An assessment of one-on-one inertial shoulder tackles in rugby league using instrumented mouthguards and qualitative video analysis.

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The candidate confirms that the work submitted is his own and that appropriate credit has been given where reference has been made to the work of others.

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Abstract

In the recent past, head injuries in rugby union and league have been a cause for concern. Making use of instrumented mouthguards enables the recording of head acceleration events (HAE) experienced by athletes in real time. Increased magnitude and frequency of HAEs increases the risk of head injuries such as concussion. The aim of the study was to examine the effect different tackle types have on peak linear acceleration (PLA), peak angular acceleration (PAA) and peak angular velocity (PAV) during one-on-one inertial shoulder tackle HAE.

The study involved the analysis of 80 tackles on male Leeds Rhinos players collected over 27 Super League matches. PLA, PAA and PAV of the head were recorded using the instrumented mouthguards with a 5 g or 500 rad/s^2 triggering threshold, which is lower than the recommended threshold of 10 g. Qualitative video analysis was done to classify the tackles based on tackle height, tackler and ball carrier speed, tackler and ball carrier body position, tackle direction and whether the tackle was active or passive.

Multiple linear regression was done to assess whether the different tackle types influenced PLA, PAA or PAV. Tackler speed and tackle height were determined to be significant predictors of PAV ($p < 0.05$). An increase in tackle height ($B= 2.1$, $p < 0.05$) and tackler speed ($B= 1.08$, $p< 0.05$) led to increased PAV. None of the other independent variables significantly predicted PLA, PAA or PAV. Majority of the impacts were below the recommended triggering threshold of 10 g.

In conclusion, this study found that an increase in tackler speed and tackle height led to an increase in PAV. In addition to this, the study highlights the efficacy of using a combination of qualitative video analysis and instrumented mouthguards to quantify and categorise HAE in rugby league.

Abbreviations

Abbreviation	Phrase
ATD	Anthropometric Test Dummies
CCC	Concordance Correlation Coefficient
Centre of gravity	CG
CMDs	Common Mental Disorders
CSDM	Cumulative Strain Damage Measure
CTE	Chronic Traumatic Encephalopathy
DAI	Diffuse Axonal Injury
DOF	Degrees of Freedom
GFT	gForce Tracker
HAE	Head Acceleration Event
HIT	Head Impact Telemetry
IMU	Inertial Measurement Unit
MADYMO	Mathematical Dynamical MOdels
NRL	National Rugby League
PAA	Peak Angular Acceleration
PAV	Peak Angular Velocity
PLA	Peak Linear Acceleration
RUVAC	Rugby Union Video Analysis Consensus
TBI	Traumatic Brain Injury
WSTC	Wayne State Tolerance Curve

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1. INTRODUCTION

Rugby league is a popular contact sport played throughout the world. It is most popular in Australia and the United Kingdom (Gardner et al, 2015a). Given its physical nature, participating in rugby league carries an inherent risk for injury (Gardner et al, 2015a). Concussion is a common injury in rugby league with between 8 and 17 concussions sustained per 1,000 playing hours (Gardner et al, 2015b). Concussions are caused by acceleration or deceleration of the head because of linear or rotational forces (Gardner et al, 2015a). Approximately 90% of concussions in sport occur without loss of consciousness which makes detecting concussions difficult (Guskiewicz et al, 2000).

There is currently no way of detecting concussions during a live game therefore the magnitude of head acceleration events (HAE) is measured instead (Patton, 2016). Increased magnitude or frequency of HAE increases the risk of concussion (Stemper et al, 2019). By identifying what aspects of the game increase the magnitude or frequency of HAE, more research can be conducted to identify how to modify these aspects of the game.

Wearable head kinematic devices can be used in the measurement of HAE (Ng et al, 2006). These devices include helmets, instrumented mouthguards, headbands and skin patches. They make use of accelerometers and gyroscopes which measure the angular and linear velocity and acceleration at the head’s centre of gravity (Tierney, 2021). This study used instrumented mouthguards to record the HAE experienced during one-on-one inertial shoulder tackles in professional male rugby league matches.

Research has been conducted on direct head impacts in rugby union and league, however there is limited research on inertial impacts. Inertial impacts involve the rapid acceleration or deceleration of the body following an impact resulting in head movement by inertia (Smith et al, 2012). The HAE experienced during these tackles will be quantified and categorised using a combination of instrumented mouthguard data and qualitative video analysis. The aim of this study is to identify which tackle types affect PLA, PAA or PAV.

2. BACKGROUND LITERATURE

This section gives a background on concussion and its effects, and head acceleration events.

2.1 Concussion and sub-concussion

Concussion

Concussion is defined as a pathophysiological process which affects the brain and is induced by traumatic biomechanical forces (McCrory et al, 2013). Concussion represents low velocity injuries that cause the brain to shake causing clinical symptoms that are not related to a pathological injury (McCrory et al, 2013). It has been characterised as force transferred to the brain resulting in the disruptive stretching of neuronal cell membranes and axons which results in a series of metabolic and pathophysiologic events (Barkhoudarian, 2016).

Research has shown that the highest shear forces are transferred to the deep midbrain level of the head (Pellman et al, 2003). It has been assumed that the forces transferred to the mesencephalon and corpus callosum regions of the brain are responsible for concussion symptoms (Pellman et al, 2003). Concussion has been found to be the most common head injury sustained throughout the world, with over half of these concussions being sport-related (Patton et al, 2013).

Concussions have been linked with a few complications such as early-onset Alzheimer's disease, depression, dementia and chronic traumatic encephalopathy (Breedlove et al, 2012). Sports related concussion has also been found to be a leading cause of disability among the youth which indicates how severe the effects can be (Seifert, 2013).

Concussions are a functional rather than structural injury (Patton et al, 2013). This makes it difficult to diagnose concussion due to the lack of objective tests, reliance on self-reported symptoms and symptoms that can be caused by other common conditions (Clugston et al, 2019). Symptoms associated with concussion include headaches, mood changes, dizziness and fatigue (Clugston et al, 2019). Given that the symptoms need to be self-reported, some athletes may give false reports to avoid losing playing time (Sahler and Greenwald, 2012).

Sub-concussion

Sub-concussive head impacts have been defined as head impacts which do not result in symptoms typically experienced after concussion. For example, the loss of consciousness, amnesia, confusion, and headache (Merchant-Borna et al, 2016). Sub-concussive impacts can be caused following inertial head loading because of the rapid acceleration or deceleration of the body which has been defined as the slosh phenomenon (Smith et al, 2012).

There is a challenge in defining the lower and upper bounds of sub-concussive impacts. The upper boundary of sub-concussive impacts is dependent on the accuracy in the detection of concussion (Tooby, 2021). The lower boundary has also not been established but studies typically exclude impacts lower than 10g to avoid HAE caused by non-contact events (Ng et al, 2006).

2.2 Long term effects of concussive and sub-concussive impacts

Sub-concussion

Accumulated sub-concussive impacts to the head can lead to neurodegenerative brain disorders in the long term (Miller et al, 2020). A study reported neurophysiological changes, such as decreased frontal lobe activation, in high school football players despite having no observable symptoms of concussion (Talavage et al, 2014). Similarly, three ex-NFL players developed chronic traumatic encephalopathy (CTE) without any history of concussion (Omalu et al, 2010). CTE is one of the severe results of traumatic brain injury (TBI) that some retired athletes from contact sports suffer.

It is defined as trauma to the brain which is either repetitive, occurring in episodes or a single event which results in the progressive degeneration of the brain (Tharmaratnam et al, 2018). The primary risk factor for CTE is repetitive trauma to the head (Asken et al, 2017). As of now, autopsy examination remains the only way to diagnose CTE (Stewart et al, 2016). Evidence has shown that repetitive concussive and sub-concussive trauma can both have lasting effects on brain function and can lead to its neurodegeneration (Breedlove et al, 2012).

In both soccer and boxing, sub-concussive impacts have been associated with acute changes in brain function, white matter changes, biomarkers of neuronal injury and short-term cognitive changes (Tierney et al, 2018). In addition to this, they have also been associated with long term white matter changes, lower brain volume and long-term cognitive defects (Tierney et al, 2018).

Concussion

Like sub-concussive impacts, repeated concussion is associated with development of neurodegenerative diseases such as CTE (Stewart et al, 2016). A study revealed that retired American footballers who reported having experienced one or two concussions in their career were 1.5 times more likely to report depression than those with no concussion history.

Those who experienced three or more sports related concussions were three times more likely to report concussion (Guskiewicz et al, 2007).

Experiencing three or more concussions has been found to have a detrimental effect on cognitive function (Gardner et al, 2010). A study analysing 73 male rugby union players found that those who had reported three or more concussions performed significantly worse in cognitive tests and had slower processing times than those who had not reported any concussions (Gardner et al, 2010).

Former athletes who had reported four or five concussions were approximately 1.5 times more likely to experience common mental disorders (CMDs) than those who had reported no concussions (Gouttebarge et al, 2017). Some of the CMDs include symptoms of distress, anxiety and depression, sleep disturbance and substance abuse/dependency (Gouttebarge et al, 2016).

2.3 Head acceleration events

Head acceleration event (HAE) exposure is a term used when measuring the head kinematics experienced in sport (Beckwith et al, 2013). It can occur directly from head contact (Direct HAE), indirect through body impact (indirect HAE) or from voluntary movement (voluntary HAE) (Bailes et al, 2013). HAE kinematics can be recorded using wearable head sensors, mouthguards or skin patches that are instrumented with accelerometers and/or gyroscopes (Wu et al, 2016).

Typically, HAE with a magnitude of less than 10 g are excluded from analysis (King et al, 2016). This is because non-impact activities such as running, walking and jumping can contribute to small HAE which trigger false-positive events (Ng et al, 2006). A false-positive event is when an impact is triggered but there is no HAE occurring in the video at the stated time whereas a true positive event is when an event is triggered, and a HAE occurs at the stated time on the video. When analysing HAEs, all the triggered events need to be cross-referenced with video-footage to determine whether they are true positive or false-positive events (Tierney et al, 2021).

Some studies have discovered that increased HAEs have the potential to reduce an athlete's tolerance to head injury therefore making them more susceptible to concussion from head impacts of a lower magnitude (Stemper et al, 2019). Stemper et al found that 72% of concussed athletes in their study had the most or second most HAEs when compared with the control group. It is therefore important to monitor HAEs to reduce the potential injury risks. It has however been found that the HAE magnitudes are dependent on the sensor used for measurement as helmet and headband-based sensors had higher magnitudes than biteplate-based sensors (Miller et al, 2020).

3. LITERATURE REVIEW

In this section is a literature review of the methods used to measure head acceleration events and studies analysing concussion and head acceleration events in various sports. The method used in this study is a combination of qualitative video analysis and instrumented mouthguards. However, a background on other instrumented head kinematic devices has also been discussed.

3.1 Methodological approaches to concussion injury biomechanics research

Several approaches have been used to investigate the mechanism of concussion. These approaches have been split into observational approaches and biomechanical approaches. An example of an observational approach is video analysis. A biomechanical approach is the use of instrumented wearable devices.

3.1.1 Video analysis

General video analysis

General video analysis gives analysts an opportunity to distinguish between injury and non-injurious events and be able to identify the potential risk factors for injury by making use of video footage (Meeuwisse, 2009). Video analysis has been used to analyse concussions in both rugby league and hockey (Gardner et al, 2015b) (Hutchison et al, 2013).

Two-dimensional kinematic video analysis

Video footage from a single camera view can be calibrated to provide one or two-dimensional kinematic estimates of the athlete's motion (Frechede and McIntosh, 2009). However, movements that are out of the frame cannot be accounted for (Frechede and

McIntosh, 2009). A study made use of two-dimensional kinematic video analysis to investigate the dynamics of concussive head impacts in rugby and Australian football (McIntosh et al, 2000).

They used video analysis to estimate the closing speed and change in head linear velocity. This was done by using the dimensions of the field and calculating the time taken to cover the distances. For two-dimensional video analysis to be more accurate, it requires the movement to occur on a calibrated plane perpendicular to the camera (McIntosh et al, 2000). When measuring speed using two-dimensional video analysis, parallax error can only be minimised but cannot be eliminated.

3.1.2 Instrumented wearable devices

Inertial measurement units

Wearable devices are typically instrumented with devices used to measure kinematics referred to as inertial measurement units (IMUs).

Accelerometer

Accelerometers are used to measure the head kinematics during impacts. To be specific, they measure the impact acceleration to the head to derive PAA or PLA (Higgins et al, 2007). Instrumented mouthguards use tri-axial accelerometers which measure impact acceleration in 3 axes: anterior-posterior (x), inferior-superior (y) and medio-lateral (z) (Higgins et al, 2007). Use of accelerometers has been found to be useful when characterising the kinematic load to the head. It is however not useful when diagnosing concussion. This is because of the lack of an identified concussion injury threshold (Brennan et al, 2017).

Gyroscopes

Gyroscopes are devices that are used to measure changes in angular velocity (Passaro et al, 2017). Gyroscopes are typically used together with accelerometers when creating head-impact devices. For example, the X-Patch designed by X2 Biosystems makes use of a gyroscope and 3-Axis accelerometer to record angular velocity and linear and angular acceleration (O'Connor et al, 2017). A study attempting to quantify postural stability made use of tri-axial gyroscope as it was able to measure rotation of the body which is particularly useful in less planar movements (Alberts et al, 2015).

