Using In-Mouth Sensors to Measure Head Kinematics in Rugby

Emily E. Kieffer, Chase Vaillancourt, P. Gunnar Brolinson, and Steven Rowson

Abstract Sports-related concussions were once thought to present only transient symptoms, but recent research has shown the potential for long-term neurological impairments. Previous research has instrumented helmeted athletes with sensors to measure head kinematics and concussive biomechanics on-field. However, expanding studies beyond helmeted players will help us to understand sex-specific differences and unhelmeted loading conditions and their relationship to impact exposure and tolerance. Athletes on the men's and women's club collegiate rugby teams at Virginia Tech were custom fit with instrumented mouthguards and monitored over a year to quantify sub-concussive and concussive biomechanics in unhelmeted athletes. The objective of this study was to quantify head kinematics sustained from inertially driven (body impacts) and direct head impacts in collegiate rugby players. The majority of the impacts collected were low in magnitude, and head impacts were associated with higher PLAs than body impacts ($\Delta = 2.5$ g, $p < 0.005$). Head impacts had a median PLA of 15.2 g and a median ΔRV of 6.7 rad/s; body impacts had a median PLA of 13.6 g and a median ΔRV of 6.6 rad/s. For each impact type, men and women sustained similar peak linear accelerations (head impact: $\Delta = 1.9$ g, $p = 0.219$; body impact: $\Delta = 0.5$ g, $p = 0.280$). Men sustained higher changes in rotational velocities for head impacts ($\Delta = 1.4$ rad/s, $p = 0.003$) and longer impact durations for both head ($\Delta = 2.6$ ms, $p < 0.001$) and body ($\Delta = 2.8$ ms, $p = 0.023$) impacts. These findings can be used to advise best practices for in-mouth sensors and better understand the mechanisms of concussion in unhelmeted sports.

Keywords Acceleration, Biomechanics, Concussion, Female, Impact

I. INTRODUCTION

Not only do concussions present acute cognitive impairment at the time of injury and long-term complications, but a history of repetitive head impact exposure may be similarly detrimental [1]. Instrumenting American football players with sensors to measure head impact kinematics has improved our understanding of head impact exposure and biomechanics [2-6]. However, we still do not understand sex-related differences in exposure and head impact tolerance. There is also much unknown about exposure in other sports, especially in those with unhelmeted athletes, who likely undergo different loading conditions than helmeted athletes. Concussive biomechanics in unhelmeted sports have not been quantified because instrumenting these athletes is challenging. Different types of sensors have been used on-field, including those worn in headbands [7-9], on skin-mounted patches [10-13], in the ear [14], and embedded into mouthguards [15-16]. Of these, mouthguards offer the best solution as they allow for a nearly rigid coupling to the skull, which improves head kinematics measurement accuracy [17].

Similar to American football, rugby players have a high exposure to head impacts [18]. Worldwide, rugby is a popular full-contact team sport and its physicality and competitiveness lead to a high concussion rate [19]. The only protective equipment worn by rugby players is a mouthguard, which does not reduce the risk of concussion [20]. About 13–17% of rugby players will sustain a concussion during a season, though this varies by the play style. Both men's and women's teams play at various levels of competition, under the same rules [20].

Athletes can play Rugby-7s (7s), or Rugby-15s (15s). Rugby-7s is played with seven players on the field (per team) and is played for seven-minute halves. Rugby-15s is played with 15 players on the field (per team) and is played for 40-minute halves. A previous study from Ireland found the incidence and severity of concussion to be higher in Rugby-7s [19]. The same study found that the leading causes of concussion were tackling (7s) and collisions (15s) [19]. According to video review, concussion risk for a tackler is elevated in a collision tackle, when the initial contact is made by the tackler's head/neck, when the tackler does not use their arms, or when the

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tackle is broken by the ball carrier [21]. King *et al.* quantified head impacts in junior league rugby and amateur rugby union (Rugby-15s), measuring resultant head accelerations similar to youth football and some college football, but lower than those in female youth soccer and professional American football [18][22]. The junior league players were instrumented with skin-mounted sensors worn behind the ear, and the union players wore moulded instrumented mouthguards [18][22].

Most of these studies did not include female athletes (the junior league had five girls who were under 11), nor did they characterise head kinematics by impact type. Additionally, King noted limitations with his devices, including a lack of consistent reliability studies with the skin-mounted sensor systems and saliva ruining components in the instrumented mouthguards [18][22]. Instrumentation approaches have improved since, with new technologies coming to market. The objective of this study was to quantify head kinematics and impact durations sustained from both inertially driven and direct impact head acceleration events experienced by men and women collegiate rugby players. For this study, collegiate rugby players wore custom-fit instrumented mouthguards. We hypothesised that head impacts would have higher peak kinematics and shorter impact durations compared to body impacts, and that kinematics would not differ between sexes.

II. METHODS

During the spring and fall of 2019, athletes were recruited from the Virginia Tech men's and women's club rugby teams to participate in this study, which was approved by the Virginia Tech Institutional Review Board. A total of 26 males (age: 20.6 ± 1.1 years, height: 1.79 ± 0.06 m, weight: 88.8 ± 16.3 kg) and 21 females (age: 20.7 ± 1.1 years, height: 1.66 ± 0.08 m, weight: 74.6 ± 18.5 kg) provided written consent and participated in the study. Each athlete was instrumented with a custom-fit mouthguard that contained accelerometers and angular rate sensors to measure head kinematics with six degrees of freedom. The athletes wore their mouthguards for each game and contact practice. Two different mouthguards were used, Prevent Biometrics Impact Monitoring Mouthguard (IMM) (Fig. 1) and Wake Forest retainer, both of which have been validated in laboratory testing [15-16]. The Prevent Biometrics mouthguard had an R^2 = 0.99 for linear acceleration and R^2 = 0.99 for angular acceleration in lab [15]. The Wake Forest retainer had an R^2 = 0.95 for linear acceleration and R^2 > 0.99 for angular velocity in lab, and overall sensitivity of 69.2% and a positive predictive value of 80.3% in the field [16]. A compliant material was added to the Wake Forest retainer to fit around the athletes' teeth to modify the device into a mouthguard (Fig. 2).

