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Quantifying and Comparing the Head Impact Biomechanics of Different Player Positions for Canadian University Football

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A thesis submitted in partial fulfillment of the requirements for the Master of Science degree in Kinesiology

QUANTIFYING AND COMPARING THE HEAD IMPACT BIOMECHANICS OF DIFFERENT PLAYER POSITIONS FOR CANADIAN UNIVERSITY FOOTBALL

Thesis Monograph

by

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Abstract

Differences between Canadian and American football could affect the magnitudes of head impacts and risk of concussion to Canadian players. This study sought to quantify and compare the number, magnitude, and location of impacts that Canadian University football players of different positions experienced during games and practice in a season. A kinematic measuring device collected the linear accelerations and rotational velocities of the head from impacts experienced by players competing in practices and games. The impact magnitudes that were experienced in games were significantly larger than in practice. The offensive back position and wide receiver position had significantly larger peak linear and rotational accelerations than the offensive linemen position. The magnitudes of impacts experienced by the wide receiver position in Canadian football were larger and not consistent with previous American studies, likely due to the pass style offence that is common in Canadian football. We observed that the head impact magnitudes vary by position, and session type in Canadian football.

Keywords

Concussion, head impacts, peak linear acceleration, peak rotational acceleration, football, biomechanics

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1 Introduction

There are an estimated 3.8 million sports related head injuries every year in the United States (Langlois, Rutland-Brown, & Wald, 2006); Football players account for almost half of these head injuries (Hootman, Dick, & Agel, 2007). The majority of reported head injuries are mild traumatic brain injuries (mTBI) or concussions. The Concussion in Sport Group defined concussion as a complex pathophysiological process affecting the brain, induced by traumatic forces, where symptoms can vary between occurrences for an individual and between people (McCrory et al., 2013). Often concussions go unreported and untreated since coaches and trainers struggle with proper symptom identification (Gessel, Fields, Collins, Dick, & Comstock, 2007). Identification, reporting, and proper treatment of concussions is very important; repetitive head impacts and concussions can lead to detrimental long-term effects on brain function and decreased quality of life (Guskiewicz et al., 2005; Omalu et al., 2005). The mechanisms of concussion injury are related to excessive linear and rotational acceleration of the head from impacts.

Early studies of football players' head accelerations used football helmets instrumented with linear accelerometers (Moon, Deedle, & Kovacic, 1971; Naunheim, Standeven, Richter, & Lewis, 2000; Reid, Epstein, O’Dea, & Louis, 1974) . They reported a range of helmet accelerations and concluded that it was difficult to define an acceleration threshold for concussion. However, they proved that head accelerations of football players could be measured while they compete in games and practices. Later, a series of projects examined the biomechanics of concussion in the National Football League (NFL; Pellman, Viano, Tucker, Casson, & Waeckerle, 2003). They used kinematic data of football players that received concussive impacts to determine the angle of impact, speed of impact and player head kinematics. Using these data the authors were able to recreate the impacts in a controlled setting using human representative crash test dummies (the Hybrid III anthropometric testing device; HIII) and a linear pneumatic impactor. Data from the impact recreation showed these players experienced a mean peak linear acceleration of 98 ± 28 g and a mean peak rotational acceleration of 6,432 ± 1,813 rad/s^2 (Pellman et al., 2003). Re-creation of impacts using the HIII provides insight on concussive impacts in football games; however the large variability in peak linear accelerations was consistent with previous research (Moon et al., 1971; Naunheim et al., 2000; Reid et al., 1974). Being able to collect other impact parameters through in vivo testing would help to better inform how concussions occur.

The need for in vivo testing prompted the development of the Head Impact Telemetry (HIT) system. The HIT system measures real-time head accelerations experienced by football players. The system uses an array of six linear accelerometers mounted in the crown of a football helmet to measure the linear accelerations and location of impacts experienced by players (Duma et al., 2005). A proprietary algorithm determines the peak linear and rotational acceleration at the centre of mass of the head and provides severity indexes of the impact (Chu, Beckwith, Crisco, & Greenwald, 2006; Crisco & Chu, 2004). This instrumentation has been applied to the helmets of players at various levels from youth to college in the United States (US). Currently there is no research quantifying head accelerations of Canadian (CAN) Football players at any level of play.

The Canadian game has substantial differences compared to the American game. Three major differences are the size of the field (CAN=110yds x 65yds, US=100yds x 53yds), the number of downs to achieve first down (CAN=3 downs, US=4 downs), and the number of players in play (CAN=12, US=11). These differences may affect the concussion rate and head impact severity. For example, the larger field size in Canada may result in increased head accelerations to the ball carrier during the tackle due to a potential increase in closing distance between the tackler and ball carrier (Ocwieja et al., 2012). The differences of the number of downs and players influence play selection; typically the Canadian game uses more passing plays than in the American version. A run style offensive scheme was associated with more head impacts to players than a pass style scheme; however, the pass style scheme was associated with higher magnitudes of head accelerations (Martini, Eckner, Kutcher, & Broglio, 2013). The differences in play styles may lead to different magnitudes of head accelerations in Canadian football players and subsequently may affect the risk of concussive head injuries. Accordingly, it is important to evaluate the magnitude of head impacts in Canadian football because American data cannot be generalized to the very different Canadian game.

2 Literature Review

2.1 Concussion: What is it and what are its symptoms?

Concussion has been defined many different ways. The Congress of Neurological Surgeons defined concussion nearly 50 years ago as "a clinical syndrome characterized by immediate and temporary impairments of neurological function such as alterations of consciousness, disturbance of vision, and disturbance of equilibrium due to mechanical forces" (Congress on Neurological Surgeons, 1966). This definition was limited as it did not account for a number of common symptoms associated with concussion, and it also did not include minor impact injuries that result in persistent cognitive and physiological symptoms.

The American Congress on Rehabilitation Medicine presented its definition of concussion in 1993. This time concussion was categorized as a mild traumatic brain injury. A mTBI is,

> "a traumatically induced disruption of brain function as manifested by: loss of consciousness (LOC), any loss of memory for events before or after the accident, any alteration in mental state at the time of the accident and focal neurological deficits that may or may not be temporary but where the severity of the injury does not exceed the following: LOC of approximately 30 minutes, after 30 minutes of the injury an initial Glasgow Coma Scale of 13-15, and post traumatic amnesia not greater than 24 hours." (Mild Traumatic Brain Injury Committee, Interest, & Group, 1993).

This definition created a controversy regarding the distinction between concussion and mTBI. Is concussion the same as mTBI or do they represent different conditions? As a general rule, concussion is used in a sports context, while people receiving concussions outside of sports will be diagnosed with an mTBI (Tator, 2009). There are practical implications to this controversial wording. For example, research has shown that patients received greater amounts of care from health care professionals if their injury was described as an mTBI rather than concussion (DeMatteo et al., 2010).

The controversy of concussion or mTBI was addressed in 2012 at the 4th International Conference on Concussion in Sport. This conference provided one of the more complete definitions of concussion, stating that concussions are a brain injury and a subset of traumatic brain injury (TBI), and that there is no difference between a mTBI and a concussion (McCrory et al., 2013). They added this clarification of concussion and mTBI to their previous definition of concussion, which defined concussion as a complex pathophysiological process affecting the brain induced by traumatic biomechanical forces (Aubry & Cantu, 2002). Several common biomechanical, clinical and pathological features help define a concussive head injury. The first feature is that concussions may be caused by a direct blow to the head, neck, face or elsewhere on the body with an impulsive force transmitted to the head. The second feature is that concussions usually result in the rapid onset of short-lived impairment of neurological function that resolves spontaneously. However, signs and symptoms can grow over a number of minutes to even hours after contact. The third feature is that concussions may result in neuropathological changes, but the acute clinical symptoms reflect a functional disturbance rather than a structural injury. Accordingly concussion does not result in abnormalities that are apparent through conventional neuroimaging techniques such as magnetic resonance imaging, or computerized tomography. However, studies show that techniques such as diffuse tensor imagining and functional magnetic resonance imaging can be used to detect structural differences in the brain for concussed individuals (Jantzen, Anderson, Steinberg, & Kelso, 2004; Koerte et al., 2012; Murugavel et al., 2014). Finally, concussion results in a graded set of clinical symptoms that may involve LOC, or no LOC but feeling dazed. Symptoms usually resolve over a week or two; however, in some cases symptoms may be prolonged up to months or even years (McCrory et al., 2013). The concussion in sport group definition of concussion was one of the first definitions that did not emphasize LOC in determining a concussive head injury.

Symptoms of a concussion usually manifest immediately after receiving the injury. Signs and symptoms can fall into one of four categories (CDC, 2013). The first are physical or somatic symptoms including headache, blurred vision, dizziness, fatigue, drowsiness, sensitivity to light and noise, balance problems and finally nausea or vomiting. Postural instability can be another symptom of a concussion and can be used to diagnose concussion. The next two categories are cognitive and emotional symptoms. Cognitive symptoms usually present as difficulty thinking clearly, feeling slowed down, difficulty concentrating and remembering new information. Emotional symptoms include irritability, sadness, feeling more emotional, nervous or anxious. Sleeping more or less than usual is the last category of the signs and symptoms of a concussion (CDC, 2013). Patients can experience more than one symptom and experience symptoms from more than one category. Therefore, clinical assessment of this injury can be difficult because symptoms vary from person to person. A battery of tests evaluating symptom scores, postural stability and cognitive function are used to diagnose concussion and using these tests together is more sensitive at detecting an injury than any one alone (Guskiewicz & Register-Mihalik, 2011).

2.2 Concussion: How does it happen?

Although there has been disagreement over the definition of concussion, there is consensus that concussions result from impacts to the head (Congress on Neurological Surgeons, 1966; McCrory et al., 2013; Mild Traumatic Brain Injury Committee et al., 1993). The definition supplied by the Concussion in Sport Group attributes concussion resulting from biomechanical forces causing trauma, such as a direct blow to the head, neck and face. It can also occur when an impact occurs to the body and the impulsive energy from the impact is transferred to the head (McCrory et al., 2013). One of the first studies investigating movement of the brain after receiving an impact occurred in 1946. Researchers replaced the top of the skull of live Macaque monkeys with a clear plastic skull analog top (Pudenz & Shelden, 1946). The monkeys' heads were impacted with small metal projectiles and the motion of the brain was captured on video. From the video they quantified gliding motions of the brain relative to the skull after the energy from the impactor was transferred to the skull and brain. The types of impulsive motions that can be transferred to the head from impacts are evident as accelerations. There are two types of acceleration associated with concussive impacts, linear and rotational acceleration (Meaney & Smith, 2011).

Linear head acceleration is the change in linear velocity of the head. Accelerations can be along single axes, or in combinations of the sagittal (forward/backward), coronal (side to side), and transverse planes (up and down; Figure 2.1a). Linear head accelerations occur when the perturbation force is applied through the centre of mass of the head. Linear acceleration of the head is measured in g's, where 1 g is equivalent to 9.8 m/s^2.

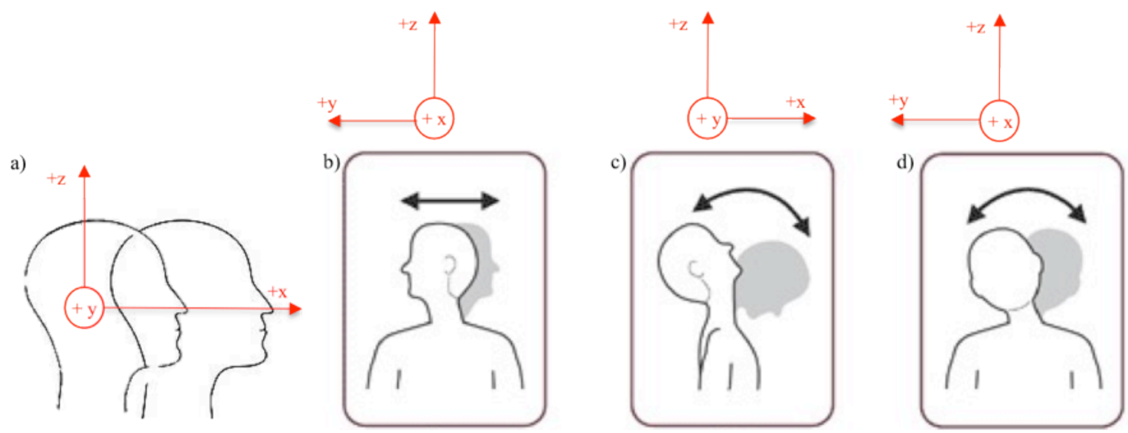

Figure 2.1: Illustration of the various forms of head acceleration: a) an impact to the back of the head causing linear acceleration along the x-axis of the head in the sagittal plane, b) rotation of the head in the transverse plane (yaw), c) rotation of the head in the sagittal plane (pitch) and d) rotation of the head in the coronal plane (roll). The coordinate system conventions are shown in red, and the circled axis is directed into the page.

A combination of linear acceleration and rotational acceleration will occur when an external force is applied eccentrically with respect to the centre of mass of the head. Rotational acceleration is the change in rotational velocity of the head. Rotational accelerations are measured in radians/s^2 (rad/s^2) and they cause the head to rotate about the neck. The head can rotate in a single plane or in combinations of planes: yaw or spin (transverse plane, where the head rotates toward either shoulder; Figure 2.1b), pitch (sagittal plane, where the head rotates forwards or backwards toward the chest or back; Figure 2.1c), and roll (coronal plane where the head rotates from left to right towards the shoulders; Figure 2.1d).

2.2.1 The Role of Linear Acceleration in Concussion

Early studies examining the role of linear acceleration in concussion focused on correlating measures of peak linear acceleration of the head to intracranial pressures measured inside the heads of human cadavers (Gurdjian, Lissner, Evans, Patrick, & Hardy, 1961; Thomas, Roberts, & Gurdjian, 1966). Pressure gradients inside the skull increased with increasing linear accelerations of the head. The pressures were greatest at the site of the impact while the other side of the skull, the countercoup side, experienced a decreased pressure. The differences in high and low intracranial pressures created a pressure gradient inside the skull. The researchers hypothesized that this pressure gradient caused the brain to move in the skull and caused shear stresses to develop and injure the brain tissue (Thomas et al., 1966).

Recently, researchers have developed finite element (FE) models of the human skull and brain to investigate strains acting on specific brain structures. The Wayne State University Brain Injury Model (WSUBIM) was created to understand brain injury from both direct impacts to the head as well as head accelerations from blows to the body (Zhang, Yang, & King, 2001). This model has confirmed that pressure gradients caused by linear acceleration causes strains on brain structures. The magnitude of the strains were positively correlated to the magnitude of the peak linear acceleration, and also correlated to the direction of impact. Larger intracranial pressures were created when impacts occurred to the side of the head versus impacts to the front of the head. Impacts to the side of the head produced larger strains in the corpus callosum and brain stem of the model. It also provided insight that levels of strain injuries to the brain from pressure gradients are also dependent on impact location (Zhang et al., 2001).

2.2.2 The Role of Angular Acceleration in Concussion

A recent review identified that intracranial pressure gradients are primarily caused by linear acceleration, however brain tissue deformation (strain) is primarily associated with rotational head impact accelerations (Meaney & Smith, 2011). The initial research evaluating the effects of rotational acceleration on the brain was conducted on primates (Ommaya, Rockoff, Baldwin, & Friauf, 1964). They observed that the probability of experimentally inducing concussion to Macaque monkeys increased when their cervical spines were not supported with an external brace. This produced larger rotational accelerations of the head when it was impacted from a metal projectile, even at lower velocities, and increased the incidence of concussion. Researchers attributed this result to a larger amount of shearing strain and tensile strain getting transmitted to the cervical spinal cord of the monkey. However, when monkeys were braced, the shearing and tensile strains to the cervical spinal cord were reduced, and the incidence of concussion was reduced even at high velocity impacts and larger linear accelerations of the head after impact.

Gennarelli (1972) expanded on the work from Ommaya and clarified the roles of linear and rotational acceleration in producing concussion (Gennarelli, Thibault, & Ommaya, 1972). Squirrel monkeys received impacts to the head in the sagittal plane that produced predominately linear acceleration or predominately rotational acceleration of the head. All monkeys that received isolated rotational acceleration of the head received concussions, while monkeys that received isolated linear acceleration of the head did not experience concussions. The researchers proposed that rotational acceleration produced larger shearing strains and tensile strains in the cervical spinal cord and brain stem of the monkey than did linear accelerations.

