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Original Research

History of concussion is associated with higher head acceleration and reduced cervical muscle activity during simulated rugby tackle: An exploratory study

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ABSTRACT

Objectives: The objective of this study is to explore the effect of concussion history on head control during front on rugby tackles in non-professional male and female Rugby Union players.
Design: An observational cross-sectional design.
Setting: A laboratory-based study of simulated front on rugby tackles with a dynamic relatively-weighted (~50% body-mass) tackle bag.
Participants: 27 club Rugby Union players were recruited; 10 (7female) with no-concussion history, 10 (3female) within 12-months of concussion and 7 (3female) with 24 + months since last concussion.
Main outcome measures: Linear and rotational head acceleration measured with a skin mounted CSx® triaxial accelerometer. Normalized Surface EMG amplitude of the bilateral sternocleidomastoid (Scm), upper trapezius (Trap) and splenius capitis (Spl). All outcome measures were synchronised with moment of impact with the tackle bag.
Results: A mixed-model analysis showed that players with 12-Month concussion history had the highest head acceleration (females = 48.6g, males = 68.3g, $p < 0.05$) with lower Trap (6.9–11.7%, $p < 0.05$) and Spl (3–12%, $p < 0.05$) amplitudes compared to athletes with No-concussion.
Conclusions: These findings suggest probable disruption to neuromuscular control of the head in athletes with concussion history as a potential mechanism for recurrent concussion incidence.

1. Introduction

The incidence of sports related concussion rugby union is one of the highest across all full contact sports (Gardner, Iverson, Williams, Baker, & Stanwell, 2014; Kemp et al., 2015). In non-professional rugby players the incidence has been identified as 19.1 per 1000 h (King, Hume, Gissane, & Clark, 2017). With respect of time lost due to injury, concussions accounted for 55% of total days' absence (Kemp et al., 2015; Kemp, Hudson, Brooks, & Fuller, 2008). During match play the tackle has been accounts for 50% of all concussions (Cross et al., 2017; Fuller, Taylor, & Raftery, 2015). With up to 70% of those occurring to the player initiating the tackle (Cross et al., 2017). Cross et al. (2017) identified several significant risk factors related to the tackler including head-to-head contact and magnitude of tackle entry acceleration of both players while Hendricks et al. (2015) identified poor tackle technique as a critical mechanism for concussion injury in under 18 players (Cross et al., 2017; Hendricks et al., 2015).

One of the most significant risk factors for sports related concussion is a previous history of concussion (Abrahams, Fie, Patricios, Posthumus, & September, 2014; Hides, Franettovich Smith, Mendis, Treleaven, et al., 2017; Schmidt et al., 2018). Hollis et al., found that non-professional rugby players who sustained one concussion were 1.75 times more likely to suffer another during the following 12 months while those with 2 or more concussions were 1.65 times more likely to suffer another concussion during the same season (Hollis et al., 2009). Studies have shown that recent

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concussion history may significantly impact the sensorimotor system (Hides, Franettovich Smith, Mendis, Smith, et al., 2017; Hides, Franettovich Smith, Mendis, Treleaven, et al., 2017) which may place athletes at higher risk for other musculoskeletal injuries such as lower extremity injury up to 24 months post-concussion (Kardouni, Shing, McKinnon, Scofield, & Proctor, 2018). Given the impact concussions can have on a participants overall health, cognitive and psychological function concerted effects are being directed towards primary prevention (Alexander, Shuttleworth-Edwards, Kidd, & Malcolm, 2015; Register-Mihalik & Kay, 2017) and understanding the mechanics of these injuries (Eckner, Oh, Joshi, Richardson, & Ashton-Miller, 2014; Mihalik, Bell, Marshall, & Guskiewicz, 2007) and the impact injury may have on neuromuscular control of the neck (Schmidt et al., 2014). Yet, there is a lack of research on how concussion history may affect head impact mechanics in rugby players during tackle situations. Neuromotor alterations may affect the player's ability to control the head in subsequent tackle situations leading to further risk of concussion.

Whilst there has been very little data related to concussion risk in female rugby players, a recent review has shown female hockey and football players had higher concussion risk compared to their male counterparts (Prien, Grafe, Rössler, Junge, & Verhagen, 2018). The concussion disparity between males and females may be associated with typical sexual dimorphism resulting in anatomical and biomechanical differences. Investigation of gender differences relating to dynamic stabilization of the head-neck region in response to an external force application has reported greater head-neck angular acceleration in the female participants (Mansell, Tierney, Sitler, Swanik, & Stearne, 2005). The authors attributed this finding to decreased levels of cervical strength, neck girth and head mass in the female participants, resulting in reduced head-neck segment stiffness during impact events when compared to males (Tierney et al., 2005). Thus, it would be plausible to expect head mechanics to differ between genders during rugby tackle situations.

