Head Acceleration in Men's University Rugby Union and the Effect of Neck Strength Training

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Abstract

There is increasing concern regarding concussion and exposure to repeated head impacts in rugby union due to the associated long-term health consequences. To date, measurement systems associated with a high degree of measurement error have been utilised to research head impacts. Moreover, increases in neck strength have been shown to reduce the risk of concussion risk. The aim of this thesis was to investigate the relationship between neck strength and head acceleration in Rugby Union players.

Maximum isometric strength data were collected from 27 male university rugby players at the start of the competitive season and following neck-specific resistance training completed throughout the season. The training programme was completed two times per week and consisted of deep neck stabiliser exercises, weighted isometric training, and dynamic resistance training. The bespoke isometric apparatus utilised four, 150 kg load cells, measuring neck strength in flexion, extension, and left and right lateral flexion. Linear and rotational head acceleration data were recorded throughout the season using mouthguards that were instrumented with a nine-axis inertial motion unit and an additional triaxial accelerometer.

The neck strength training programme resulted in improvements in all outcome parameters (5.5 – 18.8%), with significant improvements for all, except extension ($p < 0.05$). A median (IQR) of 13 *g* (11 - 18 *g*) and 849 rad•s^{-2} (642 - 1,115 rad•s^{-2}) were observed for peak linear and rotational acceleration, respectively. Results revealed that participants with greater neck strength experienced lower head acceleration values throughout the season ($p < 0.05$).

The neck-specific training programme was effective in increasing isometric neck strength. The head acceleration values recorded in the current thesis were substantially lower than those previously recorded. Findings indicate that increasing neck strength may be effective in reducing head inertial load experienced during rugby matches.

Declaration 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.

Statement 1

This thesis is the result of my own investigations, except where otherwise stated. Where correction services have been used, the extent and nature of the correction is clearly marked in a footnote(s). Other sources are acknowledged by footnotes giving explicit references. A bibliography is appended.

Statement 2

I hereby give consent for my thesis, if accepted, to be available for photocopying and for inter-library loan, and for the title and summary to be made available to outside organisations.

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Date ....02/10/2020...

Contents

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List of Figures

List of Tables

List of Abbreviations

AF – American Football

ARF – Australian Rules Football

BESS – Balance Error Scoring System

BUCS – British Universities and Colleges Sport

CARE – Concussion Assessment, Research and Education

CON – Control

CSA – Cross-sectional area

CTE – Chronic Traumatic Encephalopathy

DNS – Deep Neck Stabilisers

FEM – Finite Element Modelling

HIA – Head Impact Assessment

HITS – Head Impact Telemetry System

iMG – Instrumented Mouthguard

IMU – Inertial Motion Unit

mTBI – Mild-Traumatic Brain Injury

MVC – Maximum voluntary contraction

NFL – National Football League

OA – Osteoarthritis

PLA- peak linear acceleration

PRA – peak rotational acceleration

Rep – Repetition

RHI – Repeated head impacts

RM – Repetition maximum

rugby – Rugby Union

SAC – Standard Assessment of Concussion

SCAT – Sport Concussion Assessment Tool

SCM – Sternocleidomastoid

SRC – Sport-Related Concussion

STA – Soft Tissue Artefact

TBI – Traumatic Brain Injury

UT – Upper Trapezius

Chapter 1 : Introduction

1.1 Background and Context

1.1.1 Risks of Rugby Union

Due to the high frequency of contact events and resultant head impacts, brain injuries can frequently occur in contact sports (Cunningham, Broglio, O'Grady, & Wilson, 2020; Tierney & Simms, 2017a). Indeed, an injury surveillance study from 2001-2012 suggests that the prevalence of sport-related brain injury is increasing, with males experiencing an increase of over 105% (Coronado et al., 2015). Rugby Union (hereafter referred to as rugby), a field-based contact sport played all over the world (Brooks, Fuller, Kemp, & Reddin, 2005), has a high prevalence of repeated head impact events, and, subsequently, a high risk of sport-related concussion (SRC; Cross, Kemp, Smith, Trewartha, & Stokes, 2016; King, Hume, Brughelli, & Gissane, 2015; Tierney & Simms, 2017b). Exposure to repeated impacts and brain injury is thought to lead to long-term health implications, which may result in severe depression, cognitive decline and premature death (Bazarian et al., 2014; Broglio, Eckner, Paulson, & Kutcher, 2012; Omalu et al., 2006). Increasing the breadth of knowledge regarding rugby head impacts and potential risk of injury contributes to the creation of evidenced-based injury prevention strategies in the sport. This study design was limited to one cohort of male university rugby union players, while parallel projects focused on female players. In order to include the breadth and depth of variables presented in this study, inter-cohort comparisons would be beyond the scope of one masters thesis.

1.1.2 The Head Impact Telemetry Field

To design effective interventions to reduce injury exposure, the biomechanical inputs associated with an injury, and the context in which the injury is sustained, must be accurately assessed. Current injury metrics and thresholds associated with head impacts have been derived from head acceleration data, obtained from accelerometers and gyroscopes embedded in helmets, attached to head-bands, or adhered to the head (Brennan et al., 2017). Within all areas of biomechanics, and indeed head impact telemetry, the presence of soft tissue artefact (STA) can result in a degree of measurement error (Lucchetti, Cappozzo, Cappello, & Della Croce, 1998; Shultz, Kedgley, & Jenkyn, 2011). Specifically, telemetry systems adhered to the head via the mastoid process have been reported to substantially over-estimate head acceleration on impact due to skin movement (Wu et al., 2016). Significant measurement error has also been observed by helmet- and headband-based telemetry systems due to insufficient sensor skull coupling, producing excess sensor movement relative to the head (Cummiskey et al., 2017; Wu et al., 2016). Consequently, currently available injury criteria based on head acceleration data may be inaccurate. Nonetheless, inertial motion units (IMU) embedded in mouthguards have recently been developed to improve sensor-skull coupling to obtain more accurate measures of head acceleration (Greybe, Jones, Brown, & Williams, 2020; Wu et al., 2016). Limited research, however, has been conducted in rugby using such mouthguard-based systems. Additionally, inconsistencies surrounding data processing and verification may have led to errors with regards to impact magnitude and frequency within the existing data pool (Greybe, Arora, Jones, & Williams, submitted 2020; King et al., 2015).

1.1.3 Reducing Head Acceleration

The high exposure to repeated head impacts in rugby, and associated health concerns, mean that it is imperative to devise strategies and interventions to reduce the risk of injury and protect player welfare. One potential strategy to reduce the risk of brain injury in rugby is to increase players' neck strength. Indeed, the musculature that surrounds the cervical spine supports the control and stabilisation of the head (Falla, Debora; Jull, Gwendolen; Dall'Alba, Paul; Rainoldi, Alberto; Merletti, 2003). The vestibular and cervicocollic systems are thought to mitigate head acceleration through the activation of muscles acting in the opposite direction and controlling the muscles acting in the same direction as perturbation, respectively (Stensdotter, Dinhoffpedersen, Meisingset, Vasseljen, & Stavdahl, 2016). Whilst this relationship has been well established within soccer, research surrounding neck strength and head acceleration within contact sports remains equivocal, with a dearth of research specifically in rugby (Dempsey, Fairchild, & Appleby, 2015; Eckersley, Nightingale, Luck, & Bass, 2017; Mihalik et al., 2011; Peek, Elliott, & Orr, 2020; Schmidt et al., 2014).

1.1.4 Increasing Neck Strength

Given that increased neck strength may reduce the inertial load of the head-on impact, strength training programmes designed to increase players' maximal neck strength may be an effective strategy to reduce injury risk. However, there is no consensus regarding the optimal training programme, with varying durations, resistance loads, frequencies and exercise types limiting inter-study comparisons and precluding firm conclusions (Barrett et al., 2015; Lisman et al., 2010; Naish, Burnett, Burrows, Andrews, & Appleby, 2013; Salmon, 2014; Salmon et al., 2013). Moreover, the limited research often reports contradictory results (Barrett et al., 2015; Conley, Stone, Nimmons, & Dudley, 1997; Geary, Green, & Delahunt, 2014; Mansell, Tierney, Sitler, Swanik, & Stearne, 2005).

1.2 Aims

The aim of this thesis was to obtain accurate head acceleration data for men's university rugby and determine the effect that neck strength has on this. The efficacy of a neck-specific strength training programme, completed regularly throughout the season, was also evaluated.

Chapter 2 : Literature Review

2.1 Sport of Rugby Union

Rugby Union (rugby) is a popular team contact sport played worldwide (Brooks et al., 2005). The field-based sport, played over two 40 minute halves, elicits a high level of physical contact between two teams of 15 individuals (Duthie, Pyne, & Hooper, 2003). Therefore, high levels of strength, speed and power are required (Duthie et al., 2003). Each team is split into forwards and backs, with each position associated with roles that require differing physiological demands and require a range of physical characteristics (Nicholas, 1997; Takamori et al., 2020).

During a rugby game, multiple contact events exist that can cause head acceleration. A recent expert consensus provided consistency regarding rugby event definitions (Hendricks et al., 2020). Tackling is the most common form of contact, which involves one or more players endeavouring to stop a ball carrier, regardless of whether the ball carrier is brought to ground (Hendricks et al., 2020). The tackle event is reported to have the highest incidence of injury in the male game as well as the greatest likelihood of resulting in a concussive event (Cross et al., 2016; Fuller, Brooks, Cancea, Hall, & Kemp, 2007). Scrums are a set-piece, which involves eight players from each side binding together in a controlled situation, pushing and hooking the ball with their feet to win possession (Hendricks et al., 2020). Rucks occur often after a tackle is made whilst the ball is on the ground, where one or more players from each team are on their feet in physical contact over the ball (Hendricks et al., 2020; World Rugby, 2020). Contact may also occur in the form of a maul. A maul involves a ball carrier and at least one or more players from each team bound in contact whilst on their feet (Hendricks et al., 2020; World Rugby, 2020). A lineout is also an event that regularly occurs in rugby. Lineouts are formed on the touchline, with teams forming singular lines parallel to and 1 m away from each other. A lineout requires at least two players from each team (Hendricks et al., 2020). The ball is thrown into the lineout by the attacking team, and both teams compete for the ball in the air. Whilst limited through law, a level of contact between opposition players is present at this event (World Rugby, 2020).

2.2 Brain Injury

2.2.1 Brain Injury Nomenclature

Traumatic brain injury (TBI) refers to any change in brain function or other brain pathology as a result of an external force (Menon, Schwab, Wright, & Maas, 2010). This altered brain function may reflect a variety of neurological deficits such as loss of consciousness, memory loss, confusion, loss of balance and altered vision (Menon et al., 2010). TBI is usually split into two broad categories, acute or chronic. Chronic TBI encompasses the long-term effects of single or multiple TBI events, whilst acute refers to injuries and symptoms that imminently follow an impact event (Jordan, 2013).

A concussion is a classified as a form of mild TBI (mTBI) and is a result of inertial loading of the head via direct or indirect contact (Hoshizaki, 2013; McCrory et al., 2017). A concussion that occurs within a sporting context is referred to as a Sports-Related Concussion (SRC) and is often regarded as one of the most complex injuries to identify, evaluate and control (McCrory et al., 2017). Nevertheless, SRC is usually characterised by short-duration neurological impairment, with clinical symptoms becoming apparent from minutes to hours after an injury, that may not necessarily have resulted in loss of consciousness (McCrory et al., 2017).

Sub-concussive impacts also occur regularly in sport. Sub-concussion refers to head impact events that are not diagnosed as concussive at a clinical level, showing no observable signs or symptoms of neurological dysfunction (Bailes, Petraglia, Omalu, Nauman, & Talavage, 2013). Despite the lack of immediate symptoms, the experience of repeated sub-concussive head impacts (RHI) is thought to have detrimental consequences in later life (Bazarian et al., 2014; Broglio et al., 2012; Cross et al., 2016; Omalu et al., 2006, 2005).

2.2.2 Primary Mechanisms of Brain Injury

The common consensus is that SRC is primarily a result of rapid linear and rotational acceleration transferred to the brain (Bian & Mao, 2020; Meaney, Morrison, & Bass, 2014; Rowson et al., 2019). Linear acceleration is thought to contribute to brain injury as a result of a transient intracranial pressure gradient (Hardy et al., 2007; King et.al., 2003; Unterharnscheidt, 1971). Specifically, following an impact, a linear pressure gradient is generated within the brain. Given the large dilatational wave speed of grey and white matter, this pressure response can move through the brain 10 times per millisecond (Pearce & Young, 2014). Early research demonstrated correlations between peak intracranial pressure at the time of injury and subsequent neurological dysfunction (Nahum, Smith, & Ward, 1977). However, more recent studies suggest that rotational loading may be the more important mechanism in the occurrence of brain injury and SRC (Bian & Mao, 2020; Patton, McIntosh, & Kleiven, 2013; Tierney & Simms, 2017a). King, Yang, Zhang, and Hardy (2003) reported that when subjected to linear acceleration, the motion of the brain was limited to 1 mm compared to 5 mm when subjected to rotational acceleration. Indeed, the brain is highly resistant to changes in shape due to its high bulk modulus (Fernandes & Sousa, 2015; Meaney & Smith, 2011), however, brain tissue has a low shear modulus, meaning that it has a high sensitivity to rotational loads (Tierney & Simms, 2017). Rapid rotational head movements result in the production of high shear forces, leading to shear-induced deformation and tissue damage (Fernandes & Sousa, 2015; Meaney & Smith, 2011). Recent research, using finite element modelling (FEM), has also indicated that rotational kinematics are the main cause of brain strain, a predictor of TBI, following a head impact event (Bian & Mao, 2020).

2.2.3 Factors Affecting Brain Injury Severity

Many intrinsic factors may influence an individuals' risk of brain injury. These include, but are not limited to; demographic factors such as age, sex and race, neurodevelopment factors such as behavioural disorders and learning difficulties, or health history such as the presence of co-morbid conditions (Choe, Babikian, Difiori, Hovda, & Giza, 2012; Danelson, Geer, Stitzel, Slice, & Takhounts, 2008; Houck, Asken, Bauer, & Clugston, 2019; McCrea, Broshek, & Barth, 2015). Extrinsic factors such as team performance, opposition difficulty, fitness levels and game duration may also influence the experience of SRC (Emery, Kang, Schneider, & Meeuwisse, 2011; Gabbett, 2004, 2007; Hollis et al., 2011; King, Hume, Gissane, Kieser, & Clark, 2018). Several biomechanical factors also interact with head acceleration to influence injury tolerance, specifically, neck strength and impact duration, frequency, density, direction and location (Pearce & Young, 2014; Rowson et al., 2019).

Neck Strength

Neck strength has been identified as a potential factor in reducing the inertial load placed on the head during contact situations, with research reporting that for every pound (0.45 kg) increase in neck strength, measured as the maximal force applied to a handheld tension scale, the risk of sustaining an SRC was reduced by 5% (Collins et al., 2014). The cervical musculature is thought to limit the occurrence of SRC through the reduction of head acceleration, mitigating energy transfer to the brain (Streifer et al., 2019). Tierney et al., (2005) investigated sex differences in head-neck dynamic stabilisation. Females were observed to experience a 50% higher angular acceleration and 30% greater displacement than males. These differences were hypothesized to be a result of females having significantly lower isometric neck, neck girth and head mass which subsequently resulted in 29% lower neck stiffness compared to males. In addition, Reynier et al., (2020), using electromyography (EMG), demonstrated that maximal unilateral contraction of cervical muscles resulted in decreased head kinematics compared to a passive muscle condition. Research has reported that muscle stiffness is regulated by vestibular and cervicocollic reflex systems which work reciprocally to maintain head-neck stability (Blouin, Descarreaux, Bélanger-Gravel, Simoneau, & Teasdale, 2003; Stensdotter et al., 2016). Through the projection of vestibular neurons, the vestibular system mitigates the acceleration of the head through the activation of neck muscle forces acting in the opposite direction to perturbation. In contrast, the cervicocollic system acts through means of proprioception, controlling the activation of the muscles acting in the same direction as perturbation (Stensdotter et al., 2016).

It is also proposed that awareness and anticipation of an impact or acceleration contribute to greater stabilisation. Seminati, Cazzola, Preatoni, & Trewartha (2017), simulating different rugby tacking scenarios, reported that cervical muscle pre-activation occurs prior to impact. These authors reported that this pre-activation enables greater cervical stiffness and correct body segment orientation, which may allow for greater head-neck control when subjected to high biomechanical loads. This pre-activation, however, was reported to take up to 300 ms. This, given the relatively short duration of head impact events in rugby, suggests that anticipation of impending impacts is important to allow sufficient time for cervical muscle activation and, thus, effective stabilisation of the head. Additionally, Kumar, Narayan, & Amell (2000) found that the expectation of a perturbation to reduce head-neck acceleration by 30%, which was consistent across increasing magnitudes.

The literature surrounding neck strength and head acceleration is discussed more detail in Chapter 2.5.

Head Impact Frequency and Density

Experiencing a high frequency of head impacts may influence the risk of brain injury (Cross et al., 2016; Rowson et al., 2019). In United States Service Cadets, it was reported that sustaining a previous concussion was a consistent risk factor for sustaining future concussions (Van Pelt et al., 2019). Furthermore, using matched controls, it was found that those who experienced a concussive injury had sustained a greater impact frequency before the injury (Rowson et al., 2019). Similarly, in American football (AF) players, it was reported that experiencing three or more concussions within 7 years increased the risk of sustaining another concussion three-fold (Guskiewicz et al., 2003). This indicates a potential cumulative effect of impacts leading to greater injury risk.

Broglio, Lapointe, O'Connor, & McCrea, (2017) reported that it may not only be the number of impacts that increases the risk of injury, but also the impact density. They observed no difference in the total number of impacts in the 24 hours leading up to the injury, or the magnitude of the final 20 impacts prior to injury, sustained by concussed vs non-concussed individuals. Yet those who experienced a concussive event experienced a significantly greater impact density compared to non-concussive controls. Impact density was defined from the final 20 impacts, dividing cumulative impact magnitude by the time from the previous impact. Greater impact density is thought to result in insufficient time, between one impact to the next, for ion balance within the cerebral tissue to return to baseline levels. Therefore, a subsequent impact causing additional ion efflux may reduce the magnitude of acceleration required to damage neural tissue (Broglio, Lapointe, O'Connor, & McCrea, 2017).

The extensive list of intrinsic, extrinsic, and biomechanical risk factors has important implications for the management of brain injury, where a more individualistic approach is required to assess or predict the risk of injury and implement prevention strategies. Furthermore, data formulated with more accurate measurement systems are required to better inform these strategies.

2.3 Brain Injury in Rugby

2.3.1 Brain Injury and Head Impact Incidence

Research has reported a high SRC prevalence in rugby. A prospective study across two seasons of English Premiership rugby observed an estimated match concussion incidence of 8.9/1000 playing hours (Cross et al., 2016). Additionally, Bathgate, Best, Craig, & Jamieson (2002), reported SRC to account for approximately 5% of all injuries in Australian rugby. Moreover, Rafferty et al., (2018) reported that after playing 25 matches of rugby in a single-season, players were likely than not to sustain an SRC. This may be due to the high RHI exposure in rugby. It has been reported that amateur rugby players sustain an average of 564 ± 618 impacts throughout a season (King et al., 2015).Whilst these figures indicate a high frequency of impacts, the standard deviation reported is greater than the mean. This suggests that the data used may have included significant outliers, leading to a large variation in the data set. , This RHI prevalence per game is reported to be the higher compared to other sports such as lacrosse, AF and Australian rules football (ARF; Nguyen, Brennan, Mitra, & Willmott, 2019).

2.3.2 Head Impact Assessment Protocols in Rugby

Currently, within rugby, side-line protocols are used to attempt to diagnose and manage the experience of SRC. If a player is thought to have sustained a concussive impact, they are required to complete a HIA. The HIA is a standardised medical assessment that aims to evaluate a number of SRC symptoms to assess whether an SRC has been sustained. SRC, however, has varying symptoms with fluctuating timelines, with some symptoms taking up to 48 hours to become apparent (Raftery, Kemp, Patricios, Makdissi, & Decq, 2016). This means that the HIA alone may not be enough to accurately diagnose an SRC.

Despite being widely used, the HIA depends on a degree of subjectivity as it requires a medical professional to accurately assess concussive symptoms. This assessment may be enhanced by the analysis of available video footage. Efforts have been made to achieve a consensus regarding rugby video analysis descriptors and definitions to aid in effective injury surveillance (Hendricks et al., 2020). Accurate information on the context of the event leading to injuries such as event type, impact location, presence of direct head contact and contact intensity may allow medical professionals to make more informed decisions. Similarly, observing SRC symptoms at the exact point of injury may provide valuable insight. Apart from the use of video footage, no other objective measures are currently used in the professional game if a player should be removed from play.

Recently Garcia et al., (2019) attempted to develop a data-driven framework to objectively determine if a player had a possible, probable, or definite SRC. These authors used demographic and injury data from the Concussion Assessment, Research and Education (CARE) Consortium to inform their predictive model. This included 24,561 individuals with 1,950 SRC cases from a wide range of sports. Injury data included baseline and post-injury (< 6 and 24-48 hours) Standard Assessment of Concussion (SAC), Sports Concussion Assessment Tool (SCAT) and Balance Error Scoring System (BESS) scores. Time-injury characteristics such as loss of consciousness, post-traumatic amnesia and retrograde amnesia were also used. Garcia et al., (2019) authors reported that their model was successful in classifying up to 92% of diagnosed SRCs as high risk (probable or definite), with up to 81% of non-concussive individuals correctly classified as low risk (unlikely or possible). Whilst this suggest that this data-driven approach may be effective in the diagnosis of SRC, this study relied on the initial diagnoses of SRC to inform the CARE consortium to train the model. These diagnoses were dependent upon various cognitive assessments built upon a variety of subjective assessments. Additionally, this method may only be effective in diagnosing concussion after-the-fact with data collected up to 48 hours post-injury. Furthermore, this approach does not consider the biomechanical variables associated with head impact exposure that have been reported to be a significant risk factor of SRC. The addition of biomechanical exposure may increase the effectiveness of such data-driven models. Consequently, there is a requirement for more objective measures of SRC to enable informed, in-game, decisions to protect player welfare.

2.4 Head Impact Telemetry

Within sport, one of the more easily controlled risk factors of brain injury is the experience of head impact events, including head impact number, magnitude and density. Currently, available technology allows for the measurement of correlates of brain injury (Rowson et al., 2016). As the head experiences both linear and rotational acceleration on impact and both are thought to play a primary role in the occurrence of brain injury, peak linear (PLA) and peak rotational (PRA) acceleration are often recorded as the main outcome measures of a head impact event (Broglio et al., 2010; King, Hecimovich, Clark, & Gissane, 2017; Meaney & Smith, 2011; O'Connor, Rowson, Duma, & Broglio, 2017). Recording these values allows for a deeper understanding of the dynamics of head impacts, which can be used to implement more biomechanically informed prevention strategies.

2.4.1 Thresholds

When measuring head acceleration in sport, an appropriate threshold must be used for an event to be classified as an impact. This ensures that only accelerations that are due to an impact event are registered, ignoring those due to 'normal' activities. Research has reported that activities such as walking, running, jumping and sitting produce head accelerations less than 10 *g* (Ng, Bussone, & Duma, 2006). This would suggest that a threshold of 10 *g* may be appropriate to filter out those events and this value is commonly used, with 42% of studies using a 10 *g* threshold to report head impacts (King, Hume, Gissane, Brughelli, & Clark, 2016). However, there is still a lack of consistency with regard to the thresholds used, with approximately 30% of studies reporting impacts according to a threshold of 14.4–20 *g* (King et al., 2016). Discrepancies in threshold values may lead to variation in the number of impacts recorded, effecting the apparent head impact prevalence.

2.4.2 Measurement Techniques and Magnitude

Various head impact telemetry systems are used to assess the biomechanical determinants of head impact events. Largely the research in this area is based upon data collected using the Head Impact Telemetry System (HITS) in AF (Brennan et al., 2017). HITS is comprised of six single axis, spring-loaded accelerometers that are usually embedded in an AF helmet (O'Connor et al., 2017). Using this system, it has been reported that high school level AF players sustain a mean PLA and PRA of 25.9 ± 15.5 *g* and 1,694.9 ± 1,215.9 rad•s^{-2} respectively. These values are consistent across the AF HITS literature both at high school and collegiate level (Crisco et al., 2011; Mihalik, Bell, Marshall, & Guskiewicz, 2007; Rowson, Brolinson, Goforth, Dietter, & Duma., 2009), with research in youth AF using HITS reporting maximum values of 126 *g* and 5,838 rad•s^{-2} in 9-12-year-olds (Cobb et al., 2013). Similarly, using a similar methodology 14-18-year-olds were reported to sustain a maximum of 153 *g* and 7,701 rad•s^{-2} (Urban et al., 2013).

Laboratory reconstructions of sporting head injury events, using Hybrid III vehicle crash dummies and FEM, has found that sustaining impacts over 85 *g* is likely to result in irreversible brain injury (Zhang, Yang, & King, 2004). Similarly, PRA of 2,500 rad•s^{-2} has been reported to be associated with significant risk of brain injury (Post, Blaine Hoshizaki, Gilchrist, & Cusimano, 2017). Whilst it should be noted that currently, FEM can only produce brain strains and pressures that correlate to injury, not directly predict conditions in which injury will occur (Rowson, Tyson, Rowson, & Duma, 2018), these results indicate that it is unlikely that the peak values recorded previously in AF are biomechanically plausible (Cobb et al., 2013; Urban et al., 2013). Indeed, research into the accuracy and reliability of helmet-mounted systems such as HITS has observed an error of up to 298% (Cummiskey et al., 2017). This error is likely due to the insufficient coupling of the accelerometers to the skull, with suggestions that helmets may experience 10 times greater acceleration than the head-on impact (Manoogian, McNeely, Duma, Brolinson, & Greenwald, 2006). Similarly, helmets have been shown to translate 13-41 mm and rotate up to 37° more than the head on impact (Joodaki et al., 2019; Wu et al., 2016). Therefore, it is likely that acceleration values recorded previously may be more representative of helmet movement, leading to an overestimation of impact magnitude. Additionally, this may cause a greater number of impacts surpassing the threshold value, resulting in overestimation of head impact frequency and density.

2.4.3 Head Impact Telemetry in Non-Helmeted Sports

In non-helmeted sports, head-impact telemetry systems that are coupled to the skin by attachment to the mastoid process have been used to record head impact kinematics (Chrisman et al., 2016; King, Hume, Gissane, & Clark, 2016; King, Hume, Gissane, & Clark, 2017; Lynall et al., 2016). Using the *X2* X-Patch system (X2 Biosystems, Seattle, WA, USA), junior rugby players have been reported to experience median PLA and PRA values of 15 *g* and 2,296 rad•s^{-2} respectively, with three values recorded above 80 g and one at 141 g (King et al., 2016). Similarly in junior rugby league, using the *X2* system, 28 impacts over 80 *g* were recorded from 12 games (King et al., 2017). Despite these high magnitudes, there were no observed SRCs. Whilst these studies were conducted with junior rugby league players and are therefore not directly comparable with the values previously cited in adult AF, several limitations are present. Firstly, head impacts could not be verified due to an absence of video analysis; as such, it cannot be confirmed as to whether those values are representative of actual impacts. Secondly, telemetry systems mounted on the skin can produce a measurement error of up to 120%, likely as a result of soft-tissue artefact (STA; Wu et al., 2016).

STA has been shown to produce errors all forms of biomechanics, including the analysis of various gait parameters, such as knee joint kinematics and foot motion (Lucchetti et al., 1998; Reinschmidt, Van Den Bogert, Nigg, Lundberg, & Murphy, 1997; Shultz et al., 2011). STA refers to the movement of skin-mounted sensors relative to the underlying bone structures due to skin deformation (Shultz et al., 2011). Research has shown that STA may result in skin-mounted head impact telemetry systems over predicting event magnitude (Wu et al., 2016). Specifically using the ear canal as a reference point, the X-Patch was seen to displace by 4 mm, leading to measurement errors in PLA and PRA of 15 ± 7 *g* and 2,500 ± 1,200 rad•s^{-2} respectively, relative to a tightly coupled mouthguard system (Wu et al., 2016). Thus, kinematic data produced from devices with non-rigid skull coupling should be interpreted with caution.

Instrumented mouthguards (iMG) have been developed to improve the accuracy of head impact kinematics (Greybe et al., 2020; King et al., 2015). With inertial motion units (IMU) being placed in bespoke, tightly coupled mouthguards, they are directly coupled to the skull via the upper dentition. Wu et al., (2016) reported iMGs to provide tight sensor skull coupling, displacing by only 1 mm from an ear canal reference point, which was within video measurement error. This would indicate that the IMGs are more accurately recording the movement of the head. Similarly an iMG has also demonstrated systematic agreement with a Hybrid III anthropometric testing dummy in the linear acceleration and rotational velocity recorded (Greybe et al., 2020).

Earlier work from King et al., (2015) aimed to quantify the head impact load experienced by an amateur rugby team throughout a season. Using an iMG, these authors reported players to sustain an average of 95 ± 133 head impacts over 10 *g* per match, with an average PLA of 22 ± 16.2 *g* and PRA of 3,902.9 ± 3,948.8 rad•s^{-2}.The iMGs used in this study are reported to have 10% error for PLA and PRA (Camarillo, Shull, Mattson, Shultz, & Garza, 2013; Mattson, Shultz, Goodman, Anderson, &

Garza, 2012), and the accuracy of the system during certain rugby activities is unknown (King et.al., 2015). Therefore, the results presented may not accurately represent the head impact burden of rugby.

Despite King et al., (2015) using video footage to provide contextual support for impacts, limited information was provided regarding a video verification process. Moreover, only 65-85% of impacts could be accurately identified during analysis. This means that multiple impacts could not be verified, potentially leading to the inclusion of false positive impacts and impact frequency being over-reported. Similarly, the average magnitude of impacts may be misrepresented due to the inclusion of non-verified impacts in analysis false positive impacts may arise for multiple reasons such as biting and removal and insertion of the iMG. However, King et.al., (2015) did employ a 'declacking algorithm' to attempt to account for non-contact head movements and biting, which may have reduced the number of false positive impacts being recorded. Laboratory validation of the *X2* iMG reported that although the system was able to identify over 95% of impacts, it was unable to accurately measure the magnitude or impact direction (Siegmund, Guskiewicz, Marshall, DeMarco, & Bonin, 2016). This would suggest that results produced from this particular system should be treated with caution.

There are a number of limitations reported, which should be considered when using iMGs to assess head impact telemetry. Researchers have reported a potential effect of mandible motion leading to mouthguard deformations during dynamic events (Kuo et al., 2016). These authors reported unconstrained mandible conditions resulted in decreased mouthguard accuracy, and whilst this condition is unlikely on the field, it should be considered in validation testing. It has also been reported that the increased thickness of mouthguard systems, as a result of built-in electronics, may lead to inhibited communication with team-mates influencing performance (Rowson et al., 2018).

The data processing techniques used, such as filtering, should also be considered when using iMGs or any head impact telemetry systems. When calculating PLA and PRA from raw accelerometer and gyroscope time series data, filtering and calculation of the resultant of the triaxial components data are required, with the added step of differentiation for PRA. Greybe et al., (submitted 2020) investigated the effect of filtering data at various, impact specific cut-off frequencies on resultant head acceleration values. These authors found that applying a filter significantly affected the calculated resultant head impact magnitude, with significant differences also observed when using differing cut-off frequencies. Furthermore, a recent study compared five iMGs for measuring head kinematics in AF (Liu et al., 2020). All guards except one were reported to apply a 4^{th} order Butterworth filter to the raw kinematic data. Liu et al., (2020) reported that the linear acceleration data, obtained from the iMG without a filter, were associated with greater relative error compared to the other systems. It should be noted however, that these authors did not assess the event of mandible motion, which may have influenced results (Kuo et al., 2016). Nevertheless, these results would suggest that careful consideration should be given to the data processing techniques used when comparing head impact values recorded between studies.

Despite the limitations of mouthguard-based telemetry systems, research suggests that due to increased coupling to the skull, these systems allow for greater measurement accuracy than helmet, headband, or skin mounted telemetry systems.

2.5 Neck Strength and Head Acceleration

2.5.1 Anatomy and Role of Cervical Musculature

As introduced in Chapter 2.2.3, neck strength may have an important role in the occurrence of SRC, largely through its potential to affect head acceleration. The cervical spine is supported by a complex musculature that aids in the control and stabilisation of the head and neck (Falla, Jull, Dall'Alba, Rainoldi, Merletti, 2003). Cervical musculature is thought to provide 80% of the mechanical stability of the neck, with the remaining 20% provided by the osteoligamentous structures (Panjabi et al., 1998).

The joints of the second cervical vertebra facilitate movement in three planes: sagittal (flexion-extension), transverse (rotation) and frontal (lateral flexion) (Hay & Reid, 1988). Acting as a first-class lever system, the posterior musculature is responsible for the extension of the neck (Seeley et al., 2014). Originating from the trunk, these muscles insert onto the posterior surface of the skull or the cervical vertebrae (Reid, 1988). The largest and most superficial of these muscles is the trapezius, a large triangular-shaped muscle that overlays the splenius capitis (Seeley et al., 2014). The splenius capitis is largely responsible for the extension of the neck and is aided by the trapezius (Marieb, 2000). The sternocleidomastoid (SCM) is a two-headed muscle that is situated on the anterolateral surface of the neck (Marieb, 2000). Contracting unilaterally the SCM is the prime mover in left and right lateral flexion, supported by deeper lateral muscles such as the rectus capitis lateralis and the scalene muscles. Left- and right-lateral-flexion are also in part accomplished by several posterior muscles including the longissimus capitis, oblique capitis superior, splenius capitis and trapezius (Seeley et al., 2014). A combination of anterior and lateral muscles contributes to flexion of the neck (Seeley et al., 2014). When contracting bilaterally SCM is most prominent during flexion (Vasavada, Li, & Delp, 1998). Whilst the SCM is the prime mover during this action, the scalene muscles, longus capitis and rectus capitis assist in the movement.

