A Six Degree of Freedom Head Acceleration Measurement Device for Use in Football

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The high incidence rate of concussions in football provides a unique opportunity to collect biomechanical data to characterize mild traumatic brain injury. The goal of this study was to validate a six degree of freedom (6DOF) measurement device with 12 single-axis accelerometers that uses a novel algorithm to compute linear and angular head accelerations for each axis of the head. The 6DOF device can be integrated into existing football helmets and is capable of wireless data transmission. A football helmet equipped with the 6DOF device was fitted to a Hybrid III head instrumented with a 9 accelerometer array. The helmet was impacted using a pneumatic linear impactor. Hybrid III head accelerations were compared with that of the 6DOF device. For all impacts, peak Hybrid III head accelerations ranged from 24 g to 176 g and 1,506 rad/s² to 14,431 rad/s². Average errors for peak linear and angular head acceleration were 1% ± 18% and 3% ± 24%, respectively. The average RMS error of the temporal response for each impact was 12.5 g and 907 rad/s².

Keywords: mild traumatic brain injury (MTBI), concussion, biomechanics, angular, linear, rotational

Each year there are an estimated 1.6 million to 3.8 million sports related concussions in the United States (Langlois et al., 2006). About 300,000 of these concussions involve loss of consciousness, with football having the largest occurrence (Thurman et al., 1998). The high incidence of concussions in football provides a unique opportunity to collect biomechanical impact data from humans to characterize mild traumatic brain injury (MTBI). Competitive football has been used as an experimental environment for collecting human head acceleration data since the 1970s. Several studies have had football players wear headbands instrumented with accelerometers to measure head acceleration during football games (Moon et al., 1971; Reid et al., 1974; Reid et al., 1971). While laying the groundwork for future research and providing a proof of concept, these studies measured only a single player and were limited in their ability to measure head acceleration accurately.

One study has quantified head accelerations by recreating concussive impacts sustained by football players. The National Football League (NFL) reconstructed injurious game impacts using Hybrid III dummies based on game video (Newman et al., 2000; Newman et al., 1999; Pellman et al., 2003). This study was biased toward concussive events and was limited to impacts that could be clearly identified on video. With such a labor intensive testing methodology, it was not realistic to recreate each and every impact that football players experienced. More recently, two studies quantified head accelerations during impact by instrumenting helmets worn by collegiate football players using a six accelerometer measurement device integrated into football helmets (Duma et al., 2005; Funk et al., 2007). These six accelerometer measurement devices used the same technology described in this paper but with fewer accelerometers and with the sensing axes of the accelerometers oriented differently with respect to the surface of the head. Resultant linear and peak angular head acceleration for every head impact instrumented players experienced was computed from the measured accelerations, producing a large and unbiased dataset of over 27,000 events including 4 cases of diagnosed concussion.

The 6 accelerometer measurement device used by Duma et al. (2005) is part of the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH). The commercially available football HIT System measurement device consists of 6 nonorthogonally mounted single-axis accelerometers which are oriented normal to the head. Each time an impact occurs, data are transmitted wirelessly from the measurement device to the computer, which processes and displays data in real-time. The HIT System utilizes a novel algorithm for determining impact magnitude and location (Crisco et al., 2004). This algorithm is capable...
of determining the temporal response of the resultant linear acceleration of the head. In addition to resultant linear acceleration, the algorithm estimates peak x and y axis rotational accelerations based on an assumed pivot point in the neck. While this device can be used to collect valuable human head acceleration data from impacts, it cannot completely characterize the head kinematics of impacts because it computes only peak angular accelerations rather than the temporal angular acceleration response for each axis of the head. The temporal response of acceleration is desired with applications investigating the tissue-level response of the brain using computational models.

The goal of this study was to validate a six degree of freedom (6DOF) measurement device that completely characterizes the head kinematics resulting from impacts in football. This device records all head impacts sustained by football players during competitive play and computes the resulting linear and angular acceleration about each axis of the head. Such data are ideal for the validation of computational injury models and investigating the tissue-level response of the brain following impact.