Microcontrollers

Microcontrollers are devices attached to instrumented mouthguards with the purpose of transferring digital data from the off-chip analog-to-digital converter to the off-chip non-volatile memory buffer (Lund et al, 2018).

Proximity sensors

Instrumented mouthguards are typically fitted with infrared proximity sensors. The purpose of these sensors is to detect the presence of the teeth within the mouthguard (Wu et al, 2014). This is used to filter out readings where the mouthguard is not worn on the teeth (Wu et al, 2014).

The following are the instrumented wearable devices that measure head kinematics:

Helmet-based system

Helmet-based instruments using accelerometers have been used to monitor the biomechanics of head impacts in high-impact sports (Allison, 2015). The most widely used helmet-base system in research is the head impact telemetry (HIT) system which has been used in ice hockey and American football (Stemper et al, 2019b).

The HIT system is comprised of six spring-loaded single axis accelerometers attached to American Football helmets (O'Connor et al, 2017). The parameters recorded by the HIT system include PLA and PAA of the head's centre of gravity, HIC and the azimuth and elevation of the impact site (Siegmund et al, 2016). An impact is recorded when 1 of the 6 accelerometers exceeds 14.4g, however this threshold can be adjusted (O'Connor et al, 2017). Linear acceleration is recorded over 40ms at 1,000 Hz (Siegmund et al, 2016). The system records the 8ms before impact and the 32 ms after impact (O'Connor et al, 2017).

A second helmet-based system used in research is the gForce Tracker (GFT) sensor system. The system consists of a tri-axial accelerometer and a tri-axial gyroscope which are attached to a helmet (Allison, 2015). The impact data is recorded for 40ms at 3,000 Hz for linear acceleration data and 760 Hz for angular acceleration data (Allison, 2015). If the head acceleration exceeds 40 ms, the system will continue recording in intervals of 40ms until the acceleration goes below the set threshold (Allison, 2015).

A validation study on the HIT system was done which involved head impact reconstructions using a test dummy and a linear impactor striking at speeds of 3.6 to 11.2 m/s (Siegmund et al, 2016). The average absolute relative error for peak linear and angular acceleration was 32% and 35% respectively (Siegmund et al 2016). Helmets have been found to move relative to the head during impacts. A study found that helmets translated 12 to 41 mm and rotated to 37 degrees relative to the head (Joodaki et al, 2019). The peak linear acceleration of the helmet was 2 to 5 times more than the head's (Joodaki et al, 2019). This indicates that the kinematic values recorded by helmet-based systems are recording the helmet kinematics rather than the head kinematics (Joodaki et al, 2019).

Reconstructing impacts using test dummies and helmets leads to inaccuracies due to variations in the fit of the helmet. The friction between a test dummy and a helmet is significantly higher than a real human head and a helmet (Joodaki et al, 2019). Lower friction between the head and helmet lowers the rotational acceleration experienced by the head during a collision (Trotta et al, 2018). This means that using test dummies would overestimate the head kinematics experienced. This could explain why helmet-based sensors record higher magnitudes than bite-plate sensors (Miller et al, 2020).

Instrumented mouthguards

Instrumented mouthguards are believed to be the most accurate wearable device because of the skull coupling (Wu et al, 2016). The mouthguard couples to the upper dentition which is anchored in the upper jaw by stiff ligaments (Kuo et al, 2016). Mouthguards are usually either custom-fit to the teeth following a dental scan or boil-and-bite (Kieffer et al, 2020). Instrumented mouthguards are fitted with a tri-axial accelerometer and a tri-axial gyroscope (Kieffer et al, 2020).

The sampling rates used vary depending on the companies. Mouthguards by Prevent Biometrics (Prevent Biometrics, Minneapolis MN) cused a sampling rate of 3,200 Hz and record data from 10ms before the impact to 40ms after. Prevent mouthguards make use of an algorithm to filter out false-positive events (Kieffer et al, 2020). Vector mouthguards use a sampling rate of 1,024 Hz and record from 16ms before impact to 80ms after (Kieffer et al, 2020). X2 X-Guard mouthguards use a sampling rate of 1,000 Hz for PLA and 850 Hz for PAA which is interpolated to 1,000 Hz. Data is recorded from 10 ms before impact to 90 seconds after the impact (O’Connor et al, 2017).

There has been inconsistency in studies assessing the validity of instrumented mouthguards. Some studies have found high accuracy (Camarillo et al, 2013) while another study reported poor accuracy in measuring head kinematics (Siegmund et al, 2016). In the study by Camarillo, the mandibles were fixed to clench the mouthguard while in Siegmund’s study a spring-articulated mandible was used. Constraining the mandible with springs can lead to whipping which results in an impact under the mouthguard known as mandible strike (Liu et al, 2020).

A study was done to evaluate the effect of the mandible on mouthguard accuracy (Kuo et al, 2016). The study found that when the mandible was unconstrained and not clenching the mouthguard, there was poor accuracy when recording angular acceleration. It is suspected that this could be because lateral impacts lead to higher angular acceleration, therefore impacts hitting the mandible from the side result in greater mandible motion (Kuo et al, 2016)

Instrumented mouthguards have challenges measuring high angular velocities due to gyroscope saturation (Liu et al, 2020). This occurs when the angular velocity exceeds the dynamic range of the gyroscope sensor (Dang and Suh, 2014). This saturation has been observed in angular velocities exceeding 35 rad/s in a single direction leading to variations in accuracy (Liu et al, 2020).

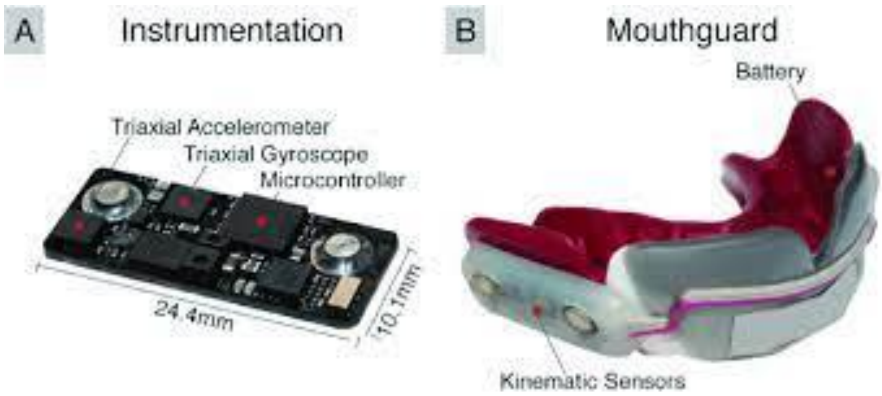


Figure 1. Instrumented mouthguard (Kuo et al, 2018)

<u>Headbands</u>

Headbands are head-mounted devices that are used for measuring head kinematics. The SIM-G (Triax Technologies, Inc., Norwalk, CT, USA) headband has been used in validation studies (Cummiskey et al, 2017). The SIM-G is composed of a tri-axial accelerometer and a gyroscope to measure linear acceleration and angular velocity respectively (Huber et al, 2021). It records over 62 ms at 1,000 Hz. Data is recorded from 10 ms before impact to 52ms after (Huber et al, 2021). It was found to be more accurate than HITS helmet mounted sensors as two headband devices recorded 20.6% and 7.9% error when recording PLA while the HITS devices recorded error of 29.5% and 14.1% respectively.

A study made use of headbands (SLICE Micro, Diversified Technical Systems, Inc., Novi, Michigan, California) to measure non-injurious head accelerations of young children (Bussone and Prange, 2014). The device contained linear accelerometers and angular rate sensors which measured linear and angular accelerations and angular velocity respectively (Bussone and Prange, 2014).

SIM-G headband sensors have been found to be highly reproducible as two devices had peak differences of 0.5% when subjected to the same impact (Huber et al, 2021). However, SIM-G was found to underestimate the peak angular velocity magnitude by 15% (Huber et al, 2021). More research is needed to investigate the skull coupling of headbands and false-positives to determine its accuracy (Huber et al, 2021)

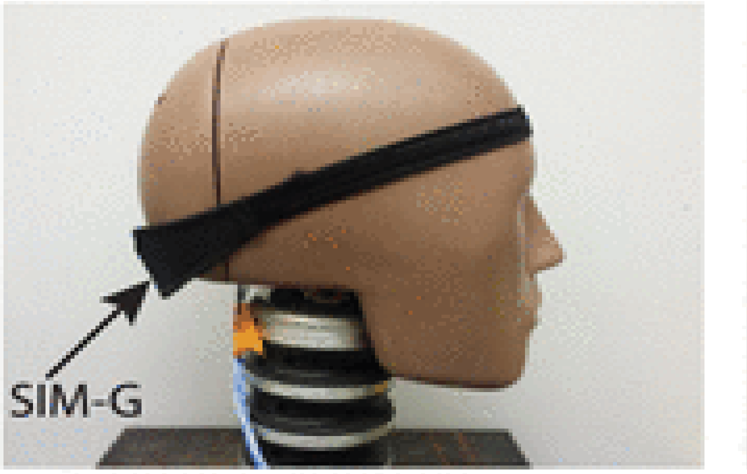


Figure 2. SIM-G Headband (Cummiskey et al, 2017).

Ear patches

Ear patch sensors are attached directly to the skin behind the right ear on the mastoid process by using a custom double-sided adhesive patch (Cummiskey et al, 2017). The xPatch sensor (X2 Biosystems Inc., Seattle, WA, USA) has been used in a few research studies. It is fitted with a tri-axial accelerometer and gyroscope (O'Connor et al, 2017). PLA is recorded at 1,000 Hz while PAA is recorded at 850 Hz. The outputted data includes PLA, PAA, HIC, impact location and direction of PLA. Recordings are made over 100ms with 10ms before impact and 90ms after (O'Connor et al, 2017).

The x-Patch sensor has been found to have measurement error of up to 50% for PLA and PAA (McCuen et al, 2015). A study was done testing the validity of an instrumented mouthguard, skin patch and headgear-mounted sensors when measuring impacts caused by heading a soccer ball. x-Patch sensors were found to overestimate PLA and PAA by 15 ± 7g and 2,500 ± rad/s^2. The over-predictions are suspected to be a result of out-of-plane motion as when compared to the mouthguard, the acceleration direction reported by the x-Patch was in a different vector direction from the instrumented mouthguards (Wu et al, 2016). In addition to this, skin patch devices have poor skull coupling because of movement of the skin the patch is attached to (Cummiskey et al, 2017).

The x-Patch has been reported to be accurate in measuring PLA, but it underestimated PAA by more than 25% and had a high number of false-positive events (Nevins et al, 2018). Due to the errors reposted, some researchers concluded that raw data from skin patch devices should not be used directly when studying injury risks (Wu et al, 2016).

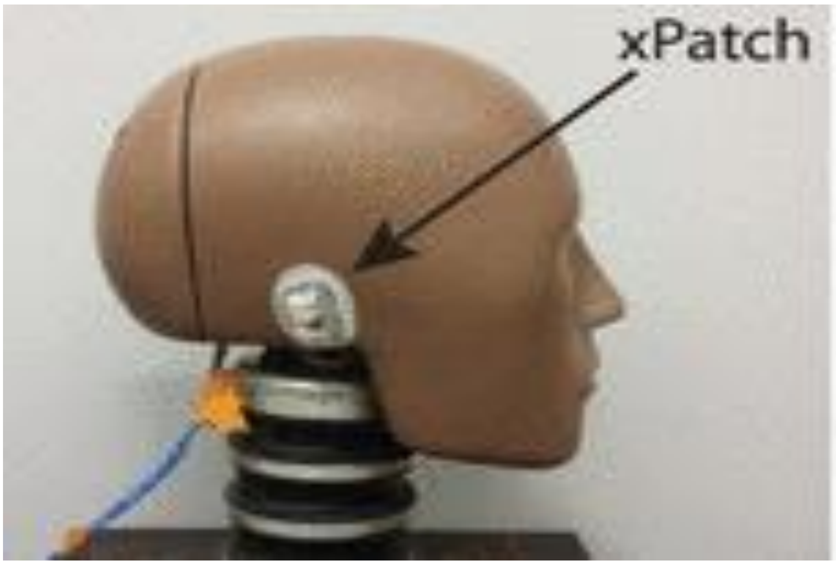


Figure 3. xPatch sensor (Cummiskey et al, 2017).

3.2 Mechanical metrics of concussion injury events

3.2.1 Linear acceleration

A few studies in American Football have found a link between likelihood of concussion and linear acceleration. One study replicated twenty-four head impact cases from NFL matches using Hybrid III Anthropometric test dummies (ATD) crash test dummies. This study found that the mean peak acceleration value for concussive impacts was approximately 98g (Newman et al, 2000). A study focusing on collegiate players found a much higher mean peak acceleration value of 151g (Funk et al, 2006). However, the studies used different methodologies as Newman's study made use of ATD reconstruction while Funk's study used instrumented helmets.