Commonly, the larger, more superficial muscles, including the SCM and upper trapezius (UT), are identified as the primary head-neck segment stabilisers (Dezman, Ledet, & Kerr, 2013; Lisman et al., 2010). Despite the primary role of the superficial muscles, it has been reported that segmental instability is more likely when movement is solely produced by the stimulation of larger more superficial muscles (Winters & Peles, 1990). A combination of deep and superficial muscle activation is thought to be a prerequisite for cervical spine stiffness and stabilisation. Using intact and injured muscular spine segments the effect of muscular forces on cervical stabilisation was investigated (Kettler, Hartwig, Schultheiß, Claes, and Wilke, 2002). It was reported that muscle forces from the longus coli stabilise the cervical spine during all loading and injury states, reducing the range of motion to less than 50%, compared to 100% without muscular stimulation. Therefore, it is important to consider both deep and superficial neck musculature when focusing on head-neck segmental stabilisation.

2.5.2 Non-Contact Sports

The relationship between neck strength and head acceleration has received a lot of attention in soccer heading. Mansell, Tierney, Sitler, Swanik, and Stearne (2005), assessed the effectiveness of an eight-week resistance training programme on head-neck dynamic stabilisation in collegiate soccer players. These authors observed male and female flexion isometric strength to increase by 15% following training; however, only female participants saw a significant increase in extension strength. Results also revealed no effect of neck strength training on dynamic stabilisation during force application. The absence of significant increases in neck extension strength likely explains the lack of change in dynamic stabilisation in males. The flexion and extension strength increases in females may have been insufficient to effect change in dynamic stabilisation during force application.

Most recently, Peek et al., (2020), in a systematic review, concluded that current research from four cross-sectional studies supported the assertion that increasing neck strength may reduce head accelerations that occur during soccer heading. Using a simulated soccer heading drill, a significant, moderate, negative relationship was seen between neck strength and subsequent head acceleration (Gutierrez, Conte, & Lightbourne, 2014). These results suggest that those who had weaker neck strength experienced a greater inertial load whilst heading the ball. The findings of this study are somewhat underpowered (N = 17), and head accelerations were measured using a triaxial accelerometer attached to a headband. This measurement technique presents potential overestimation of head acceleration due to insufficient sensor-skull coupling, as discussed in Chapter 2.4.2 (Wu et al., 2016). Despite these limitations, Caccese et al., (2018) added support to the results. These authors reported that variables such as neck size and strength explained 22% and 15% of the variance in head acceleration. These results indicate that in soccer players, those with increased head mass, neck girth and neck strength may experience lower magnitude head impacts. This study had several strengths, the sample population was large (N= 100) with a relatively even split of males and females, taken from a variety of age groups suggesting the results may be relevant to a wide sample of soccer players. The findings from both studies are specific to one sport and one specific event, which may not be applied to other sporting situations.

2.5.3 Contact Sports

The relationship between neck strength and head acceleration in contact sports is yet to be fully determined. Eckersley et al., (2017) investigated the effect of neck strength on mitigating head acceleration following a direct blunt impact to the head. Following the examination of different athletic scenarios (80 *g* helmeted impact and 40 *g* head impact), it was concluded that increasing neck strength had no critical effect on head acceleration. Eckersley et al., (2017) recommended that efforts should not be put into increasing neck strength, and focus should be directed towards other SRC prevention strategies. This study only focused on direct impacts to the head at high magnitudes.

It is possible that cervical muscle force may influence head acceleration caused by indirect impacts, transferred to the head from the upper and lower body. This study also utilised a computer model that was derived from cadaver head drop tests using helmet-mounted sensors. Therefore, this model may have been based on inaccurate metrics due to helmet-mounted sensors overestimating head accelerations (Cummiskey et al., 2017; Wu et al., 2016).

Within youth ice hockey, Mihalik et al., (2011) found no significant differences in linear and rotational acceleration between players with weak, moderate and strong cervical muscles. Interestingly, those with the weakest UT were seen to experience lower accelerations than stronger players. Like previous studies, head acceleration was measured using HITS. This may explain why neck strength was observed to have no effect on head acceleration. Having a stronger neck is unlikely to limit excess helmet movement or affect the coupling of the sensors to the skull. Similarly, Schmidt et al., (2014), utilising HITS in AF, suggested that those with stronger cervical muscles may be at greater risk of sustaining a higher magnitude head impact. These authors reported no differences in sustaining moderate and severe head impacts between those with stronger and weaker cervical muscles. However, those with stronger cervical muscles had 1.75 times increased chance of sustaining moderate compared to mild impacts. Additionally, those with greater muscle cross-sectional area (CSA) had increased odds of experiencing severe impacts. As proposed by the authors, this may be explained by a phenomenon known as risk compensation (Hagel & Meeuwisse, 2004; Hedlund, 2000). Risk compensation is the notion that individuals have a target level of risk in which they strive to maintain (Wilde, 1982). If an individual perceives that their level of risk of, sustaining high magnitude head impacts, is reduced (due to knowledge of increased neck strength) then they will attempt to change their behaviour to maintain their desired level of risk. This may also explain the results observed in youth ice hockey players (Mihalik et al., 2011).

Jin et al., (2017) using FEM, examined the role of cervical muscle activity on the risk of mTBI in AF. The head impact conditions applied to their model, were representative of a direct head-to-head collision along the transverse axis that was recorded by Viano, Casson, & Pellman, (2007). The FEM was used to compare four conditions: no muscle, late muscle activation, early muscle activation and stronger muscle. Having stronger neck muscles and early muscle activation was generally reported to reduce all calculated injury criteria. They also found stronger muscles to reduce peak rotational velocity, compared to no muscle trials. These results indicate that the strength of cervical muscles may have a role to play in reducing the rotational load experienced during an impact. This study, however, only investigated one specific head impact event from one sport. Furthermore, the maximum translational acceleration seen was upward of 110 *g*, given the average head impact acceleration values seen in previous contact sport studies (Broglio et al., 2009; King et al., 2018; King et al., 2015; Rowson et al., 2009), these metrics may have limited application to real-word sporting scenarios. These findings, however, may be supported by a recent study investigating the effect of neck-specific training on head kinematics was assessed in youth contact sport athletes (Eckner et al., 2018). Results reported significant decreases in head linear and angular velocity, following eight-weeks of neck-specific training, in all movement directions except flexion, with the largest decreases observed in angular velocity. This adds further support to the notion that increasing neck strength is effective in reducing head angular/rotational velocity when subject to an external force.

2.5.4 Rugby

Limited investigation has been conducted into the relationship between neck strength and head acceleration within rugby. Dempsey, Fairchild, & Appleby (2015) reported general correlations between an increase in neck strength and a reduction in head acceleration, using 3D motion capture during a simulated tackle event. More specifically, the strongest correlations were seen between increased flexion and extension strength and reduced medial and lateral linear and angular head accelerations. This relationship may be representative of the bilateral muscle contractions produced, during flexion and extension, by the muscles used unilaterally during lateral flexion. Results from this study provide a rationale for the use of neck strengthening programmes to mitigate head acceleration in rugby. This study had a limited sample size (N = 10) and is seemingly underpowered, which may be the reason for the limited statistical power of the correlations seen. Dempsey et al., (2015) also only focused on one specific tackle situation and cannot be generalised to all contact situations. Furthermore, only the kinematics of the ball carrier were analysed; research has found that tackling players may be at greater risk of experiencing head impacts in rugby (Tierney et al., 2016). Therefore, further research needs to be conducted analysing a range of contact situations investigating both the ball carrier and the tackler.

More recently, Bussey et al., (2019) observed that during simulated rugby tackles, males and females who had a history of concussion within the previous 12 months, experienced significantly higher magnitude head accelerations. Moreover, this elevated head acceleration was associated with reduced muscle activation within the cervical muscles. These results indicate the presence of a relationship between cervical muscles and head acceleration, suggesting that increasing the amount of cervical muscle activation during a rugby tackle may reduce subsequent inertial load. This study, however, has several limitations. Firstly 'punch bags' were used to simulate contact which may not fully represent the mechanics of an actual rugby tackle. Secondly, soft-tissue mounted sensors were used to measure head accelerations during the tackle which may be associated with measurement error (Wu et al., 2016). Similarly, there was no rotational acceleration data recorded for males so the effect of cervical muscle activation on that parameter is unknown. The absence of rotational acceleration data limits the relevance of these results as this kinematic parameter is thought to play a primary role in brain SRC (Patton et al., 2013; Tierney & Simms, 2017).

Results in soccer provide a strong argument for strengthening cervical musculature to mitigate head acceleration during head impact events (Peek, Elliot and Orr 2019). However, the relationship between neck strength and head acceleration in less clear within contact sports, with studies producing conflicting results (Eckersley et al., 2017; Jin et al., 2017; Mihalik et al., 2011). Furthermore, within rugby specifically, there are limited findings from which to draw conclusions, all of which are derived from lab-based research (Bussey et.al., 2019; Dempsey et al., 2015). This relationship requires further investigation to better evaluate the use of neck strength training as a head injury prevention tool.

2.5.5 Neck Strength Imbalances and Anthropometric Variables

Improving agonist/antagonist balance has been reported to reduce the incidence of injury in various areas such as the shoulder and hamstring (Croisier, Ganteaume, Binet, Genty, & Ferret, 2008; Yeung, Suen, & Yeung, 2009). Similarly, improving muscular balance may play an important role in reducing the magnitude of head accelerations

(Dezman et al., 2013; Morimoto, Sakamoto, Fukuhara, & Kato, 2013; Peek et al., 2020). Dezman et al., (2013) found strength symmetry in cervical flexors and extensors to reduce head acceleration during soccer heading in collegiate soccer players. It was proposed that agonist-antagonist symmetry acts to increase the effective mass of the head as well as limiting head oscillations during the heading movement.. Furthermore, Morimoto et al., (2013) observed that co-contraction of neck flexors and extensors improved head neck stability in high school rugby players during a heads-up tackle. Whilst this relationship requires further investigation, studies aiming to reduce head accelerations should consider improving neck extensor/flexor balance as well as increasing maximal and functional strength.

Anthropometric variables such as neck circumference, head mass and neck-to-head circumference ratio have been also reported to be associated with increased risk of brain injury. Tierney et al., (2005), suggested that gender differences in head angular acceleration, in soccer, may be due to significant differences in neck strength, neck circumference and head mass. This may be supported by Caccese et al., (2018) who, using 3D motion capture to measure head acceleration during a soccer heading drill, reported size variables such as head mass and neck circumference to account for 22.1–23.3% of the variance in PLA and PRA. This was seen to be a stronger predictor than strength variables. It should be noted, however, that head mass values were calculated based on a percentage of body weight, and may not accurately represent actual head mass. Relating these findings to injury risk, analysis of high school athletes from a wide variety of sports, found those who experienced an SRC to have significantly lower neck circumference and neck-to-head circumference ratio compared to their non-injured counterparts (Collins et al., 2014). However, further assessment revealed that neither of these variables were a signficant predictor of SRC risk. The current research into anthropometric neck strength variables suggests that variables such as neck cirumference, head mass and neck-to-head circumference ratio may influence head acceleration but have a limited effect on the risk of SRC. Further research is required to determine this relationship in rugby.

2.6 Neck Strength Measurements

2.6.1 Correlates of Neck Strength

There are a limited number of studies that have examined the different anthropometric variables that correlate to and may predict neck strength Hamilton et al., (2014) investigated the neck strength profiles of under-18 and adult-male front-row rugby players. These authors concluded that playing experience ($r = 0.50$), weight ($r = 0.40$) and age ($r = 0.50$) were most strongly related to neck strength, with grip strength ($r = 0.2$) showing poor association. Furthermore, a combination of playing experience and player weight was reported to account for 31% of the variance in player neck strength. In contrast to this, Salmon, Sullivan, Handcock, Rehrer, and Niven, (2018) found no significant correlation between age and neck strength in amateur, adult-male rugby players. This difference may be due to the extent of the age difference between participants in Hamilton et al., (2014) with ages ranging from 16–50 years old. It is likely that age would be a contributing factor to strength, given different adolescent maturation rates and age related decline in strength (Keller & Engelhardt, 2013; Lindle et al., 1997). Conversely, the latter study only included adults, making the contribution of age unlikely. In support of Hamilton and colleagues, the latter study also found neck strength to be significantly correlated to body weight ($r = 0.30\text{-}0.35$; Salmon et al., 2018). The agreement between the two studies supports the use of this anthropometric variable to predict neck strength.

Salmon et.al (2018) reported neck girth to be significantly correlated to neck strength ($r = 0.33\text{-}0.63$). This is likely due to the established link between muscle size and muscle strength (Maughan, Watson, & Weir, 1983). This may be supported by a study in male and female soccer players in which males had greater neck strength than females, as well as greater neck girth, suggesting a potential link between the two variables (Mansell et al., 2005). Due to the limited nature of the research in this area, further investigation is required to establish the most effective predictive variables of neck strength.

2.6.2 Neck Strength in Rugby

Rugby players' neck strength has been widely assessed at a variety of playing levels, using a range of testing methods (Geary, Green, & Delahunt, 2013; Geary et al., 2014; Naish et al., 2013; Salmon et al., 2018). Using fixed-frame dynamometry, Salmon et al., (2018) reported amateur rugby players to produce the greatest force in Ext (254 N), followed by flexion (231 N) and right- (182 N) and left- (169 N) lateral-flexion. These values are lower than those observed in professional players using similar methods (Naish et al., 2013). However, it is important to note methodological differences between the two studies in relation to the position in which participants are tested in. Specifically, Salmon et al. (2018) placed participants in a prone position to simulate a rugby contact posture, whereas Naish et al. (2013) placed participants in an upright, seated posture. Despite this difference, both testing methods have been reported to produce good reliability in the assessment of neck strength (Salmon, Handcock, Sullivan, Rehrer, & Niven, 2015; Ylinen, Rezasoltani, Julin, Virtapohja, & Mälkiä, 1999). Furthermore, greater neck strength in professional players is a consistent finding across neck strength studies (Geary et al., 2013, 2014; Naish et al., 2013; Salmon, 2014) and differences between amateur and professional players are likely reflective of the increased performance demands in professional sport.

Discrepancies, however, have been observed between the neck strength recorded in similar populations using different testing methods. Naish et.al (2013) reported neck strength values in a professional cohort of 368, 278, 362 and 376 N for extension, flexion, and left- and right-lateral-flexion, respectively. Conversely, a further study in professional players reported substantially greater values of 606, 335, 556 and 570 N respectively (Geary et al., 2014). Differences may be due to the testing methods used in each study. Naish et.al (2013) utilised fixed-frame dynamometry with participants in a seated posture, strapped to a bench with feet on an unstable surface to limit the use of the legs and recruitment of accessory muscles. Conversely, Geary et.al (2014) utilised hand-held dynamometry, with participants in a seated posture but no measures were in place to limit the use of the legs or accessory muscles. It is likely that the excess force reported in the latter study is representative of the contribution of other muscles to force production. This highlights the importance of standardised measurement techniques and protocols when comparing neck strength across studies.

2.7 Neck Strength Training

2.7.1 Importance of Specificity

Resistance training specificity refers to the notion that greater increases in strength are observed when training mirrors or is similar to the activity required during testing

(Behm, 1995; Saeterbakken et al., 2016). In other words, training should involve similar muscular co-ordination, contraction type, movement patterns and joint positions that a specific task requires (Buckthorpe, Erskine, Fletcher, & Folland, 2015; Rutherford & Jones, 1986; Saeterbakken et al., 2016). The importance of resistance training specificity with regards to improving neck strength has been previously reported (Conley et al., 1997). These authors compared the effects of a 12-week neck-specific resistance training to a generalised resistance training programme of the same duration on cervical muscle size and strength. The neck strength training group completed 3x10 repetition maximum (RM) extension exercises, three times per week for 12 weeks. Results revealed increases in neck muscle CSA and extension strength of 13% and 34% respectively. Conversely, generalised resistance training produced no significant change in either of the variables. This study only assessed extension and findings may not be applicable to multiple neck movements such as flexion or left- and right-lateral-flexion. Despite the limitations, the results of this study highlight the importance of neck training specificity if the goal is to increase cervical muscle strength.

More recently, the effectiveness of neck-specific resistance training was assessed in youth contact sport athletes (Eckner et al., 2018). These authors observed increases in neck strength in those who took part in general resistance training as well as those who participated in neck-specific resistance training. Despite neck strength gains seen in both groups, the neck-specific resistance training group recorded 2.6 times greater increases compared to general resistance training. Of note, baseline neck strength was, in general, greater in the general resistance training group compared to the neck strength group and this may have influenced the magnitude of the observed increases. Further, the sample size in Eckner et al., (2018) was small and group allocation was imbalanced with fewer participants in the general resistance training group. This may have limited the power to detect changes in neck strength.

Previous studies have suggested that participation in rugby may provide enough stimulus to facilitate increases in neck strength, without the use of specific resistance training. One study in amateur male rugby reported significant increases in neck strength in backs and forwards following a season of rugby, compared to non-rugby playing controls (Salmon et al., 2018). This study, however, also reported increases in neck pain following a season of rugby. In contrast to this, a season of rugby was seen to lead to decreases in neck strength in professional players, compared to increases that were observed in players who took part in specific neck resistance training (Salmon, 2014). Furthermore, participation in resistance training was seen to be effective in preventing increases in neck pain. This may suggest that the effect that participation in rugby has on neck strength is dependent on the demands a specific season places on the individual, as well variables such as the respective playing level, playing position and the occurrence of injuries. Despite this, it appears that neck-specific resistance training is a safer method of facilitating increases in neck strength compared to relying on rugby participation alone.

2.7.2 Specific Neck Training Protocols

Numerous studies have employed neck-specific resistance training programmes in an attempt to increase neck strength, a summary of these can be seen in **Table 2.1** (Barrett et al., 2015; Conley et al., 1997; Geary et al., 2014; Naish et al., 2013; Salmon, 2014; Salmon et al., 2013). Due to differences in training programmes, testing procedures and conflicting results, the most effective neck resistance training programmes remain unclear.

A number of studies of the same duration, prescribing only dynamic movements at a similar resistance, found their resistance training programme to result in increased neck strength (Lisman et al., 2012; Mansell et al., 2005). Mansell et al., (2005), reported neck-specific training to result in increased flexion strength (15%) in male and female collegiate soccer players, with only females reporting increases in extension strength (22.5%). The absence of significant differences in extension strength for males may be reflective of the significantly greater baseline strength compared to females. Similar neck-specific resistance training protocols were reported to lead to increased extension strength (7%) and left-lateral-flexion strength (10%) in male college AF players (Lisman et al., 2012). The lack of significant increases in all directions may be due to the low resistance (55-80% of an individual’s 10 RM). It is recommended that training loads of 60-70% one RM should be used to elicit strength gains in resistance training (American College of Sports Medicine, 2009). Further, only dynamic movements were used in these training programmes. It is recommended that a mixture of dynamic movements through eccentric and concentric contractions and isometric exercises should be included in resistance training programmes to increase strength (American College of Sports Medicine, 2009).

Naish et al., (2013) investigated the effect of a neck resistance training programme in men's professional rugby using both dynamic and isometric exercises. These authors prescribed a training programme that consisted of isometric holds and controlled movements through eccentric and concentric contractions in flexion, extension, left- and right-lateral-flexion for a total of 26 weeks (13 weeks increasing strength, 13 weeks of maintenance). The training was completed two to three times a week during the strength phase and one to two times a week during the maintenance phase, with resistance ranging from 70% one RM and maximum resistance for repetitions. Naish et al., (2013) reported non-significant increases in isometric neck strength in all directions following the first five weeks of their strengthening programme. Interestingly neck strength was not re-assessed at the end of the 13-week strength development stage, where significant increases may have been seen. These findings are supported by Geary, Green, and Delahunt (2014) who implemented a five-week neck strengthening programme in a similar population. These authors found significant increases in isometric neck strength in all directions (flexion, extension, and left- and right-lateral-flexion). The reasons for these discrepant findings may be methodological. The resistance used in the training protocol was not quantified in Geary et.al (2014), with manual pressure provided by the strength and conditioning coach. Therefore, it is possible this resistance was greater than the previous study, eliciting more significant changes in strength. Despite this method producing desirable results, it may present issues with reproducibility, reliability, and safety.

Table 2.1: Summary of neck strength training protocols and outcome measures from previous literature.

Study	Participants	Test	Exercise	Results
Barrett et.al., 2015	N = 34 (Test = 17, CON = 17) Secondary School rugby Players	MVC[e] & submax gatherer harness.	6wk, 3 days/wk isometric, 50% MVC 4 sets of 6 reps[f] 2-1-2-1	No change
Conley et.al., 1997	N = 22 Active college students	Gravity dependant head harness, max body mass for 3 sets of 10	12 wk, 4 days/wk Dynamic Ext, 3 x 10RM	↑Neck CSA (13%) ↑ Extg (34%)
Geary et.al., 2014	N = 25 Professional and Semi-Professional rugby Players	Handheld Dynamometer	5wk, 2 days/wk Isometric	↑Flxg, Ext, Rflx[i], Lflx[j]
Lisman et.al., 2012	N = 16 College AF players	Force gauge and selectorized Pro 4-way neck training machine	8wk, 2-3 days/wk Dynamic movements 60-80% 10RM 3 sets of 10 reps	↑Ext (7%) ↑Lflx (10%)
Mansell et.al., 2005	N = 36 (17 males, 19 females) Division 1 collegiate Soccer Players	Handheld dynamometer	8wk, 2 days/wk Dynamic movements 55-70% of 10RM 3 sets of 10 reps	↑Flx (15%) ↑Ext female (22.5%)
Naish et.al., 2013	N = 27 professional rugby Players	Head harness and load cell	26wk, 1-3 days/wk Isometric, 70% 1RM or max body mass for reps. 2-3 sets of 3-12 reps	Non sig ↑ in all directions
Salmon et.al., 2013	CTP[a]; n = 10 ETP[c]; n = 11 CON[d]; n = 8 Canadian Helicopter Pilots	MVC & 70% submax to fatigue Head harness	12 wk, CTP low load ISOM for DNS[b], Dynamic movements at 30% MVC, 3 sets of 10 reps ETP Dynamic movements at 30% MVC CON no neck training.	CTP; ↑Flx (13.8%), ↑Rflx (15.9%) ETP; ↑Rflx (14.4%)
Salmon, 2014	Test; n = 29 CON; n = 27 Professional New Zealand rugby Players	MVC & Submax Custom Built ISOM testing device	31wk 1-3 days/wk Combination of co-ordination, dynamic movements (30% MVC, 3 sets of 10 reps), isometric (50% MVC 15s hold for 3 reps) and impulsive loading	NG; ↑Flx, Lflx, Rflx No change in Ext CON ↓ all directions

Note, [a]CTP = co-ordination training, [b]DNS = deep neck stabilisers, [c]ETP = endurance training, [d]CON = control, [e]MVC = maximum voluntary vontraction, [f]Reps = repetitions, [g]Ext = extension, [h]Flx = flexion, [i]Rflx = right-lateral-flexion, [j]Lflx = left-lateral-flexion.

2.7.3 Importance of Deep Neck Stabilisers

Salmon et al., (2013) demonstrated the importance of isolating the deep neck stabilising muscles (DNS) as part of resistance training to enhance neck muscle function. This study in helicopter pilots compared three intervention conditions. The first was a neck strengthening programme that focused on training the larger more superficial muscles through resisted dynamic cervical movements. The second programme had three stages. The first stage focused on isolating DNS muscles during isometric contractions. The second stage integrated limb motion. The final stage focused on strengthening superficial muscles with resisted dynamic cervical movements whilst incorporating the deeper muscles using a slight chin nod. The third condition was a control group that performed no neck-specific exercises. These authors reported that incorporating deep neck muscle exercises increased isometric flexion and right-lateral-flexion by 13.8% and 15.9%. Whereas the programme that focused on solely strengthening superficial muscles was only effective in increasing right-lateral-flexion. Despite being non-significant, differences were also seen in extension endurance, with a 10.8% increase in the DNS group compared to a 4.2% increase in the superficial muscle group. These findings suggest that an neck strength training programme should focus on training DNS muscles as well as the larger more superficial muscles. It should be noted, however, that adherence to the training programme in the superficial muscle group was less than (50%) the DNS group (76%) which may have influenced improvements seen. The starting resistance used in this study was 30% of participants maximum voluntary contraction (MVC), other studies have utilised higher resistance loads and produced greater increases in isometric strength (Conley et al., 1997) and this should be considered when determining the most effective starting resistance.

Salmon, (2014) explored the efficacy of a multifaceted training program further in a cohort of professional male rugby players. In contrast to the Salmon et al., (2013), Salmon (2014) used both isometric exercises at 60% MVC as well as dynamic movements at 30% MVC. The intervention group showed increases in isometric neck strength for flexion and left- and right-lateral-flexion with extension remaining unchanged. In contrast, the control group demonstrated reductions in neck strength in all four directions. These results suggest that a season-long multifaceted neck strength training programme is effective at increasing isometric neck strength and may also mitigate the natural loss in extension strength. These improvements were only observed in those in the high adherence group (> 25.37% in season). These results suggest that, whilst a level of training compliance is required, adherence can be low and still attain some improvements in NS, potentially indicating a high level of cervical muscle sensitivity to strength training. Salmon et.al (2014) is unique in its use of a custom-built isometric neck strength testing apparatus. The equipment utilised load cells and a simulated contact posture to assess cervical muscle strength, which has been seen to be a reliable measurement tool (Salmon et al., 2015). This technique may only be relevant to contact sports and the specific testing position. Results obtained using this technique may not be directly compared to results achieved using previous measurement techniques such as hand-held dynamometry in seated postures.

Results from previous studies suggest that a programme aiming to increase neck strength should include the following.

- Isometric and dynamic (eccentric-concentric) movements (American College of Sports Medicine, 2009; Naish et al., 2013; Salmon et al., 2013).
- DNS and superficial muscle exercises (Salmon, 2014; Salmon et al., 2013)
- Resistance load of 60-80% MVC (American College of Sports Medicine, 2009; Naish et al., 2013; Salmon et al., 2013)
- Two to three training sessions a week (Geary et al., 2014; Lisman et al., 2012; Mansell et al., 2005; Naish et al., 2013; Salmon et al., 2013)

In conclusion, existing research presents unclear and contradictory results with regards to the effect of neck strength and increasing neck strength on mitigating head acceleration during impact events. Furthermore, whilst other sports such as soccer and AF have received a lot of attention in this area, rugby union is yet to be fully investigated. Studies have attempted to increase neck strength in rugby players, but few have looked at the subsequent effect on head accelerations. Similarly, the head impact measurement techniques used in previous studies are associated with high levels of measurement error, creating uncertainty in the reliability of the results.

Chapter 3 : Methodology

3.1 Participants

Overall, 31 male British University (BUCS) Super Rugby players provided written informed consent to participate in the study. Participants that had a history of neck injury and/or neck pain, were advised to seek medical clearance prior to taking part in the study.

3.2 Demographics and Anthropometric Measurements

Prior to testing, participants completed a questionnaire including their age, sex, sports participation history and injury history. Anthropometric measures including standing stature (Portable Stadiometer, Seca, 213), body mass (Digital Analogue Scale, Seca, 761)**,** head and neck circumference (Ergonomic Circumference Tape, Seca 201), and shoulder width (Tree Calliper, EIA, 2802), were obtained. Head circumference was recorded to the nearest 1 mm using anthropometric tape which was placed across the frontal bones of the skull, perpendicular to the long axis of the face and above the ears and over the occipital prominence. Neck circumference was measured below the larynx in the horizontal plane using anthropometric tape and recorded to the nearest 0.2 mm. Finally, shoulder breadth was measured as the distance between the most lateral points on the right and left acromion processes when the participant was seated with their arms relaxed by their sides.

3.3 Measurement of Inertial Loading of the Head

Head impact events sustained during BUCS matches were measured using the PROTECHT™ instrumented mouthguard (iMG) system (Sports Wellbeing Analytics Ltd, Swansea, UK). The PROTECHT™ system has shown high accuracy and reliability (Greybe et al., 2020). Dental impressions were taken from each participant to ensure the iMG was custom-fit to ensure tight sensor-skull coupling. The iMG was then worn by the participants in 13 competitive games throughout the season (November 2019 – April 2020).

The IMG system contains an embedded 9-axis IMU (LSM9DS1, STMicroelectronics, Genova, Switzerland) and an additional triaxial accelerometer (H3LIS331DL, STMicroelectronics, Genova, Switzerland). The iMG samples over a 104 ms period, at 1000 Hz (linear accelerometer) and 952 Hz (gyroscope, measuring rotational velocity), with a 16-bit resolution and ranges of ± 200 *g* and ± 35 rad•s^{-1} respectively. The raw data is then transmitted via radio frequency to a computer and stored as a time-series CSV file. The iMG measures any head impact event >10 *g* (measured with the linear accelerometer). Rotational acceleration is derived from angular velocity using a five-point stencil derivative. The system also contains a proximity sensor to ensure accelerations are only recorded when the guard is coupled to the participant's teeth. The PROTECHT™ system provides real-time maximum values for peak linear (PLA) and peak rotational (PRA) acceleration of the iMG. These maximum values were used for the analysis in this study, which were then compared to video footage obtained in each game to validate the impact and understand its context and characteristics.

3.4 Data Processing, Head Impact Verification and Event Classification

3.4.1 Filtering

Following data collection, a low pass, 4^{th} order, zero-lag, Butterworth filter was applied to the raw accelerometer and gyroscope time-series data of each recorded head impact. Variable, impact-specific filter cut-off frequencies were determined using residual analysis. These have been shown to provide more consistent results for short duration impacts than fixed filter cut-off frequencies (Greybe et al., submitted 2020; **Figure 3.1** and **Figure 3.2**).

3.4.2 Impact Verification

To determine if the impacts recorded from the PROTECHT™ system were true positives, each impact was scrutinised using an extensive classification system (**Figure 3.3** and **Figure 3.4**). This process utilised subjective and objective criteria to form the most accurate assessment of each impact. The system required each impact to progress through two main criteria (Video and System) each with several sub-criteria. The video criteria required players to be on the pitch and involved in an obvious contact event at the time of impact. The system criteria involved extensive waveform analysis to determine if the waveform produced was representative of a realistic impact event. Examples of true and false positive impacts are given in **Figure 3.5** and **Figure 3.6** respectively. If an impact lacked sufficient waveform data to make an informed decision, then it was excluded from the analysis (**Figure 3.7**). Similarly, if at any stage an impact did not meet the video or system conditions, then it was removed from the final analysis. This was to ensure that the data analysed was representative of true head impact events. It should be noted that although this process allowed data to be as accurate as possible, it is not currently possible to completely distinguish between false positive and true positive impacts, due to the complex nature of the events.

3.4.3 Impact Classification

Once verified, each impact was coded and classified based on event type and cause of head acceleration (**Table 3.1**). Videos were analysed using a coding system to ensure the accuracy and consistency of results. Definitions for event type were informed by Hendricks et al., (2020). Impacts were then grouped for comparison across broad positional groups as forwards and backs. Forwards consisted of playing numbers 1-8 and backs consisted of playing numbers 9-15. Impacts were also grouped based on specific positional groups. This consisted of front-row (numbers 1-3), second-row (numbers 4 & 5), back-row (numbers 6-8), half-backs (numbers 9 & 10), inside-backs (numbers 12 & 13) and outside-backs (numbers 11 & 15).

