

# Sex differences in neck strength and head impact kinematics in university rugby union players

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## Abstract

Globally, over three million women participate in rugby union, yet injury prevention and training strategies are predominantly based on androcentric data. These strategies may have limited generalisability to females, given the cervical spine is more susceptible to whiplash and less adept at resisting inertial loading. A total of 53 university rugby union players (25 female, 28 male,  $20.7 \pm 1.8$  years) had their isometric neck strength measured. Bespoke instrumented mouthguards were used to record the magnitude of head impact events in six female and seven male competitive matches. Mean female maximal isometric neck strength was 47% lower than male. Independent samples Mann-Whitney U tests showed no significant differences for peak linear head acceleration (female: median 11.7 g, IQR 7.9 g; male: median 12.5 g, IQR 7.0 g  $p=0.23$ ) or peak rotational head acceleration (female: median 800.2  $\text{rad}\cdot\text{s}^{-2}$ , IQR 677.7  $\text{rad}\cdot\text{s}^{-2}$ ; male: median 849.4  $\text{rad}\cdot\text{s}^{-2}$ , IQR 479.8  $\text{rad}\cdot\text{s}^{-2}$ ;  $p=0.76$ ), despite the mean male body mass being 24% greater than female. Coded video analysis revealed substantial differences in head-impact mechanisms; uncontrolled whiplash dominated >50% of all recorded female impact events and <0.5% in males. Direct head-to-ground impacts comprised 26.1% of female and 9.7% of male impacts, with whiplash occurring in 78.0% and 0.5%, respectively. Overall, the data provided in this study do not support the generalisation of male-derived training and injury-prevention data to female rugby athletes. These results suggest a considerable research effort is required to identify specific weakness of female rugby players and derive appropriate training, injury prevention and return to play protocols.

**Key Words:** biomechanics, data, gender, injury & prevention, technology

## Highlights

- Video analysis revealed substantial differences in head-impact mechanisms, with uncontrolled whiplash dominating >50% of all recorded female impact events but rarely in males.
- Isometric neck strength was 47% lower in female players than males.
- Direct head-to-ground impacts accounted for 26.1% and 9.7% of female and male impacts, with whiplash occurring in 78.0% and 0.5%, respectively.

## Introduction

Exposure to repeated head impacts and resulting traumatic brain injury (TBI) has long been associated with acute and chronic neurocognitive challenges among athletes<sup>1</sup>. Rugby union, in particular, has been highlighted as a key cause for concern, with 9.6 million players from grassroots to professional level worldwide<sup>2</sup>. Specifically, concussion accounts for 20% of all match injuries in men's professional rugby, representing 37% of all injuries incurred by tacklers<sup>3</sup>, and 15% of all injuries at the community level<sup>3,4</sup>. Moreover, incidence of concussion diagnoses has risen for seven consecutive seasons, making it the most common injury sustained during professional men's matches<sup>3</sup>. While data for the women's game is more limited, despite accounting for 29.2% of players worldwide<sup>2</sup>, concussion accounted for 19% of all match injuries for the England Premier 15's in 2017/18 and 2018/19<sup>5</sup>. This is of growing concern given that female participation in rugby continues to rise, with a reported increase of 28% in 2018 alone<sup>2</sup>.

In rugby union, existing studies that assess causative mechanisms of head impacts are almost exclusively androcentric<sup>6</sup>. Findings from male athletes are routinely used to derive rugby's injury identification frameworks<sup>6</sup> and training strategies<sup>7</sup>. These are subsequently generalised to female players, despite a paucity of supporting data to verify the applicability of these findings to females. Indeed, research demonstrated that the physiological demands placed on elite female rugby union players in England differ to their male counterparts<sup>8</sup>. Moreover, McGroarty et al.<sup>9</sup> highlighted that female players have a lower biomechanical brain injury threshold tolerance, with more severe and prolonged somatic concussive symptoms, such as headaches and dizziness<sup>10</sup>. Furthermore, females have recorded poorer post-concussion cognitive test scores than their male counterparts<sup>11</sup>, with Ferretti et al.<sup>12</sup> highlighting biological sex differences in the neurology relating to symptomology, disease progression and treatment outcomes.

Of importance, sex-specific head stabilisation strategies have been shown in response to laboratory-based perturbation<sup>13</sup>. Such differences may be due to, at least in part, dimorphisms in spinal anatomy, reportedly linked to increased head-neck movement in vehicle collisions<sup>14</sup>, increasing female susceptibility to whiplash and concussive injuries<sup>15</sup>. Specifically, the male cervical spine is reported to exhibit greater intervertebral coupling stability than females, with greater vertebral body width and disc-facet depth<sup>15</sup>. This greater comparative stability makes the male cervical spine more adept at resisting inertial loading and extreme movements of the spine<sup>16</sup>. Cervical spine dimorphisms are therefore an important consideration when developing

injury prevention and identification frameworks, training methodologies, and return to play (RTP) protocols in rugby.

