to a combination of unconventional explosives and better protection and care of the patients, increasing survivability. While the majority of the body is covered in armor, the head is left relatively unprotected. The current Advanced Combat Helmet (ACH) has been designed to protect the warfighter against ballistic impacts and impacts against a hard surface, with little to no regard to blast loading. Polyureas, a class of microsegregated, elastomeric copolymers, has been shown to be effective in shock mitigation. A combined Eulerian/Lagrangian transient nonlinear dynamics computational fluid/solid interaction analysis will be used to examine the effects of different polyurea augmentations (utilizing the polyurea in the suspension pads, as well as introducing a thin polyurea inner lining/outer coating ) to the current ACH design under blast loading conditions attributed to causing mTBI. Quantifications of shock mitigation efficacy will be determined by: (a) establishing the main forms of mTBI, (b) identifying the mechanical causes for these injuries, and (c) quantifying the magnitude changes for the mechanical causes.

Mild TBI can be broken down into three main injuries: diffuse axonal injury (DAI), subdural hemorrhage, and contusion. DAI occurs due to the stretching and shearing of axons, leading to a disruption in the cytoskeleton which inhibits neuronal transport.
Subdural hemorrhage occurs when the relative motion of the brain causes shear forces large enough to rupture blood vessels bridging the brain and the dura mater. Contusions occur due to coup-contrecoup injuries, where the coup region is the initial shock impact site and the contrecoup site is the opposite of the coup site. The shear stresses, maximum principal stresses, and relative distance between the brain and the skull will be monitored.

It was found that the addition of a polyurea inner lining was most effective in blast mitigation; however, this augmentation did not provide a significant amount of protection compared to the standard ACH.
DEDICATION

This thesis is dedicated to my parents, who have always been deeply involved in my academic pursuits. My fiancé, Micah, and my lovely beagle, Thelma, have been heavily involved in motivation and stress-relief, for which I am eternally grateful.
ACKNOWLEDGMENTS

I would like to acknowledge my advisors for continually guiding me through the process. Some of the figures in the first two chapters have been adapted with the help of Micah Guy. I would also like to thank the Office of Naval Research for support of this project under grant number: 4036-CU-ONR-1125.
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CHAPTER ONE

BACKGROUND

Combat Situations

In the past decade, wartime injuries have been primarily traumatic brain injuries (TBIs), in particular, mild TBI (mTBI). This domination of mTBI is hypothesized to be due to a combination of the prominence of roadside bombs, improvised explosive devices (IED), and improvements to personal protective equipment (PPE), the field of medicine, and military procedures, allowing patients to survive with these injuries. [1, 2] Currently, most PPE is designed to be effective against physical or ballistic impacts, with little regard to blast. [3]

Anatomy and Physiology

Brain Structures

For our purposes, the head can be divided into the cerebrum, cerebellum, brain stem, pituitary gland, cerebrospinal fluid (CSF), and the skull. The cerebrum can further be divided into four major lobes: frontal (towards the front), temporal (around the temples), occipital (on the back of the head), and parietal (surrounded by the other lobes). The cerebrum comprises the main portion of the brain, and is involved with movement, sensory processing, olfaction, language and communication, and learning and memory. The cerebellum is located under the occipital lobe and deals with motor
control as well as “supervised learning” such as language, attention and mental imagery. The brain stem, in this division, is grouped to include the pons and the medulla oblongata, but not the midbrain. The brain stem controls the travel of information between the brain and the rest of the body as well as the autonomic nervous system (ANS). The pituitary gland is located near the brain stem opposite the cerebellum and releases hormones that control growth, blood pressure, thyroid gland, and metabolism as well as other functions. [4]

Surrounding the brain are a series of three membranes (meninges), the pia mater (adjacent to the brain), arachnoid mater (the middle layer), and the dura mater (closest to and adherent to the skull). Between the pia mater and the delicate arachnoid mater is referred to as the subarachnoid space (beneath the arachnoid) and contains the CSF, which provides the brain with buoyancy, physical protection, chemical control, and blood perfusion. More physical protection of the brain is provided by the skull. [4]

Histology of Brain Matter

As a main component of the Central Nervous System (CNS), the brain is largely comprised of neural tissue. Neural tissue is comprised of neurons and glial cells, non-neuronal cells that help neurons with their function. Neurons, responsible for transmitting chemical signals, have a cell body, an axon or axons, and a dendrite or dendrites. The dendrites are cell processes that are usually close to the cell body and
receive signals from neighboring cells. Axons are usually longer, and reach out to send signals to neighboring cells. The main glial cells present in the CNS are astrocytes, which connect neuronal cell bodies to capillaries while providing filter to prohibit drugs and other substances from entering the neurons, oligodendrocytes, which produce a myelin sheath to insulate some neuronal axons, ependymocytes, which secrete CSF and make up the blood-CSF barrier, and microglia, which are macrophages. [4]

Axons can be thought of as long tubular structures supported by its cytoskeleton, comprised of neurofilaments, highly elastic long filaments that, through repulsion, control the diameter of the axon, and microtubules, tubular dimers that maintain the structure of the axon. These microtubules have highly reactive ends, either undergoing additional synthesis of monomers, undergoing depolymerization, or attracting a capping protein, thereby stabilizing the structure. Under equilibrium, microtubules contain capping proteins that stabilize the length of the microtubules. [4]

The two components of the brain are called white and grey matter, named by the color post-mortem. In the grey matter, nervous signals are received, stored, and transformed into impulses which are then conducted by the white matter. The ECM of both types of brain matter contains mostly hyaluronic acid, as well as heparin sulfate proteoglycans, lecticans (large aggregating proteoglycans), other proteoglycans, glial hyaluronic acid-binding protein, thrombospondin, and tenascin-R. This provides for a lubricated environment that resists damage from normal head movements. [6]
Grey matter is located on the superficial surface of the cerebrum and cerebellum, as well as some deep places in the cerebrum, cerebellum, and brainstem. Grey matter contains capillaries as well as neuronal tissue. This neuronal tissue consists of neuronal cell bodies surrounded by neuropil (space between the cell bodies in which unmyelinated axons, myelinated axons, and dendrites reside) as well as the glial cells mentioned earlier. The main component of grey matter is the neural cell bodies, while white matter is mostly myelinated axons. [4]

White matter mostly consists of myelinated axons and is located in the brainstem, as well as deep regions of the cerebrum, cerebellum. It appears white due to the myelination on the axons, as myelin is mostly lipid tissue veined with capillaries. White matter connects grey matter areas to each other and carries impulses between neurons. There are three different lengths of white matter axonal fibers: [7]

1) 0-3 mm very short fibers. These are the most abundant.

2) 3-30 mm longer fibers. These fibers are called U-fibers, connect cortices, and follow the contours of the grey-white matter interface.

3) 30-170 mm longest fibers. These fibers are located in deep fascicles (bundles), more superficial fascicles, and corpus callosum, do not follow the folds of the cortex, and are the least abundant.
Shock and Blast

Prior to the 20th century, it was believed that the injuries sustained by blast waves were either psychosomatic or mysterious. Mining explosions in the late 18th century suggested that injuries were possible due to ‘bad air’. This was followed in 1812 by the death of a marine due to a shot that caused no external injuries, and it was this incident that caused speculation about blast. The prevalent idea was that a sort of electromagnetic energy, converted from the chemical energy, somehow entered the body and created damaging effects to biological tissue. [8] The damages caused were not studied.

World War I produced autopsies of great clinical relevance, when small or microscopic hemorrhages were found in the cerebrum and the meninges no sign of external injury by Mott, Cohen, and Biskind. [9] In some instances, namely a mortar shell explosion and an underwater firecracker explosion, massive intracranial hematomas were found, again with no sign of external injury. Unfortunately, two of the main physicians investigating this phenomenon were convinced that the damage to the patient was neurological and psychiatric, diagnosing blast as shell shock or functional neurosis. This denotes an inability for the physicians to distinguish physical injury from emotional injury. Undeniable brain damage was thought to be caused either by carbon monoxide poisoning (if the victim was buried) or “psychic trauma” (if there was no burial or external injury). [10]
More recent approaches to analyzing blast injuries focus on what blast is and what it causes. A blast wave can ideally be represented by a biphasic Friedlander waveform, shown in Figure 1.1, in which there is an almost instantaneous rise to peak pressure, followed by an undershot decline to atmospheric pressure. These rapid changes in pressure result in material shearing (caused by the tensile component of pressure), material shear-off (caused by the deviatoric component of stress), and organ and tissue contusion (caused by the compressive component of pressure). [11] The method of propagation of these pressures can only be theorized at this point, with most agreeing on one of three methods: directly through the brain; primarily through the major vessels and secondarily to the brain; and secondarily via blast under-pressure and electromagnetic pulses. [8, 12] Primary injury has not received much attention, as it involves biological and synergistic effects in the brain. [12]
Figure 1.1. A simple Friedlander biphasic wave form. The positive phase is marked by the + sign, and the negative phase is marked by the – sign. Time of arrival is $t_a$, duration of positive phase is $t_d$, pressure behind wave is $P_s$ and ambient pressure is $P_a$, making blast overpressure $P_s - P_a$.

However, on the battlefield, boundary and ambient conditions due to the extremely rapid conversion of chemical energy into thermal, electromagnetic, acoustic, and kinetic energy can cause the shock wave to reflect and combine, resulting in up to
an eight-fold increase in amplitude. Especially if the patient is in an enclosed space, like a vehicle, blast waves reflect off the walls, floor, and ceiling, and can have an additive effect. [13] This amplification can be used to explain the causation of primary mTBI found in military situations. [8, 12]

**Traumatic Brain Injury**

A traumatic brain injury (TBI) is caused by any sort of external force, from either mechanical force, impact, blast waves, or penetration, and is classified both by extent of the injury and mechanism of injury. When blast waves are the external force, the injury is referred to as bTBI (blast traumatic brain injury). In recent combat situations, where the military personnel are subject to blast waves from explosives, the most common form of TBI is a mild (mTBI) closed (non-penetrating) injury. These blast injuries comprise around two-thirds of all injuries sustained in military operations. [9] Blast injuries can be further classified as primary (from rapid changes in air pressure), secondary (from impact from objects propelled by the blast), tertiary (from the individual being thrown against another object as a result of the explosion), and quaternary (from heat, electromagnetic pulses, or toxic byproducts). [12] Secondary and tertiary effects are similar to those of sports- and car-related collisions, which have already been extensively researched. Comparatively, the mechanism of damage induced from primary blast injury has not been studied.
TBIs are usually measured in conjunction with lung injuries, which are far easier to classify in terms of lethality. Bowen explored this connection by creating a pulmonary survivability-lethality curve for unprotected blast exposure in order to calculate initial blast conditions to compare to concussion-inducing impact conditions. [12, 14] Work done by Moore et al. showed that a LD50 blast (resulting in 50% lethality due to lung injury) could be comparable to mTBI. [12]

To date, there has not been a comparative investigation of the injuries sustained during TBI induced by blast waves and impact injury. It is known that TBI due to sports and motor vehicle accidents are marked by a lower strain rate, while blast is marked with a higher strain rate range. [8] A 12-year study of 5600 terrorist bombings found that 91% of the casualties died within 24 hours of their head injuries. These head injuries were found to be both direct (subdural hemorrhages, concussions) and indirect (cerebral infarction due to air emboli). Air emboli were found to usually cause death within the first 30 minutes after the explosion. [15]

Animal models have also been utilized to determine the primary injuries associated with an explosion. Cernak et al. [9] delivered a blast corresponding with a moderate level of lung damage to constrained rats and rabbits via a “shock tube”, and discovered histological changes in the central nervous system (CNS), consistent with diffuse axonal injury (DAI), indicative of TBI. In a related study, Singh et al. [9] exposed rats in a concrete bunker to a blast and observed microglial activation mainly in the cerebral and cerebellar cortex as well as neural degeneration in the cerebral cortex.
These rats also exhibited impaired performance on tasks post-blast compared to pre-blast. Benzinger and Dodd [9] found evidence of air emboli in animals exposed to a blast corresponding to low level lung injury.

