Design and Development of a Football Neck Support and Testing Apparatus

A Major Qualifying Project Report:

Submitted to the Faculty

of the

WORCESTER POLYTECHNIC INSTITUTE

In partial fulfillment of the requirements for the

Degree of Bachelor of Science

In Mechanical Engineering

By:

Zachary Bellion, ME

Sean Gillis, ME

Francis Lubega, ME

Blayne Merchant, ME

Collin Saunders, ME

Approved by:

____

Professor Fiona Levey

Table of Contents

Table of Contents 2
List of Figures 4
List of Tables 5
Abstract 6
1.0 Introduction 7
2.0 Background 8
2.1 Concussions: Causes and Long-term Effects 8
2.2 NFL's New Approach to Concussions 9
2.3 Current Football Equipment 10
2.4 Design Ideas of Previous Attempts at Minimizing Head and Neck Rotation 13
2.5 Shear-thickening Materials 15
2.6 Correlation of Neck Strength with Concussion Risk 16
2.7 Analysis and Testing of Head Kinematics 17
3.0 Testing Rig and Test Dummy Design and Construction 20
3.1 Design Criteria 20
3.2 Testing Rig and Test Dummy Design 21
3.3 CAD Modeling of the Testing Rig 30
4.0 Neck Support Design and Construction 32
4.1 Design Criteria 32
4.2 Neck Support Design Ideas 33
4.3 Neck Support Design 1 38
4.4 Neck Support Design 2 42
5.0 Testing 44
5.1 Test Set-up and Procedure 44
5.2 Measuring Linear and Rotational Accelerations 45
6.0 Results and Analysis 47
6.1 Linear and Rotational Acceleration 47
6.2 Head Injury Criterion and Head Impact Power 57
7.0 Discussion 63
8.0 Suggestions 65
9.0 Conclusion 67
References 68
Appendices 70

Appendix A: Properties of Poron XRD® 70
Appendix B: Survey Form and Questions..71

List of Figures

Figure 1: Kerr Collar........10
Figure 2: Neck Roll........11
Figure 3: Douglas Butterfly restrictor........11
Figure 4: Kato Collar........12
Figure 5: Kato Collar attached on Shoulder Pads........12
Figure 6: 2014 MQP Design........13
Figure 7: HALO by AEXOS........14
Figure 8: DEFLEXION™ by Dow Corning®........15
Figure 9: Poron XRD®........16
Figure 10: Equation for Peak Rotational Acceleration........18
Figure 11: Head Injury Criterion (HIC) Equation........18
Figure 12: Severity Index (SI) Equation........18
Figure 13: Head Impact Power (HIP) Equation........19
Figure 14: Virginia Tech Angles of Impact........21
Figure 15: Hybrid 3 Test Dummy Neck........21
Figure 16: Preliminary Design Sketch of Tensioning Device........22
Figure 17: SolidWorks Design of Tensioning Device........23
Figure 18: Silicon Neck Wrap........23
Figure 19: Test Dummy with Shoulder Pads........24
Figure 20: Entire Design of the Torso with Added Weight........25
Figure 21: Hand Sketch of Compressor to Cylinder........25
Figure 22: Pneumatic Cylinder Test Rig........26
Figure 23: Pendulum Testing Rig........28
Figure 24: Forward and Side Adjustment Points........28
Figure 25: Linear Sliding Rails of the Test Dummy........29
Figure 26: Final Pendulum Design after Construction........29
Figure 27: SolidWorks Simulation of the Neck........30
Figure 28: SolidWorks Testing of the Horizontal Beam........31
Figure 29: Initial Lift Support Design........33
Figure 30: 2nd View Piston Design........33
Figure 31: Neck Roll Design........35
Figure 32: T-Bracket Design Multi View........35
Figure 33: Neck Sleeve Front View (left) and Back View (right)........36
Figure 34: Sketch of Final Preliminary Design........37
Figure 35: Ribbed Foam Design........40
Figure 36: Design For 3 Piece Cut Out of 11x8in Sheet........40
Figure 37: Updated design or 3 piece cut out of 11x8in sheet........41
Figure 38: Final 3-peice Foam design sewn together........41
Figure 39: Final Brace Design........42
Figure 40: 3-2-2-2 Array of Accelerometers........45
Figures 41: 4 Accelerometers placed in weighted Styrofoam head........46

Figure 42: Testing Dummy at 0 Degrees....47
Figure 43: Testing Dummy at 45 Degrees....48
Figure 44: Testing Dummy at 90 Degrees....48
Figure 45: Constant X, Y, and Z Axes Used....49
Figure 46: Average Peak Acceleration in Neck Extension Straight On (0 degrees)....50
Figure 47: Average Peak Acceleration at 45 degrees....51
Figure 48: Average Peak Acceleration at 90 degrees....52
Figure 49: Average Rotational Acceleration at 45 degrees....53
Figure 50: Average Rotational Acceleration Straight on (0 Degrees)....54
Figure 51: Average Rotational Acceleration at 90 degrees....55
Figure 52: Absolute Value of Average Rotational Acceleration at 0 degrees....56
Figure 53: SI Equation....57
Figure 54: HIC Equation....57
Figure 55: HIC "Functional" Equation....57
Figure 56: Simplified HIC Equation....58
Figure 57: HIC at 0 degrees....58
Figure 58: HIC at 45 degrees....59
Figure 59: HIC at 90 degrees....59
Figure 60: Average HIP at 0 Degree Tests....60
Figure 61: Average HIP at 45 Degree Tests....61
Figure 62: Average HIP at 90 Degree Tests....61
Figure 63: Survey Results....63
Figure 64: Bent T-Bracket at the base of the Testing Dummy....65

List of Tables

Table 1: Material Properties of Poron XRD®....39

Abstract

In the 2017 NFL football season, 291 players were diagnosed with a concussion. The goal of this project was to create a neck support for football players to reduce linear and rotational acceleration of the head, since these accelerations have been shown to lead to concussions. The prototype neck support was designed to be comfortable, unobtrusive, attach directly to the shoulder pads, and absorb as much impact energy as possible. To test the neck brace, a testing mechanism consisting of a pendulum impact tester and a test dummy were designed to replicate the size, weight, and impact force of a football player experiencing collisions to the head. Preliminary analysis of the data showed a reduction in linear acceleration of the head by up to 30% during head on impacts to the face mask, but did not reduce rotational acceleration.

1.0 Introduction

Concussions are one of the most common injuries in football. 96% of ex-football players who donated their brains for research had developed some sort of chronic brain disease [1]. These players had developed memory and cognitive issues like dementia, Alzheimer's, depression and Chronic Traumatic Encephalopathy (CTE). The number of reported concussions has increased over the last decade and reached 291 in the 2017 NFL season [2]. Despite the increase in concussions, advancements in football equipment and rule changes have been made to make the game safer. With the focal point of a concussion being the head, a lot of attention has been focused on the production of safer helmets that are able to absorb the impact to the head and minimize damage from violent tackles. Generally, helmets are designed to prevent skull fracture and reduce direct forces to the head while having minimal effects on rotation, if any at all [3]. Neck collars have also been designed in order to reduce translational movement of the head and neck. Neck movement is at the forefront of concussion causation, specifically rotational movement. A study done by the NCAA and the United States Department of Defense [4] looked into the correlation of rotational head movement and concussions but a successful design has not yet been created and applied in the NFL.

The ultimate goal of our project was to create an effective neck support system that can be incorporated into shoulder pads to reduce the risk of concussions and the development of CTE in football. However, this device needed to be comfortable with minimal restriction to the player. A testing rig had to be designed that was able to simulate the movement of a human neck while also being able to test head movement in both translational and rotational directions.

2.0 Background

2.1 Concussions: Causes and Long-term Effects

According to the Centers for Disease Control and Prevention (CDC), a concussion is a type of Traumatic Brain Injury (TBI) that results in the disruption of normal brain function. The injury occurs when a blow, jolt, or bump to the head/body causes rapid back and forth movement of the head. This sudden movement can trigger the bouncing and twisting of the brain in the skull, leading to chemical changes in the brain and sometimes causing damage to brain cells. Furthermore, the word concussion is derived from *concutere*, a Latin word that means “to shake violently” [5].

A study by Steven Rowson and his colleagues in 2012 analyzed 31 impacts from National Football League (NFL) game film, focusing on the relationship between head kinematics and brain injury. Using previous research, the article states that rotational kinematics have a more severe impact on the brain than strictly translational kinematics. If the impact is not directed at the center of mass, the skull experiences acceleration as well as angular momentum. The farther away the impact is from the center of mass, the greater the acceleration and the larger the potential for brain injury. Rotational kinematics and brain injury are often sustained from contact sports, motor vehicle accidents, falls, and assault, among other incidents [6].

TBI accounts for approximately 1.7 million emergency department visits per year as cited by the CDC. Additionally, it is estimated that 1.6 to 3.8 million sports and recreation-related concussions occur in the United States every year [4]. In sports, they account for 5-9% of all sports-related injuries, with the majority coming from collision sports like football, rugby, and hockey. Numerous studies have reported the possibility of a higher incidence because the majority of these head injuries go undiagnosed [2].This is especially prevalent among student athletes. They may underreport their symptoms and/or falsify them due to competitiveness or in the hopes of a quicker return to play [4].

Contrary to popular belief, concussions generally cannot be seen on MRI or CT scans. As a result, people have to rely on symptoms which may or may not be perceptible. Some perceptible symptoms include balance problems, slow responses, poor concentration, and lack of recollection of the incident. Unfortunately, most symptoms are unnoticeable and can only be reported by the victim. This is another reason why the majority of concussions go undiagnosed. Imperceptible symptoms include headaches, drowsiness, sensitivity to sound and light, low energy, and anxiety among others [5].

Until recently, very little was known about the long-term effects of these head injuries. A combination of media coverage and a number of lawsuits from former NFL athletes has prompted more research on these possible effects. Some of these effects include Chronic Traumatic Encephalopathy (CTE), potential long-term behavioral changes, depression and late-life cognitive impairment [6,7]. According to a study on the long-term effects of concussions on cognitive and motor performance, some people can also experience post-concussion syndrome (PCS). PCS is characterized by an extended period of symptoms caused by an initial concussive event. The article defines long-term as one year after the concussive event. Another long term effect cited by the same article is behavioral changes. Retired athletes with a history of multiple concussions are susceptible to memory impairment, increased risk of depression and mild cognitive impairment. Retired professional athletes with a history of more than two concussions were five times more likely to develop mild cognitive impairment, memory deficits, and dementia in severe cases [7].

2.2 NFL's New Approach to Concussions

One of the biggest concerns of the NFL is how to deal with head injuries. In the 2017 NFL season, there were 291 players diagnosed with concussions, more than any season since the league began sharing the data in 2012 [2]. Before the start of the 2018 season, the NFL launched an Injury Reduction Plan [8]. The plan consisted of 3 steps that aimed to reduce the number of concussions immediately. The first step was to study preseason practices and the types of drills each team did. The second step was to prohibit the use of some lower-performing helmets. The third part of the plan was to implement and enforce new rule changes that could potentially reduce the number of concussions.

Currently, the NFL is sharing information with all teams about the causes of concussions, the helmets players wear, and injury data analysis. It is important that they work with coaches and team personnel to educate players on proper techniques that reduce concussion-causing hits. There has also been an emphasis on getting players to wear better-performing helmets. While no helmet can completely protect against head injuries, all helmets undergo lab testing to evaluate which helmets are best at reducing the severity of head impacts. The results of these tests are shared with teams so that players can make informed choices on which helmet to wear.

The first rule change the NFL made in 2018 was focused on the kickoff. Players on the kicking team no longer have a running start and must line up no more than one yard away from the location of the kick. Wedge blocks are also no longer permitted. A wedge block is defined as “two or more players intentionally aligning shoulder-to-shoulder within two yards of each other, and who move forward together in an attempt to block for the runner” [8]. The biggest rule change, however, is the new Use of Helmet Rule. This rule prohibits a player from lowering their head and initiating contact with the helmet to any part of the opponent. The contact does not have to be to the opponent's head or neck to be penalized. Players can be ejected for committing such fouls if the contact was clearly avoidable. These changes are all part of an effort to make the game safer.

2.3 Current Football Equipment

Football equipment is constantly being developed to help protect athletes as much as possible. Every year, helmets and shoulder pads are released that are lighter and stronger. Arguably, the most significant areas considered for protection are the head and neck. These two areas not only have the highest risk for the worst injuries, but are also the hardest to protect because they feature a number of sensory organs like the eyes and ears.

Figure 1: Kerr Collar

Protective gear like neck supports have been developed to try and address this issue. Neck supports are intended to help stabilize the neck, especially during impact. They do this by limiting/reducing movement in different directions. Examples include reduction in neck compression, rotation, and/or bending. One such example is the Kerr Collar, created by Dr. Patrick Kerr and produced by Kerr Sports [9] (Figure 1). Dr. Kerr studied the head and neck movement during impact and found that the two are correlated. His study looked at how hits not only affect the head, but how the neck absorbs the impact. The Kerr collar is placed inside the shoulder pads around the neck, and extends to a few inches below the bottom of the helmet. The collar is designed to reduce the axial compression of the neck, consequently reducing backwards head movement. The collar was designed to transfer the forces of the hit from the neck to the shoulders. Dr. Kerr, however, admits this is not a concussion prevention device but rather a force transfer system that can reduce the forces acting on the head and neck.

The protective neck collar, also known as the neck roll, is a device that attaches to the shoulder pads and runs outside the jersey around the neck and along the bottom of the helmet (Figure 2). The design patented in 1974 was an inflatable collar that used air to absorb impact energy. That design has been changed and developed by multiple companies. Later designs used materials such as foam and cotton stuffing. Nevertheless, the concept of the neck roll has been the same. Due to the minimal clearance between the helmet and the neck roll, head movement is limited in all directions. The disadvantages of this system out-weighed the advantages because players found that the collar was too bulky and restrictive [10]. The support is still used today but typically only by lineman who do not need full range of motion.

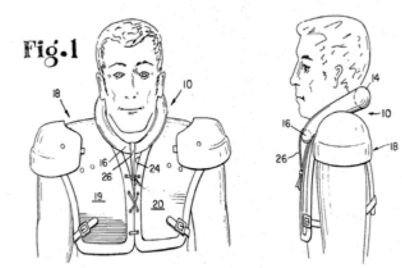


Figure 2: Neck Roll

Figure 3: Douglas Butterfly restrictor

The Douglas Butterfly restrictor is another protective device that is used to reduce the risk of head and neck injuries (Figure 3). The butterfly restrictor was designed to reduce the amount of *stingers* that occur during football hits. A stinger is categorized as a neurological injury that occurs when nerves in the neck and shoulder are stretched or compressed after an impact. This is often followed by a stinging pain, and then numbness in the athlete's extremities [11]. The butterfly restrictor is attached to the shoulder pads behind the helmet and extrudes up towards the top of the head. The butterfly restrictor reduces the motion of the head backwards, however there is still the ability to rotate and move side to side. The disadvantage of this device is the lack of support from side to side motion that can occur from an unexpected hit from the side [12].

Figure 4: Kato Collar

Figure 5: Kato Collar attached on Shoulder Pads

The Kato Collar by Guardian Athletics is a neck support system for football that was released in 2017. It is a lightweight and customizable collar that sits under the shoulder pads behind the head and neck (Figures 4 and 5). The Kato Collar was designed to reduce concussions by preventing the head from experiencing extreme movements during impacts that are likely to cause injuries like concussions. Guardian Athletics claims its product offers up to 30% head deceleration after impacts, without any vision, speed, or mobility impairment. The product is also said to prevent the stretching of nerves in the neck that lead to stingers [13].

2.4 Design Ideas of Previous Attempts at Minimizing Head and Neck Rotation

Figure 6: 2014 MQP Design

Over the last decade or so, there have been a few attempts to reduce the risk of concussions in sports by minimizing rotational acceleration of the head and neck. One such attempt was a 2014 WPI Major Qualifying Project [14] where students designed a fluid shock absorber which connected to a football helmet's rear (Figure 6). The base of the device featured a paddle inside a chamber filled with fluid. The paddle was constructed with holes which allowed for fluid displacement that created a resistive feedback force. During an impact, this resistive force would counteract any translational and rotational forces the head and neck might experience. Counteraction of these forces would reduce the risk of a concussion. A number of different fluid and hole combinations were tested to determine the best one. Fluids that were tested include shock fluid, multipurpose oil, a cornstarch & de-ionized water mixture, and lastly motor oil. These fluids were tested with a varying number of holes (1 or 2) and hole sizes (6 mm and 8 mm).

Another attempt is a smart collar by AEXOS called HALO™. It utilizes undisclosed smart materials that stiffen up to stabilize the neck and slow down head movement during impacts. As seen in Figure 7, the product features a compression shirt with a turtle neck design, worn under other protective sports equipment. The smart collar is located at the top of the design, covering the back and sides of the neck. It also has rigid plates of an unnamed polymer material that forms a flexible exoskeleton-like structure for support. The interior of the collar and compression shirt features silicone bands intended to improve posture and support the athlete's core. In the absence of high impact forces that cause whiplash, the collar's materials are soft and flexible, allowing for full range of motion in the neck [15]. Manufacturers claim the collar reduces whiplash by up to 30% during impacts. Materials with similar attributes are extremely valuable when designing a device that does not hinder athletic performance but is effective in reducing high impact forces.

