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Isometric Neck Strength In Concussed And Non-Concussed High School Football Players

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THE ROLE OF ISOMETRIC NECK STRENGTH IN PREDICTING CONCUSSIONS SUSTAINED BY HIGH SCHOOL FOOTBALL PLAYERS

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THE ROLE OF ISOMETRIC NECK STRENGTH IN PREDICTING CONCUSSIONS

SUSTAINED BY HIGH SCHOOL FOOTBALL PLAYERS

by

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ABSTRACT

Football accounts for a significant portion of the 1.275 million concussions per year. Neck strength has been theorized to possibly assist in the attenuation of severe impacts and possibly prevent football-related concussions. The purpose of this study was to investigate potential differences in isometric neck strength in flexion, extension, and lateral flexion right/left between concussed high school football players and matched non-concussed football players. Sixteen high school players that suffered a football-related concussion during the 2010 competition season comprised the research group, and were later matched to a player that did not suffer a concussion, on the basis of age, height, weight, and position of play. Measurements included: 1) maximal isometric neck strength in flexion, extension, lateral flexion right/left; 2) range of motion in the same directions; 3) neck length and girth; and 4) height and weight. A multiple analysis of variance was performed to determine if between-group differences existed in isometric neck strength. In addition, covariate analyses were performed with neck strength serving as responses, group membership as the independent variable, and body weight, neck length, neck girth, and cervical range of motion as covariate separately. Results revealed no significant differences in isometric neck strength between concussed and non-concussed players. Additionally, covariate analyses revealed no significant group differences. Differences in neck strength might not have been observable due to small sample size and subjects' age and undeveloped musculature. Future research should include eccentric neck strength, rather than isometric, in older players with electromyography. Measuring neck strength eccentrically in conjunction with electromyography would yield measurements that are more applicable to the cervical response during head impacts.

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Chapter 1

INTRODUCTION

In clinical settings, a concussion is classified and referred to as a mild-traumatic brain injury. The brain is surrounded by cerebrospinal fluid (CSF) that serves as a liquid shock-absorbing barrier between the skull and the brain. A concussion occurs when the CSF liquid barrier is overcome by either an impact or impulse force that induces rapid head acceleration resulting in forceful brain contact with the skull (Sivak, Kurca, Jancovic, Petriscak, & Kucera, 2005). An impact force is characterized by blunt force trauma resulting from the head coming into contact with an object, such as striking the back of the skull on the ground. An impulse force is characterized by rapid head movement without blunt force trauma, such as whiplash (i.e. rapid cervical flexion/extension) resulting from an automobile accident. These mechanisms of concussive injuries are most commonly referred to as an "acceleration-deceleration" of the head.

Historically, the term "concussion" originates from the Latin word *concutere* which translates to "agitation or shaking" of the brain (Maroon, Lovell, Norwig, Podell, Powell, & Hartl, 2000). The definition of concussion is a controversial issue as there is currently no universal definition available. One of the first and most widely used definitions was defined by the Congress of Neurological Surgeons in 1966. They defined a concussion as a "clinical syndrome characterized by immediate and transient post-traumatic impairment of neural functions, such as alteration of consciousness, disturbance of vision, equilibrium, etc. due to brain stem involvement" (Congress of Neurological Surgeons, 1966, p. 386). An updated

definition was proposed in 1997 by the Quality Standards Subcommittee of the American Academy of Neurology, in which they defined a concussion as an altered mental state that is not necessarily characterized by loss of consciousness but the most prevalent symptoms are amnesia and confusion. However, in 2001 the first International Conference on Concussion in Sport proposed probably the most accurate and comprehensive concussion definition currently available. They defined a concussion as "a complex pathophysiological process of affecting the brain, induced by traumatic biomechanical forces" (Aubry et al., 2002, p. 6). The lack of agreement among concussion definitions increases the difficulty for determining accurate concussion incidence rates. In addition, interpreting and comparing results from other epidemiological studies concerning concussions are also affected by varying concussion definitions.

1.1 Symptoms and Grading Scale for Concussions

A forceful brain-contact with the skull induces transient alterations in brain function that are most often characterized by alterations in one or more of the following three areas, consciousness, awareness, and cognitive function. Symptoms of concussions can include, but are not limited, to headache, dizziness, confusion, disorientation, blurred vision, amnesia, neck pain, fatigue, tinnitus, loss of balance, and difficulty performing daily activities (Guskiewicz, Weaver, Padua, & Garret, 2000). Cantu (1986) created the most widely utilized concussion grading system. The Cantu scale determines the severity of a concussion based on the presence and severity of the following symptoms, consciousness level, and amnesia. The scale ranks the severity of a concussion on a scale from grade I to grade III. A grade I concussion is

characterized by temporary altered mental status without loss of consciousness or less than 30 minutes of amnesia. A grade II concussion is characterized by temporary mental status with more severe concussion-related symptoms than a grade I or lasting longer, loss of consciousness for five minutes or less, and 30 minutes to 24 hours of amnesia. A grade III concussion is characterized by loss of consciousness for five or more minutes and amnesia for more than 24 hours.

1.2 Incidence Rates of Concussion

According to the Centers for Disease and Control, concussions account for 75% of the 1.5 million brain injuries each year in the United States (Gerberding, 2003). The World Health Organization (WHO) found the rate of medical attention for concussions was 1-3 per 1000 people per year; however, the true incidence rate of concussions is estimated to be more than 6 per 1000 people per year. This doubled estimated rate is due to the WHO estimates that 50% of people who sustain a concussion do not seek medical attention. Supporting the WHO's estimation of underreporting, Iverson (2005) reported that at least 25% of people who sustain a concussion do not seek medical attention, and therefore, their injuries are not diagnosed or reported. This contributes to difficulty in determining an accurate concussion incidence rate. Furthermore, individuals often do not recognize their concussion-related symptoms, which contribute to their failure to seek medical attention (Delaney, Lacroix, Leclerc, & Johnston, 2002). In the case of sport-related concussions, it is theorized that athletes often minimize their concussion symptoms due to motivation to return to participation and competition. Widely varying definitions of concussions, failure to recognize concussion-related symptoms

and seek medical attention, and motivation to return to competition all contribute to under-reporting of concussions, which make it difficult to determine accurate concussion incidence rates.

1.3 Concussion in Sports

Concussions are much more prevalent in certain special populations, such as elderly, youth, and athletes (Ropper & Gorson, 2007). These findings are not surprising given the most common causes of concussions are falls, motor vehicle accidents, bicycle accidents, and sport injuries (Ropper & Gorson, 2007). Young children and young adults account for the highest incidence rates and number of concussions each year, respectively (Roper & Gorson, 2007; Ryan & Warden, 2003).

Sport-related concussions are a serious issue for young children and adults as these demographics account for the vast majority of both amateur and professional athletes. In fact, sport-related concussions are the second leading cause of traumatic brain injuries, after automobile accidents, in young people age 15 to 24 years (Gessel, Fields, Collins, Dick, & Comstock, 2007). Furthermore, Gessel, Fields, Collins, Dick, and Comstock (2007) found that 8.9% and 5.8% of all sport-related injuries were concussions at the high school and collegiate levels, respectively. Herring et al. (2005) found that concussions account for five percent of all sports injuries. The Centers for Disease and Control (CDC) conservatively estimates the number of sport-related traumatic brain injuries at 300,000 per year with the majority of those injuries classified as concussion. However, they only reported traumatic brain injuries that resulted in loss of consciousness. It is hypothesized that the CDC's estimate of sport-related concussions is

grossly underestimated, due to the fact that only 8.9-10.0% of concussions result in loss of consciousness (Guskiewicz et al., 2000; Cantu, 1998).

Some of the most common sports for concussions are boxing, ice hockey, rugby, soccer, and football due to their objectives and characteristics (Covassin, Swanik, & Sachs, 2003; Marshall & Spencer, 2001). Boxing is characterized by many violent impacts to the non-protected head throughout the duration of a bout. Boxing is a very common sport for concussion occurrence as often the objective is to win by inflicting a concussion, most notably referred to as “knock-out” (i.e. KO). Although only a small portion of concussions actually result in loss of consciousness, the opposing boxer most often remains unable to continue participating in the bout after sustaining a concussion.

Ice hockey is another contact sport characterized by players violently colliding into each other while skating on ice. Professional ice hockey players are well protected with safety equipment but they still have a five percent chance of sustaining a concussion each season (Tegner & Lorentzon, 1996). A common technique in ice hockey is the “check”, in which a player lowers his shoulder and collides with an opponent. This movement often results in an opposing player’s head coming into contact with the ice, wall, or another player, and therefore, increases their likelihood of sustaining a concussion.

Rugby, another contact sport, is characterized by players forcefully colliding into each other while both in the open field and in the scrum, ruck, and maul. The risk of injuries in rugby is exacerbated by the fact that rugby players, unlike football players, wear very little protective gear. In fact, most players are only protected by a soft-shell head gear and mouthpiece.

Marshall and Spencer (2001) followed two high school rugby teams for three years and found concussions accounted for 24.6% of the 69 reported injuries. They further found the incidence rate of concussion for this cohort was 3.8 per 1000 athlete-exposure. However, they reported that underreporting of concussions likely occurred due to players' three-week exclusion of competition and practice after a diagnosed brain injury, which may have contributed to their unwillingness to seek medical attention. In addition, due to administrative and logistical reasons, medical attention is not always readily available at the time of injury.

The sport of soccer is not as aggressive as the aforementioned sports but concussions regularly occur. Covassin, Swanik, and Sachs (2003) reported concussion incidence rates of 7.0% and 11.4% in male and female collegiate soccer players, respectively. Conversely, Shankar, Fields, Collins, Dick, and Comstock (2007) reported much higher incidence rates at the high school and collegiate levels as they found male and female soccer players accounted for 15.4% and 21.5% of all sport-related concussions, respectively. Gesser et al. (2007) found that 'heading' the ball accounted for 40.5% and 36.7% of soccer-related concussions at the high school and collegiate levels, respectively. In addition, females were found to more likely sustain a concussion from an impact from the ground or the soccer ball than males (Gesser et al., 2007). These concussions often result from players jumping up to 'head' the ball and receiving a violent impact from another player's head, shoulder, elbow, or forearm. Another mechanism of soccer-related concussions is players striking their head on the ground after challenging a defender for the ball in the air and losing their ability to land on their feet.

1.4 Concussion in Football

No other sport has received more attention for concussions than American football. American football has garnered such attention due to high profile professional and collegiate players sustaining a concussion resulting in loss of playing time and, in the case of professional football, loss of income. Fields et al. (2007) found that football accounted for 40.5% of concussions at the collegiate and high school levels. Powell and Barber-Foss (1999) found that football accounted for 63% of sport-related concussions at the high school level. The sport of American football accounts for a large majority of sports-related concussions due to the characteristics of the sport and the number of subjects at all levels.

Football is characterized by violent collisions between multiple players moving at a high velocities and accelerations with the majority of impacts occurring at the facemask or shell of the struck player's helmet. Impacts are most often characterized by a player striking another player with his facemask or the front portion of the crown of the helmet. These forceful helmet impacts involve large linear accelerations that induce a great change in head velocity. Contributing to violent impacts, striking players align their head, neck, and torso while driving through the tackle or block in an effort to translate as much force and kinetic energy to the struck player. In fact, Guskiewicz et al. (2000) found that 80.5% of head injuries occurred from contact with another player (i.e. teammate or opponent) while only 10.0% occurred from head contact with the ground, which illustrates the characteristics of the sport.

Gerberich, Priest, Boen, Straub, and Maxwell (1983) reported the first concussion incidence rates for high school football players. These authors reported an incidence rate of

24% for the 1977 season and they concluded that each year 20% of players would sustain a concussion. In retrospect, their findings might have been a gross-overestimation of concussion rates. However, changes over the past 27 years, such as improvements safety equipment (i.e. helmet, chinstrap, and mouthpiece), rule changes (banishment of "spear-tackling"), as well as year-long strength and conditioning programs, have undoubtedly contributed to a reduction in concussion incidence rates. The findings of Gerbich et al. (1983) served as a catalyst for research studies and publications concerning concussion in football at all levels.

Professional football players represent the smallest cohort of football players but they experience the highest incidence rate of concussion. In the NFL a 7.7% incidence concussion rate has been reported (Pellman et al., 2004), which yields to around 130 concussions per year for a total of 1700 players. Guskiewicz et al. (2003) reported a concussion incidence rate of 8% in collegiate athletes. With an estimated 68,000 collegiate players, this yields to about 5,440 football-related concussions in this demographic per year. Similarly, Covassin, Swanik, and Sachs (2003) reported a concussion incidence rate of 8.8% for collegiate players and purported concussion rate are continually rising. On the other hand, Guskiewicz, Weaver, Padua, and Garret (2000) reported much lower concussion incidence rates, 4.4%, 4.5% and 5.5%, 5.6% for division I, II, III, and high school, respectively, with an overall concussion rate of 5.1% for both collegiate and high school players.

High school football represents the largest cohort of football players and magnitude of concussions. There are an estimated 1.2 million high school football players and with similar incidence rates of concussion as collegiate players (5.6%) the estimated the number of

concussions sustained by these players is likely between 43,200 and 67,200 each year (Guskiewicz, Weaver, Padua, & Garret, 2000). Concussions sustained by this young cohort of football players are exacerbated by their undeveloped musculature and nervous system, which could lead to irreparable damage.

Concussion occurrences differ by position of play, severity, and session of play. Guskiewicz et al. (2000) reported offensive linemen, defensive backs, and linebackers were more likely to sustain a grade I concussion; however, wide receivers and special team's players were more likely to sustain a more severe concussion (i.e. grade II or III) and concussions were more likely to occur during a game rather than practice or scrimmage. Similarly, Powell and Barber-Foss (1999) reported the positions with the highest incidence rates of concussions were linebackers (14.3%), running backs (14.0%), and offensive linemen (13.4%). In addition, these authors reported that players were eleven times more likely to sustain a concussion during competition rather than practice. Covassin, Swanik, and Sachs (2003) supported the findings of Powell et al. (1999); reporting the competition-concussion incidence rate was eleven times higher than practice.

1.5 Return to Play Considerations

Once a player has sustained a concussion, an accurate and timely diagnosis is relegated either to a team's physician or athletic trainer. At the high school and middle school levels of football, a concussion diagnosis is delegated to the athletic trainer roughly 12% of the time for lack of an on-site physician (Guskiewicz et al., 2000). As aforementioned, concussions are severely underreported, in part, due players not recognizing their concussion-related symptoms

and seeking medical treatment right after the severe impact. Often players return to contact after only minutes of sustaining a concussion. Guskiewicz, Weaver, Padua, and Garret (2000) found 30.8% of concussed high school and collegiate players returned to participation on the same day of injury with an average “sit-out” time of only 13 minutes.

The current medical recommendation for abstinence from contact participation is at least one week after sustaining a concussion and the player must be asymptomatic (Cantu, 1988). Guskiewicz et al. (2000) found the average time after injury before returning to participation for the other two-thirds of players who did not return to participation on the same day of injury was only 4.2 days, well below the current medical recommendation of one week before return to participation. In addition, Powell and Barber-Foss (1999) found the median rest time after sustaining a concussion was only three days. However, the majorities (89%) of concussed players were removed from participation for at least that day and 54% were referred to medical attention.

A player that has sustained a concussion but has not received adequate rest time is at risk for the often fatal second-impact syndrome (Cantu, 1998). Second-impact syndrome (SIS) occurs when a previously concussed person receives a second concussion without sufficiently recovering from the first. This syndrome is characterized by fatal cerebral edema (Cantu, 1998).

The hypothesized underlying mechanism for SIS induced cerebral edema is failure of the cerebral arterioles’ ability to constrict resulting in excessive cerebral arteriole dilation causing excessive cerebral blood flow (Cantu, 1998). Another second-impact consideration, concussed players are three times more likely to sustain another concussion in the same season, as

compared to non-concussed players (Guskiewicz et al., 2000). Second-impact syndrome necessitates the need for an accurate diagnosis and treatment of a primary concussion and prohibition of contact before returning to competition less a player receives a secondary concussion induced by a premature return to contact.

The associated symptoms, alarming number of concussions sustained by football players, and possibility of second-impact syndrome has prompted researchers to investigate the variables associated with concussions. Understanding the underlying biomechanics of concussive impacts was the first concern of researchers in their objective of reducing the occurrence of concussions.

Chapter 2

LITERATURE REVIEW

2.1 Biomechanics of Concussive Impacts

Numerous studies have investigated the underlying biomechanical mechanisms of concussive impacts. The two most common methods of collecting biomechanical data of head impacts is collecting real-time data by utilizing the head impact telemetry system (HITS) (Simbex LLC, Lebanon, NH) or laboratory reconstructions of concussive impacts utilizing Hybrid III dummies (crash test dummies; First Technology Safety Systems, Plymouth, MI).

The HITS is comprised of two components, an encoder unit that fits inside the player's helmet and a sideline computer. The encoder unit is further comprised of four components, six single-axis accelerometers, telemetry unit, storage device, and battery pack. The accelerometers record accelerations in their respective orientation then send the head impact information to the telemetry unit, which further sends it to the sideline computer, or if the sideline computer is out of range, information is stored in the data storage unit until the sideline computer is within range.