The severity of TBIs is classified based on three criteria: the Glasgow Coma Scale score, duration of the loss of consciousness, and the duration of posttraumatic amnesia. The Glasgow Coma Scale is a method of determining neurological function by assessing control over eyes, verbal ability, and motor skills, and a table used for diagnosing the particular TBI. We will be focused on mTBI, as it is the most common form. The three main mTBI types are considered to be diffuse axonal injuries (DAI), contusions, and subdural hemorrhage. [6]

Table 1.1. Criteria for determining the severity of a Traumatic Brain Injury. [Adapted from 6]

<table>
<thead>
<tr>
<th>Severity</th>
<th>Glasgow Coma Scale</th>
<th>Loss of consciousness</th>
<th>Posttraumatic amnesia</th>
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<tr>
<td></td>
<td>(total score)</td>
<td>(duration)</td>
<td>(duration)</td>
</tr>
<tr>
<td>Mild</td>
<td>13-15</td>
<td>&lt;1 hr.</td>
<td>&lt; 24 hrs.</td>
</tr>
<tr>
<td>Moderate</td>
<td>9-12</td>
<td>1-24 hrs.</td>
<td>24 hrs. to &lt; 7 days</td>
</tr>
<tr>
<td>Severe</td>
<td>3-8</td>
<td>&gt; 24 hrs.</td>
<td>7 days or more</td>
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Diffuse Axonal Injury

Diffuse axonal injuries (DAI) were traditionally characterized by torn axons due to either mechanical failure, but recently, new research has shown that axonal damage takes place several hours or days after trauma. It is now recently believed that the
different rates of displacement of brain regions due to stretching and shearing (due to rapid acceleration and deceleration of the head) during trauma usually does not cause the tearing, but rather the disruption of the neural cytoskeleton, and hence the disruption of neuronal transport. Specifically, microtubules experience disorganization and damage, causing a buckling of the axon. Damaged microtubules undergo depolymerization in order to allow the elastic recovery of microfilaments, leading to an external axonal configuration similar to the pre-stressed configuration. However, the missing microtubules interrupt normal neuronal transport, which causes the buildup of transported products. This causes a localized swelling, which can stretch the axon to failure. The proximal axon fragment then retracts towards the soma, while the distal axon fragment undergoes degradation via biochemical cascades known as Wallerian degeneration. This process is illustrated in Figure 1.2. [16, 17]
Figure 1.2. Diagram of axonal damage. The top figure shows a normal axon; the middle figure shows misalignment of the cytoskeleton, leading to the impaired transport; and the bottom figure shows the swelling and disconnection due to a buildup of organelles containing transported materials. [Adapted from 18]
Not surprisingly, DAI is most common in areas of white matter with long axonal lengths, such as the corticomedullary junction, internal capsule, upper brainstem, and corpus callosum. These locations are shown in Figure 1.3 in pink. [10]

Figure 1.3. Common locations of DAI shown in pink. [Adapted from 10]

A recent study done by Blennow, et al. aimed to prove this occurrence by testing for CSF biomarkers: total tau (T-tau) and neurofilament light (NFL) proteins, both indicating neuronal/axonal injury, glial fibrillary acidic protein (GFAP) and calcium-binding protein S-100β, both indicating injury to glial cells (gliosis). Also monitored was the serum albumin ration in the CSF, indicating damage to the blood-brain barrier. The subjects used were members of the Swedish Armed Forces, and they were exposed to blast from weapons they fired. The overpressure corresponded to blast overpressures between 0.18 and 0.25 atm, significantly less than the lowest value used in our previous study (1 atm). As the pressure differs by a factor of at least 4, this study may not have reflected changes in CSF fluid occurring at mTBI blast overpressure levels. [19]
An older study used amyloid precursor protein (APP) immunostaining methods to identify the occurrence of DAI in closed mTBI in subjects who died from other causes. APP is usually undergoes fast axonal transport, and therefore accumulates in axons that are severed during TBI, referred to as axonal swellings. These axonal swellings were found in several patterns: random, parallel, and zig-zag. The zig-zag patterns were notably found around focal lesions, and were thought to be indicative of “stress lines”. Although this study was done with subjects that underwent non-blast injury, it is thought that the findings can be translated. [20] A separate study done by Staal, et al. focuses less on the biomarkers involved with the injury and more on drug treatments to prevent the distal axon degradation post-traumatic injury. Their group has experienced favorable results. [21] Additionally, the study performed by Tang-Shomer, et al. showed a decrease in axonal degeneration with a pre-traumatic application of a microtubule stabilizer, taxol. [16]

DAI is hard to diagnose through imaging. A specialized form of MRI, Diffusion tensor imaging (DTI) can be used to calculate the apparent diffusion coefficient (ADC) and fractional anisotropy (FA) of the white matter in the brain to determine if DAI occurs. This is done by characterizing the water diffusion through the individual nerve fibers voxel by voxel in the MR images. This diffusion can be represented as a tensor matrix, which then can be rotated to its principal coordinate system to determine the eigenvalues and eigenvectors which can then be used to calculate the ADC and FA of the white matter. In normal, healthy white matter, water diffusion is very anisotropic, as it is
controlled by the direction of the fibers. In damaged white matter, injured axons distort this local linearity and decrease the anisotropy of the material. This is most pronounced in severe TBI, but it has been detected in some cases of mTBI. [18]

Contusion

Contusions are usually the result of localized, or focal, coup-contrecoup injuries, where the coup region is the initial shock impact site and the contrecoup site is the opposite of the coup site. In blast, the shock wave deforms the skull, which “slaps” against the brain, causing the coup injury. The brain moves to the opposite side of the intracranial cavity and bounces off the skull, which is the contrecoup injury. Contusions always involve the hemorrhage of blood vessels (which may or may not be accompanied by the rupture of lymphatic vessels leading to edema), in the gray matter, caused by rapid acceleration/deceleration of the brain relative to the skull. Contusions locations are graphically shown in Figures 1.3 and 1.4 in blue. [22, 23]
Figure 1.4. Common locations of contusions shown in blue. Subdural hemorrhage common locations shown in purple. This assumes an anterior or posterior blast impact loading. [Adapted from 10]

During different time points after blast impact, negative pressures can be found in both the coup and contrecoup regions. This has led many people to speculate about the existence of cavitation, the formation of bubbles inside a liquid and the subsequent implosion of those bubbles, in those regions which would lead to more brain damage. [24] While many people have performed studies proving the existence of negative pressure in the head, the correlation between negative pressure and cavitation is still only hypothesized. [23, 25-27] Nusholtz, et al., through a simple model, showed the existence of cavitation at accelerations above 150gs. [28]

Subdural Hemorrhage (SDH)

Subdural hemorrhages (also called subdural hematomas) occur when the relative motion of the brain compared to the skull causes shear forces large enough to rupture blood vessels (mostly veins) bridging the brain and the dura mater, connecting to the dural venous sinus, which returns blood and CSF to the jugular vein. Blood pools out of the broken vessels and collects in the subdural space (located between the dura mater and the arachnoid), putting stress on the surrounding brain tissue. This causes the patient to experience a gradually increasing headache and confusion. As subdural
hemorrhages are caused by shear forces, they usually occur on the frontal and parietal regions, assuming shock loading either anteriorly or posteriorly. This is graphically shown in Figure 1.4 in purple. [10]

Advanced Combat Helmet (ACH)

The two current helmets used in combat situations are the Advanced Combat Helmet (ACH) and the Light-weight Marine Corps Helmet (LWH); this project focuses on the ACH only. The ACH helmet, shown in Figure 1.5, consists of 7.8 mm thick hard outer shell, cushioned by seven suspension pads which are fastened to the helmet via Velcro strips and secured to the head with a chin strap. The shell is comprised of a composite of Kevlar 129 fibers in a phenolic resin, and the exact material (known as “Army foam”) and the fabrication methods of the suspension pads have not been revealed by the manufacturer, but it is highly suggested that the material is an elastomeric foam, similar to ethylene vinyl nitrile. [29]
Figure 1.5. Diagram of a Standard ACH. Shell, pads, and suspension mechanisms are shown. [29]

**Polyureas**

Polyureas have been previously investigated as an alternate material to be used in both the helmet and the suspension pads. All polyureas are thermoplastic, elastomeric copolymers formed by step-growth block polymerization of an isocyanate (containing \(-\text{N}=\text{C}=\text{O}\) groups) and amines (containing \(-\text{NH}_2\) groups) in order to create urea.
linkages (-NH-CO-NH-), which are a carbon atom double bonded to an oxygen and single bonded to two nitrogens, each bonded to a hydrogen and an R or R’ group. Polyureas are convenient for this application, as they both a) have a short polymerization time of less than a minute, allowing material fabrication by a spraying method, and b) allows for a variety of microstructures through varying the chemistry and/or polymerization conditions.

In the polyurea utilized in this project, the polar urea linkages are adjacent to diphenyl methane (C₆H₅-CH₂-C₆H₅) groups and form high stiffness segments. Also found in the polyurea molecule are nonpolar, aliphatic functional groups that serve as low stiffness segments. These two segments, due to the block arrangement, are periodic in nature and separate into hard domains and a soft matrix, a phenomenon referred to as microphase separation. This separation restricts polyurea from being considered a homologous amorphous material, rather exhibiting the following mechanical responses: a) a high level of stress vs. strain non-linearity, b) an extreme strain rate sensitivity, and c) a high degree of pressure dependence. Polyureas also are able to harden under applied loading, and alter/disperse shock waves and absorb kinetic energy under dynamic loading conditions associated with blast/ballistic impact. [29]

Ethylene Vinyl Acetate Foam

The material used in the standard ACH, “Army Foam” is proprietary with a structure and material parameters unknown to us, but is well known to have similar
properties to the common Ethylene Vinyl Acetate (EVA) foam. In itself, EVA is a copolymer of ethylene \((\text{CH}_3\text{-CH}_3)\) and vinyl acetate \((\text{CH}_3\text{-CH}_2\text{-O-CO-CH}_3)\), and is capable of forming cell-like structures within which a fluid (usually air) can reside. In this condition, EVA, as all foams is considered to be comprised of a solid matrix and a fluid phase dispersed within the solid matrix. This foaming procedure allows the EVA polymer to have a larger range of material properties, extending the range of applications: thermal conductivity, energy absorption in static and dynamic loading, and buoyancy. EVA foam is most commonly used as padding in sports equipment such as football helmets and running shoes. Structures of the repeating ethylene and vinyl acetate monomer are found in Figure 1.6. [30]

\[
\begin{align*}
\text{(poly)ethylene} & \quad \begin{pmatrix}
\text{H} & \text{H} \\
\text{C} & \text{C} \\
\text{H} & \text{H}
\end{pmatrix}_n \\
\text{(poly)vinyl acetate} & \quad \begin{pmatrix}
\text{H} & \text{H} \\
\text{C} & \text{C} & \text{O} \\
\text{H} & \text{O} & \text{C=CH}_3
\end{pmatrix}_m
\end{align*}
\]