Recent recommendations from World Rugby (WR), the international governing body for rugby union, regarding the promotion of athlete health and safety, focus on concussion<sup>2</sup>. The development of evidence-based systems to detect, characterise, monitor and manage head-impact loading and timely removal from play, are therefore crucial<sup>17</sup>. Indeed, the identification of key injurious biomechanical inputs can be utilised to develop targeted, prevention-based training strategies and technologies<sup>18</sup>. Training interventions, such as increasing neck strength, have been found to reduce head-impact magnitude and the propensity for concussive injuries in men's rugby<sup>19</sup>. Longer-term head-impact monitoring may enable a distinction between the kinematics of head impacts deemed to be concussive rather than non-concussive, which remains equivocal<sup>20</sup>.

While mechanistic injury data is often gathered through injury reports and video analysis<sup>10</sup>, head-impact telemetry can provide objective data, quantifying head-impact exposure (HIE)<sup>22-24</sup>. A substantial effort is being made to quantify the inertial forces applied to the head in contact sports and to understand resulting neurocognitive and pathological changes<sup>23-26</sup>. Head-impact telemetry in contact sports has evolved over the past decade, using sensors embedded in helmets<sup>27</sup>, skin-mounted patches<sup>28</sup> and mouthguards<sup>24,29</sup>. The variation in measurement methods and scientific rigor makes the derivation of severity metrics unreliable. For example, inertial measurement unit (IMU) sensors mounted externally to the head, either on the skin or in helmets and headbands, significantly overestimate peak linear acceleration (PLA) and peak rotational acceleration (PRA)<sup>21,28,30</sup>. This is due primarily to poor sensor-skull coupling and soft tissue artefact (STA), where the skin deforms relative to the underlying bone<sup>31</sup>. The effects of skin deformation can become exaggerated during short-duration, high-magnitude, events measured by head-impact telemetry systems<sup>26,28</sup>. Miniaturisation of electronic components has facilitated the latest advances where IMUs have been integrated into custom-fit mouthguards<sup>24,25</sup>, ensuring secure coupling to the upper dentition. The aim of this study was therefore to explore sex-specific head-impact mechanisms for rugby union players.

## **Methods**

In total, 53 (25 female, 20.6±2.1 years; 28 male, 20.7±1.4 years) United Kingdom-based first XV university rugby union players provided written informed consent and participated in this observational cohort study. Head-impact telemetry, isometric neck strength and

anthropometric variables were measured at the start of the university playing season. Anthropometric data were collected using methods compliant with the International Society for the Advancement of Kinanthropometry (ISAK) guidelines<sup>32</sup>. A novel isometric neck strength testing apparatus (INSTA, see supplementary material), using fixed-frame dynamometry was used to test maximal isometric neck strength. Participants also completed a short questionnaire at the commencement of the study which included years of playing experience, playing level and injury history. Institutional ethics approval was obtained prior to the commencement of the study (Swansea University 2016-059), and the study was conducted in accord with the requirements of the declaration of Helsinki.

### *Neck Strength Testing Protocols*

Prior to maximal isometric neck strength testing, each participant completed a pre-exercise injury screening questionnaire. A standard, progressive ten-minute warm-up was then completed. This comprised of an aerobic component, then systematic activation of upper limb and cervical musculature, focusing on the deep cervical stabilisers. Participants were securely strapped to the INSTA, which was adjusted relative to their torso and upper leg length. Each participant was then asked to perform 50-70% efforts pushing on the load cell pads with their head, in each direction in preparation for testing.

To record the maximum voluntary contraction (MVC), participants were instructed to push on the load cell pads with their head as hard as possible for three seconds. This was repeated three times, in four different directions, in a randomised order: forward flexion (Flx), extension (Ext), right lateral flexion (R-Flx), and left lateral flexion (L-Flx). Between repetitions, participants were given 15-20 seconds rest, with one-minute rest between directions. Data were recorded using Hauch and Bach DOP4 software (Lyngø, Denmark).

### *Head Impact Telemetry Protocols*

A head-impact telemetry system (Protecht™, Sports Wellbeing Analytics Ltd, Wales, UK), incorporating bespoke instrumented mouthguards (iMGs), was used to record linear and rotational head acceleration in competitive rugby matches. iMGs were produced for 13 female and 14 male players, using dental impressions to ensure tight coupling to the upper dentition. Data were collected during 13 (six female, seven male) competitive matches.

