2017

Comparison of a Head Mounted Impact Measurement Device to the Hybrid III Anthropomorphic Testing Device in a Controlled Laboratory Setting

Eric Schussler
Old Dominion University, eschussl@odu.edu

David Stark

John H. Bolte

Yun Seok Kang

James A. Onate

Original Publication Citation
Schussler, E., Stark, D., Bolte, J. H., Kang, Y. S., & Onate, J. A. (2017). Comparison of a head mounted impact measurement device to the Hybrid III anthropomorphic testing device in a controlled laboratory setting. *International Journal of Sports Physical Therapy*, *12*(4), 592-600.

IJSPT

ORIGINAL RESEARCH

COMPARISON OF A HEAD MOUNTED IMPACT MEASUREMENT DEVICE TO THE HYBRID III ANTHROPOMORPHIC TESTING DEVICE IN A CONTROLLED LABORATORY SETTING

Eric Schussler, PhD, PT, ATC[1]
David Stark, MS[2]
John H. Bolte, IV, PhD[2]
Yun Seok Kang, PhD[2]
James A. Onate, PhD, ATC, FNATA[2]

ABSTRACT

Background: Reports estimate that 1.6 to 3.8 million cases of concussion occur in sports and recreation each year in the United States. Despite continued efforts to reduce the occurrence of concussion, the rate of diagnosis continues to increase. The mechanisms of concussion are thought to involve linear and rotational head accelerations and velocities. One method of quantifying the kinematics experienced during sport participation is to place measurement devices into the athlete's helmet or directly on the athlete's head.

Purpose: The purpose of this research to determine the accuracy of a head mounted device for measuring the head accelerations experienced by the wearer. This will be accomplished by identifying the error in Peak Linear Acceleration (PLA), Peak Rotational Acceleration (PRA) and Peak Rotational Velocity (PRV) of the device.

Study Design: Laboratory study.

Methods: A helmeted Hybrid III 50th percentile male headform was impacted via a pneumatic ram from the front, side, rear, front oblique and rear oblique at speeds from 1.5 to 5 m/s. The X2 Biosystems xPatch® (Seattle, WA) sensor was placed on the headform's right side at the approximate location of the mastoid process. Measures of PLA, PRA, PRV from the xPatch ® and Hybrid III were analyzed for Root Mean Square Error (RMSE), and Absolute and Relative Error (AE, RE).

Result: Seventy-six impacts were analyzed. All measures of correlation, fixed through the origin, were found to be strong: PLA $R^2 = 0.967$ $p < 0.01$, PRA $R^2 = 0.933$ $p < 0.01$, PRV $R^2 = 0.999$ $p < 0.00$. PLA RMSE was 34%, RE 31.0% $\pm$ 14.0, and AE 31.1% $\pm$ 13.7. PRA RMSE was 23.4%, RE -6.7 $\pm$ 22.4 and AE 18.9% $\pm$ 13.8. PRV RMSE was 2.2%, RE 0.1 $\pm$ 2.2, and AE 1.8 $\pm$ 1.3.

Conclusion: Without including corrections for effect of skin artifact, the xPatch® produces measurements highly correlated with the gold standard yet above the average error of testing devices in both PLA and PRA, but a low error in PRV. PLA measures from the xPatch® system demonstrated a high level of correlation with the PLA data from the Hybrid III mounted data collection system.

Level of Evidence: 3

Key words: Concussion, head acceleration, head velocity

[1] Old Dominion University, Norfolk, VA, USA
[2] The Ohio State University, Columbus, OH, USA

Disclosure
None

Acknowledgements
We thank Sam Goldman, Alex Redrow, Arrianna Willis for testing setup and data acquisition.

CORRESPONDING AUTHOR
Eric Schussler
Old Dominion University
3107 Health Sciences Bldg, Norfolk, VA 23529
E–mail: ESchussl@odu.edu

INTRODUCTION

Reports estimate that between 1.6 and 3.8 million cases of concussion occur in sports and recreation each year in the United States.[1] Despite continued efforts to reduce the occurrence of concussion, the rate of diagnosis continues to increase. The rate of occurrence has increased from 0.23 to 0.51 per 1000 high school athlete exposures in the period from 2005-2006 to 2011-2012.[2] The cause of these concussions is often contact of a player's head with the opponent's head, body, or the ground.[3] Sports related concussions affect more than 5% of high school and collegiate football players. Nearly 15% of the affected population goes on to sustain repeat concussions within the same season.[4] Concussion has been identified as a potential risk factor for neurodegenerative dementia and decreased neurocognitive performance.[5,6] Researchers utilize measurement devices to quantify the head impacts experienced by players in order to improve safety of athletes, but the technology of these devices often moves faster than the ability to independently test their accuracy.