Therefore, the primary aim of this study was to explore the relationship between concussion history and inertial head acceleration during simulated rugby tackles in non-professional club level male and female players. We hypothesise that head acceleration would be elevated in athletes with a recent history of concussion and relative to impact momentum, females would experience greater peak head acceleration compared to males. We further hypothesise that anticipatory cervical muscle activity would be reduced in those with history of prior concussion.

2. Methods

2.1. Participant selection

An observational cross-sectional study design was used to determine the association between head acceleration experienced during a simulated front-on rugby tackle and concussion history in university aged male and female amateur club rugby union players. Ethical approval was obtained from the University Human Research Ethics Committee, reference number H15/050.

A sample of convenience of thirteen female and fourteen male participants were recruited from local rugby clubs. Participants were included if on the day of testing they were deemed medically fit and available for selection and capable of making seven consecutive tackles on their preferred tackling shoulder. The exclusion criteria for the study consisted of a concussion diagnosis within the past three months (regardless of clearance to play), any long term post-concussive symptoms that the participant was currently experiencing, or any injury to the head or neck region within the past three months, that may hinder tackle performance or be exacerbated by participation in the study.

Concussion was operationally defined as those that had been diagnosed by a medical professional. Based on their concussion history participants were classified using the following criteria: (a) No-concussion: were players with no history of diagnosed concussion, (b) 12-Months: players who had their most recent concussion within the past 12 months and were medically cleared to play, (c) 24 + Months: players with most recent diagnosed concussion at least 24 months prior to recruitment.

A questionnaire was used to collect demographic information which captured the following: age, sex, height, weight, years of club rugby played, tackle shoulder preference, total number of diagnosed concussions (Table 1). Players were asked to give the date of their most recent concussion diagnosis which was used to classify concussion athletes as above. All playing positions were represented equally across groups, thus, we did not present this data in the table.

2.2. Equipment

Each participant was fitted with a CSx® triaxial accelerometer (CSx Systems Ltd, Auckland, New Zealand) positioned over the right mastoid process using Fixomull® stretch tape (5 cm × 10 m) and sampled at 3200 Hz for linear and 1000 Hz for rotational accelerations. The accelerometer used on the male participants collected linear acceleration only while the one on the females collected both linear and angular acceleration data. The accelerometer data was acquired through CSx® proprietary wireless protocol to a base station and saved to computer. Data recording was triggered when the impact event exceeded the 10 g threshold, which eliminated incidental contact noise. The system recorded data for 5 ms prior to the impact event and 45 ms afterward (Salmon, Handcock, Sullivan, Rehrer, & Niven, 2017a,b).

The muscle activity was recorded from three bilateral muscle pairs, of the upper trapezius (RTrap & LTrap), splenius capitis (RSpl & LSpl) and the sternocleidomastoid (RScm & LScm) using an eight-channel telemetry EMG system (Noraxon telemyo 900, Noraxon

Table 1
Mean participant demographic data (standard deviation).

	NO-CONCUSSION	12-MONTH	24 + MONTH	P-VALUE	
AGE (YRS)	21.0 (2.5)	21.0 (1.2)	21.1 (0.90)	0.921[a]	0.956[b]
BMI (KG/M^2)	26.14 (1.7)	28.91 (4.32)	29.03 (3.18)	0.135[a]	0.060[b]
WEIGHT (KG)	77.26 (14.88)	93.15 (17.55)	90.55 (9.13)	0.090[a]	0.089[b]
BAG-TO-WEIGHT (%)	57 (7)	51 (9)	51 (6)	0.154[a]	0.091[b]
YEARS PLAYING (YRS)[d]	3.55 (2.8)	3.63 (1.4)	4.14 (1.5)	0.867[a]	0.592[b]
MALE (N)	3	7	4	0.183[c]	
FEMALE (N)	7	3	3		

[a] Independent t-tests between No concussion and 12-Month groups.
[b] Independent t-tests between No concussion and 24 + Month groups.
[c] Chi-square results $X^2(2, N = 27) = 3.310$ show that sex was equally distributed within the population.
[d] Indicates number of years playing at highest club level.

