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# Accelerometric Analysis of Head Impacts in Amateur Wrestling: An Exploratory Analysis

Mark Hecimovich,<sup>1</sup> Doug King,<sup>2,3</sup> and Troy Garrett<sup>1</sup>

**ABSTRACT.** NCAA men's wrestlers are at risk for concussion. Current research measures linear and rotational accelerations of the head through impact sensors. The frequency, magnitude, and distribution of head impacts >10 g on three Division I NCAA male wrestlers were recorded. Measures consisted of impact duration, linear, rotational head acceleration, and risk-weighted cumulative exposure. Participants averaged  $41 \pm 4$  impacts with a resultant median peak linear and rotation acceleration of 15 g and 1,880 rad/s<sup>2</sup>, resulting in a median HIT<sub>SP</sub> and the RWE<sub>CP</sub> of 15 and 0.0004. Results indicate lower levels compared to football and rugby.

**Keywords:** head kinematics, wrestling, concussion

## INTRODUCTION

Wrestling is not classified as a collision sport but, because of its combative nature, athletes are at risk for a variety of injuries, including concussion. Due to the sport's arduous nature, reported match injury rates at the collegiate level as high as 30.7 injuries-per-1000 athlete exposures (AEs) place it second only to injury rates among college football players (Jarret, Orwin, & Dick, 1998). Among the injuries, concussion ranks fourth (Yard, Collins, Dick, & Comstock, 2008).

In the NCAA 2009–2010 to 2013–2014 academic years in men's wrestling, 86 sport-related concussions were reported with the majority occurring during competition (53.5%) (Zuckerman et al., 2015). In that period, wrestling had an overall concussion rate of 10.92 per 10,000 AEs, with the competition rate being higher than the practice rate (relative risk [RR] = 9.76; 95% CI, 6.39–14.92) (Zuckerman et al., 2015). Of the 86 sport-related

concussions, 8.1% were recurrent with a majority occurring from participant contact (62.8%), followed by surface contact (22.1%). Although wrestlers represent 1.6% of all NCAA student athletes, they account for 5.8% of the total nationally estimated sport-related concussions (Zuckerman et al., 2015).

Information relating to the cause, course, and sequelae of concussion has increased (McCrorry et al., 2013). As a result of this information, knowledge surrounding the recognition and management of concussion has also improved (Fortington, Twomey, & Finch, 2015; Hecimovich, King, & Marais, 2016). This is important as there is a concern for the immediate and long-term effects on athletes involved in sports such as wrestling who are subjected to repeated impacts to the head resulting in sub-concussive impacts (Baugh et al., 2012; Gavett, Stern, & McKee, 2011) and how these may adversely affect cerebral functions (Baugh et al., 2012; Gavett et al., 2011; Gysland et al., 2012). As a result, research into head impacts (King, Hume, Brughelli, & Gissane, 2015) and prevention strategies (Fortington et al., 2015) has increased over the years, leading to greater insight into the likely causes and the profound effects of these injuries.

One such area of current research has sought to better determine the head linear and rotational accelerations involved in concussion injuries through the use of telemetry or impact sensors (Bartsch, Samorezov, Benszel, Miele, & Brett, 2014; Hernandez et al., 2015; King et al., 2015). These impact sensors (gyroscopes/accelerometers) have

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been utilized in helmeted, and nonhelmeted, sports to understand the link between the biomechanics of head impact and clinical outcomes of concussion in athletes (Bartsch et al., 2014; Brainard et al., 2012; Crisco et al., 2010; Hernandez et al., 2015; King et al., 2015; Mihalik, Bell, Marshall, & Guskiewicz, 2007; Rowson & Duma, 2009; Wilcox et al., 2015). Of the nonhelmeted sports, soccer (Hanlon & Bir, 2012), rugby union (King et al., 2015), and recently Australian Rules football (King, Hecimovich, Clark, & Gissane, *in press*) have started to accumulate impact-sensor data. The use of this emerging technology to gain real-time data (Broglia et al., 2010; Crisco et al., 2010; Guskiewicz & Mihalik, 2011; Guskiewicz et al., 2007) on sports collisions along with the mandatory use of helmets in American football has allowed for the systematic analysis of injury biomechanics. This new technology has led to a more extensive knowledge of the forces, velocities, accelerations, and frequencies of head injuries that can be applied to football or any other circumstances where repetitive head injury can occur. However, this technology has not been incorporated into collegiate-wrestling concussion research and management.