A method of quantifying head impulses experienced during sport is to place measurement devices such as accelerometers and gyroscopes into the athlete's helmet or directly on the athlete's head. Such technology allows potentially injurious accelerations to be quantified immediately during participation as well as for the collection of cumulative impact data from multiple players over the course of a season.[7,8] Current wearable systems, often worn on or in helmets, have shown to be ineffective in accurately measuring the kinematics of the head.[9–15] The X2 Biosystems' xPatch® system (Seattle, WA) is an option to measure head kinematics directly from the scalp. This system consists of a triaxial linear accelerometer and a triaxial gyroscope contained in a 1½ in by ½ in device that attaches directly to the skin over the mastoid process of the athlete (Figure 1A). The system allows measurement of head accelerations for activities whose participants do not typically wear headgear or helmets. Previous research on the accuracy of a mouthguard based system manufactured and tested by the same company indicated the design may be capable of accurately measuring the accelerations experienced by the head[16,17] yet limited research currently exists on the ability of the xPatch® device to accurately measure head accelerations.

The purpose of this study is to determine the measurement error of the xPatch® system when compared to a gold standard Hybrid III 50th percentile male headform (HIII) in a laboratory setting. Accurate measurements of the kinematics of the head during athletic competition are important in determining the risk of concussion, alerting medical personnel of the need for secondary evaluation and for developing concussion prevention strategies and equipment. It is theorized that these devices will have error percentages equivalent to other equipment currently commercially available.

METHODS

A helmeted HIII headform (Humanetics, Plymouth MA) was impacted via a pneumatic ram at varying impact speeds and directions. The HIII headform was attached to a HIII neck secured to a 40.23 kg mass on roller bearings which approximates the mass of a human thorax.[18] The head was level (i.e. 0°of tilt) for all impacts. The xPatch® sensor was placed on the headform's right side at the approximate location of the mastoid process. Orientation of the sensor was set with the front of the device oriented perpendicular to the plane of testing (Figure 1A). The sensor was attached utilizing the manufacturer provided adhesive patch. The headform was then fitted with a Schutt Stallion lacrosse helmet (STX, Model: Stallion 500, Baltimore, MD) under which a wig comprised of human hair was placed and kept moist with a spray bottle, to simulate sweat. A Cascade STX chinstrap (Cascade Sports, Liverpool, NY) was used to secure the helmet to the headform and the helmet was fitted using manufacturer's recommendations.

A pneumatic ram weighing 23.9 kg was utilized to impact the helmet (Figure 1B). The impacting surface of the ram was an 8.25 inch diameter cylinder made of ultrahigh molecular weight polyethylene (UHMWPE). The helmet was impacted at four different speeds: 1.5, 2.5, 3.75, and 5 m/s from five different locations: frontal, side, rear, front-oblique and rear-oblique (Figure 2) to create a kinematic profile of impacts previously described to be representative of impacts experienced during play in light helmeted sports.[14,19] Ram speed was controlled to within 0.1m/s. These speeds of impact were shown to produce accelerations at the head which have been reported during sports participation.[10,20,21]

Figure 1. *A) Front-Oblique testing setup. B) Affixation of the xPatch Device to the HIII.*

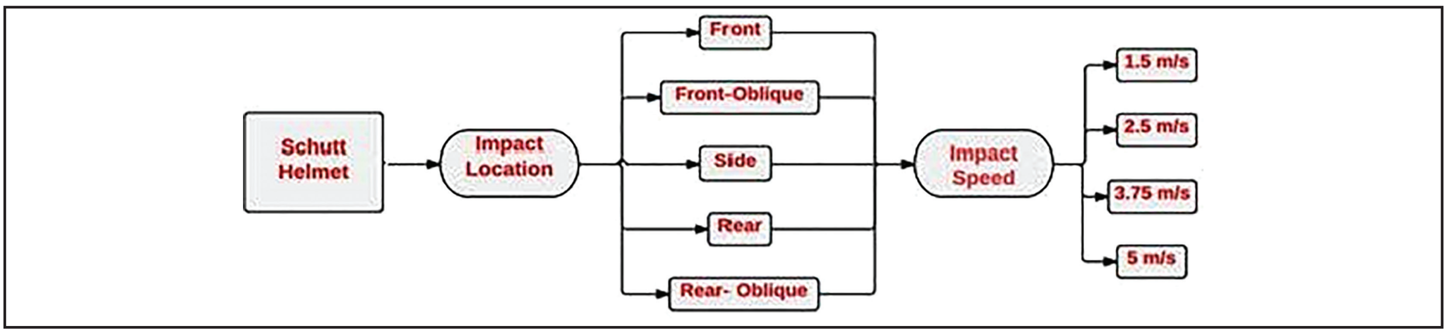

Figure 2. *Testing Matrix.*

All oblique impacts were directed 45° from the mid-sagittal plane. Each impact was directed through the center of gravity of the headform. Four impacts were performed at each selected speed for each selected direction. The number, location and velocity of the impacts were controlled rather than resultant kinematics.

