A Thesis

entitled

Mitigating Concussion: An Innovative Football Helmet

By

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Helmets are the primary safety devices for various sports, including football. A well-designed helmet is generally believed to be effective protection against head injuries and concussions. However, there are a multitude of cases where the helmet has failed in protecting the player, resulting in head injuries. These injuries include massive trauma to the brain resulting in coma, paralysis or even death. Therefore, there is the need for a more effective football helmet.

The helmet material that acts as an energy absorber is the most important parameter in helmet design. The main objective of this study is to investigate the impact response and energy absorption capability of different shock absorbing materials, in order to find an alternative safer and more effective design. The resulting design is expected to be more effective in the reducing impact force and decreased related head injuries.

To this end, various types of head injuries and their main causes have been reviewed. Theoretical and experimental techniques are used to compare the efficiency of different energy absorbing materials including the foam paddings (used in current helmets). Additionally, alternative designs are proposed. A new padding structure has been
developed and tested in a football helmet. The impact performance of an off the shelf helmet is compared to the proposed design. Drop tests based on the NOCSAE standard were performed. Results of these tests show the effectiveness of the designed padding in reducing the possibility for head injuries.
Acknowledgements

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Chapter One

Introduction

In 1888, the college football rules convention voted to allow tackling below the waist. Players and coaches soon regarded pads as essential for the game. However, helmets were the last safety device to be accepted [1]. They were not a mandatory piece of equipment in college football until 1939 and they were not made mandatory in the National Football League until 1943 [1].

From 1943, the helmets, football helmets in particular (Figure 1-1), have been improving and several studies have been focused on designing new helmets and using new technologies to improve their head protection capability.
A good design needs a thorough understanding of the concept of the product. Initially it is required to know the common parts of a helmet, the effectiveness of each part for desired application and its purpose. In order to highlight the importance of each part of the helmet in the design process, a brief description for the function of each part is given.

The helmet generally consists of three main parts (Figure 1-2):

- Shell: The main purpose of a helmet is to prevent the head from direct impact of external objects and to disperse the impact over an extended area. The hard outer surface of the helmet is called the shell. The shell’s two functions are absorbing the impact energy and dispersing the impact force. Generally, the shell does not absorb a large amount of energy.
- Liner: This part is the most important part in protecting the head from concussions and is the main energy absorber of the helmet. Padding foams are the most popular liner materials used in the helmets. A liner usually consists of two different parts. The main part is a series of padding foams or similar absorber type of foam. These absorb the impact energy. The other part is a comfort pad, which basically fills the gap between the head and inside of padding foam to make a more comfortable fit of the helmet on the head.

- Face Guard: Face guard is generally used for protecting face and chin from any direct impact.

In order to have a better general understanding of the helmet and its design concepts (and develop numerical studies), it is important to know the dynamic modeling and behavior of the helmet. Based on this model, one can calculate the force transmitted through the helmet to the head. Various dynamic modeling concepts have been developed.
A very well-known lumped-parameter model of the helmet that has been developed by Gilchrist and Mills [4] is given in Figure 1-3, for which the load paths between the object impacted and the head are shown in Figure 1-3-a and the corresponding model proposed by Gilchrist and Mills [4] is shown in Figure 1-3-b [5]. Four masses are involved: M1 the mass of the steel striker or anvil, M2 the mass of the helmet shell, M3 the mass of the helmet liner foam and M4 the mass of the headform [5].
In this model, load path 1 involves bending of the shell (parameters $K_1$ and $C_1$) and elastic deformation of the foam (parameters $K_2$ and $C_2$), whereas load path 2 represents the direct force-deflection relationship of the crushed polystyrene foam. Both paths involve the deformation of the comfort foam (parameters $K_3$ and $C_3$)[5]. Movement of all the masses was restricted to the vertical axis, so this is a one-dimensional model [5].

To apply this model and similar models, parameters such as stiffness values and damping coefficients of the helmet materials should be taken into account in order to create a new design or any modification to improve the helmet functionality. These parameters must be carefully considered. Knowing the importance of these design parameters, an attempt was made in this study to consider the effect of the stiffness values in a design of the helmet, which was prepared to improve the capability of the helmet in reducing the concussions that is more discussed in following chapters.
1.1. Standards

This section discusses the standards regarding assessing the helmet characteristics and their effectiveness in reducing the possibility of the head injury. For different types of helmets, various standards have been developed over the last 30 years, offering the instructions and experimental steps in determining the effectiveness of the helmets in mitigating head injuries. Some of the standards are related to bicycle and motor bike helmets such as DOT, Snell and ECE. Some standards are for various types of safety helmets such as ASTM and finally some other standards are related to sport helmets such as NOCSAE. The National Operating Committee on Standards for Athletic Equipment, NOCSAE, is the standard discussed in this section and contains assessment specifications for different types of sport equipment such as protective head gear for football, baseball, hockey etc.

In this study, ASTM and NOCSAE have been used separately for two different steps of the project. The ASTM quasi-static compression testing has been utilized in initial evaluation of the candidate materials for the helmet and the liner while the final design evaluation and validation has been done based on NOCSAE standard.

ASTM standard for compressive properties of rigid cellular plastics describes the process of determining the compression properties of rigid cellular materials that undergo compressive loads. Test data is obtained, and from a complete load-deformation curve it is possible to compute the desired parameters such as the energy absorption capacity of the material. Based on the ASTM standard, a testing instrument that includes both a stationary and movable members and includes a drive system for imparting to the
movable member (crosshead), a uniform, controlled velocity with respect to the stationary member (base) is needed [6]. The testing machine shall also include the following components:

- Compression Platens: Two flat plates, one attached to the stationary base of the testing instrument and the other attached to the moving crosshead to deliver the load to the test specimen. These plates shall be larger than the specimen loading surface to ensure that the specimen loading is uniform [6].

- Load Measurement System: A load measurement system capable of accurately recording the compressive load imparted to the test specimen. The system shall indicate the load with an accuracy of ± 1% of the measured value or better [6].

- Displacement Measurement System: A displacement measurement system capable of accurately recording the compressive deformation of the test specimen during testing to an accuracy of ± 1% of the measured value or better [6].

The test specimen shall be either a square or circular cross section with a minimum area of 25.8 $cm^2$ (4 $in^2$) and a maximum of 232 $cm^2$ (36 $in^2$) in area. The minimum height shall be 25.4mm (1in.), and the maximum height shall be no greater than the width or diameter of the specimen [6]. All surfaces of the specimen shall be free from large visible flaws or imperfections [6].

Thus after measuring the dimensions and preparing a valid specimen for testing, the specimen shall be placed between the compression plates, ensuring that the specimen center-line is aligned with the center-line of the compression platens so the load will be distributed as uniformly as possible over the entire loading surface of the specimen [6]. Then the test should be started by moving the crosshead in the direction to compress the
specimen with the rate of crosshead displacement of $(2.5 \pm 0.25 \text{ mm})/\text{min} \ (0.1 \pm 0.01 \text{ in.})/\text{min}$ for each 25.4 mm (1 in.) of specimen thickness [6]. The test should be continued until a yield point is reached or until the specimen has been compressed approximately 13% of its original thickness, whichever occurs first [6].

The methods of test and performance, required by NOCSAE, are based on research initiated in 1971 at Wayne State University, Department of Neurosurgery Biomechanics Laboratory [7]. NOCSAE recognizes the difficulty of formulating a laboratory standard to reduce head injury in an environment in which the injury incidence is relatively low [7]. Further, many injury mechanisms remain unknown, and no tolerable index is available for hemorrhagic injuries or subdural hematomas that are a primary cause of death and permanent injury in certain organized sports [7]. The NOCSAE drop test method defines impact limits for linear acceleration [7]. The procedure and specifications of NOCSSEA, on which the results of this study are based, is discussed in the following paragraphs.

Based on the NOCSAE standard an experimental setup generally consists of an arm on which to mount the headform and helmet, an anvil, accelerometers mounted inside the head form, sensors, gages and guidance cables for freely dropping the headform and helmet. More detailed information regarding the drop test mechanical system components can be seen in Figure 1-4 and Table 1.1.

As it is shown in Figure 1-4, the helmet without the faceguard is positioned on a proper size headform and then dropped in order to achieve an accepted free fall velocity. At impact, the instantaneous resultant acceleration is measured by the triaxial accelerometer, mounted inside the headform, and the Severity Index, a criterion for
assessing the severity of head injury, is calculated.

Physical properties of the NOCSAE headforms to be used in this standard drop test method are given in Figure 1-5 and Table 1.2:
Figure 1-4: Recommended guide and carriage assembly of NOCSAE [7]
Table 1.1: Recommended guide and carriage assembly of NOCSAE [7]

<table>
<thead>
<tr>
<th>CODE</th>
<th>DESCRIPTION</th>
<th>AVAILABILITY</th>
<th>DRAWINGS AVAILABLE</th>
<th>SIRC PART NO.</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Drop Carriage</td>
<td>SIRC</td>
<td>Yes</td>
<td>1001</td>
</tr>
<tr>
<td>2</td>
<td>½'' MEP Testing Pad</td>
<td>SIRC</td>
<td>No</td>
<td>1006</td>
</tr>
<tr>
<td>2</td>
<td>1/4'' MEP Faceguard Testing Pad</td>
<td>SIRC</td>
<td>No</td>
<td>1007</td>
</tr>
<tr>
<td>2</td>
<td>3'' MEP Calibration Pad</td>
<td>SIRC</td>
<td>No</td>
<td>1005</td>
</tr>
<tr>
<td>3</td>
<td>Hook-eye Tumbuckle, Forged Steel, 3/8'' with a 6'' take-up</td>
<td>SIRC/H</td>
<td>N</td>
<td>1043</td>
</tr>
<tr>
<td>4</td>
<td>3/8'' Wire Rope Thimble</td>
<td>SIRC/M</td>
<td>N</td>
<td>1044</td>
</tr>
<tr>
<td>5</td>
<td>1/8'' Spring Music Wire</td>
<td>SIRC/M</td>
<td>N</td>
<td>1045</td>
</tr>
<tr>
<td>6</td>
<td>1/8'' Wire Rope, Tiller Rope Clamp, Bronze</td>
<td>SIRC/M</td>
<td>N</td>
<td>1046</td>
</tr>
<tr>
<td>7</td>
<td>3/8'' 16 x 3'' Eye Bolt</td>
<td>SIRC/H</td>
<td>N</td>
<td>1041</td>
</tr>
<tr>
<td>8</td>
<td>3/8'' Forged Eye Bolt</td>
<td>SIRC/H</td>
<td>N</td>
<td>1040</td>
</tr>
<tr>
<td>9</td>
<td>Right Angle DC Host Motor</td>
<td>SIRC/G</td>
<td>N</td>
<td>2000</td>
</tr>
</tbody>
</table>

