Validation of a Wireless Head Acceleration Measurement System for Use in Soccer Play

Erin Hanlon and Cynthia Bir

Soccer heading has been studied previously with conflicting results. One major issue is the lack of knowledge regarding what actually occurs biomechanically during soccer heading impacts. The purpose of the current study is to validate a wireless head acceleration measurement system, head impact telemetry system (HITS) that can be used to collect head accelerations during soccer play. The HITS system was fitted to a Hybrid III (HIII) head form that was instrumented with a 3-2-2-2 accelerometer setup. Fifteen impact conditions were tested to simulate impacts commonly experienced during soccer play. Linear and angular acceleration were calculated for both systems and compared. Root mean square (RMS) error and cross correlations were also calculated and compared for both systems. Cross correlation values were very strong with $r = 0.95 \pm 0.02$ for ball to head forehead impacts and $r = 0.96 \pm 0.02$ for head to head forehead impacts. The systems showed a strong relationship when comparing RMS error, linear head acceleration, angular head acceleration, and the cross correlation values.

**Keywords:** football, heading, youth

Soccer heading has been studied previously with conflicting results. One major issue is the lack of knowledge regarding what actually occurs biomechanically during soccer heading impacts. The overall goal of heading is redirection of the ball. Depending on the approach of the player and the intent of the redirection, the player may move his/her head in a particular fashion. Alignment of the head, neck and torso can vary and are often dependent on the intent of the redirection i.e., clearing, passing, or contolling (Shewchenko et al., 2005b, 2005a). All of these scenarios require a specific skill level to accomplish the intent of the redirection through the use of proper techniques. Proper techniques and skill level often come with age.

It was hypothesized that soccer-related head injuries can be mainly attributed to unintentional impacts, which have a rotational component since the player is not prepared for the impact. However, the motion that occurs when heading a ball may also be attributed to technique. The importance of proper technique may be especially true in the youth population, since their skill level has not been developed to control their head motion when heading the ball. Therefore, the youth population could be at an increased risk for sustaining head injuries due to rotational acceleration.

In both proper and improper heading, the interaction between the ball and head creates both linear and angular acceleration (Naunheim et al., 2003; Shewchenko et al., 2005b, 2005a). The exact contribution of linear versus angular acceleration for a given impact when heading the ball is related to several factors. An analysis of common scenarios using both kinetic and kinematic techniques was recently conducted (Shewchenko et al., 2005b). The scenarios included tests using human subjects which were predetermined to assess various situations similar to what takes place during game play. Subjects attempted to head the ball to a target with a predetermined level of neck muscle activation: normal, pretensed, or relaxed. Soccer balls were presented to players at two different speeds to elicit a wider range of response. As part of this analysis, data were collected using both angular and linear accelerometers. The results demonstrated a difference in both angular and linear accelerations based on the scenario simulated. The authors also reported a wide variability between subjects (Shewchenko et al., 2005b). The variety of biomechanical scenarios of soccer heading will only be increased in the youth population due to the varying skill levels.

Previous studies of soccer heading have lacked the ability to measure real-time game impacts (Naunheim et al., 2000; 2003; Shewchenko et al., 2005c, 2005b, 2005a). All previous head acceleration measurements were done using recreations in a laboratory or restricted setting. By using a novel head acceleration measurement system, the Head Impact Telemetry System (HITS; Simbex, Lebanon, NH), linear and angular head acceleration can be measured during actual games. The system has been implemented and validated in both football helmets and

---

Erin Hanlon (Corresponding Author) is with the Bioengineering Center, College of Engineering, Wayne State University, Detroit, MI. Cynthia Bir is with the Bioengineering Center, College of Engineering, Wayne State University, Detroit, MI.
boxing headgear (Duma et al., 2005; Manoogian et al., 2006; Beckwith et al., 2007).