Prevent's device used a sampling rate of 3,200 Hz. When a buffering accelerometer measurement exceeded 5 g on a single axis, 10 ms of pre-trigger and 40 ms of post-trigger data were collected and stored on local memory. The Prevent iOS app was able to download the data in real-time. A low-pass filter of channel frequency class (CFC) 240 was used to filter the data. To avoid possible data loss, Prevent's algorithm to remove suspected false-positive impact events was deactivated. A generalised transformation formula was used to transform accelerations to the head's centre of gravity (CG). Wake Forest's device sampled the accelerometers at 4,684 Hz and the gyroscope at 1,565 Hz. When a buffering accelerometer measurement exceeded 5 g on a single axis, 15 ms of pre-trigger and 45 ms of post-trigger data were collected and stored on local memory. Wake Forest's device used a low-pass filter of CFC 2000 to filter the acceleration data and CFC 270 to filter angular rate data. Subject-specific transformations were calculated from the device to the head CG [23]. The head CG was estimated based on the origin of the Frankfort plane, using the infraorbital margin (IOM) and external auditory meatus (EAM) as facial landmarks, as previously described by Rich *et al* [23]. These anatomical landmarks were identified on images taken of each athlete's face.

Fig. 1. The Prevent Biometrics IMM mouthguard with instrumentation along the teeth.

Fig. 2. The Wake Forest mouthguard with instrumentation in the upper palate.

Any recorded event with a resultant acceleration below 10 g was excluded from the analysis. A 10 g threshold was used to discriminate impact events from dynamic movement events associated with activities, such as running or jumping [25]. Time-synced video from one high definition video camera was used to verify that recorded acceleration events were associated with impacts. Each verified acceleration event was categorised as either a "head impact", defined as a head acceleration resulting from a direct impact to the head (Fig. 3), or a "body impact", defined as a head acceleration resulting from an impact to the body (Fig. 4).

Fig. 3. A direct head impact during a men's practice.

Fig. 4. An inertially induced head acceleration from a body impact during a men's game.

Although peak rotational velocity was reported for each device, we reported the change in rotational velocity (ΔRV), which is the difference in maximum rotational velocity from the rotational velocity at the onset of impact. This is because we are interested in the change in rotational velocity of the head due to impact. This required Prevent Biometrics' rotational velocity to be zeroed prior to calculation. Wake Forest already zeroes their rotational velocity data. Therefore, ΔRV was computed for each impact and reported as such. We defined the peak linear acceleration (PLA) as the maximum value for the resultant linear acceleration pulse. It is important to note this because Prevent defines their PLA as the maximum linear acceleration in the 7.5 ms to 20 ms time window of data collection, with the first 2.5 ms of those as pre-trigger data. Therefore, we recalculated PLA for the Prevent data.

Impact duration was computed manually for each impact using the following approach. The start of the impact was identified by a clear increase in resultant linear acceleration for the acceleration pulse, producing the max. PLA (Fig. 5). The end of the impact was determined to be a local minimum following the maximum acceleration that seemed to indicate that an external force was no longer acting on the head. The resultant acceleration trace was used primarily for these measurements, although the axis-specific data gave context on the timing of loading (i.e. when each axis crossed zero). If there was a clear separation between multiple peaks, only the width of the pulse with the maximum resultant acceleration was considered. After reviewing all of the traces, 47 (9.4%) of the events were identified as non-typical head acceleration events (e.g. there was no clear acceleration peak, or the full width of the pulse was not captured). These impacts were not considered during the impact duration analysis. These duration-excluded impacts were still included in the impact magnitude analysis because peak values could be quantified.

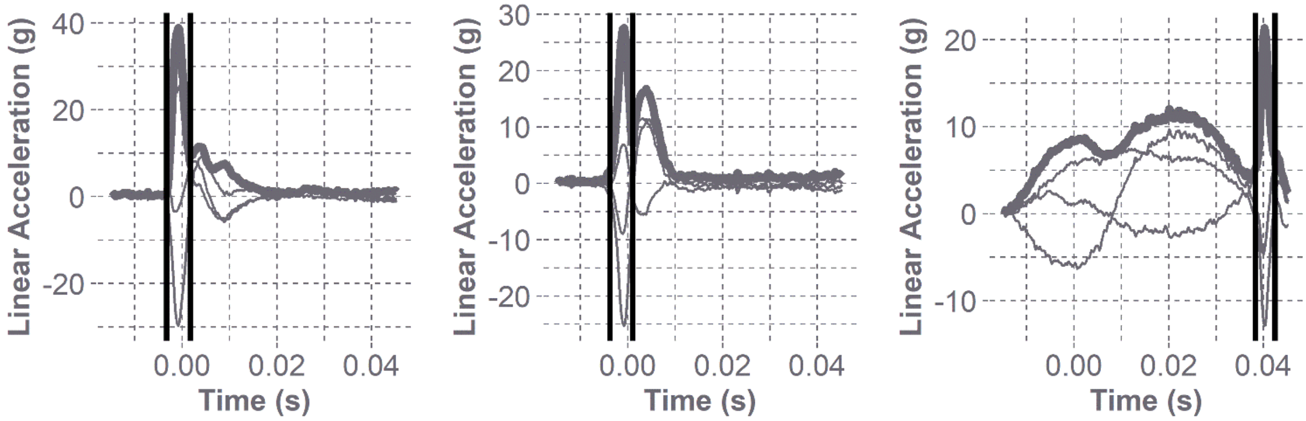

Fig. 5. Three examples of calculated impact duration. The darkest trace is the resultant acceleration and the lighter traces are the x, y and z axis-specific traces. The two lines mark the beginning and the end of the measured duration for each impact.

Impact exposure for body and head impacts was calculated per player, per session and averaged by sex. The distributions of impact exposure, PLA, ΔRV and impact duration were compared between sexes for head and body impacts. PLA and ΔRV values were log-transformed before performing a two-way analysis of variance (ANOVA) to test the effects of sex and impact type. Wilcoxon signed-rank tests were used to compare impact exposure, PLA, ΔRV and impact duration between and within impact type and sex. The mean difference and 95% confidence interval (CI) were computed to provide an estimate of effect size.

III. RESULTS

The women were instrumented for 12 games and three contact practices. The men were instrumented for eight games and 11 contact practices. In total, there were 736 video-validated acceleration events (Table I). Since data were collected from both sexes' 7s and 15s teams, a day of 7s tournament play was considered one game for this analysis. Overall, men and women sustained 0.9 [95% CI: 0.2 to 1.5] more head impacts than body impacts per session ($p = 0.012$). Men sustained 1.1 [CI: 0.5 to 1.7] more impacts per session than the women ($p = 0.002$) (Table I). Women sustained 0.5 [CI: -0.1 to 1.2] more head impacts than body impacts ($p = 0.097$). Men sustained 1.1 [CI: 0.2 to 2.1] more head impacts per session than body impacts ($p = 0.008$). Median values are presented to best capture the central tendency as the data are right-skewed.