The shearing strains on the brain from rotational acceleration produced widespread damage that was described as a diffuse axonal injury or impairment (DAI; Adams, Graham, Murray, & Scott, 1982). The human brain is made up of white matter and grey matter. Grey matter is composed of neuron cell bodies and is primarily associated with information processing and cognition. In contrast, white matter is the axonal connection between neuron cell bodies and acts as a relay to coordinate the communication and distribution of information in the brain through electrical impulses called action potentials (Fields, 2008). In a DAI white matter (axonal connections) are disrupted (Adams et al., 1982). This disruption can occur at various levels of the brain; however, white matter tracts in the corpus callosum and the brain stem are the most common sites that are injured from head impacts. These two structures distribute information from the left and right hemispheres of the brain as well as distribute information down the spinal cord and then on to various parts of the body.

DAIs also occur when large impulsive forces to the body are transmitted to the head causing it to accelerate. The direction of an impact to either the body or to the head influences the risk of receiving a concussive injury (Gennarelli et al., 1982). Monkeys whose heads were rotationally accelerated in the coronal plane had LOC lasting longer than 15 minutes, and even up to 6 hours, and larger amounts of DAI. Concussions still occurred from impacts in the sagittal and transverse plane, but required larger magnitudes of rotational acceleration. Once again, the deep structures of the brain, the corpus callosum and the brain stem, had the most DAI (Gennarelli et al., 1982). Rotational acceleration causing shearing strains and DAI illustrates the importance of this mechanism in concussions. Accordingly both the linear and rotational accelerations of the head from impacts modify the degree of the injury to the brain as well as provide insight to the location of the injuries in the brain.

2.3 The Long-Term Consequences of Multiple Concussions and Repetitive Head Impacts

In a recent review the majority of research on the long-term consequences of concussion and repetitive head impacts has come from studies involving athletes (Rivara & Graham, 2014). Contact and non-contact sports offer an opportunity to gain information on concussion because the athletes competing in them have a high likelihood for force to be transmitted to the brain. One study monitored collegiate football players to determine the incidence of concussion and time to recovery following concussion (Guskiewicz et al., 2003). They showed that football players were 3 times more likely to receive another concussion when they reported having had already experienced 3 or more concussions in their lifetime. They also showed that symptoms of a concussion resolved slower (greater than a week) in players with a history of multiple concussions. There is also evidence showing that while recovering from a concussion, a repeat injury can occur with less force and even lead to catastrophic results such as death (Bey & Ostick, 2009). Thus, receiving a concussion puts an athlete at more risk for receiving another concussion in the future and requires more time for the athlete to recover from the injury.

Studies that have attempted to understand the relationship between multiple concussions and head impacts on neurocognitive function have had mixed results. One study found no difference in pre and post season cognitive testing scores between athletes in contact and non contact sports, (McAllister et al., 2012). However, this study also reported that athletes who competed in contact sports (football and hockey) performed worse on a measure of new learning and had slower reaction times when assessed through the Immediate Post-Concussion Assessment and Cognitive Test (ImPACT). In contrast, another study found no difference on any scores from the ImPACT between athletes who had not experienced any concussions versus those that had experienced only one concussion (Covassin, Elbin, Kontos, & Larson, 2010). However, athletes that experienced two or more concussions in their lifetime scored significantly lower on the verbal memory test compared to athletes with no concussions. In addition, athletes with three or more concussions in their lifetime scored lower on visual memory scores compared to athletes with no concussions. These studies show that memory and information processing speed are the most common neurocognitive impairments from multiple concussions and head impacts.

There have been mixed findings on the effects of multiple concussions and head impacts on neurocognitive function; however, there is a consensus of its effects on clinical depression. A study on retired NFL players showed 11% of the 2,500 surveyed players were diagnosed with clinical depression (Guskiewicz, Marshall, et al., 2007). They showed a significant increasing linear relationship between the number of concussions that a player experienced in their lifetime and diagnosis of depression after controlling for a number of factors including age, playing time, and age related health disorders. Similarly, another study showed that retired football players with a history of concussions had significantly more cognitive symptoms of depression, measured by the Beck Depression Inventory II, than did an age matched and IQ matched control group of retired football players with no concussion history (Didehbani, Cullum, Mansinghani, Conover, & Hart, 2013). Both of these studies suggest that there is a correlation between the number of concussions in a lifetime and the diagnosis of clinical depression.

There have been a number of suicides of current and retired former athletes. Most of these suicides have been linked to a disease called chronic traumatic encephalopathy (CTE). CTE is the degeneration of brain matter resulting from repeated head impacts that can occur in contact sports (Stern et al., 2011). The signature of the disease is the accumulation of tau-positive neurofibrillary tangles (NFT). Tau proteins stabilize microstructures in the nervous system and brain. Tau-positive NFTs are formed when tau proteins become hyperphosphorylated after repeated trauma to the head. Features of the disease include the decline in cognitive function and memory, suicidal behavior, poor impulse control, aggressiveness, and dementia. The term CTE first emerged following the autopsy of 2 former NFL athletes. The autopsies showed no atrophy of the brain; however, tau-positive NFTs were identified when the brain was investigated microscopically (Omalu et al., 2005, 2006). Tau-positive NFTs are the primary marker for Alzheimer's disease (Selkoe, 2001); however, neither football player was diagnosed with Alzheimer's disease. Evidence of CTE was also found in the brain of a deceased 23-year-old boxer (Geddes, Vowles, Nicoll, & Revesz, 1999; Geddes, Vowles, Robinson, & Sutcliffe, 1996). Taken together, these studies suggest that the accumulation of tau-positive NFTs is a consistent feature of repeated brain injuries and repeated head impacts in athletes. CTE has now been reported in former football, soccer, and hockey players as well as boxers and military professionals; unfortunately, the disease has only been diagnosed post mortem and no method has been developed to diagnose it in living individuals (McKee et al., 2009). Regardless, the identification of a disease that is caused by repetitive impacts to the head and multiple concussions has gained the attention of doctors, researchers, parents, and sports league officials globally (Fainaru-Wada & Fainaru, 2013).

2.4 Concussion in Sports

It is difficult to estimate the incidence of concussion in sport because many concussions are not reported (McCrea, Hammeke, Olsen, Leo, & Guskiewicz, 2004). This is partly due to the lack of a clear definition for concussion, as well as relying on athletes to self-report symptoms (Daneshvar, Nowinski, McKee, & Cantu, 2011). A frequently cited study estimated that between 1.6 and 3.8 million TBI's involving LOC and concussion occur each year in sports and recreation in the United States (Langlois et al., 2006). In 1988 the National Collegiate Athletic Association (NCAA) started an injury surveillance system and collected all injury data for the next 16 years (from 1988-2004) on 15 NCAA sports (Hootman et al., 2007). This study reported over 9,150 concussions for a 16-year span and concussion rates increased significantly at an annual rate of 7%. Men's hockey, women's hockey and women's soccer had the highest prevalence (4.1, 9.1 and 4.1 concussion injuries per 10,000 athletic exposures respectively). The majority of studies define an athletic exposure (AE) as one athlete participating in one practice or one game. These data would predict approximately 4 concussions for every 10,000 AE. A similar study followed 12 sports from 25 American high schools for 11 years (1997-2008) (Lincoln et al., 2011). They reported 2,651concussions from almost 10 million AE in the 11 years. The overall prevalence of concussion increased from 1.2 concussions per 10,000 AE in 1997 to 4.9 concussions per 10,000 AE in 2008. This increase is likely due to an increase in the awareness of the symptoms and consequences of brain injury (Hootman et al., 2007). Football is associated with the greatest number of TBIs and mTBIs of all sports played in America (Daneshvar et al., 2011), and as such has received increased media attention on concussion and scrutiny on the safety of the game (Fainaru-Wada & Fainaru, 2013).

2.5 Concussion in Football

The study that monitored injuries to athletes in the NCAA found 5,016 of the 9,150 diagnosed concussions were from football players with an injury rate of 3.7 concussions per 10,000 AE (Hootman et al., 2007). Other studies investigating concussion injury rates in football at the collegiate level have found rates between 6.0 concussions per 10,000 AE (Gessel et al., 2007), and 6.3 concussions per 10,000 AE (Rivara & Graham, 2014).

Rates of concussion are similar between high school and collegiate football. A study investigating concussion injury rates of high school athletes found that half of the diagnosed concussions (1,407 of 2,651) were from football players, and football had an injury rate of 6.0 concussions per 10,000 AE (Lincoln et al., 2011). Other studies of high school football showed a similar rate of concussion injury with 4.7 concussions per 10,000 AE and of 6.4 concussions per 10,000 AE (Gessel et al., 2007; Marar, McIlvain, Fields, & Comstock, 2012). The highest reported rate of concussion in high school football was 11.2 concussions per 10,000 AE (Rivara & Graham, 2014). A study on catastrophic injuries in football estimated that there were 1.5 million high school football participants and 75,000 football participants in the NCAA (Mueller & Colgate, 2013). This study also reported that there were 225,000 football participants that were not associated with schools. Combining these numbers indicates that approximately 1.8 million football players participated in the 2009 season. This figure does not include the nearly 3 million players that participate in youth football (ages 8-12), which was recently shown to have an injury rate of 17.6 concussions per 10,000 AE (Kontos et al., 2013). In light of the high rate of injury and large number of participants in football, and the likelihood the true injury rates have not been realized, the Centers for Disease and Control Prevention has referred to sports related concussion as an epidemic (CDC, 2013).

2.6 Head Impact Biomechanics in Football

The quantity of research performed on football and concussion has increased in the last decade as a result of the high concussion injury rates and the long term detrimental effects associated with multiple concussions and multiple head impacts. One area of research has focused on quantifying the linear and rotational accelerations (head impact kinematics) of football players' heads as they compete during games and practices. This area is known as Head Impact Biomechanics. One study quantified the peak linear accelerations of football players' heads from impacts by attaching linear accelerometers to a headband that the players wore while competing in games. The peak linear accelerations measured during games were in excess of 1,000 g, and were larger than the injury tolerance estimates at the time (Moon et al., 1971). A similar study quantified head impacts to football players by attaching linear accelerometers to the suspension inside of one player's football helmet (Reid et al., 1974). Over 650 impacts were recorded in 30 games, with the peak linear accelerations ranging from 40 to 530 g. Their study measured a concussive impact and suggested that the injury tolerance of the brain was in the range of 180-400 g (Reid et al., 1974). Both studies were the first to relate the peak linear acceleration of the head from impacts in football to brain injury tolerance levels, but could not define an injury threshold because of the variability of accelerations that they measured.

A more recent study instrumented the football helmets of an offensive lineman and a defensive lineman with a triaxial linear accelerometer and recorded the impacts to the players during 3.9 hours of practice. The average peak linear acceleration was 29.1 g and ranged from 10 g to 120 g; however, it was difficult to determine an injury threshold because no injuries occurred from 3.9 hours of practice time (Naunheim et al., 2000). The early works of Moon (1971), Reid (1974), and Naunheim (2000) reported a wide range of linear head accelerations players receive while playing the game. Their studies were important in proving the concept that head impact kinematics can be measured while football players compete.

Rather than studying typical players, some researchers have focused on players that experience concussions. One study used video data to determine the impact velocity and direction of impact for NFL players that received a concussion while playing in football games(Pellman et al., 2003). The results from the investigation were used to recreate the impacts in a laboratory setting using HIII dummies. The HIII dummies measured the peak linear acceleration and peak rotational accelerations that occurred at the centre of mass of the head using an array of 9 linear accelerometers mounted within the headform (NAP; DiMasi, 1995). This study showed that the average peak linear acceleration associated with concussion was 98 ± 28 g with the lowest linear acceleration of 52 g producing concussion (Pellman et al., 2003). The average peak rotational acceleration associated with concussion was 6,432 ± 1,813 rad/s^2. Data from this study helped to propose injury thresholds for concussion from peak linear and rotational accelerations with data collected from a real world scenario. However, the study was limited to 31 cases of concussion at the highest level of football and therefore it is difficult to generalize the results to other levels of football.

2.7 Determining Head Impact Biomechanics in Football using the Head Impact Telemetry System

The football field provides an opportunity to measure in-vivo head impact kinematics of players as they compete in games and practices. However, a research tool was needed that would determine the peak linear and rotational accelerations occurring at the centre of mass of the head when players received impacts. This led Simbex LLC (Lebanon, New Hampshire, USA) to create the Head Impact Telemetry System – a dedicated instrumentation system for measuring head accelerations within helmets during games and practices.

A HIT system unit is comprised of 6 single axis spring-loaded linear accelerometers, a wireless telemetry unit, a battery and an onboard storage unit mounted in a U shaped encoder. The encoder can be inserted into the crown of Riddell football helmets (Riddell, Elyria, Ohio, USA). The spring-loading ensures that the accelerometers are in constant contact with a player's head so that they are measuring head accelerations rather than helmet accelerations. Data are collected for 40 ms at 1,000 Hz when any of the accelerometers detects an acceleration that exceeds the user programmable threshold, typically 10 g. To ensure that the entire waveform of the impact is captured, 12 ms of data is stored pre-trigger and 28 ms of data is collected post trigger. Data are time stamped and wirelessly transmitted to a sideline receiver and laptop. Impact data from the 6 accelerometers are processed with a proprietary algorithm that calculates the resultant linear acceleration and impact location (Crisco & Chu, 2004). The algorithm also determines the peak resultant linear acceleration, peak resultant rotational acceleration at the centre of mass of the head (Chu et al., 2006) and calculates two severity indexes of the impact: the Gadd Severity Index (GSI; Gadd, 1966) and Head Impact Criterion (HIC; Versace, 1971). The GSI uses an exponentially weighted integral of the acceleration-time impulse to determine head injury risk (Gadd, 1966). The HIC also uses an exponentially weighted integral of the acceleration-time impulse; however, it identifies the maximum exponentially weighted integral within a 15 ms window (HIC_{15}) or a 36 ms window (HIC_{36}; Versace, 1971)

The HIT system was validated using a series of impact tests delivered by a linear pneumatic impactor to an instrumented HIII dummy head fitted with a helmet equipped with a HIT system unit. The results of the validation testing showed the peak linear acceleration from the HIT system were strongly correlated (r^2=0.90) to the peak linear accelerations measured by the NAP of the HIII, and the slope was 1.09. However, the peak rotational acceleration from the HIT system was only moderately correlated (r^2=0.53, slope=0.94) to the peak rotational accelerations measured by the NAP of the HIII (Beckwith, Greenwald, & Chu, 2012). Regardless of the HIT system's limited ability to determine the peak rotational acceleration of the head from impacts, the system has been a commonly used research tool in quantifying the head impact biomechanics of football players as they compete throughout games and practices.

The first to use the HIT system to quantify the head impact biomechanics of collegiate football players was reported in 2005 (Duma et al., 2005). They outfitted 8 players at a time with a HIT system encoder for the 2003 football season. Players wore the encoder for 2 games and 6 practices before a new set of players were selected to wear the system. A total of 3,312 impacts were recorded; 1,198 were experienced in 10 games and 2,114 were experienced in 35 practices. The mean peak linear acceleration was 32 ± 25 g and the majority of the impacts (89%) were below 60 g's. Five concussions were diagnosed during the season and one of the concussions occurring while a player was wearing an encoder. The peak linear acceleration of the head from the concussive impact was 81 g and the impact occurred to the right side of the head. This was a landmark study as it not only quantified head impact data for a season for a variety of players but it was also the first study to capture head impact data for a diagnosed concussion during a game.

Since 2005 a number of other US colleges have used the HIT system to monitor head injury risks of football players. Data from over 1.2 million impacts and 64 concussions were recorded from the year 2005- 2010 (Rowson et al., 2014). Additional studies have quantified and compared head impact exposure of different player positions, and compared games and practices (Brolinson & Manoogian, 2006; Crisco et al., 2010, 2011; Crisco & Wilcox, 2012; Mihalik, Bell, Marshall, & Guskiewicz, 2007), and level of play (high school versus college; Schnebel, Gwin, Anderson, & Gatlin, 2007), as well as investigating the relationship between closing distance to make an impact and head impact magnitude (Ocwieja et al., 2012). Other studies have focused on linking head impact biomechanics to clinical outcome measurements of concussion (Guskiewicz & Marshall, 2007; Gysland et al., 2012; Harpham, Mihalik, Littleton, Frank, & Guskiewicz, 2013; McAllister et al., 2012; McCaffrey, Mihalik, & Crowell, 2007). The data collected with the HIT system have lead to the development of concussion risk curves and detection of concussion through head impact data (Beckwith et al., 2013; Funk, Rowson, Daniel, & Duma, 2012; Greenwald, Gwin, Chu, & Crisco, 2008). In addition, HIT data have also been used to relate concussion to helmet temperature (Rowson & Duma, 2012) and to inform new standards for football equipment testing (Gwin et al., 2010).