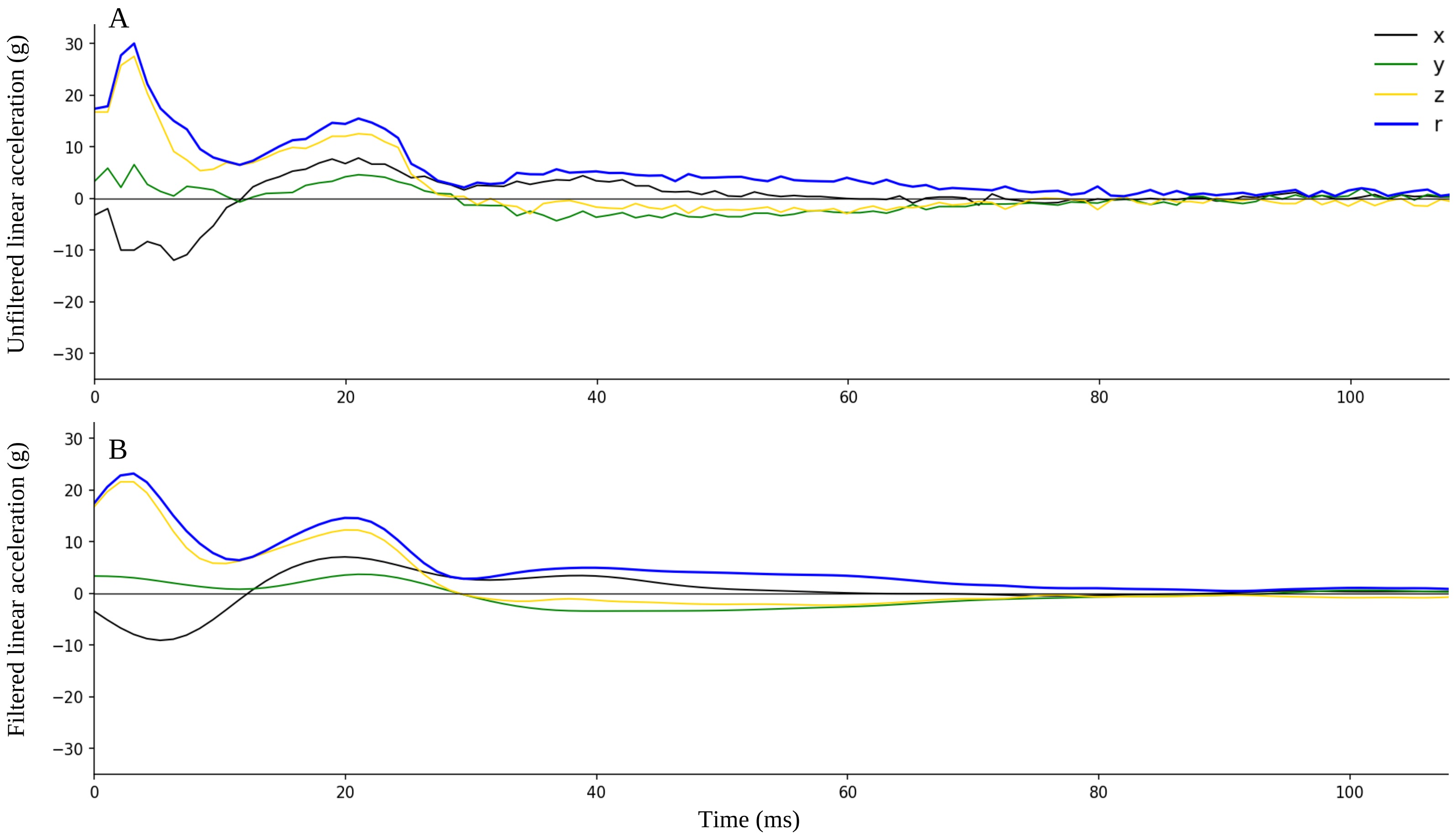

Figure 3.1 An image from the filtering software showing an example of an unfiltered (A) and filtered linear acceleration (B) waveform. Note, r = the resultant of the x, y and z components of linear acceleration.

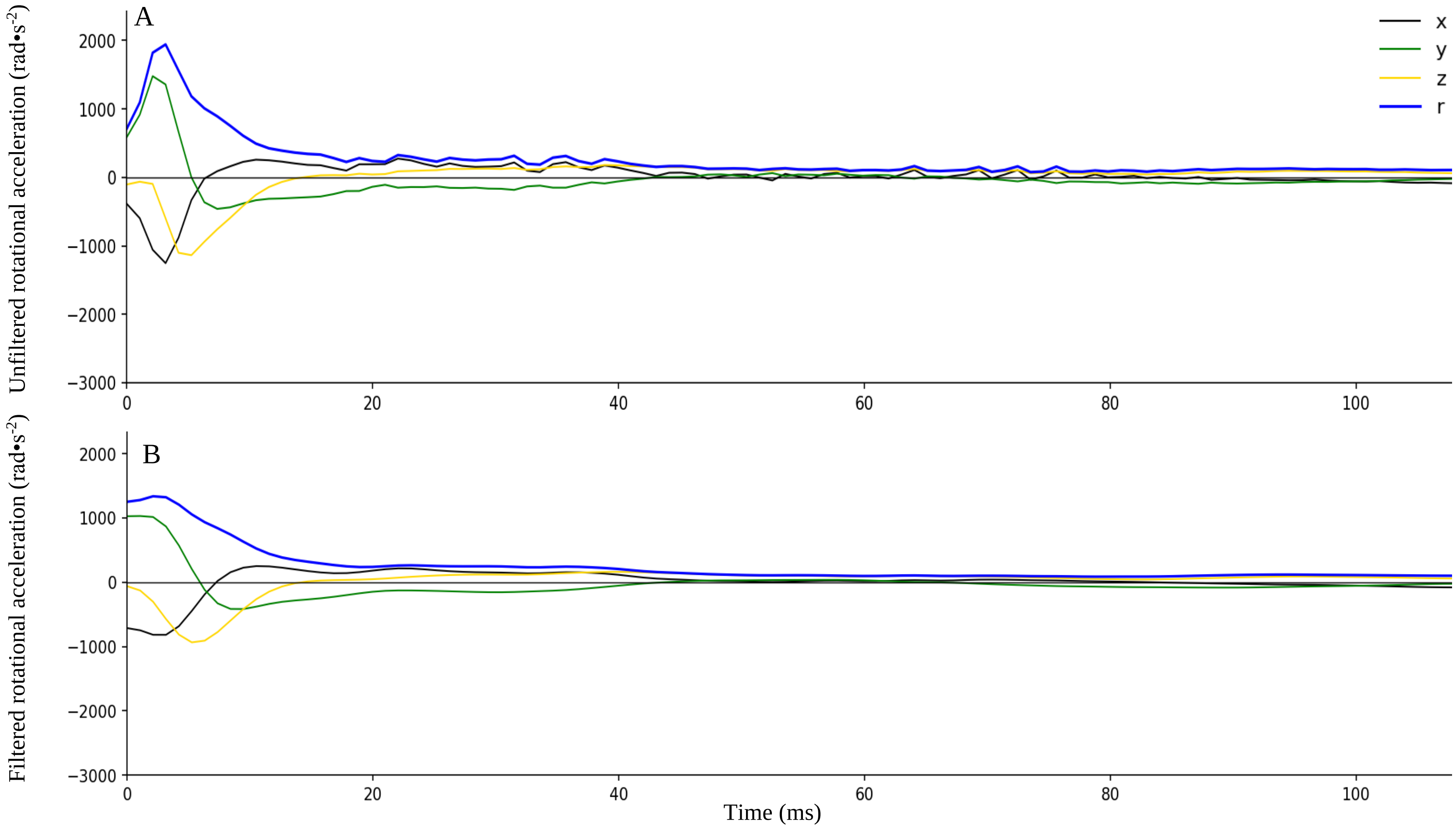

Figure 3.2: An image from the filtering software showing an example of an unfiltered (A) and filtered rotational acceleration (B) waveform. Note, r = the resultant of the x, y and z components of linear acceleration.

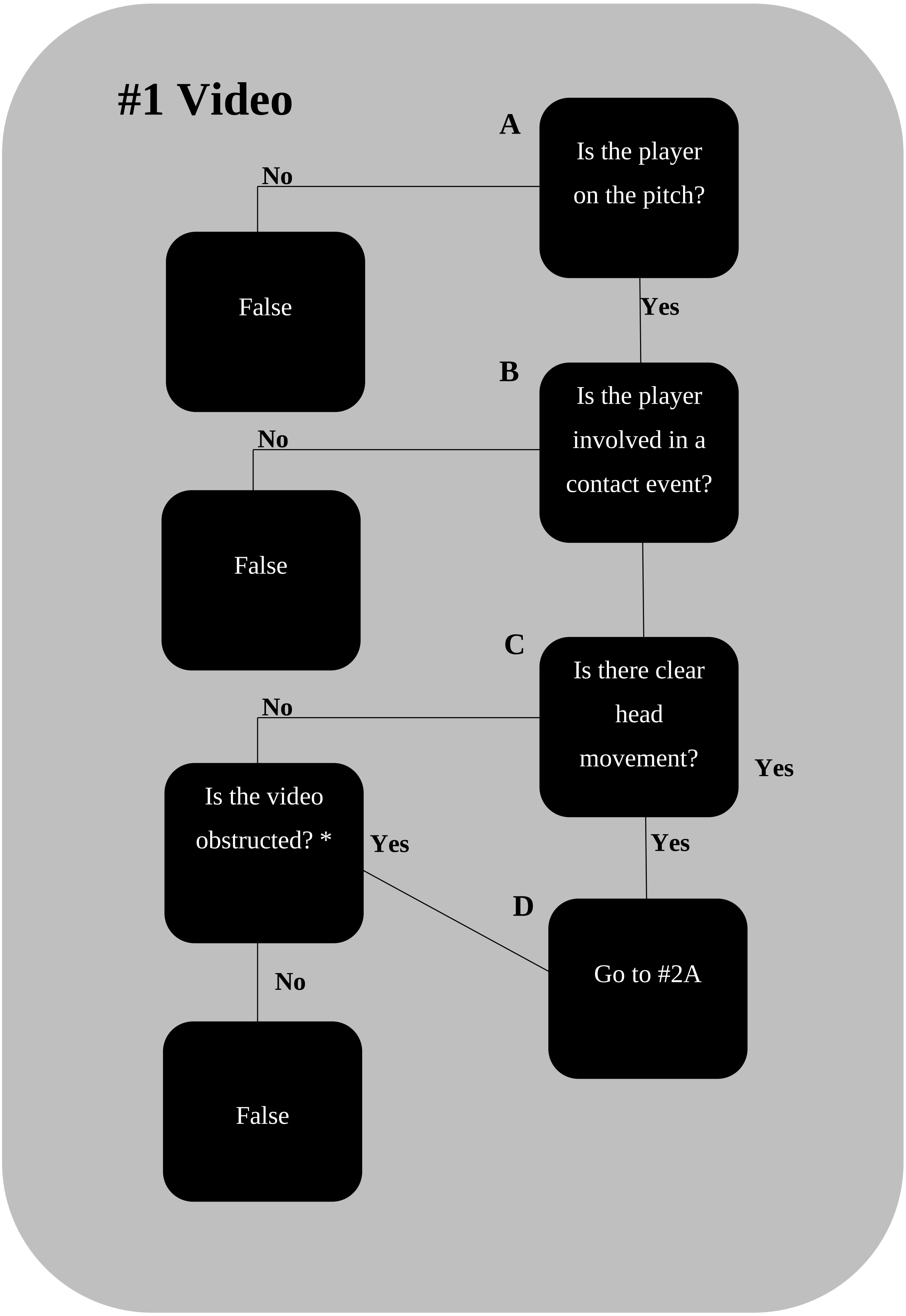

Figure 3.3: Stage One of the head impact verification process. Note, *Is the players head obscured from view such as being in a ruck or maul?

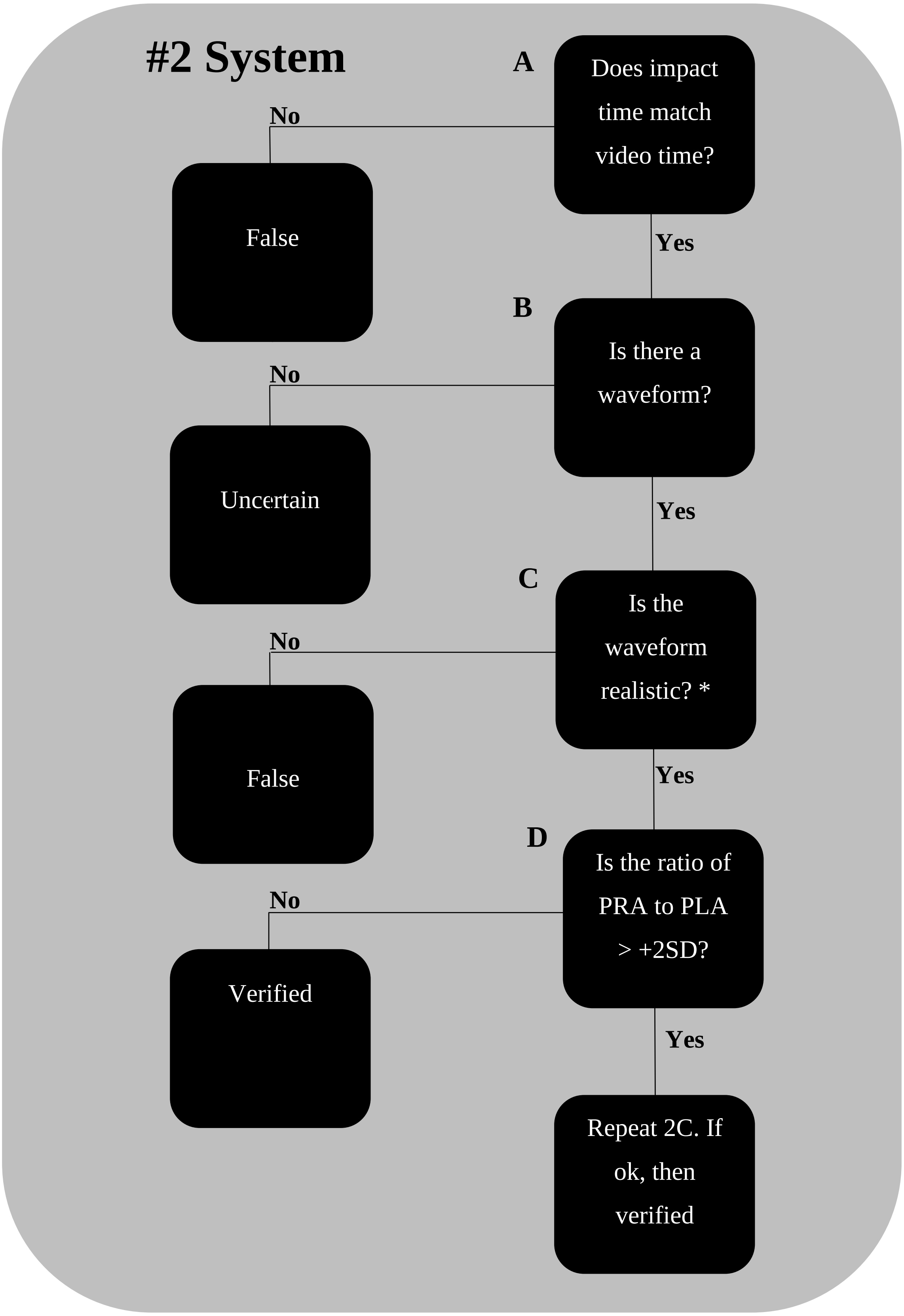

Figure 3.4: Stage Two of the head impact verification process. Note, *see **Figure 3.5** and **Figure 3.6.**

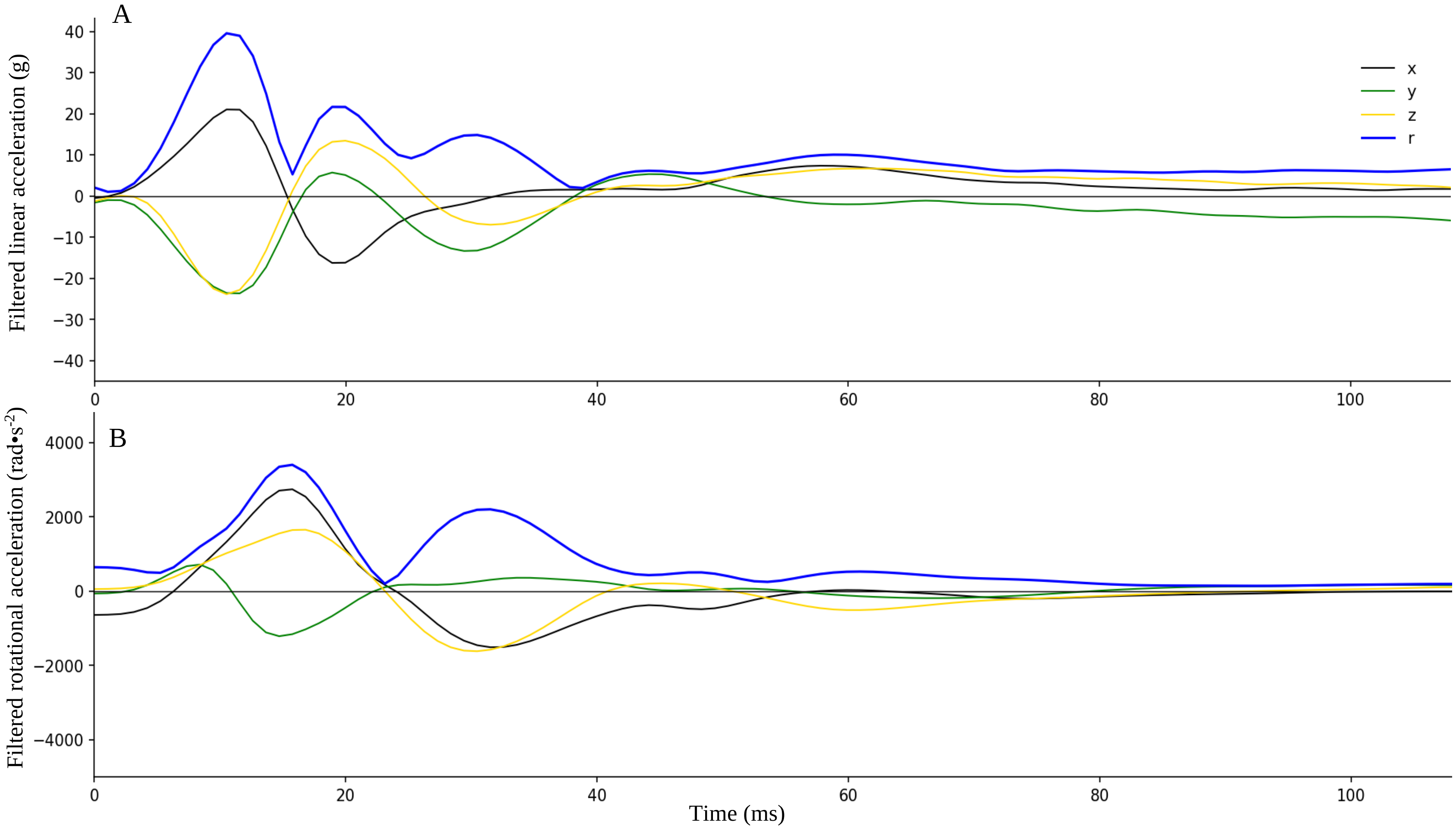

Figure 3.5: An image from the filtering software showing an example of an filtered linear acceleration (A) and filtered rotational acceleration (B) waveform from a true positive impact. Note, r = the resultant of the x, y and z components of linear acceleration.

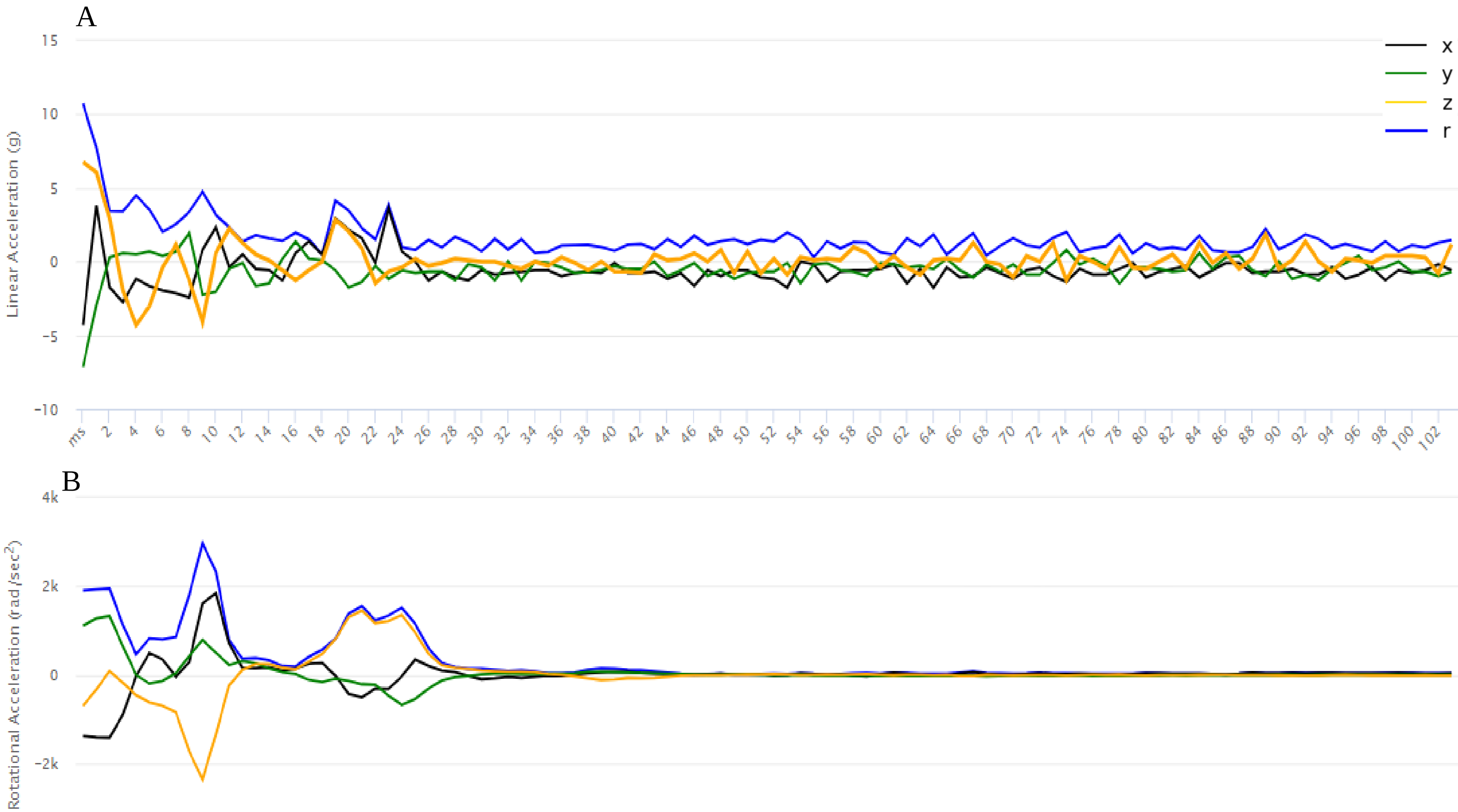

Figure 3.6: An image from the mouthguard software showing an example of linear acceleration (A) and rotational acceleration (B) waveform from a false positive impact. Note, r = resultant of the x, y, z acceleration components

Figure 3.7: An image from the mouthguard software showing an example of an impact event that was excluded from the final analysis due to the absence of sufficient linear acceleration (A) and rotational acceleration (B) waveform data. Note, aMax = maximum acceleration value

Table 3.1: Head Impact coding system to characterise impacts based on the event type and cause of acceleration.

Code		**Activity**
Event Type	1	Tackle (tackle as tackler)
	2	Carry (tackle as ball-carrier)
	3	Ruck
	4	Maul
	5	Lineout
	6	Scrum
Cause of Acceleration	1	Indirect (indirect impact to head)
	2	Soft (direct head impact to 'soft' body part*)
	3	Hard (direct head impact to 'hard' body part**)
	4	Ground (direct head impact to ground)
	5	Other

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

3.5 Assessment of Isometric Neck Strength

3.5.1 Testing Equipment

Isometric neck strength was measured using custom-built equipment, adapted from similar testing methods that have been shown to produce reliable results when assessing isometric neck strength and in rugby players (Salmon et al., 2015). A full specification is given in **Appendix A**, but briefly, the equipment was designed to place the participant in a simulated contact posture (**Figure 3.8**). Lying prone, with their torso supported, the participants were required to place their head in the centre of four adjustable 150 kg Tedea-Huntleigh load cells. Load cell placement was adjusted for each participant to ensure the correct head, neck, and spinal alignment. Participants were secured to the apparatus using a racing harness to limit the recruitment of accessory muscles and enhance measurement repeatability. Similarly, participants were instructed to keep their feet off the floor during trials to prevent them from pushing into the ground.

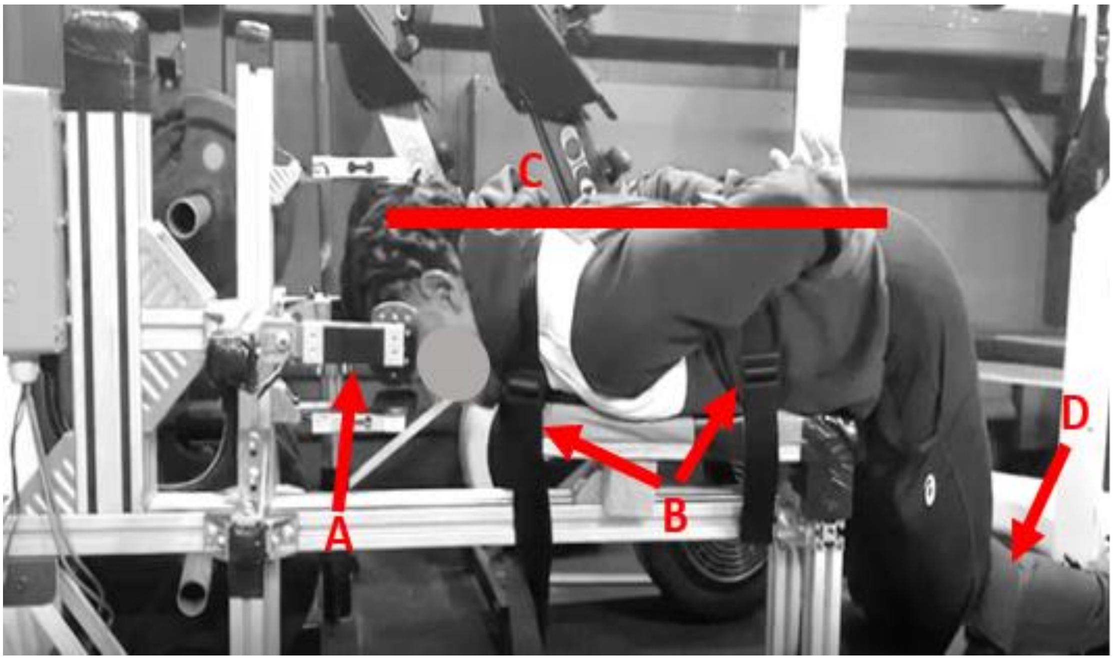

Figure 3.8: Image showing the custom-built neck strength testing equipment and participant set up and positioning during testing. Note, A) load cells. B) racing harness securing the upper body to the equipment. C) line showing the correct head, neck, and spinal alignment; and D) straps securing legs in position.

3.5.2 Warm-Up

Prior to conducting the strength tests, participants completed a standardised warm-up. Specifically, the warm-up consisted of five minutes of moderate-intensity activity on a cycle or rowing ergometer, followed by three sets of 10 shoulder shrugs, shoulder circumduction's, shoulder protraction and retraction and neck rolls. Subsequently, two deep neck stabilising (DNS) muscle pre-activation exercises, lying supine on a mat, tucking their chin, and lifting the head off the floor, and prone cervical retraction, were completed. Both exercises were performed for three sets of five-second holds. Upon completion of the warm-up, participants were positioned in the testing equipment with adjustments made to ensure correct posture and positioning of each participant.

3.5.3 Maximum Voluntary Contraction Trials

Participants were asked to complete three familiarisation trials, followed by three maximum voluntary isometric contraction (MVC) trials for each direction (flexion, extension and left- and right-lateral-flexion), in a randomised order. Participants were instructed to push isometrically into the relevant load cell at 50, 60 and 70% effort for three seconds for the familiarisation, and at maximal effort for the MVC trial. Participants were asked to employ a slight chin tuck during each trial to engage DNS muscles. Each MVC trial was repeated three times in each direction, with 20 seconds rest between individual trials and 30 seconds rest between directions. The maximum value across the three trials was taken to be the participant's MVC for the specific direction. Total neck strength was measured as the sum of MVC in each direction. Participants were provided with a consistent level of encouragement by the researcher. MVC testing was repeated following five and 17 weeks of neck-specific resistance training. All testing was scheduled at least 48 hours after or proceeding matches to limit the effect on performance.

3.6 Neck Strength Training Programme

The neck-specific training programme consisted of three stages, which were progressively introduced throughout the season (**Appendix B**). Resistance training was completed twice a week during the participants' regular, predetermined, strength and conditioning sessions.

3.6.1 Stage One: Deep Neck Stabiliser Training

Stage One focused on training and activating DNS muscles. Within this stage, participants were required to progress through three sub-stages adapted from Hanney and Kolber, (2007). Each sub-stage was as follows; i) Participants were required to lie supine on an adjustable weight bench inclined at 60°, tuck their chin and lift their head off the bench by 5-8 cm, for 10 seconds (**Figure 3.9**). This process was then repeated

10 times with 10 seconds rest between each repetition (rep). Following 10 consecutive reps, the incline on the bench was lowered by 10° and the process was repeated until they reached a 0° incline. ii) Participants performed the same chin-tuck and head-lift process as in part A whilst lying supine on a flat bench positioned at 0° (). Once they could perform 10 reps of 10 seconds they could progress to the next stage. iii) The final stage involved the participants performing a prone cervical retraction. Participants were required to lie prone on a weights bench, and retract their shoulder blades, whilst tucking their chin and simultaneously extending their lower cervical spine (**Figure 3.11**). This position was then held for 10 seconds. Once the participant could hold this position for 10 consecutive reps interspersed with 10 seconds rest, they moved on to the next stage of the programme.

3.6.2 Stage Two: Isometric

Stage Two introduced the performance of isometric holds in each direction (flexion, extension, and left- and right-lateral-flexion; **Figure 3.12**). Exercises were performed using elastic Therabands, either attached to an immovable frame or held by the researcher. Participants were required to perform three sets of 15 second holds in each direction at 60% of their MVC, as identified by their baseline test measurements, with 15 seconds rest between each set. An extra repetition was prescribed where significant imbalances were identified. Participants were instructed to perform each movement with a slight chin tuck to engage DNS muscles. The length of the holds increased by five seconds every two weeks up to 30 seconds. After 30 seconds, the resistance was increased by 5% and the length of the holds returned to 15 seconds.

3.6.3 Stage Three: Dynamic Movements

The final stage involved the use of controlled dynamic movements through eccentric and concentric contractions (**Figure 3.13**). Movements were performed at a resistance of 30% MVC for three sets of 10 reps in each direction at a tempo of 2:1:2. Once participants were able to perform three sets of 12 reps the resistance was increased by 5%. Resistance for these movements was provided by a custom-made head harness attached to a pulley system and a weights plate (**Figure 3.13**).

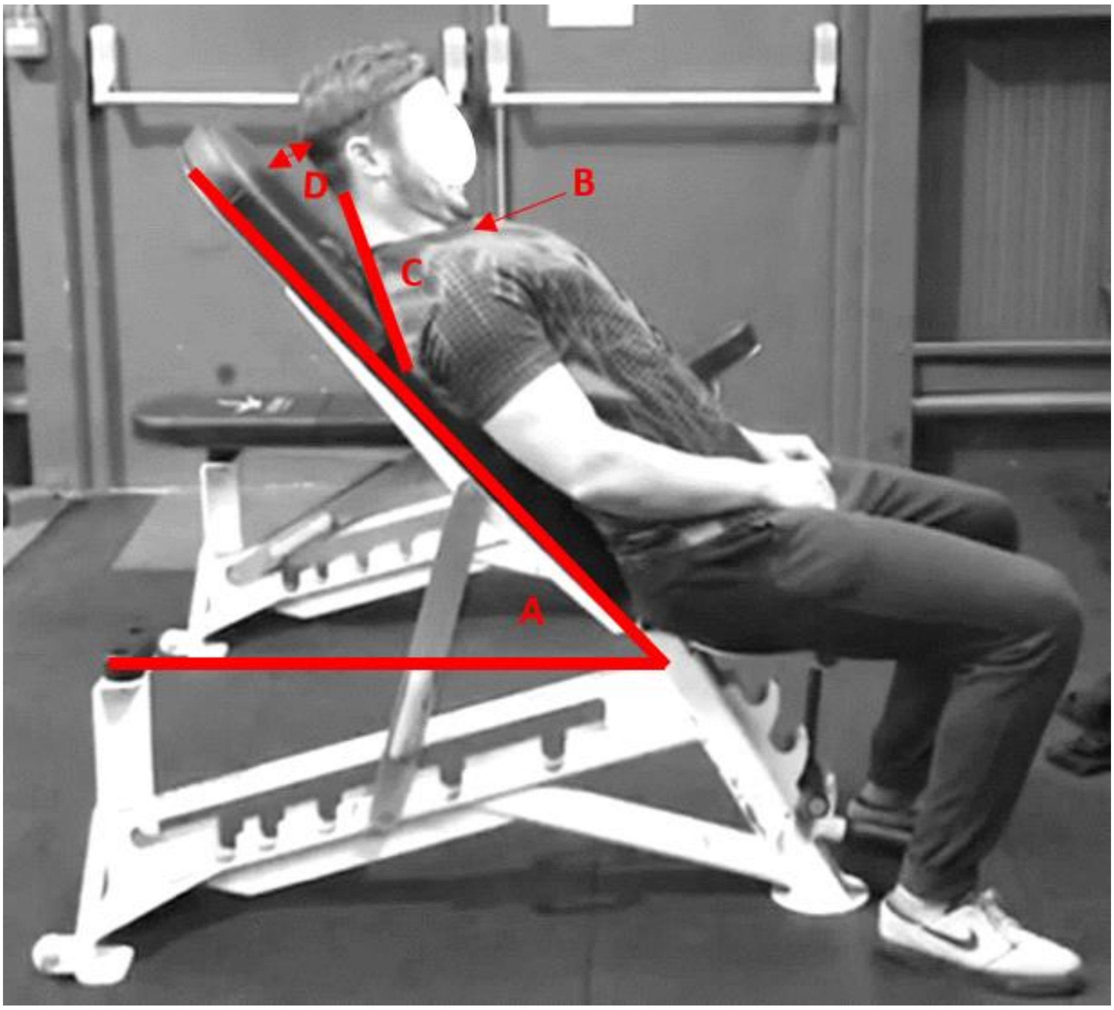

Figure 3.9: Stage One of the neck strength training programme, deep neck stabiliser exercise part i. Note, A) bench positioned at approximately 60°. B) engagement of DNS. C) correct head, neck, and spinal alignment; and D) head approximately 5-8cm off the bench.