TABLE I
IMPACT COUNTS, MEDIAN, AND 95TH PERCENTILE HEAD KINEMATICS FOR PEAK LINEAR ACCELERATION AND CHANGE IN ROTATIONAL VELOCITY EXPERIENCED BY MEN AND WOMEN FOR BOTH IMPACT TYPES.

				PLA (g)		ΔRV (rad/s)	
Sex	Impact	Count	Impacts/Session	Median	95%ile	Median	95%ile
Female	Body Impact	75	1.5 ± 0.8	13.9	41.3	6.5	20.8
Female	Head Impact	122	2.1 ± 1.2	15.2	50.4	5.5	16.5
Male	Body Impact	175	2.4 ± 1.7	13.4	44.3	6.7	19.8
Male	Head Impact	364	3.5 ± 1.6	15.2	48.4	7.2	17.9

The distributions of peak head kinematic values for both sexes were right-skewed for both body and head impacts. Non-parametric probability density functions were estimated from kernel density estimates (Figs. 6 and 7). For PLA values, 79% of events were less than 25 g, 96% were less than 50 g, and 98% were less than 75 g. For ΔRV values, 74% of events were less than 10 rad/s, 91% were less than 15 rad/s, and 96% were less than 20 rad/s. Head impacts accounted for 66% of the dataset, had a median PLA of 15.2 g [IQR: 11.7 to 24.2 g] and a median ΔRV of 6.7 rad/s [IQR: 4.4 to 10.0 rad/s]. Body impacts had a median PLA of 13.6 g [IQR: 11.0 to 19.7 g] and a median ΔRV of 6.6 rad/s [IQR: 4.0 to 10.3 rad/s]. Overall, head impacts were associated with 2.5 g [CI: 0.5 to 4.8 g] higher PLAs than body impacts ($p < 0.005$). The ΔRV of body impacts were 0.3 rad/s [CI: -0.6 to 1.2 rad/s] greater than that of head impacts ($p = 0.790$). Exemplar traces for head and body impacts can be seen in Figs. 8 and 9.

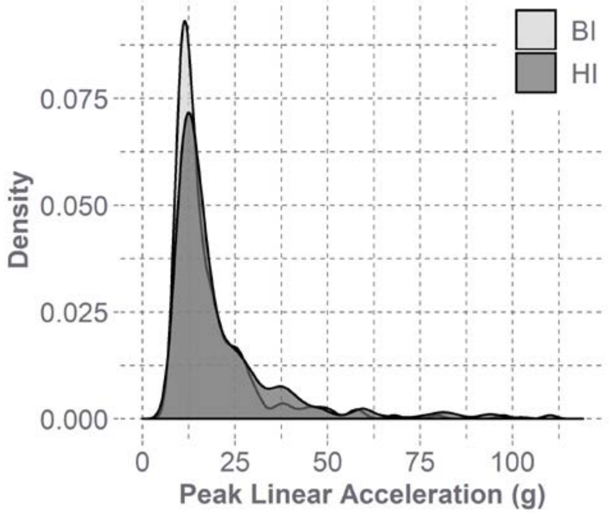

Fig. 6. Probability density function (PDF) for PLA, with HI in dark gray and BI in light gray. HIs tended to result in greater PLA ($p < 0.005$).

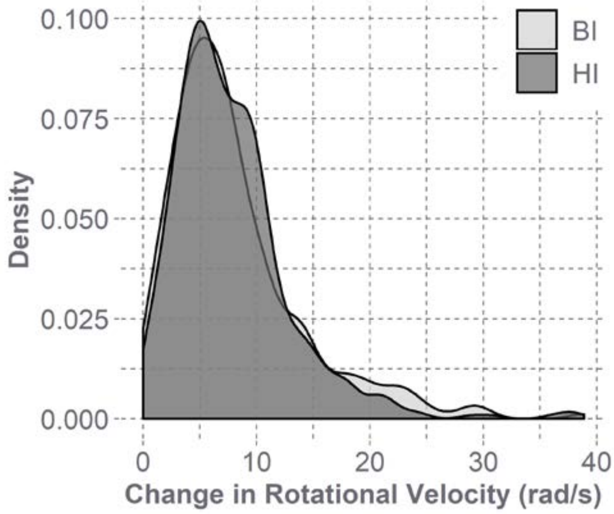

Fig. 7. PDF for ΔRV, with HI in dark gray and BI in light gray. There was little evidence suggesting that ΔRV values differed between the impact types ($p = 0.790$).

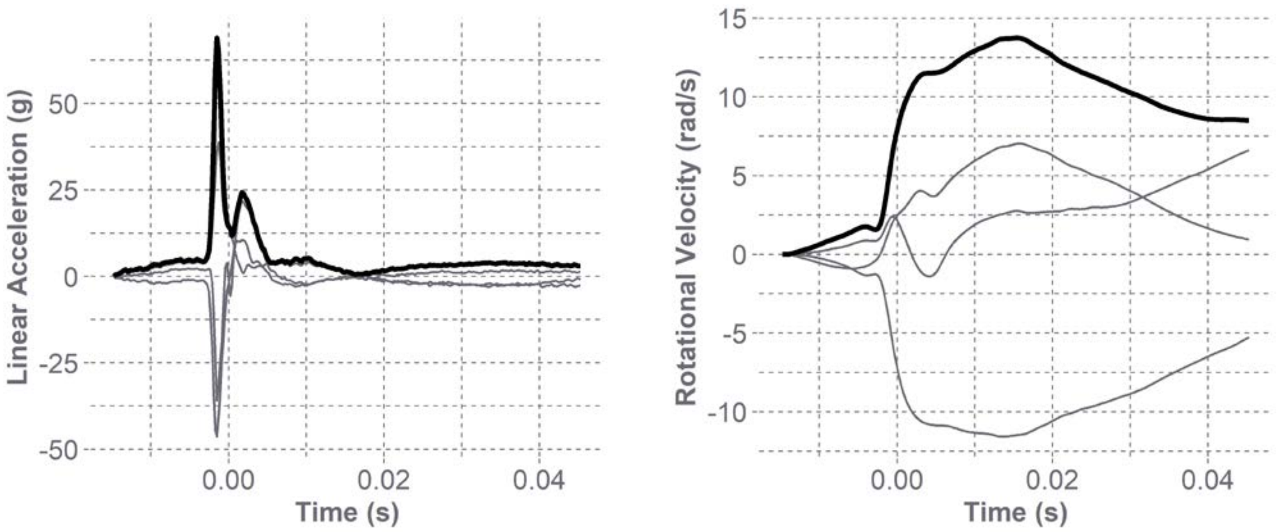

Fig. 8. Exemplar linear acceleration and rotational velocity traces for a head impact. Resultant trace is black and axis specific traces are gray.