Figure 1.6. Components of the EVA copolymer. \(n\) and \(m\) denote the number of monomers used. [Adapted from 30]

The properties of EVA depend on the ratio of monomeric components used. Polyethylene is semi-crystalline and thermoplastic with a very low glass transition
temperature, leading to a flexible structure with good impact resistance. Polyvinyl acetate, on the other hand, is not crystalline, is polar, and also has a low glass transition temperature. Therefore the copolymer EVA has a lower crystallinity than PE, with EVAs with VA percentages greater than 40% having a completely amorphous structure, better flexibility, clarity, and impact strength, and a lower hardness. These effects are diagrammed in Figure 1.7. [30]

![Figure 1.7. Effects of increasing the Vinyl Acetate content in EVA. [Adapted from 30]](image)

**Finite Element Analysis**

The finite element method utilizes the principle of discretization, subdividing an object to a number of smaller objects for simplicity, in order to attain approximate solutions to boundary-value problems. These smaller objects are referred to as *finite elements*, a finite number of objects that usually consist of a regular geometry (frequently tetrahedral or hexahedral in three-dimensional analyses) that, together,
represents a complex geometry. This can be visualized in two dimensions by representing a circle as a polygon with a large number of sides. The more sides, the closer the polygon comes to mimicking the circle. The elements consist of nodes or nodal points, the vertices of the object, nodal lines connecting these nodes, and nodal planes representing sides of the object. Not all elements contain these components, as elements range from one (lines) to two (triangles and quadrilaterals) to three dimensional (tetrahedral or hexahedral). The components, as a whole, are generally referred to as the finite element mesh. [31]

The finite element method generally contains 7 steps: [31]

**Step 1: Discretize and Select Element Configurations**

This step involves breaking the complex geometry investigated into nodes and elements, as described earlier. If desired, nodal lines do not have to be linear, but must be described by an equation. If the line is extremely irregular, a Taylor series approximation can be used. However, this added level of complexity can increase the computational cost of the problem, and usually is omitted. [31]

**Step 2: Define Strain (Gradient)-Displacement (Unknown) and Stress-Strain (Constitutive) Relationships**

This step specifies the relationship between various input and measured quantities for the given model. In a simple elastic model, this would be Hooke’s law:

\[ \sigma_i = E_i \varepsilon_i \]
where $\sigma_i$ is stress in the $i$ direction, $E_i$ is the Young’s modulus in the $i$ direction, and $\varepsilon_i$ is strain in the $i$ direction. [31]

**Step 3:** Derive Element Equations and Assemble to Obtain Global or Assemblage Equations and Introduce Boundary Conditions

In this step, the user specifies constraints, such as conservation of mass, momentum, and energy and boundary and initial conditions. The first ensures that the analysis follows the basic laws of physics, while the latter gives the conditions of the analysis, namely what happens at all nodes at the beginning of the simulation ($t=0$) and what happens at some critical nodes (usually outer) during all times of the simulation. [31]

**Step 4:** Solve for the Primary Unknowns

This step involves generating a set of simultaneous equations. Simple problems may only contain a few algebraic equations, while more complicated problems can contain many partial differential equations, not capable of solving by hand. [31]

**Step 5:** Solve for Derived or Secondary Quantities

In this step, dependent quantities (such as von Mises stress) can be solved for. [31]

**Step 6:** Interpretation of Results

This final step is the process of acquiring the output data and piecing it together to figure out what happened during the analysis. This could include analyzing tables, graphs, or even color-coded animations. [31]
For simple problems the finite element method could be solved for by hand. However, when complex geometries and equations are considered, a computer program is necessary to solve the analysis. There are several different software packages that are capable of handling Finite Element Analyses, each with its own set of advantages and disadvantages. The program utilized in the present work is ABAQUS/Explicit. [31]

Implicit vs. Explicit Analysis

There are two main ways of solving a Finite Element Analysis: implicit and explicit. Implicit, as the name suggests, contains dependent and independent variables not isolated to separate sides of the equations. That is to say, ABAQUS must solve for all elements at any given time step, involving many unknowns and a large system of equations. Implicit analyses generate more accurate results, but are extremely computationally costly, requiring an unreasonable amount of time to solve. Explicit analyses, on the other hand, evaluates elements based on results obtained from the previous time step, which involves only one unknown. This method assumes a small difference between results for consecutive time steps. Additionally, this method requires a small enough time step to produce a stable, converging result. The following example demonstrates the difference between implicit and explicit analyses in a simple two dimensional problem:
\[ \frac{\partial Y}{\partial t} = K \frac{\partial Y}{\partial X} \]

Solve for \( Y(X, t) \). To solve, replace the dels with deltas, assuming an infinitely small difference:

\[ \frac{\Delta Y}{\Delta t} = K \frac{\Delta Y}{\Delta X} \]

Initial and boundary conditions are also needed to solve:

Initial Condition: \( Y(X) = Y_I \text{ at } t = 0 \)

Boundary Condition: \( Y(X = X_0, t) = Y_B(t) \)

The implicit solution is as follows:

\[ \frac{Y_{1,1} - Y_{1,0}}{\Delta t} = K \frac{Y_{2,1} - Y_{0,1}}{2\Delta X} \]

where \( Y_{i,j} \) is the value of \( Y \) at \( X = i \) and \( t = j \). The value of \( Y_{1,0} \) is known from the initial condition, and the value of \( Y_{0,1} \) is known from the boundary condition. \( \Delta t \) and \( \Delta X \) are also known, leaving \( Y_{1,1} \) and \( Y_{2,1} \) unknown, and requiring another equation to solve. It can be seen that all the equations representing \( Y_{i,j} \) must be solved at all locations \( i \) for any time \( j \).

On the other hand, the explicit solution is as follows:

\[ \frac{Y_{1,1} - Y_{1,0}}{\Delta t} = K \frac{Y_{2,0} - Y_{0,0}}{2\Delta X} \]

In this case, \( Y_{1,0}, \Delta t, \) and \( \Delta X \) are still known, as are \( Y_{0,0} \) and \( Y_{2,0} \), which are also obtained from the initial condition. This leaves only \( Y_{1,1} \) as the unknown. This method is easier, however it assumes that the \( Y_{2,0} - Y_{0,0} \) is comparable to \( Y_{2,1} - Y_{0,1} \), which
requires a time step that is small enough to ensure that these values are comparable. If the time step is too big, the dissimilarity between the differences in successive time steps yield an unstable solution that does not converge, giving a nonsensical answer. A diagram of results obtained through implicit and explicit analyses is shown in Figure 1.8.

![Diagram showing differences in solutions of implicit and explicit analyses. The unstable explicit analysis involves a time step that is too large.](image)

**Figure 1.8. Differences in solutions of implicit and explicit analyses. The unstable explicit analysis involves a time step that is too large.**

### Linear vs. Nonlinear

Finite element analyses can also be linear or nonlinear. Linear analyses are usually employed in simple cases involving minute changes in parameters, such as displacement, rotation, temperature, etc. While not accurate for most physical problems, utilizing a linear approximation does carry the benefits of being having a low computational cost. However, when the problem contains significant intrinsic
nonlinearities, coupled with large displacements, rotations, changes in temperature, etc., a linear analysis cannot be used. These intrinsic nonlinearities can be classified into two major types, geometric and material nonlinearities. [32]

Geometric nonlinearities occur in kinematic quantities and the relationships between such quantities. An example is a bar attached to a torsional spring, found in Figure 1.9. [32]

![Figure 1.9. A bar of length $l$ attached to a torsional spring with spring constant $k_T$. The angle of the bar to the perpendicular of the wall is $\theta$, and the force applied to the end of the bar is $F$. [Adapted from 32]](image)

A moment balance on the bar yields: [32]

$$M = Fl \cos \theta$$
where $M$ is the moment at the end of the bar attached to the wall, $F$ is the applied force at the other end of the bar, $l$ is the length of the bar, and $\theta$ is the angle of the bar to the perpendicular of the wall. Expressing the moment as a function of the torsional spring constant yields: [32]

$$ M = k_\tau \theta $$

where $k_\tau$ is the torsional spring constant. Solving for the force, $F$ yields: [32]

$$ F = \frac{k_\tau \theta}{l \cos \theta} $$

This expression is nonlinear with respect to $\theta$. It can be seen that in cases of small rotation ($\theta$ approaches 0), $\cos \theta$ approaches 1, so the expression becomes linear: [32]

$$ F = \frac{k_\tau \theta}{l} $$

Again, these nonlinearities can usually be ignored in cases of small differences in these quantities. However, in cases of large differences, results are usually extremely incorrect. This is illustrated in Figure 1.10. [32]
Figure 1.10. Comparison of Linear and Nonlinear solutions. The results are similar when $\theta$ is less than around 30 degrees, but get increasingly different as $\theta$ increases. [Adapted from 32]

Material nonlinearities occur in the characterization of the behavior of the material. In the previous example, this could be achieved by expressing the spring’s behavior as a function of rotation: [32]

$$M = (k_0 + k_1 \theta) \theta$$
where \( k_0 \) is the rotation-independent aspect of the spring behavior, and \( k_1 \) is the rotation dependent aspect of the spring behavior. In this example for variable spring behavior, the force equation becomes: [32]

\[
F = \frac{(k_0 + k_1 \theta)\theta}{l \cos \theta}
\]

This equation contains both the material and geometric nonlinearities. The geometric nonlinearity can be removed again in the case of small rotations, and the equation becomes: [32]

\[
F = \frac{(k_0 + k_1 \theta)\theta}{l}
\]

Figure 1.11 updates Figure 1.10 with the addition of the material nonlinearity. The introduction of the material nonlinearity shows different solutions around \( \theta \) values of over 5 degrees (.1 radians), while the geometric nonlinearity adds variation around \( \theta \) values of over 17 degrees (.3 radians). [32]
Figure 1.11. Comparison of Linear and Geometric and Material Nonlinear solutions. The results are similar when $\theta$ is less than around 5 degrees, but get increasingly different as $\theta$ increases. [Adapted from 32]

Again, in cases of small differences in displacement, rotation, temperature, etc., these nonlinearities can be ignored and linearized, but in cases of large differences, the results are severely affected. [32]

Lagrangian/Eulerian

To deal with these large deformations, two approaches have been formulated: Lagrangian and Eulerian, both with advantages and disadvantages in different scenarios.
and materials. These two methods differ by three characteristics: 1) mesh description and function, 2) stress tensor and momentum equation, and 3) strain measure. A comparison between Lagrangian and Eulerian methods can be found in Table 1.2. [32]

Table 1.2. Differences between Lagrangian and Eulerian methods. [Adapted from 32]

