Biomechanics of injury events associated with diagnosed concussion in professional men's rugby league

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ABSTRACT

Concussions are a problem in competitive sports with growing concern over the acute and long-term consequences of repetitive head trauma. Participation in sport increases risk of concussion, particularly contact sports including rugby, hockey and football (Harmon et al., 2013). In rugby league, there are between 8.0-17.5 concussions/1000 player hours, representing roughly 10-15% of all injuries in the sport (Gardner et al., 2015). Shoulder, head, hip and knee are reported to be the most common regions that impact the head and are responsible for the greatest number of concussive injuries in rugby (Cusimano et al., 2013; Fuller et al., 2010; Gardner et al., 2014; Toth, Mcneil, & Feasby, 2005). In each of the common injury events reported in elite men's rugby, there are unique combinations of impact conditions which include effective mass, compliance, velocity and location of impact. The head-to-head event represents a low mass, low compliance event, whereby the hip and shoulder-to-head collisions represents high mass, high compliance events. Scientists have conducted research in an effort to describe incidence and mechanisms of concussive injury in rugby, however, little is known about the biomechanics of head injury in the sport (Fréchède & Mcintosh, 2009; Fréchède & McIntosh, 2007; McIntosh et al., 2000). The purpose of this thesis is to characterize dynamic response and brain tissue deformation for (1) hip-to-head, (2) shoulder-to-head, (3) knee-to-head, and (4) head-to-head concussion events in men's rugby.

Twenty-nine (29) impact videos of diagnosed concussive injuries associated with the four common injury events were reconstructed in the Neurotrauma Impact Science Lab. Head-to-head impacts were reconstructed in this study using a pendulum system, while hip, shoulder and knee to head impacts were reconstructed using the pneumatic linear impactor. Results of this study demonstrate that the common injury events resulting in concussion in elite men's rugby have different dynamic response characteristics. Head-to-head events produced significantly greater peak linear and peak rotational acceleration, however no significant differences in maximum principal strain between the injury events. Results of this study can be useful in reducing rates and severity of concussive injury in rugby.

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CHAPTER 1: INTRODUCTION

1.1 PROBLEM STATEMENT

Concussions have gained increasing media attention in recent years with an estimated 1.6 to 3.8 million sports-related concussions in the United States annually, accounting for 5-9% of sports-related injuries (CDC, 2003; Harmon et al., 2013; Powell & Barber-Foss, 1999). It has been projected that the rates of concussion, or mild traumatic brain injury (mTBI) are on the rise (Clay, Glover, & Lowe, 2013; Harmon et al., 2013), however it is unclear as to whether incidence of injury is increasing or whether greater exposure and awareness of concussion is leading to increased diagnoses. While the symptoms of concussion typically resolve within seven to ten days of the injury, in some cases symptoms may persist for months or even years (McCrory et al., 2009). Studies have described potential associations between mTBI and long-term health problems, such as a persistent decline in cognitive functioning as well as emotional and motivational impairments; there is also evidence of a link between repeated concussion and long-term neurological disorder such as Chronic Traumatic Encephalopathy (CTE) (Gardner et al., 2010, 2013; Gasquoine, 1997; Mckee et al., 2009; Shuttleworth-Edwards and Radloff, 2008). It has been widely reported that participation in sport increases risk of concussion, particularly in athletes participating in contact sports (Cusimano et al., 2013; Echemendia & Julian, 2001; Harmon et al., 2013; Koh, Cassidy, & Watkinson, 2003). Rugby, hockey and football have been reported to be amongst team sports with the greatest risk of concussive injury (Echemendia & Julian, 2001; Koh et al., 2003).With the known consequences of concussion there is growing concern over the long term health and safety of those that participate in sport.

Rugby league is a full contact sport played with minimal game stoppage with the most popular league in Australia, the National Rugby League (Gardner, 2015). There is more contact in rugby league: 69% active game play, compared to 56% active game play in rugby union, the type of rugby typically played in Canada (Gabbett, 2013). In rugby league, it has been documented that there are between 8.0-17.5 concussions/1000 player hours, representing roughly 10-15% of all injuries in the sport (Gardner et al., 2015a; Hinton-Bayre, Geffen, & Friis, 2004; Stephenson et al., 1996). However, it has been suggested that rates may actually be higher than reported because as many as 50% of concussions go unreported (Gardner et al., 2015a; Harmon et al., 2013; Hoskins et al., 2006). Video analyses of various matches have identified the most common events that lead to concussive injury, where 64%-80% of all injuries were reported to have occurred during the tackle (Best, McIntosh, & Savage, 2005; Waller et al., 1994). Shoulder, head, hip and knee are reported to be the most common regions that impact the head and are responsible for the greatest number of concussive injuries in the sport (Cusimano et al., 2013; Fuller et al., 2010; Gardner et al., 2014; Toth, Mcneil, & Feasby, 2005). Cerebral response following an impact is influenced by parameters such as: mass, compliance, location of impact and inbound velocity (Fréchède & Mcintosh, 2009; Hoshizaki, 2013; Karton, Hoshizaki, & Gilchrist, 2014; Kleiven, 2007). Head, shoulder, hip and knee impacts to the head during the rugby tackle each have unique combinations of these parameters creating unique loading conditions and risk in the sport.

Scientists have conducted extensive research in an effort to describe incidence and mechanisms of concussive injury in rugby, and have found that rugby is among the team sports with the greatest number of concussions (Best, 2005; Cusimano et al., 2013; Fuller et al., 2010; Gardner et al., 2014; Harmon et al., 2013; Hinton-Bayre et al., 2004; Hoskins et al., 2006; Stephenson et al., 1996; Toth et al., 2005; Waller et al., 1994). Research has also been targeted at decreasing incidence of concussive injury through use of equipment with the introduction of mouth guards and scrum caps. While scrum caps have been shown to reduce contusions and lacerations on the head and mouth guards have led to a reduction of oral injuries; they were ineffective at preventing concussions (Barbic, Pater, & Brison, 2005; Blignaut & Lombard, 1987; Jennings, 1990; Kemp et al., 2008; Knouse et al., 2003; Marshall et al., 2005; McIntosh & McCrory, 2000, 2001; McIntosh, 2004; McIntosh et al., 2009). While the incidence of concussion and mechanisms of injury in rugby have been described, little is known about the biomechanics of the impacts that create head injury in the sport; therefore it is unknown what strategies to implement in order to protect athletes (Fréchède & Mcintosh, 2009; Patton, Mcintosh, & Kleiven, 2013; Rock, 2016). Before interventions can be created aimed at reducing incidence of injury, it is important to understand how the events create risk of injury and whether there are differences between the events. To date researchers have yet to compare the mechanics behind the events that lead to concussion in rugby. Understanding each of the injury events and how they relate to risk will help inform decisions about interventions and is the first step in developing strategies to reduce the incidence of concussion in rugby.

1.2 SIGNIFICANCE

Rugby is a popular sport played by roughly 7.2 million players in 120 countries, with growing participation worldwide (World Rugby, 2014). While the incidence of concussion in rugby has been well-reported (Best, 2005; Cusimano et al., 2013; Fuller et al., 2010; Gardner et al., 2014; Harmon et al., 2013; Hinton-Bayre et al., 2004; Hoskins et al., 2006; Stephenson et al., 1996; Toth et al., 2005; Waller et al., 1994), little is known about the injury events that lead to concussion and how each event relates to risk. Physical reconstructions of shoulder-to-head injury events conducted by Rock (2016) is a starting point to describe the events that cause concussion in rugby. Further research must be conducted in order to provide a comprehensive description of injury events in the sport to identify head acceleration and brain tissue deformation during impact in order to compare risk of injury across events. Previous research has largely involved epidemiological studies describing incidence of events, or used mathematical modelling of impacts to the head (Fréchède & Mcintosh, 2009; Fréchède & McIntosh, 2007; McIntosh et al., 2000). Reconstructing concussive events sustained in rugby collisions as proposed in the current study will contribute to the body of knowledge by describing the relationship between the event and the ensuing brain injury. Research will be of benefit to current athletes, coaches, stake holders and governing bodies to provide a better understanding of the events that result in concussion in hopes of reducing the rate of concussions in rugby. The outcomes of this study will provide a better understanding of the four common injury events and how unique impact loading conditions from each event relate to risk and influences injury. Understanding how rugby players are obtaining concussive injuries will inform decisions about interventions and will contribute to the development of safety equipment and strategies to prevent injury.

CHAPTER 2: LITERATURE REVIEW

2.1 CONCUSSION IN THE NATIONAL RUGBY LEAGUE

Rugby league is a full contact sport played with minimal game stoppage in two 40-minute halves, with the most popular leagues in Australia and the UK (Gardner, 2015). Each team has 13 players playing both offensive and defensive roles with up to four interchange players (i.e. 17 players per match). Each team is allowed six tackles with the ball before turnover and unlike rugby union, tackles are not contested and there are no rucks or mauls (Gardner, 2015). Given these differences, there is more contact in rugby league: 69% active game play, compared to 56% active game play in rugby union (Gabbett, 2013).

In an attempt to reduce the incidence of concussion in the sport, the National Rugby League implemented the Concussion Interchange Rule (CIR) in 2014. This rule allows and encourages players suspected of having a concussion to be removed from the field to undergo an evaluation by the team physician and the interchange would not be tallied against the team (Gardner et al., 2015c). A study of the use of the Concussion Interchange Rule found that while the rule was being used as intended in many cases, there was video evidence of what appeared to be a player experiencing signs of a concussion, though the athlete was immediately cleared to return to play (Gardner et al., 2015b). While the CIR is aimed at protecting athletes once they have sustained an injury, there must be a greater focus on preventing concussions from occurring. In order to develop strategies to reduce the occurrence of concussion, we must first understand each injury event and how it relates to risk so that informed decisions can be made about interventions and potentially distinct interventions for each unique injury event.

2.2 CLASSIFICATION OF HEAD TRAUMA

Head trauma can result from various modalities in sports including falls, collisions and projectiles. Injuries that result from biomechanical forces following a head impact can be classified by severity, typically traumatic brain injury (TBI) and mild traumatic brain injury (mTBI), which includes concussion. Concussion is defined as "a complex pathophysiological process affecting the brain, induced by traumatic biomechanical forces" (McCrory et al., 2009). Following an impact to the head, there is a disruption of neuronal cell membranes and axonal stretching which influences a neurometabolic cascade of molecular changes in the brain which has an overall negative impact on performance and increases vulnerability to a repeat injury (Barkhoudarian & Hovda, 2011).

Concussion is described by a wide variety of symptoms including: loss of consciousness (LOC), dizziness, unsteadiness, headache, blurred vision, nausea and fatigue, amongst others (Chachad & Khan, 2006; Echemendia & Julian, 2001; Harmon et al., 2013; Haseler, Carmont, & England, 2010; NRL, 2015). As the presence of signs and symptoms of injury may be subjective, assessment is open to interpretation and potentially misdiagnoses and the masking of symptoms (Harmon et al., 2013).

2.3 INCIDENCE AND DISTRIBUTION OF INJURY

The head and neck are the most frequently injured regions (38 injuries /1000 player hours), representing roughly 31% of all injuries in rugby, including: lacerations, concussions, haematomas, contusions and fractures (Gabbett, 2000; King et al., 2010; McIntosh et al., 2010a; Stephenson et al., 1996; Waller et al., 1994). It has been reported that 8.5% of all injuries in men and 15% of all injuries in women are concussions, however the gender-specific differences were found to be non-significant (Gardner et al., 2014). Knouse et al., (2003) reported that concussions account for up to 25% of all rugby injuries, greater than previously documented findings. It has been reported that concussions are the third most common type of injury in rugby for all player positions, and lead to the greatest number of games missed of all injuries reported (Kemp et al., 2008).

It has been estimated that concussive injury in rugby league ranges from 8.0-17.5 injuries/1000 player hours (Gardner et al., 2015a; Hinton-Bayre et al., 2004), considerably higher than rates reported in rugby union, with 4.5/1000 player hours (Gardner et al., 2014; Kemp et al., 2008). Incidence of concussion has been documented to increase with increasing levels of play, however differences failed to reach statistical significance (Hinton-Bayre et al., 2004).

