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Sex Differences in Head Impact Magnitude, Neck and Head Size and Neck Strength in University Rugby Union

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Abstract

Concussion has consistently been reported as the most significant injury in rugby union and is an area of increasing concern. Female athletes are reported to suffer from a greater concussion incidence and worse outcomes than males. Increased neck strength has been associated with a reduction in concussion risk and requires further investigation. The aim of this thesis was to investigate sex differences which may affect brain injury susceptibility, primarily neck strength.

The magnitude of head acceleration during impact events was recorded by instrumented mouthguards. University first team rugby players (31 male and 22 female) were measured university for 13 and seven competitive matches respectively. All impacts were video and waveform verified and impact kinematics classified. Anthropometrics and isometric neck strength were measured prior to the season beginning.

Male players had significantly larger head, neck and shoulder anthropometrics than female players, as well as significantly greater neck strength in all four directions. Positional differences in size and strength were much more prominent in males than females. Head impact magnitude was found to be similar in both sexes, despite the significant differences in size and strength. Negative correlations for peak rotational head acceleration with neck flexion and extension strength, and for peak linear head acceleration with neck extension strength in the male players.

Successful growth of female rugby requires a focus on female-derived data to develop laws, training techniques and coach education, rather than relying on the traditional androcentric data.

Declarations and Statements

This work has not previously been accepted in substance for any degree and is not being concurrently submitted in candidature for any degree.

Signed [redacted]

Date 03/09/2021

This thesis is the result of my own investigations, except where otherwise stated. Other sources are acknowledged by footnotes giving explicit references. A bibliography is appended.

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The University's ethical procedures have been followed and, where appropriate, that ethical approval has been granted.

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Abbreviations

ATD = anthropometric testing device

CTE = chronic traumatic encephalopathy

HIA = head impact analysis

HIE = head impact event

HIII = Hybrid III dummy

HIT= head impact telemetry

IQR = interquartile range

mTBI = mild traumatic brain injury

PLA = peak linear acceleration

PRA = peak rotational acceleration

RTP = return to play

SCM = sternocleidomastoid

TBI = traumatic brain injury

Chapter 1: Introduction

Rugby union (rugby) is one of the most popular contact sports in the world, with over eight million players; over 2.7 million of which are women (Hume et al., 2017, World Rugby, 2019). Although the men's game became professional at the elite level in 1995 (Gardner, Iverson, Williams, Baker, & Stanwell, 2014), the women's game is only just beginning professionalisation. England women's rugby team became the first full-time professional team in 2018, however, the number of professional female players is still low. Following the professionalisation of the men's game, research suggests that rugby has become more physical, with players becoming larger and more powerful (Duthie, Pyne, & Hooper, 2003; Hendricks & Lambert, 2010). Rugby is the only contact sport where the equipment, rules and laws of the game are identical for both sexes at the adult level (Gabb, 2018). Training strategies and injury prevention techniques, also the same, are based on androcentric data. As females are not represented in the datasets on which these are derived, females may be at a greater risk of injury (Bradley, Board, Hogg, & Archer, 2020; Carson, Roberts, & White, 1999; Doyle & George, 2004; Gabb, 2018; King et al., 2019; Schick, Molloy, & Wiley, 2008).

Concussion has consistently been reported as the most common injury in men's professional rugby, accounting for 20% of all match injuries (Kemp et al., 2018). Concussion epidemiology has been widely researched in the male game (Kemp et al., 2018; Rafferty et al., 2018). Concussion was reported to account for 6.2% of injuries in the 2006 Women's Rugby World Cup and 10% in 2010 (Schick et al., 2008; Taylor, Fuller, & Molloy, 2011). The literature surrounding female rugby is limited and these studies are outdated. There is an increasing body of evidence highlighting the urgent need to develop an evidence base for female-specific rugby training and injury management practices (Dollé et al., 2018; Ferretti et al., 2018; Smith et al., 2021). This is of particular concern given the significant increases in female participation and the progressive professionalisation of the women's game.

Concussions occur when a force to the head or body results in transient changes in neurocognitive function of the brain, causing symptoms including memory and balance impairments (Abrahams et al., 2019; McCrory et al., 2017). Concussive injuries can produce somatic, emotional and cognitive symptoms, including loss-of-consciousness, dizziness, memory dysfunction and difficulty concentrating (Abrahams et al., 2019; Ganly & McMahon, 2018; McCrory et al., 2017). Repetitive brain trauma has been linked with the neurodegenerative changes characteristic of chronic traumatic encephalopathy (CTE; Smith, Johnson, & Stewart, 2013; Stewart, McNamara, Lawlor, Hutchinson,

& Farrell, 2016). CTE is associated with a progressive decline in cognition and differences in brain metabolism that are disproportionate to a person's age (Koerte et al., 2015; Stern et al., 2011). Clinical symptoms of the disease may not manifest until over a decade after the impacts that caused it and are indistinguishable from Alzheimer's until post-mortem examinations (Stein et al., 2015). This means that diagnosis during life is not currently possible, meaning preventative strategies are imperative.

Literature examining sex differences in concussion incidence is mixed, with a number of earlier studies suggesting males are at a greater risk than females (Barnes et al., 1998; Boden, Kirkendall, & Garrett, 1998). More recent literature on sex-comparable sports at the high school level suggests the opposite, reporting females to be at a greater risk of sustaining a concussion than males (Frommer et al., 2011; Gessel, Fields, Collins, Dick, & Dawn Comstock, 2007; McClincy, Lovell, Pardini, Collins, & Spore, 2006). This research also reports females to report different symptoms and worse outcomes than their male counterparts following a concussion. Further research in sex-comparable high school sports reported females to be 1.6 times more likely to sustain a sports-related concussion than males (Bretzin et al., 2018). Differences in concussion incidence between sexes have been suggested to be due to anatomical and biomechanical differences (Mansell, Tierney, Sitler, Swanik, & Stearne, 2005).

The tackle is the most common form on contact in rugby and also the most injurious event in the professional game (Burger et al., 2016; Fuller, Brooks, Cancea, Hall, & Kemp, 2007; Kemp et al., 2018). In the professional game, 76% of head impact events are reported to occur in the tackle. Research focussing on the tackle is therefore essential to better understand the mechanics of head impacts and therefore aim to improve player safety. Studying head contact type, impact direction and other impact dynamics are also important to study to gain a better understanding of concussion biomechanics. While some research has been done to understand the mechanisms of injury in rugby, this has almost exclusively been male based (Cross et al., 2019; Hendricks et al., 2020; Kemp et al., 2018). These findings have been generalised to cover the sport as a whole, forming the basis of the laws and training methods of the game.

The ability to quantify the exposure to head impact events above a potentially injurious threshold is essential to enable associations between head impact exposure and neurocognitive outcomes (Bartsch, Samorezov, Benzel, Miele, & Brett, 2014; Camarillo, Shull, Mattson, Shultz, & Garza, 2013; Wu et al., 2016). Being able to correlate movement kinematics with head impact magnitude is also crucial to identify high risk events. Safety protocols can then be informed by relevant, objective data. For decades, attempts have been made to measure head acceleration in sports and motor vehicles. The first reasonable values, however, were not obtained until the late 1990s (Patton, 2016). Since this time, many types of wearable head impact telemetry systems have been developed. These include instrumented helmets, headgear, skin patches, earpieces and mouthguards (Rowson, Tyson, Rowson, & Duma, 2018). While each system has advantages and limitations, direct on-field measurements can provide the most relevant information relating to injury outcomes associated with training and games (Rowson et al., 2018). Systems with solid sensor-skull coupling have also been shown to provide the most accurate head impact data (Greybe, Jones, Brown, & Williams, 2020; Patton, 2016).

Increases in neck strength have been associated with decreased head kinematics during low impulse loading (Eckner, Oh, Joshi, Richardson, & Ashton-Miller, 2014; Simoneau, Denninger, & Hain, 2008). Research in high school athletes from a mixture of sports reported that smaller neck circumference and weaker neck strength was correlated with a greater incidence of concussion (Collins et al., 2014). While little research exists comparing neck strength to head impacts in rugby itself, findings from other contact sports indicate that neck strength training will also be beneficial in rugby. Sex differences in head acceleration in laboratory settings have previously been attributed to females having lower isometric neck strength, smaller neck circumference and smaller head mass, therefore a lower head-neck stiffness than males (Tierney et al., 2005). The male neck is reported to exhibit greater intervertebral coupling stability than females (Mohan & Huynh, 2019). This makes the male neck more proficient at resisting inertial loading and extreme movements (Stemper et al., 2009). Neck strength and head impact associations are therefore an important area for further study. Neck strength-based training programmes could be a relatively simple way to increase the safety of rugby without substantial rule modifications.

Chapter 2: Literature Review

2.1 Rugby Union and Women

Rugby is a team sport with over eight million participants across 129 countries worldwide, making it the most popular high-impact collision sport (Hume et al., 2017; Pollock, 2014). While rugby is traditionally a male-dominated sport, female participation is increasing by almost 30% per year, with over 2.7 million female players as of 2019 (World Rugby, 2019). While some research has been completed to better understand the mechanisms of injury in rugby, this has almost exclusively been focussed on the men's game (Cross et al., 2019; Hendricks et al., 2020; Kemp et al., 2018). These findings have been generalised and applied to the sport as a whole with a lack of female representation (Fuller, Taylor, & Raftery, 2015; Tierney & Simms, 2017), creating a gender data gap (Costello, Bieuzen, & Bleakley, 2014). Considering these studies form the basis of rugby's laws and training methods, a substantial effort to understand female mechanics of injury is crucial to increase the safety of the sport.

2.1.1 The Game

The game is played over two 40-minute halves, although the ball is actually typically only in play for 30 minutes during a game (Duthie, Pyne, & Hooper, 2003; McLean, 1992). The remainder of the 80-minute game time is taken up by conversions, penalty kicks or the ball being out of play (Morton, 1978). Rugby is unique in that it is a combination of strength, power, speed, agility and endurance in a game of contact and collision, with little use of protective equipment (Frounfelter, 2008). The game is territorial, where during attacking phases of play the team attempts to advance the ball closer to the opposition's try line to score points (Tierney, Denvir, Farrell, & Simms, 2018). Meanwhile, the defending team attempts to prevent this forward movement of the other team through tackles and other high impact collisions (Gabbett & Kelly, 2007; Hendricks & Lambert, 2010).

2.1.2 The Physicality of the Game

The men's game became professional at the elite level in 1995, following the men's World Cup (Gardner, Iverson, Williams, Baker, & Stanwell, 2014). Men's professional full-time training has therefore resulted in increased player skill, strength, power and fitness (Hendricks & Lambert, 2010). The average player body mass has also increased due to increases in training and recovery time, training technology and expertise (Duthie et al., 2003). This greater physicality has increased physical demands on the players, particularly the greater speed and force of contact events (Eaves & Hughes, 2003). This has resulted in the number of contact events, such as tackles and rucks, per game quadrupling at the elite level since 1995 (Austin, Gabbett, & Jenkins, 2011; Eaves & Hughes, 2003; Quarrie & Hopkins, 2007). England's women's rugby team became the first professional full-time test team in 2018, 23 years after the professionalisation of the men's game. The number of female professional players, however, is still extremely low.

Rugby is the only collision sport that has the same rules and equipment for both the male and the female adult game (Gabb, 2018). The rules, training strategies and injury prevention practices have all been based on male-derived data, with a dearth of literature regarding female rugby (Bradley, Board, Hogg, & Archer, 2020; Carson, Roberts, & White, 1999; Doyle & George, 2004; Gabb, 2018; King et al., 2019; Schick, Molloy, & Wiley, 2008). Therefore, the game may pose a greater risk of injury for female players.

2.1.3 Rugby Positions

A rugby team consists of 23 players, with 15 players being on field at a time. Each player has a designated positional number (World Rugby, 2021). These playing positions are collated into two distinct subgroups: forwards and backs.

Forwards tend to be in continual close contact with the opposition, predominantly in charge of competing for possession of the ball and gaining ground in contact events such as rucks, scrums and mauls (Duthie et al., 2003; Quarrie & Wilson, 2000). Forwards have consistently been reported to have greater body mass, and possess greater fat masses (both absolute and relative) than backs (Zemski, Slater, & Broad, 2015). Greater body mass has also been suggested to be associated with greater team competition success (Olds, 2001; Adrien Sedeaud et al., 2012), as well as being associated with a greater force output during contact events, such as the scrum (Quarrie & Wilson, 2000). Momentum is the product of mass and velocity, meaning that a higher body mass produces a higher momentum. This is beneficial during the many contact events that forwards are involved in, where possession of the ball is contested, or tackles are attempted to be broken (Higham, Pyne, Anson, Dziedzic, & Slater, 2014).

Backs generally control possession of the ball and are responsible for gaining ground and creating scoring opportunities for the team, while also giving cover in defence (Duthie et al., 2003; Zemski, Keating, Broad, Marsh, & Slater, 2019). They tend to be the quicker, more agile players on the team (Duthie et al., 2003). Backs have been reported to have a higher percentage of lean body mass and to be more aerobically fit compared to their forward counterparts (Duthie et al., 2003; Higham et al., 2014), which is beneficial considering their main roles within the team.

2.2 Concussive Injuries

2.2.1 Identifying Concussion

Concussion has been defined as “a complex pathophysiological process affecting the brain, induced by traumatic biomechanical force” (McCrory et al., 2017). Concussions occur when an impact force to the head or body result in transient changes in neurocognitive function of the brain, leading to neurological deficits such as impairments to balance, memory and cognition (Abrahams et al., 2019; McCrory et al., 2017). The initial identification of concussion has consistently been one of the most challenging aspects of injury assessment and management concerning brain injuries (Broglio, Lapointe, O’Connor, & McCrea, 2017). To date, no precise brain injury mechanisms or tolerance criteria have been identified within concussion literature (Guskiewicz & Mihalik, 2011; Hardy et al., 2007), making the diagnosis of concussion a generally subjective decision.

World Rugby has outlined a three-stage diagnostic process for concussion. This process involves a head injury assessment (HIA) immediately post-injury (HIA 1), a repeat assessment within three hours of the injury (HIA 2) and a follow-up assessment at 36- to 48-hours post-injury (HIA 3; Raftery, Kemp, Patricios, Makdissi, & Decq, 2016). Each of the HIA stages are based on the Sports Concussion Assessment Tool 5 (SCAT5; Echemendia et al., 2017) and are outlined by Raftery et al. (2016). This process was introduced due to recognition that the signs and symptoms of concussion may be transient, fluctuating or delayed, helping to minimise misdiagnoses (Raftery et al., 2016).

Current regulations in the game stipulate that any adult player with a concussion or suspected concussion should be immediately removed from training or play. They should then rest for at least a week before beginning the 6-phase graduated return to play (RTP) programme (Liston, McDowell, Malcolm, Scott-Bell, & Waddington, 2018). These regulations, however, are based androcentric data, meaning they may not be as suitable or effective for female players.

2.2.2 Short-Term Effects of Concussion

Concussive injuries can produce somatic, emotional and cognitive symptoms, such as loss-of-consciousness, visual disturbances, dizziness, memory dysfunction, difficulty concentrating, irritability and confusion (Abrahams et al., 2019; Ganly & McMahon, 2018; McCrory et al., 2017).

This is often in the absence of any evident structural abnormality observed using standard neuroimaging techniques (Ganly & McMahon, 2018).

Prospective studies have reported that the majority of athletes recovery completely from symptoms, cognitive dysfunction and other impairments over approximately seven to ten days following concussion (Belanger & Vanderploeg, 2005; McCrea et al., 2003; Pellman, Lovell, Viano, & Casson, 2006). Returning to play following a concussion in the same season, however, has been linked to a 60% increased risk of subsequent injury (Cross et al., 2016). This was accompanied by the finding that players returning from concussion exhibited a shorter time until subsequent injury than players returning from other injuries (Cross et al., 2016).

2.3 Brain Injury in Contact Sports

2.3.1 Brain Injury Nomenclature

Brain injuries in contact and collision sports have become an area of increasing concern worldwide (Covassin, Elbin, & Sarmiento, 2012; De Beaumont, Brisson, Lassonde, & Jolicoeur, 2007; McKee et al., 2009). Both the immediate and long-term effects of the repeated impacts to the head that athletes experience during contact sport are of growing concern due to the potential adverse effects on cerebral function (Baugh et al., 2012; Gavett, Stern, & McKee, 2011; Gysland et al., 2012).

Sport-related concussion can be defined as "a traumatic brain injury induced by biomechanical forces" during sports (McCrory et al., 2017). Traumatic brain injury (TBI) has been defined as "an alteration in brain function, or other evidence of brain pathology, caused by an external force" (Menon, Schwab, Wright, & Maas, 2010).

Sub-concussive impacts are head impacts that involve the transfer of mechanical energy to the brain at enough force to injure axonal or neuronal integrity but not be expressed in clinical symptoms (Bailes, Petraglia, Omalu, Nauman, & Talavage, 2013; Belanger, Vanderploeg, & McAllister, 2016). It has been suggested that repetitive sub-concussive impacts may have cumulative neurophysiological effects (Shultz, MacFabe, Foley, Taylor, & Cain, 2012) and that brain injuries may be the result of both acute concussive events and the accumulation of sub-concussive impacts (Spiotta, Shin, Bartsch, & Benzel, 2011; Talavage et al., 2014). Due to the absence of observable clinical symptoms, sub-concussive impacts often go unnoticed and resulting neurodegenerative symptoms may have an insidious onset.

There is currently no international consensus of what minimum magnitude constitutes a 'head impact'. In head impact literature, a head impact is generally classified as when acceleration of the head exceeds 10 g (Ng, Bussone, & Duma, 2006). The general consensus is that this is the threshold that will eliminate typical head acceleration from normal, non-injurious activities, such as running or jumping (Bussone & Duma, 2010; Crisco et al., 2011). For the purpose of the current research, the term 'head impact' encompasses all head accelerations exceeding 10 g, including both concussive and sub-concussive impacts. Different measurement methods, however, have been shown to be associated with varying degrees of accuracy and reliability (Patton, 2016; Wu, Nangia, et al., 2016). Given the variability in the methods used to measure head impact magnitude in the literature, the minimum threshold is somewhat arbitrary.

2.3.2 Concussion Epidemiology in Male Rugby

Concussion has consistently been reported to be one of the most common injuries in the professional men's rugby game (Kemp et al., 2018). It has been reported to account for 20% of all injuries in the professional game, and 37% of injuries occurring to tacklers (Kemp et al., 2018). Kemp and colleagues reported 17.9 injuries per 1,000 player-hours in the 2017-18 season of the English Premiership in rugby (Kemp et al., 2018). This incidence is much greater than the reported incidence of 8.9 per 1,000 player-hours for the same level in 2016 (Cross, Kemp, Smith, Trewartha, & Stokes, 2016). The contrast is even greater when compared to the incidence of 1.4 per 1,000 player-match-hours reported in South African professional rugby in 2006 (Holtzhausen, Schwellnus, Jakoet, & Pretorius, 2006). This increased incidence, however, could be due to an improved awareness and reporting of concussive injuries rather than an increased incidence itself (Cross et al., 2016).