Each iMG contained an embedded 9-axis IMU and an additional tri-axial accelerometer (LSM9DS1 and H3LIS331DL, respectively; STMicroelectronics, Geneva, Switzerland). The

iMG sampled at 952 Hz over a 104 ms period post trigger, with a 12-bit resolution and ranges of  $\pm 200$  g and  $\pm 35$  rad·s<sup>-1</sup> for the linear accelerometer and gyroscope, respectively<sup>24,25</sup>. The raw data was transmitted via radio frequency and stored as a time-series .csv file. The iMG measured any head-impact event  $>10$  g (on the linear accelerometer). The system also contained a proximity sensor to ensure accelerations were only recorded when the iMG was coupled to the participant's teeth. The iMG system has undergone validation testing in laboratory-based settings<sup>24,25</sup> and records inertial values at the position of the sensor, located lateral to the left fifth molar<sup>24</sup>.

### *Post-Processing Protocols*

All data were filtered using a 4th order, zero lag, low-pass Butterworth filter to remove high-frequency noise. A single cut-off frequency was not found to be appropriate for all impacts, due to variability in the underlying signal components<sup>33</sup>. Consequently, impact-specific, optimal cut-off frequencies were determined for each impact using residual analysis. Filtering was applied to vector component data. Rotational acceleration was calculated from the rotational velocity data using five-point differentiation. PLA values below 9.6 g post filtering were excluded to prevent the recording of accelerations from natural movements, such as jumping<sup>34</sup>.

### *Head Impact Verification Protocol*

Matches were video recorded to verify head-impact sensor data, enabling the identification of corresponding impact mechanisms and events. To minimise false positives<sup>26</sup>, a two-stage (video and timeseries waveform) verification protocol was used for each automatically-detected impact (Figure 1). Video verification included observed impact events occurring up to ten seconds prior to the iMG system impact timestamp, due to occasional delays in iMG signal transmission. Subsequently, laboratory testing was conducted for waveform classification to identify characteristics commonly responsible for false-positive impacts, with controlled, ground-truth events. These included shouting, biting, sucking, and inserting and removing the iMG. Waveform classification testing also included simulated head-impact contact events, with the properties of corresponding iMG waveforms analysed and characterised. A sub-sample of five available participants completed a total of 160 biting events, 120 removals, 120 insertions, 50 sucking and 75 rugby tackles with hit shields. Various filtering processes were retrospectively applied to these data to determine optimal data-processing protocols to minimise false positives, whilst ensuring the retention of legitimate

head impact events. Figure 1 shows characteristic waveforms from a bite (Figure 1A), a shout (Figure 1B), and a true non-filtered (Figure 1C), or filtered (Figure 1D) head-impact event. A legitimate head-impact waveform should have a smooth bell-shaped curve (Figure 1D). Waveforms caused by biting, shouting and inserting/removing artefacts have distinct, sharp peaks that represent unrealistic acceleration profiles for the human head (Figures 1A and 1B). Impacts were excluded from the dataset if waveforms were not representative of feasible head acceleration or if incomplete waveforms were recorded, where data had been dropped during transmission. A feasible head acceleration was determined by a singular impact peak with a duration of at least 5 ms, and if subsequent peaks were present, these were no less than 10 ms apart (Figure 1C and D). Impact peaks of shorter duration and closer proximity reflected the waveforms observed in laboratory-based biting and shouting events (Figures 1A and B respectively).

#### *Statistical Analyses*

All statistical analyses were performed using SPSS Statistics 26 (IBM Corp, Armonk, NY). The alpha value for all analyses was set at  $p > 0.05$ . Shapiro-Wilk tests were used to assess the distribution of all data. Sex differences in age and playing experience were assessed using Mann-Whitney tests. Independent t-tests were used to determine differences in anthropometric and neck strength results between sexes and between forward and back positional groups for females and males. Directional imbalances in neck strength were compared using paired-t-tests. Head impact magnitudes between sexes and positions were assessed with Kruskal-Wallis tests. Pearson's correlations between neck strength, anthropometric and head impact data were conducted to establish patterns.