Previously HITS was implemented in football helmets (Duma et al., 2005), and is now commercially available for use (Duma et al., 2005). The data processing algorithm, previously developed by Crisco et al. (Crisco et al., 2004), allows for calculation of both linear and angular head acceleration (Chu et al., 2006). System and algorithm validation was performed using a Hybrid III dummy instrumented with a 3-2-2-2 accelerometer setup for both football and boxing (Crisco et al., 2004; Duma et al., 2005). Correlations were found to be strong with an $R^2 = .97$ and $R^2 = .91$ for football and boxing respectively (Duma et al., 2005; Beckwith et al., 2007).

This system uses six wireless accelerometers which were placed inside a football helmet along with a wireless transceiver, data acquisition, and on-board memory (Duma et al., 2005). The accelerometers are spring-mounted so that they are closely coupled to the head (Duma et al., 2005). This ensures that head acceleration is measured as opposed to helmet acceleration. Recording of impact data occurs when any accelerometer registers above the threshold of 10 g. When the threshold is reached, 40 ms of data are recorded. This information is then time stamped and downloaded to the sideline computer for later processing using the algorithm (Crisco et al., 2004).

The HIT system has recently been modified for use in boxing headgear (Beckwith et al., 2007). The system is much like the football system, but the boxing headgear has a total of twelve accelerometers as opposed to six. Accelerometer placement was more of a concern because the headgear has no outer shell and the accelerometers had to be placed where it was least likely for impacts to occur. Therefore, the accelerometers were placed toward the back of the headgear. The battery pack and transmitter were placed in the back panel. This system was validated using a Hybrid III (HIII) head and neck (Beckwith et al., 2007).

Linear head acceleration, angular head acceleration, impact location, Gadd Severity Index (GSI), and the Head Injury Criterion (HIC) were calculated for both systems. HIC and GSI, both developed primarily for automotive impact, were based on the Wayne State Tolerance Curve (WSTC; Gurdjian et al., 1966). These criteria take into account acceleration over a period of time to assess single impact events (Gadd, 1966; Newman et al., 2000). Although high correlations, $r^2 = .91$, for both linear and angular head acceleration were found, it was found that a limitation of the headgear development was the need to place accelerometers in the back of the headgear. While this potentially creates error, it is necessary for avoiding direct accelerometer impact. This could also be a problem during development of the soccer headgear because impacts will take place in the forehead region and there will be no padding. Therefore, accelerometers will need to be placed in the rear portion of the headgear.

Implementation of this device into a headband system that can be worn during normal soccer play would allow for collection of real-time game head accelerations without restricting player movement. The headband system will be modeled after a commercially available headgear, but will provide none of the protective effects to the players. Validation of the headband system is required before use in the field. It is hypothesized that the headband system and algorithm used to calculate injury criteria are correlated with the current laboratory standard using biomechanical surrogates and established protocols. These data are essential to ensure the impacts recorded with the system represent the true accelerations seen during the play of soccer.

**Methods**

A soccer headband HITS (Simbex Inc., Lebanon, NH), similar to those commercially available, was instrumented with 6 single-axis linear accelerometers ($\pm 250$ g) placed tangentially to the center of gravity of the head (Analog Devices, Inc.; Figure 1). To measure forces normally seen during the play of soccer, no padding was placed in the headband. All accelerometers were placed in the back of the headband to avoid ball contact during heading events. The battery pack, placed in the back of the headband, is a

---

**Figure 1** — HITS headband with circles indicating accelerometer placement.
rechargeable nickel metal-hydride battery which allows for extended use, 1–2 weeks depending on use, and minimal additional weight, with the entire headband system weighing 147 g. The headband has a threshold level of 10 g, meaning that when any accelerometer registers a reading of 10 g or greater, the impact will be downloaded. Once an impact above the threshold is recognized, 8 ms before the impact and 32 ms postimpact will be recorded. Data are downloaded to the sideline computer as long as players remain within range, approximately 200 yards. If players are out of range, up to 100 impacts can be stored within the headgear itself until the player returns within range.

