Title: Head impact exposure from match participation in women's rugby league over one season of domestic competition

Running title: Head impacts in women's rugby league

Abstract

Objectives: To quantify the magnitude, frequency, duration and distribution of head impact exposure in a women's rugby league competition.

Design: Prospective cohort study.

Methods: Twenty-one players had a wireless impact measuring device (X2Biosystems XPatch) behind their right ear during match participation. Head impact data were collected and downloaded for analysis. Median peak linear and rotational accelerations and impact locations between player positions were assessed using a Friedman repeated measures ANOVA on ranks with a Wilcoxon signed-rank test for post hoc analysis with a Bonferroni correction.

Results: A total of 1,659 impacts to the head >10*g* were recorded (range 10*g* to 91*g*) over the nine competition matches. There was a mean of 184 ±18 impacts per-match resulting in a mean of 14 ±12 impacts per-player per-match. The No. 8 prop recorded a mean of 29 ±27 impacts per-match, the No. 12 second-row forward recorded the highest median peak resultant linear acceleration (16*g*) per-match and the No. 11 second-row forward recorded the highest median peak resultant rotational acceleration (3,696 rad/s^2).

Conclusion: Our cohort of 21 female rugby league athletes were exposed to repetitive sub-concussive head impact exposure with an average of 14 impacts per-player per-match. Forwards were exposed to more impacts per-match than backs and these impacts were of higher magnitude. Most impacts occurred on the side of the head and were sustained during the second half of the game. Clinicians, coaches and players should be aware of the rates and magnitude of head impacts in female rugby league athletes.

Keywords: Head impact; rugby league; peak linear acceleration; peak rotational acceleration.

Introduction

As an intermittent collision-based team sport, rugby league is played at junior, amateur (male and female), semi-professional and professional levels of competition.[1] Rugby league is a challenging contest for players to compete in, comprising intense frequent bouts of high-intensity activity (e.g. sprinting) and collisions (e.g. offensive ball carrying and defensive tackling), interspersed with short bouts of low-intensity activities (e.g. walking, jogging).[2] As a result of these activities, players can experience 29 to 74 physical collisions (tackles and ball-carries) per game.[3, 4] From these physical collisions there is an inherent risk of injury , including concussion[5] to the players involved, as impacts to the body and head happen.[1] A pooled analysis of rugby league concussions[6] reported an incidence of 7.7 per 1000 match hours. Males had a higher incidence of concussion than females (7.7 vs 6.1 per 1,000 match hours) and amateurs recorded the highest concussion incidence (19.1 per 1,000 match hours) when compared with semi-professional (5.9 per 1,000 match hours) and professional (7.1 per 1,000 match hours) rugby league participants. To date there are no published studies reporting on head impacts in men's or women's rugby league.

Sports-related concussion is a common injury and many go unrecognised.[5] It has been reported that approximately 90% of concussions do not result in loss of consciousness.[5] Also, they are sometimes not detected or undiagnosed and underreporting rates are estimated to be as high as 50% to 90%[5] Knowledge of the potential metabolic and ultrastructural consequences of impacts to the head has increased, placing a greater focus on the possible deleterious effects of repetitive concussive and sub-concussive impacts in some individuals.[7] Technology, such as accelerometers in the helmets of American football players,[8-10] mouth guards of amateur rugby union players[11] and patches on junior rugby union players[12] have increased the knowledge and analysis of injury biomechanics of the forces, accelerations, frequencies and velocities of head injuries.[13] Despite these studies, to date there are no published head impact studies reporting on women's contact sports such as rugby league. Therefore, this study quantified impacts to the head via an instrumented patch worn behind the ear for women rugby league players over a single domestic competition season in New Zealand.

Methods

A prospective cohort design examined the frequency, magnitude and duration of head impact exposure during women's rugby league matches. The researcher's university ethics committee (AUTEC 16/35) approved all procedures in the study and all players gave informed consent prior to participating in the study.