From the head impact data from each accelerometer, resultant linear acceleration, angular acceleration, and location of impact are determined and reported by the automated features in the HITS software. The HITS has been validated against Hybrid III crash dummies with a correlation of $r = .98$, a 4% error rate when estimating resultant linear and angular accelerations, and location of impact within ± 0.41 cm (Duma et al., 2005). The HITS has been

widely utilized to capture head real-time head impact data at the high school and collegiate levels (Broglio, Sosnoff, Shin, He, Alcaraz, & Zimmerman, 2009; Mihalik, Bell, Marshall, & Guskiewicz, 2007; Brolinson, Manoogian, McNeely, Goforth, Greenwald, & Duma, 2006; Duma et al., 2005). The advantage of utilizing this system is that real-time head impact data is collected, but a major limitation is the lack of availability due to high cost of the HITS. In addition, in order to collect meaningful data, thousands of impacts must be recorded, which translates to many players being outfitted with the system. In addition, it is difficult to collect multiple concussive head impact data because of their rare occurrence on a single team.

The HITS collects real-time head impact data throughout a course of a season for a group of outfitted players, whereas, utilizing laboratory reconstructions with Hybrid III dummies collects retrograde head impact data of concussive impacts. Newman, Shewchenko, Beusenberg, Fournier, Withnall, and Barr (2005) described the validation studies for the use of football helmeted Hybrid III test dummies to ascertain mild traumatic head injury impact data and reported error analyses for this method.

Film records (most often competition film) of concussive impacts are reviewed from multiple angles in order to determine the velocities, accelerations, head kinematics, and location of impact for both the concussed and striking players. Once these variables are determined, the impact is then reconstructed on helmeted Hybrid III dummy that is fitted with nine internal head accelerometers in a 3-2-2-2 pattern. The Hybrid III dummy software yields linear and rotational accelerations. Pellman et al. (2003) reported that potential errors in translational accelerations did not exceed 2.2%; however, errors in rotational accelerations

could be as high as 6.7%. Error rate could be much higher if head kinematics of involved players were not accurately determined from film footage. This laboratory reconstruction method has been widely used by researchers to determine the underlying biomechanics of concussive head impacts (Viano, Casson, & Pellman, 2007; Pellman et al., 2003; Newman et al., 1999).

Biomechanical variables of interest when investigating concussive impacts are location of impact, linear and rotational accelerations, and change in head velocity. The location of impact is important when characterizing head impacts because it is possible that impacts to certain helmet locations (i.e. top vs. front) might increase the likelihood of sustaining a concussion. Linear head acceleration is the change in head velocities directly before and after contact with the mechanism of injury (i.e. ground, opposing player, etc.). Linear head acceleration is arguably the most significant predictor for sustaining a concussion and many researchers attempt to quantify thresholds for concussions based on linear acceleration (Pellman et al., 2003). Rotational acceleration is defined as the change in angular head velocities before and after contact with the mechanism of injury. Rotational acceleration is an important variable of interest to quantify when discussing biomechanics of concussive impacts; however, it is a debatable issue whether rotational acceleration is a significant predictor of sustaining a concussion.

Change in head velocity refers to change in head velocity *after* a player has come into contact with the mechanism of injury. Researchers have theorized that change in head velocity is possibly a more significant predictor of concussion than linear acceleration (Viano, Casson, &

Pellman, 2007). A large change in head velocity contributes to a large change in head displacement, which increases the time and space the brain has to sustain a concussion.

In summary, biomechanical investigations of concussive impacts are most often investigated by the utilizing the two aforementioned methods. In order to collect retrospective impact data, researchers reconstruct concussive impacts utilizing helmeted Hybrid III crash test dummies. In order to collect real-time impact data, researchers implant accelerometers and a telemetry unit (HITS) into player's helmets. Both methods generate a greater level of understanding concerning concussive impacts; however, both are limited by their availability due to their cost. Biomechanical variables of interest for concussive impacts are impact location, linear and rotational accelerations, and change in head velocity. Quantifying the aforementioned impact variables are important when examining the underlying biomechanics of concussive impacts; therefore, they are discussed in detail in the following sections.

Location of impact.

Location of impact refers to the whether a concussed player was struck on the front, back, top, or side of the helmet. Defining the orientation of each helmet location is important for understanding how the location of impact affects the likelihood of sustaining a concussion. There is a high level of agreement in defining helmet location in reviewed literature. Generally, the front of the helmet most often refers to the facemask area while the back of the helmet refers to the portion of the helmet shell that covers the posterior portion of the head. The sides of the helmet are the portions of the helmet shell that are between the front and back

locations. Lastly, the top of the helmet is most often designated as the helmet shell that is above the facemask area.

The location of impact is important for understanding the biomechanics of head impacts experienced at different levels of play, high school, collegiate, and professional. Frequency of impacts and location of head impact differs according to position and level of play (Broglio et al., 2009; Mihalik et al., 2007). While characterizing location of head impact by position is beyond the scope of this review, considering level of play is imperative.

Mihalik et al. (2007) found 57% of their recorded concussions occurred from top impacts. This statistic is not surprising given that Broglio et al. (2009) reported that top impacts generate the highest linear accelerations and Pellman et al. (2003) found that the most correlated variable with concussion is linear acceleration. Schnebel et al. (2007) and Broglio et al. (2010) reported their highest recorded linear accelerations (145.7 g and 146 g) that induced a concussion occurred at the top of the helmet. However, Broglio et al. (2010) the majority (62%) of their recorded concussive impacts occurred at the front of the helmet despite reporting their highest linear accelerations of concussive impacts occurred at the top of the helmet. This possibly suggests a lower threshold for sustaining a concussion from a front impact vs. a top impact.

The fact that a significant proportion of concussions occur at the front of the helmet is significant because high school-impacts to the front of the helmet occur 1.79, 2.88, and 3.32 times more frequently than impacts to the back, side, and top of the helmet, respectively (Broglio et al., 2009). At the collegiate level, players experienced 10% less frequent impacts to

the front of the helmet and more frequent impacts to the top and back of the than high school players (Mihalik, Bell, Marshall, and Guskiewicz, 2007). The aforementioned findings suggest that collegiate players sustain impacts with greater linear accelerations more frequently from top impacts, but high school players are more likely to sustain a concussion due to the increased frequency of front impacts. The fact that collegiate players experience smaller incidence rates of concussions than high school players supports this theory (Guskiewicz et al., 2000).

In order to quantify location of concussive impacts at the professional level, Pellman, Viano, Tucker, Casson, & Waeckerle (2003) analyzed concussive impacts in NFL game films and found that only 29% of the 174 analyzed concussive impacts involved loading of the facemask with the remainder occurring at the helmet shell. This finding, in conjunction with Mihalik et al. (2007), supports the notion that as level of competition increases facemask impacts decrease. This suggests that as level of competition increases struck players are more adept at preventing impacts to their facemask, which might account for the reduction in concussion incidence rates from high school to college. In addition, these authors found that 61% of concussive impacts originated from the striking player’s helmet with the remainder originating from impacts from shoulder pads, arms, or the ground. These authors concluded that concussions were, in part, generally characterized by impacts to the facemask at an oblique or lateral angle by an opposing player’s helmet, impacts from an opposing player’s body regions (i.e. shoulder pads, knee), and ground impacts to the back of the head (Pellman et al., 2003). Further investigation is needed to determine if impacts from a striking player’s helmet generate higher linear

accelerations than impacts sustained from ground contact or an opposing player's body (i.e. shoulder, forearm, etc).

Although impacts to the back of the helmet are rare, as compared to the frequency of other impact locations, they must also be taken into consideration. Schnebel et al. (2007) and Broglio et al. (2010) reported concussive impacts with their lowest linear accelerations (81 g and 81.5 g, respectively), occurred at the back of the helmet. Similar to front impacts, it is possible that less linear acceleration is needed to induce a concussion when the impact occurs to the back of the helmet and higher linear accelerations are needed to induce concussions when impacts occur at top of the helmet. Further research is warranted to investigate the theory of altered linear acceleration concussion thresholds based on impact location.

In summary, concussive impacts most often occur at the front and then the top of the helmet. High school players experience the greatest frequency of impacts to the front of the helmet while collegiate players experience less frequent impacts to the front but more frequent impacts to the top and back of the helmet. Meaning, collegiate players sustain greater frequency of impacts with very large linear accelerations, but high school players sustain the largest frequency of frontal impacts, which induce the most concussions. Therefore, high school players, as compared to collegiate players, might be at an increased risk for sustaining a concussion because of the increased frequency of front impacts, but the difference might not be significant. In addition, lower linear accelerations might be needed to induce concussion at the back and front of the helmet than those needed to induce concussion at the top of the

helmet. Lastly, more research is needed to validate claims of an impact-location and concussion interaction effect.

Linear acceleration.

Linear acceleration is the quotient of change in head velocity and impact duration. Linear acceleration of impacts in football is dependent on the velocities of the striking and struck players before and during impact. An increase in either player's velocity before and during contact will elicit an increase in linear acceleration at the head's center of gravity. Quantifying linear acceleration of concussive impacts is critical as linear acceleration is the most significant predictor of a concussion in regression analysis (Pellman et al., 2003). Pellman et al. (2003) reported linear acceleration was the most correlated impact variable with sustaining a concussion ($R = 0.71$). Researchers reconstructed 31 concussive impacts sustained by NFL players in order to quantify concussive linear accelerations of head impacts at the professional level (Pellman et al., 2003). They found mean linear acceleration of reconstructed concussive head impacts were 98 ± 28 g, which were significantly greater than linear accelerations of impacts suffered by non-concussed struck players, 60 ± 24 g. Linear acceleration differs according to level of play, position of play, and location of impact.

Linear acceleration and level of play.

In order to quantify non-concussive linear accelerations of head impacts at the high school level, Broglio et al. (2009) utilized the Head Impact Telemetry System (HITS) to record real-time head impact data throughout an entire football season (i.e. preseason games, practice, games) in 35 high school varsity football players. Their variables of interest were

linear acceleration, impact duration, impact force, jerk, and impulse. Impact duration is the time, most often reported in milliseconds, of force application to a player by an opposing player. Impact force is the product of linear acceleration at the head's center of gravity and the estimated mass of a player's head. Jerk is the change in head acceleration divided by the time to peak head acceleration. Impact is the area under the linear head acceleration curve from initiation of impact to cessation of impact. All impact variables increased in severity for games, as compared to practice, and all impact variables differed significantly by position (offensive skill, defensive skill, offensive line, and defensive line). These authors reported an average linear acceleration of 24.76 ± 15.72 g.

Duma et al. (2005) collected impact data utilizing HITS on 38 collegiate players throughout a full season (35 practices and 10 games). They analyzed 3312 impacts and reported an average resultant linear acceleration of 32 ± 25 g. Duma et al. (2005) reported much higher linear accelerations than those reported by Broglio et al. (2009). Observed differences were most likely attributed to Duma et al. (2005) collected impact data on collegiate players, whereas Broglio et al. (2009) collected impact data on high school players. Collegiate players, on average, are heavier and have a developed musculature, which contributes to their ability to deliver impacts with higher linear accelerations than their lighter, weaker and undeveloped counterparts. In addition, observed differences in linear acceleration could be due to the number of collected impacts.

Broglio et al. (2009) collected a total of 19,224 impacts, whereas, Duma et al. (2005) collected only 3,312 impacts. Both studies reported that distribution of impacts was positively

skewed. Therefore, it is reasonable to assume the more number of collected impacts would lead to a much lower average linear acceleration values. In line with this reasoning, Brolinson, Manoogian, McNeely, Goforth, Greenwald, and Duma (2006) collected substantially more impacts, 11,604, and reported a lower mean resultant linear acceleration, 20.9 ± 18.7 g, than those reported by Duma et al. (2005). In addition, Mihalik, Bell, Marshall, and Guskiewicz (2007) collected 82,026 impacts on collegiate players and found average linear acceleration was between 21-23 g. According to the aforementioned studies, it would appear that the number of collected impacts affects average linear acceleration more than the level of play; however, the non-normal distribution of linear acceleration impacts prevents any such generalization.

In order to account for observed non-normal distribution of impacts, Schnebel, Gwin, Anderson, and Gatlin (2007) compared the top 1%, 2%, and 5% of all impacts between a high school and collegiate team based on linear acceleration. Researchers recorded 54,154 collegiate-impacts and 8,326 high school-impacts, and as anticipated 79.4% of all impacts were less than 30 g generating a non-normal distribution. Collegiate players sustained impacts with higher mean linear acceleration for the top 1%, 2%, and 5% of all impacts, as compared to high school players. In addition, collegiate players sustained impacts >60 g and >98 g more frequently (4.7% and 1.0%) than high school players (3.5% and 0.7%). Researchers chose to utilize the 60 g and 98 g thresholds because they represent the 90th percentile of impacts and the level at which 75% of impacts are expected to result in a concussion, respectively (Pellman et al., 2003).

Duma et al. (2005) reported much higher frequency of >60 g (11%) collegiate-impacts than Schnebel et al. (2007) (4.7%). Frequency differences in >60 g impacts between the two aforementioned studies are most likely due to number of collected impacts and the non-normal distribution of impacts. The incidence rates of >60 g and >98 g impacts reported by Schnebel et al. (2007) support the similar incidence rate of severe impacts reported by Brolinson et al. (2006). Brolinson et al. (2006) utilized a 75 g threshold to quantify the incidence rate of severe collegiate-impacts, and found 2.5% of impacts were greater than this threshold. Their 2.5% (>75 g) incidence rate of severe impacts falls between the 4.7% (>60 g) and 1.0% (>98 g) incidence rates reported by Schnebel et al. (2007).

In summary, mean linear acceleration does not seem to differ between high school and collegiate levels; however, collegiate players suffer more frequent severe impacts than high school players when the non-normal distribution of impacts is taken into consideration. The aforementioned findings of this section would lead to the logical conclusion that collegiate players would experience greater concussion incidence rates than high school players because they greater frequency of high linear acceleration impacts (>98 g), but high school players usually experience greater concussion incidence rates. Impact location, frequency of severe impacts, age, and musculature development could account for some of the observed differences between high school and collegiate players' concussion incidence rates.

Linear acceleration and position of play.

Linear acceleration differs on the basis of position of play, or the position that a player plays the most during games. Players are most often categorized into two categories, skill

players and linemen. The positions that fall under the skill player category are wide receivers, quarterbacks, running backs, defensive backs, etc. The positions that fall under the linemen category are offensive (center, guard and tackle) and defensive (defensive end and defensive tackle) linemen.

Linemen experience the greatest number of impacts (57.5% - 76%) but the majority of these impacts are characterized by low linear accelerations (20-25 g) (Mihalik et al., 2007; Schnebel et al., 2007). Conversely, skill players experience the least number of impacts (24% - 42.5%) but sustain the most severe impacts (>98 g) (Mihalik et al., 2007; Schnebel et al., 2007). In fact, skill players sustained an impact >98 g once out of every 70 impacts, whereas linemen sustained an impact >98 g once out of every 125 impacts (Schnebel et al, 2007). To support this, offensive skill players sustained the greatest frequency of >80 g impacts than any other defensive skill players and linemen (Mihalik et al., 2007). It stands to reason that offensive skill players would receive the greatest frequency of severe impacts as they predominantly possess the ball and the goal is for defenders to tackle them at high velocities, in order to thwart the offensive player's progress.

Linear acceleration and concussion.

As linear acceleration is the most correlated variable with sustaining a concussion, it is important to provide examples of linear accelerations of concussive impacts. Broglio et al. (2010) reported an average linear acceleration of 105 g for 13 concussive impacts sustained by high school players. Guskiewicz et al. (2007) reported an average linear acceleration of 102.77 g for 13 concussive impacts in collegiate football players. Duma et al. (2005) reported a single

concussive event characterized by a linear acceleration of 81 g; however, 11% of all recorded impacts were >60 g but only one concussion occurred. Similarly, Brolinson et al. (2006) collected impact data for three concussive events with linear accelerations of 55.7 g, 117.6 g, and 136.7 g, but 99% of impacts that were >75 g did not result in concussion. In addition, Mihalik et al. (2007) reported seven concussive impacts and only 0.35% of those impacts >80 g resulted in concussion.

In summary, linear acceleration is the most correlated variable with sustaining a concussion, but it is unclear why a head impact with a high linear acceleration does not induce a concussion while low acceleration impacts induces a concussion. Sustaining a concussion is likely brought on by the interaction of several impact variables that are within an optimum range to induce a concussion.

Severity Index and Head Injury Criteria.

Linear acceleration is not only utilized to predict a concussion in regression analysis but is also used to determine the severity of an impact and the risk of head injury. Severity Index (SI) and Head Injury Criterion (HIC) are two such measures of impact severity that are dependent on linear acceleration at the head's center of gravity and duration of acceleration. The Gadd (1966) method for computing SI is as follows,

$$SI = Ta(t)^{2.5}dt$$

where T is the time duration, a(t) is the resultant translational acceleration at the head center of gravity, d is displacement of the head's center of gravity, and t is the duration of

impact. The National Operating Committee on Standards for Athletic Equipment (NOCSAE) establishes safety standards for athletic equipment. NOCSAE utilizes SI to determine if equipment (i.e. helmet, shoulder pads, etc.) is safe for athletic use. For example, a helmet is rejected if the helmet's shell becomes compromised (i.e. cracks) after an impact equal to an SI of 1200 during NOCSAE testing. NOCSAE adopting impact severity standards (i.e. SI and HIC) has forced athletic equipment manufactures to adhere to these standards. In addition to NOCSAE, researchers often report SI values in order to quantify the severity of head impacts at all levels both during laboratory reconstructions and real-time data collection.

