FACTORS PREDICTING HEAD IMPACT BIOMECHANICS IN YOUTH TACKLE FOOTBALL

by

RACHEL KHINH LE

(Under the Direction of Julianne D. Schmidt)

ABSTRACT

Statement of the Problem: Previous research has focused on mitigating head impacts to improve sport safety through rule changes and enforcing tackling techniques. However, there is limited information on pre-season factors that could be modified to further reduce head impact severity. We aimed to predict head impact biomechanics on factors such as: (1) head protection time, (2) cervical strength, and (3) anthropometric characteristics and helmet dimensions.

Methods: Twenty-seven youth tackle football players (age=12.2±0.4yrs, height=161.5±9.7m, mass=55.5±15.2kg) participated in this study. Participants completed a pre-season protocol to analyze all objectives. The first objective included a motion capture analysis of a head protection reaction time and eye movement time protocol during a single- and dual-task performance using a light stimulus. The head protection time consisted of a player strategy, head turn, and hand reaction time (s) while eye movement time (s) was recorded using videography. The second objective used a Multi-Cervical Unit to assess cervical isometric strength (N/kg) and time to peak (s) in flexion, extension, left and right lateral flexion. We also captured time to peak in all 4 directions. The third objective captured anthropometric characteristics and helmet dimensions. We used univariate linear regression model with random intercepts to determine if there were associations between our predictors and head impact outcomes. We also performed separate linear regression models to determine the association of predictors and number of impacts.

Results: Our results from the first aim suggests slower eye movement time is associated with an increase in rotational velocity. Our results for aim two indicated stronger cervical strength in the left and right lateral flexion were associated with decreased peak rotational velocity. Slower time to peak left lateral flexion was associated with increased linear acceleration. Lastly, slower time to peak flexion and extension were associated with increased rotational velocity. For our third aim we discovered several anthropometric characteristics and helmet dimensions to influence head impact severity and frequency.

Conclusions: Several factors were found to be associated with increased head impact frequency and severity. Our study found potential modifiable factors, interventions, and designs to improve sport safety and mitigate head impacts among youth players.

INDEX WORDS: Head impact sensors, youth football, biomechanics, head injury, exposure

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DEDICATION

I dedicate this dissertation to my parents and brother. I would not have been able to achieve this accomplishment without your support and encouragement during my academic career for the last decade.

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CHAPTER 1

INTRODUCTION

Sports-related head injuries are a growing concern across varying skill levels in contact and noncontact sports (Mizobuchi & Nagahiro, 2016; Waltzman, Womack, Thomas, & Sarmiento, 2020). Approximately 3.8 million sports-related concussion occur each year with an additional 50% being unreported (Langlois, Rutland-Brown, & Wald, 2006; McCrea, Hammeke, Olsen, Leo, & Guskiewicz, 2004). A concussion often results from a blow to the head or body causing a force transmitted to the brain (McCrory et al., 2017). Concussed athletes experience cognitive and motor control deficits, and increased symptoms of headache, nausea, dizziness, and fogginess (Covassin & Elbin, 2010; Guskiewicz et al., 2005). Moreover, the age of first exposure to contact sports may increase youth athlete vulnerability to the onset of neurobehavioral symptoms later in life (Alosco et al., 2018; Schultz et al., 2018; Stamm et al., 2015). It may be possible that age of exposure increases risk for neurobehavioral issues, but the data are inconclusive (Iverson, Caccese, Merz, Büttner, & Terry, 2021; Zuckerman et al., 2015). However, youth athletes are more susceptible to sustaining concussions which is a concern for neurocognitive development during childhood (Gessel, Fields, Collins, Dick, & Comstock, 2007; Patel, Shivdasani, & Baker, 2005).

A growing body of literature suggests that repetitive sub-concussive blows to the head may impact long-term neurologic health (Alosco et al., 2017) while other research have found that repetitive head impacts do not contribute to long-term consequences (Rose et al., 2021). Most injured athletes who experience increased symptoms and cognitive deficits typically recover within 10 – 14 days (McCrory et al., 2017). However, athletes who experience multiple concussions are at an increased risk for neurological and neuropsychiatric disorders such as depression (Chrisman & Richardson, 2014; Kerr, Marshall, Harding, & Guskiewicz, 2012), mild cognitive impairment (MCI) and Alzheimer's Disease (Guskiewicz et al., 2005). In the absence of concussion, exposure to repetitive head impacts associated with sport participation also raises concern for similar long-term effects and potential harm. In some animal models, repetitive mild head impacts can cause axonal damage (Shultz, MacFabe, Foley, Taylor, & Cain, 2012; Slemmer & Weber, 2005). Higher magnitude impacts from football and ice hockey studies revealed negative effects on brain structure and function (McAllister et al., 2014; McAllister et al., 2012). Thus, there is a critical need to identify interventions that will reduce exposure to repetitive head impacts in sport.

While the research points to the importance of identifying ways to decrease repetitive blows to the head; most head impact related research has focused on high school, collegiate, or professional athletes (Broglio, Martini, Kasper, Eckner, & Kutcher, 2013; Broglio et al., 2010; Broglio, Surma, & Ashton-Miller, 2012; Mihalik, Bell, Marshall, & Guskiewicz, 2007; Schmidt et al., 2016). Yet, children ages 6-13 years old account for an estimated 3.5 million participants in football and have the greatest participation rate compared to high school, college, or professional football (Daniel RW, Rowson S, & SM, 2012; Guskiewicz, Weaver, Padua, & Garrett, 2000; Powell & Barber-Foss, 1999). Youth football players can also sustain head impact frequency and severities similar to their older counterparts (Campolettano, Gellner, & Rowson, 2017; Daniel RW et al., 2012; Kontos et al., 2013; Munce, Dorman, Thompson, Valentine, & Bergeron, 2015). Children are undergoing rapid brain development, which continues throughout the adolescent timeline into early adulthood (Blume, Lucas, & Bell, 2011). During this time, there is constant improvement in cognitive and cortical development as new skills and abilities are acquired (Meehan, Taylor, & Proctor, 2011). Head impacts sustained during this critical period of development could pose a significant threat to brain maturation, holding potential implications for long-term academic performance and social functioning (Blume et al., 2011; Luna, Paulsen, Padmanabhan, & Geier, 2013).

Increased attention has been directed toward possible rule changes and enforcing proper tackling techniques in order to reduce repetitive head impacts and promote player safety in youth tackle football (Broglio et al., 2013). Such interventions have the potential to reduce the severity of head impacts, risk of concussion, and ultimately, late-life neurological and neuropsychiatric disorders (Broglio et al., 2013). An athlete's ability to prepare for and respond to an oncoming collision may depend on their ability to quickly protect their head, visually identify the collision with quick eye movements, utilize their mass and body to absorb force, and quickly activate the cervical musculature. Understanding these features will help us determine whether they warrant intervention to reduce the head impact burden in youth football.

Head Protection Reaction and Eye Movement Time

Head Reaction Time

Sport participation in football requires dynamic motor control to avoid injury with other players in the athletic environment (Eckner, Lipps, Kim, Richardson, & Ashton-Miller, 2011). Athletes commonly fixate on a target and may have decreased odds of sustaining severe head impacts because they can prepare for oncoming or anticipated collision by protecting their head (Kamp, 2011). Functional or sport specific reaction time is crucial and necessary for protective maneuvers (Eckner, Lipps, et al., 2011). Athletes with low visual and sensory reaction time performance sustained a higher number of severe head impacts compared to those with high visual and sensory performance in football (Harpham, Mihalik, Littleton, Frank, & Guskiewicz, 2014). A slower head protection reaction time may increase the injury risk and may result in greater head impact exposure because athletes have inadequate time to respond to oncoming collisions. Athletes with slower reaction time have been correlated with decrease lower extremity kinematics with a history of concussion (Avedesian et al., 2021). Just as individuals with slower functional reaction time may have slower motor responses to protect themselves from lower extremity injury, individuals with slower head protection reaction time may be slower to protect their head making them more prone to repetitive head impacts. However, there is limited research investigating head protection reaction time (Eckner, Lipps, et al., 2011). Researchers have found collision characteristics, such as anticipating impact and facing the oncoming collision, may reduce the severity of head impacts (Caswell, Lincoln, Almquist, Dunn, & Hinton, 2012; Mihalik, Greenwald, et al., 2010). Unanticipated head impacts in youth ice hockey resulted in increased severity compared to anticipated collisions (Mihalik, Blackburn, et al., 2010). Therefore, anticipating or looking in the direction of collisions may help reduce head impact severity. Football players that quickly identify oncoming collisions and respond by protecting their head more quickly may be better able to sustain the blow of impact or avoid it entirely. Results of this study could warrant interventions of head protection reaction time to be implemented in drills and practices in youth football.

Eye Movement Time

Though numerous studies have investigated neurological deficits following head injury, there is still a lack of eye tracking assessments. Eye tracking assesses brain function and represents activation of the central nervous system and its integrity (Samadani, 2015). Smooth pursuits and saccades are often evaluated clinically to detect deficits in concussion assessment batteries (Hunfalvay et al., 2019). Saccades are rapid eye movements of the fovea between fixation points that are established in the horizontal and vertical planes (Hunfalvay et al., 2019). Moreover, saccades are influenced by cognitive and motor processes that are characterized by their velocity, duration, accuracy, and initiation time (Hunfalvay et al., 2019).

Eye movements are the foundation for how humans gather information about the environment (Stuart et al., 2020). In return, this information is used to navigate a safe task or performance (Stuart et al., 2020). Eye tracking is vital in the sport environment especially in contact sports where there is a constant need for performing safety maneuvers. An athlete with slower saccades may sustain head impacts of greater severity because they will be slower in identifying the oncoming collision.

Cervical Muscle Characteristics

Cervical muscle characteristics, such as muscle strength, have been proposed as a preventative strategy to reduce the risk of concussion and severity of repetitive head impacts (Cantu, 1996; Mihalik et al., 2011; Tierney et al., 2008; Viano, Casson, & Pellman, 2007). The osteoligaments and surrounding muscles support the weight of the head and external loads. The ligaments of the cervical spine only contribute to 20% of dynamic stability (Panjabi et al., 1998), with the cervical musculature contributing the remaining 80% (Panjabi et al., 1998). Striking players who are delivering a tackle are able to line up their head, neck, and torso to impart maximum force on their opponent (Viano et al., 2007). Contracting the cervical muscles is an attempt to create more rigidity to counter this force and reduce head acceleration (Viano & Pellman, 2005). When the cervical muscles are relaxed, or when an athlete sustains a collision that is unanticipated, they aren’t aware enough to voluntarily contract cervical muscles to counteract the external force - resulting in rapid head acceleration (Viano & Pellman, 2005).

There is even more concern for adolescent and youth athletes because their cervical muscles are not fully developed and may not reduce head acceleration when sustaining an impact (Mihalik et al., 2011). Previous research has shown that youth football players sustain head impacts of similar severity to high school and college players (Campolettano et al., 2017; Daniel RW et al., 2012; Kontos et al., 2013). This may be linked to youth athletes having insufficient cervical musculature causing rapid acceleration of the head. Youth athletes may lack the ability to counter the external force and may sustain head acceleration causing similar severity to their older counterparts.

A previous study indicated that high school football players who were struck sustained head impacts of higher severity (Schmidt et al., 2014). In the same study, players with greater cervical stiffness had reduced odds of sustaining moderate or severe head impacts compared with players with less cervical stiffness (Mihalik et al., 2011; Schmidt et al., 2014). However, cervical muscle strength did not mitigate head impact severity in their sample of high school and collegiate football (Schmidt et al., 2014). The authors concluded that the dynamic response to perturbation may be of greater value in head protection than just strength alone. It is possible that cervical muscle strength may play a more valuable role in youth athletes. Cervical strength varies according to age (Hildenbrand & Vasavada, 2013). College athletes have stronger cervical muscles compared to high school athletes (Hildenbrand & Vasavada, 2013). Youth athletes may continue this same trend by age and have even weaker cervical strength.

In one study, for every one pound increase in cervical strength, the odds of sustaining concussion decreased by 5% (Collins et al., 2014). There is mixed literature to suggest cervical strength influences head impact severity. One reason may be due to existing literature measuring cervical strength in a variety of ways (i.e. hand-held dynamometers, patient positioning, etc).

However, none have used the gold standard, the Multi-Cervical Unit (MCU, BTE Technologies, Inc., Hanover, MD) (Pearson, Reichert, De Serres, Dumas, & Côté, 2009). Further research is needed to determine if cervical strength is associated with head impact biomechanics in youth football.

Anthropometric Characteristics and Helmet Dimensions

Most youth tackle football teams are created based on age restrictions, though some leagues create teams based on mass restrictions. Players of the same age can vary in physical maturity by as many as 4 years (Linder, Townsend, Jones, Balkcom, & Anthony, 1995). In youth football, height and mass are vary considerably within age groups. Yet, in age-based leagues, two players with vastly different mass could face off against one another on the football field. A player with more mass will have greater momentum and impart greater force relative to a player of smaller mass travelling at the same velocity. In a scenario where these two players strike on another, the smaller player is left to absorb a greater impact load, likely resulting in greater head impact severity. As a solution, some sport organizations have moved towards weight-restricted leagues where the player's mass must fall within a defined range. In weight-restricted league, players may be playing with others of various ages, but have similar weight classifications to promote fair and safe competition (Kerr et al., 2015). In contrast, most leagues are age-restricted, where players are categorized by their age with no reference to their mass. A benefit from age-restricted or age-only leagues is that players are not removed from their age-based peer groups that would happen in weight-restricted leagues (Kerr et al., 2015).

There has been several investigations studying anthropometric data and physical performance among high school, collegiate, and professional football players, but little research regarding youth football players exist (Caswell et al., 2016). Players with increased body mass index and higher stages of maturity had increased injury patterns such as sport-specific injuries (player contact, surface contact, field apparatuses, etc.) among high school football players (Caswell et al., 2016) (Linder et al., 1995) A previous study revealed youth tackle football players who were taller and heavier had higher head impact severity compared those who were shorter with less mass (Yeargin, Kingsley, Mensch, Mihalik, & Monsma, 2018). Further investigations may aid in determining whether age-based or weight-based groups with similar anthropometric data could be advantageous in youth football. Comparing head impacts between shorter vs. taller or heavier vs. lighter football players may help youth football leagues in creating teams to facilitate equal playing performance and reduce the risk of injury.

In addition to the anthropometrics of the body and head, helmet equipment dimensions may also play a role in mitigating head impacts in youth tackle football. There is potential that helmets may reduce the severity of head impacts due to energy absorbing material within the helmet. Different helmet brands also introduce variation in helmet designs. There is no research investigating helmet dimensions such as face mask length, face mask mass, helmet depth, helmet mass, and helmet circumference related to head impact biomechanics in youth football athletes. Tackle football players may sustain head impacts at various locations of the helmet including the face mask. Head impacts to the side of the face mask may result in an increase of rational acceleration because of the longer lever arm. This could also be true for head impacts the occur to the superior or inferior surfaces of the face mask causing an increase in impact severity when the head sustains a force causing rapid flexion and extension. Additionally, football players with heavier face masks may be more prone to sustaining a higher proportion of head impacts to the top of the head suggesting players being more able to lower their head into opponents (Schmidt et al., 2018). A face mask that is longer and distributed further away from the head’s center of mass distributes the mass further from the axis of rotation. This makes it harder to initially accelerate which also means it more difficult to decelerate the head once accelerated. This could be critical for youth players where a large head mass must be supported by underdeveloped cervical musculature making it more difficult to control rapid head acceleration. Therefore, further research is needed to determine helmet dimensions are associated with head impact biomechanics.

Specific Aims

1. To determine the influence of (a) head protection time on head impact biomechanics among youth tackle football players.
2. To determine the influence of cervical muscle strength on head impact biomechanics among youth tackle football players.
3. To determine the influence of (a) anthropometric characteristics and (b) helmet dimensions on head impact biomechanics among youth tackle football players.

Variables

Independent Variables

1. Specific Aim 1a: Head protection time on the following tasks:
    a. Single task player strategy reaction time (s)
    b. Dual task player strategy reaction time (s)
    c. Single task head turn reaction time (s)
    d. Dual task head turn reaction time (s)
    e. Single task hand reaction time (s)
    f. Dual task hand reaction time (s)

    g. Single task eye movement time (s)
    h. Dual task eye movement time

2. <u>Specific Aim 2:</u> Cervical musculature strength during an isometric task:
    a. Peak force by direction (N/kg) (directions: flexion, extension, and right and left lateral flexion)
    b. Time to peak strength (sec) (directions: flexion, extension, and right and left lateral flexion)
3. <u>Specific Aim 3:</u> Anthropometric characteristics and helmet dimensions on the following characteristics:
    a. Height (cm)
    b. Mass (kg)
    c. True Leg length (cm)
    d. Neck Length (cm)
    e. Neck Circumference (cm)
    f. Head Circumference (cm)
    g. Head Mass (kg)
    h. Helmet Depth (cm)
    i. Helmet Circumference (cm)
    j. Length of the face mask in the X direction from the sella turcica (outer ear: fossa of antihelix crura) (cm)

    Normalized by player mass

    k. Total Helmet Mass
    l. Face Mask Mass

Dependent Variables

1. Research Aim 1-4: Head Impact Severity
    a. Linear Acceleration
    b. Rotational Velocity
2. Research Aim 1 – 4: Head Impact Frequency
    a. Number of Impacts

Research Questions

This study focused on the three following head impact biomechanical outcomes during sessions:

1. Linear Acceleration
2. Rotational Velocity
3. Frequency

Research Question 1: Head Protection Time During Single and Dual Tasks

Independent and dependent variables will remain continuous for statistical analyses.

a. Does *player strategy reaction time* during a single- or dual-task performance predict head impact severity and number of head impacts?

b. Does *head turn reaction time* during a single- or dual-task performance predict head impact severity and number of head impacts?

c. Does *hand reaction time* during a single- or dual-task performance predict head impact severity and number of head impacts?

d. Does *eye movement time* dual a single- or dual-task performance predict head impact severity and number of head impacts?

Research Question 2: Cervical Muscle Strength

a. Does *cervical peak strength* in (a) flexion, (b) extension, (c) left and (d) right lateral flexion predict head impact severity and number of head impacts?
b. Does *time to cervical peak strength* in (a) flexion, (b) extension, (c) left and (d) right lateral flexion predict head impact severity and number of head impacts?

Research Question 3a: Anthropometric Characteristics

a. Does *height* predict head impact severity and number of head impacts
b. Does *mass* predict head impact severity and number of head impacts?
c. Does *true leg length* predict head impact severity and number of head impacts?
d. Does *neck length* predict head impact severity and number of head impacts?
e. Does *neck circumference* predict head impact severity and number of head impacts?
f. Does *head circumference* predict head impact severity and number of head impacts?
g. Does *head mass* predict head impact severity and number of head impacts?

Research Question 3b: Helmet Dimensions

a. Does *total helmet mass* predict head impact severity and number of head impacts?
b. Does *face mask mass* predict head impact severity and number of head impacts?
c. Does *helmet depth* predict head impact severity and number of head impacts?
d. Does *helmet circumference* predict head impact severity and number of head impacts?
e. Does *length of face mask* from the center of skull mass predict head impact severity and number of head impacts?

Research Hypotheses

Research Hypotheses for Aim 1: Head Protection Time

a. Slower head protection player strategy, head turn, and hand reaction time during a single- and dual-task will have increased head impact severity and number of head impacts.

b. Slower time with eye movements during a single- and dual-task performance will have increased head impact severity and number of head impacts.

Research Hypotheses for Aim 2: Cervical Muscle Strength

a. Youth football players with weaker cervical peak strength in all 4 directions will have increased head impact severity and number of head impacts.

b. Youth football players with slower time to peak in all 4 directions will have increased head impact severity and number of head impacts.

Research Hypotheses for Aim 3a: Anthropometric Characteristics

a. Youth football players with a shorter height will have increased head impact severity and number of head impacts.

b. Youth football players with less mass will have increased head impact severity and number of head impacts.

c. Youth football players with shorter true leg lengths will have increased head impact severity and number of head impacts.

Normalized to their body mass:

d. Youth football players with longer neck lengths will have increased head impact severity and number of head impacts.

e. Youth football players with a larger neck circumference will have increased head impact severity and number of head impacts.

f. Youth football players with smaller head circumferences will have increased head impact severity and number of head impacts.

g. Youth football players with larger head mass will have increased head impact severity and number of head impacts.

Research Hypotheses for Aim 3b: Helmet Dimensions:

a. Youth football players with greater total helmet mass will have greater head impact severity and number of head impacts.

b. Youth football players with greater face mask mass will have greater head impact severity and number of head impacts

c. Youth football players with greater helmet depth with have greater head impact severity and number of head impacts.

d. Youth football players with greater helmet circumference will have greater head impact severity and number of head impacts.

e. Youth football players with greater face mask length will have greater head impact severity and number of head impacts.

Operational Definitions

1. *Triax Smart Impact Monitors (SIM-G; Triax Technologies Inc., Norwalk, CT):* Each sensor contains a triaxial accelerometer that measures linear acceleration and gyroscope that measures rotational velocity.
2. *Player Strategy Reaction Time:* This will be calculated from the time it takes to turn or rotate one’s head in the left or right, raise their hands in a blocking mechanism or move

their eyes in the direction of a visual light stimulus. Whichever occurs first will be used for the player strategy reaction time.

3. *Head Turn Reaction Time:* This will be calculated from the time it takes to turn or rotate one's head in the left or right direction to face the visual light stimulus once cued.
4. *Hand Reaction Time*: This will be calculated from the time it takes to raise one's hand/arm as performing a stiff-arm motion in football. Participants are required to face straight ahead and to only raise their hand left or right depending on the corresponding light stimulus.
5. *Eye Movement Time*: This will be calculated from the time it takes for the pupil to reach the corner of the eye in the direction of left or right depending on the light stimulus. We counted number of frames between light stimulus and when the pupil reached the end range then divided by the seconds passed.
6. *Single Task:* This condition will consist of players/participants completing the head protection reaction time while responding the visual light stimulus.
7. *Dual Task:* This condition will consist of players/participants completing a cognitive task (i.e. spelling words backwards) in addition to responding to the visual light stimulus.
8. *Peak Force by Direction:* peak force from the four directions of cervical flexion, extension, right and left lateral flexion.
9. *Time to Peak:* Defined as time it takes for the participant to reach cervical peak output.

Limitations

1. The Triax SIM-G has mixed evidence regarding its validity of head impacts.
2. Results of this study do not translate over to concussive events.
3. The results of this study are not generalizable to different age groups of youth tackle football.

Delimitations

1. Youth tackle football players were recruited from a single recreational department in Southeastern United Stated.
2. We did not examine head impacts that resulted in concussion.
3. Data were collected across a single season in 2021.

Clinical Significance of the Study

The purpose of this study is to determine if head protection reaction time and eye movement time, cervical muscle characteristics, and anthropometric characteristics and helmet dimensions predict head impact biomechanics in youth tackle football players. The primary role of head protection reaction time is to better understand mitigating head impact severity. This has not been studied in youth football players and proposed that preparing for oncoming collision in sport could reduce the odds of sustaining a more severe head impact. Results of this study could encourage coaches to optimize players' head protection reaction time and eye movement time through interventions or drills for safety and mediating collision severity. Focusing on looking at collisions and bracing for impact may mitigate injury. Youth players have developing cervical musculature. It is not known however if cervical characteristics influence head impact biomechanics with this age group. The results of this study may influence coaches and stakeholders to design cervical strengthening and conditioning programs. There is little evidence to suggest that anthropometric data of youth players influences head impact biomechanics. However, results of our study may warrant for youth football players to be placed in either age- or weight-restricted teams to allow for improved safety. Lastly, face mask lengths have not been examined in football as it pertains to head impact rotational velocities. Results of our study could allow for future studies to examine different types of face masks to reduce head impact severity. Future research from this study could further improve helmet design for youth football players.

