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Characterization and evaluation of head impact sensors and varsity football helmets

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CHARACTERIZATION AND EVALUATION OF HEAD IMPACT SENSORS AND VARSITY FOOTBALL HELMETS

A Thesis

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by

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To my father, Daniel Cummiskey, who always encouraged me to continue my academic pursuits and taught me the joy of learning.

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ABBREVIATIONS

TBI	Traumatic Brain Injury
mTBI	Mild Traumatic Brain Injury
CTE	Chronic Traumatic Encephalopathy
NFL	National Football League
ImPACT	Immediate Post-Concussion Assessment and Cognitive Testing
HIC	Head Injury Criterion
GSI	Gadd Severity Index
fMRI	Functional Magnetic Resonance Imaging
NOCSAE	National Operating Committee on Standards for Athletic Equipment
HITS	Head Impact Telemetry System
PTA	Peak Translational Acceleration
PAA	Peak Angular Acceleration
H3H	Hybrid III Headform
CoM	Center of Mass
NAP	Nine Accelerometer Protocol
CNC	Computer Numerical Control
RMSE	Root-Mean-Square Error
SRS	Sideline Response System
PF	Peak Force
IM	Impulse Momentum

ABSTRACT

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An increased understanding of the effects of brain injury in recent years has led to greater attention being given to the topic. A desire to investigate the causal agents of these injuries in athletes has led to the development and use of several devices that track head impacts as well as improving helmet technology to protect players from said impacts. In order to determine which devices are able to best measure head impacts, a Hybrid III headform was used to quantify the accuracy for translational and angular accelerations. Testing was performed by means of administering impacts to a helmet on the headform, with each device mounted according to manufacturer instruction, using an impulse hammer. For peak translational acceleration, the worst locational root-mean-square error for a head mounted device was 74.68% while the worst for a helmet mounted device was 297.62%. Head mounted devices outperformed those mounted in helmets and should be the basis of future sensor designs. For the sake of determining the effectiveness of recent helmet innovations, several helmet models were fastened to the headform in order to measure the response accelerations from impacts. The impulse hammer provided transient force data which allowed for the comparison of the input blow and output accelerations for each impact, and several metrics were defined and evaluated to determine helmet impact mitigation ability. Relative helmet effectiveness between models varied by region. The lowest peak translational acceleration metric was 0.31, and the highest was 0.57. The corresponding angular acceleration metric had a low of 0.23 and a high above one at 1.71. The helmets evaluated were more consistent in mitigating peak translational acceleration than peak angular acceleration.

1. INTRODUCTION

1.1 Motivation

A 2010 study of traumatic brain injury (TBI)-related emergency department visits, hospitalizations, and deaths reported a total 2.5 million cases, up from 1.7 million three years prior [1]. These head traumas typically result from motor vehicle collisions, falls, and sports-related injuries [2]. For a variety of reasons, athletes are particularly at risk [3]. Estimates have placed the number of sports-related TBIs in the United States between 1.6 and 3.8 million per year, including those cases that did not seek out medical care. [4]. A 2013 epidemiological study reported that 49.2% of reported TBI in sports are determined to be concussions, and that the most common mechanism (38.1%) of sport-related TBI was being either struck or kicked in football or rugby [5]. [6].

In addition to concussions, it has been further shown that a large percentage of high school football players experience substantial changes in neurophysiology in the absence of easily recognizable symptoms as a result of repeated head impacts [7] [8] [9]. Several recent studies have sought to better understand the nature of the problem through questionnaires [10], quantification of impact mitigation by helmets [11], neurocognitive testing [12], head impact tracking and recording [13] [14] [15] [16], medical imaging, or a more comprehensive combination of these methods [17] [18] [19] [20]. In order to relate head impacts to changes in neurophysiology, accurate reconstruction of head accelerations during practices and games is critical.

For these purposes, the Head Impact Telemetry System (HITS) was the first commercially available system to estimate and record translational and angular accelerations. Additional systems have since been developed which possess a variety of attachment mechanisms and differing electronics packages to interpret and record accelerations. It should also be noted that many of them have been modified for use in other contact sports, including soccer [21], hockey [22], rugby, and lacrosse, resulting in a large range of acceleration measurement systems.

Motivated by the sport-related TBI statistics, there is interest in reducing the frequency and severity of head impacts experienced by athletes participating in contact sports. The primary protective system used in football, lacrosse, and hockey is the helmet, which has evolved dramatically over the years. Potential correlations have been investigated between improved helmet design and reduced incidence of concussion [23], although the studies to date have been inconclusive [24] [25] [26] [27]. A comparison of modern varsity football helmets and their leather predecessors was able to demonstrate that many modern helmets have been optimized to perform well under certain conditions but do not necessarily do so under those subconcussive impacts commonly experienced by football players [11]

1.2 Objectives

Although an early study demonstrated that the HITS system provided accurate results for CoM impacts under highly idealized conditions [13], a later study, under more realistic conditions, demonstrated a root-mean-square error (RMSE) for the translational acceleration as high as 189.7% for one particular location on the helmet [28]. The variability of sensor deployments and their methods for estimating the accelerations of the head's CoM, combined with limited assessment data, motivate a need for their simultaneous comparison. As such, one of the major objectives of this study was to not only determine the relative accuracies of each device, but also to make recommendations for future sensor design aimed at improving head impact measurement accuracy.

While helmet designs continue to evolve, there is little published data documenting the effectiveness of recent innovations. To date, independent verification of helmet effectiveness has evaluated helmets through the administering of highly idealized impacts. This has been through the use of drop towers [29] and pendulums [11], but these data are rarely released apart from in a form that has little physical significance. It should also be noted that impacts experienced by football players are not directed only normal to the surface of the helmet. There is also great variety in the locations of head impacts experienced by football players. Prior studies have taken neither of these facts into consideration in their experimental designs. To this end, the other major objective of this study was to evaluate the locational effectiveness of several commonly-used helmets in their ability to mitigate the severity of resultant kinematics exhibited by the head under variable impact load conditions.

2. LITERATURE REVIEW

2.1 Head Injury

Damage to the brain is one of the most difficult to understand, making diagnosis, treatment, and prevention difficult. Several forms of brain injury have been classified over the years. Traumatic brain injury (TBI), and its less severe form mild traumatic brain injury (mTBI), are injuries caused by externally applied mechanical trauma. Chromic traumatic encephalopathy (CTE) is a degenerative disease of the brain that results in individuals who have experienced head injury. These injury and disease states have profound economic consequences [30] in addition to the personal difficulties faced by the individuals who experience them.

Concussions are the most commonly diagnosed type of TBI, and is considered synonymous with mTBI. A concussion is described as a temporarily partial loss of brain function, which is sometimes evidential through headaches, difficulty with concentration, hindered coordination and balance, and memory loss [31]. Diagnosis involves a physician and requires checking the individual for said symptoms, though they can manifest in a wide variety of ways and thus can be difficult to detect. The primary mechanism of treatment is rest of the individual, both physical and cognitive, until symptoms are completely gone and potentially longer [32].

2.2 Return to Play Criteria

Football has been described as being in the middle of a 'Concussion Crisis' as coined by one author [33] and perpetuated by others [34]. An initial report that looked at the epidemiology of concussion in football players indicated that over five percent of players were receiving at least one concussion during the monitored season, and that those who received at least one concussion were three times more likely to receive another during the same season as the initial [35]. Recent study of the effects of these concussions on retired National Football League (NFL) players has shown a prevalance of CTE in these players post-mortem with the first case discovered in 2005 [36] and several more following suit [37].

Spurred on by these findings, the policy behind returning football athletes to play began to fall under reconsideration [38]. Concucssed players would receive evaluations that sought to diagnose the presence or absense of posttraumatic amnesia as a criteria by which to determine safety in return to play decisions [39]. One study had previously suggested using a pre-season baseline from which to compare a player's performance post-injury so as to have an objective means of determining a player's state [40]. An NFL study determined that neuropsychological testing had an important place in on-field evaluation of players [41], but also reported that no statistical association could be made for likelihood of concussion between same game return to play and over seven days delay prior to return to play [42]. Computer-based neuropsycholocial testing that built on this principle then began to grow in popularity. One such test that became somewhat widely used was the Immediate Post-Concussion Assessment and Cognitive Testing (ImPACT) which had the player complete the computerized assessment and his score would be compared against his own baseline [12] [43]. Questions regarding the diagnosticity of these tests have been raised due to high false positive rates in flagging controls. In light of this growing body of research, the NFL has since modified its policy on return to play criteria to require neurological examination by an independent neurological consultant in addition to the team physician [44].

2.3 Subconcussive Impacts

TBI has been studied for many years, but classical investigations have been centered around catastrophic events such as vehicular collisions [45]. Attempts at developing metrics that could quantify impact severity were accomplished in the form of the Head Injury Criterion (HIC) [46] and Gadd Severity Index (GSI) [47]. However, there was little investigation into head impacts other than the ones that directly caused brain injury until a study that coined the term 'Second Impact' [48]. This was among the first publications to formally postulate that sequential subconcussive impacts could cause injury through compounding means rather than treating them as isolated events.

One study with hockey players was able to utilize event-related potentials and post-concussion syndrome self reports to show statistically significant changes between players who experienced no concussions and those with three or more [49]. More recently, studies have sought out to characterize the quantity and magnitude of head impacts experienced by football players at both the high school and collegiate levels [50]. Subconcussive impacts have been characterized to cause cognitive declines that would typically be associated with aging [51]. CTE remains elusive in its connections with subconcussive impacts, but searches for biomarkers enabling its diagnoses in alive athletes and thus its correlations with these impacts are underway [52].

Several studies have performed statistical analyses to determine what sort of correlations exist between subconcussive impacts and various measures taken from the players. Functional magnetic resonance imaging (fMRI) scans and computer-based neuropsychological testing have become popular means of detecting neurological changes in players [18] [17] [19]. Effects of subconcussive impacts on brain connectivity have been expounded on [7] [8]. Recent investigations have even begun to make predictions of deviant brain metabolism based on measured head impacts [20]. Spectroscopic evidence of brain injury has been determined in those athletes who went the entire football season with no concussion diagnoses [9]. The effects of subconcussive impacts are growing, and our understanding is becoming clearer as a consequence.

2.4 Helmet Development

One of the means of protecting the athlete's head has been through the use of personal protection equipment. For football, the use of the helmet can be traced back over 100 years if one includes their leather predecessors [53], but they did not become mandatory unti 1939 for the collegiate level [54] and 1943 for the NFL [55]. Helmet developments followed with the change to primarily unibody molded plastic shells, which increased helmet longevity as compared to their leather predecessors. Air bladders came to replace traditional padding materials, and helmets began to be fastened more securely through the advancement from two to four attachment point chin straps.

More recently, studies have begun to investigate the epidemiological effects of helmet design on football players. One study evaluated two populations of high school football players, one which wore new Riddell Revolution helmets and the other which wore more traditional helmets. The findings were such that concussion rates decreased from 7.6% to 5.3% with statistical significance, and these changes were attributed to the technology and design of the Revolution helmets [23]. Several groups have since sought out to characterize the varying degree of effectiveness with which helmets accomplish the task of protecting a player's brain. The National Operating Committee on Standards for Athletic Equipment (NOCSAE) originally designed a certification program by which helmets could be deemed safe to use or not based on a defined GSI threshold [56]. Biokinetics and Associates Ltd. performed a validation of the methodology commonly used to measure headform kinematics in dummy heads [57]. One group developed a system of evaluation by which to rate various commercially available football helmets [29]. Another study sought to evaluate modern varsity football helmets compared to their leather predecessors and found that they often failed to outperform their less technologically advanced bretheren in cases of subconcussive head impacts, and the authors attributed this to the fact that many helmet manufacturers were designing their helmets to perform well on NOCSAE drop tests which put helmets through high-severity linear impacts rather than for the majority of impacts sustained by the helmets' wearers [11].

2.5 Head Impact Sensor Development

With the growing concern over head impacts in football, it became evident that a system was needed that could monitor the impact sets being taken by players. Among the first to emerge on the market was the Head Impact Telemetry System (HITS). Some studies have found it to represent impacts with high accuracy [13]. One paper showed that HITS performed better in estimating Peak Translational Acceleration (PTA) than Peak Angular Acceleration (PAA) [58]. One experiment made an effort to more accurately represent the fit of the helmet and found that the HITS device does not perform as well as previously reported under realistic field conditions [28]. There are kinematic shortcomings of determining the kinematics of a rigid body using only six uniaxial accelerometers, requiring the searching of a database to determine impact accelerations rather than directly computing them from measured accelerations. The HITS system assumes that all rotation in the vertical axis is zero, an unrealistic simplification.

Due to the growing interest in monitoring players' head impacts, other devices have begun to surface. The SHOCKBOX is a device that utilizes binary force switches to emeasure impacts with linear accelerations of over $30g$ and has been used to monitor head impacts in Pop Warner football [59]. Impacts in this young age group averaged, on a per player basis, 1.5 per fractice and 3.7 per game. Another recent development has been a mouthguard containing both a triaxial accelerometer and gyroscope, which is therefore capable of recording PAA with higher accuracy than an underconstrained six single axis accelerometer device [60]. In order to determine when the device was securely in place, an infrared proximity sensor was incorporated to detect that the player was biting down on the mouthguard so as to ensure that impacts measured were in fact head impacts and not jostling of the device [61]. This was a common problem with HITS, which could generate impacts when a player dropped or threw his helmet on the ground. Verifiation of the kinematic accuracy proved that it has higher fidelity of impact representation than previously available devices [62].

3. GENERAL SETUP

3.1 The Hybrid III Headform

The Hybrid III 50th Percentile Male Crash Test Dummy was originally developed by General Motors (Detroit, MI) and is now under the ownership and development of Humanetics (Plymouth, MI). It is a crash test dummy, and for the purpose of evaluating safety restraints in frontal crash tests, nothing else has been more widely used throughout the world. The Hybrid III Headform (H3H) is thus representative of an average male in size and mass. It is constructed of a single piece of aluminum with a vinyl skin overlay. The neck that the headform sits on is made out of a butyl rubber pieces with aluminum segments between them, and an adjustable center cable enables alteration of the neck's stiffness. It is described to exhibit an accurate representation of the dynamic rotational and flexion responses [63].

The H3H was deemed desireable for the purpose of head impact testing due to the versatility of impacts that can be administered to it. While the use of drop towers allows for precise linear impacts to be delivered to a headform, the neck joint is rigid and thus provides an unrealistic response during impact. However, the H3H is attached to a neck that allows for a much more realistic response motion of the headform. In order to secure this headform and neck, a neck mount was machined and attached to a large steel block. This allowed for the portability of the headform while also preventing motion of the neck's base due to severe impacts.

3.2 Kinematic Data Acquisition

The primary purpose of using the H3H was to determine the kinematic response of the headform during impacts. To this end, a sensor system was needed that would determine both the translational and angular accelerations of the headform's center of mass (CoM). Triaxial accelerometers are the obvious choice for translational acceleration measurement, but measurement of angular acceleration has a pair of options.

The first option is to utilize a triaxial gyroscope, the advantages of which are the ability to place the sensor anywhere on the rigid body, only three channels of data acquisition necessary, and no development of complex mounting hardware. However, this option also holds a significant disadvantage in that the data acquired is angular velocity. In order to determine angular acceleration, one must numerically differentiate the angular velocity data. Small errors in angular velocity, which can be due to sensor inaccuracy, acquisition hardware inaccuracy, or signal degredation, become larger errors through the process of numerical differentiation. As a consequence, the data acquired is less accurate.