Linear accelerations and angular velocities of the HIII were recorded utilizing sensors placed within the headform. Piezoelectric accelerometers (Meggitt's Endevco, Model #:7246C-2000, Irvine CA) were used to measure linear acceleration while Angular Rate Sensors (ARS) (DTS ARS P18K Pro, Seal Beach CA) were used to measure angular velocity. Accelerometers were organized within the headform in a 3-2-2-2 configuration described by Padgaonkar[22] typical for acquiring head impact data. Linear accelerations and angular velocities were stored using data acquisition system TDAS G5® (DTS, Seal Beach CA). Data were imported into DIAdem (National Instruments, Austin TX) where it was zeroed and filtered using CFC 1000 filters as per standard practices described in the Society of Automotive Engineers standard J211-1.[23] While angular accelerations were not directly measured within the headform, they were calculated using algebraic equations as described by Padgaonkar.[22]

Immediately prior to data collection, the xPatch® units were synced with the data recording system which set the time stamp on each device to the computer generated time. This same computer was utilized to run to HIII data acquisition system. Data from the xPatch® sensor were downloaded from the device into the manufacturer's software at the completion of testing. This software automatically converts the linear acceleration data from the lateral aspect of the head (a_p) to the center of gravity (a_{CG}) (Equation 1) where ω is angular velocity, $\dot{\omega}$ is angular acceleration and $r_{p\rightarrow CG}$ is the geometrical relationship between point P, the location of the device on the head and Q, the location of the center of gravity of the head.[24]

$$a_{CG} = a_p + \omega \times (\omega \times r_{p\to CG}) \dot{\omega} \times r_{p\to CG} \qquad \text{(Eq. 1)}$$

These calculations take place within the system utilizing manufacturer preset $r_{p\to CG}$ distance and orientation which are not altered by the user. All data were transferred to excel spreadsheets to coordinate time stamps. The impacts measured by the xPatch® and sensors within the HIII were matched utilizing the time stamp on the xPatch® device and the data acquisition system regardless of the system determination of an actual impact or a non-intentional impact. The xPatch® system identifies false impacts through two methods including comparison to a set of wave form parameters and comparison to a reference waveform using cross-correlation.[25] Two hundred and forty eight impacts were recording during testing.

STATISTICAL METHODS

Correlational analysis of peak linear acceleration, peak rotational acceleration and velocity were computed utilizing SPSS version 24 (IBM Armonk, NY). Error between the devices was calculated in Percent Relative Error (Equation 2), Percent Absolute Error (Equation 3) and Root mean square error (RMSE) (Equation 4) for each of the peak accelerations and velocities[14].

$$\text{Percent Relative Error} = \frac{\text{HIII Measure} - \text{xPatch Measure}}{\text{HIII Measure}} * 100 \qquad \text{(Eq. 2)}$$

$$\text{Percent Absolute Error} = \left| \frac{\text{HIII Measures} - \text{xPatch Measure}}{\text{HIII Measure}} \right| * 100 \qquad \text{(Eq. 3)}$$

$$RMSE = \sqrt{\frac{\sum_i^N (\text{Relative Error})^2}{N}} \qquad \text{(Eq. 4)}$$

RESULTS

Seventy-six tests were utilized for comparison of the xPatch® and HIII systems. Front and front oblique low speed tests included two impacts each that were below the 10*g* threshold of the xPatch® system as confirmed by the HIII system, thus they have been excluded from analysis. All remaining tests were identified as impacts by the xPatch® system and were not determined to be non-impact signals referred to by the manufacturer as "clacks".

Resultant peak linear accelerations from the HIII headform ranged from 7.1 to 134.5g (average = 40.4*g* $\pm$27.5), resultant peak rotational accelerations ranged from 606.8 to 8328.6 rad/s^2 (average = 2862.9 rad/s^2 $\pm$1889.2) and resultant peak rotational velocity ranged from 7.5 to 42.5 rad/s (average = 22.7 rad/s $\pm$9.7).

Analysis of the data indicated high correlations between the xPatch® and the HIII system. A correlation of linear acceleration fixed through the origin was found to be strong ($p < 0.01$, $R^2 = 0.967$). The results of rotational velocity were strongly correlated ($p < 0.00$, $R^2 = 0.999$) along with rotational acceleration ($p < 0.01$, $R^2 = 0.933$) when regressed through the origin (Figure 3).

Analysis of the RE, AE and RMSE for Linear, and Rotational Acceleration and Rotational Velocity are presented in Table 1. RMSE for PLA in all combined directions was 34%, 2.8% for PRV and 23.4% for PRA. Percent relative error by averaged HIII and xPatch® measurements indicate an average error of 31% $\pm$14.1 PLA, -6.7% $\pm$22.6 PRA and 1.7% $\pm$2.2 PRV (Table 1).

Bland-Altman Plots are presented (Figure 4) for the percent error of each of the measures at the average measure of each of the devices. Average percent error for PLA was 31.0 (Limit of Agreement; 58.6, 3.3), average percent error for PRA was -6.7 (Limit of agreement: 37.6, -51.0), indicating an under estimation of the acceleration by the xPatch® system and average percent error for PRV was 1.7 (Limit of agreement: 6.0, -2.6).