2.4 COLLISIONS IN RUGBY

Video analyses of various matches have identified the most common events that lead to concussive injury. 64%-94% of all injuries were reported to have occurred during the tackle (Best et al., 2005; King, Hume, & Clark, 2012; Waller et al., 1994). Further, it has been reported that of all injuries that were recorded in association with illegal match play, 29% were concussions, compared to only 9% during legal match play (Gardner et al., 2015a). These findings highlight the importance of stringent rule enforcement with the intent of minimizing concussive injury in rugby in addition to implementing strategies to reduce incidence of injury. World Rugby provisions of Law 10.4 (e) indicates a dangerous tackle (high tackle) is when an opponent tackles above the shoulders, even if the tackles starts below the line of the shoulders (World Rugby, 2017). Depending on the circumstances of the high tackle, sanctions range from a penalty kick to the player receiving a red card (elimination from the game) (World Rugby, 2017).

It has been documented that during the rugby tackle, players are obtaining concussive injury from shoulder-to-head, head-to-head, knee-to-head, and hip-to-head impacts (Cusimano et al., 2013; Fuller et al., 2010; Gardner et al., 2014; McIntosh et al., 2000; Quarrie and Hopkins, 2008; Toth et al., 2005). Shoulder-to-head collisions are frequently cited as the most common mechanism of injury, followed by head and hip. McIntosh et al. (2000) reported that 86% of impacts resulting in mTBI were to the tempero-parietal region to the anterior half of the head, impacted by the shoulder to thorax region. Researchers conducting a video analysis of National Rugby League game footage found the most common injury events to be: shoulder-to-head impacts, accounting for 35% of concussions, 20% were caused by head-to-head impacts and 20% by knee-to-head impacts, and 10% of concussions resulted from hip-to-head collisions (Gardner et al., 2015). It has been reported that the majority of tackles occurred in the hip region (bottom of the chest to the pelvis) and that 40% of overall injuries to the head and neck were caused by head to head contact, either between the tackler and ball carrier or between two tacklers (Quarrie & Hopkins, 2008). Fuller and colleagues reported that concussions occur most frequently from head impacts with another head, followed by upper limb, trunk and lower limb (Fuller et al., 2010).

While researchers have reported the incidence of injury along with the location of impacts, there have been few reports on the mechanism of injury that has resulted in concussion (Best, 2005; Cusimano et al., 2013; Fuller et al., 2010; Gardner et al., 2014; Harmon et al., 2013; Hinton-Bayre et al., 2004; Hoskins et al., 2006; Stephenson et al., 1996; Toth et al., 2005; Waller et al., 1994). McIntosh and colleagues (2000) analysed video of unhelmeted head impacts in rugby to determine impact characteristics. Many assumptions were made with respect to calculating moment, velocity and change in momentum, and when it was unsure as to which body segment impacted the head, the lower mass segment was selected (McIntosh et al., 2000). The focus of this study was to compare dynamic response values of rugby collisions by grade of concussion using loss of consciousness (LOC) (i.e. Grade 1: no (LOC), Grade 2: LOC less than 1 min, Grade 3: LOC greater than 1 min). Collisions resulting in LOC lasting longer than one minute had statistically higher peak linear accelerations than collisions with no LOC. Rather than a comparison between mechanisms of injury, this study focused on any concussive impact to the unhelmeted head. Data from this study were then used in an investigation by Fréchède et al., (2009) who reconstructed those concussive injuries using MADYMO, a rigid body simulation software. This study was then followed up by an investigation of brain tissue deformation of those impacts using a finite element analysis (Patton et al., 2013). Results of these studies were used to define average levels where risk of injury was observed and will be described in further detail in section 2.5.5 of this thesis. Previous research by Rock (2016) investigating differences in shoulder-to-head impacts between the defensive tackler and ball carrier found there were no differences in dynamic response or maximum principal strain between these two player positions. To our knowledge this was the first investigation to perform physical laboratory reconstructions of collision events causing concussion in men's rugby league. Findings from the Rock (2016) study investigating shoulder-to-head collisions were used to inform input parameters for reconstructions of shoulder-to-head impacts in the current investigation.

2.5 BIOMECHANICAL CONSIDERATIONS OF mTBI

2.5.1 BIOMECHANICAL PREDICTORS OF CONCUSSION

When the head is impacted, for instance in a shoulder-to-head collision during a rugby tackle, the head experiences acceleration caused by the impact forces. While linear and rotational accelerations of the head do not directly cause deformation to brain tissue, these variables are associated with risk of concussive injury (Holbourn et al., 1943; King et al., 2003; Rowson & Duma, 2013; Zhang, Yang, & King, 2004). Holbourn and colleagues (1943) hypothesized that shear and tensile strains generated by rotation were the cause of cerebral concussion. While strain that ensues following an impact is difficult to measure *in vivo*, input variables including linear and rotational accelerations are measured in order to characterize risk of injury (King et al., 2003). To examine differences between injury events in sport, head acceleration following an impact can be analyzed by capturing peak linear and rotational acceleration values of the head form. Dynamic response is a reliable method to test for brain injury by calculating head accelerations with the use of accelerometers mounted in the head form (Guskiewicz & Mihalik, 2011; King et al., 2003).

2.5.2 LINEAR ACCELERATION

High peak resultant linear accelerations have been shown to predict traumatic brain injury whereby pressure gradients in the skull created by impact are associated with focal head injuries, such as brain tissue lesion and skull fracture at the site of impact (Gennarelli et al., 1972; Holbourn et al., 1943; King et al., 2003). Early work studying animal models on mongrel dogs (Gurdjian et al., 1963) and monkeys (Ommaya et al., 1966) found that rotational acceleration alone did not cause brain injury; it was hypothesized that concussion was created by a combination of linear and rotational acceleration following a head impact. Initially linear acceleration was the sole method for analyzing and assessing head impacts and was thought to be the principal cause of concussion (Gurdjian et al., 1963; Ommaya et al., 1966; 1971; Ono et al., 1980), despite the hypothesized influence of rotational acceleration (Holbourn et al., 1943). Rotational acceleration is now accepted as a greater contributor in the generation of concussive injury (Deck & Willinger, 2008; Fréchède & Mcintosh, 2009; Fréchède & McIntosh, 2007; Holbourn et al., 1943; Kimpara & Iwamoto, 2012; King et al., 2003; Kleiven, 2007; Post & Hoshizaki, 2012; Withnall et al., 2005; Zhang et al., 2004).

2.5.3 ROTATIONAL ACCELERATION

Rotational acceleration of the head is associated with diffuse and focal brain injury (King et al., 2003). While linear and rotational acceleration are believed to be correlated, it has been speculated that rotational acceleration of the head is the better predictor of mild traumatic brain injury over purely linear acceleration (Holbourn et al., 1943; King et al., 2003; Post & Hoshizaki, 2012; Post et al., 2013a, 2013b; Zhang et al., 2004). Considering the correlation between rotational acceleration and shear-stress of the brain tissue, and that shear forces caused by linear acceleration are minimal, it is believed that rotational accelerations created by an impact are responsible for concussions (Holbourn et al., 1943). Zhang and colleagues (2006) analyzed the effect of purely linear and purely rotational accelerations using a finite element model of the head. This study reported that rotational acceleration had a greater influence on levels of strain detected in the brain tissue where linear translations did not have the same effect (Zhang et al., 2006). As these variables do not exist on their own but rather as interactions with one another, both peak linear acceleration and peak rotational acceleration will be examined in this study in order to characterize the risk of concussion.

2.5.4 BRAIN TISSUE DEFORMATION

While dynamic response is a relatively direct and reliable method to collect data on kinematics of the head during impact, Finite Element (FE) analysis provides further information regarding brain tissue deformation (McAllister et al., 2012; Patton et al., 2013). The analysis provides quantification of brain tissue deformation using maximum principal strain (MPS), which provides a measure of brain tissue stress and strain; it has been reported to have a higher correlation to brain injury prediction than using dynamic response metrics alone (Bain & Meaney, 2000; Horgan & Gilchrist, 2003, 2004; King et al., 2003; Kleiven, 2007; Nahum et al., 1977; Willinger & Baumgartner, 2003; Zhang et al., 2004; Zhang et al., 2003). Finite element models allow for the simulation of head and brain injuries using linear and rotational loading curves from physical reconstructions, while maintaining geometry, anatomy and material properties of a human head. The head geometry of the models are developed from computed tomography (CT) and magnetic resonance imaging (MRI) scans of adult male human cadavers and validated against cadaveric and real world TBI reconstructions (Horgan and Gilchrist, 2003; 2004; Nahum et al, 1977). While simulations using an FE model are a highly repeatable method of reporting brain injury, models are validated against cadaveric head impact research (Horgan and Gilchrist, 2003; 2004; Nahum et al, 1977). They may therefore not be an accurate representation of live human brain tissue deformation as material and viscoelastic properties differ between live human and cadaveric tissues (Zhang et al., 2001). MPS is a possible indicator of concussion and is commonly used in concussion research analyzing biomechanics of head injuries (Kleiven, 2007; Willinger & Baumgartner, 2003; Zhang et al., 2004). Combined with dynamic response obtained from physical models, finite element modelling has the highest correlation of predicting injury (Kleiven, 2007; Willinger & Baumgartner, 2003; Zhang et al., 2004).

FINITE ELEMENT MODEL (UCDBTM)

The University College Dublin Brain Trauma Model (UCDBTM) is a finite element brain model used to determine brain tissue deformation by measuring maximum principal strain (MPS) in the cerebrum. The head geometry of this model was developed from computed tomography (CT) and magnetic resonance imaging (MRI) scans of adult male human cadavers and validated against cadaveric and real world TBI reconstructions (Horgan and Gilchrist, 2003; 2004; Nahum et al, 1977). The model was validated using comparisons of real world brain injury events and cadaveric impacts performed by Nahum (1977) and Hardy (2001). Cadaveric head impacts were simulated using the UCDBTM and results demonstrated a good correlation with the cadaver experiments and the modelled simulation across a range of locations.

2.5.5 PROPOSED RISK OF INJURY

Attempts have been made to associate mechanism of injury with dynamic response and brain tissue deformation in defining proposed risk of injury. Researchers have used video analysis along with simulations and laboratory reconstructions of head impacts to estimate thresholds of injury in different sports, often resulting in a proposed 25%, 50% and 80% probability of concussion. Zhang et al., (2004) and Willinger & Baumgartner (2003) duplicated on field head-to-head collisions in professional football using in-laboratory reconstructions and FE modelling. Injury and non-injury groups were used for comparison of head kinematics at a duration of 15ms. Zhang and colleagues (2004) proposed injury thresholds of 25%, 50% and 80% risk of concussion to be 66, 82, 106g respectively for maximum resultant linear acceleration. With regards to rotational acceleration, thresholds of 4600, 5900, and 7900 rad/s^2 correspond to 25%, 50% and 80% probability of sustaining a mTBI (Zhang et al., 2004). Willinger & Baumgartner (2003), found that mTBI was incurred at slightly lower values: 28.3g and 3000-4000 rad/s^2.

With respect to MPS values obtained from finite element analysis, Kleiven (2007) determined there was a 50% chance of injury in adults associated with strain values of 0.21 or 0.26, depending on the location of injury. Zhang and colleagues, (2004) proposed a strain rate in the brain stem of 0.14, 0.19, 0.24 corresponding to 25, 50, 80% risk of injury respectively whereas Willinger and Baumgartner (2003) determined threshold values of strain rate for a 25, 50, and 75% risk of injury in adults to be 0.25, 0.37, and 0.49 respectively. Values obtained from finite element models in addition to dynamic response are useful to researchers as they aid in interpreting results of real world reconstructions and in quantifying associated risk of injury.

