Head Impact Exposure: The Biomechanics of Sports-Related Concussions

By

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Preface and Acknowledgements

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Chapter 1

Introduction

1.1 Overview

While mild traumatic brain injuries (MTBI) or concussions are the most common type of brain injury, they are exceedingly complex injuries that are difficult to diagnose and manage. It is estimated that approximately 1.6 to 3.8 million sports-related MTBIs occur annually in the United States. Recently, concerns about the long term effect of these injuries and the potential cumulative effects of repeated injuries have arisen. Concerns have also been raised that repetitive sub-concussive impacts are a possible cause of chronic brain injury. While it has been accepted that concussions result from an acceleration of the head after a direct or indirect impact to the head or body, the exact relationship between the biomechanics of these impacts and their clinical outcomes is unknown. In contact sports, there are a number of different, sport-specific head impact mechanisms, or circumstances in which head impacts occur (e.g. head contact with ice, boards, etc. in ice hockey). Head impact mechanisms and their resulting impact biomechanics are not well understood, but could improve the understanding of concussion injury and inform prevention strategies. It is also not well understood why certain risk factors, including sex, age, and previous history of MTBI have been associated with increasing an individual's susceptibility to MTBI. Epidemiological evidence suggests that female athletes may be at a greater risk of concussion than their male counterparts, but the exact reasons why remain unclear.

The primary objectives of this study are to quantify and examine *in vivo* biomechanical head impact data in several populations of contact sport athletes, examine these variables in relation to head impact mechanisms, correlate this data with clinical outcomes related to concussion, and assess the differences between males and females in these measures.

1.2 Background

1.2.1 Concussion

The word "concussion" is derived from the Latin words *concutere* and *concussus* which mean "to shake" and "to clash or strike together", respectively. While the study of concussion injury began more than 2,000 years ago, there still exists confusion and disagreement regarding the exact definition of this injury. Historical accounts suggest that the first description of a concussion was recorded by a Persian physician named Rhazes in the 10th century who appreciated that this head injury could occur independently of any gross pathology or skull fracture [1]. By the turn of the 16th century, concussions were identified as *commotio cerebri* or a "shaking of the brain" that resulted in transient symptoms [1,2]. In general, for the past two millennia the term concussion has been used to describe both a clinical state and the events that bring about that clinical state. Interestingly, today's medical field has yet to accept a single, universal clinical definition for concussion injury, and the exact mechanism of this injury is still not understood [3]. For the purpose of this dissertation the terms concussion and mild traumatic brain injury (MTBI) are used synonymously and are defined according to the American Academy of Neurology's definition; a concussion is a trauma-induced alteration in mental status that may or may not include loss of consciousness [4].

For an injury that is not well understood, concussions have a relatively high incidence and can affect a wide population of individuals of varying age and health. From elderly people at risk of falling, to individuals involved in motor vehicle accidents, to our troops serving overseas who are at risk of exposure to improvised explosive devices (IEDs); anyone in any different number of circumstances can sustain an impact to the head that may result in an MTBI. Athletes, especially those in contact sports who are exposed to repetitive head impacts, are at particular risk. It is estimated that 1.6 to 3.8 million sports-related concussions occur annually in the United States [5]. The National Center for Injury Prevention and Control has identified both severe and mild sports-related brain injury as an important public health problem because of the high incidence, the relative youth of those sustaining the injuries with possible long term disability, and the potential cumulative effects of repeated injuries [6] .

Concussions do not result in gross vascular damage or identifiable tissue pathologies that can be seen with standard structural neuroimaging techniques [3]. Current methods of diagnosing sports-related concussions rely heavily on self-reporting by the injured player. Unfortunately, studies have shown that athletes may be unwilling to report symptoms due to the possibility of lost playing time and that as many as 50% of sports-related concussions may go undiagnosed [7,8]. Further confounding the complexities associated with identifying this injury, concussions cause a wide variety of physical, cognitive, and emotional symptoms that vary with individual injury. Up until recently an MTBI was thought to only have occurred if the individual lost consciousness, but it has been established that up to 90% of concussions do not result in loss of consciousness (LOC) [6,9]. The symptoms range from very obvious signs of injury such as poor coordination, slurred speech and LOC to less obvious and more subjective symptoms including headaches, dizziness, and fatigue [10]. The varying signs and symptoms associated with these injuries has led to the proposal that the current diagnosis of "concussion" actually represents a spectrum of very different injuries [11].

In the majority of individuals that have sustained a concussion, symptoms appear rapidly after the causal incident, are short-lived, and resolve spontaneously [3]. Yet some individuals experience a delayed onset of symptoms in the day or days following the concussive event [11]. Others are diagnosed with post-concussion syndrome when their acute symptoms associated with concussion injuries become chronic, lasting longer than three months [12,13]. Distinguishing between mild concussions that will resolve quickly and more severe injuries that may have long term effects is challenging.

Early detection and accurate diagnosis is critical to the management of MTBI. It has been reported that subsequent impacts following an initial concussion are 4-6 times more likely to result in a second, often more severe brain injury [14]. Other studies have shown that individuals who have sustained a concussion are 3-5 times more likely to sustain another concussion that those with no history of the injury [15]. It has also been reported that there is an increased risk for a repeat MTBI within the first 10 days after an initial concussion [16]. There has been growing concern about the possible effects of repeated concussive and sub-concussive impacts, or impacts that do not result in symptoms, leading to long term cognitive deficits [17–20]. Recent imaging studies have shown that contact sport athletes without symptoms or diagnosis of concussion exhibit white matter alterations and metabolic changes in their brains after a single season of play [19,20].

Epidemiological evidence also suggests that there may be individuals who are predisposed to concussion injuries depending upon risk factors such as sex, age, genetics, or history of MTBI, but the exact relationship between these factors and MTBI is unknown [21].

1.2.2 Sex Differences in MTBI Incidence and Outcome

Female participation in contact sports has been steadily increasing in the United States since the inception of Title IX in 1972. Epidemiological evidence suggests that female athletes may be at a greater risk of concussion than their male counterparts [22,23]. Research conducted in sports where males and females participate at the same level, such as collegiate ice hockey, has shown that female athletes sustain concussions at a higher rate (0.82/1000 athletic exposures (AEs)) than males (0.72/1000 AEs). It has also been reported that females experience a greater number of symptoms associated with concussion than males and that these symptoms are often more severe [24–26]. Additionally, recent studies have indicated that females may be at a greater risk of post-concussive syndrome that males [27].

Several sex-specific characteristics have been proposed as the rationale for these differences in MTBI incidence and outcome including both physiological and psychological differences [23]. It has been postulated that females may be at a higher risk of concussion because their smaller size and weaker neck muscles may lead to higher accelerations of the head after impact [28]. Studies have identified that hormonal differences between males and females affect clinical outcomes after concussion and traumatic brain injury [29,30]. Interestingly, these studies note that hormone regulation is based on extent of damage, so outcomes may differ by biomechanical input with hormonal factors playing a role only in response to the injury. It has also been proposed that cultural tendencies may influence females to be more honest than males in reporting injuries [23]. If females are more likely to report their symptoms than males, and diagnosis of concussion is largely depends on a patient self-reporting symptoms, currently available MTBI incidence data could be biased [8,31].

Consideration should also be given to rule variations between men's and women's sports. For example, checking or purposeful body contact of an opposing player is an important part of the game in men's ice hockey whereas it is illegal in women's. It has been postulated that due to the aggressive nature and faster pace of the game in men's ice hockey, males should be at a greater risk of concussion than females [28,31]. Accordingly, previous studies have reported that male helmeted athletes generally have a higher frequency of head impacts and experience impacts that are greater in magnitude than females [32]. These findings led to the proposal that females may be less prepared when contact occurs, which is important to consider when it has also been reported that anticipated collisions tend to result in less severe head impacts than unanticipated [32,33]. It has also been speculated that because females experience head impacts less often, they may have a heightened awareness of them when they occur [32]. Interestingly, there are studies that have shown that athletes that play positions that tend to sustain less contact, for instance a catcher in baseball, sustain concussions at a biomechanical input that is lower than suggested injury thresholds [34]. It may be that for sports or player positions where individuals have a high exposure to head impacts, players that have a lower tolerance to head impacts are "weeded out" or quit playing by the time they get to the collegiate level. This could serve as a self selection of what might be more tolerant athletes in sports with high impact exposure than in sports where there is less contact.

In order to understand what role these factors play in the differences between males and females in concussion incidence and outcome, there needs to be a better understanding of the mechanism of concussion injury and how it differs between the sexes.

1.2.3 Head Acceleration and Mild Traumatic Brain Injury (MTBI)

Concussions are usually caused by an acceleration of the head after a direct or indirect impact to the head or body [3]. The relationship between force, mass, and linear acceleration of a head impact can be described by Newton's Second Law of Motion (**Equation 1.1**).

$$F = ma$$

where:
F = magnitude of the force applied to the head
m = mass of head
a = linear acceleration of the head

Equation 1.1: Newton's Second Law of Motion: Linear Acceleration

A simple conceptualization is that as the magnitude force from a striking player increases, the head acceleration of the struck player increases because the mass of the head is a constant [35]. Analogously, rotational acceleration, or rotation around a fixed point, can also be computed according to this law (**Equation 1.2**).

$$\tau = I\alpha$$

where:
$\tau =$ total torque exerted on head
$I =$ mass moment of inertia
$\alpha =$ angular acceleration

Equation 1.2: Newton's Second Law of Motion: Rotational Acceleration

While linear and rotational acceleration are both hypothesized to be factors in MTBI, the scientific community has debated whether they contribute equally or one plays a bigger role in concussion injuries [36–39]. It is also not understood at what threshold of linear or rotational acceleration, if one exists, concussion injuries occur. Until recently, most of the biomechanical data on head injury thresholds have been extrapolated from animal models, cadaver experiments, or laboratory reconstructions. Early studies of the biomechanics of head impact used animal models to describe movement of the brain within the skull after impact and examined linear and rotational acceleration mechanisms of injury [37,40–42]. These studies characterized severe traumatic brain injury, where gross vascular damage was present and straightforward to assess. Several brain injury tolerance criteria for severe injuries have been developed based on these studies including The Wayne State Tolerance Curve (WSTC). The WSTC is a tolerance curve for skull fracture to linear acceleration and time duration [43,44]. Two important measures in the standardization of protective devices for head injury, the Head Injury Criterion (HIC_{15})[45] and the Gadd Severity Index (GSI)[46] are based on the WSTC.

$$HIC_{15} = \left\{ \left[\frac{1}{t_2 - t_1} \int_{t1}^{t2} a(t)\, dt \right]^{2.5} (t_2 - t_1) \right\}_{max}$$

where:
a = linear acceleration of the head
t_1 = initial time
t_2 = final time
*Note: the maximum time duration for HIC_{15}, (t_2-t_1), is limited to 15ms

Equation 1.3: Head Injury Criterion

$$GSI = \int_{t_1}^{t_2} a(t)^{2.5}\, dt$$

where:
a = linear acceleration of the head

Equation 1.4: Gadd Severity Index

HIC is commonly used to assess safety related to motor vehicles while GSI is used in setting safety standards on sports equipment. The current National Operating Committee on Standards for Athletic Equipment (NOCSAE) standard for football helmets is a GSI of less than 1200 [47]. For a specific criterion value of HIC or GSI, which consider both linear acceleration and duration of the acceleration (**Equations 1.3, 1.4**) large accelerations can be tolerated for short periods of time. More recent studies using cadavers have used high-speed bi-plane x-ray to examine brain displacement and deformation during head impacts [48,49]. This work showed that as linear and rotational head kinematics increased, the intracranial pressure gradients and brain motion relative to the skull increased. While these studies have established that head acceleration is the injury mechanism in severe traumatic brain injury [41,50] and it is hypothesized to be the mechanism for MTBI, the precise relationship of head acceleration and MTBI remains unknown. Extrapolating data from these studies on severe head brain injury using animal models and cadavers to concussion injury in humans is difficult and has led to several studies using surrogate models and laboratory constructions.

A 2003 study by Pellman et al. [36,51] quantified impact magnitude and location for head impacts that were associated with diagnosis of concussion for 31 professional football players by using video reconstruction to drive laboratory simulations. While this study provided important information on the biomechanics of concussion injury, the data were limited to a low number of impacts for a relatively small number of NFL players and required extensive reconstruction using a standard crash test dummy (Hybrid III). Limitations also exist in these surrogate studies in that sports-related concussions are usually diagnosed by the self-reporting of symptoms which cannot be easily deduced from these models; additional variables that influence kinematic response to impact such as contact force, equipment, or player anticipation are not accounted for; and an athlete's cumulative exposure to head impacts, which may alter an athlete's tolerance to injury, is not considered [52]. In order to better understand the relationship between head acceleration and concussion injury, it is necessary to expand the research on this subject beyond laboratory reconstructions, animal models, and cadaver experiments to *in vivo* studies of the biomechanics of sub-concussive and concussive head impacts.

There have been several efforts to measure head impacts in contact sports dating back to the 1970's [53–55]. These studies were limited by the technology available at the time, and captured data from a limited number of athletes per session with obtrusive instrumentation that interfere with normal play. In order to study the biomechanics of head impacts associated with concussion, it is necessary to monitor a large number of players and impacts because of the relatively low incidence of MTBI per exposure and the fact that etiology may depend on several biomechanical factors [51,56–58].

1.2.4 Head Impact Telemetry (HIT) System

The HIT System (Simbex, Lebanon, NH), an accelerometer-based impact monitoring system mounted inside of sports helmets (**Figure 1.1**), was developed through an NIH SBIR award (2R44HD40473) specifically for the purpose of enabling widespread *in vivo* biomechanical head impact data collection during all practice and game situations [56,59–61]. The HIT system was designed and validated to measure head acceleration, not helmet acceleration. This system directly measures and records the magnitude of head acceleration and location of impact on the helmet for all head impacts that individual players sustain during practices and games without interfering with normal play. The playing field offers a unique laboratory for studying MTBI because of the high frequency of head impacts, the large population at risk of injury, on site medical personnel, access to matched controls and access to athletes for serial testing (including baseline and post injury follow-up)[62].

The HIT system consists of an array of 6 single axis accelerometers (Analog Devices, Inc., Cambridge, MA, range ± 250*g*) mounted on the inside of sports helmets (**Figure 1.1A and 1.1B**). These sensors are elastically mounted within the liners of helmets to maintain contact with the head and decouple shell vibrations [63,64]. The HIT System is available for football (Riddell VSR-4, Revolution, or Speed models (Riddell, Chicago IL)) and hockey (S9 Easton (Van Nuys, CA) or CCM Vector (Reebok-CCM

Hockey, Inc., Montreal, Canada)) helmets, two sports that have been reported to have some of the highest rates of concussion for all collegiate sports [65]. The system samples at a frequency of 1000 Hz and data recording is only triggered if one of the six accelerometers exceeds a threshold of 14.4*g*. For each head impact, forty milliseconds of acceleration data for all six channels are collected; 8 seconds of data are recorded pre-trigger and 32 seconds are recorded post trigger. Accelerometer data is transmitted wirelessly to a sideline controller (**Figure 1.1C and 1.1D**) via radiofrequency telemetry. Instrumented helmets have the ability to store acceleration data for up to 100 impacts if signal interference with the sideline controller occurs.

All data are sent to a secure server for post processing. Data are reduced in post processing to exclude any impact event that results in a peak linear acceleration of less than 10*g*. This is to eliminate events that that have been determined to be inconsequential or non-impact events, such as running or jumping. Any events in which the data from the 6 accelerometer traces did not match the theoretical pattern for rigid body head acceleration (i.e. dropping or throwing of a helmet) are also excluded. Resultant linear and rotational acceleration of the head, duration of impact, and the impact location on the helmet are calculated for each impact [56]. The algorithm used in the football system (**Equation 1.5**) was developed to directly measure linear acceleration. Peak rotational acceleration is then estimated from the linear acceleration vector and an assumed point of rotational 10 cm inferior to the center of gravity (COG) of the head [66].

$$\sum_{i=1}^{n}(\|H\|(\cos\alpha_i \cos\alpha_H \ \cos(\theta_i - \theta_H) + \ \sin\alpha_i \sin\alpha_H) - \|\mathbf{a}_i\|)^2$$

where:
sensing axis of accelerometers is normal to the surface of the head
i = number of accelerometers
$\|H\|$ = acceleration of the head (at center of gravity of the head modeled as center of sphere)
α_i = elevation angle (accelerometer$_i$)
α_H= elevation angle (impact location)
θ_i= azimuth angle (accelerometer$_i$)
θ_H= azimuth angle (impact location)
$\|\mathbf{a}_i\|$ = acceleration magnitude at accelerometer$_i$ (location (θ_i , α_i))

Equation 1.5: Algorithm used in football HIT System. Acceleration of the center of gravity of the head (||H||) and impact location (θ_H,α_H) are calculated by minimizing least-square error.

For the development of the hockey system, this algorithm was expanded to include direct calculation of rotational acceleration by reorienting the linear accelerometers to be mounted with their sensing axis tangential to the head [61]. By assuming rigid body dynamics and a minimal contribution of centripetal acceleration (because the accelerometers are positioned tangentially to the head), linear acceleration and rotational acceleration are directly calculated by minimizing the sum of the square error between each accelerometer value and the expected acceleration (**Equation 1.6**).

$$min\sum_{i=1}^{n}\left[\|a_i\| - \vec{r}_{a_i} \cdot \overrightarrow{H}_E + \vec{r}_{a_i} \cdot (\overrightarrow{\alpha_E} \times \vec{r_i})\right]^2$$

where:
sensing axis of accelerometers is tangential to the surface of the head
n = number of accelerometers
a_i = acceleration at any point i on the head
$\vec{H}$ = linear acceleration vector at the head center of gravity
$\vec{r}_{a_i}$= acceleration on sensing axis of the accelerometer
$\vec{r}_i$= position vector of point i relative to the center of gravity of the head
$\vec{\alpha}_E$ = estimated rotational acceleration vector

Equation 1.6: HIT System Hockey Algorithm is based on this Equation

Additionally, a non-dimensional measure of head impact severity, HITsp (**Equation 1.7**) is computed. HITsp transforms the computed head impact measures of peak linear acceleration and peak angular acceleration into a single variable using Principal Component Analysis, and applies a weighting factor based on impact location [38]. HITsp has shown to be more predictive of diagnosed concussions than linear acceleration, rotational acceleration, or HIC alone [38].

$$HIT_{sp} = 10 * L_{wf}\ ([0.478 * sGSI + 0.4742 * sHIC + 0.4336 * sLIN + 0.2164 * sROT] + 2)$$

where:
$sX = \left(X - \frac{mean(X)}{SD(X)}\right)$
GSI = Gadd Severity Index (see **Equation 1.4**)
HIC = Head Injury Criterion (see **Equation 1.3**)
LIN = Peak Linear Acceleration
ROT = Peak Rotational Acceleration
Location weighting factor (L_{WF}) coefficients:
Side = 1.00, Front = 0.95, Top = 0.62, Back = 0.48

Equation 1.7: HITsp Equation. HITsp is scaled by 10 and offset by 2 in order to obtain a score greater than zero and in the range of other classical measures of head impact severity.

Impact locations to the helmet are computed as azimuth and elevation angles in an anatomical coordinate system relative to the center of gravity of the head [56] and then categorized as front, side (left and right), back, and top (**Figure 1.1C**). Front, left, right and back impact locations are four equally spaced regions centered on the mid-sagittal plane. All impacts above an elevation angle of 65° from a horizontal plane through the CG of the head are defined as impacts to the top of the helmet.

Validation of the system has been completed in both laboratory testing and on the field. Laboratory testing was conducted at Brown University using an ASTM standard linear drop system, at Wayne State University using a linear impact ram, and at Riddell, Inc. (Evanston, IL) using a pendulum type impactor. Standard NOCSAE testing was performed at the Southern Impact Research Center (Knoxville, Tennessee). Helmet to helmet laboratory tests at Virginia Tech have demonstrated that the HIT System measures head acceleration, not helmet acceleration [67]. In these laboratory tests, the HIT System data was also found to be highly correlated with data from a Hybrid III, a standard crash test dummy, ($r^2 > 0.92$) for both linear impact and pendulum test [67,68]. During the 2003-2004 Virginia Tech football seasons, an on-field study was conducted to validate the HIT system for on-field application [59]. For the hockey system, on-ice accuracy has been corroborated through video review across several independent research institutes [33,69,70].

Several early studies that utilized the HIT System in football and ice hockey reported the number of impacts and the magnitude of the resulting head accelerations aggregated within teams and across different levels of play, but did not examine the relationships between these measures of exposure [62,71–78]. A multi-factorial measure of exposure is critical because the specific biomechanical variable or combination of variables and factors such as player position, session type (game injury rates are reported to be higher than in practices), etc. that correlate with risk of injury is unknown [79]. In our approach to understanding the biomechanical basis of MTBI, we have defined *head impact exposure* as a multi-factorial term that includes the frequency, magnitude, and impact location of all head impacts for individual athletes [80].

Other studies have utilized the HIT system to evaluate the biomechanics of single head impacts associated with diagnosed concussions. Broglio et al. postulated that if the appropriate biomechanical variables of concussion were identified then concussed players could be immediately recognized and removed from the field of play [35]. Unfortunately, a biomechanical threshold for MTBI has been difficult to quantify because of the difficulty of identifying the injury and quantifying symptoms; the varying magnitudes and locations of impacts that result in MTBI; as well as other factors including frequency of sub-concussive impacts, history of prior concussions, and underlying variation in tolerance to this injury from one person to the next [76,81].

While a single biomechanical injury threshold may be difficult to establish, the relationship between head impact exposure and concussion incidence can be discerned using tools such as the HIT System. A recent study in high school and collegiate football players reported that individual players sustained more head impacts and impacts of greater severity on days of diagnosed concussion than on days without diagnosed concussion [82]. This indicates that a relationship exists between head impact exposure and concussion injuries, and has led to the proposal that reducing an individual's head impact exposure is a practical approach for reducing the risk of brain injuries [83].

Currently, rate of injury is expressed in athletic exposure (AE), where exposure is defined as the opportunity, regardless of the degree, to participate [84,85]. While AE is a useful metric to evaluate overall risk of injury as a function of participation (i.e. one participation equals one AE), it does not account for direct exposures, such as the number of head impacts received in an event. Utilizing tools such as the HIT System allows us to quantify an individual's direct exposure to head impacts and correlate it with clinical outcome measures associated with MTBI.

With the many difficulties and complexities associated with the recognition and diagnosis of MTBI, there is a clear need for the development and validation of on-field diagnostic tools. This, along with concerns over the acute and chronic impairments resulting from concussions, has led to the development of a variety of strategies to enhance immediate diagnosis. These diagnostic tools range from sideline cognitive tests to neurophysiologic measures. Biomechanical data from the HIT system can be integrated with existing and newly developed clinical outcome measures and diagnostic tools for MTBI. There is potential for these tools to not only be an on-field biomarker for the diagnosis of concussion, but also a tool for identifying players who have sustained repeated sub-concussive impacts and may be at risk of injury or long term cognitive deficits. The efficacy and feasibility of these tools and other clinical outcome measures related to MTBI could be evaluated by studying the relationship between these variables and head impact exposure.

Besides the ability to quantify head impact exposure individual athletes, examining the biomechanics of impacts associated with MTBI, and evaluating the feasibility and efficacy of diagnostic tools, there are other potential areas in which data from the HIT system is useful in furthering our understanding of concussion injuries. Specifically, synchronizing data from the HIT system with video to evaluate the biomechanics of head impacts associated with different impact mechanisms could inform specific strategies for decreasing individual athletes' head impact exposure.

1.2.5 Mechanisms of Head Impact

The high rate of injury in contact sports, including concussions, can be attributed to some unique, sport-specific factors of each game. For example, in ice hockey, the playing area is made of solid ice and enveloped by rigid boards, players manipulate pucks that when shot can exceed 80 mph, players travel at high speeds of up to 30 mph and purposefully collide with opponents [69,86]. These unique factors allow for a number of different head impact mechanisms, or circumstances in which a head impact occurs (e.g. head contact with ice, boards, etc. in ice hockey). Specific head impact mechanisms and their resultant impact biomechanics have not been studied, but could improve the understanding of concussion injury and could inform prevention strategies.

Currently, there is a lack of data quantifying the biomechanics of head impacts as a function of the different impact mechanisms that occur in ice hockey. Previous studies have reported injury epidemiology, including diagnosed concussions, by specific injury mechanisms in collegiate ice hockey [84,86–89]. Agel et al. utilized the NCAA Injury Surveillance System (ISS) to report concussion mechanisms in men's and women's collegiate games [87,88]. Diagnosed concussions were classified into one of seven concussion mechanisms that included: contact with another player, contact with the ice surface, contact with the boards or glass, contact with the goal, contact with the stick, contact with the puck and no apparent contact. Another study classified injury mechanisms in professional players in the National Hockey League (NHL) by reviewing video footage from games in which diagnosed concussions occurred [90]. The most common mechanism that resulted in diagnosed concussions for both studies was player to player contact [87,88,90]. While these studies provided important information on injury and concussion mechanisms in ice hockey, the collection and analysis of the impact biomechanics that result from these mechanisms was beyond the scope of their study designs.

One of the most reliable and effective records of how an injury occurs is the use of archived video imagining [91]. Use of video replay is practical in that it affords the clinician or researcher the opportunity to thoroughly document and observed events leading up to impact or injury. Synchronizing video with the biomechanics of head impacts would provide a quantitative approach to evaluating head impact mechanisms and the biomechanics of the resultant impacts. This would provide data that could inform plans for reducing head impact exposure and strategies for injury prevention.

1.3 Significance

The collection of *in vivo* biomechanical head impact data, quantification of head impact exposure for individual athletes, examination of head impact mechanisms, and evaluation of impacts associated with diagnosed concussion will provide a better understanding of the complex injury of MTBI. These data will also aid in the understanding of the acute and long-term effects of repeated concussive and sub-concussive impacts. Furthermore, these data will contribute to a better understanding of the differences between males and females in concussion incidence and clinical outcome. The outcomes of this study may allow for the development and validation of improved on-field MTBI management (including early detection and accurate diagnosis), prevention strategies, and interventions. Additionally, data from this study may inform and improve future helmet design.

1.4 Specific Aims

1.4.1 Specific Aim 1

Quantify and compare the biomechanics of head impact exposures in men's and women's collegiate ice hockey players.

The first objective was to quantify and compare <u>head impact exposure</u>, a multi-factorial term that includes frequency, magnitude, and impact location of all head impacts for individual players, in men's and women's collegiate ice hockey players. We collected head impact data using the HIT System on ninety-nine (41 male and 58 female) players from two men's and two women's National Collegiate Athletic Association (NCAA) hockey programs over the course of 3 seasons of play. We quantified the head impact exposure for individual athletes and investigated the differences in exposure by sex, player position, session type, and team. We hypothesized that males would have significantly higher head impact exposures than females. We also made the null hypothesis that head impact exposure would not differ by session type, position, or team.

1.4.2 Specific Aim 2

Examine and compare head impact mechanisms in men's and women's collegiate ice hockey players.

The objective of the second aim was to examine and compare mechanisms of head impact (e.g. contact with the ice, contact with the boards, etc.) in men's and women's collegiate ice hockey players. To address this aim, video footage from 53 games was synchronized with biomechanical data from head impacts collected with the HIT System. Head impacts were classified into 8 head impact mechanism categories: contact with another player, the ice, boards or glass, stick, puck, goal, indirect contact, and contact from celebrating. We hypothesized that the biomechanical measures of head impacts would differ among impact mechanisms. We also hypothesized that head impact mechanisms and the resulting impact biomechanical measures would differ between male and female players.

1.4.3 Specific Aim 3

Evaluate differences between males and females in concussion rate and the biomechanics of head impacts associated with diagnosis of concussion.

The first objective was to evaluate differences in concussion rates between male and female athletes. Injury rate was calculated as rate per athletic exposure and rate per head impact. We hypothesized that females would have a significantly higher incidence of concussion than their male counterparts.

The second objective of the third aim was to evaluate differences between males and females in single impact severity measures (peak linear acceleration, peak rotational acceleration, and HITsp) of impacts associated with MTBI and head impact exposure on days of diagnosed concussion. We hypothesized that sex would be a significant factor in the biomechanics of single head impacts associated with concussion and head impact exposure on days of diagnosed concussion.

Each chapter in the body of this thesis (Chapters 2-4) corresponds to and accomplishes these specific aims. Chapter 2 focuses on head impact exposure in men's and women's collegiate ice hockey players. The next study, presented in Chapter 3, focused on examining the per-game frequency and magnitude of head impacts associated with various impact mechanisms (head contact with ice, boards, etc.), in men's and women's collegiate ice hockey players. Chapter 4 focuses the differences between males and females in concussion rate and the biomechanics of head impacts associated with diagnosed concussion. Findings are synthesized in Chapter 5. Related studies and future directions that further address the aims are presented in Appendices A-E.

1.5 References

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Figure 1.1: (A) Football and (B) hockey helmets instrumented with HIT System encoder consisting of six accelerometers (a), telemetry electronics (e), and a battery (b). (C) The HIT System is comprised of instrumented helmets, a sideline receiver, and a laptop computer. (D) The hockey sideline receiver in use during a game.

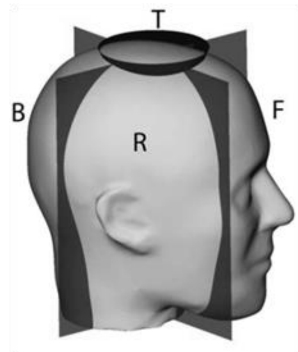

Figure 1.2: An illustration of regions that define the front (F), right side (R), back (B), and top (T) impact locations on the helmet and facemask. Figure by J.J. Crisco.

Chapter 2

Head Impact Exposure in Male and Female Collegiate Ice Hockey Players

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2.1 Abstract

The purpose of this study was to quantify head impact exposure (frequency, location and magnitude of head impacts) for individual male and female collegiate ice hockey players and to investigate differences in exposure by sex, player position, session type, and team. Ninety-nine (41 male, 58 female) players were enrolled and 37,411 impacts were recorded over three seasons. Frequency of impacts varied significantly by sex (males: 287 per season, females: 170, $p<0.001$) and helmet impact location ($p<0.001$), but not by player position ($p=0.088$). Head impact frequency also varied by session type; both male and female players sustained more impacts in games than in practices ($p<0.001$), however the magnitude of impacts did not differ between session types. There was no difference in 95th percentile peak linear acceleration between sexes (males: 41.6g, females: 40.8g), but 95th percentile peak rotational acceleration and HITsp (a composite severity measure) were greater for males than females (4424, 3409 rad/s^2, and 25.6, 22.3, respectively). Impacts to the back of the helmet resulted in the greatest 95th percentile peak linear accelerations for males (45.2g) and females (50.4g), while impacts to the side and back of the head were associated with the greatest 95th percentile peak rotational accelerations (males: 4719, 4256 rad/sec^2, females: 3567, 3784 rad/sec^2 respectively). It has been proposed that reducing an individual's head impact exposure is a practical approach for reducing the risk of brain injuries. Strategies to decrease an individual athlete's exposure need to be sport and gender specific, with considerations for team and session type.

2.2 Introduction

Sports related concussions are a growing public health problem that affects millions of individuals in the United States [1]. Of particular concern are athletes who participate in contact sports, who are not only at risk for multiple concussions whose cumulative effects are not know [2], but who are also exposed to repetitive head impacts, which have been suggested as a possible cause of chronic brain injury [3–5]. Female participation in contact sports has been steadily increasing in the United States since the inception of Title IX in 1972. Research conducted in contact sports where male and females participate at the same level, such as ice hockey, has shown that females are at a higher risk of concussion (0.82/1000 athletic exposures (AE)) than their male counterparts (0.72/1000 AE) [6], but the reasons for this are not well understood.

It has been accepted that the mechanism of concussion is related to accelerations of the head after a direct or indirect impact to the head or body [7], but the exact relationship between the biomechanics of head impacts and clinical outcome is unknown [8]. While the kinematics of head impacts associated with injury are important to understand, it has been proposed that it is equally important to examine impacts that are not associated with diagnosis of concussion [9]. Evaluating the biomechanics of all impacts may lead to a better understanding of the relative risk of head impacts, while also allowing for the evaluation of the relationship between repetitive impacts and long term cognitive deficits.

Several recent studies have focused on understanding the biomechanics of head impacts sustained in contact sports by utilizing an accelerometer-based head impact monitoring device, the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH). The HIT System allows researchers to monitor and record head impacts sustained by individual helmeted athletes during play. Utilizing this system and the unique laboratory that the playing field provides, researchers have directly measured the frequency, magnitude, and location of head impacts in a variety of sports (boxing, soccer, skiing), however the majority of studies have focused on football and ice hockey [10–15]. In our approach to understanding the biomechanics of concussions, we have used data collected by the HIT System to quantify head impact exposure, a multifactorial term that includes the frequency, magnitude, and impact location of head impacts for individual athletes. Previously, we have quantified and reported head impact exposure by specific player positions in collegiate football players [16]. In a subsequent study, where this analysis was expanded to impacts associated with diagnosed concussions, we found that a relationship exists between head impact exposure and diagnosis of concussion [17]. Based on these findings, we have proposed that reducing an individual's head impact exposure is a practical approach for reducing the risk of brain injuries [18].

Considering the high rate of concussions in ice hockey, the relative youth of its players, and the fact that both males and females participate, the expansion to and application of our previously used methods for quantifying head impact exposure to hockey is warranted. The only previous study of collegiate hockey players compared distributions of head impact exposure between sexes and reported that males experience a higher number of impacts than females and also sustain head impacts greater in magnitude [19]. While the study provided valuable insights into sex differences in the biomechanics of head impacts sustained in collegiate ice hockey, it did not provide a player-specific, detailed analysis of the exposure to all head impacts for *individual* players by sex, position, session type, or team.

The aim of this study was to quantify the frequency, magnitude, and location on the helmet of all head impacts sustained by individual collegiate male and female ice hockey players. Specifically, we tested the hypothesis that male hockey players would have a higher frequency of head impacts and would sustain head impacts that resulted in greater magnitudes than female players. We also tested the null hypotheses that head impact frequency, location, and magnitude sustained by individual athletes would not differ by player position, session type, or team.

