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A Comparison of Dynamic Response and Brain Tissue Deformation for Ball Carriers and Defensive Tacklers in Professional Rugby Shoulder-to-Head Concussive Impacts

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ABSTRACT:

The long-term consequences of repetitive mild traumatic brain injuries (mTBIs), or concussions, as well as the immediate acute dangers of head collisions in sport have become of growing concern in the field of medicine, research and athletics. An estimated 3.8 million sports-related concussions occur in the United States annually, with the highest incidence having been documented in football, hockey, soccer, basketball and rugby (Harmon et al., 2013). The incidence of concussion in the National Rugby League (NRL) corresponds to approximately 8.0-17.5 injuries per 1000 playing hours, with tackling having been identified as the most common cause (Gardner et al., 2014; King et al., 2014). The highest incidence of rugby concussive impacts is a result of shoulder-to-head collisions (35%) during tackles and game play (Gardner et al., 2014). Shoulder-to-head concussive events occur primarily on the ball carrier and secondarily on the tacklers (Hendricks et al., 2014; Quarrie & Hopkins, 2008). While some studies report that the ball carrier is at a greater risk of sustaining a concussion (Gardner et al., 2015; King et al., 2010, 2014), others have demonstrated a greater incidence of tacklers being removed from play for sideline concussion evaluation (Gardner et al., 2014). Given this discrepancy, the purpose of this study was to compare dynamic response and brain tissue deformation metrics for ball carriers and defensive tacklers in professional rugby during shoulder-to-head concussive impacts using in-laboratory reconstructions. Ten cases with an injured defensive tackler and ten cases with an injured ball carrier were reconstructed using a pneumatic linear impactor striking a 50th percentile Hybrid III headform to calculate dynamic response and maximum principal strain values. There was no significant difference between the two impact conditions for peak resultant linear and rotational accelerations, as well as brain tissue deformation. Differences between metrics in this research and past research where the impacting system was not reported were discussed. These differences reflect the importance of accounting for impact compliance when describing the risk associated with collisions in professional rugby.

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Equation 1. Shear characteristics of the viscoelastic behaviour of the brain

$$G(t) = G_\infty + (G_0 - G_\infty)e^{\beta t}$$

Equation 2. Hyperelastic Law

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-\frac{t}{0.008}} + 1103e^{-\frac{t}{0.15}}(Pa)$$

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CHAPTER 1: INTRODUCTION

1.1 Overview

The long-term consequences of repetitive mild traumatic brain injuries (mTBIs), or concussions, as well as the immediate acute dangers of head collisions in sport have become a growing concern in the field of medicine, research and athletics. Campaigns promoting prevention, education, and rule modifications, such as the *BokSmart* intervention program in rugby, are being used as risk reduction strategies for both athletes and support staff (Brown, Gardner-Lubbe, Lambert, Van Mechelen, & Verhagen, 2015; Harmon et al., 2013; McIntosh et al., 2009). The First International Conference on Concussion in Sport (Vienna 2001) provided a consensus statement on concussion in an athletic environment (Aubry et al., 2002); eleven years later, the fourth consensus, originating from the Zurich 2012 conference, has generated sports concussion assessment tools and world-wide coverage for adults and youth, which spans from amateur to professional sports groups, such as the International Rugby Board (IRB) (Harmon et al., 2013; McCrory et al., 2013). The IRB has attempted to decrease the risk of injuries in rugby by rule enforcement for illegal, or “dangerous”, tackles, protective equipment regulation, and stricter application of the graduated return to play protocols following diagnosed concussions, as established by the consensus statement (Best, McIntosh, & Savage, 2005; McCrory et al., 2013; McIntosh, 2003, 2004; McIntosh, McCrory, Finch, Chalmers, & Best, 2003; Rugby, 2014). In order for the IRB to develop appropriate player safety strategies, the risk associated with the concussive impact events must be described, including the potential differences in risk for the tackler and the ball carrier during the concussive collisions (Gardner et al., 2015; Gardner, Iverson, Stanwell, et al., 2014; King, Gissane, Clark, & Marshall, 2014; King, Hume, Milburn, & Guttenbeil, 2010).

A concussion is a trauma-induced transient or long-term physiological disturbance in brain function caused by biomechanical forces where, unlike other traumatic sports injuries where objective analyses, medical imagery and symptomology can be used in order to establish a diagnosis, healthcare professionals rely solely on symptoms as their main diagnostic tool for these brain injuries (Harmon et al., 2013; McCrory et al., 2013). A medical history of previous concussions has been associated with a greater risk of repeated mTBIs, and recent evidence suggests chronic traumatic encephalopathy to be a long-term neurological complication of repeated concussions, as reported in high-contact sport athletes post-mortem in sports such as American football and boxing (Harmon et al., 2013; McCrory et al., 2013; McKee et al., 2010; Omalu et al., 2005; Zhang, Heier, Zimmerman, Jordan, &Uluğ, 2006). Hence, a multifactorial approach is needed in concussion management in order to diminish risk to athletes, especially with regards to contact sports where mTBIs are a prominent medical concern (Benson et al., 2013).

An estimated 3.8 million sports-related concussions occur in the United States annually, with the highest incidence having been documented in football, hockey, soccer, basketball, and rugby (Harmon et al., 2013). Lack of education and knowledge, as well as the masking of symptoms, account for as many as 50% of concussions going unreported or undetected in athletes (Gardner, Iverson, Levi, et al., 2014; Harmon et al., 2013). Biomechanical analyses of head impacts in professional American football and professional hockey have been conducted in laboratory settings to analyze cerebral response and concussive risk during game play (Delaney, Puni, & Rouah, 2006; Kendall, Post, & Gilchrist, 2012; Oeur, Zanetti, Gilchrist, & Hoshizaki, 2014; Pellman et al., 2003; Post, Oeur, Hoshizaki, & Gilchrist, 2011; Rousseau & Hoshizaki, 2015; Rousseau, Post, & Hoshizaki, 2009; Rousseau, 2014; Takhounts, Craig, Moorhouse, McFadden,

& Hasija, 2013; Viano et al., 2005; Withnall, 2005; Zhang, Yang, & King, 2001). Due to the subjective nature of concussion diagnoses (Harmon et al., 2013; McCrory et al., 2013), the analyses of this cerebral response and the kinematics of head impacts in sport allow for quantification of brain tissue strain, which are used as brain injury predictors (Deck & Willinger, 2008; Fréchède & McIntosh, 2009; Kimpara & Iwamoto, 2012; King, Yang, Zhang, Hardy, & Viano, 2003; Kleiven, 2007; McAllister et al., 2012; McIntosh, 2004; Patton, McIntosh, & Kleiven, 2013; Rowson & Duma, 2013; Willinger & Baumgartner, 2003; Zhang, Yang, & King, 2004). As rugby has the third highest incidence of mTBIs in sport, following American football and hockey, attempts have been made by the IRB to decrease risk of injury through rules and regulations for tackling technique, however little is known regarding the mechanics of concussive rugby tackling events, their differing impacting systems, and the ensuing cerebral response during rugby game play (Fréchède & McIntosh, 2009; Gardner, Iverson, Levi, et al., 2014; Harmon et al., 2013; Hendricks, Matthews, Roode, & Lambert, 2014; Hollis et al., 2011; King et al., 2010, 2014; McIntosh, McCrory, & Comerford, 2000; McIntosh, Savage, McCrory, Fréchède, & Wolfe, 2010; Patton et al., 2013; Rugby, 2014). Researchers have analysed forces and accelerations involving rugby injury characteristics using mouthguards and tackling bags, as well as measuring dynamic response and brain deformation, independently reporting location and velocity, during collision reconstructions using live video footage, Mathematical Dynamic Models (MADYMO) software and finite element analyses (Hendricks et al., 2014; King, Hume, Brughelli, & Gissane, 2015; McIntosh et al., 2000, 2010; Patton et al., 2013; Seminati, Cazzola, Preatoni, & Trewartha, 2010; Usman, McIntosh, & Fréchède, 2011). However, the impacting system, shoulder-to-head collisions for example, was not reported for these impact reconstructions, nor was the player’s role, as ball carrier or tackler.

The incidence of concussion in the National Rugby League (NRL) corresponds to approximately 8.0-17.5 injuries per 1000 playing hours, with tackling having been identified as the most common cause (Gardner, Iverson, Levi, et al., 2014; King et al., 2014). Gardner et al. (2014) reported that 35% of the rugby concussive impacts were a result of shoulder-to-head collisions during tackles and plays. Shoulder-to-head concussive events occur primarily on the ball carrier as they are being tackled in an arm-out shoulder tackle with head moving away from ball carrier's body, as well as on the tackler themselves during a "ball carrier fend" where the offensive player's shoulder, in a tucked-arm position, and forward head collides with the defenseman (Hendricks et al., 2014; Quarrie & Hopkins, 2008). While studies report that the ball carrier is at a greater risk of sustaining a concussion (Gardner et al., 2015; King et al., 2010, 2014), the 2014 Concussion Interchange Rule analysis has demonstrated a greater incidence of tacklers being removed from play for sideline concussion evaluation (Gardner, Iverson, Stanwell, et al., 2014). Given this discrepancy, as dynamic cerebral response and brain tissue strain have been reported to be brain injury predictors, and in order to contribute to the IRB's attempts to decrease risk of injury, analyses of ball carrier and defensive tackler concussive injury are required in order to evaluate differences in peak resultant linear and rotational acceleration as well as maximum principal strain (Deck & Willinger, 2008; Fréchède & McIntosh, 2007, 2009; Kimpara & Iwamoto, 2012; King et al., 2003; Kleiven, 2007; McAllister et al., 2012; McIntosh, 2004; Patton et al., 2013; Rowson & Duma, 2013; Willinger & Baumgartner, 2003; Zhang et al., 2004).

The biomechanical characteristics of shoulder-to-head concussive impacts in professional rugby, the events which lead to the greatest incidence of concussions in rugby, have not been quantified. The present study is designed to compare dynamic response and brain tissue deformation in ball carriers and tacklers during rugby shoulder-to-head concussive impacts in professional rugby players.

1.2 Research Question

Do dynamic response and brain tissue deformation values differ between ball carriers and tacklers for professional rugby shoulder-to-head concussive impacts?

1.3 Objectives

1. To compare peak resultant linear acceleration values of ball carriers and tacklers from diagnosed concussive shoulder-to-head impact reconstructions in professional rugby players.
2. To compare peak resultant rotational acceleration values of ball carriers and tacklers from diagnosed concussive shoulder-to-head impact reconstructions in professional rugby players.
3. To compare brain tissue deformation (maximum principal strain) values of ball carriers and tacklers from diagnosed concussive shoulder-to-head impact reconstructions in professional rugby players.