2.6 HEAD IMPACT CHARACTERISTICS

It has been documented that concussions are occurring most frequently from shoulder-to-head, hip-to-head, knee-to-head, and head-to-head impacts (Cusimano et al., 2013; Fuller et al., 2010; Gardner et al., 2015b; Gardner et al., 2014; Toth et al., 2005). The resulting brain injury associated with these events are influenced by parameters including: impact location, mass, velocity and compliance of the impacting surface (stiffness of the hip, head, knee and shoulder) (Fréchède & Mcintosh, 2009; Hoshizaki, 2013; Karton et al., 2014; Kleiven, 2007). Concussion risk can be estimated experimentally by performing reconstructions of head impacts obtained from video footage. In order to ensure accuracy in the reconstructions of these events, each of the above parameters must be considered (Fréchède & Mcintosh, 2009; Karton et al., 2014).

2.6.1 IMPACT LOCATION

Location of impact has been shown to have an influence on the head kinematics and resulting brain injury. Studies of animal and human models have found that impacts to the side of the head are prone to greater head and brain response than impacts to the front of the head (Delaney, Puni, & Rouah, 2006; Gennarelli et al., 1982; Kleiven, 2003; Zhang et al., 2004). Impact locations are expected to vary between the injury events and impacts to the side of the head in rugby tackles are likely to cause a greater risk of concussion in rugby than other impact locations. Video of concussive impacts must carefully be analyzed in assessing location of impact for laboratory reconstructions.

2.6.2 MASS

Impact mass has been shown to affect magnitude of the event with increasing masses generally causing an increased risk of injury (Karton et al., 2014; Post & Hoshizaki, 2012; Rousseau & Hoshizaki, 2015). The mass of the impactor or striking body has been shown to influence dynamic impact response of a Hybrid III head form as measured by peak resultant linear and rotational accelerations as well as brain tissue deformation (Karton et al., 2014). At low masses (below 10.3 kg), as impactor mass increased by 2 kg increments, peak resultant linear and rotational accelerations increased. However, as impactor mass increased from 10.3 to 14.3 kg, linear accelerations for the centric impact location attained a plateau (Karton et al., 2014).

2.6.3 INBOUND VELOCITY

Velocity has been shown to influence dynamic impact response and brain tissue deformation values. Linear and rotational accelerations and brain tissue deformation increased with increasing velocity in studies investigating falls, projectiles and collisions (Kendall et al., 2012; Post et al., 2013b; Rousseau et al., 2009).

A study by Hendricks et al., (2012) investigating velocity of rugby hits found that in controlled settings, inbound velocities during tackle events ranged from 1.5-4.6 m/s for the tackler and 1.5-5.9m/s for the ball carrier. Using video analysis during competitive rugby matches, authors reported inbound velocities of a front tackle to range from 4.8-5.2 m/s for the ball carrier, and 5-

6.4m/s for the tackler (Hendricks et al., 2012) while McIntosh et al., (2000) reported the average velocity to be roughly 6m/s (range 3.0-11.4 m/s).

2.6.4 COMPLIANCE

Compliance of an impacting surface influences the duration of the acceleration loading curve and varies from one injury event to another. Deformable materials absorb energy during impact and have been shown to have an influence on dynamic response and brain tissue deformation (Post, Hoshizaki & Gilchrist, submitted). Compliance of the impacting material is determined by the duration of the event that is established by examining the impact loading curve. Softer, less compliant materials have longer durations as they absorb impact energy over a longer period of time (Post et al., 2012).

Shoulder compliance was determined by Rock (2016), in a similar fashion as Rousseau and Hoshizaki (2015) whereby participants struck either a suspended (Rousseau & Hoshizaki, 2015) or fixed (Rock, 2016) head form with their shoulder, designed to replicate collisions in the respective sport. Compliance of the event was determined as the duration taken to impart acceleration of the head form and was determined to be roughly 25-30ms (Rock, 2016; Rousseau & Hoshizaki, 2015).

2.7 VIDEO ANALYSIS

Video analysis of head injury events from game footage allows for a biomechanical analysis of characteristics of the impacts. Particularly in professional sport in which games are filmed, athletes are assessed by medical staff and video is available in high resolution (Fréchède & Mcintosh, 2009). This approach has been used to obtain characteristics of player collisions and falls in professional American football, Australia rules football, hockey and rugby (Fréchède and

McIntosh, 2007; Fréchède and Mcintosh, 2009; Gardner et al., 2015c; Gardner et al., 2015b; McIntosh et al., 2000; Pellman et al., 2003; Quarrie and Hopkins, 2008; Rousseau and Hoshizaki, 2015). Video quality, number of cameras, and camera orientation can influence the accuracy of obtaining kinematic data using this method. Studies performing analyses using video footage of injury events in hockey and rugby reconstructions have reported an average 10% error in calculating velocity in video compared to real world velocity (McIntosh et al., 2000; Post et al., 2015). In order to manage calculation error, a range of velocities can be presented for each injury reconstruction, ± 10% of the calculated mean, in order to create a corridor of response of best and worst case scenarios. Outcomes of video analyses help to inform characteristics of the tackle that result in overall injury as well as concussive injury.

CHAPTER 3: METHODS

RESEARCH DESIGN

3.1 RESEARCH QUESTION

Are there differences in dynamic response and brain tissue deformation of shoulder-to-head, head-to-head, hip-to-head, and knee-to-head events that result in concussion in elite men's rugby?

3.2 OBJECTIVES

1. To determine values of peak resultant linear and rotational accelerations in shoulder-to-head, head-to-head, hip-to-head, and knee-to-head collisions causing concussion in elite men's rugby.
2. To compare maximum principal strain of collision events causing concussion in men's rugby.
3. To identify magnitude and duration of the impact loading curves in describing each of the common injury events.

3.3 VARIABLES

3.3.1 INDEPENDENT VARIABLES

1. Injury event
    a. shoulder-to-head collisions;
    b. head-to-head collisions;
    c. hip-to-head collisions;
    d. knee-to-head collisions

3.3.2 DEPENDENT VARIABLES

1. Dynamic response
    a. Peak resultant linear acceleration
    b. Peak resultant rotational acceleration

2. Brain Tissue Deformation
    a. Maximum Principal Strain (MPS)

3.4 HYPOTHESES

i. It is hypothesized that the head-to-head event will result in the greatest peak resultant linear acceleration of the head form, followed by knee-to-head, shoulder-to-head and hip-to-head collision events.

ii. It is hypothesized that the head-to-head event will result in the greatest peak resultant rotational acceleration of the head form, followed by knee-to-head, shoulder-to-head and hip-to-head collision events.

iii. It is hypothesized that the head-to-head event will result in the greatest maximum principal strain values, followed by knee-to-head, shoulder-to-head and hip-to-head collision events.

3.5 NULL HYPOTHESES

i. Injury event will have no effect on peak resultant linear acceleration.
ii. Injury event will have no effect on peak resultant rotational acceleration.
iii. Injury event will have no effect on maximum principal strain.

METHODOLOGY

3.6 STUDY POPULATION: INCLUSION/EXCLUSION CRITERIA

A total of 297 impact videos obtained from the 2013-2015 seasons of National Rugby League game footage (www.nrl.com) were reviewed of which twenty-nine (29) concussive impact videos met inclusion criteria for this study. All concussions were diagnosed by the team physicians. Subjects are all male elite rugby players, aged 18-35 years old. All video depicting direct shoulder, head, hip, and knee impacts to the head that resulted in concussion were considered for analysis. Collisions in which the mechanism or location of the impact or markings on the field (for velocity calculation) were not clearly visible, were excluded. Analysis of 297 National Rugby League videos determined that 11 shoulder-to-head, 9 head-to-head, 7 hip-to-head and 2 knee-to-head impacts met the inclusion criteria. Given the low quantity of knee-to-head impacts that met inclusion criteria, knee impacts were excluded from the statistical analysis, however they were reconstructed to compare dynamic response and brain tissue deformation to other injury events.

3.7 VIDEO ANALYSIS

Cases were selected based on the mechanism of injury. Video analysis using Kinovea software 0.8.20 (open source, Kinovea.org) was used to determine velocity and location of impact as well as whether head gear was worn. Diagnosed concussive impacts involving these mechanisms of injury were obtained from high quality concussion impact videos from Australia's National Rugby League (www.nrl.com) and recorded at 25 frames per second. The impact velocity was determined by applying a perspective grid of known distances from markings on the field in the plane of impact, to establish the players' displacement (Figure 1). Displacement over time was calculated to determine the impact velocity between the impact sites on each player (Figure 2).

Location of impact was also determined from video analysis using head form location grids (Figure 3).

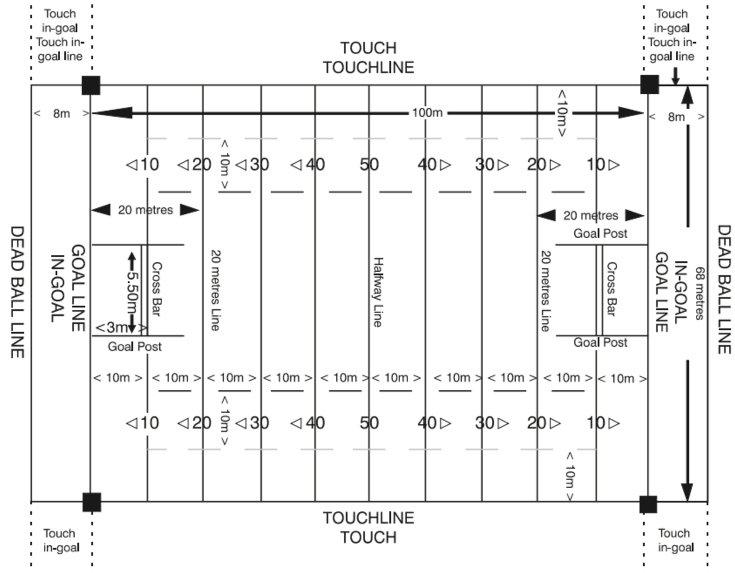

Figure 1: Official NRL field dimensions used to calibrate Kinovea in velocity calculations

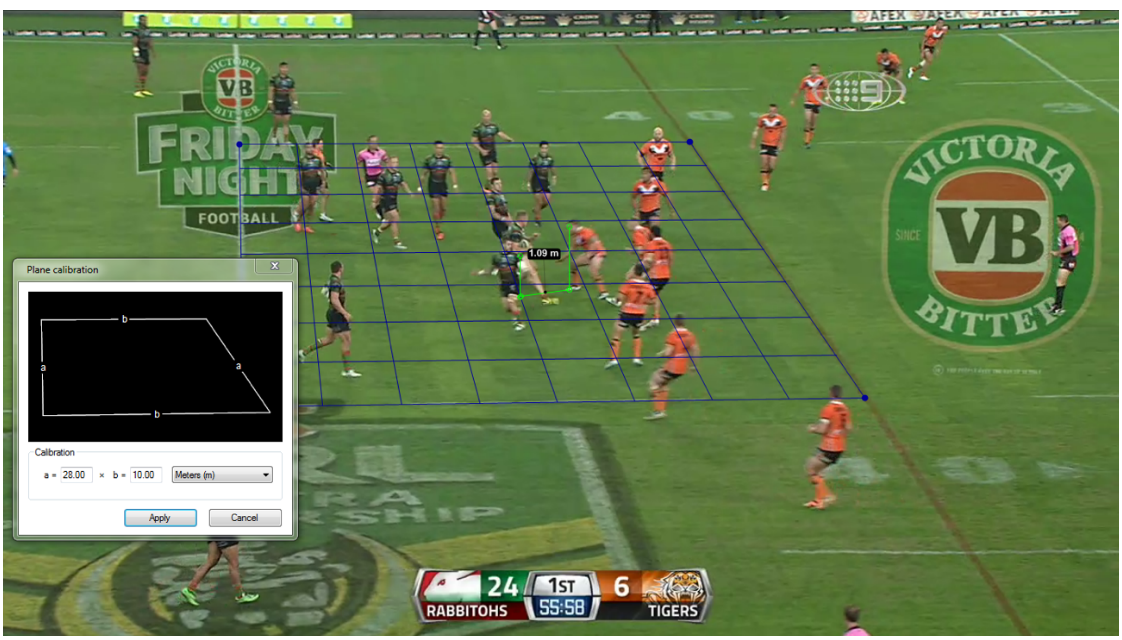

Figure 2: Example of grid used to create scaling reference system in calculating velocity