3.2.2 Angular acceleration

McIntosh et al. performed a study estimating head impact cases among Australian rules football players by making use of rigid body simulations using the MADYMO human facet model. Analysis found that angular acceleration values of 1,747 rad/s^2 and 2296 rad/s^2 led to a 50% and 75% likelihood of concussion (McIntosh et al, 2014).
A study was done investigating head impact exposure over a season of women's rugby league. By using skin patches, they found that PAA values experienced by backs were higher than forwards (King et al, 2018). A study involving a junior rugby league team wearing skin patches also concluded that backs experienced higher PAA values than forwards (King

et al, 2017a). In contrast, a study involving a senior amateur rugby league team wearing skin patches found that forwards experienced higher PAA values than backs (King et al, 2017b).

3.2.3 Changes in head linear and angular velocity

A study focusing on the dynamics of concussive head impacts in rugby and Australian rules football found the mean change in head linear velocity in 97 impacts to be 4m/s (McIntosh et al, 2000). In a study on American football, concussed players experienced a change in head linear velocity of 7.2 ± 1.8 m/s while the change for uninjured struck and striking players was 5 ± 1.1 m/s and 4 ± 1.2 m/s respectively (Pellman et al, 2003).
By making use of FE models to reconstruct impacts, researchers found that the maximum change in head angular velocity for concussive head impacts was 33 rad/s (Patton et al, 2012).

3.2.4 Head acceleration exposure in American football using wearable head sensors

Head acceleration exposure in American football has been explored by several researchers. Reynolds et al. analysed 20 collegiate players over an entire season and investigated whether the type of practice had an impact on the number of HAE (Reynolds et al, 2016). The practice types were helmet-only, shell practices, full-pad practice, and competitive games. Impacts were collected using the xPatch earpatch sensor using a 10g linear acceleration threshold. A total of 890 exposure events were recorded.
Majority of HAE were from competitive matches followed by full-pad practice while helmet-only has the fewest HAE (Reynolds et al, 2016). Using a similar threshold, Stemper et al. used instrumented helmets on 342 collegiate athletes to find whether there was a difference in number of HAE events before and after eliminating two practice sessions on the same day. The number of HAE events increased even after eliminating the two training sessions (Stemper et al, 2019).
Broglio et al. analysed 42 high school athletes over a season using instrumented helmets collecting data at a 14.4 g threshold. Athletes in the linemen position sustained the highest number of HAE. Majority of HAE occurred during matches, followed by contact training and then non-contact training. Reducing contact training to once a week would have reduced HAE by 18% and eliminating it would reduce HAE by 39% (Broglio et al, 2013).
An earlier study done by Broglio et al. also used instrumented helmets to measure impacts at a threshold of 15g. Similarly, it was found that impacts were more frequent and intense during competitive matches compared to training sessions (Broglio et al, 2009).

4. AIMS AND OBJECTIVES

By measuring and analysing the HAE, this study aims to investigate the effects of the different tackle types on PLA, PAA and PAV. This involves the analysis of one-on-one inertial shoulder tackles in rugby league by combining instrumented mouthguard data and qualitative video analysis. This will enable tackle types that result in a higher magnitude of PLA, PAA or PAV to be identified. By identifying these tackle types, sporting bodies can make interventions to reduce the incidence of these tackles.

Another aim of this study is to investigate the effect of using a lower triggering threshold for the instrumented mouthguards. The triggering threshold used in this study was 5g compared to the recommended 10g. Therefore, this study investigates the effects of using a lower triggering threshold for analysis of HAE experienced during inertial tackles.

5. METHODS

5.1 Participants

The study involved the analysis of 14 male athletes from Leeds Rhinos using instrumented mouthguards. 27 Super League matches from 2019 to 2021 were used for analysis. The athletes had a mean age of 23 (± 4.1) years, mean height of 184 (± 6.4) cm and a mean weight of 97.9 (± 8.7) kgs.

Ethical approval for the study was provided by the University Research Ethics Committee of the University of Leeds (BIOSCI 18-023). The athletes involved provided written consent for their inclusion in the study.

5.2 Video analysis and the analysis framework

Only one-on-one inertial shoulder tackles were analysed for this paper. A tackle is defined as an event where one or more defenders attempt to stop a ball carrier (Hendricks et al, 2020). All the tackles were inertial impacts meaning there was no direct impact to the head. Only shoulder tackles were included, these are tackles where the first point of contact with the ball carrier is the shoulder. Finally, only impacts to the ball carrier were analysed.

The impacts were analysed based on the tackle height, tackler and ball carrier speed, tackle direction, tackler, and ball carrier body position, and whether it was an active or passive tackle. A qualitative video analysis framework was developed based on the framework developed by the Rugby Union Video Analysis Consensus (RUVAC) group (Hendricks et al, 2020). Table 2 gives details on the analysis framework. One-on-one inertial shoulder tackles recorded by the instrumented mouthguards were classified based on this framework.

Broadcasted match footage was used for the analysis. Kinovea software was used for the video analysis to analyse the impacts frame by frame. This is useful as high-speed impacts could be slowed down for more accurate analysis. Identifying the impacts on video prevents any potential false-positives. The impacts recorded by the mouthguard were time-stamped and synchronised with the video data to confirm that the impacts occurred. The video analysis involved going through the time-stamped impacts from all the matches and identifying the tackles which fit the criteria. A Microsoft Excel spreadsheet was then used to record information on the analysed tackles.

5.3 Instrumentation

The players involved in this study were equipped with instrumented mouthguards provided by the company Prevent Biometrics (Prevent Biometrics, Minneapolis MN). An image is shown in figure 1 in section 3.1.2. Instrumented mouthguards were chosen over other wearable head kinematic devices as they are more accurate due to their skull coupling (Wu et al, 2016). The mouthguards are coupled to the skull through the upper dentition (Kuo et al, 2016). By making use of upper dentition impression moulds, the mouthguards were custom fitted for the players (Tierney et al, 2020).
These mouthguards are fitted with tri-axial accelerometers and gyroscopes which measure the linear and angular acceleration and the linear and angular velocity respectively at the head's centre of gravity (CG) (Tierney, 2021). Proximity sensors in the mouthguard detected the presence of the teeth in the mouthguard (Wu et al, 2014). Algorithms by Prevent Biometrics transformed the head kinematics measured to the head's CG (Tooby et al, 2022).

In a study investigating the accuracy of 8 instrumented wearable head devices, the mouthguard by Prevent Biometrics was found to be the most accurate (Kieffer et al, 2020). The study involved two phases. In phase one, laboratory impacts using a linear impactor on a dummy were done while phase two involved on-field video validation of HAE measured by these devices. Custom-fit Prevent mouthguards were found to have a concordance correlation coefficient (CCC) of 0.95 in phase one and a positive predictive value (PPV) of 96.4% in phase two during active playing minutes. Vector mouthguards had a CCC of 0.81 in phase one and a PPV of 85.7% in phase two during active playing minutes (Kieffer et al, 2020).

The tri-axial linear accelerometer and tri-axial gyroscope had measurement ranges set at ± 200 g and ± 35 rad/s respectively. The trigger threshold was set at 5 g whereby HAE were recorded when linear acceleration exceeded 5 g on any of the axes. The mouthguards recorded 10 ms prior to the impact and 40 ms after. HAE were processed by Prevent Biometrics.

The processing involves transforming the linear acceleration data to the head's centre of gravity and applying a 4-pole zero phase low-pass Butterworth filter to reduce the noise using a frequency of 400 Hz (Tooby, 2021). Given that poor adherence to the teeth can increase noise, Prevent Biometrics run the HAE through a machine learning model which determines whether the HAE has minimal noise (Class 0), moderate noise (class 1) or severe noise (class 2). HAE in class 1 and 2, are filtered again at frequencies of 100 Hz and 50 Hz to reduce the noise (Tooby, 2021).

5.5 Statistical analyses

To analyse the data, multiple linear regression analysis was done. This was used to assess the potential associations between the dependent, also called outcome, variables and independent, also called predictor, variables. Three separate multiple linear regression tests were done for each dependent variable which are PLA, PAA and PAV. These were analysed against the independent variables which were tackle height, tackler and ball carrier speed, tackle direction, tackler and ball carrier body position, and whether the tackle was active or passive. All the independent variables were analysed together against each dependent variable in three separate tests.

Analysis was done using IBM SPSS Statistics Version 23. To determine whether the model was a good fit for regression, the Shapiro-Wilks test for normality was done. This was done to reduce error from non-uniform data. PLA and PAA data were log-transformed as they had non-normal distributions. PAV data was normally distributed. As shown in figure 4, given that the p value of the Shapiro-Wilk test was greater than 0.05, the data is normally distributed.

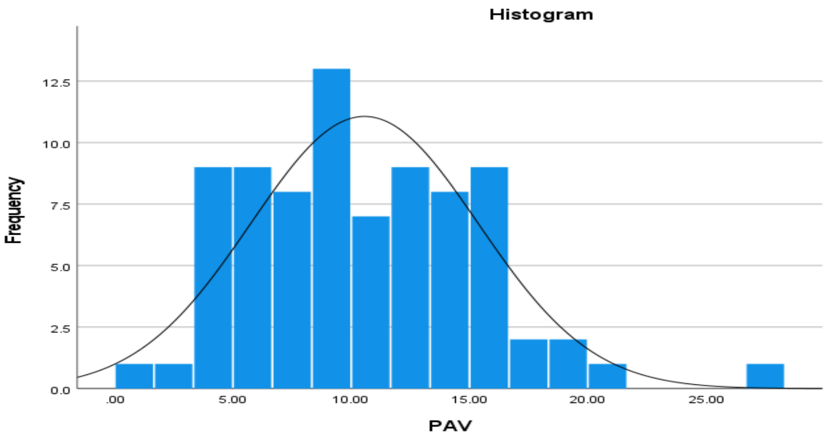
Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
PAV	.068	80	.200*	.970	80	.055
Log10PAA	.056	80	.200*	.989	80	.740
Log10PLA	.086	80	.200*	.974	80	.100

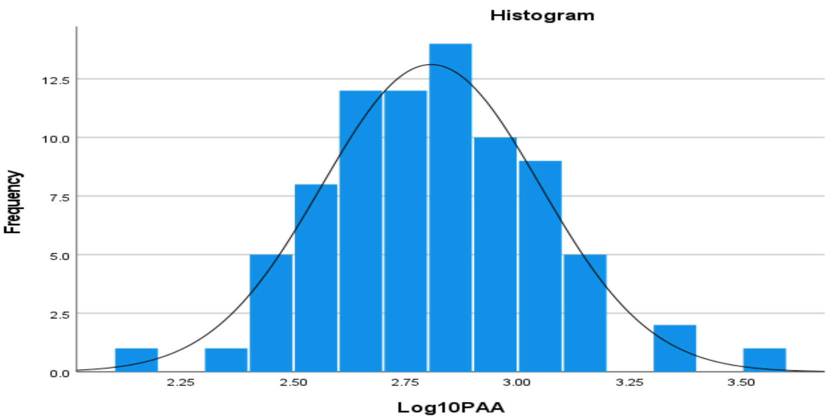
*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

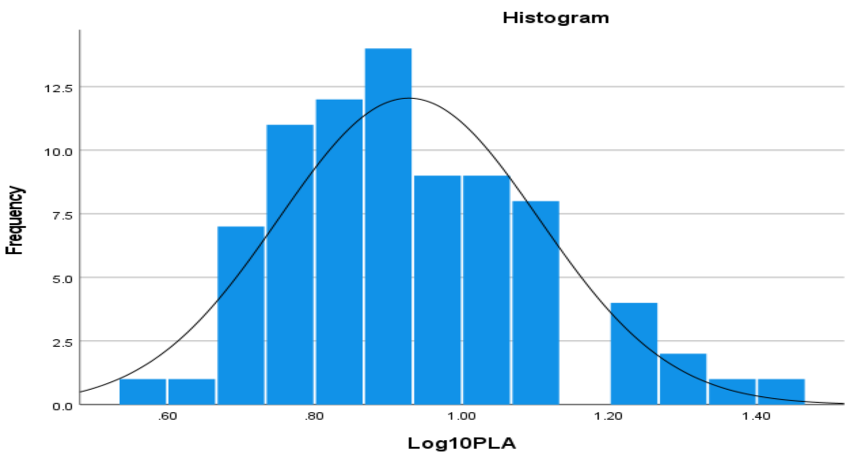
a)



b)



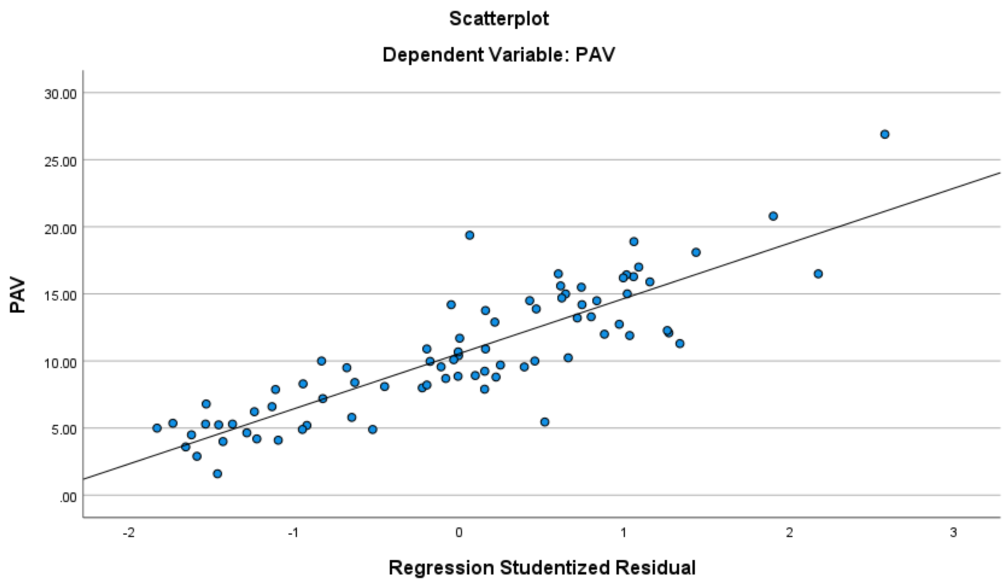
c)