2.3 Methods

Ninety-nine (41 male and 58 female) players from two men's and two women's National Collegiate Athletic Association (NCAA) hockey programs (Brown University and Dartmouth College, teams denoted arbitrarily as M1, M2 for males and F1, F2 for females) participated in this observational study after informed consent was obtained with institutional review board approval. Teams M1, F1, and F2 participated during the 2009-2010, 2010-2011, and 2011-2012 hockey seasons, while team M2 participated in a single season (2010-2011). Thirty males and 19 females were monitored during one season, 5 males and 20 females during two seasons, and 6 males and 19 females during three seasons. Players were categorized into one of two positions, forward or defense. Goalies were not included in this study. Of the 41 male players, 16 were defenders and 25 were forwards. The 58 female players included 21 defenders and 37 forwards.

Players wore S9 Easton (Van Nuys, CA) or CCM Vector (Reebok-CCM Hockey, Inc., Montreal, Canada) helmets instrumented with the HIT System. The HIT System measures and records biomechanical data from head impacts including linear and rotational acceleration at the head center of gravity (CG) and impact location on the helmet. The instrumented helmets were equipped with six single-axis accelerometers arranged tangentially to the head and mounted elastically within the helmet's foam liner to maintain contact with the head and decouple shell vibrations [15,19,20]. The system collects acceleration data at 1 kHz, time stamps and stores the data on the helmet. Data are then transmitted by radiofrequency telemetry to a computer and entered into a secure database. System design, validation, accuracy, and data reduction methods have been previously described in detail [14,15,21–26].

Head impact exposure, including frequency of head impacts, magnitude of head impacts and impact location on the helmet for individual players, was quantified. This was accomplished using previously established methods used to quantify head impact exposure in collegiate football players [12,13,16]. A session was defined as either a practice or a game. An individual participated in a session when the player was present and partook in a game or practice, regardless of whether they sustained an impact during that particular session. Practices were sessions where players wore protective equipment with the potential of head contact. Game sessions included both competitions and scrimmages. Five measures of impact frequency were computed for each player: practice impacts, game impacts, impacts per season, impacts per practice, and impacts per game. Practice impacts and game impacts are the total number of head impacts for a player during all practices and all games, respectively. To calculate the number of impacts per game and per practice, the frequency of impacts players received was normalized by the number of sessions the player participated in. This accounted for differences in schedules and player attendance. Impacts per season, per game, and per practice are the *average* number of head impacts for a player during all sessions in a single season, during all games, and during all practices, respectively.

Impact magnitude variables included peak linear acceleration (g), peak rotational acceleration (rad/s^2), and HITsp. HITsp is a composite measure of head impact severity that includes linear and rotational acceleration, impact duration, and impact location [27]. Each individual player's distribution of peak linear acceleration (g), peak rotational acceleration (rad/s^2), and HITsp were quantified by the 50^{th} and 95^{th} percentile value of all seasonal impacts. Additionally, impacts were further reduced for analysis by computing the 50^{th} and 95^{th} percentile value of all seasonal impacts at each location.

Impact location variables were computed as azimuth and elevation angles relative to the center of gravity of the head [21] and then categorized as front, side (left and right), back, and top. Four equally spaced regions centered on the mid-sagittal plane make up the front, left, right and back locations. Impacts to the top of the head were defined as all impacts above an elevation angle of 65° from a horizontal plane through the CG of the head [27].

Statistical Analysis

Results were expressed as median values and [25-75% interquartile range] because study variables were not normally distributed (Shapiro-Wilk test; $P < 0.05$). Differences in impacts per season among team and sex were examined separately using a Kruskal-Wallis one-way ANOVA on ranks with a Dunn's post-hoc test for all pairwise comparisons. The significance of the differences in sex and player positions in impact frequency (impacts per practice, impacts per game, and impacts per season) and in severity measures (50^{th} and 95^{th} percentile peak linear and rotational acceleration, and HITsp) were examined using a two-way ANOVA with a Holm-Sidak post-hoc test for all pairwise comparisons. Statistical significance was set at $\alpha = 0.05$ and the reported *p*-values are those for the post hoc test. An identical approach was used to examine the significance of the differences among sex and player positions in frequency and severity measures at each location. Statistical comparison among impact location were performed with a Friedman repeated measures ANOVA on ranks. All statistical analyses were performed using SigmaPlot 12.0 (Systat Software, Chicago, IL).

2.4 Results

2.4.1 Overall Impact Distributions

A total of 37,411 head impacts were analyzed in this study with 19,880 impacts sustained by males and 17,531 by females. These data were collected during a player median of 109 [96-113] practices and 36 [27-43] games for males and a player median of 142.5 [77.5-174] practices and 53.5 [32-68] games for females. Distributions of the magnitudes of all impacts by peak linear acceleration, rotational acceleration, and HITsp were skewed towards lower values (**Figure 2.1**). The total number of impacts received by an individual male player during a single season was a median of 287 [200-446] with a maximum of 785 and females received a median of 170 [116-230] and a maximum of 489. The percentages of players receiving any given number of season impacts are plotted by sex and by team using cumulative histograms (**Figure 2.2**). The number of season impacts for male players was significantly higher than for female players ($p<0.001$) (**Figure 2.2A**). The number of season impacts for players on Team M2 was higher than the number of season impacts for players on team M1, F1, and F2 ($p<0.001$)(**Figure 2.2B**). After normalizing for differences in the number of sessions, there was no difference in the number of impacts per game or impacts per practice between M1 and M2 ($p>0.05$) or F1 and F2 ($p>0.05$).

2.4.2 Impact Frequency

Males experienced a significantly higher number of impacts per game 6.3 [3.5-9.0] than impacts per practice 1.3 [1.0-1.7] ($p<0.001$). Similarly, females had a significantly higher number of impacts per game 3.7 [2.5-4.9] than impacts per practice 0.9 [0.6-1.0] ($p<0.001$). Males experienced a significantly higher number of impacts per season, impacts per practice, and impacts per game than females (**Table 2.1**). Across all players and within sex, there were no statistically significant differences in the frequency of impacts per game or per practice between forwards and defenders (**Figure 2.3**).

2.4.3 Impact Magnitude

When compared to females, males were found to sustain impacts with greater 50^{th} percentile peak linear and peak rotational acceleration, as well as HITsp (**Table 2.2**). Males also sustained impacts with greater peak 95^{th} rotational acceleration and HITsp, but the trend in differences in peak linear acceleration did not reach significance. Position was not a factor in magnitude of head impacts for males or females (**Table 2.3**). There were no increases from practices to games in head impact magnitude regardless of sex or position (**Figure 2.3**).

2.4.4 Impact Location

For both male and female players, frequency and magnitude of head impacts varied by impact location on the helmet. The lowest percent of impacts occurred to the top of the helmet ($p<0.001$)(**Table 2.4**). There were no statistically significant differences between males and females in the frequency of impacts to different locations on the helmet (*p*-values ranging from 0.31-0.87).

Location was found to be a factor in impact magnitude for both male and female players. Males experienced greater 95th percentile peak linear acceleration from impacts to the back of the helmet than impacts to the front or side ($p=0.002$) (**Figure 2.4A**). Males also experienced impacts to the side of the helmet that were greater in 95th percentile peak rotational acceleration when compared to impacts to the front ($p=0.002$). For females, impacts to the back and top of the helmet were greater in 95th peak linear acceleration than impacts to the front and side ($p<0.001$) (**Figure 2.4B**). Females also experienced greater 95th percentile peak rotational acceleration from impacts to the side and back of the helmet than impacts to the top or front ($p<0.001$). Impacts to the front of the head and side of the head were significantly greater in 95th percentile peak linear acceleration for males than females ($p=0.028$, $p<0.001$ respectively) (**Figure 2.4A**). Impacts to all four locations (front, side, top, back) were significantly greater in peak rotational acceleration for males compared to females (p=<0.001, $p<0.001$, $p=0.028$, $p=0.024$ respectively) (**Figure 2.4B**). Position did not play a factor in frequency or magnitude of impact location for males or females.

2.5 Discussion

The purpose of this study was to quantify head impact exposure in individual male and female collegiate ice hockey players and then examine the relationships between head impact frequency, location, and magnitude as a function of sex, player position, session type, and team.

Male players were found to have a higher frequency of head impacts per practice, per game and per season than female players. The difference in impact frequency between sexes can most likely be attributed to gender-specific rules in ice hockey. Checking, or purposeful body contact of an opposing player, is allowed in men's hockey whereas it is illegal in women's. In a previous study in the same subject population, we found that while head impacts that resulted from contact with another player occurred at a higher rate per game for males compared to females, contact with another player was the most frequent head impact mechanism for both sexes [26]. Similar to previous studies in collegiate football and youth hockey, both male and female players were found to have higher impacts per games than impacts per practice [13,15,16]. Accordingly, overall injury rates in collegiate ice hockey have been reported to be eight times higher in games than in practices for males and five times higher for females [28,29]. The number of impacts per practice, per game, and per season for male and female hockey players were considerably lower than those previously reported in collegiate football [13,16]. When considering the difference in head impact frequency between the two sports, it should be noted that, in this study, the number of head impacts were normalized by the number of sessions individual players participated in, regardless of whether they received a head impact or not. Crisco et al. normalized the frequency of head impacts by the number of sessions where a player sustained at least one impact [13,16]. For an individual athlete, the number of AE will be higher than or equal to the number of sessions where a single impact occurs.

Male hockey players were found to sustain head impacts that resulted in greater acceleration magnitudes than females. When one considers that impacts of greater magnitude have more associated risk for concussion and that females are reported to have a higher incidence of concussion, the data presented here may seem counterintuitive [6,17]. Several additional factors may contribute to the differences in biomechanics of head impacts and concussion incidence between males and female players in collegiate ice hockey, including physiological differences, psychological factors, and rule variations within the sport by gender. Mihalik et al. [14] reported an average linear acceleration in male youth hockey players of 18.3*g* for defensemen and 18.4*g* for forwards which was substantially higher than our median 50th percentile linear acceleration for male and female defensemen and forwards, which ranged from 15.0-15.8g. This discrepancy may be associated with differences in analysis; in that study average values were computed, while in the present study we computed the median value to account for the positively skewed, non-normally distributed data. While the 95th percentile peak rotational acceleration for male hockey players in this study, 4424 rad/sec^2, was comparable to the reported value of 4378 rad/sec^2 in collegiate football, both of these values are considerably greater than what we found in female hockey players, 3409 rad/sec^2. The 95th percentile peak linear acceleration, 62.7*g*, and HITsp, 32.6, reported in collegiate football are considerably greater than those reported in this study for both male and female hockey players. This suggests that that type of sport is likely a key factor in individual athlete's head impact exposure.

Impacts to the back of the head resulted in greater 95th percentile peak linear acceleration than impacts to the front, top, and side for both male and female players. This is consistent with previous findings in collegiate ice hockey that reported impacts to the back of the head had the greatest magnitudes [22]. Interestingly, epidemiological studies dating back to the 1960's that were used to develop the first ice helmets standard suggested that increased protection is needed for the back of the head due to the prevalence of falling backward and hitting the ice [30]. Similarly, in a previous study that synchronized data from the HIT system with game video, we found that head contact with the ice resulted in the greatest magnitudes [26]. While linear acceleration was highest for impacts to the back of the head, impacts to both the back and side of the head resulted in the highest rotational accelerations. This is similar to findings in a group of youth hockey players [14], and the authors postulated that impacts to the side of the head may elicit rotation of the head about the neck more easily than impacts to other locations. Interestingly, impact location, along with rule variations, may also explain why male hockey players were found to sustain head impacts that resulted in greater rotational, but not linear, acceleration magnitudes than females. It has been reported that males sustain a higher rate of impact with the boards than female athletes and these impacts result in higher rotational, but comparable linear, accelerations than other head impact mechanisms [26]. While linear and rotational acceleration have shown to be correlated; the data presented in this study along with previously reported collegiate football data, have shown this relationship is dependent upon impact location [12]. Anecdotally, for male players, we can speculate that the majority of head impacts that occur with the boards are a result of checking and are to the side of the head, a location associated with high rotational accelerations.

While player position was found to have the most significant effect on head impact exposure in collegiate football [16], we found no differences between player positions in collegiate ice hockey. This disparity may be attributed to the fact that player positions in football tend to be highly specialized with very different objectives. Similar to findings in previous studies in youth hockey [15], we observed no differences in frequency or magnitude of head impacts between defenders and forwards. There are conflicting data available on player position and injury rates in hockey; some studies have reported that forwards sustain higher overall injury and concussion rates than defenders, while others found no differences [9,28,29,31,32]. A limitation of this study is that we did not analyze the relationship between head impact exposure and the diagnosis of concussion.

This study has several additional limitations. There were a disproportionate number of females to males (58 vs. 41) included in this study, however we captured a relatively large number of head impacts and observed statistically significant differences, so it is unlikely that additional data collection would alter the results. There were also differences in the numbers of seasons in which each team participated. Data collection for team M1 was limited to a single season because the team accepted a new equipment contract after that season, and HIT System technology was not compatible with helmet models specified within the new contract. Additionally, laboratory validation of the hockey HIT System has demonstrated mean linear and rotational acceleration errors of 9 and 11%, respectively. The large distribution of data evaluated in this study addresses randomized single impact errors such as these.

In summary, we have shown that head impact exposure for collegiate ice hockey players is dependent upon sex, session type, and team but not on player position. Head impacts occur at a higher frequency in games than in practices but these impacts are not greater in magnitude. Impacts to the back of the helmet result in the greatest peak linear accelerations, while impacts to the side and back of the helmet are associated with high rotational accelerations. Male players have higher frequencies of head impacts and experience impacts greater in magnitude than females. Our findings also suggest, when compared to the literature, that head impact exposure for individual athletes is dependent upon which sport they play. It has been proposed that reducing an individual's head impact exposure is a practical approach for reducing the risk of brain injuries [18]. Strategies to decrease an individual athlete's exposure need to be sport and gender specific, with considerations for team and session type.

2.6 Acknowledgments

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	Impacts Per Season	Impacts Per Practice	Impacts Per Game
Male	287.0 [201.5-444.6]	1.3 [1.0-1.7]	6.3 [3.5-9.0]
Female	169.8 [119.0-230.0]	0.9 [0.6-1.0]	3.7 [2.5-4.9]
p-value	<0.001	<0.001	<0.001

Table 2.1: Median [25-75th percentile] impacts per season, per practice, per game for male and female players.

	50th Percentile			95th Percentile		
	Peak Linear Acceleration (g)	Peak Rotational Acceleration (rad/sec^2)	HITsp	Peak Linear Acceleration (g)	Peak Rotational Acceleration (rad/sec^2)	HITsp
Male	15.7 [14.8-17.1]	1630 [1454-1733]	13.6 [13.4-14.1]	41.6 [36.6-49.5]	4424 [4076-5182]	25.6 [22.7-29.5]
Female	15.0 [14.5-15.5]	1211 [1091-1353]	13.1 [12.9-13.6]	40.8 [36.5-49.9]	3409 [3152-3839]	22.3 [21.0-25.2]
p-value	0.007	<0.001	<0.001	0.366	<0.001	0.002

Table 2.2: Median [25-75th percentile] impacts per season, 50th and 95th percentile peak linear acceleration, peak rotational acceleration and HITsp for male and female players.

	Position	n	Peak Linear Acceleration (g)	Peak Rotational Acceleration (rad/sec^2)	HITsp
Male	D	16	38.5 [33.1-48.4]	4420 [3955-5115]	23.6 [22.0-29.0]
	F	25	43.4 [38.5-49.8]	4499 [4076-5221]	26.4 [24.0-29.6]
Female	D	21	40.6 [34.3-50.0]	3431 [3108-3786]	21.5 [20.9-22.7]
	F	37	40.9 [36.8-49.2]	3371 [3170-3883]	23.0 [21.3-25.5]

Table 2.3: Median [25-75th percentile] 95th percentile peak linear acceleration, peak rotational acceleration and HITsp for male and female players by position (D – defenders, F – forwards).

	Front (%)	Side (%)	Top (%)	Back (%)
Male	28.0 [21.2-33.5]	30.1 [26.5-36.4]	7.5 [5.3-10.4]	28.4 [23.2-37.1]
Female	29.3 [24.8-32.1]	27.9 [21.5-33.4]	9.1 [5.2-12.1]	31.3 [28.2-37.9]

Table 2.4: Impact location distribution, median [25-75th percentile] percentages of all head impacts to the front, side, top, and back of the helmet.

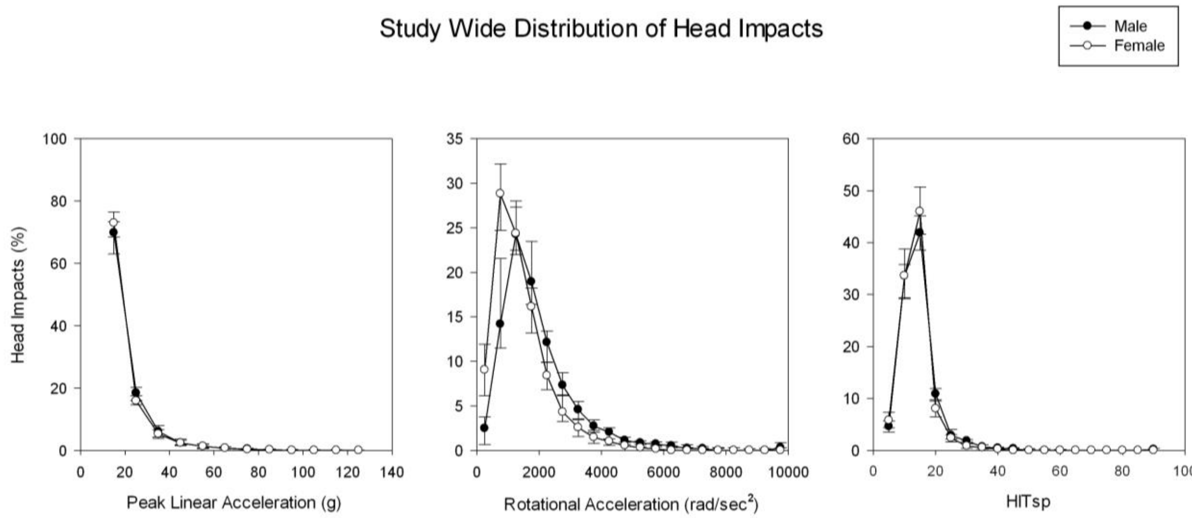

Figure 2.1: Study wide peak linear acceleration (g), peak rotational acceleration (rad/s^2) and HITsp distributions of head impacts. Data are a percentage of all impacts for individual players with median [25-75%] values plotted at each bin in the distribution.

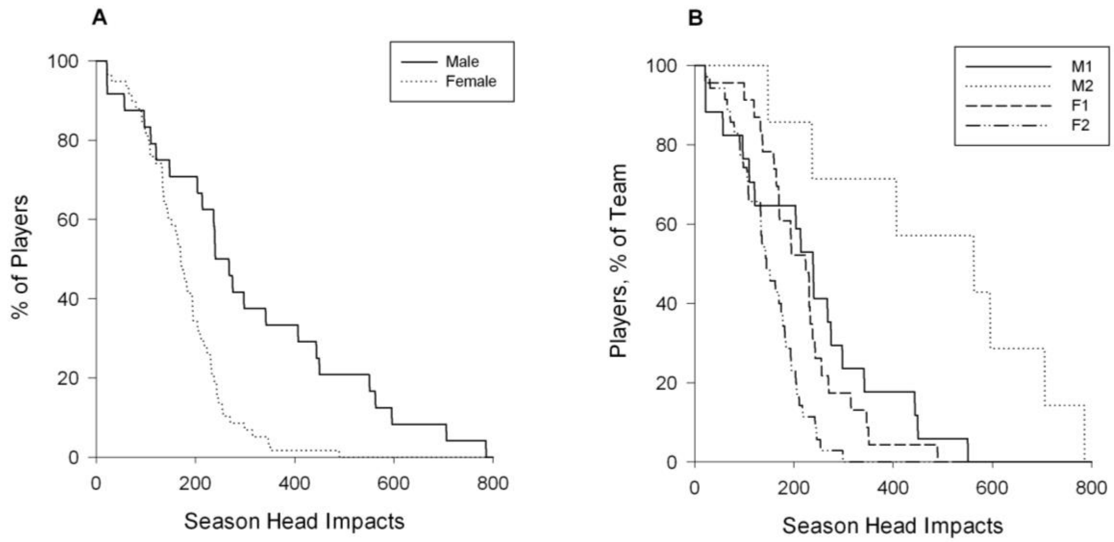

Figure 2.2: The cumulative distribution of the percentage of players for the number of impacts per season by sex (A) and team (B).

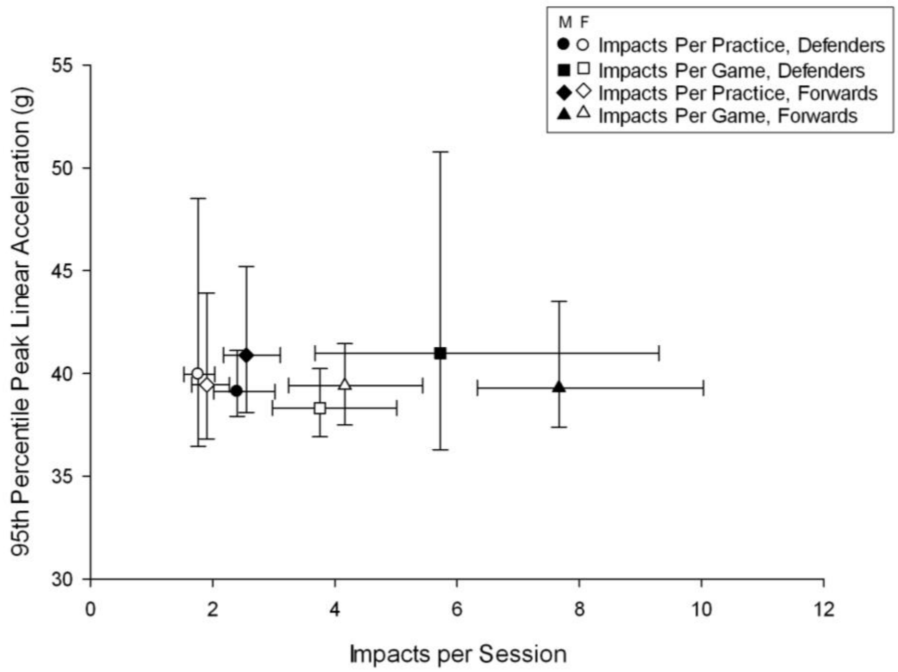

Figure 2.3: After categorizing by sex and player position, the median [25%-75%] 95th percentile peak linear acceleration as a function of the frequency of the median head impacts per session [25%-75%]. Filled markers represent men; unfilled markers represent women.

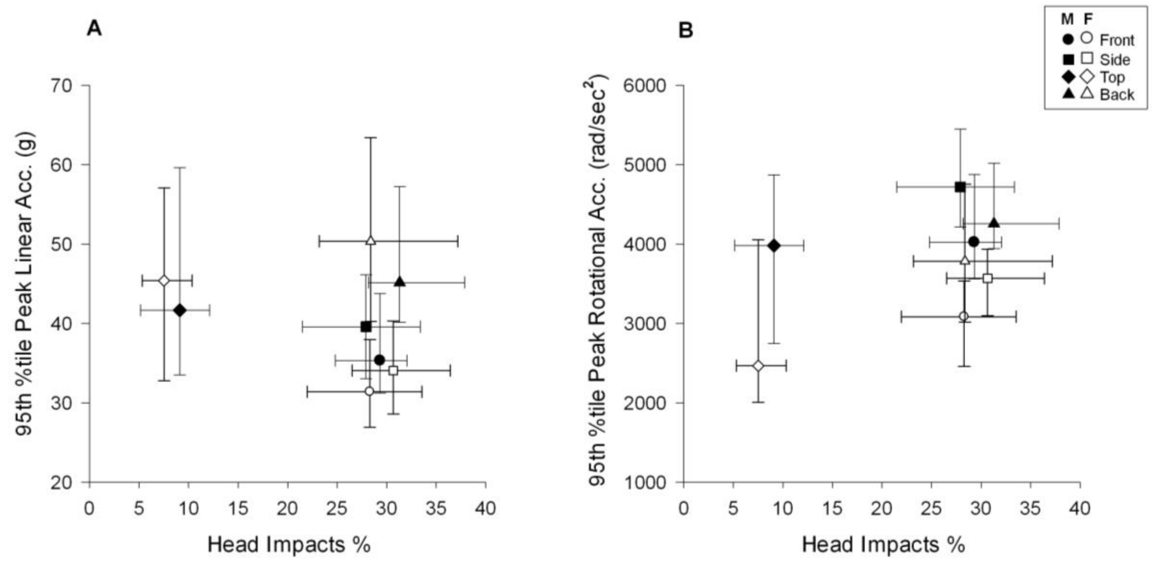

Figure 2.4: Median [25%-75%] of the 95th percentile peak linear (g) (A) and peak rotational (rad/sec^2) acceleration (B) as a function of the median [25%-75%] frequency of impacts at each helmet location and categorized by sex. Filled markers represent men; unfilled markers represent women.

Chapter 3

Head Impact Mechanisms in Men's and Women's Collegiate Ice Hockey

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3.1 Abstract

3.1.1 Background

Concussion injury rates in men's and women's ice hockey are reported to be among the highest of all collegiate sports. Quantification of the frequency of head impacts and the magnitude of head acceleration as a function of the different impact mechanisms (e.g. head contact with the ice) that occur in ice hockey could provide a better understanding of this high injury rate.

3.1.2 Objectives

To quantify and compare the per-game frequency and magnitude of head impacts associated with various impact mechanisms in men's and women's collegiate ice hockey players.

3.1.3 Methods

Twenty-three men and thirty-one women from 2 NCAA Division I ice hockey teams participated in this IRB approved study. Magnitude and frequency (per game) of head impacts per player among impact mechanisms and between sexes were analyzed using generalized mixed linear models and generalized estimating equations to account for repeated measures within players. Participants wore helmets instrumented with accelerometers in order to collect biomechanical measures of head impacts sustained

during play. Video footage from 53 games was synchronized with the biomechanical data. Head impacts were classified into 8 head impact mechanism categories: contact with another player, the ice, boards or glass, stick, puck, goal, indirect contact, and contact from celebrating.

3.1.4 Results

For men and women, contact with another player was the most frequent impact mechanism and contact with the ice generated the greatest magnitude head accelerations. The men had higher per game frequencies of head impacts from contact with another player and contact with the boards than women ($p < 0.001$), and these impacts were found to be greater in peak rotational acceleration ($p = 0.027$).

3.1.5 Conclusions

This study identified the impact mechanisms in collegiate ice hockey that result in frequent and high magnitude head impacts, providing data that could improve the understanding of the high rate of concussion in the sport and could inform prevention strategies.

3.2 Introduction

Ice hockey is a high-intensity, high-speed collision sport where the majority of injuries are caused by blunt trauma or direct contact with another player or object, as opposed to overuse injuries [1]. High rates of injury have been reported in both men's and women's collegiate ice hockey (5.95/1000 and 5.12/1000 athletic exposures (AE), respectively) and the most common injury in both populations is concussion[2]. The rate of concussion has been reported to be higher in womens ice hockey (0.82/1000 AE) than in mens (0.72/1000 AE), but the reasons for this are not well understood [2]. Concussions are usually attributed to a direct impact to the head but can also be the result of an impact to the body that results in an acceleration of the head [3].The high rate of injury, including concussions, in ice hockey can been attributed to some unique factors of the game; the playing area is made of solid ice and enveloped by rigid boards, players manipulate pucks that when shot can exceed 80 mph, players travel at high speeds of up to 30 mph and purposefully collide with opponents [4,5]. These factors allow for a number of different head impact mechanisms, or circumstances in which a head impact occurs (head contact with ice, boards, etc.), in ice hockey.

Currently, there is a lack of data quantifying the biomechanics of head impacts as a function of the different impact mechanisms that occur in ice hockey. Previous studies have quantified the frequency and magnitude of head impacts in cohorts of male and female hockey players at different levels of play using the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH) [6,7]. The HIT System measures and records biomechanical data from head impacts including the linear and rotational acceleration of the head, impact duration and impact location on the helmet [5–19]. These studies have provided valuable information on individual player's exposure to head impacts but did not identify, or examine the relationship with, mechanisms of impact. Other studies have reported injury epidemiology, including diagnosed concussions, by specific injury mechanisms in collegiate ice hockey [2,4,20–22]. Agel et al. utilized the NCAA Injury Surveillance System (ISS) to report concussion mechanisms in men's and women's collegiate games [20,21]. Diagnosed concussions were classified into one of seven concussion mechanisms that included: contact with another player, contact with the ice surface, contact with the boards or glass, contact with the goal, contact with the stick, contact with the puck and no apparent contact. Another study classified injury mechanisms in professional players in the National Hockey League (NHL) by reviewing video footage from games in which diagnosed concussions occurred [23]. The most common mechanism that resulted in diagnosed concussions for both studies was player to player contact [20,21,23]. While these studies provided important information on injury and concussion mechanisms in ice hockey, the collection and analysis of the impact biomechanics that result from these mechanisms was beyond the scope of their study designs. Synchronizing video with the biomechanics of head impacts would provide a quantitative approach to evaluating head impact mechanisms and the biomechanics of the resultant impacts.

The aim of this study was to quantify and compare the frequency and magnitude of head impacts associated with various impact mechanisms in men's and women's collegiate ice hockey players. We accomplished this by synchronizing video footage from games with biomechanical data from the HIT System. We hypothesized that the frequency and magnitude of head impacts would differ among the various head impact mechanisms and that sex would be a significant factor in both frequency and magnitude.

3.3 Methods

Twenty-three men and thirty-one women from Brown University's ice hockey teams participated in this study with Institutional Review Board (IRB) approval from Brown University and Rhode Island Hospital. The female players (ages 18-24, weight 68.0 ± 8.2 kg, height 168.9 ± 7.1 cm), ten defenders and twenty one forwards, wore instrumented helmets for three seasons (2008-2009, 2009-2010, 2010-2011). 13, 10 and 8 of the female players participated in one, two and three seasons, respectively. The male players (ages 19-25, weight 85.2 ± 4.7 kg, height 182.6 ± 4.1 cm), eight defenders and fifteen forwards, wore instrumented helmets over the course of one season (2009-2010). Data collection for the men's team was limited to a single season because the team accepted a new equipment contract after that season, and HIT System technology was not compatible with helmet models specified within the new contract. Goalies were not included in this study.

Home games were professionally video recorded at 30 frames per second using a Panasonic DVC - PRO AJ-D810ap professional broadcast camera. This camera provided 750 lines of horizontal resolution and 450 lines of vertical resolution. The lens was a Fujinon A18x7.6 mm broadcast lens with a focal length of 7.6 -137mm. The camera followed the puck during play. Video footage was collected for all home games where the camera equipment and operator were available, resulting in footage for twelve out of the fifteen men's home games and forty-one out of the forty-four women's home games.

Biomechanical data from head impacts were collected using helmets instrumented with the HIT System. Participants were fitted with S9 Easton helmets (Van Nuys, CA) that had been modified to accept the HIT System. The HIT System instrumentation, data reduction methods, and the accuracy of the HIT algorithm have been previously verified and described in detail [7–10,14–17,24]. Briefly, six single-axis accelerometers, arranged tangentially to the head, were elastically mounted within the helmet's foam liner to maintain contact with the head and to decouple shell vibrations [10,15]. Acceleration data associated with unique player IDs were collected at 1 kHz, time stamped and stored on the helmet (up to 100 impacts in static memory), transmitted by radiofrequency telemetry to a computer and then entered into a secure database. For each impact, magnitude was quantified by peak linear acceleration (g) and peak rotational acceleration (rad/s^2) of the head, as well as HITsp, a weighted measure of head impact severity that includes linear and rotational acceleration, impact duration, and impact location [15]. The biomechanical data were filtered to include only head impact events that resulted in a peak linear acceleration greater than 20 g. This threshold approximates the mean of all head impact events sustained in contact sports [5,7,12,13,17,18,25] and is well below the reported acceleration levels for diagnosed sports concussions [26–28].

Game video and biomechanical data were time-synchronized at the beginning and end of each game period by manually generating an impact event on a spare HIT System unit within the video field of view. Video was reviewed using VLC media player (VideoLAN, Paris, France; version 2.0.0). Identification of the impacts on video was aided by player ID and location of impact on the helmet recorded by the HIT System. Head impacts were classified into mechanism categories modeled after the NCAA Injury

Surveillance System's (ISS) Game Concussion Mechanisms of Injury, previously described by Agel et al. [2,20,21] and included: contact with another player, contact with the ice surface, contact with the boards or glass, contact with the goal, contact with the stick, contact with the puck, indirect contact, and contact from celebrating. An event was classified as one of these mechanisms when there was direct *head* contact with another player, the ice surface, boards or glass, stick, puck, or goal. An impact event was classified as indirect contact when an acceleration event above 20 *g* was recorded but the head did not appear to make contact with any object or player. This usually occurred as a secondary event. For example, when a player was clearly hit in the torso or fell to the ice but the player's head did not appear to make contact with any other person or object. For impacts such as these, the primary mechanism was not categorized or reported. An eighth category, contact from celebrating, was included due to the observation of relatively frequent head impacts greater than 20 *g* that resulted from teammates hitting each other in the head in a congratulatory way after a good play or a goal. Because the videographer followed the puck during play, there were head impacts recorded by the HIT System that were not captured on video. These impacts were sustained by players outside of the field of view of the camera or were obstructed from view by the angle of the camera or by an object, such as the boards or another player. These head impacts were not included in the present analysis, as the mechanism of contact could not be classified.