**Methods**

The 6DOF measurement device was designed to be integrated into Riddell Revolution (Elyria, OH) football helmets (Figure 1). The measurement device utilizes 12 single-axis, 250-g MEMS accelerometers (ADXL193, Analog Devices, Norwood, MA). The 12 accelerometers are grouped in orthogonally oriented pairs at 6 different locations within fabric padding. All accelerometers are orientated so that their sensing axes are tangential to the skull. The fabric pad system serves as an elastomeric spring and damper system that regulates the contact pressure and maintains accelerometer orientation relative to and in contact with the head during an impact event. When the helmet shifts position on the head due to impact, the fabric pad either compresses or expands to remain in contact with the player’s head ensuring head acceleration, not helmet acceleration, is measured (Manoogian et al., 2006). Data acquisition is triggered when any accelerometer exceeds 10 g. Data are collected for 40 ms at 1000 Hz, of which 8 ms are pretrigger and 32 ms are posttrigger. After each impact is recorded, the data are sent to a computer via a 903–927 MHz FHSS wireless transceiver. Following data transmission, each impact is processed for linear and angular acceleration about all three axes of the head center of gravity (CG), in addition to impact location.

Linear and angular acceleration are determined by iteratively optimizing the equations of motion for the head during impact (Chu et al., 2006). Assuming rigid body dynamics, the acceleration of any point on the surface of the head relative to the head CG can be expressed by Equation 1: |

\[ \mathbf{a} = \mathbf{H} \mathbf{a} + \mathbf{r} \times (\mathbf{\omega} \times \mathbf{r}) \]

\( \mathbf{a} \) is the acceleration magnitude at each individual accelerometer, \( \mathbf{r} \) is the orientation of the sensing axis of each accelerometer, \( \mathbf{H} \) is the head CG linear acceleration, \( \alpha \) is angular acceleration about the head CG, \( \mathbf{r} \) is the accelerometer location relative to the head CG, and \( \mathbf{\omega} \) is the angular velocity of the head.

\[ \| \mathbf{a} \| = \| \mathbf{a} \| \mathbf{H} \mathbf{a} + \mathbf{r} \times (\mathbf{\omega} \times \mathbf{r}) \] (1)

By orienting the 6DOF measurement device’s accelerometers tangentially to the skull, the centripetal acceleration term of Equation 1 becomes negligible, simplifying to Equation 2.

\[ \| \mathbf{a} \| = \| \mathbf{a} \| \mathbf{H} \mathbf{a} + \mathbf{r} \times (\mathbf{\omega} \times \mathbf{r}) \] (2)

The linear and rotational acceleration vectors are estimated by iteratively solving for the vector components using a simulated annealing optimization algorithm. The cost function for the optimization algorithm is the minimization of the sum of square error between each measured acceleration and the estimated acceleration at each accelerometer location based on the estimated head linear and rotation vectors.

The location and orientation of each accelerometer relative to the head CG were determined by digitizing their positions (Microscribe G2, Amherst, VA) within a football helmet relative to the head CG of a Hybrid III 50th percentile male head. For field use, this digitization can be repeated for all available helmet sizes, providing a scalable solution for all players; however, it is still assumed that the location of the head CG of each player is represented by a Hybrid III head. The location of the Hybrid III head CG has been previously defined and is based on the data from cadaver heads (Foster et al., 1977; Hubbard & McLeod, 1974; Walker Jr., et al., 1973).

A total of 114 impact tests were conducted to assess the accuracy of the 6DOF measurement device using an instrumented 50th percentile male Hybrid III head and neck assembly. The Hybrid III head was equipped with 9 accelerometers (7264–2000B, Endevco, San Juan Capistrano, CA) in a 3-2-2-2 orientation; which allowed linear and angular acceleration to be calculated (Padgaonkar et al., 1975). Hybrid III data were sampled at 10,000 Hz and filtered in accordance with SAE J211 using Channel Frequency Class (CFC) 1000. The head and neck were mounted on a custom linear slide table built to National Operating Committee on Standards
for Athletic Equipment (NOCSAE) specification. The linear slide table was permitted 5 degrees of freedom, allowing repeatable adjustment of the head and neck orientation. All impacts were performed using a pneumatic linear impactor that was built to NOCSAE specification (NOCSAE, 2006). At the end of the 15 kg impactor arm, a disc of high-density vinyl nitrile foam attached to a hemispherical nylon shell were used to create an impacting surface that replicated the characteristics of a typical football helmet (Figure 2) (Newman et al., 2005).