The second option is to utilize a nine accelerometer protocol (NAP) in a 3-2-2-2 orientation originally defined by Padgaonkar et al. [64] (Fig. 3.1). The advantages to this option are that translational accelerometers have higher accuracy than angular sensors, there is no compounding of errors due to numerical differentiation, and this protocol is well established in literature [13] [28] [58]. This was the method chosen.

A brief derivation of the equations used to solve for the angular acceleration follows. The origin is defined at a triaxial accelerometer placed at the CoM of the H3H with $\underline{E}_1$ going forward through the front of the face, $\underline{E}_2$ going to the side through the left side of the head, and $\underline{E}_3$ going up through the top of the head. There are six single axis accelerometers, a_4 through a_9, placed strategically along the axes of the triaxial accelerometer. The derivations for the accelerometers 4 in Equation (3.1) and 9 in Equation (3.2) are combined to solve for $\alpha_1\underline{E}_1$ in Equation (3.3), and the same process is done for the other four peripheral accelerometers in order to solve for $\alpha_2\underline{E}_2$ in Equation (3.4) and $\alpha_3\underline{E}_3$ in Equation (3.5).

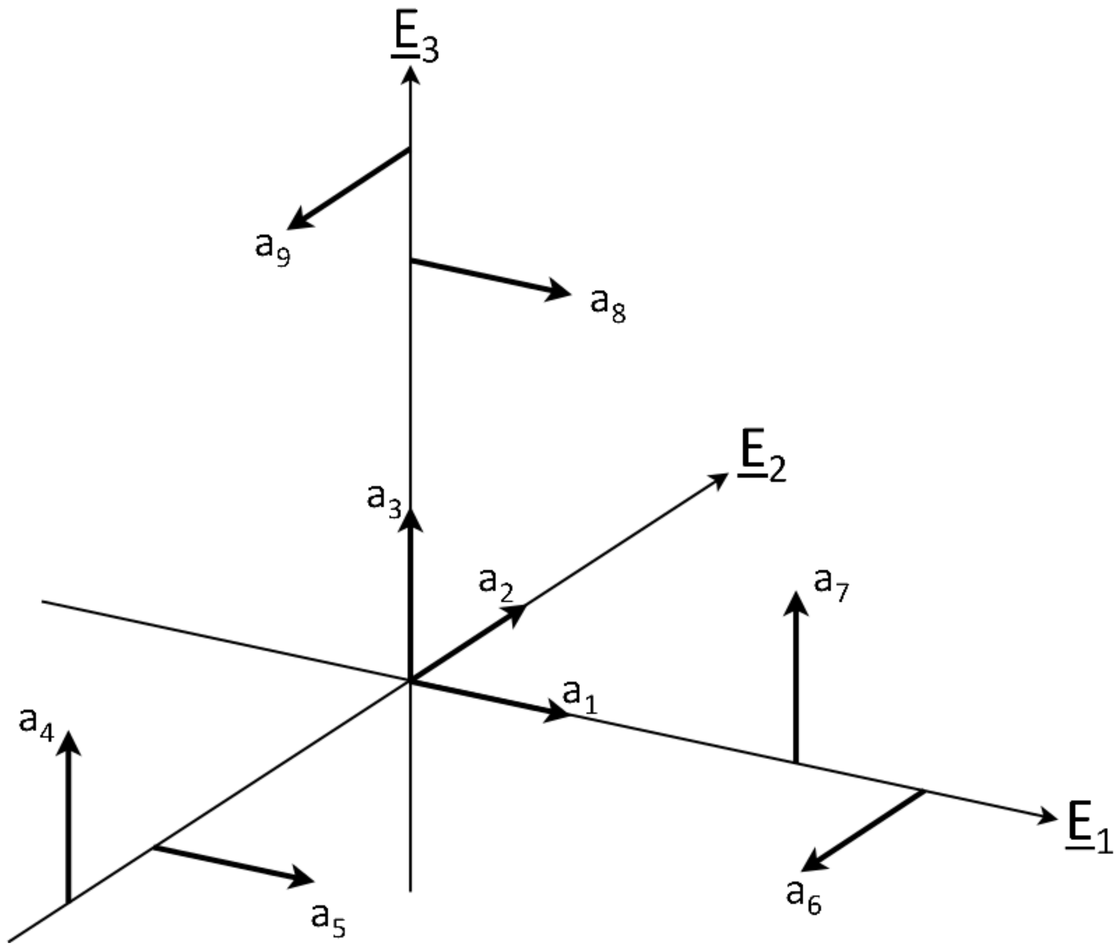

Figure 3.1. Nine accelerometer channels were arranged with a triaxial accelerometer at the H3H's CoM, while a pair of single axis accelerometers were placed strategically along three of the axes with orientations orthogonal to them.

$$\underline{r}_{O4} = -\rho_4\underline{E}_2 \qquad \rho_4 = 1.545in$$

$$\underline{r}_{O5} = -\rho_5\underline{E}_2 \qquad \rho_5 = 1.265in$$

$$\underline{r}_{O6} = \rho_6\underline{E}_1 \qquad \rho_6 = 2.070in$$

$$\underline{r}_{O7} = \rho_7\underline{E}_1 \qquad \rho_7 = 1.790in$$

$$\underline{r}_{O8} = \rho_8\underline{E}_3 \qquad \rho_8 = 1.815in$$

$$\underline{r}_{O9} = \rho_9\underline{E}_3 \qquad \rho_9 = 2.095in$$

$$\begin{aligned}
[\underline{a}_4 &= \underline{a}_O + \underline{\omega} \times \underline{\omega} \times \underline{r}_{O4} + \underline{\alpha} \times \underline{r_{O4}}] \cdot \underline{E}_3 \\
&= [\underline{a}_O + \underline{\omega} \times \underline{\omega} \times -\rho_4\underline{E}_2 + \underline{\alpha} \times -\rho_4\underline{E}_2] \cdot \underline{E}_3 \\
a_4 &= a_3 + -\omega_2\omega_3\rho_4 - \alpha_1\rho_4 \\
\alpha_1 &= (a_3 - a_4)/\rho_4 - \omega_2\omega_3 \qquad (3.1)
\end{aligned}$$

$$\begin{aligned}
[\underline{a}_9 &= \underline{a}_O + \underline{\omega} \times \underline{\omega} \times \underline{r}_{O9} + \underline{\alpha} \times \underline{r_{O9}}] \cdot -\underline{E}_2 \\
&= [\underline{a}_O + \underline{\omega} \times \underline{\omega} \times \rho_9\underline{E}_3 + \underline{\alpha} \times \rho_9\underline{E}_3] \cdot -\underline{E}_2 \\
a_4 &= -a_2 - \omega_3\omega_2\rho_9 + \alpha_1\rho_9 \\
\alpha_1 &= (a_9 + a_2)/\rho_9 + \omega_3\omega_2 \qquad (3.2)
\end{aligned}$$

$$\alpha_1 = \frac{(a_3 - a_2)}{2\rho_4} + \frac{(a_9 + a_2)}{2\rho_9} \qquad (3.3)$$

$$\alpha_2 = \frac{(a_3 - a_7)}{2\rho_7} + \frac{(a_8 + a_1)}{2\rho_8} \qquad (3.4)$$

$$\alpha_3 = \frac{(a_5 - a_1)}{2\rho_5} - \frac{(a_6 + a_2)}{2\rho_6} \qquad (3.5)$$

3.3 Accelerometer Mount

In order to place the nine accelerometers in the locations and orientations necessary for the previous derivation, a mount needed to be manufactured that would secure the accelerometers during impacts exceeding $200g$. A preliminary mount was manufactured that accomplished this through the use of one central bracket and three arms that extended along the axes of the triaxial accelerometer. All parts were machined out of aluminum using CNC mills to within 1/100th of an inch. Aluminum was chosen due to its high strength while also adding minimal weight to the H3H assembly. The accelerometers were held into place with Loctite 454 (Henkel; Dusseldorf, Germany).

Once assembled, a preliminary use of the array revealed that one of the arms was exhibiting a great deal of noise on the order of $10g$. It was determined that the E_3 arm was experiencing resonance through calculation of the resonant frequency. It was also determined that neither of the other two arms (Figs. A.2, A.3) were experiencing resonance, so the first resonant frequency was matched between the existing E_2 arm and the new design of the E_3 arm as shown in Equation (3.6). The new design was the original beam with an additional extrusion from each face, creating a plus shape geometry (Fig. A.4) while maintaining its ability to be bolted to the base (Fig. A.5). Upon incorporation with the rest of the setup (Fig. A.1), the cleanliness of the signal confirmed that the beam resonance had been sufficiently driven up to a higher frequency such that it was no longer affecting the output signal. Subsequent evaluation of head impact sensors and football helmets was performed using this finalized setup.

$$\begin{aligned} \alpha_n^2[\frac{EI_2}{\rho A_2 L_2^4}]^{1/2} = \omega_2 = \omega_3 = \alpha_n^2[\frac{EI_3}{\rho A_3 L_3^4}]^{1/2} \\ \frac{I_2}{A_2 L_2^4} = \frac{I_3}{A_3 L_3^4} \\ I_3 = \frac{I_2 A_3 L_3^4}{A_2 L_2^4} \end{aligned} \tag{3.6}$$

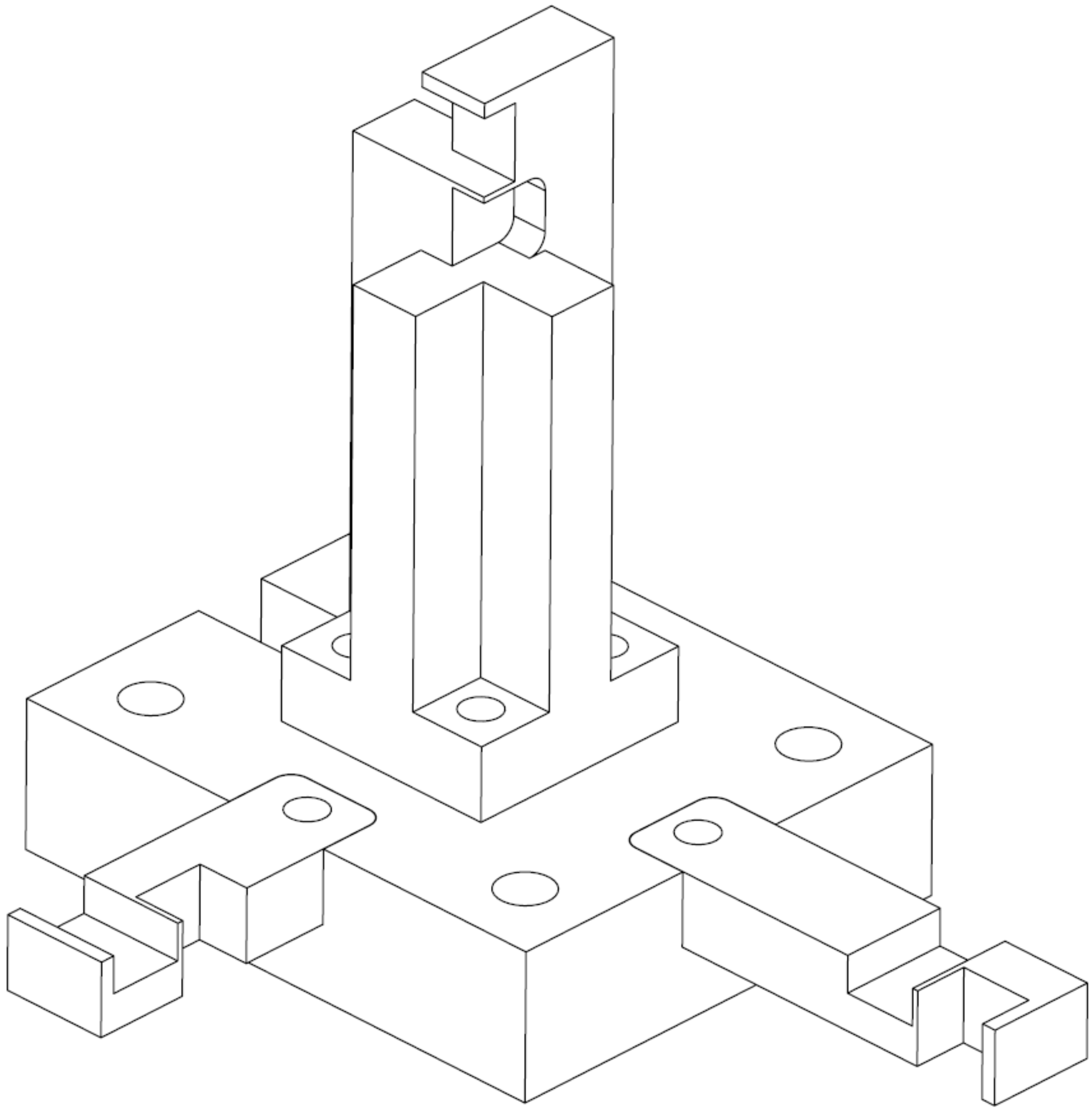

Figure 3.2. Nine accelerometer channels were arranged with a triaxial accelerometer at the H3H's CoM, while a pair of single axis accelerometers were placed strategically along three of the axes with orientations orthogonal to them.

4. SENSOR VALIDATION

4.1 Methods

This sensor study consisted of the evaluation of several commercially available sensor packages used to detect head impacts in athletes (Fig. 4.1): HITS (Simbex; Lebanon, NH), Shockbox HD (Impakt Protective Inc.; Kanata, Ontario, Canada), SIM-G (Triax Technologies Inc.; Norwalk, CT), xPatch (X2Biosystems, Seattle, WA), and the CHECKLIGHT (Reebok; Canton, MA). HITS and Shockbox HD both mount to the helmet although it should be noted that use of HITS is restricted to Riddell brands. In contrast, the remaining systems attach to or fit on the head directly. The SIM-G was enclosed in a headband and the sensing device rested on the back of the head. The xPatch was affixed behind the ear with a custom double-sided adhesive patch, and the CHECKLIGHT was integrated into a skullcap and its electronics package stretched from the back of the head to roughly the left ear along the edge of the skullcap.

4.1.1 Data Collection

All laboratory evaluations were performed with a 2012 Riddell Revolution helmet (size Large) outfitted with a 05-08C facemask and soft cup chin strap (Riddell; Rosemont, IL). The helmet was fitted on a 50^{th} percentile H3H [28].