<table>
<thead>
<tr>
<th>Lagrangian Mesh</th>
<th>Eulerian Mesh</th>
</tr>
</thead>
<tbody>
<tr>
<td>Coordinates of nodes coincide and move with the material points. Mesh encapsulates the material and remains fixed inside elements.</td>
<td>Nodal coordinates are fixed in space. Material points are free to move within the mesh. Material locations are time variant.</td>
</tr>
<tr>
<td>Boundary nodes are always on the edge, which allows for easier definition and application of boundary conditions.</td>
<td>Material boundaries may not fall inside the mesh boundaries.</td>
</tr>
<tr>
<td>The mesh distorts with the material, allowing for severe mesh distortions. This is illustrated in Figure 1.12.</td>
<td>The mesh does not distort because it is fixed in space. However, the domain must be larger to account for deformation and motion of the material within the domain. (Figure 1.13)</td>
</tr>
</tbody>
</table>

Figure 1.12. Lagrangian mesh. Mesh encapsulates material and distorts along with material. [Adapted from 32]
A combination of Lagrangian and Eulerian can be utilized to employ the benefits of both methods into a single analysis. Materials, such as air, that experience large deformations can be handled as Eulerian, while containing Lagrangian models within the Eulerian domain. This allows for a decrease in computational cost, while allowing for an accurate analysis.
CHAPTER TWO
INTRODUCTION

Traumatic Brain Injuries (especially bTBIs) are still enigmatic, with precise mechanisms of injury unknown to us. The main methods of investigating these injuries are through in vitro studies, animal models, mannequins, computational models, and humans (mostly cadavers). For each study, it is imperative that the model chosen satisfy the following criteria: 1) possess a controlled, reproducible, and quantifiable method of producing the mechanical force for inducing the injury, 2) mimic components of human conditions with the reproducible and quantifiable injury, 3) contains a relation between the injury-inducing force and the injury, as measured by morphological, physiological, biochemical, or behavioral parameters, and 4) predict the severity of the outcome by the intensity of the mechanical force. [33]

Previous In Vitro Studies

In vitro studies allow for an isolation of neurons and more confident manipulation of variables. However, no systemic effects can be attained from in vitro studies, and the interaction between different organ systems cannot be studied.

To administer a shock-like loading for primary bTBI in vitro studies, various methods have been fabricated. Between 1994 and 1998, the barotrauma chamber, consisting of a modified fluid percussion device to administer even pressure changes to
all cells to induce pressure-mediated injury, was utilized test treatments for reversing the injuries it causes. In 1991, a rapid acceleration injury (RAI) device, consisting of a flask containing cells swung on a pendulum from a specific height and impacted, was utilized to demonstrating multiple impacts; however, the pressure transient obtained from impact was not fully characterized. Extracorporeal shock wave lithotripsy (ESWL), which is used to treat kidney stones, involves shock waves generated and reflected off of an ellipsoidal shaped reflector. ESWL is currently used to study tissue damage caused by shock waves, which has been identified as due to either the high transient pressure or the generation of cavitation bubbles in fluids subjected to high tensile stress. A study done in 1997 by Howard and Sturtevant showed cellular membranes failing in fatigue by shear, with the damage threshold dependent on the shock magnitude, number of exposures, and physical properties of the membrane. [34]

A different approach was taken by Tang-Schomer, et al. in 2010, where neurons were seeded onto both sides of micropatterned channels and induced to allow axonal growth into these channels. After axonal ingrowth was observed by fluorescent staining, the channel walls were removed, and the unsupported axonal region was subjected to a pulse of air. Axonal injury was reported, as microtubule destabilization and axon relaxation was shown to prevent axonal transport and lead to neuronal death. The involvement of neurofilaments was not investigated in this study, however, and the results from this study and the previously mentioned in vitro studies have no application in injury prevention or mitigation. [16]
Another method, that is widely utilized, is the use of animals. This provides systemic data, as well as temporal data from the body’s attempts at healing. However, differences in anatomy, especially in terms of shock wave propagation throughout the body, can reduce the accuracy of these results. Animals previously utilized in these shock experiments have been rats, mice, ferrets, rabbits, cats, dogs, pigs, sheep, and primates. Despite concerns raised regarding the differences in systemic physiological and behavioral responses between rodents and humans to trauma, as well as the differences in the cortex structure, rats continue to be the prevalent species for testing. Figure 2.1 shows the animal models as well as shock models utilized in different TBI cases. [33]
To simulate injury, the animals are either subjected to static or dynamic brain injury induced via a mechanical force. Static trauma involves a known amplitude and duration, with no consideration about velocity and acceleration of loading. Dynamic trauma involves a known amplitude, duration, velocity and/or acceleration to inflict...
either direct (on the brain) or indirect (systemic) injury. As dynamic conditions reflect blast conditions better, they are used more frequently. A review of dynamic injury models is given below. [33]

Direct dynamic injury can be classified by either impact or nonimpact, and these can be further classified by the level of restraint on the motion of the head. Impact injuries are either penetrating the head/involve direct brain deformation, sustained from a high-velocity missile injury, nonpenetrating, or acceleration-based. Since the present work does not deal with penetrating mTBI, the associated models will not be discussed. [33]

The most frequently used model is the fluid percussion injury (FPI) model, involving a fluid pressure pulse via a pendulum strike in a cylindrical reservoir to the intact dura mater through a craniotomy. This method is used to study injury pathology, physiology, and pharmacology, and utilizes rats, mice, cats, pigs, rabbits, dogs, and sheep as test subjects. As expected, different results are seen with different placement of the craniotomy, and the severity of the damage depends on the magnitude of the pressure pulse. With consistent placement of the pressure pulse, results appear to be reliable and reproducible. [33]

Rat FPI experiments done by Saatman in 1998 proved the existence of a change in cytoskeletal structure in mTBI and their similarity to damages attained in moderate TBI and that there were injury thresholds to experience some of the other cellular changes found in moderate TBI. [35] A similar study by Kuehn, et al. in 2011 involved
shock loading via detonation of .22 caliber cartridges to replicate those found within armored vehicles penetrated by IEDs, and reported to have found to have produced a successful method for modeling bTBI for the purpose of testing protection methods. [36]

Controlled cortical impact (CCI) involves a more controlled impact also delivered to the intact dura mater via a pneumatic cylinder. Injuries sustained by ferrets, rats, and mice from this method more successfully mirrored mTBI in humans. DAI was seen in white matter (as opposed to focal damage seen in the previous methods), and a coma was induced. [33]

Controlled concussion models are less invasive, involving blow delivered to the coronal suture by a controlled, blunt force. An earlier model designed by Tornheim, et al. involved a metal plate propelled towards an exposed cat skull via a .22 caliber cartridge. This model proved to not be reliably reproducible, and was only utilized for studies involving cerebral contusion. A later rat model developed by Goldman, et al. incorporated a pendulum with better, more reproducible results that showed more correlation to effects seen in cases of human TBI. [33]

Impact accelerations models are either constrained or unconstrained. With different unconstrained models involving primates and sheep, reproducibility is low. This improved with constraining the head to only allow movement on one plane. Constrained models have had more successful results. A method originally created by Marmarou, et al. and modified by Cernak, et al. involved a laser-guided air-driven high-
velocity impactor loading a steel disc fixed by dental cement to an exposed rodent skull. Mortalities were primarily due to respiratory depression, and a direct correlation was found between force and induced injury. Additionally, reproducibility was good and major biochemical and neurological changes were noted. [33]

Nonimpact head acceleration models involve some means of inducing rotational acceleration and motion to induce brain injury. These models usually involve some degree of head fixation, to reduce the number of outcome variables. One method, with great outcome similarity to sever TBI, involves a pneumatic shock administered to a nonhuman primate that induces a 60 degree head rotation within 10-20 ms. A miniature swine model involved converting linear motion produced by a pneumatic actuator to rotational motion. This model was also shown to correlate with human findings, and allowed for examination of morphological, cellular, and molecular responses to blast TBI. However, these studies have not yielded a complete reproduction of human TBI pathobiology. [33]

Previous Headform Studies

Headforms present another option for TBI research. While a headform is incapable of having any complex response to stimuli, it is able to undergo true loading conditions, and possess a human geometry. Material components of the headforms are chosen to mimic properties of the simulated human tissue. Headforms also have the advantage of ability to contain sensors for intracranial head conditions.
The earliest headform incorporated in a blast test was the Hybrid III anthropomorphic test device (ATD), developed by General Motors in 1973 for use in automotive safety experiments. The Hybrid III has advantages of possessing a head with a similar mass to a human, a flexible neck, and durability under extreme conditions. Unfortunately, the headform was made of aluminum covered with a vinyl rubber, and thus did not possess a similar shock response to that of a human head. [37, 38]

The UK Defense Evaluation and Research Agency (DERA) created their Dynamic Event Response Analysis Man (DERAMan) for use in impact testing, but has been used in blast testing since. The DERAMan contained a separate flesh, skin, skull, and brain materials, all polymers specially formulated to closely represent the human counterparts. The neck, as with the Hybrid III, was flexible. Additionally, the DERAMan contained sensors capable of producing 90 inputs: 40 piezoelectric polymer pressure sensors within the brain, 45 piezoelectric ceramic pressure sensors on the inner surface of the skull, 2 accelerometers, and 1 three-dimensional force gauge. [37, 38]

John Hopkins University’s Applied Physics Laboratory (APL) is working on a headform specifically designed with blast conditions in mind, specifically pressure and acceleration responses. The headform is not complete, and little information is available about it. The first prototype contained a head with a flexible neck. [38]

A Masters Thesis from the Univeristy of Nebraska at Lincoln published in 2010 introduced a Realistic Explosive Dummy Head (RED Head), which consisted of a one-piece polyurethane skull, a two-piece polydimethylsiloxane (PDMS) skin, a silicone gel
brain cured after application of sensors, and a flexible neck attached by pins to the skull. A steel holder created to position and secure the RED Head in front of the shock tube was also created. Testing of the current prototype has revealed some dissimilarities compared to behavior of a human head. UNL is continuing development on this model for optimization. [38]

While the use of these headforms has yielded promising results, more development is required to better characterize the mechanical and structural properties of natural human tissue in head components.

**Previous Computational Models**

Computational models have the most promise for reproducibility of input parameters. Unfortunately, human tissue is complex, with many components, which is not favorable towards the computational cost. Additionally, replication of the reaction of living cells to the stimuli and injury associated with mTBI is not possible with current technology and knowledge. Within the last decade, many finite element analyses have been carried out to investigate mTBI.