Figure 7: HALO by AEXOS

2.5 Shear-thickening Materials

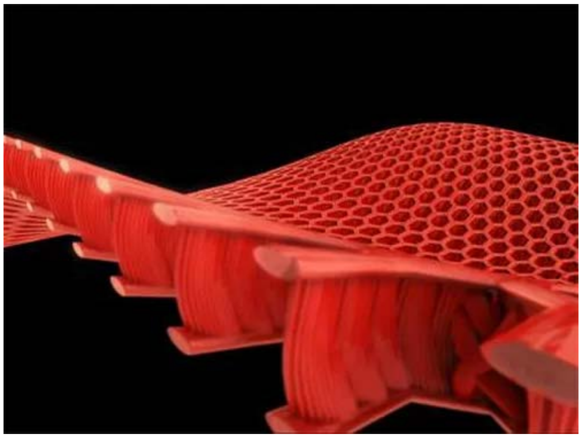


Figure 8: DEFLEXION™ by Dow Corning®

Currently, shear-thickening materials are being researched for their potential use in protective equipment. This is especially prevalent in the army, where there is the possibility of using them in bulletproof vests and other protective equipment. In a shear-thickening material, viscosity increases with the rate of shear strain. Simply put, the material thickens or gets a more solid consistency when subjected to a rapidly applied shear force. Shear-thickening plastics contain long-chain polymers suspended in a liquid lubricant that allow them to flow smoothly past one another at slow speeds. This material will replicate properties of a liquid when a slow and gentle force is applied. Conversely, high impact forces at high velocity will cause the material to harden and emulate properties of a solid [16]. When considering a contact sport like football, shear-thickening materials incorporated into protective equipment will absorb the energy of a violent hit to the head rather than the energy transferring directly to the head and neck. Surrounding the neck with such materials could reduce rotational and translational forces experienced during impacts.

D3O®, a commercially available product, and DEFLEXION™ by Dow Corning® (Figure 8), a patent-pending product, are two products that exhibit the properties of shear-thickening fluids. Both products are flexible, washable and breathable textiles that can be sewn into clothing and used as protective padding. US and Canadian skiers have used D3O® as protection against high impact forces occurring from high velocities down the slopes while DEFLEXION™ has been used in impact-absorbing motorcycle clothing. D3O® has a 34% reduction in peak transmitted force compared to standard thermoplastic rubber (TPR). These products have the potential to provide football players with proper protection necessary for the sport [16].

Another product used in protective equipment is a high impact absorption material called XRD®. The product is said to be most effective during high speed impacts where it boasts up to a 90% energy absorption rate [17]. It's technology is breathable, flexible and customizable, making it ideal for a variety of designs like elbow and knee pads. Manufacturers also vary the product's energy absorption, for low, medium, and high impacts, to accommodate other applications.

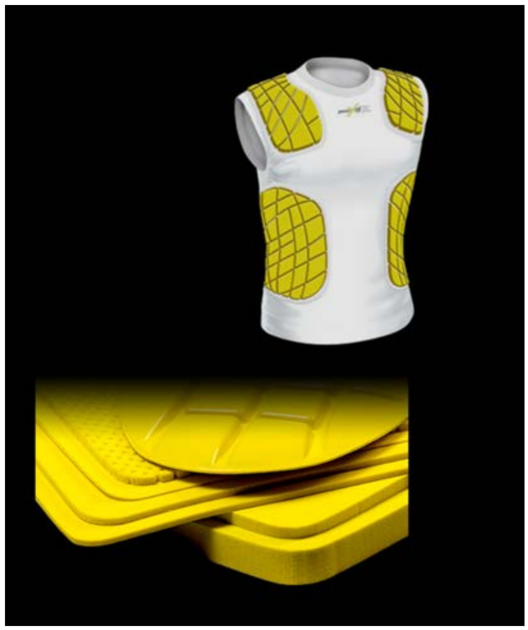


Figure 9: Poron XRD®

XRD® material is soft when at rest while above the "glass transition temperature" (Tg) of the urethane molecules. When stressed at a high rate or impacted quickly, the Tg of the material reaches the point when the urethane momentarily "freezes" - like water freezing into ice. When this happens, the material firms to form a comfortable protective shell that shields the body from impact better than other protective foams currently available [17]. XRD® materials are used in different sports like football and hockey, providing additional protection (Figure 9).

2.6 Correlation of Neck Strength with Concussion Risk

In addition to providing football players with materials that optimize protection, studies have shown that neck strength is correlated to concussion risk. A study done in 2010 and 2011, including 6,704 high school athletes across 51 high schools in the United States, found that concussed athletes had a smaller mean neck circumference and smaller mean overall neck strength than uninjured athletes [18]. Christy Collins and her colleagues found that stronger necks decrease head acceleration and that neck strengthening programs have strong potential for concussion prevention. Nowadays, helmets have become more effective in decreasing the direct force to the head but are ineffective at preventing rotational acceleration. Including neck strengthening exercises in an everyday or offseason routine could help reduce the risk of concussions.

2.7 Analysis and Testing of Head Kinematics

In the past, researchers have utilized the Head Impact Telemetry (HIT) System. This system consists of accelerometer arrays that are placed inside football helmets to gain insight into head kinematics associated with concussion analysis. A study by Steven Rowson, a researcher in Biomedical Engineering and Sciences at Virginia Tech-Wake Forest University, and his colleagues measured head acceleration for every head impact of 335 collegiate football players between 2007 and 2009. Two accelerometer arrays were used during this study, one being the commercially available HIT system and the other a custom six degree of freedom (6DOF) measurement device. The HIT system consists of six accelerometers mounted in a football helmet so that they remain in contact with the head at all times, guaranteeing that head accelerations are measured rather than the vibrations of the helmet's shell. Data was recorded for 40 milliseconds at 1000 Hertz when the accelerometer exceeded a threshold of 14.4 g. The unit, g, is referring to g-force which is the acceleration due to gravity. 1 g is equal to 9.80665 newtons of force per kilogram of mass. Peak rotational acceleration was estimated using the equation in Figure 10, where α is peak rotational acceleration, m is the mass of the head, ax is peak linear acceleration along the anterior–posterior axis of the head, ay is peak linear acceleration along the medial–lateral axis of the head, I is the moment of inertia of the head, and d is the perpendicular distance from the head's center of gravity to the impact vector. The 6DOF measurement device used 12 accelerometers, oriented and positioned differently, compared to the 6 accelerometers used in the HIT system. Other than this, the two accelerometer arrays were similar [19].

$$\alpha = \frac{m\sqrt{ax^2 + ay^2}}{I} d$$

Figure 10: Equation for Peak Rotational Acceleration

Other methods of testing head kinematics and its severity are the Head Injury Criterion (HIC) and Severity Index (SI). The 2014 MQP, mentioned in Section 2.4, analyzes shock absorbing fluid using HIC and SI. Results indicated that the 50 weight shock fluid was the most successful in reducing rotation to the head relative to the torso. The same fluid was the most effective in reducing the HIC value at 39%. This is a scalar value that determines the severity of a collision and can be related to the likeliness of an injury. The HIC is based on an acceleration curve with the equation below. The design was able to prevent the severity and likelihood of a concussion through the use of shock absorbing fluid in order to reduce translational acceleration [14]. The same study also evaluated SI using the SI equation below. The SI was used to evaluate the severity of an injury before the HIC was developed. Values for HIC and SI should not exceed 1000 or the player experiencing hits of that magnitude will suffer from extreme brain dysfunctionality and even death. HIC values over 200 are the threshold at which concussions occur. The study focused only on HIC since it includes the SI and is more advanced [20].

$$HIC \equiv \left\{ \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} (t_2 - t_1) \right\}_{max}$$

Figure 11: Head Injury Criterion (HIC) Equation

$$SI = \int_0^T \{a(t)\}^{2.5} \, dt$$

Figure 12: Severity Index (SI) Equation

In terms of rotational kinematics, Head Impact Power (HIP), shown in Figure 13, is another equation that evaluates the severity of a collision to the head, but includes rotational acceleration. Instead of solely taking linear acceleration and time into account like that HIC does, HIP accounts for linear acceleration, rotational acceleration, inertia and head mass. HIP was developed by Newman and colleagues to determine the probability that a concussion is likely to occur. The concussion probability curves that were generated permitted the determination of the specific values of each head injury assessment function that corresponded to significant concussion probabilities. From the probability curve for the HIP, a value of 12.79 kW corresponded to a 50% chance of concussion and an HIP of 20.88 kW corresponded with a 95% chance that a concussion occurred. These values were deemed to be preliminary and require additional testing [20]. As a reference, those numbers will be used in this study as a conservative estimate to determine HIP probability. A more recent study by Marjoux et. al. concluded that an HIP of 24 kW and of 30 kW corresponded to 50% and 95% risk of concussion, respectively [20].

$$HIP := m \cdot a_x(t) \cdot \int_{t_1}^{t2} a_x(t)\,dt + m \cdot a_y(t) \cdot \int_{t_1}^{t2} a_y(t)\,dt + m \cdot a_z(t) \cdot \int_{t_1}^{t_2} a_z(t)\,dt + I_x \cdot \alpha_x(t) \cdot \int_{t_1}^{t_2} \alpha_x(t)\,dt + I_y \cdot \alpha_y(t) \cdot \int_{t_1}^{t_2} \alpha_y(t)\,dt + I_z \cdot \alpha_z(t) \cdot \int_{t_1}^{t_2} \alpha_z(t)\,dt$$

Figure 13: Head Impact Power (HIP) Equation

3.0 Testing Rig and Test Dummy Design and Construction

3.1 Design Criteria

In order to accurately construct a prototype that helped reduce concussions, the testing apparatus needed to represent what a player experiences during high impacts. The testing rig and test dummy were designed as one entity but were constructed with two different purposes. The testing rig was designed to replicate the forces that a football player exerts on another player while the test dummy was designed to act as the player experiencing impact to the head. The following sections discuss the research, design and development of both the testing rig and test dummy.

Current helmet testing is being done to accurately represent the same amount of energy a football player applies during impact. Virginia Tech has a helmet-testing laboratory that tests helmet impact resistance [21]. They are replicating forces that are similar to those that occur in the field of play. Virginia Tech uses a pendulum with a ball mass of 15.5 kg, a maximum velocity of 6 m/s and a pendulum length of 1.9 meters. Using these numbers and a potential energy calculation, it was found that Virginia Tech produces 288J of energy at impact. This was used as a baseline model for the mass of the ball and the velocity the testing rig needed to have.

To calculate the amount of force needed to be applied to the head on impact, analysis of the forces applied by top NFL linebacker Luke Kuechly was considered. A study in Human Body Dynamics: Classical Mechanics and Human Movement by Aydin Tozeren [22] found that most of the force during a collision comes from the head and partial torso. The torso makes up 50% of the body weight and the head makes up 7.3% of the body weight, totaling 57.3% of the player's body weight. Based on the NFL's player statistics [23], Luke Kuechly weighs 240 pounds and has a maximum velocity of 8 m/s. Using these statistics of Luke Kuechly the kinetic energy formula was applied to calculate the energy that is created. Using 50% of his mass during a hit, Luke Kuechly can produce 1128 Joules of energy.

Applying 1128 Joules of energy to the test dummy means that the ball mass would have to be 63 kg with a velocity of 6 meters per second. Realistically, a ball mass of 63kg would be too heavy and would cause safety concerns during testing. Consequently, the impact energy was changed to replicate the testing done at Virginia Tech to serve as a point of comparison for the neck support model. By scaling back our design and using a ball of mass of 15.5 kg at a velocity of 6 m/s and a pendulum arm length of 6 feet the 288 Joules of impact energy was in line with the Virginia Tech testing.

3.2 Testing Rig and Test Dummy Design

Multiple factors were considered when determining how to test the neck support prototype. The neck portion of the testing rig had to mimic a human neck, including its rotational characteristics. After some analysis, it was determined that the use of a test dummy was the best way to test the neck support. Beginning the research for the construction of a test dummy used to replicate impact, we found that Virginia Tech was one of the leading research programs in head impacts.

Figure 14: Virginia Tech Angles of Impact

Figure 15: Hybrid 3 Test Dummy Neck

Virginia Tech Helmet Lab, which is part of Virginia Polytechnic and State University, tests and rates all the current helmets used in football. Since the NFL and college football programs use their ratings for helmet selections, we decided to look into how Virginia Tech tests their helmets. Virginia Tech uses a pendulum system to strike helmets at a varying velocity (Figure 14) [21]. The neck used on the testing rig is a Hybrid 3 neck (Figure 15) made by Humanetics Innovative Solutions [24]. The neck section they produced would have worked well for our project, however, due to high cost and a limited budget we decided to use this as a model to create our own. We analyzed how the neck was made, which uses alternating aluminum and rubber disks to create a bendable neck. The Hybrid 3 neck was tensioned using a tensioning cable, allowing the neck to bend in a similar way to a human neck. We then began to research what kind of rubber to use and how tight the neck should be tensioned. A previous MQP "Design of a Protective Device for Head and Neck Injuries in Football", created a neck that was similar to the Hybrid 3 [14]. The MQP neck used 3 inch diameter and .25 inch thick aluminum disks alternated with 2.5 inch diameter and .75 inch thick #30 durometer neoprene. The previous MQP found that the aluminum combined with #30 durometer was the best way to replicate the Hybrid 3 design and allow for similar neck movement. As shown by the design in Figure 14, the device used a 6 strand .5 inch steel cable down the center of the neck. That was tensioned to 5.41 foot- pounds to replicate the bending strength of the neck. The preliminary design sketch of the tensioning device with the Hybrid 3 design can be seen in Figures 16 and 17.

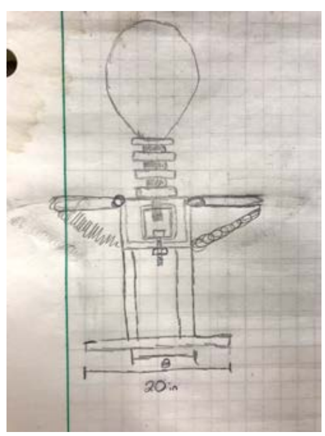


Figure 16: Preliminary Design Sketch of Tensioning Device

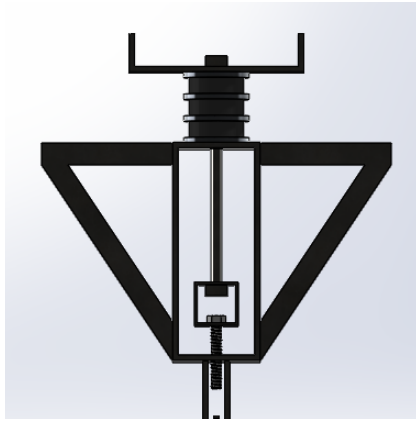


Figure 17: SolidWorks Design of Tensioning Device

After constructing the Hybrid 3 replica we designed a wrap to place around the neck that would resemble the dimensions and surface properties of a human neck. This design effectively replicated a human neck and gave us a realistic neck to perform tests, shown in Figure 18. Modeling of the neck was critical for our test rig because of the neck sleeve design of our neck support prototype. The replication of the human neck made from aluminum and rubber disks, and was much smaller in diameter than an actual human neck. Since the neck sleeve was meant to be in contact with the individual's skin, an accurate model of the neck needed to be created with properties similar to human skin. We found that silicone, which is commonly used for casting in the film industry, would be the best way to cast a human neck. After some research, LIFERITE, a skin-safe silicone rubber was chosen because it is reusable, safe to use on skin, and has a high tear strength. This product came in a kit that provided everything needed to make the mold including mixing guidelines and directions. The silicone was wrapped around the Hybrid 3 replica. The LIFERITE neck wrap allowed for the brace to attach around the neck without being in contact with the aluminum and rubber disks.

Figure 18: Silicon Neck Wrap

The tension in the tensioning cable was adjusted to take the added silicone into account during head impacts. We didn't want the silicone to restrict the head movement so a slit was cut down the front to allow for the neck to still move properly. This slit also had no effect on the way our brace would fit on the neck. The silicone neck wrap would allow for the neck brace to fit snuggly around the neck and have a realistic contact area when force was applied.

The torso of the test dummy was then designed. Simple bars were added to replicate shoulders that could be used to rest the shoulder pads on (Figure 17 and 19). The shoulder shown in Figure 17 were designed to simply hold the shoulder pads resting on them, however, a second MQP team redesigned the shoulder to analyze shoulder movement occurring during hits in lacrosse. This redesign is shown in figure 20 but didn't affect the testing of our neck brace. The torso also had to be able to house the tensioning mechanism to allow for the proper torque to be acquired. Bars were also added, shown at the bottom of Figure 20, to allow for 60lbs to be added to the torso to replicate 50% of the weight of a human that is used during a hit in football.

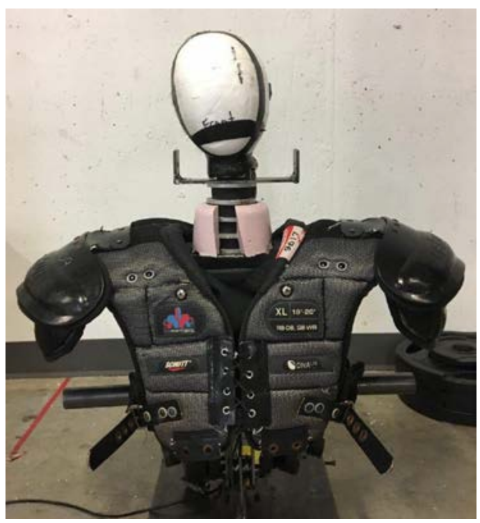


Figure 19: Test Dummy with Shoulder Pads

Figure 20: Entire Design of the Torso with Added Weight

A couple of different methods of applying impacts to the helmet were considered. Virginia Tech uses a pendulum that can be dropped from different heights, allowing for the impact velocity to be changed. The pendulum idea was put aside at first because we had access to a variety of pneumatic air cylinders applicable to the testing rig. The air cylinder would work in a similar way. A pressure regulator could adjust the total amount of air pressure applied to the cylinder. Using a switch, air could be rapidly forced into the air cylinder. Figures 21 and 22 display the hand sketch and SolidWorks sketch of the pneumatic cylinder design. Using the equation F = $p \pi d^2 / 4$ allowed us to see that varying the gauge pressure could change the speed of the piston, therefore changing the impact force applied to the head.