The trigger mechanism for event recording was set to a threshhold of 10 pounds of force as measured by an impulse hammer, and an impact recorded 200 ms split up into 70 ms pre-trigger and 130 ms post-trigger. Each of nine accelerometer channels, as well as the impulse hammer channel, were recorded over the same impact time domain. The impulse hammer and accelerometer data were acquired using model 9234

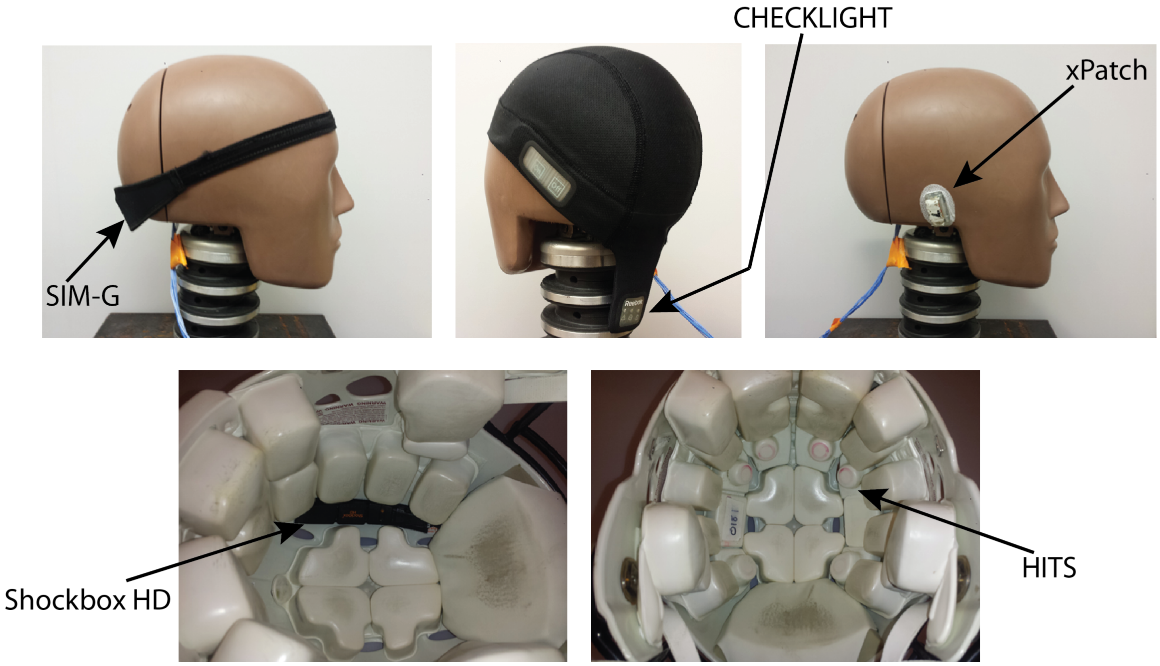

Figure 4.1. Five devices were evaluated in this study. The top row illustrates the three head-mounted devices, SIM-G (Triax Technologies), CHECKLIGHT (Reebok), and xPatch (X2Biosystems). The two helmet-mounted devices are shown in the bottom row, the Shockbox HD (Impakt) and HITS (Simbex).

data acquisition modules (NI; Austin, TX) with custom software [65] at a sampling frequency of 5120 Hz (Fig. 4.2).

A total of 140 impacts (7 locations, 20 impacts per location) were delivered per testing iteration using a modally tuned impulse hammer (PCB Piezotronics, Inc.; Depew, NY) to record the impact force for each blow (Fig. 4.2). The impacts recorded by each sensor package were compared to an accelerometer array housed inside of the H3H consisting of 9 accelerometers (six uniaxial and one triaxial) arranged in the 3-2-2-2 configuration originally proposed by Padgaonkar et al. [64]. One testing iteration was performed for each of two devices of HITS, xPatch, and SIM-G, and one device was tested twice for Shockbox HD and CHECKLIGHT.

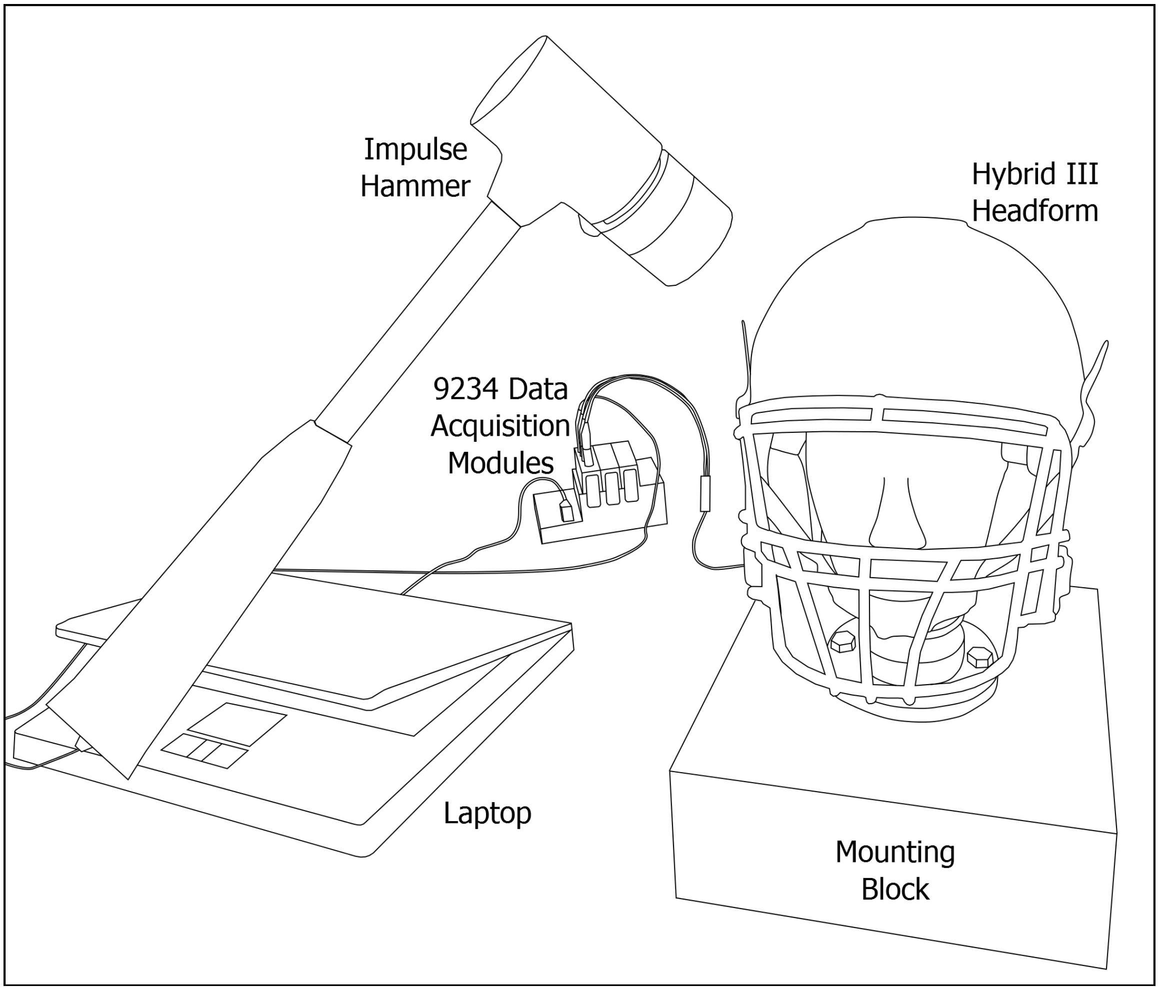

Figure 4.2. The experiments consisted of impacts delivered by an impulse hammer to a H3H outfitted with a Riddell Revolution helmet. The National Instruments 9234 Data Acquisition Modules digitized the voltage output from both the impulse hammer and the accelerometers. Custom software processed the data and provided force vs. time for the impulse hammer as well as the translational and angular acceleration of the headform's CoM.

Seven locations were chosen to determine device performance under a variety of impact conditions (Fig. 4.3). All impacts were delivered approximately normal to the helmet except for A', which was applied at an angle of approximately 45°.

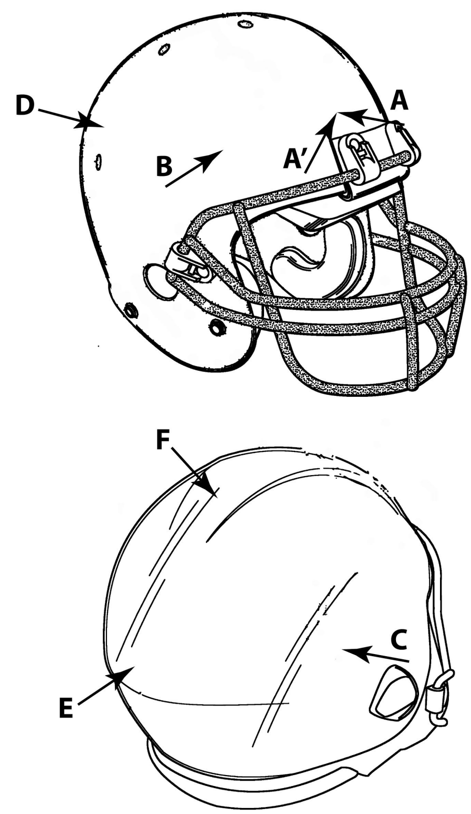

Figure 4.3. Seven impact locations were used in the current study. Impacts at locations A-F were administered at an orientation normal to the helmet surface, while those at A' were administered at an oblique orientation of approximately 45°.

4.1.2 Post-Processing

The data recordings were post-processed by solving the kinematic equations for the angular accelerations [64], and the peak translational (PTA) and angular (PAA) accelerations were then determined for the CoM of the headform. Each sensor package that reported information on individual impacts was correlated with the headform data using their timestamps (Fig. 4.4). Successful recordings were tabulated and the PTA and PAA were recorded for each head impact. Relative errors were calculated using the headform data as the datum, and from these, the absolute error and the root mean square errors (RMSEs) were calculated as shown in Equation (4.1). P_m is the peak measured translational or angular acceleration as appropriate, H_3 is the corresponding acceleration measured by the headform accelerometers, and N is the number of data points.

$$RMSE = \sqrt{\frac{\sum(\frac{P_m - H_3}{H_3} \times 100)^2}{N}} \tag{4.1}$$

The CHECKLIGHT represented a special case in the sense that it does not provide data for each hit. A green light was triggered if at least 100 "mild" impacts, a yellow light corresponded to "intermediate" impacts and red lights were considered "severe," although definitions of mild, intermediate, and severe were not provided.

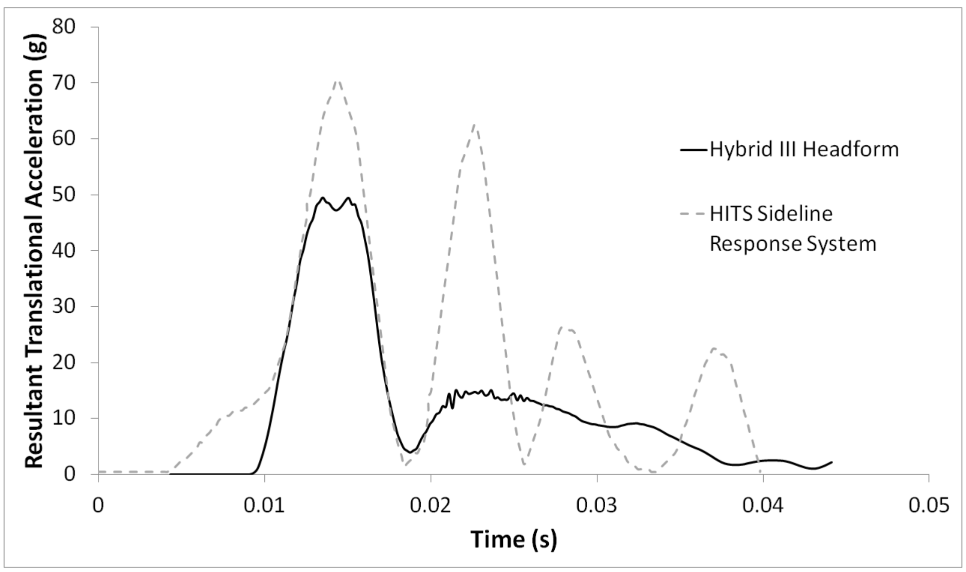

Figure 4.4. The HITS software, called the Sideline Response System (SRS), provides a plot of the translational acceleration as soon as the impact is read in. One such impact is displayed above. Also displayed is the actual resultant translational acceleration of the H3H's triaxial accelerometer at its CoM for the same impact.

4.1.3 Human Subjects

All research conducted in the currrent study was approved by an Institutional Review Board. Where applicable, parental consent and participant assent were obtained from study subjects. Seventeen male high school football players between the ages of 15 and 18 were enrolled in the study for season five during Fall of 2013, and all 17 participated throughout the entire season. For an in season game, both HITS and xPatch devices were deployed amongst these players. The HITS devices were installed in the helmets of the players, while xPatches were placed behind the players' right ears. Close attention was paid to players throughout the game so as to ensure that no artifical impacts were created by HITS through the jostling of a removed helmet as well as to verify that the xPatches were adhering well to the players' heads. The data was collected for both systems using the same generation of devices as were used in the rest of the current study.

4.2 Results

Of the four systems tested here that report on a per-impact basis, all of them recorded at least 128 of the 140 head impacts (Table 4.1). The Shockbox, SIM-G, and xPatch were all able to record impacts with no more than one event omission, while HITS failed to record five impacts for Device 1 and twelve impacts for Device 2. Device designation of impacts as Valid or Invalid attempted to convey whether an impact was a legitimate one to a player wearing the device. This was done in an attempt to filter out illegitimate hits sensed by the devices when not being worn by a player, such as HITS detecting an impact when a helmet gets dropped or xPatch detecting an impact when the device fell off of a player.

As for correctly predicting the location of the impact, xPatch performed best of the three that report such information. It is notable that the results for Shockbox were device-dependent.

Table 4.1.
The device impact recording, invalid designation, and location prediction rates give insight as to the reliability of these devices in their respective abilities to consistently record existent head impacts and depict the impacts' locations accurately.

	Impacts Recorded	Impacts Marked Invalid	Correct Location Prediction
HITS Dev. 1	96.43%	-	40.71%
HITS Dev. 2	91.43%	2.86%	53.57%
Shockbox Trial 1	100.00%	-	27.14%
Shockbox Trial 2	100.00%	-	65.00%
SIM-G Dev. 1	100.00%	-	-
SIM-G Dev. 2	99.29%	-	-
xPatch Right Dev. 1	99.29%	50.00%	77.86%
xPatch Right Dev. 2	100.00%	50.71%	80.00%
xPatch Left Dev. 1	100.00%	56.43%	86.43%
xPatch Left Dev. 2	100.00%	55.71%	91.43%

The Reebok Checklight was not included in any of the the surrounding analyses because it does not provide a per-impact event reporting system. The first trial yielded one green light corresponding to at least 100 mild severity impacts, four yellow lights corresponding to four intermediate severity impacts, and one red light corresponding to one severe impact. The most severe impact administered in the first trial had a PTA of $123g$ and a PAA of $7660rad/s^2$. The second trial yielded the same results except for zero red lights corresponding to zero severe impacts. The most severe impact administered in this second trial had a PTA of $87g$ and a PAA of $9330rad/s^2$.

4.2.1 PTA Assessment

The RMSE for the PTA varied with sensor type, impact location, and device (Table 4.2) with a minimum of 10.99% for the xPatch Device 2 placed on the right ear due to impacts at location A and a maximum of 297.62% for the first Shockbox trial due to impacts at location D (Table 4.2). Except for the oblique Location A', the lowest average RMSE was obtained by one of the xPatch devices behind either the right or left ear. The maximum RMSE was observed in the Shockbox for every location except for E, in which HITS Device 1 narrowly exceeded the worst Shockbox HD statistics (Table 4.2).