A 6DOF measurement device was installed in a medium Riddell Revolution helmet, which was fitted on the Hybrid III head. A skullcap (89% nylon, 11% spandex) was put on the Hybrid III head to reduce the friction between the head-to-helmet interface to simulate a worst-case scenario. The same helmet was used and repositioned for every test. The medium-sized jaw pads of the helmet were replaced with large jaw pads to better fit the narrow face of the Hybrid III head. A custom helmet positioning tool was used to ensure that the helmet’s fit on the head was consistent and in accordance with NOCSAE standards. This Plexiglas tool uses landmarks on the helmet and head to align the center of the facemask with the center of the nose and the top of the facemask with the brow of the Hybrid III face. A standard four-point chinstrap was used to secure the helmet on the head. An air pump was used to inflate the padding of the helmet with accordance to the manufacturer’s specification.

The helmeted Hybrid III head was struck with the pneumatic linear impactor with several combinations of impact energies and locations. The impact energies ranged from 67.5 J to 607.5 J and were chosen to replicate on-field impact data as determined by the NFL through impact reconstructions (Pellman et al., 2003). To account for the various ways a helmet can be struck, 5 impact locations were chosen based on NFL video analysis (Pellman et al., 2003b). Impact locations ranged from the front to the backside of the helmet (Figure 3). Table 1 defines each impact location in terms of azimuth and elevation. Azimuth refers to the angle that the impact location makes with the sagittal plane of the Hybrid III head. Elevation refers to the angle that the impact location makes with the transverse plane of the Hybrid III head. A negative elevation is interpreted as having the Hybrid III head tilted away from the impactor. The number of tests in each configuration can be seen in Table 2.

Regression analyses were applied to the data to determine how strongly the 6DOF measurement device

Table 1 Definitions of the 5 impact locations tested

<table>
<thead>
<tr>
<th>Location</th>
<th>Elevation</th>
<th>Azimuth</th>
</tr>
</thead>
<tbody>
<tr>
<td>C</td>
<td>−11°</td>
<td>112.5°</td>
</tr>
<tr>
<td>D</td>
<td>−11°</td>
<td>157.5°</td>
</tr>
<tr>
<td>F</td>
<td>25°</td>
<td>0°</td>
</tr>
<tr>
<td>R</td>
<td>0°</td>
<td>180°</td>
</tr>
<tr>
<td>U</td>
<td>20°</td>
<td>67.5°</td>
</tr>
</tbody>
</table>

Table 2 Impact testing test matrix

<table>
<thead>
<tr>
<th>Energy (J)</th>
<th>C</th>
<th>F</th>
<th>R</th>
<th>D</th>
<th>U</th>
<th>Total</th>
</tr>
</thead>
<tbody>
<tr>
<td>67.5</td>
<td>4</td>
<td>0</td>
<td>0</td>
<td>4</td>
<td>0</td>
<td>8</td>
</tr>
<tr>
<td>187.5</td>
<td>4</td>
<td>4</td>
<td>4</td>
<td>4</td>
<td>0</td>
<td>16</td>
</tr>
<tr>
<td>270.0</td>
<td>4</td>
<td>4</td>
<td>4</td>
<td>8</td>
<td>4</td>
<td>24</td>
</tr>
<tr>
<td>367.5</td>
<td>4</td>
<td>4</td>
<td>4</td>
<td>4</td>
<td>0</td>
<td>16</td>
</tr>
<tr>
<td>480.0</td>
<td>10</td>
<td>4</td>
<td>4</td>
<td>12</td>
<td>4</td>
<td>34</td>
</tr>
<tr>
<td>607.5</td>
<td>8</td>
<td>4</td>
<td>0</td>
<td>4</td>
<td>0</td>
<td>16</td>
</tr>
<tr>
<td>Total</td>
<td>34</td>
<td>20</td>
<td>16</td>
<td>36</td>
<td>8</td>
<td>114</td>
</tr>
</tbody>
</table>
correlated with the Hybrid III head form. Subsequently, a power-law regression using a least squares fitting technique was performed to map the 6DOF accelerations to that of the Hybrid III to get the highest correlation coefficient. These mapping functions between Hybrid III and 6DOF were desired because the input to many finite element head models used in the injury biomechanics field is kinematic data from the Hybrid III head (Takhounts et al., 2008).