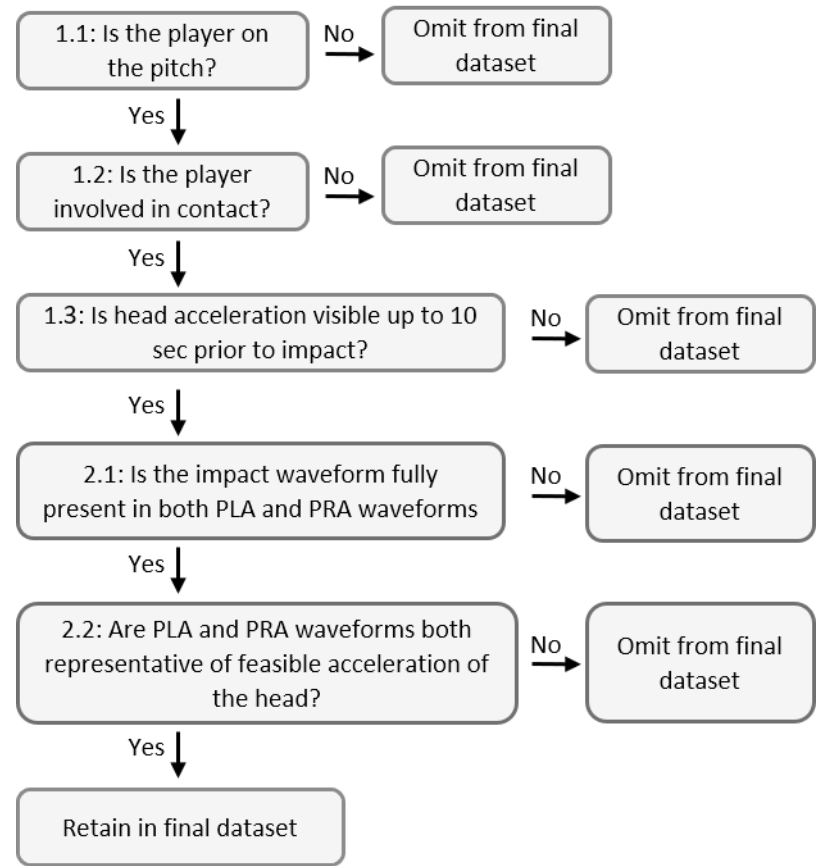
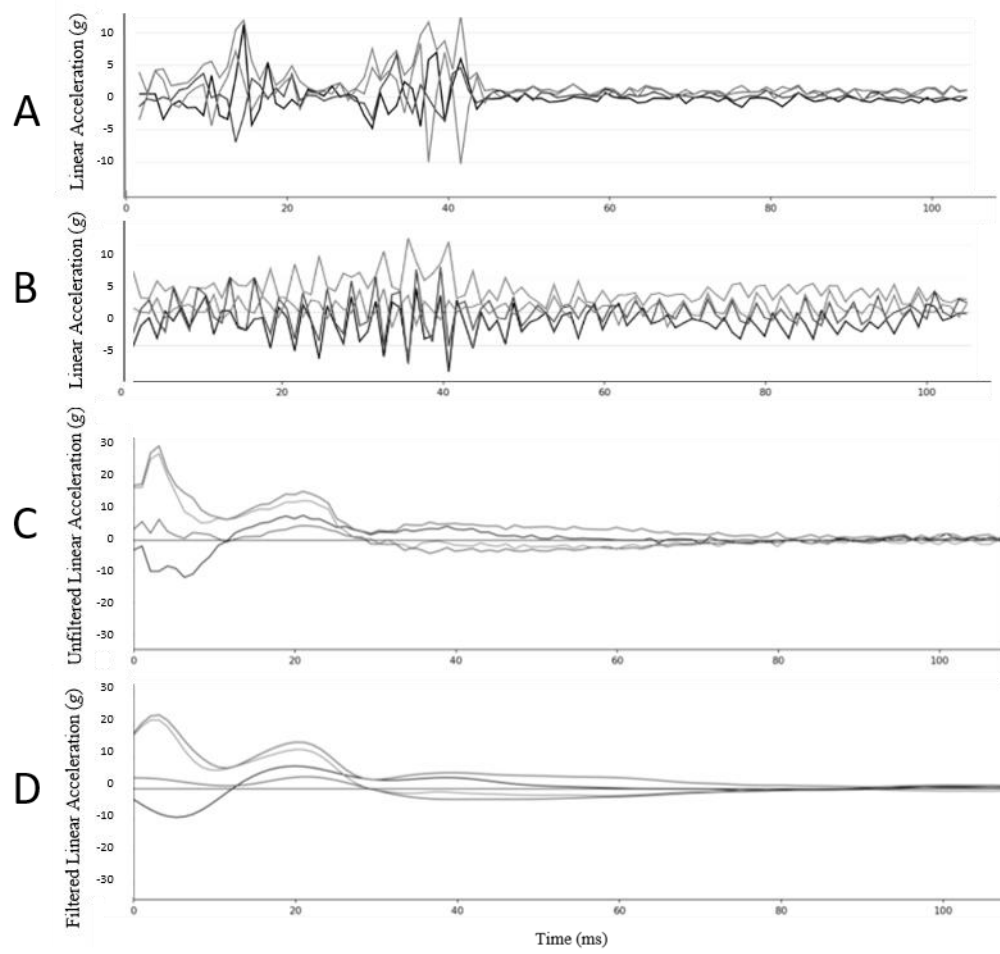


Figure 1. iMG linear acceleration timeseries waveform examples characteristic of biting (A), shouting (B), a verified, non-filtered (C), and filtered (D) head impact event.