Significantly higher error was found between devices in RE and AE between the oblique measures over the non- oblique measures of PLA (Figure 5). Significant differences in AE: Front to Front Oblique ($p = 0.029$), Front to Rear Oblique ($p < 0.001$), Front-Oblique to Rear ($p = 0.018$), Side to Rear Oblique ($p = 0.010$), Rear Oblique to Rear ($p < 0.001$). Significant Differences in RE: Front to Rear Oblique ($p < 0.001$), Front-Oblique to Rear

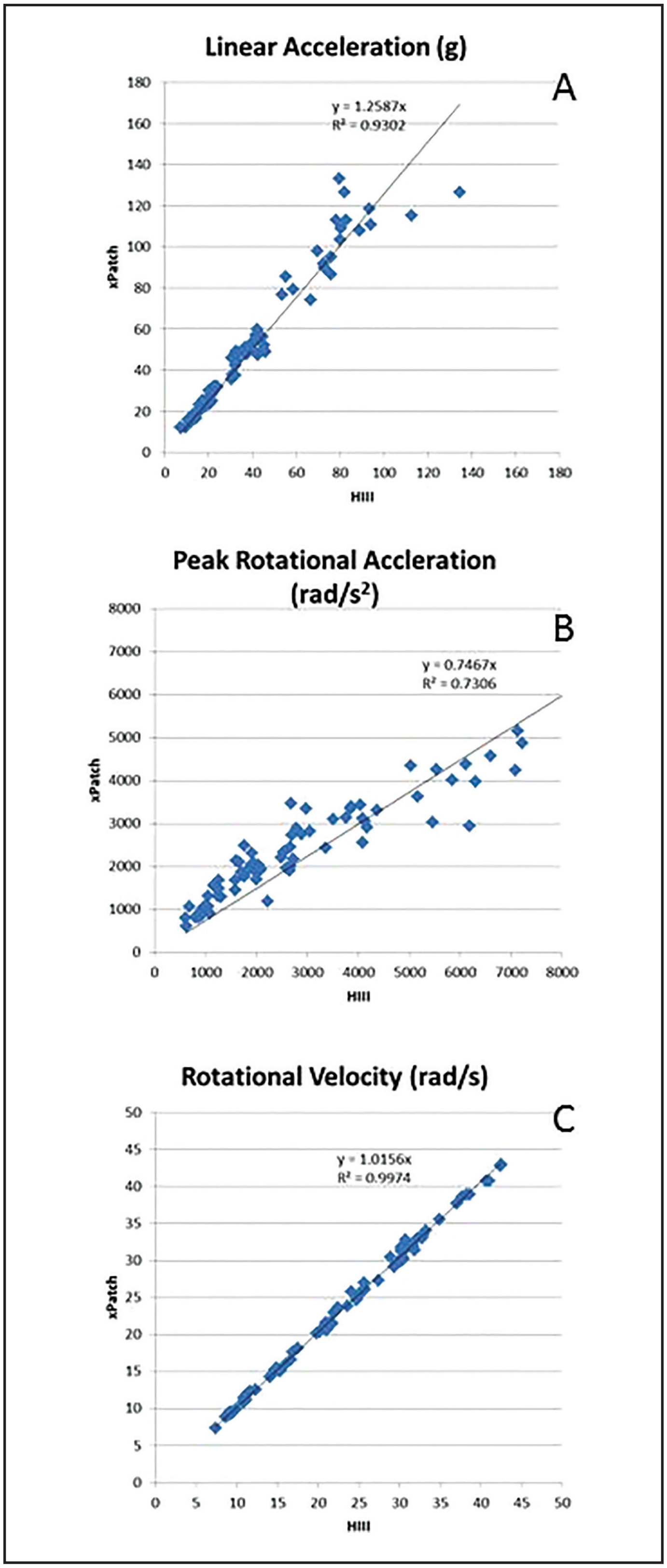

Figure 3. *Correlation of Measurements from xPatch to HIII system.*

Table 1. *Error by Type and Measurement*

	Peak Linear Acceleration	Peak Rotational Velocity	Peak Rotational Acceleration
Percent Relative Error	31.0 ± 14.0	0.1 ± 2.2	-6.7 ± 22.4
Percent Absolute Error	31.1 ± 13.7	1.8 ± 1.3	18.9 ± 13.8
Percent Root Mean Square Error	34.0	2.8	23.4