Recovery from concussion is related to player age, with high school athletes having prolonged memory dysfunction compared to collegiate athletes (Field, Collins, Lovell, & Maroon, 2003). As such, studies using the HIT system quantified head impact exposure (Broglio et al., 2009; Broglio, Eckner, Martini, et al., 2011; Schnebel et al., 2007), concussion thresholds (Beckwith et al., 2013; Broglio, Eckner, Surma, & Kutcher, 2011; Eckner, Sabin, Kutcher, & Broglio, 2011; Greenwald et al., 2008), and investigated the relationship between head impact biomechanics and the neurocognitive function (Breedlove et al., 2012; Broglio, Eckner, Surma, et al., 2011; Talavage et al., 2014) for football players at the high school level. Implications for safer practices by reducing the number of contact practices, as well as investigating the role of different offensive schemes, run or pass scheme, on head impact biomechanics were investigated at the high school level (Broglio, Martini, Kasper, Eckner, & Kutcher, 2013; Martini et al., 2013).

The biomechanics of concussive injuries also differs between adults and children patients due to the differences in relative head to body size, brain water content, as well as differences in shapes of the skull (Meehan III, Taylor, & Proctor, 2011). Given the differences in concussion biomechanics between adults and children, a few studies determined the head impact exposure in youth football (Cobb et al., 2013; Daniel, Rowson, & Duma, 2012; Young, Daniel, Rowson, & Duma, 2013). There has only been a limited amount of work characterizing head impacts in youth football even though nearly 3 million youth players compete on a yearly basis. Youth leagues offer an opportunity to monitor the effects of rule changes, equipment design changes and use of safer techniques in the game and how they relate to head impact biomechanics (Rivara & Graham, 2014).

2.8 Effect of Player Position and Session Type in Football on Head Impact Biomechanics

Mihalik et al. (2007) used the HIT system to quantify the impact biomechanics of different collegiate football positions. Seventy-two players were enrolled in the study and head impact data were collected for the 2005 and 2006 football seasons at the University of North Carolina. Offensive linemen had significantly larger average peak linear accelerations (22.89 ± 1.79 g) than defensive linemen (21.56 ± 1.56 g) and defensive backs (21.02 ± 1.78 g). Offensive backs (22.93 ± 1.83 g) and linebackers (22.67 ± 1.81 g) also had significantly larger average peak linear accelerations than defensive lineman and wide receivers (22.19 ± 1.83 g). The study also determined the difference between sessions of football (practice or game). The average peak linear accelerations were significantly larger during practices (22.65 ± 1.80 g) than in games (21.12 ± 1.73 g). Although these differences between player positions and practices versus games were statistically significant, it is important to appreciate that the magnitude of the differences were quite small. Additionally, offensive backs were more likely to sustain an impact of greater than 80 g than defensive linemen, defensive backs, offensive linemen, linebackers, and wide receivers (1.52, 1.41, 1.24, 1.17, and 1.03 times greater, respectively). This study was the first to examine differences between positions and session types of the game of football and helped to inform positions that would be at a greater risk for experiencing higher magnitudes of head impacts.

A recent series of papers expanded on the differences in head impact biomechanics between player positions in football (Crisco et al., 2010, 2011; Crisco & Wilcox, 2012). The first study recorded the frequency (number of impacts), and location of impacts to the heads, of different player positions from 3 different collegiate institutions (Dartmouth, Brown, and Virginia Tech; Crisco et al., 2010). There were almost 3 times as many impacts recorded per game than there were per practice across all the schools. The maximum number of impacts per game ranged from 58 to 86. In contrast the maximum number of impacts per practice ranged from 15 to 24. It is important that quarterbacks were hit the least in practice (2.3 impacts per practice) but presumably they were hit more in games, as wide receivers were hit the least per game (7.3 impacts per game). Defensive linemen received the most impacts per practice and per game (11.5 impacts per practice and 29.8 impacts per game respectively). In general, defensive linemen, linebackers and offensive linemen had a greater number of impacts in both practices and games compared to defensive backs, running backs, quarterbacks, and wide receivers. The highest percentage (about 40%) of hits occurred to the front location of the head for almost all positions. Quarterbacks were the only position that received the highest percentage (about 40%) of hits to the back location of the head.

More recent studies have investigated differences in impact magnitudes (peak linear acceleration and peak rotational acceleration) between player positions in football during games and practices (Crisco et al., 2011; Crisco & Wilcox, 2012). These studies calculated the 95th percentile peak linear and rotational acceleration thresholds to identify which position was experiencing the highest magnitudes of head impacts. Since concussion injury risk increases with higher magnitudes of impacts, this parameter has been used to relate the risk for head injury to player position (Rowson et al., 2012). The 95th peak percentile linear and rotational accelerations were greater for the running backs (approximately 68 g and 4,869 rad/s^2) compared to the offensive linemen (approximately 56 g and 3,799 rad/s^2), defensive linemen (approximately 57 g and 3,891 rad/s^2) and defensive backs (approximately 59 g and 4,269 rad/s^2). Offensive linemen and defensive linemen received the most impacts for the season (543 and 718 respectively); however, the magnitudes of the impacts were the least compared to the other positions (Crisco et al., 2011).

3 Purpose Statements and Hypothesis

3.1 Purpose Statements

The first purpose of this study was to quantify the number, magnitudes, and location of impacts experienced by the heads of Canadian University football players as they compete throughout a season. The following impact parameters were determined for the players:

I. Peak Linear Acceleration

II. HIC_{15}

III. Peak Rotational Velocity

IV. Peak Rotational Acceleration

V. Number of extreme impacts (impacts greater than the 95th percentile of the data set) per 1,000 impacts

The second purpose of this study was to compare the magnitude and location of impacts experienced in games to impacts experienced in practice and to compare the magnitude and location of impacts experienced by the different player positions.

3.2 Hypotheses

1) Impacts experienced during games will have larger magnitudes compared to the impacts experienced during practices, but there will be no difference in the location of impacts to the helmet in games compared to impact locations in practice.

2) Impacts experienced by the linemen positions[1] (offensive linemen and defensive linemen) will be to the front of the helmet more often and will have smaller magnitudes

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[1] These are positions that are arranged along the line of scrimmage at the snap of the ball. The majority of the positional responsibilities for this group occurs along the line of scrimmage but is not limited to the line of scrimmage.

than impacts experienced by skilled positions[2] (offensive backs, wide receivers, linebackers and defensive backs).

[2] These positions are not required to be set on the line of scrimmage at the snap of the ball. Positional responsibilities can occur along the line of scrimmage but the majority of their responsibilities occur away from the line of scrimmage (“in space”).

4 Methods

4.1 Participants

Fifty-six varsity football players were recruited from the 2013 University of Western Ontario Mustangs varsity football team to participate in the study. Of the 100 players that can be part of a roster on a Canadian university football team, only 47 of the players, the dress roster, compete in the games. Therefore, participants were selected based on input from coaching staff to reflect the different positions as well as to reflect the players that would be routinely playing in the games. The different positions included: 3 quarterbacks, 5 running backs, 3 fullbacks, 9 receivers, 8 offensive linemen, 9 defensive linemen, 8 linebackers, and 11 defensive backs. There was a narrow range for the participants' age but participants' weights and heights varied between player positions (Table 4.1). The study was approved by the University of Western Ontario's Human Subjects Research Ethics Board and all players provided informed consent (Appendix A). Inclusion criteria included members of the varsity football team during the Fall 2013 Canadian Interuniversity Sport (CIS) football season with helmets that had enough space for adhering the GForce Tracker (GFT) device to the inside of their football helmet shell. Originally 49 football players were recruited to participate in the study, but three participants were dismissed from the team just after the training camp concluded; their devices were assigned to the new players that took their spot on the dress roster. These replacement players were selected in consultation with the head football coach. Players were also withdrawn from the study if they received a season ending injury (three players). Similarly to the players that were dismissed, new participants were chosen to either reflect the injured person's position, or were referred by the head football coach. Furthermore, one of the original participant's device was reassigned to another player, as he was not on the dress roster consistently; this device was given to a participant who was seeing significantly more time in games. All together, seven additional participants participated in the study. Of the fifty-six participants, two had positional changes in response to a coaching decision to utilize the player's talent more effectively or to provide depth to a position that had suffered a lot of players lost to injury. One player was dropped from the study due to a malfunctioning device.

Table 4.1: The number of players per position and the average (and standard deviation) weights, heights and ages for each position.

Position	Number of Players	Weight (kg)		Height (cm)		Age (yrs.)	
		Mean	SD	Mean	SD	Mean	SD
Defensive Back	11	87.96	2.42	182.53	4.79	20.45	0.92
Defensive Line	9	120.61	17.63	189.37	4.69	20.67	0.94
Linebacker	8	99.45	5.44	182.88	2.11	21.88	0.93
Offensive Line	8	134.15	12.59	193.04	4.60	20.63	1.32
Quarterback	3	86.79	7.65	183.73	4.79	20.00	0.82
Fullback	3	106.14	8.02	181.19	2.16	21.00	1.41
Running Back	5	88.11	3.20	179.71	1.91	20.25	0.43
Wide Receiver	9	88.85	4.15	184.57	4.40	20.44	1.17
All Positions	**56**	**102.51**	**18.74**	**185.25**	**5.80**	**20.70**	**1.15**

4.2 Instrumentation

The GForce Tracker (GFT2, Artaflex Inc., Markham, Ontario, Canada) is a new wireless device that is adhered to the inside of sports player's helmets and measures head impact biomechanics. Much like the HIT system, the GFTs collected the linear acceleration and rotational velocity of the player's helmets when they received impacts. The device was adhered to the inside of the helmet, on the left side of the crown air bladder, using an industrial strength recloseable fastener (3M™ Dual Lock™ Recloseable Fastener SJ3551 400 Black, 3M Global Headquarters, St. Paul, MN, Figure 4.1).

Figure 4.1: Illustration showing a GFT device, circled in red, mounted on the left side of the crown bladder inside a football helmet

The GFT measures 55 mm long, 29 mm wide, and 14 mm high, and weighs 2 grams (Figure 4.2). It contains a tri-axial linear accelerometer, a tri axial gyroscope, as well as a lithium ion rechargeable battery, and on-board memory for storing up to 400 impacts. The triaxial accelerometer measures accelerations in the x, y, and z directions, have a range of $\pm$ 200 g and a 1g resolution for each axis. The triaxial gyroscope measures rotational velocity in degrees per second (˚/s) in the x, y and z directions, and has a range of ± 2,000˚/s. Data are collected when any linear accelerometer axis detects an acceleration greater than a user selected threshold. For the purposes of this study the data collection trigger threshold was set to 15 g. This reflected similar record thresholds from previous studies using the HIT system (Crisco et al., 2010, 2011; Crisco & Wilcox, 2012). To ensure that the entire impact is collected, the device records 8 ms of the impact preceding the threshold, and 32 ms of impact data following the threshold. Linear acceleration signals are passed through an onboard analog low pass filter with a cut-off frequency of 300 Hz and data are sampled at 3,000 Hz. Rotational velocity signals are recorded at 800 Hz and passed through a analog low pass filter with a cut-off frequency of 100 Hz. Data are time stamped and recorded to the onboard memory.

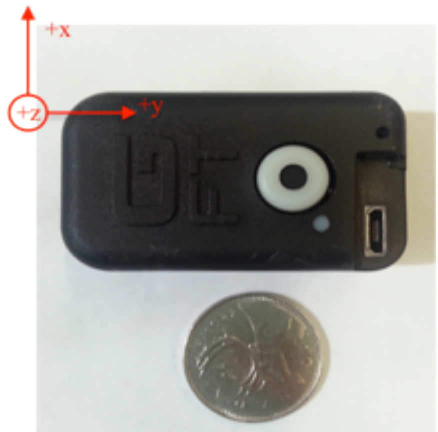
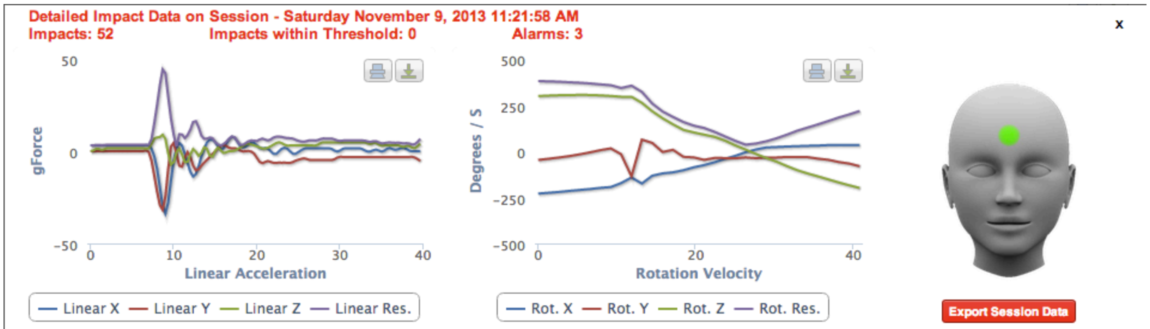


Figure 4.2: The GForce Tracker device illustrating its size relative to a quarter. The coordinate system conventions are shown in red, and the circled axis is directed into the page.

4.3 Software

Once a player had completed a session, practice or game, impact data from the session are uploaded to a cloud based-internet software by connecting the device to a laptop via a micro USB cord. The cloud software stored the data collected from the sessions and calculated the resultant linear acceleration and resultant rotational velocity for each impact. Peak linear acceleration and peak rotational velocity for the impacts are determined and severity metrics for the impacts are calculated in the form of GSI and HIC_{15}. Lastly, the software calculated the location of the impact in degrees of azimuth and elevation (Figure 4.3).



	Date	Impact Time	gForce	Location	HIC	GSI	Linear X	Linear Y	Linear Z	Rotation X	Rotation Y	Rotation Z	Rotation Resultant	Azimuth	Elevation
	11/09/2013	13:49:41.17	16.24	LEFT	1.50	2.70	15.62	3.12	-3.12	-288	-67	-116	318	293.20	-7.48
	11/09/2013	13:49:42.129	20.55	LEFT	3.10	4.90	9.38	-17.19	-6.25	111	95	-342	372	300.96	27.23
	11/09/2013	13:58:30.820	52.22	LEFT	45.80	68.80	-28.12	-43.75	-4.69	459	151	568	746	261.87	40.32
	11/09/2013	14:08:29.562	45.23	FRONT	5.80	19.40	-29.69	-32.81	9.38	-226	-41	301	379	12.53	-28.47
	11/09/2013	14:08:31.409	16.90	RIGHT	1.40	3.50	-15.62	-6.25	-1.56	241	166	-46	296	51.34	-8.88
	11/09/2013	14:12:27.7	30.06	TOP	20.30	33.80	-12.50	14.06	23.44	-67	770	-497	919	270.00	81.87
	11/09/2013	14:13:14.292	62.21	LEFT	7.00	13.20	26.56	0.00	-56.25	-322	-213	235	452	270.00	-15.95

Figure 4.3: Screen shot of the GForce Tracker cloud software and data after collecting for a session.

Azimuth is defined as the angle in the x-y plane between 0˚ reference coordinate (anterior) and the impact vector (Figure 4.4a); it is measured between 0˚ and 360˚. Elevation is defined as the angle between the x-y plane and the impact vector (Figure 4.4b). Impact vectors that are oriented above the x and y plane are defined from 0˚ to +90˚ of elevation, and impact vectors that are oriented below the x and y plane are defined from 0˚ to -90˚. Azimuth and elevation coordinates were used to categorize impacts into one of six locations: front, back, right, left, top and bottom. Azimuth angles occurring between 315˚ to 45˚ were categorized as front impacts, while azimuth angles occurring between 135˚ to 225˚ were categorized as back impacts. Azimuth angles between 45˚ to 135˚ and between 205˚ to 315˚ were categorized as right and left impacts. Lastly, impacts were categorized as top if the elevation angles were greater than 45˚ and classified as bottom if their elevation angles were less than 45˚; this overrode the category that was defined based on the azimuth of the impact vector.