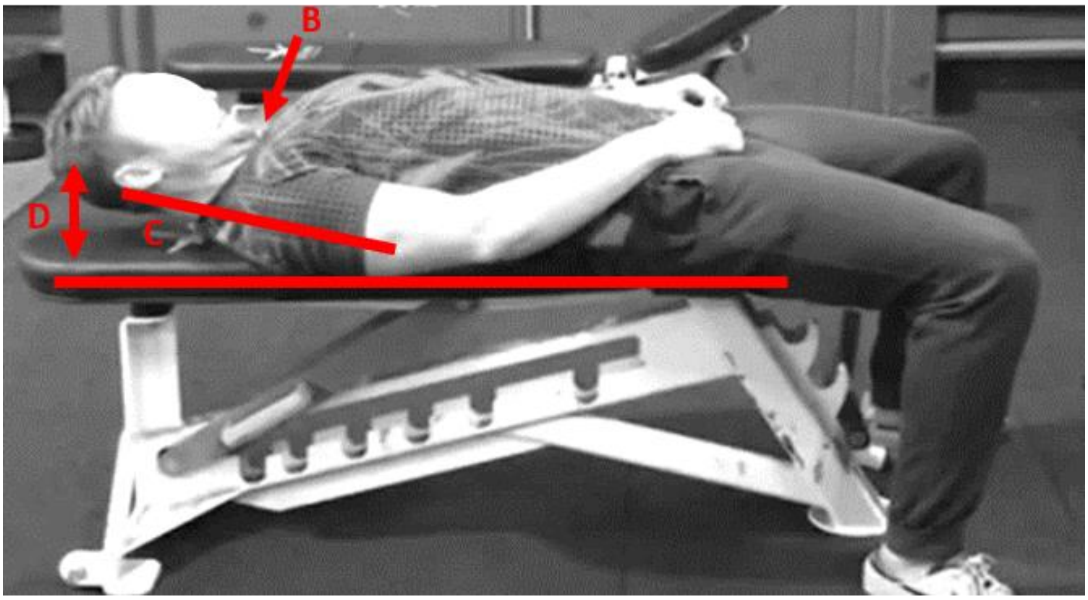

Figure 3.10: Stage One of the neck strength training programme, deep neck stabiliser exercise part ii. Note, A) Bench positioned at 0°. B) Engagement of the deep neck stabilising muscles. C) Correct head, neck, and spinal alignment. D) Head approximately 5-8cm off the bench.

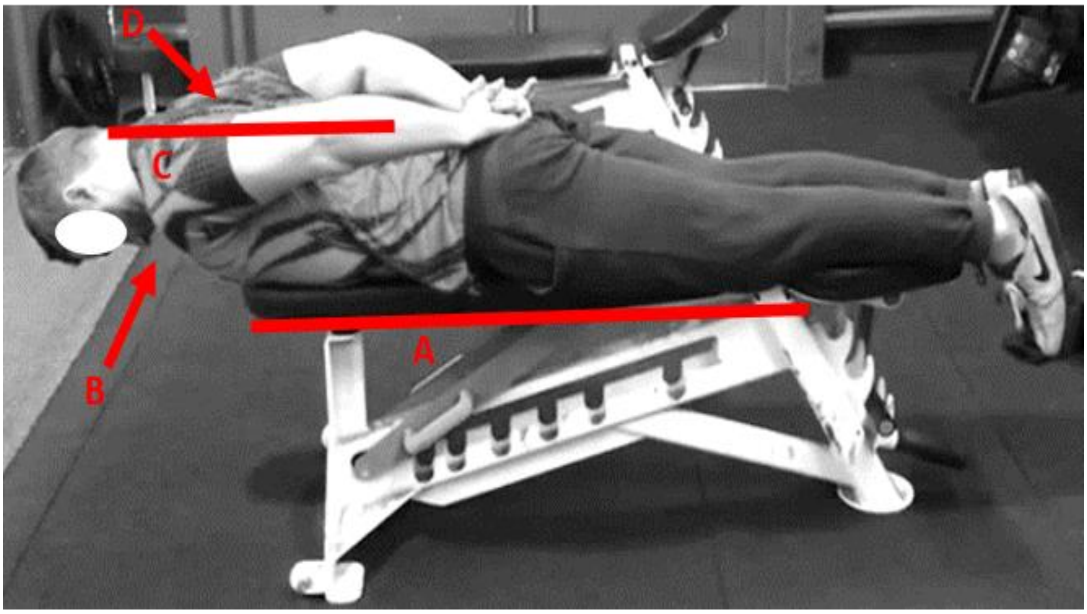

Figure 3.11: Stage One of the neck strength training programme, deep neck stabiliser exercise part iii. Note, A) bench positioned at 0°. B) engagement of the deep neck stabilising muscles. C) correct head, neck, and spinal alignment; and D) shoulder blades retracted.

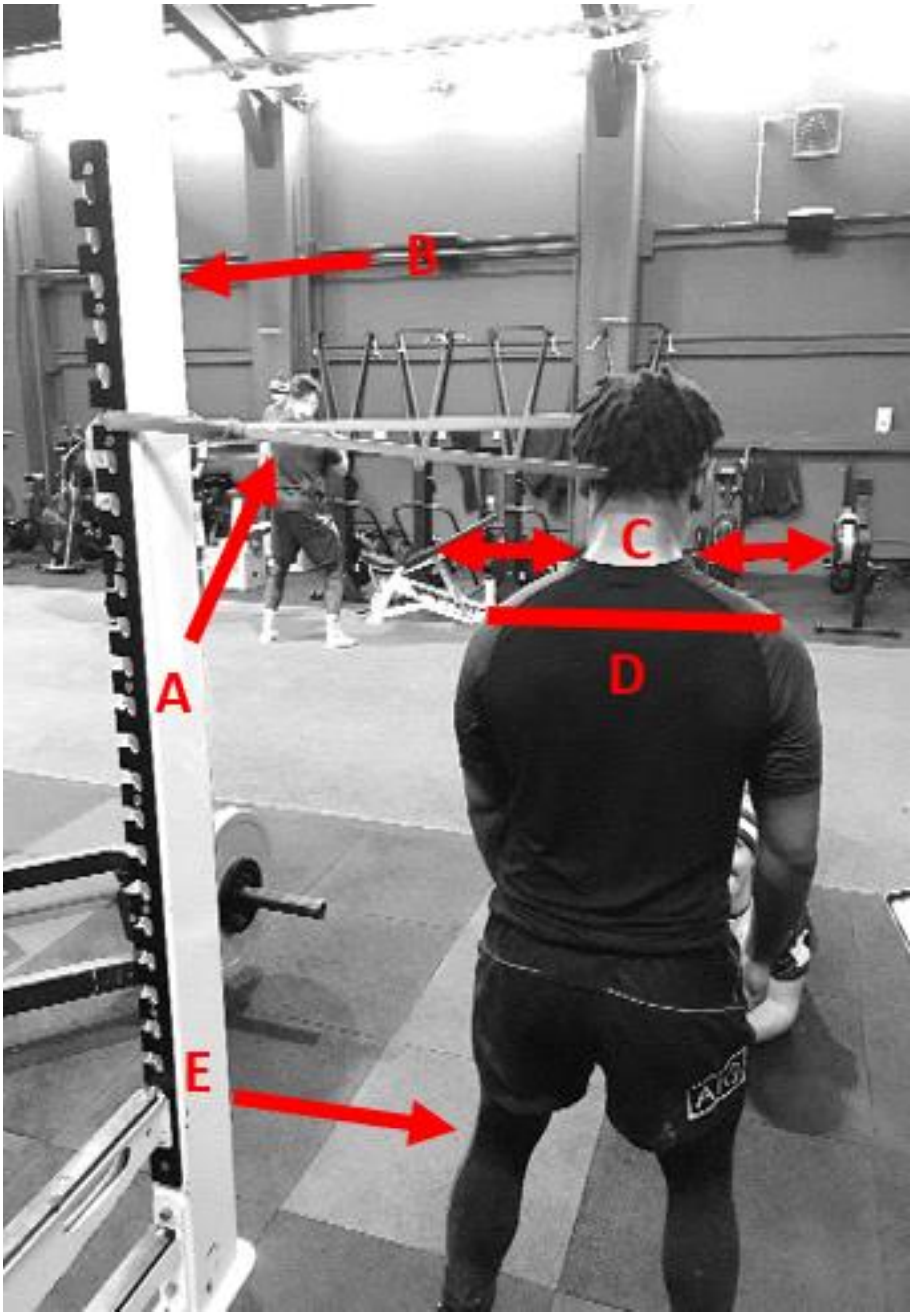

Figure 3.12: Stage Two of the neck strength training programme isometric holds using a theraband. Note, A) theraband under tension. B) fixed, immovable frame. C) neck in a neutral position. D) even shoulder alignment; and E) slightly flexed at the knees.

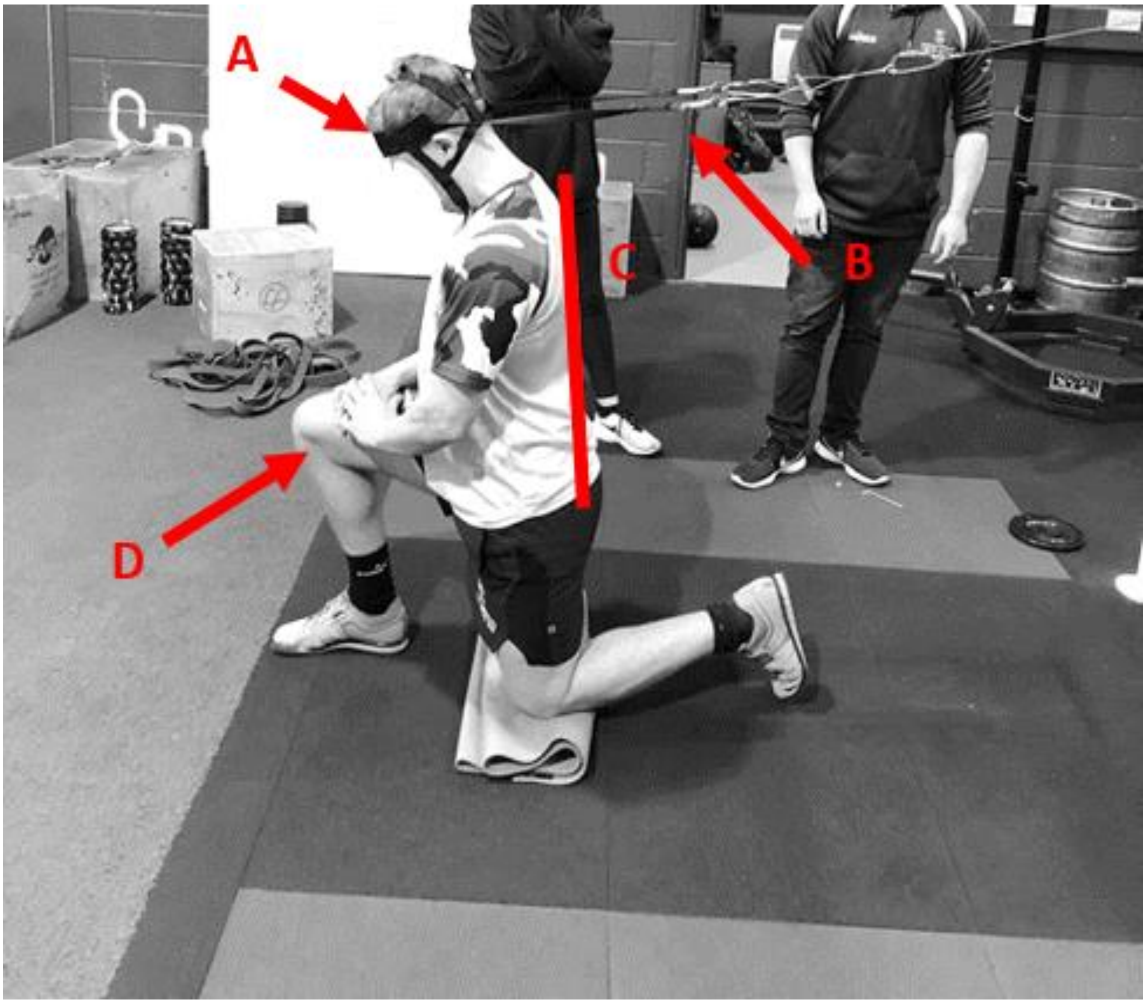

Figure 3.13: Stage Three of the neck strength training programme dynamic movements through concentric and eccentric contractions using a custom-built pulley system. Note, A) custom head harness adjusted to fit the individual. B) straps and wire connecting the harness to a pulley system. C) straight back; and D) one knee flexed in front for stabilisation.

3.7 Statistical Analysis

All analyses were completed using SPSS (IBM SPSS Statistics for Windows, Version 26.0. Armonk, NY: IBM Corp). All anthropometric, peak head acceleration and MVC data were visually assessed for normality using histograms, as well using quantitative assessments of skew and kurtosis. Similarly, a Shapiro-Wilks test was conducted to assess whether the data significantly differed from a normal distribution to ascertain whether to use parametric or non-parametric tests. All significance was set at $p < 0.05$.

3.7.1 Anthropometrics

Anthropometric variables were compared between broad positional groups using independent samples t-tests, and specific positional groups using one-way ANOVA's with Bonferroni *post-hoc* analysis.

3.7.2 Head Impacts

Differences between filtered and unfiltered data and between true and false positive impacts were assessed via Mann-Whitney's U. False positive impacts were defined as any impact recorded by the system that, following video and waveform analysis, was deemed not to have been caused by a head impact event. Pearson's correlations were also conducted to investigate the relationship between PLA and PRA in true and false positive impacts. Head impact magnitude data were analysed via one-way Kruskal-Wallis and *post-hoc* Mann-Whitney tests with a Bonferroni correction. These were conducted to assess significant differences in head impact magnitudes across event type, acceleration cause, position, and time in the game.

3.7.3 Neck Strength

Anthropometric Variables

Relationships between various anthropometric variables and baseline neck strength were explored using Pearson's correlation analysis. Differences in absolute neck strength and neck strength relative to body mass between positions were assessed via independent samples t-tests and one-way ANOVA's with Bonferroni *post-hoc* analysis.

Training

Differences in neck strength between baseline and post- five weeks of training were assessed via paired samples t-tests. Two-way mixed ANOVA's were completed to determine the effect of training adherence on changes in neck strength.

3.7.4 Head Acceleration and Neck Strength Variables

As acceleration values were obtained continuously and neck strength training was implemented out throughout the season, average directional and total neck strength were taken from baseline and mid-season scores. This was to account for any effect of training on head acceleration, as well as uncontrolled game variables that may have influenced head impact magnitude. Pearson's correlations were conducted between a range of head acceleration and neck strength variables to see if any relationships were present. Where significant correlations were found, simple, one model, regression analyses were completed with the neck strength variable as the independent variable and head acceleration as the dependant, to explore the relationship further.

Chapter 4 : Results

4.1 Anthropometrics

Anthropometric data were normally distributed *($p > 0.05$)*. Mean ± SD of all participants were as follows; age 20.3 ± 1.1 years, body mass 93.6 ± 13.3 kg, height 185.1 ± 9.5 cm, BMI 29.7 ± 2.7 kg/m^2, head circumference 58.1 ± 1.9 cm, neck circumference 41.2 ± 2.2 cm, neck-to-head circumference ratio 0.71 ± 0.03 cm and shoulder breadth 43.3 ± 2.4 cm.

4.1.1 Positions

A summary of the anthropometric differences between broad and specific positional groups is given in **Table 4.1**.

Forwards vs Backs

Forwards had a significantly higher body mass ($t_{(20)} = -5.8$, $p < 0.001$), BMI ($t_{(20)} = -2.8$, $p < 0.05$), neck circumference ($t_{(20)} = -3.9$, $p < 0.01$), neck-to-head circumference ratio ($t_{(20)} = -3.0$, $p < 0.01$) and were significantly taller than backs ($t_{(20)} = -3.9$, $p < 0.01$). No differences were observed between positions in head and shoulder breadth ($t_{(20)} = -2.0$, $p = 0.06$ and $t_{(20)} = -1.6$, $p = 0.12$, respectively).

Specific Positions

There was a significant between-group effect of specific positions for body mass ($F_{(5,21)} = 9.32$, $p < 0.001$). Front-row players were significantly heavier than half-backs ($p < 0.01$), second-row players were heavier than half-backs ($p < 0.001$) and outside-backs ($p < 0.05$) and back-row players were also significantly heavier than half-backs ($p < 0.05$). There was also a significant between-group effect for height ($F_{(5,21)} = 8.24$, $p < 0.001$). Based on *post hoc* analysis, the second-row was significantly taller than the front-row ($p < 0.01$), half-backs ($p < 0.01$), and outside-backs ($p < 0.001$).

There was also a significant between-group effect for BMI ($F_{(5,21)} = 5.66$, $p < 0.01$). *Post hoc* testing revealed that the front-row players had a significantly higher BMI than all positions except the back-row ($p = 1.0$). There was a significant between-group effect for neck circumference ($F_{(5,21)} = 4.7$, $p < 0.01$). Half-back players had a significantly smaller neck circumference than front-row and second-row players ($p < 0.05$). There was also a significant between-group effect of positions for neck-to-head circumference ratio ($F_{(5,21)} = 3.3, p < 0.05$). The front-row players had a significantly greater neck-to-head circumference ratio than the outside-backs ($p < 0.05$). There were no differences between individual positional groups for head circumference or shoulder breadth ($F_{(5,21)} = 2.6, p = 0.06$ and $F_{(5,21)} = 1.1, p = 0.39$, respectively).

Table 4.1: Anthropometric variables compared broadly, between forwards and backs, and between specific positional groups.

Position	Anthropometrics **Body mass** (kg)	**Height** (cm)	**BMI** (kg/m^2)	**Head circumference** (cm)	**Neck circumference** (cm)	**Neck-to-head circumference ratio** (cm)	**Shoulder breadth** (cm)
Forwards (n = 16)	105.7 ± 8.5	188.9 ± 9.2*	29.7 ± 2.7*	59.0 ± 1.6	42.9 ± 1.8*	0.73 ± 0.03*	44.3 ± 2.5
Backs (n = 15)	85.5 ± 8.2	179.2 ± 6.4	26.6 ± 2.3	57.5 ± 1.7	40.1 ± 1.7	0.70 ± 0.02	42.8 ± 2.1
Front-row (n = 5)	105.1 ± 6.1[a]	179.5 ± 7.8	32.5 ± 1.4[d]	58.4 ± 1.0	47.3 ± 2.1[a]	0.75 ± 0.03[e]	44.3 ± 1.8
Second-row (n = 6)	107.3 ± 9.5[b]	196.1 ± 5.6[c]	29.9 ± 1.1	59.2 ± 2.1	43.1 ± 1.4[a]	0.73 ± 0.03	44.0 ± 3.0
Back-row (n = 5)	103.4 ± 11.7[a]	186.8 ± 1.0	26.9 ± 3.3	59.2 ± 1.7	41.5 ± 1.9	0.70 ± 0.03	44.7 ± 3.1
Half-backs (n = 6)	80.4 ± 8.4	176.5 ± 3.1	25.8 ± 2.4	56.3 ± 1.6	39.3 ± 2.5	0.70 ± 0.03	41.6 ± 2.6
Inside-backs (n = 4)	92.4 ± 5.5	184.7 ± 5.5	27.0 ± 2.6	57.8± 1.2	40.8 ± 0.6	0.71 ± 0.02	43.4 ± 1.2
Outside-backs (n = 5)	87.8 ± 8.5	179.2 ± 8.5	27.4 ± 2.2	58.7± 1.4	40.7 ± 1.3	0.69 ± 0.02	44.0 ± 1.8

Note, *indicates a significant difference between forwards and backs. [a] significantly higher than half-backs. [b] significantly higher than half-backs and outside-backs. [c] significantly higher than front-row, half-backs, and outside-backs. [d] significantly higher than the second-row, half-backs, inside-backs and outside-backs. [e]significantly higher than outside-backs. All significance is given as $p < 0.05$.

4.2 Head Impact Kinematics

Throughout the season (13 games) 976 impacts were recorded using the PROTECHT™ system. Of these, 203 failed the first stage of verification, as they did not meet the video criteria and were classified as false positive impacts (**Figure 3.3**). A further 84 impacts were disregarded in the second verification stage, as they were deemed false positive impacts (**Figure 3.6**). Therefore, overall, there were 287 false positive impacts measured by the system. A further 544 met the video criteria but impacts lacked sufficient acceleration data to be fully verified (**Figure 3.7**). Hence, 144 impacts were classified as true positive, video verified impacts. These impacts were from seven different games and were recorded by 14 participants. The head acceleration data were not normally distributed *($p < 0.05$)*. Data were expressed as median, interquartile range (IQR) and maximum values.

4.2.1 Filtering

Unfiltered peak linear (PLA) and peak rotational (PRA) acceleration was significantly higher than filtered PLA and PRA ($U = 8113.0, p < 0.01$ and $U = 8757.0, p < 0.05$, respectively, **Table 4.2**).

Table 4.2: A comparison of the median, interquartile range (IQR) and maximum (Max) values for all unfiltered and filtered peak linear accelerations (PLA) and peak rotational accelerations (PRA).

	PLA (*g*)			**PRA (rad•s^{-2})**		
Filtering	**Median**	**(IQR)**	**Max**	**Median**	**(IQR)**	**Max**
Unfiltered	14**	(12 - 20)	57	943*	(742 – 1,337)	3,850
Filtered	13	(11 - 18)	50	849	(642 – 1,115)	2,973

Note, *indicates $p < 0.05$, ** $p < 0.01$

4.2.2 False Positive vs True Positive Impacts

False positive impacts had significantly higher PLA and PRA values compared to true positive impacts *($U = 17814.5, p < 0.05$,* **Figure 4.1** and $U = 12950.0, p < 0.001$, **Figure 4.2**, respectively*)*. This finding led to the investigation of the ratio between PLA and PRA (**Figure 4.3**). In true positive impacts, the average ratio of PLA to PRA was 1 g to 67 ± 30 rad•s^{-2}, with the two variables showing a significant moderate correlation ($r = 0.61, p < 0.001$). In contrast, the average ratio of PLA to PRA for false positive impacts was significantly higher (1 g to 126 ± 99 rad•s^{-2}, $U = 13555.0, p < 0.001$) with the two variables showing a significant but weak correlation ($r = 0.39, p < 0.001$).

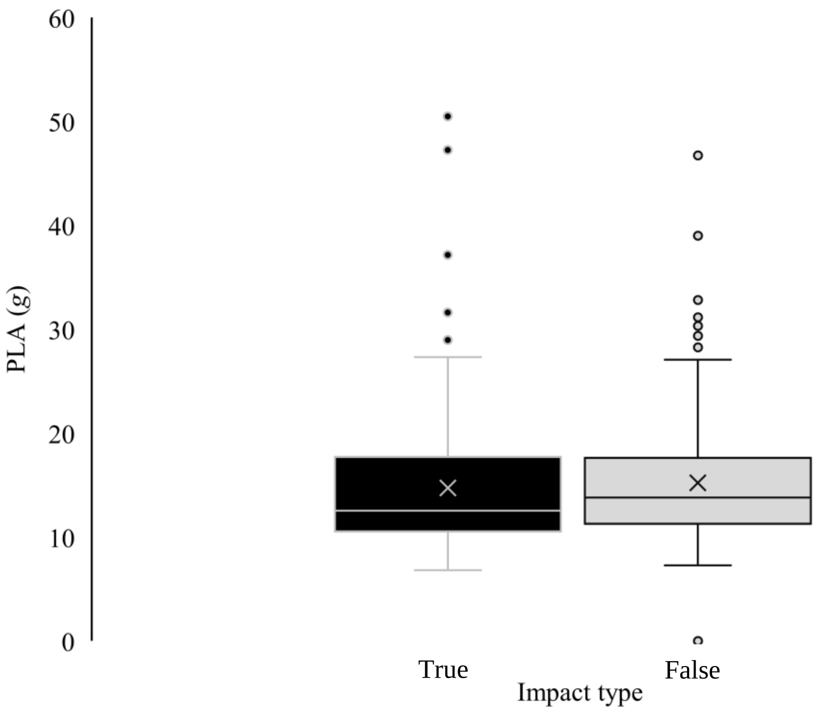

Figure 4.1: Comparing the distribution of peak linear acceleration (PLA) recorded in true (n = 144) and false (n = 287) positive impacts. Note, there was a significant difference between the two median values ($p < 0.05$). The centre, horizontal line indicates the median value, the X indicates the mean value, the box indicates the IQR, the whiskers indicate the minimum and maximum values, and the dots indicate outliers (1.5 x IQR).

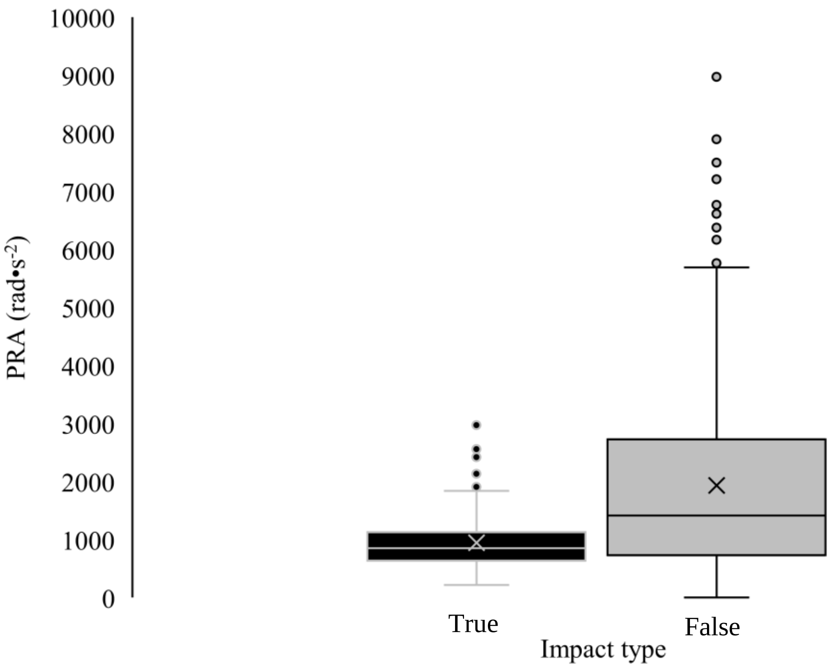

Figure 4.2: Comparing the distribution of peak rotational acceleration (PRA) recorded in true (n = 144) and false (n = 287) positive impacts. Note, there was a significant difference between the two median values ($p < 0.001$). the centre, horizontal line indicates the median value, the X indicates the mean value, the box indicates the IQR, the whiskers indicate the minimum and maximum values, and the dots indicate outliers (1.5 x IQR).

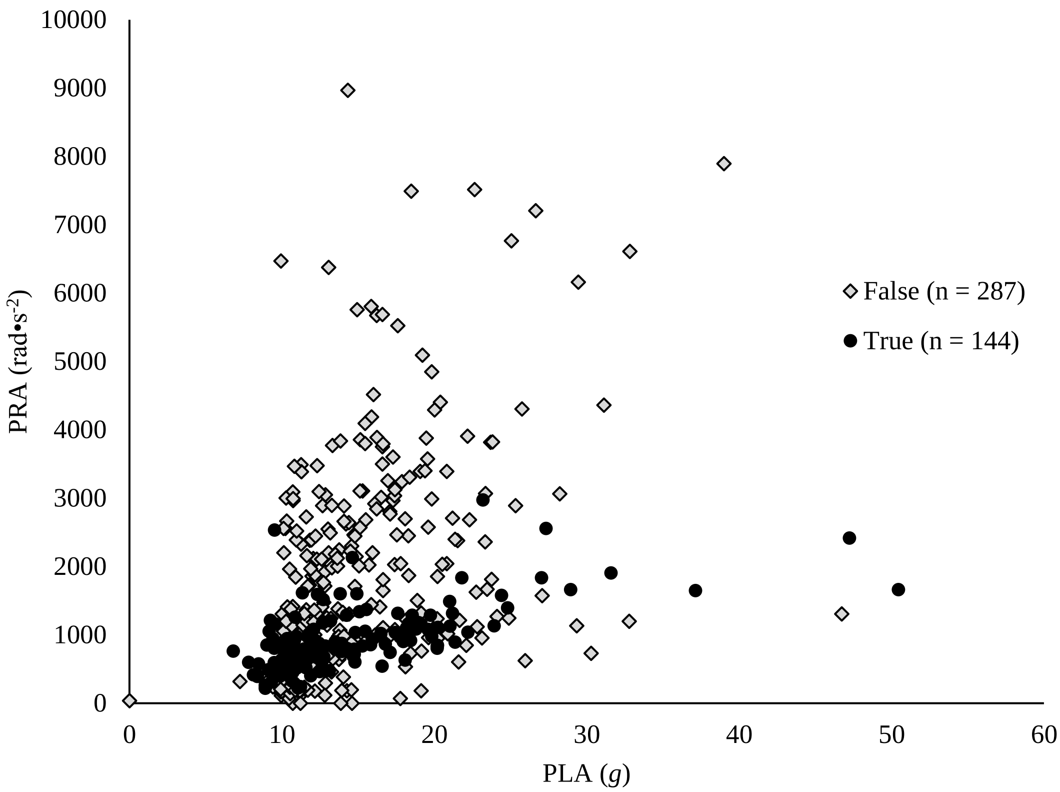

Figure 4.3: The relationship between peak linear (PLA) and peak rotational acceleration (PRA) in true and false positive impacts.

4.2.4 Head Impact Magnitude

Within this section, the data that is reported is representative of true positive, verified and filtered acceleration data. Across all games, median (IQR) PLA experienced per impact was 13 *g* (11 – 18 *g*) with a maximum recorded value of 50 *g*. Median (IQR) PRA experienced per impact was 849 rad•s^{-2} (642 – 1,115 rad•s^{-2}) with a maximum recorded value of 2,973 rad•s^{-2}.

Events

A summary of the median (IQR) and maximum values for PLA and PRA for each event can be seen in **Table 4.3.** There were no significant differences in PLA or PRA between event type *($H = 2.0$, $p = 0.58$* and *$H = 1.3$, $p = 0.75$,* respectively*).* No verified impacts were recorded during scrum or lineout events.

Table 4.3: Median (IQR) and maximum (Max) peak linear (PLA) and peak rotational accelerations (PRA) across each event type.

	PLA (*g*)			**PRA (**rad•s^{-2}**)**		
Event	**Median**	**(IQR)**	**Max**	**Median**	**(IQR)**	**Max**
Tackle (n = 57)	14	(11 - 18)	47	875	(678 - 1,174)	2,559
Carry (n = 49)	12	(10 - 18)	50	848	(848 - 1,181)	2,133
Ruck (n = 34)	12	(11 - 15)	23	819	(681 - 978)	2,973
Maul (n = 4)	14	(12 - 16)	20	874	(718 - 967)	1,032

Cause of Acceleration

A summary of the median (IQR) and maximum values for PLA and PRA for each cause of acceleration can be seen in **Table 4.4**. Cause of acceleration had no effect on resultant PLA *($H = 6.16$, $p = 0.11$).* Cause of acceleration had a significant effect on resultant PRA *($H = 11.36$, $p < 0.01$).* Direct impact to soft *($U = 605.0$)* and hard body parts *($U = 348.0$)* resulted in significantly higher PRA than indirect impacts *($p < 0.01$).*

Positions

A summary of the median (IQR) and maximum values for broad and specific positional groups can be seen in **Table 4.5.** There was no significant difference in the PLA experienced by backs and forwards *($U = 2026.0$, $p = 0.39$).* Similarly, specific position had no significant effect on PLA *($H = 5.88$, $p = 0.32$)***.** There was also no significant difference in PRA experienced by backs and forwards *($U = 2115.0$,*

$p = 0.64$). However, specific position had a significant effect on PRA ($H = 16.3$, $p < 0.01$). The front-row players experienced significantly higher PRA than outside-backs ($U = 20.0$, $p < 0.01$) and the second-row ($U = 205.0$, $p < 0.01$). Inside-backs also experienced significantly higher PRA than outside-backs ($U = 25.0$, $p < 0.01$).

Table 4.4: Median (IQR) and maximum (Max) peak linear (PLA) and peak rotational accelerations (PRA) for different causes of acceleration.

	PLA (*g*)			**PRA (rad•s^{-2})**		
Cause	**Median**	**(IQR)**	**Max**	**Median**	**(IQR)**	**Max**
Indirect (n = 45)	12	(10 - 15)	50	737	(543 - 943)	2,133
Hard (n = 42)	15	(11 - 18)	47	900[a]	(705 – 1,255)	2,973
Soft (n = 28)	14	(11 - 18)	24	975[a]	(763 – 1,244)	2,533
Ground (n = 14)	13	(12 - 18)	27	873	(527 – 1,062)	1,837

Note, [*] indicates significantly greater than indirect ($p < 0.01$).

Table 4.5: Summary of median (IQR) and maximum (Max) peak linear (PLA) and peak rotational accelerations (PRA) experienced by forwards and backs, as well as by specific positional groups.

	PLA (*g*)			**PRA (rad•s^{-2})**		
Position	**Median**	**(IQR)**	**Max**	**Median**	**(IQR)**	**Max**
Forwards (n = 10)	12	(11 - 17)	50	852	(655 – 1,083)	2,973
Backs (n = 5)	14	(10 - 18)	32	848	(643 – 1,214)	2,559
Front-row (n = 2)	15	(11 - 17)	50	946[ab]	(776 – 1,366)	2,973
Second-row (n = 4)	11	(10 - 15)	25	682	(520 – 1,008)	1,834
Back-row (n = 4)	13	(11 - 17)	47	856	(577 – 1,086)	2,418
Half-backs (n = 2)	14	(10 - 18)	21	858	(787 – 1,214)	1,372
Inside-backs (n = 3)	15	(12 - 18)	32	875[a]	(716 – 1,286)	2,559
Outside-backs (n = 1)	12	(11 - 13)	21	474	(322 - 667)	1,129

Note: [a] indicates significantly higher than outside-backs. [b] indicates significantly higher than the second-row. All significance is given as ($p < 0.01$).