Research in professional men's rugby in Wales (International and Guinness Pro14 teams) has reported that, on average, club players are at a greater risk of sustaining a concussion after 25 matches (Rafferty et al., 2018). The Guinness Pro14 league consists of 21 rounds of matches alone, with the potential for further games due to Champions Cup and International matches occurring alongside the regular season. Many players, especially international players, will therefore be approaching or exceeding 25 matches per year. At the lower levels, English community clubs reported concussion to be the fifth most common injury, with an incidence of 1.2 injuries per 1,000 player-hours (Roberts, Trewartha, England, Shaddick, & Stokes, 2013). At the collegiate level, a USA study reported concussion incidence of 2.16 per 1,000 player-hours (Kerr et al., 2008).

2.3.3 Concussion Epidemiology in Female Rugby

The English Rugby Football Union (RFU) began the Women's Rugby Injury Surveillance Program (WRISP) in 2017 which included six Premier 15s teams (Kemp et al., 2018). This was expanded to ten teams for the 2018/19 season (Kemp et al., 2019). While only focused on a relatively small number of elite women's teams, findings show that concussion was the most commonly reported match injury (6.7 per 1000 hours, or one concussion every 3.7 matches), making up 19% of all match injuries (Kemp et al., 2019). In comparison to the men's game, the ball carrier incurred a greater proportion of tackle injuries (21%) than the tackler (15%; Kemp et al., 2019).

Concussion incidence at the 2006 Women's Rugby World Cup was reported as 0.33 per 1,000 player-match hours, accounting for 6.2% of all injuries (Schick et al., 2008). This incidence increased, with concussion accounting for a 10% of all injuries, at the 2010 World Cup (Taylor, Fuller, & Molloy, 2011). Research comparing concussion incidence between male and female collegiate rugby players reported an incidence of 2.16 and 1.58 per 1,000 player-match hours, respectively (Kerr et al., 2008). Very little data on concussion incidence in women's rugby exists (Gabb, 2018). More in-depth and up-to-date data needed to enable more accurate comparisons between men and women.

2.3.4 General Sex Comparisons in Sports Concussion

Studies comparing concussion patterns and rates in gender-comparable sports at the high school level report that females sustained a significantly greater number of concussions than males (Frommer et al., 2011; Gessel, Fields, Collins, Dick, & Comstock, 2007; McClincy, Lovell, Pardini, Collins, & Spore, 2006). The research also found that females report different symptoms and worse overall outcomes in comparison to males (Frommer et al., 2011; Gessel et al., 2007; McClincy et al., 2006). Concussion disparity between sexes has been also reported in collegiate soccer, with concussion accounting for 11.4% of game injuries in the female game, compared to 7% in the male game (Covassin, Swanik, & Sachs, 2003). This difference may be attributed to anatomical and biomechanical differences between sexes (Mansell, Tierney, Sitler, Swanik, & Stearne, 2005; Chapter 2.5.3; Chapter 2.7.2).

2.4 Head Impact Mechanisms in Rugby

2.4.1 Reported Movement Kinematics and Head Impacts in Rugby

The tackle is the most common form of contact in rugby, likely explaining why it is the most injurious match event in the professional game, accounting for 52% of match injuries (Burger et al., 2016;

Fuller, Brooks, Cancea, Hall, & Kemp, 2007; Kemp et al., 2018). Studies in six major rugby competitions reported that 76% of head impact events (HIE) occurred due to tackles (Tucker, Raftery, Fuller, et al., 2017). It has been reported in the literature that 70% of concussions occur to the tackler, meaning tackler-specific research is a priority (Cross et al., 2019). Head contact type has been identified as one of the most important variables for predicting concussion, meaning a focus on this is also necessary to better understand risk factors for concussion. Direct head contact with the ground or the opposing player's head or knee has been reported to be associated with substantially greater risk of concussion when compared with all other injury locations (Burger et al., 2016). However, these scenarios do tend to occur infrequently in comparison to more common impact locations, such as the opposing player's trunk or lower limbs (Burger et al., 2016).

Research into the mechanics of head impacts could improve player safety through the implementation of rule changes, behaviour changes, better equipment and improved on-field identification of potentially injurious impacts (Siegmund, Guskiewicz, Marshall, DeMarco, & Bonin, 2016). Impact direction is a potentially important parameter to ascertain as impact-induced skull motion is related to the magnitude of brain tissue strain experienced, as well as brain injury tolerance (Bayly et al., 2005; Takhounts, Craig, Moorhouse, McFadden, & Hasija, 2013; Weaver, Danelson, & Stitzel, 2012; Zhang, Yang, & King, 2004). Motion of the skull can be described in terms of both linear and rotational motion. Linear and rotational acceleration and velocity are correlates of injury that can be measured to quantify a head impact, but are not the actual mechanism of injury (Rowson, Tyson, Rowson, & Duma, 2018).

2.4.2 Reported Sex Differences and Injury Patterns

Collins et al. (2014) reported gender to be a significant predictor of concussion in high school soccer, basketball and lacrosse athletes, with females sustaining substantially more concussions than males. Female athletes have been reported to be 1.56 times more likely to sustain a sport-related concussion than their male counterparts (Bretzin et al., 2018). Indeed, studies investigating a variety of high school sports have indicated that young female athletes are more likely to sustain a concussion than males in sex-comparable sports (Kerr et al., 2019; Schallmo, Weiner, & Hsu, 2017; Tsushima, Siu, Ahn, Chang, & Murata, 2019).

2.5 Brain Pathology and Brain Physical Properties

2.5.1 Brain Macrostructure

Concussion has been linked to damaging magnitudes of strain within the brain tissues, especially in the corpus callosum (Kleiven & Von Holst, 2002; Patton, McIntosh, & Kleiven, 2013; Post, Blaine Hoshizaki, Gilchrist, & Cusimano, 2017). The corpus callosum is the largest commissure of the brain, connecting the cerebral hemispheres (VanPutte, Regan, & Russo, 2014). The falx cerebri is located superior to the corpus callosum and is thought to affect the corpus callosum during skull acceleration, particularly during lateral impacts (Hernandez et al., 2019). The falx cerebri is the dural fold between the two cerebral hemispheres and helps to hold the brain in place within the skull (VanPutte et al., 2014). The surface of each hemisphere consists of numerous folds, called gyri, which increase the surface are of the brain. The grooves between each of these folds are known as sulci (VanPutte et al., 2014).

2.5.2 Brain Pathology and Sex Differences

Repetitive brain trauma is associated with long term neurodegenerative challenges, including chronic traumatic encephalopathy (CTE; Smith, Johnson, & Stewart, 2013; Stewart, McNamara, Lawlor, Hutchinson, & Farrell, 2016) and early onset Alzheimer’s (McKee et al., 2010). Mild cognitive impairment, neuroimaging abnormalities, a progressive decline in cognition and differences in brain metabolism that are disproportionate to a person’s age are all characteristic of CTE (Koerte et al., 2015; Stern et al., 2011).

The brain has a low shear modulus, meaning that it has a low ability to withstand changes in shape (Patton et al., 2013). Rapid rotational acceleration of the head results in shear forces in the brain, leading to deformation and shear-induced tissue damage (Tierney & Simms, 2017). There is a strong link between brain tissue strain and the tissue damage or cell death experienced by brain tissue (Cullen, Vernekar, & LaPlaca, 2011; Lamy, Baumgartner, Yoganandan, Stemper, & Willinger, 2013; LaPlaca, Cullen, McLoughlin, & Cargill, 2005; Morrison, Cater, Benham, & Sundstrom, 2006).

The two major mechanisms of brain injury frequently postulated in the literature are linear and rotational acceleration of the head (Greybe et al., 2020; Post et al., 2017). Linear acceleration of the skull relates to intracranial damage due to injurious pressure gradients within the brain tissue (Hardy et al., 2007; Jin et al., 2017; Takhounts et al., 2008). Rapid head rotation results in shear forces throughout the brain. This causes deformation and shear-induced mechanical damage to the axonal microtubules of the brain tissue (Dollé et al., 2018; Duthie et al., 2003; Tierney & Simms, 2017).

Rotational acceleration is frequently cited as the major contributor to brain tissue damage during head impact events (Fernandes & Sousa, 2015; Tierney & Simms, 2017; Zhang, Yoganandan, Pintar, & Gennarelli, 2006).

While the literature relating specifically to female TBI is sparse, recent findings suggest that female axons are more susceptible to damage following comparable trauma than male (Dollé et al., 2018; Gupte, Brooks, Vukas, Pierce, & Harris, 2019) Female axons are smaller in diameter with fewer microtubules than male (Dollé et al., 2018) and sex hormones are also thought to influence inflammation and other cellular processes (Gupte et al., 2019).

2.6 Head Impact Telemetry in Sports

2.6.1 Introduction to Head Impact Telemetry

Given the consequences associated with both concussive and sub-concussive impacts, it is important to be able to accurately quantify exposure to accelerations above a normal threshold. Ultimately, this will enable mechanistic links to be made between both absolute and cumulative head impact exposure and neurocognitive outcomes (Bartsch, Samorezov, Benzel, Miele, & Brett, 2014; Camarillo, Shull, Mattson, Shultz, & Garza, 2013; Wu, Laksari, et al., 2016). Accurate knowledge of the magnitude of forces that are transmitted to the head is essential to mitigate the occurrence of mTBI (Jadischke, Viano, Dau, King, & McCarthy, 2013). Understanding the kinematics that cause these forces is also vital to identify high risk events and attempt to reduce these risks.

Rather than using quantifiable parameters to predict injury rates for populations as a whole, these parameters can be used for injury prediction in individuals. Research has reported that concussed participants sustain a greater biomechanical load prior to concussive injury, suggesting that head impact loading data needs to be considered on an individual level (Rowson et al., 2019). Individual-specific risk analyses are able to consider a person’s impact history and other factors that may influence their tolerance to head acceleration (Rowson et al., 2019). Other factors that may influence head impact tolerance include head size and shape, age, material properties of the soft tissue of the brain and neck size and strength (Collins et al., 2014; Danelson, Geer, Stitzel, Slice, & Takhounts, 2008; Miller & Chinzei, 2002; Rowson et al., 2016).

2.6.2 History and Development of Head Impact Telemetry in Sports

Early head impact telemetry systems only included linear acceleration (Patton, 2016). The importance of angular head kinematics in TBI has been discussed in biomechanical literature since the 1940s (Gennarelli, Thibault, & Ommaya, 1972; Holbourn, 1943). From the 1960s to 1980s, attempts to measure head acceleration in American football were made, however, these attempts encountered many technical difficulties and resulted in unreasonably high measurements (Morrison, 1983; Reid, Tarkington, Epstein, & O'Dea, 1971). The first seemingly reasonable values were taken in the late 1990s using a helmet with embedded triaxial accelerometer (Naunheim, Standeven, Richter, & Lewis, 2000), with the first widely used sensor developed in the early 2000s. The Head Impact Telemetry (HIT) instrumented helmet system has been used to quantify head accelerations on-field in American football since the 2003 season (Crisco, Chu, & Greenwald, 2004; Duma et al., 2004).

Many types of wearable head impact telemetry systems exist, including instrumented headgear (helmets, headbands and skullcaps), skin patches, mouthguards and ear pieces (Rowson et al., 2018). Validation data exists for some of these devices, however, each of the studies had limitations that may affect data reliability (Siegmund et al., 2016). Injury criteria derived from cadaver studies are used for current helmet and automotive safety standards (Gurdjian, Lissner, Evans, Patrick, & Hardy, 1961; Lissner, Lebow, & Evans, 1960). This research measured head acceleration with only a single accelerometer, however, and could therefore only measure 2-dimensional acceleration, decreasing the reliability of the results (Rowson et al., 2018). The use of cadavers in head injury research means functional brain injury outcomes and injury responses are not provided, but parameters relating to injury may still be measured (Hardy et al., 2007; Rowson et al., 2018). Although each method of studying head impact biomechanics has its limitations, direct, on-field measurements from athletes provide the most information in terms of biomechanical measures with associated injury outcomes (Rowson et al., 2018).

2.6.3 Different Head Impact Telemetry Systems: An Evaluation

The Hybrid III (HIII) anthropometric testing device (ATD) was initially developed to investigate head accelerations in relation to car crashes (Bartsch, Benzel, Miele, Morr, & Prakash, 2012). It is now also considered the gold standard reference for all head acceleration data (Jadischke et al., 2013). It has been widely used in both vehicle safety literature and sports-related head impact research (Greybe et al., 2020; Liu et al., 2020). The ATD head form has been deemed to be biofidelic based on four facts (Mertz, 1985). Specifically, the shape of the forehead and mass are representative of a

50th percentile male and the rearward bending of the neck structure and the forehead stiffness are humanlike. These comparisons have been made to cadavers, however, and deceased tissue responds differently to live tissue (Kent, Patrie, & Benson, 2003). Furthermore, although the ATD is representative of an average male head, this may differ in females and athletes (Linder & Svedberg, 2019). Female ATDs do exist, however, they are just scaled versions of the male 50th percentile head form (Linder & Svedberg, 2019). These do not take into account the differences in spinal anatomy and the surrounding tissues between males and females (Yoganandan, Bass, Voo, & Pintar, 2017).

2.6.4 HIT System

The HIT system uses helmet-mounted accelerometers to determine the linear and angular acceleration of the head on-field in American football (Jadischke et al., 2013). Research using this system is mainly focused on youth and collegiate level athletes to assess the frequency and severity of the head impacts they experience (Funk & Duma, 2007). The HIT system has been evaluated multiple times in literature, with a mixture of conclusions formed (Jadischke et al., 2013; Rowson et al., 2018). Multiple studies have been performed correlating kinematic measures recorded by the HIT system and a HIII anthropometric head form in conditions that mimic head impacts experienced in the National Football League (Jadischke et al., 2013; Rowson et al., 2018; Wu, Nangia, et al., 2016). The extent to which the HIT system validation experiments represent human head-helmet interaction dynamics have been widely questioned (Beckwith, Greenwald, & Chu, 2012). Jadischke et al. (2013) further evaluated the accuracy of the HIT system. These authors reported PLA to be over-predicted at all impact speeds and locations. The root mean square error (RMSE) was greater than 59.1% for peak linear acceleration (PLA) measured with the HIT system compared to the reference ATD. Of the validation studies completed, many have used a medium-sized helmet on a HIII head (Beckwith et al., 2012; Rowson et al., 2011; Steven Rowson, Brolinson, Goforth, Dietter, & Duma, 2009). Tighter helmet-head form coupling in these laboratory validation studies may also be greater than in vivo (Beckwith et al., 2012; Rowson et al., 2011, 2009) further questioning the validity of these findings.

2.6.5 Skin Patch Head Impact Sensor Systems

An assessment of the wearable head impact sensors reported many limitations of the X-Patch® sensors (McIntosh et al., 2019). The skin patch accelerometers were compared to the HIII head, with PLA being over-predicted by 17% on average, and PRA underpredicted by 28% (McIntosh et al., 2019). The research reported the X-Patch® to have limitations in accuracy within the lab, and validity in on-field measurements. However, the lab-based assessment was limited by the number, location and severity of the impacts that were analysed. The X-Patch® derived angular acceleration was considered to be inaccurate and it was suggested that considerable caution be used when interpreting and applying the data.

Rowson and colleagues stated that the presence of skin motion renders sensors unreliable on the field, due to a time lag between the true impact and the impact recorded by the sensor (Rowson et al., 2018). Soft tissue artefact (STA), the relative motion of the sensor on the skin, has also been shown to overestimate the impact magnitude and cause numerous false-positive impacts (Blache et al., 2017). Skin-mounted sensors demonstrate non-rigid sensor-skull coupling and out-of-plane motion, therefore over-predicting skull kinematics (Wu, Nangia, et al., 2016).

2.6.6 Instrumented Mouthguards

There are significant inaccuracies associated with sensors worn on the skin or in headgear due to STA and decoupling of equipment from the skull. The instrumented mouthguard (iMG) has subsequently been designed to couple directly with the upper dentition, giving the sensors superior coupling with the skull (Kuo et al., 2016). Multiple studies, however, have reported conflicting results regarding the validity and accuracy of instrumented mouthguards (Patton et al., 2020). The discrepancies between these studies could be attributed to the use of different anthropometric test dummies (ATDs) and the employment of different mandible constraint (Kuo et al., 2016; Liu et al., 2020).

One limitation associated with using iMGs is biting. Because the lower dentition is in direct contact with the mouthguard, mandible motion could lead to deformation within the iMG during dynamic events where the upper and lower dentition close (Kuo et al., 2016). Research in amateur rugby, consisting of over 20,000 impacts exceeding 10 g, reported an average of 1,379 impacts per player, per season (King, Hume, Brughelli, & Gissane, 2015). In comparison with head impact literature in other sports, these values are significantly greater (Brolinson et al., 2006; Crisco et al., 2012; Schnebel, Gwin, Anderson, & Gatlin, 2007). The mouthguards used in these studies, however, were reported to be bulky to fit and required individual customisation, likely leading to poor sensor-skull coupling and therefore over-estimations of head acceleration.

Research comparing wearable head impact sensors reported instrumented mouthguard to display the least displacement from the skull when compared to skin patch and head-gear mounted sensors

(Bartsch et al., 2014; Wu, Nangia, et al., 2016). Due to the rigid coupling between the upper dentition and the skull, iMGs have been shown to be a valid measure of head impact kinematics (Liu et al., 2020). Various companies have subsequently further developed iMGs as a research tool. A study validating and comparing iMGS showed that all tested iMGS gave accurate measurements for PRA and PRV (Liu et al., 2020).

2.7 Neck Strength and Head Acceleration in Sports

2.7.1 Anatomy of the Neck

The musculature of the neck can be split into three broad categories: anterior, posterior and lateral muscles (VanPutte et al., 2014). Most of the neck flexors lie deep within the neck, along the anterior margins of the vertebrae. These anterior muscles are the longus capitis and rectus capitis anterior. The posterior neck muscles are attached to the occipital bone and mastoid process of the temporal bone and function as a first-class lever to extend the head/neck. These muscles are the longissimus capitis, oblique capitis superior, rectus capitis posterior, semispinalis capitis, splenius capitis and trapezius muscles. Rotation of the head is achieved through the use of both posterior and lateral muscles. The lateral muscles are the rectus capitis lateralis, sternocleidomastoid (SCM) and scalene muscles. The SCM is the prime mover of the lateral group. When contracting bilaterally it is primarily responsible for flexion of the head, and unilateral contraction causes head rotation. The scalene muscles assist the SCM in head flexion.

2.7.2 Sex Differences in Neck Anatomy and Physiology

Several systemic morphological differences have previously been noted between male and female cervical spines (Östh et al., 2017). It has been reported that male necks exhibit greater intervertebral coupling stability than female necks, with greater vertebral body width (Mohan & Huynh, 2019). This greater stability mean the spine is more proficient at resisting inertial loading and extreme movements (Stemper et al., 2009). The circumference of the neck relative to its length is smaller, as are the vertebral body sizes, in females than in males (Vasavada, Danaraj, & Siegmund, 2008). It has been reported in vehicle-impact literature that during rear-impact collisions, females exhibit greater head-neck movement than males (Stemper, Pintar, & Rao, 2011). This elevates the load placed on the neck and head, increasing injury risk due to greater soft-tissue deformation and failure (Mohan & Huynh, 2019; Stemper et al., 2011). These findings mean females are at a greater risk of injury under whiplash conditions (Stemper, Yoganandan, & Pintar, 2004).