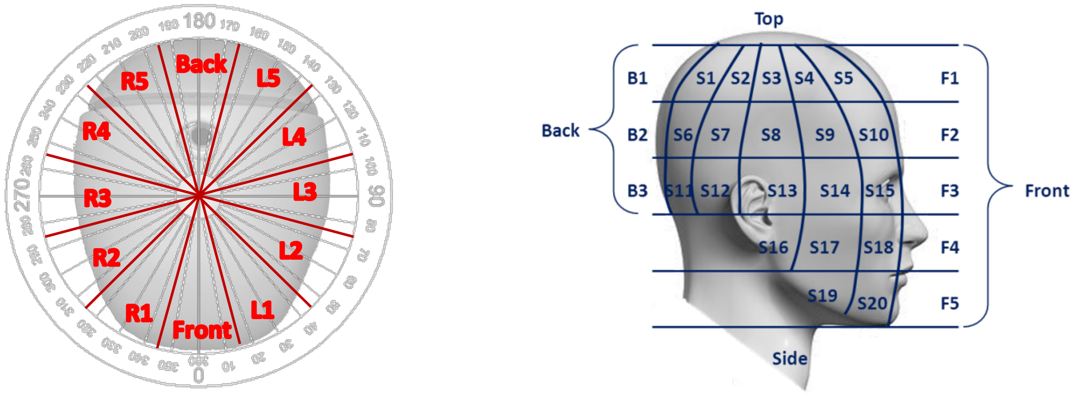

Figure 3a, 3b: Location grids for reconstructing concussive impacts

3.8 RECONSTRUCTION PROTOCOL

Four distinct head injury events associated with the most common concussive injuries in men's rugby league were reconstructed in the Neurotrauma Impact Science Laboratory: (1) hip-to-head impacts, (2) head-to-head impacts, (3) shoulder-to-head impacts and (4) knee-to-head impacts. Velocities and location of the reconstructed impacts were obtained from video of actual concussive injuries diagnosed by team doctors. The Hybrid III 50th percentile adult male head form and unbiased neck form complex were used for all injury reconstructions. The head-neck complex was affixed to a sliding table (Figure 4) and adjusted to reflect position of impact from the game analysis. Three impacts were collected for each condition and values were averaged across trials for statistical analyses. A range of velocities were presented for each injury reconstruction, ± 10% of the calculated mean, to create a corridor of response based upon the calculated impact velocity. The range of velocities reflects average percent error when performing analyses using video footage of injury events reported in hockey and rugby reconstructions (McIntosh et al., 2000; Post et al., 2015). For statistical analyses, only the reconstruction data of the low-end velocity were used to represent "best case" scenario and the most conservative value at which concussion could occur.

3.9 EQUIPMENT

3.9.1 HYBRID III HEAD FORM AND UNBIASED NECK FORM

A 50th percentile male Hybrid III head form (mass = 4.54 kg +/- 0.01 kg; FTSS, Plymouth MI) containing an aluminum shell with a vinyl cover was used for all injury reconstructions (Figure 4). The head form was equipped with nine single-axis Endevco 7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA). Accelerometers were mounted within the head form on blocks; one block with three accelerometers at the centre of gravity and three blocks each with two accelerometers on the anterior, lateral and superior surfaces of the head form (Figure 5). Accelerometers were mounted in an orthogonal arrangement following the 3-2-2-2 array to measure three-dimensional motion of the head form during impact as per the protocol developed by Padgaonkar and colleagues (1975). The TDAS ProLab Module (DTS, Seal Beach, CA) was used to collect acceleration data at 20 kHz and filtered through a low pass 1650 kHz filter for analysis of the acceleration time histories. The unbiased neck is the neckform used in this study. It is asymmetric and designed to respond the same to impacts in all directions to eliminate directional bias (Foreman & Hoshizaki, 2011). The unbiased neck was affixed to the head form, and the head-neck complex was attached to the custom sliding table (mass=12.782+/-0.001 kg; Cadex, St-Jean-sur-Richelieu, QC). Throughout testing, the head form remained fixed while the impacting arm of the linear impactor (shoulder, hip and knee reconstructions) or pendulum (head), was directed at the location and orientation determined from video.

Figure 4: Hybrid III head attached to unbiased neck form and sliding table

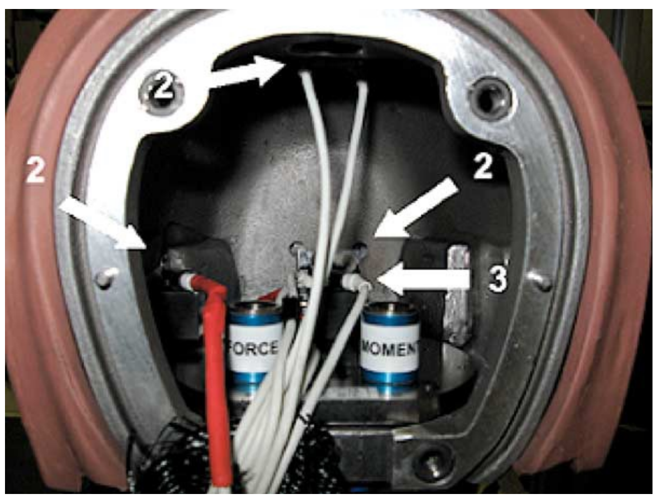

Figure 5: Hybrid III head form equipped with Endevco accelerometers in 3-2-2-2 array

3.9.2 HEAD-TO-HEAD COLLISIONS

Nine (9) head-to-head impacts were reconstructed in this study using a pendulum system. The system contains a hollow metal frame weighing 3.36 kg that is suspended by four cables attached directly above the headform on the ceiling. A circular metal plate was added in the centre of the frame to attain a mass of 4.8 kg, designed to match the average mass of an adult male head

(Zhang et al., 2001). A hemispherical aluminum disc covered with a vinyl skinform from the back of a hybrid III headform was fixed to the impacting end of the pendulum designed to represent compliance of a human head (Figure 6). The back of the frame is suspended by a 3/32 inch aviation cable attached to a latch that is released manually (Karton et al., 2014). Pendulum height was adjusted to reflect velocity of the impact established from video footage. A High Speed Imaging PCI-512 Fastcam camera (Photron USA Inc., San Diego, CA, USA) was used to monitor velocity of the pendulum at time of impact. High speed video was recorded at a frame rate of 250 frames per second and velocity was calculated 3 frames prior to impact, each frame representing 0.004 seconds. The camera was located perpendicular to the line of action and the position and zoom were kept constant throughout data collection. Table position and orientation of the head and neck were adjusted to reflect location and direction of impact.

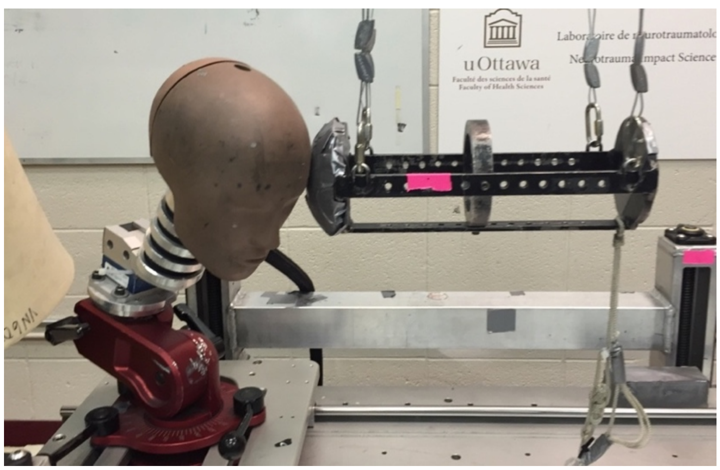

Figure 6: Example of pendulum system used to reconstruct head-to-head impacts

3.9.3 HIP-TO-HEAD COLLISIONS

Seven (7) hip-to-head impacts were reconstructed using the pneumatic linear impactor (Figure 7). The linear impactor consists of the frame, an impacting arm and a table in which the head form is fixed. The frame supports the compressed air canister, impacting arm and piston. The impacting arm is propelled horizontally by a piston from the compressed air towards the head form. The mass of the hip was determined using anthropometric scaling data (Winter, 1990) as a proportion of whole body mass such that the impacting arm reflects segment mass. Player anthropometric data were obtained from the National Rugby League official website (NRL.com) in which mass of the hip was determined to be 13.8 kg, based on an average player mass of 97.3 kg. The pneumatic linear impactor can be equipped with either a 13.1 kg or 15.9 kg impacting arm, therefore the lower mass arm (mass = 13.1 ± 0.1kg) was selected as it more closely represents the effective mass of the hip. Previous research by Karton and colleagues (2014) indicated that as impactor mass increased between 10.3 and 14.3 kg there was no significant increase in dynamic response or MPS. Results of this study support the use of a slightly lower mass impacting arm, as the effective mass of the impacting system is above 10.3 kg and the discrepancy should not have an influence on dynamic response or MPS. The impacting end of the arm was fixed with a vinyl nitrile foam replicating compliance of the hip (figure 8), determined in a pilot study to be roughly 25-30 ms. Table position and orientation of the head and neck were adjusted to reflect location and direction of impact.

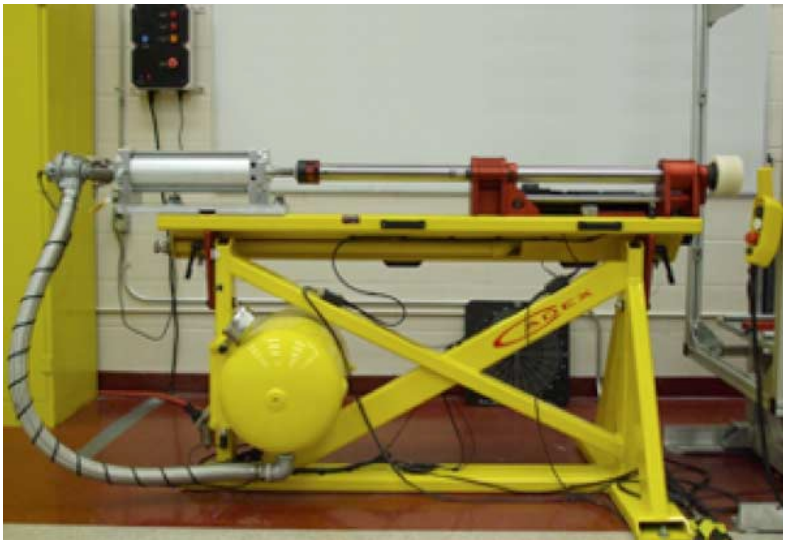

Figure 7: Cadex Pneumatic Linear Impactor

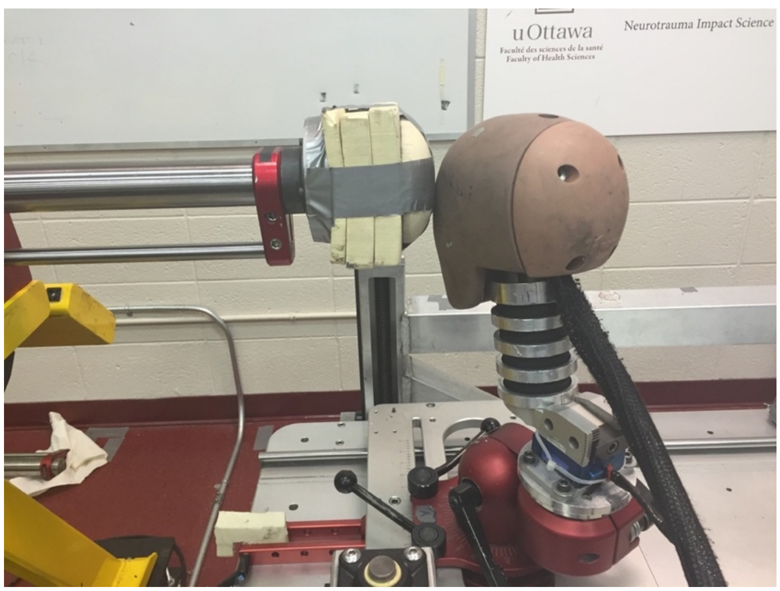

Figure 8: Linear impactor and anvil used to reconstruct hip-to-head and shoulder-to-head impacts

HIP STRIKER COMPLIANCE

Compliance of the hip was established in a pilot study by collecting three dimensional dynamic response data of hip strikes to a hybrid III head form. Eight adult male athletes (mean age 23.8 ±

4.1 years, weight 82.1 ± 4.58 kg, height 178.2 ± 4.98 cm) with no history of lower extremity pain or injury volunteered to participate in this study. All participants were either current or former rugby or football athletes with an average 3.9 ± 2.6 years of experience in the sport. All participants signed an informed consent form and completed the PAR-Q prior to participation in the study. Participants watched a National Rugby League hip-to-head concussive impact video and were given instructions to strike the hybrid III headform with their hip simulating the rugby tackle observed in the video (Figure 9). They were then given time to practice the protocol before beginning data collection. A minimum of three trials were collected for each participant. A High Speed Imaging PCI-512 Fastcam camera was used to ensure appropriate location of impact and impact velocity were attained to replicate characteristics of hip-to-head concussive injuries from NRL footage. Resultant linear and rotational acceleration data from all participants were averaged and plotted across time to determine duration of the impact and compliance of the hip. A foam surrogate was designed to replicate compliance of the hip from the participant trials by matching dynamic response peaks and durations. Acceptable compliance of the surrogate was reached when the curve characteristics and durations were matched within one standard deviation of the participant data (Figure 10). A summary of dynamic response data from participant trials can be found in table 9.