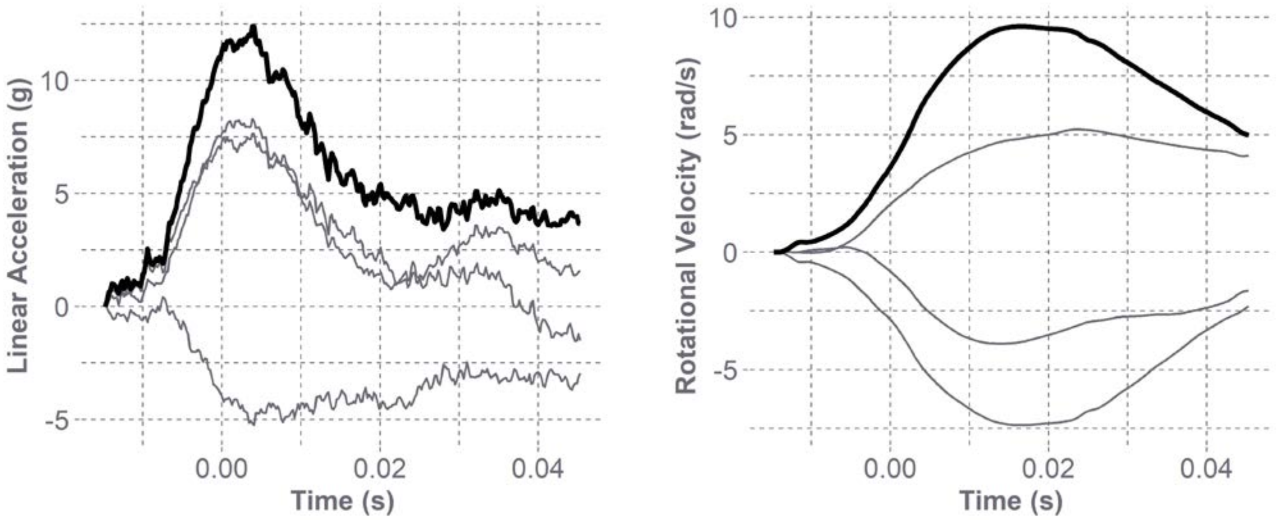

Fig. 9. Exemplar linear acceleration and rotational velocity traces for a body impact. Resultant trace is black and axis specific traces are gray.

The head impacts that the women sustained had a median PLA of 15.3 g [IQR: 12.1 to 27.8 g], which was 3.8 g [CI: -0.04 to 7.6 g] higher PLA than their body impacts (p = 0.052) (Fig. 10). For all box plots, outliers were excluded from the plot to better visualize differences in the central tendencies. The men's median head impact had a PLA of 15.2 g [IQR: 11.6 to 23.2 g], which was 2.4 g [CI: -0.3 to 5.0 g] greater than their body impacts (p = 0.004) (Fig. 10). There was no evidence to suggest that women's body impacts resulted in a meaningful difference in ΔRV compared to their head impacts, with an estimated effect size of 0.9 rad/s [CI: -0.7 to 2.5 rad/s] (p = 0.277). There was also no evidence to suggest that men's body impacts resulted in a meaningful difference in ΔRV compared to their head impacts, with an estimated effect size of 0.11 rad/s [CI: -1.0 to 1.2 rad/s] (p = 0.340) (Fig. 11).

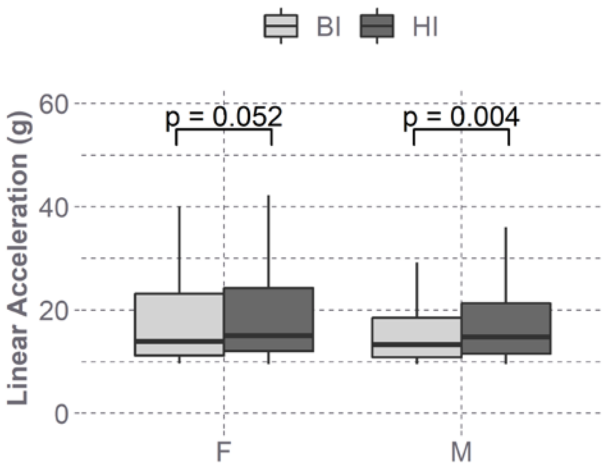

Fig. 10. Linear acceleration for impact type compared within sex. Head impacts had a higher PLA than body impacts for both men and women. Outliers are excluded from the plots.

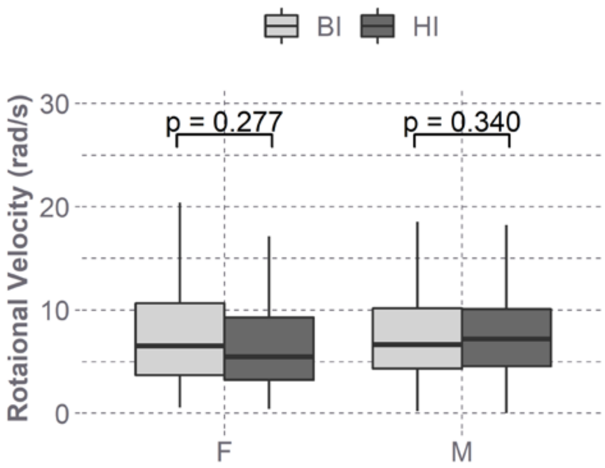

Fig. 11. Rotational velocity for impact type compared within sex. There was no difference in ΔRV between impact type within sex. Outliers are excluded from the plots.

Men and women sustained similar PLA values for head impacts, with an estimated effect size of 1.9 g [CI: -1.4 to 5.3 g] (p = 0.219) (Fig. 12). The sustained PLA values for body impacts were also similar between sexes, with an effect size of 0.5 g [CI: -2.7 to 3.7 g] (p = 0.280) (Fig. 12). Men's head impacts resulted in ΔRV values 1.4 rad/s [CI: 0.3 to 2.4 rad/s] higher than women's head impacts (p = 0.003) (Fig. 13). The difference in ΔRV between men's body impacts and to women's body impacts was 0.6 rad/s [CI: -1.0 to 2.2 rad/s] (p = 0.636) (Fig. 13).