Twenty-one female rugby league players were enrolled into the study [mean (±SD) age of 29.2 ±7.8 yr]. The players competed in nine competition matches resulting in a match exposure of 155.6 hours. Players were placed into three positional groups: (1) hit-up forwards (n= 4: 2 x prop, 2 x second row); (2) outside backs (n= 4: 2 x centre, 2 x wing); and (3) adjustables (n= 5: hooker, halfback, five-eight, loose forward and fullback).[14] Three players (prop, loose forward and centre) wore their own scrum caps during match participation. These were Canterbury Ventilator headguard (prop and centre) and Canterbury honeycomb headgear (loose forward).

Head impact exposure was measured with the XPatch (X2Biosystems, Seattle, USA). The XPatch is a 1cm x 2cm device that measures acceleration and is mounted with a single-use adhesive behind the right ear. Containing a triaxial accelerometer and a triaxial angular rate gyroscope to capture six degrees of freedom for linear acceleration and rotational velocity, the XPatch has a 4.2v battery and a small memory chip measuring continuously at 1 kHz for linear acceleration and 800 Hz for angular velocity[15] and this is triggered when impacts greater than a pre-set level of 10*g* occur. This threshold was chosen based on a review of previously published studies[11, 16] and it has been reported that running and jumping were observed to elicit a maximum of 9.54*g* of linear head acceleration.[17] The XPatches were synchronised and checked in the morning of the match. Prior to commencing their warm-up, the lead researcher applied the XPatch behind each player's right ear ensuring it was fitted over the mastoid process and loose hair was not in the adhesive. The patches were allocated to player positions and the players' names were recorded. Data collection was delimited to matches only - not team training sessions.

When an impact above this pre-set level occurs, the device saves 10-milliseconds prior to the impact and 90-milliseconds after the impact providing X, Y and Z coordinates of acceleration at 1-millisecond intervals (see Fig 1). Peak linear acceleration (PLA) was measured and peak rotational acceleration (PRA) was then calculated. The time stamp of the match was synchronised with the X2 XPatch prior to every game. The frequency, location, PLA, PRA and duration of all head impacts ≥10*g* threshold of linear acceleration were recorded by the XPatch for each match and stored on the device until uploaded. The XPatch has a strong correlation with peak linear acceleration (PLA: r^2 = 0.93) with a normalised root square error of 18%, but may over predict PLA and PRA by 15g ± 7g and 2,500 ±1,200 rad/s^2, respectively. [18] The XPatch has also been reported[15] to have good agreement with PLA, can underestimate PRA by at least 25% and has a significant statistical correlation with the Head Impact telemetry System (HITS) for PLA (r = 0.144; p<0.001), PRA (r = 0.15; p<0.001) and Head Impact Telemetry severity profile (HITsp) (r = 0.34; p<0.001).[19]”

Before the statistical analysis was conducted, the raw data were reduced As follows. Data contained on the XPatch were uploaded to the Impact Management System (IMS) provided by X2Biosystems. The data were then downloaded and filtered through the IMS to remove any spurious linear acceleration that did not meet the proprietary algorithm for a head impact.[20] The data underwent a second filtering waveform parameter proprietary algorithm during data exporting to remove spurious linear acceleration data with additional layers of analysis.[20] This included the area under the curve, the number of points above threshold and filtered versus unfiltered peaks.[20] Press[21] estimated that approximately 80% of the impacts recorded may have been removed through the analysis that X2Biosystems provide as part of the IMS program. This may be similar for the current study. The remaining data were exported onto an Excel spreadsheet (version 2013; Microsoft Corporation, Redmond, WA) for visual examination. The data were then reviewed by impact time stamps (hr:min:sec) to identify identical and sequential patterns for each player. Time stamps with multiple (≥2) linear accelerations having the same hr:min:sec time stamp in quick succession milliseconds after the preceding impact were removed. These were removed by the authors by utilising Microsoft Excel conditional formatting and duplicate values to screen for linear accelerations with the same hr:min:sec time stamp in quick succession following downloading the impacts from the IMS. The data were then screened for player number and if there were any duplicate impacts on the hr:min:sec time stamp with the same player number these were removed. No incidence of this was identified in the final screening of the data prior to analysis in the Microsoft Excel. Once the review was completed, the data estimates were adjusted to estimates of the Hybrid III headform criterion standard and all impacts <10*g* were removed (n=185) from the database. This was undertaken to remove any impacts <10*g* following the completion of the adjusted calculations in line with previous results[22] that reported the XPatch over-estimates the linear accelerations when compared with the centre of gravity of the headform criterion.