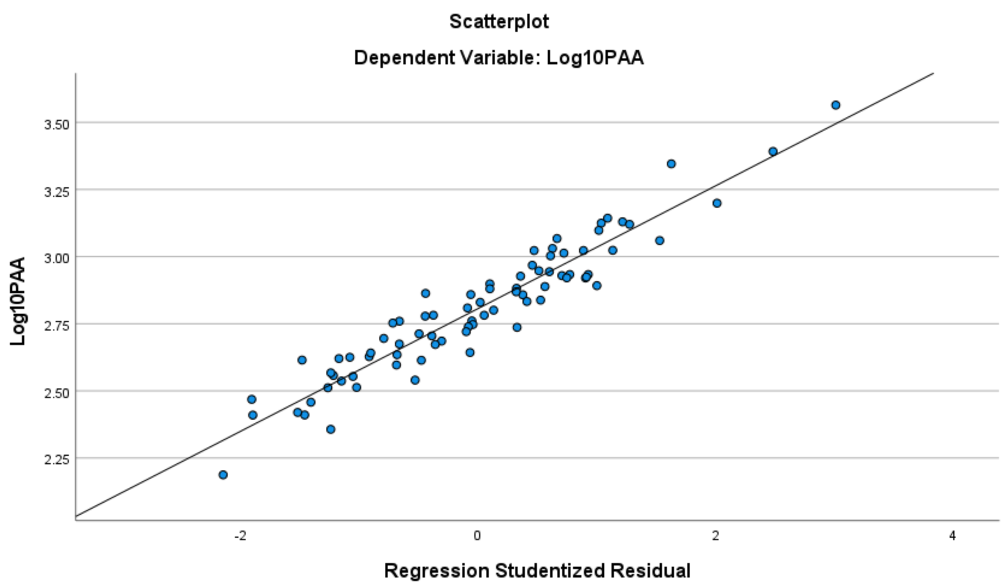
d)

Figure 4. Normality test results and histogram plots for the dependent variables.

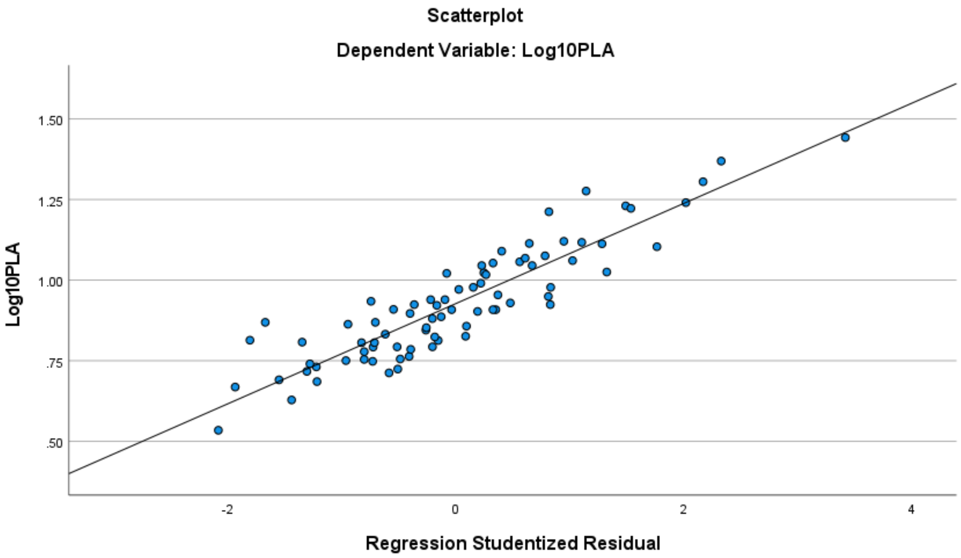
Homoscedasticity was tested using scatter plots as shown in figure 5. Given that the distance from the line of best fit was consistent, homoscedasticity was confirmed as the variance of the residuals was constant.



a)



b)



c)

Figure 5. Test for homoscedasticity by plotting the dependent variable against the standardized residuals.

Multicollinearity was confirmed to be absent by correlating the independent variables against each other. Variables were confirmed to be strongly correlated r was greater than 0.8. As shown in table 1, none of the independent variables showed multicollinearity.

Table 1. Tables showing results of the Pearson's correlation test. Multicollinearity was confirmed when values exceeded 0.8.

Correlations

		PAV	Tackler speed	Ball carrier speed	Tackle direction	Tackle height	Ball carrier body position	Tackler body position	Active or Passive
Pearson Correlation	PAV	1.000	-.259	-.078	-.094	-.172	-.058	-.045	-.213
	Tackler speed	-.259	1.000	-.011	-.083	-.167	-.056	.086	.338
	Ball carrier speed	-.078	-.011	1.000	-.123	.000	-.172	-.091	-.079
	Tackle direction	-.094	-.083	-.123	1.000	-.082	-.048	.076	.029
	Tackle height	-.172	-.167	.000	-.082	1.000	-.140	-.392	-.184
	Ball carrier body position	-.058	-.056	-.172	-.048	-.140	1.000	.238	.070
	Tackler body position	-.045	.086	-.091	.076	-.392	.238	1.000	.419
	Active or Passive	-.213	.338	-.079	.029	-.184	.070	.419	1.000

Correlations

		Log10PAA	Tackler speed	Ball carrier speed	Tackle direction	Tackle height	Ball carrier body position	Tackler body position	Active or Passive
Pearson Correlation	Log10PAA	1.000	-.083	-.044	-.141	-.125	-.038	-.146	-.088
	Tackler speed	-.083	1.000	-.011	-.083	-.167	-.056	.086	.338
	Ball carrier speed	-.044	-.011	1.000	-.123	.000	-.172	-.091	-.079
	Tackle direction	-.141	-.083	-.123	1.000	-.082	-.048	.076	.029
	Tackle height	-.125	-.167	.000	-.082	1.000	-.140	-.392	-.184
	Ball carrier body position	-.038	-.056	-.172	-.048	-.140	1.000	.238	.070
	Tackler body position	-.146	.086	-.091	.076	-.392	.238	1.000	.419
	Active or Passive	-.088	.338	-.079	.029	-.184	.070	.419	1.000

Correlations

		PAV	Tackler speed	Ball carrier speed	Tackle direction	Tackle height	Ball carrier body position	Tackler body position	Active or Passive
Pearson Correlation	PAV	1.000	-.259	-.078	-.094	-.172	-.058	-.045	-.213
	Tackler speed	-.259	1.000	-.011	-.083	-.167	-.056	.086	.338
	Ball carrier speed	-.078	-.011	1.000	-.123	.000	-.172	-.091	-.079
	Tackle direction	-.094	-.083	-.123	1.000	-.082	-.048	.076	.029
	Tackle height	-.172	-.167	.000	-.082	1.000	-.140	-.392	-.184
	Ball carrier body position	-.058	-.056	-.172	-.048	-.140	1.000	.238	.070
	Tackler body position	-.045	.086	-.091	.076	-.392	.238	1.000	.419
	Active or Passive	-.213	.338	-.079	.029	-.184	.070	.419	1.000

5.6 Reliability

To assess the validity of the qualitative video analysis, the inter-rater reliability was assessed for each of the descriptors described (Hendricks et al, 2020). This involved analysing 40 tackles with another rater. The reliability was assessed using Cohen's kappa (k) where a value above 0.8 indicates an almost perfect agreement (Landis and Koch, 1977).

Table 2. Table showing the analysis framework used to analyse the tackles in this study and the kappa values from the inter-rater reliability test.

Descriptor	Cohen's kappa	Definition
Tackler speed	0.79	**Fast** – Running or sprinting. Purposeful running with maximal effort, with high knee lift. **Moderate** – Jogging, non-purposeful slow running with low knee lift **Slow** – Stationary or walking, or no visible rapid foot movement.

Ball carrier speed	0.78	**Fast** – Running or sprinting. Purposeful running with maximal effort, with high knee lift. **Moderate** – Jogging, non-purposeful slow running with low knee lift **Slow** – Stationary or walking, or no visible rapid foot movement.
Direction	0.88	**Front** – Initial contact made within 30° of hip orientation. **Side** – Initial contact made within 30°-150° of hip orientation. **Behind** – Initial contact made within 150°-180° of hip orientation.
Tackle height	0.87	**Lower Leg** – Area below the base of the knee. **Upper Leg** – Area above the base of the knee. **Hip** – Area covered by the shorts. **Torso** – Area above short line to base of armpit. **Arm** – Area below armpit level on the upper limb. **Shoulder** – Area above the armpit level to the base of the neck.
Ball carrier body position	0.83	**Upright** – Extended hips and knees. **Bent at the Waist** – Approximately 30°+ bend at waist.

		Falling/Diving – Player is falling or diving to the ground.
Tackler body position	0.84	**Upright –** Extended hips and knees. **Bent at the Waist –** Approximately 30°+ bend at waist. **Falling/Diving –** Player is falling or diving to the ground.
Active or passive	0.88	**Active**- Initial contact is made by tacklers shoulder attempting to drive the ball carrier back followed by attempted use of the arms. **Passive**- Initial contact is made by the tacklers shoulder following the attempted use of arms to use bodyweight to drag the ball carrier down. No attempt to drive the ball carrier backwards.

Tackle type	0.95	**Arm –** Tackler attempts to impede the ball carrier with use of upper limbs. **Smother –** Tackler attempts to impede the ball carrier with the use of their chest and by wrapping both arms around. **Shoulder**- Initial contact is made by the tacklers shoulder following the attempted use of arms **Tap –** Tackler attempts to trip the ball carrier with a hand on the lower limb or knee. **No Arm –** Tackle impedes the ball carrier without the use of arms.

6. RESULTS

Over 27 Super League matches, 80 one-on-one inertial shoulder tackles occurred exceeding the triggering threshold of 5g or 500 rad/s^2. PLA, PAA and PAV of the head was recorded by the instrumented mouthguards and the independent variables were categorised using video analysis. The aim of the study was to assess which independent variables had an impact on PAA, PLA and PAV. The data was analysed by making use of multiple linear regression in which the aim was to identify which independent variables could predict PLA, PAA and PAV.

6.1 Frequency distribution

For the PLA (Median = 8.1g, Q1 = 6.3 g, Q3 = 11.1 g), PAA (Median = 638 rad/s^2, Q1= 426.1 rad/s^2, Q3= 873.8 rad/s^2) and PAV (Median = 10 rad/s, Q1 = 6.7 rad/s, Q3 = 14.2 rad/s) majority of the impacts were below the suggested triggering threshold of 10g as shown by the medians in table 3.

Table 3. Table showing the frequency distribution of the head kinematics data from the analysed tackles.

	First Quartile (Q1)	Median	Third Quartile (Q3)
PLA	6.3g	8.1g	11.1 g
PAA	426.1 rad/s^2	638 rad/s^2	873.8 rad/s^2
PAV	6.7 rad/s	10 rad/s	14.2 rad/s

Table 4. Table indicating the mean, minimum and maximum of the HAE data from the analysed tackles.

Descriptive Statistics

	N	Minimum	Maximum	Mean
PLA	80	3.42	27.70	9.2478
PAA	80	154.00	3668.00	756.2474
PAV	80	1.60	26.90	10.5606
Valid N (listwise)	80			

Table 5. Table showing the frequency distribution of the different tackle categories.

Independent variable	Categories	Frequency (n)
Tackle height	Shoulder	52 (65%)
	Arm	1 (1%)
	Torso	23 (29%)
	Upper leg	4 (5%)
Tackler speed	Fast	3 (4%)
	Moderate	25 (31%)

	Slow	52 (65%)
Ball carrier speed	Fast	17 (21%)
	Moderate	57 (71%)
	Slow	6 (8%)
Ball carrier body position	Upright	76 (95%)
	Bent at the waist	4 (5%)
Tackler body position	Upright	40 (50%)
	Bent at the waist	38 (47%)
	Falling	2 (3%)
Tackle direction	Front	45 (56%)
	Side	33 (41%)
	Behind	2 (3%)
Active or passive	Active	48 (60%)
	Passive	32 (40%)

Tackle height

In most of the analysed tackles, the ball carrier was impacted at shoulder height (65%), followed by tackles at the torso (29%) and tackles at the arm (1%). This indicates that tackles to the upper body made up 95% of the analysed tackles. For lower body tackles, impacts were only made to the upper leg (5%).

Speed at impact

The ball carrier carried the ball into contact at moderate speed in majority of the tackles (71%). In 21% of the carries, the ball carrier entered contact at fast speed. Carries at slow speed were less common (8%).

In majority of the events, the tackler completed a tackle while at slow speed (65%). In 31% of the tackles the speed was moderate while 4% of the tackles were completed at high speed.