The aim this exploratory study was to investigate the frequency, magnitude, and distribution of head impacts greater than 10 g with the use of wireless head-impact sensors over a single tournament in a small sample of adult collegiate wrestlers. This is the first study to embark on this emerging area of research within the sport of collegiate wrestling and the generation of these data potentially to serve as a control measure for future research comparing concussed and nonconcussed participants. The data-acquisition threshold was based on a review of data-acquisition thresholds utilized in previously published studies (King et al., 2015).

## METHODS

A prospective observational cohort study was conducted on three Division I collegiate NCAA male wrestlers (mean  $\pm$  SD age, height, and body mass of  $21 \pm 2$  years,  $173.5 \pm 12.5$  cm, and  $73 \pm 19$  kg, respectively) over three matches during a 2015 tournament contest.

Consent was obtained from the participants prior to enrolment in the study. The researchers' university Institutional Review Board approved all procedures in the study (16-0089). Approval was provided by the team's coach and head certified athletic trainer prior to commencing the study.

Study participants wore the XPatch impact-sensing skin patch (X2Biosystems Ltd., Seattle, Washington, USA) on the skin covering their mastoid process (right side) during each match. The headgear was positioned on each participant to avoid compression over the sensor. The XPatch

sensor sampling at 1,024 Hz was placed behind the player's right ear just before they participated in match activities and was removed immediately after the tournament was completed. The positioning of the XPatch over the mastoid process is important to ensure that the sensor was not activated by enhanced soft-tissue effects when impacts occur (Wu et al., 2016).

The XPatch contained a low-power, high-g triaxial accelerometer with 200 g maximum per axis and a triaxial angular rate gyroscope to capture six degrees of freedom for linear and rotational time history accelerations of the head's center of gravity for all impacts that occurred during match participation. The time history incorporated three axes (x, y, z) of acceleration and three axes of velocity. Standing in an upright position, these planes describe the medial-lateral, anterior-posterior, and vertical acceleration and deceleration.

Following the tournament, the XPatches were removed from the wrestlers and downloaded to the Injury Management Software (IMS; X2Biosystems). The IMS enabled the raw accelerometer data to be transformed to the head's center of gravity by using a rigid-body transformation for linear acceleration and a five-point stencil for rotational acceleration (King et al., 2015; Wu et al., 2016). The biomechanical measures of head-impact severity consisted of impact duration (ms) and linear (g) and rotational head acceleration ( $\text{rad/s}^2$ ). Resultant linear acceleration is the rate of change in velocity of the estimated center of gravity of the head attributable to an impact and the associated direction of motion of the head (Mihalik et al., 2010). Resultant rotational acceleration is the rate of change in rotational velocity of the head attributable to an impact, and its direction in a coordinate system with the origin at the estimated center of gravity of the head (Mihalik et al., 2010). False impacts were removed by the X2Biosystems proprietary "de-clacking" algorithm (King et al., 2015) by comparing the waveform of each impact to a "Gaussian-like" reference waveform using cross-correlation (King et al., 2015). Impacts with a resultant linear acceleration of  $< 10$  g were removed. The remaining impacts were downloaded to an MS Excel spreadsheet and were time filtered to include only those impacts that occurred during match participation.

Head-impact exposure including frequency, magnitude, and location of impacts was quantified using previously established methods (Crisco et al., 2010, 2011). Data collection was limited to a single tournament only — not additional competitions or practice sessions. Two measures of impact frequency were computed for each participant: *impacts per match* or the total and average number of impacts per match for all matches; *participant group impacts* or the total and average number of recorded head impacts for the three participant groups

for all matches. Video capture was not conducted on the matches; therefore, verification of the impacts in conjunction with video evidence was not possible.

An exposure measurement was established based on the Weibull probability density function (pdf). This has been previously utilized to fit helmeted-impact-exposure data (Rowson & Duma, 2011). The Weibull pdf is demonstrated in Equation 1 where  $\alpha$  is the shape parameter,  $\beta$  is the scale parameter,  $\theta$  is the location parameter, and  $x$  is the peak resultant head acceleration (Rowson & Duma, 2011):

$$W_{pdf} = \frac{\beta(x - \theta)^{\beta-1}}{\alpha^\beta} e^{-\left(\frac{x-\theta}{\alpha}\right)^\beta}. \quad (1)$$

The Weibull parameters  $\alpha$ ,  $\beta$ , and  $\theta$  were calculated from the Weibull distribution fit for each participant's resultant linear and rotational acceleration and were integrated over the respective acceleration to calculate the Weibull cumulative density function (CDF; see Equation 2) (see Figure 1).