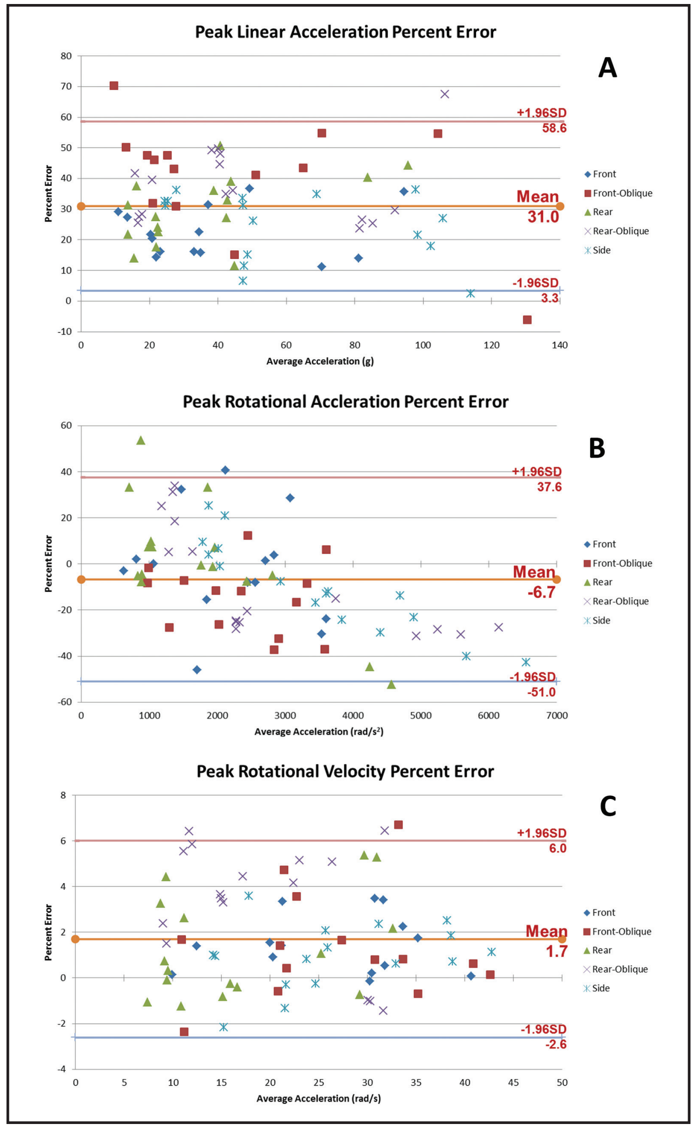

Figure 4. *Bland-Altman Plot of Percent Error in A) Linear Acceleration B) Rotational Acceleration C) Rotational Velocity*

($p = 0.044$), Side to Rear Oblique ($p = 0.016$), Rear Oblique to Rear ($p < 0.001$). Significant differences were found in PRV between rear oblique measures and all other measures in RE and AE at $p < 0.001$ (Figure 6). No significant difference was found between AE in PRA ($p = 0.199$), with the only difference in RE between Rear Oblique and Rear ($p = 0.05$, Figure 7).

DISCUSSION

As has been previously reported,[15,26,27] there is a high correlation between the measurements of linear acceleration from a gold standard system and linear acceleration reported by the xPatch® system yet higher pooled RE and AE than other devices previously studied.[15] All xPatch® measures of linear acceleration over-estimated the linear acceleration recorded by the HIII system. While pooled rotational acceleration measures were comparable to previously reported error seen in the X2 mouth guard and the Head Impact Telemetry® (HIT) System (Simbex, Lebanon, NH),[15] the rotational velocity measures were quite accurate when compared to both the gold standard and other tested systems on market. Whereas linear accelerations were routinely over estimated, the RMSE of rotational acceleration from the xPatch® was found to be 24%, with errors