Statistical Analysis

The percentage of all head impacts among each mechanism category and between sex were computed independent of player, but not used in hypothesis testing. All hypothesis testing was performed using SAS version 9.2 (Cary, NC). Alpha was set to

0.05 for all analyses. A generalized estimating equation for negative binomial data, offset with the natural logarithm of the number of games each player played in, was used to test whether frequency (per game) of head impacts per player differed among impact mechanisms and between sexes (one count per player per game per mechanism). Generalized linear mixed models for lognormal data were used to test whether the peak linear acceleration, peak rotational acceleration, and HITsp of impacts experienced by players varied as a function of sex and mechanism. Compound symmetry variance-covariance structures were used to model the nature of the within-subjects correlation by game (games have same variance and single covariance between games). Follow-up pairwise comparisons were adjusted using the Holm test to maintain overall alpha at 0.05 for all models. All models were also adjusted for model misspecification using classical sandwich estimation, making inferences robust to errors in distribution and variance-covariance structure selection.

3.4 Results

A total of 4,497 head impacts, 1,965 sustained by men and 2,532 sustained by women, were recorded during fifty-three home games. Six-hundred and sixteen head impacts had a resultant peak linear acceleration greater than 20 *g*. For the twelve men's home games, 270 impacts (81% of the 333 impacts >20 *g* recorded) were successfully captured on video and classified (**Table 3.1**). For the forty-one women's home games, 242 impacts (85.5% of the 283 impacts > 20 *g* recorded) were successfully captured on video and classified. Approximately half of these impacts were caused by contact with another player in both the men's and women's teams (50.4% and 50%, respectively). For the men's team, 31.1% of head impacts were caused by contact with the boards or glass, 7% were caused by contact with the ice, and 4.5% were caused by indirect contact and contact from celebrating. For the women's team, 17.3% of head impacts were caused by contact with the boards or glass, 15.3% were caused by indirect contact, and 11.2% were caused by contact with the ice. Contact with the stick, goal, and puck were each less than 3% of the impacts for both men and women and were not analyzed further.

In men's ice hockey, head impacts for individual players resulting from contact with another player occurred at a frequency of approximately once in every two games (0.46 per game)(**Table 3.2**). The frequency of impacts per game that were caused by contact with another player and contact with the boards (0.349 per game) were both significantly higher than the frequency of impacts by contact with ice, indirect contact, and celebrating ($p < 0.001$ for each comparison). In women's ice hockey, the frequency of impacts per game that were caused from contact with another player (0.21 per game) was significantly higher than those that were caused by contact with the boards, indirect contact, and celebrating ($p < 0.001$ for each comparison).

The mean and 95% confidence for peak linear acceleration, rotational acceleration, and HITsp for all impacts analyzed in this study was 31.2 *g* [28.9, 33.7], 2881.0 rad/s^2 [2580.0, 3217.2] and 18.8 [17.3, 30.4] for men and 28.3 *g* [26.6, 30.1], 1766.8 rad/s^2 [1508.0, 2068.8] and 16.74 [15.7, 17.9] for women. Across all mechanisms, peak rotational acceleration and HITsp were both significantly higher in men compared to women ($p < 0.001$, $p = 0.035$ respectively), while the difference in peak linear acceleration was not significant ($p = 0.054$). In men's ice hockey, peak linear accelerations caused by contact with the ice were significantly greater ($p < 0.001$) than those from contact with another player (**Table 3.3**). In women's ice hockey, peak linear acceleration was significantly greater in head impacts caused by contact with the ice than those caused by contact with another player ($p = 0.029$), contact with boards or glass ($p < 0.001$), indirect contact ($p = 0.043$), or celebrating ($p < 0.001$). Women also experienced significantly greater peak rotational acceleration in head impacts caused by contact with another player than in those caused by contact with the boards ($p = 0.03$). Head impacts sustained while celebrating in men's and women's hockey were generally lower in linear acceleration, rotational acceleration, and HITsp than the other mechanisms.

The frequency of head impacts per game resulting from contact with player and contact with the boards or glass were both significantly higher ($p < 0.001$) for men than for women. Although contact with another player was the most frequent impact mechanism in both men and women, these impacts were not the greatest in magnitude

(**Figure 3.1**). Peak linear acceleration ($p < 0.001$) and HITsp ($p = 0.003$) from contact with the boards or glass were both significantly greater for men than for women. Peak rotational acceleration was greater for men than for women in contact with another player ($p = 0.027$), contact with the boards or glass ($p < 0.001$), and celebrating ($p < 0.001$). All other comparisons of impact magnitude were not statistically significant between men and women.

3.5 Discussion

The purpose of this study was to quantify and compare the frequency and magnitude of head impacts associated with various impact mechanisms in men's and women's collegiate ice hockey. The impact mechanisms of head contact that we identified included contact with another player, the ice surface, boards or glass, stick, goal, puck, indirect contact, and celebrating. These categories were modeled from previous studies [2,20,21] that utilized the NCAA Injury Surveillance System (ISS) to report the epidemiology of injury mechanisms in men's and women's collegiate ice hockey.

We found that male players experienced head impacts greater than 20 *g* from contact with another player and contact with the boards once every two-to-three games while women were less than once every three-to-five games. Peak rotational acceleration of the impacts from these mechanisms was also greater for men than for women by approximately 25%. We attribute this difference to the fact that checking, or purposeful body contact of an opposing player, is permitted in men's collegiate ice hockey, whereas it is illegal in women's. The high rate of contact with another player and with the boards in men's hockey was expected, as checking is an important part of the game and frequently results in secondary impact of the head or body to the boards. In a study on male players at the professional level, the predominant mechanism of concussion was reported to be player to player contact [23]. Similarly player-to-player contact was reported to be the most common cause of concussions at the collegiate level, accounting for 72% of diagnosed concussions in males and 41% of those sustained by females in games [2,20,21]. While checking is not allowed by the NCAA in women's ice hockey, these data confirm, as previous studies have reported[6], that there is frequent and high magnitude player-to-player contact during women's play. It has been speculated that because many female hockey players are not taught and do not practice checking skills, they may be less prepared to absorb impacts when they are subject to collisions [2]. Studies have shown that anticipated head impacts result in less severe head impacts than unanticipated impacts in youth hockey [18]. It is possible that inconsistent enforcement of the rules leaves female players more vulnerable to impacts caused by player contact because they may not be expecting the contact [20]. Whereas this study did not identify illegal plays or infractions, a review of the penalties along with comparisons of secondary contact after player-to-player contact may aid in a better understanding of the role that checking plays in the high concussion rate in men's and women's collegiate ice hockey.

While contact with another player was the head impact mechanism that resulted in the highest rate of per game head impacts, impacts from contact with the ice were the greatest in magnitude (**Figure 3.1**). These impacts had greater mean peak linear acceleration than impacts caused by other mechanisms in both men's and women's ice hockey. This is not surprising given the fact that these players can move at speeds upwards of 30 mph [4], are falling onto the ice from a height several inches higher than their own (skates and blades can add inches to players height), and are hitting the hard ice surface. While contact with the ice occurred at a relatively low frequency when compared to other impact mechanisms, this impact mechanism occurred approximately once every ten games for individual male and female players. Given that NCAA teams are allowed to play 34 games during a regular season (not including conference post season tournaments), our findings suggest that average player is experiencing these high magnitude head impacts with the ice approximately three times per season. There were no differences between men and women in the frequency and magnitude of impacts that occur from contact with the ice. Interestingly previous studies [20,21] have reported that the contact with the ice results in 28.1% of game concussions for female collegiate ice hockey players but only 7% of game concussions for their male counterparts. While the incidence of diagnosed concussion has been reported to be higher in female hockey players than in males [2,29], the frequency and magnitude data collected in this study confirm previous reports that females sustain fewer impacts and these impacts result in lower head accelerations than males [6]. Several factors for why females have a higher rate of concussion than males have been proposed, including physiological and psychological differences, but the exact reason remains unclear [29].

Regardless of head impact mechanism, the mean peak linear accelerations reported in this study were greater than those previously reported for other studies that have utilized the HIT System, including youth hockey and collegiate football. Mihalik et al. studied a cohort of youth male hockey players and reported a mean peak linear acceleration of 18.98 g [7]. Crisco et al. reported 50th percentile peak linear acceleration of 20.3- 20.5 *g* in collegiate football [12,13]. Besides the factors of age, sex and sport, these differences can be attributed to the 20 *g* acceleration threshold selected for this analysis. A primary benefit of the HIT System is its ability to capture head acceleration events that may not be easily discernible in the fast paced environment of contact sports. To accommodate practical limitations of video review, we elected to use 20 *g* (an approximation of the mean values of head impact events sustained in contact sports [5,7,12,13,17,18,25]) as an inclusion threshold for biomechanical data. This threshold level isolated head acceleration events with a high likelihood of producing a clear physical response that could be identified with video (in comparison, an aggressive pillow fight results in approximately 20 *g* [30]) while still including events well below acceleration magnitudes associated with diagnosed sports-concussion [26–28]. There was an approximately equal proportion of impacts less than 20 *g* for both males (86.3% (1695/1965)) and females (90.4% (2290/2532). While the acceleration threshold does not affect the associations between impact mechanism and biomechanical response reported in this study, future studies should avoid direct comparisons to the mean acceleration values reported in **Table 3.3** without consideration for this threshold. It is also important to note that while accelerations of less than 20 *g* are considered relatively low in magnitude for single events [19,25], the long-term consequence of such repeated events are unknown.

This threshold was just one of several recognized limitations of the study. Classification of the impact mechanisms was ultimately subjective. While video is a relatively simple method of observing head impact mechanisms in hockey, it can be challenging because of the fast pace of the game. Approximately 15-20% of impacts above 20 *g* that were recorded by the HIT System were not captured on video because they occurred outside of the field of view of the camera or they were obstructed from view by the angle of the camera or by an object, such as the boards or another player. The study included an unequal number of seasons and games between men and women. While we only analyzed data from a series of home games as opposed to all athletic exposures (all practices and games) during each season, it has been reported that the incidence of injury is higher in games than in practices and that 77% of concussions occur in games [1,2]. The HIT system provides additional biomechanical variables that were not included in the analysis, including location of impacts on the helmet and duration of acceleration. While these variables were beyond the scope of this analysis; future analysis of these measures in relation to head impact mechanism is warranted. A final limitation is that there were no diagnosed concussions that occurred during the games included in this study.

In summary, the most frequent head impact mechanism in both men's and women's collegiate ice hockey was contact with another player, and contact with the ice was the mechanism that resulted in head impacts with the greatest magnitude. Recently, research related to head injuries in sports has primarily focused on addressing two main concerns, the high rate of diagnosed and undiagnosed concussions and the potential long term effects of repetitive head impacts. If you assume that impacts of greater magnitude have more associated risk for concussion [26], it would appear that the strategies to address these two concerns may be different. To reduce frequency of head impacts, player contact rules should be addressed. To address high magnitude head impacts, re-evaluating helmet design to protect against contact with the ice may be warranted. Sex was found to be a factor in per-game frequency and magnitude of head impacts associated with several impact mechanisms. Men experienced head impacts from contact with another player and contact with the boards more frequently than women and these impacts were generally of greater magnitude. Further study is required to better understand why female athletes are reported to have higher concussion rates compared to their male counterparts [2], given that females sustained less frequent and lower magnitude head impacts in our study. The identification of impact mechanisms in collegiate ice hockey that result in frequent and high magnitude head impacts is an important step in understanding the high rate of concussions in the sport and could inform concussion prevention strategies.

3.6 Acknowledgments

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	Men (n=270)	Women (n=242)
Another player	50.4% (n=136)	50% (n=121)
Ice	7% (n=19)	11.2% (n=27)
Boards or glass	31.1% (n=84)	17.3% (n=42)
Stick	1.9% (n=5)	2.9% (n=7)
Goal	0.4% (n=1)	0% (n=0)
Puck	0.4% (n=1)	0.8% (n=2)
Indirect	4.4% (n=12)	15.3% (n=37)
Celebrating	4.4% (n= 12)	2.5% (n=6)

Table 3.1: Percentage of head impacts (and number) for each category of head impact mechanism across the entire study for each team, independent of players.

	Men	Women	p-value
Another Player	0.464	0.208	<0.001*
Ice	0.104	0.106	0.950
Boards	0.349	0.095	<0.001*
Indirect	0.087	0.100	0.539
Celebrating	0.080	0.073	0.618

Table 3.2: Head impact frequency per game among the head impact mechanism (see Results for statistical analysis) and between men and women (*, statistically different between sexes).

	Linear Acceleration (*g*)	Rotational Acceleration (rad/s^2)	HITsp
Men			
Another Player	28.0 (26.4, 29.7)	2901.8 (2514.5, 3348.7)	19.2 (17.7, 20.7)
Ice	40.1 (31.8, 50.5)	3454.9 (2590.2, 4608.4)	22.8 (17.9, 29.1)
Boards	32.1 (29.7, 34.7)	3350.4 (2995.9, 3746.8)	21.0 (20.2, 21.8)
Indirect	31.5 (26.4, 37.8)	2873.8 (1949.8, 4235.7)	19.7 (15.2, 25.5)
Celebrating	25.9 (23.6, 28.4)	2056.3 (1707.9, 2475.7)	12.9 (11.3, 14.7)
Women			
Another Player	27.9 (26.3, 29.6)	2323.0 (2031.6,2656.9)	17.9 (16.8, 18.9)
Ice	35.2 (30.9, 40.0)	2318.9 (1644.2, 3270.4)	21.2 (17.7, 25.5)
Boards	26.8 (25.8, 27.9)	1859.5 (1587.0, 2178.8)	16.7 (14.5, 19.2)
Indirect	29.5 (25.6, 34.0)	1861.3 (1387.1, 2497.6)	19.1 (17.1, 21.4)
Celebrating	23.3 (20.1, 27.0)	923.3 (675.2, 1262.5)	10.9 (8.8, 13.5)

Table 3.3: Mean (95% confidence interval) resultant peak linear acceleration, peak rotational acceleration, and HITsp of head impacts greater than 20 *g* sustained by men's and women's collegiate ice hockey players for each category of head impact mechanism.

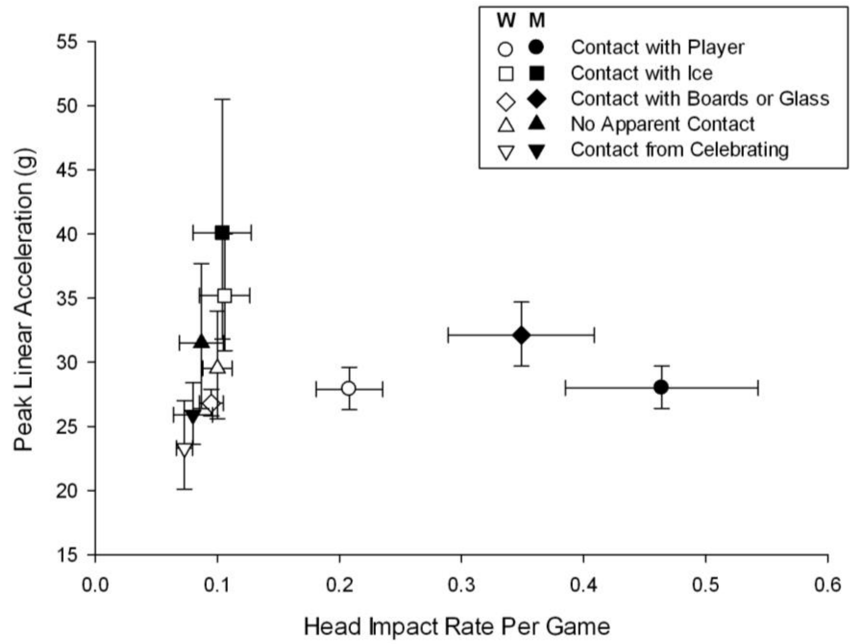

Figure 3.1: Mean and 95% CI peak linear acceleration (*g*) as a function of the frequency of head impacts per game for each head impact mechanism. Unfilled markers represent women; filled markers represent men.

Chapter 4

Female Athletes Experience Higher Rates of Concussion with Lower Head Impact Exposure

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4.1 Abstract

4.1.1 Purpose

Epidemiological evidence suggests that female athletes may be at a greater risk of concussion than their male counterparts. The purpose of this study was to determine if the rate of concussion injury, the biomechanics of single head impacts associated with concussion and head impact exposure of individual athletes on days of diagnosed concussion differ between males and females in a study population of collegiate football and ice hockey players.

4.1.2 Methods

Instrumented helmets were worn by 354 football players, 41 male ice hockey players, and 58 female ice hockey players from 3 NCAA athletic programs over a five year period. Thirty-seven concussions were diagnosed: 26 in football players, 2 in men's ice hockey, and 9 in women's hockey. Rates of concussion were calculated by number of athletic exposures and number of head impacts. Kinematic measures of single impacts associated with diagnosed concussion and impact exposure on days of concussion were evaluated.

4.1.3 Results

Female hockey players were found to have a significantly higher rate of diagnosed concussion per 1000 head impacts (0.513) than male football players (0.071, $p<0.005$). Average peak linear acceleration of single impacts associated with diagnosed concussion was significantly higher for male football players (117*g*) than female hockey players (43*g*, $p<0.05$). Male football players experienced a higher number of impacts and impacts of greater magnitude on days of diagnosed concussions than females (*p*-values range from $p<0.001$ to $p=0.04$). Consistent trends were shown for male hockey players (with relatively low number of diagnosed concussions (n=2)) compared to females.

4.1.4 Conclusions

These findings indicate that females have a higher rate of diagnosed concussion and may have a lower biomechanical tolerance to, or lower threshold for reporting symptoms of, concussion injuries.

4.2 Introduction

Epidemiological evidence suggests that female athletes may be at a greater risk of concussion than their male counterparts [1,2]. Research conducted in sports where male and females participate at the same level, such as ice hockey, has shown that female athletes sustain concussions at a higher rate (0.82/1000 athletic exposures (AEs)) than males (0.72/1000 AEs) [2,3]. It has been reported that females experience a greater number of symptoms associated with concussion and that these symptoms are often more severe [4–6]. Recent studies have also suggested that females may be at a greater risk of post-concussive syndrome, where the acute symptoms associated with concussion injuries become chronic [7]. Several gender-specific characteristics have been proposed as the rationale for these differences, including physiological differences, such as anthropometrics and hormone regulation, and psychological factors, including speculation that females tend to be more honest than males in reporting injuries [2], but the exact reasons remain unclear. Before we can understand what role these factors play in the differences between the sexes in concussion incidence and outcome, a better understanding of the differences between males and females in the mechanisms of concussion injury is warranted. While it has been generally accepted that the mechanism of concussion injury is related to head acceleration [8], the exact relationship between the biomechanics of head impacts, sex, and clinical outcome of concussion is unknown.

Recent studies have focused on understanding the biomechanics of head impacts sustained in contact sports by utilizing an accelerometer based head impact monitoring device, the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH). The HIT System allows researchers to monitor and record head impacts sustained by individual helmeted athletes during play. Utilizing this system and the unique laboratory that the playing field provides, researchers have quantified head impact exposure, a multifactorial term including frequency, magnitude and location on the helmet of head impacts, for individual athletes at different levels of play in football and ice hockey [9–12]. Studies using the HIT System have reported that male helmeted athletes generally have higher frequency of head impacts and experience impacts that are greater in magnitude than females [12,13].

Data from the HIT System has also been used to evaluate the biomechanics of head impacts associated with diagnosed concussions in high school and collegiate football players [14–18]. Beckwith et al. [14] reported that individual players sustained more head impacts and impacts of greater severity on days of diagnosed concussion than on days without diagnosed concussion. While these studies have provided valuable insights into the kinematics of impacts associated with diagnosed concussion in male football players, analysis and comparison of head impacts associated with diagnosed concussion in both male and female athletes is warranted.

The purpose of this study was to evaluate rate of diagnosed concussion and the biomechanics of head impacts associated with diagnosed concussions in a study population of collegiate football, male ice hockey and female ice hockey players. We aimed to test the hypotheses that sex would be a significant factor in the rate of diagnosed concussion, the biomechanics of single head impacts associated with diagnosed concussions and head impact exposures on days of diagnosed concussion.

4.3 Methods

Over a five year period, 354 football players, 41 male ice hockey players, and 58 female ice hockey players from 3 National Collegiate Athletic Association (NCAA) athletic programs (Brown University, Dartmouth College and Virginia Tech) participated in this observational study after informed consent was obtained with institutional review board approval. Participation in the study was voluntary with no consideration given to a player's previous history of concussion. Many players participated during multiple years of play. Of the 354 football players (height = 186.3 ± 6.2cm, weight = 104.2 ± 17.3 kg), 24 players participated in 4 seasons, 74 players participated in 3 seasons, 122 players participated in 2 seasons, and 134 players participated in a single season. For ice hockey players, 30 males and 19 females were monitored during one season, 5 males and 20 females during two seasons, and 6 males and 19 females during three seasons. This participant turnover was attributed to typical fluctuations on collegiate athletic teams. The average reported height and weight of male hockey players was 184.4 ± 4.1cm and 86.3 ± 4.9 kg and for the females, 168.9 ± 6.4 cm 67.9 ± 6.8 kg, respectively.

All participants wore instrumented helmets (Head Impact Telemetry (HIT) System, Simbex, Lebanon, NH) to collect the magnitude and location of head impacts sustained during practices and games. Football players wore Riddell VSR-4, Revolution, or Speed (Riddell, Chicago IL) helmets and hockey players wore either S9 Easton (Van Nuys, CA) or CCM Vector (Reebok-CCM Hockey, Inc., Montreal, Canada) helmets. These helmets measured and recorded biomechanical data from head impacts including linear and rotational acceleration of the center of gravity (CG) of the head, as well as location of impact on the helmet. The instrumented helmets each consist of six single-axis accelerometers mounted within the liners of helmets to maintain contact with the head and thus isolate the accelerometers from shell vibrations [19,20]. The designs of the football and hockey instrumented helmets are comparable with the exception of the orientation of the accelerometers relative to the head (normal to the head for football and tangential to the head for hockey) and processing algorithms. System design, validation, accuracy, and data reduction methods have been previously described in detail [11,19–23]. Both systems collect acceleration data at 1 kHz, time stamp each impact and store the data (up to 100 impacts in static memory). Data is then transmitted by radiofrequency telemetry to a computer and entered into a secure database.

During the period of study, concussions were generally defined as an alteration in mental status, as reported or observed by the player or medical staff, resulting from a blow to the head. Diagnosis of concussion was made by the team physician or certified athletic trainer (ATC) at each respective institution diagnosed. For each diagnosed concussion, suspected date and time of injury, anthropometrics of the injured player, and symptom presentation within 72 hours were documented. If a single impact was identified as being associated with the concussion event by the injured player or observers, this was recorded. Anecdotal descriptions of the events surrounding injury were also collected to further aid in corroborating data from the instrumented helmets with single impacts associated with diagnosis of concussion.

Rates of concussion for each subject population were calculated as both the number of injuries over the number of athletic exposures (AE) and number of concussions over the number of head impacts. An AE is defined as an individual athlete participating in a game or practice in which there is the possibility of athletic injury [24]. Due to limitations associated with keeping attendance records for the large number of football players on each team, an individual football player was defined to have one AE when at least one head impact was recorded within the specified time of the team session. For hockey players, an individual was defined to have an AE when the player was present and partook in the game or practice, regardless of whether they sustained an impact or not. All concussion rates are reported per 1000 AEs and per 1000 head impacts and 95% confidence intervals.

To evaluate the biomechanics of impacts associated with diagnosed concussion, two different approaches were taken: (1) the examination of single impacts linked to injury, and (2) the cumulative analysis of all impacts sustained on days of injury. Single impacts were identified when a concussion event was attributed by the player or observers to a specific identifiable impact. Impact magnitude variables included peak linear acceleration (g), peak rotational acceleration (rad/s^2), and HITsp [17,19,23]. HITsp is a weighted measure of head impact severity that includes linear and rotational acceleration, impact duration, and impact location [23]. Results were expressed as mean ± standard deviation. For the cumulative analysis, head impact exposure for individual players on days with and without diagnosed concussion was computed. This was accomplished using previously established methods to evaluate differences in head impact exposure on days with and without concussion injury in a large cohort of football players [14]. A single measure of impact frequency, the total number of impacts per day, was computed for individual players on days with and without diagnosed concussion.

Each individual player's distribution of peak linear acceleration (g), peak rotational acceleration (rad/s^2), and HITsp per day were quantified by the 50^{th} and 95^{th} percentile value of all impacts per day. Results are expressed as median values and 25-75% interquartile range.

Statistical Analysis

To compare concussion rates between males and females, 95% confidence intervals (CIs) were calculated using standard large sample formulas [25]. Significance was defined as non-overlapping CIs ($p \leq 0.05$). To evaluate differences in head impact frequency and magnitude variables (that were found to be non-normally distributed (Shapiro-Wilk test, $p < 0.05$)), on days with and without concussion within each sport, Wilcoxon sign-ranked test for pairs was used to test for significance. Severity measures of single impacts associated with diagnosed concussion, including peak linear acceleration, peak rotational acceleration, and HITsp, were compared among football players, male ice hockey players, and female hockey players using Kruskal-Wallis one way ANOVA on ranks (non-normally distributed data, Shapiro-Wilk test, $p < 0.05$). An identical approach was used to examine the significance of differences in head impact exposure on days with diagnosed concussions for male football players and female ice hockey players. Male ice hockey players were not included in the statistical comparisons for impact biomechanics because of the low number of diagnosed concussions (n=2) that occurred during this study. One way ANOVAs with Holms-Sidak post-hoc tests were used to compare average height and weight for players between teams and between players with and without diagnosed concussions. All statistical analyses were performed using SigmaPlot 12.0 (Systat Software, Chicago, IL).

4.4 Results

There were a total of 37 concussions that were diagnosed during the course of the study: 26 were diagnosed in 24 football players (two players sustained two concussions) over the course of 28,390 athletic exposures (AEs), 2 concussions in men's ice hockey over 3,777 AEs, and 9 in women's ice hockey over 6,503 AEs. There was no statistically significant difference in concussion rate per 1000 AEs between football players (0.92 [0.57-1.27]), male hockey players (0.53 [-0.20-1.26]) and female hockey players (1.4 [0.48-2.28]) (Figure 4.1A). However, female hockey players were found to have a significantly higher rate of concussion per 1000 head impacts (0.51 [0.18-0.87]) than male football players (0.07 [0.04-0.10]) (Figure 4.1B). This trend was consistent for rate of concussion per head impact for male hockey players (0.10 [-0.04-0.24]) compared to females. The estimated confidence intervals for male hockey were based on a relatively low number of diagnosed concussions (n=2) and resulted in a negative lower limit. These head impacts rates were computed from a total of 404,748 head impacts: 367,337 sustained by football players, 19,880 sustained by male hockey players, and 17,531 sustained by female hockey players.

Approximately half of the diagnosed concussions (19/37) were associated with an identifiable single impact (Table 4.1). Mean linear acceleration, rotational acceleration and HITsp of single impacts associated with diagnosed concussion ranged from 24% to 96% greater for male athletes (football and hockey players) than for female athletes (Table 4.1). Average peak linear acceleration of single impacts associated with diagnosed concussion was significantly higher for male football players ($p<0.05$) compared to female hockey players (Table 4.1).

Significant differences were observed in the frequency and magnitude of head impacts that players sustained on days with and without diagnosed concussion. Football players ($p<0.001$) and female ice hockey players ($p=0.023$) sustained a greater number of head impacts on days of diagnosed concussion than on days without (Figure 4.2). The 95th percentile values of peak linear acceleration, peak rotational acceleration and HITsp were significantly higher on days with diagnosed concussion than on days without for football players and for female ice hockey players (Table 4.2). These trends were consistent for male hockey players (Figure 4.2).

Statistical differences were observed in frequency and magnitude of head impacts on days with diagnosed concussion between football players and female hockey players (Figure 4.2). The number of impacts sustained on days with diagnosed concussion was significantly higher for male football players than for female hockey players ($p<0.001$)). The 95th percentile values of peak linear acceleration ($p=0.002$), peak rotational acceleration ($p=0.04$), and HITsp ($p=0.006$) were all significantly higher for football players than female hockey players on days with diagnosed concussions (Figure 4.2). Again, trends were consistent for male hockey players compared to females.

Male football players and hockey players were significantly taller and weighed more than female hockey players ($p<0.001$). There were no significant differences in height or weight between the two sports for male athletes. There were also no significant differences in height or weight between athletes diagnosed with concussions and athletes that were not diagnosed with concussion for all sports (p-values ranged from 0.57-0.978).

4.5 Discussion

The purpose of this study was to examine the biomechanics of head impacts associated with diagnosed concussion in a study population of collegiate football, male ice hockey and female ice hockey players with the objective of determining if rate of diagnosed concussion, biomechanics of single impacts associated with diagnosed concussions, and head impact exposure on days of diagnosed concussion differ between males and females.

The female athletes that participated in this study experienced a significantly higher rate of diagnosed concussion per head impact than male football players. This trend was also consistent for male hockey players compared to females. Incidence of concussion is usually expressed as rate per AE. While AE is a useful metric to evaluate overall risk of injury as a function of participation (i.e., one participation equals one AE), it does not account for direct exposures, such as the number of head impacts an individual player receives during an AE. Evaluating the rate of concussion per head impact allows us to evaluate injury rate in relation to direct exposures, and provides a better understanding of the relative risk of head impacts. While both approaches demonstrated the same trend, rate of concussion per head impact demonstrated a greater and statistically significant difference between males and females. Inverting the rates provides another perspective on the incidence. In other words, in football one concussion was diagnosed in every 1087 AE or 14,085 head impacts, whereas in women's ice hockey one concussion was diagnosed in every 725 AE or 1,883 head impacts. There is a wide range of concussion rates per AE that have been reported for collegiate athletes in football (0.37-2.34/1000 AEs), men's hockey (0.41-1.47/1000 AEs), and women's (0.82-2.72/1000 AEs) ice hockey [3,26–29]. The rates of concussion per AE that were reported in the current study fall within these ranges. The high variability in previous studies that have reported rate of concussion per AE may be attributed to, while further emphasizing, difficulties in the identification of concussion injuries. Factors may include the high rate of underreporting in athletes, the widely varying definitions of concussion, and the numerous ways that injury rate is estimated, reported, and interpreted. Comparison with the literature on the rate of concussion per head impact is limited by the lack of previous reports, as this is a relatively new method of representing concussion rate.

Substantial research has focused on single impacts associated with concussion injury with the objective of identifying a biomechanical threshold for the injury. Pellman et al. [30] determined that impacts associated with concussion had peak linear accelerations of approximately 98 ± 28 *g*. While these values were indirect estimates, based upon surrogate dummy impacts whose approach velocity was reconstructed in the laboratory using video head impacts sustained by NFL players, these values fall within the range of most of the direct measurements that have been made to date. A study in a large cohort of high school and collegiate football players reported an average peak linear and peak rotational acceleration of 112.1 ± 35.4 *g* and 4253 ± 2287 rad/sec^2 for single impacts associated with diagnosed concussion [14]. Other studies in football have reported peak linear accelerations of single impacts associated with concussions ranging from 60.51 to 168.71 *g* [16,31]. In a study of collegiate football players the average linear acceleration of concussive impacts was 145 ± 35 *g*, and a risk analysis resulted in a threshold of 100 *g* as a cutoff for identifying potential concussions [15]. In our study, the average peak linear and rotational acceleration for impacts associated with diagnosed concussion in football players, 117.1 ± 43.0 *g* and 4576 rad/sec^2, were similar to values reported in these previous studies. The average peak linear and rotational acceleration of the two impacts associated with injury in male ice hockey players (71.0 ± 14.0 *g*, 5313 ± 15.6 rad/sec^2) fell within the reported range of those in football. Interestingly, while the average peak rotational acceleration associated with diagnosed concussions for female hockey players (4029.5 ± 11.5 rad/sec^2) was also comparable, peak linear acceleration (43 ± 11.5 *g*,) was substantially lower than any values reported for football players in the literature. If the recommended cutoff for collegiate football players (100 *g*) was used to monitor this group of female athletes, not one of the impacts that were associated with diagnosed concussions would have reached this threshold. These findings suggest that biomechanical thresholds may need to be gender-specific.

Similar to previous studies, we found that head impact exposure on days of diagnosed concussion is higher than on days without diagnosed concussion, in both frequency and magnitude [14]. This supports previous assertions that diagnosis of concussion is related to head impact exposure [14,32]. Head impact magnitude (95th percentile) on days without diagnosed concussion were somewhat lower than those previously reported for uninjured collegiate football (62.7 *g*, 4378 rad/sec^2), men's ice hockey (41.6 *g*, 4425 rad/sec^2) and women's ice hockey (40.8 *g*, 3408 rad/sec^2) [10,12]. This may be due to the fact that this analysis only included injured athletes, thus factors that influence head impacts exposure, such as players position and session type [10,12], could not be controlled for. On days with diagnosed concussions the magnitudes of the recorded head impacts were higher for our cohort of injured football and male hockey players than previously reported for typical collegiate athletes in these sports [10,12], with the exception of peak rotational acceleration in football players. It should be noted that the values for peak rotational acceleration in the football dataset were adjusted from the HIT system calculated values as per Rowson et al. [17]. While these values are consistent with methods used in more recent studies that utilized the football HIT system [14,33], they have been scaled by a factor of 0.64791 when compared to earlier reports that pre-date the Rowson method [10]. Interestingly, the magnitude of head impacts female hockey players sustained on days of injury were comparable to those previously reported for typical female hockey players [12]. That is, while we found that individual female athletes diagnosed with concussion received higher magnitude head impacts on days of diagnosed concussion compared to days without, these impacts were comparable to those previously reported in a larger cohort of uninjured female ice hockey players [12].

Single impacts associated with diagnosed concussion and head impact exposure on days of diagnosed concussion (Figure 4.2) were higher for male football players than female hockey players. This trend was also consistent for male hockey players compared to females. These findings suggest that females may have a lower biomechanical tolerance to concussions than males. Several gender-specific characteristics have been proposed to explain differences between males and females in concussion injuries including physiological and psychological differences. The male athletes in our study weighed more than the female athletes and were generally taller. These differences were similar to those observed between males and females for this age group in the general population [34]. It has been postulated that females may be at a higher risk of concussion because their smaller size and weaker neck muscles may lead to higher accelerations of the head [35]. We observed that while females were smaller than their male counterparts, they experienced lower magnitude head accelerations. Studies using animal models have identified that hormonal differences between males and females affect clinical outcomes after concussion and traumatic brain injury, but have noted that hormone regulation is based on extent of damage and that different outcomes may have a biomechanical basis with hormonal factors playing a role only in response to the injury [36,37].