Figure 9: Participant performing hip-to-head rugby strike in-lab (left) simulating the rugby tackle observed in video (right).

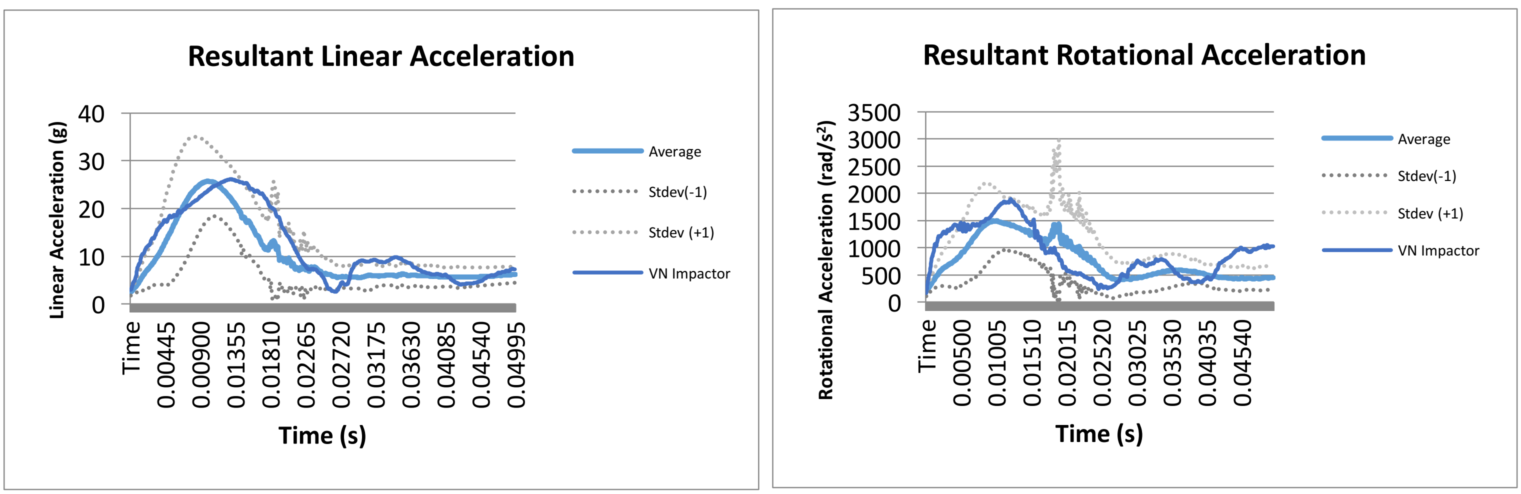

Figure 10: Vinyl Nitrile (VN) impactor matched to compliance of hip-to-head impacts within one standard deviation of average peak resultant linear (left) and rotational (right) acceleration data from all participant trials.

3.9.4 KNEE-TO-HEAD COLLISIONS

Two (2) knee-to-head impacts were reconstructed using the pneumatic linear impactor. The mass of the knee was determined by anthropometric scaling data (Winter, 1990) of the lower extremity as a proportion of the whole body mass. Player anthropometric data were obtained from the National Rugby League official website (NRL.com) in which mass of the lower extremity was determined to be 15.7 kg, based on an average player mass of 97.3 kg. The pneumatic linear impactor was equipped with the larger mass arm (mass = 15.9 kg) as it more closely represents the effective mass of the lower extremity. As the mass of the impacting arm is above 10.3 kg, the discrepancy between the calculated effective mass and mass of impactor should not influence dynamic response and brain tissue deformation (Karton et al., 2014). The end of the impacting arm was fixed with a 3.81 cm modular elastomer programmer (MEP) disc covered with a hemispherical steel cap striking the head (Figure 11), intended to replicate compliance of the knee, roughly 5-6 ms. Real-world knee compliance data were collected in-lab from a single, standing knee-to-head strike to the Hybrid III headform. As the participant experienced a large amount discomfort from the impact, data collection was terminated and compliance was determined from the single acceleration time history curve. Table position and orientation of the head and neck were adjusted to reflect location and direction of impact.

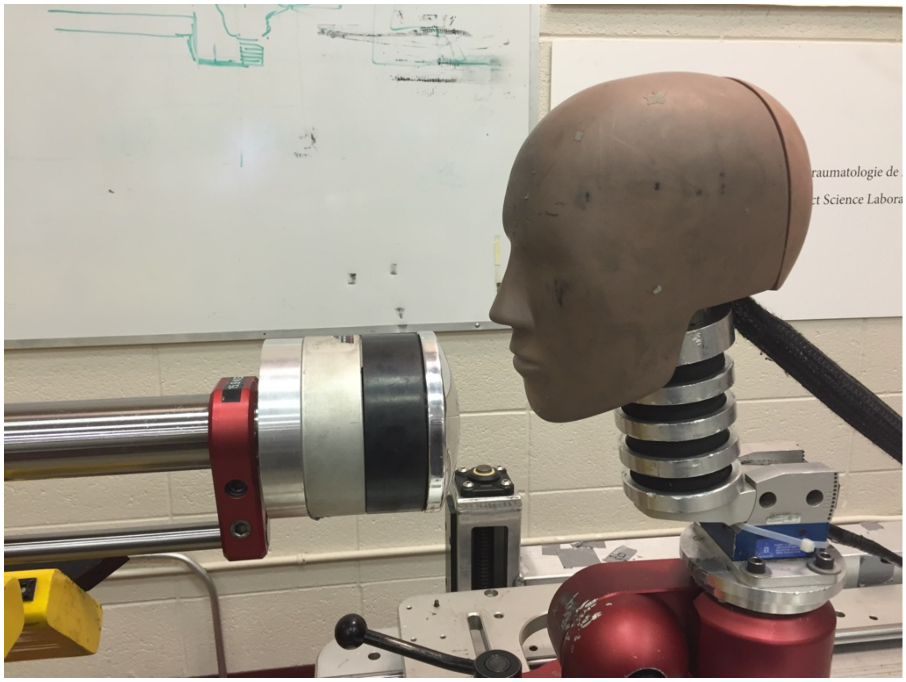

Figure 11: Linear impactor and anvil used to reconstruct knee-to-head impacts

3.9.5 SHOULDER-TO-HEAD COLLISIONS

Eleven (11) shoulder-to-head impacts were reconstructed using the pneumatic linear impactor following the shoulder-to-head collisions reconstruction protocol by Rock (2016). Impacts selected for reconstruction represent injuries to the defensive tackler obtained from tucked-arm, direct shoulder to head impacts by the ball carrier. The impactor was equipped with the lower mass arm (mass = 13.1 kg) as it more closely represents the effective mass of the shoulder. Effective mass of the shoulder (mass = 12.9 kg) was determined from shoulder-to-head collisions in men's ice hockey (Rousseau & Hoshizaki, 2015). The striker was fixed with VN foam reflecting the compliance, 25-30 ms, of shoulder to head impacts determined in previous studies of shoulder-to-head collisions in men's ice hockey and rugby (Rousseau & Hoshizaki, 2015; Rock, 2016). Table position and orientation of the head and neck were adjusted to reflect location and direction of impact (Figure 8).

Peak resultant linear and rotational accelerations from each impact were captured, with peak values averaged across trials. Values were used to input into the UCDBTM finite element model to determine strain of the brain tissue following an impact, measured as maximum principle strain in the cerebrum. Outputs of the model along with peak dynamic response values were used to compare the four common injury events in rugby.

3.10 FINITE ELEMENT MODEL (UCDBTM)

The University College Dublin Brain Trauma Model (UCDBTM) is the finite element brain model that was used in this study. The model was used to determine brain tissue deformation by measuring maximum principal strain (MPS) in the cerebrum of the concussive impacts for comparison across injury events. Maximum principal strain outputs from the model along with peak dynamic response values were compared to proposed injury thresholds from the literature.

The model is comprised of 26, 000 hexahedral elements divided into ten sections: the scalp, skull (cortical and trabecular bones), pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brain stem (Horgan & Gilchrist, 2003, 2004). The material characteristics of the model are represented below (tables 1 & 2).

Table 1: 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 and Tentorium	31.5	0.45	1130
CSF	15000	0.5	1000
Grey Matter	Hyperelastic	0.49	1040
White Matter	Hyperelastic	0.49	1040

Table 2: Material characteristics of the brain tissue for 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 behaviour of brain tissue was characterized as viscoelastic in shear, while the compressive behaviour of brain tissue was considered to be elastic. The following equation was therefore used to define the shear characteristics and viscoelastic behaviour of the brain tissue:

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

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). To account for interaction between the skull and brain, the cerebral spinal fluid was achieved by modelling, allowing it to behave fluidly, like water. A hyperelastic material was used for the brain tissue, represented by the following equation:

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

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

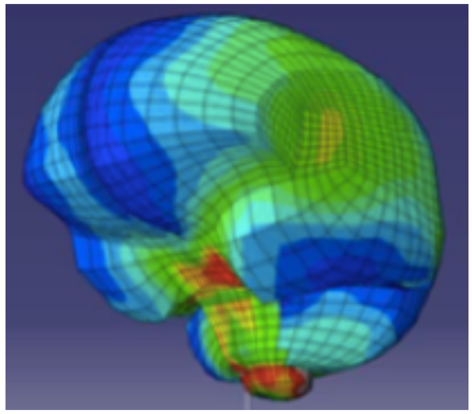

Figure 12: Example of UCDBTM output (colour gradient representing low (blue) to high (red) brain deformation)

3.11 STATISTICAL ANALYSIS

Mechanisms of injury were studied individually to compare between groups. One-way ANOVAs were used to compare peak resultant linear acceleration, peak resultant rotational acceleration, and maximum principal strain values between the three different injury events. Due to the small sample size of the knee-to-head events, this group was excluded from the analysis. The alpha level was set at $p < 0.05$. All data were analyzed using SPSS 21 for Windows (IBM Corp., Armonk, NY, USA).

A power analysis was conducted with six shoulder-to-head impacts. To achieve 80% power ($\beta = 0.2$) a sample size of 5 events is required to detect significance in peak linear acceleration and maximum principal strain and 6 events to detect significance in peak rotational acceleration.

CHAPTER 4: RESULTS

A total of 29 diagnosed concussions of direct head impacts from a collision with a (head (n=9), hip (n=7), shoulder (n=11) and knee (n=2) met inclusion criteria and were reconstructed in the Neurotrauma Impact Science Laboratory. Dynamic response data were obtained from each injury reconstruction and values were used as input for the finite element analysis in which maximum principal strain data were extracted. Dynamic response and brain tissue deformation data were presented in terms of means and standard deviations of peak resultant linear and rotational head accelerations and maximum principal strain. A summary of the dependent variables for each injury event can be found in table 3.