2.7.3 Correlations Between Neck Strength and Head Acceleration in Rugby

Increases in neck strength have been widely hypothesized to be an effective strategy to lower the risk of sports-related concussions, however, current evidence is limited (Collins et al., 2014; Eckersley, Nightingale, Luck, & Bass, 2019; Jin et al., 2017; Schmidt et al., 2014). These hypotheses are based on the idea that cervical muscle contraction increases head-to-neck coupling, increasing the effective mass of the head-neck segment. According to Newton's law of acceleration, force = mass x acceleration, meaning acceleration is proportional to both the mass of the object being accelerated and the force applied. According to this, a greater head-neck segment mass will reduce the acceleration of the head for a given force. Although multiple studies have investigated the relationship between neck strength and head impacts in other contact sports and soccer, rugby-specific research is still lacking.

2.7.4 Relationships Between Neck Strength and Concussion in Contact Sports

A number of studies have investigated the relationship between neck strength and anthropometric measurements with head and neck kinematics during low impulse loading (Collins et al., 2014; Eckner, Oh, Joshi, Richardson, & Ashton-Miller, 2014; Mansell et al., 2005; Simoneau, Denninger, & Hain, 2008). The majority of these studies determined that increases in neck muscle force led to decreased head kinematics (Collins et al., 2014; Eckner et al., 2014; Mansell et al., 2005; Simoneau et al., 2008).

Research in high school athletes, including both male and female soccer, basketball and lacrosse players, used a hand-held tension scale to obtain neck strength measurements. The athletes were monitored for concussion using the National High School Sports-Related Injury Surveillance Study (Collins et al., 2014). The research reported smaller neck circumference, smaller mean neck to head circumference and weaker mean overall neck strength to be correlated with concussion. This study reported that for every one pound (0.45 kg) increase in neck strength, the odds of concussion decreased by 5%. While factors other than neck strength are likely to contribute to this finding, this supports neck strength screening as one possible indicator of concussion risk.

Research aiming to determine the influence of neck strength and muscle activation status on the resultant head kinematics after impulsive loading demonstrated clear links between neck strength and head acceleration (Eckner et al., 2014). Eckner and colleagues studied 24 male and 22 female contact sport athletes between the ages of eight- and 30-years. The participants represented a broad range of sports, including soccer, ice hockey and martial arts, across a variety of competitive levels. The findings supported the theory that greater neck strength and anticipatory muscle activation attenuated the kinematic response of the athletes' heads to impulsive forces. This study, along with others, suggests that strong necks decrease head acceleration, rapid changes in velocity and displacement following a collision (Caccese et al., 2018; Eckner et al., 2014; Viano, Casson, & Pellman, 2007). Even a small decrease in head velocity following a collision may result in a significant reduction in the risk of concussion (Viano et al., 2007). Research involving mathematical and computational models, including FE models, agrees with the findings in human participant studies, also reporting neck muscle force to influence head kinematics (Jin et al., 2017; Viano et al., 2007). These kinematics include head acceleration, velocity and displacement. A systematic review, consisting of five cross-sectional studies, indicates that greater neck strength is significantly associated with lower linear and rotational head acceleration during soccer heading (Peek, Elliott, & Orr, 2020). The athletes in this research were all high school to collegiate level soccer players. Although these findings are not directly related to rugby impact events, the findings could be transferrable to rugby impacts. However, further rugby-specific research would be required to verify any association.

Research investigating the role that neck muscle strength plays in blunt head impact kinematics hypothesized that head kinematics would not vary due to low short-term head-to-neck coupling (Eckersley et al., 2019). The impacts used in the study were modelled using the Duke University Head and Neck model across eight sites, with four impact types and six neck conditions. No significant differences were found between neck activation conditions, suggesting that increased neck force does not influence short term head kinematics, and therefore concussion risk. Investigations into the effect of neck strength on head impact biomechanics using HIT instrumented helmets in youth ice hockey further supported this finding. No significant associations were observed between isometric cervical strength and both linear and rotational head acceleration (Mihalik et al., 2011). The HIT system has been utilised in high school and collegiate American football to investigate how cervical muscle characteristics influence head kinematics (Schmidt et al., 2014). The findings reported that greater cervical stiffness reduced the risk of sustaining high magnitude head impacts. However, neither increased neck muscle girth nor strength significantly mitigated the risk of high magnitude head impacts.

2.7.5 Research on Head-Neck Segment Mass and Head Acceleration

Tierney et al. (2005) reported that physically active females have greater head-neck segment acceleration than males when their heads are subjected to the same load. This difference has been attributed to the females having a smaller neck girth and lower head mass, and therefore lower head-neck stiffness and strength than males. These findings are consistent with earlier studies demonstrating that males have stronger necks, with greater girths than females (Garcés, Medina, Milutinovic, Garavote, & Guerado, 2002; Jordan, Mehlsen, Bülow, Østergaard, & Danneskiold-Samsee, 1999; Maeda, Nakashima, & Shibayama, 1994). It has been suggested that if neck girth and strength were increased in females, neck stiffness values would likely also increase, leading to a decrease in head-neck segment acceleration (Mansell et al., 2005).

2.7.6 Sex Differences in Neck Strength

Research has reported female neck strength to be between 20 to 49% less than that of males in a healthy population (Garcés et al., 2002; Tierney, 2004; Vasavada et al., 2008). Further research into sex strength differences has reported no significant differences in force per unit area of muscle between genders (Bell & Jacobs, 1986; Schantz, Hutchinson, Tydén, & Åstrand, 1983). This suggests that differences in strength between genders is related to muscle cross-sectional area rather than muscle function.

2.8 Objectives of This Thesis

Given the increasing numbers of females in rugby, and the androcentricity of the research behind the laws and training protocols, a focus on female-based research is critical. Concussion is one of the most common injuries in the sport, with the majority of research in other sex-comparable sports suggesting that females are at a greater risk. The mixed reports in the existing literature means further research is essential to determine if there is a significant relationship between neck strength and head impacts, particularly in rugby players. If significant relationships exist, further research needs to be completed to identify how best to train neck strength, and therefore increase player safety.

The aim of this thesis is to identify the differences in playing experience and head and neck size and strength both between sexes, and between the positional groups within the sexes. This thesis also aimed to observe differences in contact technique characteristics between the sexes, and better understand the relationships between these and head impact magnitude. The final aim of this thesis is to identify correlations between head and neck size and strength, and the head impact magnitudes experienced by the players.

Chapter 3: Methodology

3.1 Study Participants

A total of 31 male (15 forwards, 16 backs; 20.7 ± 1.4 years; 183.5 ± 9.3 cm; 94.9 ± 12.8 kg) and 22 female (9 forwards, 12 backs; 20.6 ± 2.1 years; 163.8 ± 6.1 cm; 71.7 ± 15.0 kg) university rugby players volunteered to participate in this thesis. Male and female participants competed in the British Universities and Colleges Sport Super Rugby and Women's Premier South leagues, respectively.

3.2 Ethics Approval

Ethics approval was obtained from Swansea University Applied Sports, Technology, Exercise and Medicine Human Ethics Committee (2016-059) prior to the commencement of the study, consistent with the requirements of the declarations of Helsinki.

3.3 Experimental Procedures

3.3.1 Demographics Anthropometric Assessment

Prior to testing, the participants completed a questionnaire, detailing their age, sex, sports participation history and injury history. Participant stature, body and head mass, head circumference, neck circumference and shoulder breadth were measured. Participant stature and body mass were measured using a Seca portable stadiometer (Model 213, Hamburg, Germany) and digital scale (Model 716, Seca, Hamburg, Germany) to the nearest 0.1 cm and 0.1 kg, respectively. Furthermore, head and neck circumference were measured using a Seca Ergonomic Circumference measuring tape (Model 201; Seca, Hamburg, Germany) to the nearest millimetre. Head circumference was measured horizontally, slightly above the eyebrows across the frontal bones of the skull. This was done perpendicular to the long axis of the face, above the ears and over the occipital protuberance, locating the maximal circumference. The tape was pulled tight to compress any hair. Neck circumference was measured in a horizontal plane, just below the thyroid cartilage/larynx, in accordance with previous research (Salmon et al., 2015). Shoulder (biacromial) breadth was measured with a Bahco tree calliper (Model 500 mm; SNA Europe, Cergy Pontoise, France) as the distance between the most lateral points on the left and right acromion processes. Participants were sat erect with their arms relaxes, and the measurements were taken standing posterior to the participant to the nearest 0.5 cm. Head and neck segment mass was estimated using the sex-specific estimation constants determined by Plagenhoef (men = 8.26% of body mass, women = 8.20% of body mass; Plagenhoef et al., 1983). The demographic and anthropometric data are presented as mean ± standard deviation.

3.3.2 Head Impact Telemetry System

Inertial loading of the head during 13 men's and seven women's matches was measured using the Protecht™ iMG system (Greybe, Jones, Brown & Williams, 2020). This bespoke iMG is instrumented with a 9-axis inertial measurement unit (IMU; LSM9DS1, STMicroelectronics, Genova, Switzerland) and tri-axial accelerometer (H3LIS331DL, STMicroelectronics, Genova, Switzerland). The accelerometer and gyroscope within the iMG have a 952 Hz sampling rate, with a 16-bit resolution and range of ± 200 *g* and ± 35 rad·s^{-2} over a 104 ms period. Raw data is captured by the iMG and transmitted to a receiving computer via radio frequency. The peak resultant linear acceleration (PLA; *g*) and rotational acceleration (PRA; rad·s^{-2}) were subsequently calculated from the time-series data. For the purpose of this thesis, these acceleration values were assumed to represent the acceleration of the head (Wu, Nangia, et al., 2016).

The iMGs used in this thesis were all custom formed to each individual player's upper dentition. Prior to any data collection, the fit of all iMGs was assessed by the research team to ensure they were well coupled to the teeth. Any iMGs that were not well fitted were adjusted or remade if this was not possible. This process was repeated, if necessary, until tight coupling was achieved and was vital to minimise STA (Bartsch et al., 2014; Camarillo et al., 2013; Greybe et al., 2020).

3.3.3 Video Verification Protocol and Impact Filtering

Video footage of women's matches was recorded using standard capture settings (30 Hz) via Sony video cameras at a minimum of two locations around the pitch to enable multiple angles of each event to be viewed. This minimised the chance of errors during video analysis. Men's match footage was obtained via the team performance analyst, recorded from a viewing tower using standard capture settings (30 Hz).

Video Verification.

A three-step video verification process was followed for all video and iMG-matched impacts (Figure 4, Appendix A). First, based on the video, it was verified that the player was on the pitch, they were involved in a contact event, and there was visible head acceleration. Second, the iMG data was verified by checking that the impact time aligned with the video-observed impact time, there were full waveforms for PLA and PRA, and the shape of the waveform was representative of a feasible head acceleration (Figure 5, Appendix A). Any maxima in the PLA and PRA data were excluded as no firm conclusions can be drawn as to whether these were head impact events or players biting or shouting (Figure 6, Appendix A). Specifically, the shape of the waveform cannot be evaluated, and filters cannot be applied.

Filtering

Recent research has demonstrated the importance of removing high-frequency noise from head impact data prior to analysis (Greybe et al., In press, 2021). Therefore, a fourth-order, zero-lag, low-pass Butterworth filter, with data-derived cut-off frequencies, was applied to the iMG data following video verification (Figure 5, Appendix A).

Video Analysis

Systematic video analysis was conducted in order to characterise each impact. Four analysts studied the video footage and coded impact events by activity type and cause of head acceleration (Table 1). All videos were analysed by more than one analyst and any discrepancies were re-analysed and discussed until a consensus was agreed.

For the purposes of this thesis, a tackle was defined as “when a ball carrier was contacted (hit and/or held) by an opponent without reference to whether they went to ground” (Quarrie & Hopkins, 2008). This is in line with previous research (C. W. Fuller et al., 2007; Tucker, Raftery, Fuller, et al., 2017). An “indirect impact to head” was classified as when an impact to another body part caused rapid acceleration of the head, with no direct impact to the head itself.

Table 1- Code for the characterisation of impact events for video analysis

Code		Activity
Activity	1	Tackle as tackler
	2	Tackle as ball-carrier
	3	Ruck
	4	Maul
	5	Lineout
	6	Scrum
Cause of acceleration	1	Indirect impact to head
	2	Direct head impact to 'soft' body part*
	3	Direct head impact to 'hard' body part**
	4	Direct head impact to ground
	5	Other

*Note. * 'soft' body parts include stomach, inner arm, thigh and chest, ** 'hard' body parts include head, knee, shoulder, back, elbow, shin and foot.*

To ensure iMG impacts were legitimate, a video verification protocol was followed (Figure 4, Appendix A). Video start time was recorded for all matches, meaning absolute time could be determined for all match footage. This enabled the time of observed video impacts to be aligned with the absolute time of the impacts recorded by the iMGs. Where multiple impacts were recorded in a short period of time, the impact waveforms were observed to determine which, if any, looked the most realistic by comparing them to examples of verified waveforms. Video footage was also closely analysed to determine whether multiple impacts could have been received at that time. The head acceleration data were not normally distributed ($p < .05$) and expressed as median, interquartile range (IQR), minimum and maximum values.

3.4 Neck Strength Assessment

3.4.1 Rationale

The majority of isometric neck strength research has been undertaken in either a seated or supine position (Caccese et al., 2018; Collins et al., 2014; Eckner et al., 2014). While this method has been shown to present valid and reliable measures, it has little relevance to contact sports like rugby, where contact events tend to occur while running and in a horizontal plane. Given that the primary purpose of this the present thesis was to evaluate neck strength using a method that would be functionally relevant to rugby. Consequently, the isometric neck strength testing apparatus (INSTA) was designed to assess neck strength in a simulated contact position (Salmon et al., 2015).

3.4.2 Isometric Neck Strength Testing Apparatus (INSTA)

A bespoke neck-strength-testing apparatus was designed and constructed for use in this thesis. Participants lay prone throughout the tests, kneeling on a raised platform (Figure 1). The participants feet were kept off the ground during testing to minimise the input of the leg muscles in the measures. Isolation of the neck musculature was achieved by securing the trunk with a four-point safety harness and instructing the participants to keep their arms behind their backs or keep their hands on their hips. This was designed to limit the input of accessory muscles.

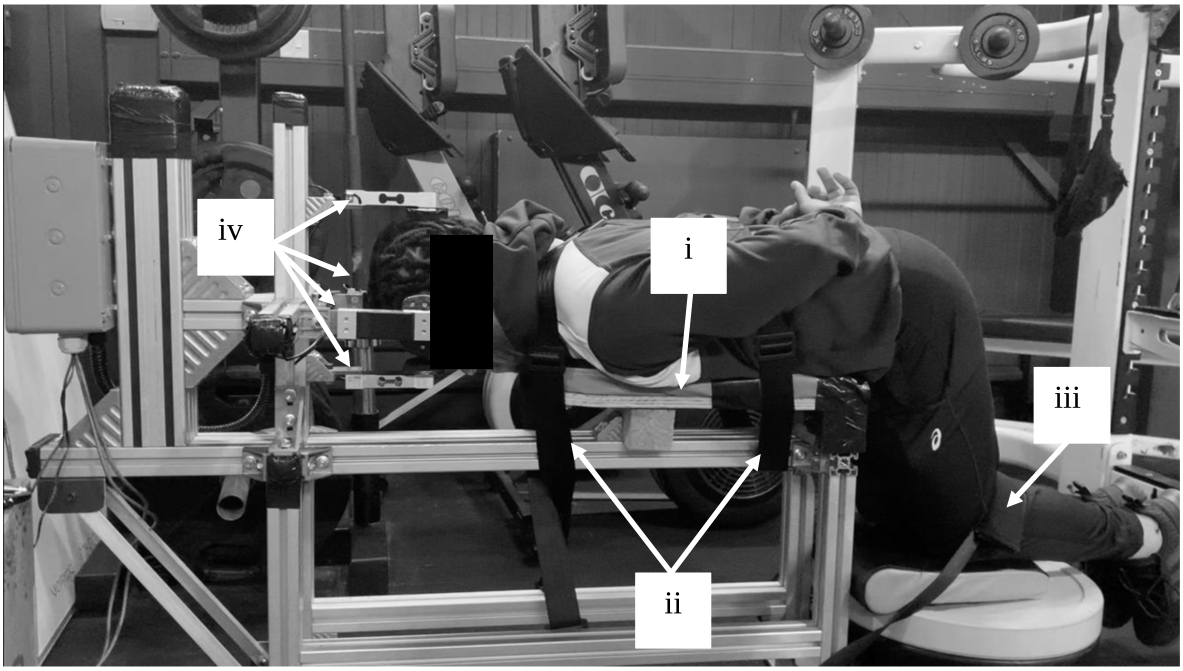

Figure 1- Participant strapped to the INSTA prior to peak neck strength testing

The INSTA consisted of: (i) an adjustable padded support bench; (ii) an adjustable four-point safety harness; (iii) an adjustable leg strap and; (iv) four adjustable 150 kg load cells (foam padded). Prior to testing, the bench, knee platform and load cells were adjusted to ensure the participants' back and neck were in a neutral position and hips were at a 90° angle. This position was standardised for all participants. A detailed description of the testing apparatus is provided in Appendix B.

3.4.3 Peak Neck Strength Testing Protocol

Prior to data collection, participants were shown a three-minute instructional video and completed a standardised warm-up, focused on the neck and upper back muscles. This consisted of a five-minute row followed by three sets of 10 repetitions of shoulder shrugs, shoulder circles, shoulder protractions and retractions, and neck half circumductions in each direction (Salmon et al., 2015).

Participants subsequently completed three familiarisation trials in each direction by performing sub-maximal contractions. Three isometric maximal voluntary contractions (MVC) were performed for three seconds in each direction: flexion (FL), extension (EXT), left lateral flexion (LL) and right lateral flexion (RL). Maximum force was recorded in Newtons (N) and was taken as the maximum from the trials for each direction. Each trial was followed by 20-30 seconds of rest, and trial directions were performed in a random order to minimise the effect of fatigue. Verbal encouragement was provided throughout each trial to encourage the participants to exert maximal effort (Mansell, Tierney, Sitler, Swanik, & Stearne, 2005). Neck-strength imbalance was defined as the difference between peak flexion and peak extension (FL – EXT; Dezman, Ledet, & Kerr, 2013). The neck strength data are presented as mean ± standard deviation.

3.5 Hypotheses

H_1: Male players have significantly greater playing experience than female players.

H_2: Male players have significantly greater stature, body mass, BMI and predicted head mass than female players.

H_3: There are significant differences in stature, body mass, BMI or predicted head mass between male forwards and backs.

H_4: There are significant differences in stature, body mass, BMI or predicted head mass between female forwards and backs.

Head, Neck and Shoulder Anthropometrics

H_5: Male players have significantly greater head and neck diameters and shoulder breadth than female players.

H_6: Male forwards have significantly greater head and neck diameters and shoulder breadth than male backs.

H_7: Female forwards have significantly greater head (a) and neck (b) diameters and shoulder breadth (c) than female backs.

H_8: Neck-to-head circumference proportion in forwards is significantly greater than that of backs in both sexes.

H_9: The proportion of neck circumference relative to shoulder breadth in both sexes is greater in forwards than in backs.

Neck Strength

H_{10}: Male players have significantly greater neck strength than female players.

H_{11}: Male forwards have significantly greater neck strength than male backs.

H_{12}: Female forwards have significantly greater neck strength than female backs.

H_{13}: There are greater differences between peak flexion and peak extension in males than females.

H_{14}: There are greater differences between peak left and right lateral flexion in males than females.

Head Impact Telemetry

H_{15}*:* Male players experience significantly greater head impact magnitudes than female players.