$$W_{cdf} = 1 - e^{-\left(\frac{x-\theta}{\alpha}\right)^\beta}. \quad (2)$$

The impact-location variables were computed as azimuth and elevation angles relative to the center of gravity (CG) of the head centered on the midsagittal plane (Crisco, Chu, & Greenwald, 2004). These were categorized as front (Left:  $\theta = 180^\circ$  to  $-135^\circ$ ; Right:  $\theta = 180^\circ$  to  $135^\circ$ ), side (Left:  $\theta = -135^\circ$  to  $-45^\circ$ ; Right:  $\theta = 135^\circ$  to  $45^\circ$ ), back (Left:  $\theta = -45^\circ$  to  $0^\circ$ ; Right:  $\theta = 45^\circ$  to  $0^\circ$ ), and top (Left:  $\theta = 180^\circ$  through negative  $\theta$  to  $0^\circ$ ; Right:  $\theta = 180^\circ$  through positive  $\theta$  to  $0^\circ$ ). Impacts to the top of the head were defined as all impacts above an  $\alpha$  of  $65^\circ$  from a horizontal plane through the CG of the head (Greenwald, Gwin, Chu, & Crisco, 2008).

Head impacts were assessed for injury-tolerance level for a concussion occurring using previously published injury-tolerance levels (Broglia et al., 2010, 2011; Guskiewicz & Mihalik, 2011) for linear ( $> 95$  g) and

rotational acceleration ( $> 5,500$  rad/s<sup>2</sup>). Head impacts were assessed for impact severity using previously published levels for linear acceleration (mild  $< 66$  g; moderate 66–106 g; severe  $> 106$  g) and rotational acceleration (mild  $< 4,600$  rad/s<sup>2</sup>; moderate 4,600–7,900 rad/s<sup>2</sup>; severe  $> 7,900$  rad/s<sup>2</sup>) (Harpham, Mihalik, Littleton, Frank, & Guskiewicz, 2013; Ocwieja et al., 2012; Zhang, Yang, & King, 2004).

Two additional risk equations were included in the analysis of the impact data to identify players at risk of a concussion. The Head Impact Telemetry Severity profile (HIT<sub>SP</sub>; Greenwald et al., 2008) is a weighted composite score including linear and rotational accelerations, impact duration, as well as impact location. The Risk Weighted Exposure Combined Probability (RWE<sub>CP</sub>; Urban et al., 2013) is a logistic regression equation and regression coefficient of injury-risk prediction of an injury occurring based on previously published analytical risk functions. The RWE<sub>CP</sub> combines the resultant linear and rotational accelerations to elucidate individual participant and team-based exposure to head impacts. As a value of 63 is a 75% indicator for a concussive injury (Broglia et al., 2011; Greenwald et al., 2008), the HIT<sub>SP</sub> values were evaluated by limits of less than 25% risk ( $< 21$ ), 25–75% risk (21–63), and  $> 75\%$  risk ( $> 63$ ). The RWE<sub>CP</sub> values were evaluated by the same values of 25% risk ( $< 0.2500$ ), 25–75% risk (0.2500–0.7500), and  $> 75\%$  risk ( $> 0.7500$ ).

All filtered data on the Microsoft Excel spreadsheet were analyzed with SPSS V.22.0.0. To test for normality, one-sample Kolmogorov-Smirnov and one sample  $t$  tests were conducted. The impact variables were not normally distributed,  $D_{(122)} = 4.86$ ;  $p < .0001$ ,  $t_{(121)} = 26.31$ ,  $p < .0001$ . Therefore, data were expressed as median (25th to 75th Interquartile Range [IQR]), and as severity measures (95th percentile linear acceleration, 95th percentile rotational acceleration) (Hopkins, Marshall, Batterham, & Hanin, 2009; Hopkins, Marshall, Quarrie, & Hume, 2007). Median peak linear and rotational accelerations and impact locations between player positions were

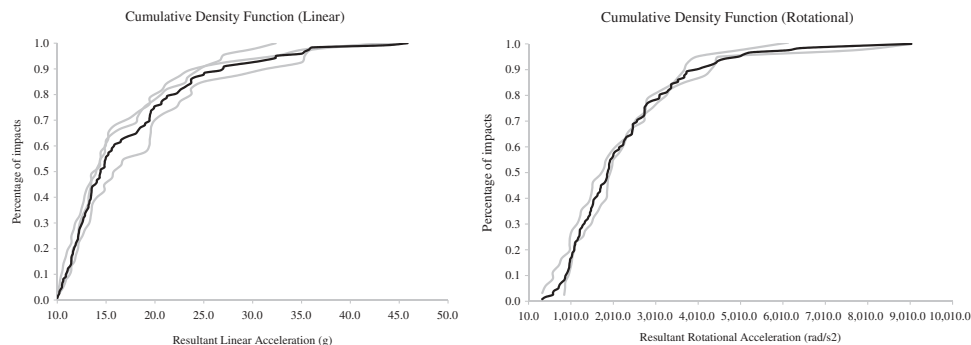


FIGURE 1 Weibull cumulative density function (CDF) of linear (left) and rotational (right) accelerations. Each participant CDF is represented in grey and the total CDF is represented in black.

