THE INFLUENCE OF THE CERVICAL MUSCULATURE, VISUAL PERFORMANCE, AND ANTICIPATION ON HEAD IMPACT SEVERITY IN HIGH SCHOOL AND COLLEGIATE FOOTBALL

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ABSTRACT

JULIANNE SCHMIDT: The Influence of the Cervical Musculature, Visual Performance, and Anticipation on Head Impact Severity in High School and Collegiate Football
(Under the direction of Kevin M. Guskiewicz)

Context: Athletes with weaker, smaller, and less stiff cervical musculature; diminished visual performance; and that do not anticipate an oncoming collision are thought to be more likely to experience rapid head acceleration during collision. **Objective**: To compare the odds of sustaining higher magnitude head impacts between athletes with higher and lower performance on cervical characteristic and visual performance measures and to compare head impact magnitudes between anticipated and unanticipated collisions. **Participants**: Forty-nine high school and collegiate football players. **Interventions**: Participants completed the cervical testing protocol and visual performance assessment prior to the season. Video footage of on-field collisions was analyzed to determine each player's level of anticipation at the time of head impact. Head impact biomechanics were captured at each practice and game. **Main Outcome Measures:** Cervical muscle strength, size, and stiffness, visual performance measures, level of anticipation, and head impact biomechanical measures. **Results:** Football players with greater cervical stiffness had reduced odds of sustaining higher magnitude head impacts, rather than head impacts in the 1st quartile, compared to players with less cervical stiffness. Surprisingly, players with stronger and larger cervical musculature had increased odds of sustaining higher magnitude head impacts, rather than head impacts in the 1st quartile, compared to players with weaker and smaller cervical musculature. Players with better near-far quickness, target capture, and reaction time performance had increased odds of sustaining higher magnitude head impacts, rather than head impacts in the 1st quartile. Head impact biomechanical measures did not differ between anticipated and unanticipated collisions. **Conclusions:** Neuromuscular training aimed at enhancing cervical muscle stiffness may be useful in reducing the magnitude of head impacts sustained while playing football. The results of this study do not support the theory that players with stronger and larger cervical musculature are better able to mitigate head impact severity. Vision and level of anticipation may play less of a role than expected for protecting against higher magnitude head impacts among high school football players. In summary, cervical stiffness plays a role in mitigating head impact severity, but the roles of cervical strength, visual performance, and level of anticipation need further study.

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INTRODUCTION

As many as 3.8 million sports-related traumatic brain injuries occur each year, not counting injuries that go unreported (Langlois, Rutland-Brown, & Wald, 2006; McCrea, Hammeke, Olsen, Leo, & Guskiewicz, 2004). Concussion is defined as “a complex pathophysiological process affecting the brain, caused by traumatic biomechanical forces” (McCrory, et al., 2009). Acutely, concussed athletes experience diminished cognitive function, altered motor control, and symptoms such as headache, nausea, and dizziness (Guskiewicz, Ross, & Marshall, 2001; McCrea, et al., 2005; McCrea, et al., 2003). Sport-related concussion is of particular concern in youth athletes because younger athletes are more susceptible to sustaining concussions (Buzzini & Guskiewicz, 2006; Gessel, Fields, Collins, Dick, & Comstock, 2007; Guskiewicz, Weaver, Padua, & Garrett, 2000). In fact, concussion incidence rates among high school football players are higher than in any of the three collegiate divisions (Guskiewicz, et al., 2000). Concussions can have severe acute and long-term consequences for youth athletes because of the ongoing neurocognitive development that occurs throughout adolescence (Patel & Greydanus, 2002).

Although most athletes recover from concussion within seven to ten days after injury (Guskiewicz, et al., 2001; McCrea, et al., 2005; McCrea, et al., 2003), a growing body of literature suggests that athletes with a history of concussion are at higher risk for depression, mild cognitive impairment, and early onset Alzheimer's later in life (Dale, Leigh, Luthert, Anderton, & Roberts, 1991; Guskiewicz, et al., 2005; Guskiewicz, Marshall, et al., 2007). Some speculate that these debilitating conditions could also result from the cumulative effects of the thousands of subconcussive (non-injurious) impacts to the head that athletes experience throughout their careers (Spiotta, Shin, Bartsch, & Benzel, 2011). For high school athletes who continue to play in college and then professionally, exposure to a high number of cumulative head impacts may increase their risk of developing neurodegenerative disorders during late-life. Animal studies have demonstrated that higher magnitude impacts to the head or body cause the brain to accelerate and decelerate rapidly within the skull resulting in greater brain tissue strain (Ommaya & Gennarelli, 1974). Extrapolating these data to humans, it seems possible that reducing the magnitude of head impacts that athletes sustain during sport participation may reduce the risk of concussion, the severity of subconcussive head impacts, and, subsequently, the risk of developing late-life cognitive declines that some have speculated are associated with concussions and repetitive brain trauma. However, little research is available addressing the modifiable factors that could help mitigate the severity of head impacts that result during sport, leaving sports medicine professionals with limited options for preventing concussion. Cervical muscle characteristics, visual performance, and the ability to anticipate impending collisions are three modifiable factors that could potentially be targeted to reduce head impact magnitude during sport.

Cervical Muscle Characteristics

Since the cervical musculature contributes 80% of the stability necessary to resist injurious forces to the cervical spine (Panjabi, et al., 1998), athletes with an insufficient cervical musculature response may be predisposed to concussion because they are less able to generate adequate internal preparatory and reactive forces to counter head acceleration (Viano, Casson, & Pellman, 2007). Contraction of the cervical musculature strong enough to make the cervical spine a rigid segment is believed to link the head, neck, and thorax as a single segment. If inadequate force is generated rapid acceleration of the head occurs. Force imparted to the athlete during a collision is theoretically dispersed over the effective mass of the head, neck, and thorax segments combined, thereby reducing head acceleration. When the cervical musculature is not fully contracted, such as when a player receives an unexpected hit, the impact force is imparted to the head rather than across the neck to the thorax. It seems possible that as cervical muscle activity increases players simultaneously experience a proportional decrease in the severity of head impact. Previous studies have manipulated neck tension in Hybrid III anthropometric head models and have observed that increasing neck tension resulted in a 35% decline in concussion risk, as measured by the head injury criterion, based on laboratory measures (Viano, et al., 2007). The role of the cervical musculature in modifying head impact forces remains unclear in human models (Mihalik, et al., 2011; Tierney, et al., 2008; Tierney, et al., 2005). Possessing certain anatomical and dynamic cervical spine characteristics may enable an athlete to better increase his or her effective head-neck-thorax mass, making the player better prepared to limit rapid head acceleration. However, the role of cervical muscle strength, physiological cross-sectional area, stiffness, and muscle activation in reducing in vivo head acceleration remains unknown.

Visual Performance

The eyes supply sensory information to the brain, the brain then decodes and integrates the visual information while also considering vestibular and somatosensory information (Zimmerman, Lust, & Bullimore, 2011). The brain then sends out an appropriate motor signal to the muscles based on the supplied sensory information. Many sports involve quick and unpredictable movement of an object, teammates, and competitors. Athletes must be able to accurately perceive and identify both static and dynamic features, scan and interpret visual information at differing contrast levels, alternate between focusing on objects at varying distances, perform efficient eye movements, and respond quickly to visual stimuli (Henderson & Hollingworth, 2003; Zimmerman, et al., 2011; Zupan, Arata, Wile, & Parker, 2006). Numerous studies conclude that athletes demonstrate better visual abilities than non-athletes, and that elite athletes have visual abilities superior to novice athletes (Hitzeman & Beckerman, 1993; Laby, et al., 1996; Stine, Arterburn, & Stern, 1982; Uchida, Kudoh, Murakami, Honda, & Kitazawa, 2012). It seems possible that enhanced visual performance would allow an athlete to better anticipate impending collisions with other players allowing them to better mitigate head impact severity.

Although the importance of visual performance in sport is widely accepted, detailed assessments are not often completed in athletic settings. Several studies have identified superior visual performance among elite athletes; however, how these differences relate to sport performance and injury prevention is not yet known (Zimmerman, et al., 2011). Although visual training in athletes is a relatively new concept, studies suggest that visual exercises improve visual performance (Maxwell, Tong, & Schor, 2012). Further research is needed to determine if visual performance influences head impact biomechanics.

Collision Anticipation

Anticipatory responses to impending head or body collisions may help mitigate acceleration of the head, thereby reducing the potential risk for sustaining a brain injury and reducing the magnitude of subconcussive impacts. An athlete that is able to foresee an impending impact will instinctively and cognitively react with anticipatory responses, such as leaning, using the arms to block the face, or recoiling the head by elevating the shoulders (Metoyer, Zordan, Hermens, Wu, & Soriano, 2008). During sport, athletes must maintain gaze fixation on a target area, such as a goal or ball, for accurate aiming. Gaze fixation may limit the athlete's ability to foresee and prepare for impending impacts (van der Kamp, 2011). In youth ice hockey, unanticipated collisions tend to result in more severe head impact magnitudes than anticipated collisions (Mihalik, Blackburn, et al., 2010). In contact sports, the striking player prepares for impending collision by aligning the head, neck, and thorax to impart maximum force on an opponent by driving through the struck player. Previous studies, that have modeled helmet-to-helmet impacts, show that the struck players, on average, experienced greater linear and rotational head acceleration than the striking player (Viano, et al., 2007). However this study used a small sample of head impacts that were reconstructed in a laboratory setting and anthropometric models that lack the ability to anticipate an impending collision. Because the striking player fully anticipates the impending collision he imparts much greater force on the struck player. Thus, further research is necessary to determine the effect of collision anticipation on head impact severity among high school football players.

Specific Aims

1. To evaluate the effect of cervical musculature characteristics measured during the preseason on head impact biomechanics sustained in-season among high school and collegiate football players.
2. To evaluate the effect of visual performance measured during the preseason on head impact biomechanics sustained in-season among high school football players.
3. To evaluate the effect of level of anticipation at the time of collision on head impact biomechanics among high school football players.
4. To determine if preseason measures of cervical musculature characteristics and visual performance, and level of anticipation at the time of collision predict head impact biomechanics among high school and collegiate football players.

Variables

Independent Variables

1. **RQ1:** High and low performance on the following cervical musculature characteristics:
    a. Composite peak torque
    b. Composite rate of torque development
    c. Composite cross-sectional area
    d. Composite stiffness
    e. Composite angular displacement
    f. Composite muscle onset latency
2. **RQ2:** High and low performance on the following visual performance:

    a. Visual acuity
    b. Contrast sensitivity
    c. Depth perception
    d. Near-Far quickness
    e. Target capture
    f. Perception span
    g. Eye-Hand coordination
    h. Go/No Go
    i. Reaction Time
3. **RQ3:** Level of anticipation
    a. Anticipated
    b. Unanticipated
4. **RQ4a:** Predicting Head Impact Biomechanics
    a. Composite peak torque
    b. Composite rate of torque development
    c. Composite cross-sectional area
    d. Composite stiffness
    e. Composite muscle onset latency
    f. Visual acuity
    g. Contrast sensitivity
    h. Depth perception
    i. Near-Far quickness
    j. Target capture

k. Perception span

l. Eye-Hand coordination

m. Go/No Go

n. Reaction Time

o. Level of Anticipation

5. **RQ4b:** Predicting Head Impact Biomechanical Profiles

   a. Composite peak torque

   b. Composite rate of torque development

   c. Composite cross-sectional area

   d. Composite stiffness

   e. Composite muscle onset latency

   f. Visual acuity

   g. Contrast sensitivity

   h. Depth perception

   i. Near-Far quickness

   j. Target capture

   k. Perception span

   l. Eye-Hand coordination

   m. Go/No Go

   n. Reaction Time

Dependent Variables

1. **RQ 1 & 2:** Categorized Head Impact Biomechanical Measures

    a. Frequency of categorized head impact magnitude by linear acceleration (1st quartile, 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, 99th percentile)
    b. Frequency of categorized head impact magnitude by rotational acceleration (1st quartile, 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, 99th percentile)
    c. Frequency of categorized head impact magnitude by Head Impact Technology Severity Profile (HITsp) (1st quartile, 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, 99th percentile)
2. **RQ 3 & 4a:** Game Head Impact Biomechanical Measures
    a. Peak linear acceleration
    b. Peak rotational acceleration
    c. Head Impact Technology Severity Profile (HITsp)
3. **RQ 4b:** Cumulative Game Head Impact Biomechanical Measures Per Play Exposure
    a. Cumulative game linear acceleration per play exposure
    b. Cumulative game rotational acceleration per play exposure
    c. Cumulative game HITsp per play exposure
4. **RQ 4b:** Cumulative Game Head Impact Frequency Per Play Exposure

Research Questions

This study focused on the three following head impact biomechanical measures: 1) linear acceleration, 2) rotational acceleration, and 3) HITsp.

Research Question 1: Cervical Muscle Characteristics

We split football players into a group of high and a group of low performers for each cervical characteristic measure.

a. Do football players with high and low preseason ***composite cervical peak torque*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

b. Do football players with high and low preseason ***composite cervical rate of torque development*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

c. Do football players with high and low preseason ***composite cervical cross-sectional area*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

d. Do football players with high and low preseason ***composite cervical stiffness*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

e. Do football players with high and low preseason ***composite cervical angular displacement*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

f. Do football players with high and low preseason ***composite cervical muscle onset latency*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

Research Question 2: Visual performance

We split high school football players into a group of high and a group of low performers for each visual performance measure.

a. Do football players with high and low preseason ***visual acuity*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

b. Do football players with high and low preseason ***contrast sensitivity*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

c. Do football players with high and low preseason ***depth perception*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

d. Do football players with high and low preseason ***near-far quickness*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

e. Do football players with high and low preseason ***target capture*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?

f. Do football players with high and low preseason ***perception span*** performance differ in odds of sustaining head impacts in 2^{nd} quartile, 3^{rd} quartile, 4^{th} quartile, 95^{th} percentile, or 99^{th} percentile, rather than head impacts in the 1^{st} quartile?

g. Do football players with high and low preseason ***eye-hand coordination*** performance differ in odds of sustaining head impacts in 2^{nd} quartile, 3^{rd} quartile, 4^{th} quartile, 95^{th} percentile, or 99^{th} percentile, rather than head impacts in the 1^{st} quartile?

h. Do football players with high and low preseason ***go/no go*** performance differ in odds of sustaining head impacts in 2^{nd} quartile, 3^{rd} quartile, 4^{th} quartile, 95^{th} percentile, or 99^{th} percentile, rather than head impacts in the 1^{st} quartile?

i. Do football players with high and low preseason ***reaction time*** performance differ in odds of sustaining head impacts in 2^{nd} quartile, 3^{rd} quartile, 4^{th} quartile, 95^{th} percentile, or 99^{th} percentile, rather than head impacts in the 1^{st} quartile?

Research Question 3: Level of Anticipation

a. Is there a significant difference in head impact biomechanical measures between anticipated and unanticipated collisions in high school football players?

Research Question 4: Predicting Head Impact Biomechanical Measures

a. Do cervical muscle characteristics, visual performance, and level of anticipation predict ***game head impact biomechanical measures*** in high school football players?

b. Do preseason cervical characteristics and visual performance predict ***cumulative game head impact biomechanical measures per play exposure*** in high school and collegiate football players?

c. Do preseason cervical characteristics and visual performance predict ***cumulative game head impact frequency per play exposure*** in high school and collegiate football players?

Research Hypotheses

Research Hypotheses for Research Question 1: Cervical Characteristics

a. Football players that are high performers on ***composite cervical peak torque*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

b. Football players that are high performers on ***composite cervical rate of torque development*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

c. Football players that are high performers on ***composite cervical cross-sectional area*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

d. Football players that are high performers on ***composite cervical stiffness*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th

quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

e. Football players that are high performers on ***composite cervical angular displacement*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

f. Football players that are high performers on ***composite cervical muscle onset latency*** will have a reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

Research Hypotheses for Research Question 2: Visual performance

a. Football players that are high performers on ***visual acuity*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

b. Football players that are high performers on ***contrast sensitivity*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

c. There will be no differences in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile between high and low performers on ***depth perception***.

d. Football players that are high performers on ***near far quickness*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

e. There will be no differences in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile between high and low performers on ***target capture***.

f. Football players that are high performers on ***perception span*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

g. Football players that are high performers on ***eye-hand coordination*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

h. There will be no differences in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile between high and low performers on ***go/no go***.

i. Football players that are high performers on ***reaction time*** will have reduced odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, and 99th percentile, rather than head impacts in the 1st quartile.

Research Hypothesis for Research Question 3: Level of Anticipation

a. Unanticipated collisions will result in significantly higher head impact biomechanical measures than anticipated collisions.

Research Hypotheses for Research Question 4: Predicting Head Impact Severity

a. Composite cervical peak torque, rate of torque development, stiffness, cross-sectional area, contrast sensitivity, near-far quickness, perception span, and level of anticipation will be significant inverse predictors of ***game head impact***

biomechanical measures. Muscle onset latency, visual acuity, depth perception, target capture, eye-hand coordination, and reaction time will significant direct predictors of game head impact biomechanical measures. Depth perception and go/no go will not be significant predictors.

b. Composite cervical peak torque, rate of torque development, stiffness, cross-sectional area, and composite visual performance raw score will be significant inverse predictors of ***cumulative game head impact biomechanical measures*** while controlling for play exposure. Muscle onset latency will be significant direct predictors of mean game head impact biomechanical measures.

c. Composite visual performance raw score will be a significant inverse predictor of ***cumulative game head impact frequency*** while controlling for play exposure. Composite cervical peak torque, rate of torque development, stiffness, cross-sectional area, and muscle onset latency will not be significant predictors.