USA inc.) collecting at 1000 Hz. Electrode placement and normalisation procedures followed SENIAM guidelines (Hermans et al., 1999). Prior to electrode placement the skin was prepared at each specific site, by shaving with a razor, lightly abrading with fine sandpaper and cleansing using 70 percent isopropyl alcohol swipes. Two 30 × 20 mm disposable Ag/AgCl electrodes (Ambu® Blue Sensor N) were placed 20 mm apart over each muscle belly. Maximum voluntary contraction elicited against manual resistance was utilised for EMG normalisation. For all MVC testing, the participant was seated in a neutral position with feet on the floor and relaxed, 2 s of resting baseline data was collected, after which, three MVC trials of 5 s duration was collected for each of the three pairs of muscles. A randomised order of muscle testing was used for each participant with a 30 s rest period following each trial. The TRP manual test followed SENIAM guidelines utilising resisted shoulder elevation and posterolateral head extension. For the SPL test, the participant was required to rotate the head to the ipsilateral side and extend against manual resistance (Almosnino et al., 2009; Criswell, 2010). The SCM was tested as per Tartaglia et al. where the participant was required to rotate the head to the contralateral side and flex against resistance (Tartaglia, Barozzi, Federico, Cesarani, & Ferrario, 2008).

A punching bag (length 1.25 m and circumference 1.09 m) was used for participants to perform a simulated tackle. The bag was instrumented with an AS-20B 20 g uniaxial accelerometer (KYOWA, Tokyo), used to identify moment of impact. The dynamic weighted punching bag was designed to mimic the effective mass of a ball carrier without considering limbs, simulating a rugby tackle situation as closely and as ethically as possible. Female and male participants used 40 kg and 50 kg bags respectively. The bag weights were chosen to represent approximately 50% body mass based on sex specific data identified in the literature (Smart, Hopkins, & Gill, 2013). The swing arc of the bag was standardised to ensure similar bag velocity at the moment of impact for each tackle (Fig. 1).

To calculate tackle entry momentum, each tackle simulation was recorded using a high-speed camera (Edgertronic SC1, Sanstreak Corporation, USA) and a Tamron AF 20–40 mm F2.7–3.5 SP Nikon Mount Lens, capturing at 500 Hz to provide a reference point to confirm impact timing and calculate the athletes linear tackle entry velocity (averaged over 10 frames prior to impact). A trigger was used to synchronise the timing of the video, EMG recording and bag accelerometer. Kinovea (version 0.8.15, Joan Charmont & Contrib., France) was used to determine the tackle entry velocity of each participant.

2.3. Procedure

The participants were instructed to make a front-on tackle with their preferred tackle shoulder in the tackle bag target zone (representing a player's navel to mid-thigh region), without taking the bag to the ground. The bag was held at a 1.6 m radius from the expected impact point, through a pulley to a stationary position (Fig. 1). The player was instructed to make contact with the bag as it reached its vertical position, there was a line marker on the floor to aid targeting. The player was then given a 5 s countdown to begin their 3 m run-up prior to the bag release. Participants were given three familiarisation trials performed at sub-maximal pace to acquaint themselves with the timing of the tackle. After familiarisation, each participant performed seven maximal effort tackles on their dominant shoulder, with a 2-min rest period between each tackle (Salmon et al., 2017a,b).

2.4. Data processing

Raw data, both EMG and acceleration, were imported into Matlab® (version R2015b, The Mathworks inc, Natick, MA, USA). Processing of the acceleration data included initial examination of the frequency spectrum using a Fast Fourier Transform (FFT). The appropriate cut-off for the low pass filter was determined to be 160 Hz for the linear signal and 100 Hz for the rotational signals. A low pass forth order Butterworth filter was used to filter the signals. Pre and post signal filter figures were visually inspected to ensure the appropriateness of the filter and preservation of the impact data topography. Peak resultant linear and rotational acceleration values were extracted for statistical analysis. Duration of the impact was determined by visual inspection of the period of the head acceleration waveform form and manual selection of the baseline time points either side of the peak waveform on the Matlab figures facilitated by graphic user interface.

EMG signals were bandpass filtered between 16 and 500Hz, amplified with gain 1000, and a 49–51 Hz band stop filter was applied to remove electrical noise. The data were baseline corrected, rectified, smoothed using a 10 HZ low pass (4th order Butterworth) filter and normalized to the respective maximum voluntary contraction (%MVC). EMG amplitude was calculated for each muscle in twenty-four 25 ms epochs from 300 ms before to 300 ms after tackle bag impact utilising the trapezoid method. The moment of impact with the tackle bag (T_0) was identified from the bag accelerometer via a computer algorithm. To examine the magnitude of preparation for impact we utilised the average of 5 epochs over 125 ms (4 epochs up to impact and 1 post impact e.g., Fig. 2). These 5 epochs represent the anticipatory postural adjustment period associated with the impact (Santos, Kanekar, & Aruin, 2010).