The mean absolute value of device error for the PTA had a low value of 7.74% for xPatch Device 2 placed behind the right ear when measuring impacts administered to location A, while the highest value was at 294.79% for the first Shockbox trial at location D (Table 4.3). The maximum mean value was observed in the Shockbox for every location including E. All devices were exposed to a similar set of impacts (Figs. 4.5 & 4.6).

Table 4.2.
The RMSE of device reported PTA varied by both device and impact location.

	PTA RMSE							
	A	A'	B	C	D	E	F	Average Rank
HITS Dev. 1	46.91%	49.10%	41.43%	60.55%	69.00%	197.80%	38.18%	3.71
HITS Dev. 2	62.27%	59.12%	60.65%	60.33%	36.95%	87.74%	33.42%	
Shockbox Trial 1	138.36%	104.84%	222.93%	211.57%	297.62%	163.81%	138.42%	5.00
Shockbox Trial 2	141.79%	92.49%	210.12%	136.21%	100.18%	195.88%	223.68%	
SIM-G Dev. 1	24.24%	12.97%	25.95%	33.80%	46.14%	52.91%	74.68%	2.43
SIM-G Dev. 2	17.91%	19.77%	23.19%	26.69%	34.95%	64.89%	68.14%	
xPatch Right Dev. 1	13.60%	21.66%	17.14%	45.42%	16.06%	23.94%	37.18%	1.71
xPatch Right Dev. 2	10.99%	20.30%	21.93%	34.13%	11.81%	21.21%	23.90%	
xPatch Left Dev. 1	22.13%	21.30%	29.25%	30.71%	58.64%	36.70%	16.05%	2.14
xPatch Left Dev. 2	18.43%	20.54%	31.90%	25.92%	49.65%	21.29%	24.50%	

Table 4.3.
The mean absolute value of the errors of device reported PTA varied by both device and impact location.

	PTA Mean Absolute Value of Error							
	A	A'	B	C	D	E	F	Average Rank
HITS Dev. 1	37.13%	35.59%	31.38%	42.98%	45.66%	149.27%	25.27%	3.71
HITS Dev. 2	50.77%	45.10%	43.89%	51.18%	31.09%	67.80%	26.32%	
Shockbox Trial 1	133.61%	96.11%	217.52%	206.13%	294.79%	133.03%	98.51%	5.00
Shockbox Trial 2	128.80%	76.92%	192.08%	106.14%	79.17%	170.13%	214.31%	
SIM-G Dev. 1	19.76%	10.30%	19.73%	24.85%	40.61%	42.82%	56.36%	2.29
SIM-G Dev. 2	15.90%	18.04%	19.57%	21.91%	28.21%	43.83%	50.36%	
xPatch Right Dev. 1	11.02%	17.75%	14.62%	40.87%	12.49%	20.07%	31.49%	1.57
xPatch Right Dev. 2	7.74%	16.88%	17.51%	29.49%	9.90%	18.87%	20.27%	
xPatch Left Dev. 1	19.25%	18.64%	27.43%	30.18%	57.94%	29.42%	12.90%	2.43
xPatch Left Dev. 2	15.11%	16.53%	30.02%	24.32%	46.02%	17.98%	17.50%	

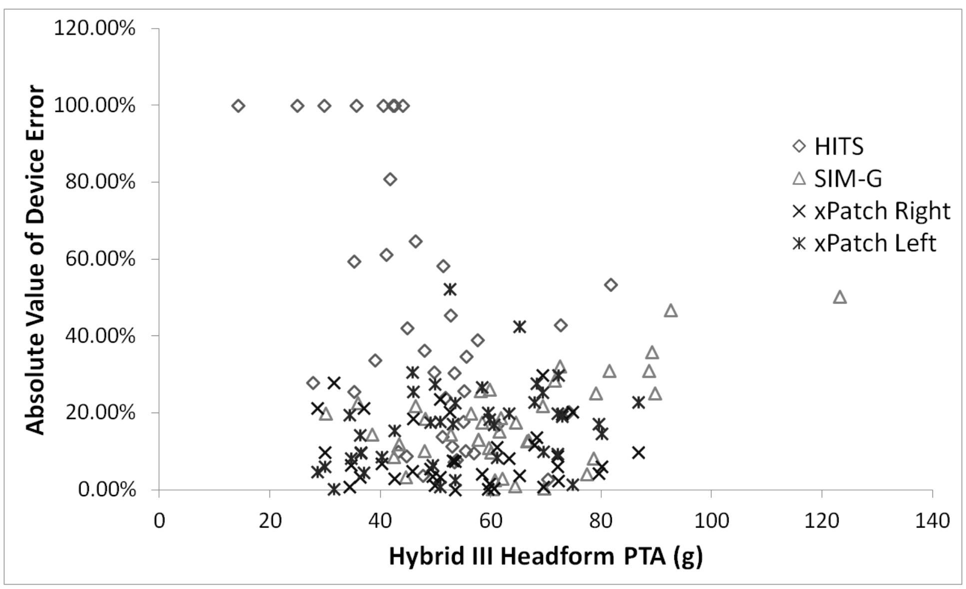

Figure 4.5. HITS, SIM-G, and xPatch exhibited different trends in errors that were dependent on the impact location but not on the magnitude of the acceleration. It can be seen that while SIM-G and the xPatch behind both ears performed similarly, HITS had a broader range of errors for impacts at location A.

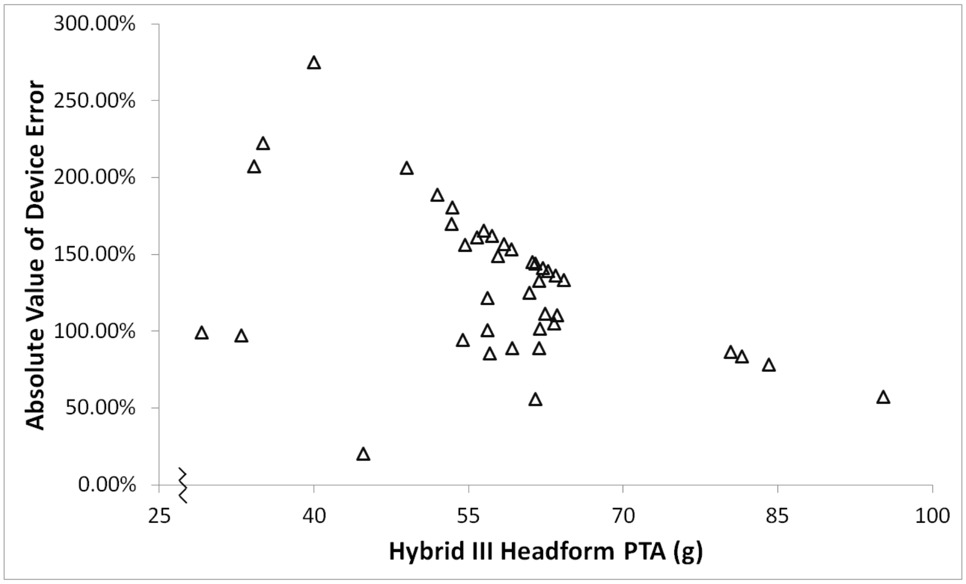

Figure 4.6. Location A impacts measured by Shockbox yielded PTA errors that ranged from 20.63% to 275.24%. It should be noted that, above 65g, the error decreased monotonically, but was consistently over 100%.

4.2.2 PAA Assessment

The RMSE for the PAA also varied with sensor type, impact location, and device (Table 4.4). The xPatch Device 1 mounted behind the left ear exhibited the lowest RMSE of all device and hit location combinations for location D, while the same device for location E had the highest. The large RMSE and mean value of absolute error statistics for the two xPatch devices behind the left ear, when measuring impacts made at location E, were caused by a subset of the impacts that produced grossly overestimated PAAs.

The mean absolute value of device error for the PAA had a low value of 9.48% for xPatch Device 1 placed behind the left ear when measuring impacts administered to location D, while the highest value was at 245.65% for the same device at location E (Table 4.5). The maximum mean value observed for each location alternated between xPatch and HITS.

Table 4.4.
The RMSE of device reported PAA varied by both device and impact location.

	PAA RMSE							
	A	A'	B	C	D	E	F	Average Rank
HITS Dev. 1	82.02%	48.36%	50.22%	44.63%	58.26%	208.97%	42.08%	2.43
HITS Dev. 2	65.23%	70.44%	60.48%	27.37%	56.56%	106.38%	86.67%	
Shockbox Trial 1	-	-	-	-	-	-	-	-
Shockbox Trial 2	-	-	-	-	-	-	-	
SIM-G Dev. 1	-	-	-	-	-	-	-	-
SIM-G Dev. 2	-	-	-	-	-	-	-	
xPatch Right Dev. 1	73.11%	61.34%	57.67%	36.56%	31.11%	32.47%	64.72%	1.43
xPatch Right Dev. 2	76.90%	50.56%	35.97%	35.20%	20.25%	40.30%	58.29%	
xPatch Left Dev. 1	81.12%	63.61%	55.09%	20.82%	10.50%	350.10%	61.26%	2.14
xPatch Left Dev. 2	72.29%	64.53%	54.61%	21.74%	18.28%	123.52%	64.50%	

Table 4.5.
The mean absolute value of the errors of device reported PAA varied by both device and impact location.

	PAA Mean Absolute Value of Error							
	A	A'	B	C	D	E	F	Average Rank
HITS Dev. 1	59.91%	41.42%	43.15%	34.46%	43.61%	159.97%	37.76%	2.14
HITS Dev. 2	56.64%	63.47%	47.99%	22.75%	49.97%	95.69%	86.06%	
Shockbox Trial 1	-	-	-	-	-	-	-	-
Shockbox Trial 2	-	-	-	-	-	-	-	
SIM-G Dev. 1	-	-	-	-	-	-	-	-
SIM-G Dev. 2	-	-	-	-	-	-	-	
xPatch Right Dev. 1	72.75%	59.41%	42.38%	23.05%	27.08%	26.85%	63.53%	1.57
xPatch Right Dev. 2	76.58%	49.61%	32.28%	28.18%	17.32%	32.45%	57.93%	
xPatch Left Dev. 1	80.93%	63.19%	54.68%	18.58%	9.48%	245.65%	60.86%	2.29
xPatch Left Dev. 2	71.99%	63.44%	54.03%	19.58%	12.78%	52.18%	63.26%	

4.2.3 Game Data

To further demonstrate the differences between two of the sensor packages, xPatch and HITS, data were analyzed from a single football game.There was HITS data for twelve players and xPatch data for fifteen. The number of impacts recorded by each system were 224 and 231 for HITS and xPatch, respectively. However, the distribution of these impacts differed substantially between the two systems (Fig. 4.7). HITS had a tendency to record impacts in the bottom bin (20g to 40g), while xPatch acquired a much broader distribution of impacts. HITS reported only 5 of the 224 recorded blows (2.23%) as being above $80g$, whereas xPatch reported 24 of the 231 recorded blows (10.4%) as being above this level.

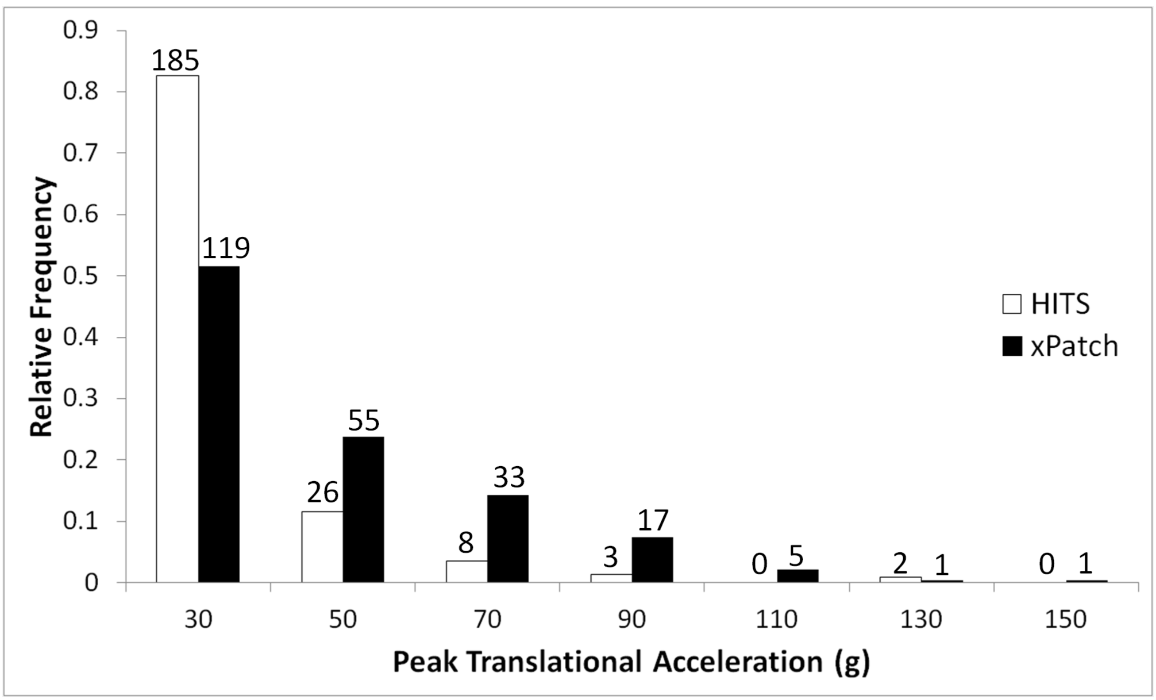

Figure 4.7. HITS and xPatch PTA impact distribution for one high school football game. The dissimilar distributions indicate that the two sensor packages produce visibly different sets of measurements in a real-world situation.

4.3 Discussion

The goal of this study was to evaluate a range of helmet-mounted and head-mounted devices used to measure head accelerations using a common testing protocol. Their accuracy was determined using two methods. The first method employed the absolute error to provide an assessment of average sensor accuracy at a given location. The second method was the RMSE, which provides a measure of the average reliability of any particular measurement. Devices with low RMSE values are likely to provide good measures of individual head impacts which would be required to determine what types of hits were sustained just prior to a diagnosis of concussion. Based on the data obtained herein, the head-mounted systems out-performed the helmet-mounted sensors in the measurement of PTA and PAA.

Few studies have attempted to measure the accuracy of head acceleration measurement systems, but of the systems commercially available, HITS has been studied the most [13] [28] [58] . One such study demonstrated much smaller errors [58], but placed a medium helmet on the same Hybrid III headform used herein. While it is technically possible to fit a medium helmet on the headform, it is challenging and requires much greater contact pressure than those generated by proper fitting of a helmet [28]. Utilizing a large helmet on the Hybrid III headform most closely represents the proper fitting of the helmets, justifying its use for this study. It should be noted, however, that the larger helmet allows for a small amount of relative motion between the helmet and the head. Therefore, it is not surprising that the head-mounted devices, which are not affected by helmet fit, were better able to reproduce the motion of the head's center of mass and would be likely to provide better data in practice and game situations.

A strength of the present study was the use of an impulse hammer, resulting in blows to the head that were not uniformly through its center of mass. Use of the hammer also allowed us to easily deliver oblique impacts. The resulting range of normal to oblique blows is certainly more consistent with the variety of impacts occurring in a game situation. The data collected in previous studies on HITS were performed with a hydraulic linear impactor that delivered the blows through the head center of mass [28] [58]. For the purposes of comparison, region C in this study corresponded to region C in Jadischke et al. (2013). In the current study, the RSME for the PTA of the two HITS devices were 60.6% and 60.3% while the Jadischke study measured RMSE value of 11.3% and 19.0%.