4.3 Neck Strength

Baseline maximum voluntary isometric contraction (MVC) represented a sample of 27 players. Due to injury and participant availability, post-training testing could not be completed for five participants, thus, data from these individuals were excluded from post-training analysis. The neck strength data were normally distributed *($p < 0.05$).*

4.3.1 Correlations

A summary of the Pearson's correlation analysis of anthropometric variables and MVC at baseline is given in **Table 4.6**. There were significant correlations between BMI and MVC, between body mass and MVC, and between neck-to-head circumference ratio and MVC in all directions and total MVC at baseline ($r = 0.38\text{-}0.70$, $p < 0.05$). There were also significant moderate positive correlations between neck circumference and MVC at baseline in flexion, left-lateral-flexion and total MVC *($r = 0.44\text{-}0.52$, $p < 0.05$).*

4.3.2 Positional Groups

Positional differences in absolute and relative MVC were assessed broadly as forwards and backs (**Table 4.7**) and specifically as front-row, second-row, back-row, half-backs, inside-backs and outside-backs (**Table 4.8** and **Table 4.9**).

Backs vs Forwards

Forwards had greater absolute baseline MVC than backs, with significant differences in flexion ($t_{(20)} = -2.1$, $p < 0.05$), left-lateral-flexion ($t_{(20)} = -2.9$, $p < 0.01$), right-lateral-flexion ($t_{(20)} = -2.1$, $p < 0.05$) and total ($t_{(19)} = -3.1$, $p < 0.01$). There were no significant differences between forwards and backs in extension ($t_{(20} = -0.8$, $p = 0.46$).

Due to significant differences in baseline body mass between position groups (**Table 4.1**), differences in MVC relative to body mass were also assessed. There were no significant differences in relative MVC between backs and forwards in any direction or total at baseline (extension, $t_{(19)} = 1.9$, $p = 0.07$; flexion, $t_{(19)}=1.7$, $p = 0.26$; left-lateral-flexion, $t_{(19)}=-0.8$, $p = 0.46$; right-lateral-flexion, $t_{(19)}=-0.3$, $p = 0.79$; total MVC, $t_{(19)} = -0.6$, $p = 0.55$).

Imbalances.

There was no significant differences between backs (36 ± 27 N) and forwards (59 ± 49 N) in flexion and extension imbalance ($t_{(20)} =-1.7$, $p = 0.09$). Similarly, there was no significant differences between backs (17 ± 17 N) and forwards (32 ± 20 N) in the imbalance between left-and right-lateral-flexion. ($t_{(20)} = -1.5$, $p = 0.16$).

Individual Positional Groups

Specific position had no significant effect on extension or right-lateral-flexion ($F_{(5,21)} = 1.1$, $p = 0.37$ and $F_{(5)} = 0.9$, $p = 0.49$, respectively). There was a significant effect of specific position on absolute flexion ($F_{(5,21)} = 4.8$, $p < 0.01$), left-lateral-flexion ($F_{(5,21)} = 3.4$, $p < 0.05$) and total MVC ($F_{(5,21)} = 3.5$, $p < 0.05$). Front-row players had significantly higher MVC than half-backs ($p < 0.01$) and inside-backs ($p < 0.05$) in flexion. Front-row players also had higher absolute MVC than inside-backs in left-lateral-flexion ($p < 0.05$) and total neck strength ($p < 0.05$).

Specific position had a significant effect on relative extension and left-lateral-flexion ($F_{(5,21)} = 2.9$, and $F_{(5,21)} = 3.2$, $p < 0.05$, respectively). However, *post-hoc* analyses were unable to detect significant differences in either direction between individual positions. There was no significant effect of specific position on relative MVC in flexion and right-lateral-flexion ($F_{(5,21)} = 2.5$, $p = 0.07$ and $F_{(5,21)}=0.8$, $p = 0.55$). There was a significant effect of specific position on relative total MVC ($F_{(5,21)}=3.9$, $p < 0.05$). *Post-hoc* analysis however was unable to detect significant differences between specific positions.

Imbalances.

Specific position had no effect on flexion and extension imbalance ($F_{(5,21)} = 1.9$, $p = 0.13$) or left-lateral-flexion and right-lateral-flexion imbalance ($F_{(5,21)} = 1.4$, $p = 0.54$) (**Table 4.10**).

Table 4.6: Pearson correlation coefficients and associated p-values of anthropometric variables to baseline maximum isometric voluntary contraction (MVC) in extension (Ext), flexion (Flx), left- (Lflx) and right- (Rflx) lateral-flexion and total MVC .

		Baseline MVC (N)				
Characteristic		**Ext**	**Flx**	**Lflx**	**Rflx**	**Total**
Age	*r*	-0.17	-0.17	-0.15	-0.16	-0.20
(years)	*p*	0.41	0.41	0.45	0.43	0.31
Height	*r*	0.15	-0.11	0.11	0.15	0.12
(cm)	*p*	0.46	0.96	0.60	0.45	0.55
Body mass	*r*	0.38*	0.49*	0.44*	0.44*	0.55**
(kg)	*p*	0.04	0.01	0.02	0.02	<0.01
BMI	*r*	0.38*	0.70**	0.49*	0.44*	0.64**
(kg/m²)	*p*	0.04	< 0.01	0.01	0.02	<0.01
Head circumference	*r*	0.10	0.17	0.07	0.03	0.12
(cm)	*p*	0.61	0.40	0.67	0.89	0.54
Neck circumference	*r*	0.36	0.49*	0.44*	0.33	0.52**
(cm)	*p*	0.07	0.01	0.02	0.09	<0.01
Neck-to-head	*r*	0.38*	0.51**	0.49**	0.40*	0.56**
circumference ratio (cm)	*p*	0.04	<0.01	<0.01	0.03	<0.01
Shoulder breadth	*r*	0.29	0.10	-0.09	-0.02	0.11
(cm)	*p*	0.14	0.62	0.97	0.92	0.59

Note, * indicates $p < 0.05$, ** $p < 0.01$

Table 4.7: Absolute and relative baseline maximum voluntary isometric contraction (MVC) in extension (Ext), flexion (Flx), left- (Lflx) and right- (Rflx) lateral-flexion and total MVC in forwards and backs.

	Absolute MVC (N)		**Relative (N/kg)**	
Direction	**Backs** (n = 13)	**Forwards** (n = 14)	**Backs** (n = 13)	**Forwards** (n = 14)
Ext	240 ± 53	257 ± 51	2.8 ± 0.6	2.4 ± 0.5
Flx	251 ± 44	297 ± 60*	2.9 ± 0.5	2.7 ± 0.4
Lflx	175 ± 33	239 ± 69**	2.1 ± 0.5	2.3 ± 0.6
Rflx	177 ± 30	225 ± 74*	2.1 ± 0.4	2.2 ± 0.7
Total	843 ± 111	1018 ± 209**	9.9 ± 1.4	9.5 ± 1.9

Note, * indicates $p < 0.05$, ** $p < 0.01$

Table 4.8: Absolute baseline maximum voluntary isometric contraction (MVC) in extension (Ext), flexion (Flx), left- (Lflx) and right- (Rflx) lateral-flexion and total MVC in specific positional groups.

	Absolute MVC (N)					
Direction	**Front-row** (n = 4)	**Second-row** (n = 6)	**Back-row** (n = 3)	**Half-back** (n = 5)	**Inside-back** (n = 4)	**Outside-back** (n = 5)
Ext	289 ± 34	243 ± 50	252 ± 50	240 ± 35	203 ± 57	260 ± 66
Flx	355 ± 66[a]	283 ± 44	262 ± 32	231 ± 25	227 ± 35	273 ± 47
Lflx	269 ± 65[b]	234 ± 58	168 ± 29	199 ± 36	151 ± 31	188 ± 34
Rflx	242 ± 80	210 ± 67	203 ± 34	189 ± 30	173 ± 30	180 ± 42
Total	1154 ± 172[b]	970 ± 180	885 ± 139	860 ± 79	754 ± 30	902 ± 148

Note, [a] indicates significantly higher than half-backs ($p < 0.01$) and inside-backs ($p < 0.05$). [b] indicates significantly higher than inside-backs ($p < 0.05$).

Table 4.9: Baseline maximum voluntary isometric contraction (MVC) relative to body mass in extension (Ext), flexion (Flx), left- (Lflx) and right- (Rflx) lateral-flexion and total MVC in specific positional groups.

	Relative MVC (N/kg)					
Direction	**Front-row** (n = 4)	**Second-row** (n = 6)	**Back-row** (n = 3)	**Half-back** (n = 5)	**Inside-back** (n = 4)	**Outside-back** (n = 5)
Ext	2.8 ± 0.3	2.3 ± 0.4	2.4 ± 0.2	3.0 ± 0.4	2.2 ± 0.6	2.9 ± 0.5
Flx	3.4 ± 0.8	2.6 ± 0.4	2.5 ± 0.1	2.9 ± 0.3	2.5 ± 0.5	3.1 ± 0.4
Lflx	2.6 ± 0.6	2.2 ± 0.5	1.6 ± 0.1	2.5 ± 0.5	2.3 ± 0.5	2.1 ± 0.4
Rflx	2.3 ± 0.7	2.0 ± 0.6	2.0 ± 0.2	2.4 ± 0.4	1.9 ± 0.3	2.0 ± 0.4
Total	11.0 ± 1.6	9.0 ± 1.5	8.5 ± 0.4	10.8 ± 1.1	8.2 ± 0.7	10.2 ± 0.9

Table 4.10: Absolute Imbalances between flexion (Flx) and extension (Ext) and left- (Lflx) and right- (Rflx) lateral-flexion in specific positional groups.

	Absolute Imbalance (N)					
Direction	**Front-row** (n = 4)	**Second-row** (n = 6)	**Back-row** (n = 3)	**Half-back** (n = 5)	**Inside-back** (n = 4)	**Outside-back** (n = 5)
Flx vs Ext	87 ± 58	62 ± 43	15 ± 12	28 ± 26	40 ± 21	44 ± 33
Lflx vs Rflx	27 ± 27	32 ± 22	35 ± 11	13 ± 13	22 ± 9	19 ± 25

4.4 Head Acceleration and Neck Strength Variables

Due to injuries and iMG malfunctions, head acceleration and neck strength correlations could only be completed for 13 participants. Both the head acceleration and neck strength data in this sample were normally distributed *($p > 0.05$)*. Where significant correlations were observed, regression analyses were conducted to evaluate the effectiveness of the variable as a predictor of head acceleration.

4.4.1 Head Acceleration and Neck Strength

Overall

A summary of Pearson's correlation results between head acceleration and average directional and total MVC can be seen in **Table 4.11**. There was a significant moderate negative correlation between extension and PRA. Variance in extension explained 40% of the variance in PRA ($R^2 = 0.40$, $F = 7.41$, $p < 0.05$, **Figure 4.4**). There was also a significant moderate negative correlation between total and PRA, with variance in total explaining 37% of the variance in PRA *($R^2 = 0.37$, $F = 6.48$, $p < 0.05$,* **Figure 4.5).**

Event Type

Correlation analyses were also carried out to see if any relationships were present between neck strength variables and average PLA and PRA experienced in the three main event types (tackle, carry and ruck). A detailed summary of the results can be seen in **Appendix C**.

There were significant, moderate, negative correlations between extension and PRA experienced during the carry ($r = -0.61$, $p < 0.05$) and PRA experienced during the ruck ($r = -0.64$, $p < 0.05$). Total MVC had a significant, moderate, negative correlation with PRA experienced during the tackle *($r = -0.58$, $p < 0.05$)*. Variance in extension explained 37% of the variance in PRA experienced during a carry *($R^2 = 0.37$, $F = 5.21$, $p < 0.05$,* **Figure 4.8**) and 41% of the variance of PRA experienced during a ruck *($R^2 = 0.41$, $F = 6.89$, $p < 0.05$,* **Figure 4.6***)*. Total neck strength explained 33% of the variance of PRA experienced during a tackle *($R^2 = 0.33$, $F = 5.47$, $p < 0.05$,* **Figure 4.7***)*. There were no significant correlations between PLA and any of the measures of neck strength.

Cause of Acceleration

A detailed summary of Pearson's correlation results between head acceleration, due to different causes of acceleration, and average directional and total MVC can been seen in **Appendix D**.

There was a significant, strong, negative correlation between extension and PRA experienced because of direct head contact to hard body parts ($r = -0.69$, $p < 0.05$). There was also a significant, moderate, negative correlation between total MVC and PRA experienced as result of direct contact to hard body parts ($r = - 0.58$, $p < 0.05$). Variance in extension ($R^2 = 0.48$, $F = 9.21$, $p < 0.05$,**Figure 4.9**) and total ($R^2 = 0.33$, $F = 4.97$, $p < 0.05$,**Figure 4.10**) MVC explained 48% and 34% of the variance in PRA experienced as a result of direct head contact to hard body parts.

4.4.2 Head Acceleration and Anthropometrics

Neck Circumference

There was no significant correlation between neck circumference and average PLA and PRA sustained across the season ($r = -0.25$, $p = 0.42$ and $r = -0.37$, $p = 0.20$ respectively).

Neck-to-Head Circumference Ratio

There was no significant, correlation between neck-to-head circumference ratio and average PLA and PRA sustained across the season ($r = -0.14$, $p = 0.66$ and $r = -0.22$, $p = 0.48$ respectively).

Table 4.11: Pearson's correlation coefficients and associated p-values for relationships between average maximum voluntary isometric contraction (MVC) in extension (Ext), flexion (Flx) and left-(Lflx) and right-lateral-flexion (Rflx) and peak linear (PLA) and rotational acceleration (PRA)

		MVC (N)				
		Ext	**Flx**	**Lflx**	**Rflx**	**Total**
PLA	*r*	-0.54	-0.12	-0.28	-0.36	-0.46
	p	0.06	0.70	0.36	0.22	0.12
PRA	*r*	-0.64*	-0.27	-0.39	-0.47	-0.61*
	p	0.02	0.37	0.19	0.11	0.03

Note, * indicates $p < 0.05$.

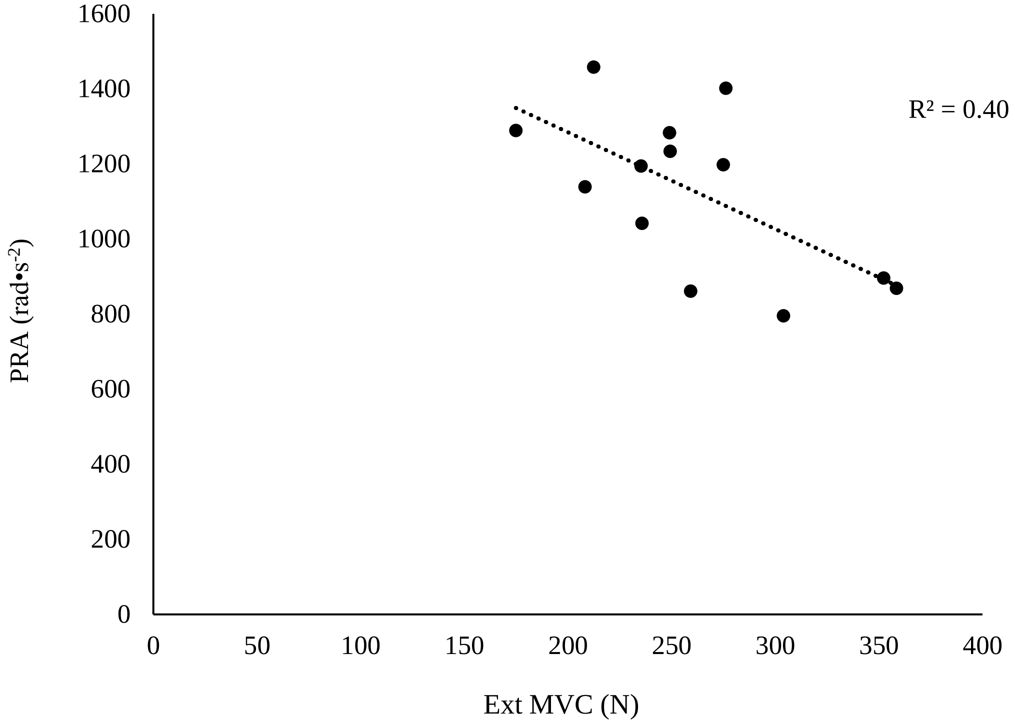

Figure 4.4: The relationship between maximum voluntary isometric contraction (MVC) in extension (Ext) and average peak rotational acceleration (PRA) sustained across the season.

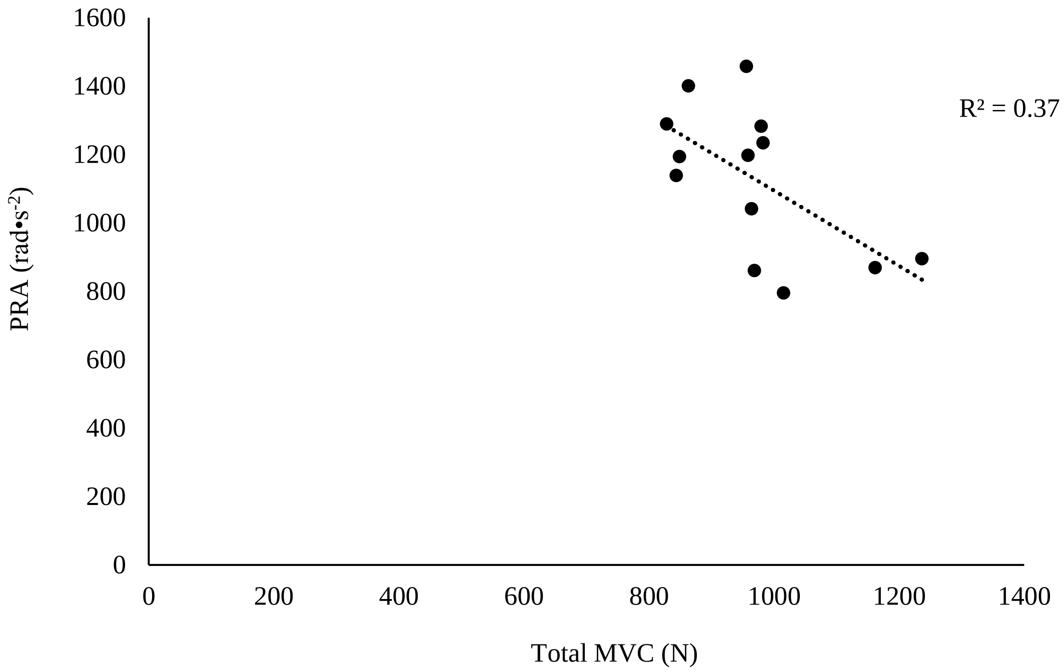

Figure 4.5: The relationship between total maximum voluntary isometric contraction (MVC) and average peak rotational acceleration (PRA) sustained across the season.

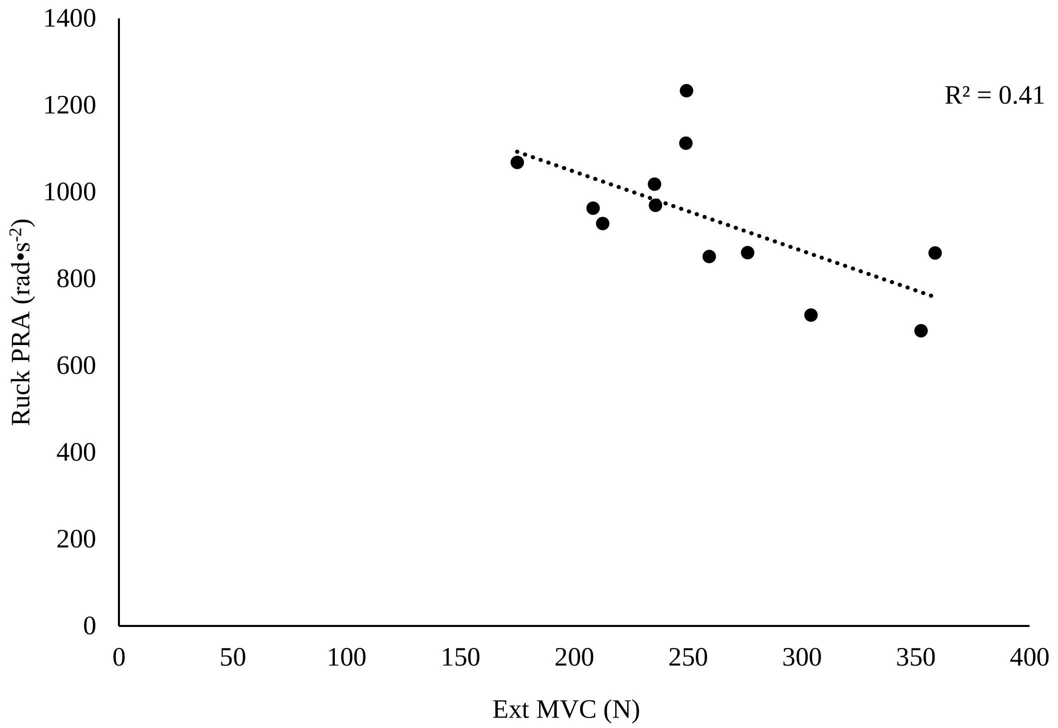

Figure 4.6: The relationship between maximum voluntary isometric contraction (MVC) in extension (Ext) and average peak rotational acceleration (PRA) sustained during ruck events across the season.

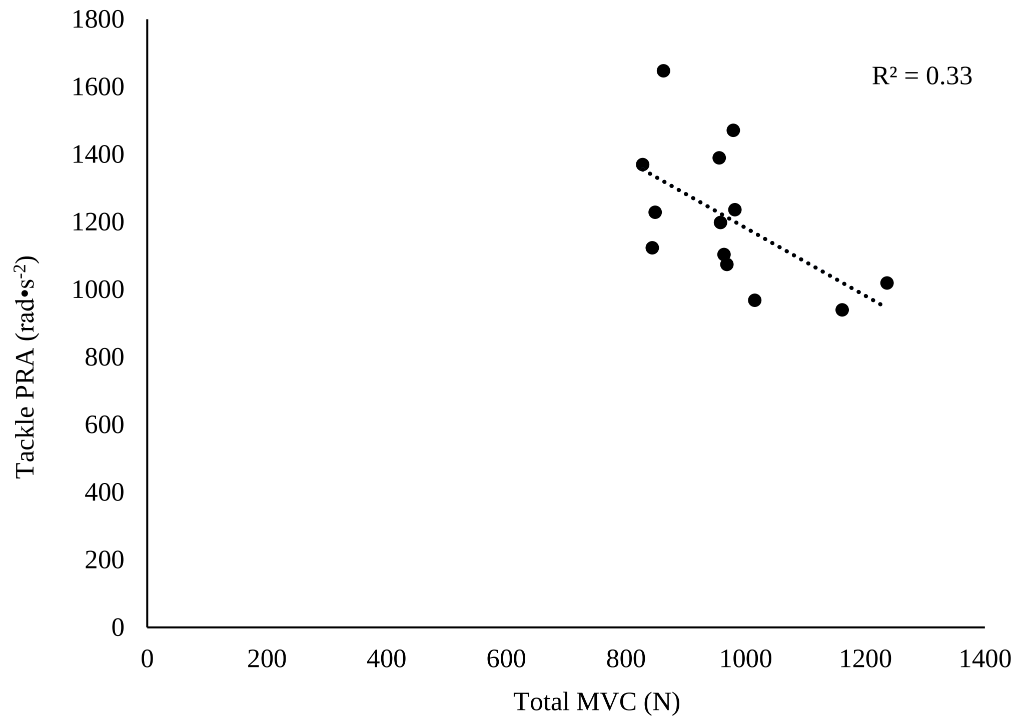

Figure 4.7: The relationship between total maximum voluntary isometric contraction (MVC) and average peak rotational acceleration (PRA) sustained during tackle events across the season.

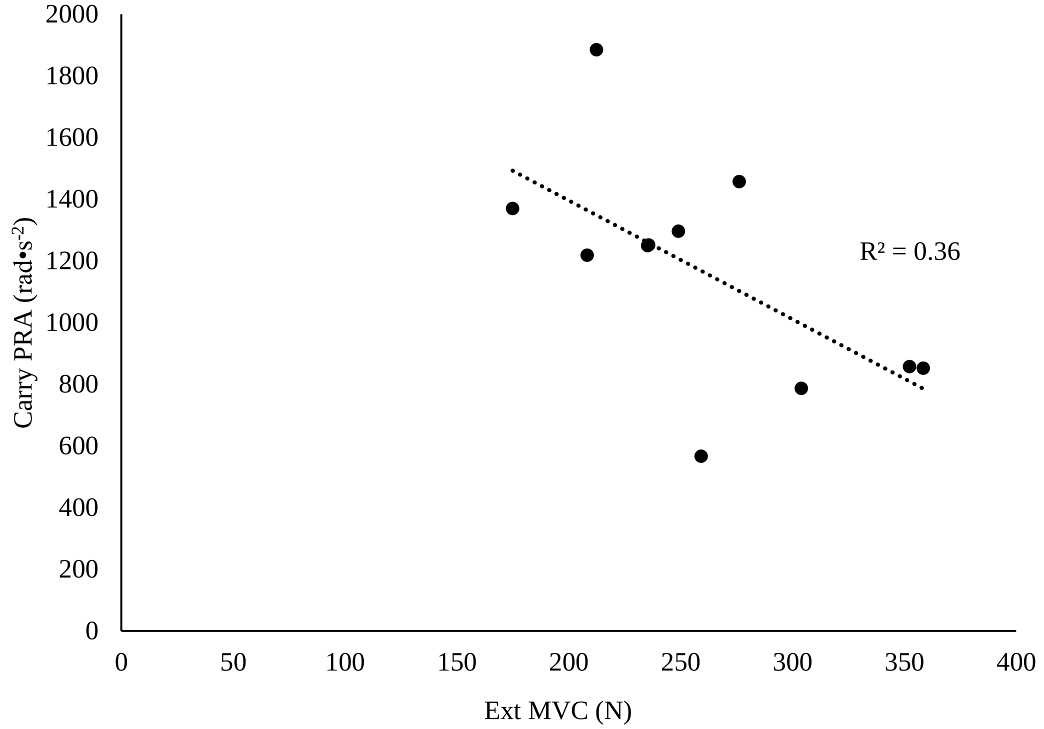

Figure 4.8: The relationship between average maximum voluntary isometric contraction (MVC) in extension (Ext) and average peak rotational acceleration (PRA) experienced during carry events across the season.

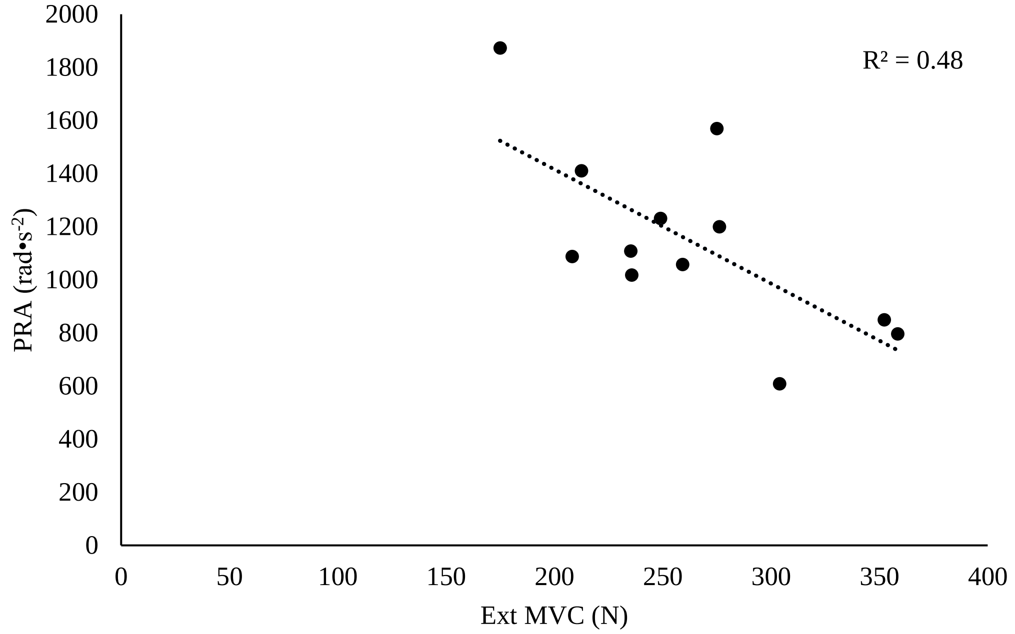

Figure 4.9: The relationship between average maximum voluntary isometric contraction (MVC) in extension (Ext) and average peak rotational acceleration (PRA) experienced, as a result of direct head contact to hard body parts, across the season.

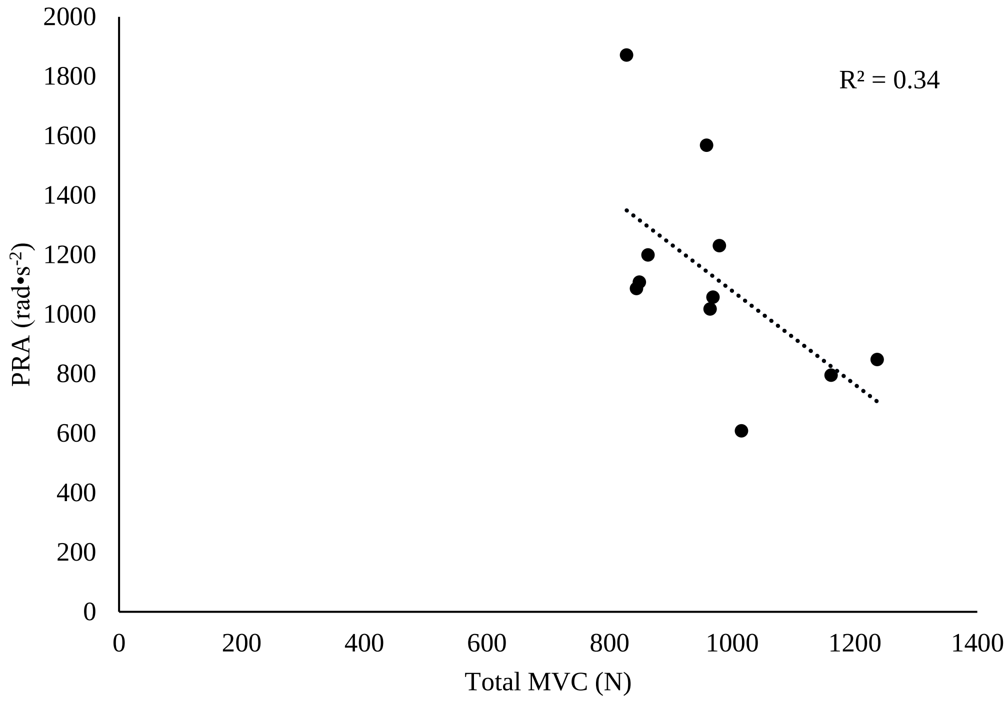

Figure 4.10: The relationship between total maximum voluntary isometric contraction (MVC) and average peak rotational acceleration (PRA) experienced, as a result of direct head contact to hard body parts, across the season.

4.5 Training

MVC significantly increased from baseline, in all directions except extension, following five weeks of neck specific resistance training (flexion, $t_{(22)} = -4.3, p < 0.001;$ left-lateral-flexion, $t_{(22)} = -3.6, p < 0.01;$ right-lateral-flexion, $t_{(22)} = -3.6, p < 0.01$, **Figure 4.11**). Extension showed trends towards increases but these were non-significant $(t_{(22)} = -1.8, p = 0.08)$. Total MVC also significantly increased from baseline (920 ± 175 vs 1030 ± 176 N, $t_{(22)} = -4.7, p < 0.001$). Absolute imbalance between flexion and extension, and between left-lateral-flexion and right-lateral-flexion did not change following training ($t_{(22)} = -1.4, p = 0.18$ and $t_{(22)} = -0.5$, p = 0.59, respectively. **Figure 4.12**).