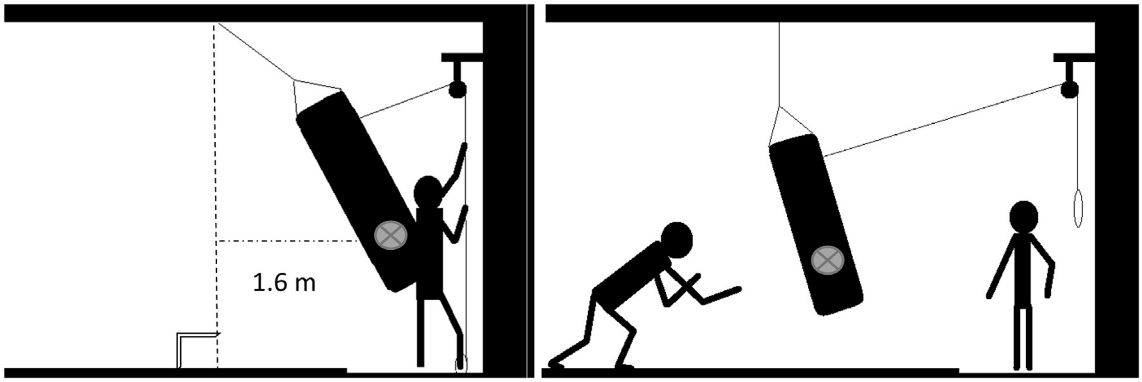

Fig. 1. Representation of the tackle simulation, the researcher controls the bag release and impact occurs when bag is 90° to the horizontal.

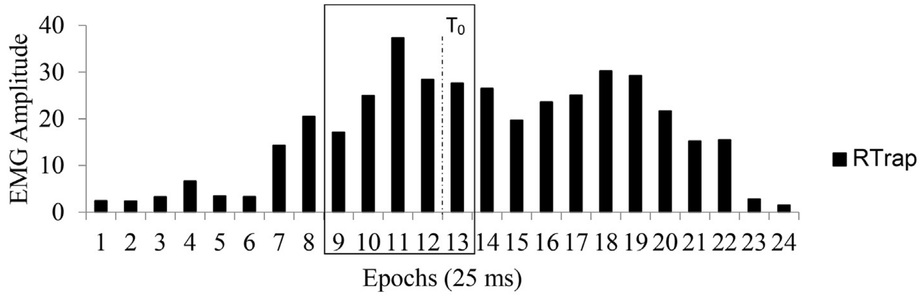

Fig. 2. Example of the muscle epochs for one participant for the right upper trapezius muscle (RTrap). The box surrounds the epochs of interest and the dashed line represents T_0.

2.5. RTrap

The statistical analysis was performed using SPSS® software (version 26, IBM® Corporation, USA). Independent t-tests were used to examine the participant demographics. Linear mixed models were fit to the data with Trials as repeated measures. “Concussion History” was entered as a categorical variable with three levels based on time since last reported concussion; No-concussion (n = 10), 12-Months (n = 10) and 24 + Months (n = 7). Separate models were run for each tackle characteristics including velocity, momentum, linear acceleration and each muscle amplitude. Each tackle characteristics were entered as a continuous outcome with fixed effects of Concussion History (3 Levels) and Sex (male, female) and Participant was incorporated as random effects and an unstructured covariance matrix was established. Where significant interactions were found, pairwise comparisons were made utilising least significant difference adjustment for multiple comparisons. Only the females provided rotational data, thus, the rotational acceleration model was run with the female data only, and had just one fixed effect (Concussion History) and one random effect (Participant). All statistical decisions regarding significance were based on Alpha <0.05.

3. Results

3.1. Descriptive statistics

Independent t-tests were used to examine descriptive statistics (Table 1) and no significant differences were noted between groups. There were multiple athletes with recurrent concussion histories, 2 females and 4 males had three or more diagnosed concussions. Twenty-five players preferred to tackle with the right shoulder, the two who preferred the left side were No-Concussion females.

3.2. Tackle characterises

The mean tackle characteristics by group are displayed in Table 2. For linear head acceleration there were significant main effects of concussion history ($F_{(2,119)} = 19.41$, $p < 0.001$) and Sex ($F_{(1,119)} = 9.36$, $p = 0.003$). Significant Sex effects were noted for peak impact duration ($F_{(1,119)} = 72.025$, $p < 0.001$), tackle entry velocity ($F_{(1,119)} = 11.08$, $p < 0.001$), and tackle entry momentum ($F_{(1,119)} = 45.53.08$, $p < 0.001$). The momentum of the bag just prior to impact averaged 129.38 kg m/s for females and 198.09 kg m/s for males. Pairwise comparisons displayed in Table 3 show the confidence intervals of the differences in these characteristics between groups within-sex. All confidence intervals include 0 apart from the intervals for head acceleration (Table 3).