The 50th percentile male HIII head was used as the standard of comparison for the linear and angular head accelerations for the HITS headband. The HIII head was instrumented with nine linear accelerometers (Endevco 7264C and 7264D) in the 3-2-2-2 setup (Padgaonkar et al., 1975) with a triaxial linear block placed at the center of gravity (CG) of the head form. HIII head acceleration data were collected at 20,000 Hz while HITS acceleration data were collected at 1,000 Hz. The headband was placed on the HIII head and Velcro straps were tightened to the manufacturer’s specifications allowing all accelerometers to make firm contact with the head form. Impacts occurred at the forehead, side, and temple of the HIII head using an air cannon and a linear impactor. No impacts were performed with an impact direction going directly through the center of gravity of the head as this would be highly unlikely in an on-field data collection scenario. Both ball to head and head to head contacts were simulated.

Ball to head impacts were performed using an air cannon with a barrel fitted to accommodate a soccer ball. The ball was shot through a three screen chronograph to obtain velocity readings. To obtain velocities representative of soccer impacts, Helium was used in the air cannon. Impacts were performed at 8 m/s, 10 m/s, and 12 m/s based on previous research performed (Withnall et al., 2005). Ten impacts were performed at each velocity to the forehead ($n = 30$), right side ($n = 30$), and left temple of the head ($n = 30$). These locations were chosen to represent various impacts seen during soccer play, as well as to provide a variety of impact locations possible in soccer games while not impacting accelerometers directly.

Head to head impacts were conducted by mounting one HIII head and neck face down, to a linear impactor and placing another on a trolley in front of the impactor (Figure 2). By placing the head and neck on the impactor, some flex was possible allowing for an impact more closely representative of an on-field situation. The head on the trolley was instrumented as described above and the HITS headgear was placed on it. Tests were run at three velocities: 2.5 m/s, 3.5 m/s, and 4.75 m/s based on previous research on head to head field impacts (Withnall et al., 2005). Ten impacts were performed at each of these velocities at two locations: the forehead ($n = 30$), and to the right side ($n = 30$). These locations represent standard impacts seen during soccer play, and provide a more severe impact condition than the ball to head impacts.

Data analysis was conducted to determine the agreement between the HIII and the HITS headgear. Linear head acceleration and angular head acceleration were calculated for both systems. The HITS system data were processed using a simulated annealing optimization algorithm, previously described in detail by Chu et al. (2006), which iteratively solves for linear and angular acceleration based on the 6 accelerometer measurements. Linear regression was used to compare the systems for the ball to head impacts, the head to head impacts, and then all impacts together. This was done for both linear and angular head accelerations.

In addition, root mean square (RMS) error was calculated using the equation below for the duration of the impacts for linear head acceleration. This provided information about specific portions of the curve and how closely they match up in value. Cross correlation was also calculated for linear head acceleration. These values provided insight into how strongly the variables are related. Cross correlation values were assessed using a scale to determine correlation strength: $>0.95$ was considered “excellent,” $>0.85$ was considered “good,” and $>0.75$ was considered “acceptable.” Due to the fact that the HIT System has built-in data acquisition and wireless communication, the HIII and HITS could not be linked. To compare HITS data and HIII data during postprocessing.

Figure 2 — Head to head impact test setup for forehead testing.
data were synchronized at the point of minimum RMS error. The two resultants were synchronized by shifting the HIII data incrementally until a 40 ms span of the HIII gave the lowest cross correlation factor.

\[
RMS \text{ Error} = \sqrt{\frac{\sum_{i=1}^{n} (x_{1i} - x_{2i})^2}{n}}
\]

where \(x_{1i}\) = HITS measurement at single time point and \(x_{2i}\) = HIII measurement at the same time point.

**Results**

Linear regressions were performed for the ball to head impacts, head to head impacts, and all impacts combined. All impact locations are combined in the linear regressions. Regressions for both linear and angular accelerations are shown below with each impact location denoted by a different shape (Figures 3–5).