Not Shown

DC Motor Speed Controller (Reversible)  | SIRC/G        | N                  | 2001          |
Single Groove Sheave (Pulley), 3 ½''     | SIRC/G        | N                  | 2002          |
Top Mount Plate                         | SIRC         | Y                  | 2003          |
18'' Top Channel Bracket                | SIRC/H        | N                  | 2004          |
Wall Mount Channel Bracket, 4 x 1 5/8''  | SIRC/H        | N                  | 2005          |
Mechanical Release System                | SIRC         | Y                  | 2006          |
Lift Cable, Wire Rope, 20' Coil          | SIRC/H        | N                  | 2007          |
Anvil Base Plate                        | SIRC         | Y                  | 2010          |
Anvil                                   | SIRC         | Y                  | 2011          |
Headform Adjuster                       | SIRC         | Y                  | 2012          |
Headform Rotator Stem                   | SIRC         | Y                  | 2013          |
Headform Threaded Lockring              | SIRC         | Y                  | 2015          |
Headform Collar                         | SIRC         | Y                  | 2014          |
Nylon Bushing                           | SIRC         | Y                  | 1803          |
Small Headform                          | SIRC         | N                  | 1100          |
Medium Headform                         | SIRC         | N                  | 1101          |
Large Headform                          | SIRC         | N                  | 1102          |
Table 1.2: Approximate measurements of NOCSAE headforms [7]

<table>
<thead>
<tr>
<th>POINTS OF MEASURE</th>
<th>6 5/8</th>
<th>7 1/4</th>
<th>7 5/8</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Head Breadth</td>
<td>5.63 (143)</td>
<td>5.98 (152)</td>
<td>6.46 (164)</td>
</tr>
<tr>
<td>2 Maximum brow width (frontal diameter)</td>
<td>4.65 (118)</td>
<td>5.20 (132)</td>
<td>5.52 (140)</td>
</tr>
<tr>
<td>3 Ear hole to ear hole (bifrontal diameter)</td>
<td>5.24 (133)</td>
<td>5.51 (140)</td>
<td>6.06 (154)</td>
</tr>
<tr>
<td>4 Maximum jaw width (biparietal diameter)</td>
<td>4.13 (105)</td>
<td>4.65 (116)</td>
<td>5.08 (129)</td>
</tr>
<tr>
<td>5 Head length (glabella landmark to back of head)</td>
<td>7.09 (180)</td>
<td>7.87 (200)</td>
<td>8.15 (207)</td>
</tr>
<tr>
<td>6 Outside eye corner (external canthus) to back of head</td>
<td>6.22 (158)</td>
<td>6.81 (173)</td>
<td>7.32 (186)</td>
</tr>
<tr>
<td>7 Ear hole (tragion) to back of head</td>
<td>3.50 (89)</td>
<td>3.86 (96)</td>
<td>4.25 (108)</td>
</tr>
<tr>
<td>8 Ear hole to outside corner of eye (tragion to ext. canthus)</td>
<td>2.72 (69)</td>
<td>2.95 (75)</td>
<td>3.07 (78)</td>
</tr>
<tr>
<td>9 Ear hole to top of head (tragion to vertex)</td>
<td>4.72 (120)</td>
<td>5.24 (137)</td>
<td>5.67 (144)</td>
</tr>
<tr>
<td>10 Eye pupil to top of head</td>
<td>4.13 (105)</td>
<td>4.53 (115)</td>
<td>4.96 (126)</td>
</tr>
<tr>
<td>11 Ear hole to jaw angle (tragion to gonion)</td>
<td>3.31 (84)</td>
<td>3.03 (77)</td>
<td>2.84 (72)</td>
</tr>
<tr>
<td>12 Bottom of nose to point of chin (subnasal to menton)</td>
<td>2.56 (65)</td>
<td>2.86 (71)</td>
<td>3.03 (77)</td>
</tr>
<tr>
<td>13 Top of nose to point of chin (nasion to menton)</td>
<td>4.45 (113)</td>
<td>4.85 (122)</td>
<td>5.39 (137)</td>
</tr>
<tr>
<td>14 Head circumference</td>
<td>21.02 (534)</td>
<td>22.68 (576)</td>
<td>24.17 (614)</td>
</tr>
<tr>
<td>15 Head weight including mounting interface</td>
<td>9.08 lb (4.12 kg)</td>
<td>10.8 lb (4.90 kg)</td>
<td>10.08 lb (4.50 kg)</td>
</tr>
</tbody>
</table>

Figure 1-5: Approximate measurements of NOCSAE headforms - inches (mm) [7]
The dropping of the helmet is oriented such that eight different locations on the helmet are impacted as shown in Figure 1-6 and Figure 1-7.

The NOCSEA drop velocities are shown in Table 1.3.

Figure 1-6: Dropping locations on the head for NOCSAE [7]
Figure 1-7: Approximate impact locations [7]

Table 1.3: NOCSAE drop velocities [7]

LOCATION - DROP velocities – ft/s (m/s)
(All drop velocities must be within +3% -0%)

<table>
<thead>
<tr>
<th></th>
<th>FRONT</th>
<th>SIDE</th>
<th>F. BOSS</th>
<th>R. BOSS</th>
<th>REAR</th>
<th>TOP</th>
<th>RANDOM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ambient</td>
<td>11.34 (3.46)</td>
<td>11.34 (3.46)</td>
<td>11.34 (3.46)</td>
<td>11.34 (3.46)</td>
<td>11.34 (3.46)</td>
<td>11.34 (3.46)</td>
<td>11.34 (3.46)</td>
</tr>
<tr>
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NOTES: The high temperature condition impacts must be done after the ambient temperature impacts.

Impact locations requiring more than two (2) impacts must be conducted in sequence from the lowest drop velocity through the highest.
For the High Temperature impact condition, the helmet shall be exposed to a temperature of $115\pm 5^\circ F$ ($46 \pm 3^\circ C$) for a minimum of four hours and a maximum of twenty-four (24) hours.

The impact velocities specified in the appropriate NOCSAE performance standards for impact testing are measured during the last 1.5in. (40mm) of free fall for each test [7]. The measured velocities shall be within the limits specified in the appropriate NOCSAE performance standards [7]. According to NOCSAE, the Peak Severity Index (SI) is the criterion for assessing the severity of head injury. Any impact of the helmet shall be calculated and it should not exceed a specific level (1200SI) [7].

1.2. Objectives

The main objective of the research is to design a new football helmet using an effective absorption technology that can absorb significantly more energy than current helmets during the impact and protect the head against probable head injuries. This helmet will perform better in standard drop testing than helmets currently commercially available.

1.3. Approach

To achieve the desired objectives the first step is to look for good shock absorbers and try to select the best candidates using specific testing methods. To this end, standard
testing methods, such as an ASTM compression testing standard are used. The energy absorption capability of each shock absorber is assessed to find the best candidates.

The second step is to prepare a design based on an alternative shock absorber. The final step is to determine which one of the new designs is more effective by comparing their performance with those of the current helmets. To this end, the NOCSAE drop-testing standard is used and the modified helmet is compared with existing commercially available ones. This provides an objective way to assess the concept in improving the head protection characteristics of the helmet and reducing the head injury probability of football helmets.

1.4. Contributions

The main contribution of the work is an alternative football helmet. Various shock absorbing materials (including the existing helmet paddings) are evaluated. A prototype helmet was manufactured and was evaluated through drop testing.

1.5. Publication

A conference paper has been published based on the results of analysis on shape memory materials and a new design for the liner part of the helmet including shape memory wires and regular helmet foams. It was presented at ASME SMASIS 2011:
Chapter Two

Head Injury

Head injuries occur in a variety of incidents. They can involve direct impact with the head striking a hard surface like the ground or any other player’s body. Head injuries are especially common in helmet-to-helmet contact. They can also occur without direct impact, as in severe whiplash from blunt force trauma to the chest.

2.1. Head injury types

This section explores head injuries by focusing on the types of injury that are sustained by the skull, the neck, and the brain. This will lead to a discussion of the Head Injury Criteria (HIC), which are the indicators used to measure the potential for head injury.

Head injuries can be classified into cranial injuries (skull fractures) and intracranial injuries (injuries to vascular and neurological tissue) [1]. Multiple types of head injury can occur simultaneously when the head receives an impact. The
physiological consequences can be determined by the anatomical location of the possible lesions and their severity. The main anatomical structures of the head and their locations are shown in Figure 2-1.

Figure 2-1: Main anatomical structures of the head and their locations inside the head [8]

Head injuries can be classified in various ways. Gennarelli [9] has provided a biomechanical description and classification system of injuries, as shown in Table 2.1. The contact injuries are known that likely to be prevented, or even excluded, by a helmet because a helmet (that meets approved standards) spreads or diffuses any contact impact force and provides energy absorption under that contact point. The avulsion of the pinna is a typical injury when the unprotected head encounters the ground or other obstacles, and a helmet easily prevents such an injury.
Table 2.1: Mechanistic types of head injury [9]

- Contact injuries (requiring impact of the head; but head motion is not necessary)
  - Skull deformation injuries
    - Local
      a) Skull fractures (suture separation, indentation, linear, depressed, comminuted, crushing, massive comminuting)
      b) Epidural haemorrhage/haematoma EDH
      c) Coup contusions, lacerations, macerations, avulsion, extrusion
    - Remote
      a) Vault and basilar fractures
  - Stress waive injuries
    a) Countercoup contusion
    b) Intra cerebral haemorrhage/haematoma ICH
  - Inertial injuries (direct impact to the cranial vault is not necessary; head acceleration necessary)
    - Surface strains
      a) Subdural haematoma SDH
      b) Countercoup contusion
      c) Intermediate Coup contusions
    - Deep strains
      a) Contusion syndrome
      b) Diffuse axonal injury DAI
      c) Intra cerebral haemorrhage/haematoma ICH

Otte et al. [10] found that injuries were mostly located on the anterior side of the head. Although there were a large number of different kinetic patterns in impact and post-impact phases, the head with the face region, especially the chin and forehead, is nearly always exposed to risks. Approximately one third of all injuries to helmet-protected heads are minor soft part injuries, such as contusions and abrasions (32.9% and 32.8% respectively) [8]. More serious soft part injuries, such as laceration, contusions, cut or scalp injuries represent another 21.9% of helmet-protected head injuries, and 25.5% of unprotected head injuries [1]. Otte [11] also found that persons who suffered a chin impact remained uninjured in only 37% of the cases, while persons with an impact to the
helmet other than chin impact sustained no injury in 70.1% of the cases. Otte's injury analysis shows that as a rule, persons with chin impact suffer soft-part injuries three times as often (49.3% of the persons), twice as many fractures (18.1% of the persons) and twice as many skull-brain injuries (39.9% of the persons) as those without impact [1]. Fractures to the base and the top of the skull and also lower and upper jaws were especially frequent with chin impacts. Otte showed that fractures sustained by the skull with chin impacts are more than twice as frequent as those sustained otherwise [11]. For fractures of all parts of the skull and facial bones, except for cheek bones, it is true. He also found that an oblique frontal impact (case II-Figure 2-2) and a more sagittal impact from the front, perpendicular to the face (case I), produce different injury patterns, with fundamentally different impact kinematics, although soft-part injuries as well as fractures are characteristic for both types of chin impacts [1].
The difference in kinematics can be analyzed from an anatomical viewpoint. In case of the oblique frontal impact, more force is transmitted by the denture.