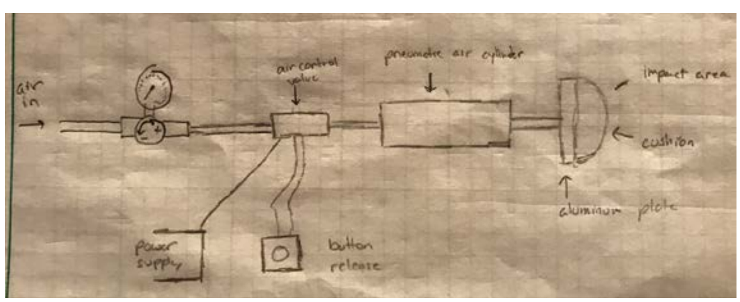


Figure 21: Hand Sketch of Compressor to Cylinder

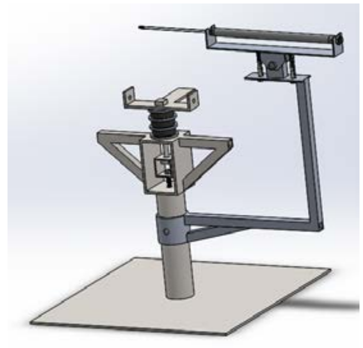


Figure 22: Pneumatic Cylinder Test Rig

The initial testing rig design utilized a pneumatic air cylinder to strike the helmet. As shown in Figure 22, the cylinder would be on a pole that had the ability to rotate around the impact zone. The available pneumatic air cylinders were tested to find the output force before fabricating the design. With an air pressure of 105 psi, a force of 5000 Newtons (as produced by NFL linebacker Ray Lewis) needed to be produced to accurately represent the impact of a NFL player. The speed of the cylinder was dependent on the diameter and the air pressure applied. Multiple sized cylinders were tested to see how much force could be applied. It was concluded that cylinders with smaller diameters would move faster but with a smaller force. The calculations are shown below.

	Equation	Variables
Force	$F = p\,A$ $= p\,\pi\,d^2 / 4$	F = force exerted (N) p = gauge pressure (N/m^2, Pa) A = full bore area (m^2) d = full bore piston diameter (m)

Cylinder Testing

Cylinder 1: 0.75DPSK08.00
Stroke length = 8”
Bore Diameter = ¾”
Pressure set at 105 psi in shop
Measured force: 140N

Cylinder 2: 1.50DSR05.00
Stroke length = 5”
Bore Diameter = 1.5”
Pressure set at 105 psi in shop
Measure force: 400N

Cylinder 3: 1.25DXPSR05.0
Stroke length = 5”
Bore Diameter = 1.25”
Pressure set at 105 psi in shop
Measured force: 300N

After testing, it was determined that the pneumatic cylinders were too slow for the force and energy needed. A reverse calculation was applied and in order to produce 5000 Newtons of force with only 105 psi of pressure, a cylinder with an 8.4-inch diameter would be needed. Even so, the limiting pressure would cause the pneumatic cylinder to produce a high force but at a very slow velocity. It was evident that this would not work with an instantaneous strike.

The team decided to use a pendulum-based testing rig, shown in Figure 23. The pendulum allowed the calculated ball mass to strike the helmet and cause a translational force directly to the helmet, which then caused the neck to absorb and react to the forces applied. The rig had a total height of 9 feet with a pendulum arm of 6 feet. The pendulum arm = had a mass of 5lbs. The contact point of the pendulum and helmet was 32 inches from the ground.

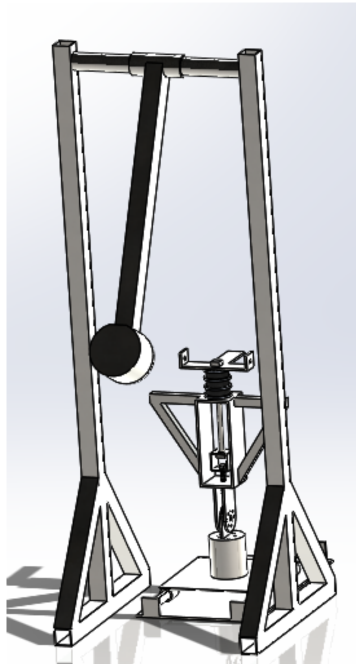


Figure 23: Pendulum Testing Rig

The testing dummy was able to move forward and backwards at about 45 degree increments, as shown in Figure 24, in addition to rotation around the vertical axis at 45-degree increments. This allowed testing to be done from multiple angles. The swing arm could also slide along the horizontal crossbar at the top of the rig, enabling the mass to strike the helmet at different points along the horizontal axis.

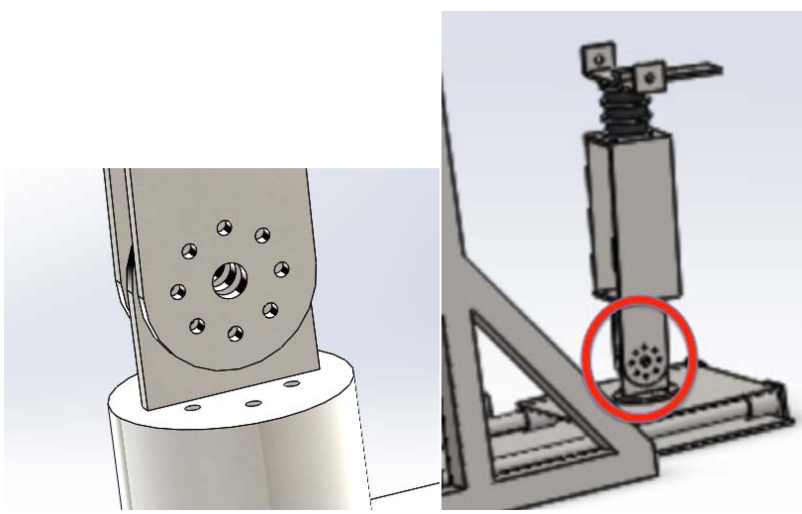


Figure 24: Forward and Side Adjustment Points

After contact, the testing dummy would need to move away from the pendulum to avoid any repeated impacts to the helmet. As shown in Figure 25, the testing dummy was designed to sit on guide rails that would allow it to slide away from the pendulum in a controlled manner. To control speed and distance traveled, set screws were placed on the rail sliders of the test dummy. If the dummy slid too easily, determining the acceleration in the testing dummy neck would be difficult.

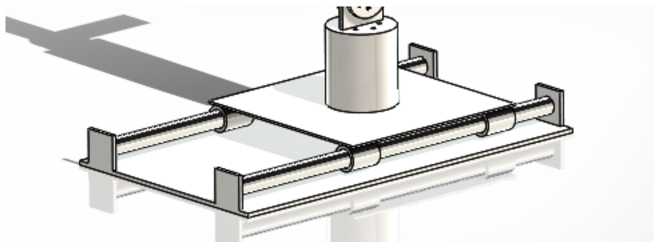


Figure 25: Linear Sliding Rails of the Test Dummy

The rest of the testing rig was made with a 3/16" mild steel plate and 2" x 2" mild steel square tubing. The rig was made primarily out of 3/16" thick A36 steel plate cut to the appropriate dimensions and welded together. The square tubing was 2 inches by 2 inches. This was welded together and strong enough to support the swing arm. The impact ball had an arm on the back that allowed for more to be added to increase the mass of the ball. The final prototype of the testing rig and testing dummy is shown in Figure 26.

Figure 26: Final Pendulum Design after Construction

3.3 CAD Modeling of the Testing Rig

SolidWorks was chosen for modeling because the program enabled multiple parts to be designed and put together into an assembly. The assembly allows for the parts to be mated, providing free range of motion for the pendulum arm. Other parts such as the square tubing were able to be placed and fixed to resemble the welds. This feature gave an understanding of how the pendulum would function.

The stresses on select parts were simulated to make sure they would react properly. The primary parts that were analyzed were the crossbar that the pendulum rotated around and the swing arm itself. The results of this simulation were confirmed using calculations shown in the next section. The other part that was analyzed was the neck movement. The neck is made of aluminum and #30 durometer rubber disks. This aluminum and rubber configuration mimics the function of a human neck. As seen in Figure 27, stress analysis in SolidWorks modeled the application of forces to the neck to certify that the neck moved as intended. The rubber compressed and absorbed some of the impact that was applied.

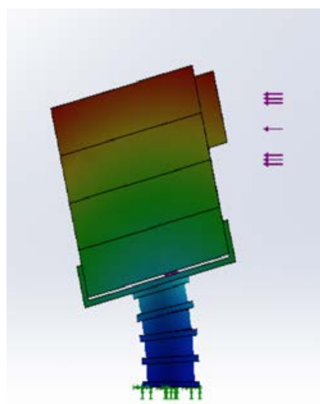


Figure 27: SolidWorks Simulation of the Neck

An analysis of the pendulum was carried out to ensure that the pendulum could support the desired weight of 16 kg. Through a SolidWorks simulation, the pendulum was positioned horizontally, at a 90 degree angle to the supports, and stress testing was performed to see how the pendulum would react when a weight of 16kg was applied to the center of the beam as shown in Figure 28. The simulation displayed that the beam would only deflect 0.0817 inches. A hand calculation was also done to confirm the beam deflection and the same result was obtained. This showed that the beam with an outer diameter (OD) of 1 inch and an inner diameter (ID) of 0.75 inches is strong enough to support the required weight.

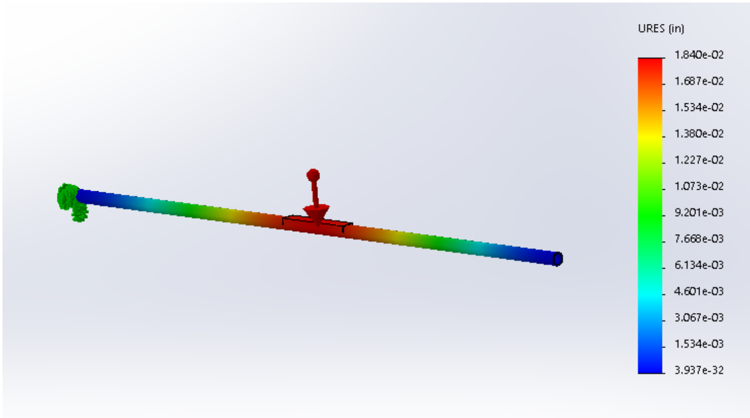


Figure 28: SolidWorks Testing of the Horizontal Beam

Horizontal bar at top of pendulum rig

OD=1in ID=.75in

L= 48in W= 35lbs

E= $29*10^6$ lb/in^2

I= Pi(OD^4 -ID^4)/64 = .034in^4 def= (L^3*W)/(48*E*I) = .0817 in.

4.0 Neck Support Design and Construction

4.1 Design Criteria

While constructing the testing rig and testing dummy, the neck support was researched and developed. The goal of this project was to reduce the risk of concussions among football players by designing a neck support that minimized rotational and translational forces as well as, if not better than, existing products. A plan was devised and followed to achieve this goal and its objectives. It entailed the identification of different design possibilities, evaluating them and determining the best one. The design possibilities were influenced by a few design constraints. The first constraint was that the design had to reduce translational and rotational acceleration of the neck, as research showed this was one of the main causes of concussions. Easy attachment to shoulder pads served as the second design constraint. Shoulder pad attachment would enable the athlete to wear the neck support with existing shoulder pads. The final design constraint was comfort. Lack of comfort has been cited as a significant reason why most athletes choose to avoid extra protective equipment.

Athletes often use compression shirts with rib, back, and/or shoulder protection. Incorporating the neck support into a compression shirt would require the athlete to choose between the two or wear both equipment at the same time. Wearing both at the same could cause discomfort, especially on hot days. After considering these factors, it was determined that an attachment to the player’s shoulder pads rather than an entire compression shirt, was the better option.

Some of the biggest constraints on our design had to do with time, budget, and availability of materials. Time was a huge limitation, because we only had a limited time to use the space that the testing rig was located. Due to the size of the rig and nature of the testing, the only space available for testing was the Robotics Pit in the Sports and Recreation Center. This space is used for a variety of events and we were only permitted to use the space for a few weeks. If there had been more time to spend testing, we could have been able to analyze the results, and then make any improvements to the design so that it could be tested again. Another constraint was the commercial availability of some materials. The group had the intentions of using a shear thickening fluid as the impact absorbing material for the neck support. Unfortunately, there were not any that were commercially available for purchase, so the design had to be altered. Another difficult material to find was the impact absorbing foam used in the design. Many companies were only interested in large purchases of material. The group was able to receive free samples of impact absorbing urethane foam from Poron XRD®. However, these samples only came in 8.5” x 11”, so the design had to be slightly adjusted to fit on one sheet.

4.2 Neck Support Design Ideas

Based on the design criteria, a variety of design ideas were brainstormed and considered to help reduce translational and rotational head accelerations. The first idea involved the application of two lift support struts attached to the back of the helmet and shoulder pads, as seen in Figures 29 and 30.

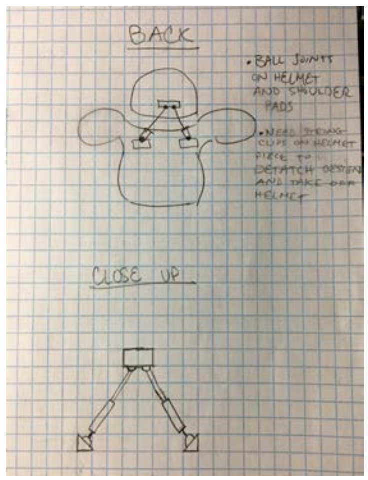


Figure 29: Initial Lift Support Design

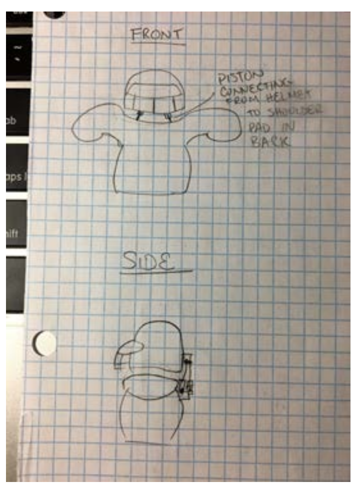


Figure 30: 2nd View Piston Design

The support struts would attach to ball joints at each of the connection points to allow for 180 degree rotation at the helmet connections, and 90 degree rotation at the shoulder pad connections. Air pressure in the support struts would minimize impact forces transferred to the head and neck. During normal head movement, the speed at which air flows through the struts would be slow, providing mobility for proper head kinematics during normal play. Air pressure would increase during extreme head movements caused by big impacts and/or collisions. This would consequently reduce linear and rotational acceleration.

This design was inspired by the lift support struts found in the trunk of a car. The struts in a car operate using a gas spring, completely sealing off pressurized nitrogen gas inside the pistons. When one pushes on the gas spring, the piston rod is forced into the cylinder, compressing the gas. Then once the gas spring is released, the gas inside the cylinder pushes the piston in the strut back out. The force that the lift support strut generates is produced by a difference in area on both sides of the piston, while the pressure on either side is the same. The force that the lift support rod can generate is calculated by the equation below:

	Equation	Variables
Force	F = A * p	F = force A = piston area (square inches) P = gage pressure

In order for the design to be successful, the two lift support struts had to produce as much force as a football player exerts during a tackle. The idea was to mimic the function of the neck and provide additional strength and support during impacts. Further analysis highlighted the impracticality of this design during football play. Since the design required fixed connections from the helmet to the shoulder pads, the athlete would not be able to take off their helmet without any assistance or the necessary equipment. From an athlete's perspective, this is an inconvenience as players like to take off their helmets on the sideline or during a timeout. Another issue was the safety risk that these pistons posed. Given the plethora of impacts during football, there was a possibility of the pistons breaking and causing further injury to the athlete wearing them as well as other athletes on the field. These potential problems made this design impractical.

The next design idea brainstormed was a silicone-lined neck roll filled with a shear-thickening fluid (STF), as shown in Figure 31.

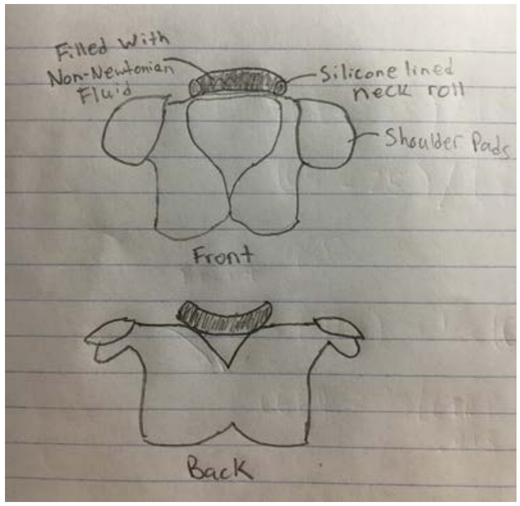


Figure 31: Neck Roll Design

This design idea was inspired by the traditional neck roll, which is attached to shoulder pads and wraps around the head and neck opening. However, the brainstormed design would feature a shear thickening fluid inside the collar as opposed to a cotton fabric. A silicone lining would hold the shear-thickening fluid inside the neck roll. Given its shear thickening properties, this design would possibly reduce more neck and head acceleration during impacts than its cotton fabric counterpart. Although the replacement design would help improve impact absorption compared to traditional neck rolls, it wouldn't solve the problem of being too bulky and restrictive. Discomfort and aesthetics are some reasons players choose not to wear traditional neck rolls; this design would not improve upon any of these factors.

The third design idea was the T-shape collar design, as shown in Figure 32. The idea behind this was minimizing head rotation during impacts with a structure surrounding the helmet. This structure would attach to the shoulder pads and go up the helmet.