Ball to head comparisons provided minimal correlation for both linear and angular acceleration, \(R^2 = .3403\) and \(R^2 = .5716\) respectively (Figure 3). This is most likely due to the fact that although various impact velocities were tested, a range of linear and angular head accelerations were not obtained. Output accelerations for both the HIII and HITS systems were very limited in range when impacted over the three ball to head impact velocities. Although the correlations are not ideal due to the lack of acceleration range, the average difference between the two acceleration measurements is minimal. This is especially true for the linear head acceleration which has an average difference of 2.25 g. The angular head acceleration has an average difference of 100.58 rad/s².

Head to head comparisons provided strong correlations for both linear and angular acceleration, \(R^2 = .8940\) and \(R^2 = .8998\) respectively (Figure 4). A much wider range of output velocities was provided from the three impact velocities used in the head to head impact conditions providing a much stronger dataset for linear regression. This demonstrates that the HITS is a very good system for measuring higher velocity impacts. The average differences between the two systems are –2.01 g for the linear acceleration measurements and –1721.05 rad/s² for the angular acceleration measurements. These differences are calculated over all three impact velocities.

Linear regressions were also performed for all head impacts combined providing a very strong correlation over the range of impact velocities that are anticipated for soccer games. This was done to determine the overall

---

**Figure 3** — Linear regression for HIII and HITS ball to head conditions a) linear acceleration b) angular acceleration.

**Figure 4** — Linear regression for HIII and HITS head to head conditions a) linear acceleration b) angular acceleration.
accuracy of the system for the range of velocities over which it will be used. Very strong correlations were found over all of the impact conditions, $R^2 = .9437$ and $R^2 = .9194$ for linear acceleration and angular acceleration respectively (Figure 5). This provides a very strong basis for using the system for future soccer research.

Root mean square error was calculated for each impact on a point by point basis. An example of the waveforms being compared is shown in Figure 6. An average was then calculated for each impact and each impact condition. The average RMS error of the linear accelerations for ball to head impacts at the 8 m/s condition was $2.04 \pm 0.25$ for forehead impacts, $3.55 \pm 0.44$ for right side, and $2.20 \pm 0.74$ for left temple impacts. Similarly RMS error for the 10 m/s condition was $2.35 \pm 0.27$ for forehead impacts, $3.86 \pm 0.62$ for right side, and $1.97 \pm 0.54$ for left temple impacts. The 12 m/s conditions had RMS errors of $2.55 \pm 0.77$ for forehead impacts, $3.08 \pm 0.94$ for right side, and $3.89 \pm 1.21$ for left temple impacts. Head to head impacts RMS errors were $2.69 \pm 0.32$ for the forehead and $5.83 \pm 0.67$ for the left side at the 2.5 m/s condition, $5.82 \pm 0.65$ for the forehead and $15.19 \pm 1.30$ for the left side at the 3.5 m/s condition, and $9.47 \pm 0.57$ for the forehead and $21.89 \pm 2.62$ for the left side at the 4.75 m/s condition.

Cross correlations were performed and demonstrate a strong relationship between the two systems. Average cross correlation ($r$) values were $0.95 \pm 0.01$ for forehead impacts, $0.88 \pm 0.05$ for right side, and $0.95 \pm 0.04$ for left temple impacts for the 8 m/s condition, $0.94 \pm 0.01$ for forehead impacts, $0.87 \pm 0.04$ for right side, and $0.96 \pm 0.02$ for left temple impacts for the 10 m/s condition, and $0.95 \pm 0.03$ for forehead impacts, $0.96 \pm 0.02$ for right side, and $0.92 \pm 0.04$ for left temple impacts for the 12 m/s condition. Head to head impacts had cross correlation values of $0.97 \pm 0.01$ for the forehead and $0.94 \pm 0.01$ for the left side at the 2.5 m/s condition, $0.98 \pm 0.00$ for the forehead and $0.88 \pm 0.04$ for the left side at the 3.5 m/s condition, and $0.94 \pm 0.00$ for the forehead and $0.83 \pm 0.02$ for the left side at the 4.75 m/s condition. All ball

---

**Figure 5** — Linear regression for HIII and HITS all conditions combined a) linear acceleration b) angular acceleration.