The other commonly used index for impact severity is HIC, which is a variation of SI. The equation for computing HIC is as follows,

$$HIC = \{[t_2 - t_1][a(t)dt/(t_2 - t_1)]^{2.5}\}_{max}$$

where a(t) is the resultant translational acceleration at the head's center of gravity and dt is the product of head displacement (d) at the head's center of gravity and duration of impact (Henn, 1998). In order to determine the highest value ($_{max}$) of HIC, 15 milliseconds is the most often utilized time interval (i.e. duration of impact; $t_2 - t_1$) for head impacts sustained during football. HIC is utilized by the National Highway Traffic Safety Administration (NHTSA) since 1975 to measure the likelihood of sustaining a head injury during automobile accidents and evaluate the effectiveness of vehicle safety features (i.e. airbags, seatbelts, etc.).

The NHTSA limits HIC to 700 for airbag during testing. HIC scores of 700 and 1000 represents a five percent and eighteen percent chance of sustaining a severe head injury (Prasad & Mertz, 1985). HIC is also commonly utilized by researchers to evaluate the

effectiveness of personal protective gear and sports equipment. In concussion-based research, HIC is commonly reported as another measure of impact severity for both concussive and non-concussive impacts.

Models for concussion prediction.

Pellman et al. (2003) found the three variables most correlated with concussion occurrence were linear acceleration, SI, and HIC. Their proposed thresholds were a linear acceleration of 70-75 g, SI $\geq$ 300, and HIC $\geq$ 250. These proposed concussion thresholds are an area of controversy among researchers.

The findings of Guskiewicz et al. (2007) support the proposed concussion linear acceleration threshold as they reported an average linear acceleration of 102.77 g for concussive impacts in 13 collegiate football players. On the other hand, Duma et al. (2005) were only able to collect data on a single concussive event but the concussive impact was characterized by a resultant linear acceleration of 81 g and HIC = 200. The linear acceleration of the reported concussive impact agrees with the proposed threshold for a concussion of 70-75 g by Pellman et al. (2003), however, they only collected impact data for one concussion. The HIC value of their reported concussion does not agree with the concussion tolerance level of HIC = 250. More importantly, Duma et al. (2005) reported 55 impacts over the Pellman et al. (2003) proposed tolerance HIC value of 250 with no concussions occurring. Similarly, Mihalik et al. (2007) collected data on seven concussive impacts and only 0.35% of impacts greater than 80 g resulted in concussion. In addition, Brolinson et al. (2006) collected impact data for three

concussive events with average translational accelerations of 55.7 g, 117.6 g, and 136.7 g; however, 288 of their recorded impacts were over 75 g but did not result in concussion.

As aforementioned, a concussion is likely induced by the interaction of several impact variables within an optimum range, which prompted new multiple linear regression models for predicting a concussion that incorporated impact variables that were previously disregarded.

Broglio et al. (2010) proposed their own concussion model, integrating rotational acceleration as well as linear acceleration, and location of impact. Seventy-eight high school players from the 2005-2008 seasons were outfitted with HITS and researchers recorded 54,247 impacts with 13 impacts resulting in concussion. They suggested a concussion model with rotational acceleration, linear acceleration, and location of impact (i.e. front, top, and back) as the best predictors of a concussion occurring. Their thresholds were a rotational acceleration >5,582.3 rad/s^2, linear acceleration > 96.1 g, and frontal, top, or back helmet impact. Their thresholds for high school players are similar to those proposed for collegiate and professional players leading to their conclusion that all players are at equal risk for concussion regardless of level of play. They further theorized that medical personnel can expect to diagnose one out of every five players that experience breaches of these proposed concussion thresholds with a concussion when players are outfitted with the HITS.

The concussion model and proposed thresholds by Broglio et al. (2010) agree with the thresholds proposed by Pellman et al. (2003) and seem to be a step in the right direction for accurately determining all concussions, but the feasibility for many teams to utilize HITS is still very far away. However, with so many discrepancies in the literature concerning linear

acceleration and impact severity scores of concussions, the widely-cited proposed thresholds set by Pellman et al. (2003) should be viewed as general guidelines for understanding the underlying biomechanics of a concussion, more specifically linear acceleration, rather than concrete standards.

In summary, concussive impacts are characterized by large linear accelerations that result in high HIC and SI scores and most often occur at the top and front of the helmet. Sustaining a concussion is most likely dependent on the interaction between location of impact and linear acceleration. For instance, impacts to the top and front of the helmet might require much larger linear accelerations to induce concussion as compared to impacts to the back of the head. Researchers have proposed concussion thresholds with location of impact, linear acceleration, and rotational acceleration as covariates; however, their proposed models are an area of controversy in the literature. More research is required in order to accurately predict if a player has sustained a concussion after suffering a severe impact.

Rotational acceleration.

Linear acceleration was discussed at length because it is the most correlated variable with sustaining a concussion ($R = 0.72$); however, discussing rotational is warranted as all concussive impacts are characterized by both their linear and rotational accelerations. The association between rotational acceleration and sustaining a concussive impacts is $R = 0.68$. Rotational acceleration is defined as the quotient of the change in head velocity around either the z (cervical rotation right/left), x (cervical flexion/extension), or y axis (cervical lateral flexion right/left) and impact duration.

Duma et al. (2005) reported average rotational accelerations of non-concussive impacts about the x and y axes as 905 ± 1075 rad/s^2 and 2020 ± 2042 rad/s^2, respectively. Broglio et al. (2009) reported resultant rotational head accelerations for competition and practice non-concussive impacts and reported they were 1670 ± 1249 rad/s^2 and 1469 ± 1055 rad/s^2, respectively. Similar to linear acceleration, average rotational accelerations differ according to level of play, position of play, and session-type (Broglio et al., 2009); however sample size seems to affect average rotational acceleration more than the aforementioned variables. The distribution of rotational accelerations of non-concussive impacts is skewed to the left, which limits generalization of average rotational accelerations results of non-concussive impacts.

Quantifying rotational accelerations of non-concussive impacts is important for comparison to concussive impacts. Broglio et al. (2010) reported an average resultant rotational acceleration of 7230 ± 1158 rad/s^2 for 13 concussive impacts sustained by high school football players with an impact duration of 9.7 ± 2.6 ms. This statistic illustrates that concussive impacts are characterized by very high rotational accelerations acting over a short duration. Pellman et al. (2003) reported an average peak rotational acceleration of 6432 ± 1813 rad/s^2 for reconstructed concussive impacts sustained by professional football players using helmeted Hybrid III crash test dummies. This rotational acceleration is greater than the average rotational acceleration observed in the striking player, 4255 ± 1405 rad/s^2.

Observed differences between Pellman et al. (2003) and Broglio et al. (2010) might be attributed to level of play or age, but an alternative theory for the observed differences is the crash test dummy neck stiffness was standardized during reconstruction, which masks the role

that neck strength and stiffness might play in severe impacts. This raises the issue of neck strength and passive stiffness in reducing accelerations suffered during concussive impacts. Viano et al. (2007) reported that as passive neck stiffness increased, exponential reductions in head displacement were observed; indicating a reduction in both linear and rotational accelerations.

In summary, concussive impacts are characterized by both high linear and rotational accelerations. However, because linear acceleration experiences a greater association with concussive impacts than rotational acceleration, rotational acceleration is not always reported in biomechanical investigations of concussive impacts by researchers. Rotational acceleration of non-concussive impacts differs according to sample size, level of play, session-type, and position of play. During the mechanism of injury, the struck player sustains greater rotational acceleration than the striking player. Lastly, the roles that neck strength and stiffness play in attenuating severe impacts are unclear and warrant investigation.

Change in head velocity.

In addition to high linear accelerations, concussions are characterized very rapid changes in head velocity. Change in head velocity is defined as the velocity change of the head’s center of gravity from initiation of impact to cessation of impact (i.e. impact duration). Linear acceleration and change in head velocity demonstrate a cause and effect relationship as high linear accelerations result in a large change in head velocity throughout contact.

Broglio et al. (2009) found that impacts to the top of the helmet produced the largest linear accelerations, which produced the highest maximum head jerks, a similar indicator as

change in head velocity, in non-concussed high school players. Specifically, maximum head jerk is the quotient of peak linear acceleration and time to peak linear acceleration. Similarly, Pellman et al. (2003) found that NFL players who suffered a concussion experienced significantly greater linear accelerations (98 ± 28 g), which induced greater changes in head velocities (7.2 ± 1.8 m/s) than uninjured struck players (60 ± 24 g and 5.0 ± 1.1 m/s). Viano et al. (2007) quantified the relationships between linear acceleration, change in head velocity, impact duration, and head displacement during replication of the Pellman et al. (2003) study.

In the replication study, Viano et al. (2007) found similar mean linear accelerations and change in head velocities as those reported by Pellman et al. (2003). After contact, researchers reported that head displacement increased exponentially while linear acceleration decreased markedly. Linear acceleration declined from maximum values (~100 g) upon contact to 19.8 ± 10.3 g 20 ms after contact. As linear acceleration dramatically decreased 20 ms after contact, change in head velocity approached maximum values (8.3 ± 2.0 m/s), which contributed to translational head displacement increasing fourfold from 10 to 20 ms (20.2 ± 6.8 to 87.6 ± 21.2 mm).

Viano et al. (2007) further quantified neck rotation around the *z* (superior-inferior plane) and *x* (anterior-posterior plane) axes. This is important as most concussive impacts occur at a lateral or oblique angle to the facemask or helmet shell. This most often causes rotations around the *z* and *x* axes away from the point of force application. The point of force application is most often forward of the head's center of gravity and *z* axis of the cervical vertebrae. The facemask and helmet shell increase the moment arm of the applied torque with the facemask

contributing to a longer moment arm as it protrudes out further from the head than the helmet shell. This is analogous to receiving a hook punch to the jaw in boxing. The jaw serves as a long moment arm, which enables the blow to generate very high linear and rotational accelerations, and subsequently, a large change in head velocity and displacement.

As with head displacement, head rotation about the *z* axis increased exponentially from 10 ms (7.8 ± 1.9 degrees) to 20 ms (33.8 ± 6.1 degrees) after contact. Ninety-five percent of reconstructions resulted in >15 degree rotation around the x axis and 79% resulted in >25 degree rotation. Similar to head displacement, displacement from lateral flexion increased fourfold from 10 ms (6.9 ± 2.5 degrees) to 20 ms (29.9 ± 9.5 degrees). Resultant head displacement increased nearly fivefold from 10 ms (22.1 ± 5.6 mm) to 20 ms (97.7 ± 20.7 mm) after contact. The exponential increase in resultant head displacement is also likely due to long impact duration (~15 ms) of concussive impacts. This long duration is due to the striking player aligning their head, neck, and torso for the purpose of increasing their effective mass and continuing to apply force to the struck player by following through the tackle (Pellman et al., 2003; Viano et al., 2007).

In summary, linear acceleration dramatically decreases while change in head velocity reaches maximum values ~20 ms after contact, which induces an exponential increase in translational head displacement around both the *x* and *z* axes. This large change in head displacement enables the mechanism of concussion to occur, an acceleration of the brain into a forceful contact with the skull.

Quantifying change in head velocity and displacement is crucial because researchers have theorized their importance in relation to head injury and concussion (Denny-Brown & Russell, 1941, Pellman et al., 2003). According to Pellman et al. (2003) concussion is not only related to translational head acceleration, SI, and HIC, but a significant correlation exists between change in head velocity and concussion. Even a small reduction in change in head velocity can drastically reduce the HIC scores (Viano, Casson, & Pellman, 2007).

Denny-Brown and Russell (1941) examined the role that head displacement after contact plays in head injury. They found animals did not suffer a concussion when their head was fixed and immovable during head impact from a pendulum, even at a pendulum speed of 8.8 m/s; however, concussions did occur when the animal's head was allowed to move just 3 mm after impact. These findings lead to the conclusion that rapid head displacement resulting from a rapid change in head velocity might play a significant role in the probability of sustaining a concussion. The aforementioned change in head velocity and displacement findings raise the issue of neck strength and the ability for the player to resist large changes in head velocity and subsequent head displacement from head impacts characterized by high translational accelerations. Large changes in head velocity might be due to insufficient neck strength because the player cannot resist the concussive impact. The ability for football players to attenuate these large changes in head velocity with neck strength could very well play a significant role in concussion prevention.

2.2 Review of Cervical Anatomy

The neck supports the weight of the head, allows for multi-planar movement of the head, and is comprised of vertebras, muscles, nerves, and blood vessels. The purpose of this

section is to review cervical anatomy, specifically cervical vertebrae and the joints they form that allow for multi-planar movement, and the cervical musculature that generates that multi-planar movement.

Cervical vertebrae and joints.

The neck is composed of seven cervical vertebras, which are the smallest of all cervical vertebras. In addition, cervical vertebrae are unique from thoracic and lumbar vertebrae as they are characterized by a foramen in their transverse processes for passage of the vertebral artery. There are seven cervical vertebras (C1-C7), but due to their special characteristics and function, two of them are discussed in detail.

The atlas directly supports the weight of the head and directly articulates with the occipital bone of the head forming the atlanto-occipital joint. Two condyloid joints comprise the atlanto-occipital forming a synovial joint, which primarily affords movement in the sagittal plane (i.e. flexion/extension) and some movement in the frontal plane (i.e. lateral flexion right/left).

The second cervical vertebra (C2) that is inferior to the head is referred to as the axis. The axis directly articulates superiorly with the atlas in order to form the atlanto-axial joint. The atlanto-axial joint is most notably comprised by the articulation of the odontoid process of the axis with the transverse ligament of the atlas. This articulation creates a pivot joint that primarily allows for movement in the transverse plane (i.e. cervical rotation right/left). The atlanto-axial joint is the only cervical vertebra joint that is not characterized by an inter-

vertebral disc. An inter-vertebral disc between the atlas and axis is not necessary as the axis has no vertebral body, but rather directly fuses with the axis.

The other cervical vertebras (i.e. C3-C7) are no less important than the atlas and axis, but they possess similar characteristics as thoracic and lumbar vertebrae with the exception of a foramen in their transverse process.

Range of motion and cervical musculature.

As aforementioned, the cervical spine affords the head movement in multiple planes. Movement in the sagittal plane (i.e. flexion/extension) primarily occurs at the atlanto-occipital joint. In addition, sagittal plane movement occurs at mid-cervical spine (i.e. C3-C5). Normal range of motion (ROM) in the sagittal plane is 45° and 65° in flexion and extension, respectively (Youdas, Garret, Suman, Bogard, Hallman, & Carey, 1992). The following muscles generate cervical flexion bilaterally: 1) sternocleidomastoid; 2) rectus capitis anterior; 3) longus capitis; and 4) longus colli (superior, inferior, and vertical. The following muscles generate cervical extension bilaterally: 1) rectus capitis posterior (major and minor); 2) semispinalis capitis; and 3) splenius muscles (cervicis and capitis).

Movement in the frontal plane (i.e. lateral flexion right/left) primarily occurs at the C3-C5 vertebrae with some movement originating from the atlanto-occipital joint. Normal range of motion in the frontal plane is 35° for both lateral flexion right and left (Youdas et al., 1992). The following muscles generate lateral flexion ipsillaterally: 1) rectus capitis lateralis; 2) oblique capitis superior, sternocleidomastoid; and 3) splenius muscles (cervicis and capitis).

Movement in the transverse plane primarily occurs at the atlanto-axis joint with some originating from rotation of the cervical vertebrae. Normal range of motion in the transverse plane for cervical rotation right and left is 65° and 60°, respectively (Youdas et al., 1992). The following muscles generate cervical rotation unilaterally: 1) sternocleidomastoid (contralaterally); 2) rectus capitis posterior (major); 3) semispinalis capitis (contralaterally); 4) obliquus capitis inferior; and 5) splenius muscles (cervicis and capitis).

2.3 Role of Neck Strength in Concussion

As HIC is significantly related to suffering a concussion, any reduction in HIC might significantly reduce the likelihood of a concussion. Chou and Nyquist (1974) assumed a half-sine head acceleration and developed various equations for quantification of different head impact variables under the HIC model. The following equation quantifies the relationship between change in head velocity (ΔV) and HIC assuming peak head acceleration remains constant (A_p),

$$HIC = 0.0132(\Delta V)A_p^{1.5}$$

The following equation determines head displacement,

$$d = 0.1246(\Delta V^2)/A_p$$

Viano et al. (2007) combined the above two equations of Chou and Nyquist (1974) and formulated the following equation which yields the relationship between HIC, change in head velocity, and head displacement,

$$HIC = 0.0058(\Delta V^4)/d^{1.5}$$

The above equation demonstrates the significance that change in head velocity and head displacement plays in HIC. Out of the two impact variables change in head velocity plays a much larger role given that it is raised to the fourth power, whereas, head displacement is raised to the 1.5 power. For example, given constant head displacement, a 10% reduction in change in head velocity yields a 34% reduction in HIC, whereas, a 10% reduction in head displacement results in a 15% reduction in HIC (Viano et al., 2007).