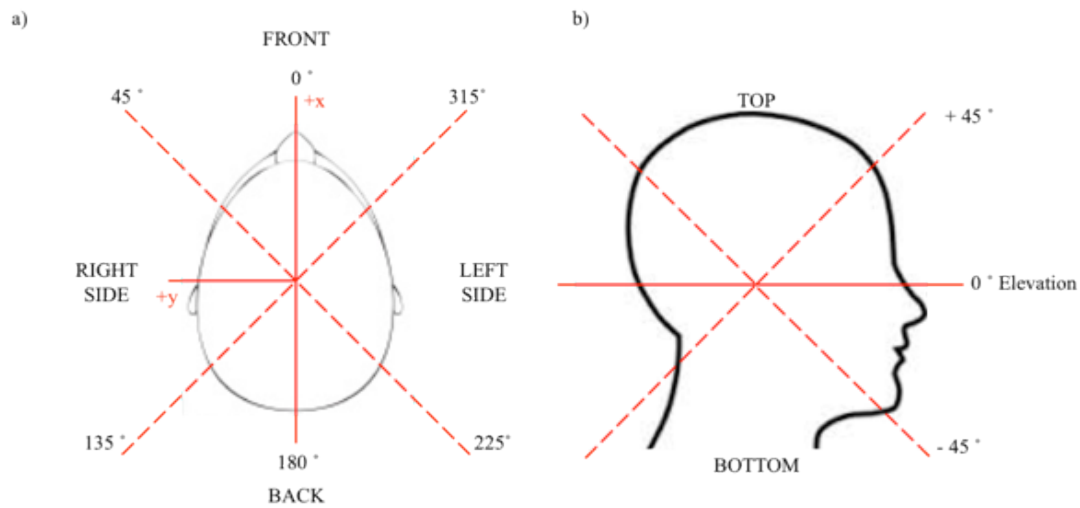


Figure 4.4: The GFT coordinate system for defining a) Azimuth angles and b) Elevation angles

4.4 Data Collection Protocol

4.4.1 Practices

The research team, before practice, was responsible for turning on all of the devices inside the player's helmets. During practice, the research team recorded the timing of session events throughout the practice. Timing of session events were used later to categorize when impacts occurred to the players and used to delete any impacts the device recorded while the player was not wearing a helmet and not in practice. Players returned to the locker room upon completing practice and placed their helmets on charging racks. The devices remained inside of the helmet while connected to a micro USB cord and wall mount adapter for charging before the next practice or game. After practice, the data from each device for the session were uploaded to the internet cloud based software for storage and a summary file for the session was exported to Microsoft Excel 2010 (Microsoft, Redmond, WA, USA). Each summary file contained the date and time of every impact from the session as well as the associated peak linear acceleration, peak rotational velocity, max HIC_{15}, GSI, and location of the hits (azimuth and elevation angles and syntax category) on one sheet. Additional sheets contained the linear acceleration time series data for all hits and rotational velocity time series data for all hits.

4.4.2 Games

The data collection protocol for games was similar to practices. The research team turned on the devices before any player went out onto the field. The timing of which players were on the field was recorded to ensure that only impacts occurring to players on the field were included in the analysis. After the game the helmets and devices were returned to the rack for charging and downloading the data to the cloud software. Summary files were exported to Microsoft Excel 2010 for further analyses.

4.5 Data Reduction and Calibration

The timing of impacts in the summary files was compared to the timing of session events from game and practice scenarios using various custom LabVIEW programs (Version 10, National Instruments Corp., Austin, TX, USA). Only impacts that were experienced while the players were competing in practices and games while wearing a helmet were included for further analysis.

The device was calibrated to predict the resultant linear acceleration and resultant rotational velocity at centre of mass of the head. This was important because the device is adhered to the inside of the helmet shell and does not use an algorithm to predict accelerations at the centre of mass of the head. The calibration involved filtering the x, y, z linear acceleration data with a CFC 180 filter (Cichos, Vogel, Otto, Schaar, & Zölsch, 2006), application of an impact location correction algorithm, and use of location dependent equations, biases and gains, to approximate the peak resultant linear acceleration and peak resultant rotational velocity at the centre of mass of the head. The resultant rotational acceleration was obtained by differentiating the resultant rotational velocity using a 5-point stencil method (Cichos et al., 2006). Like the peak linear acceleration and peak rotational velocity, the peak rotational acceleration at the centre of mass of the head from impacts and HIC_{15} was approximated using impact location dependent equations. Addition details are presented in Appendix D.

Equations for the algorithm were developed and validated from testing that used a pneumatic linear impactor to deliver repeatable impacts to a HIII head wearing a football helmet with a GFT mounted on the inside of the helmet shell (Appendix B). The calibrated peak linear acceleration, peak rotational velocity, peak rotational acceleration and max HIC_{15} of the impacts measured from the GFT were compared to the peak linear acceleration, peak rotational velocity, peak rotational acceleration, and max HIC_{15} measured at the centre of mass of the HIII head. Calibrated peak linear acceleration, peak rotational velocity, peak rotational acceleration and max HIC_{15} from the GFT using the impact calibration algorithm, were strongly correlated to the head impact kinematics measured at the centre of mass of the HIII (r^2= 0.96, 0.95, 0.87 and 0.97 respectively). Additional details are presented in Appendix C.

4.6 Statistical Analyses

Only players that experienced impacts in at least one practice and one game were included for statistical analysis. A total of 53 players met this inclusion criterion. Head impact data can be skewed because of the high number of low magnitude impacts that are measured during a season. Therefore, a Shapiro-Wilks test was used to determine the normality of the distribution of the impact parameters: peak linear acceleration, peak rotational velocity, peak rotational acceleration, max HIC_{15} and number of extreme hits. If the impact parameters were not normally distributed ($p<0.05$) then non-parametric analyses were conducted to test the hypotheses.

In the event of failed tests of normality on the impact parameters, medians and interquartile ranges for 53 players were determined for the impact parameters. An extreme hit was defined as an impact with a peak linear acceleration greater than the 95th percentile peak linear acceleration (Crisco et al., 2011; Crisco & Wilcox, 2012). The 95th percentile peak linear acceleration was calculated from the peak linear acceleration data set that contained all impacts measured across all players and all sessions. The total number of impacts for the season and total number of impacts in games and practices were determined. The median and interquartile ranges for the impact parameters for all impacts, and the 95th percentile for peak linear and rotational acceleration were determined for describing impact exposures.

Wilcoxon Sum of Ranks tests (non-parametric equivalent to a dependent t-test) were performed to assess whether there were statistically significant effects of session type on the median peak linear acceleration, median HIC_{15}, median peak rotational velocity, median peak acceleration, and median number of extreme hits per 1,000 hits. Session type was the with-in subjects factor and had two levels: game or practice.

Kruskal-Wallis H tests (non-parametric equivalent to one-way ANOVA tests) were performed to determine whether there were statistically significant effects of player position on the impact parameters. Player position was the between subjects factor and comprised of 6 levels: defensive back (DB), linebacker (LB), defensive linemen (DL), offensive linemen (OL), offensive back (OB), and wide receiver (WR). The offensive back was made up from the fullback (FB), running back (RB), and quarterback (QB) positions, and was created as we expected that these player positions would experience similar impacts and we wanted the number of players within each positional group to be more equal so that the statistical analysis would be more robust. When the Kruskal-Wallis tests revealed a main significant effect of player position on the impact parameters, Mann-Whitney U tests were performed to determine where the significant differences were between player positions. The Bonferroni adjustment is the most conservative method to control the likelihood of making a type I error. It accounts for this error by dividing the *a priori* alpha value by the number of comparisons that are being made. However, using a Bonferroni adjustment comes at the cost of reducing statistical power. Therefore a modified Bonferroni adjustment was used because it is less conservative, and represents the most powerful method for controlling multiple comparison bias while maintaining power for detecting true relationships (Olejnik, Li, Supattathum, & Huberty, 1997). The comparison with the largest effect was assessed by an alpha level of α (0.05) divided by the number of comparisons (15). The next largest effect was assessed by an alpha level α (0.05) divided by the number of comparisons minus one (14). This continued until the comparison with the smallest effect was assessed by an alpha level of 0.05.

In the event of successful tests of normality on the impact parameters means and standard deviations for the impact parameters were determined. Two-way repeated measures analysis of variance tests (ANOVAs) were performed to assess whether there were statistically significant effects of session type and player position on the mean peak linear acceleration, mean HIC_{15}, mean peak rotational velocity, mean peak acceleration, and mean number of extreme hits per 1,000 hits. Session type was the with-in subjects factor and had two levels: game or practice. Player position was the between subjects factor and comprised of 6 levels: defensive back (DB), linebacker (LB), defensive linemen (DL), offensive linemen (OL), offensive back (OB), and wide receiver (WR). The offensive back was made up from the fullback (FB), running back (RB), and quarterback (QB) positions, and was created as we expected that these player positions would experience similar impacts and we wanted the number of players within each positional group to be more equal so that the statistical analysis would be more robust. A Tukey's Honestly Significant Difference (Tukey's HSD) test was performed when there was a main significant effect of player position on the impact parameters. This controlled the likelihood that we would make a type I error from multiple pairwise comparisons.

To examine the proportions of impacts to the front, back, left and right helmet locations, two Chi-square tests were carried out comparing games and practices, and player position. These tests were used to gather information on whether players experienced a similar number of impacts to the four helmet locations during games and practices, and across positions. A Bonferroni adjustment was then used on the results of the Chi-square analyses to control for the likelihood of making a type I error due to multiple comparisons. All statistical tests were performed using SPSS 16.0 (SPSS Inc., Chicago, IL) and the level of significance was set at an alpha of less than 0.05 *a priori*.

5 Results

All impact parameters failed the Shapiro-Wilks test of normality (p<0.05). Impact parameters were not only skewed but also kurtotic (Figure 5.1). Accordingly non-parametric statistical analyses were performed. The following sections describe the data using the median and interquartile ranges, and present the results of the non-parametric tests.

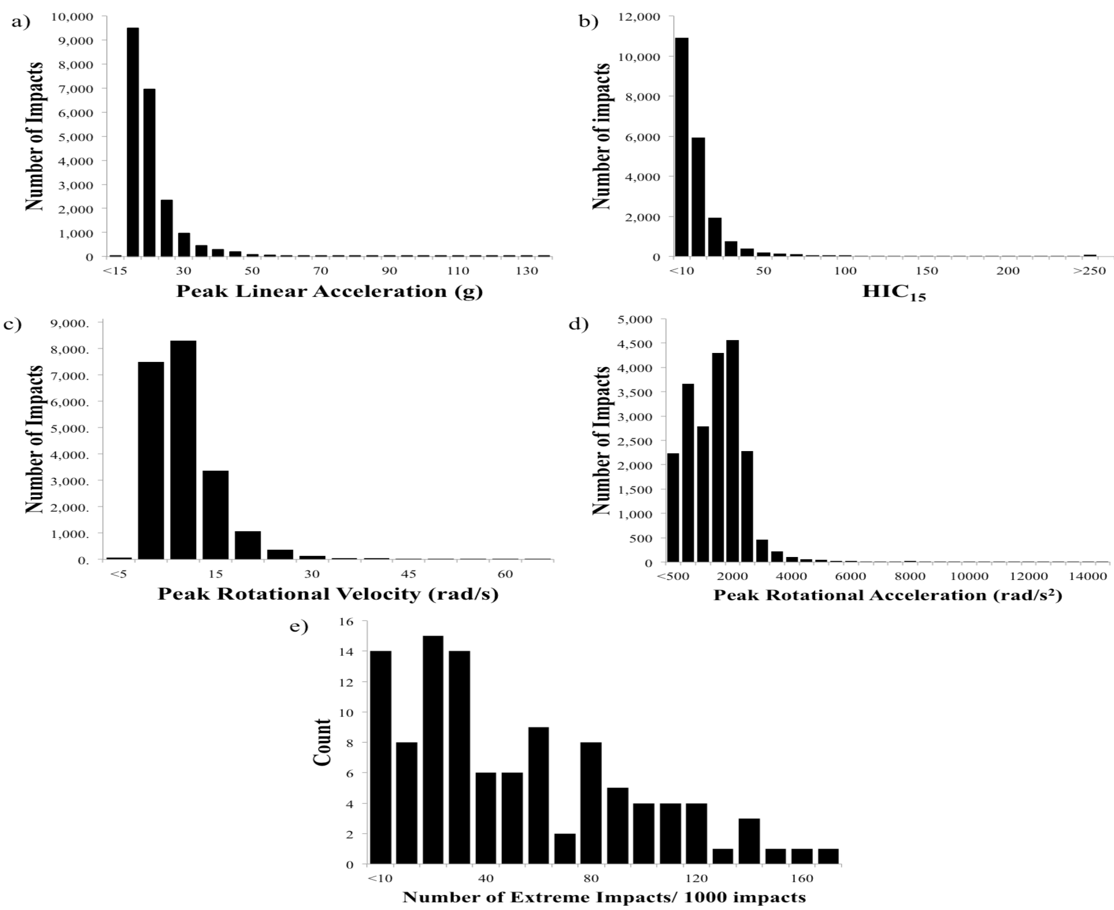


Figure 5.1: Distribution for the impact parameters a) Peak Linear Acceleration, b) HIC_{15}, c) Peak Rotational Velocity, d) Peak Rotational Acceleration, and e) Number of Extreme impacts. All impact parameters were skewed and kurtotic.

5.1 Number and Magnitudes of Impacts Experienced by The 2013 Football Team

Over the course of the 2013 CIS football season, a total of 20,924 impacts (10,528 impacts from practice and 10,396 impacts from games) were collected from 66 sessions (55 practices and 11 games). Across the study, medians for the peak linear acceleration, HIC_{15}, peak rotational velocity and peak rotational acceleration were 20.43 [interquartile range 18.39-24.07] g, 9.17 [3.99 -17.33], 11.36 [8.93-14.83] rad/s, and 1,724.44 [857.91-2,322.95] rad/s^2 respectively. The 95th percentile of the peak impact was 36.13 g and 3,029.62 rad/s^2 for the linear and rotational accelerations respectively. The largest percentage of impacts (45.8%) experienced by the players was to the front location of the helmet. There was a median of 58.08 [27.68-81.25] extreme impacts (greater than 36.13 g) per 1,000 impacts across the study.

5.2 Impact Parameters Magnitudes in Games and Practices

All impact parameter magnitudes were significantly larger in games than in practices. The median peak linear acceleration of impacts experienced during games (Figure 5.2; 21.53 [18.85-26.43] g) was 8% larger than the median peak linear acceleration of impacts measured during practice (19.94 [18.10-22.92] g, z= -5.90, p< 0.001).

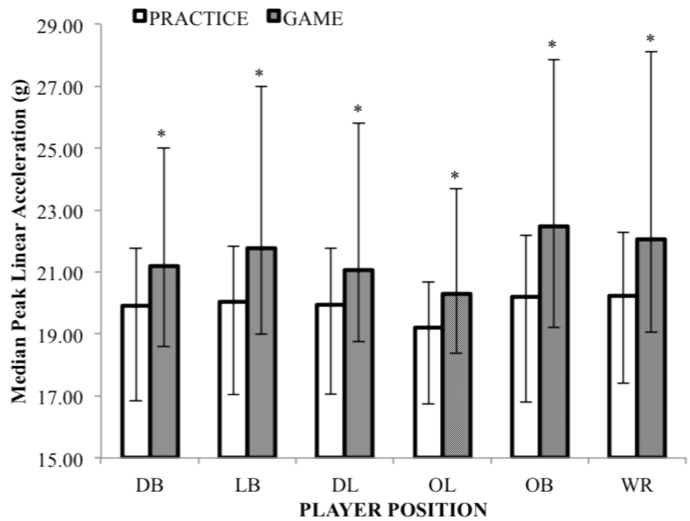


Figure 5.2: The median peak linear acceleration and interquartile ranges for practice sessions and game sessions. Games had a significantly larger median peak linear acceleration than practices (*p*<0.001). Player position groups are defined as defensive back (DB), linebacker (LB), defensive linemen (DL), offensive linemen (OL), offensive back (OB), and wide receiver (WR). The offensive back was made up of the fullback, running back, and quarterback positions.

***- Denotes a significant effect of session type; *p*<0.001**

The median peak rotational velocity of impacts experienced by players in games (Figure 5.3; 12.79 [9.68-26.43] rad/s) was significantly larger (24%) compared to practices (10.34 [8.23-13.27] rad/s; z= -5.37, p<0.001). Players also experienced significantly larger median peak rotational accelerations (20%) during games (Figure 5.4; 1846.41 [1,030.57-2,433.13] rad/s^2), than during practices (1,523.01 [759.35-2,182.99 rad/s^2; z= 4.44, p<0.001).