H_{16}*:* There are differences in HIE contact events between male and female players.

H_{17}*:* There are differences in HIE contact events between male forwards and backs.

H_{18}*:* There are differences in HIE contact events between female forwards and backs.

H_{19}*:* There are differences in HIE contact characteristics in tackles between male and female players.

H_{20}*:* There are differences in HIE contact characteristics in tackles between forwards and backs in both sexes.

H_{21}*:* There are differences in HIE contact characteristics in carries between male and female players.

H_{22}*:* There are differences in HIE contact characteristics in carries between forwards and backs in both sexes.

H_{23}*:* There are differences in HIE contact characteristics in rucks between male and female players.

H_{24}*:* There are differences in HIE contact characteristics in rucks between forwards and backs in both sexes.

Neck Strength and Head Impact Magnitude

H_{25}: Significant correlations exist between head and neck diameters and neck strength in male or female rugby players.

H_{26}: Significant correlations exist between head and neck diameters and head impact magnitude.

H_{27}: Significant correlations exist between neck strength and head impact magnitude.

3.6 Statistical Analysis

3.6.1 Descriptive Statistics and Neck Strength

All statistical analyses were completed using SPSS version 26 (IBM Corp., Armonk, NY). Prior to further analyses, all data was visually assessed for normality using histograms, as well as quantitative assessment using the Kolmogorov-Smirnov test. Significance was set a $p < .05$. Descriptive statistics were conducted for the males and females separately, as well as for forwards and backs within each group. A Mann-Whitney U test was used to assess differences in playing experience between male and females (H_1). Independent t-tests (or Mann-Whitney U where data was non-parametric) were used to determine any differences in anthropometric variables and neck strength between sex and position (H_{2} – $H1_{4,}$).

3.6.2 Contact Characteristics

Pearson's Chi-square tests were used to identify differences in HIE events and contact characteristics between sex and positions (H_{15} -,$H_{24,}$). Cramer's V correlations were used to estimate effect sizes, as the Chi-square table was larger than 2x2 and it is easily interpretable (< 0.1 = trivial, 0.1 - 0.2 = small, 0.3 - 0.5 = moderate and > 0.5 = large; Ferguson, 2009).

3.6.3 Correlations

Spearman's correlations were calculated between anthropometrics and both absolute and relative neck strength for both groups (H_{25}). Spearman's correlations were also calculated for both head and neck diameter and neck strength with head impact data (H_{26}, H_{27}).

Chapter 4: Results

4.1 Descriptive Statistics and Anthropometrics

4.1.1 Descriptive Statistics

Whilst there was no difference in age, male players had greater playing experience than females (13.1 ± 2.5 vs. 5.2 ± 3.3 years; $U = 9.5$, $p < 0.05$), therefore the hypothesis (H_1) was accepted. Sex differences were observed for stature, body mass, body mass index (BMI) and predicted head-neck segment mass ($p < .05$), as reported in Table 2. Based on these results, the hypothesis (H_2) was accepted.

Table 2- Descriptive statistics for stature, body mass, body mass index and predicted head mass: positional differences and sex differences

Characteristic	Sex	Position	n	Mean	SD	t/U	p	t/U	p
						Position Diff.		*Sex Diff.*	
Stature (cm)	M	F	12	188.9	10.2	3.0	.005*	8.8	< .001*
		B	15	179.3	6.1				
	F	F	8	163.6	6.5	-0.1	.896		
		B	13	163.9	6.2				
Body mass (kg)	M	F	14	105.4	8.2	6.6	< .001*	5.9	< .001*
		B	16	85.7	8.1				
	F	F	8	81.1	20.0	2.0	.071		
		B	13	65.9	6.9				
BMI (kg/m^2)	M	F	13	29.7	3.0	49.0	.045*	167.5	.016*
		B	15	26.8	2.4				
	F	F	8	30.1	6.0	13.5	.005*		
		B	13	24.5	1.9				
Predicted head mass (kg)	M	F	14	8.7	0.7	6.5	< .001*	6.0	< .001*
		B	16	7.1	0.7				
	F	F	8	6.6	1.6	2.1	.074		
		B	13	5.4	0.6				

*Note. Sex: M = male, F = female, position: F = forward, B = back, * = significant*

In males, significant positional differences were found for stature, body mass, BMI and predicted head-neck segment mass in male players ($p < .05$), so hypothesis H_3 was accepted. In contrast, no significant differences in stature, body mass or predicted head-neck segment mass were present between forwards and backs in the female group ($p > .05$), although the

BMI of forwards was significantly greater than that of backs ($p < .05$) so hypothesis H_4 was rejected.

4.1.2 Head, Neck and Shoulder Anthropometrics

Male players were characterised by greater head circumference, neck circumference and shoulder breadth than females, which translated to greater neck-to-head and neck circumference to shoulder breadth proportions ($p < .05$). Therefore, the hypothesis (H_5) was accepted. In males, head and neck circumference of forwards was significantly greater than backs (Figure 2), so H_6 was accepted. The head circumference of female forwards was not significantly different to that of backs ($p > .05$), so hypothesis H_7(a) was rejected. Female forwards, however, were found to have a significantly greater neck circumference and shoulder breadth than backs (Figure 2; $p < .05$), so for H_7, parts (b) and (c) were accepted.

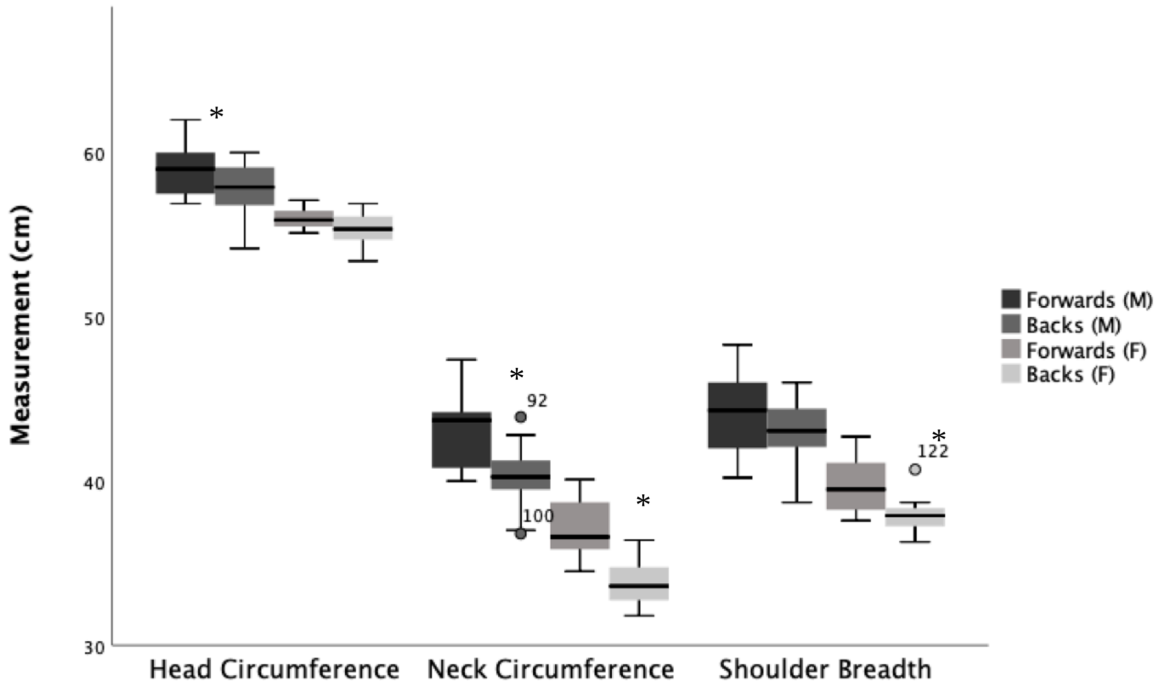

Figure 2- Box plot to show positional differences in head circumference, neck circumference and shoulder breadth in male (M) and female (F) players

*Note. * = significant*

The neck-to-head circumference proportion in forwards was significantly greater than that of backs in both sexes (males: 0.73 ± 0.03, or 73% vs. 0.70 ± 0.02, or 70%; $t(28) = 3.3$; $p < .05$; females: 0.66 ± 0.04, or 66% vs. 0.61 ± 0.02, or 61%; $t(17) = 3.4$; $p = .05$). Hypothesis H_8 was therefore accepted. Similarly, the proportion of neck circumference relative to shoulder breadth in both sexes was found to be greater in forwards than in backs, so H_9 was accepted (males: 0.99 ± 0.07, or 99% vs. 0.93 ± 0.05, or 93%; U = 62.0; p < .05; females: 0.94 ± 0.05, or 94% vs. 0.89 ± 0.03, or 89%; U = 22.0; p < .05).

4.2 Neck Strength

In total, 28 males (20.8 ± 1.4 years; 184.0 ± 9.5 cm; 95.1 ± 12.9 kg) and 22 females (20.6 ± 2.1 years; 163.8 ± 6.1 cm; 71.7 ± 15.0 kg) took part in baseline neck strength testing. Males demonstrated significantly greater absolute MVC than females in all four directions (p < .05; Table 6, Appendix C). Therefore, hypothesis (H_{10}) was accepted. Male forwards exhibited significantly greater absolute flexion and left lateral flexion than backs, but no differences were observed for extension and right lateral flexion (Figure 3). Overall average neck strength was greater in male forwards than male backs (p < .05), accepting H_{11}. No statistically significant differences were observed between female forwards and backs for absolute or directional MVC (Figure 3) or directional imbalances, so H_{12} was rejected.

Male players demonstrated significantly greater flexion-extension difference than female players (25.5 ± 55.6 vs. -0.6 ± 31.3 N; $t(48) = 2.1$; p < .05), (H_{13} accepted). No significant sex difference, however, was observed for the differences between left and right lateral flexion, rejecting hypothesis H_{14}. An incidental finding was that male players exhibited significantly greater time to peak MVC than females in all directions (p < .05; Table 6, Appendix C).

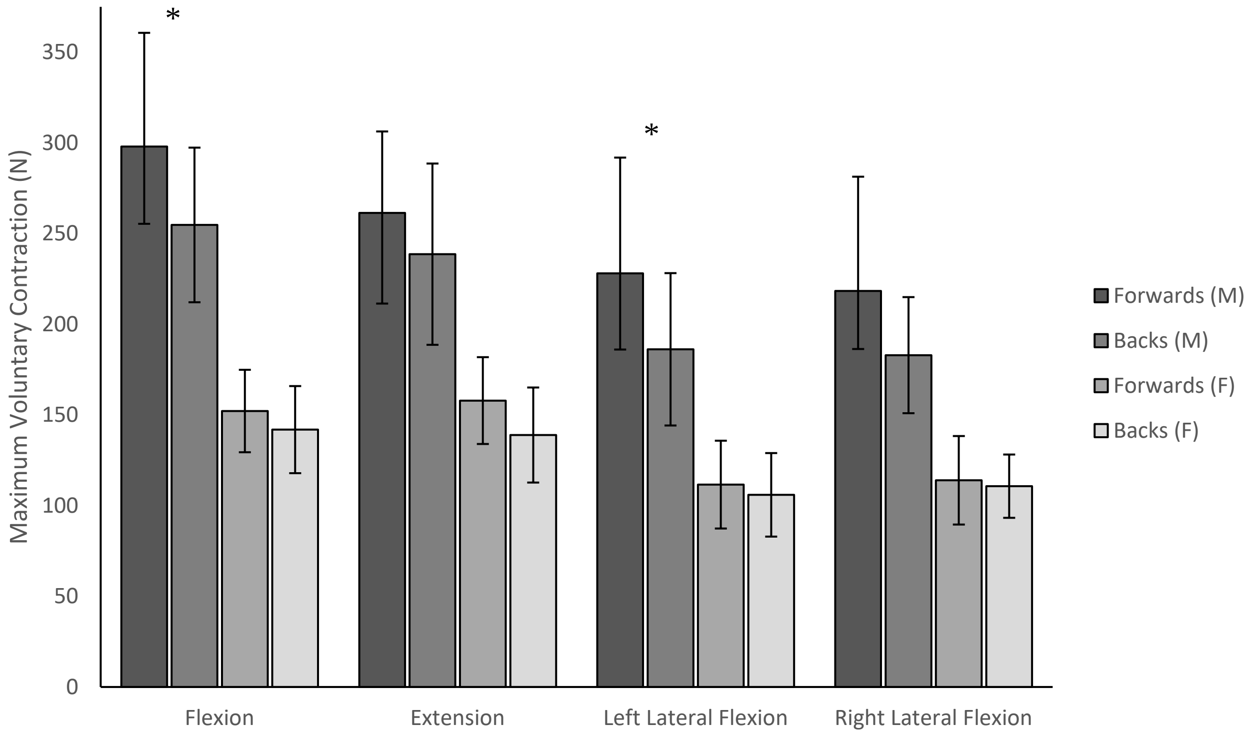

Figure 3- Absolute neck strength in forwards and backs for male (M) and female (F) players

*Note. * = significant*

4.3 Head Impact Telemetry

Bespoke iMGs were worn by 21 male (20.8 ± 1.5 years; 183.0 ± 7.6 cm; 93.4 ± 12.5 kg) and 13 female players (20.6 ± 1.9 years; 163.2 ± 4.7 cm; 68.1 ± 7.4 kg) over the 2019/20 BUCS playing season. Data were recorded over 13 and seven matches respectively, with the iMGs recording a total of 687 and 138 video verified HIEs respectively. Of these HIEs, 635 and 49 respectively were excluded following waveform-verification as timeseries data showed spurious patterns, representative of artefacts including bites and shouts. A total of 144 and 90 video and waveform-verified impacts respectively were included in the final analysis, as both PLA and PRA waveforms represented feasible head movement. Of these for males, there were 57 tackles, 49 carries, 34 rucks and four mauls and for females, 42 tackles, 31 carries, 16 rucks and one maul. Observation shows that male players experienced slightly greater median magnitude of PLA, PRV and PRA across all impacts (Table 3). Therefore, hypothesis (H_{15}) was rejected due to the non-significant difference in overall magnitudes.

Table 3- Overview of male and female HIEs

	Tackles			Carries			Rucks		
	PLA (g)	PRV ($rad \cdot s^{-1}$)	PRA ($rad \cdot s^{-2}$)	PLA (g)	PRV ($rad \cdot s^{-1}$)	PRA ($rad \cdot s^{-2}$)	PLA (g)	PRV ($rad \cdot s^{-1}$)	PRA ($rad \cdot s^{-2}$)
Male									
Median	13.9	10.7	874.9	11.6	11.0	848.1	12.2	8.1	819.3
IQR	7.1	5.3	496.6	8.0	5.8	653.1	4.4	3.9	297.1
Min	6.8	2.7	219.6	8.9	3.3	241.3	7.8	4.1	415.6
Max	47.53	25.7	2559.1	50.5	22.6	2133.0	23.2	17.4	2973.2
Female									
Median	11.9	9.0	874.9	11.7	12.2	757.3	10.2	5.6	639.2
IQR	7.3	7.6	558.1	5.6	7.5	523.4	2.7	4.5	256.1
Min	8.6	3.0	288.1	7.9	3.0	262.0	6.0	3.4	111.8
Max	44.3	29.8	3401.9	36.6	31.8	2345.9	24.5	18.2	1491.0

In 51.2% of the recorded and verified impacts for females, the player's head followed a poorly controlled whiplash action upon impact. This happened in particularly during carries, where players' heads out accelerated their bodies into the ground. Chi-square testing showed there to be no difference in HIE contact events between male and female players ($p = .520$, *ES* = trivial; Table 7, Appendix D). Therefore, the hypothesis (H_{16}) was rejected. Tackles accounted for a 7.1% greater proportion of HIEs in the female players than male players, however the distribution between other forms of contact was similar between sexes. Significant differences were observed in the occurrence of each HIE classification between male forwards and backs ($p < .05$, *ES* = small; Table 4), accepting hypothesis H_{17}. In male forwards, there was a near even distribution of HIE classification between tackles, carries and rucks. In backs however, tackles accounted for the majority of HIEs at 53.5%. There was no apparent difference in PLA, PRV or PRA between the positional groups. In females, there was no significant difference in the occurrence of each HIE classification between forwards and backs ($p = .368$, *ES* = small; Table 4), rejecting hypothesis H_{18}.. Observational analysis showed no difference in PLA or PRA between the positional groups, however, forwards experienced greater PRV than backs.

Table 4- HIE classification occurrence for male and female forwards and backs

HIE Characteristic	Forwards (n = 92)	Backs (n = 45)	X^2	p	Cramer's V	Interpretation of Effect Size
Males						
Tackle	31 (33.7%)	24 (53.3%)				
Carry	31 (33.7%)	16 (35.6%)	8.703	.034*	.252	Small
Ruck	27 (29.3%)	5 (11.1%)				
Maul	3 (3.3%)	0				
Females						
Tackle	21 (39.6%)	21 (56.8%)				
Carry	21 (39.6%)	10 (27.0%)	3.159	.368	.187	Small
Ruck	10 (18.9%)	6 (16.2%)				
Maul	1 (1.9%)	0				

*Note. * = significant*

4.3.1 Tackles

An overview of the magnitudes of the 57 and 42 observed tackle HIEs in male and female players respectively, is presented in Table 3. Of the male tackle HIEs, the majority were caused by direct impacts with 'hard' (35%) and 'soft' (27%) body parts. In female players, the majority of tackle HIEs were caused by direct impacts with 'hard' body parts (62%) and direct impacts with the ground (21%). Of the direct impacts with 'soft' body parts in male players, the majority occurred due to head contact with the upper leg (n = 10; PLA, 13.9 ± 5.9 g; PRV, 11.4 ± 4.2 rad·s^{-1}; PRA, 1206.2 ± 460.8 rad·s^{-2}). Direct impacts with 'hard' body parts in males mostly occurred with the hip (n = 8; PLA, 14.9 ± 6.2 g; PRV, 10.1 ± 4.0 rad·s^{-1}; PRA, 1145.4 ± 409.5 rad·s^{-2}). In females, the majority of direct impacts with 'hard' body parts occurred due to head contact with the shoulder (n = 8; PLA, 11.8 ± 12.9 g, PRV, 10.5 ± 12.8 rad·s^{-1}, PRA, 907.4 ± 920.9 rad·s^{-2}) or hip (n = 7; PLA, 11.9 ± 7.2 g; PRV, 9.0 ± 6.9 rad·s^{-1}; PRA, 808.8 ± 616.1 rad·s^{-2}). An overview of each major classification of tackle HIE in males and females is presented in Table 8, Appendix D.

There was no significant difference in tackle HIE cause of impact between male and female players (p < .05, *ES* = moderate; Table 5). Hypothesis (H_{19}) was therefore rejected. No difference in acceleration magnitude between males and females was observed. In tackles, direct impacts with hard body parts accounted for the greatest number of HIEs in both sexes, however they accounted for 26.8% more of tackle HIEs in females. Direct contact with soft body parts was the cause of the least HIEs in female players, whereas direct impacts with the ground were the least common classified cause of HIE in male players. No significant difference in HIE cause of acceleration frequency for tackles was observed between male or female forwards and backs (p = .281, ES = moderate; p = .515, ES = small; Table 11, Table 12, Appendix D; H_{20} rejected). Observation showed that backs experienced a greater median PRV than forwards during tackles in male players. In contrast, female forwards experienced greater median PRV than backs during tackles, but no difference in PLA or PRA.