CHAPTER 2

LITERATURE REVIEW

Introduction

Mild traumatic brain injuries (mTBI), such as concussions, can occur when there is a direct or indirect blow to the brain (McCrory et al., 2017). This injury raises a serious concern to the public health in the United States (Langlois et al., 2006). Concussions often result in increased symptoms and neurocognitive deficits that effect daily activities and work productivity (Cancelliere, Coronado, Taylor, & Xu, 2017; McCrory et al., 2017). There is also a financial burden of $17 billion for those seeking care and treatment (Cancelliere et al., 2017). Most individuals recover from concussion within 10-14 days while others may have persistent symptoms and impairments for even longer (McCrory et al., 2017).

Repetitive head impacts raise another concern regarding the health and safety of acute and long-term consequences (McAllister & McCrea, 2017). However, it is crucial to understand that head impact biomechanics are not equivalent to concussion biomechanics. Impact severities are heterogenous among concussed individuals (McAllister et al., 2014). Individuals may sustain a concussion at a lower impact severity while others sustain a concussion at a higher severity (Zhang, Yang, & King, 2004). There is little research investigating factors that could mitigate head impact frequency and severity in youth football.

Concussion Epidemiology

An estimated 1.6 to 3.8 million sports-related concussions occur each year (Langlois et al., 2006). Sports-related concussion are the second leading cause of TBI among young people between the age of 15-24 years old while motor vehicle accidents are first. (Sosin, Sniezek, & Thurman, 1996). Understanding the ongoing epidemiology of concussion is crucial for improving safety in athletics and to guide targeted preventative measures.

In the high school setting, concussions represent about 13.2% of all sport-related injuries (Marar, McIlvain, Fields, & Comstock, 2012). The overall incidence rate of concussion accounted for 6.2% of all NCAA injuries (Zuckerman et al., 2015). An epidemiological study report revealed slightly lower incidences of concussion that represented 5.5-8.9% of all injuries; however, did not include contact sports like ice hockey and lacrosse (Gessel et al., 2007). The overall concussion rate found in that study is approximately 2.5 concussions per 10,000 athlete exposure (AE; 1 athlete participating in one game or practice which he or she is exposed to injury) (Gessel et al., 2007).

Preliminary high school concussion rates were low overall, but football having the highest with about 4.5 concussion per 10,000 AEs (Guskiewicz & Valovich McLeod, 2011). Girls soccer had similar concussion rates as wrestling with about 3.0 concussions per 10,000 AEs (Guskiewicz & Valovich McLeod, 2011). When comparing gender-comparable sports in high school, females have a higher concussion rate than males (Gessel et al., 2007; Marar et al., 2012). Additionally, females have higher rates of recurrent concussions than males (Castile, Collins, McIlvain, & Comstock, 2012). When comparing concussion by age, half of all concussions among youth and adolescents resulted from participation in sport (Meehan & Mannix, 2010). This is concerning for youth athletes as there is increased vulnerability for a repeat concussion when the first concussion is sustained at a young age (Guskiewicz & Valovich McLeod, 2011).

Overall, football players sustained most concussions, new and recurrent, by player to player contact. (Castile et al., 2012; Lynall, Campbell, Wasserman, Dompier, & Kerr, 2017). In regards to gender, boys overall sustain concussion from player to player contact whereas females commonly sustain concussion from contact to the playing surface (Castile et al., 2012). During games, youth football players have a greater proportion of sustaining concussions based on surface contact compared to high school and college football (Lynall et al., 2017). Additionally, youth football players participating in games have a greater proportion of concussion occurring while being tackled compared to high school and college football (Lynall et al., 2017). Participation in games is also associated with an increased risk of concussion compared with practices in youth football players (Kontos et al., 2013). A previous study found the majority of concussion were the results of head-to-head contact which is more defined than player to player contact in existing literature (Kontos et al., 2013).

Concussion Neurophysiology

In 2014, Giza and Hovda proposed a new model to describe the neurometabolic cascade phenomena that follows concussion (Giza & Hovda, 2014). This is crucial for understanding the underlying pathophysiology that is linked to clinical characteristics of concussion such as symptoms, vulnerability to repeat concussions, and cognitive impairments (Giza & Hovda, 2014). Traumatic brain injuries, including concussion, set off a complex and interwoven sequence of ionic and metabolic events (Giza & Hovda, 2014). Even without neural damage, concussion can present with neurological signs and symptoms after a biomechanical force to the brain (McCrory et al., 2017).

Immediately following injury, there is a membrane disruption and axonal stretch (Giza & Hovda, 2014). This causes potassium channels to open which results in sodium and calcium influx into cell and an efflux of potassium from the call into extracellular space (Giza & Hovda, 2014). Glutamate releases which activates N-methyl-D-aspartate (NDMA) and D-amino-3-hydroxl-5-methyl-4-isoxazole-propionic acid receptors (AMPA). Activation of glutamate, NDMA, and AMPA further exacerbate the potassium efflux and sodium-calcium influx (Giza & Hovda, 2014). Ultimately, this exchange creates a diffuse "spreading depression-like" state that may reveal very acute post concussive impairments. In order for the membrane disruption to be restored, the sodium-potassium channels work overtime. However, this results in an energy crisis. There are increasing amounts of adenosine triphosphate (ATP) being consumed (Giza & Hovda, 2014).

Within minutes following brain injury, there is an upregulation of cellular glycolysis (hyperglycolysis) (Giza & Hovda, 2014). However, lactate becomes a byproduct of glycolysis and builds up in the neuron (Giza & Hovda, 2014). There is mismatch between blood flow and demand for glucose. There is an increased demand for energy with normal or reduced cerebral blood flow. The activation of the NMDA receptor causes a rapid and sustained influx of calcium (Giza & Hovda, 2014). This then causes the elevated levels of calcium to be sequestered by the mitochondria, but will eventually lead to dysfunction, damage, and sometimes death. In return, this further increases the cell's dependency on glycolysis-generated ATP (Giza & Hovda, 2014). The resulting energy crisis serves for an increased vulnerability for sustaining secondary injuries which is especially dangerous for the already injured brain (Giza & Hovda, 2014). The axonal dysfunction and altered neurotransmission from injury may lead to cognitive deficits and a myriad of symptoms due to white matter damage even in the absence of cell death (Giza & Hovda, 2014). Concussions are heterogenous and often have inconsistent symptom presentation and cognitive impairments (McCrory et al., 2017).

Similar neurometabolic processes have been found subconcussive, repetitive head impacts in football. During subconcussive events, asymptomatic male football players experienced significant changes in concentration of glutamate+gluatamine, and total choline containing compounds in the dorsolateral prefrontal cortex (Bari et al., 2019). These neurometabolic alterations were significantly associated with the average acceleration of head impacts exceeding 50*g* (Bari et al., 2019). In youth football players, repetitive head impacts in the absence of a clinically diagnosed concussion resulted in changes in white matter microstructure, a brain tissue responsible for transmission of electrical nerve signals (Bahrami et al., 2016). Previous studies have also shown multiple hits to the head <15g over a season lead to brain damage affecting neuronal health, white matter integrity, and impairing cognitive function (Bailes, Petraglia, Omalu, Nauman, & Talavage, 2013; Sundman, Doraiswamy, & Morey, 2015). The etiology behind chronic traumatic encephalopathy may be due to repetitive axonal injury from subconcussive head impacts (Gavett, Stern, & McKee, 2011; Morley, 2018). When an axonal injury occurs, there are also axonal disconnections in the brain (Morley, 2018). Repetitive blows to the head can generate enough force to disrupt neuronal integrity without resulting in clinically evident symptoms of concussion (Morley, 2018). It is proposed that repeated axonal injury sets up a series of metabolic, ionic, and cytoskeletal disturbance that triggers a neurometabolic cascade leading to CTE (Gavett et al., 2011; Morley, 2018).

Long Term Consequences & Negative Outcomes

There has been increasing awareness regarding the short- and long-term consequences from concussion and/or repetitive head trauma on overall brain function. Negative subconcussive impacts may trigger onset of chronic traumatic encephalopathy, cognitive impairments, and depression.

Chronic Traumatic Encephalopathy (CTE) is diagnosed as the presence of tau protein within the cerebral tissue that results in neurological deterioration (Stern et al., 2011). It has been theorized that CTE results due to axonal stretching caused by repetitive head impacts that trigger a neurodegenerative cascade and neurocognitive decline (Stern et al., 2011). Sport type, competition level, and playing career duration may all be factors that influence the risk of developing CTE (Stern et al., 2011). Interestingly, all CTE cases have a history of head trauma, but not all cases of head trauma lead to CTE. A predominant controversary exists over the risk of exposure to brain trauma and CTE due to its prevalence in cases.

The symptoms of CTE, such as cognitive, emotional and behavioral changes, are very distinct from post-concussion syndrome. Rather than immediately following injury, CTE typically presents with an even more delayed onset even decades after exposure (McKee et al., 2009). The onset of CTE is rather progressive with a gradual decline in brain function (McKee et al., 2009). Behavioral changes from CTE are often concerning involving depression, apathy, violence, anxiety, and suicidal thoughts (McKee et al., 2009) (Stern et al., 2011). Due to the concerning awareness of CTE, states have proposed bills to ban youth tackle football (DC, 2018). However, more research is needed due to current studies having biased samples of people who volunteer to donate their brain, no control groups, and very limited control of those with co-morbidities.

Approximately 80-90% of concussed individuals experience symptoms following injury but then recover within 10-14 days (McCrory et al., 2017). Post-concussion syndrome is defined as a concussed individual who have prolonged recovery beyond a month or 3 months post-injury. (McCrea, Prichep, Powell, Chabot, & Barr, 2010; McCrory et al., 2017). Approximately 15% of concussed individuals may experience symptoms of physical (e.g. headache), cognitive (e.g.

difficulty concentrating), and emotional (e.g. anxiety) domains that persist for weeks up to years following injury (Williams, Potter, & Ryland, 2010). However, definitions of post-concussion syndrome differ across diagnostic criteria when using the Procedural Classification System (PCS-ICD-F07.2) and Diagnostic and Statistical Manual of Mental Disorder (DSM-IV) (Williams et al., 2010). Additionally, symptoms associated with post-concussion syndrome are not specific to PCS due to the overlap of symptoms found in other clinical populations such as individuals with depression, chronic pain, and whiplash (Williams et al., 2010).

Williams et al. theorized that post-concussion syndrome from the injury may cause stress which triggers depression and anxiety (Williams et al., 2010). This could also be due to neuronal changes that occur in areas of the brain the modulate mood (Kerr et al., 2012). The prevalence of depression is already high in individuals after sustaining a severe TBI (Williams et al., 1997). A study by Guskiewicz et al., found that retired professional football players with a concussion history of 3 or more are at an increased risk for being diagnosed with clinical depression compared to those with no concussion history (Guskiewicz, Marshall, et al., 2007). In a study by Chrisman et al., the researchers found similar results in adolescents with a history of concussion (Chrisman & Richardson, 2014). Areas of the brain that show structural changes with depression include the hippocampus (Sheline, Sanghavi, Mintun, & Gado, 1999), amygdala (Sheline et al., 1999), orbitofrontal cortex (Lacerda et al., 2004), and basal ganglia (Baumann et al., 1999). Having a history of concussions could put individuals at an increased for recurrent concussion and major depression. Although this link remains unknown, it is rather possible that structural changes in the brain may lead to mental health disorders such as depression.

However, there is mixed-research to suggest repetitive head impacts are not linked to long-term consequences. One study found no difference between white matter following a season of youth tackle football (Nilsson et al., 2019). An additional study found no relationship between repetitive head impacts and cognitive tests or behavioral/symptom questionnaires in youth tackle football players (Rose et al., 2021). In that same study, premorbid conditions such as attention-deficit/hyperactivity disorder, anxiety, and depression were associated with worse cognitive or behavioral scores more than cumulative impacts (Rose et al., 2021).

Concussion Biomechanics

It is crucial to understand that concussion biomechanics and head impact biomechanics are not the same. Ommaya and Gennarelli were among the earliest researchers to describe the linear and rotational acceleration of concussion using animal models (Ommaya & Gennarelli, 1974). In a coup injury, sufficient force from a linear impact can cause the brain to strike against the skull in the direction the force was traveling (Barth, Freeman, Broshek, & Varney, 2001; Meaney & Smith, 2011; Ommaya & Gennarelli, 1974). In the event of a countercoup injury, the brain may also have the potential to rebound in the opposite direction (Barth et al., 2001; Ommaya & Gennarelli, 1974). Linear acceleration was thought to be less severe as these studies showed focal brain tissue stress (Ommaya & Gennarelli, 1974). In addition to linear acceleration to the brain, rotational mechanisms may result in more severe brain tissue deformation and alteration due to shear forces (Ommaya & Gennarelli, 1974). With rotational force, there are multiple sites in which the brain may contact the inner lining of the skull (Barth et al., 2001; Ommaya & Gennarelli, 1974). Rotational forces may cause a more diffuse axonal injury, such as concussion, since kinetic energy is transmitted through the skull which causes deformation and induces shear-strain injury (Barth et al., 2001; Ommaya & Gennarelli, 1974). Multiple signs and symptoms of injury may manifest from neurometabolic processes when the brain exceeds physiological or mechanical thresholds from external forces (Giza & Hovda, 2014).

Head Impact Biomechanics & Accelerometry

Using real time, on-field head accelerometers is novel way to measure linear and rotational acceleration to better understand the biomechanics of concussion and repetitive head impacts. Naunheim et al were one of the first studies to introduce and use in vivo accelerometry that were imbedded in the helmet of high school hockey and football players during games (Naunheim, Standeven, Richter, & Lewis, 2000). The results of this study showed football players sustained on average 29.2g and 35g for hockey players. These numbers are slightly higher than current estimates of peak linear acceleration (Broglio et al., 2009; Mihalik, Guskiewicz, Jeffries, Greenwald, & Marshall, 2008). Since then, there have been numerous studies analyzing on-field linear and rotational acceleration with head impact accelerometry devices. Concussions can result from varying impact severities (Guskiewicz & Mihalik, 2011). For example, an individual may sustain a concussion from a lower severity impact compared to a teammate who sustained a higher severity impact, but with no concussion. Additionally, a higher number of sustained impacts does not lead to a concussion.

Efforts have been made to understand biomechanical outcomes of head impacts and determine a threshold of concussion. Zhang et al proposed an injury threshold for mild traumatic brain injury (Zhang et al., 2004). The researchers considered ≤66g as mild, 66g> or <106g as moderate, and ≥106g severe (Zhang et al., 2004). However, the majority of head impacts tend to occur within 25-30g. Mihalik et al reported less than 0.35% of head impacts over 80g of linear acceleration resulted in concussion (Mihalik et al., 2007). Broglio et al. identified that no significant relationships existed between symptoms and cognitive performance change scores and impact biomechanics; even the number of impacts had no significance (Broglio, Eckner, Surma, & Kutcher, 2011). In a similar design, Guskiewicz et al. also found that head impact biomechanics that resulted in a concussion had no relationship on cognitive function, postural control, and symptomology (Guskiewicz, Mihalik, et al., 2007). These studies suggest that post-concussion deficits are independent of head impact biomechanics since concussions can occur from a wide range of severity (Broglio et al., 2011; Guskiewicz, Mihalik, et al., 2007).

Similar to the negative outcomes following concussion, repetitive head impacts also raise the same concern towards cognitive declines. McCaffrey et al. analyzed short-term clinical outcomes in collegiate football player who sustained low and high impact severities. The results of this study suggest that impact greater than 90g did not alter cognitive function, postural control, or symptoms (McCaffrey, Mihalik, Crowell, Shields, & Guskiewicz, 2007). However, more research is needed to understand additional factors that may influence or predict head impact biomechanical outcomes. Previous research has found concussed players to experience about 94 more head impacts, 10.2g greater head severity impacts, and about 2 times greater risk-weighted exposure compared to their matched controls (Rowson et al., 2019). Additionally, 72% of all concussed athletes had the most or second to most severe head impact exposure compared with their matched controls (Stemper et al., 2019). Also, 77% of athletes that participated in 10 or more days of contact activity had increased head impact exposure than their matched controls (Stemper et al., 2019). In combination, these studies suggest that repetitive head impact exposure lowers tolerance to concussion.

Wearable accelerometry devise quantify on-field frequency and severity of head impact in sport to further understand impact and improve safety. No on-field head impact biomechanics system is perfect. Previous studies suggest the SIM-G has inconsistent validation data (Cummiskey et al., 2017; Karton, Oeur, & Hoshizaki, 2016; Oeur, Karton, & Hoshizaki, 2016; Tyson, Duma, & Rowson, 2018). Previous investigations have reported optimal impact detection with 100% of non-helmeted impacts detected and 80% of helmeted impacts detected in one study (Oeur et al., 2016), and 99% of helmeted impacts detected in another study (Cummiskey et al., 2017). There was also no difference between head form and SIM-G linear acceleration at low and medium energy impacts in one study (Karton et al., 2016) and approximately 14.25% error on average found in another study (Cummiskey et al., 2017). A separate investigation reported less than optimal acceleration validity where there was underpredicted linear and rotational acceleration in helmeted and non-helmeted conditions (Tyson et al., 2018). We used the SIM-G because it can record head impact data from helmeted participants regardless of helmet manufacturers and brand. To overcome instrument limitations, commonly employed methods and initially-filtered manufacture parameters include removing insidious head impacts that went through a multi-step cleaning process to only include real on-field head impacts (Figure 2.1).

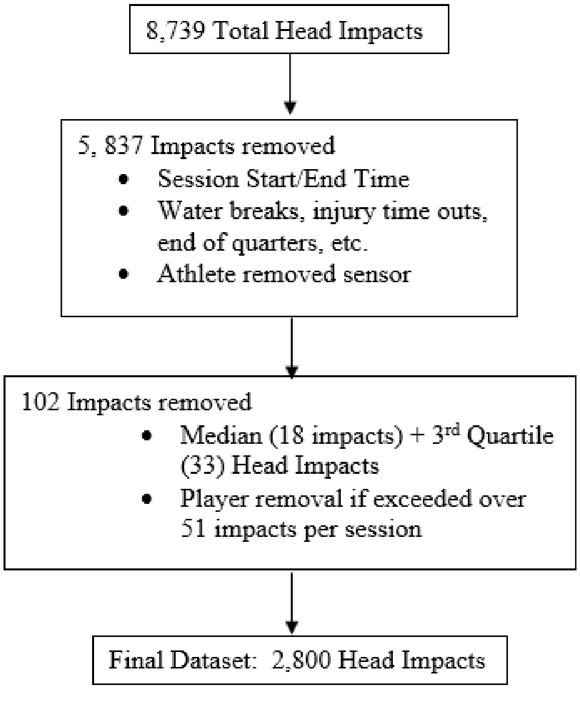

Figure 2.1. Multi-step data extraction of head impacts

Mitigating Head Impacts

Factors affecting head impact biomechanics include cervical strength, rule changes, and player on-field behavior (Mihalik, Blackburn, et al., 2010; Pellman, Viano, Tucker, & Casson, 2003; Schmidt et al., 2014). To mitigate the severity and frequency of head impacts, researchers have suggested rule changes, modifying helmet to helmet tackles, enforcing proper tackling techniques, and reducing drills with high impact rates (Campolettano, Rowson, & Duma, 2016; Daniel RW et al., 2012). State legislators have also proposed laws to ban tackle football for youth athletes. However, youth football players rarely sustain high severity impacts greater than 40g (Campolettano et al., 2017; Campolettano et al., 2016; Lynall, Lempke, Johnson, Anderson, & Schmidt, 2019). A typical threshold in head impact literature is greater than 10g.Impacts occurring under 10g are typically found in activities such as jumping and running (King, Hume, Gissane, Brughelli, & Clark, 2016). To improve the safety of youth tackle football, there are considerations to mitigate head impact severity and promote future interventions. Previous studies have suggested that flag football may be a safer alternative to tackle football. However, flag football players still sustain repetitive head impacts, although less frequent and of lower severity (Waltzman et al., 2021). Results from these studies suggest fewer head impacts found in flag football have the potential to minimize concussion risk (Waltzman et al., 2021). However, previous research states that youth flag football players were still sustaining head impacts with higher odds of sustaining increased severity of impacts than tackle football (Lynall et al., 2019).

Understanding other factors beyond mechanism or adjusting rules and regulations are warranted to explore ways to mitigate impact frequency and severity. Athletes who have slower head protection reaction time and look at an oncoming opponent may not prepare themselves to prepare for impact such as contracting the cervical musculature (Eckner, Lipps, et al., 2011).

Additionally, athletes with weaker isometric cervical muscles may not be able to counteract the external force and experience rapid head acceleration (Viano et al., 2007; Viano & Pellman, 2005). Anthropometric and helmet dimensions may also influence head impact severity in youth tackle football athletes. Mitigating head impacts through these factors may improve sport safety and reduce severity while still allowing youth football players to participate.

Head Protection & Reaction Time

Sport participation requires intact dynamic motor control and fast reaction time to avoid injury with opponents or objects in the athletic environment. In order to reduce concussion risk and repetitive head trauma, head protection reaction time, or responding to prepare for impact, is pivotal for football players to prepare for impact and reduce the risk of sustaining further injury (Eckner, Lipps, et al., 2011). Individuals with slower head protection reaction time may be slower to protect their head making them more prone to repetitive head impacts. Current guidelines for return to play after concussion stress the use of a multifaceted approach, but there is no guideline regarding functional movements (McCrory et al., 2017). Prematurely returning to play could place an athlete at increased risk for sustaining repeated head impacts (Eckner, Lipps, et al., 2011). Moreover, having an increased reaction time may not allow for an athlete to maneuver quick enough to avoid injury. Additionally, a slower head protection reaction time has the potential for football players to sustain impacts of greater severity due to the inability to prepare for impact quick enough. Research suggests increased or slower reaction time during lower extremity functional tasks such as jump landing is related to increased lower extremity injury risk (Taimela, Osterman, Alaranta, Soukka, & Kujala, 1993). This may be translatable to the head and upper extremity injury risk and head impact biomechanics. However, there is little research on head protection reaction time as it pertains to head impacts (Eckner, Lipps, et al., 2011). Clinical reaction time from computerized neurocognitive tests, which is used for determining readiness to return to sport, is typically measured by a single key on a computer keyboard or mouse clicking. But it is not clear whether this clinical reaction time measure adequately represents the multiple demands of sport (Lempke, Johnson, Schmidt, & Lynall, 2020). From a concussion perspective, this measure is moderately correlated to the number of days from initial post injury to return to play which is within 14 days of injury (Alsalaheen, Stockdale, Pechumer, Broglio, & Marchetti, 2017). However, current clinical reaction time tests do not gather information from sport-specific tasks or functional reaction time (Lempke et al., 2020).

Single tasks such as quiet standing or walking have been used to assess concussion recovery over time. Dual tasks serve to simulate multitasking in sport performance requirements such as running while listening to coaching demands, direction from teammates, and performing alternate routes in youth football (Howell, Osternig, & Chou, 2015). Cognitive motor tasks (e.g. spelling words backwards) included in combination with a functional movement (e.g. gait, balance) are pivotal for assessing impairments following concussion (Howell et al., 2015). Further, normal daily activities require cognitive and physical demands, simultaneously (Kleiner et al., 2018). If the demands of performing the dual-task exceed cognitive capacity, then performance on one or both tasks will be diminished and may lead to injury (Kleiner et al., 2018). Previous studies have found decreased gait speed and greater sway during a dual-task motor performance in those with a concussion (Howell et al., 2015). Currently, dual task assessments are not measured during concussion baseline or longitudinally. It is also unknown if healthy individuals prior to the start of the season with slower reaction time and poorer dual task outcomes sustain greater head impact severity. We will further explore how these factors might predict head impact biomechanics in youth tackle football athletes.