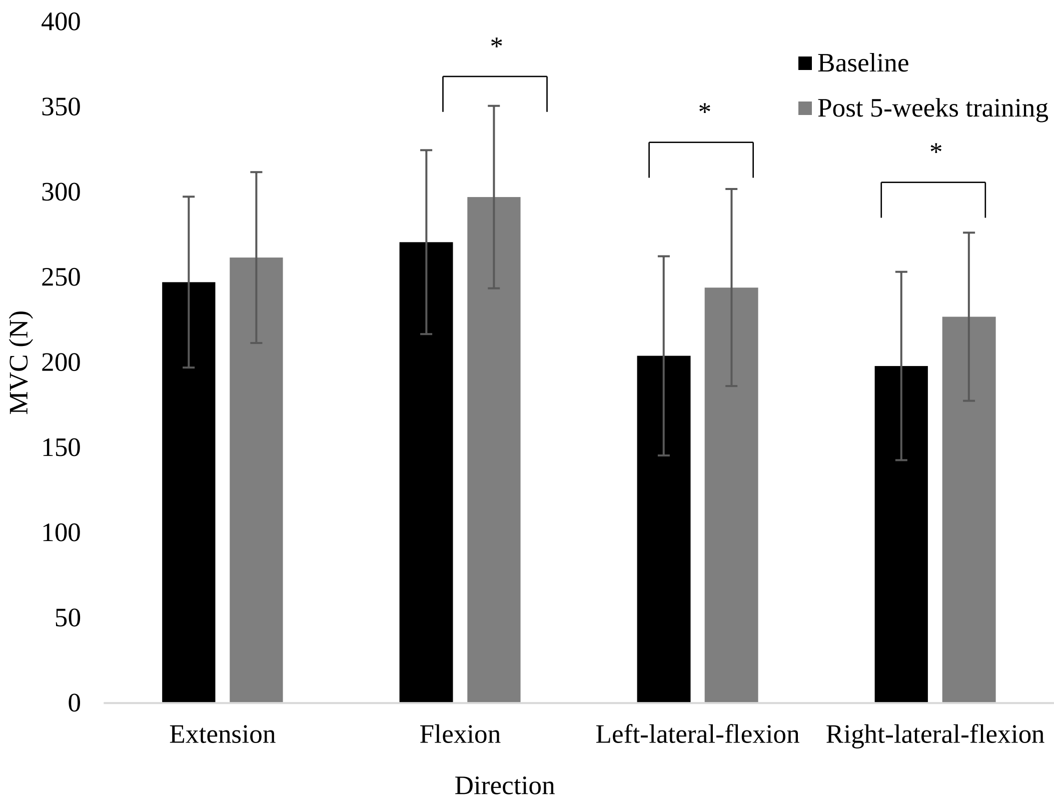

Figure 4.11: Maximum isometric voluntary contraction (MVC) in flexion (Flx), extension (Ext), and left- (Lflx) and right- (Rflx) lateral-flexion at baseline and following five weeks of training. Note, *indicates $p < 0.05$.

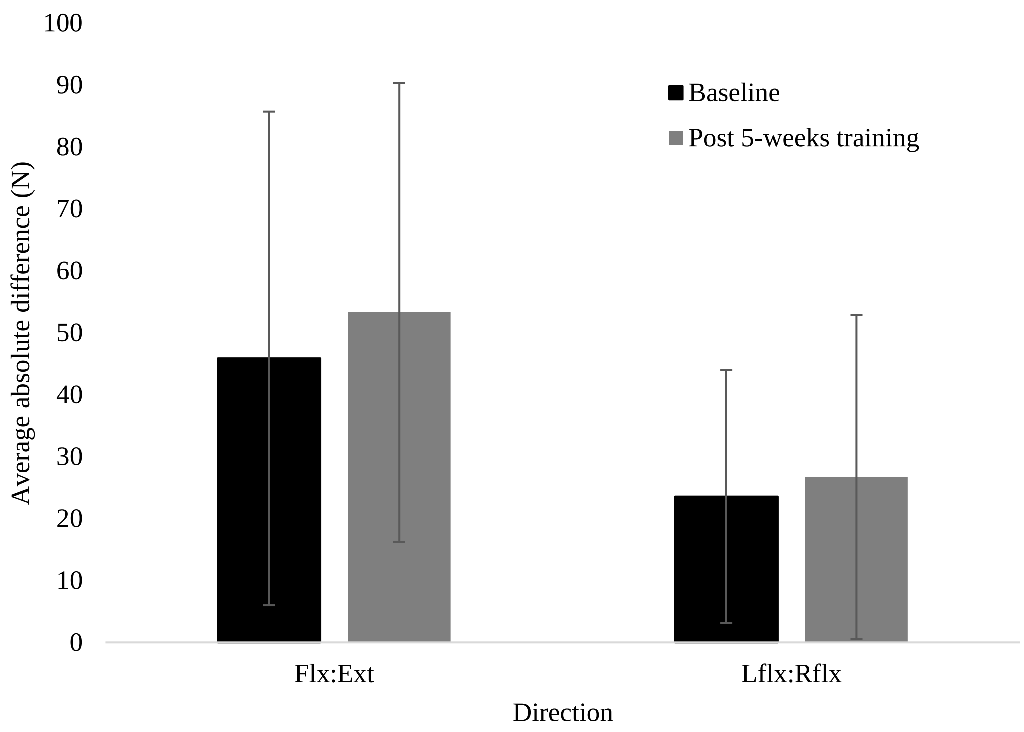

Figure 4.12: Absolute difference between maximum isometric voluntary contraction in flexion (Flx) and extension (Ext) and left- (Lflx) and right- (Rflx) lateral-flexion at baseline and post-training.

Adherence

Due to individual differences in training attendance, changes in MVC and imbalances were also assessed in relation to training adherence. Mean training adherence was 25.5 ± 18.4%, minimum adherence was 0% and maximum adherence was 70%. Therefore, participants were split into two groups based on the mean attendance, high adherence (> 25.5%, n = 11) and low adherence (< 25.5%, n = 11).

Extension

Neither high nor low adherence groups changed from baseline following five weeks of neck strength training ($t_{(10)} = -2.18, p = 0.55$ and $t_{(10)} = -0.26, p = 0.80$, respectively, **Figure 4.13**). There was no main effect of time ($F_{(1,20)} = 3.45, p = 0.08$) or group ($F_{(1,20)} = 0.12, p = 0.73$). The mean improvement seen for high adherence (27.3 ± 41.7 N) was greater than for low adherence (2.6 ± 33.6 N). However, this was not statistically significant, as indicated by the interaction ($F_{(1,20)} = 2.35, p = 0.14$).

Flexion

For flexion there was only a main effect for time ($F_{(1,20)} = 21.56, p < 0.001$). The main effect for group ($F_{(1,20)} = 0.755, p = 0.39$) and interaction ($F_{(1,20)} = 2.59, p = 0.12$), were not statistically significant. When collapsed across groups, there was an improvement from pre- to post- five weeks of training for flexion. The paired-samples t-test with respect to training adherence revealed a statistically significant improvement for high adherence ($t_{(10)} = -4.63, p < 0.01$) whilst the low adherence flexion strength remained unchanged ($t_{(10)} = -2.06, p = 0.07$) (**Figure 4.14**).

Right-Lateral-Flexion

The paired-samples t-test with respect to training adherence revealed a statistically significant improvement for high adherence ($t_{(10)} = -4.19, p < 0.01$), whilst the low adherence remained unchanged ($t_{(10)} = -1.52, p = 0.16$) (**Figure 4.15**). There was a main effect of time ($F_{(1,20)} = 10.83, p < 0.01$), with no main effect of group ($F_{(1,20)} = 0.61$ $p = 0.45$) or interaction ($F_{(1,20)} = 0.84, p = 0.37$).

Left-Lateral-Flexion

For left-lateral-flexion, there was a main effect of time ($F_{(1,20)} = 15.10, p < 0.01$), and no main effect of group ($F_{(1,20)} = 0.21, p = 0.66$) or interaction ($F_{(1,20)} = 0.47, p = 0.50$). Like flexion, the paired-samples t-test revealed a statistically significant improvement in left-lateral-flexion for the high adherence group *($t_{(10)}$= -4.19, p < 0.01),* with no improvement for low adherence ($t_{(10)}$ = -1.91, p = 0.09) (**Figure 4.16).**

Total MVC

Paired samples t-test revealed that total MVC significantly increased in both high and low adherence groups, from baseline, following five weeks of training ($t_{(10)}$= -4.77, $p < 0.01$ and $t_{(10)}$= -2.25, $p < 0.05$, **Figure 4.17**). There was a main effect of time ($F_{(1,20)} = 24.05$, $p < 0.001$), however, there was no main group effect ($F_{(1,20)} = 0.28$, $p = 0.28$) or interaction ($F_{(1,20)} = 2.58$, $p = 0.12$).

Absolute Imbalances

For imbalances observed between flexion and extension (**Table 4.12**) there was no significant effect of time ($F_{(1,20)}= 1.61$, $p = 0.22$), group ($F_{(1,20)}= 0.52$, $p = 0.48$) or interaction ($F_{(1,20)} = 0.18$, $p = 0.89$). Similarly, for imbalances observed in left-lateral-flexion vs right-lateral-flexion (**Table 4.12**), there was no significant effect of time ($F_{(1,20)} = 1.03$, $p = 0.32$), group ($F_{(1,20)} = 0.35$, $p = 0.56$) or interaction ($F_{(1,20)} = 0.58$, $p = 0.45$).

Table 4.12: Absolute differences between flexion (Flx) and extension (Ext) and between left- (Lflx) and right- (Rflx) lateral-flexion in low (n =11) and high (n =11) adherence groups.

	Low adherence absolute difference (N)		High adherence absolute difference (N)	
Direction	**Baseline**	**Post 5-weeks training**	**Baseline**	**Post 5-weeks training**
Flx vs Ext	40 ± 42	48 ± 39	52 ± 40	59 ± 38
Lflx vs Rflx	17 ± 19	27 ± 27	27 ± 20	28 ± 27

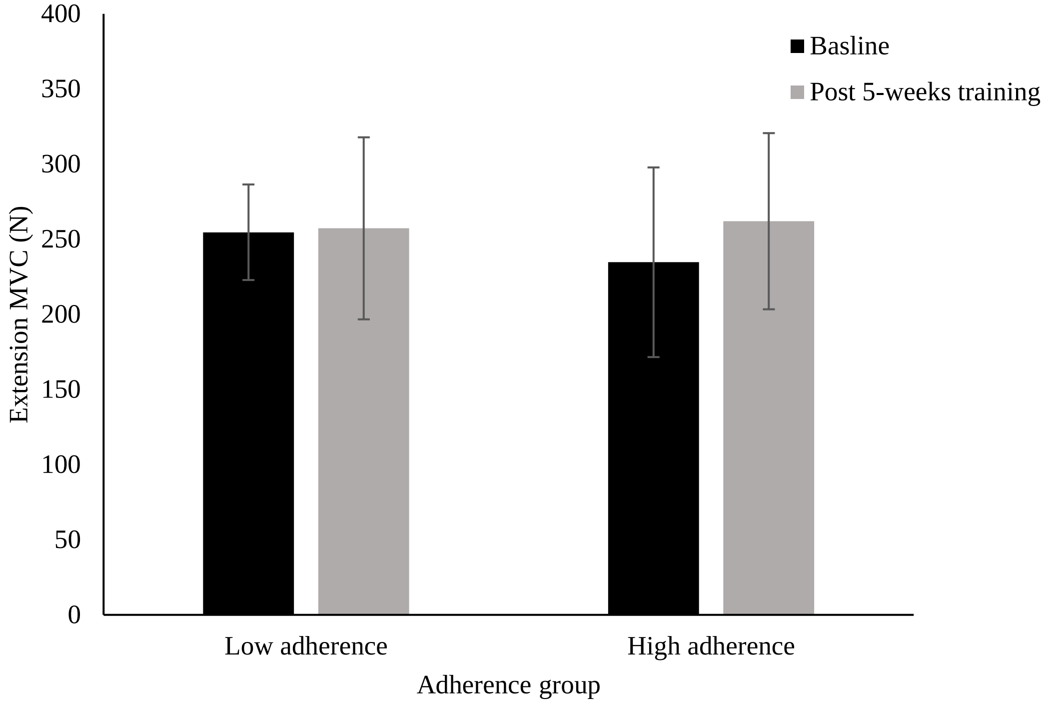

Figure 4.13: Maximum voluntary isometric contraction (MVC) in extension (Ext) at baseline and post five-weeks of training in low (n=11) and high (n=11) adherence groups.

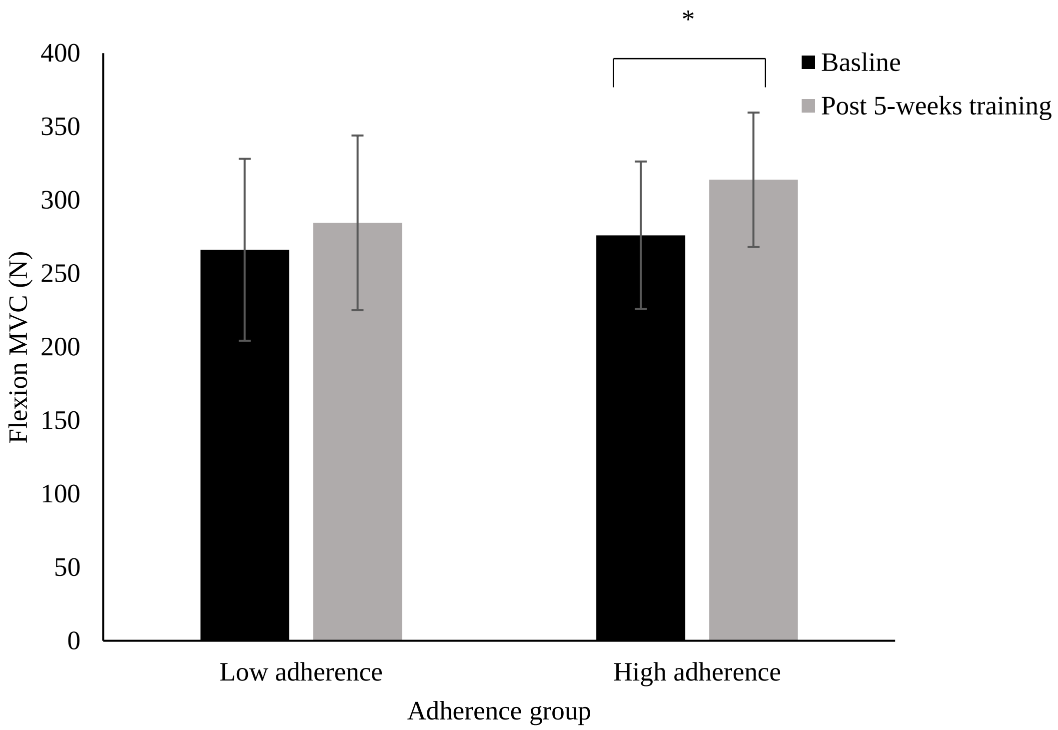

Figure 4.14: Maximum voluntary isometric contraction (MVC) in flexion (Flx) at baseline and post five-weeks of training in low (n=11) and high (n =11) adherence groups. Note, * indicates a significant difference from baseline to post five weeks of training (*p <0.05)*.

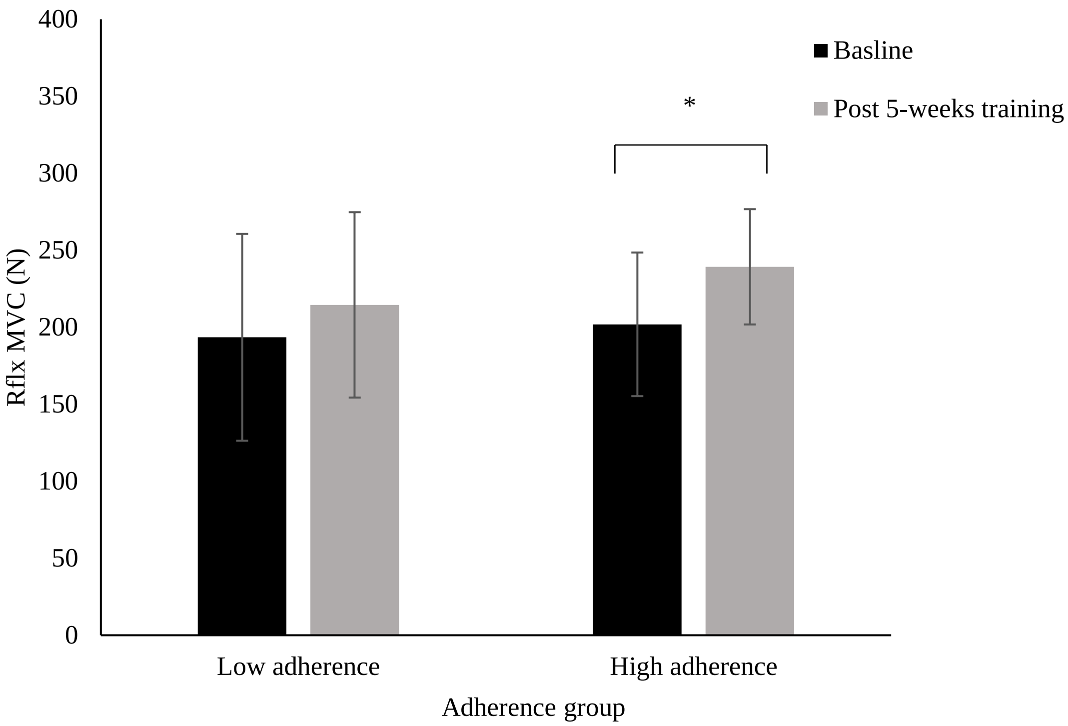

Figure 4.15: Maximum voluntary isometric contraction (MVC) for right-lateral-flexion (Rflx) at baseline and post five-weeks of training in low (n=11) and high (n=11) adherence groups. Note, * indicates a significant difference from baseline to post five weeks of training (*$p < 0.05$).*

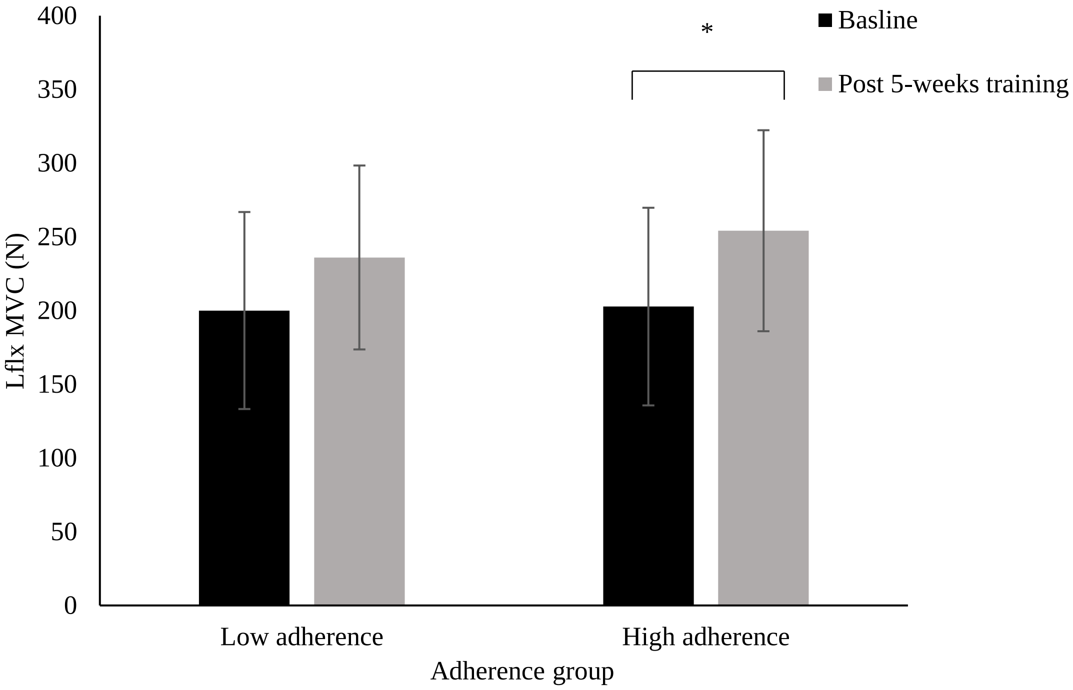

Figure 4.16: Maximum voluntary isometric contraction (MVC) in left-lateral-flexion (Lflx) at baseline and post five-weeks of training in low (n =11) and high (n = 11) adherence groups. Note, * indicates a significant difference from baseline to post five weeks of training (*$p < 0.05$).*

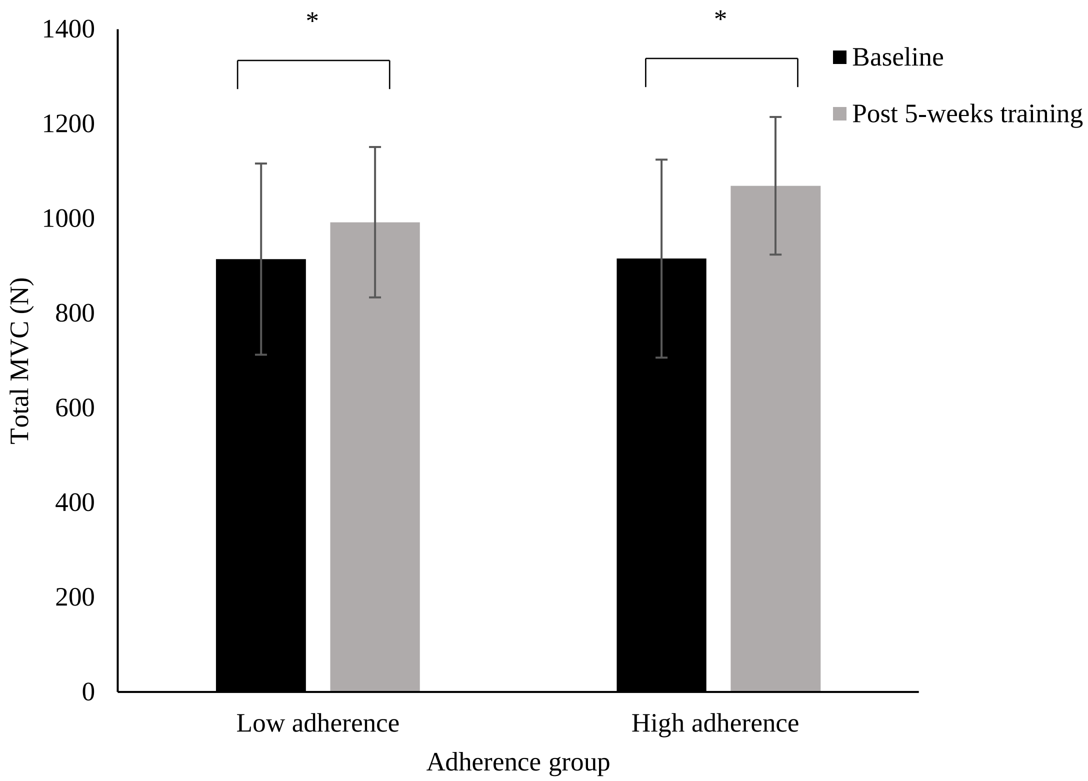

Figure 4.17: Total maximum voluntary isometric contraction (MVC) at baseline and post five-week of training in low (n =11) and high (n = 11) adherence groups. Note: * indicates significant difference from baseline to post five weeks of training (*$p < 0.05$*).

4.5.1 Case Studies

Due to the complications of COVID-19, the final testing protocol was only completed by three participants following the full 17-week training programme. The participants consisted of a front-row forward (SUM035, age, 20 years; height, 178.9 cm; body mass, 108.8 kg), a second row forward (SUM040, age, 20 years; height, 189.3 cm; body mass, 97.3 kg) and a half-back (SUM033, age, 19 years; height, 178.6 cm; body mass, 77.3 kg). SUM035 and SUM040 both completed 20% of the total sessions over the 17-week programme, whilst SUM033 completed 50%.

A summary of total MVC for each participant, at each time point, is given in **Figure 18.** Similarly, MVC in each direction, at each time point, for SUM040, SUM033 and SUM035 is given in **Figure 19**, **Figure 4.20** and **Figure 4.21**, respectively. Additionally, the percentage change from baseline in each direction following five and 17 weeks of training is given in **Table 4.13**, and absolute imbalances at each time point are given in **Table 4.14.**

Table 4.13: Percentage change in maximum voluntary isometric contraction (MVC) in extension (Ext), flexion (Flx), left- (Lflx) and right-lateral-flexion (Rflx) and total MVC from baseline following five and 17 weeks of training.

	Percentage change from baseline (%)							
	Ext		**Flx**		**Lflx**		**Rflx**	
Participant	**5 weeks**	**17 weeks**	**5 weeks**	**17 weeks**	**5 weeks**	**17 weeks**	**5 weeks**	**17 weeks**
SUM040	5.0	22.4	0.5	0.8	15.8	21.6	10.6	19.1
SUM035	14.9	24.1	10.4	16.4	-20.6	-3.9	-18.2	-8.6
SUM033	0.9	31.2	22.8	19.2	-6.8	11.5	13.3	26.5

Table 4.14: Absolute differences between flexion (Flx) and extension (Ext) and left- (Lflx) and right-lateral-flexion (Rflx) at baseline and following five and 17 weeks of training.

	Absolute difference (N)					
	Flx vs Ext			**Lflx vs Rflx**		
Participant	**Baseline**	**5 weeks**	**17 weeks**	**Baseline**	**5 weeks**	**17 weeks**
SUM040	77	68	19	4	20	14
SUM035	36	21	4	0	6	15
SUM033	13	42	61	13	30	30

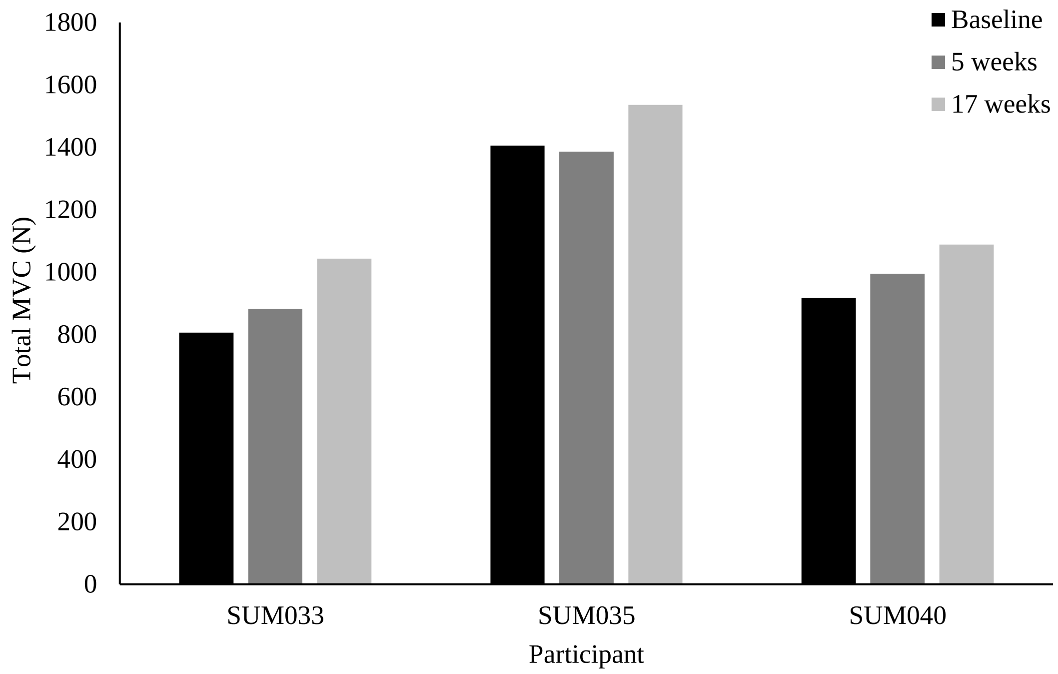

Figure 4.18: Total maximum voluntary isometric contraction (MVC) at baseline, post five and 17 weeks of training in SUM040, SUM033, and SUM035.

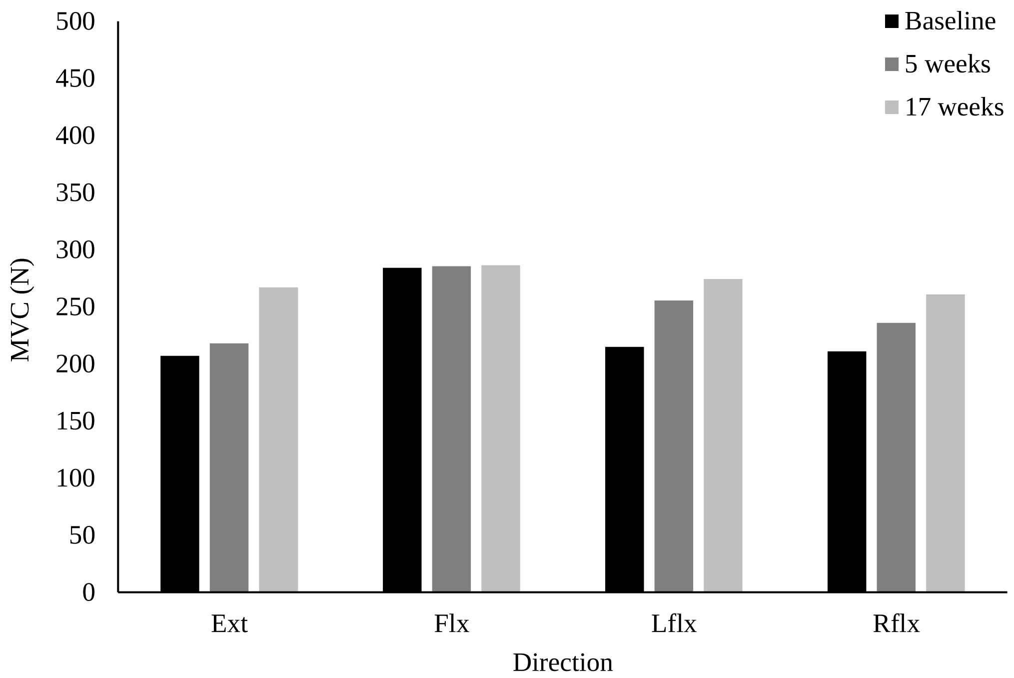

Figure 4.19: Maximum voluntary isometric contraction (MVC) in extension (Ext), flexion (Flx), and left- (Lflx) and right-lateral-flexion (Rflx) at baseline, post five and 17 weeks of training in SUM040.

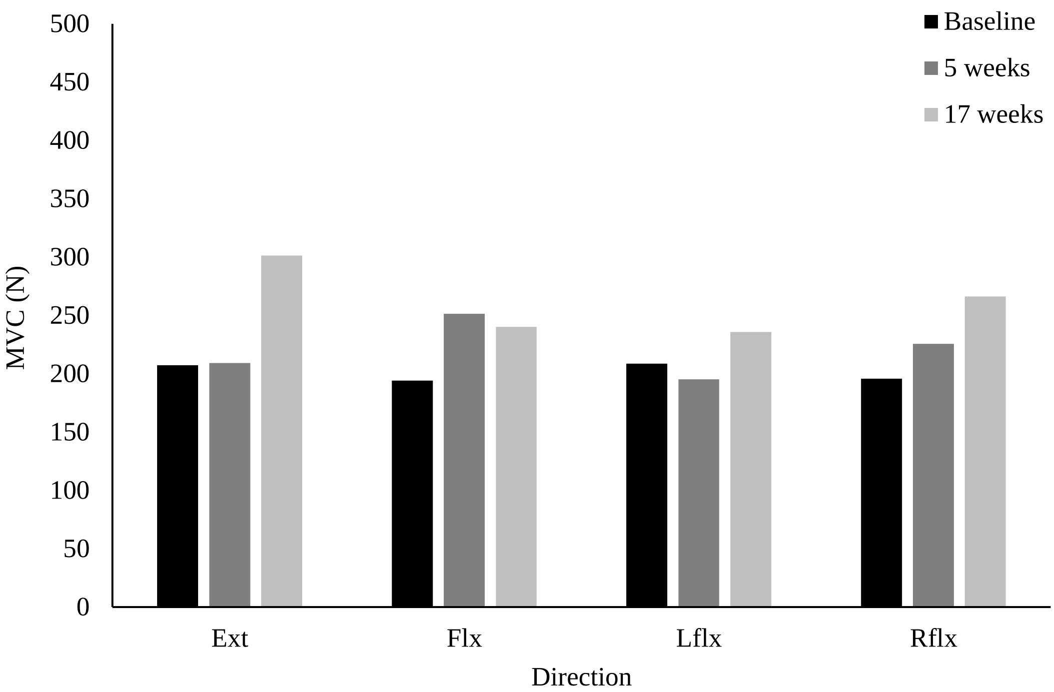

Figure 4.20: Maximum voluntary isometric contraction (MVC) in extension (Ext), flexion (Flx), and left- (Lflx) and right-lateral-flexion (Rflx) at baseline, post five and 17 weeks of training in SUM033.

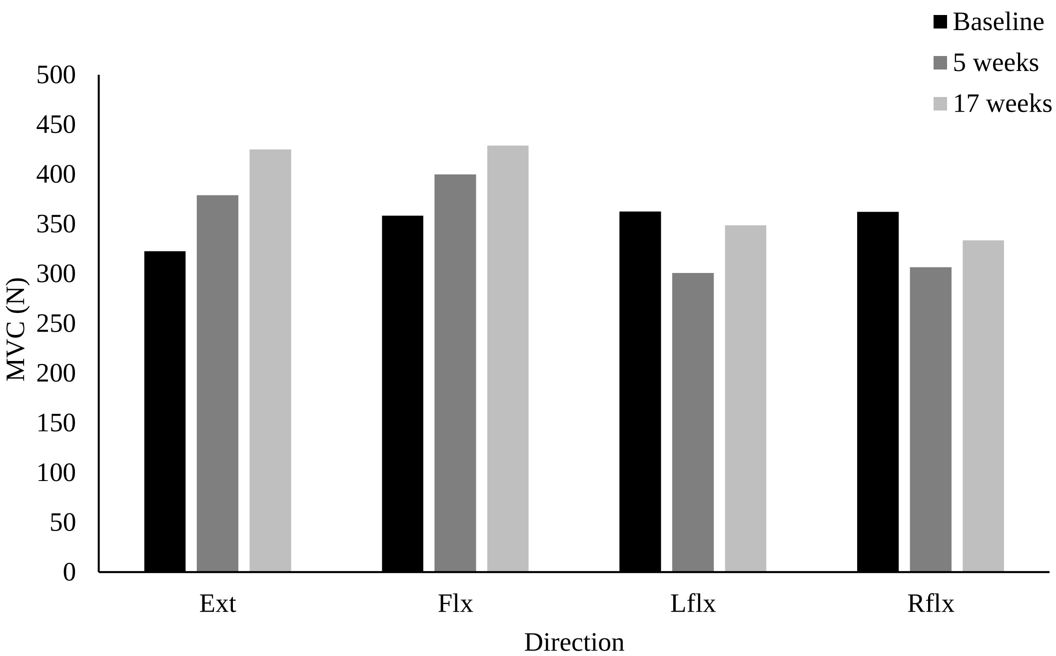

Figure 4.21: Maximum voluntary isometric contraction (MVC) in extension (Ext), flexion (Flx), and left- (Lflx) and right-lateral flexion (Rflx) at baseline, post five and 17 weeks of training in SUM035.