**Figure 6** — Linear acceleration for both HIII and HITS for one ball to head forehead impact at 12 m/s.
to head conditions fell either within the good or excellent range when looking at cross correlation values. Of the nine ball to head conditions, five of them were above the 0.95 value required for an excellent rating. Head to head conditions also provided very strong cross correlation values, with five of six conditions falling into either the good or excellent categories.

Peak linear and rotational acceleration was also calculated for each of the impacts. These values for both the HITS and HIII systems are shown in Tables 1–4. All values shown are the average for the impact condition listed.

The HITS system slightly over predicts peak linear head acceleration in the ball to head impacts. This can be seen in all impact conditions, except the forehead condition at 12 m/s. At this condition the HITS provides a slight under prediction (Table 1). Angular head acceleration for the ball to head impacts provides a different pattern for the peak values. The HITS system over predicts angular head acceleration in the ball to head impacts for all conditions except the forehead impacts. Forehead impacts at all three velocities have the HITS under predicting angular acceleration (Table 2).

Linear head acceleration for the head to head impacts shows a general over prediction by the HITS for the two lower impact velocities. At the 4.75 m/s impact condition, the HITS under predicts linear head acceleration for both the forehead and left side conditions. Although a general over prediction occurs, values are very similar as shown by the strong correlations between the two systems (Table 3). Angular head acceleration for the head to head impacts shows that the HITS under predicts the angular head acceleration slightly for nearly all the impact conditions. The HITS system only over predicts angular head acceleration in the head to head impacts for one impact condition, the left side at 2.5 m/s. This is shown in below Table 4.

**Discussion**

Although attempts have been made to determine head acceleration during soccer heading events, a system for on field data collection had not been previously available. A system has now been created for research purposes; however, validation was necessary before the system could be used to collect data during a game or scrimmage situation. The results show that the new soccer HITS system correlates well with the standard measurement system of the HIII 3-2-2-2 accelerometer system. Locations and impact velocities were chosen to simulate events that take place in normal soccer play.

Good cross correlation values were found for the linear accelerations for all conditions with two of the conditions having excellent correlation. The lowest correlation value of 0.83 ± 0.02 was for the left side during head to head impact and is not a location that is expected to be frequently impacted during soccer play. Even as the lowest correlation, it still shows an acceptable level of agreement. In addition, all other linear acceleration cross correlation values exceed the 0.85 value and shows a very well matched system.

Strong correlations were found between the systems for both linear and angular head acceleration for the head to head condition, 0.8940 and 0.8998 respectively. In addition, very strong correlation was found for overall use of the system. This was shown by performing linear regression over all conditions with results of 0.9437 for linear acceleration and 0.9194 for angular acceleration. The ball to head condition did not have a strong correlation, but this is due to the lack of velocity distribution as all of the impacts were at a very low magnitude. Although these impacts has low $R^2$ values, they did have a very small absolute difference, 2.25 g for linear head acceleration and 100.58 rad/s$^2$ for angular head acceleration. In addition, all average peak values for linear acceleration were well below 66 g which has been previously established as a 25% risk of injury (Zhang et al., 2004). This indicates that a difference of ± 2.25 g would not be clinically significant.

Root mean square error values showed a consistency between the two waveforms for each of the conditions (Figure 6). Slightly higher average RMS error values were found in some waveforms, but upon further inspection it seems as though the discrepancy in the waveforms took place in the tail of the impact or in a small secondary impact but not in the peak. Therefore, even in the impacts with a slightly higher average RMS error the peaks were still similar.