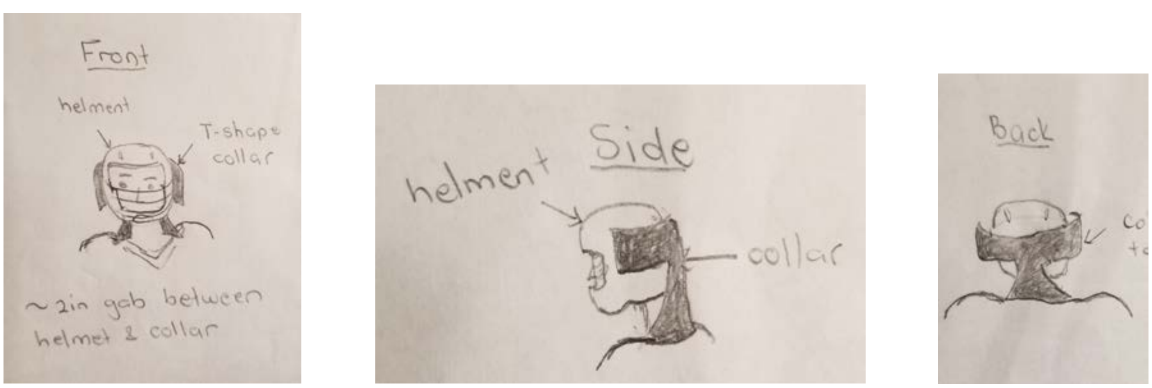


Figure 32: T-Bracket Design Multi View

For the product material, having a strong but flexible material was discussed. A soft material or padding would serve as the interior lining of the collar to absorb impact forces. As brainstorming progressed, design flaws were realized. The main issue with this design was the lack of mobility. Side to side head movement would be nearly impossible. Another issue was the effectiveness of this collar in impact absorption. To create a light collar, the amount of padding had to be minimized. Therefore impact protection would be minimized as well. Going forward, the main priority was mobility and comfort without sacrificing protection.

The next design stemmed from the idea of creating a tight-fitting but comfortable neck sleeve that could reduce harsh rotations and impacts of the neck. The original brainstormed sketches of the front and back views are shown in Figure 33.

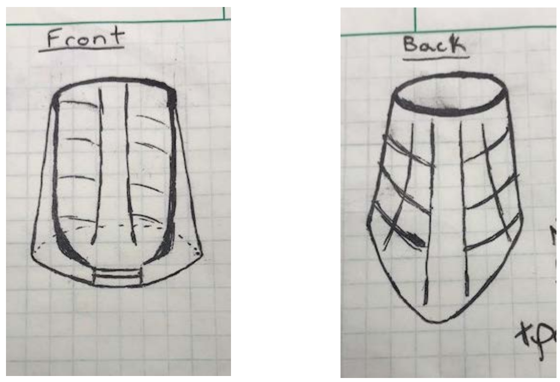


Figure 33: Neck Sleeve Front View (left) and Back View (right)

This design envisioned a turtleneck-type of sleeve that would have pouches of STF. The presence of a STF would enable the sleeve to stiffen once subjected to an impact. The front portion of this design would feature a cut-out to minimize discomfort and restriction of the wearer's throat, while the back portion would extend down onto the wearer's upper trapezoids. The neck support would have a gel, foam, or plastic that would offer some rigidity. When analyzing this design, it was clear that some factors were overlooked initially. One was how this sleeve would attach to the athlete's neck. It either had to be incorporated into shoulder pads or have some sort of support system that would allow it to securely rest on the athlete's neck. Another overlooked factor was comfort. Anything tight to the neck could possibly be restricting and uncomfortable. The goal of this design was to be non-restrictive, while reducing rotational and translational head accelerations during impact.

The previous design attempts culminated in the final preliminary design idea (Figure 34). This design would feature a compression neck sleeve that would connect to the top of the wearer's shoulder pads. A high impact absorption material would be enclosed by polyester material, making the neck support light, breathable and flexible. This would be sewn to a separate nylon piece with four connections that would allow it to be a secured to a pair of shoulder pads. Using this idea, two separate prototype designs were made to be tested, with the only difference being the material used for impact absorption.

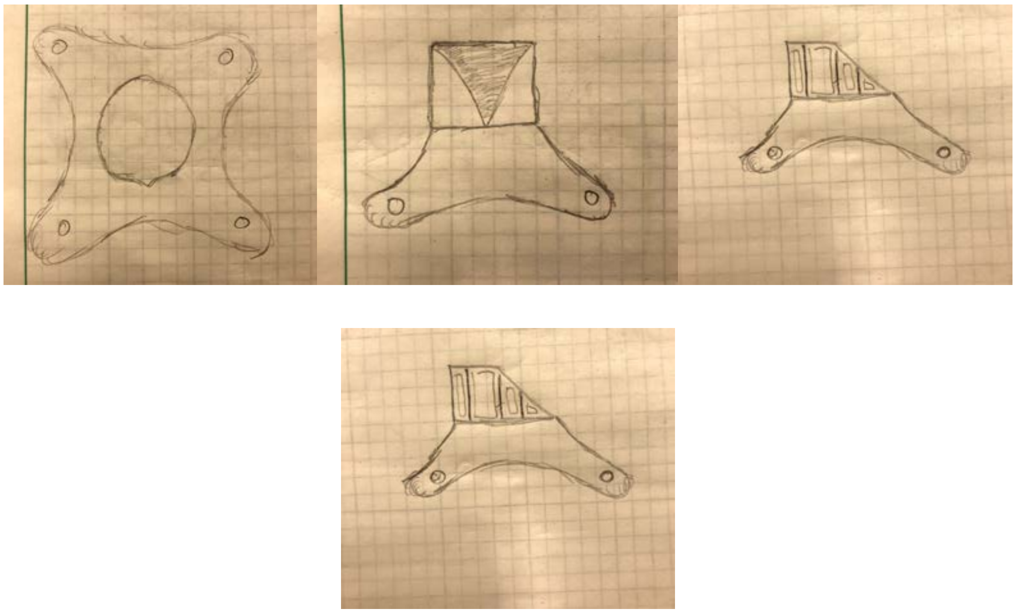


Figure 34: Sketch of Final Preliminary Design

4.3 Neck Support Design 1

Two designs were simultaneously developed and compared for effectiveness. The principal difference between the two designs was the materials they employed to reduce rotational and translational forces in the neck. Design 1 utilized an impact absorbing urethane foam called Poron XRD®. This foam mimics the properties of a non-Newtonian fluid. Specifically, it momentarily hardens when subjected to a high stress or sudden impact. An analysis of three different materials, D3O, DEFLEXION, and Poron XRD® led to the first design; all three have shear thickening properties. Due to availability, Poron XRD® was selected for the prototype. The company was able to supply multiple sample sheets of extreme impact protection Urethane foam. The samples were only available in 8.5” x 11” sheets. Physical properties of the foam are displayed in Appendix A, but Table 1 below represents material properties that were vital to the project application. A preliminary prototype was made out of each of the 5 materials listed to compare and decide which was best. Outside of tensile strength, the material properties below are similar across all three materials. Poron XRD® 15500 was expected to be the most effective material for the design since it has a much larger tensile strength, resisting tensile forces and, in turn, resisting compressive forces in the opposite direction. However, once the sheets of foam were cut to shape, it was clear that the 15500 was too stiff to be comfortable when worn around the neck. The Poron XRD® 12374 was chosen for the final prototype in order to balance impact reduction and comfort.

Material	Thickness	Density	Tear Strength	Tensile Strength	Resilience	Compression Set	Compression Force Deflection *
Poron XRD® 15500	0.5 in (12.7 mm)	15 lbs/ft^3	0.9 kN/m	483 kPa	3-5	<10%	4-9 psi
Poron XRD® 15374	0.374 in	15 lbs/ft^3	0.9 kN/m	483 kPa	3-5	<10%	4-9 psi
Poron XRD® 12500	0.5 in (12.7 mm)	12 lbs/ft^3	0.9 kN/m	310 kPa	3-5	<10%	1.5-5.5 psi
Poron XRD® 12374	0.374 in	12 lbs/ft^3	0.9 kN/m	310 kPa	3-5	<10%	1.5-5.5 psi
Poron XRD® 09374	0.374 in	9 lbs/ft^3	0.8 kN/m	207 kPa	3-5	<10%	1.1-3.4 psi

Table 1: Material Properties of Poron XRD®

The original intention was to cut strips of foam and sew them together to form the inner structure of the design. The strips would meet each other during impact, reducing rotational and translational forces. Space between the strips would allow for flexibility during normal neck movement. After a brief analysis, it was decided that cutting rectangular gaps or ribs from the sheet of foam (Figure 35), as opposed to attaching the strips together, would be more efficient. It was also deemed more durable because of the fewer points of attachment, hence less opportunity for structural tear. The foam would be enclosed in polyester and elastane fabric and attached to the athlete's shoulder pads. Both edges of the ribs were angled outward 45 degrees to limit restriction to the front of the neck.

Figure 35: Ribbed Foam Design

The high impact XRD® foam we decided to use for this design could only be provided in 8.5" x 11" sheets. The company that makes the foam usually only partners with large companies and manufacture to specific needs. The size of the sheets forced us to adapt our design. The design no longer could fit as one piece on the sheet of foam, so three separate pieces were cut out and sewn together with high strength polyester thread. The altered three piece design is shown below (Figure 36).

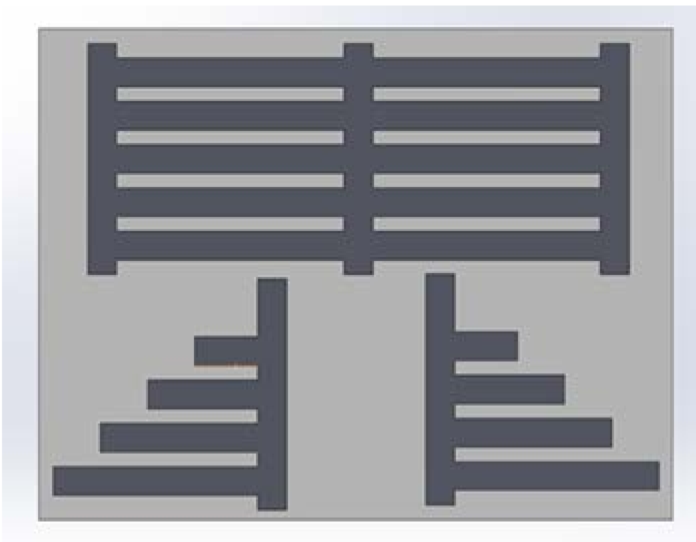


Figure 36: Design For 3 Piece Cut Out of 11x8in Sheet

After a few prototypes with the desired dimensions, we added extra structural support to the outer angled ribs (Figure 37).

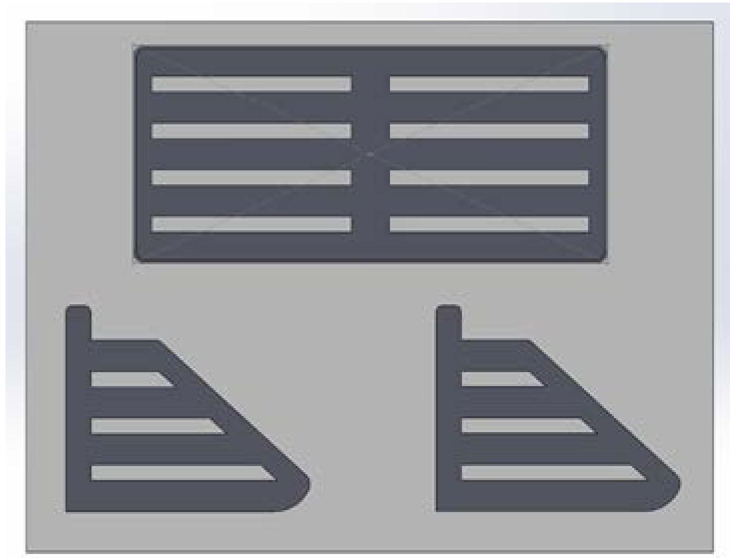


Figure 37: Updated design or 3 piece cut out of 11x8in sheet

For the final design, the dimensions were kept the same and finalized on the Poron XRD® 12374 foam sheet (Density 12lbs/ft ^{2}, Thickness .375 inches). The major change in this version of the design is that the ribs were not fully cut out. Horizontal cuts, ¼ inch apart, and approximately 50% deep were made allow proper movement of the neck during gameplay (Figure 38).

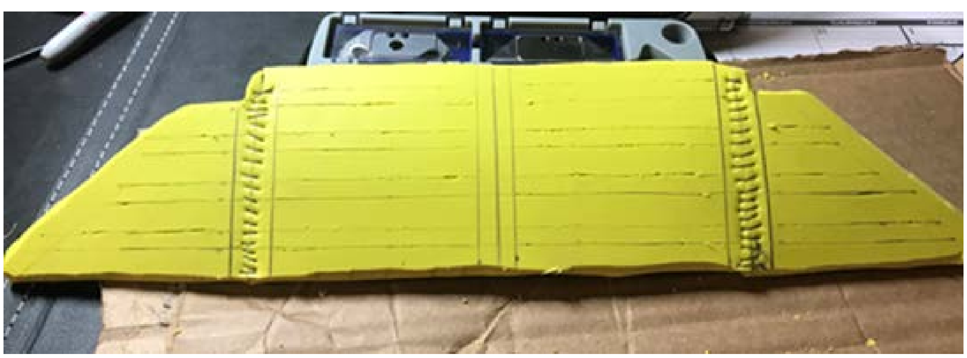


Figure 38: Final 3-peice Foam design sewn together

The cuts allowed for some bending, but during impacts of high accelerations, there would be sufficient compressive strain rate. A fabric made of 90% polyester and 10% elastane was used to surround and hold the Poron XRD® design in place. This fabric was chosen because it has similar qualities to that of Under Armour, which gives a tight and comfortable fit to the skin. It also provides a small amount of friction that will help keep the neck brace in contact with the skin. The fabric was then sewn to a 12 inch by 16 inch sheet of heavy-duty nylon that attached to the inner shell of the shoulder pads. Four holes were cut out of the nylon to attach and fasten to the shoulder pads. 5/16” grommets were inserted into the four holes so that forces acting in every direction didn’t affect the strength and quality of the material.

Figure 39: Final Brace Design

4.4 Neck Support Design 2

Similar to the above design, this neck support utilized a material with non-Newtonian fluid properties. Both neck supports (Design 1 and Design 2) start at the neck, just below the ears, and extend down to the mid chest where they are attached to the shoulder pads. Design 2 features a shear-thickening fluid (STF) instead of foam, as the primary material for impact protection. The STF was impregnated in Kevlar so that the Kevlar acts the scaffold for the fluid. The purchase of a shear thickening non-Newtonian fluid was the original intention, but there weren’t any commercially available. Extensive research revealed multiple methods of creating a STF. Creating a STF with Polyethylene Glycol and Silica nanoparticles proved to be the best method due to availability and budgetary constraints. To produce a fluid with true shear-thickening properties, mixing the two materials by hand was not enough; a more effective method of particle dispersion required ultrasound. The Chemical Engineering Department at Worcester Polytechnic Institute verified that the proper tools were accessible. In the lab, the department’s sonication bath, vibrational vortex and conventional oven were available for use.

The STF was created by mixing Silica nanoparticles of 0.5 - 10 micrometers and Polyethylene Glycol (PEG) 200 in a 55:45 ratio, respectively, by weight. In order to properly mix the solution, tools like a sonication bath, vibrational vortex and conventional oven were utilized. The first method of particle dispersion was done using the vibrational vortex to slowly mix the silica into the PEG. 100 mL of PEG, with a weight of 112.4 grams, was added to a 250 mL Erlenmeyer flask. The flask was placed on the vibrational vortex and turned on. After a vortex was created within the PEG, 137.79 grams of silica was slowly added, resulting in a mixture with a 55:45 ratio of silica to PEG. The mixture was moved, still in the flask, to the sonication bath to send ultrasound waves throughout the fluid to properly disperse the Silica particles in the PEG. Formation of Silica agglomerations prevents the mixture from obtaining non-Newtonian properties [25]. The flask sat in the sonication bath for 30 minutes to allow enough time for the ultrasound waves to break up any agglomerations. While this was happening, Kevlar was cut into sheets that would mold to the back of the neck properly. These dimension were the same as Neck Support Design 1 and were big enough to cover the back and side of the neck but small enough so that it doesn't wrap around the front of the neck, causing discomfort. Four sheets of Kevlar were sewn on top of each other so that the fabric would soak up as much of the non-Newtonian fluid as possible without being too thick, avoiding possible hardening.

Unfortunately, the silica-PEG mixture did not show any shear thickening properties, so only the prototype for Neck Support Design 1 was constructed. The prototype for this design was not worth finishing once it was known that the STF manufacturing process did not work. Otherwise, the impregnated Kevlar then would have been enclosed in a nylon fabric for comfort, and attached to the shoulder pads with the same nylon fabric used in the first design.

5.0 Testing

5.1 Test Set-up and Procedure

Since the swing arm of the pendulum is 6 feet long, a safety plan was put in place to ensure that the rig was used in a safe manner and injuries were prevented while testing. The detailed safety steps are listed below.

Safety Steps for Testing

- A 12 ft by 12 ft square of tape was placed on the ground around the center point of contact of the pendulum and test dummy. This was the restricted zone. The only person allowed in the restricted zone was the person operating the pendulum.
- Everyone around the pendulum was required to wear safety glasses.
- No one was allowed to operate the pendulum alone, at least 2 people had to be present during operation.
- The pendulum operator had to give a “dropping” call before releasing the pendulum.
- A stop with a rubber absorption pad was placed on the opposite side of the pendulum to prevent the sliding torso from hitting the wall with a high force.

To test the operation of the testing rig, weights were added to the end of the pendulum arm. 60 pounds of weight was placed on the torso to replicate the actual weight of a small football player’s torso. 8 pounds of weight were also placed inside the Styrofoam head so that it had a total weight of 10 pounds. We used a Styrofoam head to resemble a human head inside a helmet. The Styrofoam also allowed for the accelerometers to be mounted on and in the head.

We then torqued the tensioning mechanism of the neck to 65 inch pounds or 5.41 foot pounds. This tension caused the neck to fully compress to 4 inches which we used as our length for repeatability of accuracy of the tension. This was checked after every other impact to ensure that the tensioning cable remained accurate. The pendulum arm was dropped with varying weight from 90 degrees.