Chapter 5 : Discussion

5.1 Neck Strength Variables and Head acceleration

5.1.1 Maximal Neck Strength

The main finding of this study was that increased neck extension and total maximum voluntary isometric contraction (MVC) were significantly correlated to reduced peak rotational acceleration (PRA) experienced across the season. Variance in extension and total MVC also explained 33-41% of the variance in PRA sustained across the season, as well as specifically during a tackle, carry and ruck. The current findings, are supported by lab-based studies in rugby, that have reported the presence of a relationship between neck strength and head acceleration (Bussey et al., 2019; Dempsey et al., 2015). Using 3D motion capture, Dempsey et al., (2015) reported general correlations between increased neck strength and reduced head acceleration of the ball carrier during a simulated tackle. Similarly, reduced cervical muscle activation was observed in a sample of rugby players with a history of concussion and high magnitude head accelerations (Bussey et al., 2019).

The results of this thesis are similar to those that have been consistently reported in soccer heading (Caccese et al., 2018; Gutierrez et al., 2014; Peek et al., 2020; Tierney et al., 2005). In contrast, previous authors found neck flexor strength to predict reduced peak linear acceleration (PLA; Caccese et al., 2018; Gutierrez et al., 2014). This may be due to the different mechanisms causing acceleration and the dynamics of heading a ball. During soccer heading, the sternocleidomastoid (SCM) contracts eccentrically, moving the head posteriorly as the ball makes contact. This is followed by concentric contraction of the same muscles as the ball rebounds and the head moves anteriorly (Bauer, Thomas, Cauraugh, Kaminski, & Hass, 2001; Dezman et al., 2013). The extensor muscles contract at the same time to brace for impact (Caccese et al., 2018). The anterior-posterior motion creates a condition where high linear acceleration is likely. Furthermore, the dominant role of the SCM in controlling this motion may explain the relationship seen between SCM strength and PLA. Despite previous studies in soccer providing support for the relationship observed between head acceleration and neck strength, they only refer to acceleration caused by direct impact to a ball and findings have limited application to impacts that commonly occur in rugby.

In this thesis extension MVC explained a greater percentage of the variance in PRA in Rugby Union (rugby) compared to soccer heading (40% vs 17%; Caccese et al., 2018). This may be reflective of the relative contribution of the neck extensor muscles during rugby compared to soccer heading. Bussey et al., (2019) reported male rugby players to experience greater head acceleration during a simulated tackle, which was associated with reduced amplitude of the upper trapezius (UT) and splenius muscles. Research using electromyography (EMG) has also highlighted the dominant role of the UT during an American football (AF) and rugby tackle, placing the shoulder in hyperextension, elevating the scapular, and extending the cervical spine to maintain a head-up position (Lisman et al., 2012; Morimoto et al., 2013). The UT produces large moment arms due to their attachment site directly to the cervical region (Morimoto et al., 2013), therefore, they are thought to be associated with head-neck stability. The UT and splenius muscles form part of the posterior cervical muscles that are dominant in extension, acting as a first-class lever system (Marieb, 2000). These posterior muscles are also responsible for rotation and of the neck (Seeley et al., 2014).

The dominant role of the neck extensor muscles during rugby specific events, and the movements controlled by these muscles, may explain the significant correlation seen between extension MVC and PRA in this thesis. These findings suggest that increasing the strength of these muscles may be an effective strategy in increasing head-neck stabilisation in rugby, specifically with regards to rotation. Additionally, the relationship observed between total MVC and PRA provides support for increasing total neck strength as well as extension to increase head-neck dynamic stabilisation in rugby. This may have important consequences for the reduction of head impact burden experienced by rugby players, due to rotational acceleration being the dominant mechanism in brain injury (Meaney & Smith, 2011; Patton et al., 2013; Tierney & Simms, 2017a).

Contrary to the current findings, several studies within helmeted sports have reported a limited effect of neck strength on head acceleration (Mihalik et al., 2011; Schmidt et al., 2014). In a population of youth ice hockey players, those with the weakest UT muscles were seen to experience lower head impact magnitude than their stronger counterparts (Mihalik et al., 2011). Similarly, Schmidt et al., (2014) reported AF players with stronger and weaker cervical muscles to have the same likelihood of experiencing moderate and severe head impacts. This thesis used neck strength testing methods designed to reflect the demands of rugby. In comparison, the previous authors used non-sport specific tests (Mihalik et al., 2011; Schmidt et al., 2014). It is possible that these tests lack practical applicability and do not place the participant in a respective 'contact' posture. The force produced from the neck musculature during the test may not accurately reflect the force that they can produce during competitive contact. Similarly, the Head Impact Telemetry System (HITS), used to measure head accelerations, can experience 10 times the acceleration of the head on impact, due to excess translation and rotation (Joodaki et al., 2019; Manoogian et al., 2006). Consequently, differences in neck strength would not have accounted for differences in the inertial load recorded, as neck strength will not influence the degree of helmet movement during contact.

The higher magnitude head accelerations seen in individuals with stronger cervical muscles may also be explained by the phenomenon known as risk compensation. This theory would suggest that the awareness of reduced injury risk, due to greater cervical muscle strength and/or the use of a helmet, results in the engagement of higher-risk activities. This, in turn, may lead to the experience of high magnitude head accelerations (Hagel & Meeuwisse, 2004; Mihalik et al., 2011; Schmidt et al., 2014).

In this thesis, the direct effect of strength training on head impact magnitude was not assessed. However, the correlations observed between neck strength and PRA suggest that increasing neck strength may be an effective strategy to reduce head acceleration in rugby. This supported by Eckner et al., (2018) who reported eight-weeks of neck strength training to reduce head linear and angular velocity when subject to an external force. These findings, however, contradict those reported by Mansell et al., (2005) who investigated the effect of neck strength training on head-neck dynamic stabilisation in collegiate soccer players. These authors utilised an eight-week cervical resistance training programme, conducting non- and anticipated stabilisation trials pre- and post-training. They found that despite increases in neck flexor and extensor strength following training, there was no effect on any of the head kinematic variables. Similarly, Lisman et al., (2012) observed 7% and 10% increases in neck extensor and left-lateral-flexion strength, respectively, to have no effect of head-neck dynamic stabilisation during an AF tackle. The absence of a training effect could be due to head kinematic measurement techniques. In both of the previously mentioned studies, head kinematics were assessed using 2D and 3D motion capture systems with reflective markers placed on headgear (Lisman et al., 2012; Mansell et al., 2005). As these markers are not directly coupled to the skull, excess movement from the headgear may have resulted in acceleration values that were not representative of the head. Therefore, the increases in neck strength as a result of training would likely have a limited effect on the acceleration of the headgear.

5.1.2 Cause of Acceleration

An important finding of this thesis was that total MVC and extension MVC accounted for 34% and 48% of the variation in PRA respectively, for impacts resulting from direct head contact to hard body parts. Results revealed that direct head contact to any body part produced significantly higher PRA than indirect head impacts. Comparing this result with other studies is difficult as the majority of studies in rugby and other contact sports distinguish between direct impact locations, not between direct and indirect impacts (Broglio et al., 2011; King et al., 2018; King et al., 2015). Nonetheless, rotational loading is proposed to be the most dominant cause of brain injury, due to the brains low shear modulus (Meaney & Smith, 2011; Patton, McIntosh, & Kleiven, 2013; Tierney & Simms, 2018). High PRA due to direct head impacts, may result in high intracranial shear forces, tissue deformation and damage (Meaney & Smith, 2011). This may be why previous studies have reported direct contact to the head to occur in the majority of SRC cases (McIntosh, McCrory, & Comerford, 2000). The correlations between neck strength and head acceleration observed here, support the case for increasing neck strength as a method of limiting the magnitude of direct head impacts to hard body parts.

Direct head contact with the ground, although not significant, was associated with lower PLA and PRA values compared to other direct impacts. This may be representative of the players' ability to fall correctly during contact events, thus, allowing the head to be more controlled when hitting the ground. Despite not being directly measured in the current study, this may also indicate a level of anticipation of contact. Research has demonstrated that awareness of a forthcoming impact, and pre-activation of the neck musculature, allows for greater cervical stiffness and stabilisation of the head and neck (Kumar et al., 2000; Seminati et al., 2017). Similarly, rugby players who are visually unaware of imminent contact have been reported experience greater head motion (Tierney et al., 2019). Therefore, despite limited correlations observed between MVC and ground impacts, greater muscular pre-activation may result in reduced head acceleration during these events. As no measures of muscle activation were recorded in this thesis, further research is required to corroborate this.

5.1.3 Positional Differences.

Despite forwards having significantly greater total MVC compared to backs, there was no significant difference in the PRA experienced throughout the season. This finding is consistent with previous results that have been reported in rugby (King et al., 2015). In this study, no differences were present between these positions with regards to extension MVC. This may explain the lack of observed acceleration differences due to the relatively high contribution of extension strength to PRA.

The front row and inside backs experienced significantly greater PRA throughout the season compared to outside-backs. However, there were no significant differences in neck strength between these positional groups. This suggests that something other than neck strength may have influenced the observed differences in head acceleration. Research has reported outside-backs to cover a greater distance in sprinting and maximal sprinting than front-row forwards and inside-backs (Takamori et al., 2020). During sprinting, humans tend to increase trunk and head forward flexion angle during maximal velocity (Nagahara, Matsubayashi, Matsuo, & Zushi, 2014). Thus, when contact is initiated at high speeds, the ball carrier will be in a more upright posture. This creates a condition where tackles are made to the lower body, reducing the chance of direct head contact for the ball carrier. Similarly, this creates a condition where the tackler can more easily initiate contact to the waist. This may prevent the tackler from sustaining direct head contact to anatomical structures such as the hips or legs.

Front-row players and inside-backs, however, are often required to take the ball into contact from short distances, following a ruck or maul, and make upfront, first phase tackles. This creates a condition where players are likely to experience tackles to the upper body and direct head contact due to a lower centre of gravity and reduced trunk and head angle in the early phase of acceleration (Nagahara et al., 2014). Tierney and Simms (2017b) reported that tackles made to the upper body produced significantly higher head acceleration values. This indicates that teaching ball carriers to approach contact with a greater trunk and head angle may be a potential strategy to reduce inertial load. Thus, facilitating a reduction in the exposure to direct head impacts and reducing the PRA experienced on impact. Although further research is required to substantiate this speculation, this may be supported by previous authors who investigated the effect of reducing tackle height through law changes in rugby (Stokes et al., 2019). Under the new proposed laws, ball carrier and tackler behaviour changed, with ball carriers entering contact with a partially bent posture, and the tackler approaching contact with fully bent posture. Furthermore, these authors reported SRC to increase under the new proposed laws. Therefore, reduced trunk angle from the ball carrier and tackler may have had increased tackler exposure to direct contact to hard anatomical structures such as the driving knee of the ball carrier.

In contrast to this thesis, King et al., (2015) reported a number of outside-back positions to sustain the highest average PRA with inside-backs and front-row players experiencing lower magnitudes. Differences in findings may be due to sample size. In this thesis, there were a limited number of participants in each positional group. Similarly, some participants played multiple positions throughout the season and were represented in multiple positions. Thus, results may have been affected by individual characteristics as opposed to positional characteristics. Conversely, observed differences may be due to the error associated with the measurement system used in King et.al, (2015) as well as the limitations surrounding their video verification process. Therefore, comparisons drawn between the two studies should be treated with a degree of caution.

The lack of differences between broad positional groups and the presence of differences between specific positional groups in this thesis supports the notion that head impact burden and neck strength should be assessed in relation to specific positions. Furthermore, other techniques in addition to increasing neck strength should be considered when implementing strategies to reduce head acceleration in rugby.

5.1.4 Neck Strength Imbalances

Agonist/antagonist muscular imbalance has been proposed as an important factor in head injury prevention (Dezman et al., 2013; Morimoto et al., 2013; Peek et al., 2020). Due to a lack of accurate measures of head weight, this thesis was unable to investigate this relationship. The association between muscular imbalance and injury for other areas such as the hamstring and shoulder is well established (Croisier et al., 2008; Wang & Cochrane, 2001; Yeung et al., 2009). However, little is known about neck musculature imbalance and head injury. EMG study of cervical musculature has reported that head-neck stability may be improved through co-contraction of neck extensor and flexor muscles during a tackle (Morimoto et al., 2013). It has been reported that improving cervical extensor/flexor symmetry may reduce the magnitude of acceleration during soccer heading through increasing the relative mass of the head and reducing oscillations (Dezman et al., 2013; Peek et al., 2020). These authors reported significant correlations between increased neck extensor/flexor imbalance and increased PRA on impact. Whilst the resistance training programme used in this thesis was ineffective in reducing extensor/flexor imbalance after five weeks, the case study results indicate that 17 weeks of neck-specific resistance training may be effective in doing so. Further investigation is required to establish the statistical relevance of this finding and to determine the effect of this on head acceleration experienced in rugby.

5.1.5 Anthropometric Variables

The results of this study suggest a limited contribution of neck circumference to controlling head acceleration. This is different to results previously observed in soccer (Caccese et al., 2018). Discrepancies may be a result of the previous study grouping size variables. These authors reported a regression model of neck circumference and head mass to explain 22.1% of the variance in head acceleration. However, only head mass was reported to be the significant predictor of rotational acceleration (Caccese et al., 2018). Newton's Second Law of Motion (force = mass x acceleration) suggests that greater mass of the head would lead to lower linear acceleration. Equally, when an object is subject to torque, the rotational acceleration it experiences is proportional to its moment of inertia. Since the object's moment of inertia is dependent on its mass, theoretically, an athlete with greater head mass should experience reduced rotational acceleration. This suggests that in grouping the variables, the results of the regression model are more representative of the contribution of head mass to acceleration as opposed to neck circumference. This would explain why no significant relationship between neck circumference and head acceleration was observed in this thesis. Unfortunately, no accurate measures of head mass could be obtained in this thesis, as such, the relationship between head mass and acceleration was not explored. Furthermore, the previous study was in soccer, therefore substantial differences in sporting demands make comparisons between the two sets of results difficult.

Tierney et al., (2005) reported that individuals who experienced greater head accelerations to have lower neck circumference. Similarly, lower neck circumference and neck-to-head circumference ratio have been reported in those who experienced an SRC compared to non-injured individuals (Collins et al., 2014). In this thesis, these two anthropometric variables showed significant, positive correlations to neck strength and neck strength was seen to be inversely correlated to head rotational acceleration. This may suggest that higher neck circumference and neck-to-head circumference ratio are simply a biproduct of greater neck strength and do not directly affect acceleration. This is supported by Collins et al., (2014) who, despite recording significant differences between participants in neck girth and neck-to-head circumference ratio, found neck strength to be the only significant predictor of concussion risk.

The results from this thesis indicate that, of the variables measured in this study, neck strength is the strongest predictor of head acceleration in rugby. However, the limited sample size, coupled with the lack of research in rugby to draw comparisons from, limits the ability to draw definitive conclusions. This indicates a need for further investigation. Similarly, there is evidence to suggest that head mass and neck musculature imbalances may also be strong predictors of neck strength (Caccese et al., 2018; Dezman et al., 2013; Peek et al., 2020). Correlations between these variables, however, were not assessed in this thesis. Therefore, future research should consider these variables alongside neck strength when assessing predictors of head acceleration.

5.2 Neck Strength Measures

5.2.1 Anthropometric Correlates of Neck Strength

Determining anthropometric predictors of neck strength may provide an indication of strength when direct testing measures are not available. In this thesis, BMI, body mass and neck-to-head circumference ratio showed significant positive correlations to total MVC and MVC in all directions. Neck circumference was only significantly correlated to flexion and left-lateral-flexion. The positive relationship between maximal strength and muscle cross-sectional area (CSA) is well documented (Maughan et al., 1983). Hence, the association seen between MVC and neck circumference was to be expected. These findings are supported by a similar study at the highest level of amateur New Zealand rugby that observed neck circumference to be significantly correlated to neck strength in all directions (r = *0.33-0.63*) (Salmon et al., 2018). The absence of significant correlations in extension and right-lateral-flexion in this thesis is likely a reflection of the lack of statistical power in the sample size.

Research in adult and youth male front-row rugby players has generated a regression model where playing experience and body mass accounted for 31% of the variation in neck extension strength (Hamilton et al., 2014). This supports the relationship seen between body mass and neck strength in this thesis. Little is known about the relationship between BMI and neck-to-head circumference ratio and neck strength. However, the moderate correlations in this thesis indicate that these may also be effective predictors of isometric neck strength. These findings may support the use of a combination of these variables as a function of neck strength. A limitation of this thesis is a relatively small sample size. Larger cohorts are required to investigate these relationships further to determine an effective surrogate measure of strength. in

5.2.2 Neck Strength in Different Playing Levels and Sports

Baseline neck strength values recorded in this thesis were comparable to those seen previously in amateur rugby players using a similar methodology (Salmon et al., 2018). These values are lower than those previously recorded in professional rugby players using fixed frame dynamometry (extension 368 N, flexion, 278 N, Left, 362 N, right-lateral-flexion 376 N; Naish et al., 2013). The higher recorded values in professionals compared to amateurs is expected due to significant differences in playing demands at increasing levels of participation (Quarrie, Hopkins, Anthony, & Gill, 2013). In contrast, previous studies have reported substantially higher neck strength scores for professional and amateur players compared to those observed in this thesis and other studies with professionals (Geary et al., 2013, 2014; Naish et al., 2013; Salmon et al., 2018). Geary et.al., (2014) reported professional players to produce raw MVC of flexion, 335 N, extension 606 N, left-lateral-flexion 556 N and right-lateral-flexion 570 N. This difference is likely due to the measurement technique used. These authors utilised handheld dynamometry, whilst participants were seated with limited restriction of accessory muscles. Therefore, higher recorded values may reflect the ability to recruit accessory muscles during testing, resulting in greater force production. Similarly, high force values have been recorded in collegiate level AF players, using similar seated testing methods (Lisman et al., 2012). Comparison between different studies and different sports should be treated with caution due to variation in testing methods.

Within existing research, extension strength is consistently reported to be greater than neck flexion strength (Geary et al., 2013; Naish et al., 2013; Salmon et al., 2018). However, the opposite was observed in this thesis. This is likely a result of the position that participants were tested in, as the weight of the head will have affected both the extension and flexion scores. Furthermore, Salmon et al., (2018), who used a similar testing position, were able to obtain accurate measures of resting head weight and accounted for this in each direction. Unfortunately, due to time constraints, a method of accurately obtaining head weight was not available. Future studies should obtain accurate measures of head weight, a method for which would need to be developed and validated. Head weight should be considered when comparing flexion values from this study to previous research.

5.2.3 Neck Strength in Different Playing Positions

In this study, forwards were significantly taller, heavier and had a greater neck circumference and neck-to-head circumference ratio than back, in agreement with previous findings (Salmon et al., 2018). This is likely reflective of the physical characteristics that are required to sustain the demands of these positions (Takamori et al., 2020). Absolute baseline MVC in flexion and left- and right-lateral-flexion was significantly higher in forwards compared to backs. Greater extension strength was also observed in forwards, however this failed to reach significance. Similar patterns of greater neck strength and circumference in forwards have been reported previously (Salmon et al., 2018). This relationship has also been reported when comparing neck strength in various combat sports athletes, with wrestlers reporting greater muscular CSA and strength compared to judo athletes (Tsuyama et al., 2001). In wrestling, athletes keep their necks extended to prevent their shoulders from being pinned. Exposure to repeated mechanical stress of this kind may lead to physiological adaptions that result in increased strength and size. Similar cervical stress may also be present for forwards in rugby, with high reported levels of neck muscle activation during scrummaging (Cazzola, Stone, Holsgrove, Trewartha, and Preatoni, 2016), As such, the greater neck strength and circumference may be reflective of the muscular adaptation required to sustain these demands. This may also explain why front-row forwards had significantly greater neck strength than other specific positional groups, due to their heavy involvement in the scrum.

There were no differences in relative MVC (normalised to body mass) between broad positional groups in this thesis. These results are different from those seen by Salmon et al., (2018), who reported amateur forwards to have a higher relative neck strength in all directions compared to backs in. Similarly, Olivier and Du Toit (2008) reported a presence of significant relative strength differences between forwards and backs in professional players. Within the current cohort of players there was variability with regards to individual playing level. Whilst all players were members of the first team for the university, a number of players also compete or have competed at academy level. This was not, however, consistent across positions. Differences in neck strength have been recorded at differing levels of play (Naish et al., 2013). This lack of consistency in playing level may have influenced positional averages, resulting in an absence of significant differences between broad positions. This may also explain the finding that when split across individual positional groups, a main effect of position on relative neck strength was observed in several directions. Unfortunately *post hoc* analysis was unable to identify the specific differences due to a lack of statistical power in the sample size.

5.3 Neck Strength Training

5.3.1 Maximal Strength

Five weeks of neck-specific isometric resistance training resulted in significant increases in MVC in all directions except extension, with the greatest increases seen in left-lateral-flexion. The greatest increases in MVC were attained by those who had the highest adherence to the neck-specific resistance training programme. Despite an overall increase in extension, and the high adherence group showing a greater increase in extension compared to the low adherence group, no significant change was observed in any of the conditions. This may suggest that the resistance training programme failed to induce any meaningful increases in extension strength, testing in this configuration. These findings are inconsistent with Geary et al., (2014), who found five weeks of training to result in significant increases in extension strength in professional players. Several methodological differences exist between the two studies, which may have contributed to these results. For their training programme, Geary et al., (2014) used manual pressure provided by the coach. This resistance cannot be quantified and may have been greater than the fixed values used in this thesis. Geary et al., (2014) also used seated, handheld dynamometry to assess neck strength, so findings cannot be directly compared.

Consistent with the current results, Salmon (2014), using similar training and testing methods, found increases in all directions except for extension. This study reported a matched control population (rugby players with no neck training) to have decreased extension following a season of amateur rugby. This is indicative that neck strength training mitigated the loss of strength that naturally occurs through the demands of a rugby season. This may explain the results seen in this thesis. Conversely, Salmon et al., 2018) reported that a season of professional rugby - with no specific neck strength training led to increased neck strength in all directions. Differences in playing level may explain the contrasting results, however, without the presence of a control sample in the current research, it is not possible to draw a reliable conclusion. In this thesis, the increase seen for extension in the high-adherence group was only 1.6% less than the significant change seen for the same group in flexion. Similarly, results from the three case studies showed trends towards greater increases in extension following 17 weeks of training. It is possible that with more post-season data and a greater sample size, a significant change may have been present.

A number of studies have found that five and six weeks of neck-specific resistance training produced no significant changes in neck strength (Barrett et al., 2015; Naish et al., 2013). Discrepancies may be due to differences in training modality and resistance. Firstly, Barrett et al., (2015) used a starting resistance of 50% MVC; 15% lower than this thesis. Resistance training models recommend that novice individuals use an initial training load of 60-70% MVC to elicit strength gains (American College of Sports Medicine, 2009). Despite this, Naish et al., (2013) used a similar starting resistance to this thesis; this may, therefore, not account for the observed differences. Another explanation the exercise selection. Previous studies solely prescribed isometric exercises, to target superficial muscles, as part of their training programme (Barrett et al., 2015; Naish et al., 2013). This thesis used a combination of deep neck stabiliser (DNS) exercises and superficial muscle training.

EMG analysis of the cervical muscles in helicopter pilots has reported that the smaller, deeper agonist muscles are highly susceptible to fatigue during isometric movements (Harrison et al., 2009). Specifically training these muscles may therefore contribute to enhanced force production. Salmon et al., (2013), in a non-rugby setting, reported that the use of specific DNS exercises in conjunction with superficial muscle training was more effective in increasing neck strength than superficial exercise alone. This was further supported by Salmon, (2014) with professional rugby players, indicating that a multifaceted training programme utilising deep and superficial muscle training resulted in a significant increase in neck strength.

This thesis supports the findings of Salmon (2014) and Salmon et al., (2013), showing a five-week (minimum) resistance training program to increase isometric neck strength in male rugby players. Additionally, a combination of deep and superficial muscle training appears to be most effective. The results of the current case studies support previous research, indicating that programmes of longer duration are likely to elicit greater strength gains (Conley et al., 1997; Mansell et al., 2005; Salmon et al., 2013).

Total MVC significantly increased in both low and high-adherence groups. This may be reflective of the sensitivity of the cervical musculature to resistance training, with requiring a relatively low training frequency to obtain significant strength gains. All participants were also completing a general resistance training programme throughout the season. Eckner et al., (2018) reported increases in neck strength in those who completed general resistance training with no specific neck exercises. Therefore, neck strength increases in low-adherence players may be attributed to the indirect effect of non-specific resistance training. These authors also reported substantially greater increases in neck strength in those who completed neck-specific resistance training (as well as general training). This further supports the greater increases in high compared to low adherence groups in this thesis.

The majority of participants were relative novices with respect to specific neck resistance training. It has been reported that, when compared to trained, untrained individuals experience greater increases in strength as a result of training (Ahtiainen, Pakarinen, Alen, Kraemer, & Häkkinen, 2003). This may in part explain the significant increase seen in the low adherence group. Similarly, Paulsen, Myklestad, & Raastad, (2003) reported that individuals who partake in a greater volume of training produce the greatest increases in strength. This would further explain the greater increases observed in high compared to low adherence groups.

The strength gains observed in the low adherence group may also be the result of participation in rugby training and games. Previous results have shown a season of rugby, with no specific neck training, to cause significant increases in neck strength in forwards and backs (Salmon et al., 2018). Research into muscle activity during rugby tackles has reported neck musculature activity of up to 20.9% of MVC (Morimoto et al., 2013). This muscle activity may produce enough stimulus to facilitate strength adaptations. Consequently, it is possible that all players, regardless of their training history, experienced an increase in neck strength due to rugby participation, with neck strength training leading to additional increases. Salmon (2014), however, reported a reduction in neck strength following a season of rugby. Consequently, the effect of rugby participation on neck strength is not conclusive, and further research should be conducted to establish this relationship. Similarly, further research is required to identify the exact reason for the neck strength increases observed in the low adherence group. Nevertheless, the results of this thesis, supported by previous research, indicate that high adherence to neck-specific resistance training is required to elicit the greatest increases in isometric neck strength.

5.3.2 Imbalances

The training programme used in this study resulted in no significant change in the anterior-posterior and lateral imbalances that were observed at baseline. The limited change in anterior-posterior imbalance may reflect of the lack of significant change in extension compared to the significant increase in flexion. At the start of the training programme, a focus was placed on DNS training. The DNS muscles that were targeted during these exercises largely contribute to flexion of the neck. This may have led to the greater increases in strength that were observed in flexion compared to extension, subsequently maintaining the initial imbalance. Adding support to this observation, the anterior-posterior imbalance showed a decreasing trend following 17 weeks of training in two of the three case studies. This indicates that training of a longer duration, with equal focus on each muscle group, may be sufficient to reduce the imbalance between flexor and extensor muscles. This may have important consequences for head-neck stabilisation.

The lack of change in the initial imbalance between lateral neck flexors may be representative of individuals' preferred tackle side. Repeatedly tackling with the same shoulder may cause cumulative microtrauma to that region of the cervical spine

(Pelham, White, Holt, & Lee, 2005). This trauma may lead to pain and reduced force production (Pelham et al., 2005). With the majority of participants playing rugby for over 15 years, being exposed to a high frequency of collisions is likely to have predisposed this group to such a condition. Despite both left-lateral-flexion and right-lateral-flexion improving as a result of training, the extra repetition given to the weaker side was insufficient to account for the reduced force production as a result of repeated microtrauma. As the direct effect of a preferred tacked side on directional neck strength was not assessed in this thesis, future research should be conducted to investigate this relationship further.

5.4 Head Impact Verification

The results of this study suggest that the head impact magnitudes currently associated with rugby, and potentially other contact sports, may be overestimated. The median and interquartile range (IQR) values recorded in this thesis are substantially lower than those previously recorded in rugby and other sports (Broglio, Martini, Kasper, Eckner, & Kutcher, 2013; Cobb et al., 2013; Crisco et al., 2011; King et al., 2015; Mihalik et al., 2007; Rowson et al., 2009). Broglio et al., (2013) reported that high school AF players sustained average PLA and PRA values ranging from 26-28 *g* and 1,741-1,826 rad•s^{-2} across positions. Additionally, King et al., (2016) reported that rugby players sustained similar average PLA values to those reported in this thesis, but substantially higher PRA values. The reasons for differences between studies may be due to the head impact telemetry system used.

The majority of existing research used helmet-mounted sensors such as HITS or head-mounted sensors to measure head acceleration, which have been associated with a measurement error of up to 298% (Cummiskey et al., 2017). This measurement error is thought to be due to insufficient sensor skull coupling. Studies have reported that helmet-mounted sensors can translate and rotate up to 41 mm and 37° in excess to the head on impact (Joodaki et al., 2019; Wu et al., 2016). Furthermore, head-mounted sensors adhered to the skin have been shown to displace by up to 4 mm on impact, due to the presence of soft-tissue artefact (STA) (Wu et al., 2016). In contrast to this, inertial motion units (IMUs) embedded in instrumented mouthguards (iMG) have been shown to displace by less than 1 mm relative to the skull on impact, with the iMG used in this study showing systematic agreement with a Hybrid III anthropometric testing device (Greybe et al., 2020; Wu et al., 2016). This highlights the importance of using tightly coupled sensors to obtain accurate PLA and PRA measurements, to effectively inform injury prevention strategies.

Interestingly, King et al., (2015), whilst using iMGs to record head acceleration, reported substantially higher average head impact magnitudes (22 *g* and 3,903 rad•s^{-2}). A possible explanation for this is that these authors may have included false positive impacts in their dataset. These authors reported that only 65% of impacts could be video verified; multiple impacts occurred in ruck and mauls that could not be verified. The head impacts in this thesis were subject to a rigorous verification criterion, so that only true, video verified impacts were included in the analysis. The head impact verification process identified 30% of recorded impacts as false positive impacts. Furthermore, a comparison of the magnitude of false and true positive impacts showed false impacts to have a significantly higher median (40%), IQR (14–59%), and max (67%) PRA values. This suggests that the inclusion of false impacts may lead to a significant overestimation of impact magnitude. This may have important consequences regarding injury metrics and prevention strategies, highlighting the importance of having a comprehensive video verification process.

Impact verification also plays a vital role in reporting the frequency and density of head impact events. A large number of impacts in this thesis were not included in the analysis due to poor waveform quality. Whilst this decision ensured that no false positive impacts were included, it also presents the opportunity for true impacts to be omitted from the analysis. Therefore, frequency data was not reported in this thesis. However, the inclusion of false positive impacts may also lead to an overestimation of the frequency of head impact burden in contact sports. Inaccurate estimation of head impact frequency may misrepresent the risk that the sport has on cumulative impact burden and impact density. Rowson et al., (2019) reported that sustaining a high number of impacts increases an individual's risk of sustaining an sports-related concussion (SRC). Similarly, sustaining a high density of impacts in a given period is reported to predispose an individual to a higher risk of SRC (Broglio et al., 2017). Hence, frequency and density data are essential to accurately assess an individual's injury risk.

A variable that may be important in the identification of false positive impacts is the relationship between linear and rotational acceleration. In this thesis, false positive impacts produced a ratio of PLA to PRA that was 47% greater than true positive impacts. In all SRC cases, both linear and rotational acceleration is present (Meaney & Smith, 2011). Therefore, it is hypothesised that when the human head is subjected to a certain level of rotational acceleration, a proportional level of linear acceleration will be present.

Equation 1: where a = Linear acceleration at the centre of gravity (m.s^{-2}), $\vec{a}_s$ = Linear acceleration at the sensor (m.s^{-2}), $\vec{\alpha}$ = angular acceleration at the sensor (rad.s^{-2}), $\vec{\omega}$ = angular velocity at the sensor (rad.s^{-1}), $\vec{r}_s$ = displacement vector (m)

$$a = \vec{a}_s + \vec{\alpha} \times \vec{r}_s + \vec{\omega} \times (\vec{\omega} \times \vec{r}_s)$$

Equation 1, as given in Wu et al., (2016), states that there is a relationship between linear and rotational acceleration, where larger rotational acceleration will produce a larger linear acceleration. Thus, PRA may be more closely related to PLA in true impacts compared to false impacts, as observed in this thesis. However, as seen in **Figure 4.3**, there are occasions where false positive impacts have similar PLA and PRA values to true-positive impacts. Consequently, this variable may allow the classification of false impacts, but not the classification of true impacts. As such, at this current stage, it is not possible to confirm false and true positive impacts from the relationship between these variables alone. Further investigation is required to establish the nature of this relationship. Nonetheless, this may be an important variable to aid in the removal of a significant proportion of false positive impacts from head impact datasets.