As indicated in Table 2 there was a significant interaction between Sex and Concussion history in linear head acceleration. Pairwise comparisons in Table 3 show that there was a significant difference between those with 12-Month and those with No-concussion history. Mean differences within females between No-concussion and 12-Months 44.9 g (29.4–55.2 g; $p < 0.001$). The mean difference for males between No-concussion and 12-Months was 27.2 g (13.9–40.2 g; $p = 0.014$) (Table 3). Further, there was a significant difference between males and females with No-concussion with a mean difference 20.01 g (6.7–33.4 g; $p = 0.004$).

Mean rotational acceleration values are displayed in Table 3. Significant main effects ($F_{(2,62)} = 14.43$, $p < 0.001$) were found at all levels of Concussion History (Table 3). Peak rotational acceleration is significantly higher for 12-Months compared to No-Concussion by 4760.1 rads/s^2 (2519.7–7000.5 rads/s^2; $p < 0.001$) and 24 + Months was significantly lower than 12-Months −2925.3 (−5360.3 to −490.3, $p = 0.016$).

3.3. Muscle activity

Mean muscle magnitudes for the 125 ms pre-impact are presented in Fig. 3. There was a significant effect of concussion history within Sex across muscles, the mean differences per muscle by Sex is presented in Table 4. The muscles with the largest mean differences were the RTrap & LTrap and the LSpl. Mean muscle amplitudes were significantly lower in the 12-Month males (Table 4).

Table 2
Mean tackle characteristics with standard deviation (SD) by history of concussion.

History	Sex (N)	Mean Peak Accel (g)*		Impact Peak Duration (ms)		Tackle Entry Velocity (m/s)	Tackle Entry Momentum (kg m/s)	
NC	F (7)	18.49	7.31	23.04	8.72	3.61 0.55	250.71	55.73
	M (3)	38.69	14.74	11.30	2.93	4.53 0.59	421.98	64.15
12 months	F (3)	48.61	18.28	21.10	8.35	3.19 0.06	267.98	76.62
	M (7)	68.71	20.77	11.43	2.95	5.0 0.70	491.81	58.83
24 + months	F (3)	36.52	20.72	21.01	10.33	3.45 0.73	297.91	59.53
	M (4)	48.68	20.68	12.72	3.00	4.93 0.36	463.28	82.40
Group Effect		<0.001		0.967		0.919	0.336	
Sex Effect		0.003		<0.001		<0.001	<0.001	

*Significant interaction between sex and group.

Table 3
Pairwise comparisons of tackle characteristics. Mean differences and 95% confidence intervals for each of the tackle characteristics within sex. The p-values have been adjusted for multiple comparisons using LSD and stars (*) represent significant difference. Positive values represent greater values in the No-Concussion group.

Sex	Tackle Characteristics	No-Concussion – 12 Months				No-Concussion – 24 Months			
		Mean difference	95% CI		Adj p value	Mean difference	95% CI		Adj p value
Males	Peak Accel (g)	−27.18	−40.19	−13.93	0.014*	−10.61	−24.35	3.14	0.129
	Velocity (m//s)	−0.48	−1.35	0.37	0.371	−0.39	−1.28	0.51	0.37
	Momentum (kg m/s)	−69.82	−170.05	30.40	0.161	−41.29	−146.11	63.52	0.419
	Impact Duration (ms)	−2.01	−9.16	5.16	0.586	−2.62	−10.25	5.00	0.478
Females	Peak Accel (g)	−44.97	−55.63	−29.31	0.001*	−16.66	−37.33	4.00	0.113
	Peak Rot Accel (rads/s^2)	−4760.09	−7000.48	−2519.79	0.001*	−1834.77	−3641.98	−27.63	0.047*
	Velocity (m//s)	0.43	−0.41	1.25	0.297	0.16	−0.63	0.99	0.694
	Momentum (kg m/s)	−17.27	−114.32	79.76	0.713	−47.21	−144.25	49.84	0.320
	Impact Duration (ms)	2.49	−4.87	9.85	0.485	1.55	−5.82	8.92	0.664