assessed using a Friedman repeated-measures analysis of variance (ANOVA) on ranks with a Wilcoxon signed-rank test for post hoc analysis. Impact locations were analyzed by front, back, side, and top impacts using a Friedman repeated-measures ANOVA on ranks. A one-sample chi-square ( $\chi^2$ ) test was used to determine whether the observed head-impact frequency was significantly different to the expected head-impact frequency. The  $HIT_{SP}$  and  $RWE_{CP}$  were analyzed using a Friedman repeated-measures ANOVA on ranks. A post hoc analysis with Wilcoxon signed-rank tests was conducted with a Bonferroni correction applied. A one-sample chi-squared ( $\chi^2$ ) test and risk ratio (RR) with 95% confidence intervals (CI) were used to determine whether the observed impact frequency was significantly different from the expected impact frequency.

## RESULTS

Over the duration of the tournament, three individual matches were completed. A total of 122 impacts were recorded resulting in an average of  $41 \pm 4$  impacts per player to the head over the duration of the tournament. There was a total of three matches per participant totaling nine tournament matches completed over the duration of the study. None of the participants sustained a concussion.

The linear accelerations recorded over the tournament ranged from 10.0 to 45.8 g. These data were right skewed with a median (IQR) value of 14.5g (range = 12.2–20.1g) and a 95th percentile value of 34.6 g (see Figure 1). The rotational accelerations recorded over the tournament ranged from 333.3 to 9,037.0 rad/s<sup>2</sup>. These data were right skewed with a median (IQR) value of 1,880.2 rad/s<sup>2</sup> (range = 1,200.8–2,757.6 rad/s<sup>2</sup>) and a 95th percentile value of 5,066.5 rad/s<sup>2</sup>. The  $HIT_{SP}$  recorded over the duration of the tournament varied from 6.3 to 81.3 and had a median (IQR) value of 14.5 (range = 11.0–22.0) and a 95th percentile of 52.7 (see Table 1). The  $RWE_{CP}$  recorded over the duration of the tournament varied from 0.0001 to 0.3124 and had a median (IQR) value of 0.0004 (range = 0.0002–0.0010) and a 95th percentile value of 0.2625. There were no observable differences identified over the tournament for the resultant linear acceleration,  $\chi^2(2) = 0.38$ ,  $p = .8276$ , rotational acceleration,  $\chi^2(2) = 0.38$ ,  $p = .8276$ ,  $HIT_{SP}$ ,  $\chi^2(2) = 0.84$ ,  $p = .6585$ , and  $RWE_{CP}$ ,  $\chi^2(2) = 1.06$ ,  $p = .5892$ , of all the participants.

As shown in Table 2, there were more head impacts to the front of the head ( $n = 51$ ) than the back,  $\chi^2 = 14.6$ ,  $p < .0001$ , and top,  $\chi^2 = 24.1$ ,  $p < .0001$ , of the head. There were observable differences identified that the back of the head recorded higher median peak linear accelerations (PLAs) than the side (18.9 g vs. 13.7 g) of the head,  $\chi^2(1) = 6.37$ ,  $p = .0116$ , and on the post hoc analysis ( $z = -2.70$ ;  $p = .0070$ ). There were observable differences identified that

the right side of the top of the head recorded higher median peak rotational accelerations (PRAs) than the right side of the side (2,072.8 rad/s<sup>2</sup> vs. 1,529.1 rad/s<sup>2</sup>) of the head,  $\chi^2(1) = 4.50$ ,  $p = .0339$  and on the post hoc analysis ( $z = -2.38$ ;  $p = .0173$ ). As a result, the top right side of the head recorded a higher median  $RWE_{CP}$  (0.0004 vs. 0.0002) than the front right side of the head,  $\chi^2(1) = 6.0$ ,  $p = .0143$ , and on the post hoc analysis ( $z = -2.21$ ;  $p = .0273$ ).

There were four impacts (3.3%) to the head recorded above the rotational tolerance threshold ( $> 5,500$  rad/s<sup>2</sup>) with a median peak resultant rotational acceleration of 7,039.9 rad/s<sup>2</sup> (range = 6,205.6–8,931.3 rad/s<sup>2</sup>) (see Table 3). As a result, there were four impacts (3.3%) to the head above the  $HIT_{SP}$  severe ( $> 63$ ) threshold with a mean score of 78.0 (range = 73.2–81.1). The majority of impacts (99.2%) to the head had a median  $RWE_{CP}$  value (0.0003 [range = 0.0002–0.0010]) in the mild category.