Eye Movement

Oculomotor behavior can be categorized into the following eye movements: fixations, smooth pursuits, and saccades (Land & Tatler, 2009). In recent concussion assessments, smooth pursuits and saccades are often used to detect deficits. Smooth pursuits are defined when the eyes track a moving stimulus to stabilize the image on the fovea (Hunfalvay et al., 2019). Saccades are rapid eye movements of the fovea between fixation points that are established in the horizontal and vertical planes (Hunfalvay et al., 2019). Different brain regions become activated depending on the eye movement type (Hunfalvay et al., 2019). Moreover, saccades are influenced by cognitive and motor processes that are characterized by their velocity, duration, accuracy, and initiation time (Hunfalvay et al., 2019). However, several injuries and disorders such as Parkinson’s supranuclear palsy, and traumatic brain injury can affect saccadic performance in humans (Leigh & Zee, 2015). A quicker saccade may allow for players to see and react quicker to oncoming collisions. Whereas a slower saccade may put the player at greater risk of injury due to the inability to react quick and prepare for an oncoming collision.

Controlling eye and arm movement is central to sport performance and natural behavior (Dean, Martí, Tsui, Rinzel, & Pesaran, 2011). Moreover, eye-hand coordination depends on a combination of retinal signals for accurate movement (Dean et al., 2011). A quicker saccadic movement speeds up and enhances target detection (Diederich, Schomburg, & Colonius, 2012). Therefore, rapid saccadic movements are necessary in football to protect the head by means of bringing the hands towards the head or object by blocking opponents. A previous study found deficits on self-paced saccade tasks in patients with TBI or post-concussion syndrome (Diederich et al., 2012). In youth athletes, right eye skew was significantly higher in those with a concussion compared to healthy controls (Howell et al., 2018). These deficits may reveal lingering consequences translated over to sport-specific eye tracking demands such as tracking a ball or opponent. In comparison to control subjects, those with concussion displayed impaired self-paced saccades such as slower saccades with lower peak velocity or fewer saccades in timed task (Johnson, Zhang, Hallett, & Slobounov, 2015; Mulhall, Williams, & Abel, 1999; Williams et al., 1997). These tests are simple and quick; however, more investigations are needed to integrate eye-tracking assessments into the sport setting. However, it is unknown if preseason measures of eye-tracking movements can predict head impact severity.

Cervical Muscle Strength

In relation to the human brain, preparing for oncoming collision by contracting the cervical musculature may reduce the severity of rotational acceleration and aforementioned axonal injury (Collins et al., 2014). Head impact biomechanical research has found collision characteristics, such as anticipation and head direction, may influence the severity of head impacts using wearable sensors (Lincoln, Caswell, Almquist, Dunn, & Hinton, 2013; Mihalik, Blackburn, et al., 2010). Unanticipated head impacts in comparative sports, such as youth ice hockey, resulted in higher severity compared to anticipated collisions (Mihalik, Blackburn, et al., 2010). In addition, lateral or side head impacts have also been investigated to influence concussion risk and impact severity (Pellman et al., 2003). When athletes sustain a collision that is unanticipated, they lack the ability to voluntarily contract neck muscles to counteract the external force - resulting in rapid head acceleration (Viano & Pellman, 2005).

The osteoligaments and surrounding muscles support the weight of the head and external loads, with cervical spine ligaments contributing just 20% of dynamic stability (Panjabi et al., 1998). The cervical musculature contributes the remaining 80% for stability (Panjabi et al., 1998). During a cervical flexion movement with all the muscles maximally activated, the sternocleidomastoid is responsible for the majority of the movement capacity (69%), with the longus capitis and colli (17%) and scalenus anterior (14%) contributing to the remaining cervical strength output (Newtons) (Vasavada, Li, & Delp, 1998). In flexion, the moment-generating capacity comes from the semispinalis (37%) and splenius (30%) (Vasavada et al., 1998). For lateral bending, the sternocleidomastoid (28%) and trapezius (19%), with the scaleni, splenius, levator scapulae, semispinalis, and erector spinae contribute from a 5-15% range (Vasavada et al., 1998). For rotation, the moment generating capacity is greater for trapezius (32%), followed by 10-20% each from the splenius, sternocleidomastoid, semispinalis, and suboccipital muscles (Vasavada et al., 1998).

Less cervical strength is associated with greater head acceleration when being struck (Viano et al., 2007). Striking players who are delivering a tackle are able to line up their head, neck, and torso (Viano et al., 2007). A stronger or stiffer neck occurs when there is coupling of the players' torso into a collision. Contracting the cervical muscles is an attempt to create more rigidity to counter this force and reduce head acceleration (Viano & Pellman, 2005). A previous study found that collegiate athletes are stronger, more powerful, and better coordinated than high school and adolescent athletes (Baker, Horton, Robertson-Wilson, & Wall, 2003). The maturity of cervical strength should be considered between age groups. Normalized peak force values were captured by dividing peak force by body mass (N/kg) will help gather more information as it pertains to player anthropometrics.

It is unknown whether resistance training to improve cervical strength reduces head acceleration (Viano et al., 2007). Weak cervical strength has been proposed and hypothesized as a risk for injury (Pellman et al., 2003; Viano et al., 2007; Viano & Pellman, 2005). However, strength and conditioning programs targeting the cervical musculature are not commonly introduced in sports. Understanding cervical strength in youth athletes may help explain head impact biomechanics during sport participation. In high school football, players with greater cervical stiffness had decreased odds of sustaining head impacts in the moderate and severe categories compared to those with less cervical stiffness (Schmidt et al., 2014). However, football players with stronger and larger cervical muscles did not mitigate head impact severity (Schmidt et al., 2014). In youth ice hockey, there were no differences in linear or rotational acceleration across different strength groups for the cervical muscles (Mihalik et al., 2011). However, cervical strength was a significant predictor of concussions (Collins et al., 2014). A stronger neck resulted in a decreased odd of sustaining concussion (Collins et al., 2014). It is possible that cervical muscle strength may play a more valuable role in youth athletes. Every one pound increase in cervical neck strength, the odds of sustaining concussion decreased by 5% (Collins et al., 2014). Mixed results may exist due a variety of measurements used to assess cervical strength. Results from future studies may help coaches and healthcare professionals implement cervical strengthening programs to reduce the risk of injury and mitigate head impact severity.

Anthropometric Characteristics and Helmet Dimensions

Most youth tackle football leagues are established based on age-restrictions and other leagues are based on weight-restrictions. Body mass index differences among high school football players have shown injury patterns of concussion and head injury negatively for athletes of lesser mass (Caswell et al., 2016). Youth athletes who sustain concussion tend to have a longer recovery and symptoms resolution than older athletes (Field, Collins, Lovell, & Maroon, 2003). However, little research exists among youth athletes. Critical periods of growth and maturation along with variability in anthropometric characteristics occur within age groups. Not to mention, there is great variability in skill level and player development. Although youth football players have lower head impact exposure than high school and collegiate football players, their impact severity can be identical to their older counterparts (Munce et al., 2015). Therefore, some leagues offer weight-based categorization for players as a potential strategy to reduce injuries related to size. However, there is limited evidence to suggest this strategy actually reduces head impact severity. For example, a player aged 11 years old but of lighter weight may play with heavier players of the same age. An oncoming collision involving a player with greater mass may come with more velocity causing a more severe impact for the lighter player. The lighter player is left to absorb a greater impact load, likely resulting in greater head impact severity.

To date, only one study has analyzed anthropometric data among youth football players and head impact biomechanics (Yeargin et al., 2018). The researchers found boys who were in the older age category had a greater linear and rotational acceleration than those in the younger category (Yeargin et al., 2018). Rotational acceleration was lower in those who were taller (Yeargin et al., 2018). Those considered overweight based on BMI had greater linear and rotational accelerations (Yeargin et al., 2018). Future research should further investigate if additional anthropometric characteristics such as neck circumference influence head impact biomechanics in youth football.

Additionally, there are no in vivo studies regarding helmet dimensions in reference to the athlete for head impact biomechanics. Future research should gather total helmet mass, circumference, depth, face mask mass, and face mask length from the center of mass of the athlete's head and reference these outcomes to athlete's body mass. Results from this type of study design could help improve the safety of sport through equipment changes to help mitigate head acceleration. A face mask distributed further away from the center of mass of the heads may distribute the weight further from the axis of rotation and lead to head down mechanisms when tackling (Schmidt et al., 2018). This may make it harder for the head to initially accelerate, but also more difficult to decelerate once the head becomes accelerated. As mentioned previously, youth athletes do not have mature neck strength coupled with a disproportionate helmet could cause an increase in head impact severity.

Methodological Considerations

This study focused on factors influencing head impact severities in youth tackle football players. We have chosen to examine youth tackle football players because they account for the highest participation rate in football compared to high school, college, and professional but are rarely studied (Daniel RW et al., 2012). Youth football players can also sustain head impact frequency and severities similar to their older counterparts (Campolettano et al., 2017; Daniel RW et al., 2012; Kontos et al., 2013; Munce et al., 2015). Many state legislators wish to ban youth tackle football; however, the results of our study could influence methods in reducing head impact severity through saccadic training and cervical strength interventions.

We have chosen to study head protection reaction and eye movement time in youth tackle football players. Anecdotally, there are several ways players are able to block opponents when tackling or being tackled. However, functional movements have yet to be analyzed in youth players and if their outcomes can mitigate head impact severity remains unknown. If players can see and react to oncoming collisions faster, they may be able to prepare for collision by contracting their cervical musculature to reduce acceleration of the head or utilize a stiff-arm maneuver. We have chosen to study reaction and movement time because previous research has shown concussion risk and impairments increase when there is also cognitive impairment (Eckner, Lipps, et al., 2011; Howell et al., 2018; Howell et al., 2015; Lempke et al., 2020).

We have chosen to study cervical strength in youth tackle football players using the MCU because this is considered the gold standard but has not been studied. There is mixed literature to suggest cervical strength influences head impact severity. One reason may be due to existing literature measuring cervical characteristics in a variety of ways (i.e. hand-held dynamometers, custom devices, patient positioning, etc), but, none have used the gold standard MCU. It is important to note, however, that the MCU does not allow for rotational measures which is arguable the most important direction. Time to peak was included because peak force measures may not best demonstrate the role of cervical strength in preventing or mitigating rapid head acceleration.

We have also chosen to gather anthropometric characteristics among youth tackle football players. Some youth leagues use age-restricted or weight-restricted to create teams. Other state legislators would like to ban tackle football for youth athletes altogether. We've chosen to study anthropometrics to further explore head impact severity within an age-restricted team of varying heights and masses. It is unknown if withholding age-restrictions is a safer alternative for youth tackle football players.

We gathered helmet dimensions among our cohort of youth tackle football players. We have chosen to study helmet dimensions because there is no research investigating helmet dimensions as it relates to the proportion of the player's mass and head impact biomechanics.

Summary Rationale for Study

Results from this study will highlight preseason factors that may influence in-season head impact biomechanics. The role of head protection reaction time has not been previously studied. If the results of our study suggest reaction time does play a role in impact severity, then sports medicine staff could place emphasis on improving reaction time to stimulus prior to games. Additionally, this study will answer lingering research questions regarding cervical strength in youth football athletes. Results from this study could guide interventions to be integrated in sport practices or conditioning programs by coaches and stakeholders to reduce head impact severity. Lastly, there is little evidence regarding anthropometric and helmet dimensions predicting head impact biomechanics. Although some youth football leagues have changed rules for teams being built based on age- or weight-restrictions, it is unknown if this is a safe alternative to reduce head impact severity. Additionally, it is unknown if helmet dimensions play a role on in-season head impact biomechanics. The results of our study may aid stakeholders and league developers to improve the sport of youth tackle football through proper rule or equipment changes without eliminating youth football entirely.

CHAPTER 3

METHODOLOGY

Study Participants

A youth tackle football team was recruited from a single recreation department in the rural southern United States. Twenty-seven youth tackle football players participated in this study during the 2021 fall football season. The study protocol was approved by our Institutional Review Board prior to subject recruitment. We obtained child assent with at least one parent or guardian providing signed informed consent.

Inclusion Criteria: Any youth tackle football players participating on identified teams (age groups 12-13-years old).

Exclusion Criteria: Any youth tackle football players not willing to wear head impact biomechanics monitoring equipment during practices and games. For the cervical strength assessment, we will exclude players with prior cervical injury, cervical deformities, and neurological disorders such as learning disability, ADD/ADHD, psychiatric or mental health disorders.

Study Design

During this prospective cohort study, participants completed one pre-season testing session prior to the start of their first game that lasted approximately 1.5 hours. Participants completed all protocols during the testing session in Table 3.1

Table 3.1. Research Design, Timeline, Measures

Preseason Data Capture **08/24/2021 – 09/10/2021** Head protection, cervical strength, anthropometrics, and helmet dimensions were administered in a random order	**In-Season Data Capture** **09/11/2021-11/06/2021**
Head Protection Time Testing (administered first) • Player Strategy Remaining conditions were randomized: • Eye Movement Time • Hand Reaction Time • Head Reaction Time Cervical Testing in Order: • Flexion • Extension • Right Lateral Flexion • Left Lateral Flexion Anthropometric Characteristics randomized: • Height • Mass • True Leg Length • Neck Length • Neck Circumference • Head Circumference • Head Mass Helmet Dimensions randomized: • Total Helmet Mass • Face Mask Mass • Helmet Depth • Helmet Circumference • Length of Face Mask	Triax SIM-G data collection at practices and games • Linear acceleration • Rotational velocity • Number of head impacts

Measures & Instrumentation

The head protection reaction and eye movement time protocol included procedures for measuring head reaction time, hand reaction time, and eye movement time, and while reacting to a light stimulus. The cervical testing protocol included procedures for measuring cervical isometric strength: total peak force and time to peak. Anthropometric characteristics that were measured included: height, mass, true leg length, neck length, neck circumference, head circumference, and head mass. Helmet dimensions that were measured included: total helmet mass, helmet depth, helmet circumference, face mask mass, length of the face mask from the center of mass of the skull; approximately posterior to the ear. All testing protocols were completed in the Biomechanics Laboratory.

Head Protection Time

Prior to the testing session, players were asked to bring their helmets, shoulder pads, and to wear athletic clothing with no reflective material. Any reflective pieces on the helmet or shoulder pads were covered with black tape. Players were instructed to react to a left or right light stimulus placed 4.5 meters away and 45° from the player according to the following conditions: player strategy, head turn, hand reaction time, and eye movement time. Players completed the player strategy condition first and the remaining conditions were randomized. All conditions were completed under single-task (i.e. focusing on completing the task as quickly as possible) and dual-task (i.e. spelling words backwards as quickly and accurately as possible). Before each trial, players were instructed to get in an athletic stance with knees slightly flexed by the words “get set” from the research member. Players began spelling words backwards prior to any movement. Trials were repeated if participants did not initiate movement when cued or did not perform the correct movement. Players were able to complete a familiarization trial for each condition that was not included in the final analysis. The single- and dual-tasks were alternated within each condition across trials. Players completed a total of 4 single- and 4 dual-tasks for each condition. The left or right light stimulus was given at random.

All reaction and eye movement conditions occurred in a 3D motion capture space compromising 8 cameras (MIQUIS, Qualisys Systems, Goteborg, Sweden) recording at 240Hz. An additional video camera was placed on the floor and captured the eye movement condition at 480p/400Hz. Players were fitted with helmet markers consisting of 4, 14-mm retroreflective markers to track reaction time. To measure head movement, markers were placed on the back of their helmet and on the styloid process of each ulna for the hand reaction condition. A research team member visually looked at the participant and on camera to ensure they were not performing trunk movements. Trials were repeated if the participant moved their trunk or eyes were not fixated ahead. The middle of the 4 markers was placed 4cm superior from the most distal part on the back of the helmet. Single markers were placed over the left and right ulnar styloid processes. For each condition, players were instructed to take an athletic stance after being told to “get set” by the research team. For the dual-task, players were instructed to “get set” then given a word to spell backwards. A visual light stimulus (green) was triggered randomly within 2 to 5 seconds to signal movement while they were spelling words backwards. All sessions began with the player’s strategy condition followed by the three remaining conditions in random order.

Player’s Strategy

Players were instructed to face straight ahead then quickly identify the light stimulus and perform a blocking technique with their hand to protect their head as if the light represented an oncoming opponent. Reaction time was calculated by the time the light was cued to $\geq$1cm of movement of either the helmet markers or the ulnar markers. To obtain their reaction time, we took the average time across trials occurring at each helmet marker, left and right ulnar markers, and eye movement time during the single task performance. Then we determined the fastest time and that was their player strategy outcome measure. The same process was repeated for the dual task performance.

Head Turn

Players were instructed to face straight ahead. They were instructed to turn their head to face the light stimulus as soon as it was cued as if they were facing an oncoming opponent (Figure 3.1a). Reaction time was calculated by the time the light was cued to $\geq$1cm of movement of the helmet markers (Figure 3.1b) in any direction.

Figure 3.1a. Participant reacting to light with head turn reaction time condition.

Figure 3.1b. Markers placed on back of helmet.

Hand Reaction Time

Players were instructed to face straight ahead and to raise their hand as if performing a stiff-arm motion towards the corresponding light that was cued as if they were blocking an opponent and protecting their head (Figure 3.2). They were instructed to keep looking straight ahead. Reaction time was calculated by the time the light was cued to $\geq$1cm of movement of the helmet markers in any direction.

Figure 3.2. Participant reacting to light with hand reaction time condition.

Eye Movement Time

Players were instructed to look straight ahead and to only move the eyes to look at the light and stay fixated on the light until told to “relax”. Eye movement time was calculated by the time it took for the pupil to reach the end range of the saccadic movement when the light went off (Figure 3.2). The additional camera (480p/400Hz) placed in front of the player contained a view with a mirror in the background to reflect the light box that captured at 1200Hz. Trials were repeated if there was any head movement. Eye movement time was calculated by subtracting the time the light was cued from the time the eyes completed the saccadic movement.

Figure 3.3. Participant reacting to light with eye movement time condition.

Data Processing

All marker data were exported from the motion capture software (Qualisys Track Manager; Qualisys System) and processed using Visual 3D (Visual 3D; C-Motion Inc, Germantown, MD). We also recorded the strategy sequence (i.e. eye-head-hand, head-eye-hand, etc) for descriptive purposes during the player strategy condition.

Research Aim 1: To address the first aim, we used the average single and dual task times for each condition and player to predict impact severity of peak linear acceleration, peak rotational velocity, and number of head impacts.

Cervical Strength Protocol

Players were seated in the multi-cervical unit that captured force output at 20Hz (MCU, BTE Technologies, Inc, Hanover, MD). The chair and proper position of the cervical spine was placed in neutral that was obtained by adjusting the height and back of the adjustable chair. For all testing procedures, the cervical spine was placed in neutral or 0°. For lateral flexion testing, the headpiece was rotated by 90°. The pads on the headpiece were positioned just above the upper part of the eyebrows for flexion, just superior of the occiput for extension, and just superior of the earlobes for lateral flexion testing. Shoulder and waist straps were adjusted to isolate the cervical spine from the body. Players were instructed to cross their arms so hands were on opposite shoulders and feet crossed.

Maximal voluntary isometric force was measured in 4 directions in order of: flexion, extension, right and left lateral flexions. Prior to the testing protocol, players were given a two-minute warmup consisting of static stretches in all 4 directions as well as active range of motion of flexion to extension, left lateral to right lateral flexion, and head circles to include all planes of motion. Two familiarization trials at submaximal effort were performed for each direction prior to recording. Researcher’s guidance and any corrections were given during the familiarization trials. During the actual testing period, each player was asked to perform 3 consecutive trials of maximal voluntary isometric force in each direction for 3 seconds (Pearson et al., 2009). They were instructed to generate as much force as rapidly as possible and to hold that force over the duration of the trials. 1-minute rest was given between each trial and 2-minutes between each direction. Verbal directions were given as “Push as hard and fast as you can into the pads…[in the appropriate direction]” during each trial. Peak force was normalized by body mass (N/kg) and then averaged across the three trials. Time to peak (sec) was calculated by identifying the greatest peak output utilizing 50 millisecond increments supported by the MCU software (Morais de Oliveira, Greco, Molina, & Denadai, 2012).

Research Aim 2: To address the second aim, we used the peak force values and respected time to peak for each direction to predict impact severity of peak linear acceleration, peak rotational velocity, and number of head impacts.

Anthropometric Characteristics & Helmet Dimensions

Anthropometric characteristics were obtained before the testing protocol. We gathered players' height (cm), mass (kg), true leg length (cm), neck circumference (cm), neck length (cm), and head circumference (cm). We captured height (cm) by using a stadiometer (SECA) in a standing position with shoes removed. Players will be informed to be in good posture and to maintain good posture by saying "stand up tall". Mass (kg) was captured via a Med-Weigh digital scale MS-330. True leg length (cm) was measured using a tape measure from the anterior superior iliac spine to the medial malleolus from a supine position (Sabharwal & Kumar, 2008). With good posture, the superior border of the tape measure was placed on the inferior pole of the laryngeal prominence (Adam's apple) and applied perpendicular to the long axis of the neck to measure neck circumference (cm) (Ben-Noun, Sohar, & Laor, 2001; Preis et al., 2010). Neck length (cm) was measured from the base of the occiput to cervical spine 7. Head circumference and helmet circumference (cm) was measured over the most prominent part of the occiput and just above the supraorbital ridges (Harris, 2015). Head mass (kg) was captured by taking 8% of the total body mass.

Helmet dimensions included total helmet mass (kg) using the Med-Weigh digital scale, helmet depth (cm) and helmet circumference (cm) with a tape measure, face mask mass (kg) using the digital scale. Length of the face mask in the sagittal plane from the sella turcica (outer ear: fossa of antihelix crura) (cm) were measured when the player had their helmet on. Helmet depth was measured from the most distal aspect of the helmet shell to padding. We removed the face mask from the helmet to obtain its mass (kg) as well as the helmet shell mass (kg).

We normalized neck length, neck circumference, head circumference, head mass, total helmet mass, face mask mass, helmet depth, helmet circumference, and length of face mask by dividing these outcomes by body mass.

Research Aim 3: To address the third aim, all anthropometric characteristics and helmet dimensions were used to predict impact severity of peak linear acceleration, peak rotational velocity, and number of head impacts.

Head Impact Biomechanics

Head impact biomechanics were captured using the Triax Smart Impact Monitor (SIM-G) head impact sensors (Triax Technologies Inc., Norwalk, CT). Previous studies have used this same device with various sports to examine head impact biomechanics (Lamond, Caccese, Buckley, Glutting, & Kaminski, 2018; Lynall et al., 2019). The head impact sensors capture location, frequency, and severity using a 3-axis gyroscope, high-g 3-axis accelerometer, and low-g 3-axis accelerometer. In addition, data from the sensors are time-stamped, encoded, then transmitted in real-time to a sideline aggregator (SKYi) which stores all accelerometer data. An iPad or smartphone will also be used to display real time head impacts. Data will be transmitted to the SKYi and base monitoring station as long as athletes are within a range approximately 150 yards. If athletes are not within range, the sensors are able to store up to 140 impacts until the athlete returns within range. Data will be extracted from the SKYi device. For a single head impact, each axis accelerometer collects data at 1kHz for 28ms and can measure g-force levels between 3-150*g*. Frequency was another primary outcome of this study therefore data went through a multi-step cleaning process. We will exclude breaks or pauses, when players removed their head bands, and player exposure to remove false head impacts. Similar to prior studies, we calculated the threshold of 51 session head impacts by adding the 3rd quartile value (33 head impacts) to the median (18 head impacts) (Lynall et al., 2019). Any player who experienced more impacts than the high frequency threshold was identified and removed from the dataset.