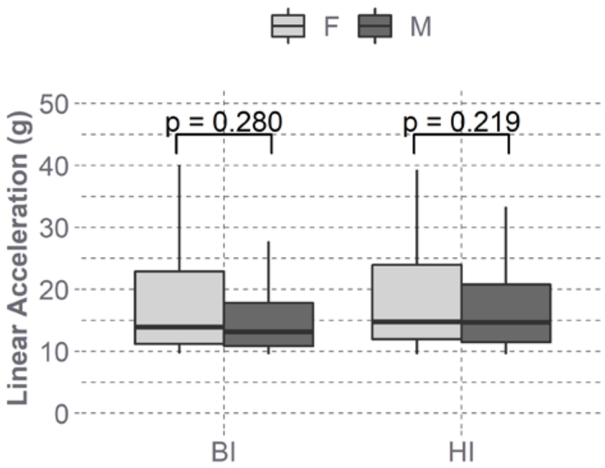

Fig. 12. Linear acceleration for impact type compared between sex. Men and women sustained similar PLA magnitudes for each impact type. Outliers are excluded from the plots.

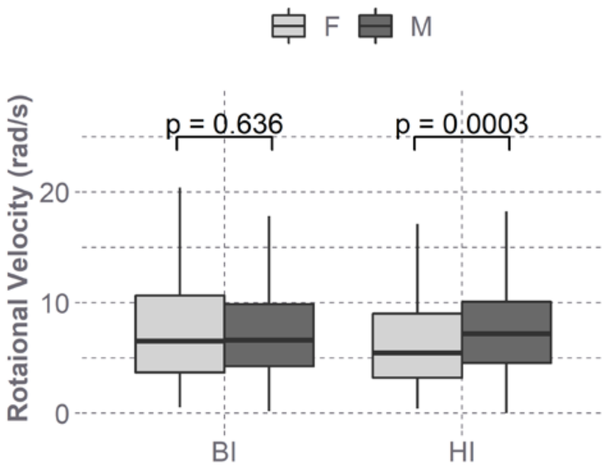

Fig. 13. Rotational velocity for impact type compared between sex. Men sustained higher ΔRV magnitudes for head impacts. Outliers are excluded from the plots.

Of the 47 events removed for the duration analysis (Fig. 14), 12 were head impacts and 35 were body impacts. This left 215 body impacts and 476 head impacts to characterise impact duration. Body impacts had a median duration of 10.9 ms [IQR: 6.5 to 18.8 ms], which was 1.1 ms [CI: -0.4 to 2.5 ms] higher than head impacts' durations ($p = 0.56$). Head impacts had a median duration of 10.4 ms [IQR: 6.7 to 16.6 ms].

Men's body impacts were similar in duration to their head impacts, only 1.3 ms [CI: -0.6 to 3.1 ms] longer ($p = 0.432$) (Fig. 15). Women's body impacts were also similar in duration, only 1.0 ms [CI: -1.3 to 3.3 ms] longer than their head impacts ($p = 0.976$) (Fig. 15).

Men experienced body impacts 2.8 ms [CI: 0.3 to 5.4 ms] longer than women did ($p = 0.023$), and they experienced head impacts 2.6 ms [CI: 1.2 to 4.0 ms] longer than women did ($p < 0.001$) (Fig. 16).

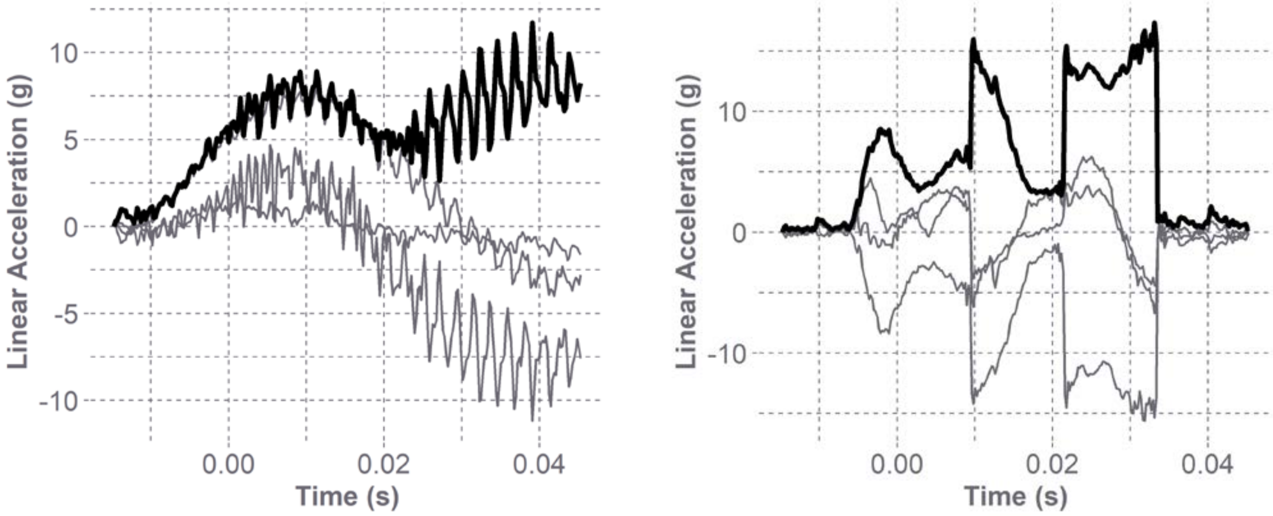

Fig. 14. Two exemplar linear acceleration traces for impacts removed from the duration analysis due to their lack of consistency to acceleration traces for an impact. Resultant trace is black and axis specific traces are gray.

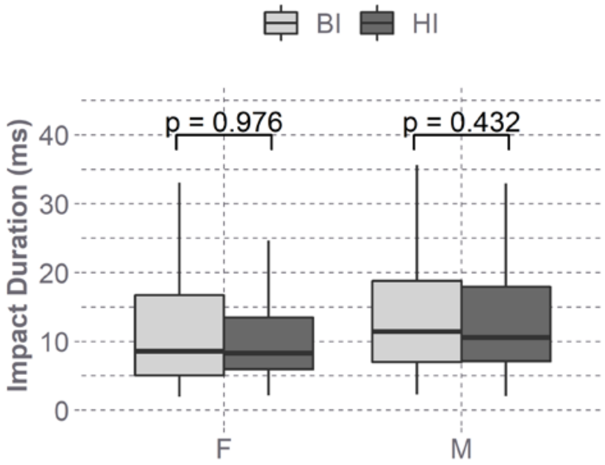

Fig. 15. Impact duration for impact type compared within sex. Body impacts were similar in duration to head impacts for both sexes. Outliers are excluded from the plots.