## DISCUSSION

This study reports, for the first time, the head-impact biomechanics experienced while participating in collegiate wrestling. The aim of this exploratory study was to quantify the frequency, magnitude, and distribution of head impacts in a small sample of wrestlers over the course of a tournament. Due to the exploratory nature of this study, small sample size, and the absence of a diagnosed concussion, the results need to be viewed with caution, as it is unclear of the generalizability of the data. However, because the participants'  $RWE_{CP}$  values were low, concussions would not be expected. The data garnered may provide for future research as a control measure comparing nonconcussed to concussed participants.

From these data, the frequency and severity of impacts to the head experienced by participants during competition can be characterized. Additionally, the inclusion of a risk-weighted cumulative exposure (RWE) measure (Urban et al., 2013) was incorporated. By adjusting the impacts' contribution to cumulative exposure according to its associated impact tolerance, the RWE for linear, rotational, and a combined (linear and rotational) probability measure can be established (Urban et al., 2013). In order to accurately predict the risk of concussion, both linear and rotational accelerations should be accounted for to determine the concussion risk (Urban et al., 2013). The RWE combined probability ( $RWE_{CP}$ ) enabled this concussion-risk prediction to be undertaken. By recording the  $RWE_{CP}$  of individual participants (King et al., 2015), the resulting values may assist with the identification of participants with a potential cumulative exposure resulting in concussion. Only one other study (Urban et al., 2013) has reported the  $RWE_{CP}$  and this was in high-school-level American football. Utilizing the same logistic regression equations and

TABLE 1 Impacts to the Head Greater than 10 g for Individual Participants and for Total Impacts Recorded in an Amateur Wrestling Tournament

Total impacts n	Impact duration (ms) Mean ± SD	PLA (g)			PRA (rad/s <sup>2</sup> )			HIT <sub>SP</sub>			RWE <sub>CP</sub>		
		Mean ± SD	Median (IQR)	95%	Mean ± SD	Median (IQR)	95%	Mean ± SD	Median (IQR)	95%	Mean ± SD	Median (IQR)	95%
1	5.7 ± 5.4	16.7 ± 7.8	14.3 (11.6–19.5)	36.8	2,082.3 ± 1,331.5	1,792.7 (1,012.1–2,802.9)	5,188.1	20.0 ± 15.2	15.2 (11.2–21.2)	57.2	0.0016 ± 0.0039	0.0003 (0.0002–0.0010)	0.0012
2	7.5 ± 5.8	18.9 ± 8.2	16.2 (12.6–22.5)	36	2,463.3 ± 1,724.0	1,959.2 (1,302.7–3,036.4)	7,495.9	20.0 ± 15.3	15.2 (10.6–20.7)	66.3	0.0111 ± 0.0511	0.0004 (0.0002–0.0011)	0.0910
3	7.5 ± 5.5	16.3 ± 5.7	14.0 (12.1–19.5)	29.2	2,161.8 ± 1,290.9	1,861.3 (1,211.5–2,707.2)	5,175.3	20.0 ± 14.8	14.0 (10.8–24.1)	59.3	0.0015 ± 0.0043	0.0003 (0.0002–0.0010)	0.0079
Total	7.0 ± 5.6	17.3 ± 7.2	14.5 (12.2–20.1)	34.6	2,236.6 ± 1,455.3	1,800.2 (1,200.8–2,757.6)	5,066.5	20.0 ± 15.1	14.5 (11.0–22.0)	52.7	0.0047 ± 0.0296	0.0004 (0.0002–0.0010)	0.0085

Note: Data are presented as mean and standard deviation ( $\pm$ SD), median (interquartile range) and 95th percentile for individual participants and total impacts, impacts per player position group, impact duration (ms), resultant linear and rotational acceleration, head-impact telemetry severity profile, and risk-weighted exposure combined probability; IQR = Interquartile (25th to 75th) percentile; 95% = 95th percentile; PLA (g) = peak linear acceleration in gravitational force (g); PRA (rad/s<sup>2</sup>) = peak rotational acceleration in radians/second<sup>2</sup>; HIT<sub>SP</sub> = head-impact telemetry severity profile; RWE<sub>CP</sub> = risk-weighted exposure combined probability.