In general, head injuries happen in different configurations, involving direct or indirect contact. Using a helmet and more effectively, improving existing helmets, a considerable number of injuries could be prevented or reduced.
2.1.1. Cranial Injuries

Skull fractures can occur with and without brain damage, but skull fractures are not an important cause of neurological injury [9]. There are two forms of skull fracture: open and closed. A break in the bone but with no break of the overlying skin is a closed fracture. An open fracture, however, is a break involving both the skin and underlying bone and is more serious because of the risk of infections. Fractures to the neuro-cranium are divided into basilar skull fractures and vault fractures (fractures to the non-base part of the skull) [1]. Basilar fractures are known as clinically significant, because they may cause dura to be torn adjacent to the fracture site and as a result increase the possibility of contamination of the central nervous system. Vault fractures are divided into linear and depressed fractures. Linear fracture (no bone displacement) is considered not severe and does not have much significance on the course of brain injury, although this subject is still controversial [1][12]. Depressed fractures (with bone displacement) are likely to be associated with neural injury and/or intracranial hematoma, especially when the depression is deeper than the thickness of the skull [13].
2.1.2. Intracranial Injuries

Impact may cause various types of brain injury. Mainly two categories are considered: diffuse injury and focal injury. In order of increasing severity, the most important types of diffuse brain injury are discussed below [14].

- Mild concussion is the type of brain injury that results in confusion, disorientation and/or minor loss of memory [1]. This injury type is reversible and may not result in loss of consciousness.

- Classical cerebral concussion is an injury type which involves temporary loss of consciousness and lasts less than 24 hours, while it is reversible.

- Diffuse white matter shearing injury (DWSI) or diffuse axonal injury (DAI) is a severe type of diffuse brain injury with prolonged loss of consciousness (more than 24 hours) and may result in dysfunction of the brainstem.

Focal brain injuries are those cases in which a lesion has occurred large enough such that it can be visualized without special equipment and always include anatomical damage. Four different types of focal brain injuries are known and they include the following:

- Epidural haematoma (EDH) directly resulting from skull deformation, are usually associated with skull fracture and concern the meningeal vessels directly underneath the skull [1].

- Subdural hematoma (SDH) the most severe of which is the acute form (ASDH). The most common cause of ASDH is tearing of the bridging veins and arteries, crossing the subdural space [1].
• Contusion, the most frequently found trauma following head impact, occurs at the site of impact (coup contusion) or at remote sites of the impact (contre-coup contusion) [1].

• Intracerebral haematoma (ICH) include homogeneous collections of blood within the brain and are distinguished from contusions by a more pronounced localization of the haematoma [1]

2.2. Head injury mechanisms

Contact impact causes various mechanical effects to the head either because of contact phenomena or inertial effects or the combination thereof. Contact phenomena mostly result in focal head injuries. Another possibly important response of the head due to contact impact is the propagation of stress waves in the skull or the brain, which may cause focal injuries distant from the site of impact (contre-coup) [1]. Generally, an impact to the head results in acceleration of the head, which leads to inertial loading of the intracranial structures [1]. Accelerations can be translational (linear) and rotational (angular) and can result in concussion and diffuse brain injury rather than focal injury [1]. Rotational acceleration is the most important cause of severe head injury [14].

Figure 2-3 shows the occurrence of the most severe head injuries as a function of angular acceleration amplitude and time duration of this acceleration [12]. The trend is that at short pulse durations, cerebral concussion can be produced (along with cortical contusion) [1]. But the strain rate sensitive bridging veins may be torn which results in subdural haematomas, as the acceleration level increases. Cerebral concussion can be
found at lower acceleration magnitudes if the pulse durations are longer, but subdural bridging vein rupture is caused by more acceleration. The incidence of cerebral contusion also further decreases with increasing pulse duration (not shown in Figure 2-3) [1]. It is thought that shearing brain injuries are caused by high angular acceleration at longer pulse durations.

![Figure 2-3: Relationship between angular acceleration and head injury](image)

**Cranial injuries**

According to Gurdjian et al., linear skull fracture is caused by skull bending [16]. As an impact occurs on the side of the head, inward bending of the skull at the site of the impact and outward bending at some distance from the impact site take place. The
fracture of the skull is a result of deformation of the bone beyond its loading capacity. Since bone is weaker in tension than in compression, cracks will appear at the skull's outer surface in the regions where the skull bends outwards and on the inner surface in regions in which the skull bends inwards. In Figure 2-4, the arrows show the sites under tension that come from skull bending [1]. Several mechanisms were proposed as the cause of these fractures in a review article on basilar skull fracture by Huelke et al. [15]. Originally, it was thought that basilar fracture results from cranial vault impacts, causing deformations remote from the impact site [16][1]. Thom & Hurt found that axial loading of the neck was significantly associated with basilar fracture for un-helmeted motorcyclists [17][1]. There are indications that basilar skull fractures can also be caused by impacts to the face and especially the mandible [1][18]. In a study by Alem et al. [1][19] they impacted the crowns of the heads of un-embalmed cadavers and results showed that a rigid impacting surface with sufficient impact energy will cause fractures at the impact site [1]. However, under the same loading the fractures occur at the base of the skull if the impact site was padded. Increasing the thickness of the padding prevented all skull fractures, but the fractures then occurred in the cervical spine [1].
Injuries based on skull deformation are when fragments of bone resulting from skull fracture or penetration into the skull have been known to be the main cause of damage to underlying meningeal and cortical tissues. The dura, the tough outermost membrane enveloping the brain and spinal cord, is connected to the inner aspect of the cranial bones and contains several blood vessels [1]. Skull deformation or skull fracture can easily cause rupture of these blood vessels, leading to an extra dural haematoma [1][20]. Direct laceration of the bridging veins or the cortical veins and arteries can cause acute subdural haematomas by penetration wounds resulting from impacts to the head [9]. Large cortical contusions resulting from skull deformation or skull penetration can
lead to subdural haematomas [9]. Various studies have addressed skull denting as a cause of cortical contusions [9][19].

Another mechanism of injury is based on relative movement between the skull and the brain. At the vertex it can be seen that the skull is smooth; however, at the base, the skull is highly irregular. Therefore, the brain slides smoothly against the internal surface of the skull at the vertex, but is impeded at the skull base [1]. This can produce high shear strains in the meningeal and cortical tissues at the skull base. Most cerebral contusions occur at the frontal and temporal lobes [21][22] regardless of whether the site of impact is frontal or occipital [23]. On the other hand, the relative movements between the skull and the brain at the vertex lead to high strains in the structures tethering the brain to the vault of the skull [1].

Rupture of bridging veins due to these high strains are considered to be the main cause of subdural, cortical haematomas in this area [9][1][24][25]. Because of the non-spherical shape of the brain and skull, relative rotation between skull and the brain presses the skull base towards the brain (Figure 2-5). This may cause a combined compression and shearing of the meningeal and cortical tissues, which can lead to increasing the effects of the sliding of the brain over the skull base [27]. When the head is subjected to a backward non-centroidal rotational acceleration or if a forward non-centroidal rotational deceleration occurs, it can be seen that the effects of this relative rotation between the skull base and the brain are most severe. Since the relative movement between the skull and the brain was the brain moving towards the site of impact, intracranial tissue is compressed at the site of impact and it is also strained at the contra-lateral site. This may cause positive pressure at the site of impact and negative
pressures at the opposite site (Figure 2-6). This effect was clearly visible in experiments by Nahum et al., where pressurized cadaver heads were subjected to frontal impact [29][1].

The relative movement within the brain also could cause injuries. The brain is inhomogeneous and it has several different parts with different material properties.

Figure 2-5: Rotation of the skull towards the brain [26]
Acceleration of the head causes different loads in different parts of the brain and as a result relative movement and deformation occurs within the brain. The brain contains several membranes (e.g., the falx and tentorium), that are much stiffer than the surrounding neurological tissues and these hinder the relative movement [1]. This leads to considerable deformations in the brain at the contact interfaces between the brain and the membranes and is thought to be the main cause of contusions [1][9]. It should be noted that the modeling of the interface of the skull and brain is very important. If the brain is fixed rigidly to the skull, the maximum deformation in a rotational acceleration test occurs remote from the skull-brain interface (Figure 2-7-a). If the brain is allowed to slip relative to the skull, the maximum deformation moves more towards the skull (Figure 2-7-b) [1]. The human head lies between these extremes [27][1].
2.3. Head Injury Criteria

Many studies have been focused on assessing the injury mechanisms causing inertial head injury during impact in order to establish associated tolerance levels of the human head. The development of injury criteria has been a very important challenge, in order to be able to determine the risk of a head injury, and to investigate the effectiveness of protection measures and facilities. Injury criteria for inertial induced head injuries can roughly be divided into three categories: injury criteria based on translational accelerations of the head's center of gravity; injury criteria based on translational and rotational accelerations of the head's center of gravity; and injury criteria based on
stresses and strains inside the brain. Each category is discussed in the following sections [30].