Consideration should also be given to rule variations within sports by gender. For example, checking or purposeful body contact of an opposing player is an important part of the game in men's ice hockey whereas it is illegal in women's. It has been postulated that due to the aggressive nature and faster pace of the game in men's ice hockey, males should be at a greater risk of concussion than females [35,38]. Accordingly, previous studies have reported that male hockey players generally have higher frequency of head impacts and experience impacts that are greater in magnitude than females [12,13]. Counter intuitively, the data presented in this study demonstrates that females sustain higher rates of concussion with lower head impact exposure.

It has been speculated that because females experience head impacts less often, they may have a heightened awareness of them when they occur [13]. Interestingly, there are also studies that have shown that athletes in player positions that tend to have less contact, for instance a catcher in baseball, sustain concussions at a biomechanical input that is lower than suggested injury thresholds [39]. It may be that for sports or player positions where individuals have a high exposure to head impacts, players that have a lower tolerance to head impacts are “weeded out” or quit playing by the time they get to the collegiate level. This could serve as a self-selection of what might be more tolerant athletes in sports with high impact exposure than in sports where there is less contact. Regardless of regulations, contact with other players and contact with the boards still occurs in women’s ice hockey [40]. It has been proposed that because of the no-checking rule in women’s ice hockey, females may be less prepared when contact occurs [13]. This is important to consider when it has also been reported that anticipated collisions tend to result in less severe head impacts than unanticipated collisions [41].

Diagnosis of concussion is difficult in that it largely depends on a patient self-reporting symptoms and it has been suggested that athletes, especially those playing at a competitive level, underreport or even hide symptoms [42]. It has been postulated that females may be more likely to report their symptoms than male players [38]. Whether this reflects actual increased tissue damage or greater symptom intensity for a given biomechanical input, increased concern or focus on potential consequences of injury, or some other factor remains speculative. Regardless, if this were the case in our study, our results could be influenced by a reporting bias.

This study has several additional limitations. While concussion rates were normalized by impact exposure and athletic exposure, there may be variables that affect concussion rates that were not accounted for. These could include players that played on the same team but did not participate in the study. While almost 15% of the female athletes were diagnosed with a concussion during the course of the study, we still had a small sample size of female concussions (n=9) to report. There were also a low number of diagnosed concussions in the male ice hockey subject population and because of this we were unable to directly compare biomechanics of head impacts associated with diagnosed concussion between males and females playing the same sport at the same level. Statistical comparisons were based on diagnosed concussions in female ice hockey players and male football players. While combining data from the male football and hockey players resulted in significant differences between the sexes across all variables, we felt that it was inappropriate to combine all male athletes due to known differences in head impact exposure between the sports [10,12] and design differences between the HIT hockey and football systems. While men's ice hockey data were not including in the statistical analysis, we do note that the trends were consistent; male hockey players sustained a lower rate of concussion with higher impact exposure than females.

The analysis of single impacts associated with concussion did not take into account the number of impacts players sustained on the days of diagnosed concussion. The high number of diagnosed concussions that could not be clearly associated with a single impact (18/37) indicates there may be a cumulative effect of head impacts. Other studies have identified the difficulties in evaluating single impacts associated with concussion [33,43]. Schnebel et al. [44] highlighted these issues, including; a large amount of head impacts occurring within a short period of time, variability in symptom presentation and time resolving, and relying on injured athletes to accurately and timely report their symptoms. An additional limitation in evaluating single impacts associated with diagnosed concussions is the inherent error associated with the use of any measurement device, in this case the HIT System. To address this concern and further examine the relationship between the biomechanics of head impacts and diagnosis of concussion, we analyzed head impact exposure cumulatively for the days that players were and were not diagnosed with concussion injuries; however, it is not known whether a cumulative measure over a longer period of time would be more appropriate than limiting the analysis to a single day.

Another important consideration in this study is that diagnosis of concussion was represented as a binary state and did not acknowledge that there exists a spectrum of concussive injuries [43]. For all of the concussions reported in this study, there were differences, within and between the sexes, in the symptoms that the injured players presented with within 72 hours of diagnosis. The type of symptoms (headache, fatigue, dizziness) and the total number of different symptoms for each injury varied. The diagnosed concussions presented in this study exhibited a spectrum of signs and symptoms that may represent very different injuries. It is unlikely that any single biomechanical characteristic is the cause of all outcomes.

This is the first study, to the best of our knowledge, to report the biomechanics of impacts associated with concussions in collegiate ice hockey, and the first to report this data in a female population in general. Females sustained a higher rate of concussion than males, but the single impacts associated with concussions and head impact exposure on days of diagnosed concussion were significantly lower for female athletes compared to their male counterparts. These findings suggest that females have a higher rate of diagnosed concussion and may have a lower biomechanical tolerance to, or lower threshold for reporting symptoms of, concussion.

4.6 Acknowledgments

This manuscript is the third in a series of communications within Medicine & Science in Sports & Exercise by the collaborating authors investigating the biomechanical basis of mild traumatic brain injury through the use of in-vivo biomechanical data obtained from on-field head impact monitoring in sports.

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	n/t	Peak Linear Acceleration (g)	Peak Rotational Acceleration (rad/sec^2)	HITsp
Football	13/26	117.1 ± 43.0*	4576 ± 2179	73.1 ± 43.7
Men's Ice Hockey	2/2	71.0 ± 14.0	5313 ± 16	52.7 ± 15.0
Women's Ice Hockey	4/9	43.0 ± 11.5*	4029 ± 1434	25.6 ± 4.8

Table 4.1: Average (± st. dev) peak linear acceleration (*g*), peak rotational acceleration (rad/sec^2), and HITsp of single impacts associated with diagnosed concussion in collegiate football, men's and women's ice hockey players. (n=number of diagnosed concussion associated with a single impact, t = total number of diagnosed concussions for each population). *$p<0.05$

	Football	Women's Ice Hockey
# Impacts	$p < 0.001$*	$p = 0.023$*
Linear Acceleration (95th)	$p < 0.001$*	$p = 0.001$*
Rotational Acceleration (95th)	$p < 0.001$*	$p = 0.015$*
HITsp (95th)	$p < 0.001$*	$p = 0.012$*

Table 4.2: P-values for comparisons between head impact exposure on days with and without diagnosed concussions. Note: Male hockey players sustained a relatively low number (n=2) of diagnosed concussions and where not included in statistical analysis.

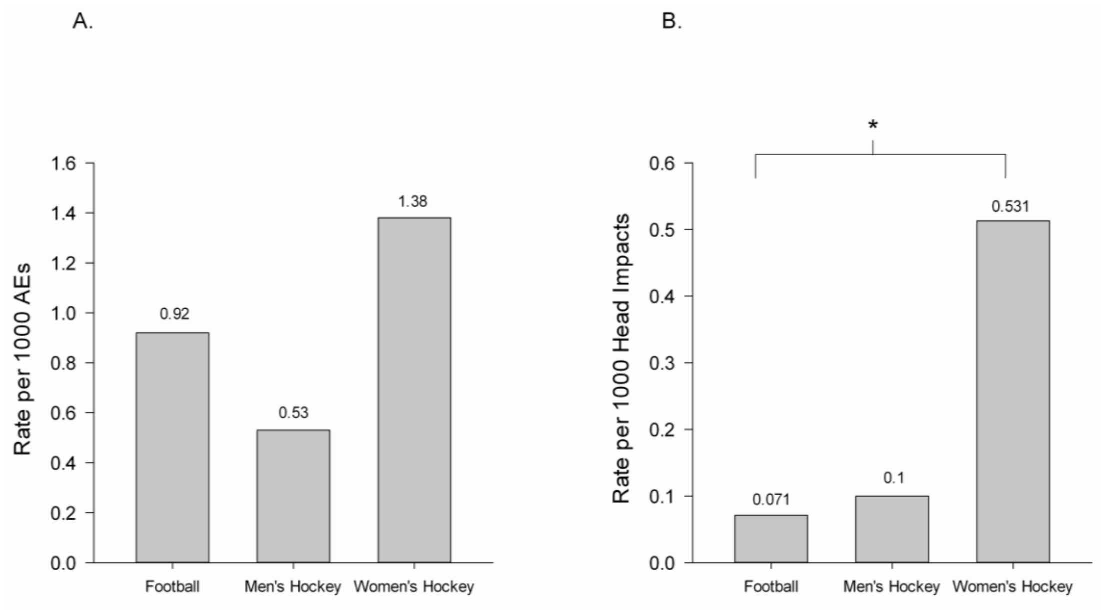

Figure 4.1: Concussion rate (A) per 1000 athletic exposures and (B) per 1000 head impacts for collegiate football, men's and women's ice hockey players. Note: Rates for male hockey are based on a small sample size (n=2).

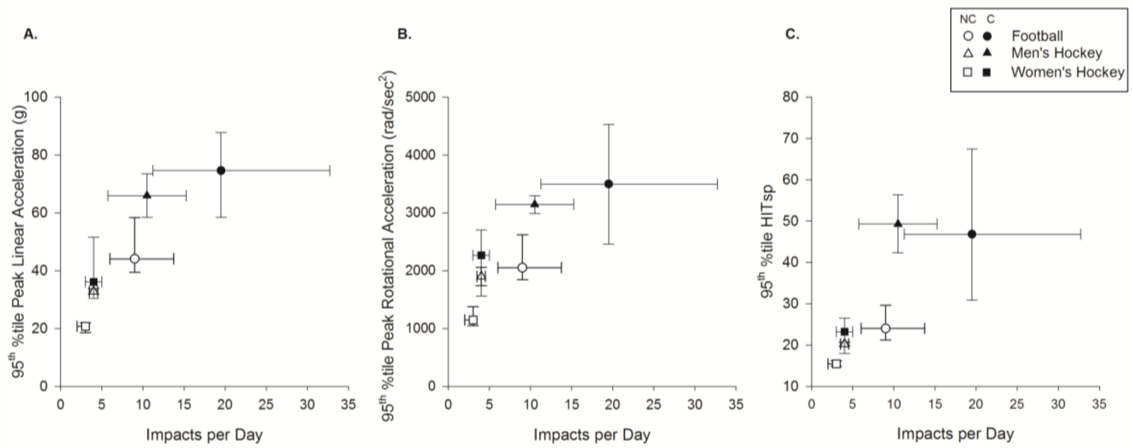

Figure 4.2: Median [25%-75%] of the (A) 95th percentile peak linear acceleration (*g*), (B) 95th percentile peak rotational acceleration (rad/sec^2), and (C) HITsp as a function of the median [25%-75%] frequency of impacts per day categorized by sport. Filled markers represent days with diagnosed concussion (C); unfilled markers represent days without diagnosed concussion (NC). See **Table 4.2** for statistical comparisons.

Chapter 5

Conclusions, Related Studies, and Future Directions

5.1 Conclusions

The studies and data presented in this dissertation describe an observational study that collected and quantified *in-vivo* biomechanical head impact data in several populations of contact sport athletes, correlated the data with clinical outcomes related to concussion, and assessed the differences between males and females in these measures. The long term objective of this work was to improve the understanding of the relationship between biomechanics of head impacts and concussion injuries to inform prevention strategies. The specific aims of the chapters presented in this thesis were to (1) quantify and compare the biomechanics of head impact exposures in men's and women's collegiate ice hockey players; (2) examine and compare head impact mechanisms in men's and women's collegiate ice hockey; and (3) evaluate differences between the sexes in concussion rate and the biomechanics of head impacts associated with MTBI. The topics covered in Chapter 1 outline the background and significance of the studies presented in this dissertation while explaining the purpose and hypotheses of the specific aims in detail.

5.1.1 Specific Aim 1

Quantify and compare the biomechanics of head impact exposures in men's and women's collegiate ice hockey players.

The first aim was accomplished with the study presented in Chapter 2. The goals of this study were to quantify head impact exposure (frequency, location and magnitude of head impacts) for individual male and female collegiate ice hockey players. Head impact exposure for collegiate ice hockey players was found to be dependent upon sex, session type, and team but not on player position. For both males and females, head impacts occur at a higher frequency in games than in practices but these impacts are not greater in magnitude. Impacts to the back of the helmet result in the greatest peak linear accelerations, while impacts to the side and back of the helmet are associated with high rotational accelerations. Male players have higher frequencies of head impacts and experience impacts greater in magnitude than females. Our findings also suggest, when compared to our previous studies in collegiate football, that head impact exposure for individual athletes is dependent upon which sport they play. We have proposed that reducing an individual's head impact exposure is a practical approach for reducing the risk of brain injuries [1]. Strategies to decrease an individual athlete's exposure need to be sport and gender-specific, with considerations for team and session type.

5.1.2 Specific Aim 2

Examine and compare head impact mechanisms in men's and women's collegiate ice hockey players.

The objective of the second aim was to examine and compare the per-game frequency and magnitude of head impacts associated with various impact mechanisms, or circumstances in which head impacts occur, in men's and women's collegiate ice hockey players. We accomplished this by synchronizing video footage from games with data from the HIT System and then classifying impacts into 8 head impact mechanism categories: contact with another player, the ice, boards or glass, stick, puck, goal, indirect contact, and contact from celebrating. While the most frequent head impact mechanism in both men's and women's collegiate ice hockey was contact with another player, contact with the ice resulted in head impacts with the greatest magnitude. Sex was found to be a factor in per-game frequency and magnitude of head impacts associated with several impact mechanisms. Men experienced head impacts from contact with another player and contact with the boards more frequently than women and these impacts were generally of greater magnitude. This difference can most likely be attributed to rule variations within sports by gender. Checking is an important part of the game in men's ice hockey whereas it is illegal in women's. It is interesting to note that, regardless of regulations, contact with another player was still the most common head impact mechanism in women's hockey.

Recently, research related to head injuries in sports has primarily focused on addressing two main concerns, the high rate of diagnosed and undiagnosed concussions and the potential long term effects of repetitive head impacts. If you assume that impacts of greater magnitude have more associated risk for concussion [2], it would appear that the strategies to address these two concerns may be different. To reduce frequency of head impacts in collegiate ice hockey, player contact rules should be addressed. To address high magnitude head impacts, re-evaluating helmet design to protect against the linear and rotational accelerations associated with head contact to the ice may be warranted. The identification of the impact mechanisms in collegiate ice hockey that result in frequent and high magnitude head impacts is an important step in understanding the high rate of concussions in the sport. These data could inform specific strategies for decreasing head impact exposure including rule changes, improved protective equipment design, and training regime development.

5.1.3 Specific Aim 3

Evaluate differences between males and females in concussion rate and the biomechanics of head impacts associated with diagnosis of concussion.

The objective of the third and final aim was to determine if the rate of diagnosed concussion, the biomechanics of single head impacts associated with diagnosed concussion and head impact exposure of individual athletes on days of diagnosed concussion differ between males and females in a study population of collegiate football, male ice hockey and female ice hockey players. There was a total of 37 concussions that were diagnosed during the course of the study, 26 in football players, 2 in men's ice hockey, and 9 in women's ice hockey. Female hockey players were found to have a significantly higher rate of diagnosed concussion (0.51 per 1000 head impacts) than male football players (0.07 per 1000 head impacts). In other words, one concussion was diagnosed in every 1,883 head impacts in women's ice hockey, whereas one concussion was diagnosed in every 14,085 head impacts in football. While females had a higher rate of diagnosed concussion than males, single impacts associated with diagnosed concussion and head impact exposure on days of diagnosed concussion were significantly lower for female hockey players than male football players. These trends were also consistent between male and female hockey players. These findings suggest that females have a higher rate of diagnosed concussion and may have a lower biomechanical tolerance to, or lower threshold for reporting symptoms of, concussion injuries.

5.2 Related Studies

There are a number of related studies and future directions that would further address the aims presented in this dissertation. The following subsections provide a brief overview of three related studies that were the foundation of this dissertation and two additional research avenues afforded by the primary and related studies presented in this dissertation.

5.2.1 Frequency and Location of Head Impacts in Collegiate Football (Appendix A).

The first related study was conducted to quantify the frequency and location of head impacts that individual players received in one season among three collegiate teams, between practice and game sessions, and among player positions. We found that an individual player can receive as many as 1400 head impacts during a single season. The average number of head impacts per game was nearly three times greater than the average number of head impacts per practice. There were significant differences in impact frequency and impact location among different player positions. We also found differences in head impact frequency among teams but it is unclear if this is related to differences among the players themselves, coaching approaches, or other factors that remain to be identified. These data document head impact exposure in terms of frequency and location sustained by individual players in college football, which varies according to practice vs. game, player position, and team. This data could aid football helmet manufacturers in establishing design specifications and governing bodies in setting testing criteria, and, with further studies, provide clinicians and scientists with a more complete understanding of the relationship between head impact exposure, concussion injury, and long-term cognitive deficits.

5.2.2 Magnitude of Head Impacts in Collegiate Football (Appendix B).

The second related study was conducted to quantify the severity of head impacts sustained by individual collegiate football players and to investigate differences between impacts sustained during practice and game sessions, as well as by player position and impact location. We found that an individual collegiate football player receives head impacts of varying magnitudes during play, and that the magnitudes of these impacts are heavily skewed towards lower values. Interestingly, the magnitude of impacts during games was not significantly greater than the magnitude of the impacts during practices except for the 95th percentile HITsp. We also found that there were significant differences in the magnitude of impacts among different player positions. Impact location was found to be a factor strongly associated with the differences in magnitude of peak linear acceleration and peak rotational acceleration. The interaction of player position and impact location also yielded statistically significant differences in impact magnitude. This study has provided a detailed description of the magnitude of head impact exposures for collegiate football players that will be critical for establishing the relationship between head impact biomechanics and the risk of concussion injury and for developing appropriate concussion injury prevention strategies.

5.2.3 Head Impact Exposure in Collegiate Football (Appendix C).

The third related study built upon the two previous (related) studies and examined head impact exposure by quantifying the frequency, magnitude, and location on the helmet of all impacts to individual collegiate football players among various player positions. Interestingly, the positions that sustained the highest frequency of head impacts where not the positions that sustained the greatest magnitude head impacts. Running backs (RB) and quarter backs (QB) received the greatest magnitude head impacts, while defensive line (DL), offensive line (OL) and line backers (LB) received the most frequent head impacts (more than twice as many than any other position). Impacts to the top of the helmet had the lowest peak rotational acceleration (2387 rad/s^2), but the greatest peak linear acceleration (72.4*g*), and were the least frequent of all locations (13.7%) among all positions. OL and QB had the highest (49.2%) and the lowest (23.7%) frequency, respectively, of front impacts. QB received the greatest magnitude (70.8*g* and 5428 rad/s^2) and the most frequent (44% and 38.9%) impacts to the back of the helmet. This study quantified head impact exposure in collegiate football, providing data that is critical to advancing the understanding of the biomechanics of concussive injuries and sub-concussive head impacts.

5.2.4 Head Accelerations from Various Stick Checks in Girls Lacrosse: A Surrogate Impact Study (Appendix D)

The first objective of the fourth related study was to determine head accelerations associated with various stick checks experienced in girl's lacrosse in a laboratory setting. This was accomplished by measuring stick speed and surrogate headform accelerations in experienced female youth players. We found that headform accelerations increased with increasing stick speed. The midrange of all swing speeds was found to be 6.3-10.3 m/s or 14.1-23.0 mph. Stick checks within the midrange swing speed resulted in average peak linear accelerations of approximately 50*g*. While 50*g* is substantially lower than accelerations typically associated with diagnosed concussions (approximately 100*g* [2–4]), previous on-field studies may not be directly comparable to our laboratory test. Additionally, factors that influence biomechanical tolerances to concussion injury, such as age and sex, are still not well understood.

The second objective was to characterize the response of various commercially available helmets to impacts of comparable magnitude. The four different helmet or headgear included; men's lacrosse helmet, rubgy scrum cap, women's soft headgear recommended for use in lacrosse, field hockey, and soccer, and an ultimate fighting (UFC) headgear. For impacts of the same swing speed, peak linear accelerations were decreased for all helmeted impacts compared to those delivered to the bare headform. For impacts to the side of the head, the highest decrease in acceleration was seen in the men's lacrosse, followed by the UFC, the soft head gear, and then the rugby scrum cap. As expected, the helmets and headgear lined with more material decreased the peak accelerations of the headform the most drastically. This is consistent with laboratory tests that have shown that increases in thickness and density of headgear foam can improve impact performance [5].

There is an inherent risk of injury in any sport. Understanding head impact exposure for individual athletes and the impact response of currently available protective equipment to comparable impact exposure is an important first step in understanding preventative strategies for head injuries. These data provide scientific evidence to assist governing bodies, rule-making bodies, coaches and parents in making informed decisions on all issues related to protective head gear in the sport of girl’s lacrosse.

5.3 Future Directions

5.3.1 Salivary Microvesicles as a Novel Biomarker for MTBI and Head Impact Exposure (Appendix E)

With the many difficulties and complexities associated with the recognition and diagnosis of MTBI, there is a clear need for the development and validation of on-field diagnostic tools. Concerns over the acute and chronic impairments as a result of MTBI, including the risk of long-term cognitive deficits due to repeated injuries [6] and second impact syndrome, has lead to the development of a variety of strategies to enhance immediate diagnosis. These experimental diagnostic tools range from sideline cognitive tests to neurophysiologic measures. Biomechanical data from the HIT system can be integrated with existing and newly developed clinical outcome measures and diagnostic tools for MTBI. Recent studies have evaluated microvesicles as possible biomarkers in different disease states, including traumatic brain injury [7–10]. Microvesicles or exosomes are fragments of plasma membrane that are shed by all cell types, under both normal and pathological conditions, and carry internal cargo representative of their cell of origin, such as membrane and cytoplasmic proteins and mRNA [11–13]. It has been proposed that vesicles in the brain significantly increase with head trauma and that these vesicles will migrate to the olfactory bulb and then to the nasal/oral cavity where they can be detected in saliva. Saliva has long been used as a diagnostic medium because it is accessible, inexpensive, non-invasive, requires minimal training for collection, and can be used in mass screening of large populations [14–16]. Microvesicles within saliva can be identified by their cell of origin and can be quantified by presence (number), size, and composition. There is potential for microvesicles to not only be an on-field biomarker for the diagnosis of concussion, but also a tool for identifying players who have sustained repeated subconcussive impacts and may be at risk of injury or long term cognitive deficits. The efficacy and feasibility of using salivary microvesicles and other clinical outcome measures related to MTBI could be evaluated by studying the relationship between these variables and head impact exposure. Details on a proposed study and results from a small pilot study can be found in Appendix E of this dissertation.

5.4 Summary

In our approach to understanding the biomechanics of concussions, we have defined *head impact exposure* as a multifactorial term that includes the frequency, magnitude, and impact location of head impacts for individual athletes. The primary objectives of this dissertation were to quantify head impact exposure in several populations of contact sport athletes, examine these variables in relation to head impact mechanisms, correlate this data with clinical outcomes related to concussion, and assess the differences between males and females in these measures. There were several key findings. We found that head impact exposure for individual athletes is sport and gender-specific, with considerations for team, session type, and position. We found that a relationship exists between head impact exposure and diagnosis of concussion. Based on these findings, we have proposed that reducing an individual's head impact exposure is a practical approach for reducing the risk of brain injuries [1]. In order to investigate strategies for reducing head impact exposure, we developed a tool that synchronized impact data with game video footage in order to evaluate the biomechanics of head impacts associated with specific head impact mechanisms (e.g. head contact with the ice in hockey) to determine which circumstances of play result in frequent or high magnitude head impacts [17]. Finally, we evaluated differences between males and females in concussion rate and the biomechanics of head impacts associated with diagnosed concussion. Because we quantified head impact exposure for individual athletes, we were able to not only evaluate rate of concussion per athletic exposure but also per head impact. Female hockey players had significantly higher rates of concussion per head impact than male football players. Approximately fifty percent of the concussions that were diagnosed during the course of the study were associated with a single impact. This indicates that there may be a cumulative effect of head impacts and led us to examine not only single impacts associate with the injury but also the head impact exposure for the injured athletes on the days of concussion. While females had higher rates of concussions than males, single impacts associated with concussion and head impact exposure on days of diagnosed concussions were lower for female hockey players compared to male football players. These findings suggest that females may have a lower biomechanical tolerance to, or lower threshold for reporting symptoms of, concussion injuries.

Ultimately, these data provide a better understanding of the relationship between biomechanics of head impacts and diagnosed concussions, and the differences between males and females in that relationship. The outcomes of this study allow for the development and validation of improved on-field MTBI management (including early detection and accurate diagnosis), prevention strategies, and interventions. Additionally, data from this study may inform and improve future helmet design.

5.5 References

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Appendix A

Frequency and Location of Head Impacts in Individual College Football Players

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Project contributions: data collection & analysis

A.1 Abstract

A.1.1 Purpose

Measuring head impact exposure is a critical step towards understanding the mechanism and prevention of sports-related mild traumatic brain (concussion) injury, as well as the possible effects of repeated subconcussive impacts. The purpose of this study was to quantify the frequency and location of head impacts that individual players received in one season among three collegiate teams, between practice and game sessions, and among player positions.

A.1.2 Methods

188 players from three NCAA collegiate football teams wore helmets instrumented with an accelerometer-based system during the fall season of 2007. The number of head impacts greater than 10*g* and location of the impacts on the player's helmet were recorded and analyzed for trends and interactions among team, session type, and player positions using Kaplan-Meier survival curves.

A.1.3 Results

The total number of impacts players received was non-normally distributed and varied significantly by team, session type and by player position. The maximum number of head impacts for a single player on each team was 1022, 1412, and 1444. The median

number of head impacts on each team was 4.8, 7.5 and 6.6 impacts per practice and 12.1, 14.6, and 16.3 impacts per game. Linemen and linebackers had the largest number of impacts per practice and per game. Offensive linemen had a significantly higher percentage of impacts to the front of the helmet, while quarterbacks had a significantly higher percentage to the back of the helmet.

A.1.4 Conclusions

The frequency of head impacts and the location on the helmet where the impact occurs are a function of player position and game versus practice sessions. These data provide a basis for quantifying specific head impact exposure for studies related to understanding the biomechanics and clinical aspects of concussion injury as well as the possible effects of repeated subconcussive impacts in football.

A.2 Introduction

Concussion injuries, most often due to head impacts, are a growing concern in sports [1–6]. However, the biomechanical factors that relate a particular impact or a series of impacts to the clinical signs and symptoms of concussion injury, second impact syndrome, or delayed cognitive sequelae have not yet been established. Several animal models suggest that repeated impacts [7,8] and direction of impact or head rotation [9–12] influence clinical and pathophysiologic consequences of injury. To investigate these relationships and the many factors potentially involved in acute and chronic effects of head impact, numerous researchers have turned to the sports field as their laboratory. Quantifying events occurring in the experimental environment of the sports field is important for understanding the factors relevant to impacts that result in acute symptoms as well as whether there might be subacute or long-term effects of repeated impacts, even when no symptoms are immediately apparent.

In a study of collegiate sports injuries by the NCAA [13], exposure to the risk of injury is measured as an athlete-exposure (AE), which is defined as one student-athlete participating in one practice or competition in which he or she was exposed to the possibility of athletic injury, regardless of the time associated with that participation. The measure of AE does not account for the magnitude or frequency of head impacts to individual players. For example, two athletes who participate in the same game, both of whom would have experienced one AE, might experience a significantly different number of head impacts. Furthermore, these same two players might receive the same number of head impacts, but the impacts may be of very different magnitudes, and at different locations to their heads. Because traumatic brain injuries are likely to occur along a broad continuum, be cumulative, and involve pathophysiologic events which may occur without evidence of acute injury symptomatology, the concept of "exposure" needs to incorporate these elements in order to best understand individual player risk and potential prevention.

We propose to define "head impact exposure" as a broad term that incorporates multiple variables. Multiple measures of head impact exposure are critical at this juncture because the specific variable or combination of variables that correlate with the risk of head injury has not yet been determined. The first important variable of head impact exposure is athletic-exposure [13], which is well suited for understanding the overall risk of head injury per session of participation. A more complete understanding of the risk of head injury requires additional quantifiable variables that we propose to include: the magnitude of the head impact, the number or frequency of head impacts, the location of the head impact, and cumulative measures of head impacts.

Obtaining detailed information on magnitude and frequency of head impacts to individual players has been challenging. Videotaping of athletic events can provide some insight into injury mechanisms, but has limited practicality because it has limited ability to continuously track all players, cannot accurately identify all impacts to the helmet, and cannot provide a direct measure of impact magnitude. The challenges associated with using video tapes to study head impacts in football are well illustrated in the studies of professional football players by Pellman et al. [14,15]. Despite the prevalence of head impacts and the amount of video coverage, only 31 of 182 of known concussive events were available for their analysis because the impacts were required to be in the open field and to be recorded from at least two unobstructed views. The magnitude of the head impacts, which could not be recorded or quantified directly from the video, were then computed by reconstructing the impact scenario in the laboratory using Hybrid-III crash-test dummies.

To measure the specific details of head impact exposure in sports participants, a variety of systems have been previously developed and implemented [16–18]. Early prototypes required players to wear obtrusive data acquisition hardware that required manual data download following each activity: consequently, these studies were limited in the number of athletes and session data that was collected. The Head Impact Telemetry (HIT) System technology [19–21] (developed by Simbex, Lebanon, NH and marketed commercially as the Sideline Response System by Riddell, Elyria, OH) was specifically designed to address these limitations by measuring the biomechanical factors associated with head impacts for a large number of players, without interfering with the play of the game. The HIT system is an accelerometer-based system mounted inside a football helmet that is able to directly measure head acceleration and location of head impacts.

Several studies have used the HIT system to study head impacts to football players [22–26]. These studies have provided new insights into the biomechanics of head impacts in football by examining the number of impacts and the magnitude of the resulting head accelerations across teams and groups of players. While each study reported the total number of impacts per team recorded in their study, detailed analyses of the head impact exposure for individual players were not reported. To increase our understanding of the biomechanics of concussion injuries and the potential cognitive effects related to single or repeated head impacts, we sought to analyze head impact exposures for individual players by focusing on two specific measures of exposure.

The purpose of this study was to quantify head impact exposure by recording the frequency and location of head impacts that individual players received in one season. We tested the hypotheses that head impact frequency and helmet impact locations would differ among three collegiate teams, between practice and game sessions, and among player positions.

A.3 Methods

Players from three NCAA football programs (Brown University, Dartmouth College, and Virginia Polytechnic Institute) were provided the opportunity to participate in this observational study after IRB approval and informed consent. Two teams play in the Ivy League of the NCAA, a Football Championship Subdivision (FCS) league that does not allow post-season play and one team plays in the NCAA Football Bowl Subdivision (FBS). During this study, all three teams participated in approximately the same number of games, but one school had almost twice as many practices as the other two schools. During the 2007 fall football season, a total of 188 players from the three teams, denoted arbitrarily as Team A (n = 65 players), Team B (n = 60 players), and Team C (n = 63 players), participated in this study. Each player was assigned a unique identification number and categorized in one of eight position units defined by the team staff as the player's primary position: defensive line (DL, n = 29), linebacker (LB, n = 29), defensive back (DB, n = 34), offensive line (OL, n = 46), offensive back (OB, n = 23), wide receiver (WR, n = 16), quarterback (QB, n = 8), and Special Teams (ST, n = 3).

All players wore Riddell football helmets (Riddell, Chicago IL) instrumented with the HIT System (**Figure A.1A**), a device capable of recording the acceleration-time history of an impact from six linear accelerometers at 1000 Hz. The HIT System continuously samples all six accelerometers during play. When a pre-set threshold for a single accelerometer channel exceeded 14.4 *g*, 40 ms of data (8 ms pre-trigger and 32 ms post-trigger) is transmitted to a sideline receiver connected to a laptop computer. From the acceleration-time histories, the severity (magnitude of linear and rotational acceleration) and duration of the head acceleration, along with the location of the impact on the helmet, are computed and stored for future analysis [19,20] Head impact data from all participating institutions were uploaded to a secure central server with a consolidated database, and subsequently exported for statistical analysis (SAS, Chicago, IL). Data were reduced in post-processing to exclude any impact event with a peak resultant linear acceleration less than 10g [24] in order to eliminate events that had been determined during initial system development to be inconsequential, non-impact events (e.g. running, jumping, etc.). Any impact event in which the acceleration-time history pattern of the six linear accelerometers did not match the theoretical pattern for rigid body head acceleration [19], such as a single accelerometer spikes which can occur during throwing or kicking a helmet, was also excluded. Finally, all impacts exceeding 125 *g* were visually reviewed to verify quality of acceleration data. These methods have been previously verified by comparing measured impacts with video footage [23,25].

A team session was defined as a formal practice (players wore protective equipment with the potential of head contact) or a game (competitions and scrimmages). A player session was defined when at least one head impact was recorded during one team session, as this provided confirmation that the given participant was present and was exposed to impact. Impacts that were recorded outside of the time of the team session, as defined by the team staff, were excluded from the analysis. Head impact data were recorded during a total of 215 team sessions (172 practices and 43 games during the 2007 fall season). The number of sessions that were analyzed for each player ranged from 1 to the maximum number of possible team sessions for each individual school (**Table A.1**).

Head impact frequency and the location of the impacts on the helmet were analyzed for each player by team, session, and position. Head impact frequency was quantified using five measures: season impacts was the total number of head impacts recorded for a player during all sessions; practice impacts was the total number of head impacts recorded for a player during all practices; game impacts was the total number of head impacts recorded for a player during all games; impacts per practice was the average number of head impacts for a player during practices; and impacts per game was the average of the number of head impacts for a player during games. These data are plotted in **Figures A.2, A.4** and **A.5** using cumulative histograms and ordinary histograms of impact events sustained by members of each team. By presenting the data as cumulative histograms, we are able to present values for number of impacts for individual players, normalized for the total number of players on each team. For example, if every player on the team received exactly 200 impacts during a season impacts, then the curve in **Figure A.2** would simply be a vertical line at the 200 value on the x-axis. As described in more detail below, the data were non-normally distributed and hence the resulting curves possessed a complex shape.