Work done by Zhang, et al. sought to investigate any differences in brain response to frontal and lateral impacts not due to blast. The geometric model consisted of a separate scalp, skull, dura, falx cerebri, tentorium, pia, CSF, venous sinuses, ventricles, and cerebrum (gray and white matter), cerebellum, brain stem, and bridging veins [39]. The brain materials were considered to be linear viscoelastic; the skull was
modeled with elastic-plastic constitutive equations, and all other tissues were assumed
to be linearly elastic, homogeneous, and isotropic. Gray and white matter were assigned
identical properties with the exception of different shear moduli. No neck was
considered, and head motion was considered to be translational, with no regard to
angular acceleration. With regards to skull deformation, brain pressure gradient, and
localized shear stress distributions, it was found that the lateral impact produced more
favorable results. [39]

El Sayed, et al. reproduced axonal damage and cavitation through inelastic
deformation due to a frontal and oblique impact, also not due to blast. His brain model
consisted of the skull without facial bones, CSF, gray matter, white matter, cerebellum,
corpus callosum, telencephalic nuclei, brain stem, and ventricles. Brain tissue was
represented with a series of elastoplastic and viscoelastic equations, while the skull and
CSF were considered to be linear elastic. As in the Zhang study, only translational
motion of the head was considered. [23]

Nyein, et al., in 2008 [3], developed a model and simulation of the fluid-solid
interaction in the case of a blast shockwave and compare the data to the Bowen curves.
This model was generated with real MR images merged with CT images and partitioned
into head components identified based on mechanical function: ventricular
cerebrospinal fluid, peri-ventricular glia, white matter, gray matter, eyes, venous
sinuses, subarachnoid cerebrospinal fluid, air sinuses, muscle, skin and fat, and diploic
skull bone. Brain tissue was modeled by the Tait equation of state (EOS) and a neo-
Hookean elastic model, and the skull was modeled with the Mie-Gruneisen/Hugoniot EOS. Overpressures of 5.2 atm and 18.6 atm, corresponding to the injury threshold and LD50 were investigated. [3]

Chafi, et al. utilized a radial instead of planar blast wave to investigate the dynamic brain response. The model consisted of the brain, falx and tentorium, CSF, dura mater, pia mater, skull, and scalp. To handle the blast simulation, an Arbitrary Lagrangian-Eulerian (ALE) formulation was utilized. The scalp was considered to be linear elastic; the CSF is modeled to be a layer of solid elements with fluid-like properties; the brain tissues were considered to be a hyperelastic material with the Mooney-Rivlin model. It was found that the dynamic response was better at predicting injury than head input accelerations. [22]

Nyein, et al. assessed the efficacy of the ACH and a conceptual face shield in 2010. The geometric model and components from the previous study was used, and the material models remained the same. It was found that the ACH had no effect on shock mitigation, but that the presence of a face shield rigidly attached to the helmet shell lowered stress magnitudes significantly. [1]

Abolfathi, et al. concentrated efforts on modeling white matter as anisotropic. Axons in random, hexagonal, and square arrangement and undulations of varying sizes were modeled and compared against known experimental data for the brainstem. It was found that the axons showed higher stresses, while the ECM exhibited higher strains. [40]
Colgan, et al. also investigated effects of modeling brain matter as anisotropic. Using DTI images, a family of axonal fiber bundles was modeled in an isotropic ECM. Statistically significant differences in shear strain were found in the brainstem and corona radiate region, but not in the midbrain, corpus callosum, or gray matter. [41]

**Previous Human Studies**

Only one blast study has been performed on live humans, due to ethical reasons. As mentioned earlier, Blennow, et al. searched for a useful biomarker in the CSF of Swedish Armed Forces members a given time after experiencing a mild blast wave due to firing heavy weapons. His results showed no significant difference in the tested biomarkers, but this could be attributed to the low level of overpressure utilized in this study. [19]

More studies have been done on cadavers that experienced mTBI along with an unrelated fatal injury. Cadavers have the advantage of possessing true material properties and real-world loading conditions. The study done by Blumbergs, which was mentioned previously, found significant patterns and locations of DAI in the case of impact TBI. The patients utilized in this experiment remained alive for a significant range of time: 1 minute to 8 days, allowing the brains in some patients to attempt to heal. Ages also varied from 8 to 89 years old, which also affects the healing response. [20]
Cadavers provide a good insight to the effects of loading conditions on the human brain. However, useful cadavers are not plentiful, variation of parameters is large, and the effects of primary, secondary, and tertiary effects cannot be isolated.

**Objective**

The present work investigates the effect of the utilization of a polyurea as the material for the helmet pads as well as the use of a polyurea inner lining or outer coating for the helmet on blast mitigation of the human head, thereby reducing the possibility of TBIs. A case of the unprotected head is modeled, as well as a case of the standard ACH-protected head, to ensure that no intrinsic aspect of the finite element analysis program, geometric model, or the analysis code of the problem could affect the data gathered. Cavitation will not be considered, as there has been no conclusive evidence to date that negative pressures formed inside the skull lead to cavitation.

A description of the experiment conducted is found in Chapter 3, including the method, models used, and the problem definition. Results and discussion of the various cases along with the efficacy of different augmentations to different aspects of mTBI is given in Chapter 4. Conclusions and direction of future work is provided in Chapter 5.
CHAPTER THREE

COMPUTATIONAL PROCEDURE

Method/Analysis

ABAQUS was utilized to solve the transient nonlinear dynamics problem of the interactions between an air-borne blast wave and a human head, involving simultaneously solving the partial differential equations for the conservation of momentum, mass and energy, along with the material constitutive equations and the equations defining the initial, boundary, and kinematic constraint and contact conditions. A coupled Eulerian/Lagrangian formulation was utilized to handle the large motions and deformations experienced by the fluid air and the smaller motions and deformations experienced by the solid head construct.

All Lagrange-Lagrange and Euler-Lagrange interactions are assigned the following contact/sliding and kinematic coupling options. Lagrange-Lagrange interactions are restricted with a penalty contact method, where interpenetration of surfaces is resisted by linear spring forces and contact pressures with values proportional to the depth of penetration. Contact pressures are only transmitted when the nodes of the predefined slave surface on one body contact the predefined master surface of another body. Shear stresses are transmitted by use of a static and kinematic friction coefficient and are bound by a maximum value of transmittable shear stress. Euler-Lagrange contacts are fluid-solid interactions. The Lagrangian surfaces define the inner boundaries for the
Eulerian region (air fills up the space not taken up by the head construct, and is only allowed to travel in regions not occupied by the head).

**Geometrical and Meshed Models**

A cube-shaped Eulerian domain was used to represent the air environment and a geometrically-irregular Lagrangian domain was meshed to resemble the human head (as well as the helmet apparatuses, when applicable), and placed in the air. The standard bioengineering coordinate system was utilized, so that the x-axis is pointing forwards, y-axis upwards, and z-axes laterally.

**Eulerian Domain**

A larger cube, 400mm in edge length, was meshed using approximately 185,000 smaller cubic elements (averaging 7mm in edge length and consisting of eight nodes). This domain was filled with air, whose material model is defined in the next section.

**Lagrangian Domain**

A previously-developed CAD model of the head, consisting of the cerebrum, CSF, cerebellum, brainstem, pituitary gland, and the skull, was modified to include a layer of skin/fat tissue. The ACH was scaled up from the previous model to account for the extra layer of material, and consisted of an outer composite shell and seven suspension pads. The ACH protected head measured 240 mmx265mmx245mm. A 2mm thick inner lining
and outer coating was also created and meshed to allow for augmented configurations of the ACH. In the case of the inner lining, the ACH was modified again to accommodate the lining. The head construct was meshed with first-order tetrahedral solid finite elements with a typical edge length of 2mm, which has been shown to have a good compromise between accuracy and computational efficacy. The unprotected head contained approximately 550,000 elements, and the augmented ACH case contained approximately 700,000 elements.

Material Models

Materials utilized must be defined by relationships between field and material state variables, represented here by an equation of state, a strength model, and a failure model. These relationships were partitioned by the relationship $\sigma_{total} = \sigma_{hydro} + \sigma_{dev}$, where the total stress tensor is equal to the sum of hydrostatic stress and deviatoric stress. Hydrostatic stress involves pressure, controls any changes in volume or density of a material, and is represented as:

$$\sigma_{hydro} = \begin{bmatrix} -P & 0 & 0 \\ 0 & -P & 0 \\ 0 & 0 & -P \end{bmatrix}$$

where $P$ is the pressure. The equation of state defines pressure, and therefore defines the hydrostatic stress.

Deviatoric stress, on the other hand, controls any changes in shape of the material, and is defined by the strength model. A failure model is not presented in this
study, as the blast loading is not strong enough to cause failure of any of the materials present. Temperature, as well as other temperature-related effects, is not considered in the present work, as blast loading conditions usually do not involve temperature. [29]

**Air Material Model**

For this application, air was assumed to be an ideal gas, with an equation of state defining pressure as: [42]

\[ P = -P_a + (\gamma - 1) \frac{\rho}{\rho_0} E \]

where \( P \) is the pressure, \( P_a \) is the ambient pressure, \( \gamma \) is the heat capacity ratio of constant pressure to constant volume specific heat \( \frac{C_p}{C_v} \), \( \rho_0 \) is the initial density of air, and \( \rho \) is the current air density. For air, \( P_a \) is set to 101.3 KPa, \( \gamma \) is set to 1.4 due to the diatomic gaseous components, \( \rho_0 \) is set to 1.225 kg/m\(^3\), and \( E \) was set to 261.2 kJ/m\(^3\) in order to achieve the initial condition of \( P = 101.3 \) kPa.

A strength model was not defined for air, as it cannot support shear stresses.

**Kevlar/Phenolic resin Composite Material Model**

In accordance with previous studies, the Kevlar/Phenolic resin composite will be presented as an orthotropic material with pressure defined as the following equation of state:
\[ P = -K_1 e_{vol} + K_2 e_{vol}^2 - \frac{1}{3}(C_{11} + C_{21} + C_{31})e_{11}^d - \frac{1}{3}(C_{12} + C_{22} + C_{32})e_{22}^d - \frac{1}{3}(C_{13} + C_{23} + C_{33})e_{33}^d \]

and \[ K_1 = \frac{1}{9}(C_{11} + C_{22} + C_{33} + 2(C_{12} + C_{23} + C_{31})) \]

where \( K_1 \) is the effective bulk modulus, \( e_{vol} \) is the volumetric strain, \( K_2 \) is a correction factor for non-linearity [29]. The \( C_{ij} \) values are derived from the stiffness tensor and can be represented with two elastic moduli. In the equation shown below, the two moduli chosen (for ease) have been the Lamé parameters, \( \lambda \) and \( \mu \).

\[
\begin{bmatrix}
\sigma_1 \\
\sigma_2 \\
\sigma_3 \\
\sigma_4 \\
\sigma_5 \\
\sigma_6 \\
\end{bmatrix} =
\begin{bmatrix}
C_{11} & C_{12} & C_{13} & 0 & 0 & 0 \\
C_{12} & C_{22} & C_{23} & 0 & 0 & 0 \\
C_{13} & C_{23} & C_{33} & 0 & 0 & 0 \\
0 & 0 & 0 & C_{44} & 0 & 0 \\
0 & 0 & 0 & 0 & C_{55} & 0 \\
0 & 0 & 0 & 0 & 0 & C_{66} \\
\end{bmatrix}
\begin{bmatrix}
\varepsilon_1 \\
\varepsilon_2 \\
\varepsilon_3 \\
\varepsilon_4 \\
\varepsilon_5 \\
\varepsilon_6 \\
\end{bmatrix}
\]

\[
\begin{bmatrix}
2\mu + \lambda & \lambda & \lambda & 0 & 0 & 0 \\
\lambda & 2\mu + \lambda & \lambda & 0 & 0 & 0 \\
\lambda & \lambda & 2\mu + \lambda & 0 & 0 & 0 \\
0 & 0 & 0 & 2\mu & 0 & 0 \\
0 & 0 & 0 & 0 & 2\mu & 0 \\
0 & 0 & 0 & 0 & 0 & 2\mu \\
\end{bmatrix}
\begin{bmatrix}
\varepsilon_1 \\
\varepsilon_2 \\
\varepsilon_3 \\
\varepsilon_4 \\
\varepsilon_5 \\
\varepsilon_6 \\
\end{bmatrix}
\]