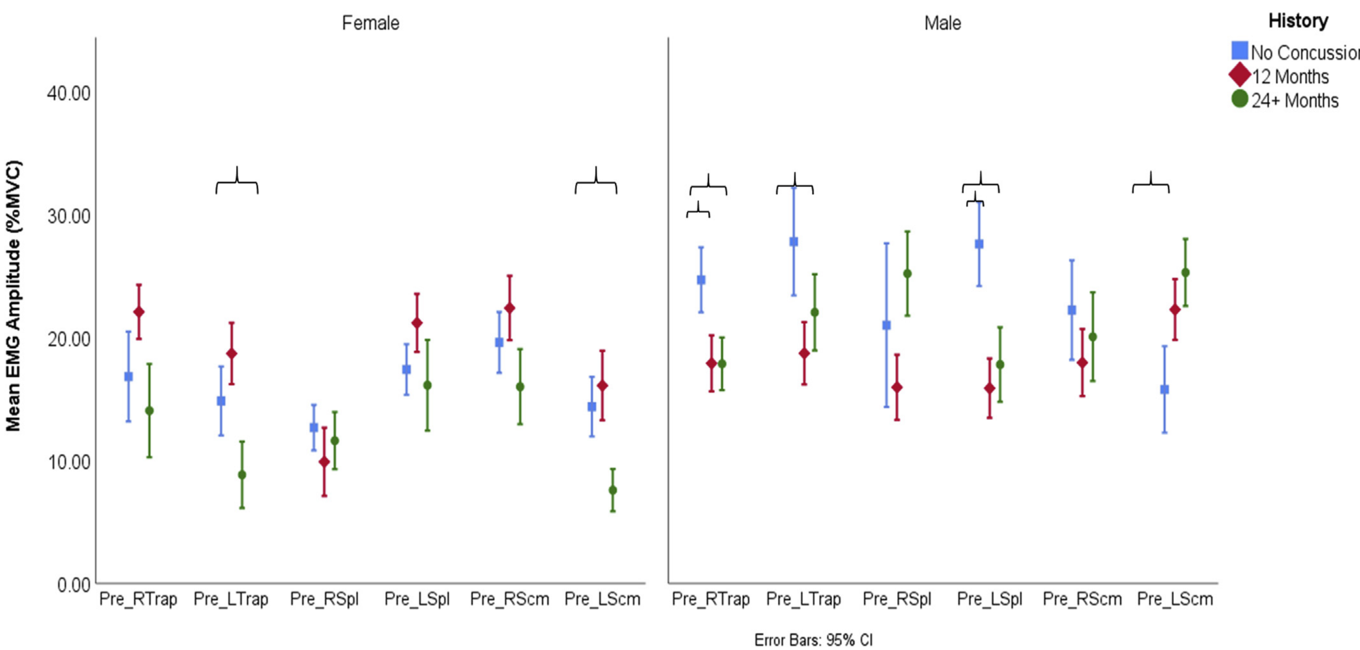

Fig. 3. Mean muscle amplitude normalized to % MVC for 125 ms pre-impact with the tackle bag. Error bars represent the 95% CI of the mean; brackets indicate adjusted pairwise significant differences ($p < 0.05$).

Table 4
Between group comparisons of mean muscle activity (%MVC) 125 ms pre-impact. Mean differences and 95% confidence intervals for each muscle within sex. The p-values have been adjusted for multiple comparisons using LSD and stars (*) represent significant difference. Positive values represent greater amplitude in the No-Concussion group.

Sex	Muscle	No Concussion – 12 Months				No Concussion – 24 Months			
		Mean difference (%mvc)	95% CI		Adj p value	Mean difference (%mvc)	95% CI		Adj p value
Males	RTrap	6.89	2.41	11.35	0.003*	6.14	1.85	11.73	0.007*
	LTrap	11.73	4.48	18.27	0.002*	6.44	−0.97	14.5	0.085
	RSpl	3.67	−5.00	12.35	0.404	−4.92	−14.67	4.054	0.295
	LSpl	12.35	5.63	19.08	>0.001*	10.56	4.99	17.77	0.005*
	RScm	−1.71	−6.89	3.28	0.499	−1.28	−6.16	3.61	0.606
	LScm	−5.61	−12.14	1.10	0.098	−9.13	−16.33	−1.92	0.014*
Females	RTrap	−3.50	−7.83	0.82	0.112	3.60	−3.89	10.31	0.360
	LTrap	−4.02	−11.58	3.287	0.251	8.04	3.32	12.35	>0.001*
	RSpl	3.26	0.6	8.0	0.175	−1.57	−3.29	3.16	0.514
	LSpl	−3.45	−7.36	0.46	0.083	1.75	−2.33	5.83	0.398
	RScm	−2.91	−7.79	−1.98	0.241	−2.16	−7.05	2.72	0.382
	LScm	−1.63	−5.74	2.48	0.434	7.67	0.69	14.65	0.032*

Comparatively, there were fewer differences across groups within the Females with just the LTrap and LScm being significantly lower in the 24 + Month group (Table 4).