Impact *locations* to the helmet and facemask were computed as azimuth and elevation angles in an anatomical coordinate system relative to the estimated center of gravity of the head [19] and then categorized as front (F), left (L), right (R), back (B), and top (T) (**Figure A.1B**). Four equally spaced regions centered on the anatomical mid-sagittal and coronal planes defined front, left, right and back impact locations. All impacts occurring above an elevation angle of 65°, where 0° elevation was defined as a horizontal plane through the center of gravity of the head were defined as impacts to the top of the helmet [26].

Statistical Analysis

To determine if season impacts, practice impacts, game impacts, impacts per practice and impacts per game for individual players were different among teams, we used the Wilcoxon test for comparing Kaplan-Meier survival curves. Comparing the survival curves provided both a compelling visualization and a valid (non-parametric) method for comparisons of the positively-skewed data of season impacts, practice impacts, game impacts. For consistency, similar analyses were carried out for the rate variables (impacts per practice and impacts per game), though these measures were not significantly skewed. Since there were only three possible post-hoc comparisons among the 3 teams, p-values were adjusted using the Bonferroni method so that statistical significance was evaluated per comparison with $\alpha < 0.0167$. Analyses were carried out using SAS version 9.2 (SAS Institute, Cary, NC).

The relationship between the number of season head impacts a player received and the number of NCAA-defined Athletic Exposures (AE) was examined using linear regression (SigmaPlot, Systat Software, San Jose, CA).

To determine if impacts per game and practice were different among players of the various positions we used a mixed linear models with fixed effects for team, position, and activity, and random effects for position and activity within each player. The percentage of impacts at various locations within position (e.g. left vs. right in WR), as well as location differentials between positions (e.g. left vs. right in WR as compared to QB), were compared using mixed linear models with fixed effects for team, position, and location, and random effects for position and location within each player. The Holm test was used to adjust p-values for multiple comparisons, owing to the large number of comparisons. Analyses were carried out using SAS version 9.2 (SAS Institute, Cary, NC).

A.4 Results

The number of *season impacts* for players on Team C were significantly higher than those for players on Team A ($P < 0.0003$) and tended to be higher than those on Team B (**Figure A.2**). The number of season impacts for players did not differ significantly between Team A and B. The maximum number of season impacts were 1022, 1412 and 1444 among players on Team A, B, and C, respectively. The median values (i.e., 50% of the players on a team had a higher number of season impacts and 50% of the players had a lower number of season impacts) were 257, 294, and 438 on Team A, B, and C, respectively. The percentage of players receiving any given number of season impacts are plotted in the cumulative histograms of **Figure A.2A**, while the percentage of players receiving season impacts in bins of 200 impacts are plotted in the ordinary histogram of **Figure A.2B**.

Across all players in the study, the number of season impacts increased significantly ($P < 0.001$; $R^2 = 0.415$) with Athletic Exposure (AE), defined by the NCAA as 1 player in 1 session in which the athlete is exposed to the possibility of athletic injury [13]. However, the variability in the number of season impacts for a given AE increased substantially as the number of AE increased (**Figure A.3**). For example, the number of season impacts for players with an AE value of 50 ranged from 175 to 1,405.

The number of *practice impacts* were significantly less for players on Team A than those for players on Team C ($P = 0.0022$) (**Figure A.4A and A.4C**). There were no differences in the number of practice impacts between players on Teams B and C and between players on Teams A and B (**Figure A.4A**). The maximum number of practice impacts were 761, 811 and 910 among players on Teams A, B and C, respectively. The median value for practice impacts was 160, 207 and 210 among players on Teams A, B and C, respectively.

The number of *game impacts* were significantly higher for players on Team C than for players on Team A ($P < 0.0001$) and for players on Team B ($P = 0.0012$) (**Figure A.4B and A.4D**). Game impacts for individual players on Teams A and B did not differ. The maximum number of game impacts were 351, 601 and 775 among players on Teams A, B and C, respectively. The median values (i.e., 50% of the players had a higher number and 50% of the players had a lower number) among players on Teams A, B and C were 79, 102, and 173, respectively.

The *number of head impacts per practice* (the number of practice impacts normalized by the number of practice sessions for each individual player) was significantly lower ($P < 0.0001$) for players on Team A than for either Team B or C (**Figure A.5A**). Impacts per practice were not different between Teams B and C. The maximum (and median, in parentheses) values for the number of impacts per practice f were 15.6 (4.8), 18.9 (7.5), and 24 (6.6) among players on Teams A, B and C, respectively. In contrast to practices, the *number of head impacts per game* did not differ by team (**Figure A.5B and A.5D**). The maximum (and median) values for head impacts per game were 58.5 (12.1), 66.8 (14.6), and 86.1 (16.3) among players on Teams A, B and C, respectively.

After grouping players by their position, the number of impacts per practice and per game did not differ among teams. The number of impacts per practice ranged among positions from 3.2 (QB) to 11.5 (DL) (**Figure A.6A**). The number of impacts per game ranged from 7.3 (WR) to 29.8 (DL) (**Figure A.6B**). This significant ($P < 0.001$) increase in the number of head impacts per game, as compared to head impacts per practices, was relatively constant for all positions with a coefficient of 2.4 times ($r^2 = 0.92$). In general, DL, LB and OL had a greater number impacts per practice and impacts per game than did players at the other positions.

Across all players, the highest percentage of impacts occurred to the front of the helmet (**Figure A.7**). The back of the helmet received the second highest percentage of impacts. When examined by positions, DB, DL, LB, and OL had a significantly higher number of impacts to the front than to the back of the helmet ($P<0.0001$). The OL had the highest percentage of impacts to the front of the helmet compared to any other position. Conversely, QB had a significantly higher percentage of impacts to the back than to the front of the helmet ($P = 0.0015$). The percentage of impacts to the front and the back were not different for WR. There was no difference between impacts to the left and to the right side at any position. Impacts to the top of the helmet occurred more often than impacts to either side, but this difference between top and sides reached statistical significance only for OL ($P < 0.0001$).

A.5 Discussion

In this study we sought to quantify head impact frequency and location for individual players on three collegiate football teams during a single season. We focused on these two measures of head impact exposure because of the lack of data on individual exposures and on concussion injury mechanisms. To date the only reported exposure measure for individual players is the risk of injury through participation defined using athlete-exposures (AE), in which one AE is one practice or game [13]. Although AE is a useful factor for comparing the risk of injury across sports, gender and other environmental factors, it has limited applicability to the study of injury mechanisms.

The ability to directly measure head impacts of individual players is critical to establishing the relationship between head impacts and concussion injury, and to examining the potential effects of cumulative subconcussive impacts. Only a few players of the 188 players enrolled in this study received an impact in all team practices and games. More typical in our study were head impacts that occurred in approximately one-half to two-thirds of the team sessions. Head impact frequency recorded over the entire season for practices and games varied by team. This was not unexpected given that players on one team had substantially more practices sessions, but this team also had the lowest median number of average impacts per practice. While the number of team sessions certainly influenced the number of individual head impacts, the structure of the practice plan and the philosophies of the coaching staff were also a likely factors that are difficult to quantify. Interestingly, the number of impacts a player received per game did not vary by team. We presume this is due to the controlled and timed nature of football games, which are less dependent on a team's style of play or specific practice tendencies. The total number of impacts players received during all practices and during all games were comparable (**Figure A.4**); however, after accounting for the number of sessions of each, the number of impacts per game were 2 to 3 times greater than those per practice, which is consistent with the findings that injury rates are higher in games than in practices [27]. The number of impacts recorded per practice and per game for an individual player reached a maximum of 24 and 86, respectively. The median values for these players were 6.3 impacts per practice and 14 impacts per game. For some individual players the values we recorded may be underestimates of the actual impacts because a player may have started a practice or game but not completed the session, and we were not able to instrument all players on each team.

Previous studies have not reported head impact measures for individual players so direct comparison with our data is limited, but instructive. Using an earlier version of the instrumented helmet technology, Duma et al [25] reported 2,114 impacts in 35 practices while monitoring 38 different players (up to eight players a session) giving a value of approximately 7.6 impacts per player per practice. In 10 games, they recorded 1,198 impacts for an estimated 15.0 impacts per player per game. Brolinson et al. [23] recorded 11,604 impacts over 84 sessions of games and practices. During each session up to 18 players were monitored, with a total of 52 different players wearing the instrumented helmets over the two-season period. From their results we estimate that the average number of impacts per player per session were approximately 4, which would be in the lower 20% of the 188 players from our study. It is likely, however, that this prediction of impacts per player per session is an underestimate considering that 18 players were not instrumented each day for the entire duration of the study. Schnebel et al. [22] , using similar technology, reported a total of 54,154 impacts for 40 players over 105 sessions at a single NCAA Division 1 school during a single season. Their overall average number of player head impacts per session was approximately 13, which was greater than our median value of 9.4 impacts per player per session. Mihalik et al. [24] reported that the total number of impacts sustained in full-contact practice (28,610) were about twice as many as those sustained in games (12,873). This ratio is roughly consistent with our findings.

We found player position affected both head impact frequency and location. Previous studies have suggested similar trends. Schnebel et al. [22], reported their non-linemen ("skill positions") received only 25% of the total impacts in contrast to linemen, who received 75% of the total impacts. In another study of one collegiate football team over two seasons, the largest percentage of impacts were recorded in offensive and defensive linemen (36% and 22%, respectively) [24], which is consistent with our findings. In that study linebackers received only a 1/3 of the number of impacts of the linemen, whereas in our study linemen and linebackers received approximately the same number of impacts per practice and per games.

We found that the majority of impacts occurred to the front of the helmet for all player positions, except QB. OL had the highest percentage of impacts to the front of the helmet, which is consistent with the observation that OL are more likely to initiate and control the site of impact than other position groups. The highest percentage of impacts to the back of the helmet occurred in QB, suggesting that the QB were most often hit from behind or were tackled falling backwards, hitting the back of their head on the ground. These explanations are based upon general observation of football and are not yet confirmed by video analysis. Mihalik et al. [24] did not examine impact location by player position, but their overall results on impact location are in general agreement with our findings for all players.

In the present study we focused our analysis on head impact frequency and impact location for individual players. We chose this focus because this analysis for individual players has not been previously reported and the resulting data are crucial in establishing baseline exposures for the mechanism and the risk of concussion injury, as well as the risk, if any, of cumulative subconcussive injury. Accordingly, a substantial amount of data from our project was not reported in the present study. The severity (magnitude) of the linear and rotational acceleration and the time duration of head acceleration during impacts were not reported, as these are the subject of an ongoing analysis of specific biomechanical input variables and their relationship to symptoms and cognitive function. Additionally, cumulative measures of head impacts have yet to be formulated and hence were not included in this analysis. Concussion injuries and any measure of long-term cognitive deficits were not reported here. Our study is also limited to three teams during a single football season. Our current multi-year study is ongoing and differences, if any, between seasons will be analyzed as the study continues. The impact frequency reported here may slightly underestimate the true frequency for individual athletes given that there were instances when the helmet instrumentation ran out of battery life, or where there was data loss during transfer from the helmet to the sideline computer. We selected a lower range cutoff of 10 g's of peak linear acceleration of the head for inclusion as an impact to be consistent with data collection thresholds across the three test sites. In our statistical analyzes, given the size of the dataset and number of levels of within-subject (5 x location x 2 x activity) and between-subjects (3 x team) factors, not all sources of heterogeneity of variance could be tested. While we believe the numbers of samples would minimize this effect, it is possible that heterogeneity of variance across some factors could impact the mixed model analyzes.

Head impact in sports continues to be a significant and growing concern at all levels of football and other sports due to the known adverse outcomes in some cases and the potential for long term detrimental cognitive effects. At present, the exact mechanisms for and variability of concussion signs, symptoms, and long-term sequelae from head impacts are not well understood, particularly in helmeted sports. There are few data to address possible differences in mechanisms and susceptibilities among athletes of different ages, including children, and between the sexes. Estimates of concussion injury thresholds based upon laboratory reconstructions using animal, cadaver, and manikin surrogates [15,28,29] have proven to be inadequately predictive of injury when compared with measurements of actual head acceleration [26]. In order to appropriately evaluate the risk of concussion injury and the potential for interventions likely to reduce the incidence of concussions in sports, as well as the potential role of accumulated subconcussive events, a detailed understanding of the exposure and the mechanism of injury is needed. Animal models suggest that multiple factors likely influence the risk of neurologic and somatic symptoms after “concussive” head impacts, and these may include prior head impact events, site or direction of head impact, and other mechanical and physiologic factors [7–12]. The data presented in this manuscript begin this process of quantifying head impact exposure in collegiate football by focusing on head impact frequency and location.

In summary, we found that an individual player can receive as many as 1400 head impacts during a single season. The average number of head impacts per practice was 5 to 8, and these values nearly tripled for games. There were significant differences in impact frequency and impact location among different player positions. We also found differences in head impact frequency among teams but it is unclear if this is related to differences among the players themselves, coaching approaches, or other factors that remain to be identified. There was no difference among teams in the average number of head impact per game. These data document head impact exposure in terms of frequency and location sustained by individual players in college football, which varies according to practice vs. game, player position, and team. This data could aid manufacturers of football helmets in establishing design specifications and governing bodies in setting testing criteria, and, with further studies, provide clinicians and scientists with a more complete understanding of the relationship between head impact exposure, concussion injury, and long-term cognitive deficits.

A.6 Acknowledgments

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Team	Players	Team Sessions		Player Sessions	
		Practices	Games	Practice [max. (median)]	Game [max. (median)]
Team A	65	76	14	55 (32)	9 (6)
Team B	60	48	14	48 (28)	12 (6)
Team C	63	48	15	46 (37)	15 (10)
Total	**188**	**172**	**43**		

Table A.1: Data collection occurred in a total of 215 team sessions (172 practices and 43 games) across the three teams. A player session was defined as 1 session (practice or game) in which a player received at least 1 head impact. The maximum number of sessions for an individual player from each team ranged from 1 to the number of team sessions (practices plus games).

Figure A.1: (A) Football players wore instrumented helmets during practices and competitions to record the frequency, magnitude, and location of head impacts. The HIT System is comprised of an in-helmet unit containing six accelerometers, a sideline receiver, and a laptop computer. (B) An illustration of the regions that defined the front (F), right side (R), back (B) and top (T) impact locations on the helmet and facemask.

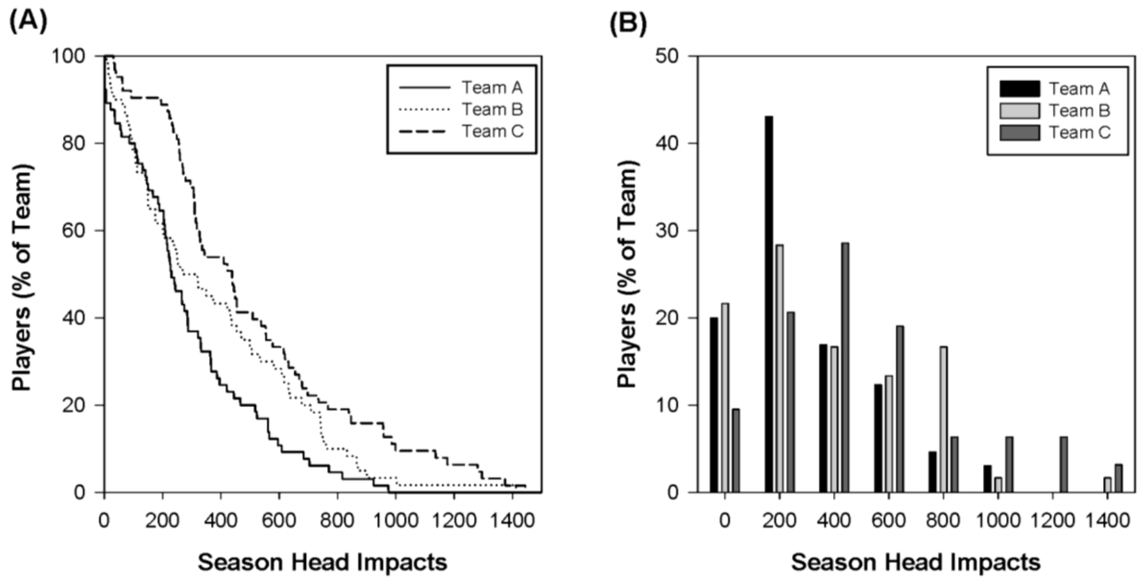

Figure A.2: The total number of head impacts for individual players during the season differed significantly with team. On the left (A) the complete distribution of the number of season head impacts is plotted for all players of each team. The data are plotted as a cumulative sum; on the x-axis is the number of season impacts and on the y-axis are the number of players, as a percentage of the team, with the given number of season impacts, or greater. On the right (B) is an ordinary histogram of the same data.

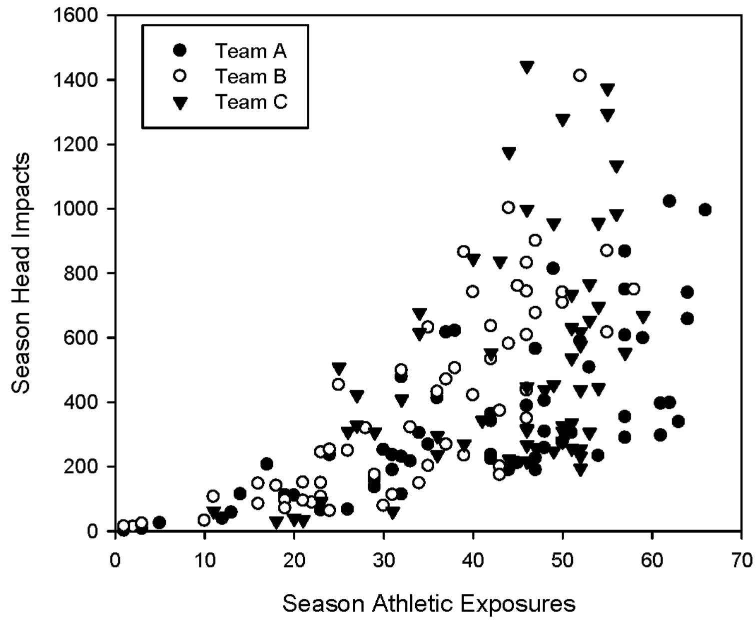

Figure A.3: Season head impacts increased with Athletic Exposure (AE), defined as 1 player in 1 session in which he or she is exposed to the possibility of athletic injury; however, AE was a poor predictor (R^2 = 0.412) of the number of season head impacts head impacts for any given player.

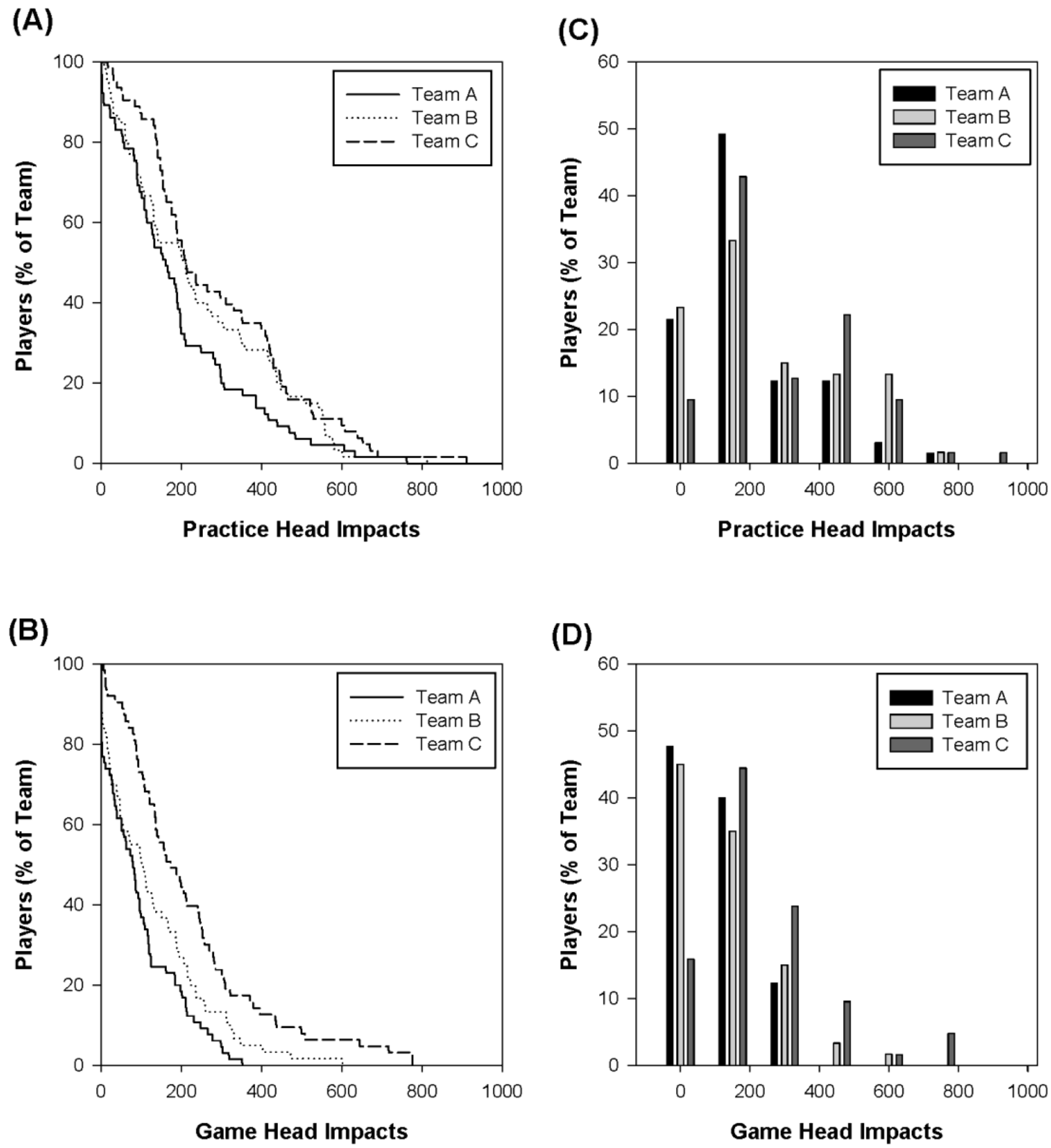

Figure A.4: The number of head impacts for individual players of each team during all practices (A and C) and games (B and D). On the left the data are plotted as a cumulative sum with the number of practice (A) and game (B) impacts plotted on the x-axis and the percentage of players on each team with the number of impacts, or greater, plotted on the y-axis. On the right (C and D) is an ordinary histogram of the same data.

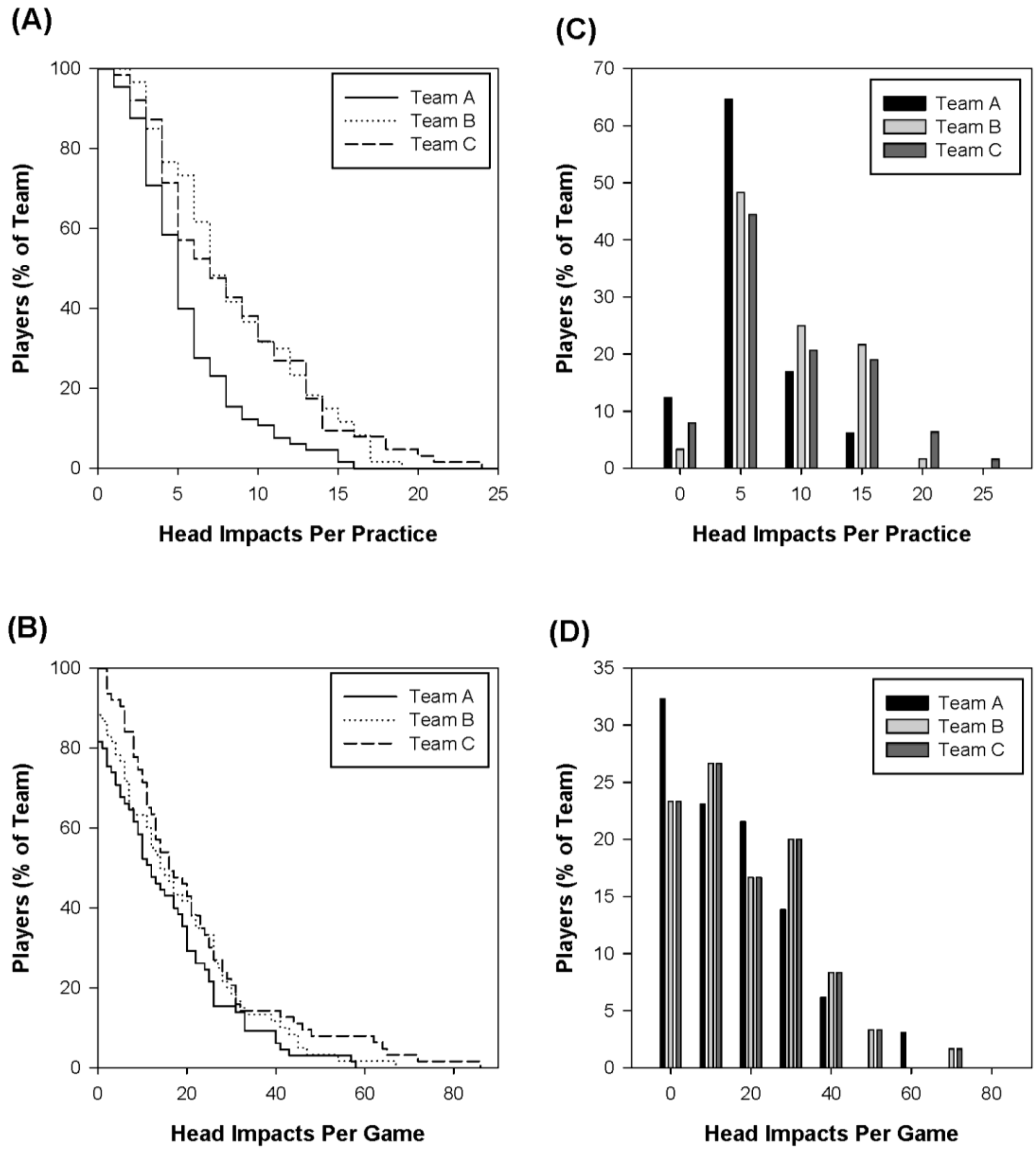

Figure A.5: The number of head impacts for individual players per practice (A and C) and per game (B and D) for players on each team. On the left the data are plotted as a cumulative sum with the number of impacts per practice (A) and per game (B) plotted on the x-axis and the percentage of players on each team with the number of impacts, or greater, plotted on the y-axis. On the right (C and D) is an ordinary histogram of the same data.

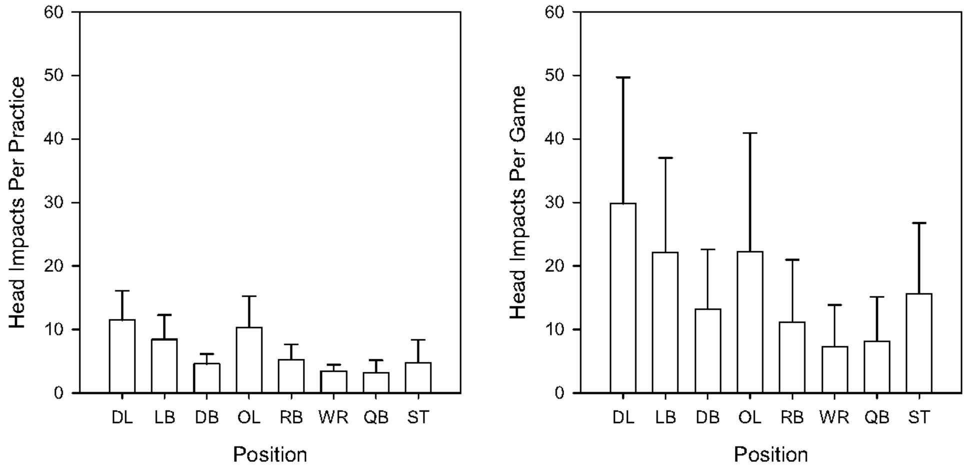

Figure A.6: The mean (one SD) number of impacts per practice (6A) and per game (6B) across player positions did not differ with team, and were grouped together. Impacts per game were typically 2.4 times greater than the impacts per practice across these various positions. Player positions were defined as Defensive Linemen (DL), Linebacker (LB), Defensive Back (DB), Offensive Linemen (OL), Running Back (RB), Wide Receiver (WR), Quarterback (QB), Special Teams (ST).

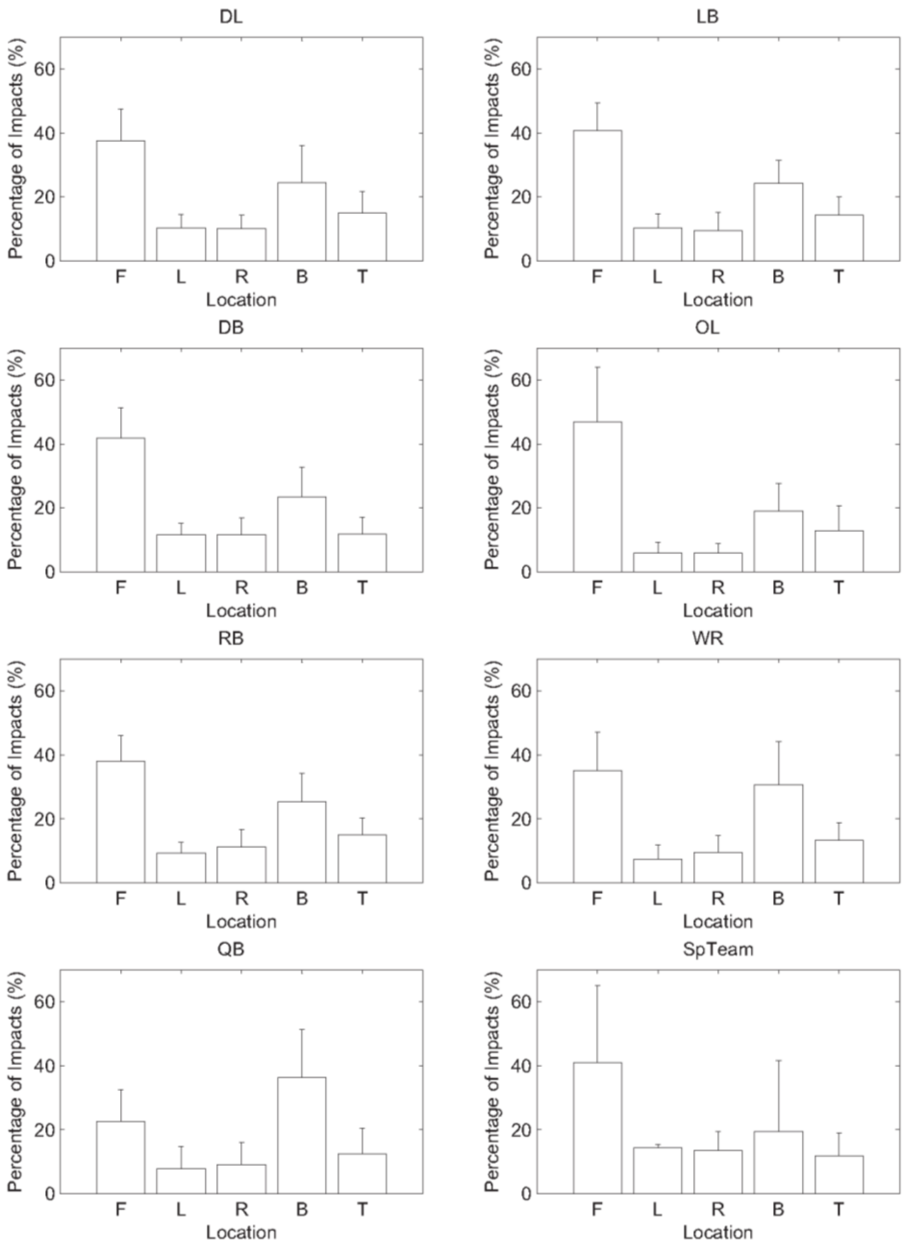

Figure A.7: The mean (one SD) percentage of season head impacts at each helmet location (F: front, L: left, R: right, B: back, T: top). The majority of players had the highest percentage of impacts to the front of the helmet. Offensive lineman (OL) had the greatest percentage of impacts to the front of the helmet, while quarterbacks (QB) had the greatest percentage of impacts to the back of the helmet.

Appendix B

Magnitude of Head Impact Exposures in Individual Collegiate Football Players

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B.1 Abstract

The purpose of this study was to quantify the severity of head impacts sustained by individual collegiate football players and to investigate differences between impacts sustained during practice and game sessions, as well as by player position and impact location. 184,358 head impacts were analyzed for 254 collegiate players at three collegiate institutions. In practice, the 50^{th} and 95^{th} percentile values for individual players were 20.0 g and 49.5g for peak linear acceleration, 1187 rad/s^2 and 3147rad/s^2 for peak rotational acceleration, and 13.4 and 29.9 for HITsp, respectively. Only the 95^{th} percentile HITsp increased significantly in games compared to practices (8.4%, $p = 0.0002$). Player position and impact location were the largest factors associated with differences in head impacts. Running backs consistently sustained the greatest impact magnitudes. Peak linear accelerations were greatest for impacts to the top of the helmet, whereas rotational accelerations were greatest for impacts to the front and back. The findings of this study provide essential data for future investigations that aim to establish the correlations between head impact exposure, acute brain injury and long-term cognitive deficits.

B.2 Introduction

Concussion injuries are a growing and important health care problem in sports [1], affecting approximately 5% of athletes of all ages and at all levels of participation [1–9]. The current definition of concussion injury is a change in cognitive state preceded by an impact to the head [10]; however, to date the specific association between the biomechanics of a head impact and a concussion injury remains unclear. Developing and documenting measures of head impact biomechanics is one critical step to understanding the cause of concussion injuries.