Statistical Analyses

We will perform separate linear mixed-effects models with a random intercept to predict head impact severity with $\alpha = 0.05$. We will also include lineman and non-lineman as a predictor variable in the model to control for player positions when there is a significant difference in predictor variables.

Table 3.2. Summary and Data Analysis Table

Aims	Independent Variables		Dependent Variables	Data Analysis
To determine the influence of head protection, hand, player strategy reaction time and eye movement time on head impact biomechanics.	• Player Strategy • Head Turn Reaction • Hand Reaction Time • Eye Movement Time	Player Position Controlled for Aim 2 Analyses (Cervical Strength Protocol): • Lineman • Non-Lineman	Head Impact Biomechanics: • Linear Acceleration* • Rotational Velocity* *log transformed	Univariate Linear Mixed-Effects with a Random-Intercept for head impact severity outcomes
To determine the influence of cervical muscle strength on head impact biomechanics.	• Peak force by direction (Flexion, Extension, Left lateral flexion, Right later flexion) • Time to Peak by direction (Flexion, Extension, Left lateral flexion, Right later flexion)		• Number of Head impacts	Univariate linear regression models for predicting number of head impacts
To determine the influence of anthropometric characteristics and helmet dimensions on head impact biomechanics.	• Height • Mass • True Leg Length • Neck Circumference • Neck Length • Head circumference • Helmet Depth			

	• Total Helmet Mass • Face Mask Mass • Length of face mask from COM in X direction			

Supplemental Data Capture Separate from Dissertation

In addition to the outlined methodology, but separate from the dissertation document, we gathered players’ tackle football experience, overall sport participation experience, sports played during the calendar year other than football, helmet type, and listed size (Appendix A).

CHAPTER 4

MANUSCRIPT I

THE EFFECT OF FUNCTIONAL HEAD PROTECION TIME ON HEAD IMPACT BIOMECHANICS

Context: Interventions during youth tackle football are needed to reduce repetitive head impacts and promote player safety. However, there has been limited research in understanding if preseason measures of head protection reaction time are associated with head impact severity.

Objective: To determine whether functional head protection reaction time under single- and dual-task (motor and cognitive task concurrently) predicts on-field head impact severity and frequency.

Methods and Materials: Twenty-seven youth tackle football players (males=100%, age=12.2±0.4yrs, height=161.5±9.7m, mass=55.5±15.2kg) participated in this study. During pre-season, participants reacted to a light stimulus using the following strategies under single- and dual-task: (1) player strategy (self-selected by participant), (2) eye movement (move eyes only to stimulus), (3) head turn (move entire head towards stimulus), and (4) hand (raise hands to block stimulus). Participants wore a Triax SIM-G accelerometer throughout the season that captured number of impacts, peak linear acceleration (PLA; g), and peak rotational velocity (PRV; rad/s). Univariate linear regression models were used to determine if head protection reaction time and eye movement time were associated with number of head impacts. Separate, univariate, linear regression models with random-intercepts were used to determine the effect of log-transformed head impact severities. The α value was set to *P*=0.05 a priori.

Results: None of the conditions associated with number of head impacts ($p>0.05$). Eye movement time in single- ($R^2=0.103$, $P=0.012$) and dual-task ($R^2=0.126$, $P=0.021$) predicted PRV. With every 1 second increase in single and dual task eye movement time, there was an increase in PRV of 4.4 rad/s ($P=0.013$) and 5.2 rad/s ($P=0.021$), respectively.

Conclusions: Our study identified eye movement as a potential target for future interventions aimed at reducing head impact severity.

Introduction

Youth football players sustain head impacts of similar frequency and severity relative to their older counterparts (Campolettano et al., 2017; Daniel RW et al., 2012; Kontos et al., 2013; Munce et al., 2015). However, modifiable risk factors have not been as extensively studied in this group. Understanding risk factors essential for developing and implementing preventative strategies and equipment design (Rowson, Tyson, Rowson, & Duma, 2018). While the research points to the importance of identifying ways to decrease repetitive blows to the head, most head impact related research has focused on high school, collegiate, or professional athletes (Broglio et al., 2013; Broglio et al., 2010; Broglio et al., 2012; Mihalik et al., 2007; Schmidt et al., 2016). Previous studies that utilized youth football have used interventions such as limiting high impact frequency drills to reduce the number of head impacts and potential higher levels of severity (Campolettano et al., 2017; Campolettano et al., 2016).

Increased attention has been directed toward possible rule changes and enforcing proper tackling techniques in order to reduce repetitive head impacts and promote player safety in youth tackle football (Broglio et al., 2013). An athlete’s ability to prepare for and respond to an oncoming collision may depend on their ability to quickly protect their head, visually identify the collision with quick eye movements, and utilize their mass and body to absorb force (Harpham et al., 2014). Previous research has found a player’s anticipation behavior (unanticipated vs. anticipated) may influence the severity of head impacts. A player who did not anticipate or see the on-coming collision experienced a higher impact severity (Mihalik, Blackburn, et al., 2010). Understanding these features serves as a necessary first step towards determining whether outcomes of reaction time and eye movement time warrant intervention to reduce the head impact burden in youth football.

Recently, there has been a growing body of research analyzing the effect of functional reaction time on injury risk. Traditionally, reaction time has been assessed with computerized neurocognitive testing (Eckner, Kutcher, & Richardson, 2011), but full body motor responses to a stimulus have emerged (Lynall et al., 2018). Reaction time is critical to sport performance and protective maneuvers. Football players must respond quickly to catching a ball or covering an opponent to complete the task or avoid injury (Eckner, Lipps, et al., 2011). Athletes with low visual and sensory reaction time performance sustained a higher number of severe head impacts compared to those with high visual and sensory performance in football (Harpham et al., 2014). Typically, when the head follows the eyes, a slower head protection reaction (i.e. time it takes to react to a stimulus) time may increase the injury risk and may result in greater head impact exposure because athletes have inadequate time to respond to oncoming collisions (Eckner et al., 2014). Just as individuals with slower functional reaction time may have slower motor responses to protect themselves, individuals with slower head protection reaction time may be slower to protect their head making them more prone to repetitive head impacts.

However, there is limited research investigating head protection reaction time (Eckner, Lipps, et al., 2011). Researchers have found collision characteristics, such as anticipating impact and facing the oncoming collision, may reduce the severity of head impacts (Caswell et al., 2012; Mihalik, Greenwald, et al., 2010). Unanticipated head impacts in youth ice hockey resulted in increased severity compared to anticipated collisions (Mihalik, Blackburn, et al., 2010). Therefore, anticipating or looking in the direction of collisions may help reduce head impact severity as they are able to brace for impact sooner. Football players that quickly identify oncoming collisions and respond by protecting their head more quickly may be better able to sustain the blow of impact or avoid it entirely. The purpose of this study is to determine whether a head protection protocol consisting of a player’s strategy, head turn reaction, hand reaction, and eye movement time to predict head impact severity and frequency. We hypothesized that youth tackle football players with slower head protection reaction time and eye movement time would sustain more severe head impacts.

Methods

Study Participants

A youth tackle football team was recruited from a single recreation department in the rural southern United States. Twenty-seven youth tackle football players (males=100%, age=12.2±0.4yrs, height=161.5±9.7cm, mass=55.5±15.2kg) participated in this study during the 2021 fall football season. The study protocol was approved by our Institutional Review Board prior to subject recruitment. We obtained child assent with at least one parent or guardian providing signed informed consent.

Head Protection Reaction & Eye Movement Time

Players completed the testing protocol prior to the start of the season before the first game. All aspects of the protocol were completed in a research laboratory setting. Prior to the testing session, players were asked to bring their helmet and to wear athletic clothing with no reflective material. Any reflective pieces on the helmet were covered with black tape. Players were instructed to react to a left or right light stimulus placed 4.5 meters away and 45° from the player according to the following conditions: player strategy, head turn, hand reaction time, and eye movement time. Each player completed the player strategy task first with the order of the remaining conditions being randomized. All conditions were completed under single-task (i.e. focusing on completing the task as quickly as possible) and dual-task (i.e. spelling words backwards as quickly and accurately as possible).

Before each trial, players were instructed to get in an athletic stance with knees slightly flexed by the words "get set" from the research member. During dual task trials, players began spelling words backwards prior to any movement. Trials were repeated if participants did not initiate movement when cued or did not perform the correct movement (e.g. moved their head when only instructed to move their eyes). Players were required to complete a familiarization trial for each condition. The single- and dual-tasks were alternated within each condition. Players completed a total of 4 single- and 4 dual-tasks for each condition. The light stimulus occurred randomly between 2-5 seconds. Left and right light stimulus were also implemented at random.

All reaction and eye movement conditions occurred in a 3D motion capture space compromising of 8 cameras (MIQUIS, Qualisys Systems, Goteborg, Sweden) recording at 240Hz. An additional camera with a sampling frequency of 480p/400Hz was placed in front of the player that contained a view with a mirror in the background to reflect the light box that was sampling at 1200Hz. Players were fitted with 14-mm retroreflective markers to track reaction time. To standardize marker placement, the middle of the helmet markers was placed 4cm superior from the most distal part on the back of the helmet. The markers created a square shape where each marker was spaced out 4cm from each other (Figure 4.1b). Single markers were placed over the left and right ulnar styloid processes upon palpation by the primary research investigator. To measure head movement, markers were placed on the back of their helmet and on the styloid process of each ulna for the hand reaction time condition. A research team member visually looked at the participant and on camera to ensure they were not performing trunk movements. Trials were repeated if the participant moved their trunk or moved their head during the eye movement only condition.

Conditions:

Players were instructed to face straight ahead in an athletic stance then quickly identify the light stimulus and perform the designated technique to protect their head as if the light represented an oncoming opponent. Players started with hands on their thighs being in an athletic stance after being told to "get set" by the research team. For the dual-task, players were instructed to "get set" then given a word to spell backwards and given new words subsequently. A visual light stimulus (green) was triggered randomly either left or right by the research member within 2 to 5 seconds after the "get set" instruction to signal movement. All sessions began with the player's strategy condition followed by the three remaining conditions in counterbalanced order. Additional details provided for each condition are as follows:

Player's Strategy: Players were instructed to react to the light stimulus as quickly as possible as if they were to block an on-coming opponent in a practice or game. Players could react however they wanted but were told to react as quickly as possible. This commonly consisted of players turning their head to face the light stimulus and both hands being raised to their chest.

Head Turn: Players were instructed to face straight ahead. They were instructed to turn their head, only, to face the light stimulus as soon as it was cued as if they were facing an oncoming opponent (Figure 4.1a).

Hand: Players were instructed to face straight ahead and to raise their hand as if performing a stiff-arm movement towards the corresponding light that was cued as if they were blocking an opponent and protecting their head (Figure 4.2). They were instructed to keep looking straight ahead.

Eye: Players were instructed to look straight ahead and to only move the eyes to look at the light and stay fixated on the light until told to "relax" (Figure 4.3).

Figure 4.1a. Participant reacting to light with head turn reaction time condition.

Figure 4.1b. Helmet markers placed on back of helmet.

Figure 4.2. Participant reacting to light with hand reaction time condition.

Figure 4.3. Participant reacting to light with eye movement time condition.

Head Impact Biomechanics

Head impact biomechanics were captured using the Triax Smart Impact Monitor (SIM-G) head impact sensors (Triax Technologies Inc., Norwalk, CT). The sensor is inserted into either a headband or skullcap then placed around the base of the skull. Each sensor was uniquely assigned to the player. A member of the research team distributed sensors to participants before the start of an event when they were instructed to wear all equipment and collected the sensors afterwards. We recorded attendance of players being preset and wearing their headband. SIM-G linear acceleration (PLA;g) data is first measured at the sensor and is then transformed to the center of gravity of the head using the measured rotational velocity (rads/s) (Yoganandan, Pintar, Zhang, & Baisden, 2009). A 14-g threshold was established such that data 10ms before the impact and 52m after impact were recorded when meeting or exceeding 14-g (Lynall et al., 2019). We recorded start and end times of every event. We also took diligent notes and recorded start and end times if a sensor was removed temporarily and placed back on and if there were any breaks in the events (i.e. waterbreaks, halftime).

No on-field head impact biomechanics system is perfect. Previous studies suggest the SIM-G has inconsistent validation data (Cummiskey et al., 2017; Karton et al., 2016; Oeur et al., 2016; Tyson et al., 2018). Previous investigations have reported optimal impact detection with 100% of non-helmeted impacts detected and 80% of helmeted impacts detected in one study (Oeur et al., 2016), and 99% of helmeted impacts detected in another study (Cummiskey et al., 2017). There was also no difference between head form and SIM-G linear acceleration at low and medium energy impacts in one study (Karton et al., 2016) and approximately 14.25% error on average found in another study (Cummiskey et al., 2017). A separate investigation reported less than optimal acceleration validity where there was underpredicted linear and rotational acceleration in helmeted (linear acceleration slope=0.31; rotational velocity=0.74) and non-helmeted conditions (linear acceleration slope=0.37; rotational velocity=0.94)(Tyson et al., 2018). We used the SIM-G because it can record head impact data from helmeted participants regardless of helmet manufacturers and brand.

Data Analysis

All marker data were exported from the motion capture software (Qualisys Track Manager; Qualisys System) and processed using Visual 3D (Visual 3D; C-Motion Inc, Germantown, MD). Reaction time data was determined by which movement occurred first in either the X (antero-posterior) and Z (longitudinal) planes. We determined the player's strategy by identifying which segment they first moved (head, eyes, or hands). For the player strategy we averaged time across trials per condition. We then identified which segment moved first. Head and hand movement from the markers were determined as the time between light stimulus and head or hand movement >1cm from the resting position (averaged over the 0.5s prior to stimulus). We defined eye movement by totaling the number of frames between light stimulus and when the pupil reached the end range of the saccadic movement divided by the seconds passed (Figure 4.2). We used data from more than one rater on the research team who also viewed the videos to report inter-class correlation results (single-task: ICC (2,*k*)=0.948; 95%CI=0.628-0.994; dual-task: ICC (2, *k*)=0.882; 95%CI=0.602-0.987).

Head impact data went through a multi-step process before being analyzed where session was defined as either a practice or game. First, head impact went through an interval validity check within the Triax software which identifies and labels false impacts (Lynall et al., 2019). Head impacts were removed if they occurred outside of each defined practice or game session time. We also removed head impacts that occurred in any pauses or breaks in the session such as water breaks, quarter breaks, injury start/end times, and time outs. To further ensure only real-time head impacts were included in the dataset, we excluded individual player sessions where a large number of head impacts occurred, most likely where a player removed and placed back on the headband often or was not worn appropriately. Similar to prior studies, we calculated the threshold of 51 session head impacts by adding the 3rd quartile value (33 head impacts) to the median (18 head impacts) (Lynall et al., 2019). Our dependent variables consisted peak linear acceleration (PLA;g), peak rotational velocity (PRV;rad/s), and number of head impacts.

Statistical Analysis

Descriptive statistics were assessed for player predictors and head impact outcomes (Table 4.1a&b). All reaction time and eye movement conditions were normally distributed among single and dual task performance.

Head impact biomechanics data were log transformed to approximate for a near-normal distribution then back transformed for interpretation. Separate, univariate random intercepts linear regression models were used for each condition to predict head impact severity. We compared independent variables between player position (lineman vs non-linemen) using an independent t-test to determine if controlling for player position was necessary. No significant differences were observed between player positions, so we did not include player position in the models (*P* range = 0.127-0.438). All regression model assumptions were assessed for violations before statistical interpretation. Separate, univariate linear regression models were used to determine if player strategy, head turn, hand reaction time and eye movement time were associated with number of head impacts. All statistical analyses were performed using R Studio (version 4.1.2; PBC) with $\alpha = 0.05$.

Results

There was a significant difference between single task and dual task performance among all conditions (Table 4.1a). Correlation analysis of single- and dual-task performance can be found on Appendix B.

Table 4.1a. Descriptive Statistics of Predictors

Head Protection Protocol	**Single Task Mean±SD (Range)**	**Dual Task Mean±SD (Range)**	**P-value**
Player Strategy (s)	0.332±0.049 (0.251-0.436)	0.472±0.066 (0.364-0.632)	0.008*
Eyes-Head-Arms	N=16, 60%	N=11, 41%	
Eyes-Arms-Head	N=5, 18%	N=6, 22%	
Head-Eyes-Arms	N=2, 7%	N=3, 11%	
Head-Arms-Eyes	N=1, 4%	N=3, 11%	
Arms-Head-Eyes	N=2, 7%	N=2, 7%	
Arms-Eyes-Head	N=1, 4%	N=2, 7%	
Eye Movement Time (s)	0.351±0.055 (0.257-0.461)	0.491±0.073 (0.385-0.634)	0.019*
Head Turn Reaction Time (s)	0.480±0.036 (0.410-0.572)	0.593±0.076 (0.419-0.743)	0.014*
Hand Reaction Time (s)	0.472±0.050 (0.392-0.601)	0.584±0.060 (0.456-0.693)	0.022*

***Single and dual task differed for all tasks**

Table 4.1b. Descriptive Statistics of Head Impact Biomechanics

	Linear Acceleration (g) Mean±SD (Range)	**Rotational Velocity (rad/s) Mean±SD (Range)**	**Season No. Impacts Per Player Mean±SD (Range)**
All impacts (n=2,800)	37.6±13.5 (16.01 – 113.22)	19.1±10.0 (2.70 – 50.24)	107.7±19.3 (71 – 150)

Table 4.2. Beta coefficients (back transformed) and corresponding P-values for Head Protection Condition and Head Impact Biomechanics.

Condition	Model Summary	Single-Task (PLA; g)		Model Summary	Dual-Task (PLA; g)		Model Summary	Single-Task (PRV; rads/s)		Model Summary	Dual-Task (PRV; rads/s)	
		Beta	P		Beta	P		Beta	P		Beta	P
Player Strategy	R^2=0.177 P=.679	-0.01	.679	R^2=0.165 P=.942	-0.001	.942	R^2=0.073 P=.603	-1.20	.603	R^2=0.095 P=.057	-0.63	.057
Eye Movement Time	R^2=0.045 P=.089	-1.8	.089	R^2=0.189 P=.072	-4.5	.072	R^2=0.103 P=.013	4.4	.013*	R^2=0.126 P=.021	5.2	.021*
Head Turn Reaction Time	R^2=0.421 P=.581	-2.1	.581	R^2=0.048 P=.842	0.26	.842	R^2=0.116 P=.125	.89	.125	R^2=0.322 P=.072	-0.17	.072
Hand Reaction Time	R^2=0.199 P=.133	9.2	.133	R^2=0.022 P=.080	-1.34	.080	R^2=0.316 P=.068	12.03	.068	R^2=0.243 P=.085	-2.92	.085

*significant predictor of head impact severity

Head Impact Frequency

We analyzed a total of 2,800 head impacts across 611 athlete exposures. Head protection reaction time and eye movement time were not associated with number of head impacts (Table 4.2).

Head Impact Severity

The single-task eye movement time predicted peak rotational velocity (R^2=0.103, *P*=.013), such that every additional 1 second was associated with an increase in rotational velocity of 4.4 rad/s (*P*=.013). Dual-task eye movement predicted peak rotational velocity (R^2=0.126, *P* =.021), such that every additional 1 second was associated with an increase of 5.2 rad/s (*P*=.021). No other models were significant (*P*>0.05).

Discussion

All conditions in the single-task were significantly faster than in the dual-task. Youth tackle football players with slower eye movement times preseason went on to sustain head impacts of higher rotational velocities throughout the season. Hand and head turn reaction times did not explain variance in head impact severities.

Previous research shows that individuals with a faster visual and sensory performance are able to respond quicker to their environment (Zimmerman, Lust, & Bullimore, 2011). In our study, we found that eye movement predicted head impact severity of peak rotational velocity. Players who are able to fixate on their target quicker gives themselves more time to extract the appropriate information and prepare for impact (Savelsbergh, Van der Kamp, Williams, & Ward, 2005), which may allow them to adopt other strategies to lessen head impact severity.

A quicker saccade may allow for players to see and identify the characteristics of oncoming collisions more quickly. Whereas a slower saccade may be associated with greater head rotational velocity following impact due to the inability to react quick and prepare for an oncoming collision. Sport participation in football requires dynamic motor control to avoid injury with other players in the athletic environment (Eckner, Lipps, et al., 2011). Athletes who can identify their target or opponent quickly may have decreased odds of sustaining severe head impacts because they can prepare for oncoming or anticipated collision by protecting their head (Kamp, 2011). Functional or sport specific reaction time is crucial and necessary for protective maneuvers (Eckner, Lipps, et al., 2011). College football athletes with low visual and sensory reaction time performance sustained a higher number of severe head impacts compared to those with high visual and sensory performance (Harpham et al., 2014). Our results are similar in that we reveal that slower eye movement time was associated with an increase of 4.4 and 5.2 rad/s in single- and dual-task conditions, respectively. The conditions we used for this study are closely related to the Nike SPARQ sensory station such as Target Capture and Eye-hand coordination (Harpham et al., 2014). Results from their study revealed lower Target Capture and Eye-hand coordination in outcomes being associated with higher rotational acceleration (Harpham et al., 2014). The dual-task performance may be more closely related to football mechanics as players are anticipating where to go when the ball is snapped based on the play-call. However, not being able to identify a stimulus or opponent quicker may lead to a more severe head impact.

Eye movements are the foundation for how humans gather information about the environment (Stuart et al., 2020). In return, this information is used to navigate a safe task or performance (Stuart et al., 2020). Eye tracking is vital in the sport environment especially in contact sports where there is a constant need for performing safety maneuvers. Future studies should evaluate the effectiveness of eye movement time on head impact biomechanics as an intervention.

In contrast, the head, hand, and player strategy reaction times did not predict head impact biomechanical outcomes. Eye movements may occur first as deliberate or intentional reaching movements with the extremities comes second. Saccadic movements are present since infancy and continue to improve with higher speeds of tracking through childhood into adolescence (Luna, Velanova, & Geier, 2008). Additionally, the neuronal transmission from the eyes opposed to head or hand movements is faster (Luna et al., 2008). This is supported by our findings where the eyes were able to move faster and react to the light stimulus, in most trials, as opposed to the head or hand.

Previous head impact related research has found collision characteristics such as anticipation or not anticipated to have mixed findings. Collegiate and high school football studies found few impacts as unanticipated (Mihalik et al., 2008; Schmidt et al., 2016). Whereas in youth ice hockey, unanticipated head impacts resulted in higher magnitudes compared to anticipated impacts (Mihalik, Blackburn, et al., 2010). Of note, these studies did not examine eye movement as a method to define anticipation behavior which may explain null findings. Many studies use body position or head direction to define anticipation vs. unanticipated (Mihalik, Blackburn, et al., 2010; Schmidt et al., 2016). Video based analyses of player anticipation do not account for this important element of eye movement. The repetition of plays during practices and games may contribute to player’s getting more familiar with eye movement and tracking as a method to anticipate a collision when the ball is snapped.