Shockbox did not perform as well as the other helmet-mounted system, HITS. Several of its impacts were labeled as $> 150g$ despite the fact that none of the blows were reported by the headform to have approached that level of acceleration. For the purposes of analysis, these impacts were assumed to be exactly $150g$, yielding a conservative estimate of the total error. It should be noted, however, that the large number of these inaccurate recordings resulted in a particularly high RMSE. It is possible that the contact made between HITS and the player's head limits motion to some degree, but without a detailed description of the algorithm used by each system it is difficult to discern the source of the differences.

The xPatch data reflected a much more accurate depiction of the accelerations of the Hybrid III headform's center of mass. Most notable was the discrepency between the devices' abilities to accurately describe the impacts between right and left locations. The data indicate that there is a fundamental issue with using the xPatch behind the left ear, but the cause of this is not currently known. It is possible that the mapping is more accurate for the right patch than the left one, but regardless of the reason, we recommend only using the xPatch behind the right ear.

The data collected for the SIM-G device also demonstrated improvement over the helmet-mounted devices. While it did not perform as well as the xPatch devices in RMSE average rank, it outperformed the left xPatch in mean absolute value of error average rank and the helmet mounted devices in both statistics. The headband-based mount appears to minimize the relative motion when held in place by a football helmet. It should be noted, however, that our group's past experience with headband-mounted sensors indicated that removing and repositioning the headband can generate false impact readings. This study did not address this concern, but it would be worth investigating whether those motions have characteristics that would allow them to be filtered.

This study originally planned to include Brain Sentry in addition to the other five sensor packages, but both purchased devices failed early in the testing process.

With regards to the in-game data, the noted difference in reported fraction of impacts above the $80g$ threshold could meaningfully affect future determinations of safe and unsafe levels of contact exposure based on which system's data is being considered.

Among the devices examined here, only HITS and xPatch attempt to designate whether a hit was legitimate or not based on the characteristics of the acceleration waveform, although it should be noted that neither algorithm was supplied. Unfortunately, the only way to characterize the accuracy of such an algorithm is to record head impacts delivered to athletes in practice and game situations and then compare each one to video recordings in order to quantify the system's accuracy. Another area that is in need of exploration pertains to devices that do not have a clear mounting location. For devices such as the xPatch and Shockbox, there is variability with the precise location to which the device is applied. An investigation ought to be performed studying the effect of slight perturbations of these devices relative to their defined ideal locations, particularly for head mounted devices applied to heads of variable size. Compared to the RMSE reported here, however, small changes in position are not likely to affect the accuracy substantially. Another way to look at device reliability would be to look at more than two of each device.

It should also be noted that many sensor packages claim to be able to accurately locate the point of application of contact force, although potentially damaging acceleration events can be caused without direct head impact by blows to the body that produce a whiplash type event. Even if it is known that a force was delivered to the head, there is a fundamental limitation to what information can be gleaned about the force. When a force is applied to the head, there is necessarily a concomitant force and moment at the neck (Fig. 4.8). There are therefore three unknowns associated with each of the following vector quantities: translational acceleration, angular acceleration, impact force, position vector to the point of impact, resultant force at the neck, and resultant moment at the neck. It is possible to reduce this set of quantities from 18 unknowns to 12 using Euler's equations to relate the translational acceleration to the applied forces and the angular acceleration to the sum total of the applied moments. If one further assumes that the geometry of the helmet is known then only two components are required to unequal determine the location of the blow, reducing the number of unknowns to 11. Accurately measuring the six acceleration components leaves us with five unknowns. Consequently, it is not possible to uniquely determine the location or the magnitude of the impact (or whether the head was even struck at all) without additional data.

Overall, future head impact sensor design should focus on a head mounted sensor design. Improvements may be possible through the use of tandem accelerometers in different locations to provide redundancy and reduce errors. Higher sampling rates would also be beneficial to data integrity, though power consumption is a genuine concern in these portable devices. Future work needs to refine the algorithms used to estimate the translational and angular accelerations and examine technologies that better locate the source of the impact.

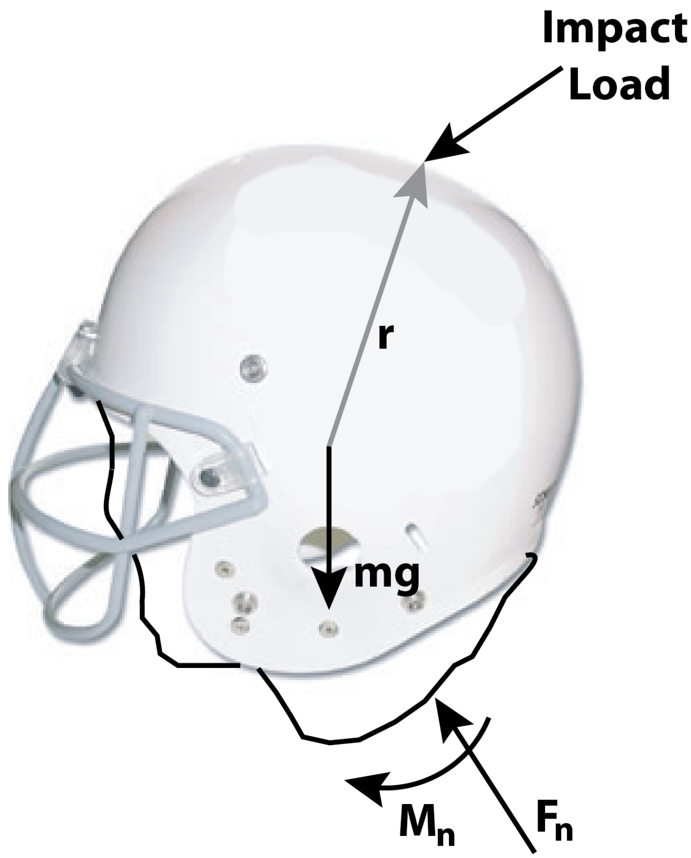

Figure 4.8. Free body diagram of a helmeted head suffering an impact. At the instant of impact, there are 18 unknowns: 3 components each of translational acceleration, angular acceleration, impact force, position vector to the point of impact, resultant force at the neck (F_n), and resultant moment at the neck(M_n). Euler's equations eliminate six of these unknowns and, if the dimensions of the helmet are known, one component of the position vector to the point of impact, $\underline{r}$, can be eliminated, leaving 11 unknowns. Current head acceleration measurement technologies measure at most six components. Consequently, it is not possible to uniquely determine the location or the magnitude of the impact (or whether the head was even struck at all) without additional data.

5. HELMET EVALUATION

5.1 Methods

This study consisted of the evaluation of four varsity football helmet models, each made by a different manufacturer. The four helmet models used were all manufactured in either 2014 or 2015: Revolution Speed (Riddell; Rosemont, IL), AiR XP PRO (Schutt; Litchfield, IL), X2E (Xenith; Lowell, MA), and SG (SG Helmets; Brownsburg, IN) (Fig. 5.1). The use of Large size helmets was based on a previous study that, when fastened to a 50^{th} percentile H3H, demonstrated resting peak pressure values at levels within the range determined to be tolerable to football players [28].

Figure 5.1. Four helmet models were used in this study: A) Revolution Speed (Riddell; Rosemont, IL), B) AiR XP PRO (Schutt; Litchfield, IL), C) X2E (Xenith; Lowell, MA), and D) SG (SG Helmets; Brownsburg, IN).

5.1.1 Data Collection

Impacts were administered using a modally tuned impulse hammer (PCB Piezotronics, Inc.; Depew, NY) to selected regions of the helmet or headform. Nine regions were chosen for testing, and 20 impacts were administered to each location for a total of 180 impacts per helmet (Fig. 4.2). Three of each model were tested for a total of twelve helmets. Likewise, the bare headform was tested three times using the same protocol. At each impact site, impacts were defined to occur within certain force ranges that spanned from 200 lbf to 1200 lbf so as to administer a similar set of impacts across sessions. This was done to ensure fair comparison of helmets.

During each impact, the transient force data at the tip of the hammer would cause an event to be generated above a threshold of 10 pounds of force. During one of these events, nine accelerometers (one triaxial and six uniaxial) arranged at or around the CoM of the headform using the 3-2-2-2 configuration [64] produced acceleration traces. Model 9234 data acquisition modules (NI; Austin, TX) were used to measure all ten channels of data over coinciding time domains, and control of the hardware was accomplished using a custom software package. Data was sampled at 5120 Hz, well above the Nyquist rate of these impacts.

In order to evaluate each helmet over a heterogeneous distribution of impact conditions, nine locations were chosen. The orientation of delivered impacts was approximately normal to the surface of the helmet at the given location with the exceptions of A' and D', both of which were administered at an angle of inclination of 45 deg and will herein be defined as the oblique locations. The rotation of these impacts was such that the impacts originated from the right side of the head for A' and from the back of the head for D'. Impacts at location G were performed to the center of the facemask with horizontal directionality, as shown in Figure 5.2.

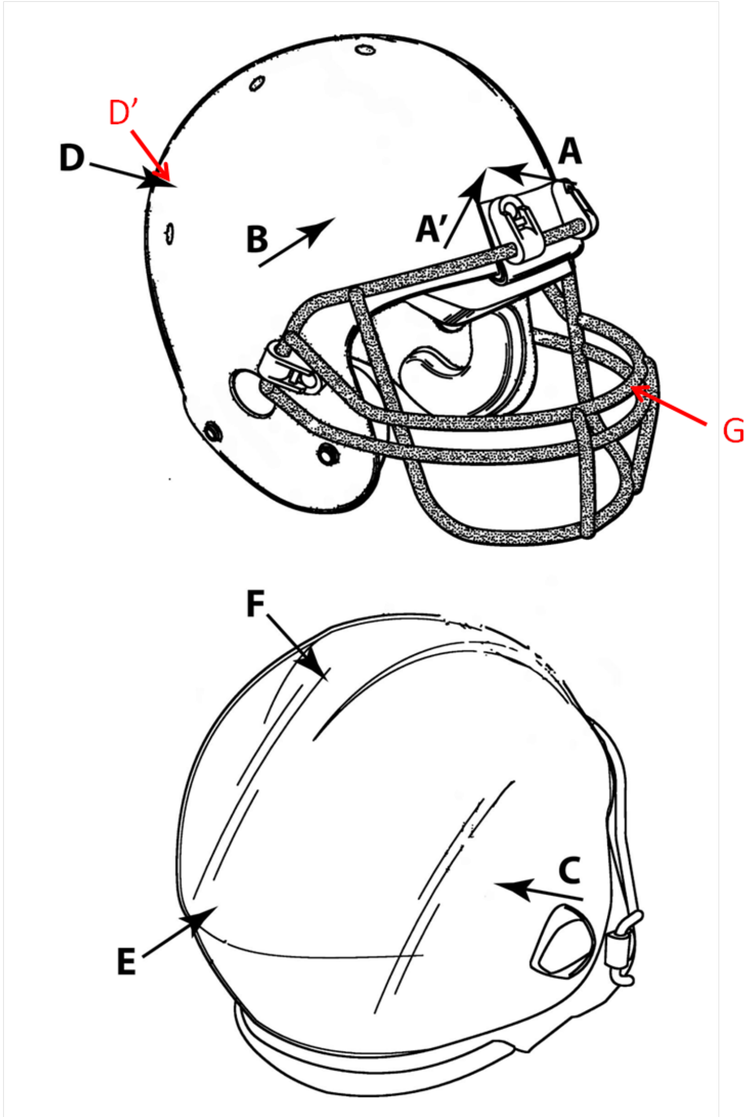

Figure 5.2. Nine impact locations were used in the current study. Impacts at locations A-G were administered at an orientation normal to the helmet surface, while those at A' and D' were administered at an oblique orientation of approximately 45°.

5.1.2 Post-Processing

Data acquired was stored in files for later post-processing. The acceleration data was processed using a low pass Butterworth filter with a cutoff frequency of 750 Hz.

The filtered acceleration traces were then used to kinematically determine transient angular acceleration components [64] and resultant translational and angular accelerations, all at the headform's CoM.

Impacts were defined as the duration during which the force was in its peak, beginning 1ms before the force reached 10 lbf and ending when the force had reached over 100 lbf and then fallen below 10lbf. This time domain is illustrated as red veritcal lines (Figs. 5.3, 5.4). The PTA and PAA, as well as the peak force (PF) and impulse (IM), were all calculated during this defined impact. IM was calculated using the trapezoidal method of numerical intergration by integrating the force data over the corresponding time data.

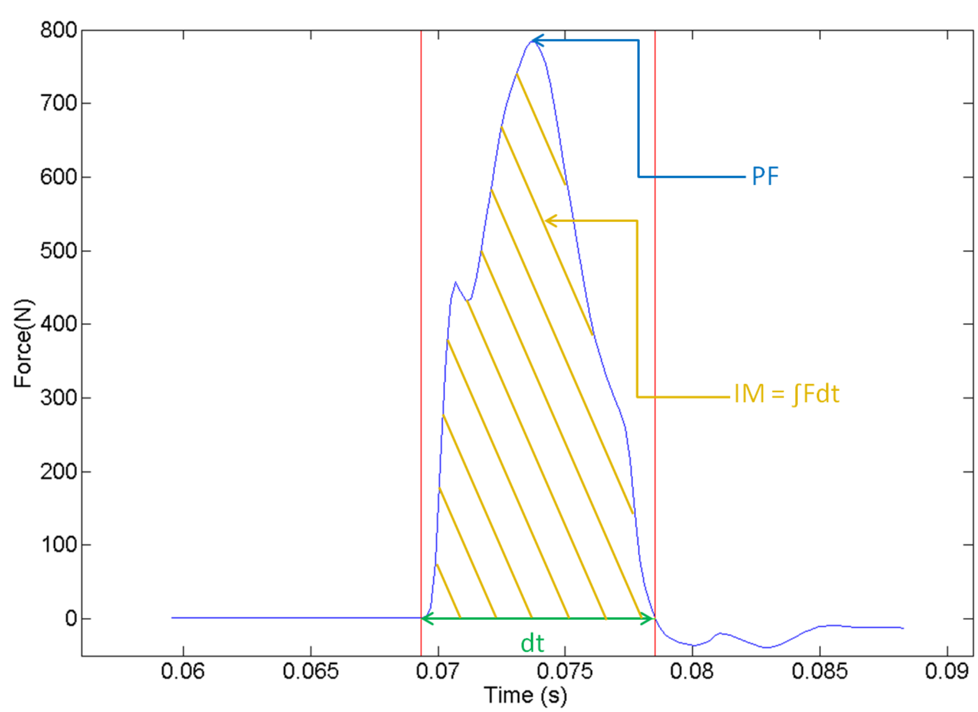

Figure 5.3. The impact administered was triggered when a force threshold reached 10 lbf and was cut off once the force fell below 10 lbf. The time domain of time impact was determined, and the IM was then calculated over it. The PF within this time domain was also determined.