TABLE 2 Impacts to the Head Greater Than 10 g by Impact Location for Total Impacts for Three Participants in an Amateur Wrestling Tournament

Location	Side	Number of impacts n	Impact Duration (ms) Mean ± SD	Total Impacts recorded			
				PLA (g) Median (IQR)	PRA (rad/s <sup>2</sup> ) Median (IQR)	HIT <sub>SP</sub> Median (IQR)	RWE <sub>CP</sub> Median (IQR)
Front	Left	22	7.8 ±6.5	17.6 (11.6–23.7)	1,913.7 (1,003.3–3,115.5) <sup>c</sup>	13.4 (9.1–24.6)	0.0004 (0.0002–0.0015) <sup>c</sup>
	Right	29	4.5 ±2.9	12.6 (11.0–16.6)	1,529.1 (1,007.2–1,991.9) <sup>d</sup>	15.1 (10.0–23.2)	0.0002 (0.0002–0.0004) <sup>d</sup>
	<b>Total</b>	<b>51<sup>cd</sup></b>	<b>5.9 ±5.0</b>	<b>14.5 (11.4–19.5)</b>	<b>1,698.8 (1,027.2–2,329.8)</b>	<b>15.3 (10.9–23.0)</b>	<b>0.0003 (0.0002–0.0009)</b>
Side	Left	5 <sup>f</sup>	6.4 ±5.6	14.3 (12.1–19.1)	2,746.0 (1,478.3–3,191.0)	19.2 (7.2–33.2)	0.0007 (0.0003–0.0013) <sup>f</sup>
	Right	35 <sup>e</sup>	7.9 ±5.7	13.5 (12.2–19.6) <sup>c</sup>	1,486.3 (1,161.8–2,746.6) <sup>d</sup>	15.2 (12.1–22.5)	0.0003 (0.0002–0.0009) <sup>de</sup>
	<b>Total</b>	<b>40<sup>cd</sup></b>	<b>7.7 ±5.6</b>	<b>13.7 (12.3–19.5)<sup>c</sup></b>	<b>1,721.6 (1,166.2–2,774.0)</b>	<b>15.3 (12.1–22.5)</b>	<b>0.0003 (0.0002–0.0009)</b>
Back	Left	4 <sup>f</sup>	5.3 ±1.3	17.7 (14.7–31.6)	1,669.9 (1,009.8–3,136.6) <sup>a</sup>	13.8 (9.8–36.4)	0.0003 (0.0002–0.0027) <sup>a</sup>
	Right	15 <sup>e</sup>	10.1 ±7.7	21.2 (13.2–30.1) <sup>b</sup>	2,578.7 (1,694.0–3,747.9)	15.2 (8.1–34.9)	0.0007(0.0003–0.0023)
	<b>Total</b>	<b>19<sup>ab</sup></b>	<b>9.1 ±7.1</b>	<b>18.9 (13.2–30.1)<sup>b</sup></b>	<b>2,434.4 (1,630.8–3,536.2)</b>	<b>11.5 (8.8–36.3)</b>	<b>0.0006 (0.0003–0.0023)</b>
Top	Left	4	5.3 ±3.4	14.5(11.2–19.9)	2,093.2 (1,261.2–3,983.6)	17.4 (7.4–35.2)	0.0005 (0.0002–0.0028)
	Right	8	5.8 ±3.8	15.4 (13.6–22.6)	2,072.8 (1,906.8–3,373.6) <sup>ab</sup>	13.4 (11.3–24.4)	0.0004 (0.0004–0.0019) <sup>ab</sup>
	<b>Total</b>	<b>12<sup>ab</sup></b>	<b>5.6 ±3.5</b>	<b>15.4 (13.3–20.8)</b>	<b>2,072.8 (1,882.1–3,373.6)</b>	<b>14.2 (11.7–22.9)</b>	<b>0.0004 (0.0003–0.0020)</b>

Note: Data are presented as mean (±SD), median (IQR) for left and right side and total impacts, impact duration (ms), resultant linear and rotational acceleration, head-impact telemetry severity profile, and risk-weighted exposure combined probability; (IQR) = Interquartile (25th to 75th) percentile; 95% = 95th percentile; PLA (g) = peak linear acceleration in gravitational force (g); PRA (rad/s<sup>2</sup>) = peak rotational acceleration in radians/second<sup>2</sup>; HIT<sub>SP</sub> = head-impact telemetry severity profile; RWE<sub>CP</sub> = Risk-Weighted Exposure Combined Probability; Significant difference ( $p < .05$ ) then (a) front, (b) = side, (c) = back, (d) = top; (e) = left; (f) = right.