Acceleration error is presented in two ways: percent error for linear and angular peak resultant accelerations, and root mean square (RMS) error for the temporal response of linear and angular resultant acceleration. In both sets of error calculations, the Hybrid III head acceleration is considered the true value of acceleration. The Hybrid III head represents the ‘gold standard’ for modeling dynamic cranial response of the human head, and is commonly used in automotive safety testing and sports injury biomechanics testing. Equation 3 was used to calculate relative error; where \( RE \) is relative error, \( 6DOF \) is the 6DOF measurement device peak acceleration predicted by the mapping functions, and \( HIII \) is the Hybrid III head peak acceleration. Relative error of the linear and angular peak accelerations are presented as an average percent error ± 1 SD. RMS error was calculated using Equation 4; where \( n \) is the number of discrete data points, \( i \) represents individual discrete data points, \( 6DOF \)
is the 6DOF measurement device acceleration predicted by the mapping functions, and \( HIII \) is the Hybrid III head acceleration. For the RMS calculation, Hybrid III data were downsampled to 1000 Hz. RMS error was calculated for the temporal response of linear and angular resultant acceleration over the acceleration pulse of interest, which equated to the first 25 ms of each impact. The remaining 15 ms of data for each impact have no relevance to the impact. This was consistent for all impacts because of the controlled impact conditions.

\[
RE = \frac{6DOF - HIII}{HIII} \tag{3}
\]

\[
RMS = \sqrt{\frac{\sum (6DOF_i - HIII_i)^2}{n}} \tag{4}
\]

**Results**

The linear and angular accelerations computed from the 3-2-2-2 array in the Hybrid III head were compared with the accelerations computed from the 6DOF measurement device. The power regression revealed strong correlations between Hybrid III and 6DOF measurement device linear

![Figure 4](image1.png)  
**Figure 4** — Nonlinear relationships between 6DOF measurement device and Hybrid III head peak resultant accelerations.

![Figure 5](image2.png)  
**Figure 5** — Typical temporal response of the Hybrid III and 6DOF measurement device resultant accelerations.
error in this study. The accelerations of the Hybrid III head with the stated power regression transformation are representative of 6DOF measurement device data undergoing the modeling applications that have traditionally used Hybrid III measurements, which is preferred for computational the 6DOF measurement device to be mapped to Hybrid and angular acceleration. The resulting equations allow average relative error for peak resultant linear acceleration was 1% ± 18%. The average relative error for peak resultant angular acceleration was 3% ± 24%.

Linear: $A_{\text{Hybrid}} = 0.8695 \times A_{\text{6DOF}}^{1.0622}$ \hspace{1cm} (5)

Angular: $\alpha_{\text{Hybrid}} = 0.0638 \times \alpha_{\text{6DOF}}^{1.3652}$ \hspace{1cm} (6)

Figure 5 compares the temporal acceleration responses for the Hybrid III and 6DOF measurement device for an impact energy of 480 J at location D. For all impacts, peak Hybrid III head accelerations ranged from 24 g to 176 g and 1,506 rad/s² to 14,431 rad/s². The average RMS errors for the resultant temporal response of all impacts were 12.5 ± 8.32 g and 907 ± 685 rad/s². RMS error for the temporal response was not greater for any specific axis of the head for either linear or angular acceleration. Furthermore, RMS error was not significantly influenced by impact location. RMS error increased linearly with impact energy, which is expected.

Discussion

A nonlinear relationship between 6DOF measurement device acceleration and Hybrid III head acceleration was observed with angular acceleration. It is thought that the reason for this nonlinear trend is relative movement between the Hybrid III head and the accelerometers of the 6DOF measurement device; resulting in the 6DOF measurement device to under-predict angular acceleration at high severities. Although the helmet was properly fitted to the Hybrid III head, the helmet could move relative to the head due to the low friction interface between the Hybrid III head and nylon skull cap. This type of helmet movement relative to the head could be seen on professional impacts in the game of football. The linear acceleration trend was practically linear, but a power regression was performed to transform the linear acceleration data in a method consistent with that used for angular acceleration. Furthermore, the power regression analysis resulted in high $R^2$ values between 6DOF measurement device and Hybrid III head for both linear and angular acceleration. The resulting equations allow the 6DOF measurement device to be mapped to Hybrid III measurements, which is preferred for computational modeling applications that have traditionally used Hybrid III data. 6DOF measurement device data undergoing the power regression transformation are representative of the accelerations of the Hybrid III head with the stated error in this study.