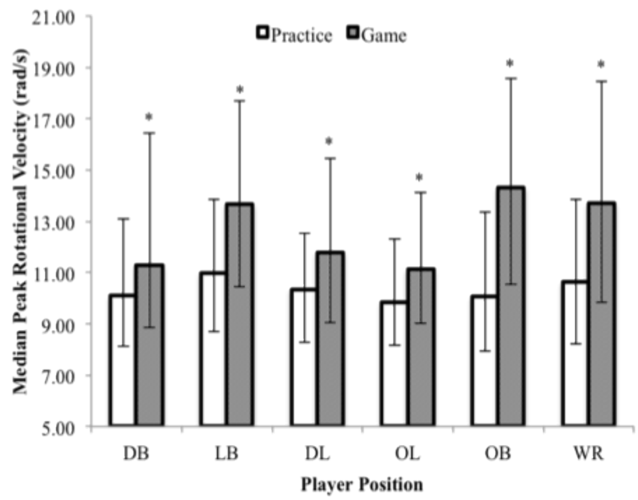


Figure 5.3: The median peak rotational velocity and interquartile ranges for practice sessions and game sessions. Games had a larger median peak rotational velocity than practices (effect of session type: *p*<0.001). Player position abbreviations are explained in the caption for Figure 5.2

***- Denotes a significant effect of session type.**

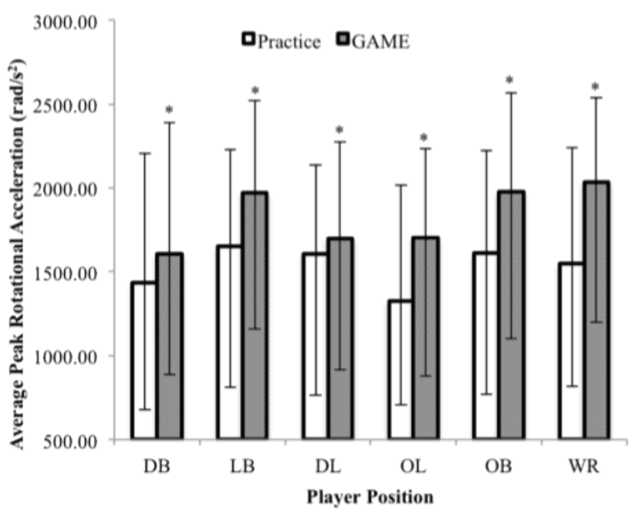


Figure 5.4: The median peak rotational acceleration and interquartile ranges for practice sessions and game sessions. Games had a larger median peak rotational acceleration than practices (effect of session type: *p*<0.001). Player position abbreviations are explained in the caption for Figure 5.2

***- Denotes a significant effect of session type.**

The impact severity parameter HIC_{15} had the second largest percent difference between games and practices of all of the impact parameters (Figure 5.5). Impacts experienced by players competing in games had a 48% larger median HIC_{15} (11.29 [4.81-21.14]) than practices (7.63 [3.62-15.38], z= -5.39, p< 0.001).

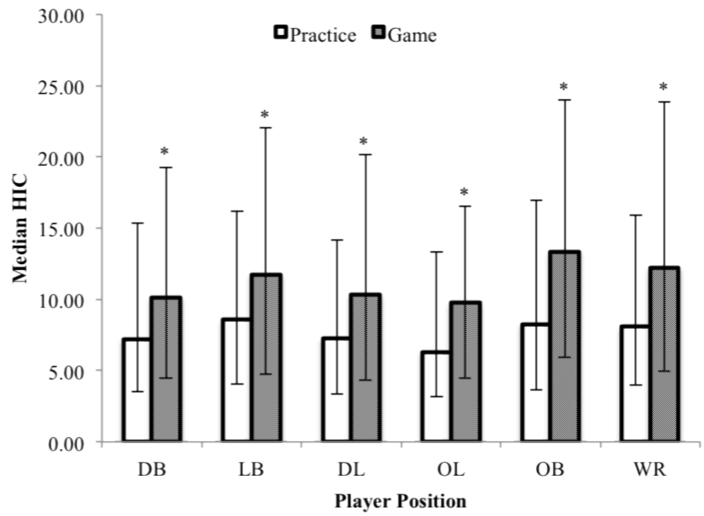


Figure 5.5: The median HIC_{15} and interquartile ranges for practice sessions and game sessions for the different player positions. Games had a larger median HIC_{15} than practices (effect of session type: *p*<0.001). Player position abbreviations are explained in the caption for Figure 5.2

***- Denotes a significant effect of session type; *p*<0.001**

Finally, players were 2.3 times more likely to experience an extreme impact during games (Figure 5.6; 78.17 [55.89-102.70] extreme impacts/ 1,000 impacts) than practices (34.15 [14.06- 46.50] extreme impacts/ 1,000 impacts). This difference between games and practices was the largest among the impact parameters (z= -5.46, p< 0.001).

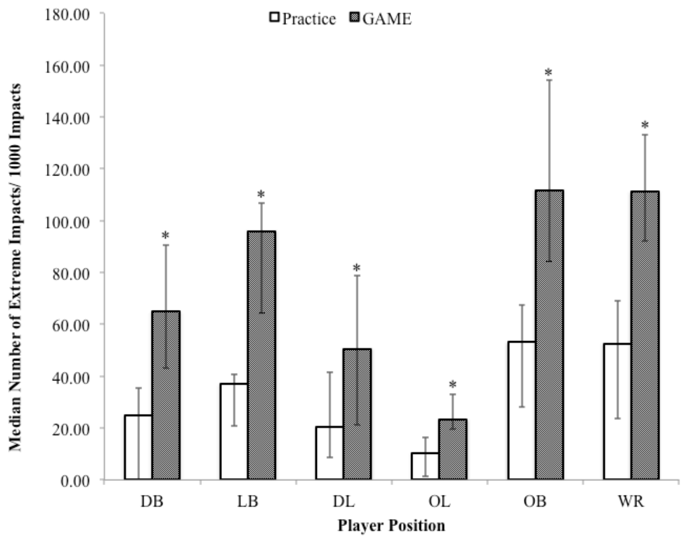


Figure 5.6: The median number of extreme impacts per 1,000 impacts for practices and games. Games had significantly more extreme impacts than practices (*p*<0.001). Player position abbreviations are explained in the caption for Figure 5.2

***- Denotes a significant effect of session type on the median number of extreme impacts per 1000 impacts; *p*<0.001**

The locations of impacts to the helmet experienced in games were significantly different than impacts experienced in practice (Figure 5.7; χ^2 (3)= 59.26, $p<0.001$). Significantly more impacts to the front helmet location were experienced during practice than in games (z= 4.8, $p<0.05$). Additionally, there were significantly more impacts to the left helmet location during games than during practices (z= 7.5, $p<0.05$). Impacts to the back and right locations did not significantly differ between games and practices (z= 0.9 $p>0.05$; z= 0.6 $p>0.05$).

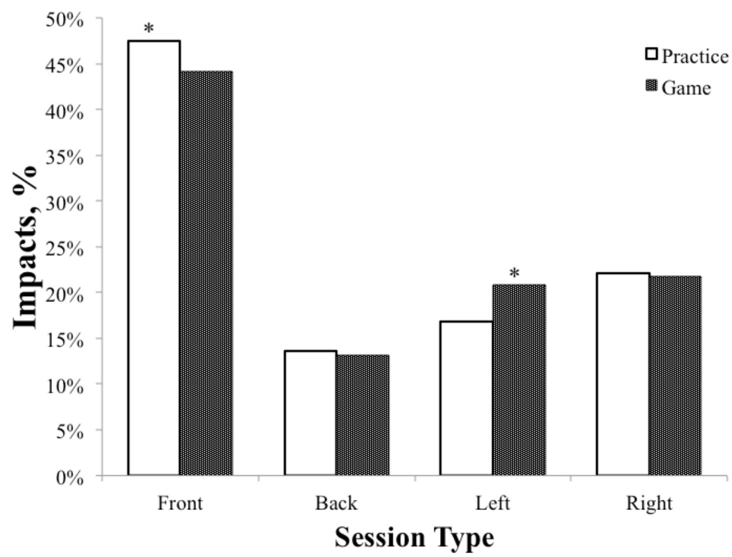


Figure 5.7: The percentage of impacts experienced in practices and games to the front, back, left, and right helmet locations. Practices had a significantly greater proportion of impacts to the front location, but games had a significantly greater proportion impacts to the left.

***- Denotes a significant larger proportion of impacts at the location between games and practice, $p<0.05$**

5.3 Magnitudes of Impact Parameters for the Different Player Positions

5.3.1 Peak Linear Acceleration

The median peak linear accelerations were significantly different between player positions (Figure 5.8; χ^2 (5)= 18.54, $p<0.05$). Post-hoc Mann-Whitney U analyses showed that the OB position had a significantly larger median peak linear acceleration (13%) compared to the OL position (z =-3.39, $p<0.05$). Post-hoc analysis also showed that the WR position had a significantly larger median peak linear acceleration (7%) compared to the OL position group (z = -3.08, $p<0.05$). The LB position showed a trend for a larger median peak linear acceleration than the OL position (4%), but it was not significantly different (z = -2.66, p>0.05). The DB, LB and DL positions were not significantly different from any other player positions (p>0.05).

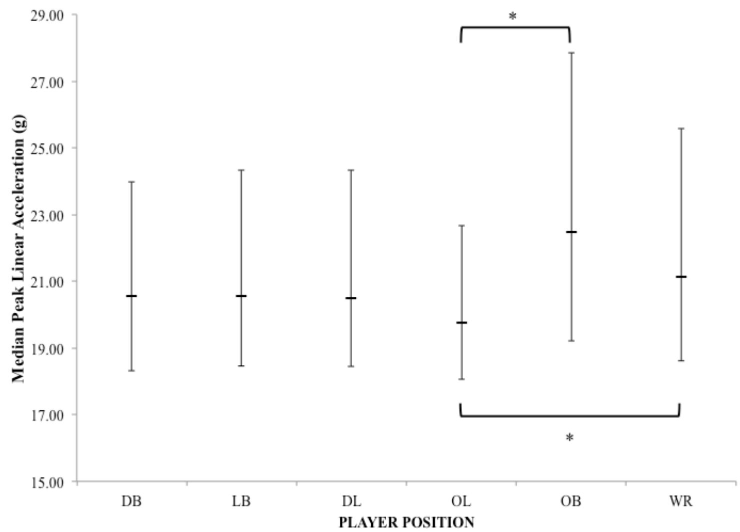


Figure 5.8: The median and interquartile ranges of the peak linear acceleration for different player positions. The median peak linear accelerations for the OB and WR positions were significantly larger than the OL. Player position abbreviations are explained in the caption for Figure 5.2.

*** Denotes a significant difference in the median peak linear acceleration between position groups; *p*<0.05.**

5.3.2 HIC_{15}

The impact magnitude parameter HIC_{15} was significantly different between player positions (Figure 5.9; χ^2 (5)= 13.26, p<0.05). Post-hoc Mann-Whitney U analyses showed that the OB position had a significantly larger median HIC_{15} (66%) compared to the OL group (z =-3.30, p<0.05). Although the OB position showed a trend of having a larger median HIC value (46%) compared to the LB position, it was not statistically significant (z = -2.49, p>0.05). The median HIC_{15} values were not significantly different between any other player positions (p>0.05).

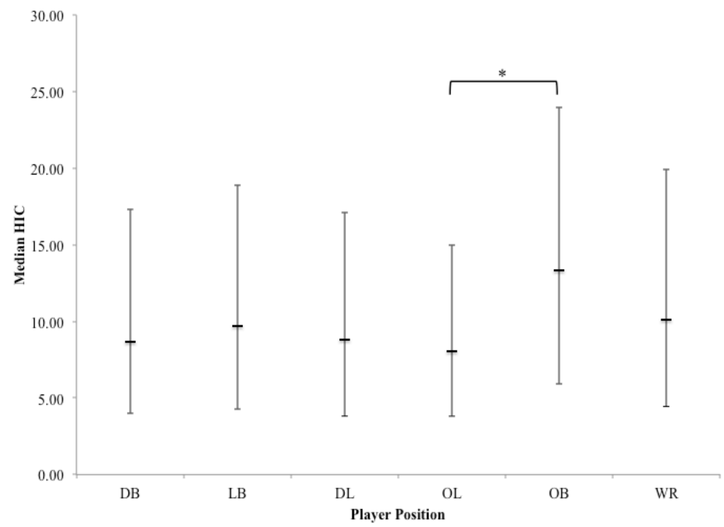


Figure 5.9: The median and interquartile ranges of the HIC_{15} values for different player positions. The OB player position had a significantly larger median HIC_{15} than the OL. Player position abbreviations are explained in the caption for Figure 5.2

***Denotes a significant difference in median HIC_{15} between player position groups; *p*<0.05.**

5.3.3 Peak Rotational Velocity

The median peak rotational velocities were significantly different between player positions (Figure 5.10; χ^2 (5)= 12.96, $p<0.05$). However, post-hoc analysis through the Mann-Whitney U test did not reveal which of the player positions were significantly different. Both the OB group (36%) and LB group (12%) showed a trend for having a larger median peak rotational velocity than the OL group, but neither group reached statistical significance ($z = -2.64$, $p>0.05$; $z = -2.66$, $p>0.05$).

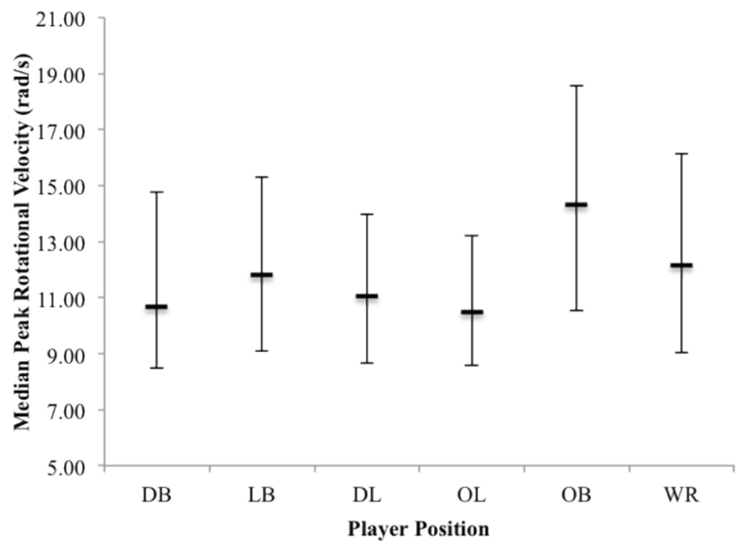


Figure 5.10: The median and interquartile ranges of the peak rotational velocity for different player positions. Player position had a significant effect on the median peak rotational velocity, however post-hoc analysis did not reveal which positions were significantly different. Player position abbreviations are explained in the caption for Figure 5.2.

5.3.4 Peak Rotational Acceleration

The median peak rotational accelerations were significantly different between player positions, (Figure 5.11; χ^2 (5)= 14.40, p<0.05). Post hoc analyses showed that both the OB (31%) and the WR (18%) positions had significantly larger median peak rotational accelerations than the OL position (z = -3.22, p<0.05, z = -2.98, p<0.05). The DB, LB, and DL were not significantly different from other positions (p>0.05).

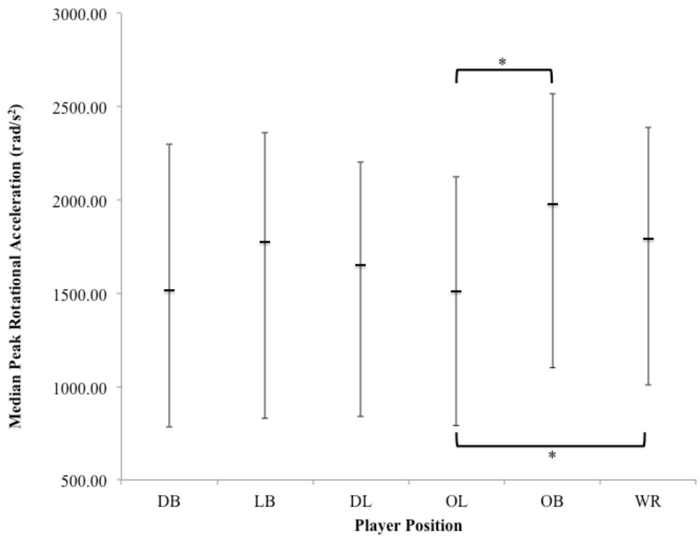


Figure 5.11: The median and interquartile ranges of the peak rotational acceleration for different player positions. The OB and WR position had significantly larger median peak rotational accelerations than the OL Position. Player position abbreviations are explained in the caption for Figure 5.2.

***Denotes a significant difference in median peak rotational acceleration between player position groups; *p*<0.05.**

5.3.5 Number of Extreme Impacts

The median number of extreme hits per 1,000 impacts was significantly different between player positions (Figure 5.12; χ^2 (5)= 26.46, $p<0.001$). The post-hoc analyses with the Mann-Whitney U tests showed that the median number of extreme impacts for the OB, WR and LB positions was not significantly different from each other (z = -.34, $p>0.05$; z = -2.12, $p>0.05$; z = -2.27, $p>0.05$). The OB, WR, and LB positions had significantly more extreme impacts than the OL group (z = -3.47, $p<0.05$; z = -3.46, $p<0.05$; z = -3.24, $p<0.05$). The LB position experienced 2.2 times more extreme impacts, the WR position experienced 3.98 times more extreme impacts and the OB position experienced 4.4 times more extreme impacts than the OL position. The number of extreme impacts for the DB and DL positions were not significantly different from other positions (*p*>0.05).