## Results

### *Age and Playing Experience*

Mann-Whitney U tests found no significant sex difference in age ( $U=123.5$ ,  $p=.32$ ). However, female players had significantly fewer years of playing experience ( $5.2 \pm 3.3$  years;  $U=9.5$ ,  $p<.05$ ) than males ( $13.1 \pm 2.5$  years). During their careers, 58.8% of females and 60.0% of males reported having suffered with at least a single diagnosed concussion, with 35.3% and 38.0% respectively, having experienced multiple concussions. Anthropometric data were normally distributed, with the mean  $\pm$  SD for female ( $n=25$ ) and male participant data ( $n=28$ ) provided in Table I.

### *Neck Strength Data*

Results for isometric neck strength are given in Table I. Female MVC scores for Flx, Ext, R-Flx and L-Flx were 50%, 43%, 47% and 50% lower, respectively than males (Table I). For both sexes, the average MVC in the anteroposterior direction (Flx and Ext) was significantly greater ( $p<0.001$ ) than mediolateral direction (R-Flx and L-Flx).

### *Head Impact Telemetry*

A total of 1723 (747 female; 976 male) impacts were recorded with the iMG system. Of these, 305 (102 female; 203 male) were excluded through video verification, a further 151 (67 female; 84 male) were excluded through waveform analysis, additional to the 1049 (505 female; 544 male) which were excluded due to having incomplete waveforms. Table II shows the number of total verified impacts, and the median, interquartile range (IQR) and maximum values for PLA and PRA. Shapiro-Wilk tests identified that the head acceleration data was not normally distributed ( $p<0.05$ ). Independent samples Mann-Whitney U tests found no significant overall differences between females and males for PLA ( $p=0.23$ ) or PRA ( $p=0.76$ ), or between female forwards and backs (PLA  $p=0.44$ , PRA  $p=0.57$ ) or male forwards and backs (PLA  $p=0.39$ , PRA  $p=0.64$ ). For both females and males, Kruskal-Wallis test results found no significant differences in PLA ( $p=0.53$  and  $p=0.58$ , respectively) or PRA ( $p=0.34$  and  $p=0.75$ , respectively) between tackle events, carries and rucks. Figure 2 shows the distribution of PLA and PRA for female and male players.



Table I. Results and statistical analyses for anthropometric and neck maximum voluntary contraction (MVC) tests for female and male forwards (Fwd) and backs (Bk).

							Female vs Male		Female Fwd vs Bk		Male Fwd vs Bk	
	Female	Female Fwd	Female Bk	Male	Male Fwd	Male Bk	<i>t</i>	<i>p</i>	<i>t</i>	<i>p</i>	<i>t</i>	<i>p</i>
<b>Body Mass (kg)</b>	73.5±8.3	81.1±18.7	65.9±6.6	95.6±11.6	105.7±8.5	85.5±8.2	7.11	<0.001	2.3	0.03*	5.8	<0.001*
<b>Height (cm)</b>	163.7±6.0	163.5±6.1	163.9±5.9	184.1±8.1	188.9±9.2	179.2±6.4	11.75	<0.001*	0.4	0.71	3.9	<0.01*
<b>BMI (kg/m<sup>2</sup>)</b>	27.4±5.7	28.7±5.3	26.1±5.9	28.2±3.6	29.7±2.7	26.6±2.3	0.8	0.94	2.3	0.03*	2.8	<0.05*
<b>Head Circ. (cm)</b>	56.1±1.0	56.2±0.9	56.0±1.4	58.3±1.6	59.0±1.6	57.5±1.7	6.45	<0.001*	0.4	0.69	2	0.06
<b>Neck Circ. (cm)</b>	35.3±1.5	36.2±2.8	34.4±2.9	41.5±2.1	42.9±1.8	40.1±1.7	9.1	<0.001*	2.8	0.03	3.9	<0.01*
<b>Neck to Head (%)</b>	62.5±2.8	64.0±5.0	61.0±5.0	71.6±3.4	73.0±3.0	70.0±2.0	8.46	<0.001*	1.6	0.12	3	<0.01*
<b>Shoulder Breadth (cm)</b>	38.4±2.0	38.8±1.8	38.0±2.1	43.6±2.1	44.3±2.5	42.8±2.1	9.69	<0.001*	1.1	0.26	1.6	0.12
<b>Neck MVC</b>							<i>t</i>	<i>p</i>	<i>F</i>	<i>p</i>	<i>F</i>	<i>p</i>
<b>Flexion (N)</b>	144.0±28.9	152.7±29.0	137.4±27.8	271.1±22.5	300.5±58.8	250.8±44.1	10.87	<0.001*	2.2	0.15	5.14	0.035*
<b>Extension (N)</b>	148±28.9	159.7±25.1	139.0±29.0	244.7±49.8	251.3±47.9	240.1±52.5	8.81	<0.001*	4.2	0.05*	2.6	0.620
<b>R-Lat. Flexion (N)</b>	113.7±20.3	118.4±17.6	110.2±22	197.7±56.6	227.3±73.5	177.2±29.8	7.52	<0.001*	1.2	0.28	4.95	0.038*
<b>L-Lat. Flexion (N)</b>	112.3±24.1	119.8±21.2	106.5±25.2	201.4±58.6	239.0±68.8	175.3±32.3	7.53	<0.001*	2.3	0.14	8.554	0.008*