4. Discussion

The primary purpose of this study was to examine the relationship between concussion history and head acceleration in simulated rugby tackles. Our hypothesis that athletes with most recent history of concussion would have increased head acceleration compared to those with no history of concussion, was supported. However, our hypothesis that females would have greater relative head acceleration compared to males was not supported. While our third hypothesis regarding preparatory muscle activity, was partially supported in that there were significant differences in muscle amplitude between No-concussion and those with history of concussion but these differences appeared to be sex specific.

Our research indicates a significant higher head acceleration in athletes within 12 months of their most recent concussion. This finding was similar for both males and females, indicating it is not likely associated with the tackle entry momentum or velocity, which differed significantly between sexes but not with respect to concussion history (Tables 2 and 3). Thus, for given tackle entry momentum the athletes with recent history (within 12 months) of concussion had higher head accelerations (49–68 g) compared to their No-concussion counterparts (18–38 g). While there is some difficulty in determining the threshold of head acceleration that may be safe for athletes (McCrory et al., 2017), in general, brain strain injury is known to be associated with the magnitude and duration of head acceleration (Gennarelli, 1983, pp. 1–13; Gurdjian, Lissner, Webster, Latimer, & Haddad, 1954; Post, Blaine Hoshizaki, Gilchrist, & Cusimano, 2017). For instance, McIntosh et al. (2014) identified concussion incidents in male Australian football players associated with an average linear head acceleration of 103.4 g compared to non-concussive impacts which averaged 59.0 g. Although, we acknowledge that inertial head accelerations measured during the tackle simulations in the current study differed from the situations observed by McIntosh, who identified head accelerations associated with direct head impacts. We would still assume a greater risk for brain strain injury in athletes with higher head acceleration for a given impact.

Athletes with history of concussion also had significantly reduced preparatory activation of the cervical muscles prior to impact. Although, this finding was not uniform between males and females nor across muscle groups. Males had a greater number significant findings between concussion and No-concussion groups with the concussion athletes having reduced amplitude in the Trap and Spl muscles. Yet, the male concussion athletes had higher muscle activity in the left Scm. Contrarily, there were fewer differences between concussion and No-concussion females in muscle amplitude with just the LTrap being lower in the 24+-Month group. Increasing muscle pre-activation has been shown to increase joint stiffness and stability (Cholewicki, Simons, & Radebold, 2000; Gardner-Morse & Stokes, 2001; Simoneau, Denninger, & Hain, 2008). Greater neck stiffness has also been linked to reduced head acceleration (Eckner et al., 2014; Tierney et al., 2005). Further, Schmidt et al., found that greater cervical muscle stiffness combined with lower head acceleration during head perturbations significantly reduced concussion risk in American Football players (Schmidt et al., 2014). Because lower neck stiffness has been associated with concussion we hypothesised that we would find reduced neck muscle activity in athletes with history of concussion. This hypothesis was only partially supported because the findings were less clear in the female athletes. We were surprised by this because Tierney et al. (Tierney et al., 2005) found females exhibited higher head accelerations and lower neck stiffness compared to their male counterparts during external head perturbations. However, the female population in the current study, trained rugby players, differed significantly from that of Tierney et al., who were described as “physically active”. Further, the head perturbation mechanisms differed significantly. Clearly the central nervous system control strategy for tackle related perturbation differs from that of the external perturbation but the tackle perturbation, while less controlled, is more ecologically valid when examining concussion mechanisms.

One of the most significant risk factors for sports related concussion is a previous history of concussion (Abrahams et al., 2014; Hides, Franettovich Smith, Mendis, Treleaven, et al., 2017). According to Hollis et al., rugby players who have suffered a concussion may have a 75% higher risk of a second concussion incident within 12 months of their initial injury (Hollis eta l., 2009). Although the mechanism of increased injury risk after sports related concussion is not known (Hides, Franettovich Smith, Mendis, Treleaven, et al., 2017). The results of the current study would support the hypothesis that recent concussion history may impact head control in a way that may increase the risk of concussion in Rugby players. Particularly, the males were found to have significantly increased head acceleration associated with a decrease in the amplitude of muscle activation in the upper trapezius and splenius muscles. It stands to reason that players with decreased amplitude of muscle activity prior to impact would have decreased neck stiffness compared to those with higher magnitudes of activation, thus may be less capable of attenuating an impulsive load to the head and neck in a tackle situation. Moreover, anticipatory activation of the cervical muscles is seen as an independent factor from muscle strength in attenuating head impulsive loading (Eckner et al., 2014). There is a growing body of evidence identifying the connection between concussion and altered neuromuscular control (Gilbert, Burdette, Joyner, Llewellyn, & Buckley, 2016; Hides et al., 2016; Hides, Franettovich Smith, Mendis, Smith, et al., 2017; Lynall, Mauntel, Padua, & Mihalik, 2015). In a rugby tackle situation, feed-forward control of the neck muscles allows a player to anticipate the external forces of a collision, and initiate an appropriate muscular response (Falla, Rainoldi, Merletti, & Jull, 2004; Lee & Terzopoulos, 2006, pp. 1188–1198). The splenius capitis, sternocleidomastoid and the upper trapezius have been identified as the primary dynamic stabilizers of the head and neck region during impulsive loading situations (Kumar, Ferrari, & Narayan, 2005) and reduced neuromuscular function in these muscles may lead to poorer head control and place players at greater risk for further concussion injury.