1.4 Variables

1.4.1 Independent Variables

Concussed athlete's position during shoulder-to-head collision

i) Defensive tackler

ii) Ball carrier

1.4.2 Dependent Variables

i) Dynamic response

  a. Peak resultant linear acceleration

  b. Peak resultant rotational acceleration

ii) Brain tissue deformation: peak maximum principal strain

1.5 Hypotheses

1.5.1 Experimental Hypotheses

1. Peak resultant linear acceleration measures will be significantly different between tacklers and ball carriers, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players
    a. Peak resultant linear acceleration values will be significantly greater when the ball carrier sustains the concussion due to ball carriers' forward head position technique, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players
2. Peak resultant rotational acceleration measures will be significantly different between tacklers and ball carriers, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players

a. Peak resultant rotational acceleration values will be significantly greater when the defensive tackler sustains the concussion due to tacklers' head moving away from the ball carrier's body during contact, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players

1.5.2 Null Hypotheses

1. Peak resultant linear acceleration measures will not be significantly different between tacklers and ball carriers, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players
    a. Peak resultant linear acceleration values will not be significantly greater when the ball carrier sustains the concussion due to ball carriers' forward head position technique, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players
2. Peak resultant rotational acceleration measures will not be significantly different between tacklers and ball carriers, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players
    a. Peak resultant rotational acceleration values will not be significantly greater when the defensive tackler sustains the concussion due to tacklers' head moving away from the ball carrier's body during contact, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players
3. Peak maximum principal strain values will not be significantly different between tacklers and ball carriers, reconstructed from diagnosed concussive shoulder-to-head impacts in professional rugby players

1.6 Significance

The proposed study will evaluate shoulder-to-head impact reconstructions of real-life rugby game play events in the National Rugby League which resulted in diagnosed concussions. As shoulder-to-head collisions are the impacts leading to the greatest incidence of mild traumatic brain injuries in professional rugby, this is significant as rugby players are among the high-contact sport athletes who are at a greater risk of sustaining concussions and potentially suffering from long-term neurological disorders linked to sustaining repetitive impacts to the head. Past research has not analyzed the player position-dependent impact characteristics contributing to shoulder-to-head collisions; therefore this information will contribute to current rugby head injury research, guide future research methodology, as well as provide additional information regarding concussive tackling events in professional rugby, contributing to the International Rugby Board (IRB) injury prevention strategies. This study will quantify dynamic response and maximum principal strain of concussive rugby impacts based on current 2014 National Rugby League video footage, with clear images free of pixilation, providing more accurate kinematic measures than past biomechanical rugby analyses. Previous research has based their research methodology and analyses on 1990s VHS footage using two-dimensional analysis, which was deemed to be a major limitation of their methodology (Fréchède & McIntosh, 2009; McIntosh et al., 2000; Patton et al., 2013). Furthermore, current protective equipment testing follows the IRB regulations which include the use of a drop system at 0.3-0.6m, representing 2.4-3.4 m/s during reconstructions (McIntosh, 2004; McIntosh et al., 2009); these reconstructions have been reported as not representing concussive impacts in rugby (Fréchède & McIntosh, 2009). The proposed study for this thesis will use a more refined methodology for shoulder-to-head collisions using a pneumatic linear impactor.

1.7 Limitations

1. The use of the Hybrid III headform in this study, a rigid model frequently used in head impact testing, was not designed as a fully biofidelic model. Thus, the head acceleration values may be lower due to the vinyl skinform, and less variable than would be commonly observed in humans due to the rigidity of the model (Deng, 1989; Hodgson & Thomas, 1971; Kendall, Walsh, & Hoshizaki, 2012; Seemann, Muzzy, & Lustick, 1986).
2. The University College Dublin Brain Trauma Model (UCDBTM) is commonly used to represent how the brain can deform under different loading conditions, as characterized by the maximum principal strain, and may not provide a global representation of the exact brain motion following an impact to the headform (Horgan & Gilchrist, 2004). Validation of the model was accomplished by comparing the UCDBTM's responses to those of cadaveric testing conducted by Nahum et al. (1977) and Hardy et al. (2001). Brain injury reconstructions based on real-world events allowed for further validations of this model, yielding brain tissue strain and stress values which were comparable to those present in the literature (Doorly & Gilchrist, 2006; Doorly, 2007; Hardy et al., 2001; Nahum, Smith & Ward, 1977; Post et al., 2014; Rousseau, 2014).
3. The finite element model calculates the response of the brain with regards to the material properties of the regions of the head. Kleiven (2007) has proposed variation in material properties, based on hyperelastic and viscoelastic constitutive laws, however the Hybrid III response and the impact parameters during the reconstructions are assumed to be similar, thus a valid estimate of brain tissue deformation (Bain & Meaney, 2000; Horgan & Gilchrist, 2003, 2004; Horgan, 2005; King et al., 2003; Kleiven, 2007; Willinger & Baumgartner, 2003; Zhang et al., 2003; Zhang, Yang, King, & Viano, 2003).

1.8 Delimitations

1. The shoulder was analyzed as opposed to other mechanisms of injuries since the highest prevalence for concussive impacts in professional rugby came from shoulder-to-head collisions (Gardner et al., 2015).
2. The selected cases for this study were chosen from video footage from the official National Rugby League website (www.NRL.com); the cases of diagnosed concussions following shoulder-to-head collisions were chosen based on video footage that will allow for laboratory reconstructions of the impacts. This professional rugby league was chosen due to accurate video analysis being made possible due to the quality of the game film as well as the player injury reports needed for the data analysis. Footage with minimal pixilation where the impact occurred within the camera frame and where the rugby field lines could result in accurate impact characteristic measurements were used.
3. The linear impactor reconstructions are based on video footage of diagnosed concussive shoulder-to-head impacts using Kinovea software. As a result, some cases were not used based on the camera angle and view of the mechanisms of injury, impairing the reconstruction of the cases.

CHAPTER 2: REVIEW OF LITERATURE

Rugby will officially be introduced to the Summer Olympics in 2016, leading to an increased interest in this high-contact sport already played by over five million people world-wide (Rugby, 2014; Savage, Hooke, Orchard, & Parkinson, 2013). With high contact game play comes a high risk of sustaining a concussion and suffering from the ensuing transient or long-term physiological disturbance in brain function caused by a neurometabolic cascade (Harmon et al., 2013). Although tackling accounts for the highest risk of concussive injuries in rugby (Gardner, Iverson, Levi, et al., 2014; Gardner, Iverson, Stanwell, et al., 2014), few biomechanical studies have analyzed tackling mechanics leading to head injuries (Hendricks et al., 2014; McIntosh et al., 2000, 2010; Patton et al., 2013; Seminati et al., 2010; Usman et al., 2011). Understanding system-specific and player position-dependent responses during tackling events leading to concussive injuries in rugby will provide dynamic brain response and cerebral strain metrics during tackles, as well as offer insight on promoting safer game play and tackling techniques.

This review covers epidemiology, pathophysiology and biomechanics in mild traumatic brain injuries associated with professional rugby. Dynamic response (peak resultant linear and rotational accelerations), brain tissue deformation (maximum principal strain), as well as finite element modelling (University College Dublin Brain Trauma Model), will be reviewed in the following section.

2.1 Head Traumas in Sport

2.1.1 Traumatic Brain Injuries

In the field of sports medicine and biomechanics, head traumas are primarily classified as traumatic brain injuries (TBIs) and mild traumatic brain injuries (mTBIs); mTBI is a pathology often interchangeably used with the term "concussion" (Harmon et al., 2013; Kleiven, 2007; Zhang et al., 2001). Approximately 90% of concussions in sport occur without the athlete losing consciousness, thus the reporting of symptoms by the athlete is necessary in order to diagnose the majority of concussions. Therefore, predictions during in-laboratory concussive event reconstructions must be made based on concussive risk probability and the cerebral response (Bain & Meaney, 2000; Gardner, Iverson, Levi, et al., 2014; Ommaya & Gennarelli, 1974; Ommaya, Hirsch, Yarnell, & Harris, 1977; Zhang et al., 2003; 2004). Current studies have reported an association with the linear and rotational accelerations of the head on impact and mTBIs, where the best predictor of concussion attenuation was reported to be rotational acceleration and maximum principal strain (MPS) (Kimpara & Iwamoto, 2012; King et al., 2003; McAllister et al., 2012; Patton et al., 2013; Rowson & Duma, 2013). As the cerebral responses during mTBI events are dependent upon impact characteristics such as mass, compliance, velocity and location, these characteristics must be considered during in-laboratory reconstructions (Gennarelli et al., 1982; Hoshizaki, Post, Kendall, Karton, & Brien, 2013; Kendall, Post, et al., 2012; Pellman et al., 2003; Rousseau & Hoshizaki, 2015; Zhang et al., 2001).

2.1.2 Epidemiology

2.1.2.1 Incidence of Head Trauma in the National Rugby League (NRL)

Approximately four million sports-related concussions are reported annually in the United States. However, estimates of concussive incidents are stated to be as high as eight million cases per year, considering the assumption that approximately 50% of concussions are said to be unreported (Harmon et al., 2013). Concussions in sport have been documented to account for an average of 5-9% of sports-related injuries, while in rugby 15-25% of all injuries are concussions, similar to American football (22%) (Gardner, Iverson, Levi, et al., 2014; Withnall, 2005). Compared to lower-risk sports, this high-contact sport is associated with a high risk of concussion due to elevated collision and tackle rates (Gardner, Iverson, Levi, et al., 2014; Hollis et al., 2009; McIntosh et al., 2009; Withnall, 2005).

Currently, over five million rugby players participate in competitive play world-wide, with a consistently growing athlete population observed over the last decade (Gardner, Iverson, Levi, et al., 2014; McIntosh et al., 2009; Rugby, 2014). Although concussions are less common in the National Rugby League (NRL) than other soft-tissue injuries, such as contusions and abrasions, they represent the highest risk to the athletes (Gardner, Iverson, Levi et al., 2014). In 2014, a new "Concussion Interchange Rule" (CIR) was integrated into the NRL, where 167 cases were documented over the entirety of the season (Gardner, Iverson, Stanwell, et al., 2014). Due to the lack of uniformity in data collection with regards to injury rates in rugby, studies have been inconsistent in reporting incidence of injuries within the International Rugby Board (IRB), specifically Rugby League and Rugby Union, as there have been important variations regarding the injury terms and definitions used (Chalmers, Samaranayaka, Gulliver, & McNoe, 2012; Fuller et al., 2007; Gardner, Iverson, Levi, et al., 2014; King et al., 2014). Since 2011, the IRB has reached a consensus where a common definition for specific injury terms is to be used to favour research uniformity and reporting in the domain of sports medicine and rugby (Gardner, Iverson, Levi et al., 2014).

Tackling has been identified as the most common event in rugby leading to mild traumatic brain injuries (mTBIs) (King et al., 2014). Ranging from 0 to 40 concussions per 1000 player match hours played, a recent video analysis study reports an average of 14.8 concussions per 1000 player match hours (Gardner et al., 2015), with other studies estimating 8.0-17.5 concussions per 1000 player match hours (Gardner, Iverson, Levi, et al., 2014; Gardner, Iverson, Stanwell, et al., 2014; Hollis et al., 2009; McIntosh, 2003; McIntosh et al., 2009; McLellan & McKinlay, 2011). In American football, 24% of professional players are reported to have sustained three or more concussions during their professional careers, and although comparable data in professional rugby is not available, the high exposure to head impact events has resulted in 13-17% of players in the NRL sustaining a concussion each year over three consecutive seasons (Gardner, Iverson, Levi, et al., 2014; Guskiewicz et al., 2005).

2.1.2.2 Collisions in Rugby

Rugby has the third highest incidence of mTBIs in sport, with tackles being identified as the component of the game that yields the highest rate of injury, as well as being the most common cause of concussion, regardless of level of play (Best et al., 2005; Hendricks et al., 2014; Hollis et al., 2011; King et al., 2010; McIntosh et al., 2000, 2010). A reported 19-55 tackles occur per match, these rates vary depending on player position; above-waist tackles encompass the largest concussive mechanisms of injury (80-83%). Player position does influence the risk of concussions, where backs represent 60% of the concussed players (Best et al., 2005; Gardner et al., 2015; Hendricks et al., 2014). Illegal tackles and plays, mostly shoulder-to-head tackles, account for 25-29% of concussions (Brown, Brughelli, & Hume, 2014; Gardner et al., 2015; Gardner, Iverson, Levi, et al., 2014; Gardner, Iverson, Stanwell, et al., 2014; McLellan & McKinlay, 2011).

The most common concussive collisions were recently reported to be shoulder-to-head impacts (35%) followed by knee-to-head and head-to-head impacts each at 20% respectively (Gardner et al., 2015). Of these impacts resulting in mTBI, 86% were to the players' tempero-parietal region, towards the anterior half of the head, with the shoulder-thorax region of the impacting player coming into direct contact with the head in 97% of cases (McIntosh et al., 2000). Although the majority of studies report factors contributing to moment and energy transfer, such as player speed and mass, specific mechanism of injury and impacting body have not been thoroughly documented with regards to concussive risk (Fréchède & McIntosh, 2009; McIntosh et al., 2000, 2010; Patton et al., 2013). Due to the aforementioned potentially skewed injury reports, and the discrepancy in the reporting of concussion incidence in ball carriers and defensive tacklers, quantifying cerebral response based on the most commonly-occurring concussive events and their specific impact characteristics is warranted.