Operational Definitions

1. *Head Impact Technology severity profile (HITsp):* A weighted composite score including linear acceleration, rotational acceleration, impact duration, and impact location.
2. *Cervical Characteristics:*
    a. *Composite peak torque:* A calculated sum of the peak torque generated by the cervical flexors, extensors, right lateral flexors, and left lateral flexors normalized to body mass in kilograms.

b. *Composite rate of torque development:* A calculated sum of the rate of torque development of the cervical flexors, extensors, right lateral flexors, and left lateral flexors. Rate of torque development is defined as the maximal value of the slope of the force-time curve, calculated using a 50-millisecond sliding window from onset to peak force (Almosnino, Pelland, & Stevenson, 2010).

c. *Composite cross-sectional area:* A calculated sum of the cross-sectional area of the sternocleidomastoid, upper trapezius, and semispinalis capitis measured using ultrasonographic imaging.

d. *Composite stiffness:* A calculated sum of flexor and extensor stiffness. Stiffness is a measure of an elastic body's resistance to deformation. Flexor stiffness was determined by measuring the flexor muscle group's resistance to deformation during forced extension after an applied load. Extensor stiffness was determined by measuring the extensor muscle group's resistance to deformation during forced flexion after an applied load.

e. *Composite angular displacement:* A calculated sum of peak angular displacement of the head relative to the thorax following perturbation into both flexion and extension.

f. *Composite muscle onset latency:* The sum of the duration of time between force application and the onsets of myoelectric activity in the sternocleidomastoid and upper trapezius.

3. *Visual performance:* Visual performance measures in this study include: visual acuity, contrast sensitivity, depth perception, near-far quickness, target capture, perception span, eye-hand coordination, go/no go, and reaction time.
4. *Level of anticipation:* Level of anticipation was determined by evaluating video of each head impact sustained during games. The Player to Player form was used to grade the player's relative body position at the time of impact (Mihalik, Blackburn, et al., 2010; Ocwieja, et al., 2012).
    a. *Anticipated:* An impact occurring while the athlete is looking in the direction of the impending collision, is in a general athletic readiness position (knee and trunk flexion with feet shoulder-width apart), and uses their legs to drive their shoulders through the collision.
    b. *Unanticipated:* An impact occurring where the athlete is looking in the direction of the oncoming collision but is not in an athletic readiness position or an impact occurring while the athlete is not looking in the direction of the impending collision.
    c. *Unknown:* Collisions where the investigator is unable to identify the direction of gaze or the positioning of the body.
5. *Play exposure:* The number of plays that each athlete participates in during all games throughout the entire season as recorded by the primary investigator.
6. *Cumulative Game Head Impact Biomechanical Measures Per Play Exposure:*
    a. Cumulative Game Linear Acceleration Per Play Exposure: The average linear acceleration per play, computed as the sum of the linear acceleration

(g) from head impacts sustained in all games divided by the number of play exposures.

   b. *Cumulative Game Rotational Acceleration Per Play Exposure:* The average rotational acceleration per play, computed as the sum of the rotational acceleration (rad/sec^2) from head impacts sustained in all games divided by the number of play exposures.

   c. *Cumulative Game HITsp Per Play Exposure:* The average HITsp per play, computed as the sum of the HITsp from head impacts sustained in all games divided by the number of play exposures.

7. *Cumulative Head Impact Frequency:* The average number of head impacts per play, computed as the sum of the frequency of head impacts sustained in all games divided by the number of play exposures.

Assumptions

1. Preseason measures of cervical characteristics and visual performance reflect the changes that occur over the course of the season.
2. Participants gave their best efforts during pre and post-season testing sessions
3. These lab measures accurately reflect cervical function in the athletic setting.
4. The head-neck segment moves about the thorax as a rigid body
5. Athletes did not alter their sport technique due to the presence of the instrumentation and investigators.

Limitations

1. The Nike Sensory Station measures have not yet been validated.

2. Cervical isometric strength and stiffness were measured in just one plane at a time.
3. Head to thorax movement measured during cervical perturbation does not account for movement of individual vertebrae or movement of the head relative to C1.
4. The Head Impact Telemetry system does not measure rotational acceleration about the Z-axis.
5. Results from this study may not apply to athletes that participate at other levels of play or female athletes.
6. We were not able to determine if cervical characteristics, visual performance, and level of anticipation influence the odds of sustaining a concussion.

Delimitations

1. Data collection was limited to only practices and games during a single competitive season.
2. Athletes were recruited from a single high school and single collegiate institution.
3. This study did not examine impacts to the head that result in concussion.
4. Participants were all males.

Significance of the Study

The primary purpose of this investigation is to determine if cervical muscle characteristics, visual performance, and level of anticipation affect biomechanics of head impacts that high school and collegiate athletes sustain while playing football. Dynamic stabilization of the head using the cervical musculature is a potentially modifiable factor that might influence concussion risk. The results of this study may aid sports medicine clinicians in isolating important cervical characteristics that put athletes at higher odds for sustaining high magnitude impacts to the head. This study provides guidance for designing neck strength and conditioning programs for later intervention studies. The role of visual performance in mitigating head impact severity has not previously been studied. If the results of our study suggest that visual performance does play a role mitigating head impact severity, then sports medicine professionals should place an emphasis on optimizing each athlete' visual conditions, either through vision correction or training. If the results of this study agree with previous trends seen among youth ice hockey players that higher levels of anticipation reduce head impact severity, future studies could examine the utility of anticipation training for reducing head impacts.

Chapter 2

REVIEW OF LITERATURE

Introduction

Sport-related mild traumatic brain injury is a major public health concern in the United States (Langlois, et al., 2006). Concussions result from rapid acceleration and deceleration of the brain caused by biomechanical forces transmitted from an impact to the head or indirectly through the body (McCrory, et al., 2009). Athletes with insufficient cervical musculature strength may be predisposed to more severe head impacts because they are less able to generate adequate internal preparatory and reactive force to counter head acceleration (Viano, et al., 2007). The purpose of this review is to discuss relevant literature regarding concussion epidemiology, neurometabolic cascades that follow traumatic brain injury, development and recovery of the adolescent brain, negative postconcussive outcomes, head impact biomechanics, important cervical characteristics, sport visual performance, and anticipation.

Epidemiology

Understanding the epidemiology of sport-related concussions is essential for improving safety in athletics. Sports are second only to motor vehicle accidents as the leading cause of traumatic brain injury among young people ages 15-24 years (Sosin, Sniezek, & Thurman, 1996). By observing epidemiologic patterns in sport-related concussion, sports medicine professionals can guide targeted preventive measures.

As many as 3.8 million sport-related traumatic brain injuries occur annually in the United States (Langlois, et al., 2006), with evidence that many go unrecognized, unreported, and untreated (Langlois, et al., 2006; McCrea, et al., 2004; Valovich McLeod, Schwartz, & Bay, 2007). Evaluation and treatment of sport-related concussion cost approximately 60 billion dollars each year (Langlois, et al., 2006). Concussions represent 13.2% of all sport-related injuries reported in the high school setting (Marar, McIlvain, Fields, & Comstock, 2012). Earlier epidemiologic studies report slightly lower incidences of concussion ranging from 5.5-8.9% of all injuries, but this is likely because these studies have not included contact sports like ice hockey and lacrosse (Gessel, et al., 2007; Powell & Barber-Foss, 1999; Schulz, et al., 2004). Overall, the concussion rate is approximately 2.5 concussions per 10,000 athlete exposures (Gessel, et al., 2007; Marar, et al., 2012).

Pediatric Brain Injuries

Nearly half of all concussions among youth and adolescents result during participation in sport (Bakhos, Lockhart, Myers, & Linakis, 2010; Meehan & Mannix, 2010). High school athletes have a higher incidence of concussion compared to their collegiate counterparts (Gessel, et al., 2007; Guskiewicz, et al., 2000). Some researchers theorize that adolescent athletes have less protection for their developing nervous system because they have relatively decreased neuronal myelination, a greater head-to-body ratio, and thinner cranial bones (Buzzini & Guskiewicz, 2006). High school athletes that sustain a recurrent concussion are more likely to take more than one month to experience full resolution of symptoms, take longer than three weeks to return to sport, and are more likely to be medically disqualified from sport (Castile, Collins, McIlvain, & Comstock, 2011). Athletes who sustain their first concussion at a young age and continue to participate in sport on into high school and college have a longer window of time they are participating in sports which increases their exposure and therefore risk of re-injury (Guskiewicz & Valovich McLeod, 2011). Symptoms that persist following concussion in adolescent athletes are particularly concerning because these deficits can significantly affect academic performance and social function during a critical period of development (Blume, Lucas, & Bell, 2011).

Gender Comparisons

Among gender-comparable sports, females have a higher concussion rate than males (Castile, et al., 2011; Gessel, et al., 2007; Marar, et al., 2012). Likewise, females have higher rates of recurrent concussions than males (Castile, et al., 2011). Some researchers and clinicians speculate that observed gender differences could be attributable to head and cervical biomechanical differences (Mansell, Tierney, Sitler, Swanik, & Stearne, 2005; Tierney, et al., 2008; Tierney, et al., 2005). However, concussion rates may differ between genders because female athletes are generally more honest about reporting injuries than male athletes, due to cultural norms (Dick, 2009).

Injury Mechanisms

Among high school football players, the highest proportion of concussions result from player-player contact like tackling or being tackled (Castile, et al., 2011; Gessel, et al., 2007; Marar, et al., 2012). A majority of epidemiologic studies identify that concussion rates are greater during competition compared to practice (Gessel, et al., 2007; Marar, et al., 2012; Schulz, et al., 2004). This could be explained by evidence in football that collisions that take place after two players travel a longer closing distances and in ice hockey where collisions occur on the open ice result in higher magnitude head impacts (Mihalik, Blackburn, et al., 2010; Ocwieja, et al., 2012). Some studies suggest that football players sustain a greater number of impacts and more severe impacts during games compared to practices (Broglio, et al., 2009). However, a similar study in a college sample suggests that head impacts sustained during helmets-only and full-contact practices are more severe than head impacts sustained during games (Mihalik, Bell, Marshall, & Guskiewicz, 2007).

Football Brain Injuries

Among high school sports, football accounts for nearly half of all reported concussions and has the highest concussion rate (Gessel, et al., 2007; Marar, et al., 2012). High school football is followed by boy’s ice hockey and boy’s lacrosse. Concussion incidence rates among high school football players have been reported to be higher than any of the three collegiate divisions. In fact, compared to the division I setting, where athletes are thought to be stronger and faster, high school players had nearly twice the concussion rate (Guskiewicz, et al., 2000). Higher incidences of concussion among high school athletes may be caused by increased exposure often seen at the high school and division III collegiate levels, such as when a football player plays both offense and defense (Guskiewicz, et al., 2000). Estimates of concussion incidence may reflect an underestimation of the true occurrence of the injury. Among high school football players that had sustained a sport-related concussion, only 47% reported the injury at the time

(McCrea, et al., 2004). Because athletes may be largely unaware of the signs and symptoms of concussion and the seriousness of premature return to play, prevalence of concussion in high school football is likely higher than previously published epidemiology literature.

Neurometabolic Cascade Following Concussion

Post-concussion deficits occur in the absence of detectable structural pathology and typically resolve completely over time. Neuronal dysfunction following concussion result from ionic shifts, altered metabolic demand, impaired neuronal connectivity, and changes in neurotransmission (Giza & Hovda, 2001). Understanding the neurometabolic cascade that follows concussion is vital for understanding the underlying pathophysiology.

Traumatic brain injury sets off a complex and interwoven sequence of ionic and metabolic events from which damaged cells may eventually recover, or in certain cases, degenerate and dies. Membrane disruption and axonal stretch caused by a direct or indirect impact to the head, results in opening of voltage-dependent potassium channels and a subsequent efflux of potassium from cells to the extracellular space. Potassium is released into the extracellular space by leaking through the mechanically stretched cell membrane and by passing through voltage-gated potassium channels (Katayama, Becker, Tamura, & Hovda, 1990; Takahashi, Manaka, & Sano, 1981). Non-specific depolarization of neurons leads to the release of glutamate, an excitatory neurotransmitter. Glutamate activates N-methyl-D-aspartate (NMDA) and D-amino-3-hydroxy-5-methyl-4-isoxazole-propionic acid receptors (AMPA), which further exacerbate the potassium efflux. In an attempt to restore the membrane potential, sodium and potassium channels work overtime, but simultaneously consume increasing amounts of adenosine triphosphate (Mayevsky & Chance, 1974; Rosenthal, LaManna, Yamada, Younts, & Somjen, 1979). To meet elevated adenosine triphosphate requirements, there is a marked upregulation of cellular glycolysis, which occurs within minutes after brain injury (Ackermann & Lear, 1989). Hypergycolysis results in lactate byproduct, which builds up within the neuron (Nilsson & Nordstrom, 1977; Nilsson & Ponten, 1977; Yang, DeWitt, Becker, & Hayes, 1985).

In addition to potassium efflux, NMDA receptor activation permits a rapid and sustained influx of calcium. Elevated intracellular calcium can be sequestered by the mitochondria, but will eventually lead to dysfunction of oxidative metabolism, which further increases the cell 's dependence on glycolysis-generated adenosine triphosphate (Giza & Hovda, 2001). Calcium accumulation eventually leads to cell dysfunction, damage, and sometimes death. Ionic shifts and acute alterations in cellular energy metabolism occur during a period when cerebral blood flow is reduced (Yamakami & McIntosh, 1989; Yuan, Prough, Smith, & Dewitt, 1988). Imbalance between glucose delivery and glucose consumption predisposes neurons to secondary injury and secondary cell death (Giza & Hovda, 2001). After the initial period of ionic disturbance and increase in glucose metabolism, the local cerebral metabolic rate for glucose and oxidative metabolism decrease significantly below baseline (Yoshino, Hovda, Kawamata, Katayama, & Becker, 1991). Depressed glucose and oxidative metabolism does not normalize until between five and ten days following injury, which possibly limits the brain’s ability to respond adequately to subsequent changes in energy demand (Giza & Hovda, 2001).

The Adolescent Versus the Adult Brain

Cognitive and cortical growth generally occurs in cycles, with a series of sporadic spurts and drops. High school athletes are still developing in cognitive areas like concentration, memory, reasoning, and problem solving (Hunt & Ferrara, 2009). The rate of progression through each phase of cognitive and cortical development differs between individuals, but most individuals go through a common developmental process (Fisher & Rose, 1998). Passing through these phases requires reorganization and simplification, which allows the individual to move through the four different tiers: reflex, action, concrete representation, and abstraction (Fisher & Rose, 1998). Once the infant moves beyond the reflexive tier, the action tier is identifiable as the infant begins building complex sensorimotor actions, typically between three months and two years (e.g. names, emotions). Between ages two and 12, the child develops concrete representational capacities and eventually understands his or her first abstractions (understanding mathematic calculations, literary meanings, concepts of law). Optimal abstraction capacities appear between 10 and 25 years of age and produce the capacity to build principles relating multiple abstractions (Fisher & Rose, 1998). Although most cognitive and cortical development is complete by the time an individual enters college, development continues on into early adulthood (Luna, et al., 2001).

As more and more children and adolescents participate in organized sports and sustain head injuries, understanding the effects of concussion and subconcussive head impacts on the maturing brain becomes increasingly important. Two general theories exist regarding pediatric recovery following concussion. Some argue that younger brains are more resilient and recover more effectively following concussion (Giza & Hovda, 2001). Recent research using juvenile rats support this concept. Compared to adult rats, younger rats show longer periods of apneas, shorter periods of unconsciousness, present with post-percussion hypotension, and have higher mortality following traumatic brain injury (Prins, Lee, Cheng, Becker, & Hovda, 1996). Despite displaying more severe immediate response to brain injury, younger rats with mild and moderate traumatic brain injury continue to perform well on spatial learning tasks (Prins & Hovda, 1998). However, when moderately concussed juvenile rats are reared in an enriched environment, they fail to develop increased cortical thickness and enhanced cognitive performance seen in sham-injured rats raised in the same enriched environments (Fineman, Giza, Nahed, Lee, & Hovda, 2000). Brain injury that occurs in the developing brain, even without early signs of damage, may lead to impaired plasticity.

Long-term deficiencies have been observed in human research, as well. Symptom-free high school athletes with a history of two or more concussions perform similarly on neurocognitive testing to athletes who have just experienced a recent concussion (Moser, Schatz, & Jordan, 2005). It seems that the harmful effects of multiple concussions on the developing pediatric brain are cumulative, but the degree to which this may affect the youth athlete later in life is not yet known. Compared to collegiate athletes, high school athletes experience delayed recovery of cognitive function and self-reported symptoms (Field, Collins, Lovell, & Maroon, 2003; Sim, Terryberry-Spohr, & Wilson, 2008). More research is necessary to determine recovery patterns for adolescent athletes, however, full recovery should generally be expected to take longer in adolescent athletes than in collegiate athletes (Guskiewicz & Valovich McLeod, 2011).

Negative Post-Concussion Outcomes

Research regarding both the short- and long-term effects of concussion has raised considerable concern about brain function. Negative post-concussion outcomes include second impact syndrome, post-concussion syndrome, recurrent concussion, chronic traumatic encephalopathy, cognitive decline, and depression.