Despite the intentions of using 15.5 kilograms (about 34.17 pounds) for the weight that struck the testing dummy, multiple test runs at weights of 15 pounds and 20 pounds forced the tensioning cable in the dummy to dissever from the clamps holding it in place. It was concluded that a weight of 34.17 pounds would apply too much force for the testing dummy to handle, therefore tests were conducted using less weight. The pendulum arm weighed a total of 5 pounds with no added weights. Varying weights of 5,10 and 12.5 pounds (added weights of 0, 5 and 7.5 pounds) were used during impact testing. Three tests were performed with each weight, at each angle (0, 45 and 90 degrees) with and without the neck brace on the dummy.

The shoulder pads and helmet that were used were borrowed from the equipment room on campus. The helmet was a Rawlings NRG Force and the shoulder pads were Shutt DNA Lite 9617 Flat Pad. These items were intended to be discarded because of the old age, so we were allowed to use them for our testing.

5.2 Measuring Linear and Rotational Accelerations

An array of accelerometers was set up in order to determine the linear and rotations accelerations experienced by the head during impact. The accelerometers were placed within the helmet in the same configuration used in a patented 3-2-2-2 array, as shown in Figure 40, positions a triple-axis accelerometer at the center of the helmet, and three double axis accelerometers are positioned at the side, back and top of the helmet. This configuration not only provides information on translational acceleration, but provides a measure of rotational acceleration as well. The difference between the acceleration of the center accelerometer and one of the outstanding accelerometers, multiplied by the distance between them provides a measure of rotational acceleration [26]. This array was chosen because of the desire to reduce rotational movement in the neck, thus reducing rotational acceleration. We made a slight alteration to the patent's accelerometer array and used four triple axis accelerometers to make the purchasing process easier.

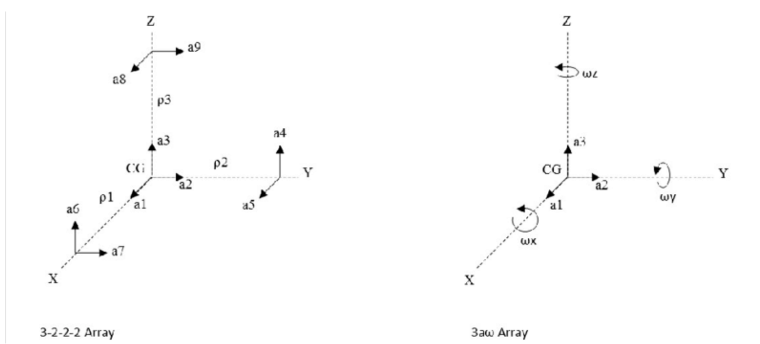
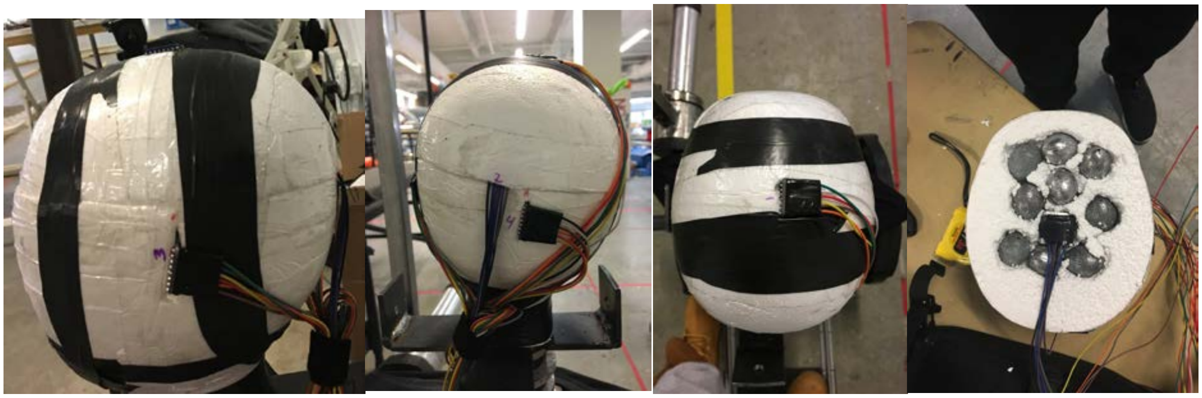


Figure 40: 3-2-2-2 Array of Accelerometers

The accelerometers were placed in the foam inside the helmet at the select locations, as seen in Figure 41. The Arduino used to convert the data into an Excel file was able to collect data at a maximum sampling rate of about 60 samples per second. This maximum sampling rate was used for all four accelerometers with the Arduino Uno, that was compatible to our computer program.



Figures 41: 4 Accelerometers placed in weighted Styrofoam head

6.0 Results and Analysis

6.1 Linear and Rotational Acceleration

The dummy was set at 0 degrees (straight on), 45 degrees and 90 degrees, as shown in Figures 42, 43, and 44, respectively. Three tests were completed per weight at each angle with and without the brace.

Figure 42: Testing Dummy at 0 Degrees

Figure 43: Testing Dummy at 45 Degrees

Figure 44: Testing Dummy at 90 Degrees

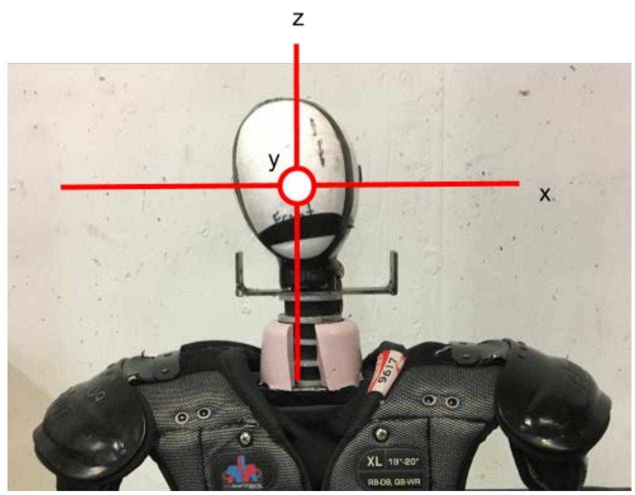


Figure 45: Constant X, Y, and Z Axes Used

Figure 45 displays the general axes that were defined for all accelerometers so that the data was consistent. Translational acceleration at 0 degrees was analyzed in the y-direction since straight-on impacts force backwards movement of the head. The accelerometers produced negative data points since the linear movement was backwards. These points were converted to positive values for visual appeal. Tests conducted from this angle with 5 and 10 pounds produced data that illustrated a reduction of translational acceleration in the y-direction. Figure 46 shows the average minimum acceleration from Accelerometer 1 with and without the brace at 5 and 10 pounds. Average minimum acceleration, in meters per second squared, was used for this analysis since the focal point was peak acceleration in the negative direction.

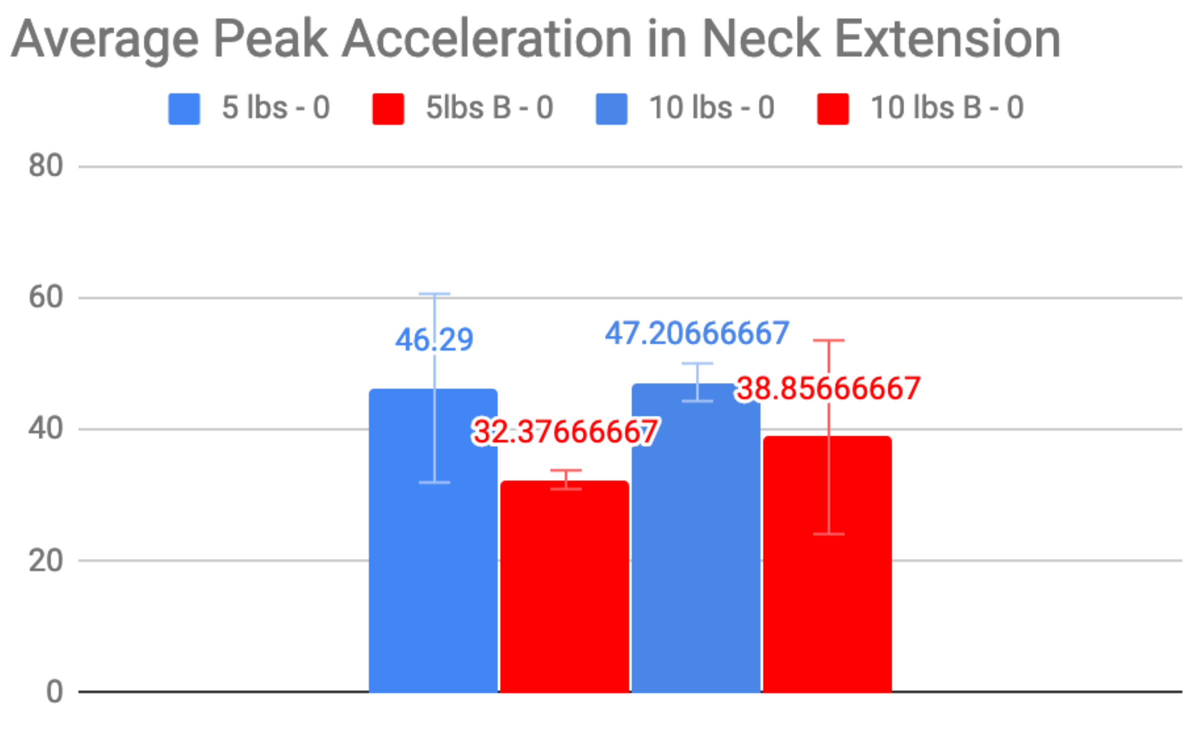


Figure 46: Average Peak Acceleration in Neck Extension Straight On (0 degrees)

Neck extension refers to the movement of the head backwards. The blue bars display the average minimum acceleration without the brace while the red bars show the average minimum acceleration with the brace on. Bars labeled with the letter "B" in the legend correlate with the red colored vertical bars and are in the same sequential order as displayed in the legend. Error bars were shown to exemplify the standard deviation of these tests. In the 5 pound tests, the neck support reduced translational acceleration by 13.91 m/s^2, a 30% reduction, and in the 10 pound tests, translational acceleration was reduced by 8.35 m/s^2, a 17.7% reduction. Data recorded at 12.5 pounds at this angle was disregarded, as Accelerometer 1 gave faulty readings during those three tests.

Tests done at 45 degrees showed markedly different results. Translational acceleration at 45 degrees was calculated using data in the y-direction multiplied by the cosine of 45 degrees to produce values of acceleration at a 45 degree angle. Figure 47 shows the average minimum acceleration in the y-direction for tests done at 45 degrees with 5 pounds, 10 pounds and 12.5 pounds.

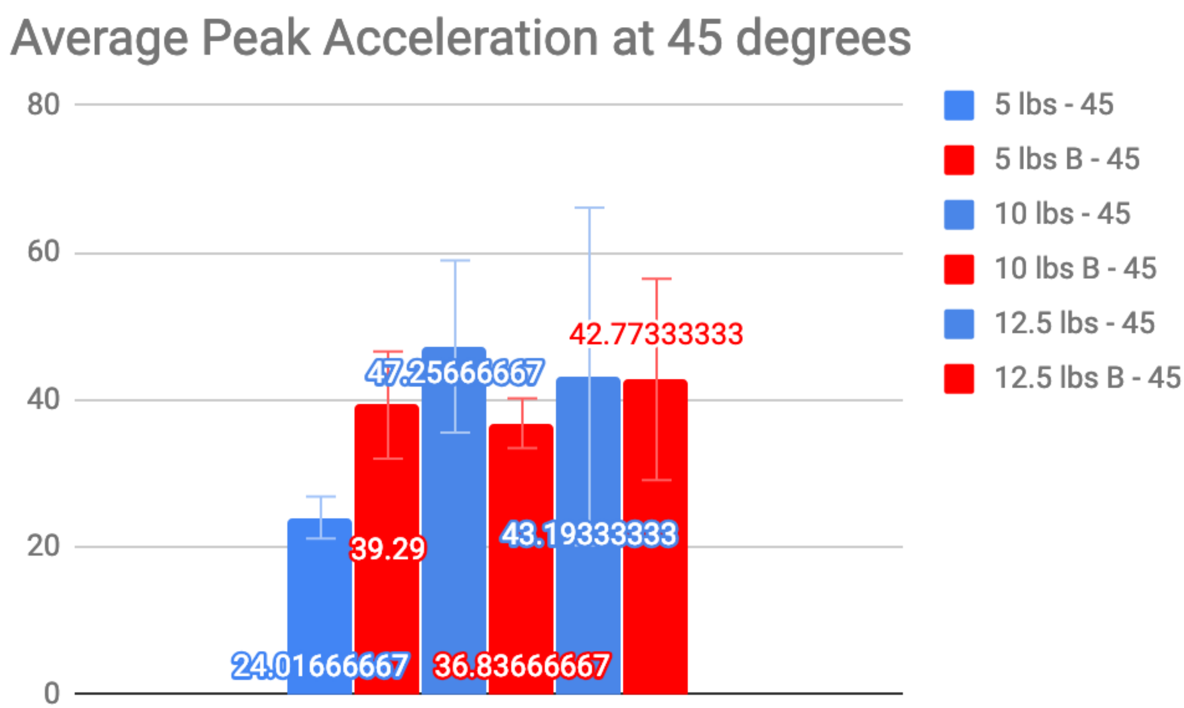


Figure 47: Average Peak Acceleration at 45 degrees

Minimum acceleration at Accelerometer 1 varies between these three tests. At 5 pounds, there was an increase of 15.27 m/s^2 or 63.6% when the brace was put on the test dummy. At 10 pounds, a decrease of 10.42 m/s^2 or 22.1% was encountered when the brace was applied and at 12.5 pounds, there was barely a decrease in acceleration at all.

There was also variation in results for tests done at 90 degrees. Accelerations in the x-direction were used for analysis at 90 degrees since the x axis produced data for lateral flexion. With 5 pounds and 12.5 pounds, the data showed that there was a reduction in average minimum acceleration, as seen in Figure 48, but with 10 pounds the data displayed the opposite. Acceleration in the 5 pound tests dropped from 36.41 m/s^2 to 24.92 m/s^2, a 31.6% decrease in lateral flexion but the standard deviation for the tests done at 5 pounds without the brace is too large to deduce anything from the data. Additionally, the standard deviation for tests done at 10 pounds with the brace is also too large to make any assumptions. Tests done at 12.5 pounds are credible, reducing translational acceleration by 64.4%, but it remains the only data at 90 degrees that shows a large reduction in translational acceleration with a low enough standard deviation.

A study done in 2012 analyzed translational acceleration in modern football helmets against older versions of helmets. The study concluded that modern day football helmets only reduce translational acceleration head-on by 7.3-14.0% [27] while our neck brace has shown to reduce translational acceleration up to 30% in flexion for moderate impacts head-on. Additionally, Guardian Athletics, producer of the Kato Collar, claims that their products reduce head deceleration by 30% after impact [13], which is comparable to our results.

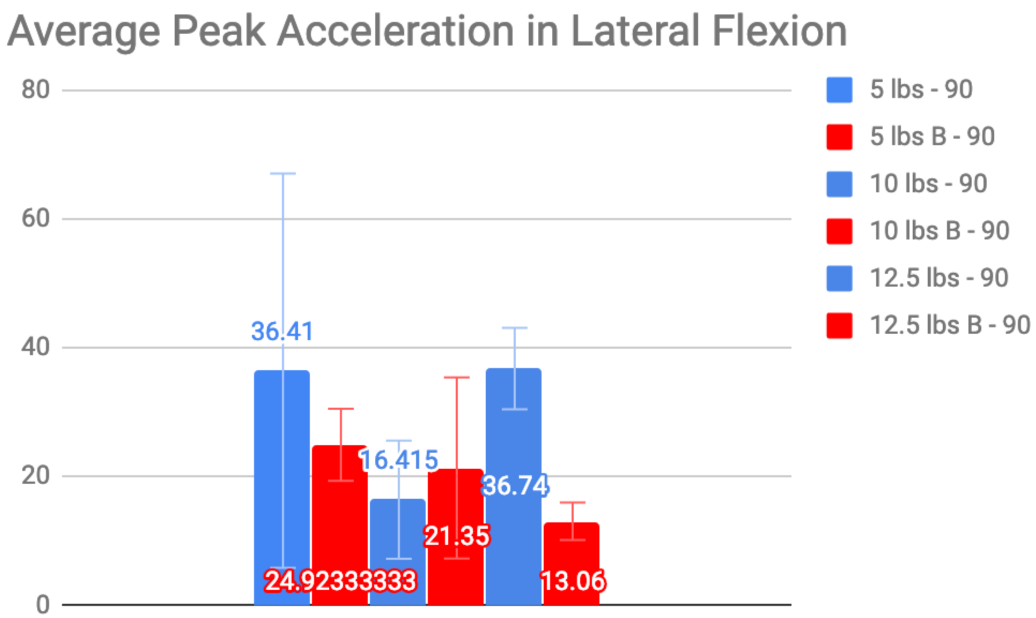


Figure 48: Average Peak Acceleration at 90 degrees

In addition to translational acceleration, rotational acceleration was also measured. By using the patent “Method and Apparatus for Testing Football Helmets” [26] (Pat.#US2004074283A1), rotational acceleration was calculated from the acceleration recorded by Accelerometer 2 (center accelerometer) and Accelerometer 3 (right ear accelerometer). The difference between both accelerometers multiplied by the distance between them, which was 3 inches or 0.0762 meters, provides a measure of rotational acceleration. The first set of tests that were measured were all tests done with 5 pounds at 45 degrees.