Limitations

Our study has limitations to be addressed. First, the Triax SIM-G has mixed evidence about its accuracy (Cummiskey et al., 2017; Karton et al., 2016; Oeur et al., 2016; Tyson et al., 2018). However, we took diligent notes to only include real-time head impacts. Video verification of head impacts may not differ from using a multi-step cleaning process employed in our study (Le et al., in review). Second, these data were collected from a small sample from a recreation department. Our findings may not be generalizable to other geographical areas, different age groups, or female players. Third, eye movement time was calculated based on video recordings and we used light cues to mimic on-coming collisions. Additionally, the additional camera used for determining eye movement time had a higher sampling rate compared to the motion capture cameras (400hz vs 240hz). Future research using advanced technology similar to motion capture analysis is warranted as well as during live-play of events.

Conclusion

Eye movement time was significant predictor during a single and dual task for PRV values. Our study suggests slower eye movements are associated with greater impact severity in rotational velocity. We did not find the remaining conditions of head and hand reaction to be significant predictors. This is possible due to players relying on visual environment prior to initiating any head or hand movement. More research is needed to explore the relationship between eye movements with head and hand reaction times during on-field events.

CHAPTER 5

MANUSCRIPT II

THE ROLE OF CERVICAL MUSCLE STRENGTH AND ACTIVATION TIME ON HEAD IMPACTS IN YOUTH FOOTBALL ATHLETES

Context: Greater cervical strength has been posed as a way to mitigate head impact severity in tackle football. However, it is unknown if younger athletes may experience greater head impact severity due to premature cervical musculature.

Objective: To determine (1) if peak isometric cervical strength in flexion, extension, left lateral flexion, and right lateral flexion predict head impact severity and (2) if time to peak strength predicts head impact severity and frequency.

Methods and Materials: Twenty-seven youth tackle football players (males=100%, age=12.2±0.4yrs, height=161.5±9.7m, mass=55.5±15.2kg) participated in this study. Participants completed a preseason maximum cervical isometric strength protocol in order of flexion, extension, left lateral flexion, and right lateral flexion on a Multi-cervical Unit. Peak strength was normalized to body mass (N/kg) and time to peak strength was calculated (s). Participants wore Triax SIM-G accelerometers that captured head impact severity outcomes (linear acceleration [PLA; g]; rotational velocity [PRV; rad/s]). Separate, univariate, linear models for strength and time to peak were used to determine impact frequency. A multivariate, non-normal regression model with random intercepts for peak force were used to determine the relationship on log-transformed PLA and PRV while controlling for player position (lineman, non-lineman).

Univariate linear regression models were used to determine the relationship of time to peak on biomechanics without controlling for player position (α =0.05).

Results: Peak left lateral flexion strength was associated with a lesser rotational velocity (R^2=0.262, P=0.040), such that every additional 1N/kg corresponded with a lesser severity of 6.49rad/s (P=0.039). Right lateral flexion strength was associated with a lesser rotational velocity (R^2=0.151, P=0.035) such that every additional 1N/kg increase was associated with a lesser severity of 2.08rad/s (P=0.026) Time to peak in left lateral flexion was associated with greater linear acceleration (R^2=0.072, P=0.041), such that every additional 1s increase corresponded with greater severity 3.10g (P=.036). Time to peak flexion was associated with a greater rotational velocity (R^2=0.128 P=0.036), such that every 1s increase corresponded with a greater severity of 4.41rad/s (P=0.006). Time to peak extension was associated with greater rotational velocity (R^2=0.224, P=0.010) such that every 1s increase corresponded with a greater severity of 3.32 rad/s (P=0.008). Models for cervical strength and time to peak were not associated with number of head impacts.

Conclusions: Youth tackle football players with stronger cervical strength had less severe head impacts and slower time to reach peak strength was associated greater severe head impacts. Future research should determine whether cervical strength interventions reduce head impact severity.

Introduction

Cervical muscle characteristics, such as muscle strength, have been proposed as a preventative strategy to reduce the risk of concussion and severity of repetitive head impacts (Cantu, 1996; Mihalik et al., 2011; Tierney et al., 2008; Viano et al., 2007). The osteoligaments and surrounding muscles support the weight of the head and external loads. The ligaments of the cervical spine only contribute to 20% of dynamic stability (Panjabi et al., 1998), with the cervical musculature contributing the remaining 80% (Panjabi et al., 1998). Contracting the cervical muscles is an attempt to create more rigidity to counter this force and reduce head acceleration (Viano & Pellman, 2005). When the cervical muscles are relaxed, or when an athlete sustains a collision that is unanticipated, they are not aware enough to voluntarily contract cervical muscles to counteract the external force - resulting in rapid head acceleration (Viano & Pellman, 2005). The effect of cervical strength on head impacts warrants further research.

There is even more concern for adolescent and youth athletes because their cervical muscles are not fully developed and may not reduce head acceleration when sustaining an impact (Mihalik et al., 2011). Previous research has shown that youth football players sustain head impacts of similar severity to high school and college players (Campolettano et al., 2017; Daniel RW et al., 2012; Kontos et al., 2013). This may be linked to youth athletes having insufficient cervical musculature causing rapid acceleration of the head. Youth athletes may lack the ability to counter the external force and may sustain head acceleration causing similar severity to their older counterparts. Cervical strength varies according to age (Hildenbrand & Vasavada, 2013). College athletes have stronger cervical muscles compared to high school athletes (Hildenbrand & Vasavada, 2013). Youth athletes may continue this same trend by age and have even weaker cervical strength.

There is mixed literature to suggest cervical strength influences head impact severity. One reason may be due to existing literature measuring cervical strength in a variety of ways (i.e. hand-held dynamometers, patient positioning, etc). However, none have used the gold standard, the Multi-Cervical Unit (MCU, BTE Technologies, Inc., Hanover, MD) (Pearson et al., 2009). Further research is needed to determine if cervical strength is associated with head impact biomechanics in youth football. The purpose of this study was to determine how cervical strength outcomes such a peak force and time to peak force predicts head impact severity. We hypothesized there would be an inverse relationship between cervical peak force and head impact biomechanics. We hypothesized there would be a positive relationship between time to peak and head impact biomechanics. We hypothesized there would be an inverse relationship between cervical peak force on impact frequency. Lastly, we hypothesized there would be a positive relationship between time to peak force on impact frequency.

Methods

Study Participants

A youth tackle football team was recruited from a single recreation department in the rural southern United States. Twenty-seven youth tackle football players (males=100%, age=12.2±0.4yrs, height=161.5±9.7m, mass=55.5±15.2kg) participated in this study during the 2021 fall football season. The study protocol was approved by our Institutional Review Board prior to subject recruitment. We obtained child assent with at least one parent or guardian providing signed informed consent.

Cervical Strength Protocol

Players were seated in the multi-cervical unit that captured force output at 20Hz (MCU, BTE Technologies, Inc, Hanover, MD). The chair and proper position of the cervical spine was placed in neutral that was obtained by adjusting the height and back of the adjustable chair. For all testing procedures, we measured for the cervical spine to be placed placed in neutral or 0°. For flexion and extension, the headpiece was placed at 0° per software directions. For lateral flexion testing, the headpiece was rotated by 90°. The pads on the headpiece were positioned just above the upper part of the eyebrows for flexion, just superior of the occiput for extension, and just superior of the earlobes for lateral flexion testing. Shoulder and waist straps were adjusted to isolate the cervical spine from the body. Players were instructed to cross their arms so hands were on opposite shoulders and feet crossed.

Maximal voluntary isometric force was measured in 4 directions in the order of: flexion, extension, right and left lateral flexions. Prior to the testing protocol, players were given a two-minute warmup consisting of static stretches in all 4 directions as well as active range of motion of flexion to extension, left lateral to right lateral flexion, and head circles to include all planes of motion. Two familiarization trials at submaximal effort were performed for each direction prior to recording. Researcher's guidance and any corrections were given during the familiarization trials. During the actual testing period, each player was asked to perform 3 consecutive trials of maximal voluntary isometric force in each direction for 3 seconds (Pearson et al., 2009). They were instructed to generate as much force as fast as possible and to hold that force over the duration of the trials. One-minute rest was given between each trial and 2-minutes between each direction. Verbal directions were given as "Push as hard and fast as you can into the pads…[in the appropriate direction]" during each trial. Peak force was normalized by body mass (N/kg) and then averaged across the three trials. We found the absolute peak force value and took the time to that measurement to identify time to peak (sec).

Head Impact Biomechanics

Head impact biomechanics were captured using the Triax Smart Impact Monitor (SIM-G) head impact sensors (Triax Technologies Inc., Norwalk, CT). The sensor is inserted into either a headband or skullcap then placed around the base of the skull. Each sensor was uniquely assigned to the player. A member of the research team distributed sensors to participants before the start of an event and collected the sensors afterwards. We recorded attendance of players being present and wearing their headband. SIM-G linear acceleration (PLA;g) data is first measured at the sensor and is then transformed to the center of gravity of the head using the measured rotational velocity (rads/s)(Yoganandan et al., 2009). A 14-g threshold was established such that data 10ms before the impact and 52m after impact were recorded when meeting or exceeding 14-g (Lynall et al., 2019). We recorded start and end times of every event. We also took diligent notes and recorded start and end times if a sensor was removed temporarily and placed back on and if there were any breaks in the events (i.e. waterbreaks, halftime).

No on-field head impact biomechanics system is perfect. Previous studies suggest the SIM-G has inconsistent validation data (Cummiskey et al., 2017; Karton et al., 2016; Oeur et al., 2016; Tyson et al., 2018). Previous investigations have reported optimal impact detection with 100% of non-helmeted impacts detected and 80% of helmeted impacts detected in one study (Oeur et al., 2016), and 99% of helmeted impacts detected in another study (Cummiskey et al., 2017). There was also no difference between head form and SIM-G linear acceleration at low and medium energy impacts in one study (Karton et al., 2016) and approximately 14.25% error on average found in another study (Cummiskey et al., 2017). A separate investigation reported less than optimal acceleration validity where there was underpredicted linear and rotational acceleration in helmeted (linear acceleration slope=0.31; rotational velocity=0.74) and non-helmeted conditions (linear acceleration slope=0.37; rotational velocity=0.94)(Tyson et al., 2018). We used the SIM-G because it can record head impact data from helmeted participants regardless of helmet manufacturers and brand.

Data Analysis

Head impact data went through a multi-step process before being analyzed where session was defined as either a practice or game. First, head impact went through an interval validity check within the Triax software which identifies and labels false impacts (Lynall et al., 2019). Head impacts were removed if they occurred outside of each defined practice or game session time. We also removed head impacts that occurred in any pauses or breaks in the session such as water breaks, quarter breaks, injury start/end times, and time outs. To further ensure only real-time head impacts were included in the dataset, we excluded individual player sessions where a large number of head impacts occurred, most likely where a player removed and placed back on the headband often or was not worn appropriately. Similar to prior studies, we calculated the threshold of 51 session head impacts by adding the 3rd quartile value (33 head impacts) to the median (18 head impacts) (Lynall et al., 2019). Our dependent variables consisted peak linear acceleration (PLA;g), peak rotational velocity (PRV;rad/s), and number of head impacts.

Statistical Analysis

Head impact biomechanics data were log transformed to approximate for a near-normal distribution then back transformed for interpretation. Cervical strength data were not normally distributed. Therefore, we used separate non-linear regression models with random intercepts for peak force at each direction. Multivariate non-linear regression models controlled for player position since we found significant differences between lineman and non-lineman on cervical strength measures. We used the Rfit package for rank-based estimation and inference for linear models. The Rfit package provides the beta coefficients in the same way as a linear regression using the same units of measure. We also used multivariate non-linear regression models to determine if cervical strength was associated with number of head impacts. Time to peak did not differ between player positions, so we did not control for player position in time to peak analyses. Time to peak values were normally distributed, so we used separate univariate linear regression models to determine if time to peak predicted head impact severity outcomes and number of head impacts. All regression model assumptions were assessed for violations before statistical interpretation and performed using R Project for Statistical Programming (version 4.1.2; PBC) with the α level set a priori at 0.05. We performed correlation matrices of peak strength and time to peak by direction (Appendix B).

Results

Impact Frequency

We analyzed a total of 2,800 head impacts across 611 athlete exposures (Table 5.1). Cervical strength (*P* range =0.087 - 0.233) and time to peak (*P* range =0.125 - 0.188) (Table 5.2) in any directions were not associated with number of head impacts while controlling for player position.

Table 5.1. Descriptive Statistics of Head Impact Biomechanics

	Linear Acceleration (g) Mean±SD (Range)	**Rotational Velocity (rad/s) Mean±SD (Range)**	**Season No. Impacts Per Player Mean±SD (Range)**
All impacts (n=2,800)	37.6±13.5 (16.01 – 113.22)	19.1±10.0 (2.70 – 50.24)	107.7±19.3 (71 – 150)

Cervical Strength

Cervical strength outcomes did not predict peak linear acceleration. Left lateral flexion strength predicted rotational velocity (R^2=0.262, P=0.043), such that every 1 N/kg increase corresponded with a lesser severity of 6.49 rads/s rotational velocity (P=0.039). Peak right lateral flexion predicted rotational velocity (R^2=0.151, P=0.035), such that every 1 N/kg increase corresponded with a lesser severity of 2.08 rads/s (P=0.026).

Time to Peak

Time to peak in left lateral flexion predicted linear acceleration (R^2=0.072, P=0.041). such that each additional 1 second increase corresponded with 3.10g higher linear acceleration (P=0.036). Time to peak flexion (R^2=0.128, P=0.036) and extension (R^2=0.224, P=0.01) predicted rotational velocity such that every 1 second increase corresponded with 4.41 rads/s (P=0.006) and 3.32 rads/s (P=0.008) greater rotational velocity, respectively.

Table 5.2. Descriptive Statistics and Beta Coefficients (back transformed) of Cervical Neck Strength and Time to Peak.

Cervical Peak Strength Normalized by Body Mass (N/kg)	**Mean±SD**	**Model Summary**	**Beta PLA(g)**	**P**	**Model Summary**	**Beta PRV (rads)**	**P**
Flexion	0.46±0.13	R^2=0.174, P=.225	-0.81	.062	R^2=0.188, P=.110	-0.98	.190
Extension	0.64±0.14	R^2=0.054, P=.124	-0.59	.071	R^2=0.009, P=.206	-1.06	.312
Left Lateral Flexion	0.46±0.14	R^2=0.084, P=.098	-0.66	.126	R^2=0.262, P=0.043	-6.49	.039*
Right Lateral Flexion	0.46±0.12	R^2=0.189, P=.265	-0.72	.098	R^2=0.151, P=.035	-2.08	.026*
Time to Peak Strength	**Mean±SD**	**Model Summary**	**Beta PLA(g)**	**P**	**Model Summary**	**Beta PRV (rad/s)**	**P**

Directions (s)							
Flexion	0.54±0.08	R^2=0.159, P=.222	1.87	.116	R^2=0.128, P=.036	4.41	.006*
Extension	0.64±0.06	R^2=0.189, P=.106	1.62	.098	R^2=0.224, P=.01	3.32	.008*
Left Lateral Flexion	0.62±0.06	R^2=0.072, P=.041	3.10	.036*	R^2=0.118, P=.260	0.91	.156
Right Lateral Flexion	0.63±0.07	R^2=0.089, P=.067	1.60	.132	R^2=0.127, P=.241	1.20	.300

significant predictor

Discussion

Our aim was to determine whether cervical strength and time to peak influenced head impact frequency and severity. Utilizing the MCU allows for the participant to be in an upright position which aids in the clinical applicability in our study. Our findings showed that youth tackle football players with stronger cervical musculature in left and right lateral flexion sustained head impacts of lower rotational velocity. Additionally, our findings showed that slower time to peak strength in flexion, extension, and left lateral flexion sustained head impacts of higher severity. However, we did not find cervical strength characteristics to predict frequency of head impacts.

Previous findings regarding cervical strength outcomes in this area have been mixed compared to our results. High school and college football players with greater cervical stiffness had decreased odds of sustaining moderate and severe head impacts compared to those with less cervical stiffness (Schmidt et al., 2014). However, in these same football players and a separate cohort of youth ice hockey players cervical muscle strength did not influence head impact severity (Mihalik et al., 2011; Schmidt et al., 2014). Mixed results may exist due a variety of measurements used to assess cervical strength. We used the multi-cervical unit, which requires the participant to be seated and the head placed in neutral. A seated position may be more advantageous in this study since football is played in an upright position. However, this equipment is limited in that it does not allow for obtaining rotational isometric strength. Additionally, the software for the MCU did not allow for altering order of directions. Other studies have used dynamometer configurations, laying down positions, or loading apparatuses and different body positioning to assess neck strength (Eckner, Oh, Joshi, Richardson, & Ashton-Miller, 2014; Schmidt et al., 2014).

Striking players who are delivering a tackle are able to line up their head, neck, and torso (Viano et al., 2007). Our findings support the commonly held theory that a stronger and stiffer cervical muscles are capable of creating more rigidity to counter the force of collision and reduce head acceleration (Viano & Pellman, 2005). A previous study found that every one pound increase in neck strength reduced the odds of sustaining a concussion by 5% (Collins et al., 2014). Concussion risk and head impact severity are not linearly related. A high impact magnitude does not necessarily result in a concussion. However, increasing cervical strength through a protocol may contribute to limiting head acceleration upon impact.

No previous studies have examined time to peak strength, but studies with similar variables have also been mixed compared to our results. A previous study found that player anticipation results in less severe head impact measures (Mihalik, Blackburn, et al., 2010). When athletes sustain a collision that is unanticipated, they may not have adequate time to voluntarily contract cervical muscles to counteract the external force - resulting in rapid head acceleration. Our findings suggest a slower time to peak strength in the flexor and extensor muscle groups was associated with greater head rotational velocity.

It seems possible that players who foresee an oncoming collision can quickly contract cervical musculature reaching their peak strength earlier and allowing their head and neck segment to be more rigid. A previous study found increased cervical stiffness protected football players against higher severity head impacts (Schmidt et al., 2014).

Cervical strength and time to peak did not decrease impact frequency. Players may not be aware of their neck strength and not willing to take the risk of creating collisions regardless of body mass. Head impacts were still occurring most likely due to the nature of football where impacts are present (Hedlund, 2000).

Limitations

This study had inherent limitations. First, the Triax SIM-G has mixed evidence about its accuracy (Cummiskey et al., 2017; Karton et al., 2016; Oeur et al., 2016; Tyson et al., 2018). We took diligent notes to only include real-time head impacts and to eliminate any spurious frequencies. We also suggest using our filtering method would not have found differences if using video verification which is commonly used to eliminate false head impacts (Le et al., in review). Second, these data were collected from a small sample from a recreation department. Our findings may not be generalizable to other geographical areas, different age groups, or female players. Third, the MCU software utilized a rather low sampling frequency of 20Hz, which limited our temporal resolution when determining time to peak across trials.

Conclusion

There have been limited studies analyzing cervical strength on head impact biomechanics in youth tackle football. Stronger isometric cervical strength was associated with decreased head impact severity and greater time to peak was associated with increased rotational velocity. Future research should determine whether cervical strength and time to peak type exercises or interventions reduce head impact severity. Results from future studies may provide valuable information for coaches and healthcare professionals when designing cervical strengthening programs to mitigate head impact severity.

CHAPTER 6

MANUSCRIPT III

THE INFLUENCE OF ANTHROPOMETRICS AND HELMET CHARACTERISTICS ON HEAD IMPACT BIOMECHANICS IN YOUTH FOOTBALL PLAYERS

Context: Critical periods of growth and maturation introduce wide variability in youth football player anthropometrics. More research is warranted to determine additional anthropometric and equipment factors to improve the safety of tackle football for youth cohorts.

Objective: To determine if anthropometric characteristics and helmet dimensions predict head impact biomechanics.

Methods and Materials: We studied 27 youth tackle football players (males=100%, age=12.2±0.4yrs, height=161.5±9.7m, mass=55.5±15.2kg) during the 2021 fall season. We captured preseason measures of anthropometric characteristics (i.e. neck circumference, head mass) and helmet dimensions (i.e. helmet and face mask mass). During the season, participants wore Triax SIM-G accelerometers that captured head impact peak linear acceleration (PLA;g), peak rotational velocity (PRV; rad/s), and total number of impacts. Univariate linear regression models with random intercepts were used to determine the effect of anthropometric characteristics and helmet dimensions on log-transformed PLA and PRV.

Results: Total helmet mass predicted number of head impacts (R^2=0.356, P=0.022) and rotational velocity (R^2=0.150, P=0.028), such that each additional 1% rise in helmet mass relative to body mass corresponded to 2.2 more impacts (P=0.029) sustained and 4.10 rad/s higher rotational velocity (P=0.022). Face mask mass predicted number of head impacts

(R^2=0.244, P=0.006) and rotational velocity (R^2=0.133, P=0.023), such that each additional 1% rise in face mask mass relative to body mass corresponded to 3.6 more head impacts (P=0.006) and an increase of 3.22 rad/s (P=0.023). Face mask length predicted number of head impacts (R^2=0.178, P=0.018) and rotational velocity (R^2=0.153, P=0.032), such that every 1 cm corresponded with 4.0 more head impacts (P=0.018) and an increase of 1.98 rad/s (P=0.005). Helmet depth predicted rotational velocity (R^2=0.180, P=0.028), such that each additional 1 cm corresponded with a decrease of 1.37 rad/s (P=0.027). Helmet circumference predicted rotational velocity (R^2=0.092, P=0.018), such that each additional 1cm increase corresponded with an increase of 1.9 rad/s (P=0.042).

Conclusions: Anthropometrics and helmet dimensions were predictors for head impact severity and frequency. Results from this study may be used inform league regulations and helmet manufacturers to improve sport safety and mitigating head impacts.

Introduction

Most youth tackle football leagues stratify players based on age, though some leagues stratify players based on mass. Players of the same age can vary in physical maturity by as many as 4 years (Linder et al., 1995). In youth football, height and mass vary considerably within age groups and teams. Yet, in age-based leagues in youth football, two players with vastly different mass could face off against one another on the football field. A player with more mass will have greater momentum and impart greater force relative to a player of smaller mass travelling at the same velocity. In a scenario where these two players travel at the same speed and then strike one another, the smaller player is left to absorb a greater impact load, likely resulting in greater head impact severity. As a solution, some sport organizations have moved towards weight-restricted leagues where the player's mass must fall within a defined range. In weight-restricted leagues, players may be playing with others of various ages, but have similar weight classifications (Kerr et al., 2015). In contrast, potential benefits from age-restricted or age-only leagues are that players are not removed from their age-based peer groups that would happen in weight-restricted leagues (Kerr et al., 2015) and players are not evaluated based on potentially sensitive issues of body mass.

Anthropometrics of youth athletes have not been examined, despite evidence that anthropometrics and physical performance characteristics influence head injuries (i.e. concussion, ankle sprains, etc.) in high school, collegiate, and professional football players (Caswell et al., 2016). High school football players with higher body mass indices and higher stages of maturity had greater injury risk (Caswell et al., 2016; Linder et al., 1995) . A previous study revealed youth tackle football players who were taller and heavier had greater head impact severity compared those who were shorter with less mass (Yeargin et al., 2018). Further investigations may aid in determining whether age-based or weight-based groups with similar anthropometric data could be advantageous in youth football. Comparing head impacts between shorter vs. taller or heavier vs. lighter football players may help youth football leagues in creating teams to facilitate equal playing performance and reduce injury risk. There is also limited evidence to suggest head circumference, head mass, neck length, neck circumference, and leg length influence head impact biomechanics.