Second Impact Syndrome

Second impact syndrome is defined as occurring when ‘‘an athlete who has sustained an initial head injury, most often a concussion, sustains a second head injury before symptoms associated with the first have fully cleared’’ (Cantu, 1998). A second insult to the brain, sometimes occurring from a seemingly innocuous hit to the head or body, that occurs prior to brain recovery is thought to results in catastrophic brain swelling. Although second impact syndrome is undoubtedly the most severe negative outcome that could occur following concussion, evidence supporting the existence of second impact syndrome remains anecdotal (McCrory, 2001; McCrory, Davis, & Makdissi, 2012; Randolph, 2011). Brain swelling is a common result from a head injury; however, it remains unknown whether a second concussive injury is a risk factor for this condition. Although it seems logical that returning an athlete to play before concussion related symptoms have resolved could increase the athletes vulnerability to negative postconcussive outcomes, the number of athletes that prematurely return to play without negative consequences is still unknown.

Post-concussion Syndrome

A majority of concussed athletes experience spontaneous recovery approximately seven to ten days following injury (Guskiewicz, et al., 2001; McCrea, et al., 2003). However, a small, but clinically significant number of athletes experience post-concussion syndrome, which consists of a complex mixture of cognitive, behavioral, and physical symptoms that persists for an extended period of time after the concussion (Jotwani & Harmon, 2010; Williams, Potter, & Ryland, 2010). Definitions of post-concussion syndrome differ across diagnostic criteria, resulting in widespread confusion about identifying and treating athletes with prolonged recoveries (Jotwani & Harmon, 2010). A reliable and consistent definition is necessary to further scientific research and provide clarity to clinical decisions regarding post-concussion syndrome.

Some authors speculate that persistent post-concussion symptoms are a consequence of psychological illness rather than brain injury (Lishman, 1988; Williams, et al., 2010). Some literature suggests the stress triggered by brain injury results in depression and anxiety, which disrupts concentration and other mental operations. In a prospective longitudinal study, Yeates et al. (Yeates, et al., 2009) observed that severity of head injury predicted post-concussion symptoms in most but not all patients contradicting the thought that post-concussion syndrome is caused by psychological illness. To date, no research has been able to biologically differentiate post-concussion syndrome from posttraumatic stress disorder (Rees, 2003). Persistent post-concussion symptoms can result for different reasons in different patients, and thus, each case of post-concussion syndrome should be evaluated differently. Clinicians still struggle to identify athletes that are at high risk for developing post-concussion syndrome. Those athletes who experience noise sensitivity, previous history of migraine headaches, or amnesia may be more likely to have prolonged symptomatology. Also, athletes with a history of previous concussion and those who have preexisting psychiatric issues could be at higher risk of developing post-concussion syndrome (Jotwani & Harmon, 2010).

Recurrent Concussion

Like many sports injuries, history of similar injury is the best predictor of recurrent injury. Epidemiologic studies have time and time again identified a history of previous concussions as a risk factor for suffering recurrent concussion (Gerberich, Priest, Boen, Straub, & Maxwell, 1983; Guskiewicz, et al., 2000; Schulz, et al., 2004). It is possible that the brain's ability to respond to traumatic insults may be compromised in previously concussed athletes making them more susceptible to another concussion. The risk of recurrent concussion in the youth and adolescent athletes is currently unknown (Guskiewicz & Valovich McLeod, 2011). Although a few studies indicate that a previous history of concussion may increase an athlete's risk of sustaining additional concussion, these trends could be attributable to the fact that these same athletes may continue to be exposed to more play-time, may exhibit risky biomechanics, or be exposed to more intense athletic activities.

Chronic Traumatic Encephalopathy

Chronic traumatic encephalopathy describes the presence of tau protein within the cerebral tissue that results in neurologic deterioration and is only observed among individuals with a history of repetitive impacts to the head (Stern, et al., 2011). It has been hypothesized that repetitive axonal stretching, caused by repetitive impacts to the head, triggers the neurodegenerative cascade. Individuals with a history of previous concussion or a history of exposure to subconcussive impacts are suspected to have undergone sustained axonal stretching and deformation that later triggers neurocognitive decline (Yuen, Browne, Iwata, & Smith, 2009). Sport type, level of competition, position, and playing career duration may all influence an athlete's risk of developing chronic traumatic encephalopathy (Stern, et al., 2011). All diagnosed cases of chronic traumatic encephalopathy have a history of brain trauma exposure, but controversy exists over the risk of exposure to brain trauma because not all individuals with exposure to repetitive brain trauma develop chronic traumatic encephalopathy.

The clinical presentation of chronic traumatic encephalopathy is distinct from post-concussion syndrome because patients do not present with unrelenting symptoms immediately following a concussion. Rather, the symptoms of chronic traumatic encephalopathy result from a progressive, but gradual, decline in neuronal function (McKee, et al., 2009). Typically, chronic traumatic encephalopathy symptoms present in midlife as cognitive, emotional, and behavioral symptoms, usually decades after exposure to repetitive brain trauma. Behavioral symptoms are often the most concerning since they present as a depressed mood, apathy, emotional instability, suicidal tendencies and behaviors, and problems with impulse control (Stern, et al., 2011).

Cognitive Decline

Higher rates of clinically diagnosed mild cognitive impairment, an intermediary stage between the normal cognitive changes and dementia, have previously been reported among retired professional football players with a history of three or more concussions (Guskiewicz, et al., 2005). Likewise, trends towards earlier onset and a higher prevalence of Alzheimer's disease have been previously been identified in retired professional football players relative to the general American male population (Guskiewicz, et al., 2005). Although additional prospective research is necessary to determine how exposure to head trauma influences the onset of dementia-related syndromes in athletes, these studies present compelling evidence that mild cognitive impairment may be initiated by multiple concussions. More acutely, a history of multiple concussions has been associated with reduced neurocognitive performance, increased symptom severity, and delayed resolution of concussion related symptoms (Collins, et al., 1999; Colvin, et al., 2009; Guskiewicz, et al., 2003; Guskiewicz, et al., 2000; Iverson, Gaetz, Lovell, & Collins, 2004). These short-term consequences of recurrent concussion support the findings regarding the more chronic consequences of years of playing football. Further research is necessary to further elucidate whether cognitive impairment, both short- and long-term, results from sport-related concussion.

Depression

Many patients that suffer a traumatic brain injury are at a high risk for developing subsequent major depression (Kreutzer, Seel, & Gourley, 2001). Although the prevalence of depression is especially high in individuals after suffering a severe traumatic brain injury (Jorge, et al., 1993), retired professional football players with a history of three or more mild traumatic brain injuries are at a threefold risk of being diagnosed with clinical depression compared with those with no prior history (Guskiewicz, Marshall, et al., 2007). Links between mild traumatic brain injury and major depression could possibly be due to neuronal changes that occur in areas of the brain that modulate mood. Neuroanatomical structures such as the hippocampus (Sheline, Sanghavi, Mintun, & Gado, 1999), amygdala (Sheline, et al., 1999), orbitofrontal cortex (Lacerda, et al., 2004), and basal ganglia (Baumann, et al., 1999) show structural changes in patients with major depression. The loss of neurons caused by recurrent concussion could put individuals at risk of depression, which results in further structural changes within regions of the brain that control mood. Many individuals also suffer from disruption of social relationships, disruptions in friendships and social support, lack opportunities to build new friendships, and often withdraw from leisurely activities (Morton & Wehman, 1995). Although the link between the pathophysiology of recurrent concussion and the lifetime risk of depression is unclear, it seems possible that recurrent mild traumatic brain injury may result in a similar structural and psychosocial impact that eventually leads to depressive disorders.

Biomechanics of Mild Traumatic Brain Injury

Research on biomechanical factors and their influence on outcomes after sport-related concussion remain inconclusive. Previous studies using animal and crash test dummy models provide preliminary evidence, but new advancements in real-time technologies may aid future research on this topic. All research regarding the biomechanics of head injury operate under the same tenet that kinetic energy from an impact to the head is transmitted to the tissue of the brain.

Animal research

Early research primarily utilized primates and other larger mammalian animal models, but changes in ethics regulations animal research in this area has been limited to the rat and other small mammalians. In one of the earliest studies of head injury biomechanics, Denny-Brown and Russell observed that brain injury is avoided when the primate head is prevented from moving when struck (Denny-Brown & Russell, 1940).

These results were later replicated demonstrating that when rotation of the head is restricted, allowing only translation, cerebral concussion does not occur (Ommaya & Gennarelli, 1974). Translational mechanisms are thought to cause focal brain tissue strain, while rotational mechanisms are thought to cause more diffuse axonal injury. Rotational acceleration of the head is thought to cause the cerebrum to rotate about the relatively fixed brainstem (Ommaya & Gennarelli, 1974). Because the midbrain and upper brainstem are responsible for alertness and responsiveness, the strain experienced during rotational mechanisms are more likely to result in loss of consciousness than linear mechanisms (Ommaya & Gennarelli, 1974). If this concept is applied to human models, contraction of the cervical musculature could limit rotational movements of the head, thereby, reducing diffuse axonal injury.

Model research

In the early 1970's, the National Operating Committee on Standards for Athletic Equipment contracted Wayne State University Department of Neurosurgery to develop standards for football helmets established standards for the impact performance of football helmets (Gurdjian, Lissner, Hodgson, & Patrick, 1964). The Wayne State University Concussion Tolerance Curve, computed from impact duration and magnitude, was used to propose a theoretical threshold of 90g of linear acceleration necessary to produce a mild traumatic brain injury. These experiments were conducted using cadavers and metal headforms, but were instrumental in developing standards for new and reconditioned helmets. Because cadaveric lack reusability, laboratory studies began utilizing the hybrid III male anthropometric test devices to reconstruct concussive head impacts observed during professional football games that were visible from two camera angles (Pellman, Viano, Tucker, Casson, & Waeckerle, 2003; Viano, et al., 2007; Viano & Pellman, 2005). The hybrid III anthropometric test device is equipped with standard accelerometers at the head center of gravity and nine linear accelerometers in a 3-2-2-2 configuration. In a series of studies regarding concussions in professional football, the National Football League Head and Spine committee observed 182 plays that resulted in a player sustaining a concussion that also had two clear views of the direction and location of the helmet impact (Pellman, et al., 2003). The authors only reconstructed 31 of the 182 plays using two helmeted hybrid III dummies and the same impact velocity, direction, and head kinematics. Linear and rotational accelerations were measured in an effort to determine the biomechanical threshold for concussion among these 31 cases. From these laboratory experiments, the authors suggested that an injury threshold of 70g to 75g existed for sustaining concussion (Pellman, et al., 2003). The proposed injury threshold from laboratory retrospective reenactments were widely criticized because the limits were estimated from just 31 collisions using game video footage with relatively low video capture speeds (Funk, Duma, Manoogian, & Rowson, 2007; Guskiewicz & Mihalik, 2011). Early standards for head injury protection focused almost entirely on measures of linear acceleration, but neglected to consider rotational acceleration (Zhang, Yang, & King, 2004). Using a finite element model that replicated the average sized adult male head and included anatomical structures including the dura mater, cerebrospinal fluid, cerebrum, cerebellum, and brainstem, the authors estimated maximum resultant rotational accelerations estimated to be 4,600, 5,900, and 7,900 rad/sec^2 for a 25%, 50%, and 80% probability of sustaining a mild traumatic brain injury, respectively (Zhang, et al., 2004).

As part of the same series of studies published in Neurosurgery, Viano et al. (Viano, et al., 2007; Viano & Pellman, 2005) evaluated the mechanics of both the struck and striking player for plays resulting in injury. Alarmingly, the authors found that the struck players, on average, experience 98g of linear head acceleration while the striking player only experienced 58.5g (Viano, et al., 2007; Viano & Pellman, 2005). Because the striking player fully anticipates the impending collision they are able to optimize their biomechanics to impart much greater force on the struck player. The striking player often delivers maximum force by lowering the head to align the head, neck, and torso. Linking the head, neck, and thorax was found to increase the effective mass of the striking athlete by up to 67% (Viano, et al., 2007). The struck player is most often the one affected by concussion because of the high inertial load imparted by the striking athlete.

In Vivo Accelerometer-Based Research

Real-time accelerometer data collection is a novel method available to researchers who are attempting to better understand the biomechanics of concussion. In one of the first studies to use accelerometry in vivo, Naunheim *et al.* (Naunheim, Standeven, Richter, & Lewis, 2000) measured head acceleration using a single triaxial accelerometer imbedded in the helmet of high school hockey and football players during actual game play. The authors measured peak linear acceleration and computed the Gadd Severity Index and Head Injury Criterion scores for head impacts sustained during actual play periods in several games over four seasons. The authors also recorded acceleration of head impacts of soccer players while heading a soccer ball while wearing the instrumented football helmet. Because of the methodological flaws in this study design, results from this study are difficult to interpret. Despite methodological differences from current research paradigms, the author estimated mean linear acceleration measured in the football and ice hockey players of 29.2g and 35.0g, respectively, just slightly higher than current estimates (Broglio, et al., 2009; Mihalik, et al., 2007).

The Head Impact Telemetry System was designed to allow clinicians and researchers to measure real-time head impact biomechanics in helmeted athletes. Helmets are equipped with six spring-loaded single-axis accelerometers. When an impact occurs to the head data are collected, time-stamped, encoded, and relayed to a near-by sideline-controller antennae and laptop computer for storage (Beckwith, Chu, & Greenwald, 2007; Broglio, Eckner, Surma, & Kutcher, 2011; Broglio, et al., 2009; Brolinson, et al., 2006; Duma, et al., 2005; Eckner, Sabin, Kutcher, & Broglio, 2011; Greenwald, Gwin, Chu, & Crisco, 2008; Guskiewicz, Mihalik, et al., 2007; Mihalik, et al., 2007). Duma et al. (Duma, et al., 2005) and Brolinson et al. (Brolinson, et al., 2006) were first to publish important descriptive data regarding head impacts in collegiate football. These authors reported a mean linear acceleration of 32g. The primary finding of this study was that the accelerometry system proved effective at collecting thousands of head impact data and that the system provide useful information to both researchers and clinicians. The invention of an in-helmet accelerometry system that allows for real-time analysis of head impact biomechanics has great potential to shed light on the biomechanical risk factors for concussion allowing for measurement of the severity, frequency, and location of impacts occurring at the head in football, hockey, and boxing (Beckwith, et al., 2007).

Since these early exploratory studies, further efforts have been made to examine the biomechanical characteristics of impacts to the head with the hopes of further describing the magnitude of impacts that occur over the course of a season and ultimately identifying a theoretical threshold of concussion (Guskiewicz & Mihalik, 2011). Despite advancements in technologies, questions regarding why some athletes withstand high magnitude impacts without sustaining a concussion, whereas others are injured by lower magnitude impacts remains unanswered. In contrast to previously published theoretical injury thresholds, Mihalik *et al.* (Mihalik, et al., 2007) reported that less than 0.35% of impacts that exceeded 80g of linear acceleration resulted in concussion. Guskiewicz *et al.* (Guskiewicz, Mihalik, et al., 2007) established that no relationship existed between biomechanical characteristics of head impacts that resulted in concussion and clinical neurocognitive, postural control, and symptom severity measures. In a similar study design utilizing high school athletes rather than collegiate athletes, Broglio et al. (Broglio, et al., 2011) observed that same results. Combined, these studies suggest that concussions occur from impacts in a wide range of magnitude and that post-concussion declines are independent of head impact biomechanics (Broglio, et al., 2011; Guskiewicz, Mihalik, et al., 2007). In an attempt to understand the dynamic nature of the injurious threshold, Eckner et al. (Eckner, Sabin, et al., 2011) evaluated the subconcussive impact profiles that preceded 20 concussive head impacts. Their data suggested that impact volume and intensity preceding a concussive event did not influence concussion threshold in high school football athletes. Although clinical outcomes measures seem to remain unaffected by head impact biomechanical measures, it seems possible that impacts occurring beyond the purposed 70 to 75g injury threshold may result in subtle neurocognitive and postural control deficits in the absence of a concussion diagnosis.

As research continues to focus on the elusive threshold of concussion, a simultaneous shift of focused has occurred highlighting the potential negative consequences of subconcussive head impacts (Spiotta, et al., 2011). McCaffrey *et al.* (McCaffrey, Mihalik, Crowell, Shields, & Guskiewicz, 2007) assessed these short-term clinical outcomes in asymptomatic collegiate football players following low and high magnitude impacts. Their findings suggested that sustaining an impact greater than 90g did not result in observable deficits in neurocognitive or postural control performance or in an increase in self-reported symptoms (McCaffrey, et al., 2007). Likewise, a similar study evaluating changes in neurocognitive, postural control, and symptom severity prior to and following a season of exposure to head impacts found that repetitive subconcussive head impacts did not appear to result in short-term neurologic impairment (Gysland, et al., 2012). Since these studies, the sensitivity of the neurocognitive measures used to identify neurocognitive deficits has been brought into question (Coldren, Russell, Parish, Dretsch, & Kelly, 2012). Recent strides have been made in understanding head impacts characteristics, but more scientific research is necessary to better understand the causes of concussions in sport and how the brain is influenced by repetitive trauma.