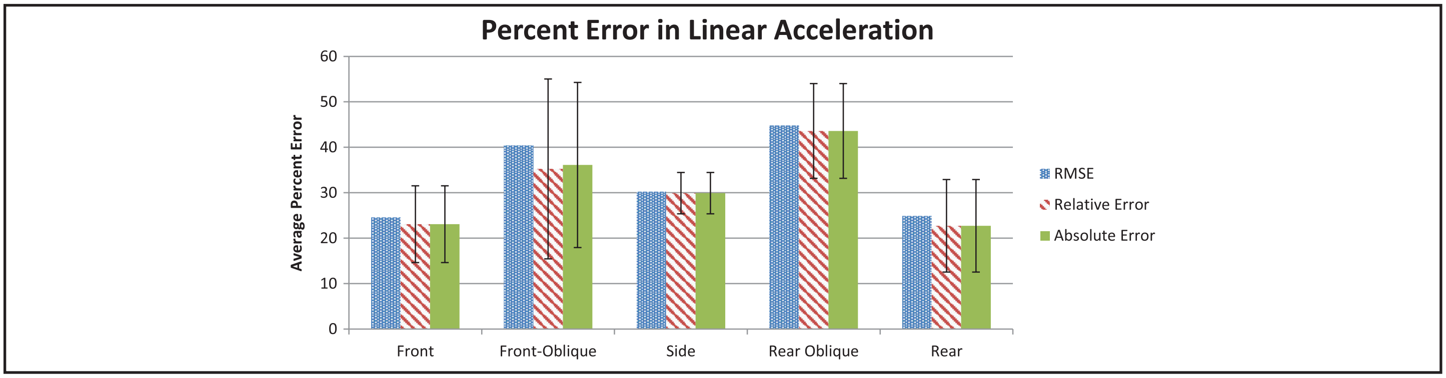

Figure 5. *Percent Error in Linear Acceleration*

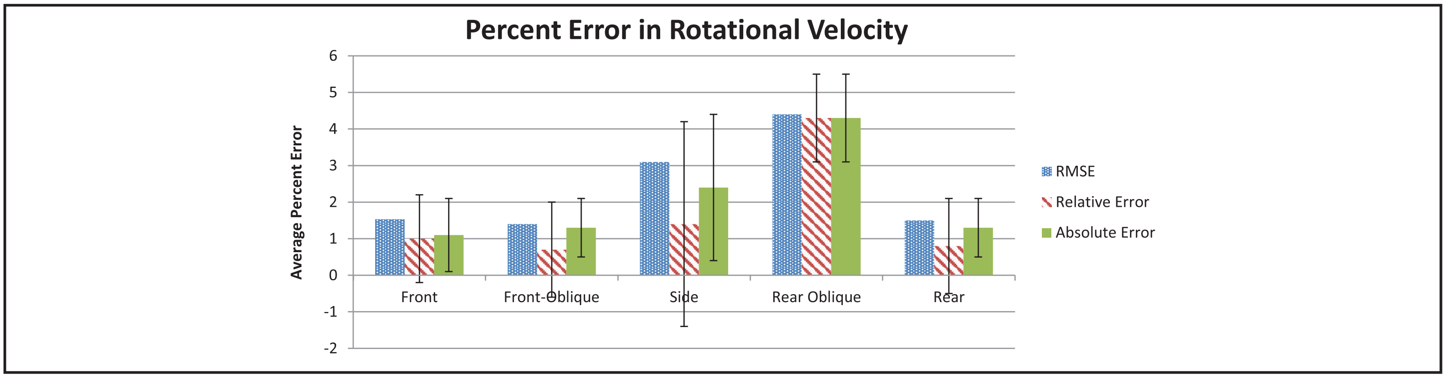

Figure 6. *Percent Error in Rotational Velocity.*

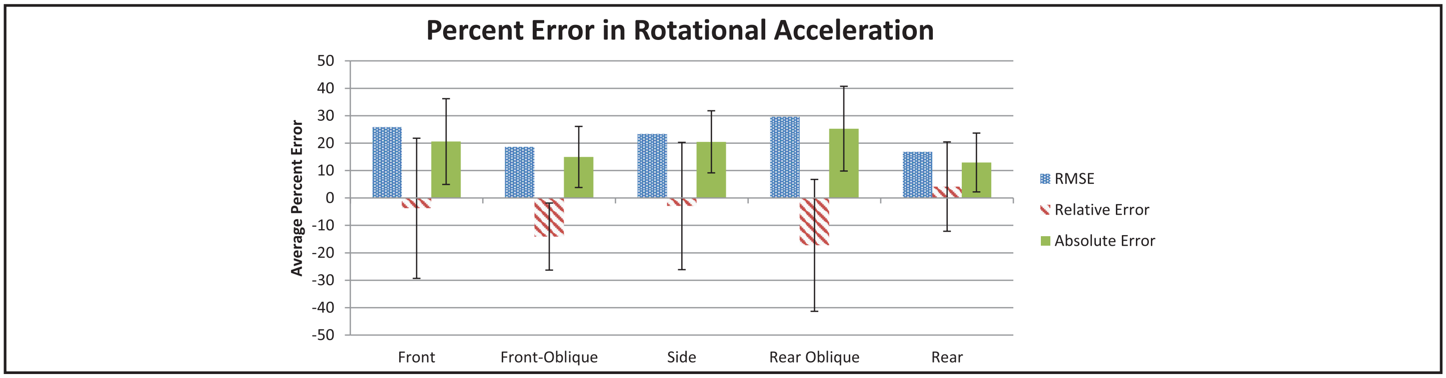

Figure 7. *Percent Error in Rotational Acceleration.*

Table 2. *Percent Root Mean Square Error (RMSE) Measures by Direction and Magnitude of Impact*

	Impact Magnitude	Front (%)	Front-oblique (%)	Side (%)	Rear-oblique (%)	Rear (%)	Average (%)
Peak linear acceleration (PLA)	10 to <45g	25 (n=13)	47 (n=10)	32 (n=11)	40 (n=10)	22 (n=9)	35
	45 to <80g	15 (n=1)		27 (n=4)	51 (n=5)	23 (n=6)	35
	>80g		12 (n=4)	37 (n=1)	61 (n=1)	38 (n=1)	29
Percent RMSE for PLA by Direction		24	35	28	49	24	34
Peak rotational acceleration (PRA)	<4000 rad/s^2	24 (n=11)	12 (n=10)	20 (n=11)	24 (n=11)	15 (n=14)	19
	4000 to 7000 rad/s^2	33 (n=3)	19 (n=2)	26 (n=3)	40 (n=5)	24 (n=2)	31
	>7000 rad/s^2		36 (n=2)	36 (n=2)			36
Percent RMSE for PRA by Direction		26	19	27	30	17	23
Peak rotational velocity (PRV)	<20 rad/s	1.2 (n=3)	0.6 (n=2)	4.0 (n=8)	3.5 (n=8)	0.7 (n=8)	2.9
	20 to 30 rad/s	2.2 (n=3)	1.9 (n=6)	2.6 (n=4)	5.2 (n=5)	1.4 (n=4)	3.0
	>30 rad/s	1.3 (n=8)	1.8 (n=6)	1.1 (n=4)	5.2 (n=3)	2.5 (n=4)	2.4
Percent RMSE for PRV by Direction		1.5	1.5	3.2	4.5	1.5	2.8

