Linear and Angular Head Acceleration Measurements in Collegiate Football

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Each year, between 1.6×10^6 and 3.8×10^6 concussions are sustained by athletes playing sports, with football having the highest incidence. The high number of concussions in football provides a unique opportunity to collect biomechanical data to characterize mild traumatic brain injury. Human head acceleration data for a range of impact severities were collected by instrumenting the helmets of collegiate football players with accelerometers. The helmets of ten Virginia Tech football players were instrumented with measurement devices for every game and practice for the 2007 football season. The measurement devices recorded linear and angular accelerations about each of the three axes of the head. Data for each impact were downloaded wirelessly to a sideline data collection system shortly after each impact occurred. Data were collected for 1712 impacts, creating a large and unbiased data set. While a majority of the impacts were of relatively low severity (<30 g and <2000 rad/s^2), 172 impacts were greater than 40 g and 143 impacts were greater than 3000 rad/s^2. No instrumented player sustained a clinically diagnosed concussion during the 2007 season. A large and unbiased data set was compiled by instrumenting the helmets of collegiate football players. Football provides a unique opportunity to collect head acceleration data of varying severity from human volunteers. The addition of concurrent concussive data may advance the understanding of the mechanics of mild traumatic brain injury. With an increased understanding of the biomechanics of head impacts in collegiate football and human tolerance to head acceleration, better equipment can be designed to prevent head injuries.
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1 Introduction

It is estimated that between 1.6×10^6 and 3.8×10^6 sports-related concussions occur in the United States each year [1]. While mild in nature, about 300,000 of these involve loss of consciousness, with football having the largest incidence of any sport [2]. The high occurrence of concussions in football provides a unique opportunity to collect biomechanical data to characterize mild traumatic brain injury (MTBI). Several injury metrics are used to predict head injury; however, all the criteria use limited data from human volunteers. Head injury criterion (HIC), peak acceleration, and severity index (SI) are injury metrics derived from linear head acceleration and are primarily based on cadaver tests with skull fractures. Rotational acceleration injury thresholds are based mostly on primate tests with severe concussion, diffuse axonal injury (DAI), or intracranial bleed.

Competitive football was used as an experimental environment for collecting human head acceleration data since the 1970s. Several early studies have had football players wear headbands instrumented with accelerometers to measure head acceleration during football games [3–5]. Another study instrumented football helmets directly to measure helmet acceleration [6]. While laying the groundwork for future research and providing a proof of concept, these older studies were limited in their ability to measure head acceleration and measured only a single player. More recently, Naunheim et al. [7] instrumented the helmets of one high school hockey player and two high school football players (both linemen) with accelerometers to measure linear head acceleration. However, there were no documented incidents of mild traumatic brain injury in this limited data set.

One study has quantified head accelerations experienced by professional football players by recreating concussive impacts in a laboratory setting. The National Football League (NFL) reconstructed injurious game impacts using Hybrid III dummies based on a game video [8–10]. The authors recreated 31 impacts, 25 of which were concussive. From the data collected in the reconstructed impacts, injury risk curves were developed for MTBI. Nominal injury values determined in this study were a peak linear acceleration of 79 g, SI of 300, HIC of 250, and peak rotational acceleration of 5757 rad/s^2 [10,11]. The main limitation of this study is that the NFL data set is biased toward concussive impacts.

Several other studies have utilized a commercially available football helmet accelerometer system, head impact telemetry system (HITS, Simbex, Lebanon NH), to measure head accelerations. HITS is a six accelerometer measurement device that is integrated into existing football helmets. The measurement device records resultant linear head acceleration for every head impact a player experiences using a novel algorithm [12]. In addition, HITS reports the impact location and estimates of rotational accelerations on the x and y axes of the head.

Duma et al. [13] presented a study to quantify head acceleration in collegiate football players by collecting over 3000 impacts from 38 players using eight HITS measurement devices, in which one concussive event was measured. A subsequent study expanded this data set to include over 27,000 impacts (four concussions) and analyzed risk using a unique statistical analysis [14]. The nominal injury values reported, representing 10% risk of concussions, were a peak linear acceleration of 165 g and HIC of 400. In separate studies, Schnebel et al. [15] presented data for over