Modifiable Factors

In an early initiative to prevent concussion, Dr. Robert Cantu suggested that measures to prevent concussion focus on changes in rules, changes in coaching technique, improvements in conditioning, improvements in equipment, and increasing medical supervision (Cantu, 1996). Changes in the rules and modes of play, such as the elimination of wedge formations, spearing, butt-blocking, helmet-to-helmet hits, and horse-collar tackles have been widely accepted and adapted into the sport of football. While safety in sport remains the primary concern of sports medicine professionals, coaches, and parents, drastic rules changes have the potential to drastically change the sport of football. Rule changes, such as eliminating tackling from youth football, may prevent some concussions by limiting exposure, but would likely encounter considerable social opposition and risk substantial change to the sport (Johnson, 2012). In contrast changes in conditioning provide an alternative to drastic rule changes. Improving the dynamic response of the cervical musculature through conditioning has promising potential for reducing head impact severity (Cantu, 1996).

The Dynamic Cervical Response

Athlete's that are better able to mitigate linear and rotational acceleration of the head, or avoid some head impacts all together, are thought to be less likely to encounter strain on the brain tissue. The cervical musculature contributes 80% of the stability necessary to resist injurious forces to the cervical spine (Panjabi, et al., 1998). Athletes with insufficient cervical musculature strength may be predisposed to concussion because they are less able to generate adequate internal preparatory and reactive force to counter head acceleration (Viano, et al., 2007). By contracting the cervical musculature, an athlete increases the effective mass to that of the head, neck, and thorax. When the cervical musculature remains relaxed, such as when a player receives an unexpected hit, the force of impact acts on the effective mass of the head alone allowing for rapid head acceleration. As adolescents undergo growth spurts, they gain significant amounts of weight and mass, which increases the force and momentum during collision. Despite increases in mass, adolescents have weaker neck muscles than adults, which could limit their ability to dissipate forces applied to the head. If the results of our study support the tenet that greater cervical strength, stiffness, muscle activation, and/or muscle size enable an athlete to reduce the acceleration of their head, then we can better guide sports medicine professionals and strength and conditioning coaches when designing cervical training programs.

Cervical Strength

The osteoligamentous structures of the cervical spine contribute approximately 20% of the minimally needed mechanical stability of the cervical spine (Panjabi, et al., 1998). This leaves a remaining 80% of the mechanical load to be managed by the cervical musculature. During trauma, the contribution of the cervical musculature becomes even more important. Much like in the sports setting, technological advancements have improved fighter plane airframe materials, propulsion systems, and flight controls, allowing fighter pilots to fly farther, faster, higher (Seng, Lam, & Lee, 2003). A large number of studies addressing cervical strength have been focused on the fighter pilot population (Alricsson, Harms-Ringdahl, Larsson, Linder, & Werner, 2004; A. F. Burnett, Naumann, Price, & Sanders, 2005; Seng, et al., 2003). Much like the modern athlete, fighter pilots are at greater risk for injury associated with their profession. Although weak cervical musculature has been proposed as a potential risk factor for concussion, strengthening programs are not emphasized in most sports.

With all muscles maximally activated, flexion moment-generating capacity is dominated by the sternocleidomastoid (69%), with additional contributions from the longus capitis and colli (17% combined) and the scalenus anterior (14%) (Vasavada, Li, & Delp, 1998). The sternocleidomastoid plays a large role in generating torque during flexion because it has the largest flexion moment arm about the lower cervical joints. In the direction of extension, the majority of moment-generating capacity comes from semispinalis capitis (37%) and splenius capitis (30%). In the upper cervical region, the semispinalis capitis, splenius capitis and upper trapezius have the greatest advantage in extension because of the magnitude of their moment arms (Vasavada, et al., 1998). Acting as primary extensors and spinal stabilizers, the right and left semispinalis capits attach to the skull between the superior and inferior nuchal lines and course down to transverse processes of the lower four cervical vertebrae (C4-C7) and the six upper thoracic spine (T1-T6). Levator scapulae, upper trapezius, erector spinae, and the suboccipital muscles also individually contribute 5-10% each to extension moment-generating capacity (Vasavada, et al., 1998). The upper trapezius dominates the moment-generating capacity for cervical rotation, contributing 32%, followed by 10-20% each from splenius, sternocleidomastoid, semispinalis capitis, and suboccipital muscles. Estimated moment-generating capacity for lateral bending is greatest for sternocleidomastoid (28%) and trapezius (19%), with the scaleni, splenius, levator scapulae, semispinalis, and erector spinae estimated to contribute 5-15% each (Vasavada, et al., 1998).

To date, only one previous studies has examined the role of neck strength in reducing in vivo head acceleration during sport activity (Mihalik, et al., 2011). However, the authors were unable to identify differences in head impact biomechanical measures between youth hockey players with strong, moderate, and weak cervical musculature. Handheld dynamometry was used to measure the isometric strength of the anterior neck flexors, anterolateral neck flexors, cervical rotators, and posterolateral neck flexors, but arguably lacks clinimetric properties such as agreement, validity and responsiveness (de Koning, van den Heuvel, Staal, Smits-Engelsman, & Hendriks, 2008). Other studies have observed that females exhibit up to 44% greater head acceleration during soccer heading tasks (Tierney, et al., 2005). Differences between genders were attributed to the observed smaller effective head mass and neck strength among the females compared with males in this study. Likewise, differences in cervical strength could also explain why high school athletes are at a higher risk of concussion compared to collegiate athletes. Previous studies suggest that collegiate athletes are stronger and more powerful than high school and junior high aged athletes (Baker, 2002; Candow & Chilibeck, 2005).

Several studies show that resistance-training programs are capable of increasing the strength of the cervical musculature (Alricsson, et al., 2004; A. F. Burnett, et al., 2005; Mansell, et al., 2005; Rezasoltani, Malkia, & Vihko, 1999). However, the relationship between increases in cervical isometric strength following resistance training and reductions head acceleration remains theoretical (Mansell, et al., 2005; Mihalik, et al., 2011; Viano, et al., 2007). Although weak cervical musculature has been proposed as a potential risk factor for concussion, strengthening programs are not emphasized in most sports. Without a clear understanding of the role of the dynamic cervical response, designing an effective cervical training program proves to be very difficult. Previous research aimed at investigating the role of an eight-week resistance training intervention failed to observe enhancements in head-neck segment dynamic stabilization, despite observing increases in isometric strength and neck girth (Mansell, et al., 2005). The authors attributed their failure to observe improvements in head acceleration resistance to the exclusion of neuromuscular control exercises, such as plyometrics, into the training program. In addition, the study used an eight-week training interval, which captures the minimal amount of time necessary to observe real strength gains.

Cross-Sectional Area

Among the cervical musculature, strength increases linearly with increases in physiological cross-sectional area (Mayoux-Benhamou, Wybier, & Revel, 1989; Rezasoltani, Ylinen, & Vihko, 2002). For every squared centimeter increase in cross-sectional area, the force output of the cervical musculature increases by approximately 10 Newton's (Mayoux-Benhamou, et al., 1989). Cross-sectional area of the cervical musculature increases significantly after a period of resistive head and neck exercise (Alricsson, et al., 2004; A. F. Burnett, et al., 2005; Mansell, et al., 2005; Rezasoltani, et al., 1999). Ultrasonographic imaging has previously been used to assess the dimensions of the splenius capitis, semispinalis captis, sternocleidomastoid, trapezius, multifidus, longus colli, deep cervical flexors as a group, deep posterior muscles as a group, rectus capitis posterior, and oblique capitis superior (Javanshir, Amiri, Mohseni-Bandpei, Rezasoltani, & Fernandez-de-las-Penas, 2010).

Imaging methods to determine dimensional size of the cervical musculature have their individual strengths and weaknesses. Magnetic resonance imaging and computed tomography are the current criterion standards for muscle size measurement, but both techniques are cost prohibitive (Javanshir, et al., 2010). Neck muscle ultrasonography is an alternative method for screening both size and function of the cervical musculature (Rezasoltani, et al., 1999). Compared to other techniques available, it is non-invasive, painless, and easily accessible. Literature regarding the reliability and validity of ultrasonography for determining cervical muscle size is scarce and contradicting (Javanshir, et al., 2010). Ultrasonography seems to have good inter- and intra-rater reliability and is a fairly valid method of measuring upper and lower trapezius muscle thickness (O'Sullivan, Meaney, Boyle, Gormley, & Stokes, 2009). Although magnetic resonance imaging and computed tomography are the criterion standards for determining muscle size, ultrasonography allows the user to move the probe to be perpendicular with the tissue of interest. Comparing cross-sectional area between criterion images and ultrasonography may be difficult because scanning planes differ between the technique (Javanshir, et al., 2010).

Cervical Stiffness

Stiffness is a measure of an elastic body's resistance to deformation. As a football player sustains an impact to the head or body, the cervical musculature, ligaments, and vertebrae deform under the applied force. Greater muscle girth and contraction of the primary stabilizing muscles increase muscle and joint stiffness (Wilson, Wood, & Elliott, 1991). Viscoelastic properties of the cervical spine enable the cervical tissues to withstand brief periods of extreme loading that would otherwise exceed static load tolerance. Preparatory muscle activation acts to stiffen the neck and to absorb energy through eccentric contraction. Mathematical models that have compared levels of neck stiffness show that linear acceleration, angular acceleration, and head injury criterion variables decrease with greater neck stiffness (Queen, Weinhold, Kirkendall, & Yu, 2003). Male participants are able to bear larger bending moment, exhibiting greater stiffness, and capacity to store more elastic energy than the female participants (McGill, Seguin, & Bennett, 1994). These dynamic response variables suggest that males have a greater resistance to injury, which is consistent with the observed rates of concussion across genders (Castile, et al., 2011; Gessel, et al., 2007; Marar, et al., 2012). Bending in flexion and passive loading of extensor tissues appear to be better tolerated when compared to extension and lateral bending. Previous studies that have used traditional cervical resistance training programs show potential to change muscle structure and increase strength, but have failed to observe change is neuromuscular plasticity that could enhance dynamic restraint and reduce head acceleration (Mansell, et al., 2005).

Muscle Activation

Neck extensors, whether acting as agonists or antagonists, are more activated than flexors during all sagittal plane movements (Cheng, Lin, & Wang, 2008). Greater activation of the extensors could be accommodating for the decreased moment-generating capacity of neck extensors. As the neck moves into flexion the neck extensors experience significant decreases of moment arms and large changes of fascicle lengths (Vasavada, et al., 1998). Strengthening of the neck extensors is suggested for preventing of neck disorders by maintaining normal level of cervical cocontraction (Cheng, et al., 2008).

During neck flexion, the extensor musculature functions to resist gravity to keep the head from falling (Cheng, et al., 2008). An athlete that is unable to activate the extensor muscle group, possibly due to fatigue, may have difficulty resisting gravity causing him to make first contact with the crown of the head. Impacts to the crown of the head are likely to be more severe than impacts to the sides, top, or back of the head (Mihalik, et al., 2007). The propensity to lower the head during contact not only has implications for head injury, but also injury to the cervical spine. As the helmeted athlete strikes another player with the crown of the head the forward momentum of the body compresses the cervical spine between the decelerated head. Force is dissipated from the crown of the head through the vertebral column until tissue failure occurs. A slightly flexed position that occurs when lowering the head, like during spear tackling, eliminates the normal lordotic curve of the cervical vertebral column placing it in a straight line, inhibiting the surrounding musculature from assisting in force absorption (Bailes, Petschauer, Guskiewicz, & Marano, 2007). The vertebrae respond to significant axial loads and compression by buckling under the pressure. Bony fragments that impede on the spinal canal can cause damage to the spinal cord. Rule changes initiated in 1976 banned the use of the head and face as the initial contact area for blocking and tackling in American football. A player that lacks cervical strength or endurance may be at a high risk for sustaining high magnitude head impacts, or worse yet, a severe axial load that could cause a catastrophic cervical spine injury.

Anthropometrics & Cervical Posture

Previous studies suggest that higher rates of concussion among female and youth athletes may be attributable to anthropometric differences (Castile, et al., 2011; Gessel, et al., 2007; Marar, et al., 2012). A fundamental question regarding the higher rate of concussions among female athletes is whether female necks are simply scaled versions of male necks, or whether there are fundamental geometrical differences. Between height matched men and women several size normalized anthropometric and strength variables differ, demonstrating that male and female necks are, in fact, geometrically different (Vasavada, Danaraj, & Siegmund, 2008). In a study of collegiate soccer players, females had 26% smaller head and neck mass than males (Mansell, et al., 2005). When a female athlete sustains an impact to the head during sport, the force applied is likely to results in a greater acceleration because of her smaller head mass. Females demonstrate greater angular acceleration and displacement of the head and neck when heading a soccer ball, despite displaying earlier activity of the sternocleidomastoid (Tierney, et al., 2008).

These differences between head and neck angular acceleration during soccer heading may also be explained by differences in cervical strength between genders. The combined effect of a smaller head mass, smaller neck girth, and weaker cervical musculature may limit female and youth athletes from stabilizing the head.

Using mathematical modeling, Queen et al. (Queen, et al., 2003) demonstrated that children with smaller head mass were more likely to experience greater linear acceleration of the head. In epidemiologic studies, athletes that have a body mass index below the 20th percentile have a moderately increased risk of concussion (Schulz, et al., 2004). Since head and neck mass are computed as percentage of mass (Dempster, 1955; Shan & Bohn, 2003), it seems possible that individuals with smaller relative total mass will have a smaller head mass as well. However, simply computing head mass as a percentage of body mass may not accurately reflect an athlete's full anthropometric profile. For example, if an athlete that increases his mass by gaining significant amounts of adipose there may not be any real change in head mass. In general, anthropometric variables like head mass and head-neck segment length are not often modifiable. However, Mansell et al. (Mansell, et al., 2005) found that women's neck girth increased by 3.4% following an eight week resistance training program.

Cervical posture measurements provide an external approximation of the position that the cervical anatomy adopts when supporting the head against gravity (Grimmer-Somers, Milanese, & Louw, 2008). Good posture allows for muscular and skeletal balance, which protects against injury and progressive deformity. Muscles function most efficiently when the optimum positions are afforded (Grimmer-Somers, et al., 2008). Forward head posture is the anterior translation of the head in the sagittal plane so that the head is placed anterior to the trunk (Silva, Punt, Sharples, Vilas-Boas, & Johnson, 2009). A forward head posture could increase the antigravity load on cervical structures. It seems possible that athletes that assume a more forward head posture may be less able to resist the draw of gravity causing them to have a tendency to lower the head when being struck or striking another player. Poor cervical resting posture is reported increase the amount of effort required to balance the head against the forces of gravity (Edmondston, Sharp, Symes, Alhabib, & Allison, 2011).

Visual performance

The eyes supply sensory information to the brain; the brain then decodes and integrates the visual information while also considering vestibular and somatosensory information. The brain then sends out an appropriate motor signals to the muscles based on the supplied sensory information. Numerous studies have identified that athletes that demonstrate better visual abilities than non-athletes, and that elite athletes have visual abilities that are superior to novice and less successful athletes (Hitzeman & Beckerman, 1993; Stine, et al., 1982). Many sports involve quick and unpredictable movement of objects, competitors, teammates, and the athlete themselves. These movements often occur simultaneously. Athletes must be able to accurately perceive and identify both static and dynamic features within their field of view.

Both visual acuity and contrast sensitivity are commonly thought to be the fundamental to visual performance (Zimmerman, et al., 2011). Visual acuity refers to the acuteness and clarity of vision. Most clinicians ensure that athletes are able to see at 20/20 to ensure that visual correction is not necessary. However, many elite athletes present with enhanced visual acuity compared to amateur and non-athletes and may need more extensive examination (Laby, et al., 1996). Contrast sensitivity measures the visual system's ability to process spatial or temporal information about an object and its background under differing lighting conditions. Most sports require athletes to scan and interpret visual information at differing contrast levels. It seems possible that athletes that are better able to discern object from their background under differing lighting conditions will perform better.

Many sports require athletes to determine the distance and spatial location of an object. Stereopsis is the ability to judge depth when a scene is viewed with both eyes. Visual information that athletes use during sport does not all occur at one distance. Most athletes need to alternate between looking between near, far, and intermediate distances. Transition between distances requires rapid accommodative-vergence responses. Previous research on this visual performance among athletes is limited. Efficient eye movements are necessary for an athlete to move and respond successfully. Most sports require eye movement in a variety of directions. Information from the retinal periphery informs the brain that there is something of interest. Saccadic eye movements direct visual fixation towards the objects of interest (Henderson & Hollingworth, 2003; Zupan, et al., 2006). Saccade efficiency can be retained and stored in visual memory (Henderson & Hollingworth, 2003).

Although the importance of visual performance in sport is widely accepted, detailed assessments are not often completed in the athletic setting. Several studies have identified superior visual performance among elite athletes, however, how these differences relate to on-field performance or injury prevention is not yet known (Zimmerman, et al., 2011). Although visual training in athletes is a relatively new concept, studies suggest that visual exercises have the ability to improve visual performance (Maxwell, et al., 2012). It seems possible that an athlete with diminished visual performance relative to their opponent may be less likely to see an oncoming collision, leaving them unable to anticipate and prepare, and more prone to injury. Further research is needed to determine if head impact biomechanical measures are influenced by visual performance.

Anticipation

The phenomenon referred to as risk compensation hypothesizes that each person has a target level of risk they are willing to accept. When applied to sports, an athlete that perceives an intervention, such as a new helmet design, has lowered their level of risk, the athlete will change their behavior in a way that brings them back to their desired risk level (e.g., playing more aggressively) (Daneshvar, et al., 2011; Hagel & Meeuwisse, 2004; Hedlund, 2000). It seems possible that interventions to improve the dynamic cervical response may ultimately cause athletes to engage in more risk taking behaviors. One modifiable variable, that when intervened upon would not likely result in a risk compensation response, is anticipation.