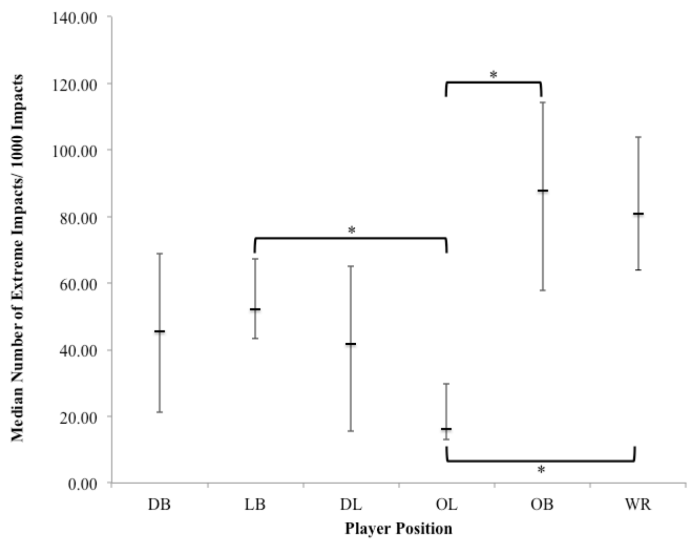


Figure 5.12: The median and interquartile ranges of the number of extreme impacts per 1,000 impacts for different player positions. The OB, WR, and LB positions all had significantly more extreme impacts than the OL position. Player position abbreviations are explained in the caption for Figure 5.2.

***Denotes a significant difference in the median number of extreme impacts between player position groups; *p*<0.05.**

5.3.6 Location of Impacts for the Different Player Positions

Position groups significantly differed on the number of impacts to the front, back, left, and right helmet locations (Figure 5.13; χ^2 (15)= 539.77, *p*<0.001). The OL position had significantly more impacts to the front location than expected compared to all other positions (*z*= 20.7, *p*<0.05). In contrast, the WR position had significantly fewer impacts to the front location than expected compared to all other positions (*z*= -11.8, *p*<0.05). The DB, LB, DL, and OB positions did not significantly differ from each other on the expected proportion of impacts to the front (*p*>0.05).

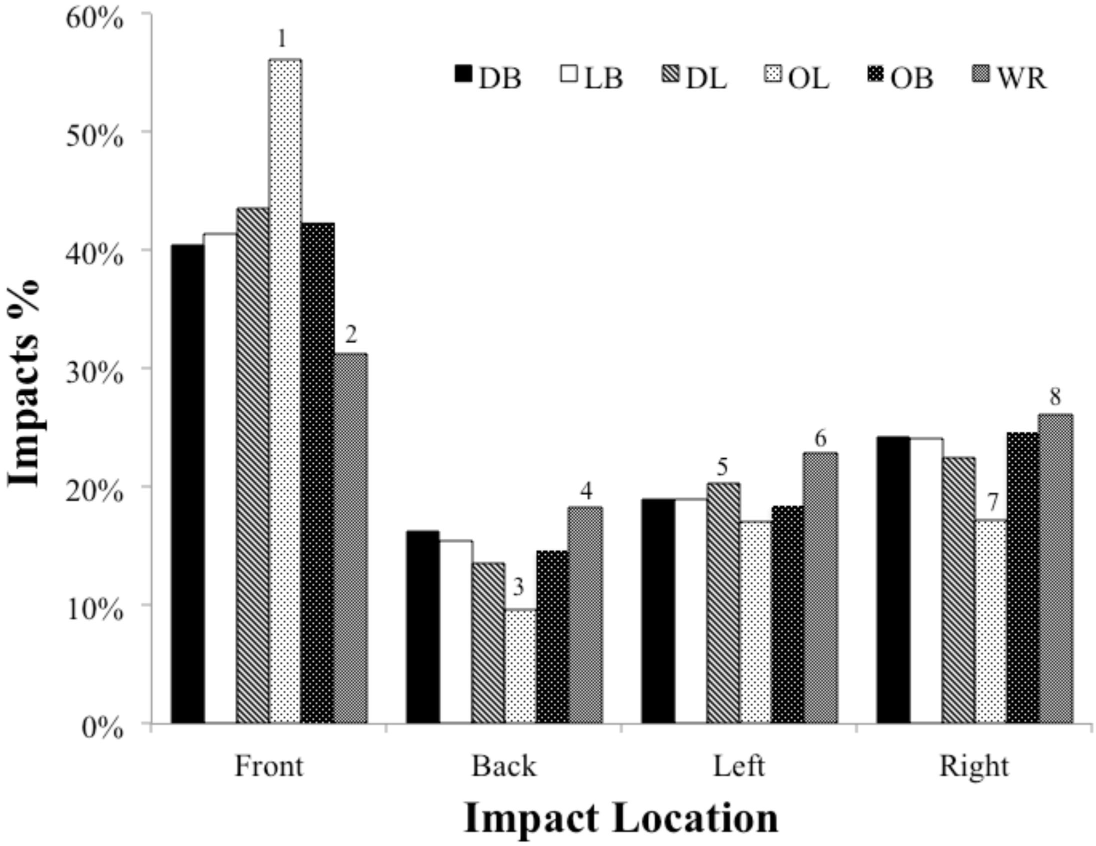


Figure 5.13: Percent of impacts experienced by each player position (graph series) to the front, back, left, and right location (different groups of bars). The numbers denote specific significant differences, p<0.05 (1) the OL position experienced a significantly larger proportion of impacts to the front compared to all other positions, (2) the WR position experienced a significantly smaller proportion of impacts to the front compared to all other positions, (3) the OL position experienced a significantly smaller proportion of impacts to the back compared to all other positions, (4) the WR position experienced a significantly larger proportion of impacts to the back compared to the OB, OL, and DL positions, (5) the DL position experienced a significantly larger proportion of impacts to the left compared to the OL position, (6) the WR position experienced a significantly larger proportion of impacts to the left compared to the OB, OL, LB, and DB positions, (7) the OL position experienced a significantly smaller proportion of impacts to the right compared to all other positions, (8) the WR position experienced a significantly larger proportion of impacts to the right compared to the DL position.

The DB, LB, and WR positions had significantly more impacts to the back than expected compared to the OL ($z= 3.9$, $p<0.05$; $z= 3.5$, $p=<0.05$; $z= 6.4$, $p<0.05$). Additionally, the WR position had significantly more impacts to the back than expected compared to the DL, and OB positions ($z= 6.4$, $p<0.05$). The OL position had fewer impacts to the back than expected and was significantly different from all other positions ($z= -11.1$, $p<0.05$). The DB, LB, DL, and OB did not significantly differ from each other on the expected proportion of impacts to the back ($p>0.05$).

At left location, the WR position had significantly more impacts to the left than expected compared to the DB, LB, OB, and OL positions ($z= 4.7$, $p<0.05$). The DL position also had significantly more impacts to the left than expected compared to the OL position ($z= 2.6$, $p<0.05$). The DB, LB, DL, and OB positions did not significantly differ from each other on the proportion of expected impacts to the left ($p>0.05$).

Finally, the OL position had significantly fewer impacts to the right location than expected compared to all other positions ($z= -11.4$, $p<0.05$). The WR position had significantly more impacts to the right than expected compared to the DL position and OL position ($z= 4.6$, $p<0.05$). The DB, LB, OB, and WR position did not significantly differ from each other on the proportion of expected impacts to the right ($p>0.05$).

6 Discussion

The first purpose of this study was to quantify the number, magnitude, and location of impacts experienced by the heads of Canadian university football players as they compete throughout a season of play. A head impact kinematic measuring device, the GForce Tracker, was adhered to the inside of 56 football players' helmets, and measured the linear acceleration and rotational velocity of impacts experienced by the players helmets as they competed in games and practices for the 2013 CIS football season. The following impact parameters were determined from the data collected throughout the season: median peak linear acceleration, median HIC_{15}, median peak rotational velocity, median peak rotational acceleration, and the median number of extreme impacts per 1,000 impacts.

The second purpose of this study was to compare the magnitudes of the impact parameters collected in games to those collected in practices and to compare the magnitudes of the impact parameters and location of impacts experienced by the different player positions. It was hypothesized that the magnitudes of impacts experienced in games would be larger than the magnitudes of impacts experienced in practice, but there would be no difference in the location of impacts to the helmet compared between games and practices. It was also hypothesized that the positional groups routinely playing along the line of scrimmage (linemen: OL and DL) would have a larger proportion of impacts to the front of helmet and smaller magnitudes of the head impact parameters compared to those positional groups routinely playing away from the line of scrimmage (skilled: OB, WR, LB, and DB).

6.1 Quantifying the Number and Magnitudes of Impacts Experienced by the 2013 Football Team

The 53 players experienced a total of 20,924 impacts during the 2013 CIS football season; 10,528 impacts were experienced during 55 practices and 10,396 impacts were experienced during 11 games. Comparing the current study to previous studies is difficult because some studies conducted their research for multiple years, or on multiple teams, or on more players than the current study. One previous study, collected almost 57,000 impacts over the course of two seasons; therefore the team averaged about 27,000 impacts per season (Mihalik et al., 2007). In that study the record threshold for the HIT system was set to 10 g. In contrast, the current study's record threshold was set to 15 g to match more recent studies (Crisco et al., 2010, 2011; Crisco & Wilcox, 2012). This difference in record threshold could be the reason for the difference for number of impacts, as the current study showed that a large number of low magnitude impacts occur during a season of football. These low magnitude impacts would have been captured with a record threshold of 10 g and would have increased the total number of impacts for the current study. A series of studies has investigated the number, location and magnitudes of head impacts to players of different positions for 3 different football teams (Crisco et al., 2010, 2011; Crisco & Wilcox, 2012). A total of 286,636 impacts were collected from the 3 teams over the course of 3 seasons (a median of 50 practices and 12 games per team). There was an average of about 31,000 impacts per season per team. These studies averaged more impacts per season than the current study. One reason for this discrepancy could be the different number of games played throughout the season by each team. In the current study each player experienced an average of 24 impacts per game. However, if this rate of impacts was maintained, but the season included one more game to match previous studies (Crisco et al., 2011; Mihalik et al., 2007), then the total number of impacts would increase from 20,924 to 22,034. Differences in practices may also contribute to these discrepancies. Other studies reported that there was a 2:1 ratio of impacts experienced during practice to impacts experienced during games (Crisco et al., 2010; Mihalik et al., 2007), while we observed a ratio that was closer to 1:1. In the current study a total of 4 practices are held prior to each game throughout the regular season. Three of these practices involved full contact and players were in full pads. The coach consciously intended to reduce the amount of contact in practice as the week progressed. A study on high school football players showed that there were significantly more impacts experienced by players in contact practices than in non-contact practices (Broglio et al., 2013). Additionally, another study reported that a youth football team had significantly fewer total number of impacts compared to two other youth football teams

(Cobb et al., 2013). The authors attributed this difference to the number of contact practices each team had during the season. The two teams with significantly more impacts had 2.1- 2.3 times more contact practices than the team with fewer impacts. The coaching philosophy of reducing the amount of contact in practices leading up to games may explain the discrepancy in the ratio of impacts in practice to impacts in games for the current study-- coaching philosophies on contact in practice differ from coach to coach (Cobb et al., 2013).

The magnitudes of peak linear acceleration and peak rotational acceleration are related to concussion, and have been commonly investigated in head impact biomechanics studies of football players. The data collected for this study was highly skewed with many low magnitude impacts and few high magnitude impacts; this data distribution is consistent with previous literature (Brolinson & Manoogian, 2006; Crisco et al., 2010; Duma et al., 2005; Mihalik et al., 2007). The median peak linear acceleration for all impacts collected was 20.43 g. and was consistent with the median peak linear accelerations of 20.30 g and 20.50 g from previous studies (Crisco et al., 2011; Crisco & Wilcox, 2012). In the current study the median peak linear acceleration was also closely related to the mean peak linear acceleration of 22.25 g from an earlier study (Mihalik et al., 2007), although they used a 10 g threshold for identifying impacts.

The HIT system was modified in 2006 to also calculate the peak rotational acceleration of impacts (Chu et al., 2006). Therefore, we can only compare the rotational accelerations from the current study to published data after 2009 when the first study that quantified peak resultant rotational acceleration to players was published (Broglio et al., 2009). The median peak rotational acceleration in this study was 1,724.44 rad/s^2. This was slightly larger but consistent with other studies that showed median peak rotational accelerations of 1,400 rad/s^2, and 1,320 rad/s^2, (Crisco et al., 2011; Crisco & Wilcox, 2012), and a mean peak rotational acceleration of 1,430 rad/s^2 (Ocwieja et al., 2012). A study comparing offensive schemes found significantly larger mean peak rotational accelerations for the pass first offense compared to run first offense (1,777.58 ± 1,266.61 vs 1,675.36 ± 1,183.94 rad/s^2 respectively; Martini et al., 2013). In Canadian football there is tendency for a pass first offense because of the 3 down rule. Canadian teams only have 3 attempts to gain 10 yards for a first down while American teams have 4 attempts. Although run to pass ratios were not documented in the current study, a greater reliance on the pass first offense may have lead to the larger median peak rotational acceleration compared to previous (American) studies.

The 95th percentile of peak linear and rotational acceleration shows the upper 5% of the distribution of magnitudes of impacts. These magnitudes are larger and are more representative of injury risk as these larger magnitudes are causing concussions (Pellman et al., 2003; Rowson et al., 2012). The 95th percentile peak linear and rotational accelerations from the current study were 36.13 g and 3,029.62 rad/s^2. These values were much lower compared to the 95th percentile peak linear (approximately 62 g) and rotational accelerations (approximately 4,300 rad/s^2) from previous work (Crisco et al., 2011; Crisco & Wilcox, 2012). Additional investigations should focus on clarifying why the current study's 95th percentile impact magnitudes were lower compared to previous research (Crisco et al., 2011; Crisco & Wilcox, 2012).

The rotational velocity of football players' heads when they receive impacts while competing in games and practices has not been quantified in previous literature. The GForce Tracker device collected this impact parameter using its tri-axial gyroscope. The median peak rotational velocity for the current study was 11.36 rad/s and was heavily skewed to the right, like the peak linear and rotational acceleration data. These data can be compared to the rotational velocities of concussive and non-concussive impacts of struck and striking players quantified by reconstructing impacts experienced by NFL players with HIII dummies (Pellman et al., 2003). They report that the average peak rotational velocity for non-concussive impacts was 26.3 rad/s, and 34.8 rad/s for concussive impacts. Only 31 total cases made up the data set for this study, of which 6 cases were non-concussive injuries for struck players. There are many low magnitude rotational velocity impacts that occur during practices and in other parts of games. Accordingly the average peak rotational velocity that they report on their relatively small number of reconstructed impacts was likely not representative of what players experience throughout an entire season. Rotational velocity is not a common measure in head impact biomechanics, but research has indicated rotational velocity from an impact can influence injury mechanisms to the brain (Hardy, Mason, & Foster, 2007; Sullivan et al., 2013; Weaver, Danelson, & Stitzel, 2012). Amounts of strain damage and structural damage to the brain were recently attributed to the magnitude of the angular velocity and also the direction of the impact causing the head to rotate (Weaver et al., 2012). Injury severity metrics are now shifting from the contributions of linear acceleration (HIC_{15} and GSI) in brain injury to the contributions of rotational velocity and acceleration, direction of impact and impact duration (Sullivan et al., 2013). A case can be made for quantifying rotational velocities for future studies, given its association with concussions.

HIC values in the current study were also severely skewed and the median HIC_{15} from all impacts was 9.17. The HIC has not been reported in recent studies of football player head impact biomechanics since the development of a new impact parameter, exclusive to the HIT system, to measure impact severity, the HITsp. The HITsp provides a severity measure of an impact by weighting the peak linear acceleration, rotational acceleration and impact location and these impact parameters are more sensitive to detecting concussion than each of them individually (Rowson & Duma, 2013) . The mean HIC reported in the first study using the HIT system was 26 (Duma et al., 2005) which is considerably larger than the current study's median HIC. There are likely two reasons for the discrepancies in HIC values between the two studies. First, the previous study reported an average HIC. HIC, like linear and rotational acceleration, can be highly skewed from many low magnitude and few high magnitude HIC values. Outliers in a distribution can influence mean and in the current situation larger magnitudes of HIC values (outliers) in a distribution of predominantly low magnitudes would increase the mean relative to the median. Given its skewed distribution, presenting the median would be a more appropriate measure of central tendency, as it is not influenced by outliers in a distribution. Second, impact data from 53 players in one season were used in the current study, and most of the participants had the GFT device for the majority of the season. The previous study was limited to only 8 HIT system units at a time; they had to rotate the units every 2 weeks in order to collect from a variety of positions (Duma et al., 2005). This meant that their data might not be representative of the true impact data for an entire season from an entire team.