In the present work, the Young’s and Shear Moduli as well as Poisson’s Ratios have been utilized to define the two Lamé parameters. These values are listed in Table 3.1. \( K_2 \), the quadratic correction constant is positive in this material, as shown in Table 3.1, allowing the Kevlar composite to support shock.
Table 3.1. Parameters, symbols, units, and values utilized in the material model for the Kevlar/Phenolic resin composite. [Adapted from 29]

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>Unit</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Young’s Modulus 11</td>
<td>E_{11}</td>
<td>Pa</td>
<td>1.799E10</td>
</tr>
<tr>
<td>Young’s Modulus 22</td>
<td>E_{22}</td>
<td>Pa</td>
<td>1.799E10</td>
</tr>
<tr>
<td>Young’s Modulus 33</td>
<td>E_{33}</td>
<td>Pa</td>
<td>1.948E9</td>
</tr>
<tr>
<td>Poisson’s Ratio 12</td>
<td>ν_{12}</td>
<td>--</td>
<td>0.080</td>
</tr>
<tr>
<td>Poisson’s Ratio 23</td>
<td>ν_{23}</td>
<td>--</td>
<td>0.698</td>
</tr>
<tr>
<td>Poisson’s Ratio 31</td>
<td>ν_{31}</td>
<td>--</td>
<td>0.075</td>
</tr>
<tr>
<td>Shear Modulus 12</td>
<td>G_{12}</td>
<td>Pa</td>
<td>1.860E9</td>
</tr>
<tr>
<td>Shear Modulus 23</td>
<td>G_{23}</td>
<td>Pa</td>
<td>2.235E8</td>
</tr>
<tr>
<td>Shear Modulus 31</td>
<td>G_{31}</td>
<td>Pa</td>
<td>2.235E8</td>
</tr>
<tr>
<td>Bulk Modulus</td>
<td>K_{1}</td>
<td>Pa</td>
<td>5.18E9</td>
</tr>
<tr>
<td>Quadratic Correction</td>
<td>K_{2}</td>
<td>Pa</td>
<td>5.0E10</td>
</tr>
</tbody>
</table>

The strength model for this Kevlar composite is just defined with a generalized Hooke’s Law:

\[ [\sigma_{\text{dev}}] = 2\mu [\varepsilon_{\text{dev}}] \]

where the deviatoric stress matrix \((\sigma_{\text{dev}})\) is equal to two times the second Lamé parameter times the deviatoric strain matrix \((\varepsilon_{\text{dev}})\). The coefficient \(2\mu\) originates from the value of \(C_{44}, C_{55},\) and \(C_{66}\) in the stiffness tensor, where \(\mu_{ij} = E_{ij}/[2(1 + \nu_{ij})]\).

Polyurea Material Model

The material model for polyurea was also acquired from previous studies [29, 42]. Polyurea was considered to be elastic in terms of the hydrostatic response, with an
additional consideration for large deformations and displacements of the material. The equation of state was defined to be:

\[ P = -K(T) \frac{\ln(J)}{J}, \text{ and } K(T) = K(T_{\text{ref}}) + m(T - T_{\text{ref}}) \]

where \( K \) is the bulk modulus, \( T \) is the temperature, \( m \) is a material parameter and \( J \) is the determinant of the deformation gradient, \( F \), and \( \ln(J) \) is the volumetric strain. During compression, \( J \) decreases, while the effective bulk modulus, \( \frac{K(T)}{J} \), increases, allowing polyurea to support shock. Values for the parameters are shown in Table 3.2.

Table 3.2. Parameters, symbols, units, and values utilized in the material model for the polyurea. [Adapted from 29]

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>Unit</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Reference Parameter</td>
<td>( T_{\text{ref}} )</td>
<td>K</td>
<td>273</td>
</tr>
<tr>
<td>Time Shift Parameter</td>
<td>A</td>
<td>--</td>
<td>-10</td>
</tr>
<tr>
<td>Time Shift Parameter</td>
<td>B</td>
<td>K</td>
<td>107.54</td>
</tr>
<tr>
<td>Pressure Shift Coefficient</td>
<td>CTP</td>
<td>K/Pa</td>
<td>7.2E-9</td>
</tr>
<tr>
<td>Constant Volume Specific Heat</td>
<td>CV</td>
<td>MJm(^{-3})K(^{-1})</td>
<td>1.97</td>
</tr>
<tr>
<td>Bulk Modulus Slope</td>
<td>( m )</td>
<td>Pa/K</td>
<td>-1.5E7</td>
</tr>
<tr>
<td>Reference Bulk Modulus</td>
<td>( K_{\text{ref}} )</td>
<td>GPa</td>
<td>4.95</td>
</tr>
<tr>
<td>Long-term Shear Modulus</td>
<td>( G_{\infty} )</td>
<td>MPa</td>
<td>22.4</td>
</tr>
<tr>
<td>Prony Series Coefficient</td>
<td>( p_1 )</td>
<td>--</td>
<td>0.8458</td>
</tr>
<tr>
<td>Prony Series Coefficient</td>
<td>( p_2 )</td>
<td>--</td>
<td>1.686</td>
</tr>
<tr>
<td>Prony Series Coefficient</td>
<td>( p_3 )</td>
<td>--</td>
<td>3.594</td>
</tr>
<tr>
<td>Prony Series Coefficient</td>
<td>( p_4 )</td>
<td>--</td>
<td>4.342</td>
</tr>
<tr>
<td>Relaxation Time</td>
<td>( q_1 )</td>
<td>s</td>
<td>0.463</td>
</tr>
<tr>
<td>Relaxation Time</td>
<td>( q_2 )</td>
<td>s</td>
<td>6.41E-5</td>
</tr>
<tr>
<td>Relaxation Time</td>
<td>( q_3 )</td>
<td>s</td>
<td>1.16E-7</td>
</tr>
<tr>
<td>Relaxation Time</td>
<td>( q_4 )</td>
<td>s</td>
<td>7.32E-10</td>
</tr>
</tbody>
</table>
Polyurea was assumed to have a time-dependent, viscoelastic deviatoric response, and was represented with the strength model as follows:

\[
\sigma'(t) = 2G_\infty \frac{T}{T_{ref}} \int_0^t \left(1 + \sum_{i=1}^n p_i \exp \left(-\frac{(\xi(t)) - (\xi(\tau))}{q_i}\right) D'(\tau)\right) d\tau
\]

where \(G_\infty\) is the 'long-term' shear modulus (at time approaches infinity), \(n\) is the number of Prony series coefficients, \(\xi\) is the reduced time, and \(D'\) is the deviatoric deformation tensor, and \(\sigma'\) is the time-dependent deviatoric stress. The reduced time, \(\xi\), accounts for the effect of temperature (not relevant in this study) and pressure on the relaxation kinetics.

\[
\xi(t) = \int_0^t \frac{dt}{10^A(t - C_{TP}P - T_{ref})/(B + T - C_{TP}P - T_{ref})}
\]

where \(A\), \(B\), and \(C_{TP}\) are material constants specified in Table 3.

**Skull Material Model**

The skull was considered to be isotropic with a relatively low compressibility, and was modeled according to a previous study [29]. The Mie-Gruneisen equation of state, often employed for a shock-compressed solid, is used in the form:

\[
P = \frac{\rho_0 C_0^2 \left(1 - \frac{\rho_0}{\rho}\right)}{\left[1 - s \left(1 - \frac{\rho_0}{\rho}\right)\right]^2}
\]

where \(\rho_0\) is the initial (reference) density, \(C_0\) is the initial (reference) sound speed, and \(s\) is defined as:
\[ U_s = C_0 + s U_p \]

where \( U_s \) is the shock speed, and \( U_p \) is the resulting particle velocity. In this equation of state, as the material undergoes a high compression, the bulk modulus increases, thereby allowing for supporting of shock. The values used for the parameters in the equation of state are displayed in Table 3.3.

Table 3.3. Parameters, symbols, units, and values utilized in the material model for the skull. [Adapted from 29]

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>Unit</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density</td>
<td>( \rho )</td>
<td>Kg/m³</td>
<td>1412</td>
</tr>
<tr>
<td>Young’s Modulus</td>
<td>( E )</td>
<td>GPa</td>
<td>6.50</td>
</tr>
<tr>
<td>Poisson’s Ratio</td>
<td>( v )</td>
<td>--</td>
<td>0.22</td>
</tr>
<tr>
<td>Sound Speed</td>
<td>( C_0 )</td>
<td>m/s</td>
<td>1850</td>
</tr>
<tr>
<td>Shock Speed vs. Particle Velocity Slope</td>
<td>( K_2 )</td>
<td>--</td>
<td>0.94</td>
</tr>
</tbody>
</table>

The skull does not experience large shear strains, due to a high shear rigidity, and was therefore given the same linear elastic strength model as that of the Kevlar composite.

Cerebrospinal Fluid (CSF) and Brain Matter Material Model

The CSF and brain matter were given the same material models with values of material parameters, as shown in Table 3.4. The equation of state was assigned under the assumption of isotropy, homogeneity, and elasticity of the materials. Since CSF and brain matter are mostly comprised of water and primarily undergo compressive loading
during shock conditions, a Tait equation of state was used, as shown below:

\[ P = B \left[ \left( \frac{\rho}{\rho_0} \right)^{\Gamma_0 + 1} - 1 \right] \]

where \( B \) and \( \Gamma_0 \) are Tait parameters set to the values for water.

Table 3.4. Parameters, symbols, units, and values utilized in the material model for the cerebrum and CSF. [29]

<table>
<thead>
<tr>
<th>Structure</th>
<th>Density ( \rho ) [kg/m(^3)]</th>
<th>Bulk Modulus ( K ) [GPa]</th>
<th>Tait EOS Parameters</th>
<th>Shear Modulus ( \mu ) [kPa]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cerebrum</td>
<td>1040</td>
<td>2.19</td>
<td>306.3</td>
<td>6.15</td>
</tr>
<tr>
<td>CSF</td>
<td>1040</td>
<td>2.19</td>
<td>306.3</td>
<td>6.15</td>
</tr>
</tbody>
</table>

The CSF and cerebrum have a time-dependent deviatoric response; however, the relaxation times calculated by a previous study are significantly longer than the times investigated in this study. Therefore, the Neo-Hookean strength model utilized assumes time-independency as well as hyper-elasticity due to the large deformations and motions incurred during loading:

\[ \sigma' = J^{-1} F \left[ \mu (\log \sqrt{C})^{\text{dev}} \right] F^T \]

where \( F \) and \( J \) were defined in the Polyurea model, \( \mu \) is the shear modulus, and \( C \) is the right Cauchy-Green deformation tensor defined by \( C = F^T F \).
Ethylene Vinyl Acetate (EVA) Foam Material Model