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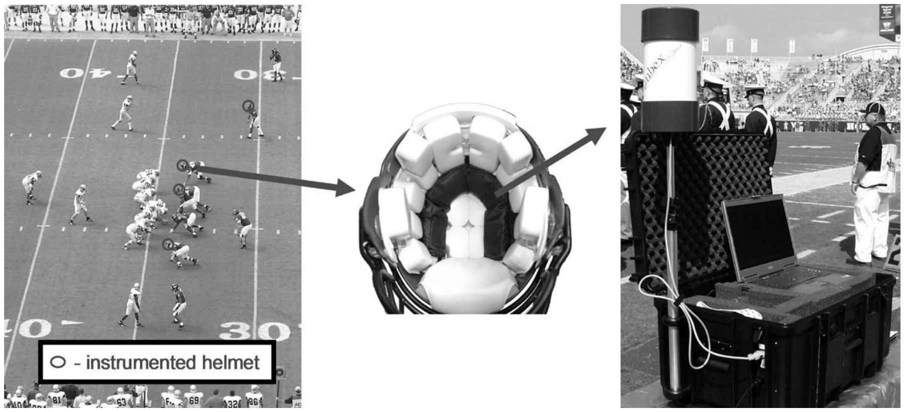

Fig. 1 Instrumented helmets communicate wirelessly with a computer on the sideline

62,000 head impacts (six concussions) experienced by 56 collegiate and high school football players recorded using HITS. Guskiewicz et al. [16] collected over 104,000 impacts (13 concussions) from 88 collegiate football players. Mihalik et al. [17] analyzed more than 57,000 impacts from 72 players to look at positional differences in impacts. Greenwald et al. [18] then compiled over 289,000 impacts (17 concussions) from 449 players at 13 organizations (seven college and six high school) to look at the predictive capabilities of a composite variable composing of linear acceleration, estimated rotational acceleration, HIC, Gadd severity index (GSI), and impact location. The authors found that the composite variable was more sensitive to the incidence of concussion than any single biomechanical measure. The main limitation of all these studies is that the HITS measurement device could not record the angular kinematics of the head during impact, which are thought by many to be a principal cause of brain injury [11].

The goal of this study was to utilize a newly developed measurement device to record six degree of freedom (6DOF) head accelerations for every head impact experienced by collegiate football players, producing a large and unbiased data set. Data collected in this experiment have applications in validating computational models and the development of injury risk curves.

2 Methods

A new 6DOF measurement device was developed, capable of measuring linear and angular accelerations for each axis of the head. The measurement device is compatible with the existing HITS technology used in previous studies [13,15–18]. The 6DOF measurement device was designed to be integrated into Riddell (Elyria, OH) Revolution football helmets. It is composed of 12 single-axis high-g iMEMs accelerometers (Analog Devices, Norwood, MA) that are encapsulated in the fabric padding. The accelerometers are positioned in orthogonal pairs at six locations. Each accelerometer is oriented so that the sensing axis is tangential to the skull. Within the fabric pad, the accelerometers are located at the face of the pad where it contacts the head. The padding between the accelerometer and helmet serves as a spring so that the accelerometers remain in contact with the head at all times. This ensures that head accelerations, and not helmeted acceleration, are measured [19]. The fabric pad is attached to the vinyl casing, which is secured to the space between the standard padding in the Revolution helmets. The vinyl casing also houses the remaining electronic components of the measurement device: a wireless transceiver (903–927 MHz), on-board memory (up to 120 impacts), and data acquisition capabilities (8 bit, 1000 Hz/channel). Data acquisition is triggered any time an accelerometer experiences 10 g or greater. Data are collected for 40 ms (8 ms pretrigger and 32 ms post-trigger). Once the data are recorded, the wireless transceiver sends the data to the HITS Sideline Controller (Riddell, Elyria, OH), which is composed of an antenna and laptop computer with specialized software. If communication cannot be established, the on-board memory stores the impact until communication is re-established.

Unlike the existing HITS measurement device, the 6DOF measurement device cannot be used for real-time impact assessment because the data are incompatible with the existing Sideline Controller software. All 6DOF data must be postprocessed using a novel algorithm to determine the linear and angular accelerations for each axis of the head [20]. The measurement device and algorithm were validated through dynamic impact testing using a 50th percentile male Hybrid III anthropomorphic test dummy headform. The Hybrid III headform was instrumented with nine accelerometers in a 3-2-2-2 orientation [21]. A 6DOF measurement device was installed in a medium Riddell Revolution helmet, which was fitted on the Hybrid III head. The head and neck of the Hybrid III were mounted on a linear slide table and struck with a pneumatic linear impactor at several combinations of impact locations and severity. The acceleration of the headform was compared with the 6DOF measurement device, which was shown to have an average error of $1\% \pm 18\%$ for linear acceleration and $3\% \pm 24\%$ for angular acceleration. In addition to this testing, Riddell Revolution helmets with the 6DOF measurement device installed were tested at a third party laboratory and passed the National Operating Committee on Standards for Athletic Equipment (NOCSAE) football helmet performance standard required for all helmets worn at the collegiate level.