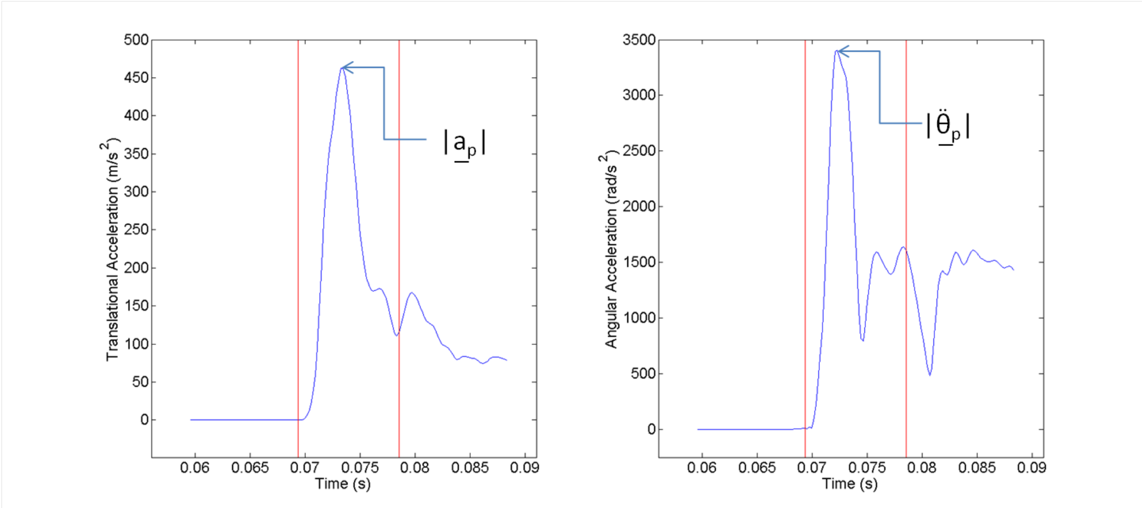

Figure 5.4. The accelerations measured by the H3H allowed for the calculating of PTA from the translational acceleration (left) and the PAA from the angular acceleration (right). These peak values were calculated during and always occurred within the same time domain as was defined for the impact.

5.1.3 Helmet Evaluation

Metrics used in evaluating the impacts include several ratios. All of these metrics relate the peak acceleration, whether PTA or PAA, to the IM. The first ratio, shown in Equation (5.1), divides the PTA measured at the CoM of the heaform, $|\underline{a_p}|$, by the IM that caused it, which is calculated by integrating the force F measured at the tip of the impulse hammer over the duration dt of the impact. The second ratio, shown in Equation (5.2), does the same with PAA, $|\underline{\ddot{\theta}_p}|$.

$$R_1 = \frac{|\underline{a_p}|}{\int F dt} \tag{5.1}$$

$$R_2 = \frac{|\underline{\ddot{\theta}_p}|}{\int F dt} \tag{5.2}$$

In order to normalize the ratios for each location, the metrics previously defined for helmet impacts were compared to the same metrics measured for the H3H impacts without a helmet. The average value for each of these was defined for the bare H3H as a datum in Equation (5.3) where a is the base metric from Equations (5.1) and (5.2) used, X is the location of the impact, and n is the number of impacts at location X (60 for the combined three trials). The metrics from each helmeted impact are then divided by their corresponding locational H3H average value datum in Equation (5.4).

$$\bar{RH}_{a,X} = \frac{\sum_{i=1}^{n} R_{a,X}}{n} \tag{5.3}$$

$$R_{a',X} = \frac{R_{a,X}}{\bar{RH}_{a,X}} \tag{5.4}$$

5.1.4 Human Subjects

All research conducted in the current study was approved by an Institutional Review Board. Where applicable, parental consent and participant assent were obtained from study subjects. Two male high school football players were enrolled in the study for season six during Fall of 2014, and both participated throughout the entire season. Head impact data was collected using the xPatch device mounted on the head behind the players' right ears using a custom adhesive patch. Attention was paid to the players so as to ensure that the devices did not come loose and create artificial impacts, and impacts created outside of the start and end times of practices and games were cropped out from the data.

5.2 Results

The masses of the tested helmets are listed for the sake of reference (Table 5.1). The three models that utilized polycarbonate shells and a solid steel facemask were similar in mass, and the model that instead utilized a composite weave and a hollow, Chrome-Moly tubular facemask weighed roughly 40% less than the other three.

Table 5.1.
The masses were similar for the three mainstream helmets, but the SG helmet was much lighter.

Helmet	Mass 1 (kg)	Mass 2 (kg)	Mass 3 (kg)	Mean Mass (kg)
Speed	1.84	1.82	1.84	1.83
Air XP PRO	1.78	1.84	1.80	1.81
X2E	1.90	1.92	1.92	1.91
SG	1.08	1.12	1.12	1.11

5.2.1 Metric Analysis

The two metrics normalized by the H3H, calculated for each of the four helmets at each of the nine locations, provide values that allow for comparison of the helmets. Due to what these metrics represent, namely peak acceleration measured given an applied input, lower values represent superior mitigation ability. Within a metric, rank was assigned to the four helmets for a given impact location. The ranks for a helmet at all nine locations were averaged to give the average rank.

Looking at the normalized metric one (Table 5.2), rank was not consistent across locations. At location A, both the Air XP PRO and SG exhibited statistically significantly ($p < 0.05$) smaller metric values than the Speed but likewise statistically significantly larger values than the X2E. When looking at the average rank for the three helmets, the X2E was the only one to stand out from the other three with the lowest of the four.

Table 5.2.
The metric $R_{1'}$, which has been normalized by the H3H mean on a per location basis, depicts the fraction of PTA experiened by the headform while wearing the helmet.

	$R_{1'}$ means									
	A	A'	B	C	D	D'	E	F	G	Average Rank
Speed	0.46	0.44	0.40	0.34	0.43	0.54	0.39	0.46	0.57	2.78
Air XP PRO	0.39	0.43	0.39	0.47	0.46	0.55	0.40	0.54	0.45	2.89
X2E	0.31	0.38	0.38	0.39	0.41	0.49	0.43	0.44	0.56	1.67
SG	0.41	0.40	0.35	0.45	0.47	0.50	0.48	0.56	0.39	2.67

Rank again had little trend for the normalized metric two (Table 5.3). One notable trend was that the SG received the worst rank for six of the nine locations. There was less clustering for the average rank as compared to $R_{1'}$, and the X2E again had the lowest.

Table 5.3.
The metric $R_{2'}$, which has been normalized by the H3H mean on a per location basis, depicts the fraction of PAA experiened by the headform while wearing the helmet.

	$R_{2'}$ means									
	A	A'	B	C	D	D'	E	F	G	Average Rank
Speed	0.71	0.34	0.34	0.35	0.40	0.75	0.68	1.46	0.26	2.67
Air XP PRO	0.48	0.27	0.29	0.48	0.50	0.80	0.48	0.75	0.24	2.22
X2E	0.46	0.23	0.29	0.39	0.41	0.56	0.56	0.90	0.31	1.89
SG	0.71	0.35	0.27	0.61	0.60	0.87	0.70	1.71	0.23	3.22

A visualization of the helmets' performances relative to the H3H was created by plotting the two unnormalized metrics against their corresponding IMs (Figs. 5.5 - 5.8). The H3H data shows a tendency for both metrics to increase as a function of the input IM, whereas the helmets tend to show fairly flat trends if any exist at all. The helmets also tend to have lower metric values at a given IM than does the H3H (Figs. 5.5 - 5.7), but this was not always the case (Fig. 5.8).

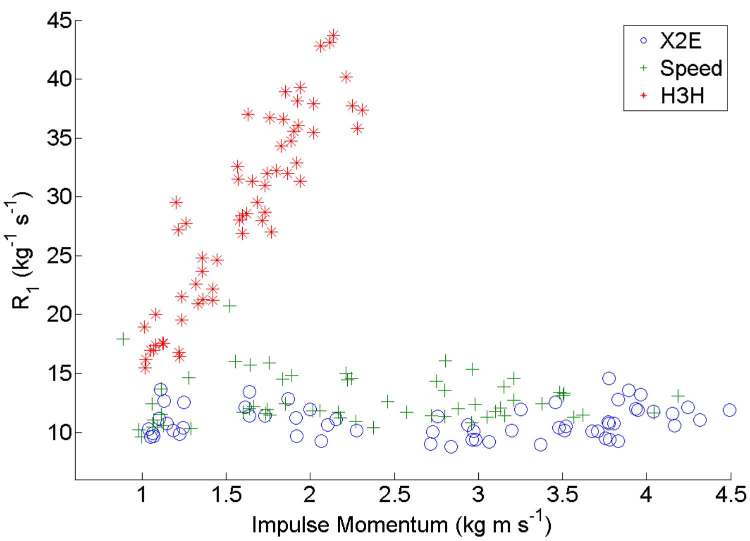

Figure 5.5. Location A' impacts show a relatively flat R_1 distribution as a function of IM for both the Speed and X2E, with the X2E lying slightly below the Speed. The H3H shows a clear linear trend.

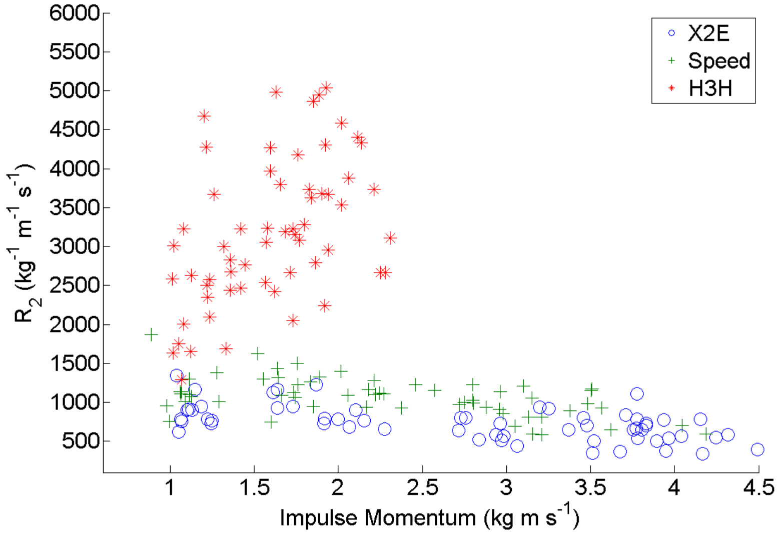

Figure 5.6. Location A' impacts show a relatively flat R_2 distribution as a function of IM for both the Speed and X2E. The H3H shows a linear trend with a good bit of variability.

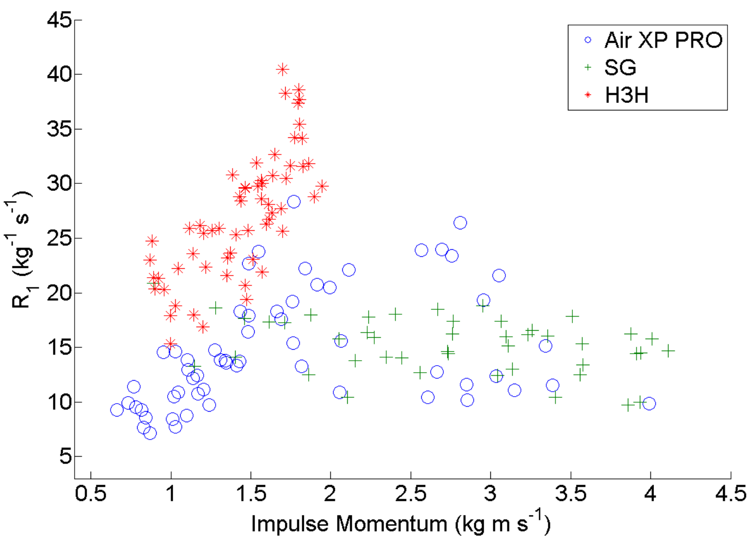

Figure 5.7. Location F impacts show a R_1 distribution as a function of IM that is slightly positively correlated for the Air XP PRO, more so positively correlated for the H3H, and fairly flat for the SG.

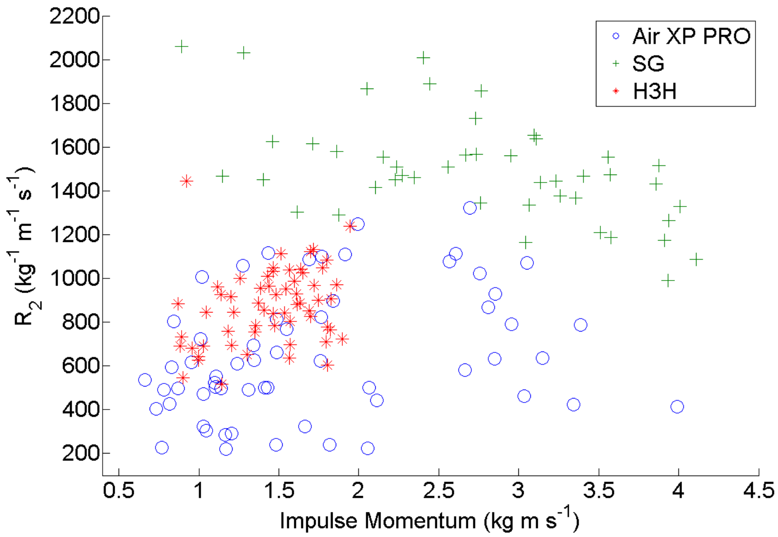

Figure 5.8. Location F impacts show a R_2 distribution as a function of IM that is positively correlated for both the Air XP PRO and H3H, and negatively correlated for the SG. At a given IM, the SG has a higher R_2 as compared to the Air XP PRO and H3H.

5.2.2 Ecological Validity of Testing

The SG Helmets experienced significant degredation over the course of the testing process (Fig. 5.9 D). This was consistent across all three devices, while none of the other helmets tested experienced more than a pair of small cracks.