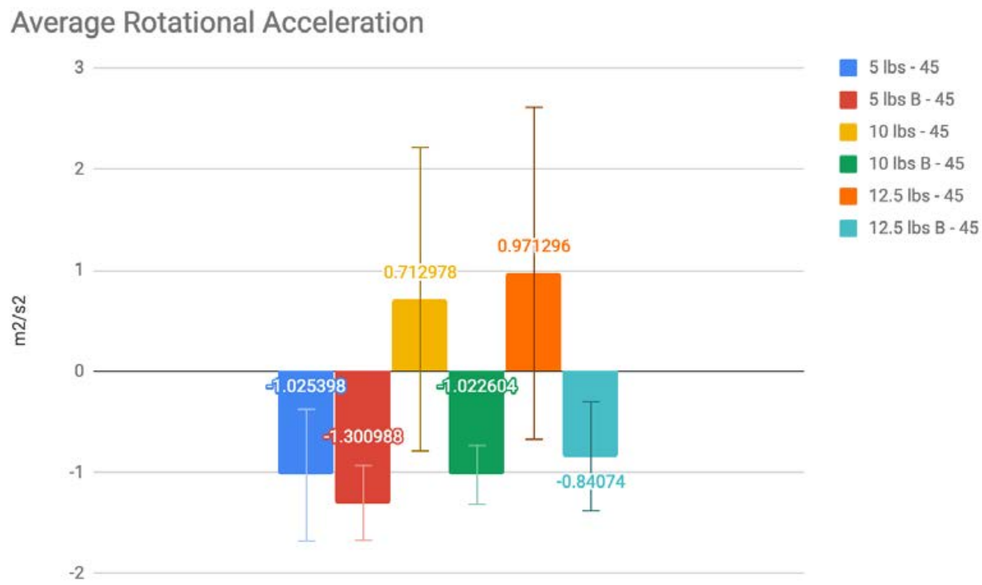


Figure 49: Average Rotational Acceleration at 45 degrees

Based on Figure 49, there was too much differentiation in rotational acceleration between all three tests with and without the brace. The figure displays the average rotational acceleration taken from the tests done at 45 degrees. Positive values refer to rotation in the clockwise direction while negative values refer to counterclockwise rotation. All units for rotational acceleration are displayed in m^2/s^2. There was a 21% increase in rotational acceleration for tests done at 5 pounds, while tests done at 10 and 12.5 pounds revealed a total change in the direction of the rotation, changing from clockwise without the brace to counterclockwise with the brace. Error bars are located on each vertical column to represent the standard deviation of these tests. Evidently, there was too much variation among these tests to make any conclusions.

In comparison, Figure 50 displays rotational acceleration calculated from tests done at 0 degrees. Again, the standard deviation of these tests was too large to deduce any conclusions. Additionally, averaging the rotational acceleration of each test results in another increase when the brace is applied. 5 pound tests revealed an increase of 44.6% and 10 pound tests revealed an increase of 85.3%. As previously stated, tests done at 12.5 pounds were not used in calculations because of the skewed data.

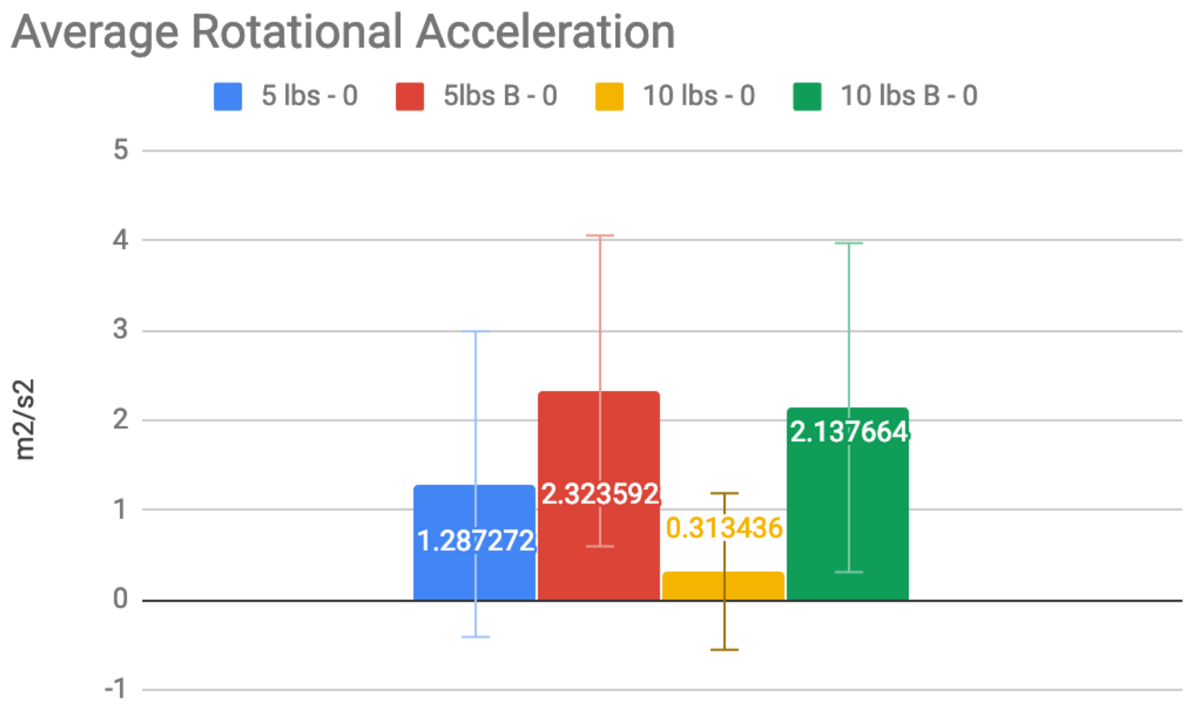


Figure 50: Average Rotational Acceleration Straight on (0 Degrees)

Lastly, rotational acceleration was also calculated for 90 degree tests, as displayed in Figure 51. Despite the variation in standard deviations, this remains the only set of tests that developed a particular trend. Even though the direction of the rotation is not consistent from test to test, there is an obvious reduction in the average rotational acceleration when the neck brace is applied. 5 pound tests revealed a 91.9% decrease in average rotational acceleration in the clockwise direction. 10 and 12.5 pound tests revealed a reduction in average counterclockwise rotational acceleration of 27.8% and 68.6%, respectively. Despite the lack of precision, results are comparable to the 2012 study that analyzes modern day football helmets, concluding a reduction of 8.4-15.9% in rotational acceleration against older football helmets [27]. Another noticeable element of the 90 degree tests is the increase in magnitude as the weight increases, making the results slightly more promising. Although the error bars for the 5 and 10 pound tests are quite large, the error bars for the 12.5 pound tests are small enough to infer that the neck brace reduces rotational acceleration during impacts at 90 degrees.

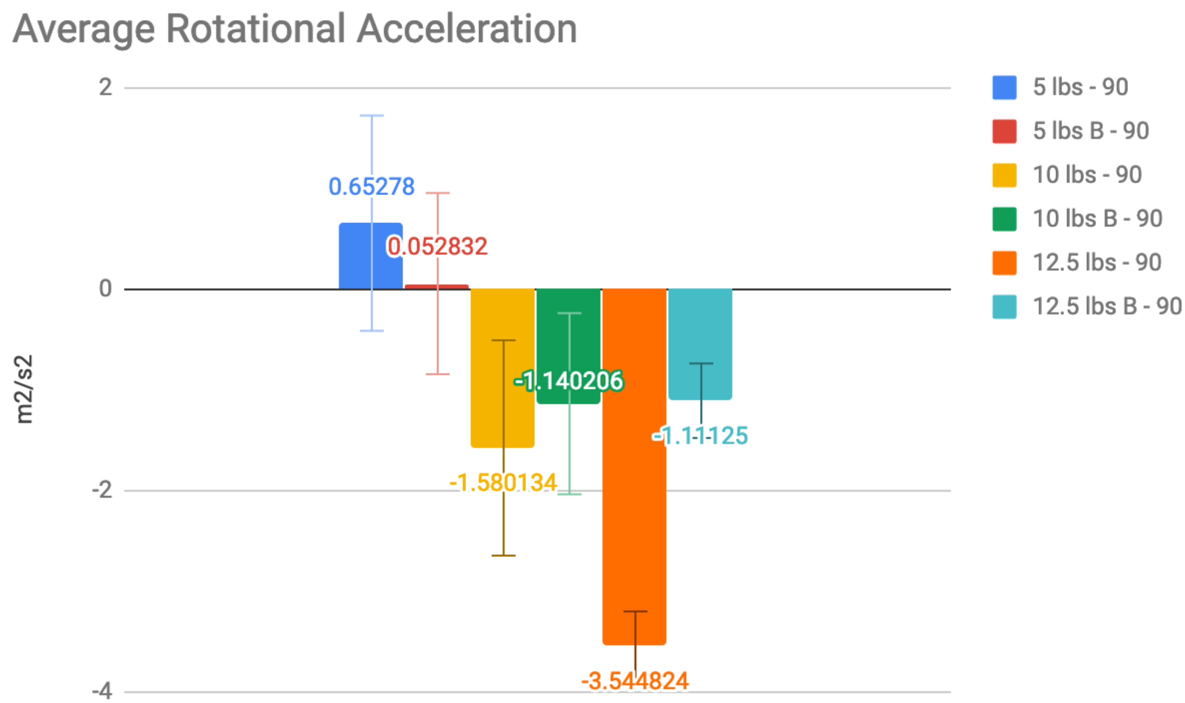


Figure 51: Average Rotational Acceleration at 90 degrees

Since there weren't any clear trends in rotational acceleration, the absolute value was taken from the results to focus primarily on magnitude rather than direction and magnitude. Results then indicated that there was a clear trend in reduction of rotational acceleration about the y-axis at 0 degrees, as shown in Figure 52. For 5 pound tests, there was an average reduction of 26.6% when the brace was applied. 10 pound tests revealed a 72.4% reduction in rotational acceleration and 12.5 pound tests revealed a 6.4% reduction in rotational acceleration. This happened to be the only set of tests that showed a clear reduction in the absolute value of rotational acceleration in tests at all weights. Calculations about the y-axis at 45 and 90 degrees did not show any trends in reduction. Similarly, calculations for 0, 45 and 90 degrees about the x and z-axis did not show any trends in rotational acceleration. In order to make the results of the impact tests relevant to other studies of the same subject, Head Injury Criterion and Head Impact Power was calculated.

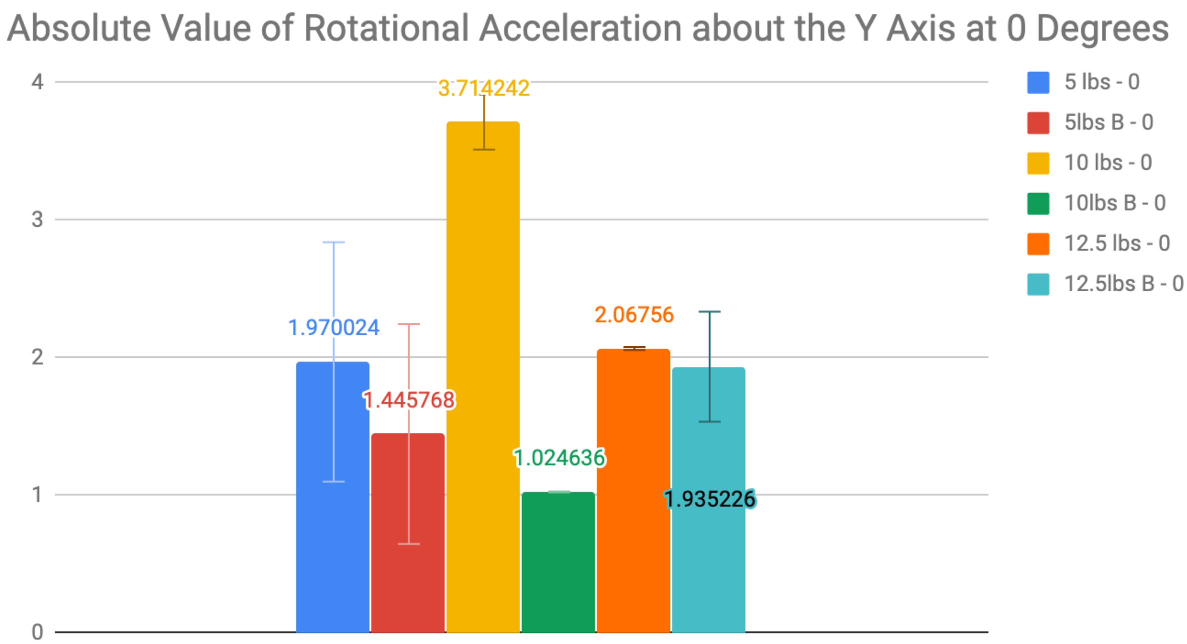


Figure 52: Absolute Value of Average Rotational Acceleration at 0 degrees

6.2 Head Injury Criterion and Head Impact Power

Values of acceleration with respect to time were used to evaluate HIC and HIP for all tests. Equations for HIC and HIP are located in Figure 53 and 54, respectively. Both equations are a measure of the likelihood for a concussion to occur, but HIP incorporates head mass, rotational acceleration and inertia in the x, y and z-directions.

$$HIC \equiv \left\{ \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} (t_2 - t_1) \right\}_{max}$$

Figure 53: SI Equation

$$HIP := m \cdot a_x(t) \cdot \int_{t_1}^{t2} a_x(t)\,dt + m \cdot a_y(t) \cdot \int_{t_1}^{t2} a_y(t)\,dt + m \cdot a_z(t) \cdot \int_{t_1}^{t_2} a_z(t)\,dt + I_x \cdot \alpha_x(t) \cdot \int_{t_1}^{t_2} \alpha_x(t)\,dt + I_y \cdot \alpha_y(t) \cdot \int_{t_1}^{t_2} \alpha_y(t)\,dt + I_z \cdot \alpha_z(t) \cdot \int_{t_1}^{t_2} \alpha_z(t)\,dt$$

Figure 54: HIC Equation

To simplify the calculation of HIC, a method described in the scholarly article, “The Head Injury Criterion (HIC) functional” [28], uses the equation shown in Figure 55 and is put into the equation found in Figure 56, where τ is t_2-t_1.

$$V = \int_{t_1^*}^{t_2^*} a(t)\,\mathrm{d}t.$$

Figure 55: HIC “Functional” Equation

$$\mathrm{HIC} = \left(\frac{V}{\tau}\right)^{0.5} \frac{V^2}{\tau},$$

Figure 56: Simplified HIC Equation

Values from Accelerometer 1 were used to keep the results consistent with the previous calculations for translational acceleration. Typically, HIC is calculated at 15 milliseconds from the time of impact but since the accelerometers could only produce data at about 60 samples per second, which is a sample every 16.6 milliseconds, we used values at about 17 seconds after impact.