Body position

In majority of the tackles, the ball carrier carried the ball in an upright body position (95%) and was bent at the waist for the rest of the carries (5%). There were no cases involving a falling ball carrier. The tackler made a tackle while upright in 50% of the cases and bent at the waist for 47.5% of the tackles. In 2.5% of the analysed events, the tackler completed a tackle while falling.

Tackle direction

In this study, majority of the tackles were front-on tackles (56%), followed by tackles on the side of the ball carrier (41%). Tackles made behind the ball carrier were less common (5%).

Active or passive

Majority of the tackles analysed were active (60%) while the rest were passive (40%).

6.2 Head kinematics in the tackle

Three separate multiple linear regression tests were done to investigate whether tackle height, tackler speed, ball carrier speed, ball carrier body position, tackler body position, tackle direction and whether the tackle was active or passive were able to significantly predict PAV, PAA and PLA.

6.2.1 Peak angular velocity (PAV)

The results of the PAV regression analysis showed that the predictor variables used in the study explained 25% of the variance given that the R^2 value was 0.25 as shown in table 6. This model was a significant predictor of PAV as $F(9, 70) = 2.6$ ($p<0.05$) shown in table 7.

Table 6. R and R squared value of the multiple linear regression.

Model Summary[b]

Model	R	R Square
1	.503[a]	.253

Table 7. ANOVA results of the model. Significant when $p < 0.05$.

ANOVA[a]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	461.091	9	51.232	2.629	.011[b]
	Residual	1364.090	70	19.487		
	Total	1825.182	79			

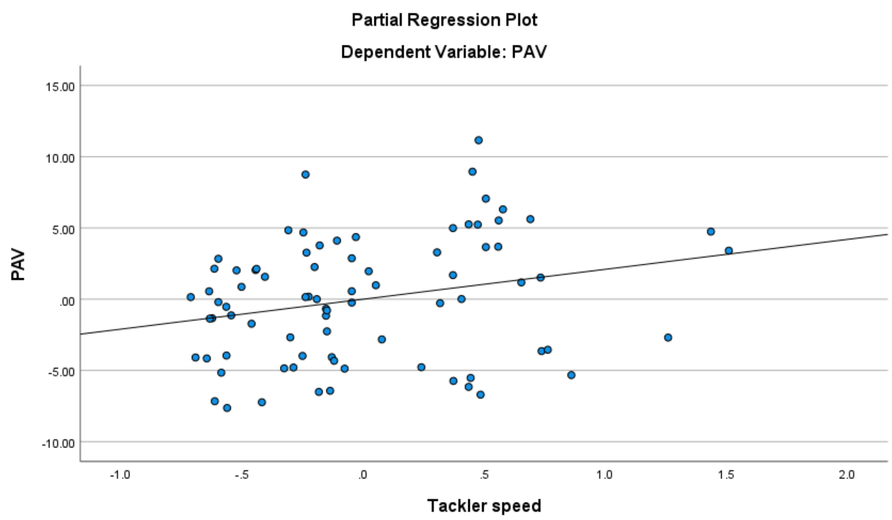
As indicated in table 8, tackler speed (B = 2.1) and tackle height (B = 1.08) contributed significantly to the predictive model as $p < 0.05$. An increase in tackler speed and tackle height by one unit is associated with a subsequent increase in PAV by 2.1 rad/s and 1.08 rad/s respectively as indicated by the positive beta coefficients in table 8. None of the other independent variables significantly predicted PAV given that p was greater than 0.05.

Table 8. Standardized and unstandardized beta coefficients and significance of the independent variables in the model. Beta coefficients show the effect of the predictor variables on the outcome variable.

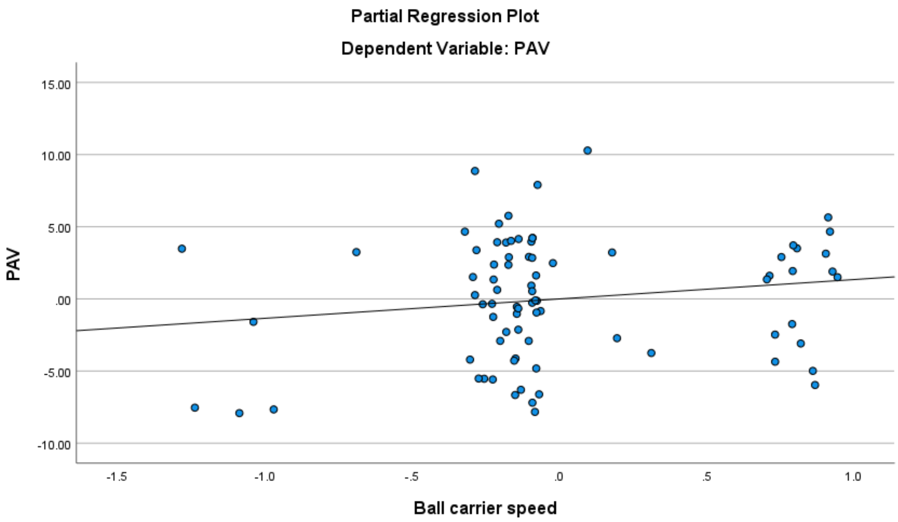
Coefficients[a]

Model		Unstandardized Coefficients B	Unstandardized Coefficients Std. Error	Standardized Coefficients Beta	t	Sig.
1	(Constant)	8.748	5.908		1.481	.143
	Tackler speed	2.100	.957	.246	2.194	.032
	Ball carrier speed	1.343	.977	.146	1.374	.174
	Tackle direction	-1.802	.971	-.202	-1.857	.068
	Tackle height	1.080	.408	.339	2.646	.010
	Ball carrier body position	-2.404	2.413	-.110	-.996	.323
	Tackler body position	-.595	.635	-.123	-.937	.352
	Active or Passive	-1.647	1.182	-.169	-1.393	.168

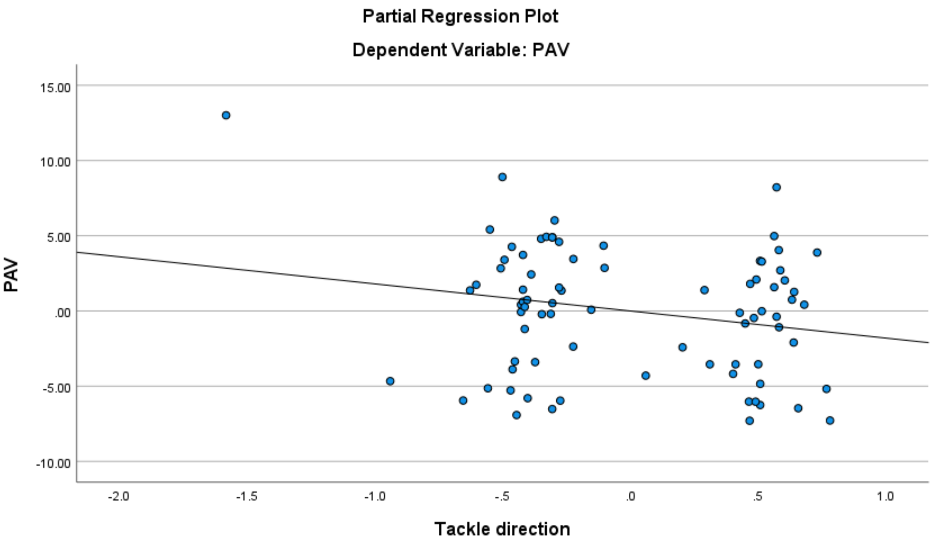
Tackler speed and tackle height were found to be the only independent variables that significantly predict PAV. As shown in plot a and d of figure 6 below, with an increase in the tackler speed and tackle height there was a subsequent increase in PAV. This is indicated by the upward trendline and the positive B coefficient in table 8 above. None of the other independent variables significantly predicted PAV as highlighted above in table 8.



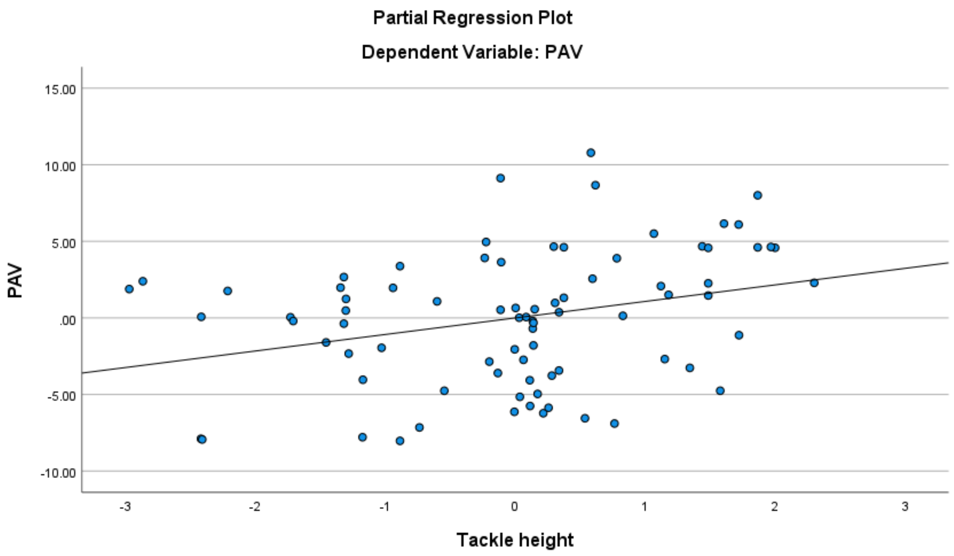
a)


Partial Regression Plot
Dependent Variable: PAV
PAV
Ball carrier speed
15.00
10.00
5.00
.00
-5.00
-10.00
-1.5
-1.0
-.5
.0
.5
1.0

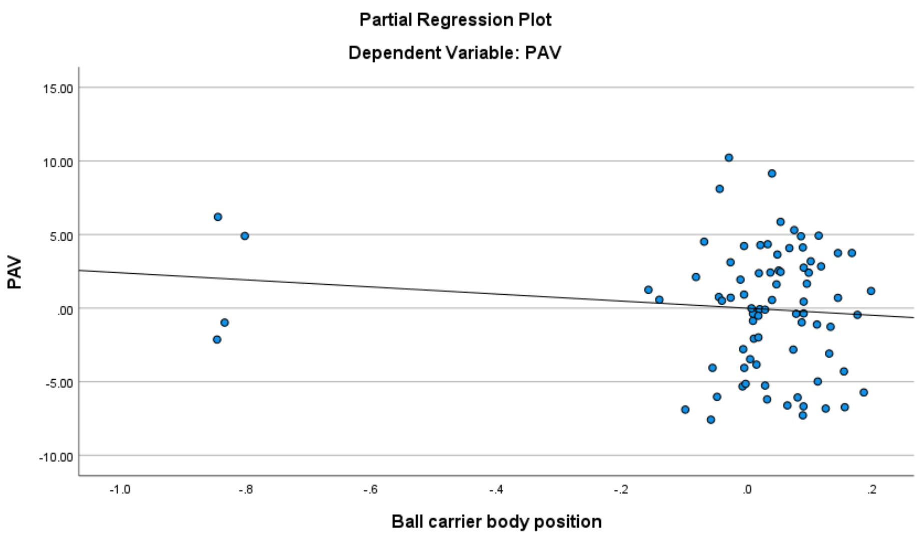
b)


Partial Regression Plot
Dependent Variable: PAV
PAV
Tackle direction
15.00
10.00
5.00
.00
-5.00
-10.00
-2.0
-1.5
-1.0
-.5
.0
.5
1.0

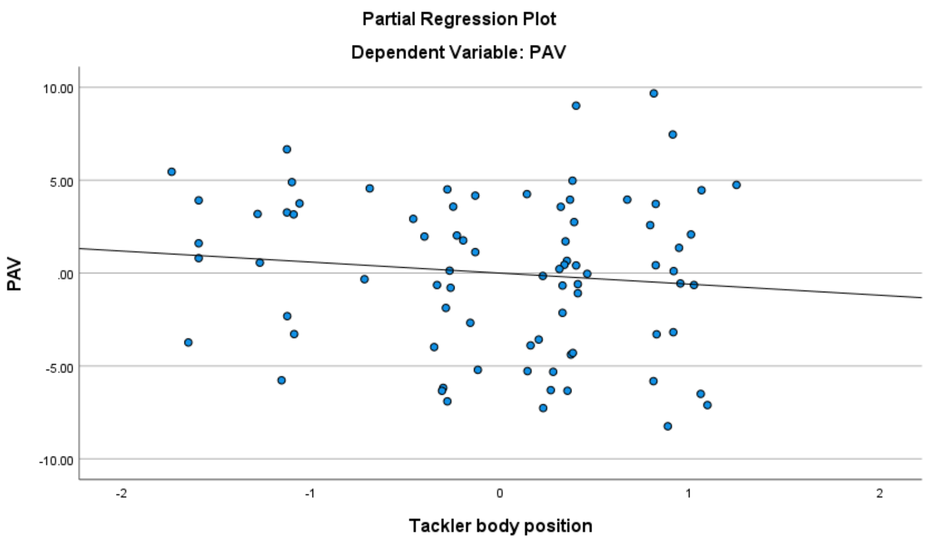
c)