Expert athletes present with an enhanced ability to identify subtle changes in the kinematics used by their opponent (Canal-Bruland, Mooren, & Savelsbergh, 2011; Ida, Fukuhara, Sawada, & Ishii, 2011). American football players have greater efficiency in running through narrow apertures because they are able to rotate their shoulders at smaller magnitudes and later (Higuchi, et al., 2011). Skilled athletes are more accurate in their anticipation and decision-making judgments compared with less skilled players (Roca, Ford, McRobert, & Mark Williams, 2011). As predicted, the underlying processes of vision and cognition were used in a quantitatively different manner between groups. Skilled athletes use visual search strategies that involve more fixations of shorter duration, alternating their gaze more frequently between the player in possession of the ball, the ball itself, and other areas of the field of play. The observation of body movements is known to activate the superior temporal sulcus, the posterior inferior frontal gyrus, the rostral inferior parietal lobule, and the intraparietal sulcus. These regions of the brain are proposed as the core network of the mirror-neuron system, that respond both when a particular action is performed and when the same action performed by another individual is observed (Decety & Grezes, 1999; Filimon, Nelson, Hagler, & Sereno, 2007; Gallese & Goldman, 1998). Moreover, expert athletes show greater activation across the mirror-neuron system than novices. In sports anticipation tasks, expert athletes show stronger neural activations than novice athletes in brain areas that are associated with visual attention and the analysis of body kinematics (Wright, Bishop, Jackson, & Abernethy, 2011). Novice athletes show stronger neural activation in the occipital cortex, which suggests a greater allocation of resources to low-level visual processing.

Observed differences between novice and expert athletes suggest that anticipatory responses to sport related tasks are a trainable attribute. Anticipatory responses to impending head or body collisions may help mitigate acceleration of the head, thereby reducing the potential for sustaining a brain injury and reducing the magnitude of subconcussive impacts (Kumar, Narayan, & Amell, 2000; Mihalik, Blackburn, et al., 2010). Previous research regarding the role of awareness on head neck acceleration in automobile accidents suggests that awareness of the impending impact serves to significantly reduce the level of accelerations of head and neck (Kumar, et al., 2000). An athlete that is able to foresee an impending impact will reflexively and cognitively react with anticipatory responses, such as leaning, using the arms to block the face, or recoiling their head by elevating their shoulders (Metoyer, et al., 2008). During sport, athletes must maintain gaze fixation on a target area, such as a goal or ball, for accurate aiming. Gaze fixation may limit the athlete's ability to foresee and prepare for impending impacts (van der Kamp, 2011). In youth ice hockey players, unanticipated collisions tended to result in more severe head impact magnitudes than anticipated collisions (Mihalik, Blackburn, et al., 2010). In contact sports, the striking player prepares for impending collision by aligning the head, neck, and thorax to impart maximum force on an opponent by driving through the struck player. Previous studies, that have modeled helmet-to-helmet impacts, show that the struck players, on average, experience 98g of linear head acceleration while the striking player only experienced 58.5g (Viano, et al., 2007). Because the striking player fully anticipates the impending collision they impart much greater force on the struck player.

Methodological Considerations

Rationale for Participant Population

This study focused on the influence of the dynamic cervical response on head impact severities among high school and collegiate football athletes. Concussions occur at alarming rates in the high school setting (Marar, et al., 2012). Compared to collegiate athletes, high school athletes have a higher risk of concussion and have a higher risk of experiencing adverse outcomes after being injured (Castile, et al., 2011; Gessel, et al., 2007; Guskiewicz, et al., 2000). High school athletes consist of two adolescent age groups, ages 14-16 and ages 17-18. Cervical characteristics are likely to vary widely across this age group, due to difference in physical maturation (Baker, 2002; Candow & Chilibeck, 2005). Adolescent athletes who sustain their first concussion at a young age and continue to play on into high school and college have a longer window of time they are participating in sports and therefore a longer window of time in which to sustain a subsequent injury (Guskiewicz & Valovich McLeod, 2011).

We've chosen to examine high school and collegiate football athletes because football players regularly engage in contact, and sustain a large number of impacts to the head over the course of a single season (Broglio, et al., 2009; Castile, et al., 2011; Gessel, et al., 2007; Marar, et al., 2012). Football accounts for nearly half of all reported concussions (Gessel, et al., 2007; Marar, et al., 2012). Although the proposed study did not examine the biomechanical variables that result in concussion, we chose to focus on a group in which the rate of concussion is high. High school football players are diagnosed with concussions at a higher rate than their collegiate counter parts. In fact, compared to the division I setting, where athletes are thought to be stronger and faster, high school players had nearly twice the concussion rate (Guskiewicz, et al., 2000). Despite differences in concussion risk between genders, our study consisted of all male participants (Gessel, et al., 2007; Marar, et al., 2012). The results of our study provide preliminary direction for future research that aims to reduce the incidence of concussion among other vulnerable populations, such as females. Although we would like to evaluate the cervical musculature, visual performance, and anticipation in females, reliable technology for measuring in vivo head impact biomechanics in these samples are not yet commercially available.

Rationale for Measurements and Instrumentation

Previous investigators measuring isometric cervical strength have used varying methods including handheld dynamometry (Mihalik, et al., 2011), isokinetic dynamometers (Seng, Lee Peter, & Lam, 2002), and custom devices (Almosnino, et al., 2010; Strimpakos, Sakellari, Gioftsos, & Oldham, 2004; Ylinen, Rezasoltani, Julin, Virtapohja, & Malkia, 1999). We've chosen to use an isokinetic dynamometer because of the observed difficulties during early pilot testing in truly administering break tests to collegiate football athletes. We've chosen not to use a custom device because these devices are not commercially available, limiting our external validity. Cervical isometric strength was measured in four directions: flexion, extension, right lateral flexion, and left lateral flexion. Because the HUMAC setup does not allow for comfortable examination of the cervical rotators, we are unable to obtain these measures. Muscle groups that contribute to cervical flexion, extension, right lateral flexion, and left lateral flexion also contribute to cervical rotation (Vasavada, et al., 1998). We also analyzed rate of torque development by identifying the maximal slope of the force-time curve, calculated using a 50-millisecond sliding window from onset to peak force. This measure was included because peak force measures may not best demonstrate the role of the cervical musculature in preventing rapid head acceleration. Cervical rate of torque development measures have previously been observed to have good reliability (Almosnino, et al., 2010). Quantifying the time dependent force-generating capacity of cervical musculature might provide better insight into the damping response of the neck.

During cervical perturbation, we expect to see a startle response during the first anticipated and unanticipated trials (Siegmund, Blouin, & Inglis, 2008). This response plays a role in explaining whiplash injury, but since football athletes sustain several head impacts over the course of a season and over the course of a career, we suspect that these athletes habituate to their cervical response (Broglio, et al., 2009). Therefore, we treated the first trials as familiarization trials and use the second through fifth trials to compute stiffness and muscle onset latencies. We completed a thorough review of cervical spine and brain injury literature to determine that the force applied during our proposed stiffness testing does not approach injury thresholds. Our calculations of energy indicate that participants would, at most, encounter 2.1g of acceleration at the spine. Given these calculations we are assured that this force delivery would not exceed injurious thresholds and would pose minimal risk to the participants. The energy absorbed at the spine is considerably lower than previously reported injury thresholds for whiplash (5g) (Ito, Ivancic, Panjabi, & Cunningham, 2004), intervertebral disc strain (3.5g) (Panjabi, Ito, Pearson, & Ivancic, 2004), and soft tissue injury (8g) (Pearson, et al., 2005). This protocol is less dangerous than previously reported methodologies used by Reid et al (Reid, Raviv, & Reid, 1981) who applied loads ranging from 0.5-21.5kg dropped from heights ranging from 20-100cm with relaxed cervical musculature. Reid et al. reported a peak force of 170 N, which for a head-neck mass of about 5.5 kg would produce an acceleration of about 3.1g, which far exceeds the values we expect to observe, yet were still referred to a “low intensity” by the authors (Reid, et al., 1981). The force encountered during the stiffness testing are similar to forces encountered during everyday activities that both sedentary and physically active individuals complete (Funk, et al., 2011).

Our visual performance assessment was completed using the Nike Sensory Station (Nike, Inc., Beaverton, Oregon). Based on previous studies that have evaluated the reliability of the Sensory Station measures we expect to see practice effects in near-far quickness, eye-hand coordination, and go/no go, but not in visual clarity, contrast sensitivity, depth perception, target capture, perception span, and reaction time (Erickson, et al., 2011). The Nike Sensory Station has not yet been validated.

We used the Head Impact Telemetry System to measure head impact biomechanics at all games and practices over the course of the season. The HIT System has been previously validated using hybrid III dummies equipped with football helmets in a laboratory setting (Duma, et al., 2005; Manoogian, McNeely, Duma, Brolinson, & Greenwald, 2006). Acceleration-time series data provided by the six single-axis accelerometer configuration accurately estimates the magnitude of the linear acceleration by the triaxial accelerometer of the hemispherical headforms (Crisco, Chu, & Greenwald, 2004). We've chosen to utilize the Head Impact Technology severity profile because it is a weighted component of several biomechanical inputs that is thought to be more predictive of concussion than traditional biomechanical measures (Greenwald, et al., 2008). A known limitation of the HIT system is the inability to measure rotational acceleration about the z-axis. Measures of rotational acceleration may be inexact because rotation is approximated about a fixed point in the neck (Greenwald, et al., 2008).

We evaluated anticipation using the Player to Player evaluation form (Mihalik, Blackburn, et al., 2010; Ocwieja, et al., 2012). We've chosen to use this evaluation form because previous researchers have successfully been able to identify varying aspects of collision.

Summary of Rationale for the Study

This study attempts to bridge the gap pertaining to the role of the cervical musculature in mitigating head impact severity among high school and collegiate football players. Results from this study guide future intervention programs to improve the dynamic cervical response and anticipation. The role of visual performance in mitigating head impact severity has not previously been studied. If the results of our study suggest that visual performance does play a role mitigating head impact severity, then sports medicine professionals should place an emphasis on perfecting each athletes' visual conditions. If the results of this study agree with previous trends that higher levels of anticipation reduce head impact severity, future studies could examine the utility of anticipation training for reducing head impacts. Dynamic stabilization of the head using the cervical musculature is a modifiable factor that potentially influences concussion risk. The results of this study aid sports medicine clinicians in isolating important cervical characteristics that put athletes at higher risk for sustaining more severe impacts to the head.

Chapter 3

METHODOLOGY

Study Participants

Forty-nine American football players participated in this study (34 high school, 15 collegiate) over the course of the 2012 fall football season. Institutional Review Board-approved informed consent documents were delivered to high school players and their parents/legal custodians at an informational meeting prior to the initiation of data collection. High school athletes under the age of majority (18 years old) were only included in the study if they and their parents both consented to participate. Institutional Review Board-approved informed consent documents were delivered to collegiate player by the team's clinical athletic training staff.

Study Design

During this prospective cohort study, participants completed two separate testing sessions that lasted approximately 1.5 hours each. The first session took place prior to the start of preseason practices and the second session took place within two weeks after the last regular or postseason game. During both sessions, participants completed the cervical testing protocol and a visual performance assessment. We utilized post-season measures of cervical characteristics and visual performance to assess maturational changes that occur between pre- and post-season. We did not observe any changes in cervical characteristics between pre- and post-season, however, near-far quickness, eye-hand coordination, and go/no go performance improved over the course of the season. A research timeline is presented in Table 3.1.

Table 3.1. Research Design, Timeline, and Measures.

Preseason High School: Jul 24-Aug 23 Collegiate: Sept 3 – Oct 8	**Regular Season** High School: August 17 – Nov 9 Collegiate: Sept 1 – Nov 24	**Postseason** High School: Nov 12 – Dec 6 Collegiate: Nov 27 – Dec 5
(n=49) Cervical Testing Visual performance	HIT System data collection at practices and games Video analysis of game footage to determine level anticipation‡	(n=28) *Cervical Testing *Visual performance

‡ Video analysis of game footage was completed for high school games only
***Post-season measures were used to assess maturational changes that occur between pre- and post-season, but were not used for this study**

Measurements & Instrumentation

The cervical testing protocol included procedures for measuring cervical isometric strength, ultrasonographic imaging of muscle size, cervical perturbation, and anthropometric/posture characteristics. Anthropometric measurements included: head mass, head circumference, neck circumference, and head neck segment length were recorded and stored, but were analyzed for descriptive purposes only. The visual performance assessment included of measures of visual clarity, contrast sensitivity, depth perception, near-far quickness, target capture, perception span, eye-hand coordination, go/no go, and reaction time. The cervical testing protocol was completed in both the Neuromuscular Research Laboratory and the Sports Medicine Research Laboratory. The visual performance assessment was completed in the Matthew Gfeller Sport-Related Traumatic Brain Injury Research Center. The same test order was followed at the post-season test session.

Cervical Testing Protocol

Prior to cervical testing, all participants completed a brief examination of neck range of motion and stability to determine the general health of the athlete's neck. Each athlete completed a brief range of motion assessment where he was asked to maximally flex, extend, laterally flex (right and left), and rotate (right and left) his neck. Individuals with visibly noticeable limited range of motion did not complete the cervical testing protocol for safety reasons and were excluded from participation in this study (n=1). Cervical stability was evaluated prior to testing by completing the Sharp-Purser test (Uitvlugt & Indenbaum, 1988), Aspinall transverse ligament test, lateral shear test, and alar ligament stress test. These four special tests were considered positive for cervical instability if the patient experienced one or more of the following symptoms: a loss of balance in relation to head movement, unilateral pain along the length of the tongue, facial lip paraesthesia, or bilateral or quadrilateral limb paraesthesia, or nystagmus. No positive tests were observed for any of these clinical assessments (n=0). Following the cervical function and stability assessment participants completed a neck warm-up including ten neck circles clock-wise, ten neck circles counter-clock-wise, and manually resisted flexion, extension, right lateral flexion, and left lateral flexion. The items of the cervical testing protocol, described below, were completed in a block-randomized order. Because the ultrasound unit and the motion capture system are located in a separate laboratory from the isokinetic dynamometer, cervical perturbation and ultrasound imaging were always performed together with isometric strength testing taking place either immediately before or immediately after.

Isometric Strength

Isometric strength was measured using an isokinetic dynamometer, the HUMAC NORM Testing & Rehabilitation System (CSMi Medical Solutions, Inc., Stoughton, MA). The HUMAC NORM is an electromechanical instrument controlled by a microcomputer, which allows for objective and quantitative evaluation of muscle functions such as strength, power, and resistance. Torque data were sampled at 2000 Hz, transmitted from the isokinetic dynamometer to a Biopac MP150 Data Acquisition System and host computer, and instantly viewed in the associated AcqKnowledge 4.0 Software (Biopac Systems, Inc., Goleta, CA). We measured the peak torque and rate of torque development of the cervical flexors (supine), extensors (prone), right lateral flexors (side lying), and left lateral flexors (side lying). All isometric strength measurements were assessed in the neutral position (0°) because this optimizes cervical musculature muscle fascicle length allowing for the strongest contraction (Suryanarayana & Kumar, 2005). A strap was wrapped circumferentially around each athlete’s thorax and shoulders at the level of the spine of the scapula to stabilize the segment and prevent the participant from using compensatory trunk musculature strength (Rezasoltani, Ylinen, Bakhtiary, Norozi, & Montazeri, 2008). A three-inch thick upholstered pad was placed beneath each participant’s head during right and left lateral flexor trials. During all trials, participants pushed directly against the padded strain gauge of the isokinetic dynamometer (Figure 4.1 –Manuscript I).

Two familiarization trials with gradually increasing force were performed in each direction to acquaint participants with the testing position and measurement. Participants were instructed to generate their maximal force as rapidly as possible and to sustain the force over the duration of the trial (Almosnino, et al., 2010). Participants were verbally encouraged to exert maximal effort during the three trials, each lasting three seconds. Participants rested for a minimum of 30-seconds between trials, but were allowed to rest for as long as they desired after each maximal voluntary contraction.

Ultrasonographic Cross-Sectional Area

Ultrasound images of the sternocleidomastoid (SCM), upper trapezius (UT), and semispinalis capitis (SSC) were obtained using an ultrasonographic imaging device (M-Turbo ultrasound system, SonoSite Inc., Bothell, WA USA) with a 7 MHz linear-array transducer that was four centimeters wide. The SCM, UT, and SSC were chosen because of their superficial location and role in stabilizing the head in multiple directions (Bauer, Thomas, Cauraugh, Kaminski, & Hass, 2001; Tierney, et al., 2005; Vasavada, et al., 1998). Previously published head impact data from collegiate and high school athletes suggest that football players sustain a majority of impacts to the top, front, and back of the head (Broglio, et al., 2009; Mihalik, et al., 2007), which we believe to be the most likely to engage the SCM, UT, and SSC. For image consistency and time efficiency, all images were taken on the athlete's right side because previous research suggests that cervical cross-sectional area does not differ between the right and left sides (Arts, Pillen, Schelhaas, Overeem, & Zwarts, 2010; O'Sullivan, et al., 2009).