One limitation of this system is that it requires the player to wear some type of headband to allow for player instrumentation to take place. The system has previously been used in sports that require helmets or headgear of some type, but in soccer this is not the case. Although headgear is available for soccer players, it is not a required piece of equipment. Therefore, it could be more challenging to find players willing to wear the system during play. An additional concern with the system is movement during play (Beckwith et al., 2007). Although this is a concern, movement would most likely just alter the accuracy of the impact location (Beckwith et al., 2007). Headband slippage was not a problem in laboratory tests, and impact location is not a primary interest for use with this system. Therefore, headband slippage is not considered a major concern but on-field research is warranted to assess these concerns.

During soccer play, many impact scenarios exist, and although the current study made every effort to recreate scenarios typically seen during soccer play, it was impossible to include all scenarios. One limitation is the limited impact locations and velocities included. Soccer has a wide range of impact locations and although they are not all included, the range included provided sufficient simulation of events to validate the system. An additional limitation is the simulation of on-field scenarios accurately. Using two HIII heads in head to head impacts as opposed to just an impactor ram provided some give due to the neck flexion on the impactor, but may not be an exact replication of on-field impacts. The
### Table 1  Average peak linear accelerations for ball to head conditions

<table>
<thead>
<tr>
<th>Impact Location</th>
<th>8 m/s</th>
<th></th>
<th>10 m/s</th>
<th></th>
<th>12 m/s</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>HITS (g)</td>
<td>HIII (g)</td>
<td>HITS (g)</td>
<td>HIII (g)</td>
<td>HITS (g)</td>
<td>HIII (g)</td>
</tr>
<tr>
<td>Forehead</td>
<td>15.20 ± 0.49</td>
<td>12.14 ± 0.64</td>
<td>18.23 ± 1.51</td>
<td>16.57 ± 0.70</td>
<td>18.70 ± 2.54</td>
<td>21.09 ± 1.35</td>
</tr>
<tr>
<td>Right Side</td>
<td>18.03 ± 4.85</td>
<td>13.31 ± 0.76</td>
<td>19.93 ± 3.20</td>
<td>17.67 ± 0.87</td>
<td>22.98 ± 4.02</td>
<td>21.13 ± 0.93</td>
</tr>
<tr>
<td>Left Temple</td>
<td>18.03 ± 3.53</td>
<td>13.39 ± 0.46</td>
<td>18.72 ± 3.26</td>
<td>18.08 ± 0.46</td>
<td>21.64 ± 4.33</td>
<td>21.45 ± 1.11</td>
</tr>
</tbody>
</table>

### Table 2  Average peak angular accelerations for ball to head conditions

<table>
<thead>
<tr>
<th>Impact Location</th>
<th>8 m/s</th>
<th></th>
<th>10 m/s</th>
<th></th>
<th>12 m/s</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>HITS (rad/s²)</td>
<td>HIII (rad/s²)</td>
<td>HITS (rad/s²)</td>
<td>HIII (rad/s²)</td>
<td>HITS (rad/s²)</td>
<td>HIII (rad/s²)</td>
</tr>
<tr>
<td>Forehead</td>
<td>822.07 ± 244.4</td>
<td>833.12 ± 74.40</td>
<td>857.74 ± 259.28</td>
<td>954.86 ± 95.32</td>
<td>949.57 ± 362.40</td>
<td>1450.81 ± 252.08</td>
</tr>
<tr>
<td>Right Side</td>
<td>1416.29 ± 94.57</td>
<td>1343.23 ± 197.30</td>
<td>1828.96 ± 334.27</td>
<td>1509.16 ± 152.15</td>
<td>1882.67 ± 318.00</td>
<td>1715.39 ± 378.91</td>
</tr>
<tr>
<td>Left Temple</td>
<td>1958.79 ± 425.65</td>
<td>1321.29 ± 179.61</td>
<td>1453.86 ± 535.28</td>
<td>1281.71 ± 282.16</td>
<td>1917.18 ± 348.35</td>
<td>1695.22 ± 347.64</td>
</tr>
</tbody>
</table>