In addition to the anthropometrics of the body and head, helmet dimensions may play a role in mitigating head impacts in youth tackle football. Different helmet brands introduce variation in helmet designs. There is no research investigating helmet dimensions such as face mask length, helmet depth (measured from top of the head padding inside of the helmet to the shell just below the ear holes), helmet mass, and helmet circumference related to head impact biomechanics in youth football athletes. Tackle football players may sustain head impacts at various locations of the helmet including the face mask (Crisco et al., 2010). A face mask that is longer and distributed further away from the head's center of mass distributes the mass further from the axis of rotation. This makes it harder to initially accelerate which also means it may be more difficult to decelerate the head once accelerated. In addition, head impacts that occur further from the center of mass of the head would generate greater rotational velocity as they act over a longer lever arm. Therefore, helmet designs that allow surfaces that are further distributed from the center of mass or distribute mass differently around head center of mass may influence head impact biomechanics. For example, football players with heavier face masks may be more prone to sustaining a higher proportion of top of the head impacts suggesting that the additional mass may influence players to lower their head during collision (Schmidt et al., 2018). This could be critical for youth players where a large head mass must be supported by underdeveloped cervical musculature making it more difficult to control rapid head acceleration (Collins et al., 2014; Yoganandan et al., 2009). These issues could further inform future helmet design and considerations. Therefore, further research is needed to determine helmet dimensions are associated with head impact biomechanics. The purpose of this study is to determine whether anthropometric characteristics and helmet dimensions predict head impact biomechanics in youth tackle football players. We hypothesized players with greater anthropometric units of measure would be associated with greater head impact severity and frequency. We hypothesized players with greater helmet dimensions would be associated with greater head impact severity and frequency.

Methods

Study Participants

A youth tackle football team was recruited from a single recreation department in the rural southern United States. Twenty-seven youth tackle football players (males=100%, age=12.2±0.4yrs, height=161.5±9.7m, mass=55.5±15.2kg) from 2 teams participated in this study during the 2021 fall football season. The study protocol was approved by our Institutional Review Board. We obtained child assent with at least one parent or guardian providing signed informed consent.

Anthropometric Characteristics

Anthropometric outcomes included: height (cm), mass (kg), true leg length (cm), neck circumference (cm), neck length (cm), and head circumference (cm). We captured height (cm) by using a stadiometer (SECA) in a standing position with shoes removed and players were instructed to “stand up tall”. Mass (kg) was captured via a Med-Weigh digital scale MS-330. True leg length (cm) was measured using a soft-tape measure (brand: Gdminlo) from the anterior superior iliac spine to the medial malleolus from a supine position (Sabharwal & Kumar, 2008). Neck circumference (cm) was measured with the player standing? in good posture using a tape measure applied perpendicular to the neck at the inferior pole of the laryngeal prominence (Adam's apple) (Ben-Noun et al., 2001; Preis et al., 2010). Neck length (cm) was measured from the base of the occiput to cervical spinous process #7. Head circumference was measured over the most prominent part of the occiput and just above the supraorbital ridges (Harris, 2015). Head mass (kg) was captured by taking 8% of the total body mass (Ben-Noun et al., 2001; Preis et al., 2010). All anthropometric measures were taken during pre-season.

Helmet Dimensions

Total helmet mass (kg) and face mask mass were measured using the Med-Weigh digital scale. We removed the face mask from the helmet to obtain its mass (kg) as well as the helmet shell mass (kg). Helmet depth (cm) and helmet circumference (cm) were measured with the aforementioned tape measure. Helmet depth was measured from the most distal aspect of the helmet shell that would cover the lower part of the jaw to the inside of the helmet where top of the head padding would protect the head. Helmet circumference was measured just above the bolts from the top of the face mask and around the helmet so that the bottom of the helmet as an extension of the occiput was included. Length of the face mask in the sagittal plane from the sella turcica (outer ear: fossa of antihelix crura) (cm) were measured when the player had their helmet on. We normalized helmet mass and face mask mass by body mass (Table 6.1a) and all helmet measures were taken during pre-season.

Head Impact Biomechanics

Head impact biomechanics were captured using the Triax Smart Impact Monitor (SIM-G) head impact sensors (Triax Technologies Inc., Norwalk, CT). The head impact sensor is placed in either a headband or skullcap. All participants were fitted for comfort and instructed on how to place the device on their head. The sensor rests at the lower back of the head near the occiput. SIM-G linear acceleration (g) data is first measured at the sensor and is then transformed to the center of gravity of the head using the measured rotational velocity (rads/s) (Yoganandan et al., 2009). A 14-g threshold was established such that data 10ms before the impact and 52m after impact were recorded when meeting or exceeding 14-g (Lynall et al., 2019).

No on-field head impact biomechanics system is perfect. Previous studies suggest the SIM-G has inconsistent validation data (Cummiskey et al., 2017; Karton et al., 2016; Oeur et al., 2016; Tyson et al., 2018). Previous investigations have reported optimal impact detection with 100% of non-helmeted impacts detected and 80% of helmeted impacts detected in one study (Oeur et al., 2016), and 99% of helmeted impacts detected in another study (Cummiskey et al., 2017). There was also no difference between head form and SIM-G linear acceleration at low and medium energy impacts in one study (Karton et al., 2016) and approximately 14.25% error on average found in another study (Cummiskey et al., 2017). A separate investigation reported less than optimal acceleration validity where there was underpredicted linear and rotational acceleration in helmeted (linear acceleration slope=0.31; rotational velocity=0.74) and non-helmeted conditions (linear acceleration slope=0.37; rotational velocity=0.94) (Tyson et al., 2018). We used the SIM-G because it can record head impact data from helmeted participants regardless of helmet manufacturers and brand.

Data Analysis

Head impact data went through a multi-step process before being analyzed where session was defined as either a practice or game. First, head impact went through an interval validity check within the Triax software which identifies and labels false impacts (Lynall et al., 2019). Head impacts were removed if they occurred outside of each defined practice or game session time. We also removed head impacts that occurred in any pauses or breaks in the session such as water breaks, quarter breaks, injury start/end times, and time outs. To further ensure only real-time head impacts were included in the dataset, we excluded individual player sessions where a large number of head impacts occurred, most likely where a player removed and placed back on the headband often or was not worn appropriately. Similar to prior studies, we calculated the threshold of 51 session head impacts by adding the 3rd quartile value (33 head impacts) to the median (18 head impacts) (Lynall et al., 2019). Our dependent variables consisted peak linear acceleration (PLA;g), peak rotational velocity (PRV;rad/s), and number of head impacts.

Helmet mass and face mask mass were normalized by body mass (kg) then multiplied by 100 to be expressed as a percentage of body mass. We decided to report this as a percentage since helmet and face mask mass had the same units of measurement as body mass.

Statistical Analyses

Head impact severity data (linear acceleration and rotational velocity) were log transformed to approximate for a near-normal distribution then back transformed for interpretation. We compared independent variables between player position (lineman vs non-linemen) using independent samples t-tests but outcomes did not significantly differ between player position (*P*=0.127). Accordingly, we did not control for player position. Separate, univariate, linear regression models with random intercepts were used for each anthropometric characteristic and helmet dimension to predict head impact linear acceleration (g) and rotational velocity (rad/s). Separate univariate linear regression models were used to determine if anthropometric characteristics and helmet dimension were associated with number of head impacts. All regression model assumptions were assessed for violations before statistical interpretation. All statistical analyses were performed using R Studio (version 4.1.2; PBC) with $\alpha = 0.05$.

Results

We collected a total of 2,800 head impacts across 611 athlete exposures. Descriptive statistics for our predictors and head impact biomechanics can be found on Table 6.1a and b. We also performed separate correlation analyses for anthropometric and helmet dimensions (Appendix C).

Table 6.1a. Descriptive Statistics of Predictors

Anthropometrics Characteristics	**Mean±SD**	**Normalized**	**Description**
Age (y)	12.2±0.4	N/A	Self-reported
Height (cm)	161.5±9.7	N/A	Stadiometer (SECA)
Mass (%)	55.5±15.2	N/A	Digital scale MS-330
Leg Length (cm)	88.2±6.5	N/A	True leg length from anterior-superior iliac spine to medial malleolus
Neck Length (cm)	9.4±0.7	N/A	Occiput to C7 spinous process
Neck Circumference (cm)	32.2±2.9	N/A	Perpendicular to spine around Adam's apple
Head Circumference (cm)	54.8±3.2	N/A	Just above eye brows to base of skull
Head Mass (%)	4.4±1.2	N/A	8% of body mass
Helmet Dimensions	**Mean±SD**		
Total Helmet Mass (%)*	1.8±0.1	3.39±0.91%	digital scale MS-330
Face Mask Mass (%)*	0.7±0.1	1.28±0.37%	digital scale MS-330
Helmet Depth (cm)	20.8±1.2	N/A	Measured from helmet shell at the bottom of ear holes to the inside padding that covers top of the head
Helmet Circumference (cm)	78.0±1.0	N/A	Measured from the top bolts to the base of the helmet covering occiput
Face Mask Length (cm)	20.8±0.8	N/A	Mastoid process of the head landmarked outside of the helmet to furthest distance on face mask

*Normalized to body mass

Table 6.1b. Descriptive Statistics of Head Impact Biomechanics

	Linear Acceleration (g) Mean±SD (Range)	**Rotational Velocity (rad/s) Mean±SD (Range)**	**Season No. Impacts Per Player Mean±SD (Range)**
All impacts (n=2,800)	37.6±13.5 (16.01 – 113.22)	19.1±10.0 (2.70 – 50.24)	107.7±19.3 (71 – 150)

Anthropometric Characteristics

Table 6.2 contains statistical analysis outcomes. Player height predicted number of head impacts (R^2=0.125, P=0.030), such that each additional 1 cm in height corresponded with 3.8 more head impacts (P=0.031). Player body mass predicted number of head impacts (R^2=0.373, P=0.032) such that each additional 1 kg corresponded with 2.1 more head impacts (P=0.033). Neck circumference predicted number of head impacts (R^2=0.272, P=0.003) and linear acceleration (R^2=0.142, P=0.013), such that each additional 1 cm increase corresponded with a decrease of 2.1 head impacts (P=0.003) and 3.21g (P=0.012). Head mass was a significant predictor for rotational velocity (R^2=0.131, P=0.002), such that each additional 1kg corresponded with an increase of 1.7 rad/s (P=0.002). Remaining predictors of age, height, mass, leg length, neck length, and head circumference did explain significant variance in PLA or PRV.

Helmet Dimensions

Total helmet mass predicted number of head impacts (R^2=0.356, P=0.028) and rotational velocity (R^2=0.150, P=0.021), such that each additional 1% rise in helmet mass relative to body mass corresponded to 2.2 more impacts sustained (P=0.029) and 4.10 rad/s higher rotational velocity (P=0.022). Face mask mass predicted number of head impacts (R^2=0.244, P=0.005) and rotational velocity (R^2=0.133, P=0.023), such that each additional 1% rise in face mask mass relative to body mass corresponded to 3.6 more head impacts (P=0.006) and an increase of 3.22 rad/s (P=0.023). Face mask length predicted number of head impacts (R^2=0.178, P=0.006) and rotational velocity (R^2=0.153, P=0.005), such that every 1 cm increase corresponded with 4.0 more head impacts (P=0.018) and an increase of 1.98 rad/s (P=0.005). Helmet depth predicted rotational velocity (R^2=0.180, P=0.026), such that each additional 1 cm increase corresponded with a decrease of 1.37 rad/s (P=0.027). Helmet circumference predicted rotational velocity (R^2=0.092, P=0.041), such that each additional 1cm increase corresponded with a decrease of 1.9 rad/s (P=0.042). See table 6.2 for beta coefficients and corresponding p-values for all univariate models.

Table 6.2. Beta Coefficients (back transformed for PLA and PRV) with corresponding P-values.

Anthropometric Characteristics	**Model Summary (R^2, P)**	**Frequency**		**Model Summary (R^2, P)**	**PLA**		**Model Summary (R^2, P)**	**PRV**	
		Beta	**P**		**Beta**	**P**		**Beta**	**P**
Age (y)	(R^2=0.065, P=.334)	1.3	.334	(R^2=0.063, P=.437)	15.21	.437	(R^2=0.234, P=.236)	3.05	.236
Height (cm)	(R^2=0.125, P=.031)	3.8	.031*	(R^2=0.067, P=.988)	1.96	.988	(R^2=0.177, P=.159)	-0.11	.159
Mass (kg)	(R^2=0.373, P=.033)	2.1	.033*	(R^2=0.097, P=.766)	2.83	.766	(R^2=0.078, P=.174)	-0.07	.174
Leg Length (cm)	(R^2=0.125, P=.421)	-1.22	.421	(R^2=0.082, P=.124)	10.21	.124	(R^2=0.013, P=.739)	-0.08	.739)
Neck Length (cm)	(R^2=0.231, P=.322)	0.018	.322	(R^2=0.019, P=.587)	11.15	.587	(R^2=0.098, P=.176)	-1.28	.176
Neck Circumference (cm)	(R^2=0.272, P=.003)	2.1	.003*	(R^2=0.142 P=.012)	-3.21	.012*	(R^2=0.027, P=.617)	-0.13	.617
Head Circumference (cm)	(R^2=0.114, P=.124)	1.2	.124	(R^2=0.009 P=.126)	1.10	.126	(R^2=0.091, P=.568)	-0.12	.568
Head Mass (%)	(R^2=0.125, P=.233)	1.1	.233	(R^2=0.015 P=.224)	2.22	.224	(R^2=0.131, P=.002)	1.7	.002*
Helmet Dimensions									
Total Helmet Mass (%)	(R^2=0.356 P=.029)	2.2	.029*	(R^2=0.023 P=.452)	8.55	.452	(R^2=0.15, P=.022)	4.10	.022*
Face Mask Mass (%)	(R^2=0.244, P=.006)	3.6	.006*	(R^2=0.126, P=.262)	4.66	.262	(R^2=0.13, P=.023)	3.22	.023*
Helmet Depth (cm)	(R^2=0.080, P=.101)	1.8	.101	(R^2=0.125, P=.126)	3.85	.126	(R^2=0.18, P=.027)	-1.37	0.027*

Helmet Circumference (cm)	(R^2=0.003, P=.097)	0.8	.098	(R^2=0.115, P=.224)	5.02	.225	(R^2=0.092, P=.042)	1.90	.042*
Face Mask Length from Skull COM (cm)	(R^2=0.178, P=.019)	4.0	.018*	(R^2=0.130 P=.127)	1.63	.126	(R^2=0.153, P=.005)	1.98	.005*

Discussion

Overall, players of greater height, body mass, and head mass sustained more head impacts and impact severity. However, players with greater neck circumference had decreased head impact severity. Helmet dimensions revealed a similar trend. Greater total helmet mass, helmet circumference, face mask mass, and face mask length had an increase in number of impacts and impact severity. However, greater helmet depth had a lower impact severity.

Anthropometric Characteristics

We found players of greater body mass and height sustained more frequent head impacts. Players of greater body mass and height may be more involved and active on the field. Risk compensation behavior may suggest that player who are bigger may perceive that they are more protected when tackling smaller opponents (Hedlund, 2000). Future research should consider gathering psychological measures in understanding the coaches' perspective of players of greater height and mass on tackling techniques and behavior. Similar to our findings, a study found players of greater body mass to sustain significantly higher numbers of impacts with lower severity compared to those of lesser mass (Baugh et al., 2015). Additionally, players of greater body mass experience impact with the ground more often and are the ones tackling more frequently (Wong et al. 2014). Taller youth football players tend to have their heads above their teammates and opponents. However, these players may try to reduce their height to perform proper tackling mechanics aimed at the torso of shorter players causing them to involve their head in collision more frequently. Additionally, shorter players tackling taller players at their torso may cause a whiplash mechanism for the taller players.

Our findings suggest that anthropometric characteristic of body mass and height do in fact influence head impact frequency and severity. Future administrators involved with league creation could consider matching players based on their anthropometrics of body mass and height.

We also found anthropometrics did not differ between player position (lineman vs non-lineman). Therefore, we did not determine if there was an interaction between player position and anthropometrics to predict head impact frequency and severity. However, one study found head impact severities that occurred on the line or in open field did not significantly differ (Wong, Wong, & Bailes, 2014). Future studies should consider combining anthropometrics with video analysis during events to gather a better understanding of opponent anthropometrics on head impact severity.

We found players with larger neck circumference had a decrease in impact frequency and linear acceleration. In a previous study regarding youth soccer athletes, greater neck size or neck girth was a significant predictor for lower peak linear acceleration and rotational acceleration (Caccese et al., 2018). A football player with a smaller neck circumference may sustain greater head acceleration. Another previous study found increased neck girth is positively correlated with greater muscle tissue (Tierney et al., 2005). Similarly, a previous study found that football players with greater cervical stiffness had reduced odds of sustaining higher severity head impacts (Schmidt et al., 2014). Players with a smaller neck circumference and heavier head mass may experience a bobble-head effect when the neck circumference and strength are not great enough to control the mass of the head during impact (Caccese et al., 2018). This study demonstrated cervical strength, stiffness, and girth are potential modifiable factors in mitigating head impact severity (Caccese et al., 2018).

No other anthropometric characteristics predicted head impact severity. Age was not a significant predictor, likely due to the limitation of using one age group and a small sample of participants. Additionally, predictors such as height, mass, leg length, neck length, head circumference, and head mass were not significant predictors of head impact severity possibly due to using a cohort from 2 teams of the same age group where their anthropometrics were similar. Youth players may generate lower collision speeds where their anthropometrics do not contribute to impact severity variance. In addition, we did not control for player position of opposing teams nor did we equip them with head impact sensors. We propose that players from opposing teams were also of similar anthropometrics and matched accordingly resulting in non-significant findings for head impact severity.

Our results differ somewhat from prior studies where the size of neck muscles had a non-significant relationship with head impact biomechanics (Schmidt et al., 2014). However, in our study we did not gather a cross-sectional area of cervical musculature and we studied youth athletes where the cervical musculature might play a larger role in mitigating head impact severity. Future studies should investigate cervical tissue compositions as it contributes to neck circumference and strength. Remaining anthropometric characteristics were not significant predictors of head impact biomechanics.

Helmet Dimensions

Total helmet mass relative to a smaller body mass may make youth tackle football players more prone to sustaining more severe head impacts due to the inability to control the mass of the head and helmet. A heavier helmet may be more difficult to initially accelerate, but also more difficult for a small player to decelerate once the head becomes accelerated. The heavier helmet may make it more difficult for a player to withstand an external force thus causing rapid head acceleration. It is also possible that players with heavier helmets are carrying a greater effective mass into the collision resulting in a higher impact severity.

A heavier face mask may make it difficult for the player to lift their entire helmet for a collision. We found that players with heavier face masks sustained head impact of higher rotational velocity. Sustaining a head impact at the furthest point of the face mask may result in faster head acceleration due to a longer moment arm. In addition, total helmet mass and face mask mass are highly correlated which may explain our significant findings since they are worn together during football events (r=0.757, P<0.001) (Appendix D). As mentioned previously, youth athletes may not have mature cervical musculature. When coupled with a disproportionate facemask, this could cause an increase in head impact severity. This may be a plausible explanation as linear acceleration was not significant whereas we found face mask mass and length were significant for rotational velocity. Football helmet manufacturers should consider the mass and size of the helmet for the youth population. These findings are rather different than previous studies using older counterparts (Schmidt et al., 2018). Youth football player are not like high school or college players. They produce slower collision speeds which may contribute to no differences found in severity.

Football players with longer face masks sustained more frequent head impacts of greater rotational velocity. Face masks that extend further from the center of skull mass create a longer moment arm. The moment of inertia is dependent on mass and the distance from the axis of rotation. A longer moment arm, in this case the face mask length, should result in higher rotational head kinematics when impacts are occurring at the most distal end of the facemask. Therefore, a longer face mask accompanied with an increased face mask mass resulted in increased rotational kinematics. This is supported by our findings. Future research is warranted to determine if a facemask closer to the face reduces head impact severity. Hockey and lacrosse all vary on face mask design with a shorter face mask and the same could hold true for youth football players.

Lastly, our findings suggest players with a greater helmet depth sustain less severe head impacts. Although players were adequately fitted for helmets, players with a shallower helmet may sustain head impacts of greater severity. A greater helmet circumference increased rotational velocity. Any part of the equipment that is further from the axis of rotation may make it more difficult to decelerate from an onset of rotational acceleration. Lastly, total helmet mass and helmet circumference were highly correlated (r=.928, P<.001) which may explain significant findings on rotational velocity. More helmet circumference could be indicative of more helmet mass, and more padding associated with an increase in head acceleration. Future manufacturing of youth football helmets should consider padding design to ensure helmet depth is sufficient.

Limitations

This study has several limitations. First, the Triax SIM-G has inherent mixed evidence regarding its validation (Cummiskey et al., 2017; Karton et al., 2016; Oeur et al., 2016; Tyson et al., 2018). However, we took diligent notes to only include real-time head impacts. Video verification of head impacts may not differ from using a multi-step cleaning process that was employed in our study (Le et al., in review). Second, these data were collected from a small sample from a recreation department. Our findings may not be generalizable to other geographical areas, different age groups, or female players. Future studies warrant to capture longitudinal data as youth athletes experience maturation and growth rapidly during this age group. Future research should consider if helmet modifications to reduce helmet and face mask mass mitigate head impact severity. Additionally, we did not capture impact characteristics or the players they striking which is recommended for future research.

Conclusion

Results from this study may be used to inform league commissioners to consider developing leagues based on height and mass. Additionally, helmet manufacturers should consider options for lighter materials with the same strength in the helmet and on the facemask to reduce total helmet mass. Parents may also consider the design of the facemask that may contribute to total mass such as fewer bars (i.e. H-frame at the eyes).

CHAPTER 7

CONCLUSIONS

We found that slower eye movements to be associated with increased head impact severity. A player who is able to identify their target or opponent more quickly may have the ability to prepare for impact and experience less head acceleration. Our findings suggest a potential modifiable factor of eye movements that may influence head impact severity. Future research is warranted to determine if a form of intervention reduces severity.

We found that cervical strength and how quickly it takes to reach peak outcomes is associated with head impact severity. Stronger neck strength was shown to reduce rotational velocity; whereas slower time to maximum strength was shown to increase head acceleration. Findings from this aim suggest clinicians and sports medicine stakeholders to further analyze the effect of cervical strength prior to the start of the season.

We found anthropometric characteristics and helmet dimensions to be associated with number of head impacts and rotational velocity. This suggest players of greater body mass and height sustain more head impacts. Greater neck circumference was associated with a decrease in impact frequency and severity. Helmet dimensions of greater helmet mass, face mask mass, face mask length were associated with an increase in number of head impacts and severity; greater helmet circumference was only associated with greater severity. Greater helmet depth was associated with a decrease in severity. Findings from this study suggest league commissioners and creators about different ways to develop teams based as a means to mitigate head impacts and improve sport safety. Additionally, helmet manufacturers should consider the design of football helmets to accommodate youth players to reduce head impact frequency and severity.