All filtered data on the Microsoft Excel spreadsheet were analysed with SPSS V.23.0.0. To test for normality, one-sample Kolmogorov-Smirnov and one sample *t*-test were conducted. The impact variables were not normally distributed ($D_{(1659)}$=0.23; p<0.0001; $t_{(1658)}$=67.0; p<0.0001), therefore data were expressed as median [IQR] and 95th percentile. Three measures of impact frequency were computed for each player: (1) *player position impacts*, the total and median number of head impacts recorded for the playing position for all matches; (2) *player group impacts*, the total and median number of recorded head impacts for the playing group (hit-up forwards, adjustables and outside backs) for all matches; and (3) *impacts per match*, the total and median number of impacts per match for all matches.

Player head impacts exposure were assessed utilising previously published levels for injury tolerance[23-25] (linear (>95 *g*) and rotational acceleration (>5,500 rad/s^2)), and impact severity[26-28] (linear (mild <66 *g*, moderate 66-106 *g*, severe >106 *g*) and rotational acceleration (mild <4,600 rad/s^2, moderate 4,600-7,900 rad/s^2, severe >7,900 rad/s^2)). Two additional risk equations were included in the analysis of the head impact exposure data. The Head Impact Telemetry Severity profile (HIT_{SP})[29] is weighted composite score including linear and rotational accelerations, impact duration, as well as impact location. The Risk Weighted Exposure Combined Probability (RWE_{CP})[30] 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_{CP} combines the resultant linear and rotational accelerations to elucidate individual player and team-based head impact exposure. As a value of 63 is a 75% indicator for a concussive injury[29, 31] the HIT_{SP} values were evaluated by limits of less than 25% risk (<21), 25% to 75% risk (21-63) and >75% risk (>63). The RWE_{CP} values were evaluated by the same values of 25% risk (<0.2500) 25% to 75% risk (0.2500-0.7500) and >75% risk (>0.7500). The HIT_{SP} and RWE_{CP} were analysed by player position impacts and player group impacts utilising a Friedman repeated measures ANOVA on ranks. Post hoc analysis with Wilcoxon signed-rank tests was conducted with a Bonferroni correction applied.

Total frequency impact burden per-match was analysed using a Kruskal-Wallis one-way ANOVA with a Dunn's post-hoc test for all pairwise comparisons with player positions. Although there is no accepted method to quantify total frequency impact burden,[24] the sum of linear and rotational accelerations associated with each individual head impact per-match over the course of the study was calculated for all of these parameters. The total sum of the resultant peak linear and the peak rotational accelerations recorded was undertaken for each match half, total match, player role, player position and for the duration of the study. Head impact exposure including impact duration, frequency, magnitude and location of impacts were quantified using previously established methods.[10, 32]

Median peak linear and rotational accelerations and impact locations between player positions were assessed using a Friedman repeated measures ANOVA on ranks with a Wilcoxon signed-rank test for post hoc analysis with a Bonferroni correction applied. Impact locations were analysed by front, back, side and top impacts using a Friedman repeated measures ANOVA on ranks by comparing impacts sustained in each location. A one sample chi-squared (χ^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. Statistical significance was set at $p<0.05$.

Results

Data were summarised and presented as total impacts recorded and impacts per-player position and per-player group[14] over a competition match season for injury tolerance level,[15, 32-34] impact severity limits,[26-28] head impact telemetry severity profile[29] and risk weighted cumulative exposure (combined probability)[30] by total impacts recorded, percentage of impacts recorded (%),and median [25th to 95th interquartile range] in Supplemental Table 1.