2.2 Biomechanical Considerations of Mild Traumatic Brain Injury (mTBI)

2.2.1 Concussions and Brain Injury Predictors

Studies have reported an association with the peak resultant linear and rotational accelerations on impact and mild traumatic brain injuries (mTBIs), however, the best predictor of concussive risk was recently reported to be rotational acceleration and maximum principal strain (MPS) (Deck & Willinger, 2008; Fréchède & McIntosh, 2009; Kimpara & Iwamoto, 2012; King et al., 2003; Kleiven, 2007; McAllister et al., 2012; McIntosh, 2004; Patton et al., 2013; Rowson & Duma, 2013; Willinger & Baumgartner, 2003; Zhang et al., 2004). As dynamic response is dependent upon specific impact characteristics, such as compliance and inbound velocity, it is important to ensure that the impact parameters and system during the reconstructions accurately represent the real-world event. Specifically, impact duration has been reported to affect dynamic response and brain tissue deformation, thus system compliance must be integrated as a validated impact parameter during rugby head impact reconstructions (Gennarelli et al., 1982, 1987; Hoshizaki et al., 2013; Kendall, Post, et al., 2012; Kleiven, 2003; Pellman et al., 2003; Rousseau & Hoshizaki, 2015; Zhang et al., 2001). Consequently, integrating the specific impacting system and its accurate compliance during head impact reconstructions, shoulder-to-head for example, is necessary to ensure accurate reconstructions of these events (Rousseau & Hoshizaki, 2015; Rousseau, Post, & Hoshizaki, 2009).

2.2.2 Video Analysis

Video analysis is commonly employed by researchers to obtain the kinematic parameters for impact reconstruction (Fréchède & McIntosh, 2009; Gardner et al., 2015; Gardner, Iverson, Stanwell, et al., 2014; Kleiven, 2007; McCrory et al., 2013; McIntosh et al., 2000; Pellman et al., 2003; Seminati et al., 2010; Usman et al., 2011). This method allows for variables, such as velocity and impact location, to be documented and calculated, in order to reconstruct real-world sports injuries with relative accuracy (Fréchède & McIntosh, 2009; Gardner et al., 2015; Gardner, Iverson, Stanwell, et al., 2014; McIntosh et al., 2000, 2010; Rousseau & Hoshizaki, 2015). Most recent rugby research involves 1990s video reconstructions from McIntosh et al. (2010), where pixilation and VHS-quality videos, using two-dimensional footage to perform reconstructions, are amongst the limitations. During these reconstructions, assumptions were made in some cases where portions of the tackling events occurred out of the camera frame (Fréchède & McIntosh, 2007, 2009; Patton et al., 2013). High-quality video footage along with motion capture software, such as Kinovea (open source, kinovea.org), provide tools to obtain velocity and location at impact, as well as other elements such as equipment worn by the concussed player (Rousseau & Hoshizaki, 2015). This analysis provides the variables which will guide the methodology based on the mechanism of injury of the head injury and the impact location.

2.2.3 Dynamic Response

2.2.3.1 Influence of Impact Conditions on Dynamic Response

Compliance, inbound velocity and impact location have been shown to affect the response of the brain during impacts, and must therefore be considered in this study. Defining potential differences in brain tissue deformation and dynamic response during different rugby collisions is necessary in order to attempt to ensure the validity of these head injury reconstructions (Gennarelli et al., 1982, 1987; Hoshizaki et al., 2013; Kendall, Post, et al., 2012; Kleiven, 2003; Pellman et al., 2003; Rousseau & Hoshizaki, 2015; Zhang et al., 2001). Current studies have reported an association with the linear and rotational accelerations of the head on impact and mTBIs, where the best predictor of concussion was reported to be rotational acceleration and maximum principal strain (MPS) (Kimpara & Iwamoto, 2012; King et al., 2003; McAllister et al., 2012; Patton et al., 2013; Rowson & Duma, 2013).

2.2.3.1.1 Location of Impact

Impact location has been reported to influence brain response and resulting mTBIs, having a direct effect on dynamic response and brain tissue deformation values (Gennarelli et al., 1982, 1987; Gurdjian, Lissner, Latimer, Haddad, & Webster, 1953; Hodgson, Thomas, & Khalil, 1983; Zhang et al., 2001). Although in-laboratory reconstructions have been completed based on head injuries in rugby, the impact locations have not been reported (Fréchède & McIntosh, 2007, 2009; Patton et al., 2013). There is a reported increased risk of sustaining a concussion with impact locations at the side of the head as rotational acceleration increases, as demonstrated by studies on primates and the effects of impact location (Delaney et al., 2006; Gennarelli et al., 1982, 1987; Kleiven, 2003; Zhang et al., 2004). As previously mentioned, linear acceleration as an individual variable is not an effective indicator of concussive risk, hence, direct frontal impacts should not be used to represent head impacts of varying directions (Gennarelli, Abel, Adams, & Graham, 1979; Kleiven, 2003; Zhang et al., 2001). Similarly, finite element analyses (FEA) of dynamic response derived from physical models (Hybrid III headforms) have demonstrated a greater risk of sustaining a concussive impact with a lateral impact to the head. With the greatest incidence of head impacts in rugby being reported on the player's tempero-parietal region, results yielding higher levels of shear and positive pressure at the centre of the brain with these lateral impacts must be considered (McIntosh et al., 2000; Zhang et al., 2004).

2.2.3.1.2 Inbound Velocity

Dynamic response has been reported to be influenced by inbound velocity during collisions. Peak linear and rotational accelerations are positively correlated with impact velocity, which has been reported with both hockey and football equipment on Hybrid III headforms using a pneumatic linear impactor (Kendall, Post, et al., 2012; Post, Hoshizaki, & Gilchrist, 2012; Rousseau, Post, & Hoshizaki, 2009; Rousseau, Post, Hoshizaki, et al., 2009). With mean impact closing speeds reported to be 3.2-9.3m/s during professional rugby game play, it is important to take these varying inbound velocities into consideration during head impact reconstructions (Fréchède & McIntosh, 2009; Gabbett, 2013; McIntosh et al., 2000; Patton et al., 2013). Past research on head impacts and protective equipment in rugby have used monorail drop systems at 0.3-0.6m, representing 2.4-3.4m/s during reconstructions (McIntosh, 2004; McIntosh et al., 2009), however it has been reported that these reconstructions are not representative of concussive impacts in rugby (Fréchède & McIntosh, 2009).

2.2.3.1.3 Compliance

Energy-absorbing materials have been reported to influence the aforementioned brain deformation and dynamic response metrics. Therefore, the impacting system used for the reconstruction must be taken into consideration when quantifying concussive injury risk, as a velocity modification over a greater distance increases the likelihood of a decreased acceleration response. The choice of impacting material is essential, where one must ensure that a "bottoming-out" effect does not occur (Gennarelli et al., 1982, 1987; Hoshizaki et al., 2013; Hutchinson, 2012; Kendall et al., 2012; Kleiven, 2003; Pellman et al., 2003; Rousseau, 2014; Rousseau & Hoshizaki, 2015; Zhang et al., 2001). The use of cadaveric models to quantify duration of impact, as used in past rugby head injury reconstructions, may not lead to a biofidelic impact response (Bain & Meaney, 2000; Fréchède & McIntosh, 2009; Horgan & Gilchrist, 2003, 2004; Horgan, 2005; King et al., 2003; Kleiven, 2007; Nahum et al., 1977; Patton et al., 2013; Willinger & Baumgartner, 2003; Zhang et al., 2003, 2004).

The Mathematical Dynamic Models (MADYMO) software was validated to primarily represent segment and full-body impacts during motor vehicle accidents (de Lange et al., 2005; MADYMO, 2011). In order to quantify dynamic response in rugby impacts, Fréchède & McIntosh (2009) used the MADYMO "facet mid-size male occupant model", based on post-mortem human substitutes (PMHS) responses during higher severity impacts (de Langer et al., 2005; Fréchède & McIntosh, 2007, 2009; MADYMO, 2011). This cadaveric model was validated using an impactor with the PMHS in a seated position, yielding dynamic response curve durations of approximately 12ms (de Lange et al., 2005; MADYMO, 2011; Meyer et al., 1994). Real-world shoulder-to-head dynamic response values yielded 20-25ms impact durations in a laboratory setting *(Appendix A)* (Rousseau, 2014).

As discrepancy is present with regards to the reporting of concussive risk in ball carriers and defensive tacklers (Gardner et al., 2015; Gardner, Iverson, Levi, et al., 2014; Gardner, Iverson, Stanwell, et al., 2014; King et al., 2010, 2014), and considering the role of impact duration on dynamic response and brain tissue deformation, compliance and mechanism of injury must be integrated as a validated impact parameter during rugby head impact reconstructions (Rousseau & Hoshizaki, 2015).

2.2.3.1.4 Impact Mass

The effect of the impacting mass has been reported to vary based on anticipation of impact in high contact sports such as American football, where the opportunity to prepare for an impact will lead to a decrease in the effect of the impact mass leading to measurements which favour linear acceleration (Barth et al., 2001; Pellman et al., 2003). As the impact locations to the head in professional rugby have been reported to be primarily to the tempero-parietal regions of the injured players, a region which favours the tackled player's anticipation of impact, a low effect of impact mass would be anticipated (McIntosh et al., 2000, 2010; Pellman et al., 2003).

The impact mass has been reported to influence the cerebral response with impacting masses below 10kg (Hodgson, 1967; Karton et al., 2013). Metal cylinders, weighing between 0.9kg and 7.9kg, were used by Hodgson (1967) in order to study facial bone tolerance to focal injuries. With an increased mass resulting in an increase in force, time to fracture was reported to decrease (Hodgson, 1967), however Karton et al. (2013) demonstrated that such an influence attained a plateau when the striking mass was over 10kg. MADYMO software impacting mass, as used for past rugby collision reconstructions, has been reported to be validated using an impacting mass varying from 10.3kg to 23.4kg (Fréchède & McIntosh, 2007, 2009; MADYMO, 2011). As the effective mass of the shoulder during shoulder-to-head collisions in ice hockey has recently been reported to be 13.1kg, this effective mass was used for the defensive tackler and ball carrier impacting systems during the rugby impact reconstructions in this thesis (Rousseau, 2014).

2.2.3.2 Linear Acceleration

Linear acceleration has previously been reported to be an indicator of measurement of focal head impact injuries, such as skull fractures, by means of pressure gradient quantifications (Gennarelli, Thibault, & Ommaya, 1972; Gurdjian, 1975; Holbourn, 1943; King et al., 2003; Ommaya & Gennarelli, 1974). With current research demonstrating higher prediction of concussive risk via maximum principal strain (MPS) measures (McAllister et al., 2012), linear acceleration was initially used as the sole method for head impact analyses (Holbourn, 1943). Linear accelerations favour focal, or traumatic, injuries, thus animal models were used to evaluate these linear-dominant injuries. Varying changes in intracranial pressure, caused by acceleration and deceleration, were reported during these impacts (Gurdjian, 1975). Holbourn et al. (1943) hypothesized that rotational acceleration would also influence head impact outcomes, however due to the challenges in obtaining rotational acceleration, this component was not initially used. This theory was tested on mongrel dogs, where Gurdjian and colleagues (1963) reported the necessity of measuring linear acceleration in order to consider the mechanism of injury, with further research also stating that rotational acceleration was not a stand-alone measure of mTBI (Gurdjian, Lissner, & Patrick, 1963; Ommaya & Hirsch, 1971; Ommaya, Hirsch, & Martinez, 1966).