Table 3: Comparison of dynamic response and brain tissue deformation between four injury events causing concussion in men's rugby (standard deviations in parentheses). ** Denotes significant difference

Injury Event	Velocity (m/s)	Event Duration (ms)	Peak Linear Acc. (g)	Peak Rotational Acc. (rad/s^2)	MPS
Head-to-head	5.23 (0.68)	3.10 (0.46)	205 ** (91.7)	15890 ** (8290)	0.428 (0.1519)
Hip-to-head	5.28 (0.850)	28.9 (5.20)	24.7 (12.76)	2650 (912)	0.271 (0.0818)
Shoulder-to-head	5.57 (1.239)	30.3 (4.06)	24.2 (12.12)	3280 (1210)	0.363 (0.1354)
Knee-to-head	4.44 (0.0401)	4.57 (0.0)	111.5 (39.4)	12980 (938)	0.578 (0.237)

4.1 Dynamic Response:

The *Shapiro-Wilks* test revealed that data were normally distributed. However, *Levene's F* test indicated that the homogeneity of variance assumption was not met ($p = 0$). As such, the *Welch's F* test was used at an alpha level of .05 for all subsequent analyses.

There was a statistically significant effect of injury event on peak resultant linear acceleration *Welch's* [$F(2, 43.1) = 55.3$, $p < .001$] and peak resultant rotational acceleration values *Welch's* [$F(2, 46.1) = 36.8$, $p < .001$] for the three events. Post hoc comparisons using Games-Howell test indicated that the mean peak linear head acceleration data for the head-to-head condition ($M = 205 \pm 88.2$) was significantly higher than the hip-to-head condition ($M = 24.7 \pm 12.13$), $p < .001$

and the shoulder-to-head condition (M = 24.2 ± 11.84), p <.001. Post hoc Games-Howell test indicated that the mean peak rotational head acceleration data for the head-to-head condition (M = 15890 ± 8030) was significantly higher than the hip-to-head condition (M = 2650 ± 876), p <.001 and the shoulder-to-head condition (M = 3276 ± 1185), p <.001. The hip-to-head condition did not significantly differ from the shoulder-to-head condition with respect to peak resultant linear (p = .987) or rotational accelerations (p = .987) of the headform (figures 14 & 15).

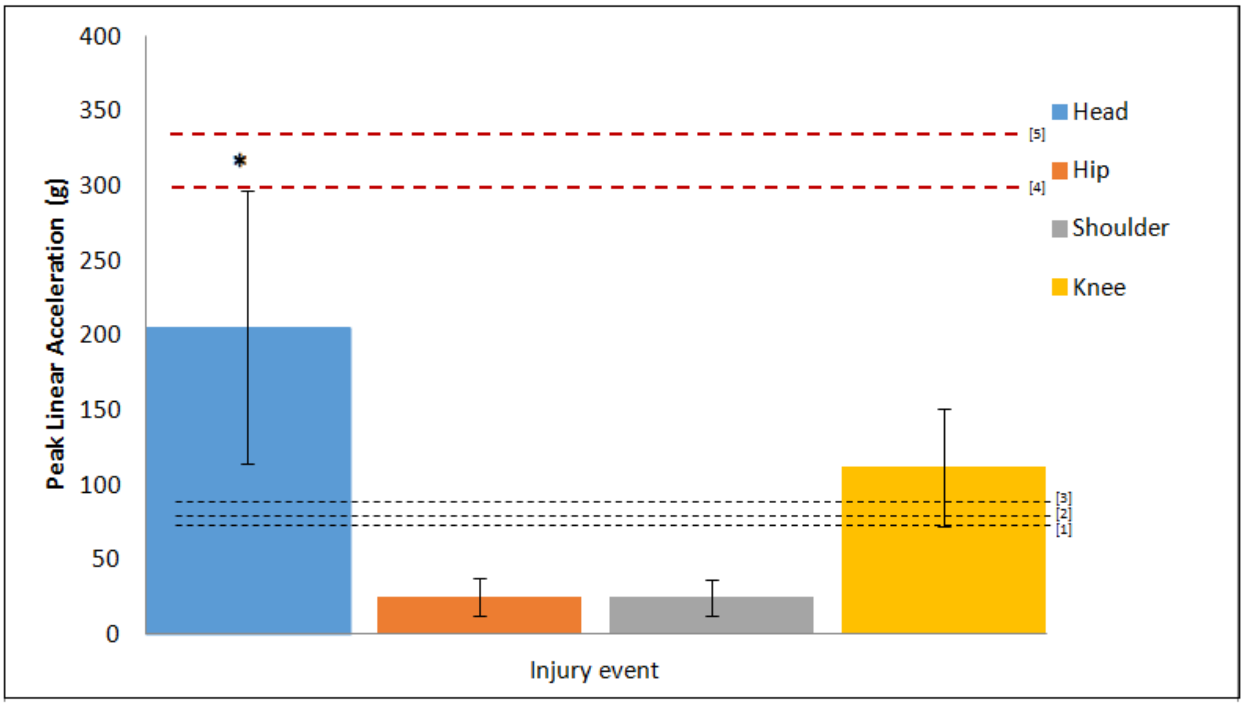

Figure 14: Peak resultant linear acceleration of three concussive injury events (head-to-head, hip-to-head and shoulder-to-head) with proposed risk of injury (black-dashed lines) (1: McIntosh et al., 2014; 2: McAllister et al., 2012; 3: Zhang et al., 2004). Wide red dashed line denotes proposed TBI threshold (4: Yoganandan & Pintar, 2004; 5: Post et al., 2015).

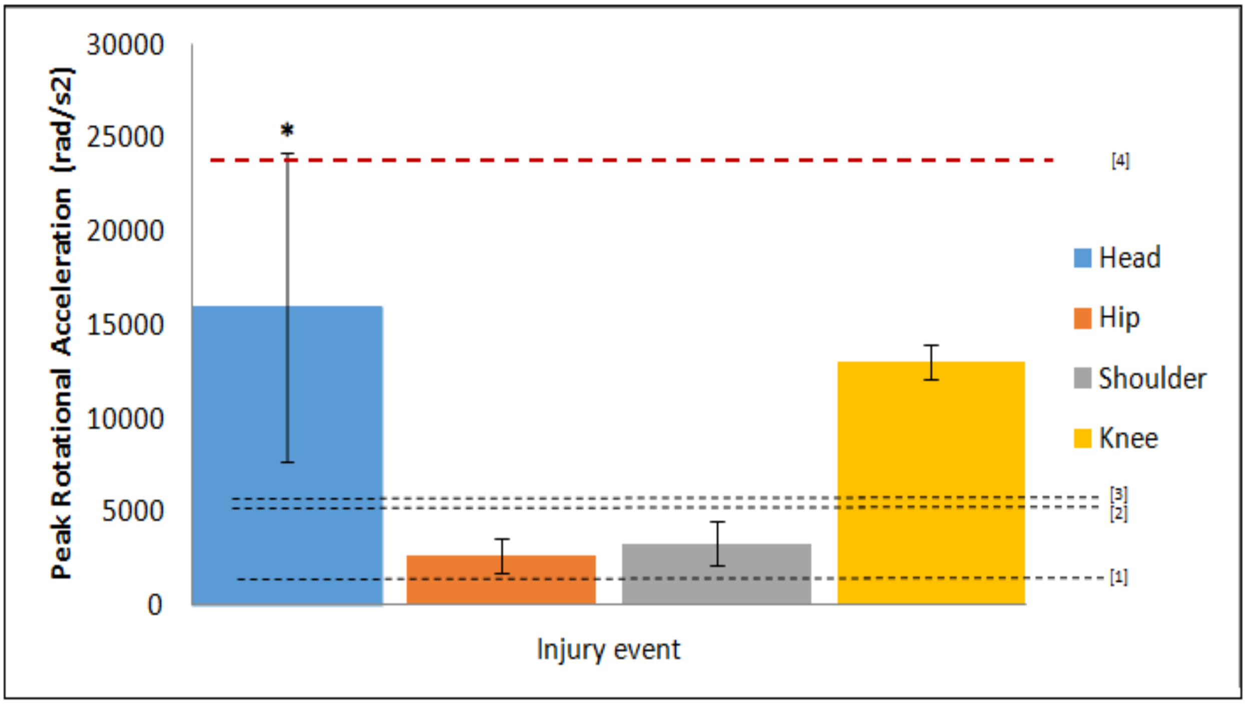

Figure 15: Peak resultant rotational acceleration of three concussive injury events (head-to-head, hip-to-head, and shoulder-to-head) with proposed risk of injury (black-dashed lines) (1: McIntosh et al., 2014; 2: McAllister et al., 2012; 3: Zhang et al., 2004). Wide red dashed line denotes proposed TBI threshold (4: Post et al., 2015).

4.2 Brain Tissue Deformation

The *Shapiro-Wilks* test revealed that data were normally distributed and *Levene's F* test indicated that the homogeneity of variance assumption was met ($p = .114$). There were no statistically significant differences between the impact events for maximum principal strain as determined by one-way ANOVA [$F(2,24) = 2.87$, $p = .076$] (Figure 16).

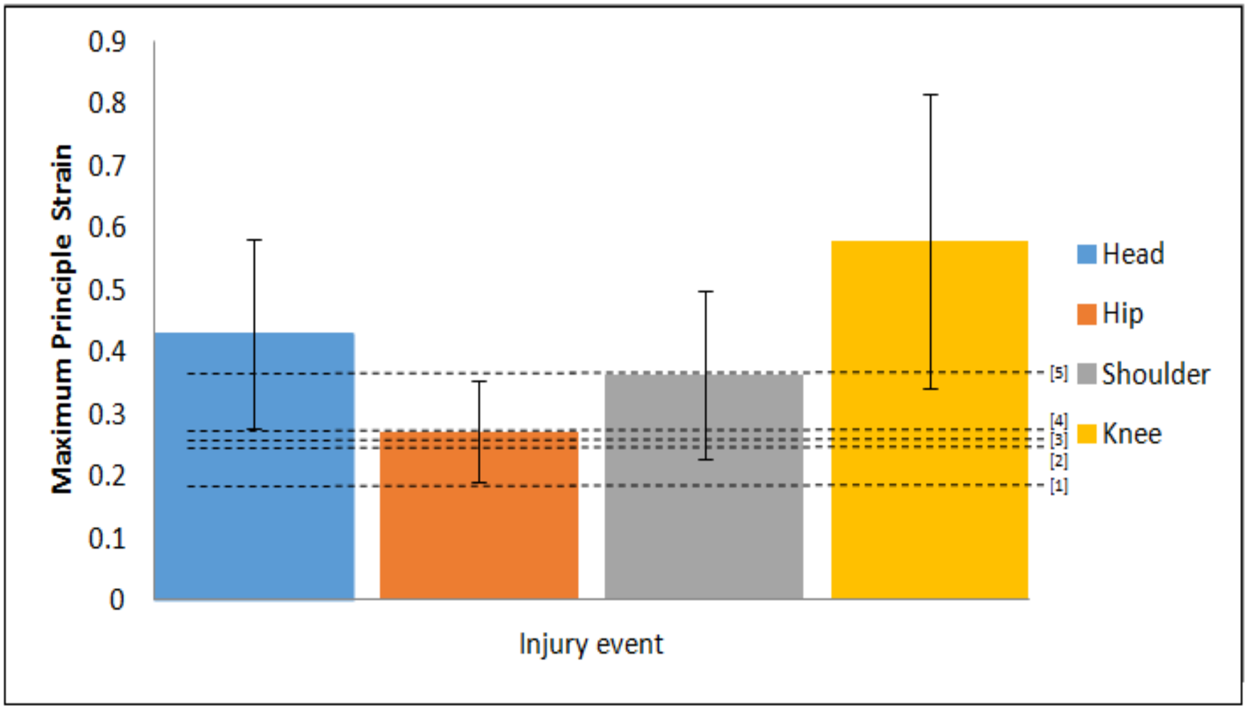

Figure 16: Maximum Principal Strain of three concussive injury events (head-to-head, hip-to-head and shoulder-to-head) with proposed risk of injury (black-dashed lines) (1: Zhang et al., 2004; 2: Kleiven et al., 2007; 3: Patton et al., 2015; 4: McAllister et al., 2012; 5: Willinger 2003).