4.3.2 Carries

An overview of the magnitudes of the 49 and 31 observed carry HIEs in male and female players, respectively, is presented in Table 3. The majority of carry HIEs in males were caused by indirect impacts (51%), whereas in females the most common HIE cause was direct head impacts with the ground (42%). In these head to ground impacts, the female players demonstrated an observable lack of control, with their heads following a whiplash action into the ground. An overview of each major classification of carry HIE for both sexes is presented in Table 9 (Appendix D).

Chi-square testing showed there to be a significant difference in carry HIE cause of impact between male and female players ($p < .05$, ES = moderate; Table 5; H_{21} accepted). Indirect impacts were the most common cause of HIE in male players, accounting for over half of all carry HIEs. In the female players, direct head contact with the ground was the most common cause of HIEs during carries at 41.9%. Observation showed that when female players were tackled and fell to the ground, they had little control of their head motion. This resulted in a whiplash action where their head out accelerated their bodies as they hit the ground. This only occurred once in the male players. No large differences in acceleration magnitude were observed between sexes. No significant difference was observed in HIE cause of acceleration frequency for carries between forwards and backs in either sex (p = .626, ES = small; p = .657, ES = small; H_{22} rejected). Observational analysis showed that male backs experienced slightly greater median PLA than forwards, and female backs experienced slightly greater median PRV than forwards, however, there were no further differences.

4.3.3 Rucks

An overview of the magnitudes of the 34 and 16 observed ruck HIEs in male and female players, respectively, is presented in Table 3. Of the ruck HIEs for males, the majority were caused by direct impacts with ‘hard’ body parts (41.2%). Similarly, in females the majority of ruck HIEs were caused direct impacts with ‘hard’ (63%) body parts. In males, direct impacts with ‘hard’ body parts occurred most frequently with the head (n = 5; PLA, 14.7 ± 2.5 g; PRV, 6.6 ± 0.5 rad·s^{-1}; PRA, 715.5 ± 221.8 rad·s^{-2}). In females, direct impacts with ‘hard’ body parts occurred more frequently with the shoulder (n = 5; PLA, 10.3 ± 1.9 g; PRV, 4.9 ± 4.7 rad·s^{-1}; PRA, 581.8 ± 495.3 rad·s^{-2}) than any other body parts. An overview of each major classification of male and female ruck HIEs is presented in Table 10 (Appendix D).

Chi-square testing showed no significant difference between male and female players for the cause of impact in ruck HIEs (p = .387, *ES* = small; Table 5; H_{23} rejected). Male players experienced greater magnitudes of PLA, PRV and PRA than females in rucks. There was no significant difference in cause of acceleration frequency for rucks between forwards and backs in either sex (p = .818, *ES* = small; p = .474, *ES* = moderate; H_{24} rejected). PRV and PRA was greater in male forwards than backs, however observation showed no difference in PLA. No differences were observed in PLA, PRV or PRA between positional groups in females.

Table 5- HIE cause of acceleration classification occurrence in tackles, carries and rucks for male and female players

Contact Characteristic	Male	Female	X^2	*p*	Cramer's V	Interpretation of Effect Size
TACKLE	57	42	15.294	.004*	.393	Moderate
Indirect impact	11 (19.3%)	4 (9.5%)				
Direct impact with 'soft' body part	15 (26.3%)	1 (2.4%)				
Direct impact with 'hard' body part	20 (35.1%)	26 (61.9%)				
Direct impacts with the ground	7 (12.3%)	9 (21.4%)				
Other	4 (7.0%)	2 (4.8%)				
CARRY	49	31	14.351	.006*	.424	Moderate
Indirect impact	25 (51.0%)	10 (32.3%)				
Direct impact with 'soft' body part	6 (12.2%)	0				
Direct impact with 'hard' body part	7 (14.3%)	7 (22.6%)				
Direct impacts with the ground	6(12.2%)	13 (41.9%)				
Other	5 (10.2%)	1 (3.2%)				
RUCK	34	16	4.142	.387	.288	Small
Indirect impact	8 (23.5%)	2 (12.5%)				
Direct impact with 'soft' body part	7 (20.6%)	4 (25.0%)				
Direct impact with 'hard' body part	14 (41.2%)	10 (62.5%)				
Direct impacts with the ground	1 (2.9%)	0				
Other	4 (11.8%)	0				

*Note. * = significant*

Statistical analysis including HIE magnitude was not performed for comparisons as only iMG data with full waveforms that had been video-verified could be included. This means that the data included in this thesis could misrepresent the full data set if all video verified HIE impacts had consisted of full waveforms.

4.4 Neck Strength and Head Impact Magnitude

4.4.1 Anthropometrics and Neck Strength

Spearman's correlations were calculated for both male and female anthropometrics and absolute neck strength (Table 13, Appendix E). There were significant positive moderate correlations between body mass and absolute neck strength for all directions in males ($p < .05$), however no significant correlations in females. Male players exhibited significant positive strong and moderate correlations between BMI and flexion and left lateral flexion, respectively ($p < .05$). No significant correlations between BMI and neck strength were present for any direction in females ($p > .05$). Significant moderate and strong positive correlations were observed between neck circumference and flexion, left and right lateral flexion in male players ($p < .05$), however no significant correlations were present in females. Neck-to-head circumference proportion was significantly moderately and strongly positively correlated with absolute neck strength for flexion, left and right lateral flexion in males ($p < .05$). In contrast, no significant correlations were present in female players. In males, there was a significant moderate positive correlation between neck circumference to shoulder breadth ratio and left lateral flexion ($p < .05$). No further significant correlations were present between anthropometrics and neck strength in either sex. The hypothesis (H_{25}) was accepted in the male group due to multiple significant correlations being present and rejected in the female group due to the absence of a statistically significant correlation.

4.4.2 Anthropometrics and Head Impact Telemetry

Spearman's correlations for head and neck anthropometrics and head impact telemetry measures are presented in Table 14 (Appendix E). A small significant positive correlation was observed between shoulder breadth and PRV in female players ($p < .05$). A small positive significant correlation was also observed between neck to shoulder ratio and PRA in male players ($p < .05$). No further significant correlations were observed between any head and neck anthropometric measures and PLA, PRV or PRA ($p > .05$). Due to a lack of significant correlations, the hypothesis (H_{26}) was accepted in both the male and female players.

4.4.3 Neck Strength and Head Impact Telemetry

Spearman's correlations for neck strength and head impact telemetry measures are presented in Table 15 (Appendix E). There was a significant small negative correlation between flexion and PRA in male players ($p < .05$). There were also significant small negative correlations between extension and PLA and PRA in male players ($p < .05$). Female players demonstrated a small positive significant correlation between both absolute and relative right lateral flexion and PRV ($p < .05$). All other correlations between neck strength and head impact telemetry measures were insignificant in both male and female players ($p > .05$). Due to a lack of significant correlations, the hypothesis (H_{27}) was accepted in both the male and female players.

Chapter 5: Discussion

5.1 Sex Differences and The Gender Data Gap

5.1.1 Rugby Rules, Protocols and the Androcentric Research Base

A substantial gender data gap exists across the sports and medical sciences (Costello et al., 2014) which is particularly significant in rugby. Rugby is the only collision sport in the world where both male and female athletes play under the same rules, with the same equipment (Gabb, 2018). The research underpinning all training and injury prevention practices in rugby, however, is highly androcentric. Meaningful data on injuries and training techniques has almost exclusively been collected from male players in the past (Cross et al., 2019; Hendricks et al., 2020; Kemp et al., 2018). The latest recommendations for video analysis to identify possible brain injuries in rugby, for example, only reference articles which include male players (Hendricks et al., 2020). The conclusions drawn from studies such as these are generalised and applied across the entire sport, including female players. These studies form the basis of rugby's law changes and training methods, which may in fact be detrimental to female players.

This thesis identified significant differences between male and female university-level rugby players in head and neck anthropometrics and isometric neck strength in four directions (both absolute and relative to body mass). Similarities between sexes in overall head impact magnitude (median PLA and PRA) were identified, despite the significant body size, neck size and neck strength differences identified. Substantial kinematic differences in head impact mechanism were identified, however, this was outside the scope of this thesis.

Given the reported importance of head and neck stabilisation for player safety in rugby (Collins et al., 2014; Eckner et al., 2018; Mansell et al., 2005), the differences identified between sexes in the current thesis are cause for significant concern. While both groups were of similar ages playing the same game with the same rules at comparable relative levels, there was a significant difference in years of playing experience (13 and 5 years for males and females, respectively). Most of the female players began playing rugby as adults upon starting university and had not graduated through the age grade pathway that male players typically follow. This pathway involves all players aged six to eighteen years and enables players to develop their skills systematically as they progress through the age grades.

Following the England Rugby player progression pathway, the sexes are mixed up to and including Under-11. This could explain the lack of uptake of female players of a young age, especially if parents feel uncomfortable with their young girls playing a contact sport with boys from age 9. Parental education strategies may increase female participation at a young age by educating parents on the minimal contact involved at a young age, and the lack of strength discrepancies between sexes at this stage.

The progression pathway includes being taught proper tackle technique and how to fall to minimise the chance of injury. Proficient tackle contact skills have been associated with a reduced risk of injury for both the tackler and the ball carrier (Burger et al., 2017, 2016; Hendricks et al., 2015, 2016). Therefore, an equal opportunities policy where both groups follow similar training programs, as in the current thesis, may be detrimental to female players with limited playing experience. The differences identified in all measured parameters, and the respective implications of these for player health and safety, are evaluated in the subsequent sections.

5.2 Head and Neck Anthropometrics: Sex Differences and Playing Position Trends

5.2.1 Anthropometrics

The findings for the anthropometrics of the male group in the current thesis were comparable with existing literature in this area (Brooks, Fuller, Kemp, & Reddin, 2005; Salmon, 2014). Male forwards were significantly taller and heavier, had a greater predicted head-neck segment mass, neck circumference and head circumference than backs. This clear distinction between the physiques of positional groups was not present in females.

While comparisons with existing literature are relevant for the male group, data on female rugby is lacking. As an alternative, the stature and body mass of positional groups for the female group can be compared to publicly available data on the world number one women's team (NZ Black Ferns 2020 squad). The average stature, body mass and BMI for forwards was significantly greater than backs (Table 16, Appendix F), whereas differences in the current group were only identified for BMI. These findings are in agreement with existing strength and condition literature, that appropriate training promotes physical adaptation to the loads sustained in rugby substantial corresponding measurable physical changes.

The differences in head and neck size between sexes is consistent with previous research, with males having greater head and neck circumferences than females (Mansell, Tierney, Sitler, Swanik, & Stearne, 2005; Chapter 2.5.3). The lack of positional anthropometric differences present in the females compared to the males may be a result of physical adaptations acquired from earlier positional specialisation in males. There are, however, many physiological differences between sexes, including hormones which also may explain the lack of positional adaptation in the female players (Brown, 2008).

5.2.2 Neck Strength

The findings for isometric neck strength for the male players in this thesis were comparable to those in the existing literature (Geary, Green, & Delahunt, 2013; Olivier & Du Toit, 2008; Salmon, 2014). The findings are consistent with other studies in which males exhibited stronger neck flexors and extensors and greater neck girth than females (Garcés, Medina, Milutinovic, Garavote, & Guerado, 2002; Jordan, Mehlsen, Bülow, Østergaard, & Danneskiold-Samsee, 1999; Maeda, Nakashima, & Shibayama, 1994; Tierney, 2004).

Research examining peak isometric cervical force using fixed frame dynamometers in male professional rugby players reported similar flexion results to this thesis, but significantly greater values for extension and lateral flexion measures (Naish, Burnett, Burrows, Andrews, & Appleby, 2013).

Male forwards exerted greater MVC in all directions than backs, with differences observed for flexion and left lateral flexion. This was not true for the female group, where no differences were observed between positional groups. The strength difference between positional groups in the male group may be reflective of the loads placed on the players' necks during contact play in training and matches (Salmon, 2014). Correlations were observed for body mass, BMI and neck circumference with absolute neck strength measures in male players, but none were observed in females. Similar findings have previously been reported for males, with body mass and neck circumference being correlated with peak force (Salmon, 2014). Comparable results have also been reported in college wrestlers and judo athletes, with positive correlations observed between cervical extensor cross-sectional area and extension peak force (Tsuyama et al., 2001). Due to the lack of research available in females, comparisons with similar literature cannot be made for the female group.

The two groups in this thesis were prescribed similar training programs. Therefore, similar relative positional differences would be expected in the female group as the male group. The lack of differences in the female group, however, may not reflect training differences, due to their training not being focused on female-derived data, and therefore not being as effective at building strength. Observations of the female training sessions also showed the females to not train as intensely as the males and may also explain the lack of positional strength adaptation.

The male players had greater playing experience than the female players and thus more advanced development in their respective positions than females. The male positional strength difference, however, could also be attributed to being selected for their positions based on size and strength prior to playing, rather than adaptations from the training itself. This difference is further exacerbated by the flexible position approach that was observed in the female group, where players tended to change position between (and even during) seasons. Within the female group, a midfield back (13) transitioned to the back row (7) at the start of the observed season. Over the same time period, a back-row player (7) transitioned to the front row (3). This highlights both the lack of playing experience and the lack of position specificity in the female game compared to the male game, which likely contribute to the lack of physical differences between female forwards and backs. The positional differences, or lack thereof, in the female group suggests that rather than following the training program that works for the male players, a training program should be developed specifically for female players. The females studied in this thesis have weaknesses in different places to the males. These should be targeted with a female-specific training program based on an objective evidence base about female athletes.

5.3 Head Impact Telemetry

5.3.1 Inertial Loading and Head Impact Telemetry in Rugby

Most of the iMG data collected in this thesis were low in magnitude, which is consistent with previous studies (Kieffer, Vaillancourt, Brolinson, & Rowson, 2020; King, Hume, Brughelli, & Gissane, 2015; King, Hume, Gissane, & Clark, 2017; Rowson et al., 2009). Kieffer and colleagues (2020) investigated sex-specific differences in club collegiate rugby teams using custom fit instrumented mouthguards over the course of a year. Similar to the current findings, the study reported males and females to experience similar magnitudes of PLA across all impacts. Relative to their strength and size, these magnitudes may have a greater propensity to cause injury in females than males. Studies comparing concussion rates report females to sustain a greater number of concussions than males (Frommer et al., 2011; Gessel, Fields, Collins, Dick, & Comstock, 2007; McClincy, Lovell, Pardini, Collins, & Spore, 2006). These findings may suggest that females tend to experience greater magnitudes of head impacts or have a lower pathological tolerance to them. Previously, concussion disparity has been attributed to anatomical and biomechanical differences between males and females, so the findings of this thesis may support this (Mansell, Tierney, Sitler, Swanik, & Stearne, 2005). More research into sex differences in head impact magnitude, and their relation to brain injuries needs to be done in order to better understand these findings.

Research using instrumented mouthguards in amateur rugby reported a mean PLA of 22 g and PRA of 3990 rad·s^{-2} (King, Hume, Brughelli, & Gissane, 2015). These results followed a similar array of magnitudes, skewed towards lower values, however the median magnitude was substantially greater than reported in the current thesis. The iMG system used by these authors consisted of a boil-and-bite moulding technique and reported a ‘bulky’ fit. This implies that the sensor-skull coupling was likely poor, potentially explaining the difference in magnitudes.

Research in youth rugby league using the X-Patch™ skin-mounted sensors reported a PLA range of 10 to 123 g and a PRA range of 89 to 22,928 rad•s^{-2} (King et al., 2017). Literature using a similar methodology in Australian-rules football reported PLA ranging from 10 to 153 g and PRA ranging from 130 to 21,890 rad•s^{-2} (King, Hecimovich, Clark, & Gissane, 2017). These values are much greater than the values reported in this thesis, highlighting the inadequacies of data generated by systems with indirect sensor-skull coupling.

The current thesis is one of very few utilising validated, well-coupled head impact sensors in rugby and including a female group (Kieffer et al., 2020). Many previous head impact telemetry studies in rugby union and league have utilised sensors worn as patches adhered to the mastoid process (King, Hume, Gissane, & Clark, 2016; King, Hume, Gissane, Kieser, & Clark, 2018; King, Hume, Gissane, & Clark, 2017). While the head acceleration values reported in these studies are indeed comparable to those published in American football using helmet-mounted sensors, there are concerns regarding both approaches (Jadischke et al., 2013; Wu, Laksari, et al., 2016). Tight sensor-skull coupling is imperative for accurate and reliable readings from head inertial sensors (Bartsch et al., 2014; Camarillo et al., 2013). Given the high-magnitude, short duration nature of head impact events, the implications of STA will be greater than for activities such as walking, where it is an acknowledged problem (Darwesh, Hafez, Aboeleneen, El-Banna, & El-Gendy, 2018; Gao & Zheng, 2008).

Literature exploring the movement of skin-mounted sensors relative to the underlying bone (STA) analysed high speed video during soccer heading (Wu, Nangia, et al., 2016). This reported poor coupling of the skin-mounted sensors to the skull, leading to over-estimations of PLA and PRV. The sensor values reported in this thesis are significantly lower than those previously reported in rugby and league (King et al., 2018; King et al., 2017), as well as American football (Steven Rowson et al., 2011). This is likely due to the mouthguards in this thesis being bespoke for each player to ensure tight coupling with the teeth and skull. This thesis also used a multi-step verification process to ensure only true positive HIEs were included in analysis. Previous studies, while mentioning video footage, do not report individual events, likely due to the complexity and time requirements needed to identify particular discrete events. Thus, the majority of data in existing head impact literature may include many false positives.

Accurate recording and reporting of head impact data is vitally important, given the importance of brain health to human health and wellbeing. Given the discrepancy between the data presented in this thesis and previously reported data, minimum standards for recording and reporting of head impact data should be introduced. This would enable the data to be deemed valid and reliable. These standards can include minimum coupling requirements, filtering settings and all head impact telemetry data should be accompanied by video verification assurances by authors. In the current thesis, it also became obvious that the analysis of waveforms for impact verification was required. Despite the addition of an infrared proximity sensor in the iMG design, many impacts resembling bite and shout artefacts were still recorded. In addition, a design feature of this system enabled the transmission of the peak value of a given impact only, in instances where many impacts were transmitted in a short time period. For these ‘maxima’ impacts (Figure 6, Appendix A), the waveform resembled a flat line at the level of the maximum numerical value and no waveform characteristics were observable. Given the inclusion of bite and shout artefacts in the data collected by the system, it was not possible to verify the impacts transmitted as maxima values, so these were excluded. For the men’s team, this meant that 779 impacts recorded while players were active on the field was reduced to 144 video and waveform-verified impacts in the final dataset. For the women's team, 173 impacts recorded in play was reduced to 90 video and waveform-verified impacts. While this thorough verification process did compromise the size of the dataset, it ensured the scientific integrity of the study. This is vital in the field of head impact telemetry in sport, where many studies include false positive impacts in addition to using poorly coupled sensors.

5.3.2 Characterising the Causal Environments of Head Impact Events

Over the course of both BUCS seasons, median PLA, PRV and PRA values were similar for the male and female groups. The biomechanics that caused the impacts, however, differed greatly between the sexes and in some cases between positions. Relatively little is known about the magnitude and influencing factors for head kinematics during the tackle in rugby (Tierney, Gildea, Krosshaug, & Simms, 2019). Although literature is beginning to emerge in the field, the data is predominantly androcentric meaning the findings may not be applicable to female players. It is only possible to investigate head kinematics if game video is directly correlated to the impacts recorded by the mouthguards, hence the dearth of literature available currently.