In sports, exposure to the risk of injury is often reported by athlete-exposure (A-E) [11], defined as one student-athlete participating in one practice or competition in which he or she was exposed to the possibility of athletic injury. A-E is valuable for assessing the risk of injury due to participation, but it does not account for specific injury mechanisms. With respect to concussion injuries, A-E cannot capture the magnitude or the frequency of head impacts. For example, two athletes who participate in the same number of games would have the same A-E, yet they would most likely be exposed to a different number and severity of head impacts.

In our approach to understanding the biomechanical basis of concussion injury we have defined "head impact exposure" as a multi-factorial term that includes the frequency of head impacts (e.g. number of head impacts per practice), magnitude of the impacts (e.g. peak linear acceleration), the location (e.g. front of the helmet), and cumulative history of head impacts for an individual athlete. A multi-factorial measure of exposure is critical at this juncture because a specific variable or combination of head impact variables that correlate with the risk of brain injury has not yet been determined.

There have been several efforts to measure head impacts in helmeted sports dating back to the 1970's [12,13]. These early efforts required football players to wear obtrusive data acquisition hardware that interfered with normal play. Consequently, these studies were limited by the number of athletes and the head impact data that were collected. More recently, an accelerometer-based system mounted inside of football helmets, the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH, marketed commercially as Sideline Response System by Riddell, Elyria, OH) [14–16], has been used to directly measure the magnitude of head acceleration and helmet impact location in football players [17–23] during practices and games without interfering with normal play. These studies have provided new insights into the biomechanics of head impacts in football by examining the number of impacts and the magnitude of the resulting head accelerations aggregated within teams and player position.

Previously we reported that the frequency of head impacts individual collegiate football players were exposed to varied significantly with player position, team session (game vs. practice) and impact location [24]. The majority of players received between 200 to 400 head impacts per season, while some players were exposed to more than 1400. The average number of head impacts sustained in games was nearly three times greater than the number of impacts received in practices. While this study was the first to report the frequency of head impacts for individual collegiate players, the magnitude of those impacts were not reported.

The purpose of this study was to quantify the severity of head impacts to which individual collegiate football players were exposed during practices and games over two seasons. We tested the hypotheses that head impact magnitude differed by team, season, session, player position, and helmet impact location.

B.3 Methods

Players from three National Collegiate Athletic Association (NCAA) football programs (Brown University, Dartmouth College, and Virginia Tech) were provided the opportunity to participate in this IRB approved observational study after informed consent was obtained. During the 2007 and 2008 fall football seasons, a total of 254 male players from the three teams, denoted arbitrarily as Team A (n = 85 players), Team B (n = 83 players), and Team C (n = 86 players), participated in this study. Of these players, 116 were monitored in both seasons. This participant turnover was expected, and due primarily to typical roster fluctuations on a collegiate team (e.g. graduation, incoming freshman, injuries, etc.). Each player was assigned a unique identification number and categorized in one of eight position units defined by the team staff as the player's primary position: defensive line (DL, n = 39), linebacker (LB, n = 38), defensive back (DB, n = 46), offensive line (OL, n = 60), offensive running back (RB, n = 29), wide receiver (WR, n = 26), quarterback (QB, n = 10), and Special Teams (ST, n = 6).

All players wore Riddell (Riddell, Chicago IL) football helmets instrumented with the HIT System (**Figure B.1A**), a device capable of recording the acceleration-time history of an impact from six linear accelerometers at 1000 Hz. Impact data from all participating institutions were uploaded to a secure central server with a consolidated database, and subsequently exported for statistical analysis. Data were reduced in post-processing to exclude any impact event with a peak resultant linear head acceleration less than 10*g* [19] in order to eliminate events that had been determined during initial system development to be inconsequential, non-impact events (e.g. running, jumping, etc.). Any impact event in which the acceleration-time history pattern of the six linear accelerometers did not match the theoretical pattern for rigid body head acceleration [14], such as a spike in a single accelerometer signal that can occur during throwing or kicking a helmet, was also excluded. These data reduction methods have been previously verified [15,18,20,22], as was the accuracy of the HIT algorithm [14]. Laboratory tests have determined that the linear and rotational accelerations measured by the HIT system were within ± 4% of a helmet-equipped Hybrid III dummy [20].

A team session (session) was defined as either a formal team practice (players wore protective equipment with the potential of head contact) or a game (competitions and scrimmages). An individual player was defined to have participated in a session when at least one head impact was recorded for that given player. Impacts that were recorded outside of the time of the team session, as defined by the team staff, were excluded from the analysis.

Head impact magnitude was quantified by peak linear acceleration (g) and peak rotational acceleration (rad/s^2). Each recorded impact event was processed using a simulated annealing optimization algorithm to solve for the linear acceleration magnitude at the head center of gravity (CG) [14]. Peak rotational acceleration was calculated as the vector product of peak linear acceleration and a point of rotation estimated to be 10 cm inferior to the CG of the head. Laboratory testing has confirmed that this location is consistent with the impact response of the Hybrid III dummy [20]. Helmet impact location for each impact was computed as azimuth and elevation angles in an anatomical coordinate system relative to the CG of the head [14] and then categorized into one of five helmet impact locations: front (F), left (L), right (R), back (B), and top (T) (**Figure B.1B**). Four equally spaced regions centered on the anatomical mid-sagittal and coronal planes defined front, left, right and back impact locations. All impacts occurring above an elevation angle of 65°, where 0° elevation was defined as a horizontal plane through the center of gravity of the head, were defined as impacts to the top of the helmet. Additionally, a non-dimensional measure of head impact severity, HITsp [21] was computed. HITsp transforms the computed head impact measures of peak linear and peak angular acceleration into a single latent variable using Principal Component Analysis, and applies a weighting factor based on impact location [21].

Statistical Analysis The 50^{th} and 95^{th} percentile values of the peak linear and peak rotational acceleration were first calculated across the entire study, independent of player. For analysis, individual players' 50th and 95th percentiles were calculated for each impact location (front, left, right, top and back) within all their practices and within all their games. For HITsp, the 50^{th} and 95^{th} percentiles were calculated for practices and for games without consideration of location. HITsp was not analyzed among impact locations because impact location is a factor in computing HITsp values. The 50th and 95^{th} percentile values were positively skewed (normality test failed, $P < 0.05$) making general linear models that assume normally distributed variances inappropriate. Therefore, generalized estimating equations (GEE) for log-normally distributed data were used to model the 50^{th} and 95^{th} percentiles, with repeated measures within players treated as having correlated error with a heterogeneous compound symmetrical variance-covariance matrix for session type x location, block diagonal by season. For peak linear and rotational accelerations the predictive factors were team, season, season x team, session type, impact location, player position, and the two- and three-way interactions among session type, impact location, and player position. The interactions between season and team were also included to allow for differences in the changes across seasons between institutions. For HITsp, the factors were team, season, team x season, session, player position, and session x player position.

Statistical significance was set at $\alpha = 0.05$. Given the large number of hypotheses, in order to minimize type II error, α was only adjusted for multiple comparisons within families of effects (e.g. differences by player position were adjusted without consideration of differences by helmet impact location). These post-hoc tests utilized the Holm-Simulated adjustment procedure. All statistical analyses were performed using SAS version 9.2 (SAS Institute, Cary, NC).

B.4 Results

B.4.1 Impacts Across Study

A total of 184,358 head impacts, recorded during 412 sessions (330 practices and 82 games), were included in the analysis. The distributions of each measure were heavily skewed ($P < 0.001$) toward lower values (**Figure B.2**). Across the study, independent of player, the 50th percentile values for peak linear acceleration, peak rotational acceleration and HITsp were 20.3 *g*, 1392 rad/s^2, and 13.7, respectively. The 95th percentile values for peak linear acceleration, peak rotational acceleration and HITsp were 62.2 *g*, 4289 rad/s^2, and 32.1, respectively.

B.4.2 Impacts Among Players

A player's position and helmet impact location were the largest factors associated with the differences in the magnitudes of 50th and 95th percentile peak linear acceleration, peak rotational acceleration, and HITsp. Season (2007 vs. 2008) and team (A vs. B vs. C) also differed significantly, but the differences were smaller (**Table B.1**).

B.4.2.1 Season and Team

Head impact magnitudes decreased from the 2007 to the 2008 season. Significant decreases were in 50th percentile peak linear acceleration (1.8 *g*, 8.5%), rotational acceleration (99.2 rad/s^2, 8.0%) and HITsp (0.93, 6.7%) (**Table B.1**). Among teams, there were no differences in the 50th peak linear accelerations, 50th and 95th peak rotational accelerations, or 95th HITsp (**Table B.1**). Differences between teams were marginally significant but small (< 3 *g*) for 95th percentile peak linear accelerations, with individual team confidence intervals overlapping: 50.8 g (95% CI: 48.6-53.0 g), 47.7 g (95% CI: 45.8-49.8 g), and 50.2 g (95% CI: 48.1-52.4 g) for Teams A, B and C, respectively. Similarly, the differences in 50th percentile HITsp with Team A (12.9, 95% CI: 12.6 – 13.1) significantly less than Teams B (13.6, 95% CI: 13.4-13.8) and C (13.6, 95% CI: 13.4 – 13.9), who did not differ. The interactions between season and team were not statistically significant (**Table B.1**).

B.4.2.2 Practice vs. Games

There were no increases (< 1%) from practices to games in the head impact magnitude, except in the 95th percentile HITsp, which increased significantly in games (**Table B.1**). In practice, the 50th and 95th percentile values were 20.0 g (95% CI: 19.7 – 20.3 *g*) and 49.5g (95% CI: 48.0 - 51.1 *g*) for peak linear acceleration, 1187 rad/s^2 (95% CI: 1166 - 1210 rad/s^2) and 3147rad/s^2 (95% CI: 3043 - 3253 rad/s^2) for peak rotational acceleration, and 13.4 (95% CI: 13.3 - 13.6) and 29.9 (95% CI: 29.1 - 30.8) for HITsp, respectively. In games, the 50th and 95th percentile values were 20.2 *g* (95% CI: 19.8 – 20.5 *g*) and 49.6 *g* (95% CI: 47.7 - 51.5 *g*) for peak linear acceleration, 1197 rad/s^2 (95% CI: 1170 - 1225 rad/s^2) and 3145 rad/s^2 (95% CI: 3011 - 3285 rad/s^2) for peak rotational acceleration, and 13.3 (95% CI: 13.1 - 13.5) and 32.4 (95% CI: 30.9 - 33.9) for HITsp, respectively. The 95th percentile HITsp during practices increased significantly, by 8.4%, when compared with games (**Table B.1**).

B.4.2.3 Positions

There were statistically significant differences in head impact magnitude among player position (**Table B.1**) (**Figure B.3**). There were no statistically significant differences in the 50^{th} peak linear acceleration among running backs (RB), linebackers (LB) or quarterbacks (QB), while each had significantly greater 50^{th} percentile peak linear acceleration than offensive linemen (OL) and wide receivers (WR) (**Figure B.3A**). The 50^{th} percentile peak linear acceleration was also significantly greater for RB than for defensive backs (DB). These significant differences ranged between 5% (LB over OL) to 16% (QB & RB over ST). RB had the greatest 95^{th} peak linear acceleration, followed by LB (3.5% less than RB), and DB (8.8% less than RB). The 95^{th} peak linear acceleration for RB, LB and DB were significantly greater than for OL, WR and ST. These differences ranged from 13% (DB over OL) to 42% (RB over ST). DL were 11% less than RB, and DL were significantly greater than WR by 20%. The pattern of differences between player positions in peak rotational acceleration was essentially the same as pattern of differences between player positions in peak linear acceleration (**Figure B.3B**).

DB, LB, OL, and RB each had significantly greater 50^{th} percentile HITsp than ST and WR (**Figure B.3C**). There were no statistically significant differences in 50^{th} HITsp among DB, LB, OL, and RB. RB also had significantly greater 95^{th} HITsp than DL, OL and WR. These differences ranged from 18.5% (RB over DL) to 25% (RB over WR). The 95^{th} percentile HITsp was 19% greater for RB than for QB, but this difference was not statistically significant.

B.4.2.4 Impact Locations

The magnitude of the head impacts differed significantly among impact locations (**Table B.1**). Peak linear accelerations were greatest for impacts to the top of the helmet, followed by the front, back, and sides (**Figure B.4A**). The 50^{th} percentile peak linear acceleration differed significantly among all helmet impact locations, except between the left and right sides. For the 95^{th} percentile data, the statistical relationships were similar to the 50^{th} percentile except impacts to the front were not significantly different from those to the back. The 95^{th} percentile peak linear acceleration for impacts to the top were 9.6% (95% CI: 5.9 - 14%) and 11.5% (95% CI: 6.6 - 16.1%) greater than the front and the back, respectively, and approximately 30% (95% CI: 26 - 35%) greater than the sides.

Impacts to the front and to the back (which did not differ from each other) had the greatest peak rotational acceleration (**Figure B.4B**). The 95^{th} percentile peak rotational acceleration for the front and back were both significantly greater (approximately 33%) than for the left and right sides (which did not differ from each other) and for the top (approximately 109% greater).

In games, impacts to the front of the helmet tended to have greater 95^{th} percentile peak linear acceleration than in practice. The 95^{th} percentile peak linear acceleration for impacts to the front of the helmet increased the most among all helmet impact locations in games compared to practice for most player positions (17% for DB, 11.5% for DL, 12.5% for LB, 8.6% for OL). For WR and ST the increases varied by location, and there was no clear pattern. There were no notable increases for the 95^{th} percentile peak rotational acceleration in games compared to practice for the various impact locations.

B.4.2.5 Impact Location by Position

There were statistically significant differences in the magnitude of impacts among helmet locations by player position (**Table B.1**). LB and RB, which did not differ significantly from each other, had the greatest 95th percentile peak linear acceleration for front impacts (**Table B.2**). RB had the greatest 95th percentile peak linear acceleration for the side impact location among all player positions, and this value was significantly greater than OL by approximately 29% and WR by 57%. QB had the greatest 95th percentile peak linear acceleration for impacts to the back of the head among all player positions, but the differences were not statistically significant, and ranged from approximately 34% for ST to 7% for RB.

The 95th peak rotational acceleration from impacts to the front of the helmet was greatest for RB and LB, which did not differ (**Table B.3**). RB and LB had significantly greater peak rotational accelerations at the front location than OL by approximately 15%. RB and LB had significantly greater peak rotational acceleration at the side impact location than OL by approximately 36% and WR by approximately 60%.

B.5 Discussion

In this study, we examined the magnitude (peak linear acceleration and peak rotational acceleration) and HITsp severity of the head impacts received players from three NCAA collegiate football teams. In our previous study, we found the median number of head impacts for individual players per practice and per game ranged from 4.8 to 7.5 and 12.1 to 16.3, respectively [24]. Additionally, we found that offensive linemen had a higher percentage of impacts to the front than to the back of the helmet, whereas quarterbacks had a higher percentage to the back than to the front of the helmet. The present study focused on impact magnitude and the relationships among magnitude, session, player position and impact location.

The question remains as to what magnitudes and/or quantities of impact accelerations are important clinically, both acutely and cumulatively. The magnitude of the majority of the impacts received by college football players in this study were less than 20 g's and 1389 rad/s^2. The 95th percentile values for peak linear acceleration and peak rotational acceleration were 61 g's and 4245 rad/s^2, respectively. These values are below brain injury tolerance levels commonly cited in the literature. Pellman et al. [25] suggested that impacts greater than 98 g had an 80% probability of resulting in a concussion in NFL players based on 25 laboratory reconstructions of NFL impacts that resulted in a diagnosed concussion. Zhang et al. [26] predicted that linear accelerations of 66, 82, and 106 g's, and rotational accelerations of 4600, 5900, and 7900 rad/s^2 were associated with 25%, 50% and 80% probability of clinical diagnosis of concussion based on computer modeling of the NFL data using the Wayne State Brain Injury Model. We propose that impact magnitude, frequency and location are all critical measures of head impact exposure, and theorize they can be used to more accurately quantify the risk of sustaining concussion or other potentially clinically consequential brain injuries in helmeted athletes than measures based upon participation levels. The current study is one important step in quantifying this risk.

Head impact magnitudes were generally not different for games compared to practices. Broglio et al. [23] reported that head impacts during games resulted in greater mean linear and rotational accelerations than in practice for high school athletes; however, these differences were small (approximately 1.5g and 200 rad/s^2). Their mean values for practices (23.3 ± 14.5g and 1469 ± 1055 rad/s^2) were similar to our 50th percentile values for individual players. We are cautious about using these values for comparison because of the skewness in the data. We did not detect differences in the magnitude of impacts between games and practices for individual players, except for the 95th percentile HITsp values, although it has been reported that injury rates are greater in games than in practices [8].

A strong association was found between head impact magnitude and player position. This is not unexpected given the different skills and strategies of play required for each position. Running backs (RB) had significantly greater 50th percentile peak linear and peak rotational acceleration than offensive lineman (OL), wide receivers (WR), special teams (ST), and defensive backs (DB), as well as the greatest 95th percentile peak linear acceleration and HITsp. These findings are consistent with those of Mihalik et al. [19], who reported offensive backs (OB) were more likely to sustain impacts greater than 80 *g*'s than defensive lineman (DL), defensive backs (DB), offensive lineman (OL), linebackers (LB), and wide receivers (WR). Similarly, Schnebel et al. [17] reported that skilled players (QB, RB, WR, LB, DB) were more likely to sustain impacts with greater magnitude than lineman (OL, DL) . Broglio et al. [23] reported that high school linemen had greater peak rotational acceleration when compared to offensive and defense skill positions. They also reported that defensive linemen and offensive skill players sustained similar-magnitude linear accelerations, but only the defensive line players had greater linear accelerations than the defensive skill and offensive line players. Whether these differences with our findings are due to differences between collegiate and high school players or to our approach in analyzing specific positions require further studies in both populations.

In addition to player position, head impact magnitudes differed by location. Consistent with previous findings [19,23,27], peak linear accelerations were greatest to the top of the helmet. Broglio et al. reported that front impacts resulted in greater rotational accelerations than any other impact location [23]. We found no difference in peak rotational accelerations between impacts to the back and to front of the helmet, but we did find that these impact locations resulted in greater peak rotational accelerations than any other impact location.

The interaction of player position and helmet impact location had a significant effect on head impact magnitude. Linebackers and running backs (LB, RB) had the greatest 95^{th} percentile peak linear and peak rotational accelerations for impacts to the front of the helmet. Running backs (RB) also had the greatest 95^{th} percentile peak linear acceleration for side impacts. For all positions (except for ST), impacts to the top of the helmet had the greatest peak linear acceleration. These data may prove useful in developing a rationale for position-specific helmet designs or protective strategies in the future.

In the present study we focused our analysis on head impact magnitude for individual players. Our previous study [24] focused specifically on head impact frequency and helmet impact location. Further analysis is needed to determine the relationship between impact frequency and magnitude. If such a relationship is found, it may be possible to reduce head impact exposures, and hence the risk of concussion injury, without significantly altering the sport. Additionally, the risk of sustaining a concussion injury may depend not only on a threshold to a single impact but also on individual's impact history. We are examining cumulative measures of repetitive head impacts as another measure of head impact exposure. Signs and symptom reports from medically diagnosed concussion injuries and other measures of cognitive deficits were not reported here, as a suitably sized cohort of players with diagnosed injuries is still being accumulated and analyzed.

In summary, we found that an individual collegiate football player receives head impacts of varying magnitudes during play, and that the magnitudes of these impacts are heavily skewed towards lower values. Interestingly, the magnitude of impacts during games was not significantly greater than the magnitude of the impacts during practices except for the 95th percentile HITsp. We also found that there were significant differences in the magnitude of impacts among different player positions. Impact location was found to be a factor strongly associated with the differences in magnitude of peak linear acceleration and peak rotational acceleration. The interaction of player position and impact location also yielded statistically significant differences in impact magnitude. This study has provided a detailed description of the magnitude of head impact exposures for collegiate football players that will be critical for establishing the relationship between head impact biomechanics and the risk of concussion injury and for developing appropriate concussion injury prevention strategies.

B.6 Acknowledgments

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Factors	Peak Linear Acceleration (g) 50th	95th	Peak Rotational Acceleration (rad/s^2) 50th	95th	HITsp 50th	95th
Season	< 0.0001*	0.1134	< 0.0001*	0.0643	< 0.0001*	0.0641
Team	0.2306	0.0468*	0.0875	0.3134	< 0.0001*	0.1862
Season x Team	0.0762	0.6151	0.1941	0.4386	0.7386	0.1999
Session	0.3972	0.9797	0.4915	0.9832	0.3507	0.0002*
Position	< 0.0001*	< 0.0001*	< 0.0001*	< 0.0001*	0.0001*	< 0.0001*
Session x Position	0.1335	0.0980	0.0869	0.1579	0.0045*	0.2317
Location	< 0.0001*	< 0.0001*	< 0.0001*	< 0.0001*		
Session x Location	0.0282*	0.2276	0.0005*	0.4000		
Position x Location	< 0.0001*	< 0.0001*	< 0.0001*	< 0.0001*		
Session x Position x Location	0.0514	0.0147*	0.0276*	0.4481		

Table B.1: Post-hoc analysis summary of statistical differences (* considered significant) by season, team, session, position and their interactions.

Impact Location	Player Position							
	DL	LB	DB	OL	RB	WR	QB	ST
Front	53.6 (3.2)	60.7 (3.7)	55.9 (3)	52.6 (2.7)	60.1 (4.2)	46.3 (3.9)	52 (6.7)	53.2 (8.7)
Left	45.3 (4.7)	48.7 (5.1)	44.8 (4.1)	36.4 (3.2)	54.1 (6.7)	33.8 (5.1)	41.2 (9)	35.5 (10.4)
Right	40.9 (3.9)	49.1 (4.8)	46.3 (4)	38.5 (3.1)	52.1 (5.9)	33.2 (4.7)	39.3 (8.3)	35.3 (9.1)
Back	55.4 (4.7)	54.9 (4.7)	55.3 (4.1)	50.7 (3.6)	56.1 (5.5)	49.1 (5.7)	59.8 (10.3)	44.5 (9.7)
Top	66.0 (7)	68.9 (7.3)	64.4 (6.1)	60.5 (5.4)	68.9 (8.6)	56.4 (8.6)	63.3 (13.9)	38 (10.7)

Table B.2: The 95th percentile (95% CI) peak linear acceleration (g) for individual players of the various positions and helmet impact locations.

Impact Location	Player Position DL	LB	DB	OL	RB	WR	QB	ST
Front	4070 (256)	4615 (294)	4359 (245)	3992 (211)	4727 (348)	3376 (293)	3746 (504)	4316 (739)
Left	3532 (411)	3817 (454)	3367 (348)	2769 (272)	4098 (567)	2434 (416)	2845 (703)	2396 (818)
Right	3125 (356)	3875 (450)	3615 (368)	2732 (262)	3995 (538)	2346 (391)	2681 (681)	2474 (758)
Back	4236 (384)	4197 (384)	4145 (332)	3916 (296)	4307 (456)	3750 (467)	4346 (805)	3555 (838)
Top	2265 (264)	2314 (271)	2245 (232)	1953 (191)	2298 (315)	1682 (284)	1890 (459)	1257 (386)

Table B.3: The 95th percentile (95% CI) peak rotational acceleration (rad/s^2) for individual players of the various positions and helmet impact locations.

Figure B.1: Football players wore helmets instrumented with the HIT system that was specifically designed to record head accelerations as a result of an impact to the helmet without interfering with play. The HIT System (**A**) is comprised of an in-helmet unit containing six accelerometers (a), battery (b), transmitting and logging electronics (e), a sideline receiver, and a laptop computer. An illustration of the regions that defined the front (F), right side (R), back (B) and top (T) helmet impact locations (**B**).

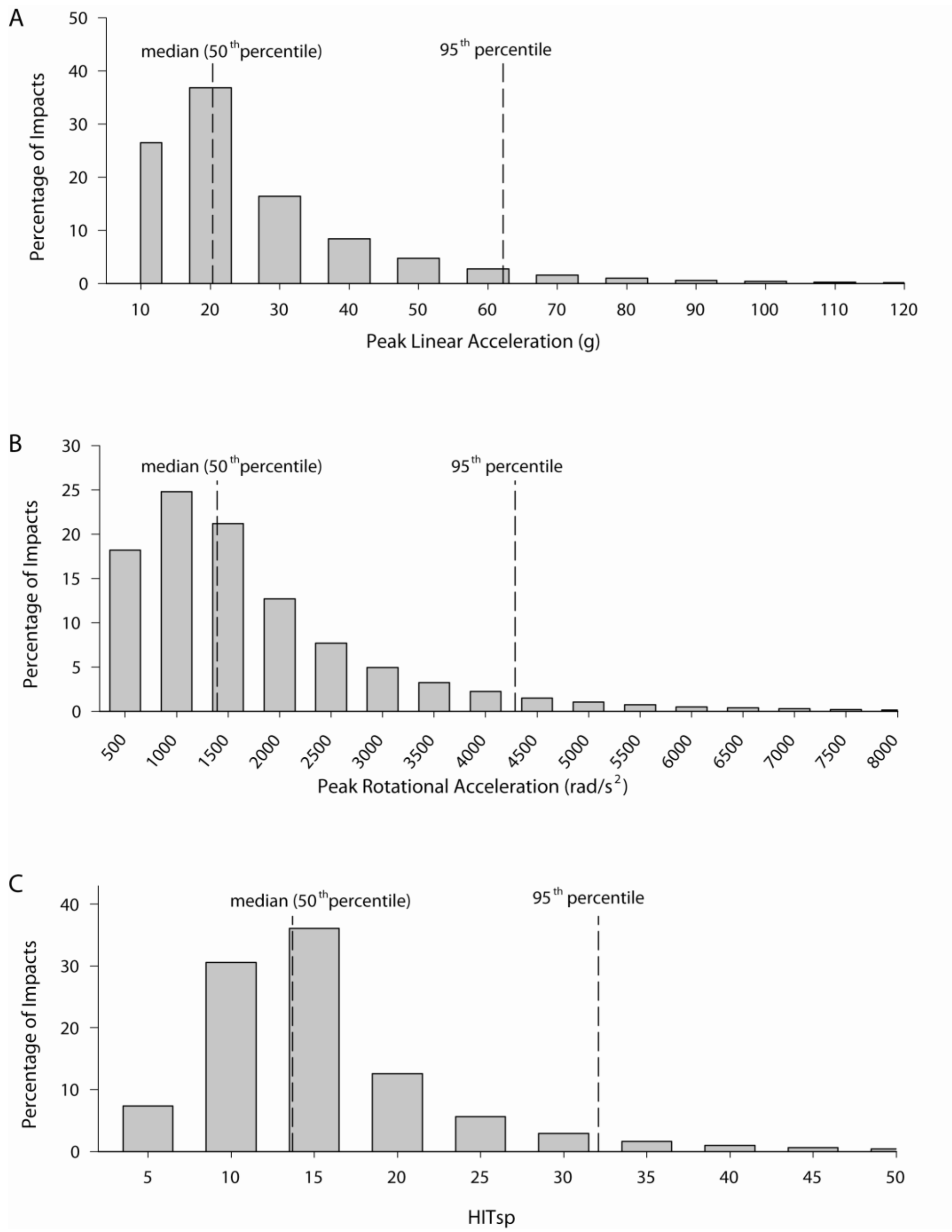

Figure B.2: The distribution of the head impact measures of peak linear acceleration (**A**), peak rotational accelerations (**B**) and HITsp (**C**) were heavily skewed towards lower magnitudes. These distributions, and their associated 50th and 95th percentile values, were computed by aggregating all impacts (n = 184,358) recorded in the study. We note the bin size for peak linear acceleration is 10 g, except for the first which binned values of 10 g to 15 g.

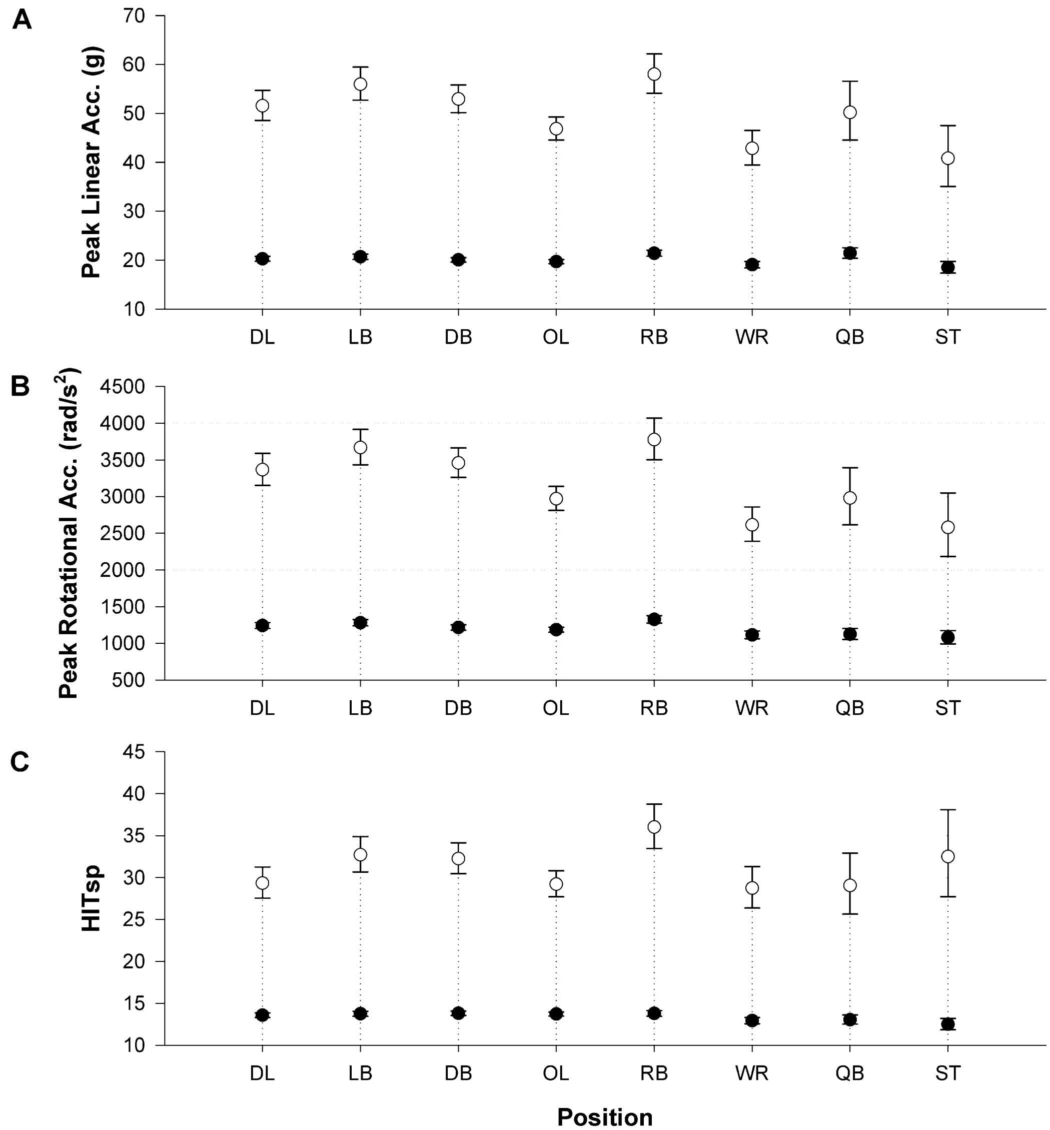

Figure B.3: The 50th percentile (filled circles) and the 95th percentile (open circles) of peak linear acceleration (**A**), peak rotational acceleration (**B**) and HITsp (**C**) differed significantly ($p < 0.0001$) among player positions. The error bars represent the lower and upper 95% confidence intervals for individual players of that position.

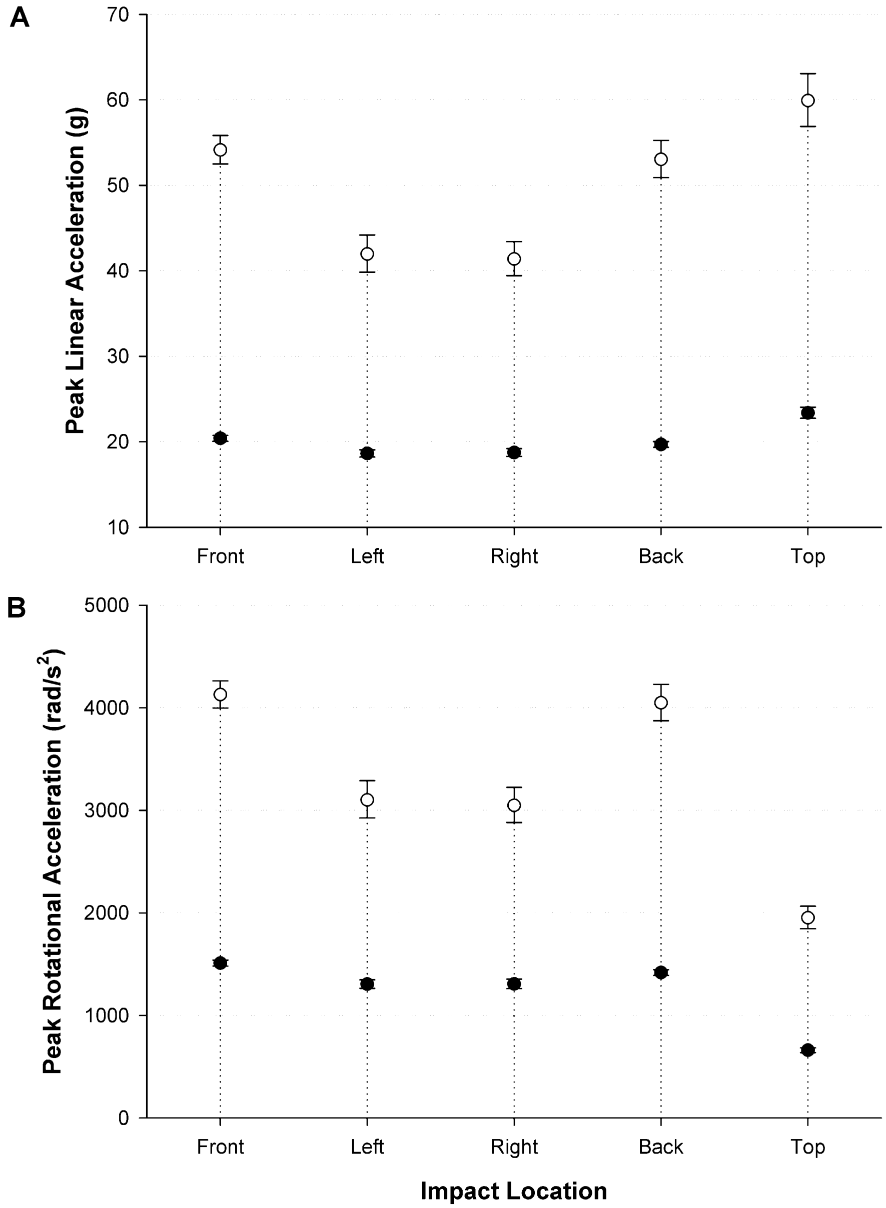

Figure B.4: The 50th percentile (filled circles) and the 95th percentile (open circles) of peak linear acceleration (**A**) and peak rotational acceleration (**B**) differed significantly ($p < 0.0001$) among impact location. The error bars represent the lower and upper 95% confidence intervals for individual players of that impact location.