PLA= Peak linear acceleration, PRA= Peak rotational acceleration, PRV= Peak rotational velocity

in both the positive and negative direction. The lack of uniformity in the error makes this measure difficult to interpret when measuring on the field accelerations, limiting clinical utility of the measure. The rotational velocity measure however was very accurate at 2.8% RMSE. The rotational velocity of the head is utilized for calculation of the Brain Injury Criterion (BrIC) which has strong support to be a predictor of brain injury in other applications such as motor vehicle accidents[28]. When analyzing the impacts from differing directions, oblique measures had higher error between the HIII and xPatch® measures for PLA and PRV. This response precludes the PLA and PRA data from being utilized to quantify peak acceleration of a single impact but does allow continued use of the pooled data to identify impact trends.

By directly placing the xPatch® on the head, this type of system may minimize the effect of interaction that has been found between the helmet and the head. Despite this, the design of the xPatch® has been shown to allow extraneous motion due to skin artifact. Accelerometers mounted on or within the helmet experience accelerations much larger than those that are experienced at the head.[29] Athletes who do not wear tightly fitting equipment[30] increase the disconnection between the accelerometers and the head, potentially introducing error.[13] Studies identifying the motion artifact of the device during in vivo use found the device displaced on average 4mm from reference with the skull.[26] This may affect the interpretation of these results, as the HIII skin may not accurately recreate the motion artifact of human skin. By placing the monitoring system directly on the head, the measurements are unaffected by the interaction of the impact and helmet design but the xPatch® unit has more extraneous motion than a mouthpiece based system.[26]

Identification of head impact severity utilizing head mounted and helmet mounted accelerometers and gyroscopes has been indicated to be unable to produce an accurate measurement of the forces experienced. The studies discussed throughout this paper indicate the usefulness of these devices may not lie in detecting a single impact injury but in quantifying the number of impacts that occur or utilization of pooled data. Future research should be performed to identify the effectiveness of these devices in monitoring total head impact exposure and the effectiveness of combined video review and impact profiles to reduce overall head contact.

Limitations

Limitations of this study include the absence of a mechanism to account for the difference in displacement of the device on human skin compared to the covering of the ATD. The xPatch® device is placed over the mastoid process and may displace up to 4mm[26] introducing error to the measure that was not replicated in this study. The data collected contains more data points at lower accelerations. While this profile is representative of the effects of impacts expected in participation, additional testing should include higher acceleration tests. This research utilized one style of helmet throughout; different styles of helmets may change the acceleration profile which can be interpreted differently by the device. Because this device was tested specifically in conjunction with lacrosse helmets, the impactor design and speeds were best suited to recreate the acceleration profile involved in light helmeted sports.[19,31] Additional work should be performed to identify the error source of this device through analysis of the raw data rather than the data as interpreted by the xPatch® software. This analysis may reveal a systematic error in the handling of the acceleration data by the software. Research should also identify the accuracy of derived measures of head acceleration such as Head Impact Criterion and Gadd Severity Index from the xPatch® compared to a gold standard measure. Future work should expand the range of impact speeds to be representative of the impacts experienced in different sports to accurately reflect the possible uses of the device.

CONCLUSIONS

Accurate measurement of head accelerations experienced during sports participation is necessary for determining the specific mechanics of concussion in sport, determining methods to reduce concussions and identification of those who have experienced an impact that may have caused a concussion. The xPatch® System provides a strongly correlated overestimation of linear acceleration and a high level of accuracy in rotational velocity when compared to a gold standard measure. As linear acceleration is often the primary injury criteria used in sport at this time, consistent over estimation of the linear acceleration makes the xPatch® system a good tool to identify those in need of secondary injury screening by a qualified medical professional.

REFERENCES

1. Langlois JA, Rutland-Brown W, Wald MM. The epidemiology and impact of traumatic brain injury: a brief overview. *J Head Trauma Rehabil*. 2006;21(5):375-378.
2. Rosenthal JA, Foraker RE, Collins CL, Comstock RD. National high school athlete concussion rates from 2005-2006 to 2011-2012. *Am J Sports Med*. 2014;42(7):1710-1715.3.
3. Zhang L, Yang KH, King AI. Biomechanics of neurotrauma. *Neurol Res*. 2001;23(2-3):144-156.
4. Guskiewicz KM, Weaver NL, Padua DA, Garrett WE. Epidemiology of concussion in collegiate and high school football players. *Am J Sports Med*. 2000;28(5):643-650.
5. Guskiewicz KM, Marshall SW, Bailes J, et al. Association between recurrent concussion and late-life cognitive impairment in retired professional football players: *Neurosurgery*. October 2005:719-726.
6. Stamm JM, Bourlas AP, Baugh CM, et al. Age of first exposure to football and later-life cognitive impairment in former NFL players. *Neurology*. 2015;84(11):1114-1120.
7. Broglio SP, Martini D, Kasper L, Eckner JT, Kutcher JS. Estimation of head impact exposure in high school football: Implications for regulating contact practices. *Am J Sports Med*. 2013;41(12):2877-2884.
8. Broglio SP, Eckner J, Martini D, Sosnoff JJ, Kutcher JS, Randolph C. Cumulative head impact burden in high school football. *J Neurotrauma*. 2011;28(10):2069-2078.
9. Cobb BR, Urban JE, Davenport EM, et al. Head impact Exposure in youth football: elementary

school ages 9–12 years and the effect of practice structure. *Ann Biomed Eng*. 2013;41(12):2463-2473.