REFERENCES

Alosco, M. L., Kasimis, A. B., Stamm, J. M., Chua, A. S., Baugh, C. M., Daneshvar, D. H., et al. (2017). Age of first exposure to american football and long-term neuropsychiatric and cognitive outcomes. *Transl Psychiatry, 7*(9), e1236. doi:10.1038/tp.2017.197

Alosco, M. L., Mez, J., Tripodis, Y., Kiernan, P. T., Abdolmohammadi, B., Murphy, L., et al. (2018). Age of first exposure to tackle football and chronic traumatic encephalopathy. *Ann Neurol, 83*(5), 886-901. doi:10.1002/ana.25245

Alsalaheen, B., Stockdale, K., Pechumer, D., Broglio, S. P., & Marchetti, G. F. (2017). A comparative meta-analysis of the effects of concussion on a computerized neurocognitive test and self-reported symptoms. *J Athl Train, 52*(9), 834-846. doi:10.4085/1062-6050-52.7.05

Avedesian, J. M., Covassin, T., Baez, S., Nash, J., Nagelhout, E., & Dufek, J. S. (2021). Relationship between cognitive performance and lower extremity biomechanics: Implications for sports-related concussion. *Orthop J Sports Med, 9*(8), 23259671211032246. doi:10.1177/23259671211032246

Bahrami, N., Sharma, D., Rosenthal, S., Davenport, E. M., Urban, J. E., Wagner, B., et al. (2016). Subconcussive head impact exposure and white matter tract changes over a single season of youth football. *Radiology, 281*(3), 919-926. doi:10.1148/radiol.2016160564

Bailes, J. E., Petraglia, A. L., Omalu, B. I., Nauman, E., & Talavage, T. (2013). Role of subconcussion in repetitive mild traumatic brain injury. *J Neurosurg, 119*(5), 1235-1245. doi:10.3171/2013.7.Jns121822

Baker, J., Horton, S., Robertson-Wilson, J., & Wall, M. (2003). Nurturing sport expertise: Factors influencing the development of elite athlete. *J Sports Sci Med, 2*(1), 1-9.

Bari, S., Svaldi, D. O., Jang, I., Shenk, T. E., Poole, V. N., Lee, T., et al. (2019). Dependence on subconcussive impacts of brain metabolism in collision sport athletes: An mr spectroscopic study. *Brain Imaging Behav, 13*(3), 735-749. doi:10.1007/s11682-018-9861-9

Barth, J. T., Freeman, J. R., Broshek, D. K., & Varney, R. N. (2001). Acceleration-deceleration sport-related concussion: The gravity of it all. *J Athl Train, 36*(3), 253-256.

Baugh, C. M., Kiernan, P. T., Kroshus, E., Daneshvar, D. H., Montenigro, P. H., McKee, A. C., et al. (2015). Frequency of head-impact-related outcomes by position in ncaa division i collegiate football players. *J Neurotrauma, 32*(5), 314-326. doi:10.1089/neu.2014.3582

Baumann, B., Danos, P., Krell, D., Diekmann, S., Leschinger, A., Stauch, R., et al. (1999). Reduced volume of limbic system-affiliated basal ganglia in mood disorders: Preliminary data from a postmortem study. *J Neuropsychiatry Clin Neurosci, 11*(1), 71-78. doi:10.1176/jnp.11.1.71

Ben-Noun, L., Sohar, E., & Laor, A. (2001). Neck circumference as a simple screening measure for identifying overweight and obese patients. *Obes Res, 9*(8), 470-477. doi:10.1038/oby.2001.61

Blume, H. K., Lucas, S., & Bell, K. R. (2011). Subacute concussion-related symptoms in youth. *Phys Med Rehabil Clin N Am, 22*(4), 665-681, viii-ix. doi:10.1016/j.pmr.2011.08.007

Broglio, S. P., Eckner, J. T., Surma, T., & Kutcher, J. S. (2011). Post-concussion cognitive declines and symptomatology are not related to concussion biomechanics in high school football players. *J Neurotrauma, 28*(10), 2061-2068. doi:10.1089/neu.2011.1905

Broglio, S. P., Martini, D., Kasper, L., Eckner, J. T., & Kutcher, J. S. (2013). Estimation of head impact exposure in high school football: Implications for regulating contact practices. *Am J Sports Med, 41*(12), 2877-2884. doi:10.1177/0363546513502458

Broglio, S. P., Schnebel, B., Sosnoff, J. J., Shin, S., Fend, X., He, X., et al. (2010). Biomechanical properties of concussions in high school football. *Med Sci Sports Exerc, 42*(11), 2064-2071. doi:10.1249/MSS.0b013e3181dd9156

Broglio, S. P., Sosnoff, J. J., Shin, S., He, X., Alcaraz, C., & Zimmerman, J. (2009). Head impacts during high school football: A biomechanical assessment. *J Athl Train, 44*(4), 342-349. doi:10.4085/1062-6050-44.4.342

Broglio, S. P., Surma, T., & Ashton-Miller, J. A. (2012). High school and collegiate football athlete concussions: A biomechanical review. *Ann Biomed Eng, 40*(1), 37-46. doi:10.1007/s10439-011-0396-0

Caccese, J. B., Buckley, T. A., Tierney, R. T., Arbogast, K. B., Rose, W. C., Glutting, J. J., et al. (2018). Head and neck size and neck strength predict linear and rotational acceleration during purposeful soccer heading. *Sports Biomech, 17*(4), 462-476. doi:10.1080/14763141.2017.1360385

Campolettano, E. T., Gellner, R. A., & Rowson, S. (2017). High-magnitude head impact exposure in youth football. *J Neurosurg Pediatr, 20*(6), 604-612. doi:10.3171/2017.5.Peds17185

Campolettano, E. T., Rowson, S., & Duma, S. M. (2016). Drill-specific head impact exposure in youth football practice. *J Neurosurg Pediatr, 18*(5), 536-541. doi:10.3171/2016.5.Peds1696

Cancelliere, C., Coronado, V. G., Taylor, C. A., & Xu, L. (2017). Epidemiology of isolated versus nonisolated mild traumatic brain injury treated in emergency departments in the united states, 2006-2012: Sociodemographic characteristics. *J Head Trauma Rehabil, 32*(4), E37-e46. doi:10.1097/htr.0000000000000260

Cantu, R. C. (1996). Head injuries in sport. *Br J Sports Med, 30*(4), 289-296. doi:10.1136/bjsm.30.4.289

Castile, L., Collins, C. L., McIlvain, N. M., & Comstock, R. D. (2012). The epidemiology of new versus recurrent sports concussions among high school athletes, 2005-2010. *Br J Sports Med, 46*(8), 603-610. doi:10.1136/bjsports-2011-090115

Caswell, S. V., Ausborn, A., Diao, G., Johnson, D. C., Johnson, T. S., Atkins, R., et al. (2016). Anthropometrics, physical performance, and injury characteristics of youth american football. *Orthop J Sports Med, 4*(8), 2325967116662251. doi:10.1177/2325967116662251

Caswell, S. V., Lincoln, A. E., Almquist, J. L., Dunn, R. E., & Hinton, R. Y. (2012). Video incident analysis of head injuries in high school girls' lacrosse. *Am J Sports Med, 40*(4), 756-762. doi:10.1177/0363546512436647

Chrisman, S. P., & Richardson, L. P. (2014). Prevalence of diagnosed depression in adolescents with history of concussion. *J Adolesc Health, 54*(5), 582-586. doi:10.1016/j.jadohealth.2013.10.006

Collins, C. L., Fletcher, E. N., Fields, S. K., Kluchurosky, L., Rohrkemper, M. K., Comstock, R. D., et al. (2014). Neck strength: A protective factor reducing risk for concussion in high school sports. *J Prim Prev, 35*(5), 309-319. doi:10.1007/s10935-014-0355-2

Covassin, T., & Elbin, R. J. (2010). The cognitive effects and decrements following concussion. *Open Access J Sports Med, 1*, 55-61. doi:10.2147/oajsm.s6919

Crisco, J. J., Fiore, R., Beckwith, J. G., Chu, J. J., Brolinson, P. G., Duma, S., et al. (2010). Frequency and location of head impact exposures in individual collegiate football players. *J Athl Train, 45*(6), 549-559. doi:10.4085/1062-6050-45.6.549

Cummiskey, B., Schiffmiller, D., Talavage, T. M., Leverenz, L., Meyer, J. J., Adams, D., et al. (2017). Reliability and accuracy of helmet-mounted and head-mounted devices used to measure head accelerations. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology, 231*(2), 144-153.

Daniel RW, Rowson S, & SM, S. (2012). Head impact expsosure in youth football. *Ann Biomed Eng., 40*, 976-981.

DC, K. (2018). Youth tackle football - proposed legislation. Retrieved from www.brainlaw.com/youth-tackle-football/.

Dean, H. L., Martí, D., Tsui, E., Rinzel, J., & Pesaran, B. (2011). Reaction time correlations during eye-hand coordination: Behavior and modeling. *J Neurosci, 31*(7), 2399-2412. doi:10.1523/jneurosci.4591-10.2011

Diederich, A., Schomburg, A., & Colonius, H. (2012). Saccadic reaction times to audiovisual stimuli show effects of oscillatory phase reset. *PLoS One, 7*(10), e44910. doi:10.1371/journal.pone.0044910

Eckner, J. T., Kutcher, J. S., & Richardson, J. K. (2011). Effect of concussion on clinically measured reaction time in 9 ncaa division i collegiate athletes: A preliminary study. *Pm r, 3*(3), 212-218. doi:10.1016/j.pmrj.2010.12.003

Eckner, J. T., Lipps, D. B., Kim, H., Richardson, J. K., & Ashton-Miller, J. A. (2011). Can a clinical test of reaction time predict a functional head-protective response? *Med Sci Sports Exerc, 43*(3), 382-387. doi:10.1249/MSS.0b013e3181f1cc51

Eckner, J. T., Oh, Y. K., Joshi, M. S., Richardson, J. K., & Ashton-Miller, J. A. (2014). Effect of neck muscle strength and anticipatory cervical muscle activation on the kinematic response of the head to impulsive loads. *Am J Sports Med, 42*(3), 566-576. doi:10.1177/0363546513517869

Field, M., Collins, M. W., Lovell, M. R., & Maroon, J. (2003). Does age play a role in recovery from sports-related concussion? A comparison of high school and collegiate athletes. *J Pediatr, 142*(5), 546-553. doi:10.1067/mpd.2003.190

Gavett, B. E., Stern, R. A., & McKee, A. C. (2011). Chronic traumatic encephalopathy: A potential late effect of sport-related concussive and subconcussive head trauma. *Clin Sports Med, 30*(1), 179-188, xi. doi:10.1016/j.csm.2010.09.007

Gessel, L. M., Fields, S. K., Collins, C. L., Dick, R. W., & Comstock, R. D. (2007). Concussions among united states high school and collegiate athletes. *J Athl Train, 42*(4), 495-503.

Giza, C. C., & Hovda, D. A. (2014). The new neurometabolic cascade of concussion. *Neurosurgery, 75 Suppl 4*(0 4), S24-33. doi:10.1227/neu.0000000000000505

Guskiewicz, K. M., Marshall, S. W., Bailes, J., McCrea, M., Cantu, R. C., Randolph, C., et al. (2005). Association between recurrent concussion and late-life cognitive impairment in retired professional football players. *Neurosurgery, 57*(4), 719-726; discussion 719-726. doi:10.1093/neurosurgery/57.4.719

Guskiewicz, K. M., Marshall, S. W., Bailes, J., McCrea, M., Harding, H. P., Jr., Matthews, A., et al. (2007). Recurrent concussion and risk of depression in retired professional football players. *Med Sci Sports Exerc, 39*(6), 903-909. doi:10.1249/mss.0b013e3180383da5

Guskiewicz, K. M., & Mihalik, J. P. (2011). Biomechanics of sport concussion: Quest for the elusive injury threshold. *Exerc Sport Sci Rev, 39*(1), 4-11. doi:10.1097/JES.0b013e318201f53e

Guskiewicz, K. M., Mihalik, J. P., Shankar, V., Marshall, S. W., Crowell, D. H., Oliaro, S. M., et al. (2007). Measurement of head impacts in collegiate football players: Relationship between head impact biomechanics and acute clinical outcome after concussion. *Neurosurgery, 61*(6), 1244-1252; discussion 1252-1243. doi:10.1227/01.neu.0000306103.68635.1a

Guskiewicz, K. M., & Valovich McLeod, T. C. (2011). Pediatric sports-related concussion. *Pm r, 3*(4), 353-364; quiz 364. doi:10.1016/j.pmrj.2010.12.006

Guskiewicz, K. M., Weaver, N. L., Padua, D. A., & Garrett, W. E., Jr. (2000). Epidemiology of concussion in collegiate and high school football players. *Am J Sports Med, 28*(5), 643-650. doi:10.1177/03635465000280050401

Harpham, J. A., Mihalik, J. P., Littleton, A. C., Frank, B. S., & Guskiewicz, K. M. (2014). The effect of visual and sensory performance on head impact biomechanics in college football players. *Ann Biomed Eng, 42*(1), 1-10. doi:10.1007/s10439-013-0881-8

Harris, S. R. (2015). Measuring head circumference: Update on infant microcephaly. *Can Fam Physician, 61*(8), 680-684.

Hedlund, J. (2000). Risky business: Safety regulations, risks compensation, and individual behavior. *Inj Prev, 6*(2), 82-90. doi:10.1136/ip.6.2.82

Hildenbrand, K. J., & Vasavada, A. N. (2013). Collegiate and high school athlete neck strength in neutral and rotated postures. *J Strength Cond Res, 27*(11), 3173-3182. doi:10.1519/JSC.0b013e31828a1fe2

Howell, D. R., Brilliant, A. N., Storey, E. P., Podolak, O. E., Meehan, W. P., 3rd, & Master, C. L. (2018). Objective eye tracking deficits following concussion for youth seen in a sports medicine setting. *J Child Neurol, 33*(12), 794-800. doi:10.1177/0883073818789320

Howell, D. R., Osternig, L. R., & Chou, L. S. (2015). Return to activity after concussion affects dual-task gait balance control recovery. *Med Sci Sports Exerc, 47*(4), 673-680. doi:10.1249/mss.0000000000000462

Hunfalvay, M., Roberts, C. M., Murray, N., Tyagi, A., Kelly, H., & Bolte, T. (2019). Horizontal and vertical self-paced saccades as a diagnostic marker of traumatic brain injury. *Concussion, 4*(1), Cnc60. doi:10.2217/cnc-2019-0001

Iverson, G. L., Caccese, J. B., Merz, Z. C., Büttner, F., & Terry, D. P. (2021). Age of first exposure to football is not associated with later-in-life cognitive or mental health problems. *Frontiers in neurology, 12*, 647314-647314. doi:10.3389/fneur.2021.647314

Johnson, B., Zhang, K., Hallett, M., & Slobounov, S. (2015). Functional neuroimaging of acute oculomotor deficits in concussed athletes. *Brain Imaging Behav, 9*(3), 564-573. doi:10.1007/s11682-014-9316-x

Kamp, J. v. d. (2011). Exploring the merits of perceptual anticipation in the soccer penalty kick. . *Motor control, 15*(3), 342-358.

Karton, C., Oeur, R. A., & Hoshizaki, T. B. (2016). *Measurement accuracy of head impact monitoring sensor in sport.* Paper presented at the ISBS-Conference Proceedings Archive.

Kerr, Z. Y., Marshall, S. W., Harding, H. P., Jr., & Guskiewicz, K. M. (2012). Nine-year risk of depression diagnosis increases with increasing self-reported concussions in retired professional football players. *Am J Sports Med, 40*(10), 2206-2212. doi:10.1177/0363546512456193

Kerr, Z. Y., Marshall, S. W., Simon, J. E., Hayden, R., Snook, E. M., Dodge, T., et al. (2015). Injury rates in age-only versus age-and-weight playing standard conditions in american youth football. *Orthop J Sports Med, 3*(9), 2325967115603979. doi:10.1177/2325967115603979

King, D., Hume, P., Gissane, C., Brughelli, M., & Clark, T. (2016). The influence of head impact threshold for reporting data in contact and collision sports: Systematic review and original data analysis. *Sports Med, 46*(2), 151-169. doi:10.1007/s40279-015-0423-7

Kleiner, M., Wong, L., Dubé, A., Wnuk, K., Hunter, S. W., & Graham, L. J. (2018). Dual-task assessment protocols in concussion assessment: A systematic literature review. *J Orthop Sports Phys Ther, 48*(2), 87-103. doi:10.2519/jospt.2018.7432

Kontos, A. P., Elbin, R. J., Fazio-Sumrock, V. C., Burkhart, S., Swindell, H., Maroon, J., et al. (2013). Incidence of sports-related concussion among youth football players aged 8-12 years. *J Pediatr, 163*(3), 717-720. doi:10.1016/j.jpeds.2013.04.011

Lacerda, A. L., Keshavan, M. S., Hardan, A. Y., Yorbik, O., Brambilla, P., Sassi, R. B., et al. (2004). Anatomic evaluation of the orbitofrontal cortex in major depressive disorder. *Biol Psychiatry, 55*(4), 353-358. doi:10.1016/j.biopsych.2003.08.021

Lamond, L. C., Caccese, J. B., Buckley, T. A., Glutting, J., & Kaminski, T. W. (2018). Linear acceleration in direct head contact across impact type, player position, and playing scenario in collegiate women's soccer players. *Journal of Athletic Training, 53*(2), 115-121. doi:10.4085/1062-6050-90-17

Land, M., & Tatler, B. (2009). *Looking and acting: Vision and eye movements in natural behaviour*: Oxford University Press.

Langlois, J. A., Rutland-Brown, W., & Wald, M. M. (2006). The epidemiology and impact of traumatic brain injury: A brief overview. *J Head Trauma Rehabil, 21*(5), 375-378. doi:10.1097/00001199-200609000-00001

Leigh, R. J., & Zee, D. S. (2015). *The neurology of eye movements*: OUP USA.

Lempke, L. B., Johnson, R. S., Schmidt, J. D., & Lynall, R. C. (2020). Clinical versus functional reaction time: Implications for postconcussion management. *Med Sci Sports Exerc, 52*(8), 1650-1657. doi:10.1249/mss.0000000000002300

Lincoln, A. E., Caswell, S. V., Almquist, J. L., Dunn, R. E., & Hinton, R. Y. (2013). Video incident analysis of concussions in boys' high school lacrosse. *Am J Sports Med, 41*(4), 756-761. doi:10.1177/0363546513476265

Linder, M. M., Townsend, D. J., Jones, J. C., Balkcom, I. L., & Anthony, C. R. (1995). Incidence of adolescent injuries in junior high school football and its relationship to sexual maturity. *Clin J Sport Med, 5*(3), 167-170. doi:10.1097/00042752-199507000-00006

Luna, B., Paulsen, D. J., Padmanabhan, A., & Geier, C. (2013). Cognitive control and motivation. *Curr Dir Psychol Sci, 22*(2), 94-100. doi:10.1177/0963721413478416

Luna, B., Velanova, K., & Geier, C. F. (2008). Development of eye-movement control. *Brain Cogn, 68*(3), 293-308. doi:10.1016/j.bandc.2008.08.019

Lynall, R. C., Blackburn, J. T., Guskiewicz, K. M., Marshall, S. W., Plummer, P., & Mihalik, J. P. (2018). Reaction time and joint kinematics during functional movement in recently concussed individuals. *Arch Phys Med Rehabil, 99*(5), 880-886. doi:10.1016/j.apmr.2017.12.011

Lynall, R. C., Campbell, K. R., Wasserman, E. B., Dompier, T. P., & Kerr, Z. Y. (2017). Concussion mechanisms and activities in youth, high school, and college football. *J Neurotrauma, 34*(19), 2684-2690. doi:10.1089/neu.2017.5032

Lynall, R. C., Lempke, L. B., Johnson, R. S., Anderson, M. N., & Schmidt, J. D. (2019). A comparison of youth flag and tackle football head impact biomechanics. *J Neurotrauma, 36*(11), 1752-1757. doi:10.1089/neu.2018.6236

Marar, M., McIlvain, N. M., Fields, S. K., & Comstock, R. D. (2012). Epidemiology of concussions among united states high school athletes in 20 sports. *Am J Sports Med, 40*(4), 747-755. doi:10.1177/0363546511435626

McAllister, Ford, J. C., Flashman, L. A., Maerlender, A., Greenwald, R. M., Beckwith, J. G., et al. (2014). Effect of head impacts on diffusivity measures in a cohort of collegiate contact sport athletes. *Neurology, 82*(1), 63-69. doi:10.1212/01.wnl.0000438220.16190.42

McAllister, T., & McCrea, M. (2017). Long-term cognitive and neuropsychiatric consequences of repetitive concussion and head-impact exposure. *J Athl Train, 52*(3), 309-317. doi:10.4085/1062-6050-52.1.14

McAllister, T. W., Flashman, L. A., Maerlender, A., Greenwald, R. M., Beckwith, J. G., Tosteson, T. D., et al. (2012). Cognitive effects of one season of head impacts in a cohort of collegiate contact sport athletes. *Neurology, 78*(22), 1777-1784. doi:10.1212/WNL.0b013e3182582fe7

McCaffrey, M. A., Mihalik, J. P., Crowell, D. H., Shields, E. W., & Guskiewicz, K. M. (2007). Measurement of head impacts in collegiate football players: Clinical measures of concussion after high- and low-magnitude impacts. *Neurosurgery, 61*(6), 1236-1243; discussion 1243. doi:10.1227/01.neu.0000306102.91506.8b

McCrea, M., Hammeke, T., Olsen, G., Leo, P., & Guskiewicz, K. (2004). Unreported concussion in high school football players: Implications for prevention. *Clin J Sport Med, 14*(1), 13-17. doi:10.1097/00042752-200401000-00003

McCrea, M., Prichep, L., Powell, M. R., Chabot, R., & Barr, W. B. (2010). Acute effects and recovery after sport-related concussion: A neurocognitive and quantitative brain electrical activity study. *J Head Trauma Rehabil, 25*(4), 283-292. doi:10.1097/HTR.0b013e3181e67923

McCrory, P., Meeuwisse, W., Dvořák, J., Aubry, M., Bailes, J., Broglio, S., et al. (2017). Consensus statement on concussion in sport-the 5(th) international conference on concussion in sport held in berlin, october 2016. *Br J Sports Med, 51*(11), 838-847. doi:10.1136/bjsports-2017-097699

McKee, A. C., Cantu, R. C., Nowinski, C. J., Hedley-Whyte, E. T., Gavett, B. E., Budson, A. E., et al. (2009). Chronic traumatic encephalopathy in athletes: Progressive tauopathy after repetitive head injury. *J Neuropathol Exp Neurol, 68*(7), 709-735. doi:10.1097/NEN.0b013e3181a9d503

Meaney, D. F., & Smith, D. H. (2011). Biomechanics of concussion. *Clin Sports Med, 30*(1), 19-31, vii. doi:10.1016/j.csm.2010.08.009

Meehan, W. P., 3rd, & Mannix, R. (2010). Pediatric concussions in united states emergency departments in the years 2002 to 2006. *J Pediatr, 157*(6), 889-893. doi:10.1016/j.jpeds.2010.06.040

Meehan, W. P., 3rd, Taylor, A. M., & Proctor, M. (2011). The pediatric athlete: Younger athletes with sport-related concussion. *Clin Sports Med, 30*(1), 133-144, x. doi:10.1016/j.csm.2010.08.004

Mihalik, J. P., Bell, D. R., Marshall, S. W., & Guskiewicz, K. M. (2007). Measurement of head impacts in collegiate football players: An investigation of positional and event-type differences. *Neurosurgery, 61*(6), 1229-1235; discussion 1235. doi:10.1227/01.neu.0000306101.83882.c8

Mihalik, J. P., Blackburn, J. T., Greenwald, R. M., Cantu, R. C., Marshall, S. W., & Guskiewicz, K. M. (2010). Collision type and player anticipation affect head impact severity among youth ice hockey players. *Pediatrics, 125*(6), e1394-1401. doi:10.1542/peds.2009-2849

Mihalik, J. P., Greenwald, R. M., Blackburn, J. T., Cantu, R. C., Marshall, S. W., & Guskiewicz, K. M. (2010). Effect of infraction type on head impact severity in youth ice hockey. *Med Sci Sports Exerc, 42*(8), 1431-1438. doi:10.1249/MSS.0b013e3181d2521a

Mihalik, J. P., Guskiewicz, K. M., Jeffries, J. A., Greenwald, R. M., & Marshall, S. W. (2008). Characteristics of head impacts sustained by youth ice hockey players. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology, 222*(1), 45-52. doi:10.1243/17543371jset4

Mihalik, J. P., Guskiewicz, K. M., Marshall, S. W., Greenwald, R. M., Blackburn, J. T., & Cantu, R. C. (2011). Does cervical muscle strength in youth ice hockey players affect head impact biomechanics? *Clin J Sport Med, 21*(5), 416-421. doi:10.1097/jsm.0b013e31822c8a5c

Mizobuchi, Y., & Nagahiro, S. (2016). A review of sport-related head injuries. *Korean J Neurotrauma, 12*(1), 1-5. doi:10.13004/kjnt.2016.12.1.1

Morais de Oliveira, A. L., Greco, C. C., Molina, R., & Denadai, B. S. (2012). The rate of force development obtained at early contraction phase is not influenced by active static stretching. *J Strength Cond Res, 26*(8), 2174-2179. doi:10.1519/JSC.0b013e31823b0546

Morley, W. A. (2018). Environmental subconcussive injury, axonal injury, and chronic traumatic encephalopathy. *Front Neurol, 9*, 166. doi:10.3389/fneur.2018.00166

Mulhall, L. E., Williams, I. M., & Abel, L. A. (1999). Bedside tests of saccades after head injury. *J Neuroophthalmol, 19*(3), 160-165.