Ethylene Vinyl Acetate, as a hyperelastic, highly-compressible, non-linear, elastomeric foam, has a behavior best described by a strain energy function:

\[ W = \sum_{i=1}^{N} \frac{2\mu_i}{\alpha_i^2} \left[ \lambda_1^{\alpha_i} + \lambda_2^{\alpha_i} + \lambda_3^{\alpha_i} - 3 + \frac{1}{\beta_i} ((J)^{-\alpha_i}\beta_i - 1) \right] \]

where \( N \) represents the number of terms in the summation and was set to be 2, \( \mu_i, \alpha_i, \) and \( \beta_i \) are material-dependent parameters, \( \lambda_i \) are stretches as defined by \( \lambda_1 = \text{trace}(U) \), \( \lambda_2 = \frac{1}{2} [\text{trace}^2(U) - \text{trace}(U^2)] \), and \( \lambda_3 = \text{det}(U) \), and \( J = \lambda_1 \lambda_2 \lambda_3 = \text{det}(F) \). Since \( \lambda \) is defined through the use of the right stretch tensor, \( U \), which is related to the deformation gradient, \( F \), it is related to the deviatoric aspect of strain. To better show an interdependency of volumetric and deviatoric terms within the strain energy equation, the equation can be rewritten as:

\[ W = \sum_{i=1}^{N} \frac{2\mu_i}{\alpha_i^2} \left[ J^{-\frac{1}{3}\alpha_i}(\tilde{\lambda}_1^{-\alpha_i} + \tilde{\lambda}_2^{-\alpha_i} + \tilde{\lambda}_3^{-\alpha_i} - 3) + 3 \left( J^{-\frac{1}{3}\alpha_i} - 1 \right) + \frac{1}{\beta_i} ((J)^{-\alpha_i}\beta_i - 1) \right] \]

where \( \tilde{\lambda}_i = J^{-\frac{1}{3}\lambda_i} \). To calculate stress, the second Piola-Kirchhoff stress tensor can be defined by the strain energy as well as the right Cauchy-Green deformation tensor \( (C = 0.5(F^T F - I)) \) as:

\[ S = \frac{\partial W}{\partial C} = 2 \frac{\partial W}{\partial \lambda_k} \frac{\partial \lambda_k}{\partial C} \]

Rewriting the partial derivative allows us to plug in the values for \( \frac{\partial W}{\partial \lambda_k} \) (obtained by differentiating the first strain energy equation), and \( \frac{\partial \lambda_k}{\partial C} \) (obtained by the summation...
of the manipulation of equations for the $\lambda_k$ values as well as $C$. The true (total) Cauchy stress in the undeformed configuration can then be defined simply as:

$$\sigma = J^{-1} F S F^T$$

As the total stress is provided, no separate equation of state or strength model was provided for this material. Values for input parameters are given in Table 3.5.

Table 3.5. Parameters, symbols, units, and values utilized in the material model for the Ethylene Vinyl Acetate. [29]

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>Unit</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shear Modulus</td>
<td>$\mu_1$</td>
<td>MPa</td>
<td>8.874</td>
</tr>
<tr>
<td>Shear Modulus</td>
<td>$\mu_2$</td>
<td>MPa</td>
<td>-7.827</td>
</tr>
<tr>
<td>Exponent</td>
<td>$\alpha_1$</td>
<td>--</td>
<td>2.028</td>
</tr>
<tr>
<td>Exponent</td>
<td>$\alpha_1$</td>
<td>--</td>
<td>1.345</td>
</tr>
<tr>
<td>Exponent</td>
<td>$\beta_1$</td>
<td>--</td>
<td>0.32</td>
</tr>
<tr>
<td>Exponent</td>
<td>$\beta_2$</td>
<td>--</td>
<td>0.32</td>
</tr>
</tbody>
</table>

Integumentary and Muscle Tissue Material Model

The material covering the skull in the head model, normally comprised of all the skin layers as well as facial muscles were grouped to be a single material and modeled as a Mooney-Rivlin hyperelastic isotropic material. These tissues do not normally possess significant differences in shock impedance; therefore homogenizing these tissues without great compromise to the results can be done. This model functioned similarly to that of the EVA model, by the use of strain energy and Cauchy stress. This strain energy equation is as follows:
\[ W = A_1(I_1 - 3) + A_2(I_2 - 3) + A_3(I_3^{-2} - 1) + A_4(I_3 - 1)^2 \]

where \( A_3 \) and \( A_4 \) are related to \( A_1 \) and \( A_2 \) through:

\[ A_3 = \frac{1}{2} A_1 + A_2 \text{ and} \]

\[ A_4 = \frac{A_1(5\nu - 2) + A_2(11\nu - 5)}{2(1 - 2\nu)} \]

The method of obtaining the second Piola-Kirchhoff stress tensor is similar to that of the EVA material, involving manipulation of the partial differential equation, and the Cauchy stress is calculated. Again, there is no separate equation of state and strength model for this material.

**Problem Definition**

Initial air pressure was set to 1 atmosphere (ambient pressure). Recent research in mTBI [1] has focused on overpressures of no more than 1 atm, the biphasic Friedlander function with a peak pressure of 1 atm overpressure was utilized to describe the pressure impulse. This pressure impulse was applied to the inflow Eulerian domain face closest to the left side of the subject (from the subject’s perspective) and allowed to propagate towards the subject. Outflow boundary conditions were applied to the Eulerian domain face parallel with the inflow face, and no flow boundary conditions were applied to the remaining four Eulerian domain faces. The neck was simulated and simplified via the use of both: a) a coupling of a set of nodes at the skull base and a kinematically constrained reference joint to form a revolute joint, and b) six axial
connectors with an elastic stiffness placed between four couplings on the skull and base of the “neck” (four nodes fixed in space) to mimic the effect of neck muscles to resist motion and induce angular motion of the head.
This chapter will be organized based on variables presented in the problem. The first-fifth sections will focus on results of all of the cases; the second section will report the crucial mechanical quantities responsible for different mTBIs; and the sixth section will explore the values of those quantities to examine the relative efficacy of shock mitigation. All images used in this section will show a coronal cut of the head shown in the anatomical position (face forward). The blast wave enters the head from the subject’s left side (right side of the image). To be succinct, the terms “proximal” and “distal” will be used in this chapter to denote the relative distance from the incoming blast wave. The subject’s left side will be referred to as the “proximal” side, and the subject’s right side will be referred to as the “distal” side.

**Unprotected Head Case**

As expected, the unprotected head case experienced high levels of intracranial stress. The blast overpressure wave is shown in Figure 4.1 during four simulation times representing key events in the temporal evolution of the pressure: (a) .20ms; (b) .25ms; (c) .35ms; and (d) .42ms. The peak pressure is shown to propagate, intracranially, in the direction of the blast, reflect, propagate in the opposite direction, reflect again, and, finally, propagate in the initial direction. The air domain allows the planar shock
pressure wave to propagate until it impacts the head construct. As expected, the pressure wave is shown to travel faster through the head than through air. Because of this, the intracranial shock wave reflects several times within the skull before the air pressure wave reaches the opposite side of the head. As a result of these reflections, the shock wave undergoes attenuation and dispersion/decomposition.

Figure 4.1. An example of the temporal evolution of the (over)pressure and its spatial distribution over the mid-coronal section in both the (air) Eulerian and the (skin/skull/brain) Lagrangian domains. Simulation times: (a) .20ms; (b) .25ms; (c) .35ms; and (d) .42ms.
An examination of the maximum principal normal stress in Figure 4.2 also reveals evidence of shock wave reflection. Prior to reflection, it is shown that the distribution and intensity of this stress (similar to the pressure) is governed by the shock wave transfer from the air to the skin then to the skull and finally into the intracranial matter. After the shock wave reaches the distal skull, the distribution and intensity is governed by the reflections at the material boundaries and interactions between the reflected shock waves. The absolute maximum principle normal stress value was found to be 600 kPa and occurred in a few dynamically changing locations in the intracranial cavity. Aside from this extreme, the shock wave front, especially when reaching and reflecting off of the skull, showed a maximum principal stress of 125 kPa, suggesting a 25% increase relative to the incident blast overpressure loading conditions. It can also be noted that the skull exhibited large values of maximum principal stress, due to the high relative stiffness of the material.
Figure 4.2. Spatial distribution of the maximum principal stress over the mid-coronal section of the head, for the unprotected head case, at three simulation times: (a) 0.24ms; (b) 0.26ms; (c) 0.28ms.

As with the maximum principal stress, maximum shear stress, displayed at three simulation times in Figure 4.3, had a high peak value (60 kPa) found only in a few dynamically changing regions of the brain, with a lower peak value (5 kPa) found more widespread in the intracranial cavity. Unlike the maximum principal stress, there was no visual representation of the “shock wave” in the intracranial cavity, which was expected due to the inviscid nature of the CSF, obstructing shear stress from being transmitted from the skull to the brain. The shear stresses that do form in the brain result from the reflections of the compressive and tensile shock waves through the different intracranial materials. Due to this, the shear stresses did not experience the initial stress peaks before attenuation and dispersion, but rather the opposite: a slow buildup of shear stresses over the simulation time. These shear stresses would attenuate and disperse.
eventually, but it was not seen in the duration of the simulation. The highest concentrations and localizations of shear stresses were found in the brainstem, due to brainstem material having the highest shear stiffness among the intracranial materials. Additionally, the skull possessed some large shear stresses, due to the high shear stiffness of the skull material.

![Spatial distribution of the maximum shear stress over the mid-coronal section of the head, for the unprotected head case, at three simulation times: (a) 0.35ms; (b) 0.44ms; (c) 0.48ms.](image)

Figure 4.3. Spatial distribution of the maximum shear stress over the mid-coronal section of the head, for the unprotected head case, at three simulation times: (a) 0.35ms; (b) 0.44ms; (c) 0.48ms.

Protected Head Case, ACH Standard

In the case of the ACH-protected head, qualitatively similar temporal evolutions of maximum principal and shear stresses were found, and therefore only differences will be discussed in this section. Maximum intracranial principal stress (275 kPa, Figure 4.4) was found to be reduced by a factor of 54% and maximum intracranial shear stress (45 kPa, Figure 4.5) was found to be reduced by a factor of 25%. Again, these maximum
values were only found in a few regions of the brain, while a maximum principal stress value of 50 kPa and a maximum shear value of 3 kPa were more ubiquitous as well as closer to the range of stresses. This maximum intracranial principal stress was found to be reduced by a factor of 60% and the maximum intracranial shear stress was found to be reduced by a factor of 40%.

The decreased maximums in stresses seems to display evidence of some shock mitigation capabilities of the helmet. A comparison of Figures 4.4 and 4.2 show a difference in shock wave propagation through the head. In the case of the unprotected head, the shock wave appears to remain planar, while in the case of the ACH-protected head, the shock wave displays two main entry mechanisms: 1) the shock wave loads the helmet and passes traverses to the distal side of the helmet, loading the distal pads, and entering from the contrecoup direction and 2) the shock wave travels under the helmet and loads the head diagonally upwards from around the jaw area. As in the previous case, once the intracranial blast wave reaches the distal skull, an increase in stress magnitude is seen as the wave experiences an additive effect. Additionally, stresses are seen to transfer from the proximal suspension pads to the proximal head. These loading mechanisms are consistent with earlier findings by Nyein that the presence of a face shield drastically decreases stresses. [1]
Figure 4.4. Spatial distribution of the maximum principal stress over the mid-coronal section of the head, for the head protected by the standard ACH, at three post blast-impact times: (a) 0.14ms; (b) 0.28ms; (c) 0.47ms.

Figure 4.5. Spatial distribution of the maximum shear stress over the mid-coronal section of the head, for the head protected by the standard ACH, at three simulation times: (a) 0.06ms; (b) 0.11ms; (c) 0.22ms.