Using a similar methodology to Duma et al. [13], 6DOF measurement devices were installed in the helmets of ten Virginia Tech football players during the 2007 season. Due to the large size of the 6DOF measurement device, only large and extra large Riddell Revolution helmets were instrumented. All ten instrumented players were either offensive or defensive linemen, which is a result of these players using the largest helmets. The average height and weight of the instrumented players were 75.6 in. ± 1.96 in. (192 cm. ± 4.98 cm.) and 292 lb ± 33.5 lb. Each player that participated in the study gave written informed consent with Institutional Review Board approval from both Virginia Tech and the Edward Via Virginia College of Osteopathic Medicine. To ensure that the collected data set was large and unbiased, head acceleration data were recorded for every practice and game each player participated in. During practices, the sideline computer was stationed next to the practice field and downloaded impacts as they happened throughout practice. During games, the sideline computer was set up at the 40 yard line and downloaded impacts as they happened throughout each game (Fig. 1). The *x*-, *y*-, and *z*-axis linear and angular acceleration traces were recorded for every impact instrumented players experienced during games and practices throughout the 2007 Virginia Tech football season.

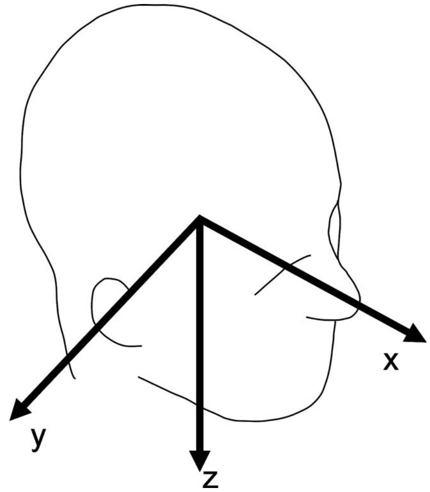

Fig. 2 SAE J211 headform coordinate system

The coordinate system referenced in this paper is that of the SAE J211 (Fig. 2). The positive x-axis runs out of the face (perpendicular to the coronal plane), the positive y-axis runs out of the right ear (perpendicular to the sagittal plane), and the positive z-axis runs out of the bottom of the head (perpendicular to the transverse plane). The presented head acceleration data describe the distributions of resultant linear and angular accelerations, as well as distributions for each axis of the head. In some cases, relationships between variables were investigated through linear regression using a least-squares approach, and R^2 is expressed as the Pearson product-moment correlation coefficient.

Table 1 Frequency of impacts above specified resultant acceleration thresholds

Linear acceleration (g)	Number of impacts	Angular acceleration (rad/s^2)	Number of impacts
>0	1712	>0	1712
>20	684	>1000	875
>40	172	>2000	339
>60	52	>3000	143
>80	11	>4000	57
>100	3	>5000	23
>120	1	>6000	12
>140	0	>7000	5
>160	0	>8000	4
>180	0	>9000	1

3 Results

A total of 1712 impacts were recorded during practices and games for the ten instrumented players during the 2007 Virginia Tech football season. 570 of the recorded impacts occurred during games, while 1142 occurred during practices. No instrumented player sustained a concussion during the 2007 season.

Table 1 displays the frequency of impacts over specified resultant acceleration thresholds for linear and angular accelerations. For resultant linear acceleration, thresholds are in 20 g increments. The majority of the impacts were under 20 g in severity. 10% of the impacts were greater than 40 g in severity. Of the 1712 impacts, 11 were greater than 79 g, which is the nominal injury value derived by the NFL study. For resultant angular acceleration, thresholds are in 1000 rad/s^2 increments. Roughly half of the impacts were less than 1000 rad/s^2 in severity. Only 143 of the impacts were greater than 3000 rad/s^2 in severity. Of the 1712 impacts, 14 were greater than 5757 rad/s^2, which is the nominal injury value derived in the NFL study.

Figure 3 displays histograms of the distributions of resultant linear and angular acceleration. Linear accelerations ranged from 9 g to 135 g. The 6DOF data set has an average peak resultant linear acceleration of 22.3 g and a median value of 17.5 g. Angular accelerations ranged from 107 rad/s^2 to 9922 rad/s^2. The 6DOF data set has an average peak resultant angular acceleration of 1355 rad/s^2 and a median value of 1017 rad/s^2. The two plots in Fig. 3 demonstrate that the majority of impacts were of low severity.