2.3.1. Translational Acceleration Based Injury Criteria

2.3.1.1. Wayne State Tolerance Curve

The Wayne State Tolerance Curve (WSTC) is considered to be one of the very first and basic criteria in head injury studies [30]. This curve evolved from the work of Lissner et al., Gurdjian et al. and Patrick et al., and gives the tolerable average acceleration magnitude in the anterior-posterior (A-P) direction as a function of the acceleration duration [30][31][32][33]. The curve is shown in Figure 2-8, and it is still an acceptable basis for most popular head injury criteria.
Slight cerebral concussion without any permanent effects was considered to be within human tolerance [30]. Only translational accelerations were used in the development of the curve, Figure 2-8. In region I, data was obtained from different experiments with post mortem human subjects with linear skull fracture as injury criterion (assumed to be highly associated with brain concussion). In region II, animals were used and intracranial pressure was measured and compared. In region III, volunteers were used with loss of consciousness as injury criterion [30]. Except for long duration accelerations (>20ms), the WST-curve has never been validated for living human beings [30].
2.3.1.2. Severity Index

Gadd argued that the response of the head to an impact cannot be sufficiently determined by the average acceleration nor by the peak acceleration observed in an impact and the results are not accurate [34]. According to Gadd, the resulting injury potential is highly dependent upon the acceleration pulse, and therefore, pulses with the same average acceleration but different shapes can have very different effects [30]. In order to consider the effect of both the acceleration pulse shape and its duration, he suggested the integration of the acceleration signal over its duration. Gadd further maintained that injury potential was a non-linear function of acceleration magnitude [30]. Therefore, he suggested that an exponential weighting factor (greater than 1) be applied to the acceleration and that the result be integrated over the duration of the acceleration [30][34]. This introduced a new injury criterion, called the Severity Index or Gadd Severity Index:

\[
(G)SI = \int_{0}^{T} a(t)^{2.5} \, dt \quad (a(t) \text{ in } g's)
\]

[1]

The weighting factor 2.5 only applies to the head and is primarily based on a straight-line approximation of the WSTC plotted on log-log paper between 2.5 and 50ms [34]. Gadd proposed a threshold (tolerance level) for concussion for frontal impact of 1000, which agreed with the WST-curve, the Eiband tolerance curve [35] and accident simulation data by Swearingen [36][30]. The (G)SI received significant scientific criticism later, because it sometimes deviates considerably from the WST-curve[30]. The WST-curve is based on the average acceleration. According to Versace, a mathematical
approximation of this curve should also be a function of the average acceleration. Therefore, he suggested the following injury criterion [30][37]:

\[ (V)SI = \left[ \frac{\int_0^T a(t) dt}{T} \right]^{2.5} T \]  \hspace{1cm} [2]

where \( a(t) \) is the acceleration of the head and \( T \) is the duration of impact.

2.3.1.3. Head Injury Criterion, HIC

\[ HIC = \left\{ \left[ \frac{\int_{t_1}^{t_2} a(t) dt}{t_2 - t_1} \right]^{2.5} (t_2 - t_1) \right\}_{\text{max}} \]  \hspace{1cm} [3]

Based on Versace's criticism of the (G)SI, it was suggested that the SI should be replaced with a slightly modified injury criterion, called the Head Injury Criterion (HIC): with \( a(t) \) the resultant head acceleration in g's (measured at the head's center of gravity and \( g=9.81 \text{ m/s}^2 \)) and \( t_1 \) and \( t_2 \) any two points in time, during any interval in the impact, that maximize HIC [30]. The value of 1000 is suggested for the HIC as tolerance level for concussion in frontal impact, similar to SI criterion. For practical reasons, the maximum time interval \( (t_1-t_2) \) which is considered to give appropriate HIC values was set to 0.036s, or for some other cases 0.015 s [38]. HIC has been shown to be a reasonable discriminator between severe and less severe injury [39]. It also correlates with the risk of cranial fracture in cadavers after impact [40]. However, HIC does not correlate well with injury severity for impact in various impact directions [41]. An important limitation of the HIC is that head rotational acceleration is not taken into
account; although rotation is debated to be the primary cause for various types of traumatic brain injury, in particular acute subdural haematoma and diffuse brain injury [42].

2.3.1.4. Maximum Resultant Head Acceleration

Maximum resultant head acceleration ($a_{max}$) is often used based on its simplicity in analysis. The thresholds for $a_{max}$ depends on its application because of the time dependent nature of the resultant acceleration with respect to head injury [30]. To account for this time dependency, this criterion can be replaced/supplemented by a value for the resultant acceleration that should not be exceeded longer than a certain time interval [30].

2.3.2. Combined Rotational and Translational Acceleration Based Injury Criteria

The previously discussed injury criteria have nothing to do with rotation of the head, and they are concerned with translational head impact response. However, the biomechanical response of the head during impact also includes rotational motion which is believed to cause injury [42], in particular acute subdural haematoma and diffuse brain injury [30]. A summary of various tolerances of the human brain to rotational acceleration (and rotational velocity) is given in Table 2.2 [13].
**Generalized Acceleration Model for Brain Injury Threshold, GAMBIT**

Newman attempted to combine translational and rotational head acceleration response into one injury criterion [43][30]. Considering these accelerations as the cause for stresses generated in the brain and resulting in brain injury, he proposed the Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) [30]. The general GAMBIT equation is:

\[
G(t) = \left[ \left( \frac{a(t)}{a(c)} \right)^n + \left( \frac{\ddot{a}(t)}{\ddot{a}(c)} \right)^m \right]^{1/s}
\]

with \( a(t) \) and \( \ddot{a}(t) \) the instantaneous values of translational and rotational acceleration respectively;

\( n, m \) and \( s \) are empirical constants selected to fit available data; and

\( a_c \) and \( \ddot{a}_c \) are the critical values of the accelerations (tolerances) [30].

On the assumption that the tolerances derived from experiments with only translational or only rotational head motion are also valid for combined head response, and on the assumption that translational and rotational acceleration equally contribute to head injury, Newman simplified this equation to become [30][43]:

\[
G = \frac{a(t)}{250} + \frac{\ddot{a}(t)}{10000} \leq 1
\]

with \( a_m \) [g] and \( \ddot{a}_m \) [rad/s²] being the mean values of linear and angular acceleration respectively;

The GAMBIT predicts injury when \( G > 1 \) and no injury when \( G \leq 1 \). However, the GAMBIT was never extensively validated as an injury criterion [30].
Table 2.2: Human brain tolerance to rotational acceleration (and rotational velocity) concerning sagittal [13]

<table>
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<tr>
<th>injury</th>
<th>tolerance</th>
<th>type of research</th>
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| cerebral concussion [Ommaya et al., 1967] | 50% probability:  
- for \( t < 20 \text{[ms]} \): \( \ddot{a} = 1800 \text{[rad/s}^2\text{]} \)  
- for \( t \geq 20 \text{[ms]} \): \( \ddot{a} = 30 \text{[rad/s]} \) | Extrapolated from animal experiments (speculative data) |
| bridging vein rupture [Löwenhielm, 1975] | \( \ddot{a} = 4500 \text{[rad/s}^2\text{]} \) and/or  
- \( \ddot{a} = 70 \text{[rad/s]} \) | Mathematical model in conjunction with experiments on bridging vein rupture |
| brain surface shearing [Advani et al., 1982] | \( 2000 < \dot{a} < 3000 \text{[rad/s}^2\text{]} \) | Mathematical model |
| brain (general) [Ommaya, 1984] | \( \dot{a} < 30 \text{[rad/s]} \): safe; \( \ddot{a} < 4500 \text{[rad/s}^2\text{]} \)  
- AIS 5: \( \ddot{a} > 4500 \text{[rad/s}^2\text{]} \)  
- AIS 2: \( \ddot{a} = 1700 \text{[rad/s}^2\text{]} \)  
- AIS 3: \( \ddot{a} = 3000 \text{[rad/s}^2\text{]} \)  
- AIS 4: \( \ddot{a} = 3900 \text{[rad/s}^2\text{]} \)  
- AIS 6: \( \ddot{a} = 4500 \text{[rad/s}^2\text{]} \) | Review of experimental and mathematical studies |

---

**Head Injury Power**

Newman et al. reasoned that the rate of change of translational and rotational kinetic energy, i.e. power, could be a viable biomechanical function for the assessment of head injury [44][30]. An empirical expression for this *Head Impact Power* (HIP), relating a measure of power to head injury, would be of the form [30]:

\[
HIP = A a_x \int a_x \, dt + B a_y \int a_y \, dt + C a_z \int a_z \, dt + \eta y a_x \int a_x \, dt \\
+ \beta a_y \int a_y \, dt + \chi a_z \int a_z \, dt
\]  

\[6\]
Each term in this relationship shows the change in kinetic energy for each degree of freedom of the system. The coefficients A, B, C, γ, β and χ denote the injury sensitivity for each of the six degrees of freedom of the head [44]. The development of this function is described in Newman et al. [44][41][30]. Based on experimental data from analyzed collisions of American Football players during games, Newman et al. validated the HIP [41]. Using TV-images and the resulting injuries, classified by qualified physicians, they reconstructed the kinematics of the head. A 50% probability of concussion at a maximum Head Impact Power (HIPmax) of 12.8 kW was found [30]. Because the experimental data is not yet available for more severe brain injuries, The HIPmax has not been validated for such cases. From their results, the authors concluded that HIPmax better correlates with mild traumatic brain injury than HIC [30]. The authors give three advantages of HIPmax over HIC to backup this conclusion: besides translational accelerations, HIPmax can also incorporate directional sensitivity, sensitivity for rotational accelerations and sensitivity for angular and translational velocities [30][41].

2.3.3. Stress and Strain Based Injury Criteria

Brain injury is reported to correlate with stress, strain and strain rate [45]. However it is difficult to measure the strains and strain rates inside the brain during impact. Advancement in technologies and computational methods has led to more accurate numerical models of the human head [30]. These models bring a detailed injury
assessment closer to reality, since they enable stresses and strains to be examined [30].

Bandak developed three measures representing the general types of brain injuries experienced in traffic accidents [46][47][30]:

- Cumulative Strain Damage Measure;
- Dilatation Damage Measure; and
- Relative Motion Damage Measure.

Stresses and strains used to compute the above injury parameters are calculated from finite elements simulations, using a finite element model of the skull and intracranial contents [30].