Table II. Number of matches and participants and overall head impact results including peak linear acceleration (PLA) and peak rotational acceleration (PRA).

	Female	Female Fwd	Female Bk	Male	Male Fwd	Male Bk
<b>Participants (n)</b>	13	6	7	14	8	6
<b>Total Verified Impacts (n)</b>	73	43	30	145	99	46
<b>PLA Median ± IQR (g)</b>	11.7±7.9	12.1±13.8	13.0±14.1	12.5±7.0	12.0±15.1	14.0±14.0
<b>PLA Max (g)</b>	44.3	44.3	36.6	50.5	50.5	31.6
<b>PRA Median ± IQR (rad·s<sup>-2</sup>)</b>	800.2±677.7	809.2±1061.4	852.2±869.0	849.4±479.8	852.0±869.0	848.0±928.5
<b>PRA Max (rad·s<sup>-2</sup>)</b>	4125.6	3401.9	4125.6	2973.2	2973.2	2559.1

For female players, 84.9% occurred during the tackle event (61.3% to the tackler and 38.7% to the ball carrier), and 15.1% within rucks and mauls. For males, 73.6% of verified impacts were sustained during the tackle event; 53.3% by the tackler and 46.7% by the ball carrier. Rucks accounted for 23.6% and the remaining 2.8% were recorded in mauls. No verified head impacts were recorded within lineouts or scrums for either females or males. However, in some cases, visual inspection was obscured by other players.

Overall, the most common general causes of head impact for females were direct impacts to a hard body part (43.7%), followed by head to ground impacts (26.1%), indirect impacts (17.8%), direct to soft body parts (8.2%) and other (4.2%). Head-to-body contact events in females were to the shoulder (24%), the hip (20%), the knee (13%), the head (13%). The remainder were hand-offs direct to the head or neck, with one incidence of contact between each of the ball, the upper thigh, upper back, hand, elbow, forearm, and boot recorded. For males, indirect (contact to body with momentum transferred to the head) impacts were the most common (31.3%), followed by direct contact with a hard body part (29.2%), direct contact with a soft body part (19.4%) and direct contact with the ground (9.7%). For males, most head-to-body impacts were to the upper leg (25%), hip (25%), shoulder (19%), head (15%) and stomach (15%).