Appendix C

Head Impact Exposure in Collegiate Football Players

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C.1 Abstract

In American football, impacts to the helmet and the resulting head accelerations are the primary cause of concussion injury and potentially chronic brain injury. The purpose of this study was to quantify exposures to impacts to the head (frequency, location and magnitude) for individual collegiate football players and to investigate differences in head impact exposure by player position. A total of 314 players were enrolled at three institutions and 286,636 head impacts were recorded over three seasons. The 95th percentile peak linear and rotational acceleration and HITsp (a composite severity measure) were 62.7g, 4378 rad/s^2, and 32.6, respectively. These exposure measures as well as the frequency of impacts varied significantly by player position and by helmet impact location. Running backs (RB) and quarter backs (QB) received the greatest magnitude head impacts, while defensive line (DL), offensive line (OL) and line backers (LB) received the most frequent head impacts (more than twice as many than any other position). Impacts to the top of the helmet had the lowest peak rotational acceleration (2387 rad/s^2), but the greatest peak linear acceleration (72.4 g), and were the least frequent of all locations (13.7%) among all positions. OL and QB had the highest (49.2%) and the lowest (23%.7%) frequency, respectively, of front impacts. QB received the greatest magnitude (70.8g and 5428 rad/s^2) and the most frequent (44% and 38.9%) impacts to the back of the helmet. This study quantified head impact exposure in collegiate football, providing data that is critical to advancing the understanding of the biomechanics of concussive injuries and sub-concussive head impacts.

C.2 Introduction

Impacts to the head are commonly identified as the cause of concussion injury during athletic play [1–3] while repetitive head impacts, even those with no acute symptoms or signs, often referred to as sub-concussive impacts, have been suggested as a possible cause of chronic brain injury [4]. At present, the relationships between head impacts and these brain injuries are not well understood. For example, studies utilizing surrogate reconstructions of documented concussive hits in the National Football League have proposed that the risk of concussion injury is associated with the peak linear acceleration of the head[5]. Others have postulated that the threshold for concussive injury may be difficult to establish because of the varying magnitudes and locations of impacts resulting in concussion, as well as other factors such as the frequency of sub-concussive impacts and the number of prior concussions [6]. This lack of consensus may be due in part to the challenges of measuring and analyzing head impacts. It also has been suggested that the location of the impact and the direction of the resulting head motion is a factor in the mechanism of concussion injury [7]. Greenwald et al. [8] determined that a weighted measure, HITsp that incorporates linear acceleration, rotational acceleration, impact duration and impact location, was more predictive of concussion diagnosis than any single biomechanical measure. Accordingly, head impact exposure is a risk factor for concussion injury that needs to be quantified, with implications for pathophysiology and for prevention. In our approach to understanding the biomechanical basis of concussion [9,10] we have defined “head impact exposure” as a multi-factorial term that includes the frequency of head impacts (e.g. number of head impacts per season), magnitude of the impacts (e.g. peak linear acceleration), impact location (e.g. front of the helmet), and cumulative history of head impacts for an individual athlete. A multi-factorial measure of exposure is critical at this time because the mechanism of acute and chronic brain injury is still not completely understood. Thus, this study is motivated by the need to fully understand and to rigorously quantify measures of head impact exposure.

There have been several efforts to measure head impacts in helmeted sports dating back to the 1970’s [11,12]. These early efforts were limited by the available technology, requiring football players to wear obtrusive data acquisition hardware that allowed data collection on only a few athletes in a few sessions. More recently, an accelerometer-based system mounted inside of football helmets, the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH, marketed commercially as Sideline Response System by Riddell, Elyria, OH) [13–15], has been used to directly measure the magnitude of head acceleration and helmet impact location in football players [8,16–22] during practices and games without interfering with normal play. These studies have provided new insights into the biomechanics of head impacts in football by examining the number of impacts and the magnitude of the resulting head accelerations aggregated within teams and player positions, and at different levels of play. Previously we analyzed the frequency [9] and magnitude [10] of head impacts for individual collegiate football players, but did not examine the relationships between these measures of exposure.

Building upon these previous studies, and expanding the data collection to three seasons, the purpose of the current study was to examine head impact exposure by quantifying the frequency of the impacts, the location of the impacts on the helmet, and the magnitude of impacts to individual collegiate football players among various player positions. Specifically, we tested the hypotheses that head impact frequency, location and magnitude would not differ by player position.

C.3 Methods

During the 2007, 2008 and 2009 fall football seasons, a total of 314 players from three National Collegiate Athletic Association (NCAA) football programs (Brown University, Dartmouth College, and Virginia Tech) participated in this observational study after informed consent was obtained with institutional review board approval. Of these players, 146, 106 and 62 were monitored during one, two and three seasons, respectively. This participant turnover was expected, and due primarily to typical roster fluctuations on a collegiate team. Each player was assigned a unique identification number and categorized into one of eight position units defined by the team staff as the player's primary position: defensive line (DL, n = 49), linebacker (LB, n = 47), defensive back (DB, n = 55), offensive line including tight ends (OL, n = 75), offensive running back (RB, n = 37), wide receiver (WR, n = 30), quarterback (QB, n = 14), and Special Teams (ST, n = 7), which were not included in this analysis because of the relatively low number of players.

All players wore Riddell VSR-4, Revolution, or Speed (Riddell, Chicago IL) football helmet models that were instrumented with the HIT System. The HIT System is an accelerometer-based device that computes linear and rotational acceleration of the center of gravity (CG) of the head, as well as impact location on the helmet [13–15]. The HIT System is specifically designed to measure head accelerations by elastically coupling the accelerometers to the head, isolating them from the helmet shell. Data were reduced in post-processing to exclude any acceleration event with peak resultant linear head acceleration less than 10g in order to eliminate head accelerations from non-impact events (e.g. running, jumping, etc.) [23]. Data reduction methods are described in detail elsewhere [15,17,19,20], as was the accuracy of the HIT algorithm [14]. Laboratory impact tests of a Hybrid III dummy fitted with all helmets instrumented with the HIT System determined that the linear and rotational accelerations measured by the HIT System were within ± 4% of those measured concurrently by the internally instrumented Hybrid III headform [19].

Head impact exposure was defined for each individual player using measures of impact frequency, location and magnitude. A team session (session) was defined as either a formal team practice (players wore protective equipment with the potential of head contact) or a game (competitions and scrimmages). An individual player was defined to have participated in a session when at least one head impact was recorded for that given player within the specified time of the team session. Five measures of impact frequency were computed: practice impacts was the total number of head impacts for a player during all practices; game impacts was the total number of head impacts for a player during all games; impacts per season was the total number of head impacts for a player during all team sessions in a single season; impacts per practice was the average number of head impacts for a player during practices; and impacts per game was the average of the number of head impacts for a player during games.

Impact locations to the helmet and facemask were computed as azimuth and elevation angles in an anatomical coordinate system relative to the center of gravity of the head [14] and then categorized as front, side (left and right), back, and top. Front, left, right and back impact locations were four equally spaced regions centered on the mid-sagittal plane. All impacts above an elevation angle of 65°from a horizontal plane through the CG of the head were defined as impacts to the top of the helmet [8].

Impact magnitude was quantified by peak linear acceleration (*g*) and peak rotational acceleration (rad/s^2) [14]. Peak rotational acceleration was calculated as the vector product of peak linear acceleration and a point of rotation 10 cm inferior to the CG of the head. Laboratory testing has confirmed that this location is consistent with the impact response of the Hybrid III dummy [19]. Additionally, a non-dimensional measure of head impact severity, HITsp [8] was computed. HITsp transforms the computed head impact measures of peak linear and peak angular acceleration into a single latent variable using Principal Component Analysis, and applies a weighting factor based on impact location [8]. It thus serves as a measure of impact severity, with weight given to factors shown in previous head injury research (linear and rotational acceleration, impact duration and location [7,24–26]) to predict increased likelihood of clinical or structural injury. Impacts were further reduced for analysis by computing the 95^{th} percentile value of all seasonal impacts for each individual player.

Statistical Analysis Results were expressed as median values and [25-75% interquartile range], because each study variable was not normally distributed (Shapiro-Wilk test; $P < 0.001$). The significance of the differences among player positions in impact frequency (impacts per season) and in severity measures (95^{th} percentile peak linear acceleration, 95^{th} percentile rotational acceleration, and 95^{th} HITsp) were examined separately using a Kruskal-Wallis one-way ANOVA on ranks with a Dunn's post-hoc test for all pairwise comparisons. Statistical significance was set at $\alpha = 0.05$ and the reported P values are those for the post hoc test. An identical approach was used to examine the significance of the differences among player positions in frequency and the 95th percentile peak linear and rotational acceleration at each location. Statistical comparison among impact location were performed with a Friedman repeated measures ANOVA on ranks. All statistical analyses were performed using SigmaPlot (Systat Software, Chicago, IL).

C.4 Results

A total of 286,636 head impacts were analyzed in this study. These data were collected during a median of 50 [28-76.5] practices and 12 [6-20] games (including scrimmages) for all players. Impact magnitudes across the study were heavily skewed to lower values ($P < 0.001$) with a 50th and 95th percentile peak linear acceleration of 20.5*g* and 62.7*g*, respectively, 50th and 95th percentile peak rotational acceleration of 1400 rad/s^2 and 4378 rad/s^2, respectively, and 50th and 95th percentile HITsp of 13.8 and 32.6 (**Figure C.1**). The total number of impacts received by an individual player during a single season was a median of 420 [217-728], with a maximum of 2492. The total number of impacts that players received in a single season during practices was 250 [131-453], with a maximum of 1807, and during games was 128 [47-259], with a maximum of 1683. The frequency of impacts players received were further analyzed by normalizing the number of impacts by the number of sessions for each individual player because of differences in team schedules and player attendance.

After grouping players by their primary position and analyzing impacts over all locations, the number of impacts per season ranged from a median of 149 [96-341] for QB to 718 [468-1012] for DL (**Figure C.2**). Across all player positions, there was a linear increase in median impacts per practice with median impacts per season (slope = 0.02, $R^2 = 0.934$) and in median impacts per game with median impacts per season (slope = 0.04, $R^2 = 0.929$). DL, LB and OL received the highest number of impacts per season, and QB and WR the lowest. The median 718 impacts per season received by DL were significantly ($P < 0.05$) more than QB, WR (157 [114-245]), RB (326 [256-457]), and DB (306 [204-419]), but not different than LB (592 [364-815]) or OL (543 [264-948]). LB received significantly ($P < 0.05$) more impacts per season than QB, WR, DB and RB. OL received significantly ($P < 0.05$) more impacts per season than QB, WR and DB.

RB received the impacts with greatest magnitude accelerations and highest HITsp values. The 95th percentile peak linear and rotational acceleration for RB were significantly ($P < 0.05$) greater than OL, DL and DB (**Figure C.3 and Table C.1**). The 95th percentile HITsp for RB and LB were also significantly ($P < 0.05$) greater than OL and DL (**Table C.1**). Although OL and DL received the most frequent impacts per season, the magnitudes of the impacts were the least of all player positions. The median 95th percentile peak linear and rotational accelerations values were greatest for QB, but these were not significantly different from the other player positions (**Figure C.3**).

The magnitudes of impacts to the front of the helmet were significantly ($P < 0.05$) greater for RB than for OL, DL, WR and DB (**Figure C.4**). LB, which were not different from RB, received significantly ($P < 0.05$) greater magnitude front impacts than OL and DL. Although the magnitudes were the lowest, OL received significantly ($P < 0.05$) more front impacts than QB, WR, DL, and LB. QB received significantly ($P < 0.05$) fewer front impacts than all player positions except WR. The magnitude of impacts to the side of the helmet was the greatest for RB, which were significantly greater than OL, DL and WR. Side impacts were significantly ($P < 0.05$) less frequent and less severe for OL than all player positions. The median 95th percentile peak linear and rotational accelerations values associated with impacts to the top of the helmet were greatest for QB, but these were not significantly different among player positions, while RB received impacts that had significantly ($P < 0.005$) greater peak linear accelerations and significantly ($P < 0.02$) greater peak rotational accelerations than OL. For all player positions, top impacts were the least frequent ($P<0.05$) impact location (approximately 13% of all head impacts), but were associated with the greatest ($P<0.05$) peak linear acceleration magnitudes of all impact locations. In contrast, peak rotational accelerations associated with top impacts had significantly ($P<0.05$) lower magnitudes than all locations for all player positions (**Table C.2**). Impacts to the back of the helmet tended to be the highest magnitude for the QB and WR, and were significantly ($P < 0.05$) more frequent for QB and WR than for all other positions.

C.5 Discussion

The purpose of this study was to quantify head impact exposure in individual collegiate football players and then examine the relationships between head impact frequency, location and magnitude as a function of player position. Quantifying head impact exposures is a critical step in achieving our long-term goals of understanding the biomechanical basis for mild traumatic brain injuries (concussion injuries), correlating head impact exposure with the clinical variables associated with these injuries, and understanding the acute and long-term effect of repeated sub-concussive impacts.

Player position had the most significant effect on head impact exposure in this study of collegiate football players. These differences across player position were considerably greater than the differences we previously reported among the teams and season in both frequency [9] and severity [10] of head impacts. The difference among teams in the median head impact frequency was approximately 100 impacts per season. This difference may simply be due to random effects or possibly to the structure of the practice plan and the philosophies of the coaching staff. The increase in impacts per game over impacts per practice was approximately a factor of two for each player position (**Figure C.2**), which is less than, but still consistent with, our previous study of a smaller cohort [9]. Our values for the number of head impacts per practice and impacts per game bracket those values reported by others when we consider the range of our values among our player positions [16–19]. Recently, Guskiewicz et al. [6] reported that the average collegiate football player experiences 950 impacts per season. Schnebel et al.

[16] reported an average of 1353 impacts per season per player in collegiate football players, while in high school players Broglio et al. [21] reported 565 impacts per season per player and Schnebel et al. [16] reported 520 impacts per season per player. The discrepancy with the median of 420 impacts per season that we report may be associated with differences in player participation, but may also be due in part to their computations used to determine the average exposure per player, which was simply dividing the total number of impact recorded in their study by the number of players in their study. Our values for frequency are computed for each individual player.

While DL, LB, and OL were found to have the lowest head impact magnitudes of all player positions, they had the greatest number of head impacts. This is in agreement with previous reports that offensive and defensive linemen sustain the most frequent head impacts [16–18], with relatively low severity [16]. QB received severe impacts (**Figure C.3**) but these were not statistically greater than the other positions, most likely because of the low sample size and the large distributions of values. Mihalik et al. reported that OL received higher linear accelerations than DL and DB [18]; however, it should be noted that these comparisons were based upon the mean values, where as our comparison were based upon those of individual players and their median 95th percentile value because of the non-normality of the positively skewed dataset (**Figure C.1**). In agreement with anecdotal observations of football games, OL received the largest percentage of impacts to the front of the helmet and the smallest to the back, while QB received the largest percentage of impacts to the back of the helmet, with magnitudes comparable to the greatest front and side impacts of all positions. RB had the greatest magnitude impacts to the front, side and top of the helmet relative to other position, and this is consistent with previous reports [16,18]. As previously reported [10,18,21,27], impacts to the top of the helmet generated the highest peak linear accelerations, while being associated with the least rotational acceleration among all player positions. The lower rotational accelerations associated with impacts to the top of the helmet is most likely due to the reduced length of the lever arm for neck sagittal and coronal rotations with impact vectors aligned with the central axis of the cervical spine.

This study has several limitations. Our study only captured the primary position as reported by the team. The number of players who play both offense and defense is low at the collegiate level, and the error due to this is considered to be small compared to the total number of impacts recorded. The differences among teams and season were small compared with the differences among player positions that we reported in this study. Post-processed linear head CG acceleration less than 10*g* were excluded from our analysis. This threshold reduces the number of impacts recorded but eliminates acceleration levels typically experienced by athletes during normal, non-injurious activities such as quickly standing up, running, and jumping [23]. The HIT System provides the location of the impact in spherical coordinates, but we reduced these coordinate to four locations (front, side, top and back). Impacts to the front of the helmet can be analyzed further by those located on the helmet and on the facemask. We found the median 95th percentile peak linear acceleration and rotational acceleration values were within 1 g and 100 rad/s^2, respectively, among these impact locations, so all front impacts were grouped. Head impact exposure metrics developed from this data set, including a highly ranked Division I team and two Ivy League teams, may be representative of different levels of collegiate football, but these data may not be readily extrapolated to other levels of play, such as high school or youth football. Finally, this study does not examine the relationships between head impact exposure and either diagnosis of concussion injury or return to play following brain injury diagnosis. We do note that injury rates have been reported higher in games than in practices [28]. While we have reported a two-fold increase in the number of head impacts per game over practice and an increase in 95^{th} percentile HITsp in games over practices, the correlations between these factors and concussion rates remains to be demonstrated.

C.6 Acknowledgements

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	Player Position						
	QB	**WR**	**DB**	**RB**	**OL**	**LB**	**DL**
Peak Rotational Acc. (rad/s^2)	4956 [3728-5451]	4297 [3478-5122]	4269 [3968-4665]	4869 [4477-5343]	3799 [3401-4380]	4359 [3823-4837]	3891 [3493-4388]
HITsp	34.5 [25.6-39.3]	31 [24.5-35.8]	31.9 [29.4-34.6]	36.1 [32.6-40.2]	29 [25.3-32.1]	32.6 [30.3-36.6]	28.9 [27-32.5]

Table C.1: The median [25-75%] of the 95th percentile peak rotational acceleration (rad/s^2) and HITsp among the various player positions. The values for 95th peak linear acceleration and the number of impacts per season are plotted in **Figure C.3**

Impact Location	Player Position						
	QB	**WR**	**DB**	**RB**	**OL**	**LB**	**DL**
Front	4548 [3399-5500]	3917 [2785-4914]	4573 [3946-4920]	4953 [4441-5786]	4033 [3487-4606]	4810 [4359-5376]	4129 [3534-4582]
Side	3956 [3236-4928]	3739 [3166-4469]	4086 [3636-4729]	5089 [4140-5815]	3129 [2638-3504]	4166 [3672-4682]	3637 [3378-3783]
Top	2802 [1536-3314]	2219 [1599-2921]	2695 [1931-3276]	2688 [2243-3270]	2125 [1717-2626]	2339 [1997-2812]	2487 [2093-2944]
Back	5428 [3985-6156]	4930 [3595-6365]	4184 [3487-5003]	4801 [3845-5492]	4145 [3456-4784]	4242 [3561-5047]	4074 [3766-5116]

Table C.2: The median [25-75%] of the 95th percentile peak rotational acceleration (rad/s^2) for each player position and impact location. Values for 95th peak linear acceleration and the percentage of impacts at each location are plotted in **Figure C.4**

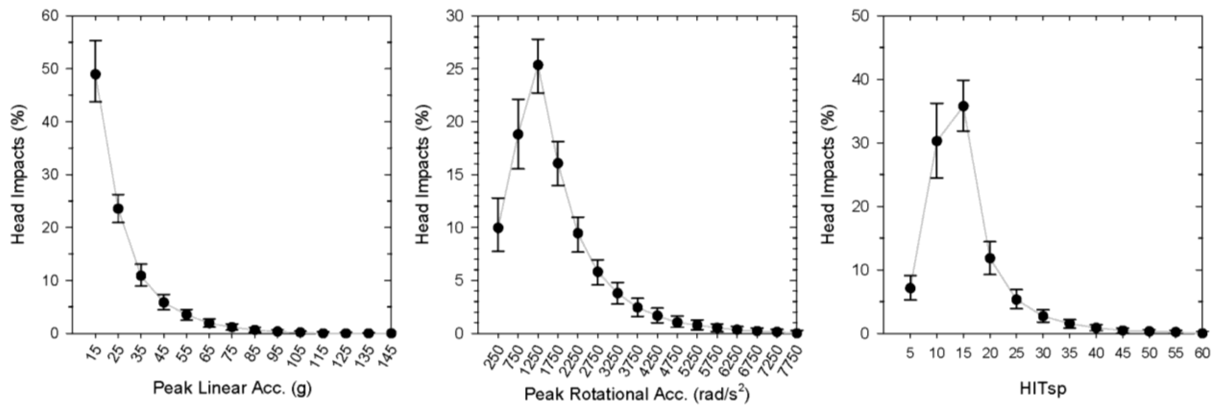

Figure C.1: Study wide peak linear acceleration (g), peak rotational acceleration (rad/s^2) and HITsp distributions of head impacts. Data are a percentage of all impacts for individual players with median [25-75%] values plotted at each bin in the distribution.

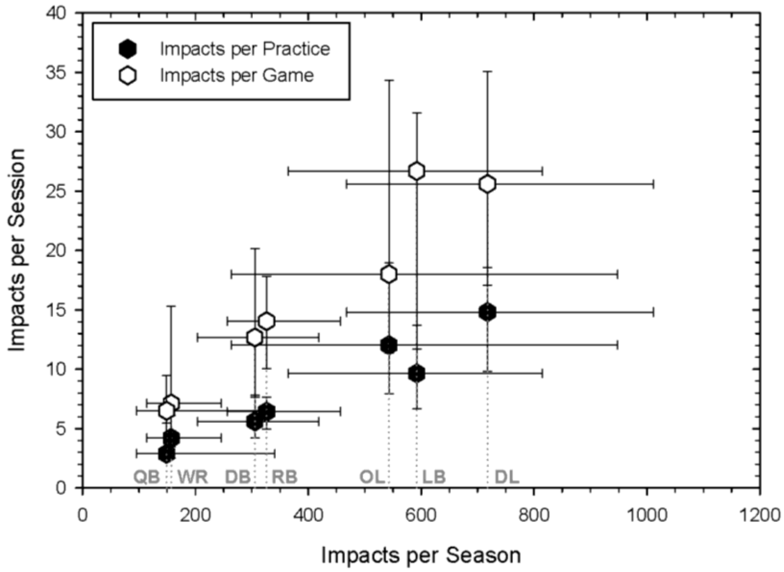

Figure C.2: After categorizing by player position, the median [25% - 75%] frequency of head impacts per practice and head impacts per game were linearly correlated with the median [25% - 75%] frequency of head impacts per season (slope = 0.02, R^2 = 0.934 and slope = 0.04, R^2 = 0.929, respectively). The vertical dotted lines identify the positions associate with each median value.

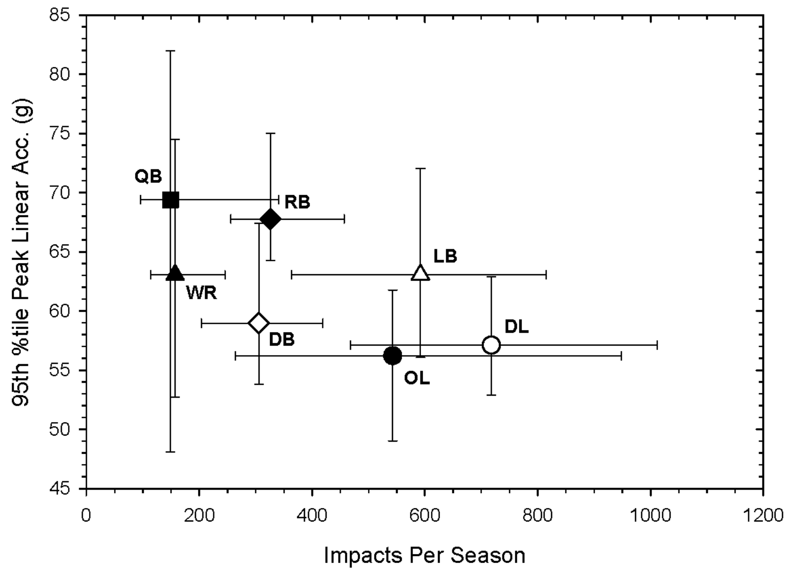

Figure C.3: The median [25%- 75%] of the 95th percentile of peak linear acceleration (g) as a function of the median [25%- 75%] number of head impacts per season and categorized by player position. Analogous values for peak rotational acceleration (rad/s^2) and HITsp are provided in **Table C.1**

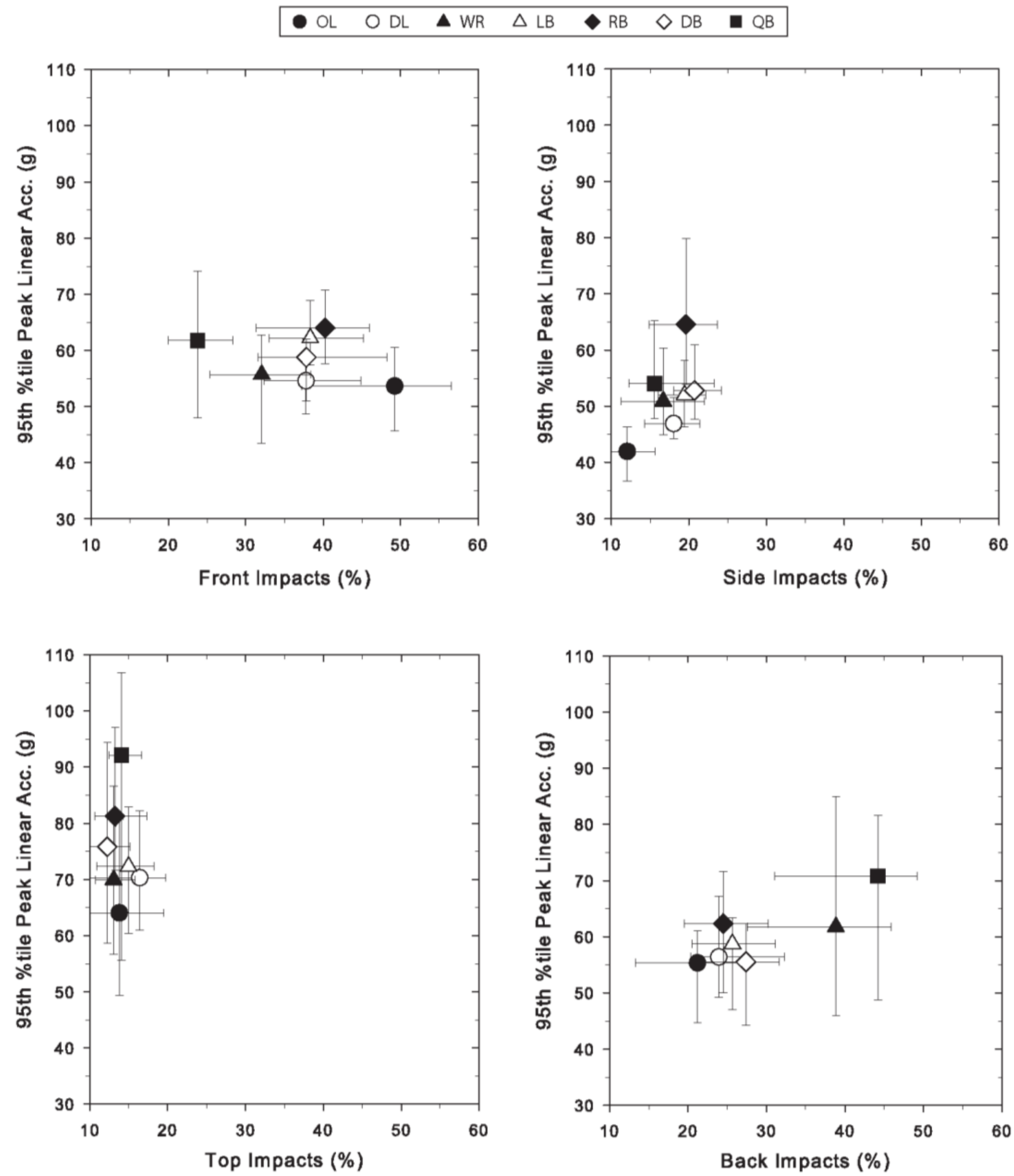

Figure C.4: The median [25%- 75%] of the 95th percentile of peak linear acceleration (g) as a function of the median [25%- 75%] frequency of impacts at each helmet location and categorized by player position. Analogous values for peak rotational acceleration (rad/s^2) are provided in **Table C.2**

Appendix D

Surrogate Head Accelerations from Stick Checks in Girls' Lacrosse

D.1 Introduction

Girls' lacrosse is an incidental contact sport in which the only protective gear that is mandated are mouth guards and eye wear. The disparity between the protective gear worn by girls and the helmet, shoulder pads, arms pads, and gloves worn by boys has heighten the public concerns over the risk of injury in girls' lacrosse. However, this heightened concern may not be warranted as the public in not always well informed about the significant differences in the rules between girls' and boys' lacrosse. Girls' lacrosse does not permit intentional stick or bodily contact, whereas the boys' sport permits such contact.

Moreover, catastrophic head injuries or severe traumatic brain injuries, injuries for which helmets are designed to prevent, have not been recorded in girls' lacrosse to date [1–9]. Nonetheless, head protection in girls' lacrosse is a current and important topic in the nation's fastest growing sport. Informed decisions by governing bodies, rule making bodies, coaches and parents on protective equipment must be based upon scientific evidence. To date there is little scientific evidence on head impacts and resulting head accelerations in girls' lacrosse.

Reducing head, face and concussion injuries is an important and ongoing effort in all sports [10,11]. Because the mechanisms of concussion injuries are poorly understood, these efforts have necessarily taken a multifaceted investigative approach that includes epidemiology, biomechanics, imaging, and neuropsychology. While cellular and biochemical alterations are certainly the direct mechanism for concussion injuries, biomechanics has been clearly established as a major factor, and moreover, biomechanics is the only underlying pathway for prevention and reduction of concussion injuries.

In girls' and women's lacrosse stick contact to the head has been reported to account for the majority of concussion injuries [4,7,12,13]. Our previous studies have measured stick velocities ranging from 13-25 m/s (30-55 mph) during shooting [14]. Given the reported association of stick check with concussion injuries, an understanding of the head accelerations associated with stick checks would provide valuable insight into the possible mechanism of these injuries.

The first objective of this study was to determine head accelerations associated with various stick checks experienced in girls' lacrosse in a laboratory setting. The second objective was to characterize the response of various commercially available protective headgear or helmets to impacts of comparable magnitude. These data will provide scientific evidence to assist governing bodies, rule making bodies, coaches and parents in making informed decisions on all issues related to protective head gear in the sport of girls' lacrosse.

D.2 Methods

D.2.1 Experiment 1

Seven (n=7) female lacrosse players (ages 12-14 years, lacrosse experience 3-7 years) participated in a headform lacrosse checking study after IRB approval and informed written consent/assent were obtained. Six stick models from three manufacturers were used in this study (**Table D.1**). Sticks were prepared for testing by applying five 9 mm spherical retro-reflective markers; two on the tip (each 3 inches from center), one at the base of the head, one at the throat/neck, and one at the shaft (11, 12, and 20 inches from tip, respectively).

Testing was held indoors in the Bioengineering Lab in the Department of Orthopaedic Research at the Warren Alpert Medical School of Brown University and Rhode Island Hospital (Providence, RI). Two headforms were used, the ASTM International (2.550 kg, Model E) and the National Operating Committee on Standards for Athletic Equipment (NOCSAE; 4.376 kg, size: small). The NOCSAE headform is a complex, multiple component silicone/urethanes structure, while the ASTM is a homogeneous solid magnesium structure. Both headforms were mounted on a custom machined neck with a stiffness of 1.8 Nm/deg. A triaxial accelerometer (PCB model 354B173, range ±500 g, PCB Piezotronics, Depew, NY) was mounted in the headforms. Each player completed a total of 36 impacts per headform or 72 hits overall. The order of the stick models was selected randomly. Not all stick models were swung by each player. Three locations were struck on each headform (front, side, top) with 2 points of contact on the stick (throat and shaft). Players were instructed to impact the headform at two intensities (medium and high).

Lacrosse stick swings were tracked at *250* Hz using a four-camera Oqus 5-series infrared sensing system (Qualysis, Gothenburg, Sweden). The cameras were mounted on tripods approximately 1.8 m (6 ft.) high and positioned to the left side of the headform. Markers on the sticks were identified within Qualysis Track Manager and exported to Matlab (Mathworks Inc., Natick MA) for subsequent analysis.

Analog data from the three channels of the accelerometer were collected at 20 kHz and were time synchronized with motion capture data. Resultant acceleration of the center of gravity (C.G.) of the headform was calculated and filtered in accordance with NOCSAE and ASTM standards using a low pass, 4-pole Butterworth filter with a 1000 Hz cutoff frequency. Peak resultant accelerations (g) were identified as the maximum of each signal. Gadd Severity Index (GSI) was calculated for each impact [15]. The impact frame was identified by the time of peak resultant acceleration of the headform. Stick swing linear and rotational velocity were calculated using previously developed algorithms [16]. Briefly, stick swing speed was computed as the speed of the stick at the impact location. The angular swing speed of the bat was computed as the time derivative of the helical rotation [17] (also referred to as the screw rotation) for the two frames prior to ball impact.