**Cumulative Strain Damage Measure**

A mechanical measure which has been shown to be very useful in the evaluation of deformation related brain injuries resulting from head impact is the Cumulative Strain Damage Measure (CSDM). It was developed to evaluate the strain-related damage within the brain. The CSDM is based on the hypothesis that diffuse axonal injury (DAI) is associated with the cumulative volume fraction of the brain matter experiencing tensile strains over a critical level [30]. At each time increment, the volume of all the elements that have experienced a principal strain above prescribed threshold values is calculated [30]. The affected brain volume monotonically increases in time during conditions where the brain is undergoing tensile stretching deformations and remains constant for all other conditions (i.e. compression, unloading, etc.) [30]. Bandak et al. found that a CSDM level
5 corresponds to mild DAI and a CSDM level of 22 corresponds to moderate DAI severity [49].

**Dilatation Damage Measure**

The second mechanical measure, Dilatation Damage Measure (DDM), provides severity determination for brain injury caused by large dilatational stresses. Dilatation Damage Measure (DDM) monitors the cumulative volume fraction of the brain experiencing specified negative pressure levels [30]. However, no direct observational evidence has been reported on the relationship between pressure mechanisms and the production of axonal, vascular or other soft tissue injury [30]. Similar to the CSDM calculation, at each time step, the volume of all the elements experiencing a negative pressure level exceeding prescribed threshold values is calculated [30]. Bandak et al. suggested a DDM value of 5 at a threshold level of 14.7 psi as an injury threshold, but also reported that further research was necessary [30][48].

**Relative Motion Damage Measure**

Relative Motion Damage Measure (RMDM) is the third mechanical measure which is used in evaluation of injury related to brain movements relative to the interior surface of the cranium. This measure monitors the tangential motion of the brain surface
resulting from combined rotational and translational accelerations of the head [30]. Such motions are a suspected cause of subdural haematoma as associated with large-stretch ruptures of the parasagittal bridging veins [30]. The rupture tolerance levels of bridging veins are a combined function of both strain and strain rate [30]. The explicit representation of the bridging veins is not required for the RMDM. However it monitors the relative displacement of several node pairs, each of which basically shows a bridging vein tethered between the skull and brain, near the parasagittal sinus. The measure accounts for the large-stretch modes of rupture while leaving open the possibility of using other micro or macro rupture-modes associated with more complex vascular tethering states [30].
Chapter Three

Helmet Design

As it was discussed in Chapter 1, the main purpose of this study is to prepare a design for football helmets in order to reduce the head injury probability. To this end, it was decided to focus on the liner part which is the main absorber part of the helmet and is responsible for dissipating energy as much as possible to reduce the head concussion possibility. Thus after investigating the energy absorption process, different types of shock absorbers and also how a proper material and design are selected for the liner in current and popular helmets. It was found that, up to now, materials with honeycomb structures, polymeric foams and cellular structures are used in popular helmets as the energy absorbing systems. As a result efforts were made to find the conventional design methods for popular helmets with respect to foam technology and also to investigate the corresponding testing methods and evaluation parameters for this technology so that the resulting performance of existing and proposed helmets could be compared and evaluated.

On the other hand, there are some unconventional shock absorber types with the capability of being used in helmets with respect to their energy absorption characteristic
and correspondingly some different testing and evaluation methods for each of them are known that should be taken into account. Thus in this study, an attempt was made to investigate the capability of other shock absorbers such as hydraulic shock absorbers, one-shot polymer shock absorbers, shape memory materials and superelastic materials. It should be noted that for each of these materials there are different methods to evaluate the energy absorption capacity. These methods include different compressive testing methods, impact testing methods and stiffness comparison methods energy evaluation methods.

After investigating all the evaluation methods, for the both conventional and unconventional technologies, it was decided to use the energy method in order to initially evaluate the energy absorption capacity of the shock absorber candidates, which is based on the energy per volume absorbed by the material and corresponds directly with the area under the stress-strain/force-displacement curve. To this end, ASTM standard compression testing, discussed in Chapter 1, has been used and the initial evaluation for the various types of shock absorbers was done based on the energy per volume criterion or specific damping energy value, $D$, using the following energy equation:

$$D = \int F \, dx$$  \[7\]

Where the integral is based on force-deflection diagram of the material and is essentially the area under the force-displacement curve of the material, it is the shaded area in Figure 3-1.
Using this method, it is possible to simply evaluate not only the conventional foam materials but also other types of shock absorbers, and the general comparative analysis can be done simply over all the candidates. After initial evaluation of the liner candidate materials based on the ASTM standard testing method, using NOCSAE standard, the drop testing method can be utilized to properly investigate the effectiveness the material as a new technology inside the helmet and also to validate the results.

After comparing all the absorber candidates, based on the best candidates, new designs were prepared for the helmet liner. To this end, design methods were first investigated for conventional foam technology, discussed in the following section, and following that an attempt was made to prepare other new designs using the other selected good candidates in the helmet.
3.1. Conventional helmet design

The possibility of controlling the stress-strain behavior by an appropriate selection of matrix material, cellular geometry and relative density makes foams a very good material candidate for such applications [50]. In order to select a proper candidate among all types of foams for the helmet and to come up with a good design, generally two conventional methods are used. They are called energy approach A and B and are discussed in the following sections.

3.1.1. Energy approach (A):

This approach is based on the cushioning curve principle and can be divided into two parts: The first, the energy density comparison method, used for the foam selection and then the analytical helmet design method [51].

3.2.1.1. The cushioning curves principle:

The cushioning curve principle is used in the packaging design for applications such as the transportation of goods. These cushioning curves are prepared as shown in Figure 3-2. The curves are based on a rectangular mass, m, which is falling from a distance, h, onto a foam block with the thickness of t. As a result the recorded peak
acceleration G of falling mass, in g’s, is graphed as a function of static stress, $\sigma_s$, which is the compressive stress experienced by the foam as the rectangular object is resting on it (mg/A). $A$ is the contact area between the foam and the object in this case. Each curve is representative of G values for different static stresses for specific thicknesses, $t$, while each graph is for a particular dropping height, $h$.

If the stress-strain curve for a specific foam and particular drop height is given as shown in Figure 3-3, the cushioning curve can be prepared based on following procedure:

$$U = \frac{mgh}{At} = \int_0^\sigma \sigma \, d\varepsilon$$  \hspace{1cm} [8]

Where $U$ is the energy density of the foam and $\sigma_m$ is maximum stress [51]. The static stress $\sigma_s$ is given as a function of $U$ as:

$$\sigma_s = \frac{t}{h} \ U (\sigma_m)$$  \hspace{1cm} [9]

Figure 3-2: Experimental cushion curves for LDPE foam density 70 kg/m3 and different thickness, $t$, for drop height of 0.6m [52]
The maximum acceleration of the impact happens when the compressive stress is maximum, $\sigma_m$. Since the unit acceleration, when the static stress $\sigma_s$ is applied via the foam to the mass $m$ that is 1g, the ratio of acceleration gives [51].

$$G = \frac{\sigma_m}{\sigma_s} = \frac{h}{t} \frac{\sigma_m}{U(\sigma_m)}$$  \[10\]

Therefore, calculating both parameters of the cushioning curve (G and $\sigma_s$) is possible from the energy density function. As a summery, for a given application (acceleration limit), a given drop height and a given stress-strain curve, the required foam thickness can be determined either from the cushioning curve or by using the previous equation [51]. In fact these may not be used as a final design tool for the helmets; rather,
it could be used as a comparative parameter for the efficacy of different foams or even different designs.

It should be noted that the energy density can be used simply as a comparative parameter between the various types of foams. What is necessary to be done in this manner is that the limiting stress that could be sustained and the stress-strain curve for the foam candidates to be compared. The limiting stress can be simply determined from the allowable limit of the head acceleration.

3.2.1.2. The Analytical Design:

In a study, Mills [53] generated a simple mathematic approach where he simplified the helmet complications to its simple elements at which the main parameter was the energy absorption capacity of the foam. His work led to estimating the effective contact area between the helmet and the striker. In reality not all the helmet surface area is effective during an impact by a striker [51]. This contact area was actually a basis for a simple approach in helmet design.

Geometry of the helmet-object impact:

Based on Mills’ work, it is assumed that the section where the impact forces act is spherical, while neither the human skull nor the outer surface of the helmet is exactly spherical. It should be noted that he also assumed that the stiffness of the shell is
negligible. This means that the helmet shell does not participate in the energy absorption during impact. However the shell can absorb up to 30% of total energy of the impact [54][51]. This could result in additional energy being absorbed but the procedure can be used as a comparative methodology for different foam material performances with the error of ignoring the helmet being cancelled out. In this method, the foam is assumed to crush at a constant yield stress $\sigma_y$ when compressed [51]. When $x$ is the maximum deflection of the foam on the outer surface of the helmet, the contact area is a circle of radius $a$ [51]. Using the Pythagoras’s theorem, the following result can be found (Figure 3-4):

$$R^2 = (R - x)^2 + a^2$$  \[11\]

If the deflection is much less than radius of the outer surface of the helmet $R$, the term $x^2$ can be ignored. So the contact area $A$ can be shown as:

$$A = 2\pi R x$$  \[12\]

The transmitted force into the foam is also defined by:

$$F = 2\pi R \sigma_y x$$  \[13\]

The above relationship can be rearranged as:

$$k = \frac{F}{x}$$  \[14\]

$$k = 2\pi R \sigma_y$$  \[15\]
The design procedure:

Using the force-deflection relationship in previous section, the design steps can be summarized as following [51].

1- According to the helmet mass and aerodynamic aspects, the maximum allowable thickness for the foam can be selected.

2- The foam yield stress and density is selected to give a loading diagram, utilizing the radius of the helmet at the impact site, as shown in Figure 3-5.

3- After calculating the energy under the loading curve, it should be checked if it exceeds the allowable value in test standards. If the allowable value is exceeded, the thickness of the foam in step1 should be increased.
3.1.2. Energy approach (B):

This approach is more foam selection rather than designing by means of relationships and equations. It is divided into two procedures. The first is called the overall picture, which is basically selecting the proper candidate from all available foams for a particular application. The second one, which is called refined selection, is a more refined solution. The refined selection procedure deals with more optimized selection of material, density and thickness of the foam for particular application.
The Overall picture:

In this method the energy per unit volume of different foams is used. According to this method, using the Figure 3-6, the best choice for foam is selected which should be able to absorb the most energy per unit volume, while limiting the load on the human head less than damaging level. Figure 3-6 shows the relation between the foam compressive strength at 25% strain against the densification strain $\varepsilon_D$ for a constant energy per unit volume as shown by the straight dotted lines [55][51].