Tackles accounted for 40% and 47% of HIE events in the male and female groups, respectively. Male players exhibited similar PLA, PRA and PRV to females for tackles. Carries accounted for 34% of HIEs in both groups, with the magnitudes of PLA, PRV and PRA all being similar. These results are consistent with literature reporting that tackles are responsible for the greatest proportion of head impacts and injury in the professional game (Tucker, Raftery, Fuller, et al., 2017). The greater proportion of HIEs to tacklers in this study is also consistent with previous research, suggesting that during tackles 72% of HIAs occur to the tackler (Tucker, Raftery, Kemp, et al., 2017a). Male players experienced a greater incidence (34% of HIES) and greater magnitudes of PLA, PRA and PRV than females (18% of HIEs) during rucks. The incidence in the male group is much greater than reported in the elite game (19% of HIEs; G. J. Tierney, Lawler, Denvir, McQuilkin, & Simms, 2016). In contrast, the 19% reported in the elite game is in line with the incidence recorded in the female group (Tierney, Lawler, Denvir, McQuilkin, & Simms, 2016).

Male players would be expected to experience greater HIE magnitudes than females due to their significantly greater size, strength and speed (Quarrie & Wilson, 2000; Sedeaud, Vidalin, Tafflet, Marc, & Toussaint, 2013). The magnitudes of the HIEs recorded in this thesis, however, did not differ substantially between sexes. This implies that females are experiencing a greater relative magnitude of inertial loading, which could be attributed to differences in contact technique. Given the reported differences in axonal diameter and structure, injury burden may be greater for females experiencing similar head impacts magnitudes to males (Dollé et al., 2018). Female axons are more prone to the rupture of microtubules during rapid stretching conditions associated with head accelerations. Consequently, for the same magnitude of HIE, female axons experience more microtubule breakage and a greater influx of ions that male axons (Dollé et al., 2018).

The contact situations in which impacts were recorded in the male group were similar to findings in the male professional game, but this is not true for female impacts (Cross et al., 2019; Tierney et al., 2019). These differences in contact technique may be a consequence of the lack of playing and training experience in the female players compared the males. They may also be attributed to training techniques being based on male data, and therefore not as applicable to the female game. When females play a match, the contact techniques they use in training may not work effectively on other female players, and therefore their contact technique may differ.

5.3.3 Tackles

Although no difference in HIE magnitude was observed, there was a difference in the cause of the tackle impacts between sexes. The majority of female tackle HIEs were caused by direct impacts of the head with a ‘hard’ body part (62%), these mainly being the ball-carrier’s shoulder followed by the hip. The median PLA experienced during direct impacts with hard body parts was greater than those experienced during other causes of impact. The cause of tackle HIEs in male players was more evenly distributed, although direct head impacts with ‘hard’ body parts were the cause of the most impacts (35%). These mainly consisted of direct impacts with the opposition’s hip.

The similarity in HIE magnitude experienced by both sexes, despite the differences in tackle technique, suggests that the female players tackle in a way that makes them more susceptible to HIEs. The findings suggest that they tend to tackle in a more upright position, making contact with the opposition’s shoulder. Upper body tackles have been reported to be the main cause of direct head impacts for the tackler (Tierney, Denvir, Farrell, & Simms, 2018; Tierney et al., 2016). Tackling in this position has a high propensity to cause direct head impacts with hard body parts, such as the shoulders and head, that would be avoided with a lower tackle height.

Tackles where the head makes direct contact with hard body parts have previously been reported to have the greatest propensity to cause a head impact assessment (Tucker, Raftery, Kemp, et al., 2017a). This is consistent with the current thesis, in which direct impacts with ‘hard’ body parts accounted for the greatest number of tackle HIEs. Previously, direct impacts to the head have been associated with greater magnitudes of PLA in both male and female collegiate rugby players (Kieffer et al., 2020). In contrast, the findings in the current thesis did not demonstrate substantial differences in head impact magnitude.

Both groups also demonstrated a large proportion of low tackles, where their head made direct contact with the ball-carrier’s hip. Tackling below the waist of the ball-carrier has been shown to results in the tackler’s head making direct contact with the hips or fast-moving legs (Hendricks & Lambert, 2010). Data suggests that tackling the ball carrier in their centre of gravity reduces the chances of an injury outcome, however, tackles to this area were uncommon in both groups (Burger et al., 2016).

5.3.4 Carries

No differences in HIE magnitude were observed between males and females for carries, but differences in cause of impact were observed. The majority of carry impacts observed in the female group were caused by a direct head impact with the ground (42%), whereas the majority of impacts in males were caused by indirect impacts (51%).

In female carries, PLA and PRA were substantially greater for direct head impacts with the ground than any other cause of impact. These magnitudes were also substantially greater than male players experienced during head to ground impacts from carries. It was directly observed that women tend to hit their head on the ground with a whiplash action. This was only observed twice in the male group, where in one instance the player was unconscious before hitting the ground. Impacts where the head makes direct contact with the ground have previously been shown to have a high risk of concussion, with a 20-fold increase in risk when compared to direct head impacts with the trunk (Cross et al., 2019). Although indirect impacts were the most common cause of carry HIE in the male group, the magnitude of the impacts was no greater than that of other causes of acceleration. Similar to current findings in the female group, Kiefer et al. (2020) reported direct impacts to the head to be associated with greater PLA magnitudes than indirect (body) impacts. In the male group, this was true for direct head impacts with the body (‘soft’ and ‘hard’ body parts) but not for impacts with the ground.

Tackles that are made to the upper trunk (but below the legal limit) can result in significantly greater inertial head kinematics for the ball carrier than lower tackles (Tierney, Richter, Denvir, & Simms, 2018; Tierney & Simms, 2017). Literature suggests that the head kinematics resulting from an impact to the body (i.e. indirect impacts) are inversely related to the distance of the impact from the head (Kim, Voloshin, & Johnson, 1994; Tierney, Richter, et al., 2018). Although tackle height was not recorded for ball-carriers in the current thesis, the lack of substantial difference between indirect impacts and other causes suggests that the majority of tackles would have been made below the upper trunk.

5.3.5 Rucks

The male group displayed slightly greater magnitudes of HIE than the female group for all measures in rucks. However, no differences in cause of impact were observed between the sexes. In both male and female players, the majority of ruck HIEs were caused by direct head impacts with hard body parts (41% and 63%, respectively). In the male group, the most commonly impacted body part was the head, accounting for 15% of recorded impacts. In the female group, the most commonly impacted body part was the shoulder, accounting for 31% of recorded impacts. This suggests that ruck technique differs between males and females and should be further investigated in combination with head impact magnitude to better understand sex-specific risks.

When compared to similar research, the male group experienced direct head-to-head impacts slightly less than the 20% observed in male professional rugby (Tierney et al., 2016). Male players in this thesis also demonstrated a lower incidence of direct contact with all other observed body parts. The female group experienced a similar incidence of head-to-shoulder impacts as the professional males (30%; Tierney et al., 2016), with no previous female data for comparison. Some rucks could, however, not be analysed due to the inability to accurately classify impacts from the video footage, so actual incidence could differ.

These results give an idea of what contact mechanisms may be associated with the greatest risk of head impacts, and therefore brain injury. However, conclusions regarding the injury risks associated with contact events requires correlations with injury data, which is beyond the scope of this thesis (Tierney et al., 2019). Although published injury data to date is very androcentric, a wider evidence base is required in both the male and female games to draw substantial evidence-based conclusions from head kinematic data. It is clear from these results that females' tackle and ruck technique differ greatly from males, influencing their head kinematics and potentially chances of sustaining brain injuries. Research focussing solely on female impact characteristics is crucial to develop effective strategies to reduce injury risk in the female game. While efforts to reduce the incidence of high-risk contact events should be made, it is also important to continually monitor the effects of any training or law changes implemented; it is possible that reducing the risks in one area of the game may in turn increase the risk in another area (Hagel & Meeuwisse, 2004).

5.4 Anthropometrics and Head Impacts

A smaller mean neck to head circumference proportion has been reported to be associated with a greater risk of concussion (Collins et al., 2014). Although correlations were observed between neck circumference to head circumference proportion and neck strength in three directions in males, no correlations were present with inertial loading measures. In order to adequately assess these relationships, a substantially greater sample size of both participants and the number of verified head impacts would be required. This is because many variables will influence the magnitude of head impact events. These include factors such as contact speed, angle of impact, angle of falling and playing surface, in addition to the size and strength of a player's neck. Previous literature reports that physically active females experience greater head-neck segment acceleration than males when their head is subjected to the same load (Tierney et al., 2005). This difference has been attributed to females having a lower head mass and neck girth than males, leading to lower head-neck segment strength and stiffness (Mansell et al., 2005). Male players would be expected to experience greater forces than females during contact events due to their larger size, strength and speed (Quarrie & Wilson, 2000; Sedeaud, Vidalin, Tafflet, Marc, & Toussaint, 2013). No substantial differences in head impact magnitude were observed between groups, despite the differences in the size and strength of the players. This suggests that, for a given impact force, females would experience greater head-neck segment acceleration than males, as suggested by Tierney et al. (2005). Based on this research, the similarity in impact magnitude between sexes in this study suggests that the force in the impact events was lower in the female players. The female players in this study has a greater incidence of observed whiplash than the male players. This was a qualitative observation so was not included in the breakdown of the impact event analysis. This requires a substantial future research effort, as there are many physical and mechanical differences in the cervical spines of male and females which may affect this.

5.5 Neck Strength and Head Impacts

Negative correlations were observed with PRA for flexion and extension and with PLA for extension in the male group. This is consistent with research showing that neck strength is important in mitigating against inertial loading of the head (Caccese et al., 2018; Collins et al., 2014). Studies report that stronger necks can decrease head acceleration and displacement following a collision, which may reduce concussion risk (Tierney, 2004; Viano, Casson, & Pellman, 2007). A systematic review indicates that greater neck strength is associated with both lower linear and rotational acceleration during purposeful soccer heading (Peek et al., 2020). It has also been suggested that a higher neck flexor to extensor ratio may reduce rotational head acceleration (Peek et al., 2020). Although not directly applicable to rugby contact impacts, this relationship suggests that a similar association may exist during rugby related HIEs. Previous research suggests that increased neck strength may enhance head and neck dynamic stabilisation, especially from the SCM and upper trapezius (Kumar, Ferrari, & Narayan, 2005; Lisman et al., 2010; Mansell et al., 2005). This may be attributed to hypertrophy of the cervical musculature that may translate to greater energy absorption during loading events (Frounfelter, 2008; Taylor, 1993). It has also been suggested that the head-neck dynamic restraint system could provide protective properties similar to those that have been documented in the knee and shoulder (Chimera, Swanik, Swanik, & Straub, 2004; Hewett, Lindenfield, Riccobene, & Noyes, 1999; Swanik et al., 2002).

Dynamic stabilisation of a joint depends on both feed-forward and feedback mechanisms to coordinate movement patterns both in anticipation of, and in reaction to, a load (Mansell et al., 2005). The feed-forward mechanism associated with head-neck joint stabilisation has been suggested to be responsible for preparatory muscle activation of the SCM and trapezius (Swanik, Lephart, Giannantonio, & Fu, 1997). This activation has been reported to increase resistance to head motion (Kumar, Narayan, & Amell, 2000; Tierney et al., 2005). The feedback mechanism of joint stabilisation is associated with reactive muscle activity and uses reflex pathways to regulate motor control (Swanik et al., 1997). These reflex responses are produced by vestibular, visual and mechanoreceptor signals to control head and neck movement (Ito, Corna, Von Brevern, Bronstein, & Gresty, 1997).

Increased neck strength and girth in females is associated with increased neck stiffness (Mansell et al., 2005). This could result in decreased head-neck segment acceleration and ostensibly, a reduced risk of concussion. Findings in collegiate American football report that greater neck stiffness reduced the risk of sustaining high magnitude head impacts (Schmidt et al., 2014).

Resistance training of the cervical muscles could improve neuromuscular recruitment patterns and control through providing a functional stimulus (Pinsault, Anxionnaz, & Vuillerme, 2010). This increases the rate and amount of force development in the muscles, and may translate into potential protective mechanisms (Mansell et al., 2005; Salmon, 2014; Swanik et al., 2002). Literature has shown that supervised and progressive isometric neck strengthening programs can increase neck strength in all directions (Naish et al., 2013). Several studies have demonstrated both increased neck strength and girth as a results of various resistance training programs (Conley, Stone, Nimmons, & Dudley, 1997; Cross & Serenelli, 2003; Mansell et al., 2005).

Research has concluded neck muscle training that elicits feed-forward and feedback motor control mechanisms is required to better utilise dynamic stabilisers for protection (Mansell et al., 2005). Anticipatory training may also be of benefit through increasing neck stiffness during head impacts. Ballistic activities have been proven to enhance neuromuscular control and dynamic stabilisation in other regions of the body, and may be an alternative training method to investigate (Chimera et al., 2004; Hewett et al., 1999; Swanik et al., 2002). Future research to determine the most effective methods to strength the neck musculature is required.

5.6 Whiplash Risk in Females

Previous studies have reported females to be more susceptible to whiplash injury in rear-impact collisions due to females exhibiting heightened head-neck movement to males (Stemper et al., 2011). The load placed on the head is therefore greater, increasing soft tissue deformation and subsequently injury risk (Mohan & Huynh, 2019; Stemper et al., 2011). This has been attributed to the morphological differences between male and female cervical spines (Östh et al., 2017; Vasavada et al., 2008). Male necks have been reported to exhibit greater intervertebral coupling stability than female necks, meaning the spine is more proficient at resisting inertial loading (Mohan & Huynh, 2019; Stemper et al., 2009). A significant observation in this study was that in over half of female HIEs, this whiplash action was present. This was particularly the case in head to ground impacts. Although head impact magnitude was similar in this thesis, the female population are at greater risk of concussion. Female axons are more susceptible to damage following comparable trauma to males due to axonal diameter being smaller and having fewer microtubules (Dollé et al., 2018; Gupte et al., 2019). More research is required to break down the kinematics during impact events to study the influence of the whiplash action, which was beyond the scope of this thesis. The future of women's rugby union must include dedicated programming for female athletes to address these specific weaknesses. Worldwide educational programs are required for coaches and athletes at all levels to educate about the vulnerabilities of the female players.

5.7 Neck Strength Measurement Methods

Neck strength has previously been measured using a variety of equipment and methodologies, meaning direct comparisons between studies often cannot be made. The different methods of assessment carry with them varying degrees of functionality and repeatability, as detailed in Chapter 2.7.3.

The purpose of this thesis was to evaluate isometric neck strength using a method that would be functionally relevant to rugby. Consequently, the INSTA was designed for participants' neck strength to be assessed in a simulated contact position, while limiting the use of accessory muscles. It is important for neck strength measurement methods to be reliable and repeatable, to ensure that accurate comparisons can be made between and within participants. This is also important to enable comparisons to be made with other literature, enabling neck strength to be measured and compared in a wider range of sports and individuals in the future. The methods used by Salmon et al. (2018) are similar to the current thesis: these authors used an apparatus with the same prone simulated contact position, but participants were standing, allowed to grip handles and could push through the ground with their feet. Salmon et al. (2018) collected data on professional male rugby players from a New Zealand franchise, which were comparable to the current thesis.

Isometric neck strength is an important factor to further recognise in rugby and other contact sports to better understand its relationship to both inertial loading of the head and head and neck injuries (Salmon, 2014). Positional differences in absolute neck strength were observed in male players in this thesis, consistent with the physiological requirements of different positional roles (Quarrie & Wilson, 2000; Zemski et al., 2015). No positional differences in neck strength were observed in the female group. This is indicative of players, particularly those in strength-dominant positions such as the front row, not being best prepared for these demands. These players could therefore benefit from targeted training programs to address female-specific weaknesses. The consistency of the results in this thesis supports the use of fixed frame dynamometry with limited use of accessory muscles to maximise repeatability of neck strength testing.

5.7 Strengths

One of the main purposes of this thesis was to measure neck strength in a functionally relevant way for rugby, this was achieved through the use of a novel test apparatus. The key aims of the design were to limit the use of accessory muscles and ensure the measurement technique was accurate and repeatable. The repeatability of the assessment methods was demonstrated through good agreement within the replicate scores for each measure. Participants adopted the same position for each test, ensuring that equipment was set up identically to their previous tests (i.e., head position, neck position, leg position, hands on hips or behind back). The apparatus was calibrated to ensure accurate and consistent measurements were made and extraneous variables were controlled for as much as possible.

To ensure iMG impacts were legitimate, a video verification process was followed for all video and iMG matched impacts (Figure 4). Verifying the video criteria included checking that the player was on the pitch, they were involved in an event and whether there was obvious head acceleration. Verifying the PROTECHT™ data included checking that the impact time matched the video-observed impact time, ensuring there were full waveforms for PLA and PRA and determining whether the waveform was realistic. Any maxima data recorded for PLA or PRA was excluded as it cannot be determined for certain whether these were head impact events or players biting/shouting. The video verification protocol and examples of verified waveforms are in Figure 4 and Figure 5 (Appendix A). A fourth order Butterworth filter, with data-derived cut off frequencies was applied following video verification to remove high frequency noise, as recommended by Greybe et al. (In Press, 2021). This eliminated peaks in the head acceleration data that occur due to the inherently noisy nature of accelerometer data.

5.8 Limitations

The purpose of the thesis was to understand the mechanisms of head impacts in men' and women's university rugby. It also aimed to compare and contrast neck strength between these two populations, as a potential area of weakness which can be addressed in women players.

An acknowledged limitation of the current thesis was the lack of neck length measurements, which limited any head and neck moment calculations. The neck moment created during the strength testing is a product of the moment arm (neck length) and isometric strength, meaning neck length will have an impact on force production (Vasavada, Li, & Delp, 1998).

Although efforts were made to isolate the neck musculature during strength assessments, a degree of movement due to soft tissue and bench cushion displacement is unavoidable (Eckner et al., 2014). This may have resulted in some measurement error associated with the values reported in the this thesis due to participants being able to potentially accelerate their head with their torso in the same direction as force measurement (Brooks & Faulkner, 1996). This inaccuracy, however, is unlikely to have a significant impact on observed relationships due to the internal consistency across all measurements for all participants.

A limitation associated with the use of video analysis in this thesis is that it is challenging to accurately determine the precise moment of impact during contact events. This limitation has previously been noted in similar literature, thus multiple video angles were used where possible and the video verification process followed to minimise the loss of data (Burger et al., 2016).

5.9 Future Considerations

Repeated loading on a single shoulder can cause an accumulation of microtrauma, and eventually lead to a macrotrauma and therefore an impairment of force production (Pelham, White, Holt, & Lee, 2005). Given the length of the season (18 weeks) and the frequency of collisions during matches, this may lead to lateral strength discrepancies. In future neck strength research, players should indicate which is their preferred shoulder when making contact and contact side taken into account during neck strength, video and head impact analysis.

The iMGs used in this thesis recorded acceleration values at the position of the mouthguard. Due to the Covid-19 lockdown, acceleration values were unable to be translated to the centre of mass of the head. Therefore, the values reported in this thesis cannot be directly considered as head centre of mass acceleration. Future research should calculate the centre of mass of the head and translate the recorded acceleration values to be representative of the head itself and be comparable to similar studies (Kieffer et al., 2020).