Partial Regression Plot
Dependent Variable: PAV
PAV
15.00
10.00
5.00
.00
-5.00
-10.00
-3
-2
-1
0
1
2
3
Tackle height

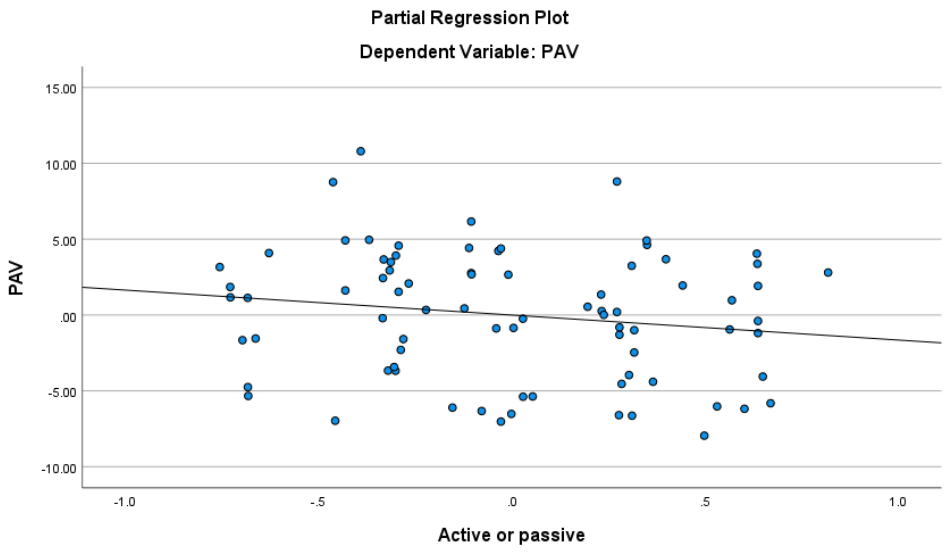
d)


Partial Regression Plot
Dependent Variable: PAV
PAV
15.00
10.00
5.00
.00
-5.00
-10.00
-1.0
-.8
-.6
-.4
-.2
.0
.2
Ball carrier body position

e)


Partial Regression Plot
Dependent Variable: PAV
PAV
10.00
5.00
.00
-5.00
-10.00
-2
-1
0
1
2
Tackler body position

f)



g)

Figure 6. The figures labelled a) to g) represent regression plots of PAV against the various independent variables. Included in the plots is a trendline.

6.2.2 Peak angular acceleration (PAA)

The results of this multiple linear regression indicated that the independent variables used explained 11% of the variance as the R^2 value was 0.11 as shown in table 9. Following an ANOVA test, the results in table 10 showed that the independent variables did not significantly predict PAA as $F(9, 70) = 0.97$, $p > 0.05$. It is significant when $p<0.05$.

Table 9. R and R squared value of the multiple linear regression

Model Summary

Model	R	R Square
1	.333	.111

Table 10. ANOVA results of the model. Significant when p < 0.05.

ANOVA[a]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.519	9	.058	.971	.471[b]
	Residual	4.161	70	.059		
	Total	4.680	79			

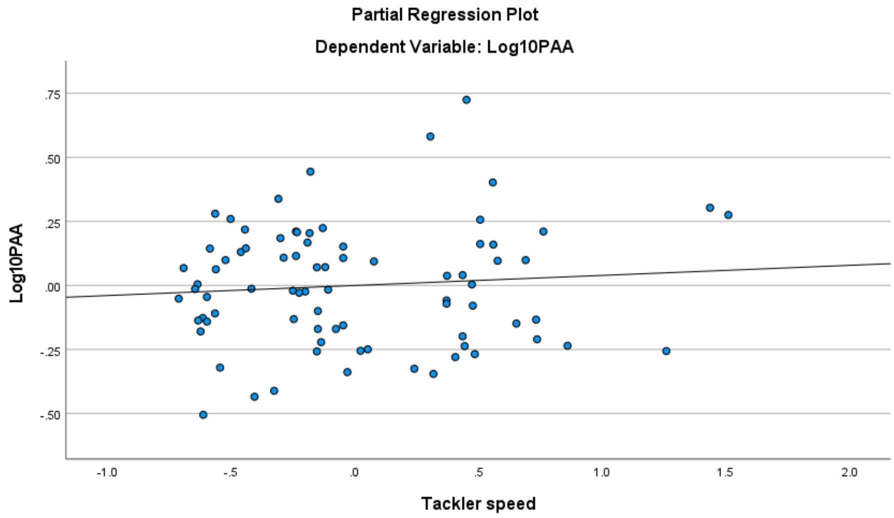
In this model, none of the independent variables significantly predicted PAA given that p > 0.05 among all the variables. A change in the tackle types did not have any effect on PAA.

Table 11. Standardized and unstandardized beta coefficients and significance of the independent variables in the model. Beta coefficients show the effect of the predictor variables on the outcome variable.

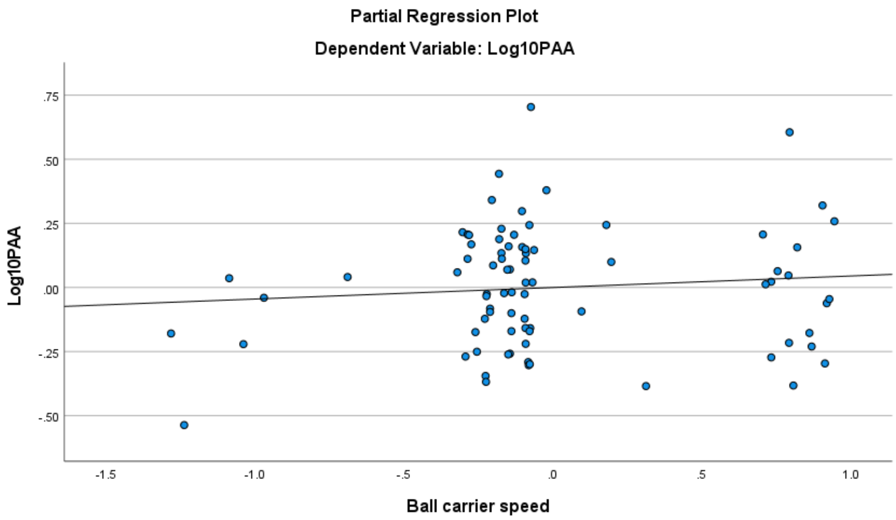
Coefficients[a]

Model		Unstandardized Coefficients B	Unstandardized Coefficients Std. Error	Standardized Coefficients Beta	t	Sig.
1	(Constant)	2.894	.330		8.774	<.001
	Tackler speed	.039	.053	.091	.736	.464
	Ball carrier speed	.045	.055	.096	.824	.413
	Tackle direction	-.092	.054	-.204	-1.695	.094
	Tackle height	.032	.023	.198	1.402	.165
	Ball carrier body position	-.043	.135	-.039	-.322	.748
	Tackler body position	-.054	.035	-.221	-1.529	.131
	Active or passive	-.012	.066	-.025	-.188	.851

In figure 8 are partial regression plots showing the relationship between PAA and the independent variables. None of the independent variables were found to significantly predict PAA.

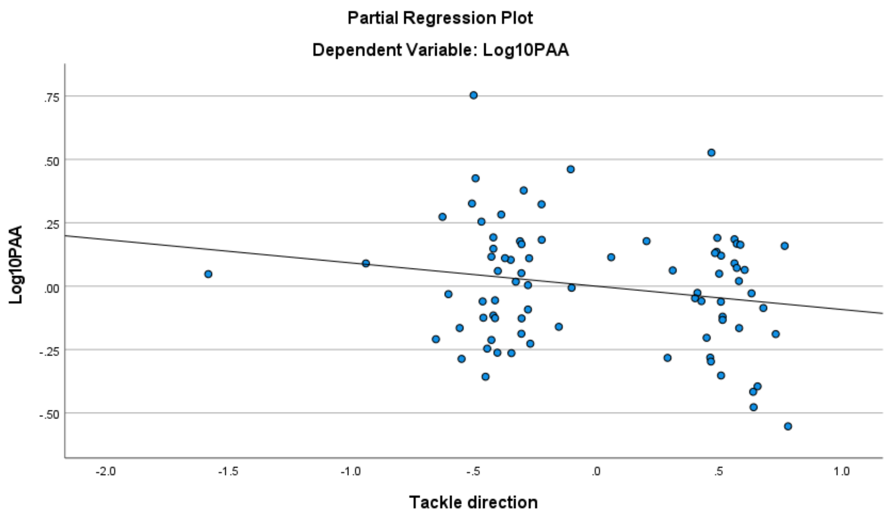


a)



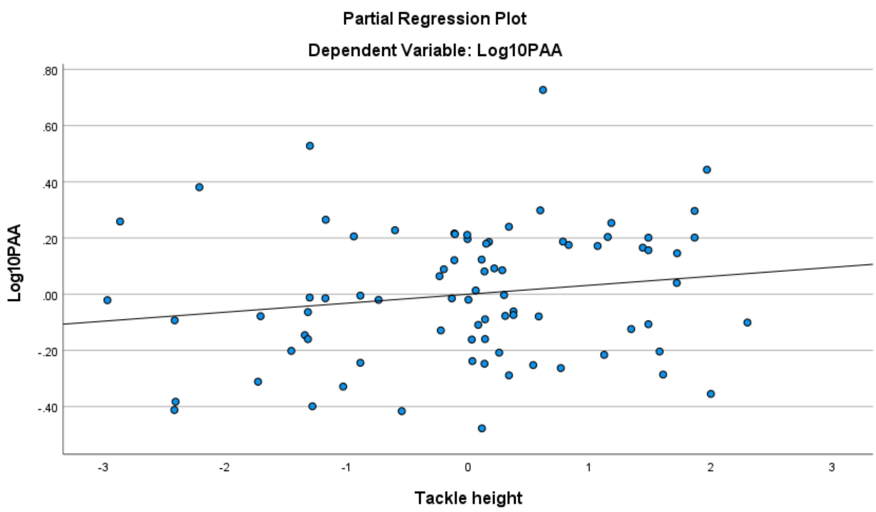
Partial Regression Plot
Dependent Variable: Log10PAA
Log10PAA
.75
.50
.25
.00
-.25
-.50
-1.5
-1.0
-.5
.0
.5
1.0
Ball carrier speed

b)



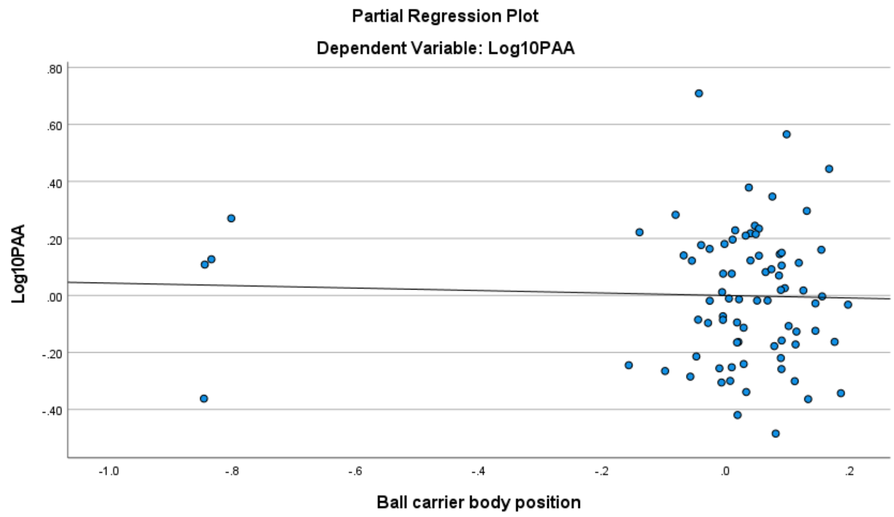
Partial Regression Plot
Dependent Variable: Log10PAA
Log10PAA
.75
.50
.25
.00
-.25
-.50
-2.0
-1.5
-1.0
-.5
.0
.5
1.0
Tackle direction

c)

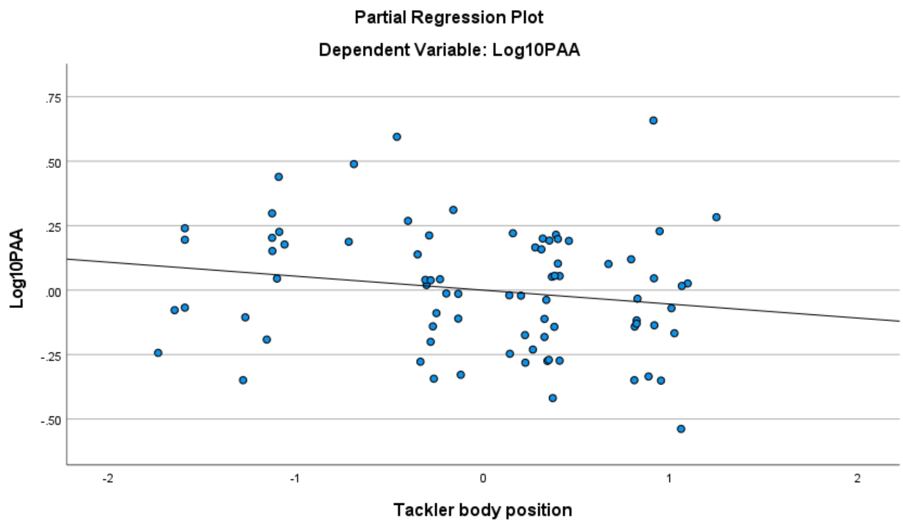


Partial Regression Plot
Dependent Variable: Log10PAA
Log10PAA
.80
.60
.40
.20
.00
-.20
-.40
-3
-2
-1
0
1
2
3
Tackle height