A single hyper-echoic marker was secured over the skin with medical tape to allow for later merging of ultrasound images. Images of the SCM were taken at 50% of the distance between a line from the mastoid bone to the clavicular margin (Figure 4.2a - Manuscript I) (Arts, et al., 2010). Images of the UT were taken by placing the transducer over the spinous process of C6 and then tilting the transducer head in line with the skin curvature until the triangular shaped medial portion of the muscle can be identified (Figure 4.2b – Manuscript I) (Andersen, et al., 2008; O'Sullivan, et al., 2009). Images of the UT were taken two centimeters lateral to the triangular myofascial junction, perpendicular to the plane of the muscle belly. Images of the SSC were taken by placing the transducer transversely at the midline over C3 (Figure 4.2c – Manuscript I) (Rezasoltani, Kallinen, Malkia, & Vihko, 1998). A permanent marker was used to mark the location of each ultrasound site in order to ensure proper placement of the transducer. The transducer head was tilted until the clearest image of the muscle tissue was observed on the monitor. Three consecutive measurements of the SCM, UT, and SSC were taken to minimize variation in echo intensity (Arts, et al., 2010).

Cervical Perturbation

We evaluated cervical stiffness by applying a load to the back of the head inducing forced extension and front of the head inducing forced flexion (Figure 4.3 – Manuscript I). All participants wore a head harness adjusted to fit snugly with two attachment points, one affixed to the front and the other affixed to the back of the harness allowing for attachment of a pulley cord. Prior to force application, the load was allowed to hang freely with all slack removed from the pulley cord so that the participant could acclimate himself to the load prior to application. The participant was asked to move into flexion or extension to move the weight up or down to get further acquainted with the weight. A strap was affixed from the chair to head harness to prevent excessive cervical spine movement. The strap was adjusted to stop movement just prior to reaching the endpoint of the participant’s natural range of motion. A block was placed beneath the site where the load was dropped to prevent it from falling further if the participant was unable to stop it. We did not observe any trials that engaged the strap or where the weight struck the box.

Tensile force, sampled at 1000 Hz, was measured throughout each trial using a load cell attached in series with the head harness and aligned with the point of force application. The external force applicator consisted of a metal frame affixed to a wall, two cords, a height-adjustable pulley affixed to a stationary wall, and two external loads equal to 1% and 2.5% of the participant's body mass. The athlete supported the external load equal to 1% of body mass throughout all trials to standardize the preload. The second external load equal to 2.5% of body mass was supported by the participant initially and then dropped from a height of 15 cm by the primary investigators following a three second count down, consistent with that of previous studies investigating neck stiffness (Reid, et al., 1981; Tierney, et al., 2008; Tierney, et al., 2005). The heights of the pulley was modified for each participant so that force was applied at 90°, perpendicular to the head-neck segment. The pulley cord was strung through an eyebolt attached to a height adjustable tripod so that the load cell remained perpendicular to the participant's head, but simultaneously would not prevent free movement of the pulley cord. Participants were instructed to activate their cervical muscle enough to support the preloaded weight and to avoid "clinching down". Participants were instructed to remain looking straight ahead (0°- neutral) and to resist the load from falling once they felt the tug. The mass equal to 2.5% of body mass was dropped by the primary investigator following a three second countdown.

An electromagnetic motion capture system (trackSTAR, Ascension Technology Corp., Burlington, VT, USA) was used to measure two-dimensional head-neck segment angular displacement in the sagittal plane. Kinematic data were sampled at a rate of 100 Hz. An electromagnetic sensor placed on the zygomatic arch tracked head movement. Another sensor, placed just below the sternal notch, tracked thorax movement (Mihalik, Beard, Petschauer, Prentice, & Guskiewicz, 2008; Petschauer, Schmitz, & Gill, 2010). Head movement was calculated relative to the thorax to derive head-to-thorax segment sagittal angular displacement as an estimate of cervical spine motion. Following sensor placement, each athlete stood still while anatomical landmarks were identified in the motion analysis system through a digitization process to recognize the head and thorax segments and orient the axes. Digitization points for the head included the bridge of the nose, middle of the chin, and the occipital protuberance. Digitization points for the thorax included the spinous process of T8, spinous process of L5, sternal notch, xiphoid process, and spinous process of the C7. Tensile force and head-neck segment displacement data were synchronized using the time of force application as the event that initiates data collection. We calculated stiffness for anticipated trials only.

Following completion of anticipated trials, each participant completed five subsequent unanticipated trials to measure muscle onset latency. Each participant first completed five anticipated force application trials in either forced flexion or forced extension to measure cervical stiffness followed by five unanticipated force application trials to measure muscle onset latency for each direction (neck flexion & extension) (Mansell, et al., 2005). During unanticipated trials, participants wore a vision blocking eye cover and noise cancelling ear-buds connected to a device playing white noise.

Participants were instructed to activate their cervical musculature enough to support the preloaded weight, but to avoid "clinching down". Participants were instructed to remain looking straight ahead (0°- neutral) and to resist the load from falling once they felt the tug. The mass equal to 2.5% of body mass was dropped by the primary investigator at a random time point following the instructions.

Sternocleidomastoid (SCM) and upper trapezius (UT) electromyography (EMG) data were collected to compute muscle onset latency. Preamplified surface EMG electrodes (Bagnoli 8 Desktop EMG System; DelSys Inc. Boston, MA) (inter-electrode distance= 10 mm; amplification factor = 10,000, 20–450 Hz; Common Mode Rejection Ratio = 60 Hz > 80 dB; input impedance > 10^{15} ohms) were used to measure electromyography activity on the right side only. For the SCM, the electrode was placed along the sternal head, centered at one-third of the distance between the mastoid process and the sternal notch (Almosnino, Pelland, Pedlow, & Stevenson, 2009; Falla, Dall'Alba, Rainoldi, Merletti, & Jull, 2002a, 2002b). For the UT, we placed the electrode two centimeters lateral to the midpoint of the C4-C5 spinous processes and oriented along the palpated anterior border of the trapezius, in line with the direction of the muscle fibers (Almosnino, et al., 2009; A. Burnett, Green, Netto, & Rodrigues, 2007) (reference electrode: dorsal wrist). These electrode placements were chosen because of their previously reported reliability. We also captured EMG data during three maximal voluntary contractions for the SCM and UT each. However, these trials are not needed to compute muscle onset latency. During SCM maximal contractions, participants were positioned supine on a treatment table in neck flexion and left rotation. The investigator performed a break test by pushing against the participant's temple in the direction of extension and right rotation. During UT maximal contractions, participants were positioned prone on a treatment table in neck extension and right rotation. The investigator performed a break test by pushing against the head near the right nuchal lines in the direction on flexion and left rotation.

Visual Performance Assessment

Visual performance was evaluated using the Nike SPARQ Sensory Station (Nike, Inc., Beaverton, Oregon). The Nike SPARQ Sensory Station consists of two high-resolution liquid crystal display monitors (a single 22-inch display and a single 42-inch touch-sensitive display) controlled by a single computer. A wirelessly connected Apple iPod touch (Apple Corporation, Cupertino, California) is used in several assessments (described below). Custom software controls the displays, input acquisition, and test procedures based on athlete responses. Prerecorded instructions play at the start of each assessment. Athletes were instructed to wear the corrective lenses that they typically wear while attending school or while playing football. Examination on the Nike SPARQ Sensory Station took approximately 20 minutes to complete. A majority of the sensory station assessments did not present with significant changes in performance over time between pre- and post-season. Near-far quickness, eye-hand coordination, and go/no go performance did improve over the course of the season, but these improvements were consistent with previously reported practice effects (Erickson, et al., 2011).

Visual Clarity

Static visual clarity was measured by having the athlete stand 16 feet away from the 22-inch display. During this test, black Landolt rings (**C** -ring that has a gap, looking similar to the letter C), with gaps at the top, bottom, left, and right, were presented in random order on a white background at preset acuity demands. Athletes were instructed to swipe the screen of the iPod touch in the direction of the gap in the Landolt ring as soon as they identified the gap. Athletes first viewed an example before the test began and then completed three practice trials. If the athlete could not easily discriminate the gap direction, the athlete was instructed to guess. The visual clarity test began with a large (20/50 equivalent) stimulus that decreased in size until the athlete did not correctly identify the stimulus. When the athlete no longer correctly identified the direction of the gap, the stimulus increased in size until it was identified correctly. This procedure continued until several reversal points were complete. The procedure was first performed with the visual occluder covering the left eye, then with the visual occluder covering the right eye, and then with neither eye covered.

Contrast Sensitivity

Four black circles were presented on a light gray background in a diamond configuration while the athlete stood 16 feet away from the 22-inch display. One circle contained a pattern of concentric rings that varied in brightness from the center to the edge. Athletes swept their finger on screen of the iPod touch in the direction of the circle with the contrasted pattern. Athletes viewed an animation example before the test began and completed three practice trials. If the athlete could not easily discriminate the circle containing the pattern, the athlete was instructed to guess. Contrast sensitivity was assessed binocularly at two spatial frequencies, six and 18 cycles per degree, using a staircase reversal algorithm. Final threshold contrast sensitivity was measured between

10% and 1.0% contrast at 6 cycles per degree, and between 32% and 2.5% contrast at 18 cycles per degree.

Depth perception

Athletes wore a pair of liquid crystal goggles (NVIDIA 3D Vision, Santa Clara, California) that were wirelessly connected to the computer while viewing the 22-inch display from 16 feet away. The liquid crystal shutter system created simulated depth in one of the four black rings presented on a white background, causing the ring to appear to float three-dimensionally in front of the screen. Athletes were instructed to swipe the screen of the iPod touch in the direction of the floating ring and were encouraged to respond as quickly as possible. If the athlete could not easily discriminate the ring depth, the athlete was instructed to guess. Athletes viewed an example before the test began and completed three practice trials.

Near-Far Quickness

Athletes stood 16 feet away from the 22-inch display holding the iPod touch 16 inches from the eyes, with the top edge of the iPod touch positioned just below the bottom of the display. Positioning and instructions were presented with an animation example, and if needed, the researcher helped the athlete with the positioning adjustments. Alternating between screens, a black Landolt ring of 20/80-equivalent was presented in a box on the far screen and then on the handheld iPod screen. Athletes were instructed to swipe the screen of the iPod touch in the perceived direction of the gap in the ring presented on each display. The assessment began with three practice trials. The first Landolt ring was always presented on the far screen, followed by a Landolt ring appearing on the handheld screen once the correct direction was chosen. Athletes were required to continually switch focus between far and near for 30 seconds, trying to correctly identify as many rings as possible.

Target Capture

Athletes stood 16 feet away from the 42-inch display with the center of the screen adjusted to their height using a ruler mounted on the right side of the Sensory Station. Athletes were instructed to fixate on a central white dot until a Landolt ring inside of a larger circle appeared briefly in one of the four corners of the screen. As before, athletes indicated the perceived direction of the gap by swiping the screen of the iPod touch. Athletes viewed an animation example before the test began and completed three practice trials. Athletes were instructed to guess if the orientation of the gap was not easily discriminated.

Perception Span

Athletes were positioned within arm's length of the 42-inch touch-sensitive display, with the center of the screen adjusted to their height using a ruler mounted on the right side of the Sensory Station. Automated instructions directed each athlete to focus on a black dot in the center of a grid pattern composed of up to 30 circles. A pattern of turquoise dots flashed simultaneously for 100 milliseconds within the grid. Athletes were instructed to touch the screen to recreate the pattern of dots. If the athlete achieved a passing score of 75% correct, the grid pattern increased in size and number of dots. The first two levels consist of six circles in the grid pattern with two and three dots, the next five levels consists of 18 circles with three to seven dots, and the last four levels consist of 30 circles with seven to ten dots. The dot patterns at each level were pseudorandomized to maintain equivalent spatial distribution of the dots for each presentation and to eliminate ‘‘clustering’’ of dots and easily recognizable patterns or shapes. Athletes viewed an animation example before the test began and completed two practice trials. If the athlete did not achieve a passing score on a level, that level was repeated. If the athlete failed a level twice, the assessment was terminated.

Eye-Hand Coordination

For this assessment, athletes held their arms parallel to the ground at shoulder height within easy reach of a grid of circles presented on the 42-inch touch-sensitive display. The grid consisted of eight columns (68.6 cm total) and six rows (44.5 cm total) of equally spaced blank circles. During the assessment, a turquoise dot appeared within one circle of the grid. Athletes were instructed to touch the dot as quickly as possible using either hand. As soon as they touched the dot, another dot was presented. A sequence of 96 dots were pseudorandomized to maintain equivalent spatial distribution within each presentation and to eliminate ‘‘clustering’’ of dots and easily recognizable patterns. Athletes viewed an animation example before the test began and completed one-practice trials.

Go/No Go

For this assessment, athletes held their arms parallel to the ground at shoulder height within easy reach of a grid of circles presented on the 42-inch touch-sensitive display. However, the dot stimulus was either turquoise or red. Although some color-deficient individuals may confuse the colors, the difference in apparent brightness of the dots allows easy discrimination. If the dot was turquoise, the athlete was directed to touch it (as described in the eye-hand coordination test). But if the dot was red, the athlete was directed not to touch it. Both the red and turquoise dots appeared at random locations for only 450 milliseconds, with no time gap between dot presentations. Athletes were encouraged to touch as many turquoise dots as possible. Athletes viewed an animation example before the test began, but there was no practice trial for this assessment. Ninety-six total dots (64 turquoise, 32 red) were presented in a pseudorandomized sequence to maintain equivalent spatial distribution within each presentation and to eliminate clustering of dots and easily recognizable patterns.

Reaction Time

For the final assessment, athletes remained at arm's length from the 42-inch touch-sensitive display. Two annular patterns appeared on the screen, consisting of two concentric circles. Automated instructions directed the athlete to place the fingertips of the dominant hand on the inner circle of the annulus on that side of the screen, with no portion of the hand extending across the boundary line marked on the screen. If the hand was aligned correctly, the control annulus changed color to turquoise. The athlete was then instructed to center the body in front of the opposite annulus and focus attention on the center of that annulus. After a randomized delay of two, three, or four seconds, the test annulus turned turquoise, and the athlete moved their hand to touch its inner circle as quickly as possible. Athletes viewed an animation example before the test began and completed two practice trials.

Head Impact Biomechanics

Head impact biomechanics were measured at each high school and college practice and game over the course of the entire season using the Head Impact Telemetry (HIT) System technology (Riddell Corp., Elyria, OH). Selected players wore a HIT System MxEncoder embedded in their helmet to measure head impact biomechanics over the course of the preseason, regular season, and postseason. The HIT System consists of MxEncoder units located in the football helmets, a signal transducer, and a laptop computer that houses the Sideline Response System (Riddell Corp., Elyria, OH). MxEncoder units embedded within the helmets are comprised of six spring-loaded single-axis accelerometers, a telemetry unit, a data storage device, and a battery power source. The MxEncoders were retrofit into the Revolution and Speed helmet designs (Riddell Inc., Elyria, Ohio). Each single-axis accelerometer collects data at one kHz for a period of 40 milliseconds (eight milliseconds prior to the data collection trigger and 32 milliseconds after the trigger). Data is time-stamped, encoded, and then transmitted in real-time to the signal transducer via radiofrequency transmission at 903–927 MHz. The signal transducer is connected through a USB port to a laptop computer, which stores all head impact data.

Measures of head acceleration are calculated and stored within the Sideline Response System, yielding measures of linear acceleration, rotational acceleration, Gadd Severity Index, Head Impact Technology severity profile (HITsp), and Head Injury Criterion. This study focused primarily on two traditional measures of head impact severity (linear acceleration and rotational acceleration) and one weighted combination of several biomechanical inputs, including linear acceleration, rotational acceleration, impact duration, and impact location (HITsp) (Greenwald, et al., 2008). The HIT System transmits accelerometer data from distances well in excess of the length of the standard American football field. The HIT System has been previously validated using Hybrid III dummies equipped with football helmets in a laboratory setting (Duma, et al., 2005; Manoogian, et al., 2006). Acceleration-time series data provided by the six single-axis accelerometer configuration accurately estimate the linear acceleration measured by the triaxial accelerometer embedded within the headforms (Crisco, et al., 2004).

Video Assessment of Level of Anticipation

We captured video footage of each high school home and away game using a Panasonic HMC-40 (Panasonic System Communications Company of North America, Secaucus, NJ) placed above the press box (~3 stories high) at the 50-yard line. A research assistant monitored the camcorder by adjusting the zoom and field of view as plays progressed up and down the field. Every effort was made to adjust the camera to maintain adequate zoom while also maintaining a wide field of view. The camcorders and Sideline Response System were date and time synchronized prior to each game. Collisions observable on video footage were matched to head impact biomechanical measures recorded by the HIT System based on date and time. We recorded video footage for all 13 games over the course of the high school football team's season. We did not obtain video footage for collegiate home and away games.

Play Exposure

We used the Play Exposure Log (Appendix II) to tally the number of offensive, defensive, and special teams plays that each high school and collegiate player participated in at all home and away games. The primary investigator was present at all games to record the number of times each player participated in any offensive, defensive, or special teams play. The jersey numbers of all players were written into the 11 cells of the Play Exposure Log. The play drive was indicated as either offensive by circling the letter ”O”, defensive by circling the letter ”D”, and special teams by circling the letter ”S”. The number of plays completed prior to obtaining a first down, scoring, or turning over the ball was indicated by circling the number 1 through 4 for each play drive. We validated the accuracy of the investigator’s records on the play exposure log by comparing play exposure totals in each quarter recorded during a single game to play exposure logs recorded while reviewing video footage. The primary investigator accurately identified 93.72% of play exposures in real-time compared to game video.