Protected Head Case, Polyurea Pads

As in the case of the standard ACH, qualitatively similar results were found. The main loading pathways found in the standard ACH case were found here again. Figure
4.6 displays the maximum principal stress, with a few regions possessing up to 375 kPa, and the majority of the intracranial region experiencing maximums of 75 kPa. Figure 4.7 displays the maximum shear stress, with a few regions possessing up to 48 kPa, and the majority of the intracranial region experiencing maximums of 4 kPa. Both the maximum principal and shear stress maximums were higher than the values found in the case of the standard ACH protected head, but still lower than in the case of the unprotected head.

![Spatial distribution of the maximum principal stress over the mid-coronal section of the head, for the head protected by the ACH with polyurea pads, at three post blast-impact times: (a) 0.12ms; (b) 0.21ms; (c) 0.31ms.](image)
Figure 4.7. Spatial distribution of the maximum shear stress over the mid-coronal section of the head, for the head protected by the ACH with polyurea pads, at four post blast-impact times: (a) 0.11ms; (b) 0.22ms; (c) 0.35ms.

Protected Head Case, Polyurea Inner Lining

Qualitatively similar results to the other cases were found. The main loading pathways found in the standard ACH case were found here again. Figure 4.8 displays the maximum principal stress, with a few regions possessing up to 270 kPa, and the majority of the intracranial region experiencing maximums of 45 kPa. Figure 4.9 displays the maximum shear stress, with a few regions possessing up to 42 kPa, and the majority of the intracranial region experiencing maximums of 3 kPa. Both the maximum principal and shear stress maximums were lower than the values found in the case of the standard ACH protected head as well as in the case of the unprotected head.
Figure 4.8. Spatial distribution of the maximum principal stress over the mid-coronal section of the head, for the head protected by the standard ACH with the polyurea inner lining, at three post blast-impact times: (a) 0.22ms; (b) 0.27ms; (c) 0.31ms.

Figure 4.9. Spatial distribution of the maximum shear stress over the mid-coronal section of the head, for the head protected by the standard ACH with the polyurea inner lining, at four post blast-impact times: (a) 0.07ms; (b) 0.10ms; (c) 0.27ms.
**Protected Head Case, Polyurea External Coating**

Qualitatively similar results to the other cases were found. The main loading pathways found in the standard ACH case were found here again. Figure 4.10 displays the maximum principal stress, with a few regions possessing up to 274 kPa, and the majority of the intracranial region experiencing maximums of 47 kPa. Figure 4.11 displays the maximum shear stress, with a few regions possessing up to 40 kPa, and the majority of the intracranial region experiencing maximums of around 3 kPa. The peak maximum principal stress were lower than that the values found in the unprotected and standard ACH case, but lower than in the polyurea inner lining case. The peak shear stress was the lowest in all of the investigated cases.

![Spatial distribution of the maximum principal stress over the mid-coronal section of the head](image)

**Figure 4.10.** Spatial distribution of the maximum principal stress over the mid-coronal section of the head, for the head protected by the standard ACH with the polyurea external coating, at three post blast-impact times: (a) 0.22ms; (b) 0.28ms; (c) 0.32ms.
Figure 4.11. Spatial distribution of the maximum shear stress over the mid-coronal section of the head, for the head protected by the standard ACH with the polyurea external coating, at four post blast-impact times: (a) 0.11ms; (b) 0.17ms; (c) 0.40ms.
CHAPTER FIVE
DISCUSSION

Shock Mitigation Efficacy of Different ACH Configurations

To analyze the results, the causes of the three main forms of mTBI must be explored. This section is divided into Diffuse Axonal Injury, Contusion, Subdural Hemorrhage, and Overall Efficacy. The graphs presented in this section refer to the different air/helmet/head assembly cases in the following manner: A) Unprotected head, B) Standard ACH protected head, C) ACH protected head with PU pads, D) ACH protected head with a PU inner lining, and E) ACH protected head with a PU external coating.

Diffuse Axonal Injury

As mentioned earlier, the primary mechanical causes of DAI are tensile and shear stress found in the white matter regions of the brain. The maximum tensile stresses and maximum shear stresses for the five cases are shown in Figures 5.1 and 5.2, respectively. Examination of these figures revealed that:

(a) The standard ACH case (Case B) significantly reduces the probability and severity of blast-induced DAI compared to the unprotected head (Case A), since the peak maximum tensile and shear stresses are reduced by ~40-50%. However, the ACH in its standard configuration is not considered as providing adequate protection against
this type of injury, since the stresses still present are significant. Hence, for any proposed ACH alteration to be considered effective, it must substantially reduce the peak tensile and shear stress levels below those found in the standard ACH case;

(b) Replacement of the EVA suspension pads (Case B) with the polyurea pads (Case C) has an adverse effect, since the resulting peak maximum tensile stress and peak maximum shear stress are higher in Case C by ~30% and ~10%, respectively;

(c) Addition of a polyurea internal lining (Case D) to the standard ACH design (Case B) only offers a minimal shock mitigation benefit of ~2% and ~6% reductions in peak maximum tensile stress and peak maximum shear stress, respectively;

(d) Addition of a polyurea external coating (Case E) to the standard ACH design (Case B) also only offers only a minimal shock mitigation benefit of ~1% and ~8% reductions in peak maximum tensile stress and peak maximum shear stress, respectively; and

(e) Among the four ACH configurations, the two which yield the lowest probability for diffuse axonal injury are those containing additions of a 2 mm thick polyurea internal lining (Case D) and external coating (Case E).
Figure 5.1. Peak values of the maximum tensile stresses generated in the five cases.

Figure 5.2. Peak values of the maximum shear stresses generated in the five cases.
Contusion

As mentioned earlier, the primary mechanical cause of contusion is the impact of the brain against the skull, which would cause a relative reduction in CSF thickness. Maximum CSF thicknesses for the five cases are shown in Figure 5.3, respectively. Examination of this figure revealed that:

(a) A comparison of cases A and B shows that the standard ACH case significantly reduces the probability and severity of blast-induced contusion, since the peak CSF thickness reduction decreases by ~50%. However, as in the case of diffuse axonal injury, the current ACH is generally believed to be ineffective in blast mitigation and an ACH alteration can only be considered adequate and effective in blast mitigation if it leads to a substantial decrease in peak CSF thickness reduction from the current ACH configuration;

(b) Replacement of the EVA suspension pads (Case B) with the polyurea pads (Case C) has a severe adverse effect, since the peak CSF thickness reduction increases by nearly 200%. It appears that polyurea pads act as high shock-impedance wave guides which provide a more direct ingress of shock waves generated within the helmet shell into the head.;

(c) Addition of a polyurea internal lining (Case D) to the standard ACH design (Case B) offers a marginal shock mitigation benefit since the peak CSF thickness reduction decreases by ~10%;
(d) Addition of a polyurea external coating (Case E) to the standard ACH design (Case B) slightly adversely affects the ACH, as the peak CSF thickness reduction increases by ~10%; and

(e) Among the four ACH configurations, the one which yields the lowest probability for contusion is the case involving an addition of a polyurea internal lining (Case D).

Figure 5.3. Peak values of the CSF thickness reduction generated in the five cases.

Subdural Hemorrhage

As mentioned earlier, the primary mechanical cause of subdural hemorrhage is the impact of the brain against the skull, which would cause a relative reduction in CSF
thickness. Maximum CSF thicknesses for the five cases are shown in Figure 5.4, respectively. Examination of this figure revealed that:

(a) A comparison of the unprotected and standard ACH case shows that the helmet significantly reduces the probability and severity of blast-induced subdural hemorrhage, since the peak brain-surface shear stress is reduced by ~40%. However, as in the case of diffuse axonal injury and contusion, the current ACH is generally believed to be ineffective in blast mitigation and only an ACH alteration leading to a substantial decrease in peak brain-surface shear stress from the current ACH configuration could be considered to provide adequate protection against this type of injury;

(b) Replacement of the EVA for polyuria in the suspension pads (Case C) has an adverse effect, since the resulting peak brain-surface shear stress is higher by ~70%;

(c) Addition of a polyurea internal lining (Case D) to the standard ACH design (Case B) offers only a minimal shock mitigation benefit since the accompanying reduction in peak brain–surface shear stress is ~6%;

(d) Addition of a polyurea external coating (Case E) to the standard ACH design (Case B) also only offers only a minimal shock mitigation benefit since the accompanying reduction in peak brain–surface shear stress is ~3%; and

(e) Among the four ACH configurations, the one which yields the lowest probability for subdural hemorrhage is the case involving an addition of a polyurea internal lining (Case D).
Figure 5.4. Peak values of the shear stresses on the surface of the cerebrum generated in the five cases.

**Overall Efficacy**

With all three forms of mTBI, Case D, involving the 2 mm thick polyurea internal lining, performed the best, yielding the lowest stress and CSF thickness reduction values. Case E, involving the 2 mm thick polyurea external coating, also yielded lower stress and CSF thickness values than the standard ACH case. However, the additional blast-mitigation protection provided by these two cases are not significant compared to the effect provided by the standard ACH alone. Based on this finding, the ACH augmentations investigated are not adequate to provide the desired and required level of blast-induced mTBI protection.
Three polyurea augmentations (pads, inner lining, and external coating) to the current Advanced Combat Helmet were investigated regarding protection against mild traumatic brain injury in comparison to both an unprotected head and a standard ACH. The standard ACH, in itself provides a certain degree of protection against mTBI and alters the blast loading path, showing that the blast wave prefers to travel under the helmet and load the brain in a diagonal manner. The replacement of a hyperelastic foam with a polyurea as the material for the suspension pads adversely affected the minimal blast mitigating effects of the current ACH. Both the addition of the inner lining and external coating provided added protection against blast, however, this benefit was found to be minimal with respect to the initial protection provided with the presence of the current ACH design.

The present investigation was of a purely computational nature and involved the use of a series of combined Eulerian/Lagrangian transient nonlinear dynamics finite element fluid/solid interaction analyses. The use of these analyses entails construction, parameterization, and validation of detailed constitutive models for a number of structural materials as well as for a number of hard and soft-tissue biological materials. Especially in the case of biological materials, these constitutive models are based on empirical data gathered in other studies, and require simplification that, although
suggested to not affect results based on multi-scale analyses, may affect interactions with other materials. Additionally, the complex user subroutine written to characterize the behavior of polyurea may not completely capture the microstructural interactions with the specific loading conditions attributed.

The CSF was modeled in a Lagrangian manner within this study, which inhibits the brain and the skull to undergo direct contact. As explained earlier, Lagrangian elements deform with the material, and no amount of deformation is able to decrease any dimension of an element to zero. In actual physiological conditions, the CSF, encased in the meninges, are able to contact the skull, and this may allow for some unforeseen behaviors to occur. The CSF was not modeled in an Eulerian manner in this study due to the high computational cost associated with Eulerian domains in finite element analyses.

While the augmentations investigated may not have produced significant differences to the blast-mitigation capability of the ACH, a sandwiching structure (effectively combining the inner lining and external coating benefits) may be a future candidate for investigation. Other microcellular foams (such as carbon foam) could also prove to be effective in augmenting the ACH. An experiment comparing the blast energy absorption effects of different materials alone would be the best starting point for any future investigations, computational or otherwise.
REFERENCES