The work by Chou and Nyquist (1974) and Viano et al. (2007) have mathematically demonstrated the relationships between HIC, change in head velocity, and head displacement, but the question of neck strength and its role in concussion remains unclear. Tierney, Sitler, Swanik, Swanik, Higgins, and Torg (2005) examined the role that neck strength plays in peak head acceleration between genders in collegiate soccer players. They found that females experienced greater peak head-neck segment acceleration than males during external force application. Authors attributed the observed differences in peak head-neck segment acceleration in females to less head mass, neck girth, and neck strength. Therefore, they concluded that females are at a greater risk for sustaining a concussion. Their findings implied that an increase in neck strength would elicit a reduction in peak head acceleration during force application, such as in a force head impact. Mansell, Tierney, Sitler, Swanik, and Stearne (2005) found that despite 8 weeks of isotonic cervical training and increases in neck strength dynamic stabilization of the head-neck segment was unaltered during force application for collegiate male and female soccer players.

In an attempt to determine the role of neck strength in sustaining a football-related concussion, Viano et al. (2007) reconstructed concussive impacts utilizing Hybrid III crash dummies with varying neck stiffness (i.e. neck strength). Neck stiffness was adjusted in the head-neck segment of Hybrid III test dummies, according to established norms, to replicate that of a 10 year old (28 N/mm), female with a neck stiffness in the 5th percentile (5% female: 39 N/mm), a male with a neck stiffness in the 50th percentile (50% male: 80 N/m), and male with a neck stiffness in the 95th percentile (95% male: 113 N/mm). Neck stiffness values represent relaxed musculature and not contracted; therefore, several values up to 240 N/mm were utilized in order to account for large neck musculature and contraction abilities of elite football players. All other variables were held constant during laboratory reconstructions of concussive impacts except for torso mass which was increased to 17.2 kg for the adult (i.e. 5% female, 50% male, and 95% male) reconstructions.

Viano et al. (2007) found that increased neck stiffness (i.e. neck strength) reduced peak resultant translational acceleration and change in head velocity, which subsequently, reduced head displacement. Given that change in head velocity (ΔV) is raised to the fourth power (ΔV^4) in the aforementioned HIC equation, the reduction in ΔV drastically reduced HIC scores. Researchers found that increasing neck stiffness from 80 to 180 N/mm resulted in a 14% reduction in ΔV, 46% reduction in ΔV^4, 11% reduction in head displacement at 15 ms, and 35% reduction in HIC. The lesser neck stiffness in females and children resulted in a non-linear increase in ΔV, and subsequently an increase in HIC, which drastically increase the risk for suffering a concussion in these populations. The findings by Viano et al. (2007) supported the much earlier observations by Denny-Brown and Russell (1941), in which their animal model of

concussion was, in part, dictated by displacement of the head and neck and ΔV. The role ΔV plays in concussion and HIC is exacerbated by the fact ΔV is raised to the fourth power (ΔV^4) in the reduced equation of Chou and Nyquist (1974).

2.4 Summary of Reviewed Literature

In conclusion, concussions are characterized by their locations of impact, linear accelerations, and changes in head velocity. The majority of concussions occur from impacts with high linear accelerations delivered at a lateral of oblique angle on the opposing player's helmet shell rather than facemask. The impact variable most correlated with concussion is linear acceleration with concussions occurring in excess of 90 g. Impacts with high linear accelerations result in large changes in head velocity, which in turn, generates large head displacements around both the *z* and *x* axes.

Concussed players experience greater linear velocities and accelerations, changes in head velocities and impact durations as compared to both the struck but uninjured players and the striking player. Research has proposed several concussion models and proposed thresholds, but currently there is no definitive model for concussion prediction. A neglected variable in predicting concussions is neck strength as sustaining a concussion is the result of the interplay between linear acceleration, location of impact, change in head velocity, change in head displacement, and neck strength. Several researchers have proposed the importance of neck strength in attenuation of severe impacts but very little research has addressed this issue. Neck strength as a means of concussion prevention is a neglected area in both the literature and football strength and conditioning programs.

2.5 Purpose

Therefore, the purpose of this study was to investigate potential differences in isometric neck strength in flexion, extension, and lateral flexion right/left between concussed high school football players and matched non-concussed football players. It was hypothesized that players that have a football-related concussion history would exhibit less isometric neck strength than players without a concussion history.

Chapter 3

METHODS

3.1 Approach to Problem

Sixteen high school football players that suffered a concussion during this past competition season were recruited to comprise the concussed (C) research group and each of these players was matched to a non-concussed player (NP) on the basis of age, height, weight, and position of play. Athletic trainers for local area high school football teams assisted in the recruitment of subjects. Data collected from the C and NP groups were compared on the basis of neck strength, cervical range of motion, and anthropometric variables. A MANOVA was performed for between group differences on the basis of cervical flexion, extension, and lateral flexion right/left. In addition, multiple covariate analyses were performed with neck strength as responses and neck length, neck girth, ROM, and body weight as covariates separately.

3.2 Subjects

Thirty-two high school football players (Age ± SD: 16.8 ± 1.0 years, Height ± SD: 1.77 ± 0.7 m, Weight ± SD: 88.5 ± 22.0 kg) were recruited for this study. Sixteen high school football players that suffered a football-related concussion during the 2010-2011 competition season comprised the C group. Sixteen high school football players with no concussion history, but possessing similar anthropometric and position characteristics as one of the C were recruited to serve as controls.

Inclusion criteria for the C group included: 1) participation at the high school football level; 2) age between 14-18 years; 3) suffered a concussion during practice or a game during

this past competition season (2010-2011); 4) concussion was diagnosed by a certified or licensed athletic trainer and/or physician.

Inclusion criteria for the NC group included: 1) participation at the high school football level; 2) age between 14-18 years; 3) have no history of a diagnosed concussion; 4) have similar anthropometric and position characteristics as a concussed player.

3.3 Preparation for Recruitment

The researcher gave a brief presentation concerning of this research at the Sun City Athletic Training Association (SCATA) monthly meeting and solicited each high school team's athletic trainer for assistance with subject recruitment. Each trainer was provided with the researcher's contact information and informed the researcher concerning the number of concussed players on their respective team. For recruitment of non-concussed players, the researcher collaborated with the athletic trainer to identify another player that is similar to the concussed player on based on age, height, weight, and position of play. If the team did not have such a player, then the researcher recruited a matched-control player from another team.

3.4 Recruitment Procedures

The researcher obtained institutional review board (IRB) approval from the University of Texas at El Paso and El Paso and Ysleta Independent School Districts. After IRB approvals, the researcher met with athletic trainers and solicited them for their collaboration with subject identification and recruitment at the aforementioned SCATA meeting. After collaborating athletic trainers were identified, the researcher traveled to each athletic trainer's respective school in order to dispense consent and assent forms and recruit subjects. Athletic trainers were encouraged to distribute forms to the concussed and possible matched-control players.

The researcher contacted athletic trainers on a weekly basis, via email or phone, and inquired about the status of subject identification and recruitment. For subjects that were minors, athletic trainers informed subjects to give the necessary forms to their parents for their review. If the player wished to participate in the study, parents of players contacted the researcher and an appointment was scheduled for further explanation of the study and obtaining consent. An informed consent form outlining the purpose and details of the study was provided to the parents and an assent form to the player. Signature approvals were sought out from both the parents and the player. For adult subjects, informed consent-signature approval was sought out only from the subject. Copies of the dispersed informed consent, parent consent, and assent forms are found in Appendix A, B, and C, respectively. Additional subject recruitment considerations were as follows:

1) The researcher collected all names of concussed and non-concussed players from each team's athletic trainer.
2) If parents or the subject needed additional information, discussion was provided by phone, email, etc.
3) Parents and players were encouraged to respond to the researcher via email or telephone within one to two weeks.
4) If there was no reply for one to two weeks, the researcher contacted the athletic trainer to determine the status of players participating in the study.
5) When parents and player signed the consent and assent forms, the player retained the forms until an appointed time to give them to the researcher.

6) When the consent and assent forms were signed by parents and players, the researcher organized a meeting with all players on a given team. At this meeting, the researcher collected informed consent and assent forms, clarified any questions players had, and collected data.

7) All data collection occurred at the Stanley Fulton Biomechanics Lab located in the Larry K. Durham center on the University of Texas of El Paso's campus.

8) Subjects were encouraged to find their own mode of transportation to and from the laboratory facilities. However, in the case that no transportation was available for subjects, the researcher transported the subject to and from the data collection site.

3.5 Data Collection

Collected data included measurements of maximal isometric neck strength (flexion, extension, and lateral flexion right/left), range of motion, position of play, concussion history, and anthropometric measurements (height, weight, neck length, and neck girth).

Maximal isometric neck strength.

Each player underwent 4 maximal isometric neck strength measurements. Cervical strength assessments included maximal isometric strength of cervical flexion, extension, and lateral flexion (right and left) using a modified head harness and the Biodex™ 3 (Biodex Medical Systems, Shirley, New York) isokinetic dynamometer. The variable of interest for each direction of testing was peak torque (N-m)

Position of play.

Position of play was defined as the position that the player primarily played and were assigned to one of the two following categories, linemen (L) or skill players (SP). Positions considered in the L category were any offensive (i.e. center, guard, or tackle) or defensive linemen (i.e. end, tackle, or noseguard). Positions considered in the SP category were any other positions (i.e. wide receivers, running backs, quarterbacks, linebackers, or defensive backs) that do not fall under the L category. Their position of play was recorded on the subjects' respective data sheet. If a player played multiple positions, then the player selected the position that they played the majority of the time. A sample player's data sheet is presented in Appendix E.

Concussion history and injury session-type.

Concussion history was assessed in the form of a questionnaire. This form is presented in Appendix D. The questionnaire determined previous concussion history for all players. For non-concussed players, having a history of concussion was an exclusionary criterion for this group and they were not accepted in the study. For concussed players, the questionnaire additionally determined injury session-type (i.e. practice or competition) in which they suffered their concussion.

Anthropometric measurements.

Anthropometric measurements included weight, height, neck length, and neck girth. Accurate determinations of height and weight were critical as they were two of the four criteria (i.e. age, height, weight, and position of play) for matching players. Cervical anthropometric measurements were measured to ascertain their relationship to maximal isometric force generation for both the C and NP groups.

Range of motion.

Cervical range of motion measurements included the following directions: 1) cervical flexion; 2) cervical extension; 3) cervical lateral flexion right; and 4) cervical lateral flexion left.

3.6 Measurements

In order to establish test-retest reliability, the researcher performed pilot testing on five university students. Each subject was tested two times for the following measurements, height, weight, neck length, neck girth, cervical range of motion, and maximal isometric neck strength in all directions. A Pearson product-moment correlation coefficient was calculated in order to determine test-retest reliability. For establishment of test-retest reliability, a minimum $r = 0.8$ was required.

Anthropometric variables.

Anthropometric variables of height, weight, neck length, and neck girth were assessed and recorded on each subject's data sheet. A sample of this subject data sheet is presented in Appendix E.

Height and weight.

Body weight was measured using a medical weight scale and recorded in kilograms (kg) to the nearest 0.1 kg. Height was measured using a stadiometer and recorded in centimeters (cm) and recorded to the nearest 0.1 cm.

Neck length.

Neck length was measured using a metric tape measure. Neck length was defined as the vertical displacement from the seventh cervical (C7) spinous process to the occipital protuberance. The C7 spinous process and occipital protuberance was palpated in order to

accurately determine their locations. Once these bony landmarks were located, the researcher measured the vertical displacement between the C7 spinous process and the occipital protuberance using a metric tape measure and recorded the vertical displacement to the nearest 0.1 centimeters (cm). This method of neck length measurement was utilized by Olivier and Du Toit (2008) to determine neck length of senior elite rugby players. A similar method was utilized by Vasavada, Danaraj, and Siegmund (2008) in which they measured neck length as vertical displacement from the C7 spinous process to the tragus. The tragus is usually located at the level of the occipital protuberance, which would yield similar measurements as this study's utilized method of determining neck length.

Potential limitations to this method of neck length measurement were the amount of cervical flexion/extension and protraction/retraction the player had during measurement. In order to control for these potential limitations and ensure reliability, the researcher manually adjusted each player's head for the purpose of manipulating cervical flexion/extension and protraction/retraction in order to achieve a neutral anatomical head position. In addition, pilot testing was performed in order to determine test-retest reliability and it was determined that this was an accurate and reliable method for assessing neck length. Test-retest reliability was computed at $r = 0.98$ for this method of measuring neck length, which exceeded the $r \geq 0.80$ requirement.

Neck girth.

Neck girth was measured using a metric tape measure and recorded in to the nearest 0.1 centimeters (cm). Neck girth measurement consisted of a circumference measurement of the neck just above the thyroid cartilage. This method of neck girth measurement was utilized

by Mansell, Tierney, Sitler, Swanik, and Stearne (2005) to determine neck girth in collegiate soccer players. The neck was palpated in order to determine the location of the thyroid cartilage. After palpation of the thyroid cartilage, the researcher placed the tape measure above this landmark and wrapped the tape measure around the player's neck making sure the tape measure lied flat on the cervical skin and was level.

A potential limitation to this method of neck girth measurement was the tension applied to the tape measure during measurement. In order to control for this limitation and increase reliability of measurements, the tape measure was outfitted with a tensiometer that allowed the researcher to standardize magnitude of tension applied during measurements. In addition, pilot testing was performed in order to determine test-retest reliability and it was determined that this was an accurate and reliable method of assessing neck girth. Test-retest reliability was computed at $r = 0.97$ for this method of assessing neck girth, which exceeded the $r \geq 0.80$ requirement..

Range of motion.

Range of motion (ROM) was measured for the following four directions, cervical flexion, cervical extension, and cervical lateral flexion right/left. Cervical ranges of motion were assessed utilizing a Penny & Giles electrogoniometer (Penny & Giles, Dorset, United Kingdom). All range of motion testing occurred with subjects sitting fully-erect in a chair. The proximal sensor of the electrogoniometer lead was secured to the C7 spinous process using double-sided tape after the skin surface was cleaned with alcohol and lightly abraded to ensure adequate adhesion. The distal sensor was placed on the occipital protuberance and secured to the head using a tight fitting swim cap to ensure movement between the occipital protuberance and the

sensor was minimized. The researcher visually inspected both sensors during testing to ensure no extraneous movement occurred.

Each subject performed three trials for each of the following directions, flexion, extension, lateral flexion right/left. Range of motion for each direction was recorded as the average of the three trials for the respective direction of ROM testing. Each trial began with subjects sitting fully-erect and placing their head in a neutral, anatomical head position. If needed, the researcher either gave verbal clarification as to what a neutral, anatomical head position entailed or manually adjusted the subject's head in order to achieve the desired position. After achieving this desired position and after each trial, the electrogoniometer was zeroed.

For cervical flexion ROM, subjects were instructed to perform a chin-tuck and flex the neck as far as possible until a non-painful endpoint was achieved and hold that endpoint until given a verbal command to return to the neutral head position.

For cervical extension ROM, subjects were instructed to perform a chin-lift and extend their neck as far as possible until a non-painful end point was achieved and without applying excess force to the back of the chair with their back/shoulder. Subjects held this fully-extended position until the researcher instructed subjects to return to the neutral head position. As aforementioned, the electrogoniometer was zeroed when the subject achieved neutral head position after each trial.

For cervical lateral flexion right and left, subjects were instructed to laterally flex their head to the desired direction (right or left) until a non-painful endpoint was achieved. During lateral flexion ROM testing, subjects were instructed to avoid raising their shoulders. For each

direction of testing, ROM was recorded at the respective non-painful end point for all trials and recorded to the 1.0 degree (°).

Pilot testing was performed to determine test-retest reliability for this method of measuring cervical ROM in multi-planes. Average test-retest reliability for four directions of cervical ROM was computed as r = 0.91. This computed correlation coefficient exceeded the required $r \geq 0.80$.

Isometric neck strength.

Isometric neck strength in cervical flexion, extension, and lateral flexion was assessed using the Biodex™ 3 (Biodex Medical Systems, Shirley, New York) in conjunction with a modified head harness. Peak torque was recorded as isometric neck strength and was recorded to the nearest 1.0 N-m for each direction of interest. Subjects underwent isometric neck strength testing after completing the concussion history survey, anthropometric, and range of motion measurements. The Biodex™ 3 was auto-calibrated before each the beginning of each data collection day.

Modified head harness.

Subjects were properly fitted with the modified head harness before undergoing neck strength testing. The modified head harness consisted of two parts, a padded head harness for cervical resistance training and a Riddell chin strap. The four straps of the chin strap were secured to the crown of the head harness using snap-on fasteners. The head harness was adjusted to fit each subject securely around the crown of their head and rested just above the ears and eyebrows while fitting over their occipital protuberance. The chin strap allowed for proper placement of the head harness, prevented extraneous movement of the head harness,

and provided an anchor for subjects to apply force against during isometric testing. See Appendix F for an illustration of modified-head harness.

Isometric flexor strength.

For assessment of isometric cervical flexor strength, subjects were placed on the fully-extended chair of the isokinetic dynamometer in the prone position with their clavicle resting on the edge of the head rest so their head was not able to rest on the isokinetic dynamometer chair. In order to minimize extraneous activation of musculature, subjects' torso was secured to the chair using padded straps. The modified head harness was secured to the arm of the isokinetic dynamometer using a chain. The vertical displacement between the modified head harness and the arm of the isokinetic dynamometer was approximately 0.30 m. Adjustments were made to fully-extended chair and isokinetic dynamometer arm to achieve proper placement of the arm in relation to the subject's head were achieved. Proper placement of the isokinetic dynamometer arm consisted of anatomical head position and the line of pull from the modified head harness to the isokinetic dynamometer during testing was perpendicular to the spine of the subject, which allowed for optimum force generation using the SCM and provided a reference for standardization across subjects. Proper testing position was ensured through visual inspection by the researcher. See Appendix G for an illustration of the isometric flexor strength set-up.