During the competition there were 1,659 impacts (range 10*g* to 91*g*) recorded ≥10*g* (Table 1) with a mean of 184 ±18 impacts to the head per-match resulting in a mean of 14 ±12 impacts to the head per-player per-match. Players recorded a median [IQR] and 95th percentile of 15 [12 to 21] *g*, and 41*g* respectively for peak linear, and 2,886 [1,864 to 4,545] rad/s^2 and 9,348 rad/s^2 respectively for peak rotational acceleration with a mean impact duration of 11.6 ±8.1ms. There were seven impacts recorded >80*g* over the nine competition matches. Forwards recorded more impacts than backs (920 vs. 739; RR: 1.24 [95% CI: 1.16 to 1.33]; p<0.0001) (see Table 1). As a result, hit-up forwards (n=702; RR: 3.31 [95% CI: 2.89 to 3.80]; p<0.0001) and adjustables (n=745; RR: 3.51 [95% CI: 3.07 to 4.03; p<0.0001) recorded more impacts than outside backs (n=212). Forwards recorded lower resultant median [IQR] peak linear (14 [12 to 20] *g* vs. 15 [12 to 21] *g*: χ^2=518.99; p<0.0001) and rotational (3,156 [1,924 to 4,911] rad/s^2 vs. 3,331 [2,168 to 5,215] rad/s^2: χ^2=89.38; p<0.0001) accelerations but had a higher mean impact duration (11.9 ±8.5ms vs. 11.1 ±7.7ms: χ^2=8.48; p=0.0036) than backs over the nine competition matches.

There were more impacts recorded in the second (n=910; RR: 1.21 [95% CI: 1.13 to 1.30]; p=0.0001) than the first half of competition matches (n=749) (Table 1). The second half of matches recorded a higher median [IQR] peak linear (15 [12 to 21] *g* vs. 14 [11 to 19]*g*: χ^2=13.13; p=0.0003) and peak rotational (3,429 [2,183 to 5,455] rad/s^2 vs. 3,167 [1,922 to 4,669] rad/s^2: χ^2=5.07; p=0.0243) accelerations, and had a higher mean impact duration (12.4 ±8.0 ms vs. 10.5 ±8.2 ms: χ^2=14.79; p=0.0001) when compared with the first half.

The number of head impacts, peak resultant linear and rotational accelerations and the total frequency impact burden varied by player position (Table 2). In the forwards, the No. 8 prop recorded a mean of 29 ±27 impacts per-match, the No. 12 second-row forward recorded the highest median peak resultant linear acceleration of 16 [13 to 23] *g* per match and the No. 11 second-row forward recorded the highest median peak resultant rotational acceleration of 3,696 [2,981 to 6,283] rad/s^2. The No. 8 prop recorded the highest total frequency impact burden for peak resultant linear (5,572 *g*) and rotational

(1,332,128 rad/s^2) accelerations. In the backs, the stand-off recorded a mean of 30 ±15 impacts per match, the right wing recorded the highest median peak resultant linear acceleration of 18 [12 to 29] *g* per match and the left wing recorded the highest median peak resultant rotational acceleration of 4,542 [2,679 to 6,799] rad/s^2. The stand-off recorded the highest total frequency impact burden for peak resultant linear (5,445 *g*) and rotational (1,343,479 rad/s^2) accelerations.

The side of the head was the most common impact location (n=791; 48%) (Table 3). Hit-up forwards recorded more impacts to the top of the head than outside backs (χ^2=4.8; *p*=0.0278) and adjustables (χ^2=1.2; *p*=0.2733). Hit-up forwards recorded a higher median peak resultant linear acceleration (22 [15 to 35] *g*) than outside backs (12 [12 to 13] *g*; *p*<0.0001) and adjustables (18 [15 to 28] *g*; *p*=0.0286) to the back of the head. Adjustables recorded the highest mean peak resultant rotational acceleration (5,504 [3,866 to 6,818] rad/s^2) than hit-up forwards (4,409 [3,779 to 8,502] rad/s^2; *p*=0.0022) and outside backs (1,365 [968 to 8,240] rad/s^2; *p*=0.0176) to the back of the head.