4.3 Head-to-head Impacts

Head-to-head impacts produced the largest dynamic response of all injury events with mean values of 205 ± 91.7 g and 15890 ± 8290 rad/s^2, for peak resultant linear and rotational acceleration, respectively (table 4). Average maximum principal strain was 0.428 ± 0.1519, second highest of all injury events following knee-to-head impacts, however differences did not reach statistical significance. Average velocity of head-to-head events was 5.23 ± 0.68 m/s and average event duration calculated from the rotational acceleration curve was 3.10 ± 0.46 ms; the lowest compliance of all injury events. Impacts occurred primarily to the front location (R1, Front, L1) followed by impacts to the lateral portion of the head (R3, R3); figure 17.

Table 4: Complete Results Table of Head-to-Head Impact Reconstructions

Case	Velocity (m/s)	Event Duration-linear (ms)	Event Duration-rotational (ms)	Linear (g)	Rotational	MPS	Location	Forward Position	Rotation
1	5.12 (0.0)	4.05 (0.427)	3.18 (0.208)	152.9 (2.55)	22322 (685)	0.522 (0.0242)	Front-S20	0°	20°
2	5.15 (0.0)	4.48 (0.226)	2.67 (0.1258)	128.4 (4.40)	4359.6 (153.3)	0.327 (0.0235)	L1-S10	0°	35°
3	4.04 (0.0)	4.98 (0.236)	4.20 (0.050)	123.0 (1.850)	9396.1 (78.6)	0.284 (0.0035)	L1-F2	15°	340°
4	5.68 (0.1935)	3.13 (0.1041)	3.02 (0.0764)	282 (5.14)	17657 (3030)	0.417 (0.0438)	L1-F2	0°	40°
5	5.86 (0.1212)	3.12 (0.0289)	2.68 (0.1756)	331 (7.75)	24771 (1409)	0.537 (0.0507)	L3-S13	0°	90°
6	5.97 (0.0577)	3.30 (0.1)	2.95 (0.1)	267 (1.845)	16938 (336)	0.541 (0.0356)	R1-S10	0°	340°
7	5.94 (0.1270)	3.37 (0.1756)	3.13 (0.1041)	314 (16.06)	28492.4 (1052)	0.685 (0.0488)	R1-S18	0°	70°
8	4.77 (0.0635)	3.98 (0.202)	2.82 (0.0577)	148.1 (6.36)	11834 (674)	0.316 (0.0184)	L1-S10	15°	30°
9	4.55 (0.0635)	4.47 (0.1258)	3.09 (0.1528)	99.7 (2.21)	7242 (661)	0.223 (0.0004)	R3-S8	15°	320°

Note: * Refer to figure 4 for location grids used to classify impact location

* Forward Position reflects the forward rotation of the head-neck complex (rotations about the frontal axis) (Figure 17)

* Rotation reflects movement about the vertical axis.

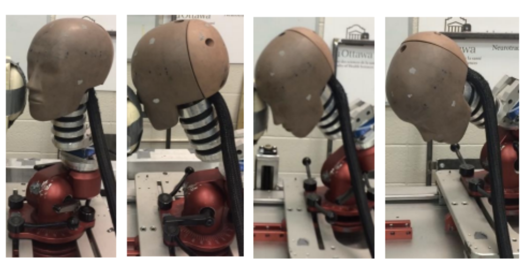

Figure 17: Forward position of head-neck complex rotation about frontal axis in a: neutral (0°), b: 15°, c: 30°, d: 45° positions

4.4 Hip-to-head Impacts

Hip-to-head impacts produced the lowest dynamic response, along with shoulder impacts, with mean values of 24.7 ± 12.76 g and 2650 ± 912 rad/s^2, for peak resultant linear and rotational acceleration, respectively (table 5). Average maximum principal strain was 0.271 ± 0.0818, lowest of all injury events, however differences did not reach statistical significance. Average velocity of hip-to-head events was 5.28 ± 0.850 m/s and average event duration calculated from the rotational acceleration curve was 28.9 ± 5.2 ms. Impacts occurred to the front location (R1, L1) with roughly the same incidence as impacts to the lateral portion of the head (R2-R4; L2-L4); figure 17.

Table 5: Complete Results Table of Hip-to-Head Impact Reconstructions

Case	Velocity (m/s)	Event Duration-linear (ms)	Event Duration-rotational (ms)	Linear (g)	Rotational (rad/s^2)	MPS	Location	Forward Position	Rotation
1	4.60 (0.0229)	29.4 (0.729)	30.1 (0.312)	9.10 (1.249)	2065 (170.8)	0.213 (0.0239)	L2-S9	60°	25°
2	5.74 (0.1300)	25.6 (0.944)	17.9 (0.815)	31.5 (0.656)	2180 (99.2)	0.211 (0.00643)	L1-S10	0°	105°
3	6.76 (0.1039)	21.2 (2.46)	26.8 (0.236)	46.0 (1.411)	4456 (146.1)	0.421 (0.01087)	R2-S9	30°	290°
4	5.83 (0.1559)	22.1 (0.702)	31.4 (0.625)	32.5 (1.504)	2172 (159.8)	0.273 (0.0555)	R1-S10	30°	310°
5	4.43 (0.0800)	34.0 (0.563)	36.1 (0.813)	17.10 (0.557)	1787 (41.6)	0.181 (0.00225)	R3-S12	30°	335°
6	4.89 (0.0520)	30.8 (2.30)	30.0 (0.454)	22.6 (0.379)	2857 (52.6)	0.274 (0.00280)	L3-S8	45°	60°
7	4.74 (0.221)	28.5 (1.561)	27.6 (0.839)	14.23 (1.050)	3038 (313)	0.320 (0.0338)	R1-S17	15°	325°

Note: * Refer to figure 4 for location grids used to classify impact location

* Forward Position reflects the forward rotation of the head-neck complex (rotations about the frontal axis) (Figure 17)

* Rotation reflects movement about the vertical axis.

4.5 Shoulder-to-head Impacts

Shoulder-to-head impacts produced the lowest dynamic response, along with the hip impacts, with mean values of 24.2 ± 12.12 g and 3280 ± 1210 rad/s^2, for peak resultant linear and rotational acceleration, respectively (table 6). Average maximum principal strain was 0.363 ± 0.1354, lowest of all injury events second only to hip-to-head impacts, however differences did not reach statistical significance. Average velocity of shoulder-to-head events was 5.57 ± 1.239 m/s and average event duration calculated from the rotational acceleration curve was 30.3 ± 4.06 ms. Impacts occurred primarily to the lateral portion of the head (R2-R3; L2-L3); followed by impacts to the front location (R1); figure 17.

Table 6: Complete Results Table of Shoulder-to-Head Impact Reconstructions

Case	Velocity (m/s)	Event Duration-linear (ms)	Event Duration-rotational (ms)	Linear (g)	Rotational (rad/s^2)	MPS	Location	Forward Position	Rotation
1	7.14 (0.1155)	24.1 (0.437)	30.7 (0.340)	35.7 (3.57)	4469 (231)	0.541 (0.0249)	R1-S20	15°	325°
2	5.37 (0.1150)	28.0 (1.069)	30.7 (1.377)	18.30 (0.954)	3716 (94.3)	0.455 (0.01582)	R1-S18	45°	45°
3	5.96 (0.0808)	21.3 (0.984)	27.9 (0.132)	29.3 (0.764)	3503 (202)	0.442 (0.0346)	L2-S9	15°	290°
4	5.70 (0.0751)	19.0 (0.752)	26.4 (1.040)	19.2 (2.98)	1178 (181.3)	0.114 (0.0262)	L3-S13	45°	330°
5	7.58 (0.1270)	19.5 (0.257)	25.0 (0.653)	50.9 (2.99)	4446 (199.4)	0.431 (0.0097)	R3-S14	30°	315°
6	3.41 (0.0)	31.7 (0.635)	36.4 (3.20)	8.37 (0.1155)	1558 (10.65)	0.180 (0.0068)	R2-S17	45°	330°
7	4.01 (0.1097)	30.1 (0.541)	36.3 (0.275)	9.97 (0.289)	1929 (86.2)	0.221 (0.01219)	R2-S17	0°	330°
8	5.57 (0.1443)	22.9 (0.1721)	34.5 (0.189)	25.3 (1.473)	3896 (259)	0.397 (0.0217)	R3-S9	15°	330°
9	4.68 (0.0)	26.1 (0.581)	28.2 (0.391)	16.00 (0.200)	2879 (46.6)	0.316 (0.0233)	L2-S17	0°	30°
10	5.45 (0.0693)	21.5 (3.25)	30.8 (0.229)	24.7 (1.559)	4049 (203)	0.439 (0.0271)	R3-S14	45°	25°
11	6.42 (0.0924)	22.7 (0.425)	25.8 (1.028)	28.4 (2.21)	4417 (436)	0.451 (0.0411)	R2-S17	30°	315°

Note: * Refer to figure 4 for location grids used to classify impact location

* Forward Position reflects the forward rotation of the head-neck complex (rotations about the frontal axis) (Figure 17)

* Rotation reflects movement about the vertical axis.

4.6 Knee-to-head Impacts

While not compared statistically due to the low number of impacts, knee-to-head impacts produced the second highest dynamic response, with mean values of 111.5 ± 39.4 g and 12980 ± 938 rad/s^2, for peak resultant linear and rotational acceleration, respectively (table 7). Average maximum principal strain was 0.578 ± 0.237, greatest of all injury events.. Average velocity of knee-to-head events was 4.44 ± 0.0401 m/s and average event duration calculated from the rotational acceleration curve was 4.57 ms. One collision occurred to the front of the head, the other to the left lateral portion of the head; figure 17.

Table 7: Complete Results Table of Knee-to-Head Impact Reconstructions

Case	Velocity (m/s)	Event Duration-linear (ms)	Event Duration-rotational (ms)	Linear (g)	Rotational (rad/s^2)	MPS	Location	Forward Position	Rotation
1	4.49 (0.1721)	7.60 (0.1500)	4.50 (0.1732)	83.7 (5.84)	13650 (1114)	0.746 (0.0472)	Front-F4	60°	270°
2	4.43 (0.08)	6.25 (0.346)	4.63 (0.382)	139.3 (3.45)	12320 (771)	0.411 (0.01434)	L4-S7	30°	285°

Note: * Refer to figure 4 for location grids used to classify impact location

* Forward Position reflects the forward rotation of the head-neck complex (rotations about the frontal axis) (Figure 17)

* Rotation reflects movement about the vertical axis.

CHAPTER 5: DISCUSSION

Head and brain injuries are a problem in our society whereby athletes, particularly those participating in contact sports such as rugby, are at a greater risk of sustaining a head injury. Recent evidence has linked concussion to long-term health problems including a persistent decline in cognitive functioning, emotional impairments, and the potential to develop long-term neurological disorders. It has been identified in previous research that rugby athletes are obtaining concussive injuries from shoulder-to-head, head-to-head, knee-to-head, and hip-to-head collisions. While research has been conducted in an effort to describe incidence and mechanisms of concussive injury in rugby, little is known about the injury events that lead to concussion and how they create risk of concussion. In order to develop effective interventions in reducing the incidence of injury, research must first be targeted at understanding the differences between injury events. Therefore, the purpose of this study was to obtain dynamic response and brain tissue deformation data of the common injury events that lead to concussion in professional men's rugby and to analyze and compare differences in these events. The hypothesis of this thesis was that the low duration events, head-to-head and knee-to-head, would result in the greatest values of both dynamic response and brain tissue deformation, followed by shoulder-to-head and hip-to-head events, lowering risk of injury with increasing levels of compliance. The results of this study confirm that head-to-head concussive impact events create large magnitude and low duration linear and rotational acceleration curves, but do not have a significant effect on maximum principal strain. Results of this study found that dynamic response was significantly higher in head-to-head collisions than shoulder-to-head and hip-to-head collisions for both linear and rotational acceleration, whereby there were no significant differences between the shoulder and hip injury groups. The head-to-head impact condition produced an average peak linear acceleration of 205 ± 91.7 g at 5.22 m/s. Not only is the short duration event producing significantly larger dynamic response values than the other injury events, linear acceleration values are approaching levels where you would expect to see Traumatic Brain Injury (TBI), such as skull fracture or subdural hematoma (Post et al., 2015; Yoganandan & Pintar, 2004). Knee-to-head collisions also produced large values of linear (111.5 g) and rotational (12983 rad/s^2) acceleration, exceeding 80% risk of mTBI injury with respect to linear and rotational acceleration and MPS (Zhang et al., 2004); however, due to the limited number of cases that met inclusion criteria, this condition was not compared statistically. The two knee-to-head impacts that were reconstructed demonstrate that even at a relatively low velocity, 4.4 m/s, knee-to-head impacts create high levels of risk.