Isometric extensor strength.

For assessment of maximal cervical extensor strength, subjects were placed on the fully-extended chair of the isokinetic dynamometer in the supine position. Subjects were positioned so that their C7 spinous process was not resting on the head rest, which prevented subjects

from resting their head on the chair of the isokinetic dynamometer. Subjects were secured to the chair using padded straps around their torso and were instructed to cross their arms over their chest during testing. The researcher secured the modified head harness to the arm of the isokinetic dynamometer, which was located approximately 0.30 m above the subject's head. The isokinetic dynamometer arm was properly positioned to anatomical position of the head and the line of pull during extensor strength testing would be perpendicular to the subject's spine. Proper extension testing position was ensured through visual inspection by the researcher. See Appendix H for an illustration of the isometric extensor strength set-up.

Isometric lateral flexor strength.

For assessment of lateral flexors, the subject lied in the lateral recumbent position with the superior portion of their deltoid muscle resting on the head rest, which prevented the subject's head from resting on the chair of the isokinetic dynamometer. The modified head harness was secured to the isokinetic dynamometer arm contralaterally to the side that was undergoing testing. The isokinetic dynamometer arm was adjusted in order to achieve both anatomical head position and a line of pull that was perpendicular to the subject's head. See Appendix I for an illustration of the isometric lateral flexor strength set-up.

Isometric neck strength protocol.

Subjects underwent dynamic cervical stretching before engaging in maximal necks strength testing. Stretching included 20 s of cervical extension/flexion, cervical lateral flexion right/left, and cervical rotation right/left each. After stretching, the neck strength testing protocol was explained to all subjects and any subsequent questions were answered. Subjects were informed that they could terminate testing at any time during the testing. In addition,

subjects were instructed to inform the research if any significant discomfort occurred and the testing would be immediately terminated. Subject safety was ensured through constant visual inspection of the testing protocol by the researcher and subject/researcher communication. Subjects did not report any complaints of pain or significant discomfort during, and therefore, testing was never prematurely terminated for subject discomfort or safety purposes.

For each direction (i.e. flexion, extension, lateral flexion), subjects performed three maximal voluntary contractions (MVC) for five seconds each with 15 seconds rest in between each repetition. In addition, at least two minutes rest was given between transitioning to a new direction of testing (i.e. flexion to extension). All neck strength testing was performed with the seat back portion of the isokinetic dynamometer in full extension, and therefore, testing was performed in the following positions, prone, supine, and lateral recumbent (right and left).

A potential limitation to isometric neck strength testing was subjects might attempt to utilize muscles other than the cervical musculature, which might have resulted in invalid isometric neck strength results. In order to control for this potential limitation and ensure validity, subjects were secured to the Biodex™ 3 chair by a strap that was placed across their torso. In addition, subjects crossed their arms over their chest. Proper subject and isokinetic dynamometer arm positioning forced subjects to localize cervical musculature and minimize core musculature involvement during testing. Proper testing positions were ensured through visual inspection. The proper subject and isokinetic dynamometer arm positioning were enforced for all C and NP subjects. As with previous measurements, pilot testing was performed in order to determine test-retest reliability and it was determined this method of

isometric neck strength produced reliable results. Computation of a test-retest reliability coefficient revealed that this method of isometric neck strength was reliable ($r = 0.92$).

3.7 Statistical Analysis

All statistical analyses were performed using SAS 9.2 with level of significance for all statistical analyses set at $\alpha = 0.05$. Before any statistical analyses were performed, the data were screened to determine if any of the assumptions of normality, linearity, and heteroscedasticity were violated. None of the assumptions were violated, and therefore, the data was not transformed for the purpose of meeting statistical assumptions.

Test-retest reliability.

In order to establish test-retest reliability, the researcher performed pilot testing on five university students. Each subject was tested two times for the following measurements, height, weight, neck length, neck girth, and maximal isometric neck strength in all directions. Through computation of a Pearson product-moment correlation coefficient it was determined that the following measurements were reliable, neck length ($r = 0.98$), neck girth ($r = 0.97$), cervical ROM ($r = .91$), and isometric neck strength ($r = 0.92$). For establishment of test-retest reliability for all measurements, a minimum $r = 0.8$ were required.

Descriptive statistics.

Descriptive statistics (Mean ± Standard Deviation) are reported for anthropometric, ROM, and isometric neck strength values. Table 1 displays each group's anthropometric data (i.e. height, weight, neck length and girth). Table 2 displays ROM values for each group. Table 3 displays isometric neck strength values for each direction of interest. T-tests were utilized to determine if differences existed between concussed and non-concussed player groups on the

basis of age, anthropometric measurements, and ROM. For between-group differences on the basis of ROM, t-tests were Bonferroni corrected to reflect a statistical significance of $\alpha = 0.01$ for each individual analysis in order to reflect an overall significance of $\alpha = 0.05$.

MANOVA*.*

A multiple analysis of variance (MANOVA) was performed to determine if between-group differences existed for isometric neck strength in flexion, extension, lateral flexion right, and lateral flexion left. The independent variable was group membership, concussed or non-concussed, and the dependent variables were the four directions of isometric neck strength.

Co-variate analyses.

For covariate analyses, three multiple analysis of covariate variance (MANACOVA) and four analysis of covariate variance (ANACOVA) were performed. The covariates of interest were body weight, neck length, neck girth, and cervical ROM. If during covariate analysis a significant group variance difference was observed then effects were not analyzed. The responses were various directions of neck strength. Not all four isometric neck strength directions served as responses for all MANACOVA's. Some effects were not analyzed due to group variance differences. Lastly, the independent variable for all covariate analyses was group membership.

MANACOVA 1.

A MANACOVA was performed to determine between-group differences in three directions of isometric neck strength with body weight serving as a covariate. Specifically, body weight served as the covariate and isometric neck strength in flexion, lateral flexion right, and lateral flexion left were the responses. Isometric neck strength in extension was not analyzed

with body weight as a covariate due to significant between-group variance differences in body weight.

MANACOVA 2.

A MANACOVA was performed to determine between-group differences in all four directions of isometric neck strength with neck length serving as a covariate. Specifically, neck length served as the covariate and isometric neck strength in flexion, extension, lateral flexion right, and lateral flexion left were the responses.

MANACOVA 3.

A MANACOVA was performed to determine between-group differences in three directions of isometric neck strength with neck girth serving as a covariate. Specifically, neck girth served as the covariate and isometric neck strength in flexion, lateral flexion right, and lateral flexion left were the responses. Isometric neck strength in extension was not analyzed with neck girth as a covariate due to significant between-group variance differences in neck girth.

ANACOVA 1.

An ANACOVA was performed to examine between-group differences in isometric neck flexion strength with cervical flexion ROM serving as a covariate.

ANACOVA 2.

An ANACOVA was performed to examine between-group differences in isometric neck extension strength with cervical extension ROM serving as a covariate.

ANACOVA 3.

An ANACOVA was performed to examine between-group differences in isometric neck lateral flexion right strength with cervical lateral flexion right ROM serving as a covariate.

ANACOVA 4.

An ANACOVA was performed to examine between-group differences in isometric neck lateral flexion left strength with cervical lateral flexion left ROM serving as a covariate.

Chapter 4

RESULTS

4.1 Descriptive Statistics

Table 1 displays the physical characteristics of concussed and non-concussed subjects. There were no significant between-group differences for age (p = 0.59), height (p = 0.52), weight (p = 0. 72), neck length (p = 0.83), or neck girth (p = 0.58).

Table 1

Mean ± standard deviation of anthropometric measurements of concussed and control high school football players*

Group	N	Age (years)	Height (m)	Weight (kg)	Neck Length (cm)	Neck Girth (cm)
Concussed*	16	16.6 ± 0.9	1.76 ± 0.06	87.1 ± 19.7	13.3 ± 1.6	38.7 ± 1.6
Control*	16	16.8 ± 1.0	1.77 ± 0.07	89.9 ± 24.7	13.1 ± 2.1	39.3 ± 3.0

*No significant group differences on the basis of age, height, weight, neck length, and neck girth (p > 0.51).

Table 2 displays group cervical range of motion (ROM) values for concussed and non-concussed subjects in flexion, extension, and lateral flexion right/left. There were no significant between-group differences for ROM in flexion (p = 0.69), extension (p = 0.69), lateral flexion right (p = 0.21), or lateral flexion left (p = 0.80).

Table 2

Mean ± standard deviation of range of motion (ROM) values for flexion, extension, and lateral flexion right/left for concussed and control players

Group	Flexion (°)	Extension (°)	Lateral Flexion Right (°)	Lateral Flexion Left (°)
Concussed*	59 ± 11	68 ± 10	41 ± 8	50 ± 9
Control*	61 ± 11	66 ± 13	44 ± 7	51 ± 11

Active ROM was assessed in the seated position with an electrogoniometer. *No significant group differences for any ROM direction (P > 0.20).

4.2 Multivariate Analysis of Variance

Table 3 displays peak isometric neck strength values (Mean ± SD) for concussed and non-concussed subjects in flexion, extension, and lateral flexion right/left. Figure 5 displays peak torque box plots for isometric neck flexion strength for concussed and non-concussed players. No significant between-group differences were observed for isometric neck flexion strength ($p = 0.83$). Figure 6 displays peak torque box plots for isometric neck extension strength for concussed and non-concussed players. No significant between-group differences were observed for isometric neck extension strength ($p = 0.82$). Figure 7 displays peak torque box plots for isometric neck right lateral flexion strength for concussed and non-concussed players. No significant between-group differences were observed for isometric neck left lateral flexion strength ($p = 0.51$). Figure 8 displays peak torque box plots for isometric neck left lateral

flexion strength for concussed and non-concussed players. No significant between-group differences were observed for isometric neck left lateral flexion strength (p = 0.79).

Table 3

Mean ± standard deviation for isometric neck strength values for flexion, extension, and lateral flexion right/left for concussed and control subjects

Group	Flexion (N-m)	Extension (N-m)	Lateral Flexion Right (N-m)	Lateral Flexion Left (N-m)
Concussed*	66.2 ± 21.0	104.4 ± 28.6	62.5 ± 20.2	63.7 ± 21.6
Control*	67.5 ± 12.0	102.5 ± 13.4	58.6 ± 12.0	65.6 ± 17.5

Isometric neck strength was assessed using an isokinetic dynamometer in conjunction with a modified head harness with subjects in the prone (flexion), supine (extension), and lateral recumbent (lateral flexion right/left) positions. *No significant group differences were observed for any neck strength variables (p > 0.50).

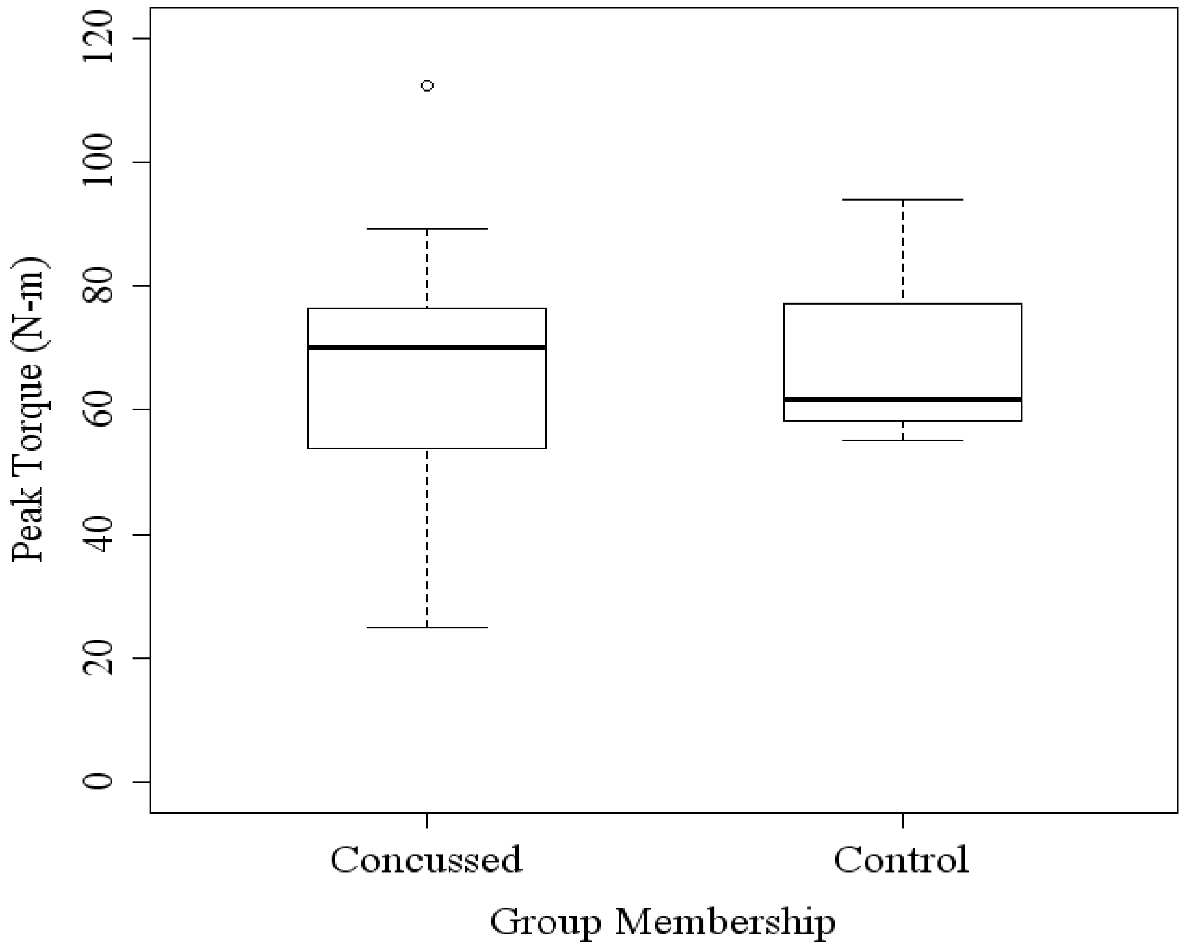

Figure 1. Isometric neck flexion strength for concussed and non-concussed groups. Box plots displays quartiles of peak torque values by group with each group's median value represented by the thick band inside the box. Statistical outliers are indicated by stand-alone circles. No significant between-group differences were observed for isometric flexor strength ($p > 0.05$).

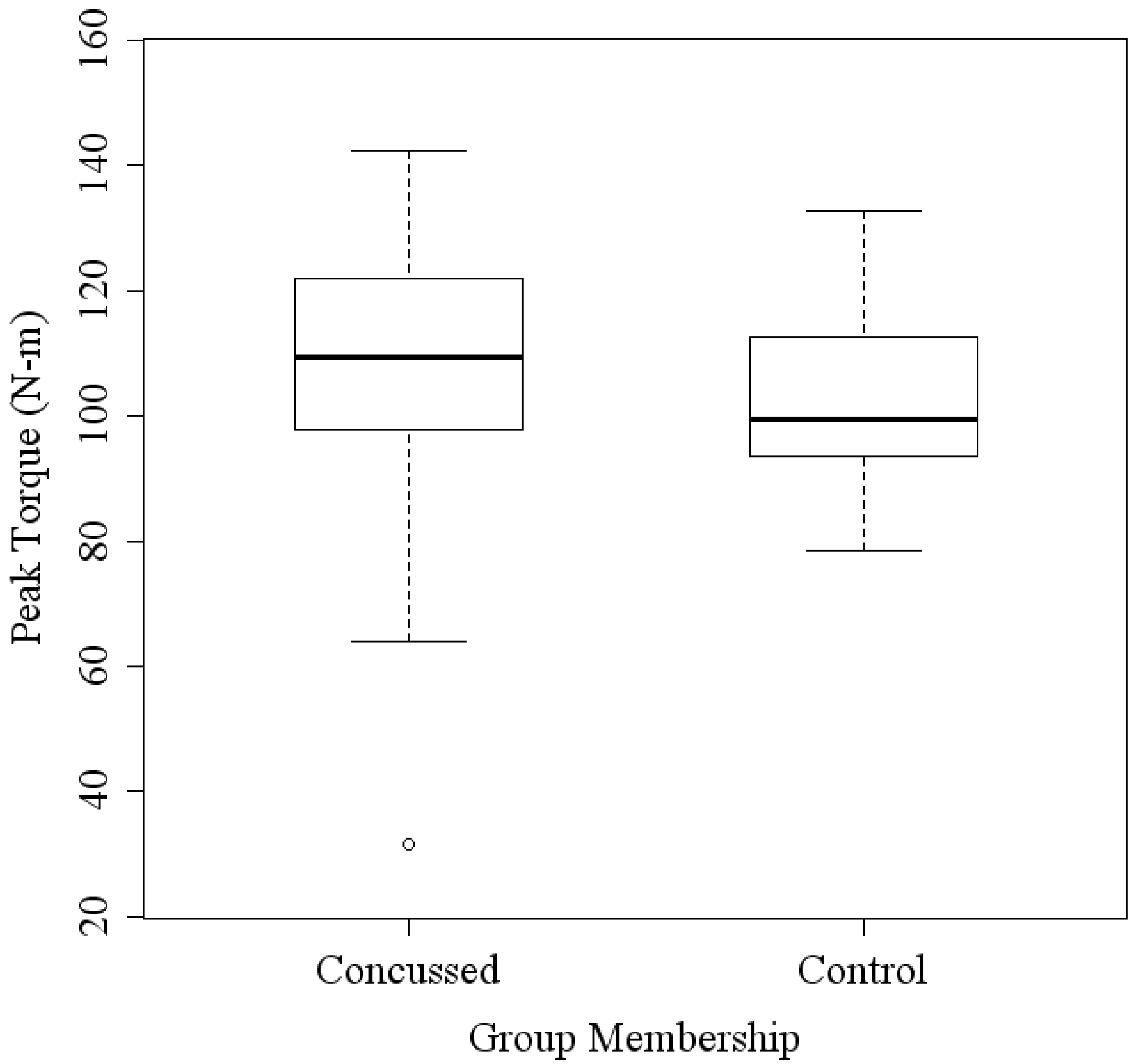

Figure 2. Isometric neck extension strength for concussed and non-concussed groups. Box plots displays quartiles of peak torque values by group with each group's median value represented by the thick band inside the box. Statistical outliers are indicated by stand-alone circles. No significant between-group differences were observed for isometric extensor strength ($p > 0.05$).