Both oral contraception and the menstrual cycle have been reported to have effects on a female's susceptibility to concussion (Gallagher et al., 2018; Wunderle, Hoeger, Wasserman, & Bazarian, 2014). Muscle function and strength has also been shown to be affected by stage of the menstrual cycle, meaning this could affect the neck strength values in this thesis (Sarwar, Beltran Niclos, & Rutherford, 1996). Future research should aim to control for the effects of the menstrual cycle and contraception in females.

The HIE sample size reported in this thesis was limited due to complications with the iMG system. Of the HIE magnitude data, a large proportion could not be verified due to missing data leading to incomplete waveforms. This reduced sample size limited the statistical power of the results. The impact verification protocol followed during data analysis prioritised data quality over quantity. This protocol ensured that only data that was representative of true impact events was included, therefore improving the reliability of the iMG results.

Conclusion

Comparisons of head impact dynamics and neck strength between males and females were complicated by the significant differences in playing experience and corresponding position-specific physical adaptations. The implications of this are far reaching and apparent at both performance and safety levels of the game. Given the numerous confounding biological sex differences, the contribution of playing experience is difficult to determine. Successful growth of the women's game internationally will require a significant investment in both research and infrastructure, focused on the development of female-specific training methods and opportunities. This includes the development and implementation of resources at the grass roots level and design and provision of training equipment appropriate for the size of the average woman player. Coach education should focus on understanding sex differences, including female physiology, factors affecting training and performance (including the menstrual cycle), and injury propensity. Crucially, research underpinning rugby laws, injury prevention and identification frameworks, training methodologies and RTP protocols must fairly represent the rugby playing populations. Cervical spine dimorphisms, for example, must be accounted for.

Female rugby training should be based on female data to address female weaknesses, rather than adopting programs based on male data. More data is also needed in many aspects of the female game to increase this objective data platform. Given the findings of the current thesis, programs such as this could be implemented in training with the aim to reduce inertial loading of the head and therefore risk of mTBI in games. Due to the substantial differences between males and females, neck strengthening programs should be sex-specific, based on an appropriate sex-specific data base.

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Appendix A

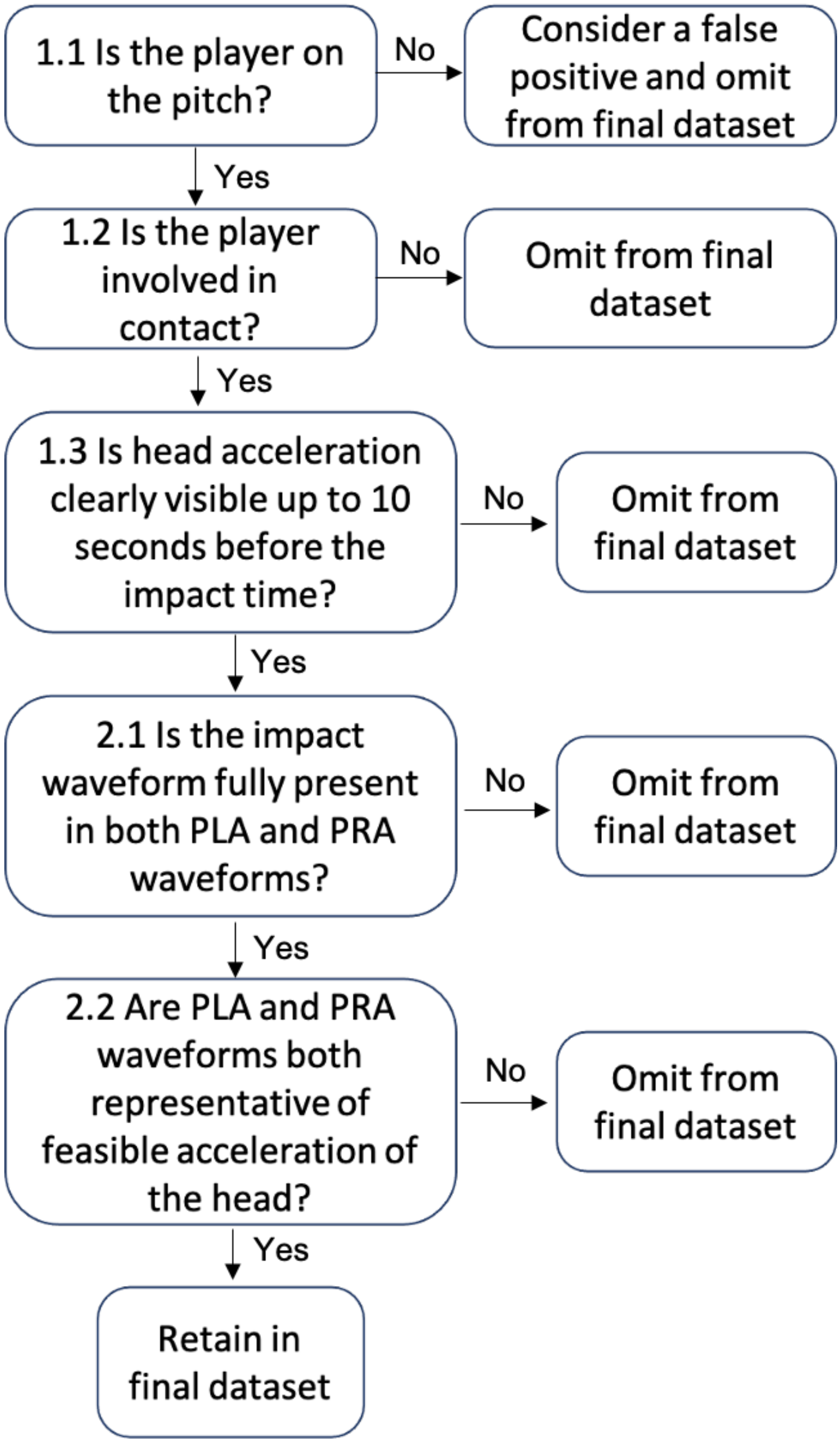

Figure 4- Video verification protocol flow chart

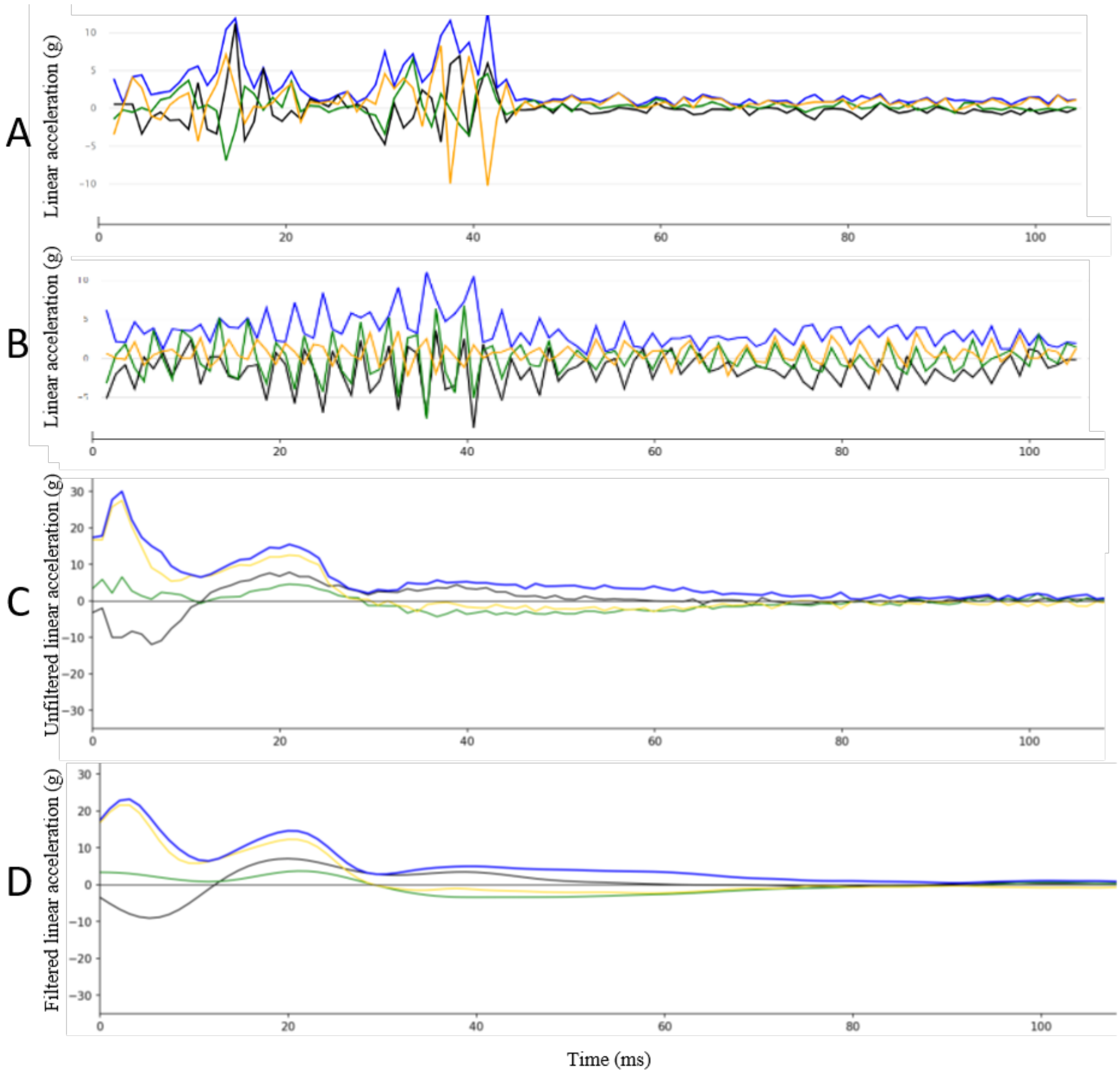

Figure 5- iMG linear acceleration timeseries waveform examples of (A) biting, (B) shouting, (C) a verified head impact event from a tackle unfiltered and (D) filtered

Figure 6- An example of an impact that was excluded from the final dataset due to the absence of sufficient (A) linear acceleration and (B) rotational acceleration waveform data from the iMG.

Appendix B

Technical Note: A Safe Isometric Neck Strength Testing and Training Apparatus (Under Review, EJSS)

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Abstract - This technical note presents the design, construction and validation of a novel, isometric neck strength testing apparatus (INSTA) for collision sport athletes. Previous methods of assessing neck strength have varied in terms of safety, validity, reliability, and repeatability. A fixed-frame dynamometry method in a simulated contact posture has previously shown to be safe, valid, and reliable to test the neck strength of rugby players. The INSTA, presented here, is a development of this, which adopts a kneeling position and bodily restraints to restrict accessory muscle involvement and improve measurement repeatability. Initial results have demonstrated good correspondence with the standing method described in previous work. The INSTA design has seven easily adjustable settings, enabling the standardised testing posture to be replicated for users with body heights between 1.3 and 2.2 metres. The INSTA can be quickly disassembled to fit in a standard motor car, requires less than two square metres floor space for operation and can be decontaminated with ease.

Introduction - The link between repetitive and grievous acute injuries to the neck and head in rugby have been well documented (J. H. M. Brooks & Kemp, 2011; M. Cross et al., 2015; C. W. Fuller et al., 2010; Danielle M. Salmon et al., 2015; Tucker, Raftery, Kemp, et al., 2017b; West et al., 2020). The risk of recurrent cervical (C. Fuller, Brooks, & Kemp, 2007) and brain (M. Cross et al., 2015) injuries in rugby remain a significant concern. Neck strength has been reported as a significant predictor of concussion in a number of contact sports (Collins et al., 2014). The cervical musculature is thought to limit concussive head acceleration, mitigating energy transfer to the brain (Caccese et al., 2018; Falla, Debora; Jull, Gwendolen; Dall'Alba, Paul; Rainoldi, Alberto; Merletti, 2003; Peek et al., 2020; Streifer et al., 2019; R. T. Tierney et al., 2005). Good stabilisation, activation and control of muscles acting in the opposite direction and those acting in the same direction as the perturbation, respectively, is required to mitigate impact severity (Stensdotter, Dinhoffpedersen, Meisingset, Vasseljen, & Stavdahl, 2016). Specific cervical weaknesses may thus predispose athletes to higher head impact magnitudes and an ongoing concern in rugby union is the C-spine degenerative changes identified in front row players (Hogan, Hogan, Vos, Eustace, & Kenny, 2010). Considerable sex differences in neck strength have also been attributed in part to lower head-neck stabilisation during head acceleration in physically active volunteers (R. T. Tierney et al., 2005). Regular surveillance of cervical musculoskeletal function may be of benefit to rugby athletes, enabling objective and ongoing monitoring of specific strength imbalances. Neck strength testing methodologies such as that described by Salmon et al., (Danielle M. Salmon et al., 2015) could be a useful addition to rugby coaching toolkits to achieve this.

Methods - As most head impacts and neck injuries in rugby union occur during the tackle when the body is in a primarily horizontal position, the relevance of evaluating neck strength in a seated or supine position has been questioned (Danielle M. Salmon et al., 2015). Salmon et al., (Danielle M. Salmon et al., 2015) developed a functionally relevant neck strength testing apparatus for rugby union. The design accommodated the horizontal loading position, with the participant standing, bent forwards at the waist in a prone position. The chest and elbows were comfortably supported, with hand holds providing stability. This apparatus has been used to test the functional neck strength of rugby players from university to professional level (Danielle M. Salmon, Sullivan, Handcock, Rehrer, & Niven, 2018). All results show excellent test-retest reliability in maximal neck strength and strength endurance measures. These authors report that repeated sessions for each participant were conducted by the same researcher (Danielle M. Salmon et al., 2015).

Apparatus Requirements - The INSTA frame must be able to support the body weight of the heaviest rugby athletes, with zero flex. The heaviest elite rugby player in the world is estimated to be approximately 145 kg. The average weight of university study population for whom the INSTA was originally constructed was 97.4 kg (SD 11.9, range 70 – 117) for men and 68.3 kg (SD 8.3, range 53.5 – 85 kg) for women. The rig must also be adjustable to accommodate athletes ranging in height from 140 to 220 cm, in an identical posture. Based on previously recorded neck strength results, the headset, including attachments and load cells, must be over-engineered to withstand repeatable application of forces in the vicinity of 500 N (Danielle Margaret Salmon, 2014). The framing for the headset and bracketing for each load cell must be able to withstand repeated force up to 50 kg (490 N) being applied. The value of 490 N is the highest reported by a previous study (Salmon, 2014) where a similar rig was used to test professional male rugby athletes (Danielle M. Salmon et al., 2018). The rig used by these authors, however, enabled accessory muscles to be recruited.

Apparatus Design Features - The design of the Salmon et al., apparatus was modified when developing the INSTA to limit extraneous variables, as testing was to be conducted by different researchers. Figure 1 shows the Salmon et al., apparatus beside the INSTA to show the modifications. Four inward-facing, 150 kg single-point load cells (Tedea-Huntleigh Model 1022) were positioned as per Figures 1 and 2 to measure the force applied by the participant’s head. Specifically, with the participant’s head positioned as per Figure 1, flexion, extension and left and right lateral flexion can be measured. The higher capacity load cells were selected to accommodate the extremes of possible human neck strength and to reduce the likelihood of metal fatigue over time. Aluminium platforms measuring 85*60 mm were attached to the loading point of each sensor, covered with neoprene pads to ensure participants could apply maximum force comfortably (Figure 2).

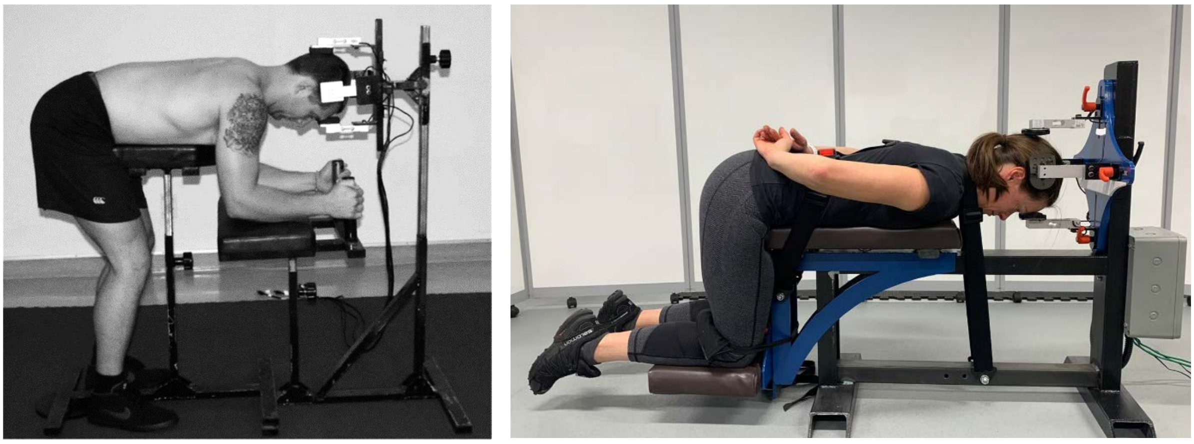

Figure 7: Photo showing a demonstration of the Salmon et al., (2018) design (left) and photo showing a demonstration of the INSTA position

The INSTA modifications from the original Salmon et al design included the following:

- Participants adopting a kneeling position, necessary to prevent accessory involvement from lower body and to enable the feet to be off the ground
- The same prone position and horizontal loading position was maintained, with the torso and knees supported with upholstered benches
- A four-point car racing harness was used to secure the torso to the INSTA bench
- An additional car seatbelt was used to secure the upper legs in a vertical orientation to the back end of the INSTA frame
- Hands were placed behind the back or on the hips during testing to eliminate accessory muscle involvement

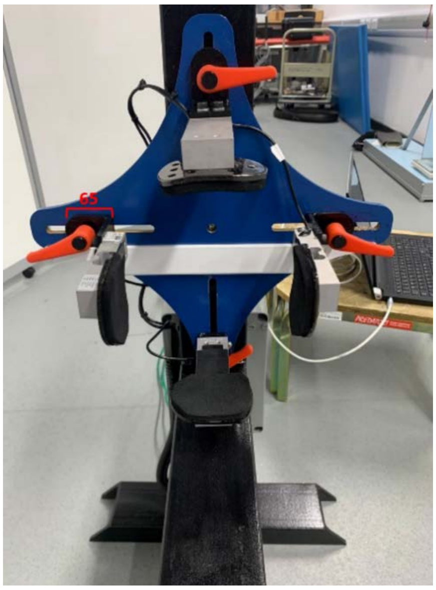

Figure 8: Photo front-on showing positions of four inward-facing load cells, with steel bracket contact length (65 mm) indicated

Figure 9: Side-view photo showing the four load cells with bracket length (65 mm) indicated

Headset Specifications and Technical Data The headset consists of the four load cells, steel brackets, adjustable connector bolts and the 10 mm steel adjustable plate (Figures 2, 3 and 4). The load cells are mounted to the brackets using 40*6 mm, 66 mm lengths of mild steel, machined for this purpose. The ultimate tensile strength of mild steel is 440 MPa and the yield tensile strength is 370 MPa (200-300 kg). The technical specifications and dimensions of the load cells are given by Pavone Systems datasheet (Pavone, 2020), diagram given in Figure 4.

