Examining Head Injury: The Efficiency and Longevity of a Modern Lacrosse Helmet

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Plastic-shelled helmets have drastically reduced the amount of "serious" injuries in contact sports. Researchers argue over the efficacy of helmets' ability to prevent mild traumatic brain injury. In this experiment, three different head models were compared: a skull model representing a control, a used helmet, and a brand-new helmet. Null hypotheses were formulated that the head models would display no difference in their performance when measuring linear acceleration, resultant acceleration, and length of time that energy causes the brain to reverberate. The null hypothesis was rejected at most locations because there was determined to be a significant protection from impact in both helmet models over the skull model. A null hypothesis was formed that there would be no difference in the head models' performances over time. The brand new helmet was predicted to decrease in performance, but both the used helmet and skull control were predicted to maintain their efficiency. The null hypothesis was rejected, and two of the alternate hypotheses were rejected as well. In each head model, a significant decline in performance was observed over time.

INTRODUCTION

In recent decades, the role of helmets in preventing concussion has become a more prevalent issue. Concussion typically occurs at a lower impact than NOCSAE requires of helmets for certification. Thus, helmets themselves may not prevent concussion at all. There is no concrete threshold in accelerations due to gravity (gs) at which concussions occur. Usual estimates for a minimum impact to cause a concussion range wildly from 60gs to 120gs. To put that number in perspective, fighter pilots usually experience about 10gs when performing a roll [34]. An extreme impact from an NFL player can generate over a half ton of force, about 150 gs [34]. According to General Motors and NASCAR injury biomechanics researchers, this amount of force can be sustained by the human body if the force is spread out over a large area; if the force is localized to one location the player being hit can experience massive damage [34].

Concussions may not be the result of one single impact. A recent study conducted at Purdue University suggests that brain injury actually results from an accumulation of impacts [28]. In their two-year study following a high-school football team, they found that 17 of 45 players exhibited brain damage after undergoing a functional MRI (fMRI) [28]. Only six of those players reported a concussion, meaning that only six reported a massive blow to the head that they thought was symptomatic. The fMRI study was paired with impact-sensor data from each player's helmet and neurocognitive tests. Researchers reported a strong correlation between the number of hits to the head that the player sustained and the changes in imaging and neurocognitive function during tests. Eric Nauman, an expert in central nervous system and musculoskeletal trauma, noted, "[t]he one hit that brought on the concussion is arguably the straw that broke the camel's back" [28]. The helmet-sensor data that they observed ranged from 20gs to over 100gs [28].

This trend suggests that concussions may not be a result of an individual hit. A growing body of evidence suggests that players who sustain multiple subconcussive hits experience more long-term effects than a person who just experiences a large, clear, and singular impact to the head. Furthermore, it is unclear whether modern helmets successfully protect the brain from low-impact forces. Not only are helmets unproven to prevent concussion, but evidence also shows that helmets rapidly decrease in their effectiveness. In a study that compared helmets manufactured in 2002 to the older "box-style" helmets that had been used since the 1980's, researchers observed a significant decrease in the helmets' ability to withstand energy from linear impacts in twenty drop-tests [35]. Twenty collisions are not many, and that number is easily surpassed in the lifetime of a helmet – twenty subconcussive impacts could be sustained by a player of any contact sport during one practice alone.

Newtonian mechanics can be directly applied to linear concussions [36]. A mass in a pendulum has potential energy when it is elevated equal to its mass multiplied by g and its height. The potential energy of the elevated mass is converted to kinetic energy after the mass is released. Assuming a completely elastic collision, which is not actually possible, but is necessary for the calculation to proceed, energy is completely transferred to the object receiving the impact. The total energy is converted to work done by the impacted object.

Linear acceleration can then be measured by an accelerometer located inside of the skull. Helmets were compared using measurements of linear acceleration, resultant acceleration, and the length of time that acceleration persisted. The null hypothesis for each variable were that the type of head model would have no effect on the measurements. The alternate hypotheses for each independent variable were that the brand new helmet would perform better than the used helmet in each independent variable, but the used helmet would still perform better than the skull model control.
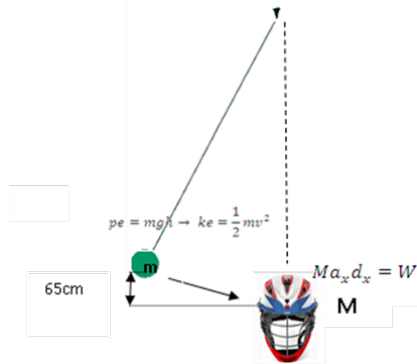

Figure 1: Diagram of How Impact Testing Was Conducted

METHODS

Constructing an Apparatus:

An apparatus was needed so that a skull model, either with a helmet or without, could be held upright and stable while being able to undergo a collision. A long pole was obtained from the Hampden-Sydney College physics department. The pole was then cut to 83.0 cm using a Powermatic™ saw. A 30.4 cm long board of pinewood was nailed to the bottom of the wooden pole so that it could stand upright with stability. Lead weights were placed on the stabilizing board to ensure that the apparatus didn't move during impact.