There were 302 impacts recorded above the rotational (>5,500 rad/s^2) tolerance threshold resulting in a median of 7,721 [6,268 to 9,567] rad/s^2 (see Supplemental Table 1). Most (97% to 99%) of linear head impact exposures were in the low (<66*g*) injury severity threshold. Backs (23%) recorded more head impact exposures in the moderate (21 to 63) severity for HIT_{SP} values than forwards (20%; $\chi^2_{(1)}$=50.58 *p*<0.0001). As a result backs (0.4801 [0.3207 to 0.6083]) recorded a higher RWE_{CP} value in the moderate severity (0.25 to 0.75) than forwards (0.4004 [0.3698 to 0.7069]; $\chi^2_{(1)}$=9.78 *p*=0.0018). The second half of matches recorded a higher median RWE_{CP} than the first half (0.0009 vs. 0.0007; $\chi^2_{(1)}$=402.8 *p*<0.0001) across the duration of the study (see Supplemental Table 2). Forwards recorded a higher median HIT_{SP} value (15.5 vs. 14.5; $\chi^2_{(1)}$ =9.13 *p*=0.0025) and a lower median RWE_{CP} value (0.0007 vs. 0.0008; $\chi^2_{(1)}$=281.01 *p*<0.0001) for the first half of matches when compared to backs. Hit-up forwards (15.3) recorded a higher median HIT_{SP} value over the duration of the competition when compared with outside backs (14.1; $\chi^2_{(1)}$=21.81 *p*<0.0001) and adjustables (15.2; $\chi^2_{(1)}$=197 *p*<0.001; z=-15.68 *p*<0.0001). As a result hit-up forwards (0.001) recorded a higher median RWE_{CP} than outside backs (0.0008; $\chi^2_{(1)}$=61.30 *p*<0.0001) and adjustables (0.0005; $\chi^2_{(1)}$=51.42 *p*<0.0001).

Discussion

This study reports, for the first time, the head impact biomechanics experienced by a team of women rugby league players during a domestic competition of nine matches. The sensor's reported a total of 3,003 impacts over the duration of the competition prior to the filtering process. The raw accelerometer data was transformed to the head centre of gravity utilising a rigid-body transformation for linear acceleration and a 5-point stencil for rotational acceleration.[35] A proprietary algorithm removed false impacts by comparing each impact to a reference waveform by way of cross correlation and impacts less than 10g were removed.[35] As a result of this process 45% of the impacts were removed from the dataset. A second analysis was undertaken utilizing the adjustment calculations reported by Chrisman et al.[22] to remove those linear impacts that were identified as <10*g*. As a result of these adjustments a total of 185 impacts were removed from the database. Players in the current cohort recorded an average of 14 impacts per-player per-match over the nine matches but this varied by player position and player-positional groups. The no. 8 prop (n=305) and the no. 6 Stand-Off (n=269) recorded the highest number of impacts for player positions over the duration of the study. Forwards recorded more impacts than backs, the adjustable player-positional group recorded more impacts than the hit-up forwards and outside backs and the side of the head was the most commonly reported impact location for all player-positional groups. Three players were identified by the team medic over the duration of the competition as having a concussive injury. These were later confirmed by the player's health professional resulting in a concussion incidence rate of 19.3 per 1,000 match hours.

The number of head impacts recorded over the duration of the competition (184 ±18) and per-player per-game (14 ±12) were fewer than those reported for senior men's amateur rugby union (564 per-match; 77 per-player per-match),[11] higher than those reported for junior rugby union (46 per-match; 10 per-player, per-match),[12] junior rugby league (116 per match; 13 per-player per-match),[36] youth ice hockey (5 per-player per-match)[37] and collegiate women's soccer (2.2 per-player per-game)[21] but similar to collegiate American football (171 per match).[38] The differences in the number of impacts per-match and per-player per-match in these studies may be related to the style of match participation, equipment utilised in the conducting of the playing of the game and in the rules involved. Rugby league players are limited in the type of protective equipment they are able to wear during match participation, and are required to tackle opposing players to the ground to stop their forward momentum. Although American football is similar, the tackle is different, they utilise additional protective equipment and have more players interchanging than rugby union and rugby league.