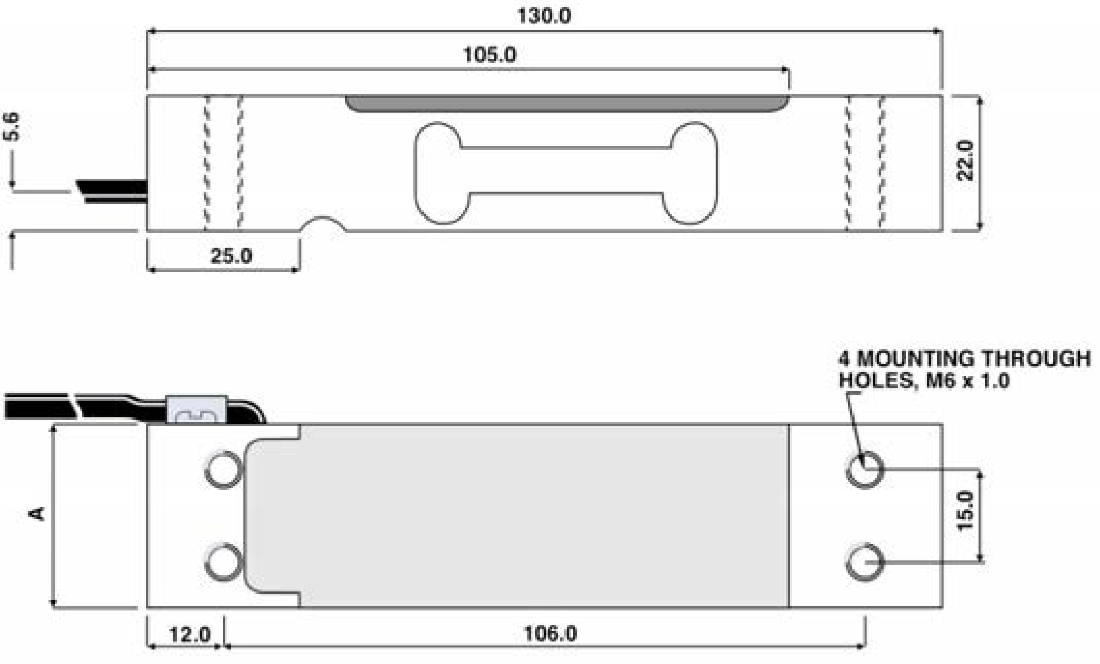

Figure 10: Measurements of Tedea-Huntleigh 150 kg load cells, Model 1022 (Pavone, 2020)

The moment arm for each load cell, including the bracket length is 159 mm and the maximum expected force is approximately 500 N. The expected maximum torque on these brackets is therefore 79.5 Nm. A finite element simulation showed with an application of 500 N, the safety factor (SF) for the bracket and load cell arrangement to be 1.13. Additional reinforcements can be added to the brackets if required and upgrading the brackets to 8 mm mild steel would improve the SF to 2. The tolerance of the 8 mm bolts holding the brackets to the plate is 250 kg, approximately 2.5 times the expected maximum applied load.

INSTA Adjustability Settings - Seven adjustable parameters were included in the INSTA design to accommodate the height range of expected equipment users, illustrated in Figure 3.

Each of the four load cells was made with 140 mm adjustability horizontally (lateral flexion) or vertically (flexion/extension) on the headset plate. This allowed load cells to be adjusted for different head sizes so the neoprene pads could be positioned at the same location on each person's head (Figure 2).

The entire headset plate, bolted to the front vertical posted, was made with x mm adjustability in the vertical direction. This was to ensure correct head, neck, and spinal alignment (Figure 5b).

The horizontal chest bench, removable from the INSTA frame, allows x mm of backward and forward adjustment, with a minimum and maximum distance between the head rest centre and end of the bench of x and y mm respectively (Figure 5c). The knee rest, detachable from the chest bench components, was made adjustable in the vertical direction to accommodate the large range of participant heights (Figure 5d).

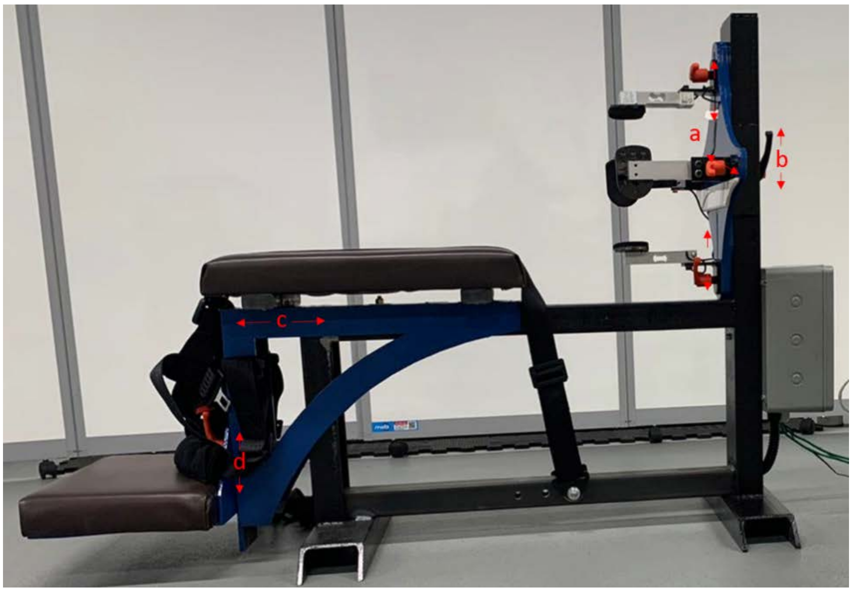

Figure 11: Photograph with adjustability settings indicated: a- the four load cell adjustments vertically or horizontally. b- the headset plate vertical adjustment. c- the chest bench horizontal adjustment. d- knee rest vertical adjustment

Safety of Electronic Components A Type B 12V power supply is required to power the load cell amplifiers. The ISNTA is electrically safe as the load cells are low voltage and correctly connected to a low powered amplifier powered by a class 2 device.

Testing Protocols and Results - Appropriate ethical approval, participant consent, injury screening and warm up procedures were completed. For each participant, the INSTA was adjusted accordingly and both the four-point harness and leg straps were securely fastened. For maximum isometric neck strength testing, participants pushed with maximum effort against each load cell in the specified direction. These efforts will be sustained for durations of three seconds and repeated three times per direction for each testing session. Endurance protocols were also completed, with peak force outputs significantly lower than the max value recorded, which was 472 N. Further results are reported in the main body of the main paper.

Discussion - Standardisation of methods across studies can be helpful in assessing neck strengthening protocols, identifying specific weakness and correlating with injury epidemiology. This method, while compromising the functionality of the Salmon et al apparatus, enables repeatable, standardised measurements with identical posture adjustable for participants between 4 ft 8 and 6 ft 10. Thus, using this method across different studies can enable comparable measurements, further facilitating advancements in knowledge in this field.

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Appendix C

Table 6- Neck strength measurements by sex

Characteristic	Sex	*n*	Mean	*SD*	*t/U*	*p*
Flexion (N)	M	28	274.8	56.3	10.9	< .001*
	F	22	146.1	23.5		
Extension (N)	M	28	249.2	48.2	9.6	< .001*
	F	22	146.7	26.5		
Left lateral flexion (N)	M	28	206.3	56.7	8.3	< .001*
	F	22	108.3	23.1		
Right lateral flexion (N)	M	28	199.4	51.2	19.0	< .001*
	F	22	112.1	30.1		
Time to peak flexion (ms)	M	28	1764	772	70.0	< .001*
	F	22	548	843		
Time to peak extension (ms)	M	28	1551	692	115.5	< .001*
	F	22	430	605		
Time to peak left lateral flexion (ms)	M	28	1866	741	104.0	< .001*
	F	22	627	1012		
Time to peak right lateral flexion (ms)	M	28	2030	676	112.0	< .001*
	F	22	549	915		

*Note. M = male, F = female, * = significant*

Appendix D

Table 7- HIE classification occurrence for male and female players

HAE Characteristic	Male (n = 144)	Female (n = 90)	X^2	p	Cramer’s V	Interpretation of Effect Size
Tackle	57 (39.6%)	42 (46.7%)	2.262	.520	.098	Trivial
Carry	49 (34.0%)	31 (34.4%)				
Ruck	34 (23.6%)	16 (17.8%)				
Maul	4 (2.8%)	1 (1.1%)				

Table 8- Overview of the magnitude of classifications of male and female tackle HIEs

	Indirect Impacts			Direct impacts with 'soft' body parts			Direct impacts with 'hard' body parts			Direct impacts with the ground		
	PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)	PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)	PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)	PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)
Male												
Median	14.8	9.7	642.9	12.7	10.8	874.9	14.5	10.5	946.7	15.5	8.1	986.2
IQR	6.7	6.8	400.0	7.3	5.5	524.3	6.9	5.6	547.7	5.7	10.4	575.1
Min	8.9	2.7	219.6	6.8	4.0	448.3	9.5	4.8	592.3	11.1	5.5	224.7
Max	31.6	14.5	1908.2	24.4	17.6	2533.0	47.3	25.7	2559.1	22.2	23.1	1065.0
Female												
Median	11.5	4.9	735.7	-	-	-	13.4	9.0	854.9	11.9	12.0	886.7
IQR	4.0	1.2	503.7	-	-	-	10.2	6.0	692.0	3.4	8.3	394.0
Min	10.2	4.2	503.7	-	-	-	8.6	3.7	288.1	9.7	3.0	481.0
Max	21.8	6.5	1221.8	-	-	-	44.3	29.8	3401.9	22.6	24.1	3304.0

Table 9- Overview of the magnitude of classifications of male and female carry HIEs

	Indirect Impacts			Direct impacts with 'soft' body parts			Direct impacts with 'hard' body parts			Direct impacts with the ground		
	PLA (g)	PRV (rad.s^{-1})	PRA (rad.s^{-2})	PLA (g)	PRV (rad.s^{-1})	PRA (rad.s^{-2})	PLA (g)	PRV (rad.s^{-1})	PRA (rad.s^{-2})	PLA (g)	PRV (rad.s^{-1})	PRA (rad.s^{-2})
Male												
Median	11.2	10.6	798.8	18.2	11.5	1130.5	12.3	11.0	908.5	11.6	13.7	682.3
IQR	14.3	7.3	672.9	7.6	2.6	910.3	9.2	7.4	800.1	11.9	11.5	1080.6
Min	9.3	3.8	241.3	10.7	9.5	454.4	9.5	3.3	326.1	8.9	5.6	262.9
Max	50.5	22.56	2133.0	21.8	14.0	1833.7	24.8	17.7	1393.5	27.0	20.3	1836.9
Female												
Median	11.3	12.6	712.6	-	-	-	10.6	11.0	666.4	15.6	12.9	1066.0
IQR	2.2	10.2	457.0	-	-	-	2.9	3.2	209.7	8.3	8.0	856.3
Min	8.4	3.9	262.0	-	-	-	7.9	4.4	476.5	9.0	3.0	405.2
Max	33.3	21.5	2345.9	-	-	-	15.2	13.1	873.4	36.6	31.8	1915.3

Table 10- Overview of the magnitude of classifications of male and female ruck HIEs

	Indirect Impacts			Direct impacts with 'soft' body parts			Direct impacts with 'hard' body parts		
	PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)	PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)	PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)
Male									
Median	10.7	8.3	694.2	13.2	7.8	969.7	14.0	7.7	819.3
IQR	5.0	9.2	266.2	323.9	4.7	8.3	4.3	3.6	356.6
Min	8.1	4.1	415.6	7.8	4.3	597.9	8.5	4.6	519.9
Max	16.5	16.5	1020.8	19.8	11.0	1204.2	23.2	17.4	2973.2
Female									
Median	-	-	-	10.2	5.5	1087.4	10.4	6.2	597.9
IQR	-	-	-	4.8	3.1	314.8	5.6	4.6	122.0
Min	-	-	-	6.0	4.6	630.8	6.9	3.4	111.8
Max	-	-	-	24.2	11.8	1295.1	24.5	12.3	1491.0

Table 11- HIE cause of acceleration classification occurrence in tackles, carries and rucks for male forwards and backs

Contact Characteristic	Forwards	Backs	X^2	p	Cramer's V	Interpretation of Effect Size
TACKLE	31	24	5.064	.281	.303	Moderate
Indirect impact	4 (12.9%)	6 (25.0%)				
Direct impact with 'soft' body part	7 (22.6%)	8 (33.3%)				
Direct impact with 'hard' body part	14 (45.2%)	5 (20.8%)				
Direct impacts with the ground	3 (9.7%)	4 (16.7%)				
Other	3 (9.7%)	1 (4.2%)				
CARRY	31	16	2.603	.626	.235	Small
Indirect impact	14 (45.2%)	9 (53.6%)				
Direct impact with 'soft' body part	4 (12.9%)	2 (12.5%)				
Direct impact with 'hard' body part	6 (19.4%)	1 (6.3%)				
Direct impacts with the ground	3 (9.7%)	3 (18.8%)				
Other	4 (12.9%)	1 (6.3%)				
RUCK	27	5	1.551	.818	.220	Small
Indirect impact	6 (22.2%)	2 (40%)				
Direct impact with 'soft' body part	4 (14.8%)	1 (20.0%)				
Direct impact with 'hard' body part	12 (44.4%)	2 (40.0%)				
Direct impacts with the ground	1 (3.7%)	0				
Other	4 (14.8%)	0				

Table 12- HIE cause of acceleration classification occurrence in tackles, carries and rucks for female forwards and backs

Contact Characteristic	Forwards	Backs	X^2	p	Cramer's V	Interpretation of Effect Size
TACKLE	21	21	3.265	.515	.279	Small
Indirect impact	2 (9.5%)	2 (9.5%)				
Direct impact with 'soft' body part	1 (4.8%)	0				
Direct impact with 'hard' body part	14 (66.7%)	12 (57.1%)				
Direct impacts with the ground	4 (19%)	5 (23.8%)				
Other	0	2 (9.5%)				
CARRY	21	10	1.612	.657	.228	Small
Indirect impact	6 (28.6%)	4 (40.0%)				
Direct impact with 'soft' body part	0	0				
Direct impact with 'hard' body part	4 (19.0%)	3 (30.0%)				
Direct impacts with the ground	10 (47.6%)	3 (30.0%)				
Other	1 (4.8%)	0				
RUCK	10	6	1.493	.474	.306	Moderate
Indirect impact	2 (20.0%)	0				
Direct impact with 'soft' body part	2 (20.0%)	2 (33.3%)				
Direct impact with 'hard' body part	6 (60%)	4 (66.7%)				
Direct impacts with the ground	1 (3.7%)	0				
Other	4 (14.8%)	0				

Appendix E

Table 13- Spearman's correlations between anthropometric measures and neck strength for males and females

		Male				Female			
Characteristic		Flx (N)	Ext (N)	LtFlx (N)	RtFlx (N)	Flx (N)	Ext (N)	LtFlx (N)	RtFlx (N)
Stature (cm)	r_s	0.051	0.054	-0.065	0.104	-0.162	-0.204	-0.008	0.094
	p	.813	.802	.762	.630	.482	.375	.971	.685
Body mass (kg)	r_s	0.447*	0.400*	0.394*	0.469*	0.017	0.308	-0.081	0.135
	p	.019	.038	.042	.013	.942	.175	.729	.559
BMI (kg/m^2)	r_s	0.558*	0.256	0.421*	0.347	0.151	0.398	-0.205	0.011
	p	.005	.228	.041	.097	.513	.074	.374	.962
Head circumference (cm)	r_s	0.122	0.139	0.214	0.180	-0.398	0.324	0.172	0.314
	p	.535	.482	.274	.358	.092	.176	.482	.190
Neck Circumference (cm)	r_s	0.469*	0.311	0.515*	0.395*	0.151	0.318	0.026	0.185
	p	.012	.107	.005	.037	.537	.185	.915	.448
Shoulder Breadth (cm)	r_s	0.131	0.286	0.047	0.085	-0.155	0.339	0.046	0.279
	p	.506	.140	.811	.667	.527	.155	.853	.248
Neck to head proportion	r_s	0.573*	0.323	0.496*	0.450*	0.262	0.195	-0.014	0.074
	P	.001	.094	.007	.016	.279	.425	.954	.763
Neck to shoulder proportion	r_s	0.347	0.122	0.431*	0.270	0.399	0.224	0.038	0.121
	p	.071	.536	.022	.165	.091	.356	.878	.623

*Note. * = significant*

Table 14- Spearman's correlations for head and neck anthropometrics and head impact telemetry measures

Characteristic		Male			Female		
		PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)	PLA (g)	PRV ($rad.s^{-1}$)	PRA ($rad.s^{-2}$)
Head circumference (cm)	r_s	-0.134	-0.038	-0.015	0.092	0.107	-0.054
	p	.119	.656	.865	.447	.377	.655
Neck circumference (cm)	r_s	-0.062	0.128	0.095	-0.090	0.019	0.020
	p	.472	.135	.268	.458	.879	.867
Shoulder Breadth (cm)	r_s	-0.084	0.003	-0.165	-0.123	0.298*	-0.117
	p	.330	.968	.054	.310	.012	.335
Neck to head ratio	r_s	-0.072	0.126	0.077	-0.121	-0.052	0.059
	p	.403	.143	.373	.319	.669	.628
Neck to shoulder ratio	r_s	0.023	0.069	0.177*	-0.119	-0.113	0.041
	p	.791	.420	.039	.327	.351	.734

*Note. * = significant*

Table 15- Spearman's correlations for neck strength and head impact telemetry measures

Characteristic		Male PLA (g)	Male PRV ($rad.s^{-1}$)	Male PRA ($rad.s^{-2}$)	Female PLA (g)	Female PRV ($rad.s^{-1}$)	Female PRA ($rad.s^{-2}$)
Flexion (N)	r_s	-0.168	-0.057	-0.206*	0.047	-0.057	0.096
	p	.068	.539	.025	.667	.599	.376
Extension (N)	r_s	-0.218*	-0.097	-0.204*	-0.085	0.075	-0.092
	p	.018	.299	.027	.432	.490	.398
Left flexion (N)	r_s	-0.109	-0.041	-0.020	-0.157	0.142	-0.040
	p	.242	.661	.831	.147	.191	.716
Right flexion (N)	r_s	-0.164	-0.102	-0.082	-0.184	0.219*	-0.035
	p	.076	.274	.278	.088	.042	.746
Relative flexion (N/kg)	r_s	-0.130	-0.029	-0.168	0.019	-0.063	0.068
	p	.160	.759	.068	.860	.560	.532
Relative extension (N/kg)	r_s	-0.167	-0.088	-0.146	-0.109	0.082	-0.123
	p	.071	.341	.115	.313	.453	.256
Relative left flexion (N/kg)	r_s	-0.066	-0.065	0.003	-0.157	0.162	-0.039
	p	.478	.482	.978	.146	.133	.720
Relative right flexion (N/kg)	r_s	-0.086	-0.102	0.031	-0.178	0.301*	-0.032
	p	.352	.270	.741	.100	.006	.766

*Note. * = significant*

Appendix F

Table 16- Independent t-tests comparing stature, body mass and body mass index (BMI) between forwards and backs for the New Zealand Black Ferns

Characteristic	Position	*n*	Mean	*SD*	*t*	*p*
Stature (cm)	F	21	174.0	7.5	2.6	.015*
	B	13	167.5	6.6		
Body mass (kg)	F	21	90.1	15.4	4.0	< .001*
	B	13	71.6	7.8		
BMI (kg/m^2)	F	21	29.7	4.5	3.3	.002*
	B	13	25.5	2.8		

*Note. F = forward, B = back, * = significant*