TABLE 3 Impacts to the Head Greater than 10 g by Injury Tolerance, Injury Severity for Resultant Linear and Rotational Accelerations, Head-Impact Telemetry Severity Profile, and Risk-Weighted Exposure Combined Probability for Total Impacts an Amateur Wrestling Tournament

		Total Head Impacts		
		<i>n</i> (%)	Mean ( $\pm$ SD)	Median (IQR)
<b>Injury Tolerance</b>				
Linear	> 95g	—	—	—
Rotational	> 5,500 rad/s <sup>2</sup>	4 (3.3)	7,312.3 $\pm$ 1,325.4	7,039.9 (6,205.6–8,931.3)
<b>Injury Severity (Linear)</b>				
Mild	<66g	122 (100)	17.3 $\pm$ 7.2	14.5 (12.2–20.1)
Moderate	66–106g	—	—	—
Severe	>106g	—	—	—
<b>Injury Severity (Rotational)</b>				
Mild	<4,600 rad/s <sup>2</sup>	114 (93.4)	1,961.8 $\pm$ 981.1	1,837.2 (1,149.7–2,601.1)
Moderate	4,600–7,900 rad/s <sup>2</sup>	6 (4.9)	5,420.9 $\pm$ 700.1	5,173.0 (4,891.1–6,205.6)
Severe	>7,900 rad/s <sup>2</sup>	2 (1.6)	8,345.6 $\pm$ 977.7	* —
<b>Head-Impact Telemetry severity profile (HIT<sub>SP</sub>)</b>				
Mild	<21	88 (72.1)	12.8 $\pm$ 3.5	12.3 (9.8–15.0)
Moderate	21–63	30 (24.6)	33.4 $\pm$ 10.9	30.9 (24.1–44.7)
Severe	>63	4 (3.3)	77.4 $\pm$ 4.1	78.0 (73.2–81.1)
<b>Risk-Weighted Exposure combined probability (RWE<sub>CP</sub>)</b>				
Mild	<0.2500	121 (99.2)	0.0022 $\pm$ 0.0092	0.0003 (0.0002–0.0010)
Moderate	0.2500–0.7500	1 (0.8)	0.3124	* —
Severe	> 0.7500	—	—	—

Note. Data are presented as number of impacts and percentage of impacts recorded; rad/s<sup>2</sup> = radians/second<sup>2</sup>; HIT<sub>SP</sub> = head-impact telemetry severity profile; RWE<sub>CP</sub> = risk-weighted exposure combined probability.

\* = unable to calculate.



regression coefficients, it can be seen that the median  $RWE_{CP}$  in high-school-level American football players was 0.497 (Urban et al., 2013) and this was higher than the current study (0.0003) recorded on wrestlers.

In this study, an average of 41 impacts per wrestler to the head occurred over the duration of the tournament. Comparisons with other sports is difficult due to vast differences in the structure of competition and the frequency of collisions associated with a specific sport is a function of the opportunity for collision to occur within the context of the sport (Powell, 2001). In the case of concussion, the likelihood of injury is a function of the number of times a participant's head sustains an impact within the context of participation. These impacts may be incidental (unintentional) and occur as a result of the nature of the sport, or they may be impacts that result from intentional acts (e.g., fighting; Powell, 2001).

In the current study, the mean, median, and 95th percentile values of the resultant linear ( $17\text{ g} \pm 7\text{ g}$ , and  $34\text{ g}$ ) and rotational accelerations ( $2,236 \pm 1,455\text{ rad/s}^2$ ,  $1,800\text{ rad/s}^2$ , and  $5066\text{ rad/s}^2$ ) were established. When compared with collegiate American football (Crisco et al., 2011), the 95th percentile resultant linear acceleration ( $62\text{ g}$ ) and the resultant rotational acceleration ( $4,378\text{ rad/s}^2$ ) were higher than those measured in the wrestlers. When compared with senior amateur rugby union (King et al., 2015), the mean resultant linear ( $22\text{ g}$ ) and rotational ( $3,903\text{ rad/s}^2$ ) accelerations were higher than those reported for the wrestlers. A possible reason for the lower linear and rotational accelerations may be due to the proximity, or closeness, of the participants. For example, in wrestling, the takedown has been reported to be the most common move leading to injury (Boden, Lin, Young, & Mueller, 2002; Hoffman & Powell, 1990; Jarret et al., 1998). The takedown consists of a rapid, high-energy struggle that ends with a wrestler being thrown to the mat in relatively close proximity (Boden et al., 2002; Hoffman & Powell, 1990; Jarret et al., 1998). Although the takedown itself is relatively brief, because all matches begin with both wrestlers standing in a neutral position and attempting a takedown and because a relatively high number of points are awarded for a successful takedown, high school and college wrestlers are encouraged to attempt multiple takedowns throughout a match (Yard et al., 2008) and thus the importance of long-term research.