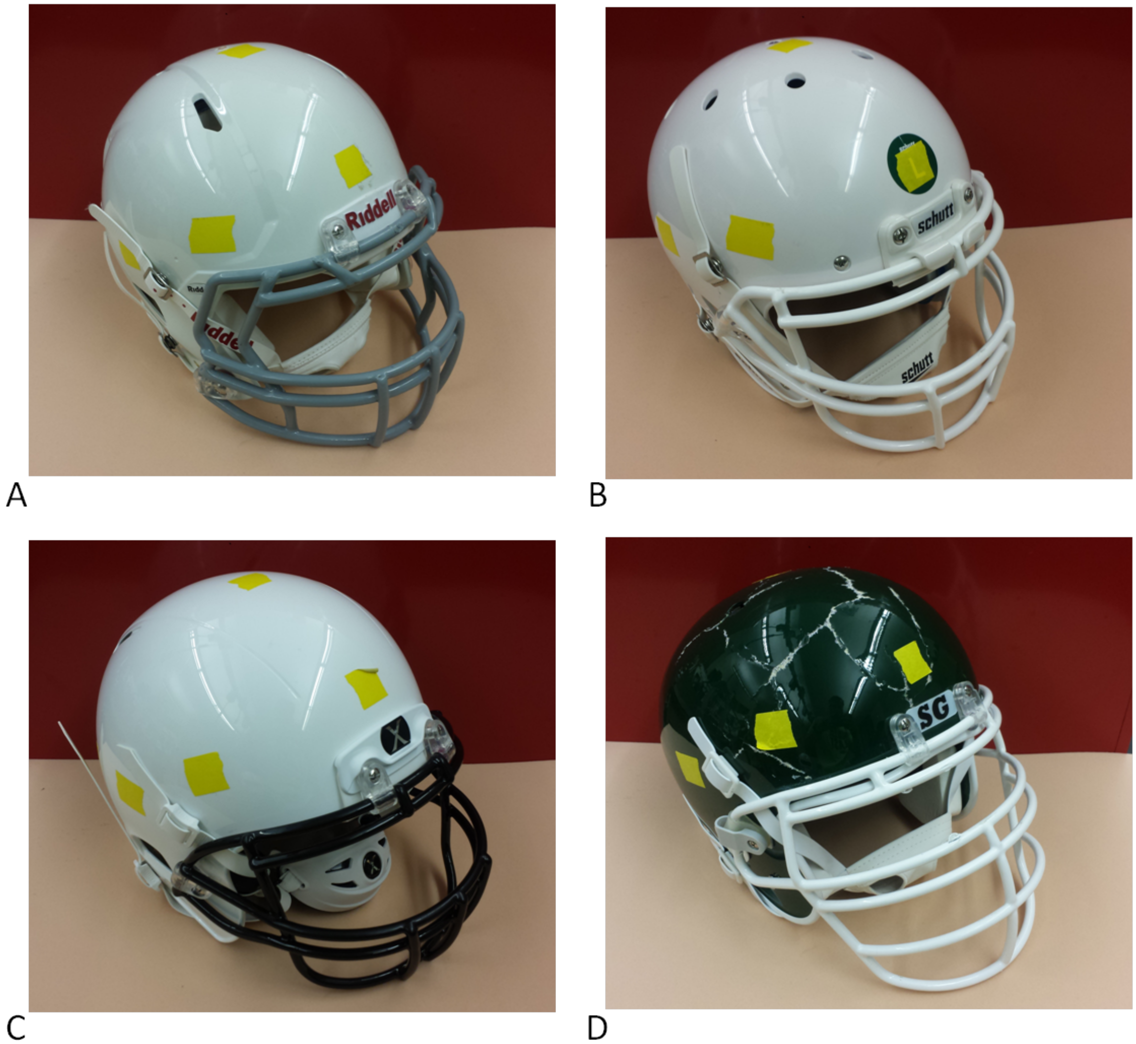

Figure 5.9. The four helmet models used in this study after one session of impacts each: A) Revolution Speed, B) AiR XP PRO, C) X2E, and D) SG. The SG helmets sustained the most damage, and the next to worst was a pair of cracks initiating from a rear vent on one of the X2E helmets.

In order to determine if the loading on the helmet is appropriate in representing that which a helmet worn by a football player would take, a histogram of PTA data was created (Fig. 5.10). This histogram contains the head impact data for the two high school football players, one of whom was chosen due to his high rate of impacts and the other for his more typical rate. These were compared to the impact distribution of one of the two SG helmets that survived the entire session. Save for the bottom bin which cropped out impacts collected by the xPatch and thus had zero for both players, the SG helmet was exposed to fewer impacts within each bin compared to even the typical high school football player in the study.

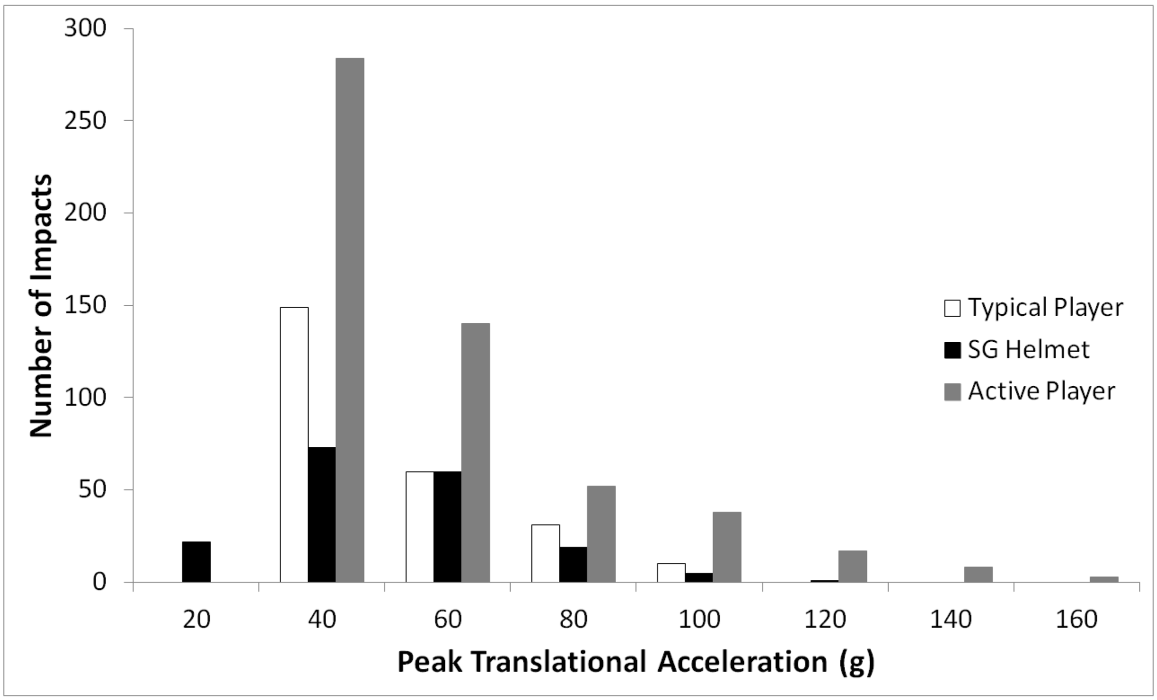

Figure 5.10. The SG helmets were exposed to a less severe set of impacts than both the active and typical players in every bin except for the one below which impacts were cropped out for the xPatches. The set of impacts administered to the second SG helmet is therefore on the conservative side of the spectrum of impacts experienced by helmets in the field.

5.3 Discussion

The goal of this study was to evaluate four commercially available varsity football helmets in their ability to mitigate impacts through the measurement of IM inputs and acceleration outputs. The two defined metrics for the helmets, once normalized by the H3H data, convey a helmet's ability to reduce the severity of an impact compared to what it would be if administered to the H3H. For these metrics, a value of one would imply that the helmet is neither increasing nor decreasing said severity. The fact that most of the metric values were below one suggest that the helmets are accomplishing their intended purpose, though the degree to which they do so is dependent on both the helmet model and the impact location.

While previous studies have evaluated the performance of football helmets under highly specified impact conditions [26] which were determined through reconstruction of impacts that caused concussion in the NFL [24], it has since been determined that several subconcussive impacts can have significant effect on brain function [17] [18] [19]. Likewise, while a previous study attempted to categorize concussion-causing impacts analyzed by video evidence [25], impacts to helmet shells can occur from oblique orientations as well as the previously investigated lateral impacts. The current study covered both oblique impacts and a wider array of impact locations than previously investigated, and did so over a range of impacts that are relevant to the study of subconcussive impacts experienced by players that don't directly result in concussion.

The metrics computed determiend that the helmets provided reduction of the severity of helmet impacts in all situations except for the PAA resulting from impacts at location F. While a helmet ought to have a metric that is less than one, which signifies a less severe response to an impact as compared to the response of a bare headform, both the Speed and the SG resulted in metrics above one. These suggest that the helmets are making these impacts more severe. While this may seem surprising, the additional distance from the base of the H3H neck to the location of the administered impact provide for a larger moment arm to be made and thus slight deviations from impacts perfectly axial to the neck may result in larger moments about the base of the neck. These would in turn explain the larger PAA values that resulted in unexpected metric values.

The helmets evaluated reflect their respective designs. In order to pass the National Operating Committee on Standards for Athletic Equipment (NOCSAE) tests and certify one's product as NOCSAE certified, these helmets must meet certain requirements related to the maximum Gadd Severity Index (GSI) which is derived from the translational acceleration. As such, it comes as no surprise that three of the helmets yielded a clustered set of average ranks for $R_{1'}$, which was based on the translational acceleration and thus related to the design criteria for these helmets in order that they could be certified. However, the same three helmets yielded a set of average ranks for $R_{2'}$, which was based on the angular acceleration and thus has no significance to their being certified, that spanned an entire rank point. This suggests that helmet designs often cater to the satisfying of certification criteria. The X2E had the lowest average rank for both metrics, potentially reflecting the increased use of viscoelastic materials rather than traditional hard foams and plastics. However, it is possible that the reduction in severity of the impacts administered to these helmets is primarily due to the additional inertia added to the head-helmet system by wearing a helmet. While it is surprising that the more compliant SG helmet had the worst performance, but it is possible that this was not because of poor design but instead because of lower mass compared to the other three helmets.

The set of impacts that were administered in this study were not unusual compared to the impacts recorded for players throughout the six seasons that Purdue Neurotrama Group has been collecting high school football head impact data. As such, a helmet ought to be able to endure the set of 180 impacts administered throughout its respective session. None of the Air XP PROs experienced any damage to the shell, though cheek pads had a tendency to detatch from the snaps that held them. One Speed developed a small crack near one of its vents, and all three X2Es did the same, though these were due to impacts administered to Locations D and D' which were near to vents on both helmet models. However, all three SG helmets developed severe fractures throughout the shell during testing. These fractures did not necessarily originate at any sort of vents or other helmet features, but would begin and propogate in the middle of continuous shell region. While anecdotal evidence has praised the technology that these helmets are based off of, said racing helmets are designed to take a single blow and fracture to absorb maximal energy and protect the wearer's head from a catastrophic blow to the head. They appear not, however, to be designed for the purpose of taking repeated impacts of a more moderate severity. The fractures developed in the SG helmets give reason to reevaluate the helmet's use in football, particularly at the high school level which has been documented to take even more severe season impact sets than this study administered.

While the current study managed to evaluate impacts administered to a helmeted H3H compared to a bare H3H by means of matching the PF of impacts across sessions, the same PF for an impact administered to a helmet resulted in greater IM than when administered to the bare H3H due to the additional compliance of, and mass added by, the helmet. As such, the range of IM for the impacts administered to the H3H is in a narrower band than those administered to the helmets (See Figs. 5.5 - 5.8). While the trends are obvious to see in the range of IM that the two overlapped, benefit would be had by including impacts at higher IM ranges such that the bare H3H impacts spanned the same IM range as those of the helmets. Future helmet evaluations should make it a goal to evaluate a helmet's ability to reduce peak accelerations through the matching of IM, rather than matching PF or impact velocity. It would be of value to perform such evaluations on newer generations of helmets as they are released, in order to determine if subsequent design iterations provide measurable improvements over their predecessors. Likewise, it would be worth evaluating helmet design features incorporated into helmets of like additional translational and angular inertia when worn by the headform in order to determine which design features reduce the severity of the impact more than others. This would remove mass as a differentiating factor between helmet designs and allow for the design features to be cross-compared directly.

Modern helmet design has served to reduce the severity of impacts experienced by football players, but the dependency of this ability is highly dependent on impact location. In addition to retaining device integrity throughout a typical set of impacts, these helmets need improvement in their mitigation abilities, particularly the PAA due to impacts at Location F. Helmet designs that result in low metric values should be pursued such that peak accelerations experienced by athletes are reduced, with the end goal being the reduction of those athletes' cognitive impairments.

6. CONCLUSIONS

Sensor packages have come a long way in the last decade, from attaching a single triaxial accelerometer to an individual hockey player to outfitting entire teams with wireless devices capable of monitoring both translational and angular accelerations of a player's head. Further development is still needed, as even the most recent devices have shown to have large errors in translational and especially in rotational acceleration measurement. Future sensor design has several options for improvement. First off, sampling rates of 1000Hz, common to several of the devices available today, are insufficient to accurately reproduce the accelerations taking place during impact events. Next, sensor designs need to focus on measuring the kinematic response of the player's head and thus need to be directly attached to it, rather than being placed inside the helmet which has relative motion with respect to the head during these impacts. Last, redundancy of sensors arranged around a player's head with noncommon axes would allow for the reduction of errors through the averaging out of marginally conflicting data, particularly when it comes to angular acceleration and the errors associated with numerical differentiation of angular velocity measured by a gyroscope. However, major obstacles that need to be overcome for these to happen are either the reduction of power consumption by these devices or the increase of battery capacity, as the duration over which these devices must remain on to cover entire sessions is a nonnegotiable design criterion.

The use of helmets has also had a great impact since the days of football commonly causing skull fractures. While it is questionable as to how effective some of the padding materials in these helmets are when it comes to dissipating blows to the head, it is inarguable that helmets do a remarkable job of spreading an impact over a larger surface area so as to prevent breaking of bones. However, much of modern helmet design has sought to satisfy NOCSAE standards rather than using a measure of impact severity reduction that is relevant to the majority of subconcussive impacts taken by football players hundreds of times per season. With the up and coming research being performed on the neurocognitive impacts of these subconcussive impacts, motivation now exists to design helmets to serve the purpose of reducing more moderate impact severity in addition to the most severe of impacts. While most materials present in helmets today have very rigid material properties, investigation into materials with a much greater ability to dissipate the sudden onset of impacts ought to be undertaken. One challenge with such materials is to include them without adding signifiant weight or size to the helmet, but added weight in the padding could potentially be offset through the utilization of lightweight composites as a replacement to the polycarbonate shells currently employed. Another challenge is to design aforementioned materials with mechanical endurance on par with that of the much harder plastics currently used. The current standard is to get helmets reconditioned once every two years of play, and any helmet designs ought to strive for this same standard of durability.

Recent attention given to the study of the epidemiology and mitigation of head impacts has been warranted but incomplete. Continued investigation into the development of head impact monitors, specifically tied to neurocognitive evaluation through techniques such as neuroimaging, will enable for the improved determination of safe head impact exposure levels for football players. Likewise, furthering helmet technology will enable players to continue participating in the sport by lowering these exposure levels on a per-impact basis. Pursuing these two avenues as a means of improving player safety will reduce health care costs, improve the quality of life for participants, and play its part in solving the concussion crisis in which we currently find ourselves.