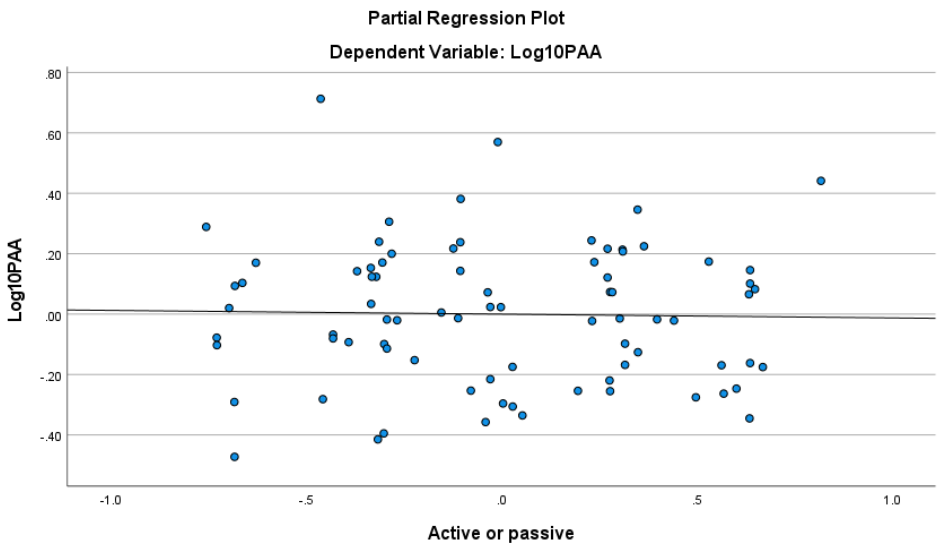
d)



e)



f)



g)

Figure 7. The figures labelled a) to g) represent regression plots of PAA against the various independent variables. Included in the plots is a trendline.

6.2.3 Peak linear acceleration (PLA)

The independent variables explained 18% of the variance in the data given that the R^2 value was 0.18 as indicated in table 12. The regression model did not significantly predict PLA as shown in table 13, F (9, 70) = 1.68, p > 0.05.

Table 12. R and R squared value of the multiple linear regression

Model Summary[b]

Model	R	R Square
1	.422[a]	.178

Table 13. ANOVA results of the model. Significant when p < 0.05.

ANOVA[a]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	.438	9	.049	1.681	.110[b]
	Residual	2.026	70	.029		
	Total	2.464	79			

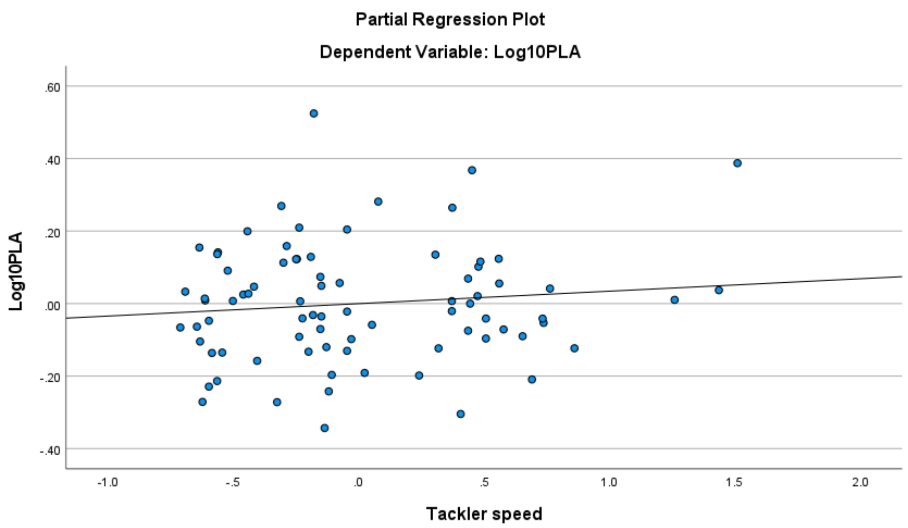
With regards to the independent variables, none of them were found to significantly predict PLA given that none of the p values of the regression test were significant as shown in table 14 below. (Significant when p < 0.05).

Table 14. Standardized and unstandardized beta coefficients and significance of the independent variables in the model. Beta coefficients show the effect of the predictor variables on the outcome variable.

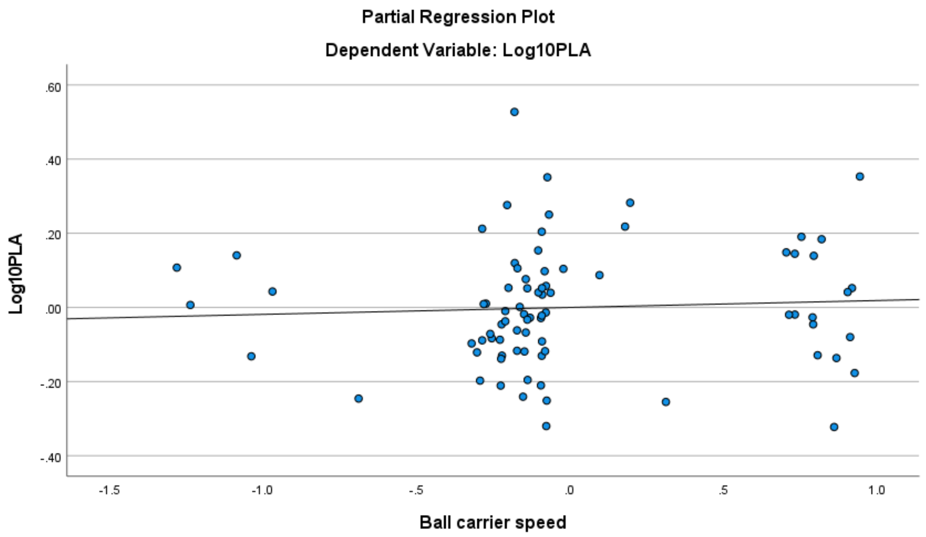
Coefficients[a]

Model		Unstandardized Coefficients B	Unstandardized Coefficients Std. Error	Standardized Coefficients Beta	t	Sig.
1	(Constant)	1.401	.228		6.139	<.001
	Tackler speed	.034	.037	.109	.928	.356
	Ball carrier speed	.019	.038	.055	.495	.622
	Tackle direction	-.043	.037	-.130	-1.137	.259
	Tackle height	.009	.016	.075	.558	.579
	Ball carrier body position	-.122	.093	-.151	-1.305	.196
	Tackler body position	-.028	.025	-.159	-1.154	.253
	Active or passive	-.068	.046	-.191	-1.496	.139

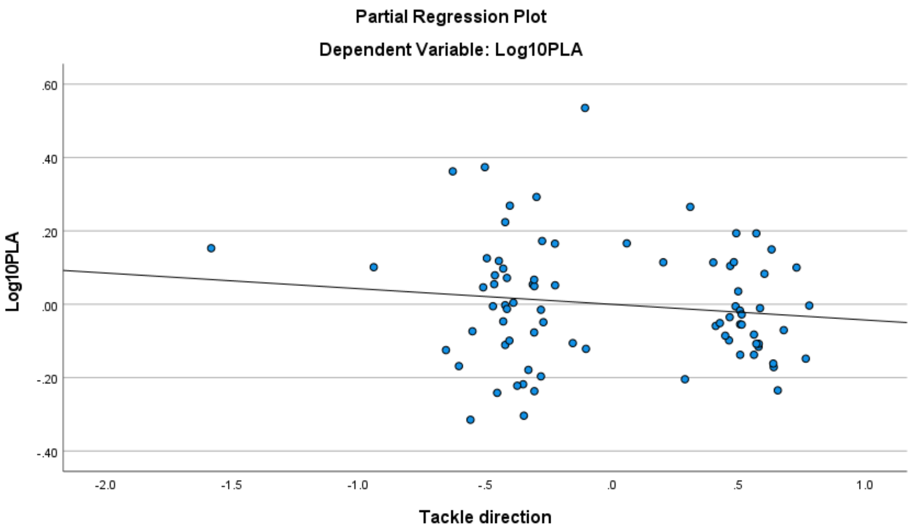
In figure 8 below are partial regression plots between PLA and all the independent variables. None of the independent variables significantly predicted PLA given that p was greater than 0.05 for all the variables as indicated in table 14.



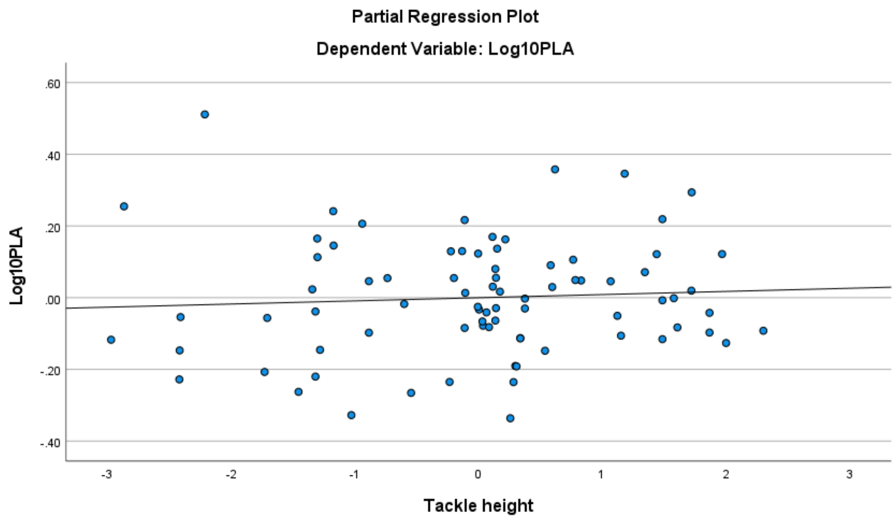
a)



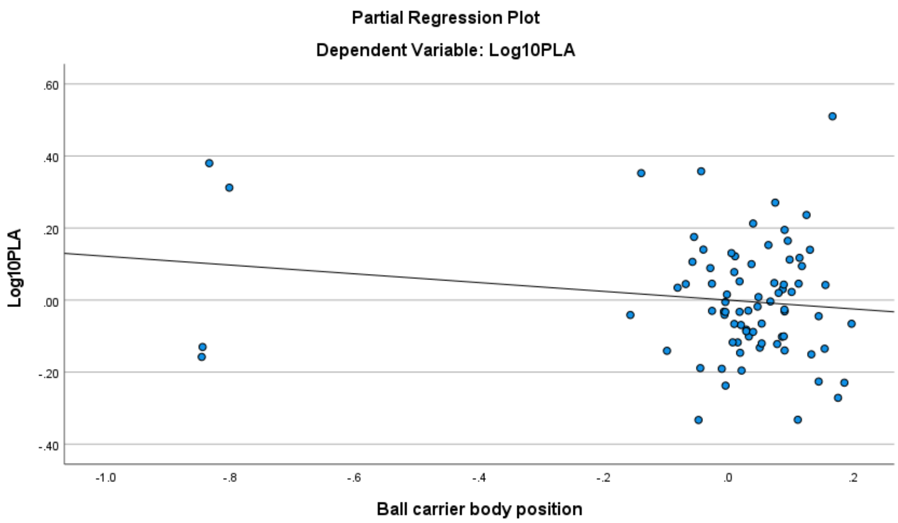
b)


Partial Regression Plot
Dependent Variable: Log10PLA
Log10PLA
.60
.40
.20
.00
-.20
-.40
-2.0
-1.5
-1.0
-.5
.0
.5
1.0
Tackle direction

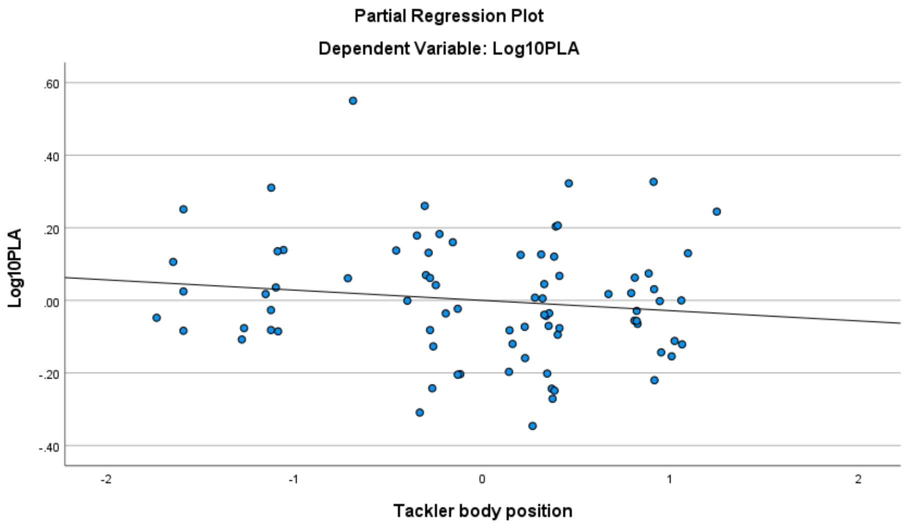
c)