The location of impacts recorded in the current study varied considerably with more head impacts to the front of the head than to the back and top, which may be due to head-up, front-facing position of the wrestlers. However, the back of the head recorded higher median PLAs than the side, which may be the result of takedown maneuvers. This may have also contributed to the higher median  $RWE_{CP}$  to the top right side of the head than the front right side. Additionally, all the participants were right-hand dominant and thus placing their head toward the right side of their

opponent in a front-aligned position, therefore creating a dominate right-sided effect.

It is known that, among a wide variety of sports, the potential for concussion is related to the number of opportunities within the sport for activities that produce collisions. For example, in football, the number of collisions involving the head is very high with some players experiencing a head impact on every play. In other sports such as ice hockey, impacts with the head are expected but not inherent in the design of the sport (Powell, 2001). Although, most concussions in wrestling are mild or Grade I (NCAA News, 1998), the sport reports the highest overall concussion rate of all NCAA sports (10.92 per 10,000 AEs). For wrestling, the most common head-injury mechanism–activity combination is wrestler contact during takedown (Zuckerman et al., 2015) that results in head-to-head or head-to-knee collisions. Concussions are also produced by contact with the wrestling mat or the floor surrounding the wrestling mat area. The fact that slamming an opponent to the mat is illegal makes serious head injury less likely via this mechanism. Future research needs to examine the head impacts with a focus on takedowns.

There are several of limitation in this study. For example, the sample size is small and, thus, the results may not represent totals across a whole season and the participants were all in a similar weight class that cannot be transferable to all wrestlers. Data collection occurred over a limited period, which impedes generalization of the results. None of the participants sustained a concussion, and, therefore, no conclusions should be made on the significance of the impact levels. The X2Biosystems XPatch accelerometer has been reported in previous studies (King, Hume, Gissane, & Clark, 2016; McCuen et al., 2015; Reynolds et al., 2016a, 2016b; Swartz et al., 2015; Wu et al., 2015) of which two studies (McCuen et al., 2015; Wu et al., 2015) tested the biomechanical validity in different settings. By comparing in vivo performance of the XPatch via video capture in a simulated low-impact soccer setting, Wu et al. studied 25 impacts, one impact location, one mastoid placement location, and one XPatch accelerometer on a single subject. It was reported (Wu et al., 2015) that the XPatch overestimated individual linear and rotational accelerations that was likely to be related to the subject's viscoelastic properties of the soft tissues. In another study, McCuen et al. evaluated the XPatch on a Hybrid III headform as a prelude to live-play soccer. There were 250 impacts over five impact locations with two mastoid placement locations and five different XPatch accelerometers. It was reported (McCuen et al., 2015) that there was significant XPatch measurement root mean square error related to individual impacts for PLA and PRA of ~50%. The study also looked at aggregate performance over a larger number of impacts and reported that the “average values over a large number of acceleration events can be determined with good

## CONFLICTS OF INTEREST

Mark Hecimovich, Doug King, and Troy Garrett declare that they have no conflict of interest.

## REFERENCES

- accuracy*" (McCuen et al., 2015, p. 3722). In another study utilizing the XPatch in collegiate football, Reynolds et al. (2016a) reported that the number and linear severity of head impacts were favorable comparable to published data reporting (Broglio et al., 2009; Crisco et al., 2011; Duma et al., 2005; Mihalik et al., 2007; Rowson, Brolinson, Goforth, Dietter, & Duma, 2009; Schnebel, Gwin, Anderson, & Gatlin, 2007) on helmet-based accelerometers. However, there were discrepancies reported (Reynolds et al., 2016b) between the rotational severity of head impacts measured by the XPatch and similar published data from helmeted systems (Reynolds et al., 2016b).
- The results in this study provide a baseline; a generation of further new knowledge may result in comparisons being drawn between weight classes, level of skill, competition (Division class), specific maneuvers (i.e., takedowns) to identify injury risk, and incidence within the sport.

## CONCLUSION

Wrestling is an aggressive contact sport and will never be free from potential injury situations. However, by examining how injuries occur, we can gain insight into their prevention (Hewitt). For the first time, the head-impact biomechanics experienced with participation in collegiate wrestling was measured. Using instrumented mastoid-based impact sensors (accelerometer) over the course of a tournament of matches, 41 impacts (> 10 g) per wrestler to the head occurred. Although the majority of the impacts recorded were under the linear and rotational injury risk limits, the effects of the frequency and location of these impacts remain unknown. The key to this study was obtaining and thus gaining initial measurements on the frequency, magnitude, distribution, and risk-weighted exposure of head impacts in collegiate wrestling in order to assist in the identification of at-risk players that will better inform medical personnel of the need to evaluate a player for concussion.

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## COMPLIANCE WITH ETHICAL STANDARDS

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