A comparison between impacting loads (direct and whiplash) was studied in monkeys, where results demonstrated that impulse loading, or whiplash, required higher mean acceleration than direct impacts. Thus, rotational acceleration was stated as being a minor contributor to mTBIs, compared to linear acceleration (Ommaya & Hirsch, 1971; Ommaya et al., 1966). Another study was conducted on monkeys, where linear and rotational accelerations were separately analyzed in order to determine their contribution to risk of head injury. This study supported linear acceleration as the prime cause of concussions from direct impacts to the head (Ono, Kikuchi, Nakamura, Kobayashi, & Nakamura, 1980). Fréchède & McIntosh (2009) have reported an average of 766 m/s^2 for linear acceleration in rugby head collisions. Although other components of dynamic response and brain tissue deformation have demonstrated higher correlations with risk of concussion, considering linear acceleration is still considered to be valid due to its role in safety equipment evaluation and the presence of injury thresholds in the literature (Rowson & Duma, 2013).

2.2.3.3 Rotational Acceleration

As previously stated, rotational acceleration has demonstrated greater association with risk of concussion due to the influence of three dimensional kinematics on brain tissue deformation and concussions (Holbourn, 1943; Kleiven, 2007; Post et al., 2014). The brain-skull interface was demonstrated by Holbourn (1943) using a rotating flask filled with water in order to provide a concrete example of water particle dissociation from the sides of the glass container. This demonstrated that rotational acceleration led to axonal damage, caused by large shear strains, as opposed to linear, or translational, forces. Further research on the linear-rotational acceleration interaction demonstrated the greater role of rotational acceleration on the shear stress and strain of the brain leading to concussions and intra-cranial haemorrhages, with little linear acceleration contribution (Gennarelli et al., 1972; Hodgson & Thomas, 1979; Holbourn, 1943; King et al., 2003; Ommaya & Gennarelli, 1974; Post et al., 2013; Zhang et al., 2001). As the location of the impact and the impacting system are not stated in past rugby head injury reconstructions, it is difficult to compare event-specific rotational accelerations, however a 7770 rad/s^2 average rotational acceleration for concussions was reported (Fréchède & McIntosh, 2009).

2.2.4 Brain Tissue Deformation

Finite element (FE) modeling, which uses maximum principal strain (MPS) to quantify brain tissue deformation, allows for an approximation of brain tissue stress and has been reported as having the highest correlation in brain injury predictions (Deck & Willinger, 2008; Horgan & Gilchrist, 2004; Nahum, Smith, & Ward, 1977; Post & Hoshizaki, 2015). Laboratory reconstructions using physical models, such as the Hybrid III headform and the Mathematical Dynamic Models (MADYMO), allow for dynamic response analyses with similar head motion patterns to that of a human. These physical models, based on human cadavers, have limited success in brain injury prediction; FE modeling has been proposed to be a more appropriate method to estimate brain strains and concussions (Brain & Meaney, 2000; Horgan & Gilchrist, 2003, 2004; Horgan, 2005; King et al., 2003; Kleiven, 2007; Nahum et al., 1977; Willinger & Baumgartner, 2003; Zhang et al., 2003, 2004). The brain's viscoelastic properties were demonstrated with animal testing, and along with its low shear modulus, make it more vulnerable to functional disturbances and cerebral strain (Galbraith, Thibault, & Matteson, 1993; Mao, Zhang, Yang, & King, 2006; Nahum et al., 1977).

King et al. (2003) used animal models in order to establish the relationship between brain tissue deformation and dynamic response in humans, followed by studies that brought forth results supporting the role of rotational acceleration in concussive injuries (Forero Rueda & Gilchrist, 2009; Zhang et al., 2006). Further research has reported a greater correlation between resultant acceleration curves and brain injuries, rather than peak resultant accelerations (Post et al., 2012). Based on previous 1990s video analyses and MADYMO simulations, finite element analyses have been conducted on rugby head injuries (Fréchède & McIntosh, 2007, 2009; McIntosh et al., 2000), with average reported maximum principal strain values of 0.27 (Patton et al., 2013).

2.2.5 Proposed Injury Thresholds

In an attempt to link mechanism of injury with specific dynamic response and brain tissue deformation variables, researchers, including Zhang (2001), have described tolerance levels and proposed injury thresholds (Kleiven, 2007; Willinger & Baumgartner, 2003; Zhang et al., 2001, 2004). Linear and rotational accelerations have been collected from real-world head impact reconstructions in order to establish concussive risk during sporting events, such as the National Football League (NFL) physical models by Zhang et al. (2004). In this suggested risk assessment, a 50% chance of concussion injury is represented by a 5600 rad/s^2 rotational acceleration (Zhang et al., 2004).

2.2.5.1 Dynamic Response and Risk

Tolerance level estimations using dynamic response values have been produced using a variety of methodologies, including in-laboratory reconstructions from video footage (Kleiven, 2007; Zhang et al., 2004). 25%, 50% and 80% probability of sustaining a concussion for peak linear acceleration were deemed to be 66g, 82g, and 106g respectively, whereas rotational acceleration tolerance levels were estimated at 4600 rad/s^2, 5900 rad/s^2 and 7900 rad/s^2 respectively (Zhang et al., 2004) *(Table 1)*. Real-world shoulder-to-head ice hockey collisions allowed for quantification of risk, where 50% probability of sustaining a concussion was reported to be at 4600 rad/s^2 at 20-25ms impact durations (Rousseau, 2014) *(Table 1)*. Other researchers reported different dynamic response thresholds, with probable concussion values as little as 28.3g and 3000 rad/s^2 (Willinger & Baumgartner, 2003) *(Table 1)*, however the proposed thresholds derived from football head impacts by Zhang et al. (2004) are most frequently referenced in the literature.

Table 1. Comparison of dynamic response values for 50% likelihood of concussion reported by previous impact collision reconstructions in sport

Sport	Source	Peak Linear Acceleration (g)	Peak Rotational Acceleration (rad/s^2)
American Football	Newman et al. (2004)	78	6822
American Football	Pellman et al. (2003)	98	6432
American Football	Zhang et al. (2004)	82	5900
American Football	Willinger & Baumgartner (2003)	28.3	3000
Ice Hockey	Rousseau (2014)	23	4600

2.2.5.2 Brain Tissue Deformation and Risk

Similar to physical models and dynamic response concussive risk probability, finite element (FE) modeling has been used to establish concussive risk based on brain tissue deformation values using von Mises stress and maximum principal strain (MPS) values. Combining these two variables to assess risk has been reported to yield the highest correlation when predicting concussive injury (Kleiven, 2007; Willinger & Baumgartner, 2003; Zhang et al., 2004). A 50% risk of moderate neurological lesions has been associated with von Mises stress values of 18 kPa and 38 kPa, and a 0.37 MPS value by Willinger and Baumgartner (2003); whereas Zhang et al. (2004) proposed a threshold value of 50% risk of sustaining a concussion to be at 7.8 kPa, with grey matter strain levels at 0.19 *(Table 2)*. Further studies proposed results of 8.4 kPa and MPS values between 0.14 and 0.30 (Kleiven, 2007; Rousseau, 2014) *(Table 2)*. Recent research by

Rousseau (2014) also integrates duration of impact in concussive injury predictions, and demonstrates different predictors for shoulder-to-head as well as elbow-to-head collisions, demonstrating the importance of using event-specific impacting systems when quantifying brain tissue deformation.

Table 2. Comparison of strain tolerance levels for 50% likelihood of concussion reported by previous finite element head model simulation studies

Sport	Source	Maximum Principal Strain
Rugby	Patton et al. (2013)	0.27
American Football	Zhang et al. (2004)	0.19
American Football	Kleiven (2007)	0.26
Ice Hockey	Rousseau (2014)	0.24

2.3 Summary

Concussions are prominent in high-contact sport, with rugby players at high risk for sustaining head injuries during game play. This chapter reviewed the direct influence of specific impact characteristic parameters on dynamic head response and brain tissue deformation metrics. As there is a discrepancy in the reporting of concussion incidence between ball carriers and defensive tacklers in professional rugby, the incorporation of player position-dependent conditions during impact reconstructions was investigated. In order to accurately represent concussive impact events for in-laboratory reconstructions, compliance was calculated for shoulder-to-head collisions. As dynamic response, especially peak resultant rotational acceleration, and brain tissue deformation were the main contributors to predicting concussive injury during direct impacts to the head, the accurate impact characteristics, such as inbound velocity and compliance, must be used in these reconstructions. These impact characteristics were not accurately described in previous studies involving rugby concussive impacts. These reconstructions allowed for a comparison of dynamic response and brain tissue deformation metrics between ball carriers and defensive tacklers and the results of this study will provide knowledge that will guide future research methodology and contribute to concussive risk reduction strategies.

CHAPTER 3: METHODOLOGY

The main objective of this thesis is to compare brain tissue deformation and dynamic response values during rugby shoulder-to-head concussive impacts between ball carriers and tacklers in professional rugby. The video footage of diagnosed concussive shoulder-to-head impacts during National Rugby League (NRL) game play was captured and analyzed using Kinovea software (open source, kinovea.org). This software was used to determine inbound velocity and location of impact during collisions resulting in concussions in order to define the shoulder-to-head impacting parameters for accurate reconstruction. Using a pneumatic linear impactor and a Hybrid III 50th percentile headform attached to an unbiased neckform (Walsh, Rousseau, & Hoshizaki, 2011), the dynamic response metrics were measured for comparison between ball carriers and defensive tacklers in professional rugby. A "3-2-2-2" array of nine single-axis accelerometers was mounted in the headform in order to measure peak resultant linear and rotational accelerations (Padgoankar, Krieger, & King, 1975). The University College Dublin Brain Trauma Model (UCDBTM), a finite element (FE) model, was used to determine maximum principal strain (MPS) and deformation response characteristics using the inputted dynamic response metrics from the impacts. This deformation response allowed to establish the stress sustained by the grey and white matter of the brain upon impact, and thus establish cerebral response during each reconstructed concussive event, based on reported concussive injury thresholds (Kleiven, 2007; Post et al., 2012; Willinger & Baumgartner, 2003; Zhang et al., 2004).

3.1 Impact Characteristic Data Collection

3.1.1 Study Population and Inclusion/Exclusion Criteria

The National Rugby League (NRL) is comprised of sixteen professional rugby teams, with male athletes ranging from eighteen to thirty-five years of age, where twenty-five regular-season games are played every season (Gardner, Iverson, Stanwell, et al., 2014). Fifty-four (54) video clips of concussive shoulder-to-head impacts were recaptured from full match replays of the 2014 regular season on the official NRL website (www.NRL.com) using WM capture software (San Anselmo, CA). Footage of these events were provided to the Neurotrauma Impact Science Laboratory of the University of Ottawa, with corresponding diagnosed concussion reports by the treating medical doctors for each respective NRL team. Videos which included only one shoulder-to-head impact, visible within the camera frame, with no apparent other head impacts or fall, were included in the data set.

Shoulder impacts leading to diagnosed concussions were captured using WM capture software at a frame rate of 25 fps and a bit rate of 6000 fps. Initial video analysis using Kinovea software was conducted in order to determine whether the video was suitable for reconstruction. Inclusion criteria included appropriate video quality, where the image was clear and free of pixel deformity, a visual of the site of impact, as well as comparative field measurements. This inclusion criteria was necessary to allow for the collection of the impact parameters needed for reconstructions – inbound velocity, angle of impact and location of impact. Professional-level rugby footage provided high-quality game videos, differing camera angles, and instant replays of concussive impacts in order to allow for appropriate reconstructions and decrease source of error (Gardner, Iverson, Stanwell, et al., 2014).

3.1.2 Video Analysis

Video analysis, a method commonly employed by researchers during impact reconstructions, was used in order to reconstruct the real-world concussive events (Fréchède & McIntosh, 2009; Gardner et al., 2015; Gardner, Iverson, Stanwell, et al., 2014; McIntosh et al., 2010; Rousseau & Hoshizaki, 2015). Kinovea software was used to determine the velocity and location of impact during the shoulder-to-head collisions; the NRL field dimensions allowed for scaled referencing during this analysis *(Figure 1).* The velocity of impact was measured using the distance between the concussed player and their opponent *(Figure 2)*. In order to account for the potential errors associated with this conversion method, a higher and lower inbound velocity (±5%) during the in-laboratory reconstructions was incorporated in the methodology (Rousseau, 2014).