D.2.2 Experiment 2

Using an identical test setup to Experiment 1, the second objective was met by impacting the NOCSAE headform with stick impacts at a comparable speed range found in Experiment 1, while it was equipped with four different commercially available helmets (**Table D.2**). These four helmets or headgear were for various sports including lacrosse, rugby, ultimate fighting (UFC), field hockey, and soccer. The men's lacrosse helmet consists of a hard plastic, non-adjustable shell with thick padding inside, a face mask made of metal bars, and a chinstrap used to secure the helmet to the head. The rugby scrum cap is made of shock-absorbing foam situated between an outer layer of Lycra and an inner layer of sweat-absorbing polypropylene. The UFC headgear head gear is made of a polyurethane material, consists of a fully padded forehead, cheeks, and chin, and lace enclosures on the top and a hook-and-loop closure in the back. The women's soft headgear is made of a thin nitrolyte foam layer and is recommended for use in lacrosse, soccer, and field hockey.

For each helmet as well as a test condition with the bare NOCSAE headform, impacts were delivered to two locations on the headform, back and side, using the shaft of the crosse. A total of 20 impacts were completed for each combination of headform and stick locations for each test condition. Headform resultant peak linear acceleration and swing speed were calculated using the same methods described in Experiment 1. Percentage of peak acceleration decrease was computed by normalizing all peak linear acceleration values by swing speed and comparing each test condition to impacts to the bare headform.

Statistical analysis

Impacts were excluded from analysis if the stick did not make square contact with the headform. These impacts were identified if the linear velocity was found to be non-normal to the desired location on the headform. To account for swing speed, only impacts that occurred from swing speeds were identified as mid-range (25th-75th percentile) were included in statistical analysis. Peak linear acceleration and GSI were expressed as mean ± standard deviation. Differences in headform type and impact location on the headform and stick were examined using a two-way ANOVA. Linear regressions were used to evaluate differences for each helmet or headgear test condition in the relationship between stick speed and peak linear acceleration. Statistical significance was set at $\alpha = 0.05$ and the reported *p*-values are those for the post hoc test. All statistical analyses were performed using GraphPad Prism (GraphPad Software, San Diego, CA).

D.3 Results

D.3.1 Experiment 1

A total of 508 impacts were analyzed, 257 impacts on the ASTM headform and 251 on the NOCSAE headform. Midrange of all swing speeds (**Figure D.1**) was found to be 6.3-10.3 m/s or 14.1-23.0 mph. Headform accelerations increased with increasing stick speed (**Figure D.2**). Overall, peak linear acceleration for the ASTM headform (68.1 ± 38.9 g) tended to be higher than the NOCSAE headform (47.4 ± 32.7 g) but significance was not reached (p=0.6). Similar trends were found in GSI (**Figure D.3**). Headform peak accelerations did not vary significantly among headform and stick locations (**Figure D.4**).

D.3.2 Experiment 2

A total of 200 impacts were analyzed, 20 for each test condition. For impacts to the back of the NOCSAE headform, peak accelerations were decreased by 83% for the men's lacrosse helmet, 31% for the rugby scrum cap, 27% for the women's soft headgear, and 23% for the UFC head gear. For impacts to the side of the headform, peak accelerations were decreased by 84% for the men's lacrosse helmet, 59% for the UFC headgear, 54% for the women's soft headgear, and 38% for the rugby scrum cap.

For impacts to the back of the head (**Figure D.5**), the relationship between peak acceleration and swing speed for each helmet model were statistically different

($p<0.0001$). While a significant difference was not found in the slopes for swing speed and peak acceleration for each test condition for impacts to the side of the head ($p=0.144$) (**Figure D.6**), there were statistical differences found between the intercept or peak acceleration values between the test conditions ($p<0.0001$).

D.4 Discussion

The first objective of this study was to determine head accelerations associated with various stick checks experienced in girls' lacrosse in a laboratory setting. This was accomplished by measuring stick speed and surrogate headform accelerations in experienced female youth players.

Stick checks to the headforms resulted in average peak linear accelerations of approximately 50*g*. While this is significantly lower than previously reported accelerations associated with concussions in male high school and collegiate football players (approximately 100*g* [18]), it is higher than the average peak linear acceleration of 43*g* in collegiate female ice hockey players [19]. The average peak linear accelerations reported in this study were also higher than those experienced in a field study of youth girls' soccer players [20]. We also note that the Gadd Severity Index (GSI) values produced in this study are far lower than those reported for impacts associated with concussion in an on-field study of male football players, 439 GSI [18]. While these comparisons are useful for putting the accelerations observed in this study into context with head accelerations experienced in other helmeted and non-helmeted sports, data from these field tests may not be directly comparable to our laboratory test. Additionally, factors that influence biomechanical tolerances to concussion injury, such as age and sex, are still not well understood.

Headform accelerations increased with increasing stick speed. Mid-range impact stick speeds for this youth cohort of experienced female lacrosse players were determined to be 6.3-10.3 m/s and 14.1-23.0 mph. While this is substantially lower than stick speeds observed during shooting in a group of 16 year old male and female lacrosse players, 13-25 m/s (30-55 mph) [14], the sticks used in this study sustained substantial damage at the points that made contact with the headforms. This damage, which is not characteristic of normal play, suggests that the stick checks in this study were unusually aggressive. The headform accelerations observed in this study most likely represent a "worst case scenario" as participants were instructed to purposely stick check the headforms, while on the field, this would be a blatant and egregious foul. In the 2010 Official Rules for Girls and Women's Lacrosse [21] there are several rules that directly pertain to reducing the risk of head injury including stick checks to the head, stick-related fouls in the area of the goal, and dangerous stick follow through. Interestingly, previous studies have found that only 25% of stick related head injuries result in a penalty [13]. While other studies have reported that the majority of head injuries in female lacrosse result from unintentional stick contact to the head [4,7,12], appropriate enforcement of the rules plays a key role in reducing stick check head injuries.

Overall, accelerations of the ASTM headform were greater than those of the NOCSAE headform for the same stick speeds. This could be due to differences in the constructs of the two headforms, the ASTM is solid magnesium while the NOCSAE is made of silicon and urethanes. The NOCSAE headform also has 40% more mass than the ASTM.

The second part of our study focused on testing several models of commercially available helmets and headgear to evaluate how they would perform for stick impacts of the same swing speed and magnitude as those produced by youth female lacrosse players. The four different helmet or headgear conditions included; men's lacrosse helmet, rubgy scrum cap, women's soft headgear recommended for use in lacrosse, field hockey, and soccer, and UFC headgear. For impacts of the same swing speed, peak linear accelerations were decreased for all helmeted impacts compared to those delivered to the bare headform. For impacts to the side of the head, the highest decrease in acceleration was seen in the men's lacrosse, followed by the UFC, the soft head gear, and then the rugby scrum cap. As expected, the helmets and headgear lined with more material decreased the peak accelerations of the headform the most. This is consistent with laboratory tests that have shown that increases in thickness and density of headgear foam can improve impact performance [22]. It should be noted that the men's lacrosse helmet is the only helmet that was evaluated in this study that has a hard plastic shell. For impacts to the rear of the head, the men's lacrosse helmet again decreased peak accelerations the most (83%) but the rugby scrum cap, soft headgear, and UFC all performed the same (ranging from 20-30%). This can be attributed to the fact that all three of these headgear enclose or tighten along the back of the head and have very little to no padding in that location. It should also be noted that not all of the helmets and headgear that were used in this study are designed specifically to protect against concussion injuries. Rubgy scrum helmets are a form of headgear used to protect ears during scrums, from injuries that lead to a condition called cauliflower ear. These have not been shown to prevent concussions [23] and the manufacturers do not claim that its headgear prevents concussions.

Currently, soft headgear is permitted for all female lacrosse players [21]. These head gear are designed to prevent contusions or lacerations of the skull and are rarely warn by players. Presently, a debate on whether to mandate female lacrosse players to wear helmets exists. It is unknown if helmets would reduce the risk of head injury in girls' lacrosse. Some observers contend that the introduction of helmets to girls' lacrosse will increase aggressive play thus leading to higher rates of injury [7,24]. This phenomenon, of more reckless play based on a misguided belief of protection from injury, is known as risk compensation and has been suggested in a variety of sports [25]. Studies have shown there may be a paradoxical increase in injury rates and it is of particular concern in younger athletes [25–27]. Nonetheless, helmets are mandatory in many contact sports and have been shown to reduce the risk of head injuries [28,29].

There are limitations associated with any laboratory experiment as the data does not directly translate to on-field scenarios. Another limitation of the study is that headform accelerations from ball impacts were not studied, but lacrosse balls weigh less than 6 oz., approximately half of what a lacrosse stick weighs, and travel at approximately the same speeds as the stick. A final limitation exists in that the female youth players did not delivered the impacts to the headforms in Experiment 2. Two engineers aimed to swing the lacrosse sticks at the approximate speeds of the impacts in Experiment 1. The relationship between swing speed and peak acceleration was considerably higher for Experiment 2 than Experiment 1. To account for this, Experiment 2 consisted of impacts to the bare headform as well as the 4 test conditions. Data was presented as percentage of decreased acceleration compared to impacts to the bare headform.

There is an inherent risk of injury in any sport. Understanding head impact exposure for individual athletes and the impact response of currently available protective equipment to comparable impact exposure is an important first step in understanding preventative strategies for head injuries. These data provide scientific evidence to assist governing bodies, rule making bodies, coaches and parents in making informed decisions on all issues related to protective head gear in the sport of girls' lacrosse.

D.5 Acknowledgements

This study was supported by US Lacrosse and the National Operating Committee on Standards for Athletic Equipment (NOCSAE). We gratefully acknowledge and thank Rhode Island Hospital Orthopaedic Foundation for all of their technical support. We would also like to thank Tarpit Patel, MS for his support on this project.

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STX Hex 10
Brine
STX AL600
STX Myth
Brine 6065
Debeer 6000

Table D.1: List of lacrosse stick models used in study.

Sport	Model
Lacrosse (males)	Cascade CLH2 (SPR fit)
Rugby	Gilbert Rugby Xact
UFC	UFC OSFM
Women's soft headgear for lacrosse, soccer, and field hockey	HRP sports SG 360

Table D.2: List of protective headgear used in study.

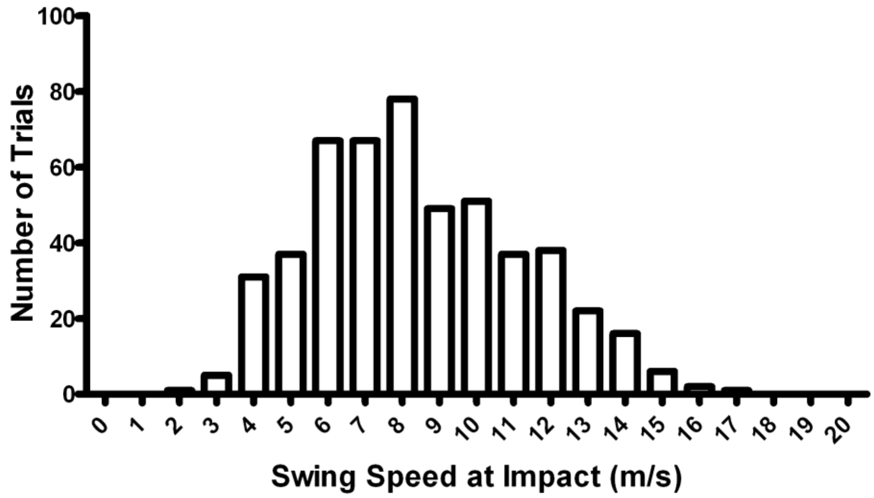

Figure D.1: Distribution of swing speeds in m/s. Midrange swing speed (25^{th}-75^{th} percentile) was identified as 6.3-10.3 m/s (14.1-23.0 mph).

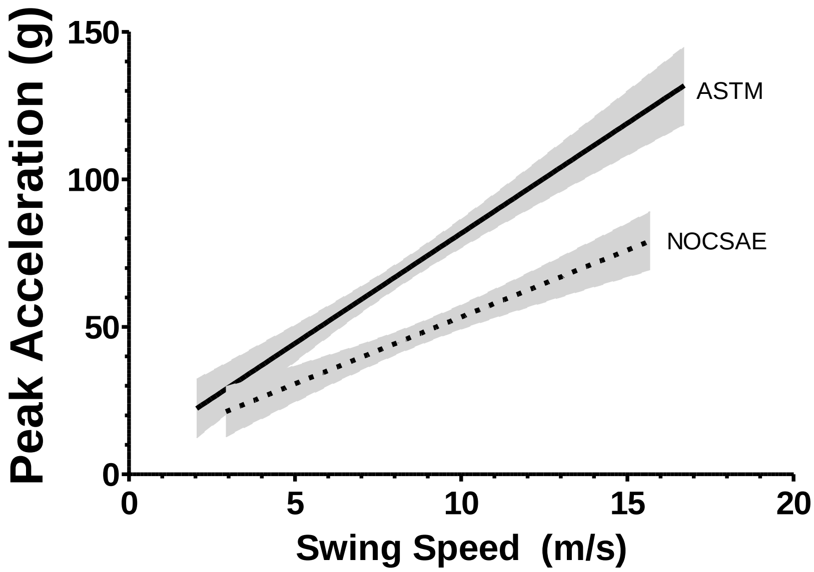

Figure D.2: Peak resultant linear acceleration (g) increases with increased swing speed.

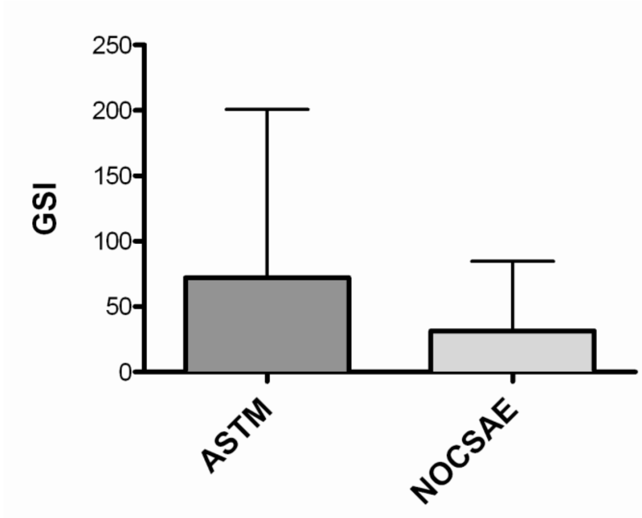

Figure D.3: Gadd Severity Index (GSI) tended to be higher for ASTM headform compared to NOCSAE for impacts in the midrange swing speed (6.3-10.3 m/s (14.1-23.0 mph)).

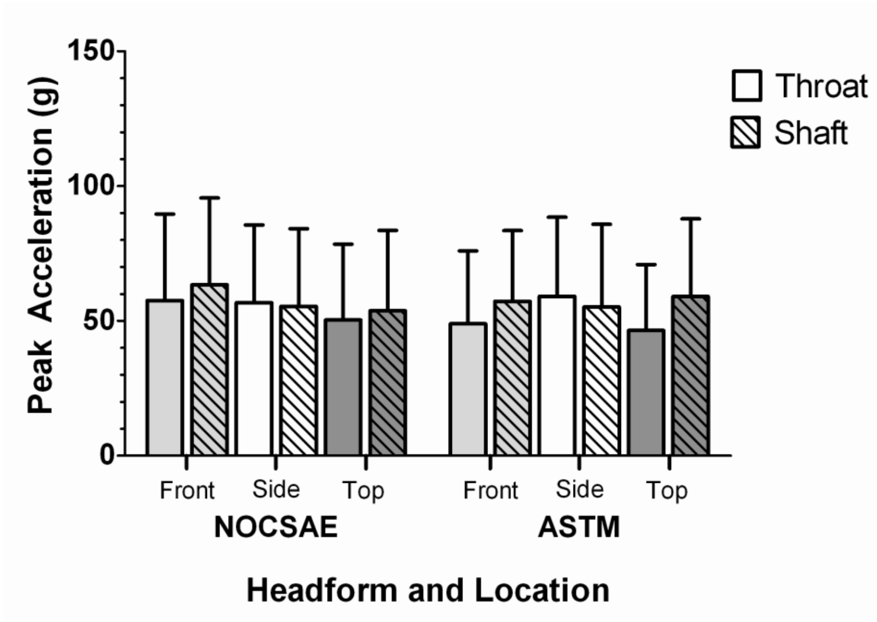

Figure D.4: Mean (standard deviation) peak linear acceleration across mid-range swing speeds (6.3-10.3 m/s or 14.1-23.0 mph) for NOCSAE and ASTM headform across front, side, and top impact locations.

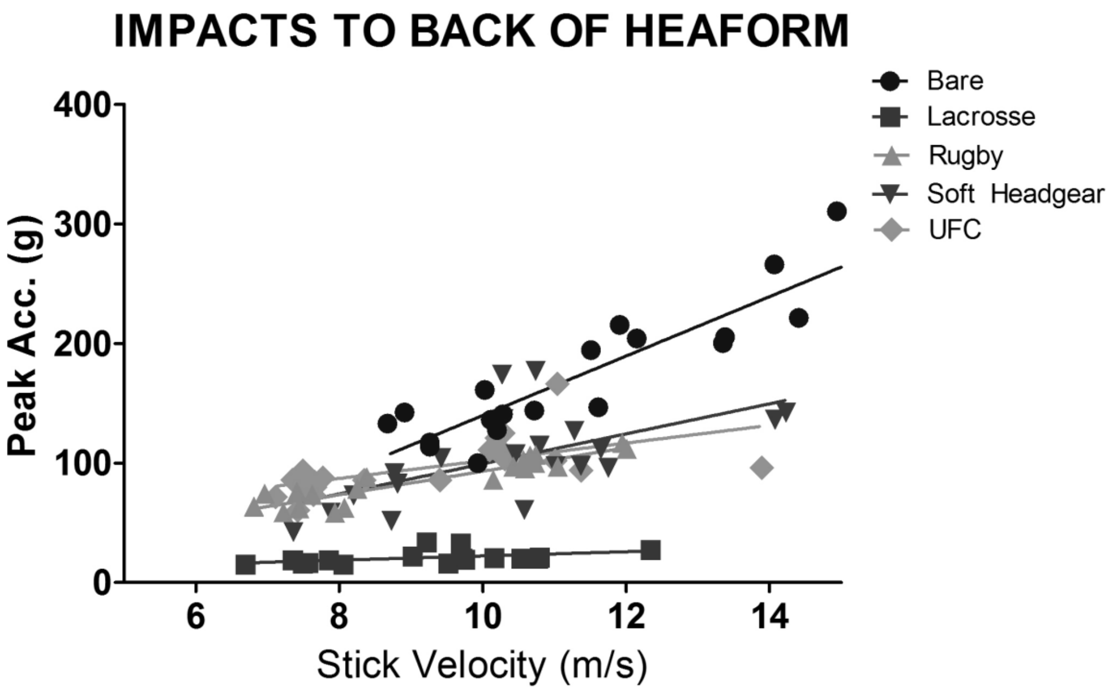

Figure D.5: The relationship between swing speed (m/s) and peak linear acceleration (g) for impacts to the back of the head for a bare NOCSAE headform and 4 different helmets/head gear. Slopes are significantly different (p<0.0001).

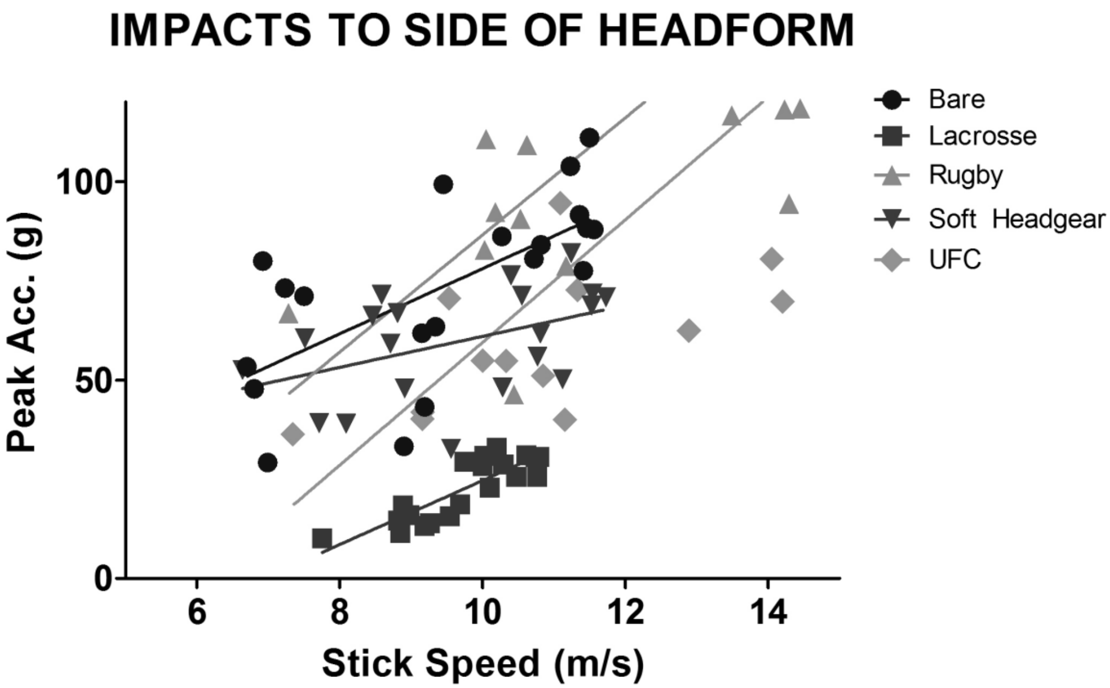

Figure D.6: The relationship between swing speed (m/s) and peak linear acceleration (g) for impacts to the side of the head for a bare NOCSAE headform and 4 different helmets/head gear. Significant differences were not found between slopes ($p<0.144$) but were found in the elevations or intercepts between different test conditions ($p<0.0001$).

Appendix E

Future Study: Salivary Microvesicles as a Novel Biomarker for MTBI and Head Impact Exposure

E.1 Introduction

With the many difficulties and complexities associated with the recognition and diagnosis of MTBI, there is a clear need for the development and validation of on-field diagnostic tools. Concerns over the acute and chronic impairments as a result of MTBI, including the risk of long term cognitive deficits due to repeated injuries during an immediate time frame [1] and second impact syndrome, has lead to the development of a variety of strategies to enhance immediate diagnosis. These experimental diagnostic tools range from sideline cognitive tests to neurophysiologic measures.

Recent studies have evaluated microvesicles as possible biomarkers in different disease states, including traumatic brain injury [2–5]. Microvesicles or exosomes are fragments of plasma membrane that are shed by all cell types, under both normal and pathological conditions, and carry internal cargo representative of their cell of origin, such as membrane and cytoplasmic proteins and mRNA [6–8]. It has been proposed that vesicles in the brain significantly increase with head trauma and that these vesicles will migrate to the olfactory bulb and then to the nasal/oral cavity where they can be detected in saliva. Saliva has long been used as a diagnostic medium because it is accessible, inexpensive, non-invasive, requires minimal training for collection, and can be used in mass screening of large populations [9–11]. Microvesicles within saliva can be identified by their cell of origin and can be quantified by their presence or number, size, and composition. There is potential for microvesicles to not only be an on-field biomarker for the diagnosis of concussion, but also a tool for identifying players who have sustained repeated subconcussive impacts and may be at risk of injury or long term cognitive deficits. The efficacy and feasibility of using salivary microvesicles and other clinical outcome measures related to MTBI could be evaluated by studying the relationship between these variables and head impact exposure.

E.2 Pilot Study

E.2.1 Methods

Saliva samples were collected pre and post game for high school (n=5) and collegiate (n=5) football players. Saliva was collected in 5 ml plastic tubes with Phosphate Buffered Saline (PBS) and stored on ice. The players were not equipped with the HIT system for this pilot study, so head impact exposure was qualitative.

Samples were subjected to 1500g and 17000g serial centrifugation with analysis of the vesicles in the final 120,000 g pellet [12]. Vesicles were then re-suspended in serum free medium 199 containing N-2-hydroxyethylpiperazine acid (HEPES) 25 mM, 1% DMSO, and high protease inhibitor single use cocktail/EDTA free (per manufactures directions). Analysis was completed by a Nanosight laser scattering device (NanoSight, Amesbury, UK). The NanoSight device estimates counts and sizes of small bodies in fluid on the basis of their Brownian motion. The readout is an empirical density function plotting the estimated particle concentration as a function of particle size (**Figure E.1**). Protein content was analyzed with a BCA assay. Samples were compared post-game and pre-game values.

E.2.2 Results

E.2.2.1 Microvesicle Count and Size

A multimodal distribution of particle sizes was found suggesting multiple distinct populations of microvesicle sizes. These vesicles fell within 10 and 1000 nm (**Figure E.1**). This data indicates several populations that were centered at roughly 150nm, 230nm, 350nm, 450nm, 580nm, and 850nm.

E.2.2.2 Protein Analysis

In two players with notable head impacts, PDE4B was elevated 20.3 and 2.5 fold, and ANXA1 was elevated 3.7 and 4.1 fold post-game. ANXA5, MAPK1 and TNFRSF1A were elevated 11.7, 2.8 and 5.1 fold compared to pre-game microvesicle profiles.

E.2.3 Summary

These pilot data indicate that evaluation of salivary microvesicles is feasible and suggests that differences that may occur with head impact exposure will be able to be discerned.

E3. Proposed Future Study

E.3.1 Objectives

The objectives of our proposed future study are to (1) determine if increases in measures of head impacts exposure will correlate with increases in vesicle measures and (2) determine if players diagnosed with concussion will have elevated levels of vesicles compared to non-concussed players.

E.3.2 Methods

Seventy (n=35 Pop Warner and n=35 Brown University) football players will be recruited after IRB approval. Forty (n=40) age and gender matched controls will be recruited from non-contact sports including cross-country and swimming.

Saliva samples will be collected pre and post season for all participants. Contact subjects will be assigned into sub-groups defined by previous head impact exposure frequency and magnitude categories [13]: (1) low frequency, high magnitude (quarterbacks and wide receivers), (2) high frequency, high magnitude (linebackers and running backs), and (3) high frequency, low magnitude (offensive and defensive linemen). Saliva will be collected from selected sets of these contact players 3 times per week (beginning, prior and immediately following each game), as well as age and gender matched non-contact controls. If a player is diagnosed with a concussion, saliva will be collected at two time points (1) within 48 hours of diagnosis and (2) once player is cleared to return to play.

Saliva for vesicle studies will be collected in 5 ml plastic tubes with Phosphate Buffered Saline (PBS) and stored on ice. Batch analysis will then be carried out on samples subjected to 1500g and 17000g serial centrifugation with analysis of the vesicles in the final 120,000 g pellet [12]. Vesicles will be resuspended in serum free medium 199 containing N-2-hydroxyethylpiperazine acid (HEPES) 25 mM, 1% DMSO, and High protease inhibitor single use cocktail/EDTA free (per manufactures directions). Analysis will be done by a Nanosight laser scattering device (NanoSight, Amesbury, UK) which will estimate the relative concentration of vesicles of varying sizes as an empirical density function. Protein content will be analyzed with a BCA assay.

Participants will wear instrumented helmets (HIT System) that record the frequency, magnitude and location of all head impacts that individual players receive in all practices and games. The HIT System is an accelerometer-based monitoring device that computes linear and rotational acceleration of the center of gravity (CG) of the head, as well as impact location on the helmet [14–16]. The HIT System is specifically designed to measure head accelerations by elastically coupling the accelerometers to the head, isolating them from the helmet shell. Data are reduced in post-processing to exclude any acceleration event with peak resultant linear head acceleration less than 10*g* in order to eliminate head accelerations from non-impact events (e.g. running, jumping, etc.). Data reduction methods have previously been described in detail [16–18]. Each player will be assigned a unique identification number and categorized into one of eight position units (see **Figure E.2**).

Head impact exposure is defined for each individual player using measures of impact frequency, location and magnitude. A team session (session) is defined as either a formal team practice (players wore protective equipment with the potential of head contact) or a game (competitions and scrimmages). An individual player is defined to have participated in a session when at least one head impact is recorded for that given player within the specified time of the team session.

Five measures of impact frequency are computed: the total number of head impacts for a player during all practices; the total number of head impacts for a player during all games; the total number of head impacts for a player during all team sessions in a single season; the average number of head impacts for a player during practices; and the average of the number of head impacts for a player during games. Impact locations to the helmet and facemask are computed as azimuth and elevation angles in an anatomical coordinate system relative to the center of gravity of the head [15] and then categorized as front, side (left and right), back, and top [19]. Impact magnitude is quantified by peak linear acceleration (g) and peak rotational acceleration (rad/s^2) [15]. Additionally, a non-dimensional measure of head impact severity, HITsp [19] is computed.

Statistical Analysis Generalized linear modeling will be used for all hypothesis testing. These models will be implemented primarily such that short–term and cumulative measures of head impact exposures will serve as predictors of brain biomarkers and neurological testing. In particular, the sizes, concentrations, protein content and mRNA expression from extracellular vesicles will be determined.

E.3.3 Limitations and Potential Pitfalls

There are several limitations and potential pitfalls associated with the proposed future study, including the inherent risk linked to the novelty of the hypotheses. While data from the pilot study showed encouraging primary results, the analysis techniques are still in the exploratory phase. Before the proposed study can be carried out, a detailed protocol, from the sample preparation phase; to the determination of microvesicle signatures based on size, shape, and number; and finally to quantifying the molecular content of RNA and microRNA, needs to be established and optimized. Consideration needs to be given to how neural injury signatures will be differentiated from nasal, facial, or other soft tissue injuries that may occur with head impact. Other confounding factors that may affect salivary vesicles other than brain injury, such as diet, fluid intake, methods of saliva acquisition, and exercise need to be evaluated. Finally, methods to evaluate the predictive ability of the identified biomarker need to be established and a detailed understanding of how the biomarker compares to pre-existing clinical predictive tools is necessary.

E.3.4 Summary

This will be the first study to evaluate salivary brain microvesicles as a potential biomarker for repeated head impacts and MTBI injury. This future study builds off of the work that has made up the Chapters and Appendices of this dissertation.

This novel work is being spearheaded by a collaboration of researchers and doctors that span several departments between Lifespan's Rhode Island Hospital *and Brown University. My contributions to the project include clinical coordination, collection and analysis of impact biomechanics, collection and transportation of saliva samples, and data analysis. Nanosight and protein analysis of saliva samples has been conducted by members of the Center for Stem Cell Biology laboratory directed by Dr. Quesenberry. Statistical analysis for the pilot study was completed by Jason Machan, Ph.D.*

E.4 References

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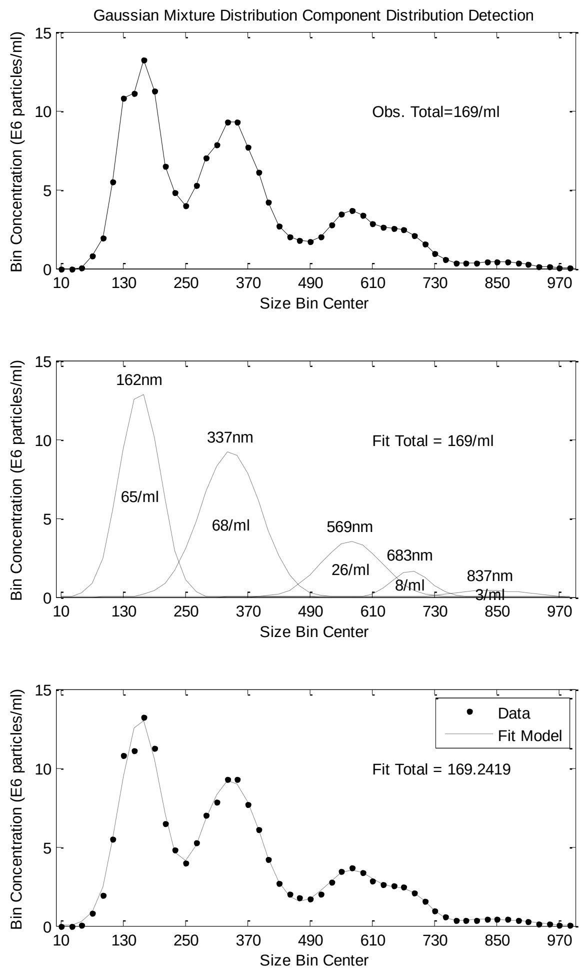

Figure E.1: NanoSight readout (top). Components of non-linear mixture model of size distributions (middle) and sum of components (bottom).

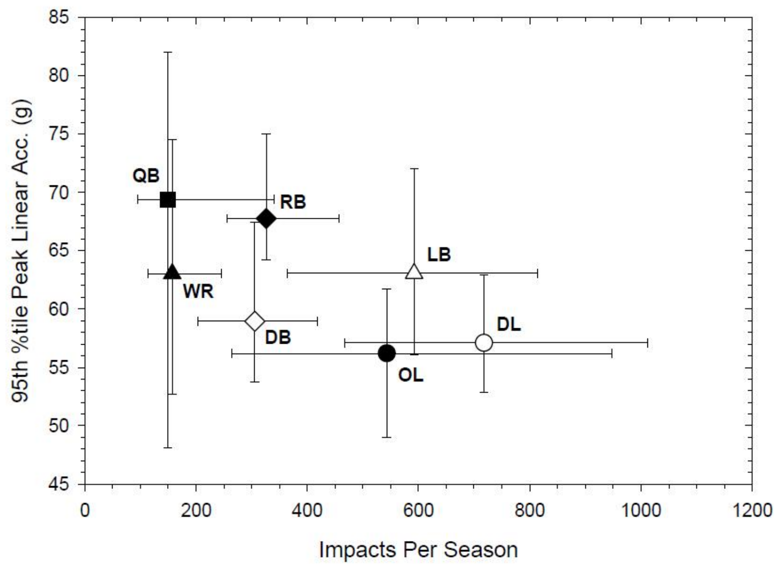

Figure E.2: From our previous studies[13], we found significant difference in the frequency (impacts per season) and severity (95^{th} % peak linear accelerations) as a function of player position. These finding will be used to group contact cohorts by frequency and severity.