Data Reduction

Head impact biomechanical measures captured during practices and games were used to address research question 1 (a-f) regarding cervical characteristics and research question 2 (a-i) regarding visual performance. We used head impact biomechanical measures captured during games only to address research question 3 regarding level of anticipation and research question 4 (a-c) regarding predicting head impact severity. Raw data captured using the isokinetic dynamometer, Motion Monitor, and the HIT system were reduced using separate custom data reduction programs in Matlab 7 (The Mathworks, Inc., Natick, MA).

Cervical Characteristics

To address research question 1 (a-f), we split participants into a group of high and a group of low performers for each cervical characteristic measure using a median split. For composite peak torque, composite rate of torque development, composite stiffness, and composite cross-sectional area, high performance meant higher values (above the median). For composite angular displacement and muscle onset latency, high performance meant lower values (below the median).

Isometric Strength

Raw torque data were zero offset and filtered with a low pass, zero lag, Butterworth filter at 10 Hz. The moment that each participant had to generate to overcome gravity's influence on the mass of the head and neck was added to each torque value. Head mass was calculated by using the following regression equation: Head & Neck Mass = Body Mass * 0.0534 +2.33 (*Anthropometric Source Book Volume I: Anthropometry for Designers*, 1978). The moment arm of the center of mass of the head was calculated as a percentage of the head-neck segment length (de Leva, 1996). We identified the maximum torque (Nm) generated during each of the three trials and then normalized by dividing by body mass in kilograms (Nm/kg). Composite peak torque was calculated by summing the normalized peak torque values across each direction (flexion, extension, right lateral flexion, and left lateral flexion).

Rate of torque development (Nm/sec) was calculated by identifying the greatest slope of the torque-time curve, using a 50-millisecond sliding window from onset to peak torque. Composite rate of torque development was calculated by summing rate of torque development across each direction.

Ultrasonographic Cross-Sectional Area

Ultrasonographic images of the SCM, UT, and SSC were exported to a public domain image processing and analysis program (Image J, National Institutes of Health, USA), merged in reference to the hyper-echoic markers, and outlined to calculate cross-sectional area. The primary investigator completed all measurements by tracing the interface between the hyper-echoic fascia and the hypo-echoic muscle tissue for each muscle. Cross-sectional area was averaged across the three images for each muscle. We calculated the sum cross-sectional area of the SCM, SSC, and UT to compute composite cross-sectional area for each athlete.

Cervical Perturbation

Kinematic data were zero offset and filtered with a low pass, zero lag, Butterworth filter of 10 Hz. Euler angles were used to calculate the movement of the head relative to the thorax. Orthogonal planes were defined in the order of flexion-extension (Y-axis), right and left rotation (Z-axis), and right and left lateral flexion (X-axis) (James, Riemann, Munkasy, & Joyner, 2004). Positive motions were flexion, left rotation, and right lateral flexion; negative motions are extension, right rotation, and left lateral flexion.

Stiffness was calculated as the ratio of the change in moment to the change in sagittal angular displacement of the head relative to the thorax between peak force and force offset (Nm/rad). We averaged the stiffness values from trials 2-5 separately for the anticipated forced flexion and extension. We observed clipping of load cell data due to capacity overload during some trials (anticipated forced extension: 115 trials, 48.32%; anticipated forced flexion: 99 trials, 41.77%). For trials where clipping was evident, we used a regression equation to estimate peak moment derived from trials where clipping did not occur. The regression equation used the participant's body weight, the last observed moment value prior to clipping, and the moment value at 50% of the peak (the moment value at the time point half way between the onset of force and the estimated time of peak moment assuming peak moment was reached at the midpoint of the clipped data) and predicted 92% of the variance in peak moment. Using trials where load cell data was not clipped, we observed good reliability between the computed peak force and the actual peak force applied ($ICC_{3,1}$=0.92, SEM=1.19 Nm). The first trials of anticipated forced flexion and extension were not included in the average across trials because of the possibility of a combined startle and postural responses causing an exaggerated neuromuscular response observed during the first exposure to a transient acceleration (Siegmund, et al., 2008). Since football athletes sustain repetitive impacts to the head over the course of the season, we speculate that they habituate their cervical neuromuscular response to these transient head accelerations (Broglio, et al., 2009; Mihalik, et al., 2007). We summed stiffness values across anticipated forced extension and anticipated forced flexion conditions to compute composite stiffness for each athlete.

Peak angular displacement (rad) was calculating by identifying the absolute value of maximum displacement of the head relative to the thorax in the sagittal plane. We averaged angular displacement values from trials 2-5 separately for the anticipated forced flexion and extension. We calculated composite angular displacement by summing angular displacement across anticipated forced extension and anticipated forced flexion conditions for each athlete.

Analog signal from EMG data were converted to a digital signal by an analog-to-digital converter card. The signal was amplified (gain 100-1000) with a single-ended amplifier and filtered with a fourth-order bandpass filter (20-350Hz) and common mode rejection ratio of 130 dB at direct current. The raw digital signal was exported to a custom data reduction program in Matlab 7 where it was rectified, zero offset, and smoothed using a root mean square algorithm over a 20ms-moving window. Muscle onset latency was calculated as the time in milliseconds between force application and the point at which myoelectric activity exceeded nine times the resting mean for the SCM and four times the resting mean for the UT. Resting EMG data for the SCM was very low requiring a high threshold to determine onset, however, resting EMG data for the UT data are higher because of the postural nature of the muscle (Sommerich, Joines, Hermans, & Moon, 2000). Muscle onset latency was calculated for unanticipated forced flexion and unanticipated forced extension trials only. We excluded trials when the onset of muscle activity was ambiguous, such as when muscle activity rises briefly, but then returns to resting (unanticipated forced extension: 32 trials, 13.06%; unanticipated forced flexion: 34 trials, 13.88%). Stiffness values were normalized to each participant's mass (N) and height (m). We computed composite muscle onset latency by summing the SCM onset latency measured during unanticipated forced extension trials with the UT onset latency measured during unanticipated forced flexion trials.

Visual Performance

To address research question 2 (a-i), this study split participants into a group of high and a group of low performers for each visual performance measure by determining if each athlete was above or below the median. For contrast sensitivity, near far quickness, perception scan, and go/no go high performance means higher raw scores (above median). For visual acuity, depth perception, target capture, eye-hand coordination, and reaction time high performance means a lower raw scores (below the median). The visual acuity, contrast sensitivity, depth perception, and target capture raw scores were identified using a custom proprietary staircase reversal algorithms embedded within the Sensory Station software.

Visual Acuity: The visual acuity raw score was calculated by identifying the threshold acuity between the demands of 20/8 and 20/99 using a staircase reversal algorithm at which the gap in the Landolt ring is barely visible from a uniform circle. We chose to the LogMar values for oculus Uterque (visual clarity using both eyes) because football athletes are not often required to complete tasks with vision occluded.

Contrast Sensitivity: The contrast sensitivity raw score was calculated by identifying the cycles per degree threshold at which the contrast between circles is barely visible from any uniform gray field. We used contrast sensitivity threshold examined during 18 cycles per degree trials because most athletes are capable of easily discriminating contrast at 6 cycles per degree.

Depth Perception: The depth perception raw score was calculated by identifying the arc second threshold between 237 and 12 arc seconds using a staircase reversal algorithm.

Near Far Quickness: The near far quickness raw score was calculated by summing the number of times each participant correctly responds by swiping towards the gap in the Landolt ring within the 30 second trial.

Target Capture: The target capture raw score was calculated by identifying the millisecond threshold between 0 at 500 milliseconds exposure duration using a staircase reversal algorithm.

Perception Scan: The perception scan raw score was calculated by summing the number of correct responses minus the number of missed responses and extra guesses.

Eye-Hand Coordination: The eye-hand coordination raw score was calculated as the total time to touch all 96 dots.

Go/ No Go: The go/ no go raw score was calculated as the sum of the number of turquoise dots touched minus any red dots touched.

Reaction Time: The reaction time raw score was measured as the elapsed time between onset of the test annulus and release of the control annulus.

Composite Visual performance Rating Scale: The composite visual performance rating scale was taken by averaging the percentile scores across all visual performance measures. This variable was used to address research question 4 (b-c).

Head Impact Biomechanics

Head impact data were exported from the Sideline Response System into Matlab 7. Consistent with previous studies, we then reduced the data to include only those impacts that register a linear acceleration greater than or equal to 10g (Guskiewicz, Mihalik, et al., 2007; Mihalik, et al., 2007; Mihalik, Blackburn, et al., 2010; Mihalik, Greenwald, et al., 2010; Mihalik, et al., 2011; Schnebel, Gwin, Anderson, & Gatlin, 2007). Previously published values for linear head acceleration during every day activities that do not involve impacts to the head result in linear accelerations less than 10g (Funk, et al., 2011). This study focused on three primary head impact biomechanical measures. The biomechanical measures of interest included (1) peak linear acceleration (g), (2) peak rotational acceleration (rad/sec^2), and (3) HITsp. In order to address research questions 1 and 2, we categorized the linear acceleration, rotational acceleration, and HITsp of each head impacts into quartiles. We also created separate categories for head impacts that occurred in the 95th and 99th percentiles. For research question 1, we excluded head impacts that occurred to the top of the head because loading transmitted directly through the spinal column does not engage the large moment-generating, superficial, cervical musculature (SCM, UT, and SSC) examined in this study (Banerjee, Palumbo, & Fadale, 2004; Swartz, Floyd, & Cendoma, 2005). To address research question 3 regarding level of anticipation, we matched game head impact biomechanical measures with the graded level of anticipation based on synchronized time-stamps obtained from the HIT system and video footage. To address research question 4(a), we used biomechanical measures from head impacts occurring in all practices and games. To address research question 4(b), we computed cumulative game linear acceleration per play exposure by summing the linear acceleration from all head impacts that each player sustained during games over the course of the season and dividing by their recorded number of play exposures. This was repeated for rotational acceleration and HITsp. To address research question 4 (c), we computed cumulative head impact frequency per play exposure by summing the number of head impacts that each player sustained during games over the course of the season and dividing by their recorded number of play exposures.

Level of Anticipation

We analyzed video footage of on-field collisions occurring during high school football games using the Player to Player Form (Appendix I) to determine each player's level of anticipation at the time of head impact (Ocwieja, et al., 2012). Each viewable collision that resulted in a head impact was determined as anticipated, unanticipated, or unknown. Collisions were deemed anticipated if the impact occurred while the athlete was looking in the direction of the impending collision, was in a general athletic readiness position (knee and trunk flexion with feet shoulder-width apart), and used their legs to drive their shoulders through the collision. Collisions were deemed unanticipated if the impact occurred while the athlete was looking in the direction of the oncoming collision but was not in an athletic readiness position or if the impact occurred while the athlete was not looking in the direction of the impending collision (Mihalik, Blackburn, et al., 2010). Collisions were deemed unknown if the investigator was unable to identify the direction of gaze or the positioning of the body. We excluded all unknown impacts, impacts that resulted from contact with the ground, and impacts that occurred outside of the field of view from our analyses. Video analysis was completed over the course of four months by five different raters and the primary investigator. Each rater was instructed on proper grading by the primary investigator and completed a reliability segment of 91 head impacts. Raters were blinded to which section was being completed to determine inter-rater reliability. We observed good inter-rater reliability for all raters (kappa: 0.309-0.376, $p<0.05$).

Play Exposure

We summed the number of exposures across all games to acquire the number of games that each athlete participated in across the entire season. Plays that did not result in physical contact between players, such as when the quarterback took a knee, were recorded, but were not included in the total.

Statistical Analyses

All statistical analyses were performed in SAS (Version 9.3; SAS Institute, Inc, Cary, North Carolina). Head impact data in previous studies have typically been skewed due to the much larger frequency of low-magnitude head impacts and relatively few high-magnitude impacts. Therefore, we evaluated skewness in our data and implemented a natural logarithmic transformation on the data to satisfy the normality assumptions. Results were considered significant at an a priori alpha level of 0.05.

Research Question 1: Cervical Characteristics

To address research question 1 (a-f) regarding cervical characteristics, random intercepts, general mixed linear, proportional odds models were used to compute odds ratios (OR) and 95% confidence intervals (CI) for each dichotomized cervical characteristic measure. Our predictor variables included each measure and composite measure of cervical isometric strength (5 peak torque measures, 5 rate of torque development measures), muscle size (4 cross sectional area measures), and perturbation (3 stiffness measures, 3 angular displacement measures, 3 latency measures). We computed the odds of sustaining head impacts in the 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile versus the reference category of head impacts in the 1st

quartile across groups of high and low performers for each cervical characteristic for each of the following categorized measures of head impact magnitude: linear acceleration, rotational acceleration, and HITsp. For all models, 1^{st} quartile head impacts and low performers were the reference categories. We included position group assignment (skill, line) in the model to control for differences in player types. Because we suspected that the collegiate athletes may have stronger, larger, and more stiff cervical musculature, despite normalization, we first analyzed group dispersions between the high school and collegiate athletes across the high and low classifications for all outcome variables using a 2 (high school, collegiate) x 2 (high, low) chi-squared goodness of fit analysis. For analyses that involved measures where dispersion was not even, we included playing level into the model as a predictor. Subgroup analyses were done among skill players and among line players separately. Results were considered significant if the 95% confidence interval about the odds ratio did not contain one. Odds ratio values greater than one indicate an increased odds among athletes categorized into the high performance group, whereas odds ratios below one indicate a reduced odds among the high performance group.

Research Question 2: Visual performance

To address research question 2 (a-i) regarding visual performance, random intercepts, general mixed linear, proportional odds models were used to compute odds ratios (OR) and 95% confidence intervals (CI) for each dichotomized (low, high) visual performance measure. We computed the odds of sustaining head impacts in the 2^{nd} quartile, 3^{rd} quartile, 4^{th} quartile, 95^{th} percentile, or 99^{th} percentile versus the reference category of head impacts in the 1^{st} quartile across groups of high and low performers for each cervical characteristic for each of the following categorized measures of head impact magnitude: linear acceleration, rotational acceleration, and HITsp. For all models, 1^{st} quartile head impacts and low performers were the reference categories. We also included position group assignment (skill, line) as a predictor variable to control for differences in player types. Results were considered significant if the 95% confidence interval about the odds ratio did not contain one. Odds ratio values greater than one indicate an increased odds among athletes categorized into the high performance group, whereas odds ratios below one indicate a reduced odds among the high performance group.

Research Question 3: Level of Anticipation

To address research question 3 (a) regarding level of anticipation we conducted three separate random intercepts general linear mixed models to assess the differences in head impact biomechanical measures between the levels of anticipation (anticipated, unanticipated). Results were considered significant at an a priori alpha of 0.05.

Research Question 4: Predicting Head Impact Severity

To address research question 4a, we conducted three separate random intercepts general linear mixed models. The models for research question 4-a included the following 15 predictors: composite peak torque, composite rate of torque development, composite cross-sectional area, composite stiffness, composite muscle onset latency, visual acuity, contrast sensitivity, depth perception, near-far quickness, target capture, perception span, eye-hand coordination, go/no go, reaction time, and level of anticipation; and the three following criterion: linear acceleration, acceleration, and

HITsp (from high school games only). We used the forward method to enter predictors. Predictors remained in the model at an a priori alpha of 0.10. To address research question 4b & c, we conducted four separate multivariate regression models. The models for research question 4-b included the following five predictors: composite peak torque, composite rate of torque development, composite cross-sectional area, composite stiffness, composite muscle onset latency, composite visual rating score; and the following three criterion: cumulative game linear acceleration per play exposure, cumulative game rotational acceleration per play exposure, and cumulative game HITsp per play exposure. The model for research question 4-c included the same predictors as 4-b and the following criterion: cumulative game frequency of head impact per play exposure.