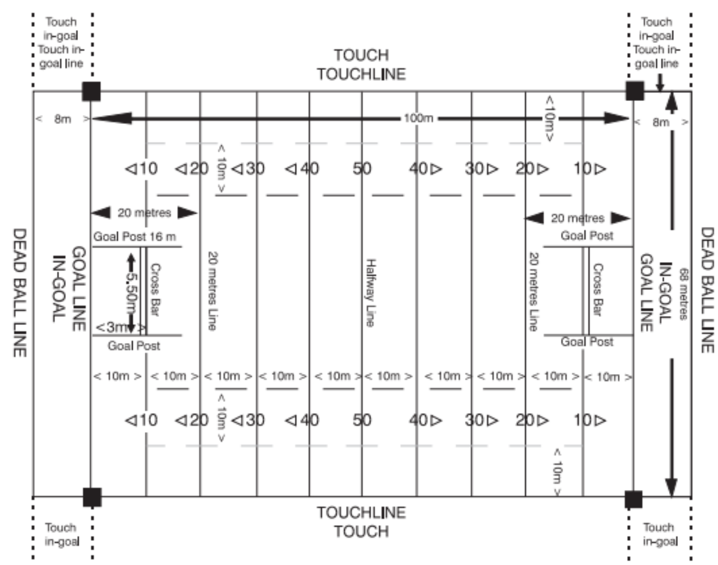

Figure 1. National Rugby League field dimensions (Rugby League Laws of the Game)

Figure 2. Kinovea velocity analysis example; distance between two players, five frames prior to impact

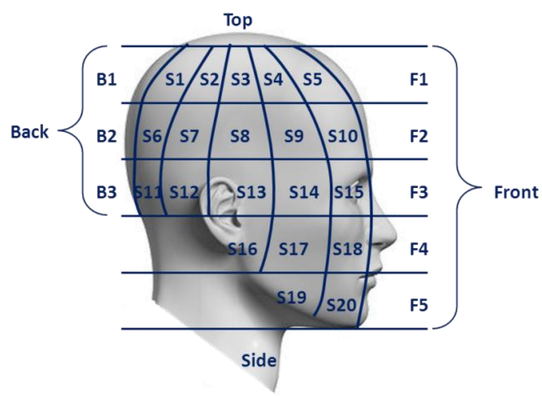

Figure 3a. The grid reference to impact location

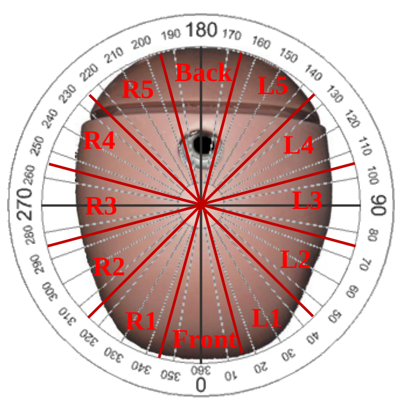

Figure 3b. The vertical offset impact location reference based on the centre of gravity

The location of impact was categorized using two references *(Figure 3a,b)* in order to facilitate in-laboratory reconstructions using the Hybrid III 50th percentile headform *(Figure 4)*. The first component of the location characteristic was a grid location based on a vertical and horizontal placement on the head, and the second component was the vertical offset with regards to the centre of gravity.

The position of the player's head and neck upon impact allowed for description of impact angle as well as orientation angle for the in-laboratory reconstructions *(Figure 4a,b,c,d)*.

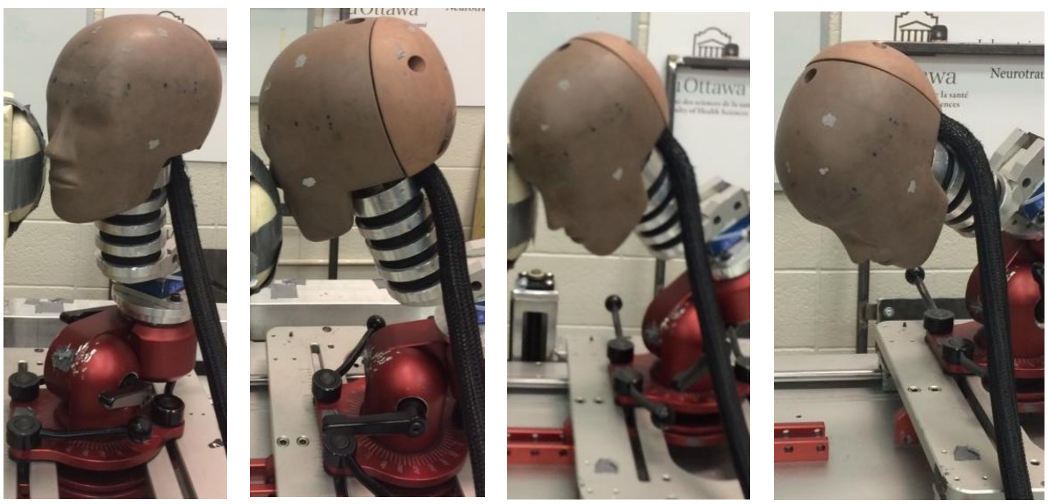

Figure 4. Hybrid III headform in a neutral (a: 0^{o}), forward (b: 15^{o}; c: 30^{o}; d: 45^{o}) position, attached to unbiased neckform

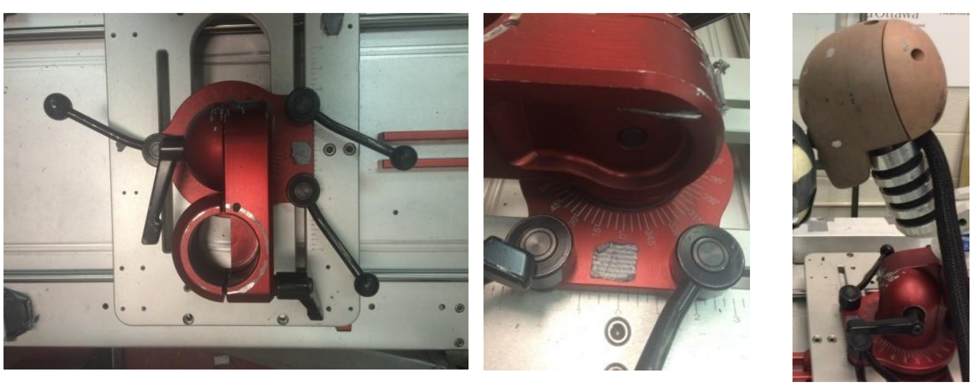

Figure 5. Hybrid III headform orientation angles (0-360^{o}) position, (a: aerial view; b: frontal view; c: headform angled at 15^{o} during impact)

3.2 Dynamic Impact Testing

3.2.1 Hybrid III 50th Percentile Headform and Unbiased Neckform

A Hybrid III 50th percentile headform fitted with nine single-axis accelerometers mounted in a "3-2-2-2" array was placed on an unbiased neckform *(Figure 5)* (Padgoankar et al., 1975; Walsh et al., 2012). In order to account for the multidirectional neck range of motion upon impact, the unbiased neck was used (Walsh et al., 2012). The physical model was positioned according to the reconstruction data collected during the video analysis, with the position of the impacting arm corresponding to the referenced locations *(Figure 3a,b)*, and the neckform angled based on the real-world event *(Figure 4)* (Walsh et al., 2011). The accelerometers' signals were collected at 20 000 Hz by the TDAS Pro laboratory software (DTS, Sealbeach, CA); using the SAE J211 Class 1000 Protocol (SAE, 2007), these signals were filtered using a 1650 Hz Butterworth dual low-pass filter. The dynamic response obtained included peak resultant linear and rotational accelerations.

Figure 6. Hybrid III headform attached to unbiased neckform

3.2.2 Pneumatic Linear Impactor

The pneumatic linear impactor was used in order to reconstruct the shoulder-to-head collisions at the calculated velocities *(Figure 6)*. The frame was mounted with an impacting arm corresponding to the position of the player's impacting upper extremity during the shoulder-to-head concussive event. The linear impactor's frame and its attached impacting arm was fitted with material analogous to shoulder compliance (20-25ms) for each respective collision *(Appendix A)* (Rousseau, 2014), where the shoulder compliance was determined during an experiment involved ten (10) competitive-level rugby player participants who impacted the Hybrid III headform using a shoulder-to-head tackling technique *(Appendix A)*, as well as an impacting arm of 13.1kg (Karton, 2013; Kendall, Post, Rousseau, & Hoshizaki, 2014; Rousseau, 2014; Rousseau & Hoshizaki, 2015). Furthermore, inter-tester methodology reliability testing has been conducted on the linear impactor, where significant differences between testers were present when placement of protective helmet equipment was required; however with a bare headform, large standard deviations and a validated test protocol, no significant differences were observed (Hoshizaki, 2013).

Current International Rugby Board headform compliance drop test specifications include a 30-60cm high, 14 J impact drop, representing a 2.4-3.4 m/s impact, which has been suggested to not be representative of concussive events in rugby (Fréchède & McIntosh, 2009; McIntosh, 2004). The pneumatic linear impactor has been used in shoulder-to-head collision reconstructions, and is more representative of such collisions and the reported 3.2-9.3 m/s inbound velocities in rugby (Gabbett, 2013; Kendall et al., 2014; Rousseau & Hoshizaki, 2015).

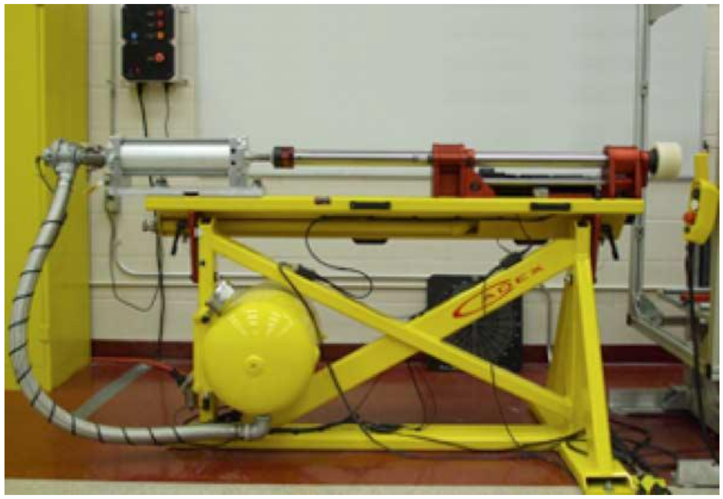

Figure 7. Pneumatic linear impactor

3.2.3 Finite Element Model

The University College Dublin Brain Trauma Model (UCDBTM) was used as the finite element (FE) model in order to analyze the brain tissue deformation of both white and grey matter during impact *(Figure 7)*. Derived from cadaveric diagnostic imaging, this model is comprised of 26 000 hexahedral elements, and allows for the quantification of material characteristics of brain tissue during head collision events, yielding the shear characteristics of the brain's viscoelastic properties (Horgan & Gilchrist, 2003, 2004; Nahum et al., 1977; Post et al., 2012). Past studies have established this model's material characteristics (Kleiven & von Holst, 2003; Ruan, 1994; Willinger, Taleb, & Pradoura, 1995; Zhou, Khalif, & King, 1995) *(Tables 3 and 4)*.