The finding that there were more impacts and higher median peak linear and rotational accelerations recorded in the second, than the first half of matches over the duration of the study was unexpected. A previous study[39] reporting on the physiological and anthropometrical aspects of elite women in rugby league players identified that they had a slower speed, agility and estimated VO2max, a greater body mass and resultant skinfold thickness when compared with other elite women sport athletes. These characteristics result in women rugby league players may having an increased risk of injury[40] and decreased skill levels[41]as a result of fatigue. This may have been the case in the current study where the players were more fatigued in the second half of matches resulting in decreased skills levels in terms of tackling, and being tackled, and there were more impacts and higher median peak linear and rotational accelerations recorded. Further research is warranted to identify the effects of fatigue in women's rugby league by match halves.

The median PLA value recorded (15*g*) was similar to the median PLA recorded for junior rugby union players and collegiate American football players (15*g*)[12, 42] but lower than the median reported for junior rugby league (XPatch: 16*g*),[36] sub-elite Australian rules football (XPatch: 17*g*),[43] American youth and high school (Head Impact Telemetry system (HITs): 22*g*) football players[30, 44] and high school (XPatch: 31*g*) and collegiate (XPatch: 32*g*) soccer players.[45] The median PRA value recorded (3,265rad/s^2) was greater than the median values reported for sub-elite Australian rules football (XPatch: 1,556rad/s^2),[43] junior rugby league (XPatch: 2,773rad/s^2),[36] and junior rugby union (XPatch: 2,296rad/s^2).[12] The median PRA in this study was higher than the median (HITs: 671rad/s^2; 1,013rad/s^2) reported in American high school football[9, 30] but similar to the 95th percentile (HITs: 2,743rad/s^2; 2,347rad/s^2).

The mean impact duration over the course of the study was 12 milliseconds. This was slightly longer than youth (8.8 ms)[33] and high school American football (8.9[8] to 10.1[23] ms) but similar to collegiate American football (14.0[38] ms) which were measured with the HIT system and New Zealand senior amateur rugby union (12.0 ms)[11] measured with an instrumented mouthguard. When viewed by player positions and player groups these varied from 11 to 13 ms. Interestingly players recorded a lower impacts duration in the first half of matches (9 to 12 ms) than the second half of matches (11 to 13 ms) which may be related to fatigue as previously discussed. Hit-up forwards recorded a lower mean impact duration (9 ms) in the first half of matches but outside backs (11 ms) recorded a lower mean impact duration in the second half of matches. Adjustables recorded the longest impact duration by match halves (12 ms vs 13 ms) and for total match (13 ms) impacts. The impact duration recorded by the HITs[30, 33] was over 40 ms which included 8ms of pre-trigger data as well as 38 ms after, the XGuard[11, 46] recorded 100 ms of data with 25 ms of pre-trigger and 75 ms after. The XPatch[22] also recorded 100 ms of data but records 10 ms of pre-trigger data as well as 90 ms after providing X, Y and Z coordinates of linear acceleration at 1ms intervals. The different impact durations recorded for the different sports may be as a result of the different pre- and post-trigger times each accelerometer has and further studies are warranted to identify if this has an influence on the recording of the impacts in time duration and resultant linear and rotational accelerations.

Total head impact distribution varied by impact location with the side of the head recording the most impacts (48%). This finding was similar to studies reporting impacts to the head for New Zealand senior amateur[11] and junior[12] rugby union, but differed compared to American high school (front of the head)[8] and collegiate (top of the head)[47] football. When viewed by positional groups, the side of the head was the most common impact location for hit-up forwards (43%) and adjustables (57%) whereas the front of the head recorded the most impacts for outside backs (43%). The difference may be reflective of the different roles of the players asoutside backs have a more open running style of match play and are often only involved in one on one tackle situations. Whereas hit-up forwards and adjustables are likely to be involved in tackle situations involving greater numbers and is reflective of the total number of impacts recorded between these different groups. The differences in player roles, tackle techniques and impacts to the head may also be a factor in the risk of concussion and future studies should consider these aspects.