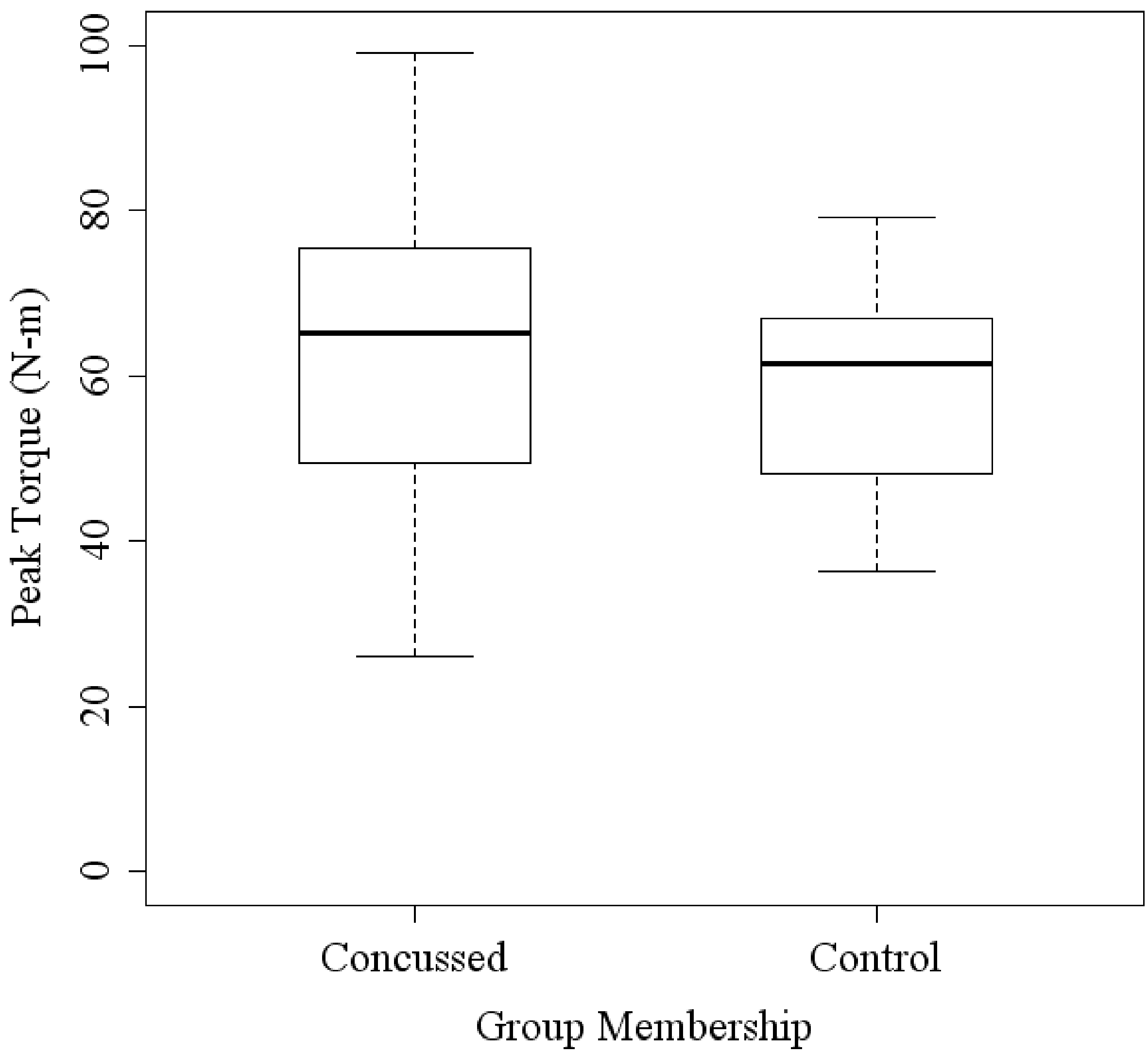

Figure 3. Isometric neck right lateral flexion strength for concussed and non-concussed groups. Box plots displays quartiles of peak torque values by group with each group's median value represented by the thick band inside the box. Statistical outliers are indicated by stand-alone circles. No significant between-group differences were observed for isometric lateral flexor right strength ($p > 0.05$).

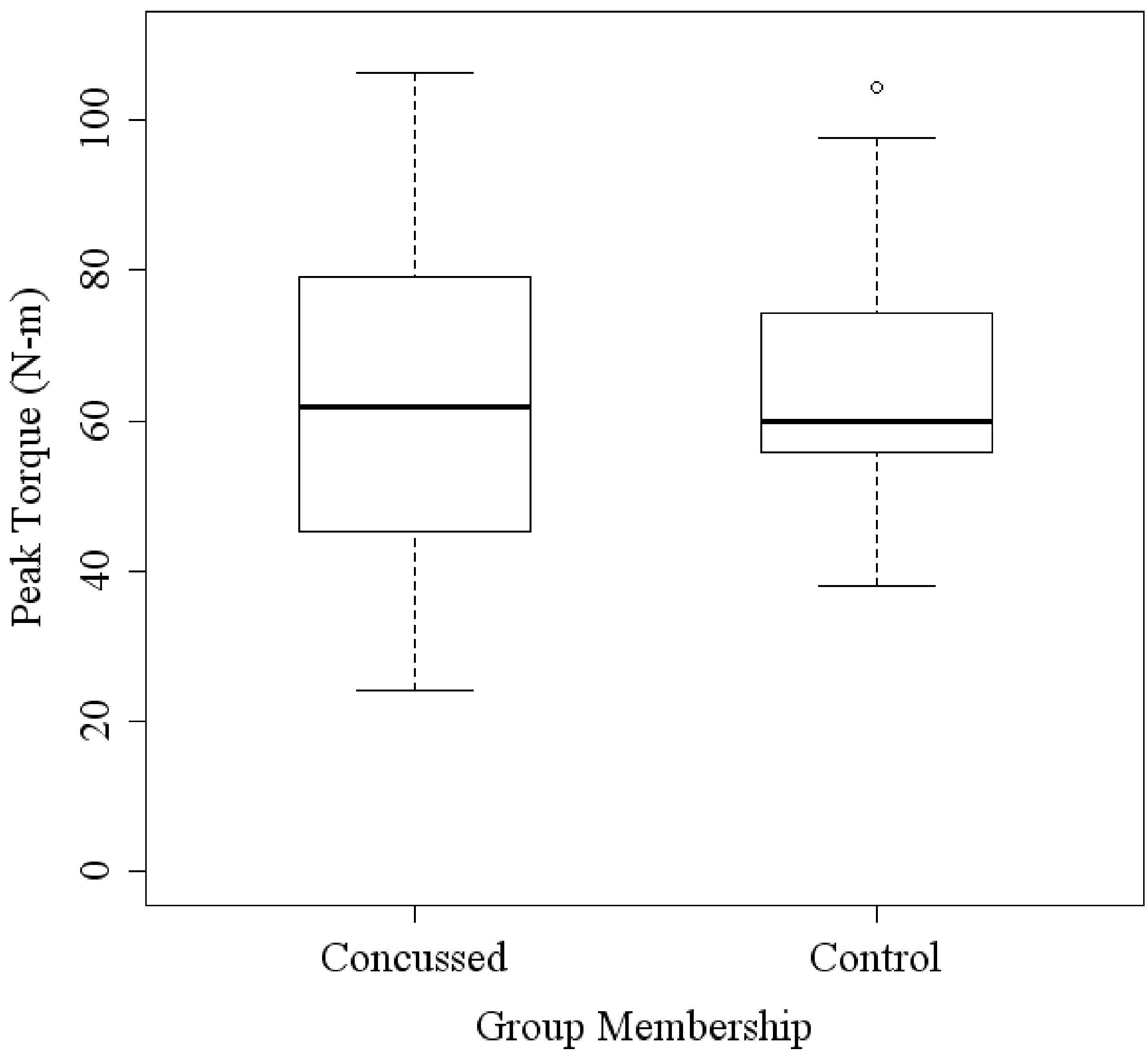

Figure 4. Isometric neck left lateral flexion strength for concussed and non-concussed groups. Box plots displays quartiles of peak torque values by group with each group's median value represented by the thick band inside the box. Statistical outliers are indicated by stand-alone circles. No significant between-group differences were observed for isometric lateral flexor left strength ($p > 0.05$).

4.3 Covariate Analyses

MANACOVA 1.

The Multiple Analyses of Covariance Test (MANACOVA) with body weight serving as a covariate showed no significant between group differences for the following directions of isometric neck strength responses: flexion ($p = 0.93$), lateral flexion right ($p = 0.44$), and lateral flexion left ($p = 0.88$). A main effect for group membership was not available for isometric neck strength in extension due to an observed significant between-group variance difference in body weight ($p = 0.02$).

MANACOVA 2.

There were no significant between group variance differences for neck length; therefore, the effects of neck length on the four directions of isometric neck strength were analyzed. The MANACOVA with neck length serving as a covariate showed no significant between group differences for the following directions of isometric neck strength responses: flexion ($p = 0.85$), extension ($p = 0.83$), lateral flexion right ($p = 0.50$), and lateral flexion left ($p = 0.82$).

MANACOVA 3.

The MANACOVA with neck girth serving as a covariate showed no significant between group differences for the following directions of isometric neck strength responses: flexion ($p = 0.95$), lateral flexion right ($p = 0.34$), and lateral flexion left ($p = 0.98$). A main effect for group

membership was not available for isometric neck strength in extension due to an observed significant between-group variance difference in neck girth (p = 0.04).

ANACOVA 1.

There was no significant between-group variance difference for cervical flexion ROM; therefore, the effect of cervical flexion ROM on isometric neck strength in flexion was analyzed. The Analysis of Covariance Test (ANACOVA) showed no significant main effect for group membership for isometric neck flexion strength with neck flexion ROM held as a covariate ($p = 0.77$).

ANACOVA 2.

There was no significant between-group variance difference for cervical extension ROM; therefore, the effect of cervical extension ROM on isometric neck strength in extension was analyzed. The ANACOVA showed no significant main effect for group membership for isometric neck extension strength with neck extension ROM held as a covariate ($p = 0.86$).

ANACOVA 3.

There was no significant between-group variance difference for cervical lateral flexion left ROM; therefore, the effect of cervical lateral flexion right ROM on isometric neck strength in lateral flexion right was analyzed. The ANACOVA showed no significant main effect for group membership for isometric neck right lateral flexion strength with neck right lateral flexion ROM held as a covariate ($p = 0.62$).

ANACOVA 4.

There was no significant between-group variance difference for cervical lateral flexion left ROM; therefore, the effect of cervical lateral flexion left ROM on isometric neck strength in lateral flexion left was analyzed. The ANACOVA showed no significant main effect for group membership for isometric neck left lateral flexion strength with neck left lateral flexion ROM held as a covariate ($p = 0.77$).

Chapter 5

DISCUSSION

5.1 Summary of Reviewed Literature, Purpose, and Hypothesis

Concussive impacts are characterized by high linear and rotational accelerations that cause large changes in head velocities, which result in large changes in head displacement (Broglio et al., 2010; Viano et al., 2007; Pellman et al., 2003). The injury mechanism of a concussion occurs during this rapid head displacement in which the brain is accelerated into a forceful contact with the skull causing a transient alteration in mental status (Sivak et al., 2005). It has been theorized that neck strength might assist in the attenuation of severe impacts causing a reduction in head displacement, and thereby, reducing the likelihood of sustaining a concussion (Viano et al., 2007). Previous research using helmeted Hybrid III crash test dummies to measure biomechanical responses with increased neck stiffness for reconstructed concussive impacts supports this theory (Viano et al., 2007). In addition, previous research has demonstrated that greater neck strength between-genders were associated with lower head-neck segment linear acceleration during force application (Tierney et al., 2005). However, significantly increasing neck strength after eight weeks of cervical resistance training did not result in a reduction of head-neck segment linear acceleration during force application (Mansell et al., 2005).

The role of neck strength in sustaining a concussion is not well understood and a neglected research topic. There is sparse research concerning the role of neck strength in the attenuation of severe impacts and the possibility of cervical resistance training as a mean of

concussion prevention. In addition, cervical resistance training is often neglected at the high school level strength and conditioning programs.

Purpose.

The purpose of this study was to investigate potential differences in isometric neck strength in flexion, extension, and lateral flexion right/left between concussed high school football players and matched non-concussed football players.

Hypothesis.

It was hypothesized that players that have a football-related concussion history would exhibit less isometric neck strength than players without a concussion history. This study aimed to determine if differences in neck strength existed between players with concussion history and those without, based on theories proposed by previous research (Viano et al., 2007; Mansell et al., 2005; Tierney et al., 2005). The rationale was that greater peak force generation for anticipated impacts would decrease the likelihood of sustaining a concussion. Therefore, it was a working hypothesis that concussed players would exhibit lower levels of isometric neck strength as they may have deficiencies in properly resisting severe impacts with force generation through cervical musculature.

5.2 Summary of Results

Findings of this study do not support the working hypothesis. There were no significant differences in neck strength in flexion, extension, and lateral flexion right/left between players with a football-related concussion history and those without. The possible confounding neck

strength variables of age, neck length, and neck girth were taken into consideration, and were therefore, utilized as matching criteria. Specifically, players were successfully matched as there were no significant between-group differences in age and anthropometric measurements.

Further analyses used body weight, neck length, neck girth, and cervical ROM separately as covariates. Performing multiple covariate analyses enabled evaluation of each covariate on their respective responses. Multiple covariate analyses revealed no significant between-group differences after adjusting for the effects of the covariates on the responses for each covariate model. Specifically, isometric neck strength did not significantly differ between groups after adjusting for body weight, neck length, neck girth, and cervical range of motion.

In summary, this study revealed no significant differences between concussed and control high school players for any directions of isometric neck strength with or without adjusting for body weight, neck length, neck girth, or cervical range of motion through covariate analyses.

5.3 Relation of Results to Past Literature

This section visits several topics in relation to this study's findings and previous concussion-based research and the implications for future research, to include: 1) in-vitro vs. in-vivo research; 2) net peak impact force; 3) passive cervical stiffness; 4) cervical musculature length and force generation; 5) cervical musculature fatigue; and 6) defensive head positioning.

Isometric neck strength.

The neck strength aspect in concussion-based research is divided between *in-vitro* and *in-vivo* data collection. *In-vitro* concussion-based research that focuses on cervical contributions primarily consists of reconstruction of concussion impacts using helmeted Hybrid III crash test dummies and theoretical computer-generated models to estimate the role of neck strength in attenuation of head impacts. *In-vivo* based concussion research examining the contribution of neck strength has primarily consists of observing kinetic and kinematic responses to force application that induces subject head displacement. *In-vitro* research supports the theory that neck strength can significantly contribute to reducing the likelihood of sustaining a sport-related concussion; however, the results of this study and previous research utilizing *in-vivo* data collection do not support this theory.

In-vitro research.

Queen, Weinhold, Kirkendall, and Yu (2003) examined the theoretical effects of heading a soccer ball with negligible vs. infinite neck-torso stiffness using the Hertz contact theory and a computer generated model. They reported reductions in linear and angular head accelerations and HIC scores with infinite neck stiffness vs. negligible neck stiffness, despite no reduction in impact force. Similarly, Viano et al. (2007) found that increasing neck stiffness in Hybrid III crash test dummies elicited exponential reductions in linear acceleration, change in head velocity, head displacement, and HIC scores. The effect of significantly reducing these head impact variables means a drastic reduction in the likelihood of sustaining a concussion. However, Viano et al. (2007) results are limited by the peak resistance force to impact was constant and did not need to be generated before or during impact, which might distinguish the

difference between the human and crash test dummy response to head impact. Similarly, the results of Queen et al. (2003) only provide a theoretical framework of the relationship between neck stiffness and impact variables using a computer generated model but not human and/or animal subjects. Admittedly, this is a very difficult research topic to utilize *in- vivo* data collection due to practical and ethical issues, and therefore, *in-vitro* data collection is substantially utilized to estimate the human response to head impacts.

In-vivo research.

In order to determine the *in-vivo* response to force application, Tierney et al. (2005) recruited male and female soccer players. They found that females experienced greater head-neck segment acceleration and displacement than males. It was theorized that these observed differences were due to males exhibited greater isometric neck strength than females, and therefore, were more adept in dynamically stabilizing the head-neck segment during known and unknown force application. This might partially account for the observed greater concussion incidence rates of concussions in females (Shankar et al., 2007).