Partial Regression Plot
Dependent Variable: Log10PLA
Log10PLA
.60
.40
.20
.00
-.20
-.40
-3
-2
-1
0
1
2
3
Tackle height

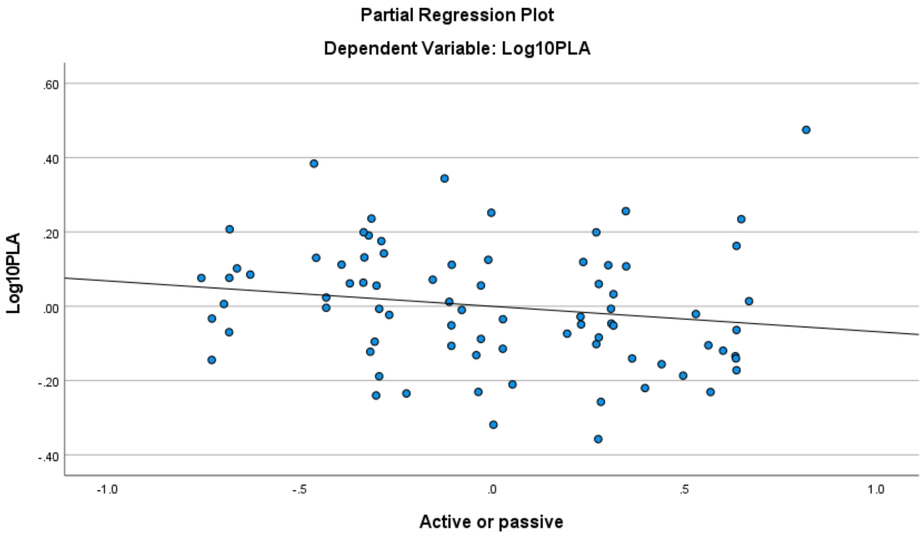
d)


Partial Regression Plot
Dependent Variable: Log10PLA
Log10PLA
.60
.40
.20
.00
-.20
-.40
-1.0
-.8
-.6
-.4
-.2
.0
.2
Ball carrier body position

e)



f)



g)

Figure 8. The figures labelled a) to g) represent regression plots of PAV against the various independent variables. Included in the plots is a trendline.

7. DISCUSSION

Tackle height

This study found tackle height to be a predictor of PAV. An increase in tackle height led to an increase in PAV as shown in table 8. Tackle height did not however significantly predict PLA or PAA as indicated in tables 11 and 14. The results of this study show that even with using a lower triggering threshold of 5g, an increase in tackle height resulted in higher HAE (PAV).

This is in line with the literature which found that tackles at shoulder height resulted in higher HAE when compared with tackles at torso height (Tierney et al, 2018). It was concluded that reducing the tackle height to below the shoulder level would reduce PLA, PAA and change in angular velocity by 35%, 61% and 40% respectively (Tierney et al, 2018).

Given that an increase in tackle height leads to an increase in PAV, more should be done to encourage lower tackle heights. Experts in World Rugby identified increased sanctions, improved tackle technique through coaching and law changes as measures to reduce the tackle height in rugby union (Raftery et al, 2021).

Between 2018 and 2019, the number of yellow and red cards given after head contact increased by 74% and 138% respectively. Subsequently, the concussion rate reduced by 28% with tackle concussions decreasing by 37%. This indicates that tougher officiating on the tackle height led to a reduction in concussion incidence (Raftery et al, 2021). These measures can potentially be adopted within rugby league in a bid to reduce concussion incidence. Although this study did not focus on concussion, reducing the magnitude of head acceleration events has the potential to reduce concussion.

<u>Body position</u>

The body position of the tackler and ball carrier was not found to predict PAV, PLA or PAA. This means that the magnitude of the HAE did not change with the different body positions. In majority of the tackles (95%), the ball carrier was observed to be carrying the ball in an upright position. However, for tacklers it was more equal as they were upright in 50% of the cases and bent at the waist in 47.5% of the tackles. It has been found that tackles involving an upright tackler had a higher propensity to result in a head injury assessment (HIA) in rugby union compared to other body positions (Tucker et al, 2017).

A HIA is when a player is removed from the pitch and assessed due to a suspected concussion (Tucker et al, 2017). A higher magnitude HAE is more likely to result in HIA than a lower magnitude HAE (Stemper et al, 2019). An upright tackler was 1.5 times more likely to experience HIA than a tackler who was bent at the waist. This is because an upright body position increases the likelihood of head-to-head contact (Tucker et al, 2017). A falling ball carrier was also found to be at the highest risk of HIA (Tucker et al, 2017). There were no tackles involving a falling ball carrier in this study which is likely why ball carrier body position was not found to be predictive of PLA, PAA and PAV. In addition to this, the previous studies also involved direct impacts while only inertial impacts were assessed for this study.

Speed in the tackle

The results of this study indicate that when the tackler's speed into the tackle increases, the PAV increases as well. This is shown in table 8 in the results section above. PLA and PAA were not found to increase with higher tackler speed as table 11 and 14 indicate that the prediction was not significant. However, the ball carrier speed did not affect PLA, PAA or PAV given that none of the results from the multiple linear regression were significant. In a study identifying the risk factors for head injury events in rugby union, it was found that an increase in the tackler's speed resulted in a higher propensity for head injury (Tucker et al, 2017).

A further study investigating the risk factors of concussion in rugby union concluded that there was a higher risk of concussion when the tackler accelerates into the tackle at high speed. Tackler speed was determined to be the second most important variable for predicting concussion outcome (Cross et al, 2019). Although concussion was not measured in this study, an increase in the magnitude of HAE increases the risk of concussion (Stemper et al, 2019).

The results of this study show that with an increase in the tackler's speed into the tackle, there is an increase in the PAV. PAV has been identified as a good predictor of brain strain (Bian and Mao, 2020). Brain strain has been identified as a significant predictor of concussion as it describes the deformation of the brain (Takhounts et al, 2008). Brain strain has also been described as the cause of the symptomatology of concussion (Post and Hoshizaki, 2015).

In both rugby league and union, when teams are defending, they aim to put pressure on the opposition by rushing up and making a tackle behind the gainline (Tierney et al, 2018b). Advancing past the gainline gives the attacking team an advantage as they can gain territory and progress to the tryline (Sayers and Washington-King, 2005). Defensive line speed (Tacklers accelerating) significantly reduces the likelihood of the attackers crossing the gainline (Hendricks et al, 2013). This is because the attackers have less time and space to execute plays when the defence rushes up (Gabbett and Kelly, 2007). Crossing the gainline has been found to be associated with the points scored against a team in rugby union (Hendricks et al, 2013).

In rugby league, the Sydney Roosters club in the NRL made use of a fast line-speed defensive strategy to nullify the attack. The team proceeded to win the 2002 NRL premiership and made it to the finals in 2003 and 2004 (Gabbett and Kelly, 2007). In response to this, many teams including amateur clubs attempted to emulate this defensive strategy (Gabbett and Kelly, 2007).

Fast line-speed has however been found to reduce tackle proficiency as excessive acceleration can compromise the tackling technique. The study found that when defenders accelerated to attempt a tackle, they were unable to keep their body square to the attacker to make an effective tackle (Gabbett and Kelly, 2007). Coaches employing defensive tactics that rely less on fast line speed could potentially reduce the magnitude of HAE as tackler speed would reduce. Another suggestion given is to reduce the distance between the two backlines during set piece plays to reduce the momentum during the collisions. However, it is unclear how much the distance should be reduced to substantially decrease the momentum (Cross et al, 2019).

Tackle direction

The results of this study show that tackle direction did not significantly predict PAA, PLA or PAV. A study on rugby union concluded that tackles from the front had a significantly higher propensity to result in HIA when compared to other directions (Tucker et al, 2017). Tackling and being tackled head-on led to the most tackle related concussions in a study covering three English Premiership rugby seasons (Kemp et al, 2008).

It is suspected that tackles from the front result in more concussions as there is increased likelihood of contact with the head (Kemp et al, 2008). This does not apply to this study as only inertial impacts were analysed hence there was no contact with the ball carrier’s head. In addition to this, there were no concussions analysed in this study.

Active vs passive tackles

In this study, there was no significant difference in the kinematics experienced during active and passive tackles. A previous study found that active tackles had a higher propensity to result in HIA when compared with passive tackles. Due to the force applied in active tackles, there is a higher energy transfer which is a risk factor for head injury (Tucker et al, 2017). Given that the previous study also included direct impacts, the results do not apply to this study as only inertial impacts were involved.

Triggering threshold

As indicated in table 3, the kinematic values are skewed to lower values. The recommended threshold for HAE is 10g to avoid including non-contact HAE (King et al, 2016). However, using a 10g threshold would result in underreporting of HAE. In this study, the data acquisition threshold was set at 5g. As shown by the results, using a 10g threshold would result in a lot of tackles being ignored as majority of the impacts were between 5g and 10g.

This can be seen as in table 3 the median values of PLA, PAA and PAV are 8.1g, 638 rad/s^2 and 10 rad/s. By combining the data from the instrumented mouthguards and video analysis, the true positive events can be identified as the mouthguard data is time-stamped. Given that different triggering thresholds lead to differences in reported HAE, when comparing HAE exposure from different studies, it is necessary to compare studies with similar triggering thresholds as there would be no underreporting of events.

7.1 Limitations and Future work

Future research should also include the tackler in the analysis. Even though ball carriers in rugby league are at a higher risk of concussion, there is also a substantial risk to the tackler (Gardner et al, 2015a). As more teams in rugby league begin to use instrumented mouthguards, research should also be done on the tackler's HAE during the tackle. It is equally important to analyse the magnitude of the HAE experienced by ball carriers.

Given the subjective nature of qualitative video analysis, there are potential errors that could occur during analysis. This is particularly prevalent during borderline cases. For example, with regards to ball carrier speed as shown in table 2, a player who is tackled while transitioning between jogging and sprinting. These cases are difficult to analyse. Development of objective parameters to measure speed has the potential to reduce some errors involved. Only inter-rater reliability was assessed for this study which could affect the validity.

Research should be undertaken on the effect of the distance between opposing backlines and the HAE during the tackle. It is suspected that with a larger distance between the players, there would be higher head accelerations due to an increased energy transfer (Cross et al, 2019). Given that no research has been done on this, it is unclear how much the distance between backlines should be reduced to reduce the head accelerations sustained by the players. Future studies should compare the effect of different distances on the HAE in the tackle. The results of these studies have the potential to influence law changes by rugby league governing bodies seeking to reduce the magnitude of head accelerations.

In this study, only tackles that triggered the mouthguards to record a HAE were analysed. Tackles that did not result in HAE were not analysed. Because of this, it is not clear whether the tackle types with the highest HAEs led to an increased likelihood of HAE or whether they simply occurred more often than other tackle types. Future studies should compare the data from tackles that resulted in HAE and those that did not result in HAE to assess the tackle characteristics with a higher propensity to result in HAE. This would enable safer tackle techniques to be identified and would inform measures to reduce the magnitude of HAE in rugby league.

7.2 Conclusion

In conclusion, an increase in tackle height ($B= 2.1$, $p< 0.05$) and tackler speed ($B= 1.08$, $p <0.05$) significantly resulted in an increase in PAV. The study found that even with using a lower triggering threshold, tackler speed and tackle height both significantly predicted PAV. With an increase in tackle height and tackler speed there was an increase in the magnitude of PAV experienced by the ball carrier. Ball carrier speed, body position at the tackle, tackle direction and whether the tackle was active or passive did not significantly predict PLA, PAA or PAV.

Majority of the HAE from these impacts were of a low magnitude (Lower than 10g). The results therefore suggest that majority of one-on-one shoulder tackles are of low magnitude. This could be a possible reason why ball carrier speed, body position at the tackle, tackle direction and whether the tackle was active or passive did not significantly predict PLA, PAA or PAV as the magnitude of HAE was low for most of these tackles.

Interventions should be made in rugby league to reduce the tackler speed and tackle height to reduce the magnitude of HAE. For tackle height, law changes and increased sanctions have the potential to change player behaviour and lower the tackle height (Raftery et al, 2021). In addition to this, using coaching strategies that rely less on fast line speed can potentially reduce tackler speed. Making these changes will potentially make the game safer for athletes as the magnitude of HAE experienced would be reduced.

Using a lower triggering threshold of 5g for the instrumented mouthguards increased the number of HAE reported. Majority of the impacts in this study were lower than the 10g triggering threshold suggested by previous studies (King et al, 2016). By combining the instrumented mouthguards with qualitative video analysis, it is possible to identify false positives caused by non-contact HAE as opposed to using a higher triggering threshold.

This study shows that using a combination of instrumented mouthguards and qualitative video analysis can be used to identify potential risk factors in sports. Video analysis is important as it categorises the HAE allowing comparison between different tackle types. Given that increased magnitude and quantity of HAE experienced by athletes increases the risk of concussion, it is necessary to monitor the HAE during games (Stemper et al, 2019). In this study, tackle height and tackler speed were both identified as areas of concern given that an increase in both led to increased PAV. To make contact sports safer, identifying factors that increase the risk of injury is important.

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