Furthermore, the 6DOF data set allows for the analysis of head acceleration data about each axis of the head. Figure 4 compares the distributions of each axis's peak linear acceleration for every recorded impact. The average linear head acceleration along the x-axis was 12.8 g and the median value was 10.0 g. The average linear head acceleration along the y-axis was 10.0 g and the median value was 8.40 g. The average angular head acceleration along the z-axis was 16.5 g and the median value was 13.0 g. Large peak linear accelerations (>60 g) were most common along the z-axis, while low peak linear accelerations (<10 g) were most common along the y-axis.

Figure 5 compares the distributions of each axis' peak angular acceleration for every recorded impact. The average angular head acceleration about the x-axis was 561 rad/s^2 and the median value was 417 rad/s^2. The average angular head acceleration about the y-axis was 625 rad/s^2 and the median was 450 rad/s^2. The average angular head acceleration about the z-axis was 1106 rad/s^2 and the median value was 781 rad/s^2. The distribution of angular acceleration about the z-axis is different than

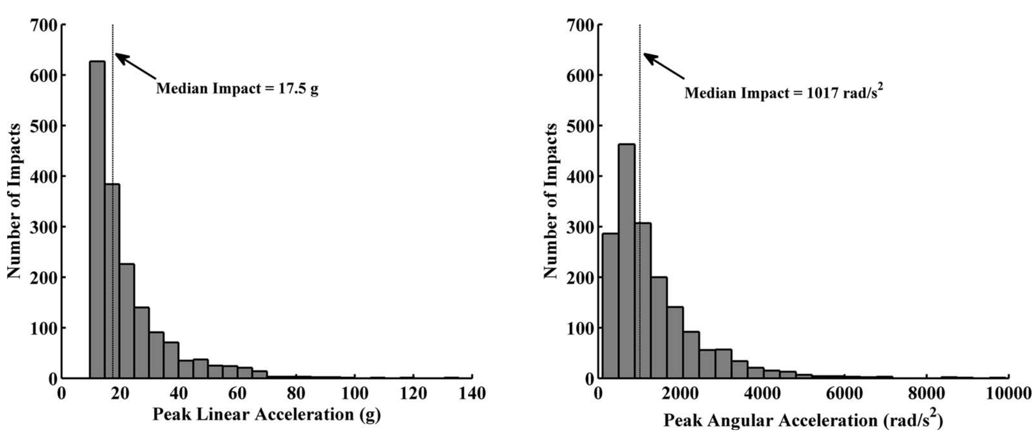

Fig. 3 Distributions of linear and angular accelerations

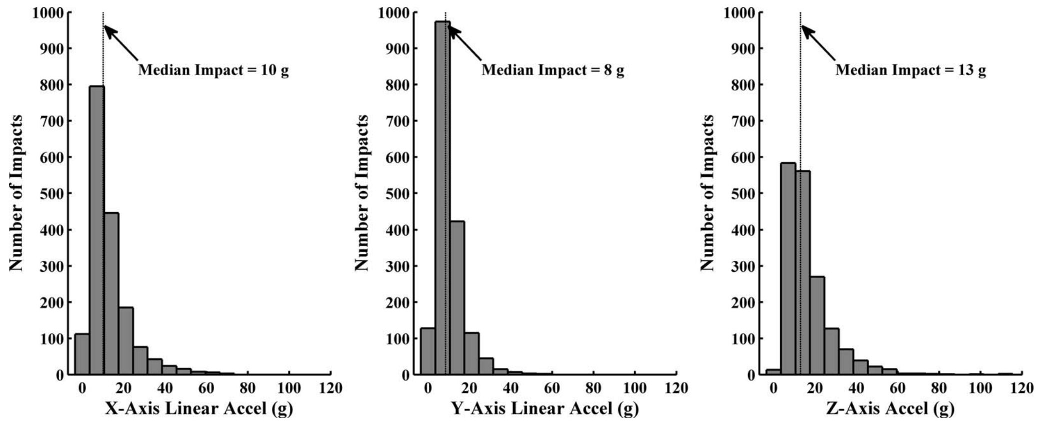

Fig. 4 Distributions of linear accelerations about each axis of the head

that of the x- and y-axes. Large peak angular accelerations ($>3000\ \mathrm{rad/s^2}$) were more common about the z–axis, while low peak low angular accelerations ($<1000\ \mathrm{rad/s^2}$) were more common about the x- and y-axes.

For each impact, azimuth and elevation are recorded to identify where the helmet was impacted. Azimuth (θ) is defined as the angle between the impact location and negative x-axis in the x-y plane. Elevation (α) is defined as the angle between the impact location and the x-y plane. The head was divided into sections to generalize each impact location. Figure 6 displays the group's impacts based on their impact location. Impacts to the front of the helmet were most common with 704 impacts. A total of 573 impacts were to the sides of the helmet. Back and top impacts were the least common with 220 and 215 impacts, respectively.