The current study suggests that isometric neck strength is not associated with an increased likelihood of sustaining a sport-related concussion. Mansell and colleagues (2005) had similar findings. They reported that significantly increasing isometric neck strength after 8 weeks of cervical resistance training did not induce a concomitant reduction in head-neck segment linear acceleration during force application, either known or unknown. These findings and the findings from the current study suggest that maximum neck strength is not related to the likelihood of sustaining a concussion. Possibly more relevant neck strength variables to

sustaining a sport-related concussion are cervical musculature onset time, time to peak torque, and eccentric neck strength.

Isometric neck strength and cervical co-contraction are likely very important for soccer players. The ability to form a rigid head-neck segment is vital for successfully stabilizing the head while heading a soccer ball. For football players, the ability to co-contract and hold the head in an isometric position while receiving a head impact is probably not as important as dynamically stabilizing the head. This raises the issue of isometric vs. dynamic strength. Isometric neck strength might resemble the cervical response to heading a soccer ball; however, it might not be representative of sustaining a football head impact. Eccentric neck strength more closely resembles the cervical response to football head impacts, and therefore, should be utilized in football related concussion based research.

This study's results are limited by the fact that neck strength was measured isometrically. To the knowledge of the researcher through an extensive literature review, there are no known studies that measure neck strength eccentrically. Neck strength has not been measured eccentrically due to ethical issues concerning participant safety. Based on this limitation to *in-vivo* research concerning neck strength, creation of a safe method for measuring neck strength eccentrically is warranted.

Net peak impact force.

The findings of this study reveal that lower levels of isometric neck strength are not associated with a concussion history. One plausible explanation is the magnitude of peak force generated by cervical musculature is not significant enough to elicit meaningful reductions in

the likelihood of sustaining a concussion. For instance, average peak torque in flexion was 67 N-m for this study and concussive impacts to the facemask, incurring eccentric flexion, regularly occur an impact force of 4.4 ± 1.2 kN (Pellman et al., 2003). These concussive impacts result in a net impact force of 4333 N-m. This reveals that the observed maximum cervical strength might be insignificant in attenuating severe impacts with such large forces. To further this point, Queen et al. (2003) reported no reduction in impact force with infinite neck-torso stiffness. Conversely, reductions in linear and angular accelerations and HIC were observed in their theoretical computer generated model. These findings suggest that the relationship between impact force and resultant head accelerations may not be as clearly understood as previously thought. The generalizations of Queen et al. (2003) are limited due to the utilization of a computer generated model and lack of *in-vivo* data collection.

However, if a player significantly increases their neck flexion strength with cervical resistance training, it is possible that they might be able to reduce head impact variables below proposed concussive thresholds. For example, would an increase of 30 N-m in flexion decrease head acceleration from 98 g to 96 g? A linear acceleration of 98 g is theorized as a concussion threshold and if a player can generate a greater resistive force then they might be able to reduce the likelihood of sustaining a concussion. Therefore, greater neck strength may or may not serve as a defense mechanism for head impacts. Further research is needed to determine if significant increases neck strength would result in a reduction in the likelihood of sustaining a concussion by reducing head impact variables below proposed concussion thresholds. In summary, the role of neck strength in attenuation of severe impacts is obviously a multi-faceted issue that warrants much more research using animal or human subjects to more

clearly understanding between impact force, neck strength/stiffness, and resultant head accelerations.

Passive cervical stiffness.

An alternative theory is that passive cervical stiffness might play a more significant role in eliciting reductions in linear accelerations and head displacement during head impacts than neck strength. McGill, Jones, and Bishop (1994) quantified passive cervical stiffness in the directions of flexion, extension, and lateral flexion. They reported tolerable limits of applied torque as less than 10 N-m for all directions with greater toleration of applied torque and elastic energy storage in extension than in flexion and lateral flexion. They concluded that average passive cervical stiffness without any cervical muscle activation in the aforementioned directions was 10 N-m.

It is unclear whether 10 N-m of resistive force originating from cervical series elastic components would be significant enough to elicit reductions in head impact variables, thereby reducing the likelihood of sustaining a concussion. Although, if a player can generate 67 N-m in isometric flexion and cervical stiffness is 10 N-m then net resistive force is 77 N-m. Additionally, if a player increases their isometric neck flexion strength to 100 N-m then it is possible that they might be able to reduce head acceleration by several gravitational force units during head impacts. A reduction in several gravitational units might reveal the difference between sustaining a concussion and no head injury.

Cervical musculature length and force generation.

Length of cervical musculature upon impact might contribute to greater cervical force generation. Lu and Bishop (1996) reported greater cervical peak electromyography activity during known force application and subsequent contralateral flexion for pre-bent head position trails vs. neutral head position trials. Greater cervical peak electromyography activity might increase muscle recruitment, and subsequently, maximal force generation during head impacts with cervical musculature in a pre-stretched position. This type of maneuver might be observed when a player that is about to be struck utilizes their cervical musculature to deliver a head impact to the striking player (i.e. striking with the head to the striking player's head.

Series elastic component and stretch-reflex.

Enhanced force generation accompanied by elongated cervical musculature is attributed to two mechanisms: 1) increased passive cervical stiffness through the series elastic component of cervical musculature; and 2) escalated muscle recruitment through the stretch-reflex initiated by intra-fusal muscle spindles. The series elastic component (i.e. passive stiffness) of cervical musculature is directly proportional to angular displacement (McGill, Bennet, & Bishop, 1994). Therefore, the greater the pre-stretch of cervical musculature through angular displacement, the greater the contribution of the series elastic component to total peak force generation, or resistance force to severe head impacts. Increased cervical electromyography activity is most likely due to increased muscle recruitment due to intra-fusal muscle spindles undergoing the stretch reflex. Despite heightening the cervical resistance force to severe impacts through the greater passive cervical stiffness and escalated muscle recruitment elicited by angular displacement which elongates cervical musculature, it is a debatable issue whether

these mechanisms can significantly reduce the likelihood of sustaining a sport-related concussion.

Cervical musculature fatigue.

Another potential confounding variable to football-related concussion is fatigue of cervical musculature. While no significant differences in isometric neck strength in the sagittal and coronal planes between players with a concussion history and control players were observed, it is possible that these groups of players differ in their endurance of cervical musculature. A player might sustain a concussion when their cervical musculature is fatigued after sustaining many impacts and unable to generate an adequate resistive force. This inability to generate an adequate resistive force might result in excessive rapid change in head displacement, thereby inducing a concussion.

Ang, Linder, and Harms-Ringdahl (2005) reported that helicopter pilots experienced greater cervical muscle fatigue than fighter pilots, as evidenced by a significant reduction in cervical musculature electromyographic activity during repeated submaximal contractions. Football players sustain repeated head impacts without adequate recovery throughout a competition or practice. Therefore, it is a reasonable assumption that football players experience greater cervical muscle fatigue than helicopter pilots, who primarily utilizes their cervical musculature for postural control and head positioning. Although not measured in this study, neck muscle fatigue would be an important variable of interest to quantify when examining differences between players with a history of concussion and those without.

Defensive head positioning.

Defensive head positioning before sustaining a severe head impact is another concussion consideration. Proper positioning of the head before impact could be a defensive mechanism to prevent concussions from occurring. Players without a history of concussion might be more adept at defensively positioning their head to prevent rapid and excessive head displacement than players with a concussion history, especially given this study's population. High school football players have limited football experience, and therefore, they may lack the skills needed to reduce biomechanical impact variables in order to reduce the likelihood of sustaining a concussion from an anticipated impact.

Newman (1997) reported that fighter pilots developing individualized approaches to reducing the effects of high gravitational forces on the cervical spine during high gravitational force (g-force) maneuvers. Individualized approaches included pre-positioning of the head before being subjected to high g-forces and utilizing cockpit structures for head immobilization. It is likely that football players prepare their head-neck/torso segments in the most optimal way to effectively absorb the impact without suffering injury. A possible defensive mechanism is positioning the head in such a manner as to receive the impact on side helmet rather than the facemask or the top of the helmet shell. This might be evidenced by the fact that more concussions occur from frontal and top impacts than side impacts (Broglio et al., 2010; Broglio et al., 2009). Older, more skilled players might intentionally or unintentionally receive impacts to the side of the helmet rather than the front or top as a defensive mechanism against sustaining a concussion. In fact, collegiate players as a whole might exhibit this defense mechanism as they experience 10% less impacts to the facemask than their high school counterparts (Mihalik et al., 2007).

5.4 Limitations, Strengths, and Suggestions for Future Research

This section is dedicated to defining the strengths and limitations of this study. In addition, suggestions for future research on the topic of neck strength contribution in head impacts. Therefore, this section is subdivided into the following subdivisions: 1) limitations; 2) strengths; and 3) suggestions for future research.

Limitations.

A limitation of this study included small sample size, which reduced statistical power of results. This reduction in statistical power might have significantly contributed to the observation of non-significant findings. From purely a statistical standpoint, the study warrants replication with a greater sample size. In fact, it was determined through a post-study power analysis that in order to accurately observe a statistically significant finding sample size should have been 2805, 2278, 282, 1718 for flexion, extension, lateral flexion right, and lateral flexion left, respectively.

Obtaining the aforementioned sample sizes was obviously impossible, but barriers to subject recruitment and retainment are discussed. The suggested sample size was, in part, not obtained due to difficulty in identifying, recruiting, and retaining players with a football related concussion history. Specifically, sample size was primarily reduced due to the fact that only a few players from the small number of collaborating high schools met the inclusionary criteria. In addition, logistic and school district administration barriers significantly contributed to subject recruitment and retainment adversities. Given that this study's population was adolescent males, logistic barriers included transportation hindrances and data collection

scheduling availability for subjects. Many subjects were currently competing in track and field or baseball during this study's data collection period. Additionally, subjects often relied on others for transportation, which likewise decreased the subject availability for data collection.

A further limitation of this study was the age of the adolescent population as the average age was only 16 years. Therefore, differences in neck strength between concussed and control players might not have been visible due to lack of consistent whole-body resistance training and biological immaturity. It is a reasonable assumption that the majority of high school football players do not participate in a consistent and regimented whole-body resistance training program that is required at the collegiate and professional levels. Moreover, young players experience undeveloped musculature due to biological immaturity that accompanies adolescence. In fact, collegiate players were on average were 12 kg heavier while only 8 cm taller than this study's subjects (McCrea et al., 2003). Differences in neck strength between concussed and control players might be observable in collegiate and professional players where biological immaturity and overall strength are not limiting factors. Future research should include replication of this study using collegiate and professional players to control for biological maturation and resistance training in order to more accurately determine if neck strength is a correlated factor with sustaining a football-related concussion.

Strengths.

A strength of this study was successfully matching players with a football-related concussion history to players without a concussion history but possessed similar age and anthropometric characteristics. Specifically, through statistical analysis it was determined that

no significant between-groups were observed for age, height, weight, neck length, and neck girth. This successful matching enabled for valid neck strength conclusions to be drawn as any observed significant between group differences for anthropometric variables would have possibly undermined isometric neck strength findings, despite no observed neck strength differences.

Another strength were the test-retest reliability coefficients of this study's measurements. Test-retest reliability coefficients were computed for neck length, neck girth, cervical ROM, and neck strength after performing pilot testing. All reliability coefficients of measurements met the $r \geq 0.80$ requirement. The measurement techniques for determination of neck length (Olivier & Du Toit, 2008; Vasavada et al., 2008) and neck girth (Mansell et al., 2005) were deemed valid assessments by previous research. Additionally, previous research has validated the use of an electrogoniometer to determine cervical ROM in flexion/extension and lateral flexion (Alund & Larsson, 1990).

Isometric neck strength was assessed using a novel technique, through the connection of a modified head harness to the arm of the isokinetic dynamometer. This method of neck strength assessment was determined a reliable measurement through computation of a test-retest reliability coefficient. The Biodex 3 isokinetic dynamometer is a valid assessment tool for measuring isometric torque with a reliability coefficient of $r = 0.99$ (Drouin, Valovich-McLeod, Shultz, Gansneder, & Perrin, 2004). In order to increase reliability of neck strength measurements, subjects were outfitted with a modified head harness and testing positions were standardized across all subjects. In addition, subjects were secured to the isokinetic dynamometer chair in order to localize cervical musculature, inhibit extraneous movement, and

prevent core musculature recruitment. Seng (2000) reported validation a similar method of assessing neck strength, Biodex isokinetic dynamometer and headgear, as the one utilized in this study. However, Seng (2000) performed neck strength assessment in a seated semi-reclined (30°) position, whereas this study utilized the prone, supine, and lateral recumbent positions.

Although, assessment of neck strength was performed using a valid and reliable method, the chosen method for assessing neck strength is a limitation of this study. Measuring neck strength isometrically might not have been the most appropriate and generalizable method of neck strength to apply to attenuation of severe impacts. Admittedly, eccentric neck strength, primarily in flexion and lateral flexion, would have been more applicable. Severe head impacts cause change in head displacement about all three axes, which most likely means that an eccentric contraction of cervical musculature is taking place. Brault, Siegmund, and Wheeler (2000) reported eccentric contractions of cervical musculature during unanticipated rear automobile impacts. Therefore, it is logical to assume that cervical musculature would be activated before or right after impact, especially if the impact is anticipated.

Future research.

Force generation in eccentric neck strength would have been more applicable for this research topic. Measuring eccentric neck strength would have yielded another strength variable that would have been of interest besides peak torque, time to peak torque. Given that no between-group differences were observed for peak torque in isometric neck strength, time to peak torque for an eccentric contraction would yield valuable information. It is quite possible that concussed players and players without a concussion history do not differ in peak

torque but differ in their response time to peak force generation. The ability to generate maximum force in a short period of time is critical for attenuation of severe impacts characterized by short impact durations. This ability might possibly reduce the magnitude of head acceleration and head displacement, and thereby, reducing the likelihood of sustaining a concussion (Viano et al., 2007).

Future research should include eccentric neck strength rather than isometric neck strength as eccentric neck strength is more generalizable to sustaining a head impact. Additionally, determining muscle onset time of cervical musculature through electromyography would be an important variable of interest to quantify. If concussed and non-concussed players do in fact differ by their time to peak torque then it is very likely that they differ in their cervical musculature onset times. If they do differ in their onset times then the role of neck strength in concussions might be additionally related to temporal activation of cervical musculature. Logically, earlier and rapid activation of cervical musculature should produce smaller time to peak force values.

Lastly, as previously discussed, fully developed players might yield different isometric neck strength results. Differences in neck strength might not have been observable due to undeveloped musculature seen with biological immaturity. Replication of this study using collegiate and/or professional players would yield meaningful results without the confounding factors of biological age. In addition, sample size should be increased to make the statistical analysis more sensitive to mean differences in neck strength.

5.5 Conclusion

This study demonstrated that players with a history of football-related concussions do not differ in isometric neck strength from players without a history of concussion. However, inadequate sample size, a limitation of this study, might have masked possible significant between-group differences. In addition, observable neck strength differences might not exist given the age and musculature immaturity of this study's population. Additionally, measuring neck strength isometrically is the most common method of cervical strength, and was therefore utilized for this study; although, measurement of eccentric neck strength is likely more representative of the cervical musculature response during severe impacts. Future research should include replication of this study with measurements of eccentric neck strength and myoelectric activity.

Even though concussed and control players do not differ in their isometric neck strength it is possible that they might differ in other strength and electromyography variable. Measuring neck strength eccentrically in conjunction with electromyography would yield the following variables of interest, eccentric peak torque, time to peak torque, muscle onset times, and peak EMG amplitude. Quantifying these variables might assist in determining naturally occurring differences between players with a history of concussion and those without, and might aid in understanding the role of neck strength in attenuation of concussive impacts. Additional considerations of this topic include net peak impact force, passive cervical stiffness, cervical musculature length, cervical musculature fatigue, and defensive head positioning. It is theorized that the combination of these factors contribute to likelihood of sustaining a football-related concussion.

5.6 Practical Applications

While neck strength is only theorized to assist in the reduction in sustaining a concussion, it is recommended that football players at all levels participate in regular cervical resistance training. Increasing neck strength has been demonstrated to reduce chronic neck pain and cervical injuries. More *in-vivo* research is needed to accurately assess the implication of neck strength in the attenuation of severe impacts.

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LIST OF APPENDIX

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A. Informed Consent Form

University of Texas at El Paso (UTEP) Institutional Review Board
Informed Consent Form for Research Involving Human Subjects

Protocol Title: The Role of Isometric Neck Strength in Predicting Concussions Sustained by High School Football Players
Principal Investigator: Cameron Raschke
UTEP : Kinesiology

In this consent form, "you" always means the study subject. If you are a legally authorized representative (such as a parent or guardian), please remember that "you" refers to the study subject.

1. Introduction

You are being asked to take part voluntarily in the research project described below. Please take your time making a decision and feel free to discuss it with your friends and family. Before agreeing to take part in this research study, it is important that you read the consent form that describes the study. Please ask the study researcher or the study staff to explain any words or information that you do not clearly understand.

2. Why is this study being done?

You have been asked to take part in a research study that will be examining the role that neck strength plays in sustaining a concussion during football at the high school level.
Approximately, 50 high school football players will be enrolling in this study at UTEP.
You are being asked to be in the study because you meet one of the following requirements: 1) you sustained a concussion during this past football season; 2) you have not sustained a concussion during this past football season, but you have similar physical and positional characteristics as a player that has suffered a concussion during this past season.
If you decide to enroll in this study, your involvement will last about one to two days.