Similar testing has been used to validate instrumented boxing head gear (IBH) for use during boxing competition (Beckwith et al., 2007). In this study, IBH acceleration was shown to correlate strongly with Hybrid III head acceleration for linear ($R^2 = .91$) and angular ($R^2 = .91$) acceleration. The 6DOF measurement device utilizes a similar algorithm (Chu et al., 2006) and sensor orientation as the IBH. Average RMS errors for the IBH were 5.9 ± 2.6 g and 595 ± 405 rad/s², compared with 12.5 ± 8.32 g and 907 ± 685 rad/s² for the 6DOF measurement device. Difference in accuracy between the two systems can be attributed to several factors. The boxing head gear had a tighter fit than a football helmet on the Hybrid III head. In addition, the 6DOF measurement device was validated over a larger range of impact energies. The maximum accelerations in the IBH validation testing were 77.3 g and 6,433 rad/s², compared with 176 g and 14,431 rad/s² in the 6DOF measurement device validation testing. When comparing similar acceleration ranges, average RMS error for the 6DOF measurement device is similar to that of Beckwith et al. 2007. For example, average 6DOF RMS errors are only 3.7 ± 4.3 g and 252 ± 267 rad/s² for impact with peak linear acceleration less than 80 g.

Frontal facemask impacts were not included in this analysis due to limitations of the linear impactor. The linear impactor was designed to mimic reconstructed head-to-head football collisions; however, the existing impactor surface was optimized for helmet shell contact and initially used for testing helmets without facemasks (Newman et al., 2005). Preliminary trials revealed an unrealistic interaction between the impactor surface and the facemask. Severe facemask bending atypical of on-field occurrence would occur even at low impact energies. In addition, at impact energies exceeding 480 J, the impactor surface would penetrate the facemask and contact the surface of the Hybrid III head. Substantial stroking was also observed on the chinstrap following each of these impacts. This is where the chinstrap slides through its grips, allowing the helmet’s position relative to the head to change. For these reasons, frontal facemask testing was not completed. In addition, inertially induced head accelerations were not examined.

Possible sources of observed error include the helmet changing position relative to the head during an impact and non-ideal orientation of the accelerometers with respect to the head. While it was not possible to measure changes in accelerometer location and orientation in this experiment, the error levels in the 6DOF accelerations are similar to that of other measurement devices and techniques. The NFL reconstruction was reported to have error as high as 17% for peak linear acceleration and 25% for peak angular acceleration (Newman et al., 2005). In addition, chest bands, which are used to measure chest deflection, can have error as high as 10% (Rath et al., 2005). Relative error for the 6DOF measurement device is 1% ± 18% for peak linear acceleration and 3% ± 24% for peak angular acceleration. Considering the vast amounts of data that can be collected with the 6DOF measurement
device on human volunteers and the error levels of other biomechanical experiments, the error inherent in the 6DOF measurement device can be considered acceptable. Furthermore, this study has provided quantification of uncertainty error that can be used for the design of experiments. Although some error is present, the data provides a distribution that can be used to help optimize design of experiments and bootstrapping techniques.

Measuring head accelerations experienced by human volunteers at potentially injurious severities has been traditionally challenging. Football provides a unique opportunity to collect such data using HIT System technology (Duma & Rowson, 2011; Rowson et al., 2009). In addition, the 6DOF measurement device used in this study provides the temporal response of linear and angular acceleration about each axis of the head due to impact, which improves applicability for validation of computational injury models, as most models are time sensitive (Rowson et al., 2008; Shain et al., 2010; Rowson et al., 2010). This technology provides the opportunity to collect a large and unbiased dataset, since every head impact that an instrumented football player experiences would be recorded including both noninjurious and concussive impacts, which can be applied to a wide array of research studies that will ultimately lead to a better understanding of the mechanisms of concussion.

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