The higher values reported in this study could be reflective of the activities undertaken in rugby league when compared with American football and rugby union. In rugby league when players are tackled and taken to the ground the defending team must maintain a 10 metre gap from where the ball is stopped and this known as the play-the-ball.[48] The tackle typically requires wrestling the ball carrier to the ground by two to three players and the focus is to dominate the tackle, while the ball carrier tries to maintain an upright position in an effort for a fast play-the-ball. The tackle and twist component of rugby league may have resulted in the high resultant PRA recorded in this study. There were similar high resultant PRA levels recorded in a study on senior amateur rugby union where similar tackle and twist components may have been undertaken but as there are no other studies reporting on rugby league head impact biomechanics the results reported here should be interpreted with caution. Further studies are warranted to identify the tackling differences between rugby league, rugby union and American football. Another possible factor to take into consideration when undertaking inter-study comparisons is the differences in the technology utilised.[49] For example, American football studies[9, 30, 43] have utilised the Head Impact Telemetry (HIT) system to record and report the head impacts in both match and practice situations. The HIT system does not measure rotational velocity or acceleration directly, it estimates the rotational acceleration based on the data received from the six helmet mounted accelerometers that are recording linear acceleration.[49] In contrast, the XPatch utilised in this study measures rotational velocity independently and derives the rotational acceleration from this.[49] As such, there are possible inherent differences in the measurement and calculation of rotational acceleration reported by these different systems[49] and interpretation of these results should be undertaken with some degree of caution.

The average total impact frequency burden for resultant peak linear (2,937 ±1,607g) and rotational (533,306 ±397,874rad/s^2) accelerations were similar to the median for junior rugby league (3,411 [3,351 to 3,605]g; 595,624 [585,834 to 599,359]rad/s^2)[36], but less that that reported for amateur senior rugby union (18,145 ±15,037g; 2,724,788 ±2,142,682rad/s^2)[11] and collegiate American football

(16,746*g*; 1,090,698rad/s^2).[24] However the collegiate American football study included both match and practice impacts, whereas we have reported on match participation only. Arguably concussions, or a combination of concussions and sub-concussive impacts to the head, may contribute to long term conditions such as cognitive impairment[50] and chronic traumatic encephalopathy (CTE),[51] but, to date, there has been no definitive link established. The total impact frequency burden of both resultant peak linear and rotational accelerations may play a role in the development of these disorders.[24] However, the total impact frequency burdens reported for the current cohort of players cannot be interpreted as evidence to support or refute any cause-and-effect relationship between impacts to the head and cognitive impairment. At no time during the conduct of this study were there any reported clinical symptoms of any of these conditions, as it has been reported that the signs and symptoms may not develop until later in life, or, may not develop at all. The reporting of the total impact frequency burden is to enable comparative analysis and should not be evaluated as an estimate for brain health or for physiological equivalency for impacts to the head.

By incorporating the RWE for linear, rotational and combined probability (RWE_{CP}), the variability of exposure due to linear and rotational accelerations can be identified.[30] Urban[30] undertook to report the RWE of head impact exposures in high school American football (14 to 18 yr.). The participants recorded a median value for resultant PLA of 22 g and a 95th percentile value of 62 g. This was higher than was recorded in the current study (15 g, 41g). As well, the median resultant PRA values reported were 1,013 rad/s^2 and a 95th percentile of 2,743 rad/s^2 which was lower than those recorded in the current study (2,886 rad/s^2, 9,348 rad/s^2). The differences may have been the use of helmeted sports compared with unhelmeted sports and the sensor array utilised (HITs vs. XPatch). More recently, in a study on junior rugby league participants,[36] it was reported that the median and 95th percentile of PLA (15 g, 57 g) and PRA (2,773 rad/s2, 11,384 rad/s2) was similar to the current study. When reviewed by RWE_{CP}, the median value for RWE_{CP} was lower (0.001) when compared with American football(0.194[30]) but similar to junior rugby league (0.001[36]). The differences seen here may be related to the use of the regression coefficients utilised and the exposure time differences between the different sporting codes. The use of the RWE may be beneficial but limited to same sporting codes comparisons. Further research is warranted to identify if there are differences between sporting codes regression coefficients and, if the regression coefficients utilised are appropriate for non-helmeted sporting activities.