Table 3. Material properties of UCDBTM			
Material	Young's modulus (MPa)	Poisson's Ratio	Density (kg/m^3)
Scalp	16.7	0.42	1000
Cortical Bone	15000	0.22	2000
Trabecular Bone	1000	0.24	1300
Dura	31.5	0.45	1130
Pia	11.5	0.45	1130
Falx	31.5	0.045	1140
Tentorium	31.5	0.45	1140
CSF	Water	0.5	1000
Grey Matter	Hyperelastic	0.49	1060
White Matter	Hyperelastic	0.49	1060

Table 4. Material characteristics of the brain tissue for the UCDBTM				
Material	G_0	G_∞	Decay Constant (Gpa)	Bulk Modulus (s^{-1})
Cerebellum	10	2	80	2.19
Brain Stem	22.5	4.5	80	2.19
White Matter	12.5	2.5	80	2.19
Grey Matter	10	2	80	2.19

The brain tissue's material behaviour was modelled as being dependent of the shear relaxation modulus with regards to the viscoelastic properties in shear with a deviatoric stress rate (Horgan & Gilchrist, 2003). The brain's compressive behaviour is considered to be elastic, therefore the following equation was used to define the nature of the shear characteristics with regards to its viscoelastic behaviour:

$$(1) \quad G(t) = G_\infty + (G_0 - G_\infty)e^{\beta t}$$

where G_∞, is defined as the long term shear modulus, G_0, is the short term shear modulus, and β is the decay factor (Horgan & Gilchrist, 2003). Simulation of the skull-brain interface was achieved by modelling the cerebral spinal fluid in order to allow it to behave like water. To do so, solid elements with low shear modulus and a high bulk were applied. A friction coefficient of 0.2 was applied, and no separation was assigned to the contact interaction at the skull-brain interface (Miller et al., 1998).

A hyperelastic model of the brain shear is represented by the following equation:

$$(2) C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-\frac{t}{0.008}} + 1103e^{-\frac{t}{0.15}}(Pa)$$

where C_{10} and C_{01} are temperature-dependent material parameters and t is time represented in seconds (Horgan & Gilchrist, 2003).

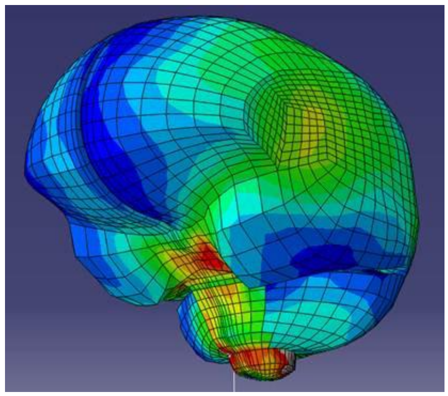

Figure 8. A brain UCDBTM model (low (blue) to high (red) deformation colour gradient)

3.3 Reconstruction Protocols

Based on the Kinovea video reconstruction analysis and the inclusion and exclusion criteria, twenty (20) shoulder-to-head concussive events were reconstructed (ten (10) of each condition (ball carrier concussion; tackler concussion)). The bare Hybrid III 50th percentile headform was impacted six (6) times at each corresponding impact velocity and location for each concussive event *(Appendix C)*; three impacts were conducted at the low-end velocity, which represents a velocity which was 5% less than the target velocity, as well as three impacts at the high-end velocity, which represents a velocity that is 5% greater than the target velocity, for a total of 120 impacts.

3.4 Statistical Analysis

The data were categorized by maximum principal strain (MPS), peak resultant linear and rotational accelerations, and impact characteristics (velocity, location and angles). The data for MPS, dynamic response and velocity were expressed as means and standard deviations. Independent sample t-tests were performed to analyze variance within the dependent variables (MPS, peak resultant linear acceleration, peak resultant rotational acceleration) and inbound velocity in the two conditions (ball carrier concussion; defensive tackler concussion), as the data was normally distributed. A secondary descriptive analysis was completed to describe variance between location of impact as well as forward and orientation angles. A power analysis was completed, and with ten (10) cases in each condition (ball carrier and defensive tackler), 80% power was obtained ($\beta = 0.2$) for peak resultant linear acceleration, peak resultant rotational acceleration and maximum principal strain. The analysis was conducted using SPSS 19 for Windows statistical software (IBM Inc., Armonk, NY, USA).

CHAPTER 4: RESULTS

Fifty-four (54) concussive shoulder-to-head cases were obtained and assessed, using the exclusion criteria, twenty (20) were accepted for this study; ten (10) defensive tackler concussive cases and ten (10) ball carrier concussive cases. For all twenty (20) concussive events, the reconstructions were based on real-world diagnosed concussive shoulder-to-head collisions; the location, velocity and angle of impact varied in each case *(Appendix B; C; D).* The twenty (20) shoulder-to-head concussive impact events were physically reconstructed using a linear impactor in order to obtain dynamic response, and finite element analysis yielded brain tissue deformation metrics. The following section includes the dynamic response results – peak resultant linear and rotational accelerations, the finite element modeling results – maximum principal strain, and the statistical analysis.

Three (3) independent sample t-tests were performed to analyze variance within the dependent variables (peak resultant linear acceleration, peak resultant rotational acceleration, MPS) in the two differing conditions (ball carrier concussion (n=10); tackler concussion (n=10)).

4.1 Dynamic Response

The complete dynamic response results for both concussive event conditions can be found in *Appendix C*; *Table 5* contains the dependent variable results, expressed in peak means, for both concussive event conditions, with standard deviations in parentheses. The reconstructions were completed based on real-world events, hence some impact characteristics varied; due to the shoulder-to-head concussive events, mass and compliance remained constant, whereas inbound velocity and impact location varied in each case.

Table 5. Dynamic response and brain tissue deformation results for defensive tackler and ball carrier shoulder-to-head reconstructions

Dynamic Response Measurement	Ball Carrier	Defensive Tackler
Velocity (m/s)	6.43 (1.19)	6.05 (1.26)
Peak Resultant Linear Acceleration (g)	34.16 (18.00)	27.67 (13.23)
Peak Resultant Rotational Acceleration (rad/s^2)	4419 (1837)	3812 (1002)
Maximum Principal Strain	0.414 (0.124)	0.408 (0.099)

4.1.1 Peak Resultant Linear Acceleration

There was no significant difference in peak resultant linear acceleration during shoulder-to-headform reconstructions, t (18) = .918, p = 0.371, when the defensive tackler sustained the concussion (M = 27.67 (13.23) g) in comparison to the ball carrier (M = 34.16 (18.00) g) *(Figure 8).*

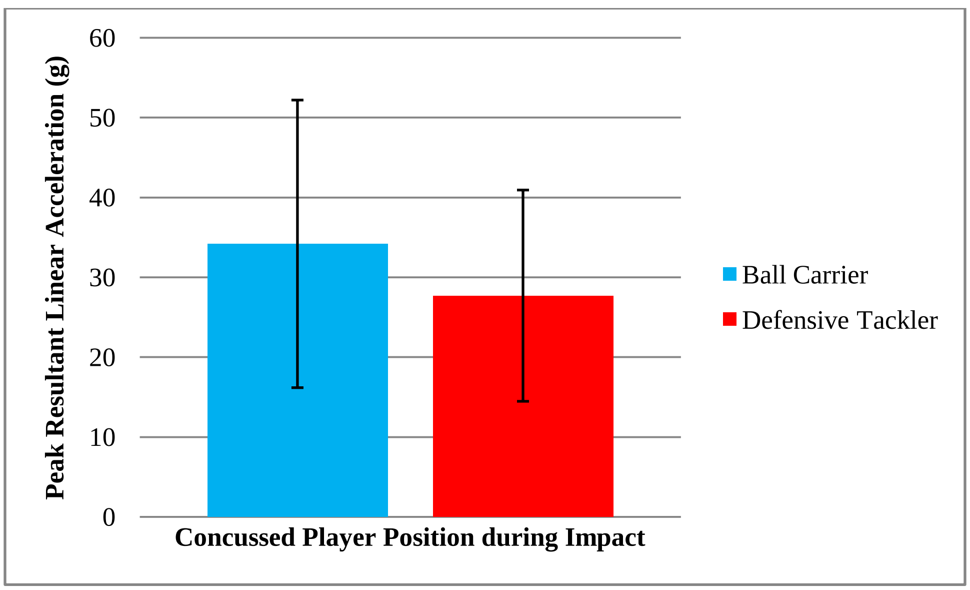

Figure 9. Mean peak resultant linear acceleration results

4.1.2 Peak Resultant Rotational Acceleration

There was no significant difference in peak resultant rotational acceleration during shoulder-to-headform reconstructions, t (18) = -.917, p = 0.371, when the defensive tackler was injured (M = 3812 (1003) rad/s^2) in comparison to the ball carrier (M = 4419 (1837) rad/s^2) *(Figure 9)*.

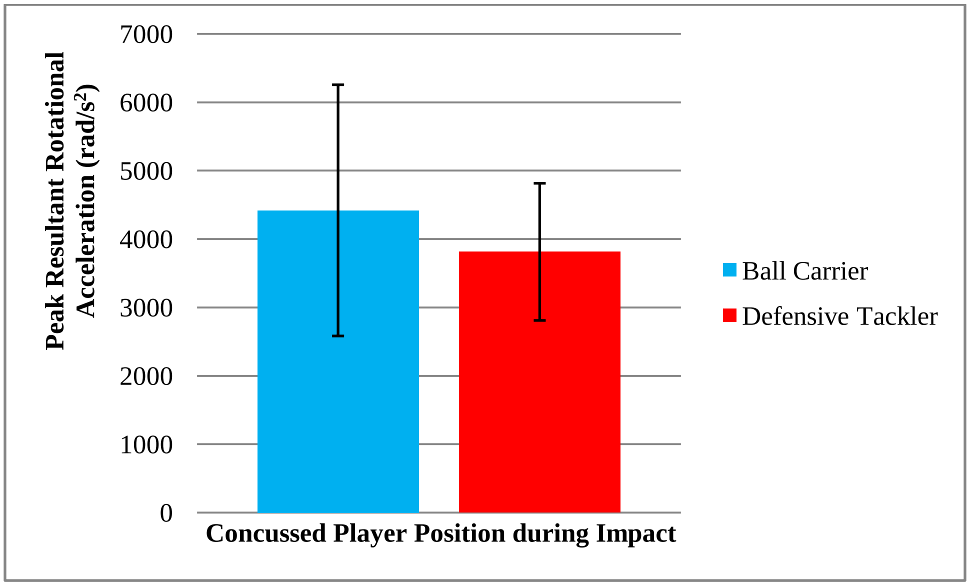

Figure 10. Mean peak resultant rotational acceleration results

4.2 Brain Tissue Deformation

Brain tissue deformation results for both concussive event conditions, characterized as maximum principal strain, can be found in *Table 5*, expressed as peak means, with standard deviations in parentheses. The maximum principal strain metrics were measured using the dynamic response values obtained during the in-laboratory reconstructions.

4.2.1 Maximum Principal Strain

There was no significant difference in maximum principal strain during shoulder-to-headform reconstructions, t (18) = -.117, p = 0.908, when the defensive tackler was injured (M = 0.408 (0.0992)) in comparison to the ball carrier (M = 0.414 (0.124)) *(Figure 10)*.

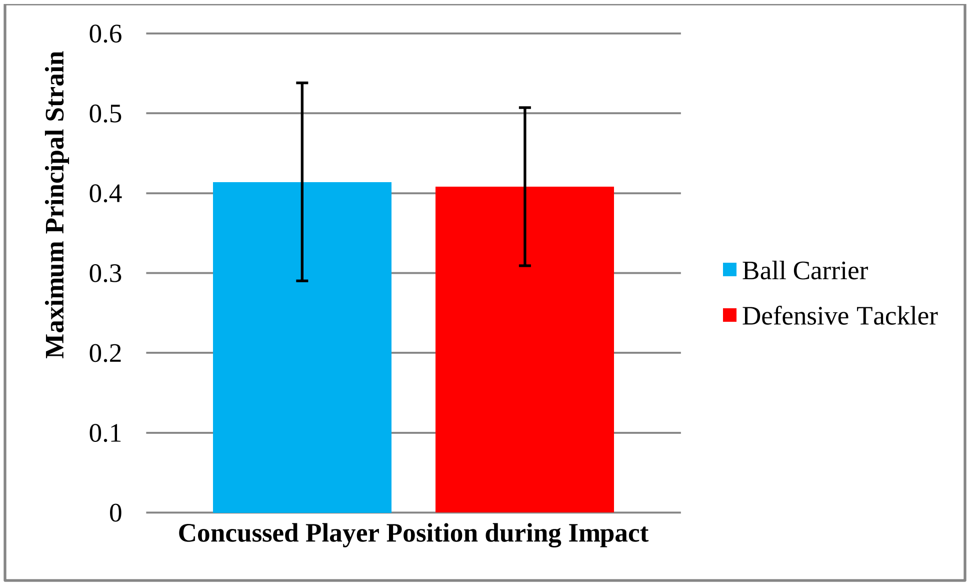

Figure 11. Mean peak maximum principal strain

4.3 Impact Characteristics

4.3.1 Inbound Velocity

Inbound velocity for both concussive event conditions, expressed as the mean of the six (6) impact reconstructions for each case, can be found in *Appendix C*. The inbound velocity for each reconstruction was captured by a time gate on the pneumatic linear impactor during the in-laboratory reconstructions.