The wooden apparatus was then fitted to the skull model that was used. The foramen magnum, where the spinal cord inserts into the skull, was used as an insertion point for the wooden pole. The pole was originally too large for the hole, and a Powermatic™ sander was used to sand down the top 3.5 cm of the pole. After insertion of the pole, another hole was needed so that a data cable could be run from an acceleration sensor to a data-reading device. Using a Clausing™ drill and ¾ inches bit, a ¾ inch hole was drilled into the base of the plastic skull, directly dorsal to the foramen magnum. The skull model could then be fitted to the wooden pole base.

Creating a Ballistics Gel

The literature suggested that ballistics gel accurately resembled the brain in density and viscosity with high biofidelity [37]. Creating a ballistics gel used by the US Army or Mythbusters requires chemicals and mixing materials that are expensive, and essentially unnecessary for creating gels. The purpose for the chemicals that they use is that they allow the gel to be stored and remain accurate for a longer period of time [37]. First, the skull model was hermetically sealed with duct tape covering any holes where water could ooze though. Water was then added through the foramen magnum with a measuring cup. It was determined that the skull model could hold about five cups of water. Five cups of water equals 1480 mL. Knox gelatine was used to make the ballistics gel. Four packets of Knox gelatine (1 oz.) were used for each cup of water. 1500 mL of water, measured with a large graduated cylinder, were added to a 2000 mL beaker and heated on a hot plate to 55 degrees Celsius. Once the water reached this temperature, 20 packets, or 5 oz., of Knox gelatine was slowly added to the water while being mixed with a stirring rod. After the solution had become homogeneous, the solution was transferred to a Tupperware™ container. To ensure that all of the air was removed from the gel, a spoon was used to remove the top-layer of bubbly film from the solution. The solution was then refrigerated at four degrees Celsius for at least two hours but no longer than overnight. The gel was then removed and placed in a 37 degree Celsius water bath to re-melt the gel. The resulting aqueous solution was poured into the skull model through the foramen magnum. The solution inside the skull model was then frozen for 30 minutes and then transferred to a 4 degree Celsius refrigerator for at least 72 hours. After the gel had formed inside the skull, a section of the gel in the middle of the skull was removed so that an accelerometer could be placed inside the brain-like gel for the experiment.

Impact Testing

In order to test different helmets' efficiency, collisions were made between a helmet and a pendulum weight. Resulting impacts were then measured and recorded in accelerations due to gravity (gs). A 0.4 kg weight was used at two distinct times about two months apart. In between the two tests, each head model was subjected to 120

impacts. A theoretical acceleration could be calculated in the linear direction. In order to calculate a theoretical acceleration, distance measurements were first calculated using the same method in which acceleration was measured. Distance measurements were obtained using a PASCO™ Motion Sensor II attached to an Xplorer GLX from PASCO™. Measurements of acceleration were obtained using a PASPORT™ PS-2119 Acceleration Sensor attached to an Xplorer GLX. Data could was then read and analyzed in Data Studio™.

The 0.4 kg weight was suspended with a string attached to an arm of a ring stand. The ring stand had a lead weight on top of it so that it could be used as a fulcrum for a pendulum off the side of a desk. A meter stick was attached to the side of the desk with the 0 cm mark located at the center of mass of the skull model. The 0.4 kg weight was

raised 65 cm and then released. The mass then impacted the head model and was immediately removed so that it could make only one collision. The mass collided with three different head models: the naked skull model, an old Cascade™ Pro 7 lacrosse helmet on top of the skull model, and a new Cascade™ Pro 7 lacrosse helmet on top of the skull model. Impacts were made at four locations to each head model. Impacts were made to the forehead area, back of the head directly in the center of the occipital bone, and both sides of the head where the parietal bone and the temporal bone meet at the squamous suture. These four locations were determined to be the site of impact for most concussions experienced in men's lacrosse [29]. Ten impacts were recorded at each location, resulting in 40 total acceleration readings for the skull model and both helmets.

RESULTS AND DISCUSSION

Preliminary Measurements: Masses of objects were first measured. Measurement of the masses was essential in order to calculate a theoretical linear acceleration   Distance measurements were then taken. The notation that is included for impact location is used throughout the paper – the letters stand for words. For example, "LSOS" means "left side of skull".

Series of Impacts: Accelerations in the linear direction were measured by the PS-2119 PASCO™ acceleration sensor. Mean values for observed acceleration in the linear direction were compared with mean expected values. Because the mean values were inconsistent for each impact location mean percentages were used to compare the magnitude of impacts. The included standard deviation represents the amount of imprecision in the acceleration measurements.