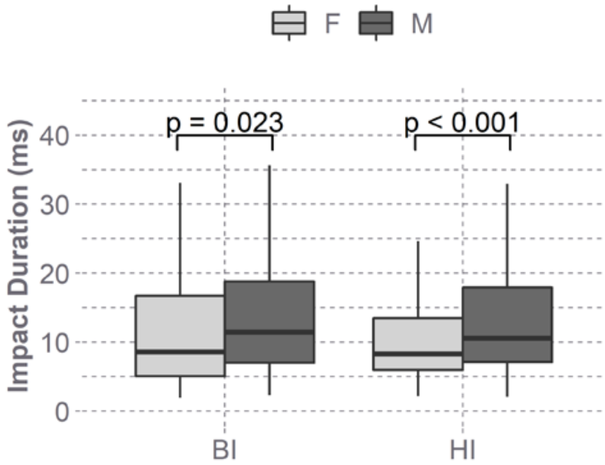

Fig. 16. Impact duration for impact type compared between sex. Men sustained longer impacts in each impact type. Outliers are excluded from the plots.

Throughout the study, three men and nine women sustained concussions. Of these, biomechanical data were collected for three impacts (two men's and one woman's). All occurred during games. One of the men was swung around by his opponent and thrown down, hitting the back of his head on the ground. His impact had a PLA of 81.6 g and ΔRV of 18.5 rad/s (Fig. 17). The second male player was making a tackle and ran his head into his opponent's chest, knocking himself to the ground. When he finally stood up, his balance was compromised and he took several unsteady steps. His impact had a PLA of 74.8 g and ΔRV of 36.2 rad/s (Fig. 18). The female player was holding onto her opponent's legs, trying to make a tackle, when she was kneed in the head by her teammate.

Her impact had a PLA of 54.6 g and ΔRV of 9.7 rad/s (Fig. 19). Data were not collected on the other concussions because they either resulted in suspected false negative events or there were device issues on the day of injury.

TABLE II

CONCUSSIVE EVENTS SUSTAINED OVER THE COURSE OF THE STUDY FOR BOTH TEAMS. PLA AND ΔRV ARE REPORTED FOR THE THREE IMPACTS THAT RESULTED IN MEASURED DATA. THE FIRST CONCUSSION'S ΔRV VALUE IS, NOTED BY *, IS LIKELY DUE TO THE INITIAL BODY IMPACT LEADING TO THE HEAD IMPACT, NOT THE HEAD IMPACT ITSELF. THE FOURTH CONCUSSION, NOTED BY $^{+}$, WAS TWO IMPACT EVENTS FOR WHICH ONLY THE INITIAL BODY TO GROUND KINEMATICS WERE CAPTURED. THESE VALUES ARE NOT CONSIDERED TO BE ASSOCIATED WITH THE CONCUSSIVE KICK TO THE HEAD, BUT ARE HIGHLIGHTED AS A LIMITATION OF THE EVENT TURNOVER TIME IN THIS DEVICE. THE OTHER MISSED CONCUSSIONS ARE DUE TO DEVICE FAILURE, OR A SUSPECTED FALSE NEGATIVE (THE DEVICE COLLECTED DATA FOR THE REST OF THE SESSION WITHOUT ISSUE).

Team	PLA (g)	ΔRV (rad/s)	Mouthguard	Notes
Men's Rugby	81.6	18.5*	Wake Forest	
Men's Rugby	74.8	36.2	Prevent Biometrics	
Women's Rugby	54.6	9.7	Wake Forest	
Women's Rugby	14.2^{+}	3.1^{+}	Wake Forest	Body to ground, then kicked in head
Women's Rugby			Prevent Biometrics	Device failure
Women's Rugby			Prevent Biometrics	Device failure
Women's Rugby			Prevent Biometrics	Suspected false negative
Women's Rugby			Prevent Biometrics	Suspected false negative
Men's Rugby			Prevent Biometrics	Suspected false negative
Women's Rugby			Wake Forest	Suspected false negative
Women's Rugby			Wake Forest	Device failure

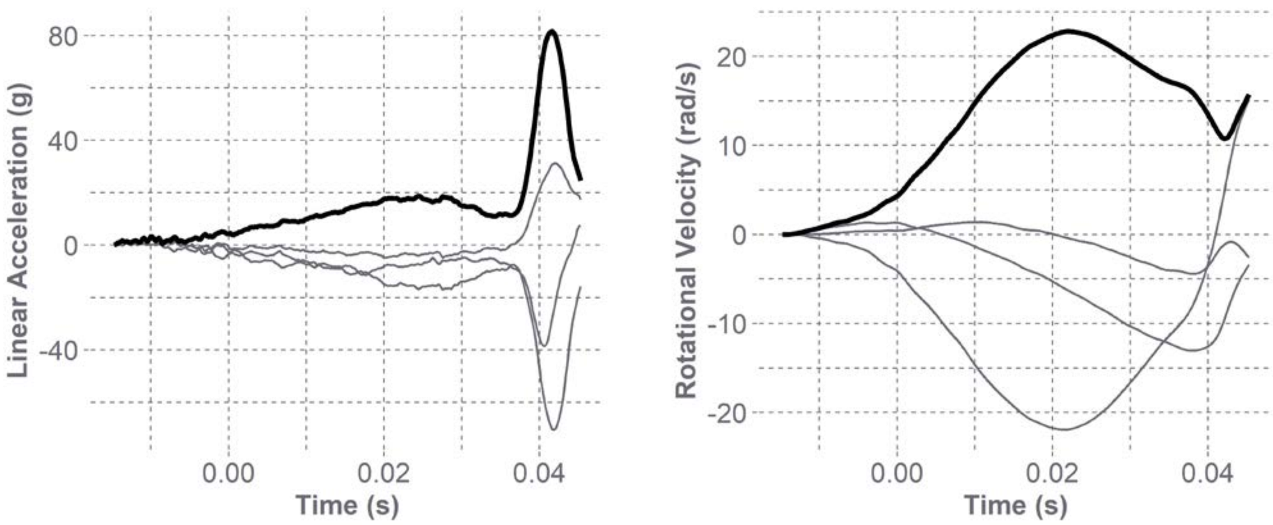

Fig. 17. Linear acceleration and rotational velocity traces for a concussion on the men's team. It is likely that the rotational velocity measured is associated with the body impact that happened before the head to ground impact, and the change in rotational velocity associated with the head impact was not captured. Resultant trace is black and axis specific traces are gray.