To our knowledge this is the first study to examine head acceleration and muscle activity in female rugby union players during simulated tackles. The accelerations, both linear and rotational recorded in the No-concussion females were within the range of values reported from female rugby league (a similar collision sport) players during game play (King, Hume, Gissane, Kieser, & Clark, 2018). However, females within 12 Months of their last concussion had significantly larger head accelerations, particularly rotational accelerations. McIntosh et al., identified concussion incidents in male Australian football players associated with rotational acceleration of 7951 rads/s^2 compared to non-concussive impacts which averaged 4300 rads/s^2. If we compare our rotational acceleration findings to those of McIntosh et al. our athletes may have been at risk of brain strain injury during this study (McIntosh et al., 2014). The current opinion is that rotational acceleration may have greater potential for causing brain strain related injuries. Recent research by Post et al. (2017) show a complex relationship between peak linear and rotational accelerations on principal strain in the brain which differ depending on the duration of impact (Post et al., 2017). Their findings show that linear acceleration components contribute to brain strain only when the duration of the impact is shorter than 10 ms. When duration of impact was longer than 10 ms strain could be contributed almost entirely to rotational acceleration components. They estimate the potential concussion risk at accelerations above 3500 rads/s^2. Our small sample of female athletes within 12 Months of concussion averaged 8265.9 rads/s^2. The rotational data for females with history of concussion was significantly larger than the non-concussive accelerations reported by King et al. (2018) which ranged between 1864 and 4545 rads/s^2. The impact duration in the current study (21 ms) was also longer than those reported by King et al. (12 ms), it is unknown how the increased duration of impact may affect peak rotational values. On a final note, similar skin patch sensors have been shown to over predict rotational (2500 ± 1200rads/s^2) acceleration due to skin artefact during impact events (Wu et al., 2016). However, even if we adjust our values for such over prediction we would still have rotational accelerations that are potentially above injury threshold in this population.

4.1. Limitations

This study has a number of inherent limitations. We interpret all data with caution due to the relatively small sample size. Further, due to the small sample size we were compelled to delimit our muscle investigation, but we believe our reported data are compelling enough to warrant further work in this area. Further, although the data are compelling, we cannot say with certainty the recent history of concussion is directly responsible for the higher head accelerations experienced in these athletes, this is a general limitation of cross-sectional study designs. Not having rotational data for males is also a limitation as this would give more insight to the potential gender differences. Further work will be conducted in this area. Finally, impact sensors attached to the skin have been shown to overestimate head accelerations in head impact studies. Certainly, this may affect the peak acceleration values, but one could assume that each study participant would be affected similarly. Therefore, we would not expect this overestimation to have a significant effect on the differences between the groups.

5. Conclusions

This preliminary study showed that rugby players who have suffered a concussion within 12 months have significantly higher head accelerations and reduced cervical muscle activation than those with No-concussion history while performing simulated front-on tackles with their preferred shoulder. Moreover, muscle activation was lower for some muscles in the athletes with 24 + Months since last concussion, indicating the need for further investigation into the effect of concussion on muscle activation and reflexes. The findings provide insight and direction for future research into a potentially important risk factor for recurrent concussion in athletes with history of concussion.

Conflicts of interest

None declared.

Ethics approval and consent to participate

This study was approved by the University of Otago Ethics Committee (reference number H15/050). All participants will sign an informed consent prior to taking part in the study.

Availability of data and material

The datasets from this study will be substantial. Potentially too large to post to online databases, however, dataset used and/or analysed during the current study are available from the corresponding author on reasonable request.

Authors' contributions

MDB was responsible for the design of the study and is the guarantor. MM, JP, RK aided in data collection, NA and JR aided in data processing and wrangling, DS provided guidance on the design and analysis planned for this study. All authors had input into revision of the manuscript for important content and approved the final version.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.ptsp.2019.03.012.

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