3. What is involved in the study?

If you agree to take part in this study, the researcher will: 1) have you fill out a questionnaire that will ask about your age, concussion history, and position of play; 2) measure your height, weight, neck length, and neck girth; 3) assess your neck strength in four directions. It should take about an hour for you to fill out the questionnaire and for the researcher to collect all of your measurements.

4. What are the risks and discomforts of the study?

There are no known risks associated with this research.

However, discomfort might occur due to neck muscle soreness and fatigue. Neck muscle soreness and fatigue might occur because you are applying as much force as possible with the muscles in your neck during neck strength measurement.
The study may include risks that are unknown at this time.

5. What will happen if I am injured in this study?

The University of Texas at El Paso and its affiliates do not offer to pay for or cover the cost of medical treatment for research related illness or injury. No funds have been set aside to pay or reimburse you in the event of such injury or illness. You will not give up any of your legal rights by signing this consent form. You should report any such injury to Cameron Raschke (915-747-5928) and to the UTEP Institutional Review Board (IRB) at (915-747-8841) or irb.orsp@utep.edu.

6. Are there benefits to taking part in this study?

There will be no direct benefits to you for taking part in this study. This research may help us to understand the role that neck strength plays in sustaining a concussion while playing high school football.

7. What other options are there?

You have the option not to take part in this study. There will be no penalties involved if you choose not to take part in this study.

8. Who is paying for this study?

Internal Funding:

Funding for this study are provided by UTEP Department of Kinesiolgy and College of Health Science Graduate Enhancement Fund.

External funding:

This study is not funded by external sources.

9. What are my costs?

There are no direct costs. You will be responsible for travel to and from the research site and any other incidental expenses.

10. Will I be paid to participate in this study?

You will not be paid for taking part in this research study.

11. What if I want to withdraw, or am asked to withdraw from this study?

Taking part in this study is voluntary. You have the right to choose not to take part in this study. If you do not take part in the study, there will be no penalty. If you choose to take part, you have the right to stop at any time. However, we encourage you to talk to a member of the research group so that they know why you are leaving the study. If there are any new findings

during the study that may affect whether you want to continue to take part, you will be told about them.
The researcher may decide to stop your participation without your permission, if he or she thinks that being in the study may cause you harm.

12. Who do I call if I have questions or problems?

You may ask any questions you have now. If you have questions later, you may call:
Cameron Raschke
Phone: (915) 747-5928
Email: clraschke@miners.utep.edu

If you have questions or concerns about your participation as a research subject, please contact the UTEP Institutional Review Board (IRB) at (915-747-8841) or irb.orsp@utep.edu.

13. What about confidentiality?

Every effort will be made to keep your information confidential. Your personal information may be disclosed if required by law. Organizations that may inspect and/or copy your research records for quality assurance and data analysis include, but are not necessarily limited to:

- The sponsor or an agent for the sponsor
- Department of Health and Human Services
- UTEP Institutional Review Board

Because of the need to release information to these parties, absolute confidentiality cannot be guaranteed. The results of this research study may be presented at meetings or in publications; however, your identity will not be disclosed in those presentations.
In order to increase confidentiality, your name will be coded with a number (John Smith = 1) as soon as you agree to participate in this study. A record sheet will be generated with all of the subjects' names and their subsequent numbers. This record sheet will be securely kept in a locked file cabinet. All record sheets and forms will be kept in a locked file cabinet in the Human Performance Laboratory. All electronic data will be on the researcher's computer and will be password protected. All records will be destroyed after the study has been concluded.

14. Mandatory reporting

If information is revealed about child abuse or neglect, or potentially dangerous future behavior to others, the law requires that this information be reported to the proper authorities.

15. Authorization Statement

I have read each page of this paper about the study (or it was read to me). I know that being in this study is voluntary and I choose to be in this study. I know I can stop being in this study without penalty. I will get a copy of this consent form now and can get information on results of the study later if I wish.

Subject Name: ______________________________ Date: ______________

Subject Signature: ______________________________ Time: ____________

Parent/Guardian Signature: ____________________________________

Consent form explained/witnessed by :

Printed name: ______________________________

Signature: __________________________________

Date: _____________ Time: _____________

B. Parent Consent Form

University of Texas at El Paso (UTEP) Institutional Review Board
Informed Consent Form for Research Involving Human Subjects

Protocol Title: The Role of Isometric Neck Strength in Predicting Concussions Sustained by High School Football Players
Principal Investigator: Cameron Raschke
UTEP: Kinesiology

This form is for the parent/guardian of potential subjects that are under the age of 18 years.

1. Introduction

Your child is being asked to take part voluntarily in the research project described below. Please take your time making a decision and feel free to discuss it with your friends and family. Before agreeing to allow your child to part in this research study, it is important that you read the consent form that describes the study. Please ask the study researcher or the study staff to explain any words or information that you do not clearly understand.

2. Why is this study being done?

Your child has been asked to take part in a research study that will be examining the role that neck strength plays in sustaining a concussion during football at the high school level.

Approximately, 50 high school football players will be enrolling in this study at UTEP.

Your child is being asked to participate in the study because he meet one of the following requirements: 1) he sustained a concussion during this past football season; 2) he has not sustained a concussion during this past football season, but has similar physical and positional characteristics as a player that has suffered a concussion during this past season.

If you decide to allow your child to enroll in this study, his involvement will last about one to two days.

3. What is involved in the study?

If you agree to allow your son to take part in this study, the researcher will: 1) have your child fill out a questionnaire that will ask about his age, concussion history, and position of play; 2) measure his height, weight, neck length, and neck girth; 3) assess his neck strength in four directions. It should take no longer than an hour for your child to fill out the questionnaire and for the researcher to collect all of his measurements.

4. What are the risks and discomforts of the study?

There are no known risks associated with this research.

However, your child might experience discomfort due to neck muscle soreness and fatigue. Neck muscle soreness and fatigue might occur because your child is applying as much force as possible with his neck muscles during neck strength measurement.

The study may include risks that are unknown at this time.

5. What will happen if my child is injured in this study?

The University of Texas at El Paso and its affiliates do not offer to pay for or cover the cost of medical treatment for research related illness or injury. No funds have been set aside to pay or reimburse your child in the event of such injury or illness. You and your child will not give up any of his legal rights by signing this consent form. You and/or your child should report any such injury to Cameron Raschke (915-747-5928) and to the UTEP Institutional Review Board (IRB) at (915-747-8841) or irb.orsp@utep.edu.

6. Are there benefits to taking part in this study?

There will be no direct benefits to you and your child for taking part in this study. This research may help us to understand the role that neck strength plays in sustaining a concussion while playing high school football.

7. What other options are there?

Your child has the option not to take part in this study, and you as the parent/guard have the right to not allow your child to participate in this study. There will be no penalties if you do not allow your child to take part in the study or your child decides not to take part in this study.

8. Who is paying for this study?

Internal Funding:

Funding for this study are provided by UTEP Department of Kinesiolgy and the College of Health Science Graduate Enhancement Fund.

External funding:

This study is not funded by external sources.

9. What are my and/or my child's costs?

There are no direct costs. You and your child will be responsible for travel to and from the research site and any other incidental expenses.

10. Will my child be paid to participate in this study?

You and your child will not be paid for taking part in this research study.

11. What if I want to withdraw my child, or my child is asked to withdraw from this study?

Taking part in this study is voluntary. You have the right to not allow your child not to take part in this study and your child has the right not to take part in this study. If you do not allow your child to take part in this study or your child does not want to take part in this study, there will be no penalty.

If you allow your child and your child chooses to take part, you and/or your child have the right to stop at any time. However, we encourage you and/or your child to talk to a member of the research group so that they know why you are leaving the study. If there are any new findings during the study that may affect whether you and/or your child want to continue to take part, you and your child will be told about them.

The researcher may decide to stop your child's participation without you and/or your child's permission, if he or she thinks that being in the study may cause your child harm.

12. Who do I call if I and/or my child have questions or problems?

You and/or your child may ask any questions you and/or your child have now. If there are questions later, you and/or your child may call:
Cameron Raschke
Phone: (915) 747-5928
Email: clraschke@miners.utep.edu

If you and/or your child have questions or concerns about your child's participation as a research subject, please contact the UTEP Institutional Review Board (IRB) at (915-747-8841) or irb.orsp@utep.edu.

13. What about confidentiality?

Every effort will be made to keep your child's information confidential. Your child's personal information may be disclosed if required by law. Organizations that may inspect and/or copy your research records for quality assurance and data analysis include, but are not necessarily limited to:

- The sponsor or an agent for the sponsor

- Department of Health and Human Services
- UTEP Institutional Review Board

Because of the need to release information to these parties, absolute confidentiality cannot be guaranteed. The results of this research study may be presented at meetings or in publications; however, your identity will not be disclosed in those presentations.

In order to increase confidentiality, your child's name will be coded with a number (John Smith = 1) as soon as you agree to allow your child to participate in this study. A record sheet will be generated with all of the subjects' names and their subsequent numbers. This record sheet will be securely kept in a locked file cabinet. All record sheets and forms will be kept in a locked file cabinet in the Human Performance Laboratory. All electronic data will be on the researcher's computer and will be password protected. All records will be destroyed after the study has been concluded.

14. Mandatory reporting

If information is revealed about child abuse or neglect, or potentially dangerous future behavior to others, the law requires that this information be reported to the proper authorities.

15. Authorization Statement

I have read each page of this paper about the study (or it was read to me). I know that being in this study is voluntary and I choose to allow my child to be in this study. I know I can stop my child's participation in this study without penalty. I will get a copy of this consent form now and can get information on results of the study later if I wish. By signing this form, I give my child permission to participate in this study.

Subject Name: ______________________________ Date: ____________

Subject Signature: ______________________________ Time: ____________

Parent/Guardian Signature: ______________________________

Consent form explained/witnessed by :
Printed name: ______________________________

Signature: ______________________________

Date: ____________ Time: ____________

C. Assent From

University of Texas at El Paso (UTEP) Institutional Review Board
Assent Form for Research Involving Human Subjects

Protocol Title: The Role of Isometric Neck Strength in Predicting Concussions Sustained by High School Football Players
Principal Investigator: Cameron Raschke
UTEP: Kinesiology

I am being asked to decide if I want to be in this research study because I have either: 1) sustained a concussion during this past football season; 2) have not sustained a concussion during this past football season, but you have similar physical and positional characteristics as a player that has suffered a concussion during this past season.
I know that to be in this study I will:

- Come to UTEP's Biomechanics lab one time for about one hour.
- Complete a questionnaire that asks questions about my age, concussion history, and my football position.
- Have my height, weight, neck length, and neck girth measured by the researcher.
- Have my neck strength measured in four different directions using a machine.
- Receive no compensation for participating in this study.

I asked and got answers to my questions. I know that I can ask questions about this study at any time.
I know that I can stop being in the study at any time without anyone being mad at me. I will not get in trouble if I stop being in the study.
I know that only the people who work on this research study will know my name.
I want to be in the study at this time. I can ask about what happened in the study.
Child's Printed Name: ______________________________
Child's Signature: ______________________________ Date: ______________
Witness or Mediator: ______________________________ Date: ______________
I have explained the research at a level that is understandable by the child and believe that the child understands what is expected during this study.

Signature of Person Obtaining Assent:

______________________________________ Date ______________

D. Concussion History/Position of Play Questionnaire

Concussion History/Position of Play Questionnaire

Subject Name: ________________________

School: ______________________________

Age: ________y

Definition of a concussion: A concussion is an impact to the head that causes a temporary change in your mental state. Symptoms of a concussion can include headache, dizziness, confusion, loss of balance, blurred vision, nausea, amnesia, loss of consciousness, vomiting.

1) *During this past season*, have you sustained a medically (physician or athletic trainer) *diagnosed* concussion?

   Yes or No

   If so, when? _________________________

2) According to the above definition of a concussion, do you think you have sustained a concussion *during this past season that was not* diagnosed by either an athletic trainer or physician?

   Yes or No

   If so, when? __________________________

3) If you suffered a diagnosed concussion, was your concussion diagnosed by a physician or athletic trainer?

   Physician or Athletic Trainer or Other: ______________________

4) For the diagnosed concussion you sustained during this past season, did the concussion occur during a game or practice?

   Game or Practice or Other: ________________________

5) For your diagnosed concussion, did the impact occur at the:

Facemask: ______________________

Top of the helmet: _______________

Side of the helmet: _______________

Back of the helmet: ______________

Other: _________________________

6) What position do you play? If you play two positions, circle the one that you *play the most* during games and practice.

Offensive Lineman = Center or Guard or Tackle

Defensive Lineman = Noseguard or Defensive Tackle or End

Offensive Skill Player = Quarterback or Running Back or Wide Receiver
Tight End or Full Back

Defensive Skill Player = Cornerback or Safety or Linebacker.

7) Are you currently participating in competitive wrestling?

Yes or No

8) Are you currently engaging in any form of neck resistance training?

Yes or No

E. Subject Data Sheet

Subject Data Sheet

Subject Number:______________________________

Group Membership: Concussed or Non-concussed

Diagnosed Concussion: Physician or Athletic Trainer or Other: _____________

Position of Play: ______________________________

Age: ______________ y

Height: ____________ cm

Weight: ___________ kg

Neck Length: _____________ cm

Neck Girth: _______________ cm

Cervical ROM:

Flexion: _____________________________________

Extension: ___________________________________

Lateral Flexion Right: ___________________

Lateral Flexion Left: ___________________________

Peak Isometric Neck Strength:

Flexor: ____________ Nm

Right Flexion: _______ Nm

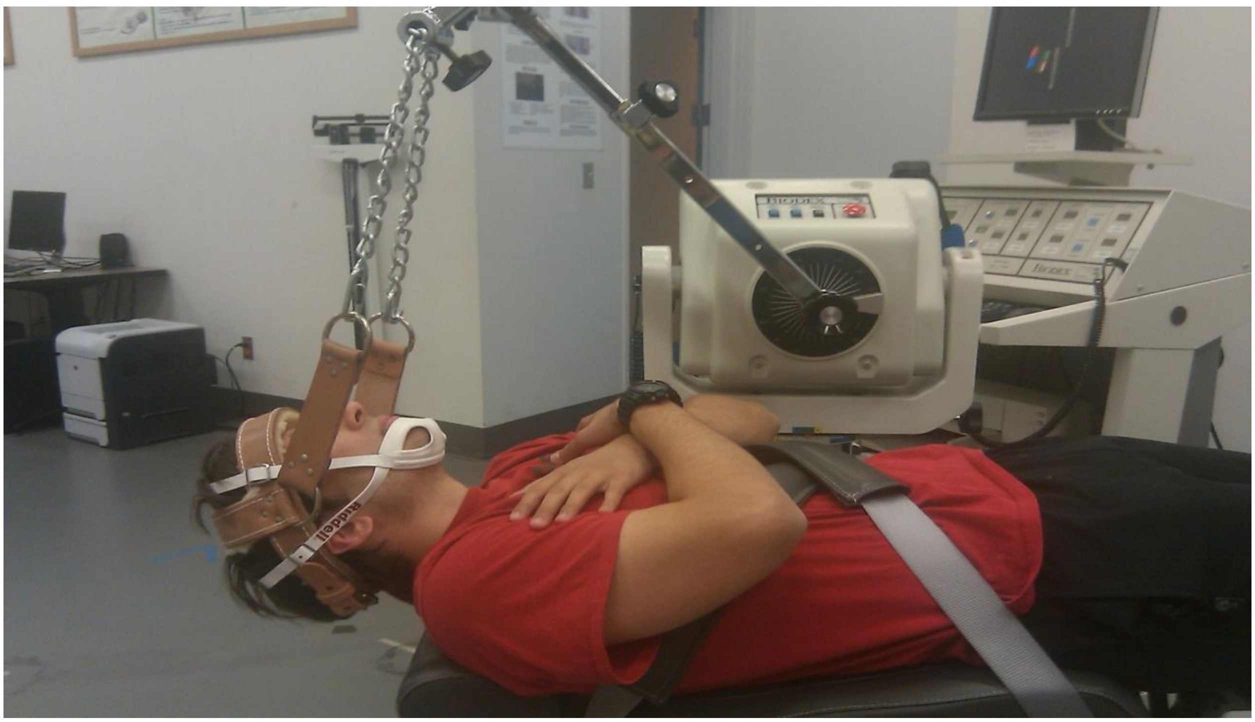
Left Flexion: ____________ Nm

Extensor: ____________ Nm

F. Modified Head Harness

G. Cervical Flexion Strength Set-up

H. Cervical Extension Strength Set-up



I. Cervical Lateral Flexion Strength Set-up

CURRICULUM VITA

Cameron Raschke was born in Kerrville, Texas. The second child of Donald and Louada Raschke, he graduated from Tivy High School, Kerrville, Texas, in the spring of 2005 and entered the University of Texas at El Paso (UTEP) in the fall with a football scholarship. While pursuing a bachelor's degree in kinesiology, he won numerous academic and athletic awards including kinesiology major of the year 2008-2009 and football all-conference honors. After earning his bachelor's degree, he entered UTEP's kinesiology graduate program in the fall of 2009 and was employed as a teaching assistant after his last year of athletic eligibility. During his graduate studies, he performed his thesis study with the assistance of a research award and presented at the South Central American Society of Biomechanics conference in 2011 with the assistance of a travel award. In the near future, he plans to pursue a doctoral degree in either biomechanics or exercise science.

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