Munce, T. A., Dorman, J. C., Thompson, P. A., Valentine, V. D., & Bergeron, M. F. (2015). Head impact exposure and neurologic function of youth football players. *Med Sci Sports Exerc, 47*(8), 1567-1576. doi:10.1249/mss.0000000000000591

Naunheim, R. S., Standeven, J., Richter, C., & Lewis, L. M. (2000). Comparison of impact data in hockey, football, and soccer. *J Trauma, 48*(5), 938-941. doi:10.1097/00005373-200005000-00020

Nilsson, K. J., Flint, H. G., Gao, Y., Kendrick, L., Cutchin, S., Pentecost, R., et al. (2019). Repetitive head impacts in youth football: Description and relationship to white matter structure. *Sports Health, 11*(6), 507-513. doi:10.1177/1941738119865264

Oeur, R. A., Karton, C., & Hoshizaki, T. B. (2016). *Impact frequency validation of head impact sensor technology for use in sport.* Paper presented at the ISBS-Conference Proceedings Archive.

Ommaya, A. K., & Gennarelli, T. A. (1974). Cerebral concussion and traumatic unconsciousness. Correlation of experimental and clinical observations of blunt head injuries. *Brain, 97*(4), 633-654. doi:10.1093/brain/97.1.633

Panjabi, M. M., Cholewicki, J., Nibu, K., Grauer, J., Babat, L. B., & Dvorak, J. (1998). Critical load of the human cervical spine: An in vitro experimental study. *Clin Biomech (Bristol, Avon), 13*(1), 11-17. doi:10.1016/s0268-0033(97)00057-0

Patel, D. R., Shivdasani, V., & Baker, R. J. (2005). Management of sport-related concussion in young athletes. *Sports Med, 35*(8), 671-684. doi:10.2165/00007256-200535080-00002

Pearson, I., Reichert, A., De Serres, S. J., Dumas, J. P., & Côté, J. N. (2009). Maximal voluntary isometric neck strength deficits in adults with whiplash-associated disorders and association with pain and fear of movement. *J Orthop Sports Phys Ther, 39*(3), 179-187. doi:10.2519/jospt.2009.2950

Pellman, E. J., Viano, D. C., Tucker, A. M., & Casson, I. R. (2003). Concussion in professional football: Location and direction of helmet impacts-part 2. *Neurosurgery, 53*(6), 1328-1340; discussion 1340-1321. doi:10.1227/01.neu.0000093499.20604.21

Powell, J. W., & Barber-Foss, K. D. (1999). Traumatic brain injury in high school athletes. *Jama, 282*(10), 958-963. doi:10.1001/jama.282.10.958

Preis, S. R., Massaro, J. M., Hoffmann, U., D'Agostino, R. B., Sr., Levy, D., Robins, S. J., et al. (2010). Neck circumference as a novel measure of cardiometabolic risk: The framingham heart study. *J Clin Endocrinol Metab, 95*(8), 3701-3710. doi:10.1210/jc.2009-1779

Rose, S. C., Yeates, K. O., Nguyen, J. T., Pizzimenti, N. M., Ercole, P. M., & McCarthy, M. T. (2021). Exposure to head impacts and cognitive and behavioral outcomes in youth tackle football players across 4 seasons. *JAMA Network Open, 4*(12), e2140359-e2140359. doi:10.1001/jamanetworkopen.2021.40359

Rowson, B., Tyson, A., Rowson, S., & Duma, S. (2018). Measuring head impacts: Accelerometers and other sensors. *Handb Clin Neurol, 158*, 235-243. doi:10.1016/b978-0-444-63954-7.00023-9

Rowson, S., Campolettano, E. T., Duma, S. M., Stemper, B., Shah, A., Harezlak, J., et al. (2019). Accounting for variance in concussion tolerance between individuals: Comparing head accelerations between concussed and physically matched control subjects. *Ann Biomed Eng, 47*(10), 2048-2056. doi:10.1007/s10439-019-02329-7

Sabharwal, S., & Kumar, A. (2008). Methods for assessing leg length discrepancy. *Clin Orthop Relat Res, 466*(12), 2910-2922. doi:10.1007/s11999-008-0524-9

Samadani, U. (2015). A new tool for monitoring brain function: Eye tracking goes beyond assessing attention to measuring central nervous system physiology. *Neural Regen Res, 10*(8), 1231-1233. doi:10.4103/1673-5374.162752

Savelsbergh, G. J., Van der Kamp, J., Williams, A. M., & Ward, P. (2005). Anticipation and visual search behaviour in expert soccer goalkeepers. *Ergonomics, 48*(11-14), 1686-1697. doi:10.1080/00140130500101346

Schmidt, J. D., Guskiewicz, K. M., Blackburn, J. T., Mihalik, J. P., Siegmund, G. P., & Marshall, S. W. (2014). The influence of cervical muscle characteristics on head impact biomechanics in football. *Am J Sports Med, 42*(9), 2056-2066. doi:10.1177/0363546514536685

Schmidt, J. D., Guskiewicz, K. M., Mihalik, J. P., Blackburn, J. T., Siegmund, G. P., & Marshall, S. W. (2016). Head impact magnitude in american high school football. *Pediatrics, 138*(2). doi:10.1542/peds.2015-4231

Schmidt, J. D., Phan, T. T., Courson, R. W., Reifsteck, F., 3rd, Merritt, E. D., & Brown, C. N. (2018). The influence of heavier football helmet faceguards on head impact location and severity. *Clin J Sport Med, 28*(2), 106-110. doi:10.1097/jsm.0000000000000437

Schultz, V., Stern, R. A., Tripodis, Y., Stamm, J., Wrobel, P., Lepage, C., et al. (2018). Age at first exposure to repetitive head impacts is associated with smaller thalamic volumes in former professional american football players. *J Neurotrauma, 35*(2), 278-285. doi:10.1089/neu.2017.5145

Sheline, Y. I., Sanghavi, M., Mintun, M. A., & Gado, M. H. (1999). Depression duration but not age predicts hippocampal volume loss in medically healthy women with recurrent major depression. *J Neurosci, 19*(12), 5034-5043. doi:10.1523/jneurosci.19-12-05034.1999

Shultz, S. R., MacFabe, D. F., Foley, K. A., Taylor, R., & Cain, D. P. (2012). Sub-concussive brain injury in the long-evans rat induces acute neuroinflammation in the absence of behavioral impairments. *Behav Brain Res, 229*(1), 145-152. doi:10.1016/j.bbr.2011.12.015

Slemmer, J. E., & Weber, J. T. (2005). The extent of damage following repeated injury to cultured hippocampal cells is dependent on the severity of insult and inter-injury interval. *Neurobiol Dis, 18*(3), 421-431. doi:10.1016/j.nbd.2004.09.022

Sosin, D. M., Sniezek, J. E., & Thurman, D. J. (1996). Incidence of mild and moderate brain injury in the united states, 1991. *Brain Inj, 10*(1), 47-54. doi:10.1080/026990596124719

Stamm, J. M., Bourlas, A. P., Baugh, C. M., Fritts, N. G., Daneshvar, D. H., Martin, B. M., et al. (2015). Age of first exposure to football and later-life cognitive impairment in former nfl players. *Neurology, 84*(11), 1114-1120. doi:10.1212/wnl.0000000000001358

Stemper, B. D., Shah, A. S., Harezlak, J., Rowson, S., Mihalik, J. P., Duma, S. M., et al. (2019). Comparison of head impact exposure between concussed football athletes and matched controls: Evidence for a possible second mechanism of sport-related concussion. *Ann Biomed Eng, 47*(10), 2057-2072. doi:10.1007/s10439-018-02136-6

Stern, R. A., Riley, D. O., Daneshvar, D. H., Nowinski, C. J., Cantu, R. C., & McKee, A. C. (2011). Long-term consequences of repetitive brain trauma: Chronic traumatic encephalopathy. *Pm r, 3*(10 Suppl 2), S460-467. doi:10.1016/j.pmrj.2011.08.008

Stuart, S., Parrington, L., Martini, D., Peterka, R., Chesnutt, J., & King, L. (2020). The measurement of eye movements in mild traumatic brain injury: A structured review of an emerging area. *Front Sports Act Living, 2,* 5. doi:10.3389/fspor.2020.00005

Sundman, M., Doraiswamy, P. M., & Morey, R. A. (2015). Neuroimaging assessment of early and late neurobiological sequelae of traumatic brain injury: Implications for cte. *Front Neurosci, 9,* 334. doi:10.3389/fnins.2015.00334

Taimela, S., Osterman, K., Alaranta, H., Soukka, A., & Kujala, U. M. (1993). Long psychomotor reaction time in patients with chronic low-back pain: Preliminary report. *Arch Phys Med Rehabil, 74*(11), 1161-1164.

Tierney, R. T., Higgins, M., Caswell, S. V., Brady, J., McHardy, K., Driban, J. B., et al. (2008). Sex differences in head acceleration during heading while wearing soccer headgear. *J Athl Train, 43*(6), 578-584. doi:10.4085/1062-6050-43.6.578

Tierney, R. T., Sitler, M. R., Swanik, C. B., Swanik, K. A., Higgins, M., & Torg, J. (2005). Gender differences in head-neck segment dynamic stabilization during head acceleration. *Med Sci Sports Exerc, 37*(2), 272-279. doi:10.1249/01.mss.0000152734.47516.aa

Tyson, A. M., Duma, S. M., & Rowson, S. (2018). Laboratory evaluation of low-cost wearable sensors for measuring head impacts in sports. *Journal of applied biomechanics, 34*(4), 320-326.

Vasavada, A. N., Li, S., & Delp, S. L. (1998). Influence of muscle morphometry and moment arms on the moment-generating capacity of human neck muscles. *Spine (Phila Pa 1976), 23*(4), 412-422. doi:10.1097/00007632-199802150-00002

Viano, D. C., Casson, I. R., & Pellman, E. J. (2007). Concussion in professional football: Biomechanics of the struck player--part 14. *Neurosurgery, 61*(2), 313-327; discussion 327-318. doi:10.1227/01.Neu.0000279969.02685.D0

Viano, D. C., & Pellman, E. J. (2005). Concussion in professional football: Biomechanics of the striking player--part 8. *Neurosurgery, 56*(2), 266-280; discussion 266-280. doi:10.1227/01.neu.0000150035.54230.3c

Waltzman, D., Sarmiento, K., Devine, O., Zhang, X., DePadilla, L., Kresnow, M.-j., et al. (2021). Head impact exposures among youth tackle and flag american football athletes. *Sports Health, 0*(0), 1941738121992324. doi:10.1177/1941738121992324

Waltzman, D., Womack, L. S., Thomas, K. E., & Sarmiento, K. (2020). Trends in emergency department visits for contact sports-related traumatic brain injuries among children - united states, 2001-2018. *MMWR Morb Mortal Wkly Rep, 69*(27), 870-874. doi:10.15585/mmwr.mm6927a4

Williams, I. M., Ponsford, J. L., Gibson, K. L., Mulhall, L. E., Curran, C. A., & Abel, L. A. (1997). Cerebral control of saccades and neuropsychological test results after head injury. *J Clin Neurosci, 4*(2), 186-196. doi:10.1016/s0967-5868(97)90072-2

Williams, W. H., Potter, S., & Ryland, H. (2010). Mild traumatic brain injury and postconcussion syndrome: A neuropsychological perspective. *J Neurol Neurosurg Psychiatry, 81*(10), 1116-1122. doi:10.1136/jnnp.2008.171298

Wong, R. H., Wong, A. K., & Bailes, J. E. (2014). Frequency, magnitude, and distribution of head impacts in pop warner football: The cumulative burden. *Clin Neurol Neurosurg, 118*, 1-4. doi:10.1016/j.clineuro.2013.11.036

Yeargin, S. W., Kingsley, P., Mensch, J. M., Mihalik, J. P., & Monsma, E. V. (2018). Anthropometrics and maturity status: A preliminary study of youth football head impact biomechanics. *Int J Psychophysiol, 132*(Pt A), 87-92. doi:10.1016/j.ijpsycho.2017.09.022

Yoganandan, N., Pintar, F. A., Zhang, J., & Baisden, J. L. (2009). Physical properties of the human head: Mass, center of gravity and moment of inertia. *J Biomech, 42*(9), 1177-1192. doi:10.1016/j.jbiomech.2009.03.029

Zhang, L., Yang, K. H., & King, A. I. (2004). A proposed injury threshold for mild traumatic brain injury. *J Biomech Eng, 126*(2), 226-236. doi:10.1115/1.1691446

Zimmerman, A. B., Lust, K. L., & Bullimore, M. A. (2011). Visual acuity and contrast sensitivity testing for sports vision. *Eye Contact Lens, 37*(3), 153-159. doi:10.1097/ICL.0b013e31820d12f4

Zuckerman, S. L., Kerr, Z. Y., Yengo-Kahn, A., Wasserman, E., Covassin, T., & Solomon, G. S. (2015). Epidemiology of sports-related concussion in ncaa athletes from 2009-2010 to 2013-2014:Incidence, recurrence, and mechanisms. *The American Journal of Sports Medicine, 43*(11), 2654-2662. doi:10.1177/0363546515599634

APPENDICES

Appendix A. Exploratory Sport Participation Survey.

Sport Participation Survey

Please fill out the following items to the best of your knowledge:

Is this your first year playing tackle football?

_____Yes   _____No

How many years have you played tackle football?

______ Years

Have you ever played flag football?

_____Yes   ______No

If yes, how many years did you play?

_______Years

Please check in the boxes for other sports you have participated in along with number of years:

☐ Flag football___Years

☐ Soccer___Years

☐ Baseball___Years

☐ Basketball___Years

☐ Swimming___Years

☐ Diving___Years

☐ Gymnastics___Years

☐ Wrestling___Years

☐ Tennis_____Years

Other__________;Years____

Appendix B. Correlation Matrix for Predictors in Research Question One and Normalized Predictors for Research Question Two.

		Player Strategy Single Task	Player Strategy Dual Task	Eye Movement Single Task	Eye Movement Dual Task	Head Turn Single Task	Head Turn Dual Task	Hand Raise Single Task	Hand Raise Dual Task	Peak Flexion	Peak Extension	Peak Left Lateral Flexion	Peak Right Lateral Flexion	Time to Peak Flexion	Time to Peak Extension	Time to Peak Left Lateral Flexion	Time to Peak Right Lateral Flexion
Player Strategy Single Task	Pearson Correlation	1	-.113	.247	.243	-.050	.002	-.112	-.464*	.174	-.125	-.356	-.232	-.037	-.238	-.242	-.342
	Sig. (2-tailed)		.583	.224	.231	.809	.992	.587	.017	.394	.543	.074	.253	.857	.242	.233	.088
Player Strategy Dual Task	Pearson Correlation	-.113	1	.302	.243	-.079	.460*	-.329	.103	.123	.040	-.152	-.150	-.215	-.075	.314	.097
	Sig. (2-tailed)	.583		.134	.232	.700	.018	.101	.616	.549	.844	.459	.464	.292	.716	.118	.637
Eye Movement Single Task	Pearson Correlation	.247	.302	1	.557**	-.056	.092	-.057	.067	.179	.072	-.150	-.206	-.408*	.189	-.029	-.274
	Sig. (2-tailed)	.224	.134		.003	.785	.654	.783	.743	.383	.725	.463	.312	.038	.355	.890	.176
Eye Movement Dual Task	Pearson Correlation	.243	.243	.557**	1	-.273	.188	-.138	-.370	-.233	.303	.051	-.055	-.328	.115	-.075	.212
	Sig. (2-tailed)	.231	.232	.003		.177	.358	.503	.063	.253	.133	.805	.789	.102	.575	.717	.297
Head Turn Single Task	Pearson Correlation	-.050	-.079	-.056	-.273	1	.028	-.135	.252	.068	-.094	-.196	-.136	-.049	.026	.191	-.267
	Sig. (2-tailed)	.809	.700	.785	.177		.891	.510	.213	.741	.646	.338	.508	.812	.899	.349	.188

Head Turn Dual Task	Pearson Correlation	.002	.460*	.092	.188	.028	1	.031	-.015	-.150	-.150	-.198	-.273	-.010	.122	.214	.065
	Sig. (2-tailed)	.992	.018	.654	.358	.891		.880	.941	.465	.466	.331	.177	.961	.552	.293	.752
Hand Raise Single Task	Pearson Correlation	-.112	-.329	-.057	-.138	-.135	.031	1	-.152	-.245	-.433*	-.289	-.301	-.147	.044	-.161	-.060
	Sig. (2-tailed)	.587	.101	.783	.503	.510	.880		.457	.227	.027	.152	.135	.472	.832	.432	.772
Hand Raise Dual Task	Pearson Correlation	-.464*	.103	.067	-.370	.252	-.015	-.152	1	.190	.186	.223	.260	.259	.058	.103	-.175
	Sig. (2-tailed)	.017	.616	.743	.063	.213	.941	.457		.354	.363	.273	.199	.201	.779	.617	.391
Peak Flexion	Pearson Correlation	.174	.123	.179	-.233	.068	-.150	-.245	.190	1	.134	.018	.071	-.294	.012	-.107	.038
	Sig. (2-tailed)	.394	.549	.383	.253	.741	.465	.227	.354		.515	.929	.729	.145	.952	.604	.852
Peak Extension	Pearson Correlation	-.125	.040	.072	.303	-.094	-.150	-.433*	.186	.134	1	.645**	.784**	.203	.314	.205	.503**
	Sig. (2-tailed)	.543	.844	.725	.133	.646	.466	.027	.363	.515		.000	.000	.319	.118	.316	.009
Peak Left Lateral Flexion	Pearson Correlation	-.356	-.152	-.150	.051	-.196	-.198	-.289	.223	.018	.645**	1	.812**	.338	.119	.097	.218
	Sig. (2-tailed)	.074	.459	.463	.805	.338	.331	.152	.273	.929	.000		.000	.091	.562	.636	.285
Peak Right Lateral Flexion	Pearson Correlation	-.232	-.150	-.206	-.055	-.136	-.273	-.301	.260	.071	.784**	.812**	1	.463*	.155	.228	.255
	Sig. (2-tailed)	.253	.464	.312	.789	.508	.177	.135	.199	.729	.000	.000		.017	.450	.263	.209

Time to Peak Flexion	Pearson Correlation	-.037	-.215	-.408*	-.328	-.049	-.010	-.147	.259	-.294	.203	.338	.463*	1	.082	-.105	-.135
	Sig. (2-tailed)	.857	.292	.038	.102	.812	.961	.472	.201	.145	.319	.091	.017		.692	.609	.512
Time to Peak Extension	Pearson Correlation	-.238	-.075	.189	.115	.026	.122	.044	.058	.012	.314	.119	.155	.082	1	-.054	.149
	Sig. (2-tailed)	.242	.716	.355	.575	.899	.552	.832	.779	.952	.118	.562	.450	.692		.793	.466
Time to Peak Left Lateral Flexion	Pearson Correlation	-.242	.314	-.029	-.075	.191	.214	-.161	.103	-.107	.205	.097	.228	-.105	-.054	1	.123
	Sig. (2-tailed)	.233	.118	.890	.717	.349	.293	.432	.617	.604	.316	.636	.263	.609	.793		.550
Time to Peak Right Lateral Flexion	Pearson Correlation	-.342	.097	-.274	.212	-.267	.065	-.060	-.175	.038	.503**	.218	.255	-.135	.149	.123	1
	Sig. (2-tailed)	.088	.637	.176	.297	.188	.752	.772	.391	.852	.009	.285	.209	.512	.466	.550	

*. Correlation is significant at the 0.05 level (2-tailed).

**. Correlation is significant at the 0.01 level (2-tailed).

Appendix C. Correlation Matrix for Research Question 3: Anthropometrics and Helmet Dimensions.

Anthropometric Correlations

		Height	Mass	Head Circumference	Neck Length	Neck Circumference	Head Mass
Height	Pearson Correlation	1	.823**	.350	.095	.603**	.823**
	Sig. (2-tailed)		.000	.080	.644	.001	.000
Mass	Pearson Correlation	.823**	1	.418*	.088	.734**	1.000**
	Sig. (2-tailed)	.000		.033	.668	.000	.000
Head Circumference	Pearson Correlation	.350	.418*	1	.200	.496**	.418*
	Sig. (2-tailed)	.080	.033		.326	.010	.034
Neck Length	Pearson Correlation	.095	.088	.200	1	-.008	.084
	Sig. (2-tailed)	.644	.668	.326		.968	.682
Neck Circumference	Pearson Correlation	.603**	.734**	.496**	-.008	1	.735**
	Sig. (2-tailed)	.001	.000	.010	.968		.000
Head Mass	Pearson Correlation	.823**	1.000**	.418*	.084	.735**	1
	Sig. (2-tailed)	.000	.000	.034	.682	.000	

**. Correlation is significant at the 0.01 level (2-tailed).

*. Correlation is significant at the 0.05 level (2-tailed).

Helmet Dimension Correlations

		Helmet Circumference	Helmet Mass	Face Mask Mass	Helmet Depth	Face Mask Length
Helmet Circumference	Pearson Correlation	1	.928**	.684**	.155	.497**
	Sig. (2-tailed)		.000	.000	.451	.010
Helmet Mass	Pearson Correlation	.928**	1	.757**	.143	.535**
	Sig. (2-tailed)	.000		.000	.486	.005
Face Mask Mass	Pearson Correlation	.684**	.757**	1	.011	.466*
	Sig. (2-tailed)	.000	.000		.956	.016
Helmet Depth	Pearson Correlation	.155	.143	.011	1	-.276
	Sig. (2-tailed)	.451	.486	.956		.172
Face Mask Length	Pearson Correlation	.497**	.535**	.466*	-.276	1
	Sig. (2-tailed)	.010	.005	.016	.172	

**. Correlation is significant at the 0.01 level (2-tailed).

*. Correlation is significant at the 0.05 level (2-tailed).