Figure 3-6: Densification strain $\varepsilon_D$ against the plateau stress $\sigma_c$ (the compressive stress at 25% strain) for commercially available foams. The contours show energy absorption $\varepsilon_D \sigma_c(25\%)$ per unit volume. [55]
The Refined Selection Procedure:

Since this method is used for packaging design, when using it for helmet design, it is assumed the head is a delicate good to be packaged. This method is again based on the energy absorption graphs (Figure 3-7 and Figure 3-8) [55]. Given the material and density, the diagram identifies the normalized peak stress and energy absorbed per unit volume at which it is best used [55][51]. For more details please refer to Gibson and Ashby [55].
Figure 3-7: Energy absorption curves for two electrometric foams constructed by measuring the area under the stress-strain curves. The envelope liner relates $W$, $\sigma_p$ and relative density for the optimum choice of foam at the strain rate $10^{-2}s^{-1}$ [55]
Figure 3-8: Energy absorption curves for two plastic foams constructed by measuring the area under the stress-strain curves. The envelope liner relates $W$, $\sigma_p$, and relative density for the optimum choice of foam at the strain rate $10^{-2}s^{-1}$ [55]

3.2. Helmet design and modifications

As it was discussed initially, using the ASTM standard testing method and energy per volume criterion or specific damping energy, the effectiveness of various types of shock absorbers has been evaluated. Based on the best candidates of different shock absorbers and also regarding their effectiveness in energy absorption applications, helmet
designs have been prepared. The designs have been prepared either by modifying the conventional liner part of existing helmets or by completely redesigning the liner as a totally different part. The candidate shock absorbers are Riddell foam, one-shot polymer shock absorber and shape memory alloy materials.

3.2.1. Design Based on Shape Memory Wire Mesh Structure

Shape Memory Alloys

One of the candidates capable of being used in the liner of the helmet as the energy absorber is SMAs (Shape Memory Alloys). Shape memory alloys are a specific type of smart materials which can recover their strain upon heating based on specific cyclic phase transformations. Increasing the temperature of the material, results in recovery of the introduced strain into the material even when the applied force is relatively large which is known as shape memory effect. This property makes the energy density high in shape memory alloys. Also under specific conditions the material undergoes a hysteresis reversible transformation which enables the material to dissipate energy for energy absorption applications. These properties make shape memory alloys a good candidate for a variety of applications from aerospace and automotive to biomedical applications. However, low frequency response of the material is a disadvantage which restricts its applications. A brief overview of the behavior and characteristics of shape memory alloys is discussed in the following sections:
Phase diagram and transformation

The phase diagram is a schematic representation of the transformation regions for shape memory alloys [59]. The lines in the phase diagram show the phase boundaries that separate the two solid phases of the shape memory alloy. Usually, the phase diagram for shape memory alloys has temperature along the abscissa axis and stress is shown along the ordinate axis. SMAs have two solid states in typical operating temperatures. The high temperature phase which is known as austenite (A) and the low temperature phase which is known as martensite (M). The mechanism in which the crystalline structure of the austenite phase transforms to martensite phase is called lattice distortion and the transformation is called martensitic transformation. The martensite crystal can also be formed as twinned or detwinned in its reoriented forms. The reversible transformation of the material from austenite, which is known as the parent phase to martensite, makes the special thermo-mechanical behavior of shape memory alloys. The well-known phase diagram of shape memory alloy materials is shown in Figure 3-9. The pseudoelasticity and shape memory effect as the special thermo-mechanical properties of shape memory alloys are described in the following sections.
Shape memory effect

Shape memory effect is the ability of the alloy to recover a certain amount of strain upon heating. This phenomenon happens when the material is loaded such that the structure reaches the detwinned martensite phase and then unloaded while the temperature is below the austenite final temperature (As). Heating the material at this stage will lead to strain recovery of the material and the material will regain its original shape. This phenomenon can be better understood in the combined stress-strain
temperature diagram as shown in Figure 3-10. This diagram shows the typical thermo-mechanical behavior of shape memory alloys.

![Figure 3-10: Shape memory effect path of a shape memory alloy material](image)

Starting from point A, the material is initially in the austenite phase. Cooling down the alloy to a temperature below its martensite final temperature (Mf) will form the twinned martensite crystals. At this point, loading the alloy at the same temperature will lead the crystals to transform to the detwinned martensite phase at point C. This path is nonlinear because of the transformation phenomenon. Unloading the applied stress of the material at the same temperature will be linear to point D and maintains the detwinned martensite phase. At this point, the material keeps a residual strain. By heating the alloy, when the temperature passes the austenite start temperature (As) at point E, the transformation from detwinned martensite to Austenite crystalline phase starts. This transformation recovers the residual strain of the alloy which will be fully recovered at
point F, where the alloy passes the Austenite final temperature. Increasing temperature more, the material will reach the starting phase and temperature at point A.

This shape memory behavior is used for the actuation applications. The actuation behavior of shape memory alloys is shown in the Figure 3-11. In this figure a constant dead weight is applied to one end of a SMA spring. From left to right, the SMA spring actuator starts from the detwinned martensite phase in which the material is stretched and has the maximum displacement. Upon heating, the SMA temperature is increased which causes the gradual transformation to austenite while passing the martensite-to-austenite transformation region as can be seen in Figure 3-12. This transformation leads to recovery of the strain in the material, which can be seen as the recovered displacement in the spring. Cooling down the spring will cause the temperature to pass the austenite-to-martensite band which leads the forward transformation and generating transformation strain in the material and making the material crystal to convert to the detwinned martensite phase.
Figure 3-11: SMA spring in actuation [60]

Figure 3-12: Actuation path in phase diagram [60]
**Pseudoelasticity**

The pseudoelasticity or superelastic behavior is associated with the stress-induced transformation of the shape memory alloys, which generates strain in the material and recovery of this strain upon unloading at temperatures above Af (Figure 3-13). The superelastic thermo-mechanical behavior starts from temperatures above austenite final temperature (Af) where stable austenite exists. The alloy is loaded to make the detwinned martensite crystal form. During the transformation from austenite to martensite the forward transformation happens and the transformation strain is generated. Upon unloading, the generated strain upon forward transformation is fully recovered in the backward transformation and the original form is achieved. As shown in Figure 3-14, the pseudoelasticity effect involves recovering large deformations. The stress-strain cycle of such behavior is shown in Figure 3-15.
Figure 3-13: Pseudoelasticity path in phase diagram [60]

Figure 3-14: Large deformation and recovery of strain in SMA wire above Af [60]
According to all that has been discussed about the behavior of SMAs, in particular, there are two remarkable features of SMAs: noticeable hysteresis, or in other words, good energy absorption characteristic and also their capability of having different stiffness after transformations, makes them one of the good candidates of this study for designing the liner of the helmet. Therefore, an attempt was made to prepare a new design based on SMA materials for the helmet. The design attempted to take advantage of the good energy absorption capability of the SMA wires but also to give the helmet an extra stiffness adjustment feature based upon the stiffness changes in the SMA wire after transformations when subjected to impact forces. The stiffness adjustment feature would
have an effective role specifically when the helmet is subjected to a severe hit specifically in the cases that the impact force is huge and it may result in the foam compression beyond the bottoming out point (i.e. compression of the foam up to about 80-90% of its original thickness).

It should be noted that existing helmets and their conventional foam technology have a foam hardening problem that makes the foam much stiffer than usual when a severe impact force is applied. It happens when the foam reaches its bottoming out point during compression. At this point, the hardening of the foam results in increasing the force level on the head, or in other words, increasing the acceleration level of the head. Bottoming-out likely increases the probability of concussion for the player.

The new design was prepared based on the combination of the SMA wires and conventional foams in a series arrangement. It attempted to modify the current foam technology and come up with a new design based on using SMA wires in a mesh structure, as shown in Figure 3-17, under the foam part in order to not only use the energy absorption characteristics of SMA but also have the stiffness adjustment (or liner softening) feature for the liner package, utilizing the stiffness change of the SMA wires during the impact. In this design, the impact to the helmet initially leads to compression of the foam piece, and it causes the foam to expand a little bit in other directions. The expansion of the foam, resulting from its compression, in turn results in expansion of the SMA wire mesh structure thus, producing an increasing tension in the SMA wires. As a result, when a severe impact force enters to the system, the tensions in the wires can reach the transformation points and the stiffness of the wires can change accordingly (Figure 3-18). In this case, not only has the energy absorption capacity of the liner
package been improved, but also the stiffness adjustment feature has been added to the system that can be effective in case a severe impact leads to the bottoming out point of the foam. It should be noted that the concept softening process in the design is based on using two shock absorbers in a series arrangement in which the total stiffness of the system would be lower than the stiffness of each individual. It follows the concept that, for instance, if multiple (n) shock absorbers or springs with the same stiffness (k) are used in series arrangement, the total stiffness of the system will be as shown in Figure 3-16 [57]. If a mesh structure with changing stiffness is used in a series arrangement with the foam, the total stiffness is smaller and it could be more helpful in producing a better head protection during the severe impact, when the foam is getting close to its bottoming out point.

\[
\frac{1}{K_t} = \frac{1}{K_1} + \frac{1}{K_2} + \frac{1}{K_3}
\]

Figure 96: Total stiffness of a system including multiple shock absorbers in series arrangement [57]
Figure 10: The schematic view of the design based on shape memory wire mesh structure

Figure 11: Schematic view of how the design based on SMA wire mesh structure works
3.2.2. Design based on new polymer shock absorber:

While searching for new candidates for the liner in the helmet, a new one-shot polymer based shock absorber (Figure 3-19) was found with a behavior that is very close to what is known as an ideal shock absorber (Figure 3-20). It should be noted that the ideal shock absorber is an absorber that keeps the level of force constant during impact, or in the other words, the ideal shock absorber has a Force/Strain vs. Displacement/Strain graph, as shown in Figure 3-21. In this case, the absorbed energy per volume of the shock absorber is maximum and there is not any rebound energy to the head. Based upon the highly desirable force-displacement characteristic of the one-shot shock absorber an attempt was made to create a design based on using the one-shots.