A limitation to this study was not having multi-angled video footage of the matches to enable correlation between the head impacts recorded and physical contacts during match participation. Although the first few matches were videoed with a hand-held camcorder (Sony HDR-PJ540 camcorder) standing on the sideline, the quality was poor and it was not possible to identify which player was tackled and whether the contact was to the body, or when the body impacted with the ground, particularly in field positions that were on the other side of the field from the camera. The use of the camcorder was stopped after the first two matches as the comparison was limited to impacts that occurred within a 20-m range of where the person with the camcorder was standing. As such, we were unable to establish whether the impacts were from body contact or from contact with the ground and hence, the results must be interpreted accordingly. Future head impact studies should use high quality multiple angled cameras in an elevated position to enable verification of the impacts recorded.

Coupling between the skull and the XPatch can affect the linear and rotational acceleration transformation to the centre of the gravity of the head. In evaluating wearable head impact sensors, it was reported[18] that the XPatch over-predicts PLA by 2,500 ±1,200 rad/s^2 and this can result in high levels of error. However, the studies[15, 22] undertaken to validate the XPatch have been done on head-forms requiring the impact to be from various angles to an upright stationary object enabling the differences between the sensor and the head-forms centre of gravity to be identified. Lennon[18] utilised mild impacts from a soccer ball launched from a ball launcher to the head of a test participant at approximately 7 m/s wearing several different sensors to determine a correlation between the sensors. These studies showed different results for the measurement of the PLA and PRA with some,[18] but not all[15, 22] studies, reporting an over-prediction of the accelerations recorded when compared with the heads centre of gravity. This may have been the situation in the current study with the resultant high PRA results reported and these should be interpreted cautiously The manner of participation in contact sports such as rugby union, rugby league and Australian Rules football require the players to tackle the opposition player to the ground in a twisting manner in order to dominate the tackle situation where as the contact and tackle situation in other sports such as Lacrosse and American football do not require this to occur. These types of tackle / contact situations are not replicated in the testing validation environment and may contribute the high PLA recording reported in this study. Further research is warranted to evaluate the XPatch, and other wearable sensors, in an environment where helmetless contact sports conditions such as the tackle situation are replicated.

Conclusion

Female rugby league athletes were exposed to repetitive sub-concussive head impacts with an average of 14 significant impacts per-player per-match. Forwards were exposed to significantly more impacts per-match than backs and these impacts tended to be of greater magnitude. Most impacts occurred on the side of the head and were sustained during the second half of the game.

Practical implications

Clinicians, coaches and players should be aware of the rates and magnitude of head impacts in female rugby league athletes. Until the effects of such impacts are understood, training, fitness and technique should be optimised to limit the burden of repetitive head injuries. Awareness of these risks should allow recognition and optimal management of these athletes in order to reduce any possible deleterious concussive injury.

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Tables

Table 1: Impacts to the head greater than 10*g* in women's rugby league for total impacts recorded, impacts per match-half and impacts per-player positional group over one season of matches for total impacts, impacts per-match, impacts per-player per-match, impact duration, resultant peak linear and peak rotational accelerations. Data are presented as median [25-75th interquartile range], 95th percentile and total frequency impact burden.

Table 2: Impacts to the head greater than 10*g* in women's rugby league by player position over one season of matches for total impacts, impacts per-player per-match, impact duration, resultant peak linear and peak rotational accelerations. Data are presented as median [25-75th interquartile range], 95th percentile and total frequency impact burden.

Table 3: Impacts to the head greater than 10*g* in senior amateur rugby league by player positional group over one season of matches for impact location, total impacts, impact duration, resultant peak linear and peak rotational accelerations. Data are presented as median [25-75th interquartile range], 95th percentile and total frequency impact burden.

Supplementary Tables

S Table 1: Impacts to the head greater than 10 g in women's rugby league for total impacts recorded and impacts per-player position and per-player group[14] over a competition match season for injury tolerance level,[15, 32-34] impact severity limits,[26-28] head impact telemetry severity profile[29] and risk weighted cumulative exposure (combined probability)[30] by total impacts recorded, percentage of impacts recorded (%),and median [25th to 95th interquartile range].

S Table 2: Impacts to the head greater than 10 g in women's rugby league for total impacts recorded, impacts per match-half over one season of matches for Head Impact Telemetry severity profile[29] and Risk Weighted Cumulative Exposure (combined probability)[30]. Data are presented as median [25-75th interquartile range] and 95th percentile.

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