There was no significant difference in inbound velocity during shoulder-to-headform reconstructions, $t\ (18) = -.697$, $p = 0.495$, when the defensive tackler was injured ($M = 6.05$ (1.26) m/s) in comparison to the ball carrier ($M = 6.43$ (1.19) m/s).

4.3.2 Location of Impact

The location of impact for each concussive shoulder-to-head event varied, as the reconstructions were conducted based on real-world concussive events. The locations of impacts were categorized based on a grid reference and a vertical offset impact location reference based on the centre of gravity *(Figure 3a,b)*. The distribution of the impact locations is presented in *Appendix D*. The locations of impact for the ball carrier were evenly distributed between the frontal and lateral portions of the headforms, whereas the defensive tackler impact locations were mainly impacting the lateral portion of the headform; this corresponds to the reported incidence of tempero-parietal concussive impacts in rugby (McIntosh et al., 2000).

CHAPTER 5: DISCUSSION

Epidemiological studies allow for quantification of mild traumatic brain injury cases in high-contact sports, as well as the associated symptomology. However, in order to attempt to develop concussion prevention strategies, such as rule, technique and protective equipment modifications and regulations, the biomechanical considerations of these high-risk events must be analyzed to decrease the risk of injury. Consequently, this study conducted the biomechanical analyses of the events leading to the highest rates of concussions during rugby game play: shoulder-to-head impacts (Gardner et al., 2014). As there was a reported discrepancy between the incidence of cerebral injury between the ball carrier and defensive tackler, the purpose of this study was to conduct a comparative analysis for concussive events between these two player positions to characterize risk (Gardner et al., 2015; Gardner, Iverson, Levi, et al., 2014; Gardner, Iverson, Stanwell, et al., 2014; King et al., 2010, 2014). In order to assess a potential difference in impact characteristics and risk of injury to either player position, reconstructions then allowed for characterization of cerebral response and brain tissue deformation. Although player position-dependent concussion rate differences had been previously reported (Gardner et al., 2015; Gardner, Iverson, Levi, et al., 2014; Gardner, Iverson, Stanwell, et al., 2014; King et al., 2010, 2014), this study yielded no significant differences between ball carrier and defensive tackler concussive events.

The cerebral responses resulting from impacts to the head during concussive events in sport are dependent upon impact characteristics such as mass, compliance, velocity and location (Gennarelli et al., 1982; Hoshizaki et al., 2013; Kendall, Post, et al., 2012; Pellman et al., 2003; Rousseau & Hoshizaki, 2015; Zhang et al., 2001). During the in-laboratory reconstructions, the impacting mass and compliance of the system remained constant, however the impact location and angle, as well as the inbound velocity, varied for each real-world event. Although the impacting technique changes between a defensive shoulder tackle and a ball carrier fend *(Appendix A)*, these results suggest that the technique and arm position in defensive tacklers and ball carriers do not influence the magnitude of the cerebral response during concussive events. A comparative analysis of these results and those previously reported in the literature will be conducted in the following section.

5.1 Dynamic Response

Dynamic response values were not significantly different between ball carriers and defensive tackler concussive events, however some cases yielded great differences in dynamic response with almost identical impact characteristics. Case 1 and Case 4 ball carrier concussive event reconstructions *(Table 9)* consisted of identical impact locations and forward angles, as well as similar inbound velocities, however the dynamic response and MPS values differed (38.27(3.82)g, 4922(564)rad/s^2, 0.547 and 16.58(0.66)g, 3045(193)rad/s^2, 0.306 respectively). The orientation angle difference of 25^o explains this difference in dynamic response and brain tissue deformation, whereby although the location of impact was identical, the forces transmitted to the headform resulted in different vector directions.

The mean peak resultant linear acceleration during concussive event reconstructions was significantly smaller, $t(41.76) = -10.68$, $p < .01$, in this present study (M = 30.92 (15.73) g) in comparison to the reported mean values by Fréchède & McIntosh (2009) (M = 101.36 (28.98) g). Similarly, the mean peak resultant rotational acceleration during concussive event reconstructions was significantly smaller, $t(36.73) = -4.99$, $p < 0.01$, in this present study (M = 4116 (1474) rad/s^2) in comparison to the reported mean values by Fréchède & McIntosh (2009) (M = 7910 (3560) rad/s^2). These differences are likely a result of the choice of system compliance (Gennarelli et al., 1982; Hoshizaki et al., 2013; Kendall, Post, et al., 2012; Pellman et al., 2013; Rousseau & Hoshizaki, 2015; Zhang et al., 2001). A 12ms duration was used in the MADYMO model during the Fréchède & McIntosh (2009) reconstructions as opposed to the 20-25ms used in this study specifically to represent shoulder-to-head collisions *(Appendix A)* (Rousseau, 2014). This 8-13ms decrease in duration results in higher dynamic response values, as the difference in energy-absorbing characteristics influences these parameters (Gennarelli et al., 1982, 1987; Fréchède & McIntosh, 2009; Hoshizaki et al., 2013; Hutchinson, 2012; Kendall et al., 2012; Kleiven, 2003; Pellman et al., 2003; Rousseau, 2014; Rousseau & Hoshizaki, 2015; Zhang et al., 2001).

5.1.1 Impact Characteristics

5.1.1.1 Compliance

As the dynamic cerebral response is directly influenced by compliance, as well as location, velocity and mass (Gennarelli et al., 1982; Hoshizaki et al., 2013; Kendall, Post, et al., 2012; Pellman et al., 2003; Rousseau & Hoshizaki, 2015; Zhang et al., 2001), it is important to consider the influence of this characteristic during biomechanical reconstructions of concussive events. Differing energy-absorbing materials have been reported to influence brain tissue deformation and dynamic response metrics (Gennarelli et al., 1982, 1987; Hoshizaki et al., 2013; Hutchinson, 2012; Kendall et al., 2012; Kleiven, 2003; Pellman et al., 2003; Rousseau, 2014; Rousseau & Hoshizaki, 2015; Zhang et al., 2001). The duration of impact that resulted from the MADYMO software reconstructions by Fréchède & McIntosh (2009), based on rigid-body cadaveric models, was reported to be 12ms, compared to the 20-25ms obtained in-laboratory with real-world shoulder-to-headform impacts using participants *(Appendix A)* (Fréchède & McIntosh, 2007, 2009; Rousseau, 2014). This can be explained by the magnitude-duration relationship, whereby short duration, high magnitude events, or inversely long duration, low magnitude events, can lead to brain injury (Gennarelli, 1982; Horgan & Gilchrist, 2003; Kleiven, 2005; Kleiven & von Holst, 2003; Willinger et al., 1995).

This study examined the mechanics of shoulder-to-head impacts in rugby using in-laboratory reconstructions based on real-world concussive events, and it would seem that shoulder impacts in rugby result in a longer duration, lower magnitude event, as opposed to the inverse relationship reported in previous research. However, it is important to consider that previous biomechanical analyses which quantified dynamic response of concussive impacts in rugby did not specify the impacting system that collided with the head during impact (Fréchède &McIntosh, 2009; McIntosh et al., 2000; Patton et al., 2013). These specific impact characteristics will directly influence the dynamic cerebral response values, and consequently, the brain tissue deformation metrics (Gennarelli et al., 1982; Hoshizaki et al., 2013; Kendall, Post et al., 2012; Pellman et al., 2003; Rousseau & Hoshizaki, 2015; Zhang et al., 2001).

5.2 Brain Tissue Deformation and Risk of Injury

5.2.1 Maximum Principal Strain

The mean peak maximum principal strain (MPS) value in this current study (0.411 (0.116)) was significantly greater than values reported in past finite element analyses of concussive impacts (Kleiven, 2007; Patton et al., 2013; Rousseau, 2014; Zhang et al., 2004) *(Figure 11; Table 2)*. These reported values represent a 50% likelihood of sustaining a concussion, hence the values in this study are likely higher due to the reconstruction of actual real-world rugby impacts which led to physician-diagnosed concussions.

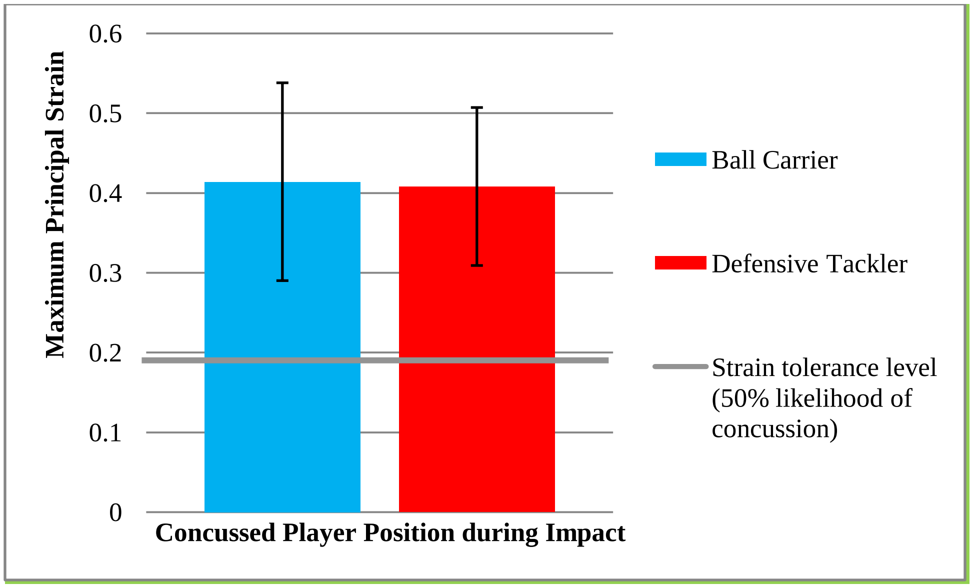

Figure 12. Mean peak maximum principal strain results and strain tolerance level

5.3 Conclusion

The main objective of this thesis was to compare dynamic response and brain tissue deformation during concussive impacts between ball carriers and defensive tacklers in professional rugby to determine if an association existed between tackling technique, player position and reported rates of concussion. The results of this study yielded no significant differences in dynamic response or brain tissue deformation between impacting offensive or defensive players during concussive shoulder-to-head events. Although the reconstructions conducted in this study did not demonstrate any difference in the two conditions, significant differences in cerebral response and brain tissue strain were reported upon comparison with previously reported rugby biomechanical analyses (Fréchède & McIntosh, 2009; Kleiven, 2007; McIntosh et al., 2000; Patton et al., 2013; Rousseau, 2014; Zhang et al., 2004).

As the impact parameters in this study closely represented rugby game play inbound velocity, impact locations, compliance, and mass, the impact characteristics used to conduct these reconstructions should be considered for future rugby impact testing reconstructions (Best, McIntosh, & Savage; 2005; McCrory et al., 2013; McIntosh, 2003, 2004; McIntosh, McCrory, Finch, Chalmers, & Best, 2003; Rugby, 2014). Differences between this study and past rugby collision reconstructions (Fréchède & McIntosh, 2009; McIntosh et al., 2000; Patton et al., 2013) reflect the importance of accounting for impact compliance when describing the risk associated with collisions.