A further explanation for the high magnitudes reported in previous studies may be the use of unfiltered time series data. This study applied a 4th order, zero lag, Butterworth filter to remove high frequency noise from the raw accelerometer and gyroscope time-series data (Greybe et al., submitted 2020). Filtering techniques can significantly affect resultant head impact magnitudes (Greybe et al., submited 2020;Liu et al., 2020). As demonstrated in this thesis unfiltered PLA and PRA values are significantly higher than their filtered counterparts. An example of this is shown in **Figure 3.2**, where filtering removed a substantial artefact in the waveform and resulted in a PRA reduction of 602.38 rad•s^{-2}. Similarly, with the impact shown in **Figure 3.1**, filtering removed an artefact in the waveform and resulted in a PLA reduction of 6.82 g. This highlights the importance of applying an appropriate data derived filter to head acceleration data and how reporting unfiltered data can result in significantly overestimated PLA and PRA values data. Additionally, data processing techniques should be specifically described in studies to allow for accurate comparison of results.

An important stage in effective injury prevention is to apply the assessed biomechanical inputs to various physical and computational models. This allows the investigation of the brain response to those inputs, to define human tolerance levels (Meaney et al., 2014). The lower values reported in this thesis may have important implications with regards to existing brain injury metrics. Previously published values in helmeted adult sports suggest that PLA < 66 *g*, and PRA < 4,600 rad•s^{-2} can be classified as 'mild' impacts, with computational models suggesting these magnitudes present a 25% chance of sustaining a mild-traumatic brain injury (mTBI) (Broglio et al., 2011; King et al., 2015; Zhang, Yang, & King, 2004). Additionally, SRC events in youth AF are reported to be associated with average PLA and PRA values of 62.4 ± 27.9 *g* and 2,609 ± 1,591 rad•s^{-2} (Campolettano et al., 2020). These classifications have been formed based on data obtained from previously described inaccurate HITS (Joodaki et al., 2019), and whilst they may be accurate when using this type of system, the relatively low magnitudes recorded in this thesis suggest that previous values may not be relevant when using tightly coupled sensors. Additionally, the data obtained from inaccurate systems does not represent actual head accelerations, due to the excessive movement between the sensor and skull (Joodaki et al., 2019; Wu et al., 2016). Therefore, these data are unlikely to accurately predict injury tolerance when using complex finite-element models (FEM).

Further research is required, employing a new minimum standard for recording and reporting head impact data, to develop a more accurate picture of human tolerance and injury thresholds. It is recommended that head impact telemetry systems have a minimum coupling requirement to minimise the effect of STA. Similarly, studies should apply appropriate post-processing steps such as impact specific filtering and rigorous impact verification processes.

5.5 Limitations

The primary limitation of this thesis was the lack of data available for final neck strength testing following 17 weeks of resistance training. The unforeseen circumstances of a global health pandemic, along with the time-sensitive nature of data collection, meant that final neck strength data could only be collected for three participants. As a result, the efficacy of a 17-week neck-specific resistance training programme and the direct effect of neck strength training on head acceleration could not be assessed. Despite this, the general relationship observed between higher neck strength and reduced head acceleration provides a strong rationale for further investigation into the direct effect of training. Similarly, the results of the three case studies provided an important insight into the effects of a 17-week neck-specific resistance training programme on neck strength.

As with any research working with human participants, compliance with the training programme was a limitation within this study. Half of the participants included in the study completed less than 25% of the available training sessions, with one participant having a maximum attendance of 70%. This means that the full effects of the resistance training programme may not be represented within the results. This low compliance from half of the population, however, presented an opportunity to effectively analyse the effect that training adherence had on neck strength adaptations.

Complications with the iMG system resulted in limited participant and head impact sample size. Throughout the season hardware issues resulted in an inconsistent number of sensors used in each game. Furthermore, a high proportion of head impact events recorded could not be verified due to limited waveform data, so were not included within the head impact analysis. This reduced number of head impact events recorded, and subsequently limited the statistical power of the results. Similarly, due to the use of field-based measures and the nature of the game of rugby, injuries, substitutions, and opposition ability, could not be accounted for, which may have affected results and the head impact frequencies observed. Therefore, frequency data was not reported in this thesis. Nonetheless, the rigorous verification criteria employed in this study ensured that the data were only representative of true impact events, thus improving the reliability of the magnitudes reported.

The limited number of camera angles coupled with the obstructive contact nature of rucks and mauls meant the direct mechanism causing acceleration could not be completed for 10.4% of the impacts recorded. Whilst these account for a relatively low proportion of impacts, it may have affected the differences seen between the magnitude of different causes of acceleration. Finally, the results given are only representative of an amateur senior men's university rugby team, and findings should not be generalised to other levels of play, age, sex, or other sports.

Chapter 6 : Conclusions and Future Directions

This thesis was the first examination of the effect of neck strength on head acceleration in rugby union using field-based measures of head acceleration. The results of this study indicate that increasing neck strength may be an effective strategy to reduce head acceleration experienced during competitive amateur rugby matches. In particular, focus should be given to increasing neck extension and total neck strength to reduce peak rotational acceleration. This study has demonstrated that a five week, multifaceted, neck-specific resistance training programme, focusing on deep neck stabilising and superficial muscles, is effective in increasing the strength of the cervical musculature. Results from the three case studies also suggest that a resistance training programme of a longer duration may elicit greater strength adaptations. Future studies of larger sample sizes should focus on investigating how changes in neck strength, as a result of training, directly affect the head impact magnitude experienced. Similarly, whilst this thesis highlights the effect of neck strength on impact magnitude, it was unable to ascertain the effect on impact frequency or direct injury risk. Future research should investigate this relationship to determine the effect of neck strength on cumulative head impact burden and risk of brain injury.

The current findings also suggest that the currently accepted values surrounding head impact events in contact sports may be over-estimated. This has important consequences for the development of injury metrics and prevention strategies. The overestimation is likely due to a combination of factors. Firstly, the head impact telemetry systems used to quantify head acceleration are associated with a high degree of measurement error due to insufficient sensor skull coupling and soft-tissue artefact. Thus, future research in this area should utilise tightly coupled, reliable systems to collect accurate head acceleration data. Secondly, the majority of studies in this area do not report data processing techniques, specifically with regards to filtering raw-time series data. The results of the current thesis have demonstrated that applying low pass, 4^{th} order, zero-lag, Butterworth filter with variable, impact-specific filter cut-off frequencies to raw-time series data significantly reduces head acceleration magnitude. Therefore, it is recommended that studies apply appropriate impact specific filters to their data, as well as reporting data processing techniques to allow for accurate comparison of results.

Finally, findings suggest that previous studies may have included false positive impacts within their final analysis due to errors in impact verification techniques. This study has demonstrated that, compared to true impacts, false positive impacts are associated with significantly higher peak linear (PLA) and peak rotational (PRA) acceleration values. Including these impacts may lead to a significant overestimation of head impact magnitude. This study has highlighted the importance of, and proposes, an extensive video verification system to ensure that reported acceleration data is representative of actual head impact events. Additionally, the ratio between PLA and PRA has been identified as a variable that may be important in the identification of false positive impacts. Future research should further investigate this relationship to establish its efficacy as an identification variable.

Appendices

Appendix A: Mechanical specifications: Neck Strength Testing Equipment.

Mechanical Specifications for the Safety of a Bespoke Isometric Neck Strength: Testing Rig for Rugby Athletes

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October 6th, 2019

Overview

The objective of this project is to measure the isometric neck strength and strength endurance of rugby athletes. This forms part of a wider initiative to minimise head inertial loading in training and competition. A test rig has been constructed to enable this testing. This rig is designed to facilitate repeatable test measures, ensuring that accessory muscles are restricted, so that only the muscles of the neck can be recruited. Four 35 kg Tedea-Huntleigh load cells have been used to measure neck strength in four directions; flexion, extension, left and right lateral flexion. This document describes the mechanical specifications of the rig as part of the risk assessment required to carry out testing protocols.

Figure 1: The neck strength test rig with a person demonstrating the required position

Requirements of the Rig

The testing position is shown in Figure 1. The participant is in a prone position, with their torso strapped to the horizontal bench with a car racing harness. Feet will be off the ground with their knees resting on a cushion with the height adjusted for each person. The head is positioned in the centre of the four inward-facing load cells and each load cell has a neoprene pad attached via a 85*60 mm aluminium platform. For testing, participants will push with maximum effort against each load cell in the specified direction. These efforts will be sustained for durations of between 2 and 6 seconds and will be repeated between three and five times per direction for each testing session. The frame of the rig (Figure 1) must be able to support the body weight of the heaviest rugby athletes, without flexing at all. The heaviest elite rugby player in the world currently is 142 kg. The average weight of our current university study population is 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 accommodate athletes ranging in height from 150 cm to 195 cm. The horizontal bench is adjustable in a forwards and backwards direction. The entire headset, in the box marked B in Figure 2, can also be adjusted forwards and backwards. The portion of the headset in box C in Figure 2 can be adjusted in a vertical direction. When adjusting for each individual, the position of the neoprene pads must be positioned to the same location on each person's head.

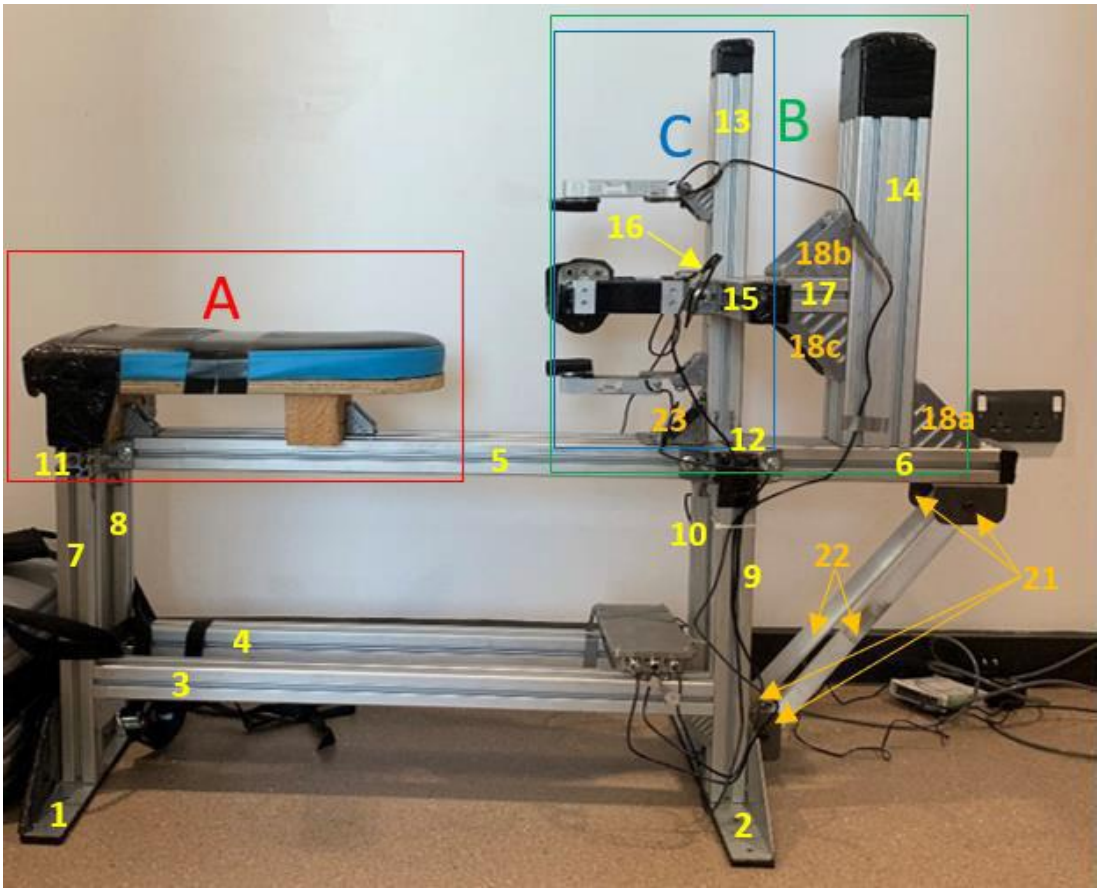

Figure 2: A side view of the neck strength rig showing the position of the bench and the headset with the mounted load cells. A indicates the horizontal bench with forwards-backwards adjustment. B indicates the entire head piece which can be adjusted forwards and backwards. C indicates the headset which can be adjusted vertically. Yellow numbers relate to frame components listed in Table 1 and orange numbers relate to the connectors listed in Table 1.

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. The rig used by these authors, however, enabled accessory muscles to be recruited which is expected to result in higher neck strength readings

Rig Design and Construction

The design and construction of this neck strength test rig has been completed with the assistance of Roberto Sotgiu, who is a qualified mechanical design engineer (MEng (hons), Bath, 2000). Roberto has significant experience in the special purpose machinery industry, primarily in the design of bespoke test/assembly/feature-checking machines for the manufacturing sector.

The frame of the neck strength rig has been entirely constructed with Bosch Rexroth aluminium profile extrusion products, which can be viewed here:

(https://www.boschrexroth.com/en/xc/products/product-groups/assembly-technology/topics/aluminum-profiles-solutions-components/aluminum-profiles-products/index

Each strut is fastened with a minimum of two rigid brackets and fasteners have been torqued to the required manufacturer's specification. This makes the frame completely rigid and capable of withstanding the loads required for the testing of rugby athletes neck strength. This will be the case so long as all fastenings are torqued to 100% and positioned as per the specifications in Figure 2. Table 1 provides a list of all structural components shown in Figure 2.

Table 1: List of all structural components which are indicated in Figure 2

Item No.	Description
	Frame length components
1	Steel foot stand bracket 500*100*45mm
2	Steel foot stand bracket 500*100*45mm
3	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 800mm length
4	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 800mm length
5	Bosch Rexroth extrusion 90*45 mm, 10mm slot, 800mm length
6	Bosch Rexroth extrusion 90*45 mm, 10mm slot, 800mm length
7	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 450mm length
8	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 450mm length
9	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 450mm length
10	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 450mm length
11	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 200mm length

12	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 200mm length
13	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 500mm length
14	Bosch Rexroth extrusion 90*90 mm, 10mm slot, 500mm length
15	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 220mm length
16	Bosch Rexroth extrusion 45*45 mm, 10mm slot, 220mm length
17	Bosch Rexroth extrusion 90*45 mm, 10mm slot, 120mm length
	Angle Brackets and Connectors
18	Bosch Rexroth Strut Profile Angle Bracket, strut profile 90 mm
	3 brackets: joining 14 to top of 6 (a), 14 to top of 17 (b) and 14 to bottom of 17 ©
19	Bosch Rexroth Strut Profile Angle Bracket, strut profile 45 mm
*4	20 brackets: 1 each joining 7, 8, 19 a 10 to 3 and 4 respectively
*4	joining 7, 8, 9 & 10 to the inside of 11 and 12 respectively
*4	joining 11 & 12 to 5 respectively, with one on either side of 5
*2	joining 12 to either side of 6
*2	joining 2*4 timber supports of flat bench (A) to both grooves of 5
*4	joining each load cell to items 13, 15 and 16 via the mild steel fittings
20	Bosch Rexroth Strut Profile T-Head Bolt
	4* each of 18a, b and c (12)
	2* each of item 19(40)
	*4 joining 1 and 3 & 4 and 2 with 9 and 10
	*4 joining 23 with 13 and 5
21	Purpose-built steel angle brackets to secure 45 degree support struts
22	Aluminium angle support struts (420*24*12)
23	Purpose built steel angle bracket supports
24	M6 machine screws
	*4 connecting each load cell to aluminium head support and mild steel fittings (16)

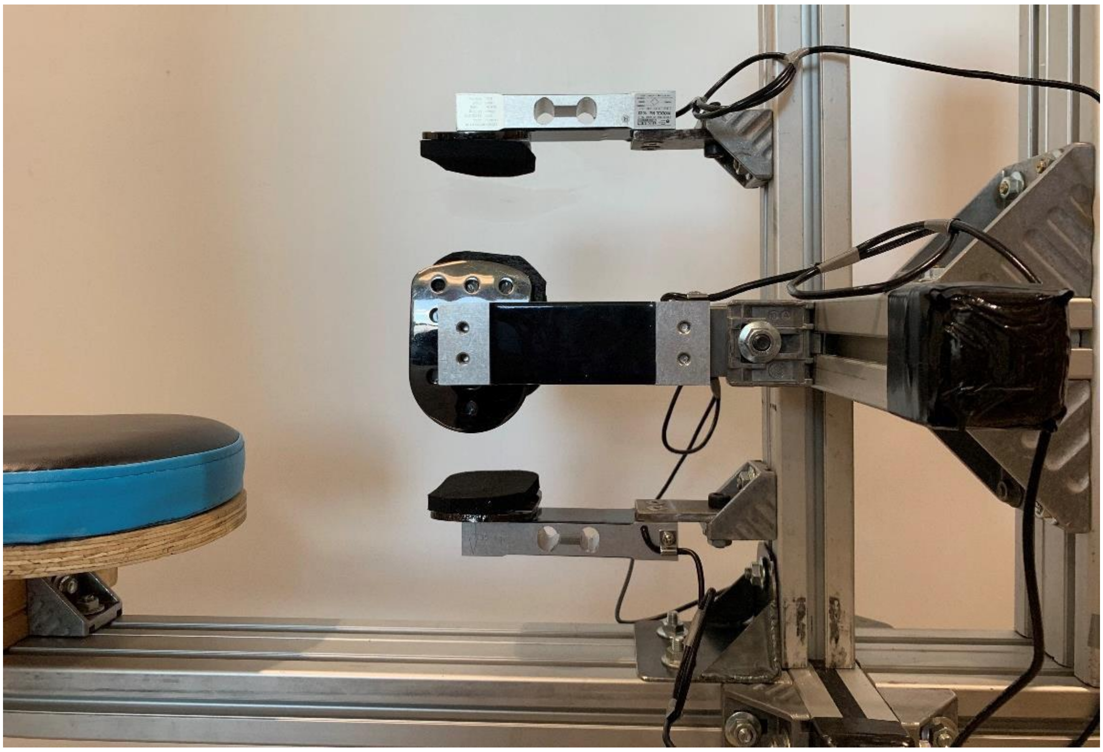

Figure 3: Side view of the load cells fixed to the head piece frame with brackets. Neoprene foam pads are visible on the inside of the aluminium platforms where force is applied

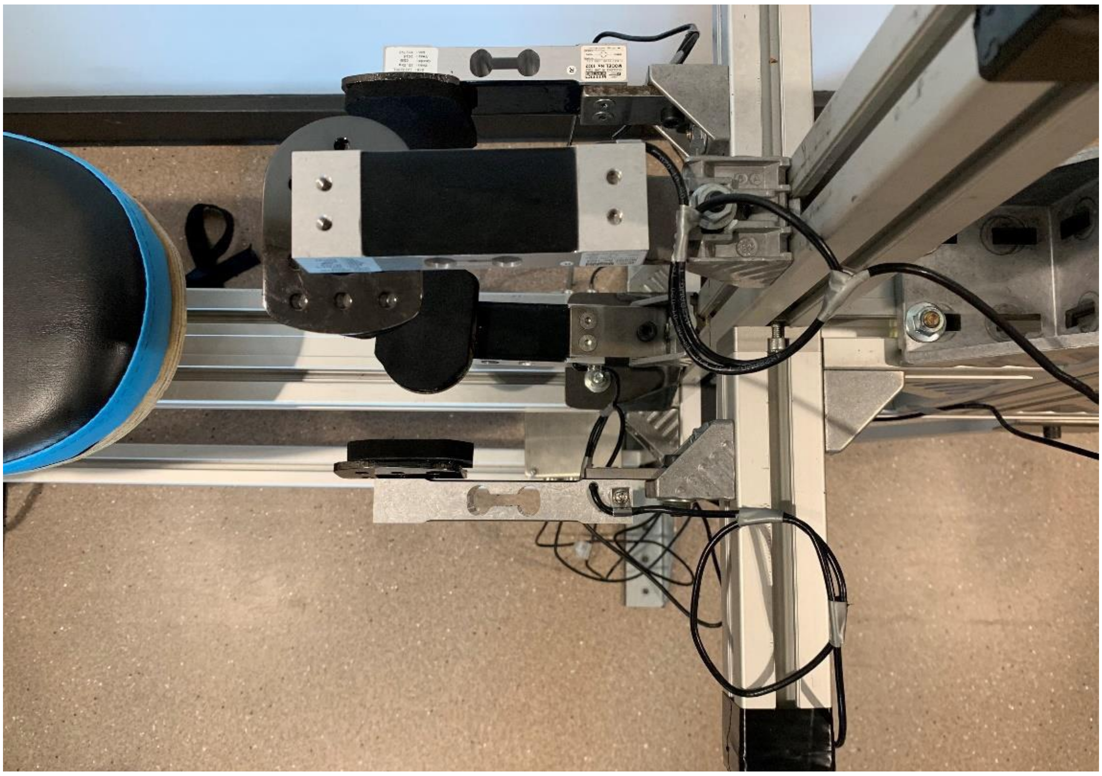

Figure 4: Top view of the head piece, showing the brackets used to fix load cells to the head piece

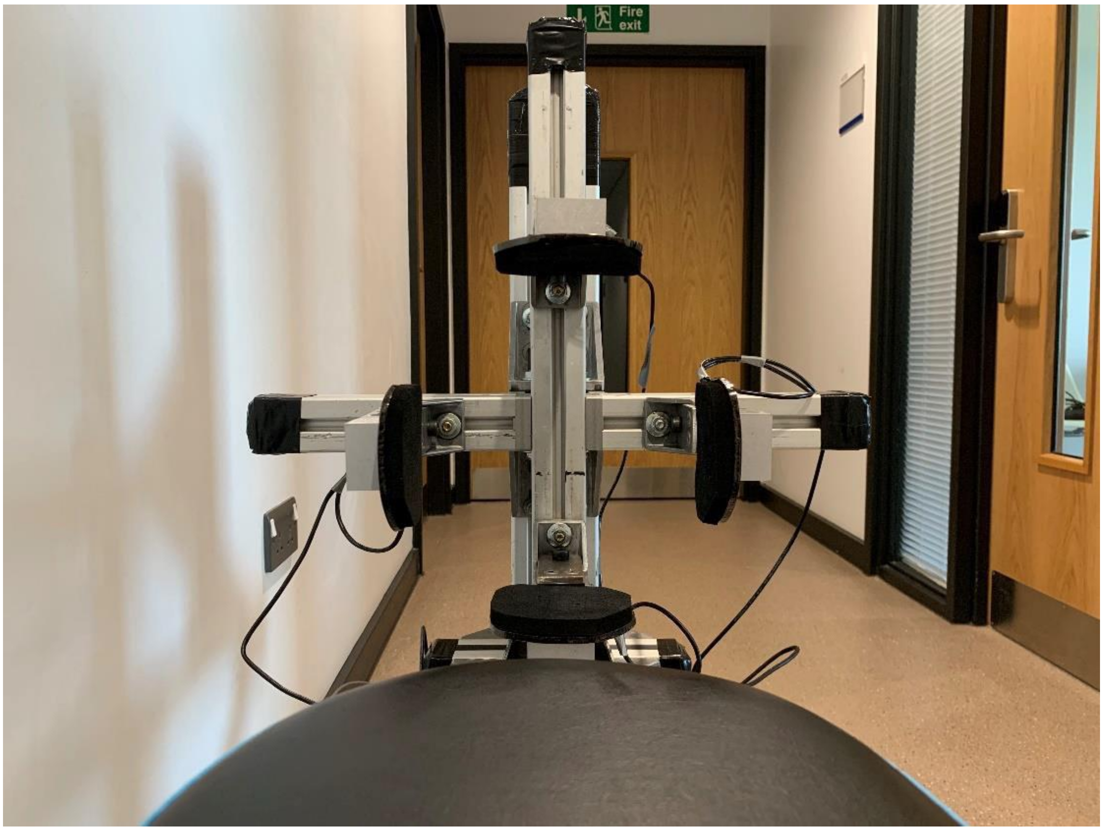

Figure 5: End-on view showing the head piece with load cells, also visible is the horizontal bench where the participant's torso will be strapped down

Headset Specifications and Technical Data

The four Tedea-Huntleigh Load Cells were positioned as per Figures 3, 4 and 5, so that when the participant's head is positioned as per Figure 1, neck flexion, extension and lateral flexion can be measured. Each load cell is mounted to the Rexroth frame using Rexroth brackets, the technical data for these is provided in Figure 7. The angle of force applied to these brackets via the load cells is consistent with the third position shown in Figure 7, which can withstand 160 Nm. Figure 6 shows that the moment arm in question is 0.16 m long and as stated above, the maximum expected force is 490 N. There expected maximum load on these brackets is therefore 78.4 Nm. The capacity of these brackets is more than double what the maximum expected load.

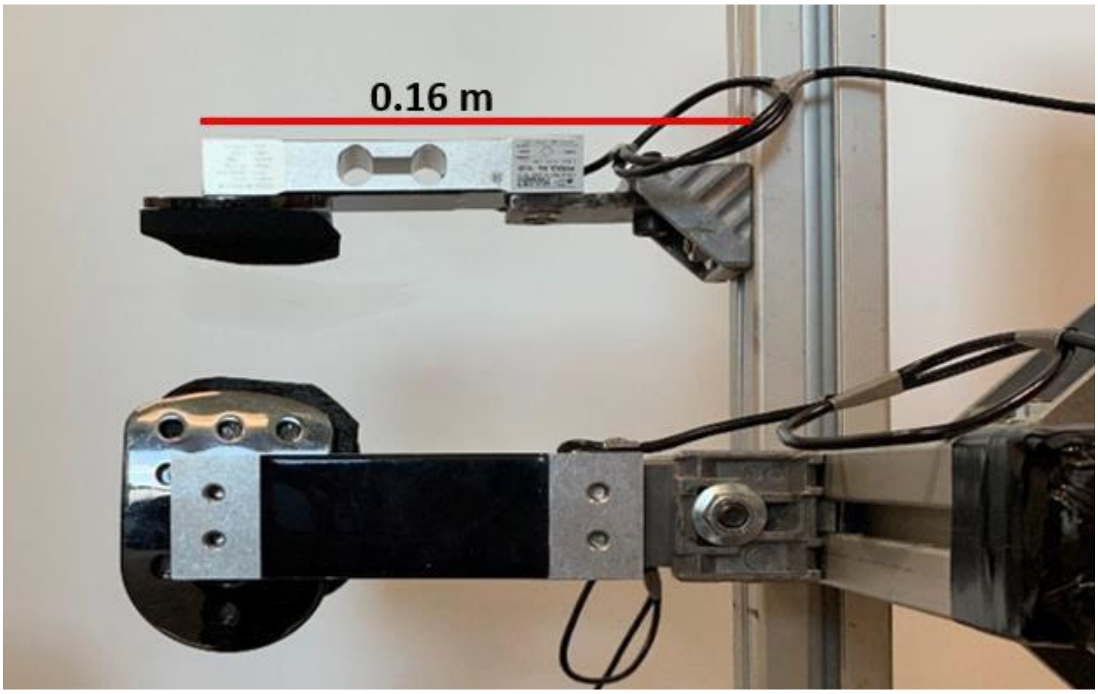

Figure 6: Distance from bracket mount to distal end of load cell where force is applied

Technical data

	Groove	ESD	Material entry	F_{max}	M_{max}	M_{max}
				N	Nm	Nm
Bracket	10		Bracket: Diecast aluminum, vibratory ground Fastening material: steel; galvanized	3000	60	160
	10		Bracket *designLINE*: Diecast aluminum; vibratory ground, painted (RAL 9006) Fastening material: steel; galvanized	3000	60	160
	10		Bracket: Diecast aluminum, vibratory ground	3000	60	160

Figure 7: Technical data for the brackets used to fix load cells to extrusion

The load cells are mounted to the brackets using 35*6mm, 66mm lengths of mild steel, machined for this purpose. The mechanical properties of mild steel can be found here:

https://www.azom.com/article.aspx?ArticleID=6115

Importantly, the ultimate tensile strength of mild steel is 400 MPa and the yield tensile strength is 370 MPa (200-300 kg). Given the loads to be applied to this apparatus, this is well over-engineered.

The load cells themselves have a rated capacity of 35kg, a safe overload capacity of 150% of this rated capacity, maximum overload 200% and ultimate overload 300% (so ultimate overload being 105 kg). This data is available here: https://www.loadcells.com/products/load-cell-1022/

The ultimate overload of these load cells is more than double the expected maximum load to be applied to each load cell.

Safety of Electronic Components

A Type B 12V power supply is required to power the load cell amplifiers. Electronics engineer Mr David Moody (Swansea University) has checked all electronic components and wiring and has considered them safe. An email from Mr Moody states “I can confirm that the rig is electrically safe as the load cells are low voltage and correctly connected to a low powered amplifier powered by a class 2 device. This Class 2 device will need the usual insulation resistance test in a PAT test as it’s a plug-in power supply, but this is carried our annually by a contractor for estates”. It has been registered online with states to be added to the annual PAT testing list.

References

Salmon, D. M. (2014). *An Examination of Neck Strength , Endurance , Neck Pain and Neck Stiffness in Rugby Union Players*. PhD Thesis, University of Otago, New Zealand

Appendix B: Complete 17-Week Neck Specific Training Programme.

Weeks 1-2 (2 sessions/wk)	
Session 1-4, DNS stage 1, 50° incline, 10x10s, 10s rest. Progression – lower incline 10° each session	
Week 3 (2 sessions/wk)	
Session 1 *Exercise/Movement:* DNS stage 2. *Resistance:* 0° incline. *Reps:* 10x10s. *Rest:* 10s rest	***Session 2*** *Exercise/Movement:* DNS stage 3, Prone cervical protraction. Resistance: 0° incline. *Reps:* 10x10s. *Rest:* 10s rest
Weeks 4-8 (2 sessions/wk)	
Exercise/Movement: Isometric holds in Flx, Ext, Lflx & Rflx. *Resistance:* 60% MVC. *Reps:* 3x15s. *Rest:* 15s rest. *Progression:* increase hold 5s every two weeks. Once reach 30s, increase resistance 5%.	
Weeks 9-17 (2 sessions/wk)	
Session 1 *Exercise/Movement: Isometric holds* in Flx, Ext, Lflx & Rflx. *Resistance:* 60% MVC. *Reps:* 3x15s. *Rest:* 15s. *Progression:* increase hold 5s every two weeks, after 30s increase resistance 5%.	**Session 2** *Exercise/Movement:* Dynamic (Eccentric/Concentric) in Flx, Ext, Lflx & Rflx. *Resistance:* 30% MVC. *Reps:* 3x10. *Tempo:* 2:1:2. *Rest:* 60s. *Progression:* if 10 reps completed increase resistance by 5%.

Note, DNS = deep neck stabiliser, Flx = flexion, Ext = extension, Lflx = left-lateral-flexion, Rflx = right-lateral-flexion, MVC = maximum voluntary isometric contraction, Reps = Repetitions

Appendix C: Correlation Table for Neck Strength and Head Acceleration Across Event Types

Appendix C: Pearson's correlation coefficients and associated p-values for relationships between average maximum isometric voluntary contraction (MVC) in extension (Ext), flexion (Flx) and left- (Lflx) and right-lateral-flexion (Rflx) and peak linear (PLA) and rotational acceleration (PRA) experienced in a tackle, ruck and carry.

			MVC (N)				
Event type			**Ext**	**Flx**	**Lflx**	**Rflx**	**Total**
Tackle	**PLA**	*r*	-0.23	0.00	-0.06	-0.16	-0.16
		p	0.48	0.99	0.85	0.61	0.60
	PRA	*r*	-0.50	-0.37	-0.33	-0.5	-0.58*
		p	0.08	0.21	0.27	0.08	0.04
Ruck	**PLA**	*r*	-0.30	0.01	-0.37	-0.45	-0.35
		p	0.35	0.97	0.24	0.14	0.27
	PRA	*r*	-0.64	0.01	-0.4	-0.45	-0.51
		p	0.03*	0.99	0.20	0.15	0.09
Carry	**PLA**	*r*	-0.58	-0.21	-0.26	-0.30	-0.47
		p	0.06	0.53	0.45	0.37	0.14
	PRA	*r*	-0.61*	-0.31	-0.27	-0.26	-0.50
		P	0.04	0.35	0.43	0.44	0.11

Note: * indicates $p < 0.05$.

Appendix D: Correlation Table for Neck Strength and Head Acceleration Across Causes of Acceleration

Appendix D: Pearson's correlation coefficients and associated p-values for relationships between average maximum isometric voluntary contraction (MVC) in extension (Ext), flexion (Flx) and left- (Lflx) and right-lateral-flexion (Rflx) and peak linear (PLA) and rotational acceleration (PRA) as a result of direct contact to hard body parts, soft body parts, the ground and indirect contact.

			MVC (N)				
Cause of acceleration			**Ext**	**Flx**	**Lflx**	**Rflx**	**Total**
Hard	**PLA**	*r*	-0.45	-0.07	-0.27	-0.25	-0.36
		p	0.14	0.83	0.40	0.44	0.25
	PRA	*r*	-0.69*	-0.35	-0.38	-0.23	-0.58*
		p	0.01	0.27	0.22	0.47	0.04
Soft	**PLA**	*r*	-0.70	-0.06	-0.18	-0.25	-0.10
		p	0.86	0.88	0.65	0.52	0.80
	PRA	*r*	-0.24	-0.09	-0.09	-0.20	-0.00
		p	0.54	0.82	0.82	0.61	0.99
Indirect	**PLA**	*r*	-0.17	-0.06	-0.36	-0.46	-0.32
		p	0.59	0.84	0.24	0.13	0.32
	PRA	*r*	-0.19	-0.03	-0.35	-0.46	-0.31
		p	0.55	0.93	0.27	0.13	0.32
Ground	**PLA**	*r*	-0.38	-0.8	-0.16	-0.22	-0.30
		p	0.25	0.82	0.64	0.51	0.40
	PRA	*r*	-0.43	-0.35	-0.11	-0.26	-0.41
		p	0.12	0.29	0.74	0.43	0.22

Note: * indicates $p < 0.05$.

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