Comparison of maximum principal strain values found no significant differences between the injury events, however head-to-head and knee-to-head impacts did produce the greatest values (0.428 and 0.578 respectively). This finding is not surprising considering that all injury events in the present study were reconstructions of clinically diagnosed concussions. While the hip and shoulder impacts produced statistically lower dynamic response values than the head-to-head impacts, there were no differences in maximum principal strain between the events. This is due to the high compliance of the hip and shoulder impacting systems, which created longer time to peak than the head-to-head events, thereby producing similar strain values. Previous research has demonstrated that when magnitude is kept constant between two conditions of varying compliance (time to peak), the longer duration event will create a greater MPS response, since the model takes into account the duration and magnitude of the event, creating an equalizing affect of the brain model (Post et al., 2012). Compliant impacting surfaces manage some of the impact energy to reduce peak linear and rotational accelerations but also influence the duration of impact (Post et al., 2012a; Post et al., 2012b). While the head-to-head and knee-to-head events have a very low compliance and short time to peak, they produce larger magnitude events, explaining why these particular injury events produced large MPS values.

Each of the common injury events have different combinations of the input variables: mass, compliance, velocity, and location of impact. Analysis found that all injury events had similar average resultant velocities for each group: head: 5.22 m/s; hip: 5.28 m/s; shoulder: 5.57 m/s, knee: 4.46 m/s (range: 3.41-7.65 m/s). These values are consistent with average tackle velocities of 3.0-11.4 m/s which was previously reported in the literature (McIntosh et al., 2000). Head-to-head collisions have the lowest compliance (~3ms) and mass (4.8kg) of the four injury events, whereby the knee-to-head injury event contains large mass (15.9 kg) and slightly higher compliance (~5-7 ms). Both the hip and shoulder injury groups are characterized by high compliance (25-30 ms) and moderate mass (13.1 kg). The shoulder and hip loading conditions resulted in different dynamic response impact curves from other injury events, but no differences in MPS. There were no differences between the hip and shoulder events however, because both events were characterized by similar effective mass, velocity and event duration.

Impact location varied with respect to the injury event and the type of tackle. In hip-to-head collisions, players are being struck most often to the temporo-parietal region, characteristic of defenders tackling the ball carrier at hip level, moving their head to the side. Similarly, shoulder impacts were primarily striking the lateral portion of the head. This is consistent to what has previously been documented as the most common impact location creating concussion during the rugby tackle (McIntosh et al., 2000). Location of head-to-head impacts are slightly more variable as they are often accidental impacts, occurring most frequently to the face and lateral aspects of the head. No pattern of knee-to-head impacts emerged as there were only two recorded instances in which location of impact could be identified; to the front of the jaw and to the lateral head. A summary of impact locations can be found in Table 8; Figure 17.

Current risk reduction protocols address detecting and diagnosing concussions on field, however few address reducing incidence of injury. The Concussion Interchange Rule (CIR), implemented in 2014, aims at detecting concussions, by allowing players suspected of having a concussion to be removed from the field to undergo an evaluation without an interchange tallied against the team. While the CIR may help to better protect athletes once they have sustained an injury, there must be greater focus on preventing the initial trauma from occurring. Given that each of the injury events are created differently, this study reveals that harm reduction strategies aimed at reducing the incidence of concussions in rugby must consider event-specific interventions.

Shoulder-to-head injury events are frequently cited as the most common injury event creating concussion and account for roughly 35% of all concussions in the sport (Gardner et al., 2015). Video analysis led to the observation that the majority of these collisions involve direct shoulder to head impacts, that are considered illegal high-tackles, however players are not being penalized. Therefore, in order to reduce incidence of concussion caused by shoulder to head collisions, there must be a greater enforcement of penalties for illegal high tackles. Head-to-head and knee-to-head events each represent roughly 20% of concussions in the sport (Gardner et al., 2015). While they are reported to occur less frequently, they both reflect high risk events, with acceleration values approaching levels where you'd expect to see Traumatic Brain Injury (Post et al., 2015; Yoganandan & Pintar, 2004). These types of impacts also appear to be accidental impacts, therefore in order to protect athletes, it is recommended that governing bodies implement rule changes that ban and assign penalties for all head impacts. Hip-to-head collisions represent roughly 10% of concussions in the sport (Gardner et al., 2015). These collisions are associated with legal match play as the defender goes to tackle the ball carrier at hip level, the ball carrier moves to avoid the tackle and he strikes the defender with his hip. In order to reduce incidence of hip to head collisions, it is recommended that there be a modification of tackling technique that protects the head or rule changes that penalize players for head contact.

Now that each of the common injury events have been studied individually, it is possible to make informed decisions about interventions relating to improved protective equipment, athletic training or referee rule enforcement, with respect to each unique injury event. This research has implications for developing protective equipment and also provides a better understanding of differences between the common injury events creating concussion in men's rugby.

Future research could investigate differences between injury events in rugby union and 7s rugby as well as amongst female and youth athletes. It would also be beneficial to obtain a larger sample of knee-to-head injuries in order to statistically compare dynamic response values to other common injury events. This research was also limited to diagnosed concussive injury events and it would therefore be beneficial to explore differences in injury events and cerebral response of head impacts occurring at sub concussive levels.

LIMITATIONS

The Hybrid III 50th percentile head form used in this study is commonly used as a physical model in impact reconstructions due to its reliability. However, this head form is comprised of metal and rubber, making it less compliant and deformable than a human head, therefore it does not provide a biofidelic response to impacts (Deng, 1989; Kendall et al., 2012). The University College of Dublin Brain Trauma finite element model used in this study was validated against cadaveric head impact research and therefore may not be an accurate representation of live human brain tissue deformation following an impact. Impact variables (mass and compliance) were selected for each category to represent characteristics of the collision and remained constant for event reconstructions within each group (ie. constant mass and compliance for all head-to-head impacts). For instance, mass of the pendulum remained constant for all head-to-head events at 4.8 kg. This methodology would therefore not account for individual differences in head mass between athletes, or differences in effective head mass for different collision events.

DELIMITATIONS

The reconstructions performed in this study will recreate concussive impacts taken from National Rugby League video whereby high quality video was available for analysis. The results of this study may therefore only apply to elite male athletes and may not reflect incidence or circumstances of injury of other age groups or non-elite athletes. Further, only impacts that occur in clear view and allow for calculation of velocity will be included in the analysis therefore they may be of higher magnitude than impacts causing concussion whereby the impact was obstructed from view and the mechanism was unclear.

CONCLUSION

Better understanding of the biomechanics of injury events and how they relate to risk will help inform decisions about interventions and is a step towards developing strategies to reduce the incidence of concussion in rugby. Results of the current study demonstrate that each of the common injury events are characterized by different combinations of velocity, mass, compliance and location. While maximum principal strain did not differ between groups, head-to-head impacts produced significantly greater dynamic response values, with linear accelerations approaching traumatic brain injury thresholds. There were no differences between shoulder-to-head and hip-to-head events with respect to dynamic response or brain tissue deformation. Harm reduction strategies aimed at reducing the occurrence and severity of concussions in rugby should implement event-specific interventions. This research has implications not only in elite men's rugby, but across contact sport, highlighting the importance of creating event specific strategies to reduce risk of injury.

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APPENDIX A

Distribution of Impact Locations

Table 8: Distribution of impact location across injury groups

Injury Event	Number of Front Impacts	Number of Lateral Impacts	Number of Rear Impacts
Head-to-head (n=9)	7	2	0
Hip-to-head (n=7)	3	4	0
Shoulder-to-head (n=11)	2	9	0
Knee-to-head (n=2)	1	1	0

Front: R1, Front, L1
Lateral: R2-R4; L2-L4
Rear: R5, Back, L5

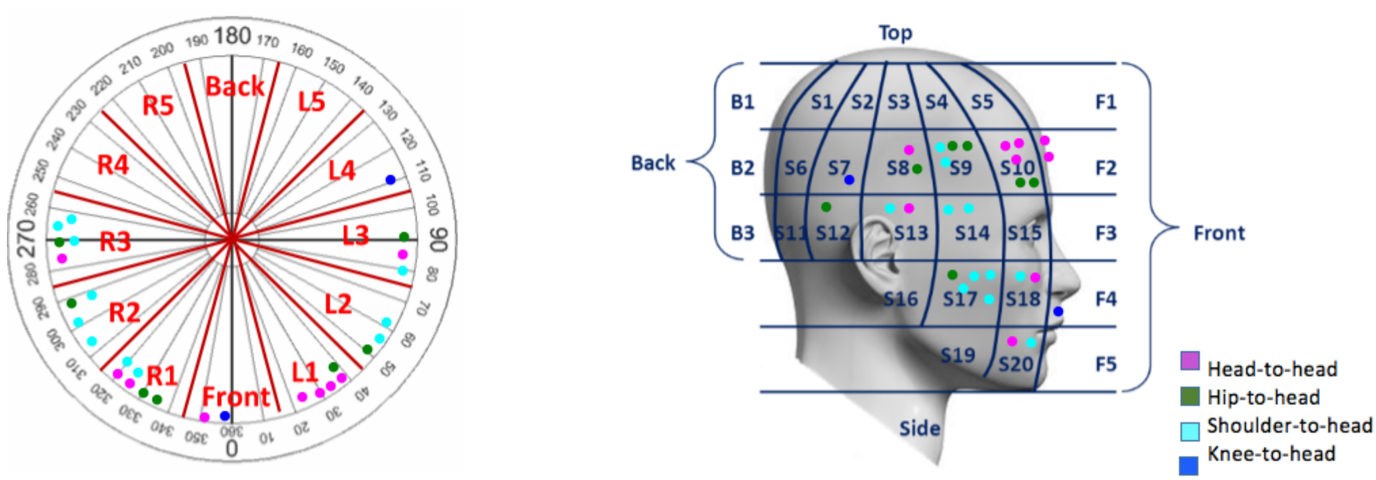

Figure 17: Distribution of impact location across injury groups

APPENDIX B

Complete Results Table of Hip Strikes to Hybrid III Headform

Table 9: Dynamic Response Data from Hip Strikes to Hybrid III headform in determining Hip Striker Compliance

Participant	Velocity (m/s)	Event Duration-linear (ms)	Event Duration-Rotational (ms)	Linear Acceleration (g)	Rotational Acceleration (rad/s^2)
1	3.80 (0.557)	24.0 (2.17)	28.0 (2.49)	19.3 (5.44)	1776 (941)
2	3.80 (0.612)	25.7 (0.975)	27.6 (1.882)	36.6 (5.06)	2456 (558)
3	7.58 (0.541)	21.4 (5.07)	25.5 (4.08)	47.5 (11.27)	3416 (1371)
4	4.29 (0.046)	20.8 (3.53)	26.9 (2.18)	28.7 (7.08)	1594 (451)
5	4.27 (0.023)	28.2 (3.18)	27.4 (2.20)	26.7 (3.37)	1747 (55.9)
6	3.94 (0.589)	23.4 (2.71)	31.0 (1.59)	30.2 (6.06)	1664 (147.0)
7	3.92 (0.630)	28.2 (1.301)	27.8 (2.23)	22.6 (6.97)	1396 (406)
8	7.44 (0.075)	26.5 (2.27)	31.1 (4.85)	44.3 (16.97)	5579 (2370)