The percentage of the expected acceleration was used to compare the accelerations of impact. A one-way analysis of variance (ANOVA) was conducted to compare the skull model, used helmet, and new helmet for each impact location. The Holm-Sidak method was used to compare the different head models pairwise. The results of the ANOVA and the Holm-Sidak test can be seen in Figure 2. Figure 2 uses the letters "A" and "B" to denote the separate groups that each head model conforms to. The "Front" and "Right" impact locations resulted in a significant difference between the skull model and both helmet models. The "Back" and "Left" impact locations resulted in a significant difference between

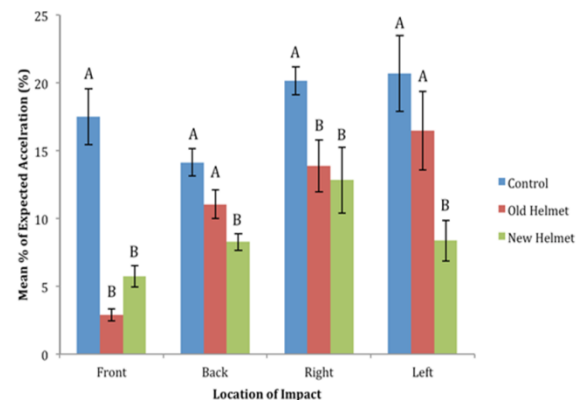the new helmet and both the skull model and old helmet.

Figure 2: Observed Linear Acceleration Percentages

Accelerations in the x, y, and z direction were measured by the PS-2119 PASCO™ acceleration sensor, and a resultant acceleration could be obtained. No theoretical acceleration could be calculated, but comparisons could still be made directly between acceleration observed after impacts. Table 1 below shows the mean observed resultant acceleration for each impact location. A post-hoc analysis using Dunn's method indicated a significant difference for the "Front" impact location between the skull model and both helmet models. The same method revealed significant difference between the new helmet and both the skull and used helmet model at the "Left" impact location.

Table 1:Observed Resultant Acceleration

Impact Location	Mean Observed Acceleration (g)
BOS	0.2869
BOO	0.1973
BON	0.3142
LSOS	0.2948
LSOO	0.211
LSON	0.2495
RSOS	0.4095
RSOO	0.1737
RSON	0.0697
FOS	0.8247
FOO	0.1823
FON	0.1837

The length of time that the acceleration persisted was compared between head models. Significance was found between impacts at all locations, but further post-hoc analysis using Dunn's method revealed that there was only a difference between the skull model and both helmets, and no difference between the different helmets

Table 2: Length of Time of Acceleration

Impact Location	Length of Observed Acceleration (s)
FOS	3.92
BOS	2.06
LSOS	2.08
RSOS	1.49
FOO	1.51
BOO	1.53
LSOO	1.46
RSOO	0.74
FON	1.56
BON	1.19
LSON	1.33
RSON	0.92

Discussion: In most every case there was a significant difference between the skull model and the new helmet on the first series of impacts. When looking at the back and left impact locations, it was determined that there was a significant difference between the old helmet and the new helmet in the first series of impacts. There was no difference between the used and the new helmet when observing the front and back impact locations. In fact, the old helmet actually performed better than the new helmet at preventing energy to the brain when impacts were made to the front of the head form. This is probably due to the high expected theoretical value for the "FOO" impact location. The resultant accelerations measured by the acceleration sensor did not indicate as much of a statistical difference. Significance was only indicated by the ANOVAs in two locations: the left and the front of the head. A post-hoc analysis using Dunn's method indicated that there was only an observed statistical difference between the used helmet and the brand new helmet when a collision was made to the left impact location.

Acceleration in the linear direction and resultant acceleration should be proportional to each other. The discrepancy in the results indicated that the acceleration sensor located inside the brain-like gel was experiencing more impact than expected at its other axes. The sensor was expected to measure accelerations in every direction because when the skull is impacted the brain-like gel will vibrate inside the skull in every direction, not just linearly. However, that vibration should be proportional to the magnitude of impact. The results introduced the idea that in the split-second of impact there could be rotational forces experienced by the accelerometer as well. Rotational forces, (unaccounted for), best explain the large discrepancy in linear and resultant acceleration.

The length of time that acceleration was measured was significant in all cases between the skull model and both helmets. The used helmet and the brand new helmet showed no difference at any impact location. Additionally, no significance could be garnered from the rates of decrease in acceleration in each head model. For all response variables, the skull model without a helmet performed worse at dissipating acceleration when compared to the two helmet models. This significant difference indicated that the helmets did protect the brain to some extent at low levels of impact. This result strongly agrees with a 2005 study that found that helmets are extremely important in preventing head injuries [38]. It also lends credence to the idea that concussions are grossly underreported in sports like rugby, and that concussion incidence rates are similar to that of contact sports that use helmets [39].

The results showed that lacrosse helmets decrease relatively rapidly. A lacrosse player can sustain many of these low-impact hits to the head over the course of a season, or even just a few practices. Currently, there is no requirement by the NCAA or NOCSAE that mandates renewal or recertification of a lacrosse helmet [18]. Based on the findings and previous results [8] there needs to be a stipulation of an expiration date for lacrosse helmets. Currently there is a ten-year expiration period on football helmets, but even that length of time is far too long [18]. Furthermore, the recertification of all sport-related helmets is insufficient. The method for recertification consists of visual observation for cracks and performing impact tests to a sample of helmets using NOCSAE protocols [42]. NOCSAE provided millions of dollars towards head, concussion, and helmet research in 2012 in order to work towards making a helmet that can better protect against concussion [18].

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