![One-shot polymer shock absorber](image)

Figure 12: One-shot polymer shock absorber [56]
Figure 13: The behavior of different sizes of one-shot shock absorbers (Sizes 1-7) [56]
Figure 14: A sample for the force/stress vs. displacement/strain graph of an ideal shock absorber

It was first decided to use one-shots in the liner of the helmet, but based on the sizes of available one-shots and knowing that the spacing limits for the liner of the helmet is a critical design parameter, this design obviously was not suited for the desired football application, because the larger one-shot sizes should be used in order to absorb the impact energy in the football helmet and the larger sizes could not fit in the available room inside the helmet. Another design was based on using a combination of one-shots and existing foams in a proper arrangement so that it would be well suited and work properly. Two different designs have been prepared: one where the foam and one-shots are parallel to each other and the other one is where the foam and one-shots are in a series arrangement Figure 3-22 and Figure 3-23.

Comparing the results and also considering the important design requirements such as packaging limits, the parallel arrangement has been selected since the series
arrangement did not perfectly meet the requirements, specifically the one regarding the spacing and the size of the liner for the helmet.

For the parallel arrangement design, the one-shot size 1 (the smallest one) has been selected based on its force level on force-displacement graph (Figure 3-20) and its dimensions. It was the only one-shot that could be used together with the foam in the helmet considering the spacing limits of the design. The stroke of this type of one-shot is about 6mm. The design has been prepared so that first only the foam is compressed and after the foam is compressed to the point that the foam only had more 6 mm to its predefined bottoming-out point, the one-shot shock absorber begins to be compressed. To this end, based on the predetermined bottoming out point of each foam part located on different sites of the helmet, filler backseat parts shim as shown in Figure 4-1 were used such that the one-shot compression occurred only during the last 6 mm of the foam right before the bottoming out point. In the last 6 mm both the foam and one-shots were involved in compression.

As an example, considering the foam part on the forehead of the helmet with 28 mm thickness, it was decided to take 80% of the total thickness, 22.4 mm, as the bottoming-out point and the shim has been prepared such that the one-shot is involved on the last 6 mm stroke of the foam which is basically starting from 16.4 mm of the foam compression (Figure 4-2).

After modifying the helmet according to the design, the modified helmet was subjected to drop testing, to verify the design. The data from the drop testing was used for the final assessment of the effectiveness of the design in reducing the head concussions has been made. The testing and results are discussed in the next chapter.
Figure 15: Design based on combination of foam and one-shot is parallel arrangement.

Figure 16: Design based on combination of foam and one-shot is series arrangement.
Figure 4-1: A Schematic of a sample for the Backseats for the helmet used in design based on combination of the foam and one-shot

Figure 17: The backseat design concept for the combination of the foam and one-shot for forehead site of the helmet
Chapter Four

Analysis and Testing

In this study, different steps, including two different testing methods, have been utilized in order to determine the capability of the candidates for the helmet liner part. These two testing methods were Material Testing, including quasi-static compression testing based on the ASTM standard method, and the Drop Testing Method, which has been done based on the NOCSAE standard. Where the behavior or the capability of each candidate has been initially determined based on the ASTM standard and after comparing the results, the designs for the liner of the helmet were prepared based upon the experimentally determined results then the NOCSAE standard drop testing method was used to experimentally determine the modified helmet performance.

4.1. Material Testing

The candidates, selected and tested in the first round of tests, were different types of existing foams in market, different types of shock absorbers such as the polymer based one-shot shock absorber and hydraulic shock absorbers. It should be noted that some
existing foam technologies in the recently-used helmets and also some unconventional technologies used in the helmets such as air chambers have also been tested for result comparison. In order to perform the compression testing based on ASTM standard, two experimental sets have been used: a BOSE machine for tests with smaller displacement stroke conditions and the INSTRON compression testing set for larger displacement stroke condition tests.

### 4.1.1. Foam Testing

The compression testing has been performed on a range of different types of the existing foams in market including open cell and closed cell foams and with different densities and other specific characteristics. The foams were compressed up to a point close to their bottoming out point. The general comparison has been made on the results for compression with 1mm/s rate of displacement and based on Specific Damping Energy criterion. In addition, It should be noted that the combination of the different foams has been tested (i.e. the series and parallel arrangements of the foams). The results of the tests are presented in Chapter 5.
4.1.2. Shock Absorber Testing

The one-shot shock absorber, new air chamber shock absorber and some small hydraulic shock absorbers were tested under the same identical testing conditions and with the same experimental set under different rates of displacement. However, in these cases, the comparison has been made for 1mm/s rate of displacement and based on the Specific Damping Energy criterion.

The results show that the best candidate is the one-shot shock absorber, which was used in the design that has been explained in Chapter 3.

4.2. Drop Testing

The last round of testing in effectiveness evaluation of the liner designs, Drop Testing, has been performed according to NOCSAE standard as discussed in previous chapters. First the helmet has been tested without any modifications and it was tested again after being modified, using one-shot shock absorbers. The modified helmet was subjected to the same droppings and the result verification and effectiveness comparison has been made afterward. The parameter for the comparison analysis was the Severity Index, SI, exactly as has been defined in the NOCSAE standard.
Chapter Five

Results

5.1. Material Testing

As it was discussed, the preliminary compression testing on materials and candidate shock absorbers has been performed and the results are presented in the following paragraph.

The first shock absorber tested was a small type of hydraulic shock absorber. Figure 5-1 shows the results of the compression testing for the hydraulic shock absorber for different rates of displacement up to its maximum compression stroke.
Figure 18: The compression testing on the hydraulic shock absorber for two different rates of displacement

The second shock absorber tested was a polymer based one-shot shock absorber, and it was tested based on ASTM compression testing standard. There have been various sizes for the one-shot shock absorbers and the smallest one was tested because its small dimensions were more compatible with use inside the helmet. Figure 5-2 shows the results of the compression testing for one-shot shock absorber for 1mm/s rate of displacement.
The third shock absorber was the air chamber shock absorber which is used in the Xenith helmet. Figure 5-3 shows the results of compression testing on the air chamber shock absorber and also the comparison between the results of the one-shot shock absorber and the air chamber shock absorber.
Figure 20: The compression testing results for air chamber shock absorber and the one-shot shock absorber for 1mm/s rate of displacement

The forth shock absorbing technology was the existing helmet foam technology. This is a soft and dense foam used in Riddell helmets. Figure 5-4 displays the results of compression testing with a 1mm/s rate of displacement for the Riddell foams and one-shot in order to have a sense of comparison between the results.
Figure 21: The compression testing results for a soft (Red Line) and a dense (Blue Line) foam from Riddell helmets and the one-shot shock absorber for 1mm/s rate of displacement

The last candidate shock absorber tested were different types of available foams in market with different characteristics densities and characteristics. The results have been separated into three different graphs. The separation was done randomly and just for easier comparison analysis, Figure 5-5 and Figure 5-6 and Figure 5-7.

For more convenient comparison analysis of the results of the compression testing on all the foam types, the foams with noticeable hysteresis, good specific damping energy value, have been graphed separately in Figure 5-8, together with the one-shot shock absorber results.

It should be noted that the combination of the shock absorbers have been also tested in series and parallel arrangements. Based on the results and the hysteresis of each
candidate, foam #12 has been selected and the compression testing has been done on the combination of foam #12 and the other foams with noticeable hysteresis in series arrangement. The results are shown in Figure 5-9. The other set of compression tests on the combination of the shock absorbers have been done on the series and parallel combination of the Riddell foam and the one-shot shock absorbers. The results for the series arrangement are shown in Figure 5-10. For the parallel arrangement the compression of the one-shot occurred during the last 6mm of the total displacement of the foam in order to get advantage of the maximum capacity of the combination. The results are shown in Figure 5-11.

Figure 22: The compression testing results on the available foams in market for 1mm/s rate of displacement (group I)
Figure 23: The compression testing results on the available foams in market for 1mm/s rate of displacement (group II)
Figure 24: The compression testing results on the available foams in market for 1mm/s rate of displacement (group III)
Figure 25: The compression testing results of the foams with noticeable hysteresis selected from the results of the available foams in market.
Figure 26: The results of compression testing on foam #12 in series arrangement to the other foams with noticeable hysteresis.

Figure 27: The results of compression testing on the combination of Riddell foam and one-shot shock absorber for 1mm/s rate of displacement (Series arrangement).
Figure 28: The results of compression testing on the combination of Riddell foam and one-shot shock absorber for 1mm/s rate of displacement (parallel arrangement)

Figure 29: The comparison on the compression testing results of the parallel and series arrangement of the Riddell foam and one-shot shock absorber for 1mm/s rate of displacement
Among all the devices and materials tested, the best candidates were found to be the one-shot shock absorber and the existing foams with respect to their Specific Damping Energy. The combination of these two best candidates was tested (Figure 5-10, Figure 5-11, Figure 5-12), and finally the parallel arrangement of Riddell foam and one-shot shock absorber was selected to be used in the helmet based on their force characteristics and their size. This arrangement lets the foam compress first and after approaching the bottoming out point of the foam the one-shot shock absorbers are compressed as shown in Figure 5-11.

5.2. Drop Testing

Based on the NOCSAE standard conditions, three rounds of drop testing have been done in order to evaluate the effectiveness of new designs for the helmet liner part in reducing the head injury possibility.

The first round of drop testing was on a Riddell helmet without any modification or any change in the structure in order to use the results in preparing a baseline for required calculations for new liner part designs. For each impact site of the helmet, using the head acceleration graphs and SI levels, an attempt was made to prepare the compression results of the helmet such that it was possible to come up with the force-compression/force-displacement results of the helmet for each site. According to force-displacement results of the helmet for each site and also considering the corresponding SI
levels, it was possible to calculate the specific damping energy level and use all this information to perform calculations for new designs or modifications.

As it was discussed, the combination of the foam and one-shot shock absorbers was selected as the new design for the helmet. The results of first round of drop testing were used in calculating the number of required one-shot shock absorbers for each site of the helmet. As an example, if the front side of the helmet is considered, the required number of one-shot shock absorbers was found using the following procedure. The acceleration graph reported from drop testing on the front of the head and helmet is shown in Figure 5-13. From the graph of acceleration of the head, using integration, the head and helmet velocity and displacement were found as shown in Figure 5-14, Figure 5-15. Thus, knowing the mass of the head and helmet the force-displacement graph can be determined from the graph of acceleration of the head and also the graph of displacement as shown in Figure 5-16.