The average duration of the 1712 impacts was 14 ms (Fig. 7). Impact duration and acceleration magnitude are ultimately responsible for the change in velocity of the head. Changes in linear and angular head velocities are displayed in Fig. 8 as a function of their respective peak accelerations. Change in linear velocity ranged from 0.3 m/s to 6.1 m/s and did not correlate strongly with peak linear acceleration ($R^2=0.49$, $p=0$). This may be a reflection of the 10 g trigger for data acquisition. Change in angular velocity ranged from 0.5 rad/s to 42.5 rad/s and correlated more strongly with peak angular acceleration ($R^2=0.68$, $p=0$).

4 Discussion

While the original HITS measurement device is capable of measuring resultant linear acceleration and impact location, its applications are limited by its inabilities to measure angular acceleration accurately and produce data traces for each individual axis. The 6DOF measurement device provides individual data traces for linear and angular acceleration. Using this measurement device, large and unbiased data set on human head acceleration were compiled. The biomechanical response of the human head to impact can be investigated at the organ level by examining the acceleration data. Computational models can be used to determine the tissue level response of the head to impact by examining the resulting stresses and strains of each impact.

Over 27,000 impacts were recorded using the original HITS measurement device throughout the 2003–2006 Virginia Tech football seasons [13,14]. Figure 9 compares the resultant linear accelerations collected throughout the 2003–2006 seasons using the original HITS measurement device and 2007 season using the 6DOF measurement device. Since there were a different number of impacts in each data set, frequencies were normalized to be a percentage. The previous HITS data are viewed as the standard of which to compare the 6DOF linear accelerations to. Comparison of these impact distributions shows similar distributions between the two data sets, which suggests that the 6DOF linear acceleration measurements are consistent with previous head impact data from football players. The higher percentage of lower magnitude impacts (<20 g) might display a positional effect between the data sets. It should be noted that the 6DOF data only include offensive and defensive linemen, while the previous HITS data include both linemen and skill position players. However, Schnebel et al. [15] reported no statistical difference in the distributions between linemen and skill position players.

The 6DOF data set shows that football players routinely experience head accelerations up to 40 g and 3000 $\mathrm{rad/s^2}$. While no clinically diagnosed concussions were measured, several impacts