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APPENDIX

A. NINE ACCELEROMETER PROTOCOL

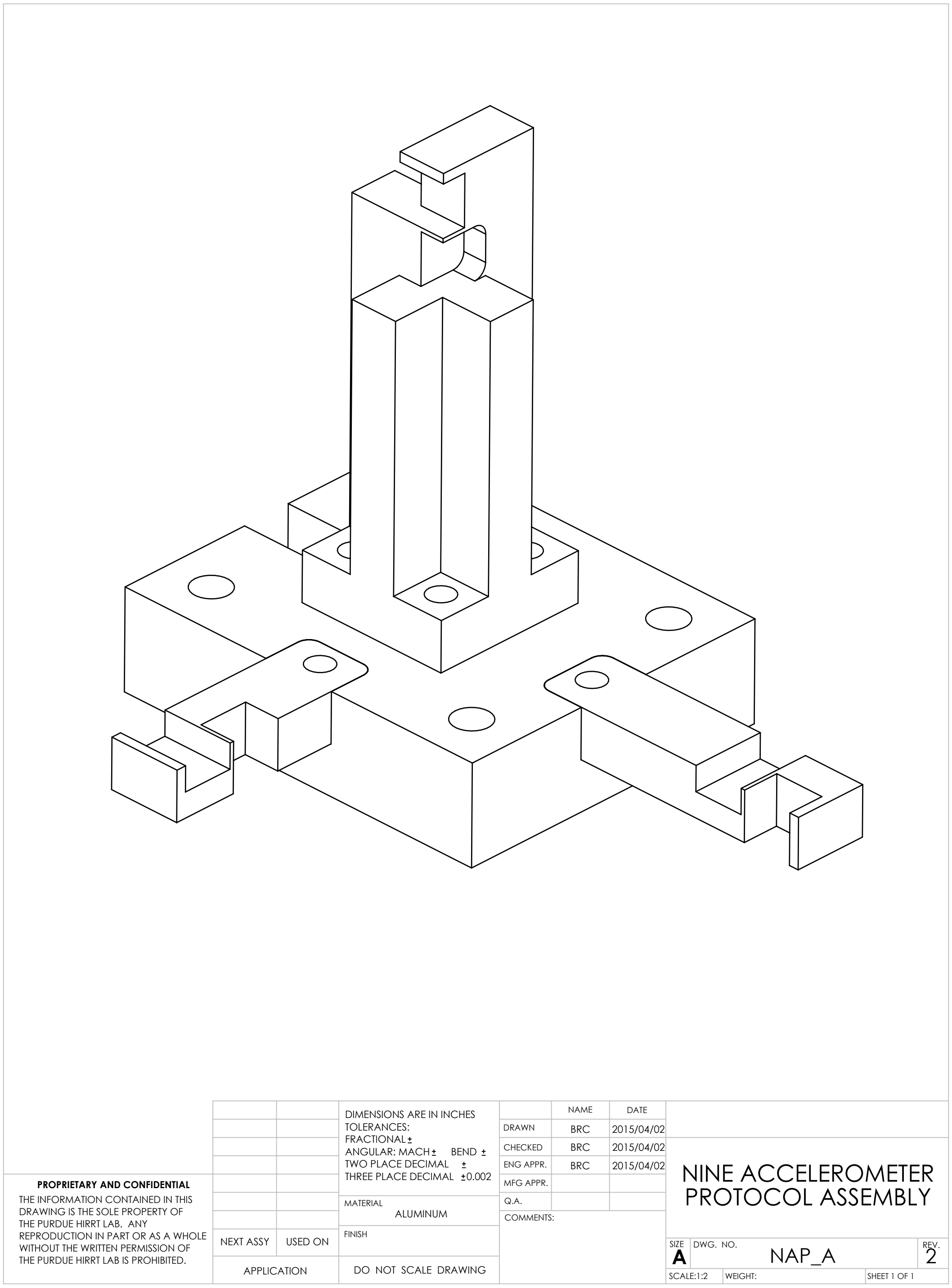

Figure A.1. A mount was created to position the nine accelerometer channels in the locations necessary to calculate the PTA and PAA at the H3H CoM.

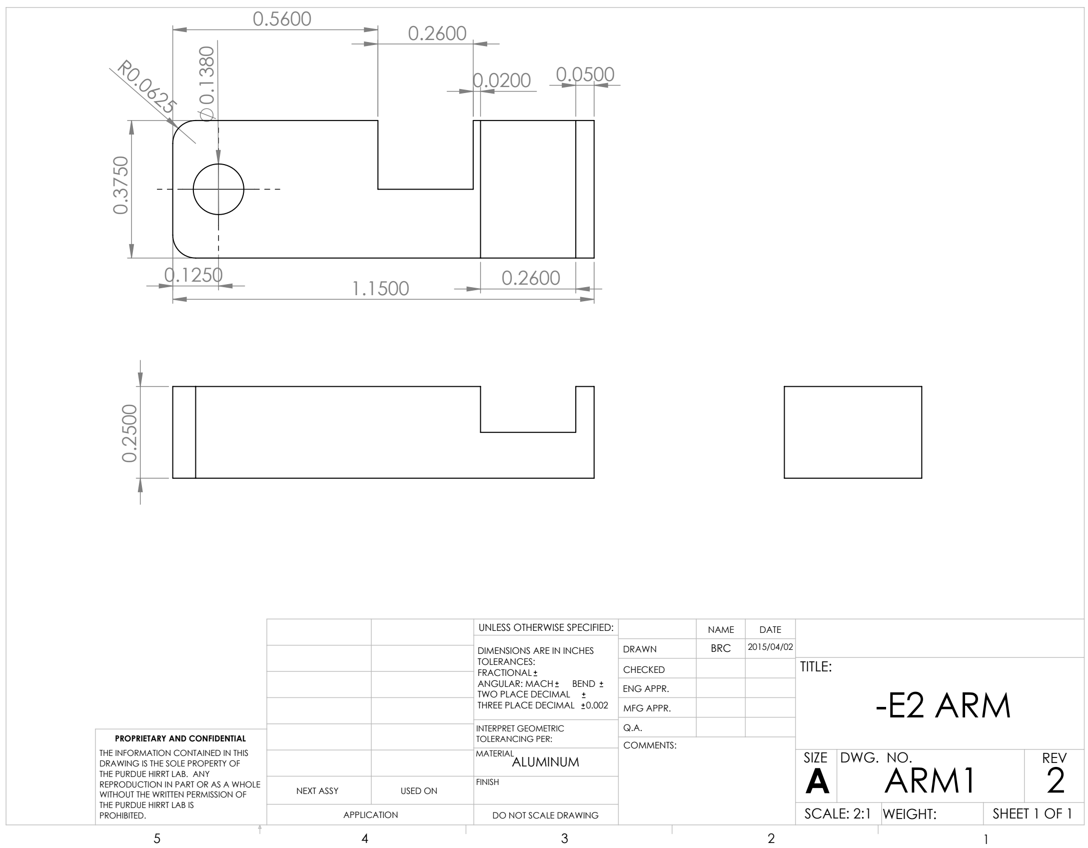

Figure A.2. The accelerometer mount arm in the $-E_2$ direction had slots for accelerometers that were oriented in the E_1 and E_3 directions.

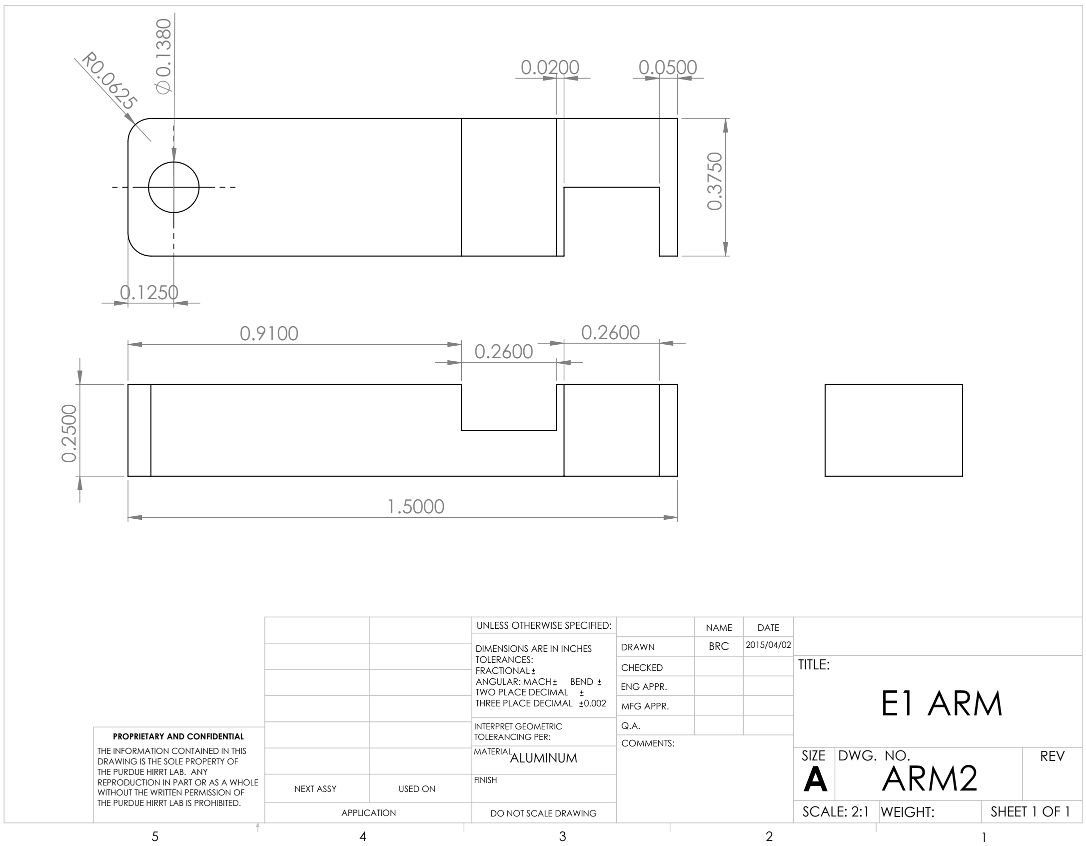

Figure A.3. The accelerometer mount arm in the E_1 direction had slots for accelerometers that were oriented in the $-E_2$ and E_3 directions.

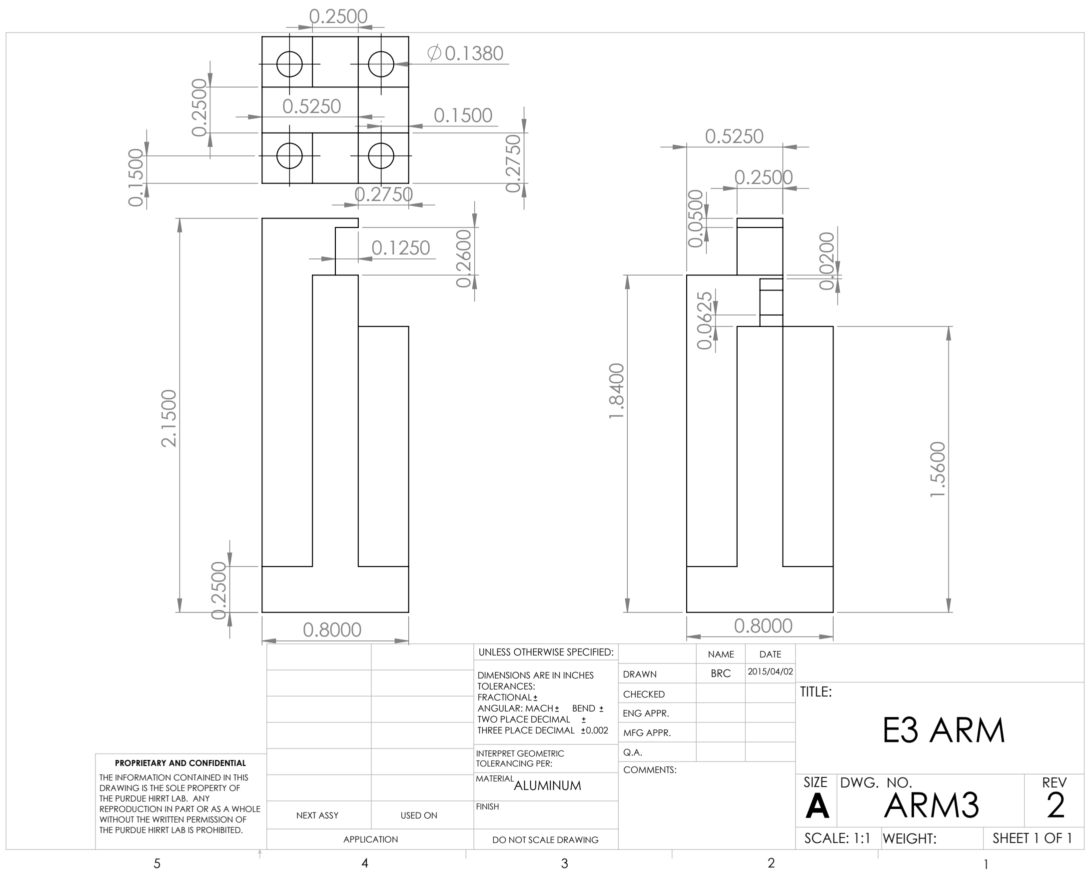

Figure A.4. The redesigned accelerometer mount arm in the E_3 direction had slots for accelerometers that were oriented in the E_1 and $-E_2$ directions.

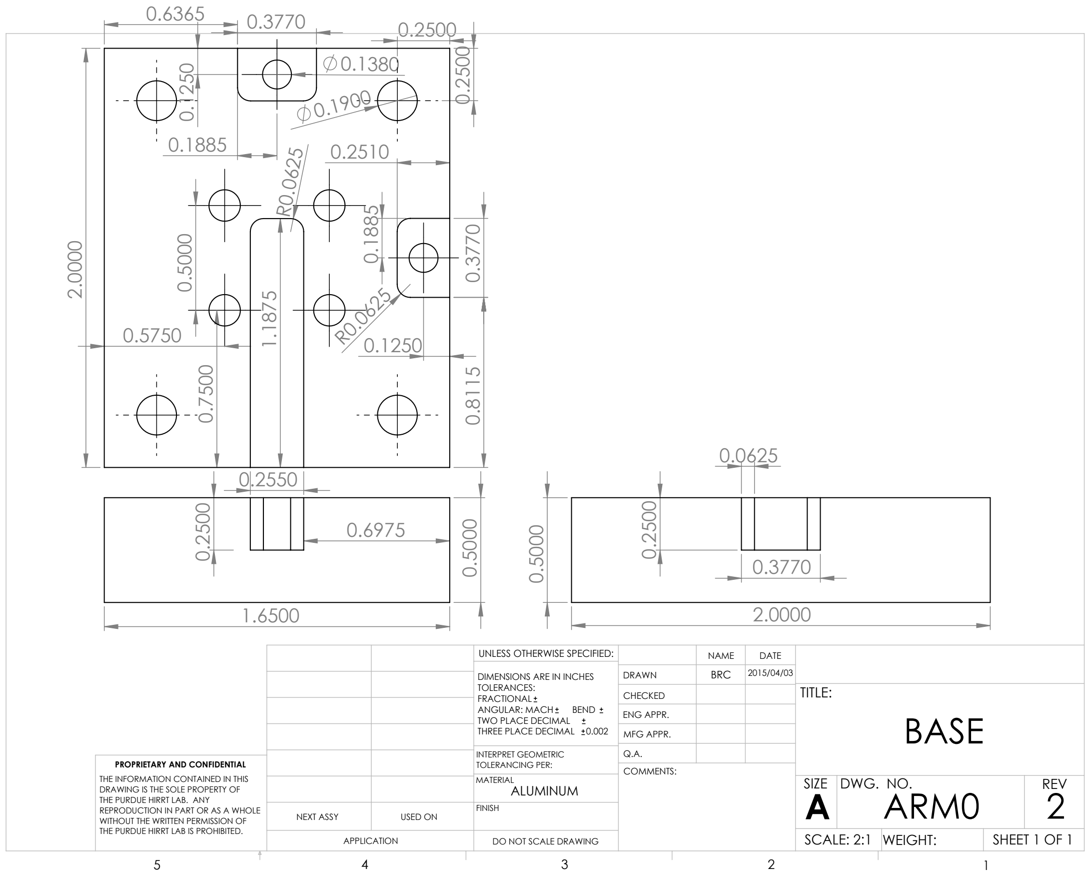

Figure A.5. The accelerometer mount base had locations for the securement of all three accelerometer arms, a slot for the triaxial accelerometer, and holes through which to fasten it to the H3H.