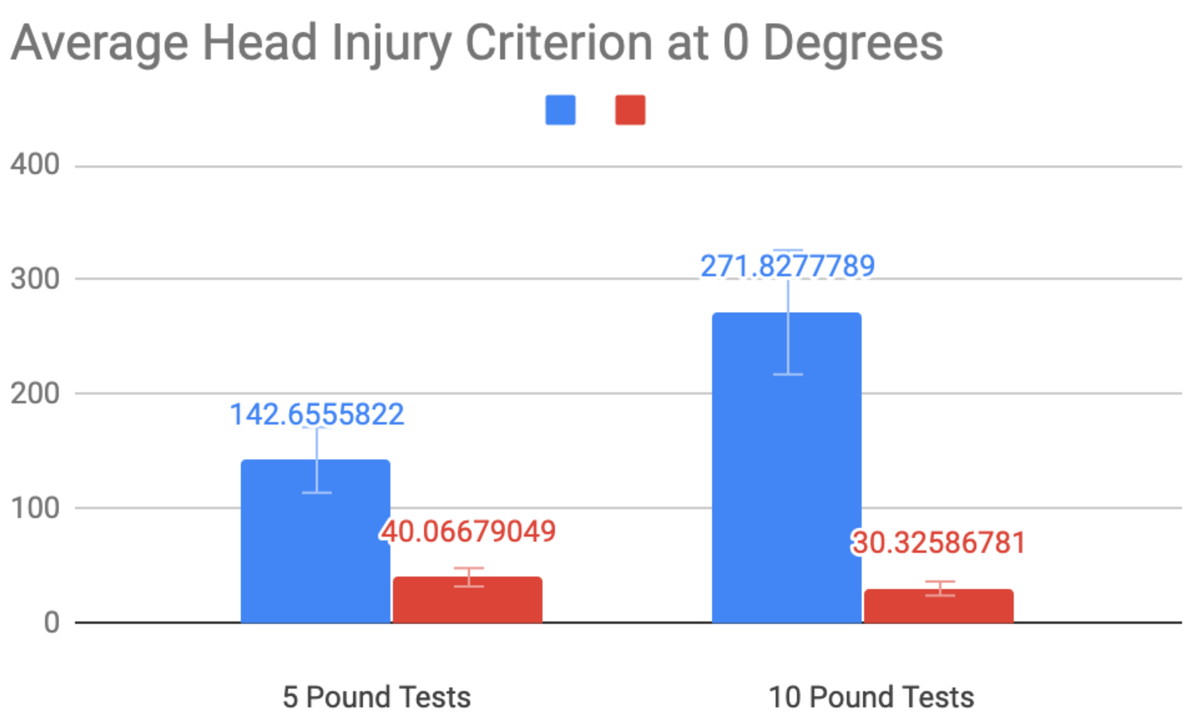


Figure 57: HIC at 0 degrees

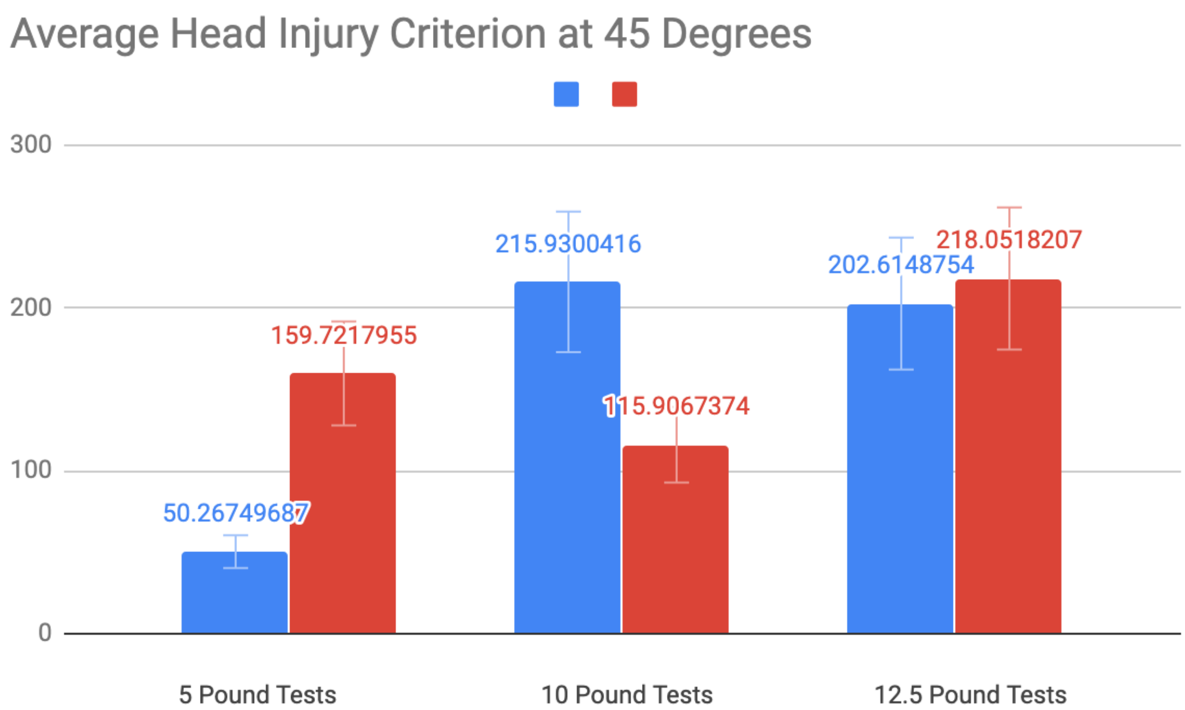


Figure 58: HIC at 45 degrees

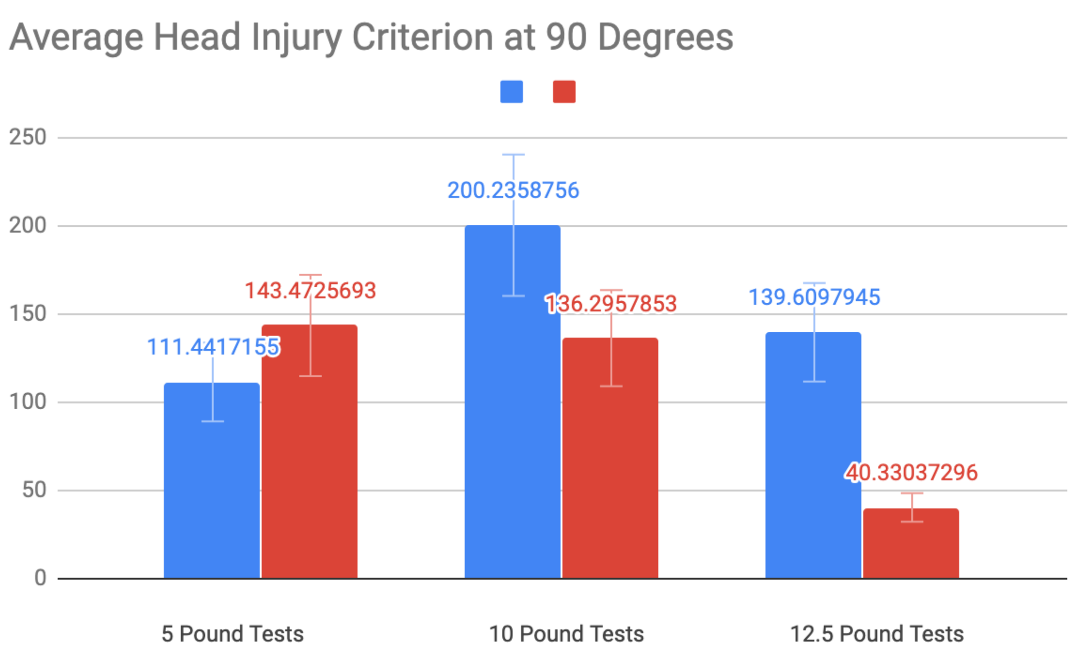


Figure 59: HIC at 90 degrees

In regards to HIC, there were trends in reduction in tests done at 0 degrees, as displayed in Figure 57. Bars in blue are tests done without the brace while bars in red are tests done with the brace. HIC was lowered by 102.589 (71.9%) in 5 pound tests and 241.502 (88.8%) in 10 pound tests. Figures 58 and 59 show average HIC at 45 and 90 degrees, respectively. The only decrease in average HIC at 45 degrees occurred during 10 pound tests, reducing the average HIC by 100.02 (46.3%). Decreases occurred in 10 and 12.5 pound tests at 90 degrees, resulting in a reduction of 63.94 (31.9%) and 99.28 (71.1%), respectively. As stated in Section 6.1, HIC values that exceed 1000 are detrimental to brain functionality. As shown in Figure 57-59, all HIC values remain under the threshold and do not exceed 466. The large difference in HIC leads to some skepticism, as a study done by WPI students in 2014 on reducing neck injuries showed to decrease HIC by only 39% [14]. Additionally, a study that analyzes modern helmet technology against older helmets concludes that newer helmets only reduce HIC by 14.6-21.9% [28].

Figures 60, 61 and 62 display the average HIP of tests done at 0 degrees, 45 degrees and 90 degrees, respectively.

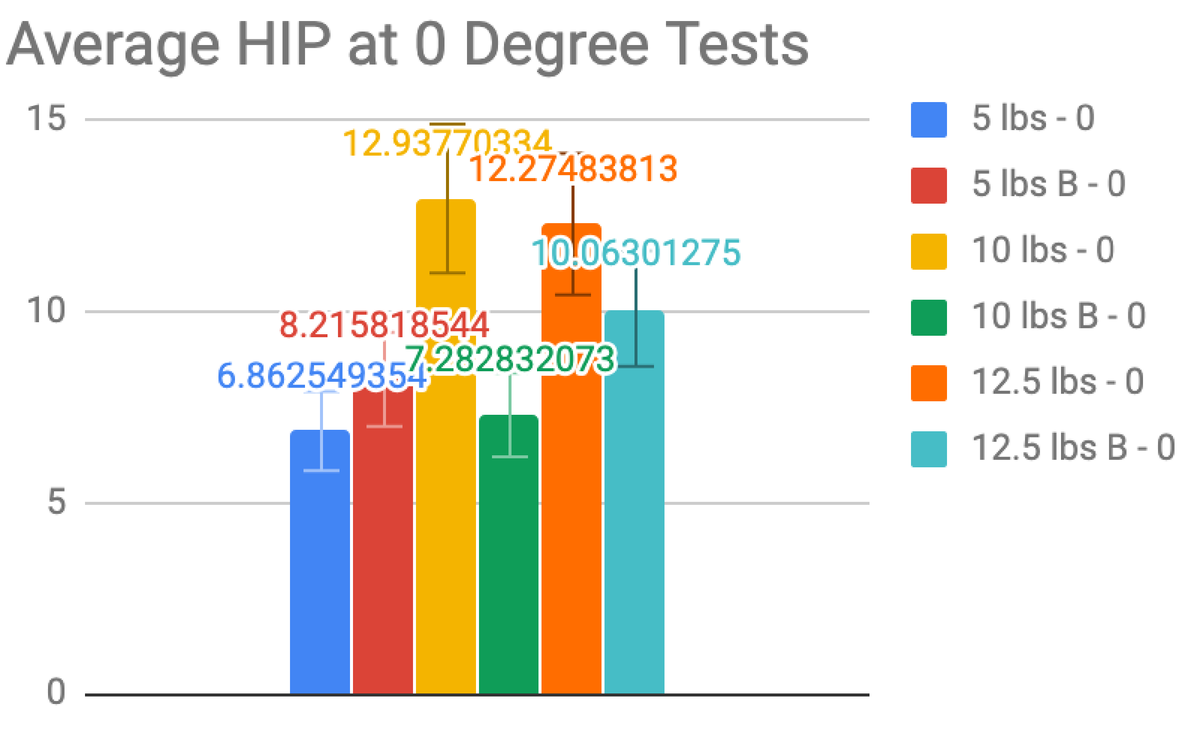


Figure 60: Average HIP at 0 Degree Tests

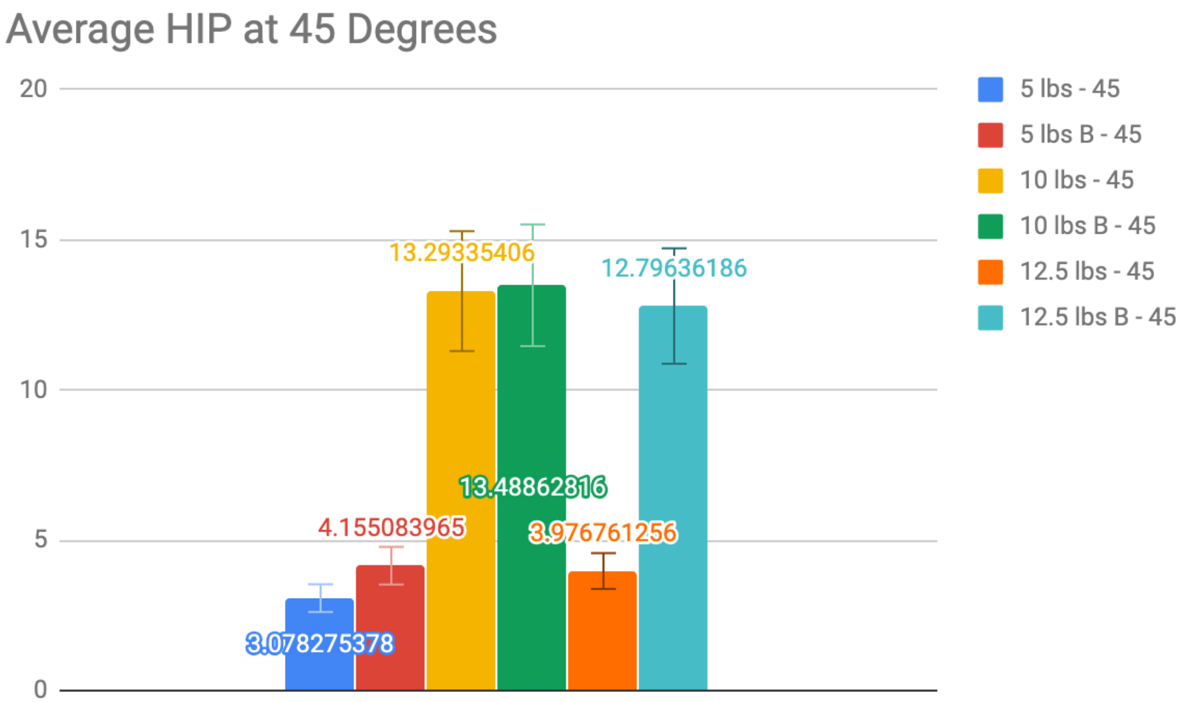


Figure 61: Average HIP at 45 Degree Tests

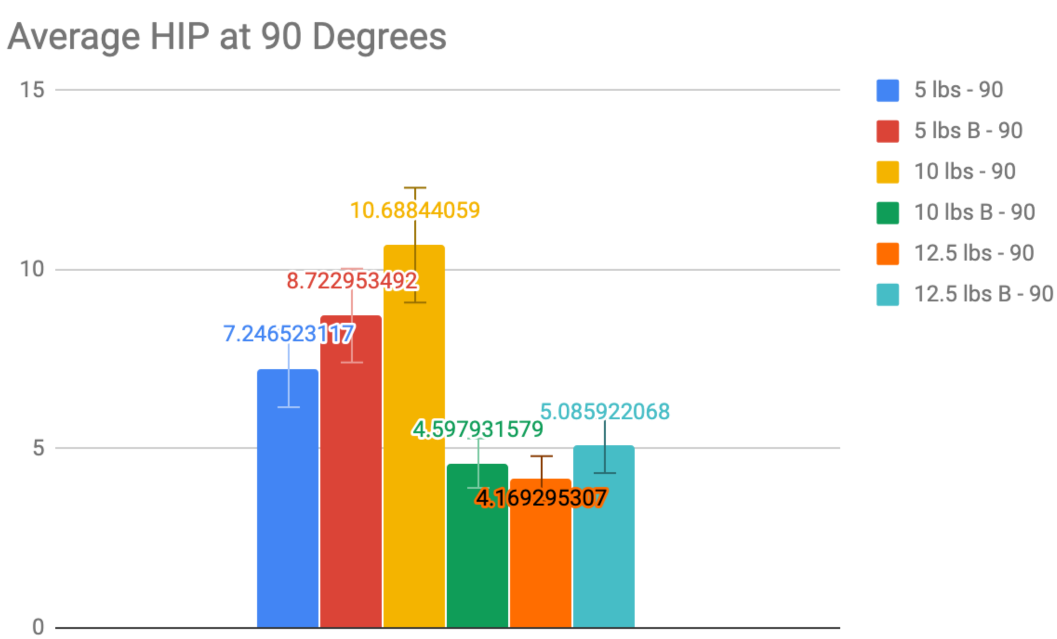


Figure 62: Average HIP at 90 Degree Tests

The HIP equation was developed and tested over time. Newman et al. first discovered the equation and determined that an HIP 12.79 kilowatts corresponded to a 50% chance of concussion and an HIP of 20.88 kilowatts corresponded to a 95% chance that a concussion occurred [29]. Marjoux et al. conducted a more recent study that concluded an HIP of 24 kilowatts corresponded to a 50% change of concussion risk and an HIP of 30 kilowatts corresponded to a 95% change of concussion risk [20]. In both cases, the average HIP of tests done in this study remain below the 95% threshold. To remain consistent, this study refers to values concluded by Newman et al. [29].

Average HIP was determined to be about 12.94 kilowatts in tests done without the brace at 10 pounds and 0 degrees, resulting in greater than 50% chance of a concussion since it is above the 12.79 kW threshold developed by Newman et al. [29]. When the neck brace was applied during the same tests, HIP was decreased to about 7.28 kW.

Although below the 50% threshold of 12.79 kW, HIP was also reduced when the neck brace was applied during tests with 12.5 pounds at 0 degrees, as seen in Figure 60. HIP decreased by about 2.21 kW, making the possibility of a concussion from impact even lower. Unfortunately, HIP results for tests done at 45 degrees did not reveal any reduction when the neck brace was applied. Even with the brace, as shown in Figure 61, average HIP is still 13.49 kW for tests with 10 pounds at 45 degrees. Therefore, we cannot deduce anything from impacts occurring at 45 degrees. Even though the testing dummy was rotated 45 degrees, values of acceleration were taken from the same axes as if the dummy was set at 0 degrees. The same thing was done at 90 degrees. Calculations were done with the same axes as if the dummy was set at 0 degrees. This kept all HIP values consistent with each other.

On the other hand, average HIP with the neck brace (4.60 kW) is calculated to be lower than the average HIP without the neck brace (10.69 kW) for tests done with 10 pounds at 90 degrees (Figure 62). This is a decrease of 6.09 kW. Results for tests with other weights at 90 degrees were shown to increase when the neck brace was applied. There is not a reason to believe that the neck brace increases HIP at 90 degrees due to the variability in accelerometer outputs, as described in Section 8.1.

7.0 Discussion

As shown in the results section, the neck brace proved effective against straight-on hits only, reducing linear acceleration up to 30%. In the field of play, this would be comparable to the traditional cowboy collar which reduces acceleration by 10% [30]. There was no noticeable linear acceleration reduction in the 45 and 90-degree impacts, nor was there a consistent reduction in rotational acceleration in any of the tests. HIC results from tests at 0 degrees, or straight on, display a reduction in HIC values once the brace was on. The large variation in HIC calculations and testing, reveals that more conclusive evidence is needed. Although the brace did not work as expected, this new design can be adjusted further and tested again in the future.

Results indicated that placing an adjustable protective device behind the neck would minimize linear head acceleration for head on collisions. However, rotational acceleration would not be reduced. Further research and analysis is needed to design a product that would minimize this kind of acceleration without restricting the athlete's side to side range of motion.

Multiple surveys (Figure 1) conducted among WPI football players revealed that the prototype has potential for future applications in football. Survey responses indicated that the neck brace was comfortable and could be used routinely during practices and games. There was no change in time it took a player to put on shoulder pads. We did learn, however, that some players thought the style of the design was poor. The prototype needs to be redesigned to make it more aesthetically appealing. Other results indicated that WPI football players would prefer multiple colors to match the uniform of the team.

	Avg.
Comfortability	4.1
Initial Ease of Use	4.3
2nd try Ease of use	4.9
Style	3.6
Functionality	4.3

Figure 63: Survey Results

Overall, the project was a success because we were able to design and build a testing rig that simulates the impact of a live football hit. The design of the testing rig allowed us to properly use a patent for testing football helmets [26] (Pat.#US2004074283A1), providing us with a credible way to measure linear and rotational acceleration. The neck support worked in the head-on direction, however a redesign would be needed to create a brace with a better looking style and the ability to reduce linear acceleration from side impacts and rotational acceleration from all angles. By furthering this project, a more effective brace could be made with the focus on reducing rotational acceleration. The research behind concussions is endless and there has yet to be a device created that can completely prevent concussions. The transition to applications that strengthen the neck are promising for future testing in concussions.

8.0 Suggestions

After testing was completed, some issues in the rig became apparent; these could have been improved with more time and money. For instance the original design of the testing rig didn't have additional bars to hold the weights. Over time, the increased stresses on the bars and repeated hits from the side caused bending of the T-bracket on the bottom of the test dummy. The bracket (Figure 61) was made with a 3/16 inch thick steel sheet. Ideally, it should be updated to a 5/16 inch thick plate to reduce the amount of flexibility in the bracket.