10. Crisco JJ, Wilcox BJ, Beckwith JG, et al. Head impact exposure in collegiate football players. *J Biomech*. 2011;44(15):2673-2678.
11. Daniel RW, Rowson S, Duma SM. Head impact exposure in youth football: Middle school ages 12 to 14 years. *J Biomech Eng*. 2014;136(9):094501.
12. Beckwith JG, Greenwald RM, Chu JJ. Measuring head kinematics in football: correlation between the Head Impact Telemetry System and Hybrid III headform. *Ann Biomed Eng*. 2012;40(1):237-248.
13. Jadischke R, Viano DC, Dau N, King AI, McCarthy J. On the accuracy of the Head Impact Telemetry (HIT) System used in football helmets. *J Biomech*. 2013;46(13):2310-2315.
14. Allison MA, Kang YS, Bolte JH, Maltese MR, Arbogast KB. Validation of a helmet-based system to measure head impact biomechanics in ice hockey: *Med Sci Sports Exerc*. 2014;46(1):115-123.
15. Siegmund GP, Guskiewicz KM, Marshall SW, DeMarco AL, Bonin SJ. Laboratory validation of two wearable sensor systems for measuring head impact severity in football players. *Ann Biomed Eng*. 2016;44(4):1257-1274.
16. Camarillo DB, Shull PB, Mattson J, Shultz R, Garza D. An instrumented mouthguard for measuring linear and angular head impact kinematics in American football. *Ann Biomed Eng*. 2013;41(9):1939-1949.
17. Hernandez F, Wu LC, Yip MC, et al. Six Degree-of-Freedom Measurements of Human Mild Traumatic Brain Injury. *Ann Biomed Eng*. December 2014.
18. Mertz HJ, Jarrett K, Moss S, Salloum M, Zhao Y. The Hybrid III 10-year-old dummy. *Stapp Car Crash J*. 2001;45:319-328.
19. Mihalik JP, Guskiewicz KM, Marshall SW, Blackburn JT, Cantu RC, Greenwald RM. Head impact biomechanics in youth hockey: comparisons across playing position, event types, and impact locations. *Ann Biomed Eng*. 2012;40(1):141-149.
20. Daniel RW, Rowson S, Duma SM. Head acceleration measurements in middle school football. *Biomed Sci Instrum*. 2014;50:291-296.
21. Broglio SP, Surma T, Ashton-Miller JA. High school and collegiate football athlete concussions: A biomechanical review. *Ann Biomed Eng*. 2012;40(1):37-46.
22. Padgaonkar AJ, Krieger KW, King AI. Measurement of angular acceleration of a rigid body using linear accelerometers. *J Appl Mech*. 1975;42(3):552.
23. Society of Automotive Engineers. J211-1 Instrumentation for impact test - Part 1 - Electronic instrumentation, in surface vehicle recommended practice. 1995.
24. X2Biosystems. Detection of head accelerations with an instrumented mouthguard and virtual projection sensors. 2011.
25. King D, Hume PA, Brughelli M, Gissane C. Instrumented mouthguard acceleration analyses for head impacts in amateur rugby union players over a season of matches. *Am J Sports Med*. 2015;43(3):614-624.
26. Wu LC, Nangia V, Bui K, et al. In vivo evaluation of wearable head impact sensors. *Ann Biomed Eng*. 2016;44(4):1234-1245.
27. Nevins D, Smith L, Kensrud J. Laboratory evaluation of wireless head impact sensor. *Procedia Eng*. 2015;112:175-179.
28. Takhounts EG, Craig MJ, Moorhouse K, McFadden J, Hasija V. Development of brain injury criteria (BrIC). *Stapp Car Crash J*. 2013;57:243-266.
29. Manoogian S, McNeely D, Duma S, Brolinson G, Greenwald R. Head acceleration is less than 10 percent of helmet acceleration in football impacts. *Biomed Sci Instrum*. 2006;42:383-388.
30. Greenhill DA, Navo P, Zhao H, Torg J, Comstock RD, Boden BP. Inadequate helmet fit increases concussion severity in american high school football players. *Sports Health Multidiscip Approach*. 2016;8(3):238-243.
31. Lincoln AE, Caswell SV, Almquist JL, Dunn RE, Hinton RY. Video incident analysis of concussions in boys' high school lacrosse. *Am J Sports Med*. 2013;41(4):756-761.