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APPENDIX A: Rugby Shoulder-to-Head Collision Compliance Quantification

Participants

Ten male participants (mean 26.7, +/- 4.1 yrs, 94.5, +/- 8.9 kg, 180.2, +/- 8.5 cm) with no history of upper extremity pain or injury volunteered to participate in this study. All ten participants had competitive (college, university, community, or semi-professional leagues) rugby experience (mean 6.4, SD 3.9 yrs).

Experimental Procedure

A competitive level rugby player shoulder tackled a fixed Hybrid III 50th headform producing responses in linear and rotational accelerations. Two different shoulder impact conditions (a tucked-arm offensive 'ball carrier fend' approach and a horizontally-abducted-arm defensive 'shoulder tackle' approach) were employed using the participants' left shoulders (McIntosh, Savage, Mccrory, Fréchède, & Wolfe, 2010; Usman, McIntosh, & Fréchède, 2011)*(Figure 12a,b; Figure 13).*

Figure 13a. Typical arm-out defensive rugby shoulder tackle (left); **b.** typical tucked-arm ball carrier fend shoulder-to-head collision (right)

Figure 14. Arm-out shoulder-to-headform tackle using a 3-step approach in the laboratory

Each participant was asked to complete three shoulder tackles of each respective technique on the headform using a self-selected pace. Participants struck the headform using first the defensive arm-out ‘shoulder tackle’ approach *(Figure 13),* followed by the offensive ball carrier fend approach as they held a rugby ball, in order to account for possible different movement mechanics for the ball carrier (Brown, Brughelli, & Hume, 2014; McIntosh et al., 2010).

The zygomatic region of the headform (R2-S17*, Figure 3a,b*) was the targeted region for the anterior deltoid-to-headform collisions, as this location was the most commonly impacted area based on full match replay video analysis of 54 concussive shoulder-to-head collisions. The mean recorded shoulder-to-head collision velocity during this analysis was 6.33m/s, with inbound velocities ranging between 3m/s and 7.9m/s. A rectangular target (h: 3.4cm; w: 5.0cm) was used with the headform angled at 45^{o} in the x-plane (R2, *Figure 3b*). The Photron Fastcam-512PC high-speed camera (Photron USA Inc., San Diego, CA, USA) was used to confirm that the impact location of the participant’s anterior deltoid corresponded to the R2-S17 site *(Figure 3a,b)* as well as to measure impacting velocity (Rousseau & Hoshizaki, 2015). Participants were asked to repeat the trials for impacts which did not result in the striking of the designated target, based on the video footage obtained by the Phototron Fastcam-512PC1, until a total of three successful trials were obtained. The headform represented the striking player’s head (defenseman) during tucked-arm shoulder-to-head impacts, and the ball carrier’s head (offensive player) during arm-out shoulder-to-head impacts. The protocol was reviewed and approved by the University of Ottawa Office of Research Ethics and Integrity *(File number: H06-12-23)*.

Results

The results for peak linear and rotational accelerations during shoulder-to-headform collisions for the two different impacting conditions are presented in *Tables 6.*

Table 6. Dynamic response during shoulder-to-headform collisions using two different tackling techniques

	Striking velocity (m/s)	Linear acceleration (g)	Linear acceleration duration (ms)	Rotational acceleration (krads/s^2)	Rotational acceleration duration (ms)
Ball carrier fend	4.79 (+/- 0.91)	29.01 (+/- 13.89)	22 (+/- 6.86)	2.23 (+/- 0.89)	22 (+/- 6.86)
Defensive tackle	5.20 (+/- 0.95)	29.69 (+/- 8.47)	20.95 (+/- 4.35)	2.32 (+/- 0.85)	20.95 (+/- 4.35)

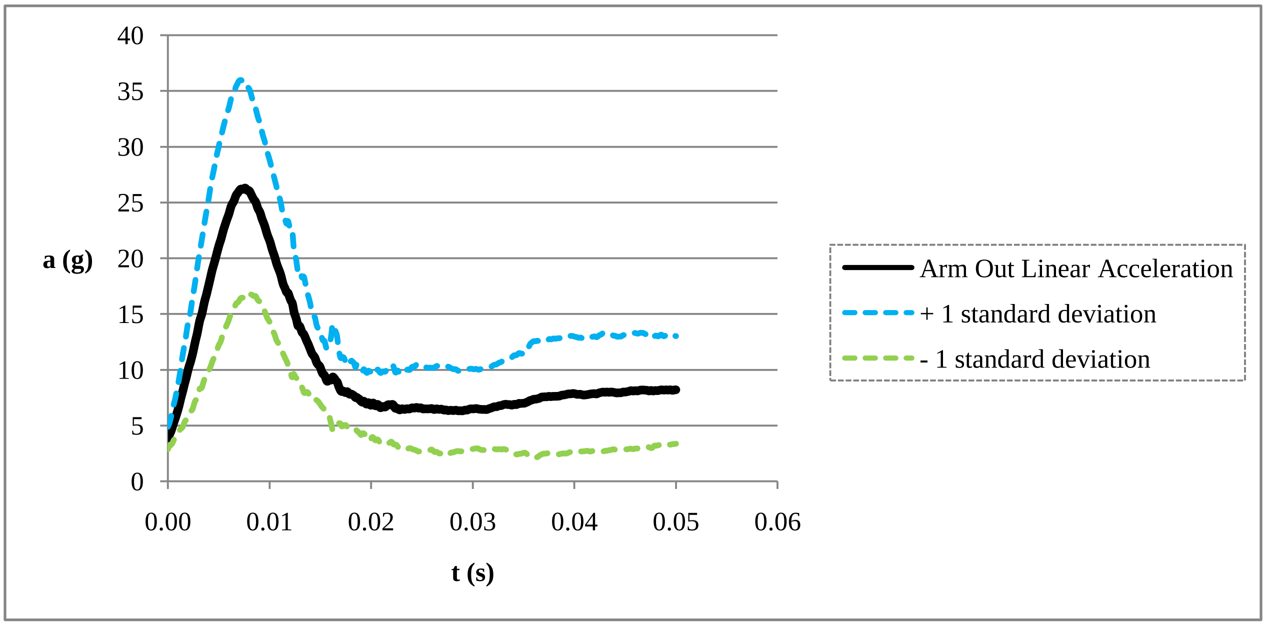

Figure 15. Linear acceleration curve of adefensive tackle

APPENDIX B: Impact Characteristics Descriptions

Table 7. Description of impact characteristics for shoulder-to-head impacts (reconstruction variables)

Impact Characteristic	Description
Defensive Shoulder Tackle	When the defensive player's shoulder is the first point of contact with the ball carrier (Hendricks et al., 2014)
Ball Carrier Fend	When the ball carrier's arm is the first point of contact with the tackler (Hendricks et al., 2014)
Velocity	3.2-9.3 m/s (Fréchède & McIntosh, 2009; Gabbett, 2013; McIntosh et al., 2000; Patton et al., 2013) Calculated for the reconstructions based on the video reconstructions of shoulder-to-headform impacts and the resulting dynamic response curves.
Location	Tempero-parietal region, towards the anterior half of the head (McIntosh et al., 2000); L1/R1-S17 (*Figure 3a,b)*
Impact Mass	13.1kg (Rousseau, 2014)
Compliance	20-25ms duration (*Appendix A)*

APPENDIX C: Complete Results Tables

Table 8. Results for defensive tackler shoulder-to-head impact reconstructions and FE modeling

Case	Peak Resultant Linear Acceleration (g)	Peak Resultant Rotational Acceleration (rad/s^2)	Maximum Principal Strain	Velocity (m/s)	Impact Location	Forward Angle (^{o})	Duration (ms)	Orientation Angle (^{o})
1	28.15 (2.17)	4882 (264)	0.498 (0.0268)	7.52 (0.42)	R1-S20	15	25	325
2	66.85 (14.06)	5696 (1133)	0.534 (0.0582)	5.89 (0.32)	R1-S20	0	20	325
3	26.27 (4.37)	4851 (707)	0.477 (0.102)	5.73 (0.32)	R3-S14	45	25	25
4	29.23 (1.66)	3752 (219)	0.406 (0.0335)	4.82 (0.15)	L2-S17	0	30	30
5	25.85 (2.88)	3820 (322)	0.455 (0.0380)	6.85 (0.48)	R2-S17	30	25	315
6	35.43 (3.00)	4111 (321)	0.427 (0.0283)	5.89 (0.36)	R3-S9	15	25	330
7	65.42 (18.70)	8019 (2661)	0.460 (0.0778)	5.77 (0.44)	R1-S18	45	25	45
8	30.23 (3.11)	5379 (384)	0.498 (0.0268)	6.30 (0.37)	L2-S9	15	25	290
9	10.75 (0.90)	2193 (303)	0.242 (0.0283)	8.04 (0.42)	R3-S14	30	20	315
10	23.43 (5.54)	1488 (444)	0.145 (0.0411)	3.67 (0.28)	R2-S17	45	25	330

*Refer to Figure 3a,b for impact location classification

*The angle is classified as the forward position of the unbiased neckform during impact *(Figure 4)*

Table 9. Results for ball carrier shoulder-to-head impact reconstructions and FE modeling

Case	Peak Resultant Linear Acceleration (g)	Peak Resultant Rotational Acceleration (rad/s^2)	Maximum Principal Strain	Velocity (m/s)	Impact Location	Forward Angle (^{o})	Duration (ms)	Orientation Angle (^{o})
1	38.27 (3.82)	4922 (564)	0.547 (0.0414)	6.17 (0.35)	L1-S18	30	25	45
2	18.03 (1.62)	2798 (175)	0.378 (0.0632)	7.55 (0.37)	R2-S9	45	20	330
3	27.2 (4.13)	4265 (333)	0.485 (0.0456)	5.94 (0.36)	Front-F5	15	25	5
4	16.58 (0.66)	3045 (193)	0.306 (0.00933)	5.95 (0.19)	L1-S18	30	25	20
5	29.88 (3.67)	4514 (550)	0.464 (0.0471)	5.64 (0.30)	R2-S14	0	25	335
6	29.67 (4.89)	4299 (475)	0.395 (0.0557)	6.71 (0.22)	R2-S18	0	25	330
7	20.62 (2.62)	4099 (427)	0.432 (0.0458)	8.28 (0.70)	R1-S9	45	20	300
8	30.97 (2.75)	3695 (459)	0.407 (0.0373)	7.79 (0.39)	R1-S18	15	20	75
9	56.48 (6.66)	4747 (384)	0.471 (0.0431)	4.21 (0.23)	R2-S17	0	30	330
10	9.02 (0.80)	1739 (206)	0.198 (0.0203)	6.06 (0.40)	L3-S13	45	25	330

*Refer to Figure 3a,b for impact location classification

*The angle is classified as the forward position of the unbiased neckform during impact *(Figure 4)*

APPENDIX D: Distribution of Impact Locations

Table 10. Distribution of impact locations for defensive tackler shoulder-to-head impact reconstructions (n=10)

Number of Front Impacts	Number of Lateral Impacts	Number of Rear Impacts
3	7	0

* The impact locations are classified according to three categories, based on the vertical offset reference *(Figure 3b)*

- Front impacts (R1, Front, L1)
- Lateral Impacts (R2-4; L2-4)
- Rear Impacts (R5-6; Back; L5-6)

Table 11. Distribution of impact locations for ball carrier shoulder-to-head impact reconstructions (n=10)

Number of Front Impacts	Number of Lateral Impacts	Number of Rear Impacts
5	5	0

* The impact locations are classified according to three categories, based on the vertical offset reference *(Figure 3b)*

- Front impacts (R1, Front, L1)
- Lateral Impacts (R2-4; L2-4)
- Rear Impacts (R5-6; Back; L5-6)