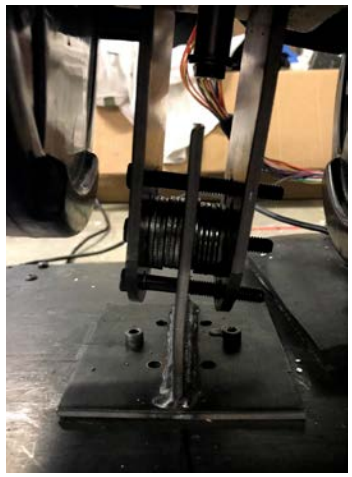


Figure 64: Bent T-Bracket at the base of the Testing Dummy

Another aspect that could have been improved was the swing arm itself. There were some slight bends after welding the "ball" to attach to the arm. This caused the swing arm to drop at an angle, and not straight, so the location of the test dummy had to be moved to adjust for the angle. This can be corrected by welding the "ball" again to drop directly at the center of the face.

The tensioning device could also be improved with a slight redesign. It worked properly, but we had to make a custom clamp to squeeze the steel cable at each end. This proved to be a problem because once the cable was tensioned to 5.4 foot-pounds there wasn't enough friction to hold the cable in place. This meant that the tensioning cable had to be checked for proper tension after every few tests. To avoid this happening again a high strength rope could be used to tension the neck. A rope could be tied at both ends and would reduce the slip that occurred from the low friction steel cable.

Another spot identified for improvement was the test dummy. A Styrofoam head filled with fishing weights was used to get a total weight of 8lbs. However, after periodic testing the Styrofoam began to deteriorate. This meant that the head was only able to last for a limited number of tests. Given more time to complete the project, a head could have been 3D printed or molded with a high density rubber with spots to place weight inside. This head could also have a precise mounting point for various accelerometers to test the movement and acceleration.

In addition to the improvements mentioned, there were also some errors that may have caused issues throughout the testing procedure. Human error and the sampling rate of the accelerometers stand out as underlying possibilities for the testing inconsistencies. The pendulum was held upright by someone standing on a ladder and two other people verified that the pendulum was both straight and raised to 90 degrees. This was done visually and could have been more consistent if there was a mechanical switch attached so that the pendulum dropped from the same exact position every test. Accelerometer sampling rate was a significant issue as the Arduino could only record up to 60 samples per second. Impacts of this caliber happen within milliseconds. A sampling rate of only 60 samples per second was likely to slow to record the accelerations during impact. Another improvement for future projects is to purchase higher quality accelerometers to get a higher sampling rate for more accuracy.

Another possible issue relates to manufacturing errors. An example would be the possibility that the upper crossbar of the testing rig was welded at a slight angle, causing the pendulum to swing down at a different angle each test as opposed to a consistent impact target.

9.0 Conclusion

The goal of this project was to create an effective and comfortable neck support that can be incorporated into shoulder pads to reduce the risk of concussions and the development of CTE in football. As a result, we designed a neck brace with Poron XRD® foam that wrapped around the neck, leaving the throat exposed for comfort and breathability. The foam was encased with a fabric made of polyester and elastane that attached to a sheet of nylon, in turn attached to the shoulder pads. Accelerometer testing revealed that the neck brace reduced linear/translational acceleration only when the impact is directly at 0 degrees to the helmet. Reductions were 30% and 17.7% in linear acceleration. Tests at 45 and 90 degrees did not display any trends in the linear acceleration data to make any conclusions. Additionally, there were no trends in rotational acceleration that were clear enough to make conclusions.

Another conclusion can be made relating to Neck Support Design 2. The procedure for mixing silica and PEG was ineffective in producing a shear thickening fluid. This may be the fault of the equipment that was used, as there wasn't a vacuum sealed room available that could extract all the air bubbles out of the mixture. Regardless, the procedure was ineffective and should be avoided in future projects. Despite the inability to provide concrete conclusions other than a reduction in linear acceleration of direct impacts in the y-direction, the results are still significant as they provide insight for future work.

The original project objective of building a testing rig that is able to test head movement in both linear and rotational directions was a sucess. However, our other objective of creating a neck support that reduces linear and rotational accelerations of the head when impact occurs was only partially successful, as it was inconclusive as to whether it helps reduce rotation. Using the knowledge gained from this project and the already existing testing rig, future work could be done to redesign the neck brace and make the necessary improvements to the testing rig and testing procedure.

References

1.M. Faul, L. Xu, M. M. Wald, and V. G. Coronado, “Traumatic Brain Injury in the United States: Emergency Department Visits, Hospitalizations, and Deaths 2002-2006,” US Department of Health and Human Services CDC Mar. 2010.

2.“2017 Injury Data.” [Online]. Available: https://www.playsmartplaysafe.com/newsroom/reports/2017-injury-data/. [Accessed: 25-Sep-2018].

3.Collins, C. L., Fletcher, E. N., Fields, S. K., Kluchurosky, L., Rohrkemper, M. K., Comstock, R. D., and Cantu, R. C., 2014, "Neck Strength: A Protective Factor Reducing Risk for Concussion in High School Sports," J. Primary Prevent, pp. 309-319.

4.S. P. Broglio, M. McCrea, and T. McAllister, “A National Study on the Effects of Concussion in Collegiate Athletes and US Military Service Academy Members: The NCAA–DoD Concussion Assessment, Research and Education (CARE) Consortium Structure and Methods,” 2017.

5.W. C. Medicine, “What is a Concussion.” [Online]. Available: https://concussion.weillcornell.org/about-concussions/what-concussion. [Accessed: 10/5/18].

6.C. A. Robbins et al., “Self-reported concussion history: impact of providing a definition of concussion,” vol. 5, pp. 99–103, 2014.

7.D. N. Martini and S. P. Broglio, “Long-term effects of sport concussion on cognitive and motor performance: A review,” vol. 132, pp. 25–30, 2018.

8.“The 2018 Injury Reduction Plan: Initiatives to Advance Player Health and Safety,” 07-Aug-2018. [Online]. Available: https://www.playsmartplaysafe.com/newsroom/videos/2018-injury-reduction-plan-initiatives-advance-player-health-safety/. [Accessed: 25-Sep-2018].

9.D. P. Kerr, “The Kerr Collar;,” 2012. [Online]. Available: http://www.kerrsports.com.

10.D. N. Ettinger, “Protective Neck Collar.” Available: patentimages.storage.googleapis.com Patent Number:3855631

11.R. Wilkerson, “Burners and Stingers - OrthoInfo - AAOS.” [Online]. Available: https://www.orthoinfo.org/en/diseases--conditions/burners-and-stingers/.[Accessed: 02-Apr-2019].

12.“Butterfly Restrictor;” [Online]. Available: https://www.sportsunlimitedinc.com/douglas-dp-series-football-butterfly-Restrictor.html.

13.“Kato Collar.” [Online]. Available: https://guardianathletics.com/about/.

14.T. Hanna, Z. Hennings, M. Mensing, E. Perry, and R. Rookey, “Design of a Protective Device for Head and Neck Injuries in Football,” May 2014.

15.R. S. Aouf, “Halo smart collar designed to protect athletes from concussion,” 2018. [Online]. Available: https://www.dezeen.com/2018/08/15/halo-aexos-smart-collar-athletes-concussion-sports-technology/. [Accessed: 25-Sep-2018].

16.J. Ding, W. Li, and S. Z. Shen, “Research and Applications of Shear Thickening Fluids,” Recent Patents on Materials Science vol. 4, no. 1, pp. 43-49, Jan. 2011.

17.“XRD®.” [Online]. Available: http://www.xrd.tech/howitworks/index.aspx. [Accessed: 25-Sep-2018].

18.C. L. Collins et al., “Neck Strength: A Protective Factor Reducing Risk for Concussion in High School Sports,” The Journal of Primary Prevention vol. 35, no. 5, pp. 309–319, Jun. 2014.

19.S. Rowsonet al., “Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion,” Annals of Biomedical Engineering vol. 40, no. 1, pp. 1–13, Jan. 2012.

20.Marjoux, D., Baumgartner, D., Deck, C., and Willinger, R., 2008, "Head injury prediction capability of the HIC, HIP, SIMon, and ULP criteria," Accident Analysis & Prevention, 40(3), pp. 1135-1148.

21.V. T. H. Lab, “Adult Football STAR Methodology.” [Online]. Available:vtechworks.lib.vt.edu

22.A. Tozeren, Human Body Dynamics: Classical Mechanics and Human Movement. Springer Science & Business Media, 1999
23.NFL, “Luke Kueckly Statistics.” Available: NFL.com
24.H. I. Solutions, “Brand Harmonized Parts Catalog.” Available: www.humaneticsatd.com
25.R. Joselin and W. Wilson, “Investigation on Impact Strength Properties of Kevlar Fabric using Different Shear Thickening Fluid Composition,” Defense Science Journal vol. 64, no. 3. Defense Scientific Information & Documentation Centre, New Delhi, pp. 236–243, May 2014
26.B. T. D and W. C. R.P, “Method and apparatus for testing football helmets,” Available: worldwide.espacenet.com
27.D. Viano, C. Withnall, and D. Halstead, “Impact Performance of Modern Football Helmets,” Annals of Biomedical Engineering vol. 40, no. 1, pp. 160–174, Jan. 2012.
28.J. Hutchinson, M. J. Kaiser, and H. M. Lankarani, “The Head Injury Criterion (HIC) functional,” Applied Mathematics and Computation vol. 96, no. 1, pp. 1–16, 1998.
29.Newman, J., Barr, C., Beusenberg, M., Fournier, E., Shewchenko, N., Welbourne, E., and Withnall, C., "A New Biomechanical Assessment of Mild Traumatic Brain InjuryPart 2-Results and Conclusions", Biokinetics and Associates Ltd., Ottawa, Ontario, Canada. Sept 1999.
30.S. Rowson and S. M. Duma, “Evaluation of the Load Limiting Capabilities of Football Neck Collars Using Anthropomorphic Test Devices,” p. 4. Available: ibrc.osu.edu

Appendices

Appendix A: Properties of Poron XRD®

PORON® Urethane Foams

PORON® XRD™ Extreme Impact Protection – Physical Properties

PROPERTY	TEST METHOD	PRODUCT				
***Density**, lb./ft³	ASTM D 3574-95 Test A	9	12	15	20	25
Specific Gravity		0.14	0.19	0.24	0.32	0.40
Tolerance, %		± 10				
***Standard Thickness**		See Product Availability				
Tolerance, %		± 10				
Standard Color		65 - Vivid Yellow				
Air Permeability	Internal using Gurley Densometer	Open Cell - Breathable				
***Compression Set**, % max.	ASTM D 3574 Test D @ 158°F (70°C)	< 10				
***Compression Force Deflection**, psi, kPa)	0.2"/min. Strain Rate Force Measured @ 25% Deflection	1.1 - 3.4 (8 - 23)	1.5 – 5.5 (10 - 38)	4 – 9 (28 – 62)	5 – 12 (34 – 83)	10 – 20 (69 – 138)
Hardness, Durometer	Shore "O"	10	19	32	**	
Hydrolysis Resistance, Compression Set, % Max	ASTM D 3574 Test J / Test D after autoclaved 5 hrs @ 250°F (121°C)	**				
Resilience, Shore Instrument Resiliometer, avg (Ball Rebound Tester)	ASTM D 2632-96, Vertical Rebound	**				
Water Vapor Transfer, Typical g/ft²/24hrs (g/m²/24hrs)	Sample Thickness, inches (mm)	0.158 (4.0)	0.118 (3.0)	0.118 (3.0)	**	
	Based on ASTM E96-00 Upright / 37°C / 0% RH	4150	3400	3100	**	
	Leakage – Inverted	Yes	Yes	Yes	**	
Water Absorption, % Wt Gain	Based on ASTM D 570 – 2hr water immersion @ room temperature	Typical Value 10				
Skin Contact	Primary Skin Irritation – FHSA. Based on ISO 10993-10 (2002), ISO 10993-12 (2007), ISO/IEC 17025 (2005)	Negligible Irritant. Primary Irritation Index = 0				
Tear Strength, pli, min. (kN/m)	ASTM D 624 Die C	4.5 (0.8)	5 (0.9)	5 (0.9)	10 (1.8)	14 (2.5)
***Tensile Elongation**, % min.	ASTM D 3574 Test E	> 145				
***Tensile Strength**, psi, min. (kPa)	ASTM D 3574 Test E	30 (207)	45 (310)	70 (483)	100 (689)	140 (965)
Restricted Substances Compliance	Based on Adidas-Salomon policy for control and monitoring of hazardous substances.	Pass				
Chemical Resistance		*PORON Cushioning Materials are unaffected by mild organic acids and bases. They show modest swelling with oils and greases and other linear hydrocarbons. Strongly polar solvents will greatly swell PORON Materials. In most cases, physical properties recover to a great extent as the solvents evaporate.*				

Notes:
1. All metric conversions are approximate.
2. Additional technical services are available.
3. Information listed based on typical physical properties.
4. * Standard testing property; Certificate of Compliance available per lot.
5. ** Indicates testing in progress to confirm reported results.

The information contained in this Data Sheet is intended to assist you in designing with Rogers' PORON XRD Extreme Impact Protection and should not be used to create a specification. The data expressed is not intended to and does not create any warranties, express or implied, including any warranty of merchantability or fitness for a particular purpose or that the results described or shown on the Data Sheet will be achieved by a user for a particular purpose. Each user must develop its own design and should determine the suitability of Rogers' products for that design.

WARNING: No impact absorbing material can prevent all injuries that may occur when the body is subjected to impact. Rogers makes no representation or warranty that PORON XRD Extreme Impact Protection will prevent such injuries. The user of protective gear containing Rogers' materials should be aware of the limitations of the gear and should exercise reasonable care and caution in the undertaking of activities that may result in impact to the body.

Appendix B: Survey Form and Questions

Informed Consent Agreement for Participation in a Research Study
Investigator(s): Zach Bellion, Sean Gillis, Francis Lubega, Blayne Merchant, Collin Saunders
Contact Information:
Email Alias: gr-ins-mqp@wpi.edu
Saunders Phone Number: 201[redacted]
Title of Research Study: Innovative Neck Support
Sponsor: Fiona Levey

Introduction: You are being asked to participate in a research study. Before you agree, however, you must be fully informed about the purpose of the study, the procedures to be followed, and any benefits, risks or discomfort that you may experience as a result of your participation. This form presents information about the study so that you may make a fully informed decision regarding your participation.
Purpose of the study: The purpose of this study is to gather opinions of football players about the comfortability and user-friendliness of our neck brace. We are trying to see whether or not our prototype is something that can be worn comfortably during gameplay.
Procedures to be followed:

- The participants of this study will be needed for approximately 10 minutes to try on the neck brace and answer questions.
- First, we will investigate the user-friendliness of the prototype. Subjects will be given 1 minute to try on neck brace without any knowledge of its application.
- Information about the application of the brace will be given to the participant. They will then try on the brace a second time and will only be given 15 seconds.
- After trying on the prototype, participants will be asked to move around in motions similar to that of football gameplay. Comfortability will be measured on a scale from 1-5, 1 being the least comfortable and 5 being the most comfortable. Measurements are relative to the comfortability of football equipment without the neck brace.

Time constraints are given to account for practicality of its application before practices and games

Risks to study participants: There are no risks associated with this study. The neck brace is made of materials like nylon, a fabric made of polyester and elastane, and Poron XRD® foam. The foam does not come in contact with the skin. Metal grommets are also used to attach the brace to the shoulder pads. These grommets are not sharp and do not come in contact with the skin. The neck brace is designed so that it does not restrict respiration.
Benefits to research participants and others: There are no benefits for the participants of this study.

Record keeping and confidentiality: No interview subject will be required to provide their name, for anonymity is guaranteed. Each subject will be given a code name to maintain confidentiality. Responses will be indicated on a number scale. There will be a number of categories and subjects will be asked to rank the neck support based on how they think it measured up. Only the MQP team members and the advisor, Fiona Levey, will have the access to the raw data from the interviews. Data and information from these interviews will be recorded on a Google Document and will be generalized for report purposes.
Compensation or treatment in the event of injury: Our interviews and surveys will not involve minimal risk of injury or harm. You do not give up any of your legal rights by signing this statement.

For more information about this research or about the rights of research participants, or in case of research-related injury, contact: See information provided at the top of page 1 or contact IRB Chair (Professor Kent Rissmiller, Tel. 508-831-5019, Email: kjr@wpi.edu) and the Human Protection Administrator (Gabriel Johnson, Tel. 508-831-4989, Email: gjohnson@wpi.edu.)

Your participation in this research is voluntary. Your refusal to participate will not result in any penalty to you or any loss of benefits to which you may otherwise be entitled. You may decide to stop participating in the research at any time without penalty or loss of other benefits. The project investigators retain the right to cancel or postpone the experimental procedures at any time they see fit. By signing below, you acknowledge that you have been informed about and consent to be a participant in the study described above. Make sure that your questions are answered to your satisfaction before signing. You are entitled to retain a copy of this consent agreement.

___________________________ Date: ___________________
Study Participant Signature

____
Study Participant Name (Please print)

____________________________________ Date: ___________________
Signature of Person who explained this study

MQP Neck Brace Survey

The participants of this study will be needed for approximately 10 minutes to try on the neck brace and answer questions. First, we will investigate the user-friendliness of the prototype. Subjects will be given 1 minute to try on neck brace without any knowledge of its application. We will then explain how the brace works.
The participant will then try the brace on again.
After trying on the prototype, please move around in motions similar to that of football gameplay. Characteristics are to be **measured on a scale from 1-5, 1 being the least comfortable and 5 being the most comfortable**. These measurements are relative to the comfortability of football equipment without the neck brace.

Survey Participant (sign) :______________________________

Comfortability: 1 2 3 4 5

Initial Ease of Use: 1 2 3 4 5

2nd Try Ease of Use: 1 2 3 4 5

Style: 1 2 3 4 5

Functionality: 1 2 3 4 5

Do you see any improvements to note?

Did the brace make your neck feel overly restricted?

How would wearing this brace affect you getting ready to practice or play in a game?