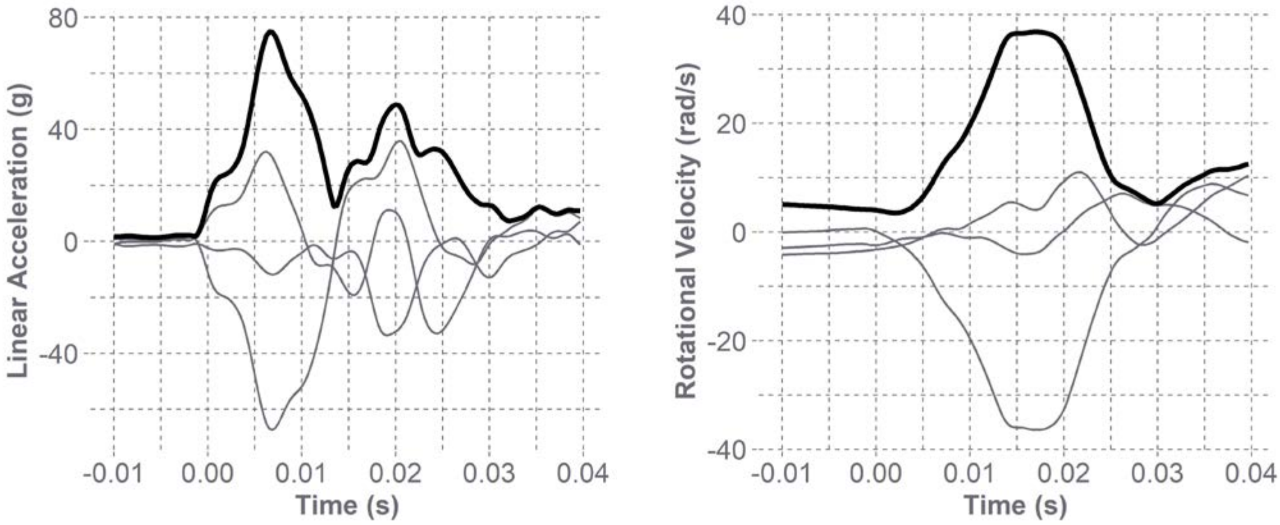

Fig. 18. Linear acceleration and rotational velocity traces for a concussion on the men's team. The two peaks in linear acceleration likely correspond to two, nearly instantaneous, impact events, potentially a head then shoulder contacting the opponent. This could explain the atypical shape of the rotational velocity curve, with the second impact stopping the head rotation that was initiated with the first impact. Resultant trace is black and axis specific traces are gray.

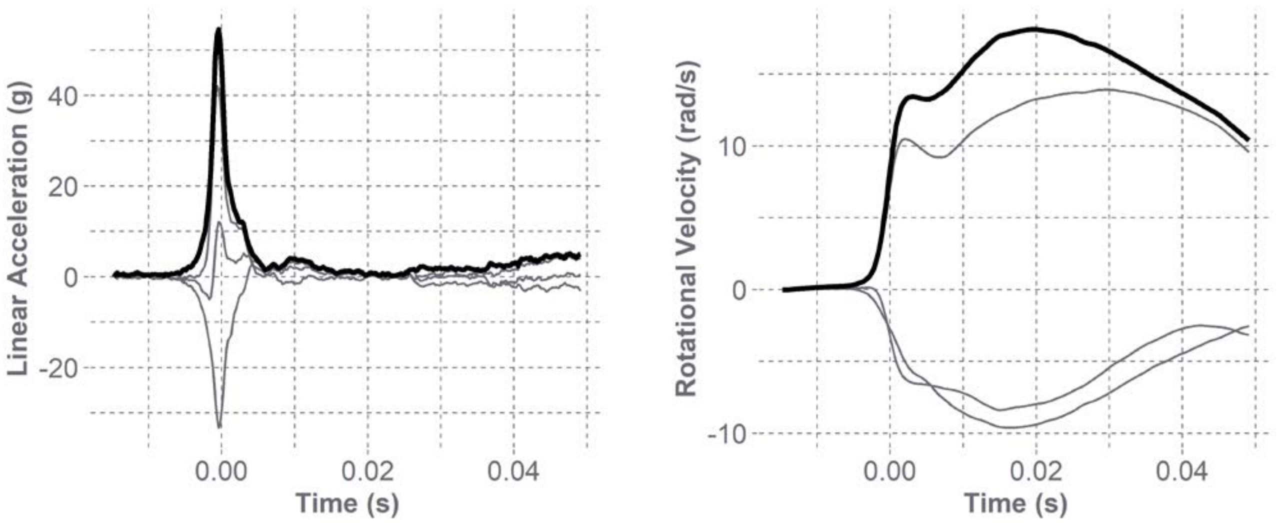

Fig. 19. Linear acceleration and rotational velocity traces for a concussion on the women's team. These traces similar shapes to those seen in lab impacts. Resultant trace is black and axis specific traces are gray.

IV. Discussion

This study aimed to quantify head kinematics associated with head and body impacts in collegiate rugby players. Most of the kinematic data collected were low in magnitude, which is consistent with other studies [18][22][25]. The head impacts in this study had an average PLA of 20.9 g and a median PLA of 15.2 g. King *et al.* reported PLA values with a mean of 22 g and median of 16 g for a junior league [18]. Similarly, his mean linear acceleration for union players was 22.2 ± 16.2 g [22]. Reconstruction of 13 non-injurious head impacts in unhelmeted Australian football players resulted in a mean PLA of 59.0 g (SD 37.2 g) and ΔRV of 18.4 rad/s (SD 7.6 rad/s) [26]. The median PLA for instrumented collegiate football players has been reported to be 17.5 g (average 22.3 g), with the majority of impacts less than 20 g [25]. Impact surface and location were not considered in this analysis and may affect the reported kinematics. The data used in this analysis were aggregated for men's and women's 7s and 15s teams. Differentiating them from each other may yield different results.

Although there was evidence suggesting that the men's team sustained more impacts per session than the women did, there was not strong evidence suggesting a difference in PLA between sexes. Although it is often suggested that men play contact sports more aggressively than women, it has been shown that women are as willing to subject themselves to similar physical risks in sports [27]. The men's ΔRV values were slightly higher than the women's ΔRV values and slightly higher than sub-concussive ΔRV values (5.5 rad/s) seen in football players [28]. We do not know if these values are clinically meaningful regarding acute or cumulative effects.