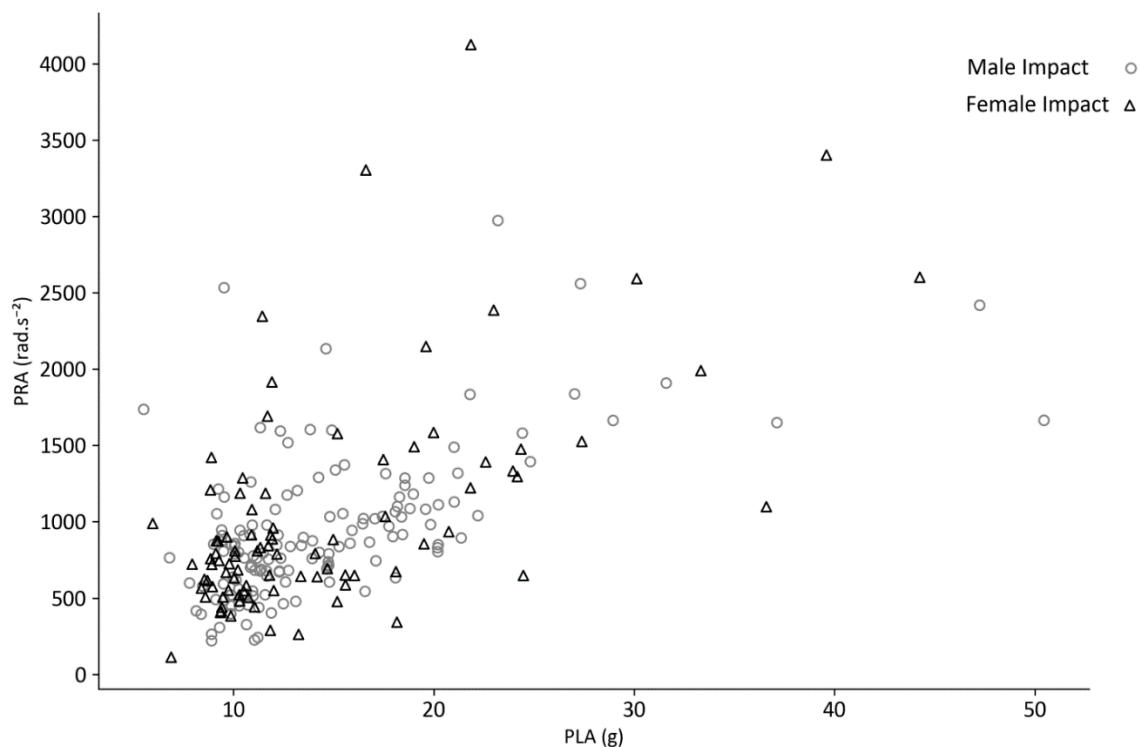


Figure 2. Scatterplot showing peak linear acceleration and peak rotational acceleration of the head for female and male participants in six and seven competitive matches, respectively.

For female tacklers, 57% of all impacts were caused by direct head to hard body part and head to soft body part accounted for 3%, head to ground contact 23.7%, indirect head contact 11% and other mechanisms 5%. For male tacklers, direct contact between the head and hard body parts accounted for 33%, direct head contact to soft body parts 27%, indirect impacts 19%, direct head contact with the ground 14%. For the remaining 7%, the exact mechanism could not be confirmed.

For impacts sustained by the ball carrier in females, head to ground impacts were the most dominant (45%), followed by indirect impacts (31%), head to hard body parts (18%), head to soft body parts (5%) and 2% incidental events including one incidence of ball to chin. For male ball carriers, indirect impacts accounted for 51%, head to ground contact 14%, direct contact with hard body parts 13%, direct contact with soft body parts 12% with the remaining 10% unclear.

Whiplash style head kinematics were observed in 51.2% of women's head impact events (78% of head-to-ground and 47% of body contact impacts), and players exhibited a lack of control and positional awareness of their head in space. This was recorded on only one occasion in a male, where the player appeared unprepared for an impact and fell backwards with the head impacting the ground (PLA 50.5 g, PRA 1663.8 rad·s<sup>-2</sup>). This whiplash head-to-ground action was also observed in another male player (not instrumented) who was knocked unconscious before contacting the ground.

For females, Kruskal Wallis tests showed no differences for either PLA or PRA were observed between different causes of head acceleration (PLA  $H=4.4$   $p=0.445$  PRA  $H=3.4$   $p=0.282$ ). The cause of acceleration in male players to have a significant effect on PRA ( $H=11.36$ ,  $p<0.01$ ). Direct head contact with hard body parts, 900 rad·s<sup>-2</sup> (705–1,255) and soft body parts, 975 rad·s<sup>-2</sup> (763-1,244) produced significantly higher PRA to indirect impacts (737 rad·s<sup>-2</sup>, 543-943), ( $U=348$  and  $U=605$  respectively,  $p<0.01$ ). No significant differences for PLA were found between the different causes of acceleration in males ( $H=6.16$ ,  $p=0.11$ ).

## **Discussion**

The current study objectively compared sex differences in head impact kinematics and neck strength in university rugby union players. Congruent with recent research<sup>35</sup>, no significant overall differences were found between the head impact magnitudes for either PLA or PRA,

despite significant sex differences in body size. The head impact mechanisms, however, differed considerably. Direct head-to-ground impacts accounted for 38.5% and 9.7% of female and male recorded head impacts, but the head kinematics were different. Uncontrolled whiplash action was observed in over half of all recorded female impact events, including 80% of head-to-ground impacts, but only once in a male. Statistical comparison of mechanisms was also complicated by this whiplash action. These mechanistic differences reflect previous research, which has reported the female cervical spine to be less proficient at resisting inertial loading<sup>36,37</sup> than male. This is largely due to sexual dimorphisms in cervical spine anatomy, compromising the intervertebral coupling stability in females<sup>15</sup>.

The current findings demonstrate that these well-established, underlying physical sex differences are dominating factors in rugby head-impact dynamics, particularly regarding cervical spine stability. Overall, these findings indicate that a vast array of research, focusing on female-specific injury mechanisms and epidemiology, is required to safely advance the women's game. While rugby union uses an evidence-based approach for the development of concussion management protocols<sup>38</sup>, these are predominantly based on androcentric data lacking female representation<sup>6</sup>. When female players have been included, this is usually restricted to the elite demographic<sup>5</sup>, with the very recent inclusion of collegiate players<sup>35</sup>. For example, the extent to which cervical spine weaknesses and instabilities are trainable in female players requires further investigation. Well-developed neck strength and stability has been reported as a predictor of brain injury in a number of contact sports<sup>19</sup>, limiting head acceleration and mitigating energy transfer to the brain<sup>39,40</sup>. To mitigate impact severity, good stabilisation, activation and control of both agonist and antagonist muscles, with respect to the direction of perturbation, are required<sup>13</sup>.