Across all players, and for both games and practices, the highest percentage of impacts occurred to the front of the helmets; this was consistent with another study (Crisco et al., 2010). Impact locations do significantly differ between player positions. Subsequently, these differences between positions affect the risk for concussion (Zhang et al., 2001; Zhang, Yang, & King, 2004). Differences between positions will be discussed in a later section.

6.2 Impact Magnitudes in Games and in Practices

Magnitudes for all head impact parameters (median peak linear acceleration, median HIC, median peak rotational velocity, median rotational acceleration and median number of extreme impacts) collected during games were significantly larger than the magnitudes for the impacts parameters collected during practices. Differences between games and practices ranged from 8% for the median peak linear acceleration up to 222% for the number of extreme impacts. Results from the current study were not consistent with previous literature. One previous study showed that there was no significant difference in the median and 95th percentile peak linear and peak rotational acceleration between games and practices (Crisco et al., 2011). In contrast, another study showed that the average peak linear acceleration experienced during practices (22.65 g) was significantly larger than impacts experienced in games (21.12 g); however, this difference was very small between session types and might not be important clinically. Currently, there is no previous literature to compare our results of median HIC and median peak rotational velocities in games versus practices. No studies have quantified the number of extreme impacts per 1,000 impacts, but there are similar measures such as the number of impacts above 80 g (Mihalik et al., 2007). However, in that study the number of impacts greater than 80 g was determined for position types but not for different session types (game versus practice). Other studies determined the 95th percentile of the peak linear accelerations as it provides an indication of the magnitudes of impacts at the upper end of the distribution (Crisco et al., 2011; Crisco & Wilcox, 2012). However, these studies do not indicate the number of impacts at the upper end experienced by players in each session. It is difficult then to compare results for the current study for the number of extreme impacts experienced during practice and games to previous literature.

For the purposes of the current study comparing its results to published literature has been limited to studies that quantified the head impact biomechanics of collegiate football players, because both levels would represent relatively similar skill levels. However, one study done on high school football players with a similar amount of games (13) and practices (55) to the current study, found that both the average peak linear and peak rotational accelerations were larger in games than in practices (Broglio et al., 2009). That study attributed the differences in magnitudes to the way games and practices are structured, as described below.

Coaches will structure a practice in a way that will limit the number of impacts and the intensity or magnitudes of impacts to reduce the chance of injuries to the players. In the current study, the head coach kept the majority of hitting limited to one practice a week. In that practice intense hitting, similar to games, was limited to a period for the offense to work on run blocking schemes against the defense lasting no more than 20 minutes. The running drills during practice were kept within constrained space to limit the closing distance a player had to make a tackle. A study investigating the relationship between closing distance and head impact magnitudes showed that the peak linear and peak rotational accelerations were significantly larger when players were making impacts with a closing distance greater than ten yards (Ocwieja et al., 2012). Thus, in a session during practice where impacts are structured to occur, the magnitudes of these impacts are reduced because of the space constraint. Otherwise, any other hitting that occurs during practice is limited to thudding. Thudding is where the player making a tackle will slow down before contacting the ball carrier with their shoulder in a controlled manner and not bring the ball carrier down to the ground. Thudding is a safer way to perform a tackle during practice. Some of the objectives for practice are to have players work on timings of plays, as well as prepare for the upcoming opponent by being able to recognize formations and specific plays. Therefore, practices are designed to prepare for an upcoming opponent in a way that keeps the risk for injury low.

In contrast to practices, games offer an environment for players to experience larger magnitudes of impacts (Broglio et al., 2009). Players are able to use the full range of the field (110 yards by 53 yards) to execute their positional duties; they are not limited by the same stipulations as practices. Therefore, the closing distances between the tackler and the ball carrier vary and players can be at a maximum running velocity when experiencing an impact. Consequently, impacts during games result in larger magnitude impacts (Ocwieja et al., 2012). Players are no longer thudding a teammate but running full speed to deliver impacts by blocking an opponent or taking opposing ball carriers to the ground when making a tackle (Broglio et al., 2009).

Players experienced a higher proportion of impacts to the front of the helmet during practices compared to games. No previous studies have quantified differences in proportions of impacts to various locations on the helmet between games and practices. As mentioned earlier in this section, scheduling of sessions during practice allowed the coach to control how players received impacts. With a higher proportion of impacts to the front of the helmet, players can see and anticipate when the impacts are coming. A study showed that when hockey players anticipate an impact, the resulting linear acceleration tended to be lower than unanticipated impacts (20.7 g vs. 22.6 g; Mihalik et al., 2010). They also showed that for severe impacts (impacts in the 75th percentile for HITsp), anticipated impacts had significantly less rotational accelerations than unanticipated impacts (1,215.11 rad/s^2 vs. 1,465.7 rad/s^2). In the current study players could anticipate the impacts during practice and the magnitudes of impact parameters were significantly lower in practice than during games. In contrast, lower proportions of impacts to the front, but higher proportions of impacts to the left side were experienced during games. With impact locations more variable in games than in practices, players cannot anticipate when an impact is coming; consequently, the magnitudes of the impact parameters during games were significantly higher than during practices.

While the magnitudes of impacts were larger during games, there is a question of whether an 8% difference in the median peak linear acceleration in games to practice is clinically significant? A study on the injury rates of football players showed that players had higher rates of injuries (36 per 1,000 AE) in games than in practices (4 injuries per 1,000 AE; Dick et al., 2007). More importantly football players had higher rates of concussion while playing in games (2.34 per 1,000 AE) than while participating in practices (0.21 per 1,000 AE). Larger magnitudes of peak linear acceleration and rotational acceleration are predictors for concussion (Funk et al., 2012). Therefore, there is a relationship between the magnitudes of impacts in games and risk of concussion during games and there would be implications for rule changes in games to reduce the magnitudes of impacts and risk of injury. The NFL has become much stricter on direct impacts to players heads, either by penalizing players during the game or by penalizing them after the game with fines or suspensions. Hopefully this will lead to a culture change where intentional direct head impacts are not part of football.

6.3 Impact Magnitudes of Different Player Positions

Previous studies showed that impact magnitudes will vary depending on the position of the player (Crisco et al., 2011; Crisco & Wilcox, 2012; Mihalik et al., 2007). These studies attribute the differences in magnitudes to the different demands of each position. Therefore, the demands of each of the player positions will be a common theme that will be discussed in the following section as they relate to the various impact parameters.

6.3.1 Peak Linear Acceleration

The OB, and WR positions had significantly larger median peak linear accelerations than the OL group. These differences ranged from 7%, between the WR and OL group, up to 13% between the OB and OL group. A previous study showed similar results and determined that the running back, quarterback, and linebacker position had significantly larger median peak linear accelerations than the OL position; however, there was no significant difference in the median peak linear acceleration between the OL and WR position (Crisco & Wilcox, 2012). The Canadian game of football has higher frequencies of passing, as mentioned in section 6.2, and the players at the WR position are allowed to be in motion behind the line of scrimmage when the ball is snapped. This position can be running at maximum speed at the snap of the ball by running up to the line of scrimmage before the ball is hiked. In contrast, the players at the WR position in the American game must have their feet set at the snap of the ball and will take longer to get to a maximum speed. A study showed that the WR position from a passing style offence had a larger mean peak linear acceleration than a WR from a running style of offence (27.2 ± 17.4 g vs. 24.7 ± 15.7 g respectively; Martini et al., 2013) . This study reported that the WR position in a passing style of offence is spread out across the field more than a run style and may be able to reach higher running velocities before impact. This will increase the linear acceleration when players at the WR position experience an impact. The current study's 7% difference in the median peak linear acceleration between the WR and OL position may not be clinically significant. Future studies should determine the 95th percentile impact magnitudes for of the WR position group as the higher magnitudes of peak linear accelerations are more related to concussion than the median (Funk et al., 2012).

Another study found the opposite to the current study; they showed that the mean peak linear accelerations for the OL position was not significantly larger than the OB, LB and WR group (percent differences ranging between 0.17% for OB vs OL to 3.2% for OL to WR; Mihalik et al., 2007). The offensive line position protects the OB group; they either block and protect on passing plays or block and create lanes on running plays. These assignments are executed in close proximity to the opposing teams DL and sometimes LB positions. The short closing distance will keep the impact magnitudes more consistent and on average lower in magnitude than the skilled positions (OB and WR; Ocwieja et al., 2012). In contrast, the roles for the skilled positions, OB, and WR give them the opportunity to get to higher running velocities to execute their positional duties. Consequently, these positions are exposed to a larger range of impact magnitudes, and on average higher magnitudes of impacts than the OL and DL positions. As a result, the OB and WR positions had larger median peak linear accelerations than the OL group.

6.3.2 HIC_{15}

The OB position had a significantly larger median HIC_{15} than the OL Group. Once again, it is difficult to compare the current study's finding to previous literature since the HIT system evolved away from quantifying the HIC_{15} parameter. The technique for blocking used by the OL may account for the differences between the OL and OB groups. Offensive linemen are taught to use three points of contact when blocking a player. The hands provide two of the three points of contact and the forehead area on the player's helmet provides the third. The impact using the three points of contact is used to stun the opposing player and gain control of them to execute a block. The HIC_{15} not only takes the magnitude of the linear acceleration that the head experiences from an impact, but also accounts for the duration of the acceleration. The quick and sudden impacts from the OL position while they are blocking could lead to the HIC values remaining low as the accelerations may be briefer than 15 ms. In contrast the OB position, made up of running backs, fullbacks and quarterbacks, usually doesn't get the opportunity to control when and how they get hit. Most of the time these positions are carrying the ball and are usually the positions being hit from other players on the field. This will expose these players to more varying magnitudes and durations of impacts, and could influence the median HIC_{15} for this position grouping. A study recreating football impacts using HIII dummy heads showed that the average HIC of a player being tackled (the struck player) was 308 while the average HIC was 121 for the player that was tackling the ball carrier (the striking player; Viano, Casson, & Pellman, 2007; Viano & Pellman, 2005). The majority of these impacts recreated from this study (25 of 31) were impacts that produced concussion to the struck player and represent the upper distribution of HIC values. However, this study suggests that there are differences between striking players and struck players in the magnitude of the head impacts.

6.3.3 Peak Rotational Velocity

The median peak rotational velocities were significantly different between positions; however, we could not determine which pair(s) of positions were significantly different. The OB (14.31 rad/s) and LB (11.79 rad/s) group had the largest median peak rotational velocities and were 34% larger and 12% larger than the OL group. Likely this is where the differences lay between position groups; however, the controlling for multiple comparisons bias from the modified Bonferroni correction may have reduced the power of the statistical test and the inability to locate the significant difference(s).

6.3.4 Peak Rotational Acceleration

The OB (34%) and WR (16%) position had significantly larger median peak rotational accelerations during impacts than the OL position. This was consistent with another study that showed the running back, quarterback, and linebacker positions had significantly larger median peak rotational accelerations than the OL position (Crisco & Wilcox, 2012); however, that study showed that the running back, quarterback, and linebacker positions also had a significantly larger peak rotational acceleration than the WR position. The differences between positions on impact location will be discussed in section 6.3.6, but the WR position experienced significantly fewer impacts to the front of the helmet than expected, and significantly more impacts to the back of the helmet than expected, compared to the OL position. We touched on the role of anticipation of impacts in section 6.2, and this theme is consistent with the rotational acceleration for the OL and WR position. A study on the effect of neck strength and anticipation of an impact showed that anticipating an impact significantly reduced the change in velocity of the head than unanticipating the impact; change in velocity of the head anticapting an impact ranged from 5% less up to 11% less than an unanticipated impact (Eckner, Oh, Joshi, Richardson, & Ashton-Miller, 2014). In this study, subjects engaged their cervical neck muscles in preparation for an impact. The cocontraction of the neck muscles increased the necks resistance to head motion from an external force (the impact). In the current study, the OL position experienced a larger proportion of impacts to the front of the helmet; therefore, they could anticipate and reduce the rotational acceleration of the impact by fully contracting their cervical neck muscles (Mihalik et al., 2010). In contrast, impacts to players' helmets in the WR position were more consistent across all of the helmet locations and had a higher percentage of impacts to the back and sides than the OL position. Consequently, impacts were experienced to the back and sides of the helmet by the WR were likely unanticipated; therefore, this position had larger peak rotational accelerations than the OL, because they may not have been fully contracting the neck muscles in preparation for the impact (Mihalik et al., 2010). A similar reason can be proposed for the OB position that had significantly fewer impacts to the front than expected compared to the OL. Additionally, Canadian football allows the players at the WR position to be in motion at the snap of the ball. This allows the WR position in the Canadian game to be at a higher running velocity before experiencing an impact, and the magnitude of rotational acceleration of the head from an impact is increased (Broglio et al., 2009). In contrast, the players at WR position in American football requires players in that position to have their feet set in place at the snap of the ball. It would take longer for the WR position for American football to reach a maximum running velocity before experiencing an impact. The difference in the WR position motioning before the snap of the ball likely lead to larger magnitudes of peak rotational acceleration in the current study compared to a previous study (Crisco & Wilcox, 2012).

6.3.5 Number of Extreme Impacts

The median number of extreme impacts was the greatest for the OB (87.89 impacts per 1,000) and WR (80.81 impacts per 1,000) positions, with both receiving significantly larger impacts than the OL. The LB position had a significantly larger median number of extreme impacts than the OL position. The OB position experienced a higher number of larger magnitude impacts, which is consistent with an earlier study (Mihalik et al., 2007). In that study, the OB position experienced significantly more impacts above 80 g than all other positions (DL, OL, LB, DB, and WR). The OB position group could have some of the highest rates for concussion. In the current study, this position is composed of multiple positions: fullback (FB), running back (RB), and quarterback (QB) and a previous study showed that the quarterback and running back positions experienced the highest percentage of concussions (27.6% and 17.2%) compared to all other positions (Dick et al., 2007). Another study evaluated the risk of concussions with HIT system data and showed that the number of large magnitude impacts (above 100 g) correlated well with previous epidemiological data on concussion rates in football (Funk et al., 2012). In that study, the probabilities for concussion based on the number of high magnitude impacts were the highest for the running back and quarterback positions. The previous study suggests that it is not only the magnitude of the impacts that affect the risk for concussion but also the number of the larger magnitude impacts. The risk for concussion was not determined in current study; however there might be a larger risk to the OB position based on the number of extreme impacts from the current study and the evidence from previous research.

In contrast to one study (Mihalik et al., 2007), the WR position in the current study experienced a high number of larger magnitude impacts. The reason for this discrepancy could be the definition used in current study to define an extreme impact: any impact that exceeds the 95th percentile peak linear acceleration for the distribution of all impacts. The 95th percentile peak linear acceleration for the current study (36.13 g) was lower than the 95th percentile peak linear accelerations (62.7 g and 62.2 g) from previous literature (Crisco et al., 2011; Crisco & Wilcox, 2012). This would have lead to a higher number of impacts being described as "extreme", when in actuality these impacts are lower in magnitudes and may not be relevant to injury risk. A study using an FE model determined that there was a 25% probability of sustaining a concussion from an impact with a peak linear acceleration of 66 g and a peak rotational acceleration of 4,600 rad/s^2 (Zhang, Yang, & King, 2004). Probabilities of sustaining concussions increased to 50% and 75% when magnitudes of impacts increased to 82 g and 5,900 rad/s^2, and 102 g and 7,900 rad/s^2. Using 66 g and 4600 rad/s^2, as impact magnitudes to determine an "extreme" impact may be more appropriate for determining injury risk. However, a benefit of the current study's injury risk metric, the average number of impacts per 1,000 impacts, is that it quantifies the number of impacts. Magnitudes of head accelerations are important for injury risk, but the number of impacts is also important for injury risk and potential long-term detrimental effects. For example, a previous study showed that decreases in verbal learning and reaction time are related to higher magnitudes of impacts but also a larger cumulative amount impact magnitudes (i.e. number of large magnitude impacts; McAllister et al., 2012). Future studies should continue to report concussion injury risk and clinical outcome measurements relative to number of impacts, magnitude of impacts, but also the number high magnitude impacts.