Direct head impacts were associated with slighter higher PLA and significantly higher ΔRVs than body-driven events. However, body impacts still resulted in notable head kinematic magnitudes. Accordingly, limiting exposure analyses to only confirmed head impact, and not considering all video-verified acceleration events, will underestimate the true exposure of athletes. Although contact phenomenon has shown importance in brain injury tolerance [29-31], the clinical symptoms that accompany an inertially induced acceleration event can be as severe as those associated with a direct impact, and their effect on cumulative exposure should not be overlooked [32]. Impact type is a confounding variable that should be accounted for given differences in loading duration. We saw durations 8% greater for body impacts than for head impacts. It is important to note that the change in angular velocity is occurring over a different time interval than the duration reported, which hampers its interpretation compared to peak linear acceleration.

Five acceleration events were removed from the initial 741-event dataset based on perceived video-observed impact severity and mouthguard-measured magnitude discrepancies. One impact was sustained from a men's player and resulted in a PLA recording of over 290 g. The video showed he was in a ruck, and an opponent's arm struck his head, but 290 g is not consistent with the impact severity observed. The video was of a sufficient quality to identify this discrepancy. A similar scenario occurred for a female player whose mouthguard measured a 135 g acceleration when tackling her opponent without making any head contact. This magnitude is not consistent with any of the body impacts we have seen. The third event removed was a 127 g measurement for a ball-to-face impact from one of the women's players. After the ball bounced off her face, she caught it and continued to play. Previous studies have measured soccer ball headers to be around 20 g [33], nearly a fifth of the magnitude measured. The fourth is a 187 g impact from the men's team who was lying on the ground in a ruck, and was hit with a forearm. He appeared unaffected by the impact. The fifth and final event removed had a magnitude of 158 g and occurred when a women tackled an opponent to the ground and braced on her head on the ground after the impact. She continued to play without any sign of significant injury, which would be expected with a PLA as high as this.

Good coupling to the skull is an advantage of using instrumented mouthguards to measure head impacts in athletes. However, this approach is reliant on athletes wearing the mouthguards properly. If a mouthguard is not set correctly on the teeth during an impact, the measured kinematics will be erroneous, potentially explaining the inflated kinematics in the three events removed from the dataset. We removed 47 impacts from the duration analysis given their atypical head acceleration traces. Poor coupling to the teeth is one explanation for being suspicious of these events' data quality.

The three concussive kinematic values in this study are similar to the concussion magnitudes found in two union rugby players which were 94.8g and 5319.8 rad/s^2 and 54.9g and 9935.2 rad/s^2 [22]. The concussion magnitudes measured in the current study are also within the range of observed in football players [29]. There is very little data on concussion in rugby and as we collect more, we expect the kinematics resulting in concussion in unhelmeted rugby players to be less than that of helmeted football players, partly due to different frequency content and contact phenomena.

There was another concussion on the women's team resulting from being kicked in the head after diving to the ground. Her mouthguard measured impact kinematics of 14.2 g and 6 rad/s. For this case, there were likely two acceleration events: the athlete's body impacting the ground; and the athlete's head being kicked. The Wake Forest devices require 300 ms between recording events. The kinematic values attributed to this concussion are likely the result of the ground impact. The kick occurred approximately 167 ms after the initial impact, which is during the time the mouthguard was preparing for the next impact. In comparison, the Prevent mouthguard only has a 25 ms delay between impacts. Understanding this turnover time between impacts is another critical component of properly verifying on-field kinematic data. Rugby players often undergo body-then-head acceleration events during tackles. This can complicate video validation if it is difficult to know if a body impact before a head impact is enough to trigger data acquisition and if events are outside the device turnover time. King *et al.* also noted complications with video validation in rugby, as athlete head position is difficult to distinguish in some tackles, rucks, mauls and scrums [22]. The double-impact scenario was something evident when inspecting the traces in our dataset. There were many events observed that had secondary peaks within the data-collection window, sometimes having higher values than the initial peak.

An obvious concern is the seven concussions for which data were not recorded. Four of these cases were suspected false negative events (three Prevent Biometrics, one Wake Forest) and three were device issues on the day of injury. Missing data from these concussions raises awareness for other potential false negatives that the mouthguards may be missing, thus underestimating impact count estimates in exposure studies. There were 17 instances of the Prevent mouthguard not functioning, and four instances of the Wake Forest mouthguard not functioning. Sometimes the issue was Bluetooth connectivity, either before or after the game, which complicated setting the devices for acquisition and downloading data. Wake Forest's devices also had intermittent gyroscope failures and some of the internal clocks did not sync, resulting in data that were erroneous or unusable. The battery life on the Wake Forest devices was shorter and there were issues with the charging cases not working, causing several mouthguards to die mid-session. As the seasons progressed, both companies made durability and usability improvements to reduce data loss.

Another limitation of this dataset is the inability to summarise the data on a per-player basis. This is because not every player in the study sustained enough impacts for an individualised analysis. In summarising the overall data, those athletes who experienced more impacts exert more weight in the analysis than their counterparts with fewer impacts. Continued data collection will allow for the characterisation of head kinematics on a per-player basis, which would be critically important for defining exposure in rugby.

Potential systematic differences between the mouthguards were explored in the dataset presented. Impacts measured by Wake Forest devices had a median PLA of 13.9 g [IQR: 11.2 to 21.1 g] and a median ΔRV of 5.6 rad/s [IQR: 3.5 to 9.0 rad/s]. Impacts measured by Prevent Biometrics devices had a median PLA of 15.2 g [IQR: 11.6 to 23.9 g] and a median ΔRV of 7.5 rad/s [IQR: 4.9 to 10.5 rad/s]. Attempting to stratify the dataset to reduce confounding factors (like sex, session type, and individual players' aggression) decreases the sample size and resolution for comparison. The slight differences seen between devices are more likely due to variance as a result of confounding factors than as a result of systematic differences between the sensors.

On-field use of the in-mouth sensors proved challenging and interpretation of data nuanced, especially with the missed concussions. Instrumented mouthguards offer great potential to collect on-field kinematic data for women and unhelmeted athletes, but much consideration needs to be taken when collecting and analysing this data. Carefully monitoring each device and its functionality is imperative before and after each session. Video validation is critical because each device recorded many false-positive events.

V. CONCLUSIONS

On-field head kinematic data were collected for male and female collegiate rugby players using instrumented mouthguards. More head impacts per session than body impacts were sustained, and men sustained more impacts per session than the women did. Head impacts had greater PLA values than body impacts. Women sustained head impacts similar in PLA, lower in ΔRV, and shorter in duration to their male counterparts. Analysing and characterising the axis-specific traces of each impact type provides more insight into different kinematic signatures of acceleration events. This study suggests that users of in-mouth sensors should carefully consider the technical specifications of sensors and further highlights that video verification of acceleration events is required for meaningful dissemination of data.

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