It should be noted that the previous calculations were done for the case, in which the helmet failed in the drop testing condition regarding the NOCSAE standard, i.e. the SI values were greater than 1200. The calculated force-displacement and the acceleration-time curves were both for the dropping velocity of 6m/s for frontal impact, for which the SI level reached the 1200 limit. According to the results of force-displacement of the helmet, assuming that the head and all the helmet parts except the liner are rigid, it can be observed that the foam has been already bottomed-out since the bottoming-out point of the helmet is about 80% of 28mm or 22.4mm for the frontal of the helmet, and the displacement results proved that the compression has been high and the foam has already bottomed-out. The specific damping energy of the foam up to its bottoming-out point
have been found first and the rest of energy up to the failure point of the helmet was left to be absorbed by one-shot shock absorbers. In this case, for the frontal impact, for instance, four one-shot shock absorbers were needed to absorb the rest of impact energy. The required number of one-shots was determined from the following calculations:

The absorbed energy up to the bottoming-out point of the foam for the front of the helmet was determined as 59.1J and the total energy of impact calculated from the drop testing results for the case that the helmet failed, SI value was greater than 1200, determined as 86J. The energy calculation was performed numerically based on force-displacement curve of each test sample as discussed in Appendix A.

Knowing that each one-shot could absorb 6.76J, the required number of one-shots for front of the helmet was found from the following equation:

\[
\text{Number of one shots} = \frac{(86 \text{ J} - 59.1 \text{ J})}{6.76 \text{ J}} = 3.98 \approx 4
\]

For the top of the helmet it was found as:

\[
\text{Number of one shots} = \frac{(100 \text{ J} - 48.1 \text{ J})}{6.76 \text{ J}} = 7.68 \approx 8
\]

For the sides and the back of the helmet it was also very close to the results of the front of the helmet. So the required numbers of one-shot shock absorbers are as follows:

Front: 4 one-shot shock absorbers,

Sides: 4 one-shot shock absorbers,

Back: 4 one-shot shock absorbers and

Top: 8 one-shot shock absorbers.
Figure 30: Head and helmet drop testing results for the frontal - the graph of acceleration of the head based on g level ($g = 9.81$) and SI = 1287

Figure 31: Head and helmet drop testing results for the frontal - the graph of velocity of the head and SI = 1287
Figure 32: Head and helmet drop testing results for the frontal - the graph of displacement of the head and $SI = 1287$

Figure 33: The force-displacement results of the head and helmet, calculated from the results of the second round of drop testing for the front side of the helmet - $SI = 1287$
The second round of drop testing was done on the new design with the combination of the foam and one-shot shock absorbers. In this round of testing, for each side of the helmet, the required number of one-shot shock absorbers was positioned inside the helmet in parallel with the foam according to what had been designed, as discussed in Chapter 2. Drop testing was on each side with and without the one-shot shock absorbers. After drop testing was done, it was observed that the one-shot shock absorbers did not compress at all. However, after analyzing the force-displacement graph of the head and helmet, calculated by the same method as the previous section, for the drop testing results of the helmet with foam and one-shot shock absorbers and comparing them to the results of the helmet without any one-shots, a jump in the level of the force can be observed in the results with one-shot shock absorbers, which result from the contact of one-shot shock absorbers although the one-shots were not compressed to the point of permanent deformation (Figure 5-17). The problem is the increase of the stiffness of the system including both foam and multiple one-shot shock absorbers in parallel, when involving in compression together. After compression of the foam up to the point where the one-shot shock absorber is also compressed, the level of force has been decreased and this level of force is no longer capable of compressing multiple one-shot shock absorbers together.
Figure 34: The force-displacement results of the head and helmet, calculated from the results of the second round of drop testing on the new modified helmet with and without one-shot shock absorbers

The third round of testing was done on the new modified helmet in two steps: first, without any foam on one-shot shock absorbers and second, with one-shot shock absorbers positioned inside the helmet but without foam. This round of testing was done to see the effectiveness of one-shot shock absorbers individually and without any foam.

When the helmet without any foam and without one-shot shock absorbers was tested, the SI level has been reported as 1500. Based on calculated force-displacement graph the absorbed energy by the helmet in this case was 26.2J (Figure 5-18). In the case
with just one one-shot positioned inside the helmet, the SI decreased to 1100 and the absorbed energy, calculated from force-displacement graph Figure 5-19, was 33.2J.

And in the case that two one-shots settled inside the helmet, the SI decreased to 780 and the absorbed energy from force-displacement graph (Figure 5-20) was calculated as 44.7J. However, in this case, the impact severity was not high enough to compress the one-shot shock absorbers completely, and they had almost more 2mm for compression.

As can be seen using each one-shot shock absorber was effective in reducing the SI levels about 400 per compressing each one-shot. It should be noted that for the case with two one-shots the results show the reduction of 320 in SI level but they did not completely compress. If in this case the one-shots have been also compressed completely, the SI level reduction would have been about 390 which is very close to the expected 400. It was calculated based on the assumption that if one shot compressed completely the changes in compression is linearly proportional to acceleration and the SI level proportionally changes with the power of 2.5 with acceleration, regarding the definition of the SI. If the displacement changes, the SI also changes with the power of 2.5. Besides, it is interesting that after inserting one one-shot 7J more energy was absorbed by the helmet, which was basically absorbed by the one-shot shock absorbers. It is very close to what has been found as the Specific Damping Characteristic of a one-shot, which is 6.76J. The difference, however, can be caused by the fact that when a one-shot inserted in to the helmet and it was dropped, the compression of the one-shot was complete, and also, because of severity of the impact, the one-shot has also buckled and it might have caused absorbing more energy for the system. For the case with two one-shots, knowing that they did not completely compressed, the absorbed energy for both one-shots has been
calculated as 11.5J. Assuming that both one-shots were compressed identically, it can be concluded that each one-shot has been absorbed 5.75J of impact energy, which is also close to 6.76J Specific Damping Energy of a one-shot, and the difference here is based on incomplete compression of the one-shots.

![Force-Displacement results calculated from drop testing results corresponding to the helmet without any foam and one-shot shock absorber inside the helmet](image)

Figure 35: Force-Displacement results calculated from drop testing results corresponding to the helmet without any foam and one-shot shock absorber inside the helmet
Figure 36: Force-Displacement results calculated from drop testing results corresponding to the helmet with one one-shot shock absorber inside the helmet but without any foam.
Figure 37: Force-Displacement results calculated from drop testing results corresponding to the helmet with two one-shot shock absorbers inside the helmet but without any foam.

More testing with higher impact velocity was done to see the effectiveness of more than two one-shot shock absorbers, but again, because of using multiple one-shots and increasing the stiffness of the multiple one-shots, being active in parallel to each other, the one-shot shock absorber set did not permanently compress at all and the SI level did not change either.

It was found that in the case having multiple small one-shots, regarding the stiffness of their parallel arrangement, there is an optimum in the number of effective one-shots in order to be used as the impact protection layer in the helmet and using more one-shots may not work well in absorbing the total energy in the helmet and reducing the SI level.
The solution found was to use the larger sizes of the one-shot shock absorbers instead of the smaller one and find a way to package them properly inside the helmet; for instance, find a way to position them in an oblique position, and find a way to transfer the impact force properly in that direction. The solution with larger one-shots is appealing because the larger one-shot shock absorbers have a larger compression stroke and can absorb much more energy than the smaller ones (Figure 3-20). There would not need to be as many of them in parallel to each other to absorb all the energy, and this would solve the problem of the increasing stiffness.
Chapter Six

Conclusion and Future Work

6.1. Conclusion

The underlying questions of this study have been:

- What are the main reasons for the head injury and how the severity of the head injury can be estimated?
- Is it possible to design a new football helmet or find a new technology that is used in current helmets in order to reduce the severity of the head impact and reduce the probability of any severe head concussion?

To answer the first question, we have researched for all the probable reasons of the head injury and also the specific parameters and standards in assessing the severity of the head injuries, and it was found the most important cause of the head injury is related to acceleration of the head, including both rotational and translational accelerations. The most popular and acceptable criterion to specifically determine the severity of impact and
assess the probability of head concussion is Severity Index (SI), which is related to integration of the head acceleration over the impact duration, where a weighting factor of 2.5 is also applied to the acceleration values, and it is the main criterion that is an assessment base for such popular standard in designing the sport related facilities like NOCSAE.

For the next question, firstly, various types of shock absorbers have been considered and different testing methods have also been carried out in order to find the best candidates with the best energy absorption characteristic. Secondly, various designs were considered, incorporating the best shock absorber to absorb as much energy as possible during the impact and thus reduce the head injury. The Riddell foams and one-shot shock absorbers have been selected as the best candidates and the effectiveness of using each of these candidates individually and also in combination has been determined based on the NOCSAE standard and drop testing method. It was observed that in using multiple small size one-shot shock absorbers in parallel to each other and also in parallel to the foam, there is an optimum regarding the stiffness of the whole system and if the number of one-shot shock absorbers increases, the stiffness may also drastically increase. It may also cause a jump in the force, subjected to the head, without any compression-or in other words, no energy absorption in shock absorbers. A sudden increase in head injury probability may be occurred. It is also concluded that in impact based studies, not only it is important to pay attention to the energy analysis of the materials and designs, but also force analysis of the materials and impact resistive designs should be done and carefully considered.
6.2. Future Work

This work can be continued in different aspects. First, the design of the combination of the foam and one-shot shock absorber can be refined using the larger sizes of one-shot shock absorbers. The probable packaging problems could be solved by oblique positioning of the one-shot shock absorbers inside the helmet, for instance.

Moreover, other types of new shock absorbers can be sought and other technologies in order to find a better alternative for the liner part of the helmet. One that seems to have a good potential in this application is metallic foams like Aluminum Foams that have such noticeable energy absorption characteristic together with a very light weight characteristic.

Additionally, novel designs can be considered based on shape memory alloy materials that have very good energy absorption and shock resistive characteristic, while having the ability to return to its original shape after the impact and large deformations. It might be possible to made a new metallic foam out of shape memory materials and not only take advantage of shape memory alloy damping characteristics and metallic foam energy absorption characteristic, but also make a design capable of reaching its original shape and structure after deformations caused by impact.
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Appendix A

Numerical Calculation

In order to determine the energy absorption capacity of different materials and shock absorbers tested, the numerical calculation was performed on the results of force-displacement curve of each material.

The following MATLAB code was used in order to calculate the energy absorbed by each material or shock absorber tested in this study:

```matlab
clc
k(1)=0;
for i=1:1:max(length(f))-1
    k(i+1)=k(i)+ f(i+1).*(x(i+1)-x(i));
end
subplot(2,1,1);
plot(x,f);
subplot(2,1,2);
plot(x,k);
```
Based on this code, $k$ is the energy that we are looking for and the energy absorbed by each material or shock absorber is equal to the last value calculated for $k$. 