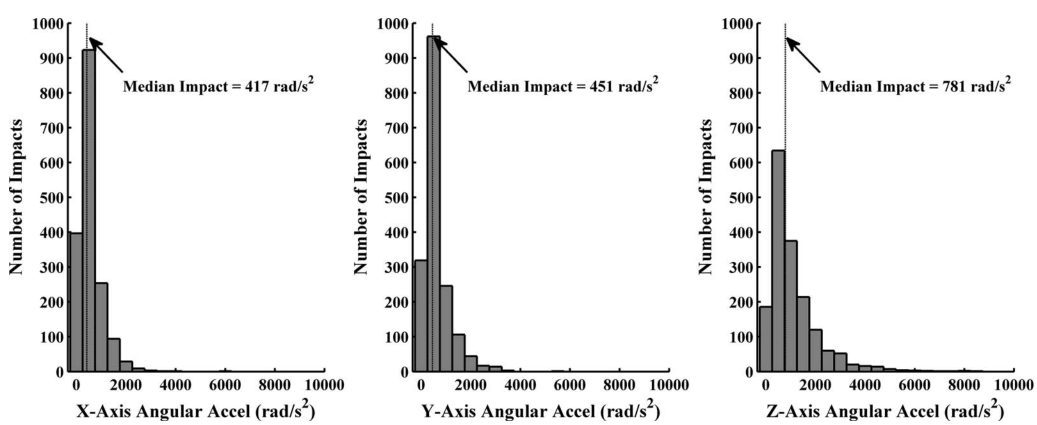

Fig. 5 Distributions of angular acceleration about each axis of the head

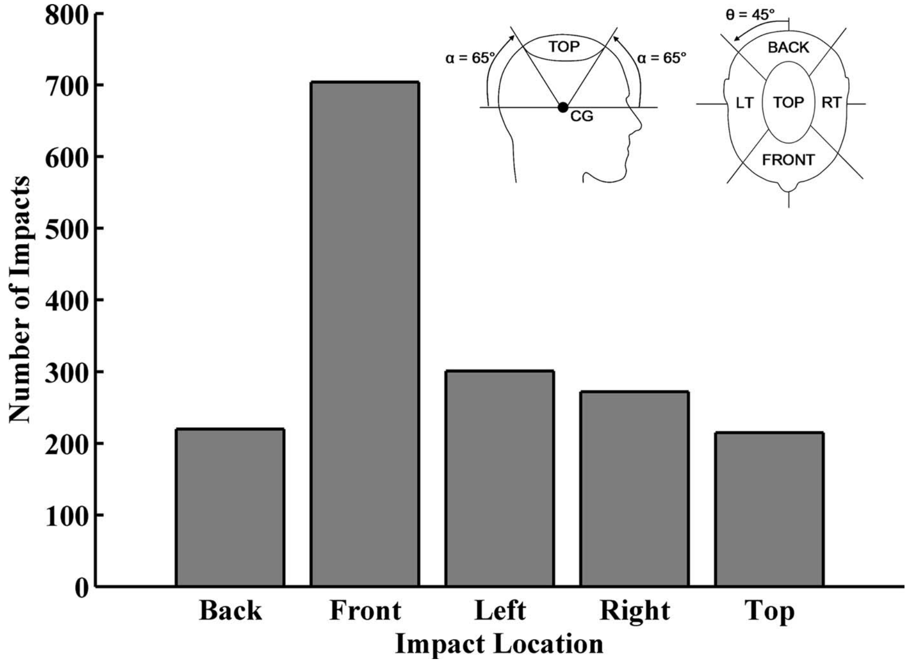

Fig. 6 Distribution of impact locations broken into back, front, left, right, and top bins. Bins are defined in the top right corner of the histogram.

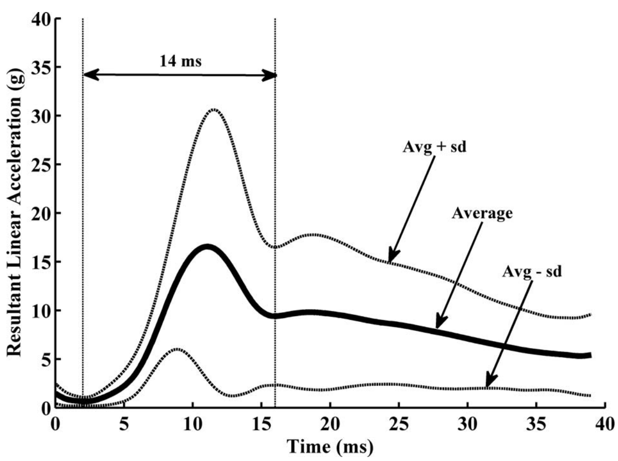

Fig. 7 Average linear acceleration response for the 1712 impacts. The average impact duration was 14 ms.

within this data set are within the range of previously reported injury data. Logistic regressions of the NFL reconstructions produced injury thresholds of 79–82 g and 5757–5900 rad/s^2, representing 50% risk of concussion [11,22]. Similarly, two separate tolerance curves defined 90 g as the injury threshold for concussion based on linear acceleration and impact duration [23,24]. Less conservative injury thresholds that were based on head impact data from human volunteers defined 165 g and 9000 rad/s^2 as representing 10% risk of concussion [14]; however, these values are a result of a limited injury data set and estimated rotational accelerations. Other studies have focused on the rotational kinematics to define injury thresholds. Ommaya [25] related the angular acceleration to angular speed and specified 4500 rad/s^2 with change in angular velocity below 30 rad/s as an injury threshold for y-axis rotation. These values are based on scaled primate data. Another separate study investigated DAI through x-axis rotation and defined 16,000 rad/s^2 and 46.5 rad/s as injury inducing based on a strain threshold [26]. These thresholds were also scaled from primate to human. Figure 5 displays that only one impact exceeded 4500 rad/s^2 about the y-axis and no impacts exceeded

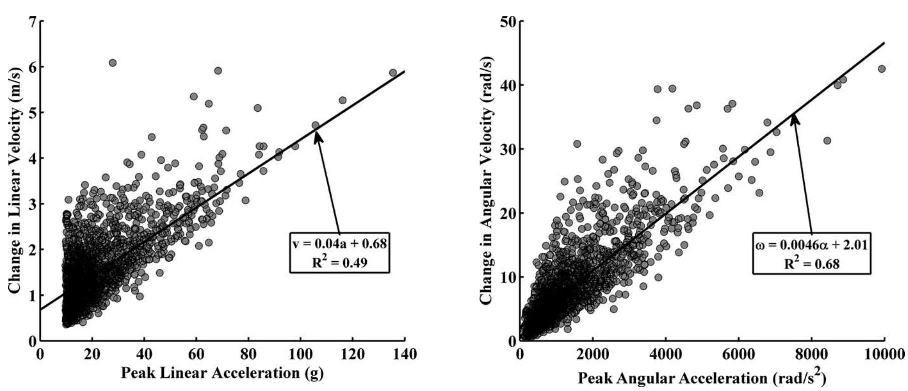

Fig. 8 Relationship between angular change in velocity of the head versus peak linear acceleration (left). Relationship between change in angular velocity of the head and peak angular acceleration (right).