Sex differences in game dynamics are also reflected in the relative proportion of direct and indirect impacts, with 17.8% of recorded indirect impacts in females, compared to 31.3% in males. While only video and waveform-verified impacts are included in these figures, and the actual proportions may differ, tactics are an important area for future investigation. A confounding factor in determining the extent that physical sex differences affect head impact kinematics is rugby experience. Despite both groups in the current cohort being the same age, male players had an average of eight years more playing experience than females. It is thus possible that males have progressed through levels of development, learning fundamental movement skills at a younger age including falling and tackling. This disparity in rugby

experience limits the applicability of sex-specific impact dynamics comparisons and further questions the appropriation of male data-derived training strategies to female players.

While not without controversy<sup>26</sup>, the use of head impact telemetry in sports is becoming more commonplace, with a parallel increase in the technological competency of these systems<sup>23–25</sup>. Steps are also being taken to limit soft tissue artefact through tight coupling to the teeth<sup>28</sup>. Nonetheless, there remains no consensus on the minimum recording, analysis or reporting processes for head impact data. Future advances in machine learning algorithms to characterise timeseries waveform data, with larger datasets, are required to optimise the practicality and accuracy of this task.

The head impact magnitudes and numbers reported in this study are typically lower than previous rugby union studies<sup>35</sup>. This is likely due to a combination of factors. Specifically, secure sensor-skull coupling achieved with bespoke mouthguards, minimises soft tissue artefact relative to externally-worn sensor patches<sup>41</sup>. This may also have reduced the number of reported impacts. Indeed, skin patch sensors have been found to overestimate peak linear acceleration with a normalised mean square error of 18% and 290% for PLA and PRA, respectively<sup>28</sup>. Furthermore, this study utilised video verification of each impact event with corresponding iMG waveform, converse to previous head impact telemetry studies, which 64% failed to do, likely resulting in substantially overestimated head impact exposure<sup>26</sup>. It is also pertinent to acknowledge the waveform verification of each impact, where all bite and shout artefacts were removed, resulting in omission of typically higher magnitude waveforms. This has not previously been included in the post-processing of data of this nature, which may have erroneously included, for example, bite artefacts, akin to Figure 1a, in the final analysis. Finally, post-processing application of a 4th order, zero lag, low-pass Butterworth filter, with impact-specific, optimal cut-off frequencies determined through residual analysis. While 74% of previous studies included in the Patton et al. review<sup>26</sup> reported using filtering algorithms, most studies do not report the details of their data filtering procedures, possibly due to commercial proprietary restrictions. Without this information, it is difficult to evaluate the accuracy of the reported data, as low-pass filters can significantly affect recorded impact magnitudes.

Unfortunately, the COVID-19 pandemic resulted in a premature end to the season, limiting opportunities for data collection. Furthermore, in situations where player density was high, such as in mauls, signal transmission failures meant that some recorded impacts were likely

lost. This can be improved in future studies with further evolution of sensor communication protocols. Researchers were not lenient when applying the verification criteria. Thus, a non-zero false negative rate is likely. Consequently, this study cannot comment on the overall number and incidence of head impacts. Instead, it reports a smaller number of verified head impact magnitudes. Importantly, it is unlikely that there were different false negative rates across female and male players. Inclusion of any concussion diagnoses, neuroimaging results or other medical reports were beyond the scope of the current study, where the focus centred on the head impact biomechanics. Future studies should take a multidisciplinary approach to understand the implications of repetitive head impacts experienced by contact sport athletes.

## **Conclusion**

The sex differences in head impact mechanisms observed in the current study do not support the generalisation of male-derived injury prevention data to female players. Nonetheless, it is pertinent to note that females in the current study had fewer playing opportunities and an average of eight years less playing experience than their male counterparts. These results suggest that a considerable effort into identifying the specific differences of female rugby players and deriving appropriate training interventions to overcome these is required. Improving the accessibility of rugby union and providing quality, informed, targeted coaching to female players may also contribute towards mitigating the high and inequitable rate of reported concussions. Non-scalable sex differences in cervical spine size, curvature strength and stability have been well established in other scientific fields and must be acknowledged in rugby union. These differences, previously attributed to lower head-neck stabilisation during head acceleration, are likely key to understanding the different head impact mechanisms in this study.

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## **Declaration of Interest Statement**

None of the authors have any conflicts of interest.

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