6.3.6 Impact Location Differences Among the Player Positions

The OL position experienced significantly more impacts to the front of the helmet than expected, compared to all other positions. Accordingly, this position experienced significantly fewer impacts to the back of the helmet than expected compared to all other positions. This was consistent with another study that quantified the number and location of impacts to different player positions (Crisco et al., 2010). In that study, the OL position experienced significantly more impacts to the front of the helmet than the QB, WR, DL and LB positions. Another study quantified the magnitudes of the impacts associated with each impact location and showed that the impact magnitudes to the front location for the OL position was lower compared to all other positions (QB, RB, WR, DL, LB, and DB; Crisco et al., 2011). While the current study did not determine the average impact magnitudes per helmet location, the OL position had the lowest median peak linear and rotational accelerations and significantly lower magnitudes compared to the OB and WR position. Previous research showed that concussion risk and the amount of DAI increases when impacts are directed towards the side of the head as compared to the front (Weaver et al., 2012; Zhang et al., 2001). Additionally, another study showed that 7.7% of diagnosed concussions were to the OL position but their rate of concussion was the lowest compared to other positions (QB, RB, WR, DL, LB, and DB; Dick et al., 2007). Therefore, it may be concluded that while the OL position sustains a large number of hits to the front of the head, the magnitudes of these hits and their potential to cause concussions are low. In football, the OL and DL positions experience the highest number of impacts compared to other player positions (QB, RB, WR, LB, and DB; Crisco et al., 2010). These linemen positions set up opposite to each other on the field and their positional duties usually require them to engage in contact with each other on almost every play throughout the game. Given these positional duties, a large proportion of impacts occur to the front of the helmet for these positions. However, the current study showed no significant difference in the proportion of impacts to the front location for the DL position compared to the DB, LB, OB, and WR positions. This was likely due to an additional duty for the DL position; this position not only routinely engages in contact with the OL position, but is also tasked with tackling the opposing team's ball carrier. As a result, the impacts experienced by the DL position are more evenly distributed around the helmet compared to the OL position.

In contrast the WR position experienced significantly fewer impacts to the front of the helmet than expected and significantly more impacts to the back of the helmet than expected compared to all other positions. While one study showed no difference in the percentage of impacts to the front compared to the back for the WR position (Crisco et al., 2010), another study showed a similar result to the current study and showed that the WR position experienced more impacts to the back of the helmet compared to all other positions except the QB position (Crisco et al., 2011). That study, showed that the impact magnitudes for impacts experienced at the back of the helmet were significantly larger in the WR position compared to all other positions except the QB position. We showed that magnitudes of the impacts the WR position were significantly larger than the OL position, suggesting that larger magnitudes of accelerations are associated with impacts not to the front location for the WR position. Subsequently, this could put this position at risk for concussion.

6.4 Limitations

One limitation of this study was the calibration algorithm used to predict magnitudes of accelerations occurring at the centre of mass of players' heads. The algorithm was developed and validated by inducing repeatable head impacts to a HIII dummy wearing a football helmet equipped with a GFT and a relatively large number of impacts (123; Appendix B). Although this algorithm was effective (coefficients of determination > 0.87 for the peak linear and rotational accelerations, peak rotational velocity and HIC15; Appendix C), it was limited to a single helmet with a single style of facemask, and a single placement location of the GFT. This helmet was the Riddell Revolution Speed Helmet (Riddell, Elyria, OH), as majority of the team, and the players that were included into the statistical analysis, wore this style of helmet. The Riddell Speed helmet had appropriate space for mounting the GFT to inside of the shell just to the left of the crown air bladder (Figure 4.1). This mounting location was kept consistent across all helmets.

Therefore, the calibration algorithm was specific to the Riddell Speed helmet and that GFT mounting location. A study recently showed that newer models of helmets can reduce the magnitudes of peak linear accelerations from head impacts (Steven Rowson et al., 2014). Therefore, the calibration algorithm should only be made generalizable to players wearing the Speed Helmet and the left side GFT mounting location. Future work should develop algorithms for different styles of helmets and different styles of facemask.

Other limitations were the length of the season and length of the study. For the current study, there were only 8 games in the regular season. Fortunately 3 additional games in the playoff allowed for more opportunities to quantify head impact exposures to players; however, we fell short of 12 games, which is the average number of games for an NCAA football team. Future studies that compare head impact exposures of Canadian university football players to American collegiate football players should ensure that there are an equal number of games to compare the two styles of football; alternatively future studies should report the average number of impacts per game so that the different teams with different schedule lengths can be compared. Additionally, previous studies have quantified head impact exposure over many seasons (Crisco et al., 2010, 2011; Crisco & Wilcox, 2012; Mihalik et al., 2007). Although there is a trend for the OB and WR position being exposed to larger magnitudes of impacts, future studies should expand on the current study to gain a more informed understanding of the number and magnitudes of impacts to Canadian University football players.

7 Conclusion

Results showed that the magnitudes of impacts experienced in games were significantly larger than the magnitudes of impacts experienced during practices. This result was expected, and is consistent with the hypothesis. However, players experienced impacts to the front of the helmet more than expected during practices compared to games; this result was not consistent with the hypothesis. Coaches will schedule a practice to work on timing of plays, develop skills for the players, and evaluate talent all in effort to prepare for an upcoming opponent. All of this is performed in a controlled environment to minimize the risk of injury. Thus players' experience a larger proportion of impacts to the front of the helmet during practice and smaller magnitudes of impacts. In contrast, there is no scheduling of events in games. Impacts occur in an uncontrolled environment and different factors, such as player intensity and closing distance to make an impact, increase the magnitudes of impacts experienced in games and impacts to the helmet are more consistent across all locations compared to practices. Therefore, the game offers an opportunity to modify rules and styles of play that can hopefully reduce the magnitudes of impacts and risks of concussion.

As expected, and consistent with the hypothesis, there are differences in the magnitudes of impacts to different player positions. Each position has specific roles that they perform during practice and games. Linemen positions were exposed to consistently low magnitudes of impacts, because of the proximity between each other (OL and DL) on the football field. This also forces the linemen positions, especially the OL group, to experience a higher proportion impacts to the front of the helmet compared to skilled positions. Skilled positions will be exposed to larger magnitudes of impacts, because they play off the line of scrimmage and are able to get to maximum running velocities before experiencing impacts.

The OB position consistently had larger magnitudes of the impact parameters than the OL position. These impact parameters included median peak linear acceleration, median HIC_{15}, median peak rotational acceleration and median number of extreme hits. The OB position was composed of quarterbacks, running backs and full backs, and these positions are usually carrying the ball. The opposing players are striking the OB position more often in the open field, or impacting this position after using a large closing distance. As a result, the magnitudes of the impacts vary and this position will receive larger magnitudes of impacts. The OB position could be at the same level of risk for concussion to American data, given the magnitudes of impacts the position experienced and how these magnitudes relate to concussion.

The magnitudes of impacts to the WR position did not agree with previous literature. Future studies should quantify the number of pass plays in a season by a Canadian football team to help to inform why the WR position in Canadian football is exposed to larger magnitudes of impacts than its American counterpart.

The current study was conducted for only one season; however, it has been important in filling a gap in the literature of head impact biomechanics to Canadian football players. The Canadian game and the American game offer different rules that subsequently change head impact exposures to different positions. The question of if the Canadian style of game reduces head injury risk or increases it has still to be fully explored.

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Appendices

Appendix A: The University of Western Ontario Human Research Ethics Board Approval Form

Western Research

Research Ethics

Use of Human Participants - Ethics Approval Notice

Principal Investigator: Dr. Jim Dickey
File Number:103727
Review Level: Delegated
Approved Local Adult Participants:75
Approved Local Minor Participants:0
Protocol Title: Biomechanical Analysis of Head Impacts to Canadian University Football Players of Different Positions
Department & Institution: Health Sciences\Kinesiology,Western University
Sponsor:
Ethics Approval Date: May 02, 2013 **Expiry Date:**April 30, 2014
Documents Reviewed & Approved & Documents Received for Information:

Document Name	Comments	Version Date
Other	References for Background information	2013/04/09
Letter of Information & Consent	Letter of Information and Consent Form	2013/04/09
Western University Protocol		2013/04/10
Recruitment Items	revised recruitment poster	2013/05/01
Other	summary of revisions to the protoccol and the recruitment poster	2013/05/01

This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced revision(s) or amendment(s) on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the University of Western Ontario Updated Approval Request Form.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

[redacted]

Ethics Officer to Contact for Further Information

Erika Basile	✓ Grace Kelly	Shantel Walcott
[redacted]		

This is an official document. Please retain the original in your files.

[redacted]

Appendix B: Validation of a calibration algorithm for the GFT to predict head impact kinematics occurring at the centre of mass of the head.

Background: GFTs were adhered to the inside of football helmets; therefore, linear accelerations and rotational velocities of the helmet were measured when a player experienced an impact. The GFT measures kinematics of the helmet shell and does not use an algorithm to predict head impact kinematics occurring at the centre of mass of the head.

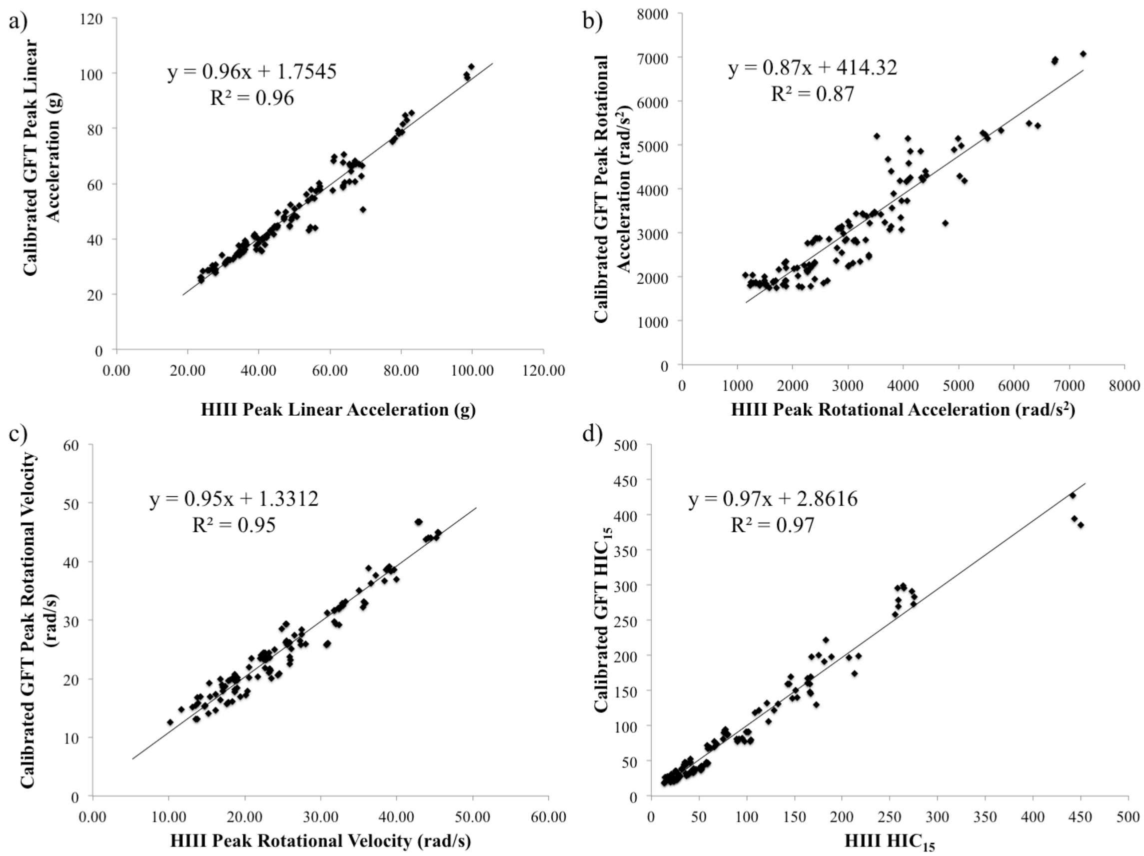
Objective: To develop and validate a calibration algorithm that predicts head impact kinematics at the centre of mass of the head from measurements made by the GFT device.

Methods: Repeatable impacts were delivered to a HIII dummy head wearing a Riddell Revolution Speed Helmet (with a GFT mounted inside) by pneumatic linear impactor; this testing was performed at private testing facilities in Ottawa (Biokinetics and Associates Ltd; Beckwith et al., 2012). Impacts were delivered to 7 impact locations on the helmet (front, left and right front Boss, left and right Sides, back, and facemask), at 6 velocities (3.0, 3.7, 4.2, 5.5, 7.0, and 8.5 m/s). Each impact velocity and helmet location was repeated 3 times for a total of 123 impacts (the facemask was the only location that did not receive the 8.5 m/s velocity). Coefficients of determination (r^2) and slopes of the line of best fit (m) were determined to evaluate the comparison between the calibrated measurements from the GFT to the gold standard HIII on the peak linear acceleration, peak rotational velocity, peak rotational acceleration, and HIC15 from impacts measured by each system.

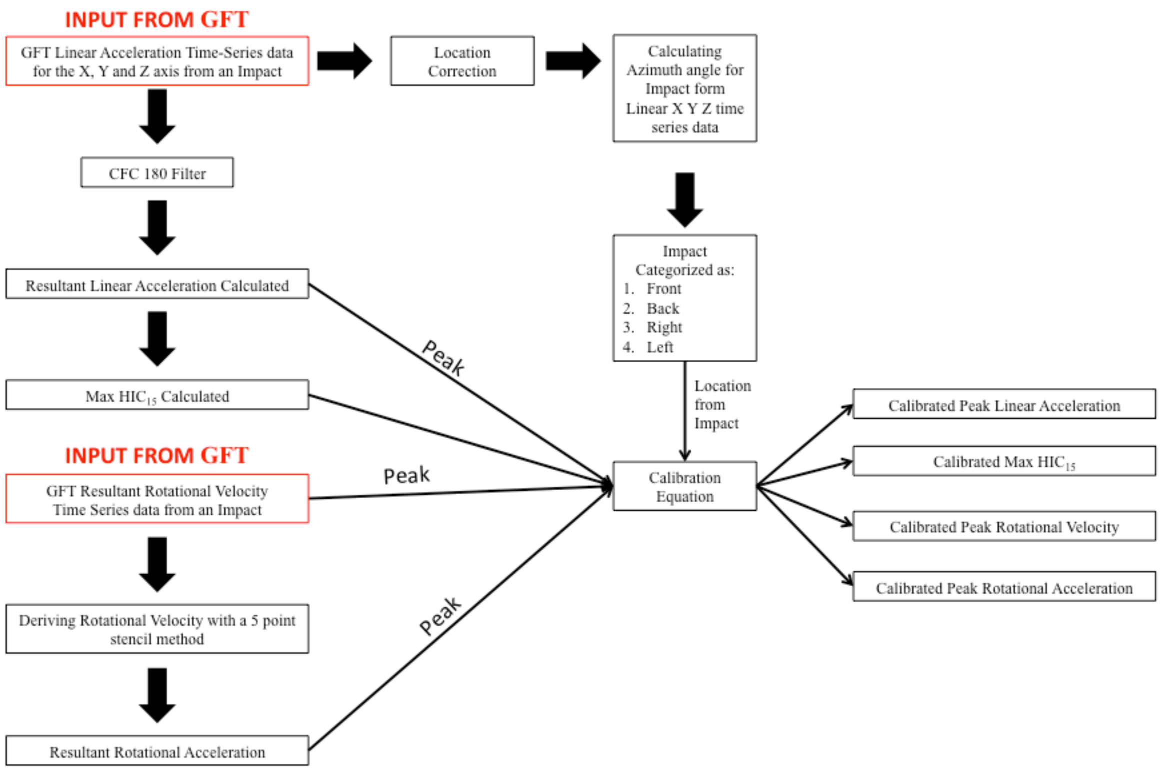
Results: All impacts parameters measured by the calibrated GFT were strongly correlated to the impact parameters measured by the HIII (Appendix C).

Discussion: Coefficients of determination and slopes were comparable to previous a previous study that validated the HIT system against a HIII dummy head (Beckwith et al., 2012). The HIT system is a heavily used research tool in determining head impact biomechanics to football players. The results from the validation suggest that the GFT can be used as a research tool in a similar manner to the HIT system.

Appendix C: Linear regression analysis between the a) peak linear accelerations measured by the GFT and HII, b) peak rotational accelerations measured by the GFT and HIII, c) the peak rotational velocities measured by the GFT and HIII, and d) HIC15 measured by the GFT and HIII



Appendix D: Flow chart illustrating the steps in the calibration algorithm



Curriculum Vitae

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