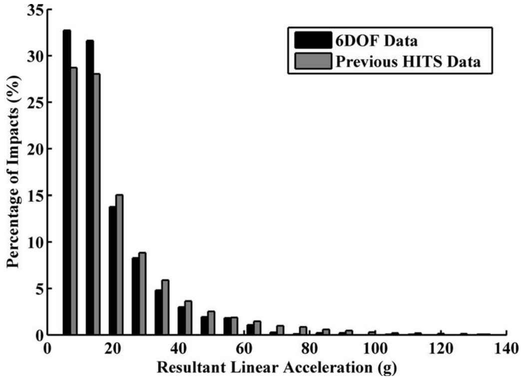

Fig. 9 Comparison of accelerations collected with the 6DOF measurement device during the 2007 Virginia Tech football season and the original HITS measurement device during the 2003–2006 seasons

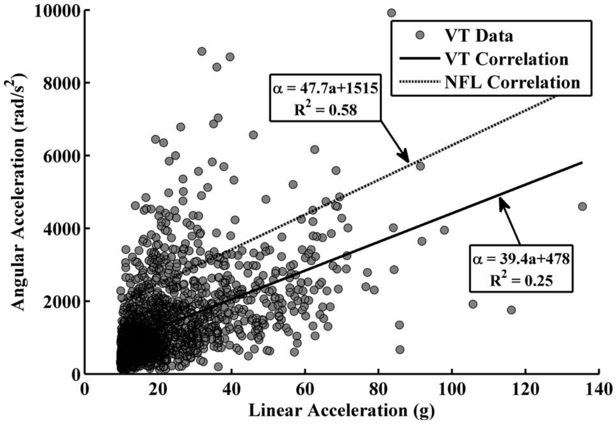

Fig. 10 Peak angular acceleration as a function of peak linear acceleration. The VT data suggest that no correlation exists ($R2=0.25$). The dashed line overlays the correlation reported by the NFL study.

16,000 rad/s^2 about the x-axis. However, it is important to note that all these impacts involved rotation about each axis. In this study, change in angular velocity ranged from 0.5 rad/s to 42.5 rad/s. The 6DOF data are in agreement with the NFL data because all impacts fell within or below the range of the change in angular velocity of the noninjured players (9.8–55.8 rad/s) [10]. Concussive impacts in the NFL data ranged from 12.8 rad/s to 80.9 rad/s. Furthermore, less than 2% of these impacts exceeded any of the thresholds previously reported in the literature (24 impacts exceeded the most conservative criteria), which is consistent with other HITS data that included much larger data sets [15,18]. Another study investigating head impacts in boxing saw angular accelerations as high as 16,000 rad/s^2 with angular speeds of 25 rad/s with no injury [27]. Interestingly, single skull kinematic parameters may not even be the best predictor of concussions. Greenwald et al. [18] showed that a composite variable of the kinematics may be a better predictor of concussion than any single biomechanical measure. Zhang et al. [22] ran finite element model (FEM) simulations of the NFL reconstructions and determined that shear strain around the brainstem region was the best predictor of MTBI. Kleiven [28] also ran FEM simulations of an expanded NFL reconstruction data set and found that the maximum pressure within the gray matter was the best predictor and stated that strain based injury predictors are very sensitive to the choice of stiffness for the brain tissue. It is thought that brain injury is much closer related to the response of the brain than the global inputs to the head, and that a computer model is needed to describe this response [11]. The outputs of the computer models are dependent on the brain properties and the validation data they are based on. Human volunteer data from studies such as this may be valuable in validating these models, as such impacts are easily modeled and can include an injury response.

When comparing the accelerations about each axis, large angular accelerations about the z-axis were most common (Fig. 4). A hypothesis for this is that the high z angular accelerations are due to the large moment arm resulting from the facemask's distance away from the center of gravity of the head. Supporting this, the majority of impacts were to the front of the helmet. However, the principal direction of force is unknown for each of these impacts. Impacts to the front of the helmet that have a line of force through the center of gravity of the head will result in low angular accelerations. Conversely, the further away from the center of gravity that the direction of the force is, the higher the angular accelerations will be. A more detailed analysis is needed to evaluate this hypothesis, which was mentioned in this paper as only an observation of the data.