### Table 3  Average peak linear accelerations for head to head conditions

<table>
<thead>
<tr>
<th>Impact Location</th>
<th>2.5 m/s</th>
<th></th>
<th>3.5 m/s</th>
<th></th>
<th>4.75 m/s</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>HITS (g)</td>
<td>HIII (g)</td>
<td>HITS (g)</td>
<td>HIII (g)</td>
<td>HITS (g)</td>
<td>HIII (g)</td>
</tr>
<tr>
<td>Forehead</td>
<td>33.68 ± 0.55</td>
<td>32.54 ± 0.69</td>
<td>74.26 ± 4.10</td>
<td>66.93 ± 1.85</td>
<td>115.45 ± 8.02</td>
<td>125.14 ± 2.80</td>
</tr>
<tr>
<td>Left Side</td>
<td>37.14 ± 3.46</td>
<td>30.58 ± 2.06</td>
<td>69.58 ± 10.12</td>
<td>63.18 ± 3.31</td>
<td>95.82 ± 13.02</td>
<td>119.65 ± 4.94</td>
</tr>
</tbody>
</table>

### Table 4  Average peak angular accelerations for head to head conditions

<table>
<thead>
<tr>
<th>Impact Location</th>
<th>2.5 m/s</th>
<th></th>
<th>3.5 m/s</th>
<th></th>
<th>4.75 m/s</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>HITS (rad/s²)</td>
<td>HIII (rad/s²)</td>
<td>HITS (rad/s²)</td>
<td>HIII (rad/s²)</td>
<td>HITS (rad/s²)</td>
<td>HIII (rad/s²)</td>
</tr>
<tr>
<td>Forehead</td>
<td>1245.63 ± 109.99</td>
<td>1544.50 ± 32.89</td>
<td>2013.79 ± 176.65</td>
<td>3177.79 ± 205.66</td>
<td>6930.20 ± 530.95</td>
<td>9414.49 ± 218.99</td>
</tr>
<tr>
<td>Left Side</td>
<td>2795.55 ± 179.72</td>
<td>2366.77 ± 456.59</td>
<td>5629.58 ± 550.05</td>
<td>6020.31 ± 365.15</td>
<td>9801.20 ± 1413.74</td>
<td>16218.41 ± 855.65</td>
</tr>
</tbody>
</table>
Current study was not designed to recreate exact scenarios, but to provide reasonable recreations to determine a correlation between the acceleration measurement systems. Therefore, the system needs to be used in on-field situations to determine fully its ability to accurately measure soccer head impacts.

Rotational acceleration for left side, head to head impacts had the largest difference measurement for any test method. For these impacts, the HITS under predicted the HII by 6417.21 rad/s^2. While this discrepancy is of concern, it is unclear how translatable these results will be to in vivo data collection. These test conditions are intended to be representative of on-field events; however, the complex biomechanical interactions that take place during live impacts may not be completely captured by our simulated event as it was impossible to recreate every possible impact scenario. Due to the high correlation found for all other test combinations, we suggest the HITS is a viable method for recording impacts during competition, however, while linear acceleration measures appear acceptable, caution should be taken when evaluating rotational acceleration for impacts similar to our head to head condition. As part of this ongoing work, future studies will address this concern by identifying head to head impacts through video analysis and comparing on-field measures with those recorded here.

In conclusion, this system provides a much needed method to measure head acceleration in soccer players during normal play. It allows for accurate measurements to be taken which could potentially lead to an injury threshold specific to certain soccer impacts. In addition, this system will allow a comparison between different types of impacts that occur during soccer play.

Acknowledgments

This study was funded by the National Operating Committee on Standards for Athletic Equipment (NOCSAE). The work was also supported in part by the Anthony and Joyce Danielski Kales Scholarship. The authors would also like to acknowledge Jonathan Beckwith and Bulent Ozkan.

References