Figure 10 plots peak resultant angular acceleration against peak resultant linear acceleration for every recorded impact. This plot suggests that there is no strong correlation between peak linear and angular acceleration. Figure 10 has data points with low linear and low angular acceleration, data points with low linear and high angular acceleration, data points with high linear and low angular acceleration, and data points with high linear and high angular acceleration. There is too much scatter for any correlation to exist.

The 6DOF data produced in this study suggest that there is little correlation ($R^2=0.25$, $p=0$) between linear and angular acceleration. This is inconsistent with relationships reported in the literature. Pellman et al. [10] reported a linear relationship between linear and angular acceleration ($R^2=0.58$). That correlation should only be applied to the NFL data, as only selected impacts in the concussive severity range were included in that study. Furthermore, this may be a result of the biofidelity of the Hybrid III neck. The neck will give a consistent response for impacts, and due to its stiffness, promotes rotation more so than head translation. It is possible for no correlation to exist between linear and angular acceleration when looking at a complete range of impacts. Varying impact location and principal direction of force for the same linear acceleration input can result in different angular accelerations. Theoretically, it is possible for impacts to exist with pure linear acceleration (no angular acceleration) and with pure angular acceleration (no linear acceleration). Pellman et al. [10] reported a correlation for specific impacts. When looking at a full range of impacts with varying impact location, principal direction of force, and impact severity, the 6DOF data suggest that no correlation exists.

The head acceleration data produced in this study were collected by measuring helmeted head impacts on human volunteers. While having applications in real world scenarios with padded impacts, these data may not be able to accurately model head impacts resulting in skull fractures. The resulting average duration of these helmeted head impacts was 14 ms. This is in agreement with the 15 ms impact duration experienced by the Hybrid III dummies in the NFL study [10]. This 14–15 ms range of impact duration is unique to helmeted head impacts. When comparing this to real world durations of impact in motor vehicle crashes, it is between that of head impacts to vehicle structures (>6 ms) and an airbag with seatbelt restraints (<40 ms) [10].

There is some inherent error within the measurement of head acceleration using the 6DOF measurement device. The 6DOF measurement device has an average error of $1\% \pm 18\%$ and $3\% \pm 24\%$ for linear and angular acceleration, respectively. Sources of this error in the validation testing include the helmet changing position relative to the head throughout an impact, as well as nonideal orientation of the accelerometers. In an attempt to simulate a more realistic interaction between the Hybrid III head and the football helmet, a synthetic skull cap commonly used in football was fitted to the Hybrid III head. The skull cap is composed of 89% nylon and 11% spandex, which substantially reduced the friction between the head and helmet when compared with the high coefficient of friction of the vinyl Hybrid III skin. This allowed for some sliding of the helmet with respect to the head during testing. The error measured in the validation testing should be reflective of the error associated with on-field impacts. While possible sources of error were accounted for in the validation testing, it is not possible to account for all sources of error with data collection, such as improper helmet fit and usage seen with players. However, this error is comparable to the error ranges of other measurement devices. The video analysis conducted for the NFL reconstructions reported error as high as 15% [29], the original HITS measurement device has an error of $8\% \pm 11\%$, and chest bands used to measure chest deflection can have error as high as 10% [30]. Considering the vast amounts of data that can be collected with the 6DOF HITS measurement device on human volunteers and the error levels of other biomechanical experiments, the error inherent in the 6DOF measurement device is acceptable. While the 6DOF error is considered acceptable, additional methods can be used and be applied to the data set to minimize the statistical effects of data scatter. The standard deviations contribute to the majority of the error, as the average errors are minimal. Funk et al. [14] presented a unique methodology for adjusting a similar large data set to minimize the effect of data scatter. Using this same technique, the effects of data scatter in a 6DOF data set can be incorporated into an injury model. Given the stated error, such an analysis would require a large number of subconcussive and concussive impacts.

5 Conclusion

Football presents a unique opportunity to quantify the biomechanical response of the head to impact with human volunteers. The helmets of ten Virginia Tech football players were instrumented with 6DOF measurement devices throughout the 2007 season, resulting in head acceleration data for 1712 impacts. The data set is large and unbiased, as head impacts were recorded for every game and practice during the 2007 season. Future collection of concurrent concussive data will result in a better understanding of the pathomechanics of mild traumatic brain injury. With an increased understanding of the biomechanics of head impacts in football and human tolerance to head acceleration, better equipment can be designed to prevent head injuries. Insight into improved tackling techniques to reduce the risk of injury may also be gained.

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