Table 3.2. Data Summary Table for Research Questions 1-3

	Research Questions	Data Source	Comparison	Method
RQ1 Cervical Characteristic Chapter IV Manuscript I	Do football players with high and low preseason: ***a) composite peak torque*** ***b) composite rate of torque development*** ***c) composite cervical cross-sectional area*** ***d) composite cervical stiffness*** ***e) composite cervical angular displacement*** ***f) composite cervical muscle onset latency*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?	<u>Cervical Characteristics:</u> Measures obtained during the preseason cervical testing protocol <u>Head Impact Biomechanics</u>: Measured at all practices and games using the HIT system	<u>Cervical Characteristics:</u> High vs. Low (ref) performers <u>Categorized head impact severity:</u> 1st quartile (ref) 2nd quartile 3rd quartile 4th quartile 95th percentile 99th percentile	Random intercepts, general mixed linear, proportional odds models were used to compute odds ratios (OR) and 95% confidence intervals (CI) for each cervical characteristic variable
RQ2 Visual performance Chapter V Manuscript II	Do high school football players with high and low preseason: ***a) Visual acuity*** ***b) Contrast sensitivity*** ***c) Depth perception*** ***d) Near-Far quickness*** ***e) Target capture*** ***f) Perception span*** ***g) Eye-Hand coordination*** ***h) Go/No Go*** ***i) Reaction Time*** performance differ in odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile?	<u>Visual performance:</u> Measures obtained during the preseason visual performance assessment <u>Head Impact Biomechanics</u>: Measured at all practices and games using the HIT system	<u>Visual performance:</u> High vs. Low (ref) performers <u>Categorized head impact severity:</u> 1st quartile (ref) 2nd quartile 3rd quartile 4th quartile 95th percentile 99th percentile	Random intercepts, general mixed linear, proportional odds models were used to compute odds ratios (OR) and 95% confidence intervals (CI) for each visual performance variable
RQ3 Level of Anticipation Chapter VI Manuscript III	Is there a significant difference in biomechanical measures of head impact severity between anticipated and unanticipated collisions among high school football players?	<u>Level of Anticipation:</u> Measures at all high school games using video analysis <u>Head Impact Biomechanics</u>: Measured at all high school games using the HIT system	<u>Level of Anticipation:</u> Anticipated vs. unanticipated <u>Game biomechanical measures of head impact severity (High School Only):</u> Linear acceleration Rotational acceleration HITsp	Three separate random intercepts general linear mixed models to assess the differences in biomechanical measures of head impact severity between the levels of anticipation

Table 3.3 Data Summary Table for Research Question 4

	Research Questions	Predictor Variables	Criterion Variable(s)	Method
RQ4A Chapter VII Overview I	a: Do cervical characteristics, visual performance, and level of anticipation predict ***game biomechanical measures of head impact severity*** among high school football players?	Cervical characteristics: Composite peak torque Composite rate of torque development Composite CSA Composite stiffness Composite onset latency Visual performance: Visual acuity Contrast sensitivity Depth perception Near-Far quickness Target capture Perception span Eye-Hand coordination Go/No Go Reaction Time Level of anticipation Anticipated Unanticipated	Game biomechanical measures of head impact severity (High School Only): Linear acceleration Rotational acceleration HITsp	Three separate random intercepts general linear mixed models
RQ4B Chapter VII Overview II	b: Do cervical characteristics and visual performance predict ***cumulative game biomechanical measures of head impact severity*** while controlling for play exposure?	Cervical characteristics: Composite peak torque Composite rate of torque development Composite CSA Composite Stiffness Composite Onset Latency Visual performance: Composite Visual Rating Score	Mean game biomechanical Measures of Head Impact Severity: Mean linear acceleration per play exposure Mean rotational acceleration per play exposure Mean HITsp per play exposure	Three multivariate regression models using the enter method
RQ4C Chapter VII Overview II	c: Do cervical characteristics and visual performance predict ***cumulative game head impact frequency*** while controlling for play exposure?	Cervical characteristics: Composite peak torque Composite rate of torque development Composite CSA Composite Stiffness Composite Onset Latency Visual performance: Composite Visual Rating Score	Mean game head impact frequency per play exposure	One multivariate regression models using the enter method

Manuscript Legend

Research Question 1 a-f, regarding cervical characteristics, is addressed in manuscript format in Chapter IV.

Research Question 2 a-i, regarding visual performance, is addressed in manuscript format in Chapter V.

Research Question 3, regarding level of anticipation, is addressed in manuscript format in Chapter VI.

Research Question 4 a-c, regarding predicting head impact biomechanics, is addressed in the format of two separate overviewss (Overview I: a, Overview II: b&c) in Chapter VII.

For the purpose of this document, table and figure numbers are referenced by chapter number, followed by a period, and then followed by the sequence. For example, the first table in Chapter V is referenced as Table 5.1.

Chapter 4

MANUSCRIPT I

The Influence of Cervical Muscle Characteristics on Head Impact Biomechanics

Introduction

Context: By contracting the cervical musculature, an athlete is thought to reduce head acceleration following impact by increasing the effective mass to that of the head, neck, and thorax. **Objective**: To compare the odds of sustaining higher magnitude head impacts between athletes with higher and lower performance on cervical characteristic measures. **Design:** Prospective quasi-experimental. **Setting**: Laboratory/On-field. **Patients or Other Participants**: Forty-nine high school and collegiate American football players. **Interventions**: Athletes completed the cervical testing protocol, which included measures of cervical isometric strength, muscle size, and response to cervical perturbation prior to the season. Head impact biomechanics were captured for each player using the Head Impact Telemetry System. **Main Outcome Measures**: Each player was classified as either a high or low performer using a median split for each measure of isometric strength, muscle size, and response to cervical perturbation. We computed the odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile between players that were high performers relative to those that were low performers for each of the cervical characteristic measures. **Results**: Football athletes with stronger right and left lateral flexors and greater cervical muscle cross-sectional area had

increased odds of sustaining higher magnitude impacts compared to players with weaker cervical musculature and smaller muscle size. However, players with greater cervical stiffness and lower angular displacement following perturbation had reduced odds of sustaining higher magnitude head impacts compared to players with less cervical stiffness. **Conclusions**: Neuromuscular training aimed at enhancing cervical muscle stiffness may be useful in reducing the magnitude of head impacts sustained while playing football. The results of this study do not support the theory that players with stronger and larger cervical musculature are better able to mitigate head impact severity.

Contact sport athletes that are better able to mitigate head acceleration following impact are thought to be less likely to encounter brain tissue strain (Ommaya & Gennarelli, 1974). By contracting the cervical musculature, an athlete is thought to increase the effective mass of the head to that of the head, neck, and thorax (Mihalik, et al., 2011; Tierney, et al., 2008; Tierney, et al., 2005; Viano, et al., 2007). When the cervical musculature remains relaxed (e.g. when a player receives an unexpected hit), the impact force acts on the head's smaller effective mass allowing rapid head acceleration. For this reason, it has been speculated that athletes with insufficient cervical musculature strength are less able to generate adequate internal preparatory and reactive forces necessary to counter head acceleration and prevent concussion. However, the relationship between increases in cervical isometric strength following resistance training and reduced head acceleration remains theoretical (Mansell, et al., 2005; Mihalik, et al., 2011; Viano, et al., 2007).

Many researchers and clinicians theorize that cervical strength differences between adult and adolescent athletes may explain why high school football players experience a nearly twofold higher concussion rate than college football players (Gessel, et al., 2007; Guskiewicz, et al., 2000; Mihalik, et al., 2011; Tierney, et al., 2008; Tierney, et al., 2005; Viano, et al., 2007). Male college athletes possess stronger cervical musculature compared to female college athletes, and both male and female high school athletes (Hildenbrand & Vasavada, 2013). In fact, male high school athletes' neck musculatures are approximately 25% weaker than their college counterparts, which could limit their ability to dissipate forces applied to the head (Hildenbrand & Vasavada, 2013). Only one previous study has examined the role of neck strength in reducing in vivo head acceleration between players with strong, moderate, and weak cervical musculature during sport activity; however, the authors did not identify any significant differences between strength groups (Mihalik, et al., 2011; Viano, et al., 2007). Further evidence is needed to support the use of cervical strength and conditioning programs.

The cervical musculature's dynamic response following head impact is not determined by muscle strength alone. Cervical musculature, ligaments, and vertebral disks deform under the applied force when a football player sustains a head or body impact. Greater muscle girth and contraction of the primary stabilizing muscles increase muscle and joint stiffness (Simoneau, Denninger, & Hain, 2008; Wilson, et al., 1991). Viscoelastic properties of the cervical spine enable the cervical tissues to withstand brief periods of extreme loading that would otherwise exceed static load tolerance (McGill, et al., 1994). Preparatory muscle activation stiffens the neck and absorbs energy through eccentric contraction. Mathematical models comparing neck stiffness levels demonstrate that linear acceleration, angular acceleration, and head injury criterion variables decrease with increased neck stiffness (Queen, et al., 2003).

The purpose of this study was to determine whether high school and collegiate football players with superior cervical muscle characteristics—stronger, larger, and stiffer muscles—are have reduced odds of sustaining higher magnitude head impacts relative to players with inferior cervical characteristic measures.

Methods

Study Participants

Forty-nine football players (34 high school, 15 college) participated in our study. Demographic data are presented in Table 4.1. Participants were excluded if they reported a history of neurological disorder; prior cervical spine injury; current neck pain; had sustained a severe head injury within a year prior to study enrollment; had unexplained pain, upper or lower extremity weakness, numbness, gait disturbance, stiffness or spasm of the neck, or headaches; or had been previously diagnosed with Down Syndrome, rheumatoid arthritis, Klippel-Feil syndrome, or any abnormality of the cervical spine. All participants signed informed consent forms approved by our Institutional Review Board. Legal guardians of minor high school athletes also signed informed consents forms. All participants completed a brief examination of neck range of motion and stability to determine general neck health. Players were excluded from the study if they exhibited limited range of motion (n=1) or cervical instability (n=0).

Table 4.1. Demographic information for both high school and collegiate football players

	High School (n=34)		Collegiate (n=15)	
Demographic	Mean	SD	Mean	SD
Age (yrs)	16.6	0.9	20.5	1.4
Height (cm)	180.4	6.4	189.4	5.1
Mass (kg)	87.2	19.0	109.5	18.4
Neck Circumference (cm)	38.8	2.8	42.9	2.3
Head Circumference (cm)	58.4	2.0	59.9	2.3
Head-Neck Segment Length (cm)	25.0	1.9	25.8	1.9
Year (Athletic)				
Freshmen	0		3	
Sophomores	9		6	
Juniors	10		2	
Seniors	15		4	
Position Group				
Skill (offense, defense)	21 (7, 14)		7 (3, 4)	
Line (offense, defense)	13 (9, 4)		8 (5, 3)	

Measurements & Instrumentation

All participants completed the cervical testing protocol prior to the fall season. The cervical testing protocol consisted of an isometric strength assessment, ultrasonographic measures of cervical muscle size, and a cervical perturbation protocol. Participants completed a neck warm-up including ten neck circles clock-wise, ten neck circles counter-clock-wise followed by the following exercises in flexion, extension, right and left lateral flexion: manually resisted isokinetic muscle contractions through the full range of motion, 30 seconds of stretching, and a 20-second static physioball hold against a stationary wall (Mansell, et al., 2005; Tierney, et al., 2008; Tierney, et al., 2005). The cervical testing protocol components described below were completed in a block-randomized order because the ultrasound unit and the motion capture system were located in a separate laboratory from our strength-testing apparatus. Cervical perturbation and ultrasound imaging were always performed together with isometric strength testing taking place either immediately before or immediately after.

Isometric Strength

Isometric strength was measured using the HUMAC NORM Testing & Rehabilitation System (CSMi Medical Solutions, Inc., Stoughton, MA). Torque data were sampled at 2000 Hz, transmitted to a Biopac MP150 Data Acquisition System and host computer, and instantly viewed in AcqKnowledge 4.0 Software (Biopac Systems, Inc., Goleta, CA). We measured the peak torque and rate of torque development of the cervical flexors (supine), extensors (prone), right lateral flexors (side lying), and left lateral flexors (side lying) (Figure 4.1). All isometric strength measurements were assessed in the neutral position (0°) to optimize cervical muscle fascicle length (Suryanarayana & Kumar, 2005). A strap was wrapped circumferentially around each participant at the level of the scapular spine to stabilize the segment and prevent the participant from using compensatory trunk musculature strength (Rezasoltani, et al., 2008). During all trials, participants pushed directly against the padded strain gauge. The padding covering the strain gage was rigid enough to resist significant deformation, but soft enough to provide comfort to the participant encouraging them to put forth their maximal effort. The inferior border of the padded strain gauge was placed at a standardized location on the head for each direction (flexion: most inferior portion of nasal bone, extension: inferior border of the external occipital protuberance, right and left lateral flexion: most inferior portion of the ear lobe). A three-inch thick upholstered pad was placed beneath each participant's head during right and left lateral flexion trials.

Figure 4.1. Participant positioning for cervical spine isometric (A) flexor, (B) extensor, (C) right lateral flexor, (D) and left lateral flexor strength measures

Three familiarization trials with gradually increasing force were performed in each direction to acquaint participants with the testing position and measurement. Participants were instructed to reach their maximal force as quickly as possible and to sustain the force over the duration of the trial (Almosnino, et al., 2010). Participants were verbally encouraged to exert maximal effort during the three trials, each lasting three seconds. Participants rested for a minimum of 30 seconds between trials, but were allowed to rest for as long as they desired after each maximal voluntary contraction.

Ultrasonographic Cross-Sectional Area

Ultrasound images of the sternocleidomastoid (SCM), upper trapezius (UT), and semispinalis capitis (SSC) were obtained using an ultrasonographic imaging device (M-Turbo ultrasound system, SonoSite Inc., Bothell, WA USA) with a 7 MHz linear-array transducer that was 4 cm wide. The SCM, UT, and SSC were chosen because of their superficial location and role in multi-directional head stabilization (Bauer, et al., 2001; Tierney, et al., 2005; Vasavada, et al., 1998). For image consistency and because cervical cross-sectional area does not differ between the right and left, all images were taken on the athlete's right side (Arts, et al., 2010; O'Sullivan, et al., 2009). Sternocleidomastoid images were captured while participants were supine; and UT and SSC images were captured while participants were prone with their chest elevated on a bolster and face down in a c-shaped facial cushion (Figure 4.2b and 4.2c).

A single hyper-echoic marker was secured over the skin with medical tape to allow for later merging of the medial and lateral images of the muscle. A permanent marker was used to mark the location of each ultrasound site to ensure proper transducer placement. The hyper-echoic marker allowed for merging of the medial and lateral views of each muscle because none of the three muscles examined fit within the view of a single transducer width. Sternocleidomastoid images were taken at 50% of the distance between a line from the mastoid bone to the clavicular margin (Figure 4.2a) (Arts, et al., 2010). Upper trapezius images were taken by placing the transducer over the C6 spinous process and then tilting the transducer head in line with the skin curvature until the triangular shaped medial portion of the muscle was identified (Figure 4.2b) (Andersen, et al., 2008; O'Sullivan, et al., 2009). Upper trapezius images were taken two centimeters lateral to the triangular myofascial junction, perpendicular to the plane of the muscle belly. Semispinalis capitis images were taken by placing the transducer transversely at the midline over C3 (Figure 4.2c) (Rezasoltani, et al., 1998). The transducer head was tilted until the clearest muscle tissue image was observed on the monitor. Three consecutive measurements of the SCM, UT, and SSC were taken to minimize variation in echo intensity (Arts, et al., 2010).

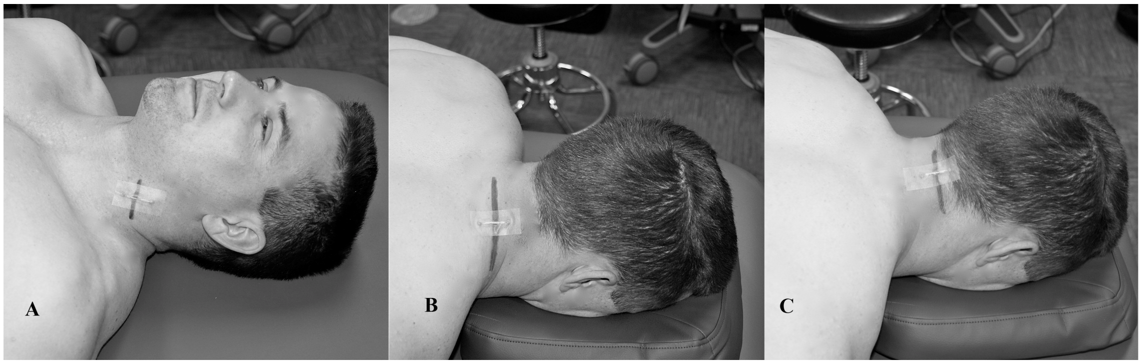

Figure 4.2. Cervical ultrasound set-up for measurement of (A) Sternocleidomastoid (B) Upper Trapezius (C) Semispinalis Capitis cross-sectional area.

Cervical Perturbation

We evoked a cervical perturbation by applying a load to the back of the head inducing force extension and front of the head inducing forced flexion (Figure 4.3). All participants wore a head harness adjusted to fit snugly with two attachment points; one affixed to the front and the other affixed to the back of the harness allowing for a pulley cord attachment. Prior to force application, the participants supported the load hanging freely with all slack removed from the pulley cord, and voluntarily moved into flexion or extension so that the participant could acclimate to the load that would be applied.

Tensile force, sampled at 1000 Hz, was measured throughout each trial using a load cell (Honeywell, International Inc., Morristown, NJ) attached in series with the head harness and aligned with the point of force application. The external force applicator consisted of a metal frame affixed to a wall, a height-adjustable pulley affixed to the metal frame, a pulley cord, and two external loads equal to 1.0% and 2.5% of the participant's body mass. The athlete supported the external load equal to 1.0% of body mass throughout all trials to standardize the preload. The second external load equal to 2.5% of body mass was dropped from a height of 15 cm, consistent with that of previous studies investigating cervical muscle stiffness (Reid, et al., 1981; Tierney, et al., 2008; Tierney, et al., 2005). The height of the pulley was adjusted for each participant so that force was applied perpendicular to the head-neck segment, which was visually confirmed using a bubble line level affixed to the pulley cord. The pulley cord was strung through an eyebolt attached to a height adjustable tripod so that the load cell remained perpendicular to the participant's head, but simultaneously would not prevent free movement of the pulley cord. Participants were instructed to activate their cervical musculature enough to support the preloaded weight and to avoid "clinching down". Participants were instructed to remain looking straight ahead (0°- neutral) and to resist the load from falling once they felt the tug. During anticipated trials, the mass equal to 2.5% of body mass was dropped from a 15 centimeter height by the primary investigator following a three second audible countdown.

Figure 4.3. Example cervical perturbation set-up during forced extension.

An electromagnetic motion capture system (trackSTAR, Ascension Technology Corp., Burlington, VT, USA) was used to measure three-dimensional head-neck segment sagittal plane angular displacement at 100 Hz. An electromagnetic sensor placed on the zygomatic arch tracked head movement. Another sensor, placed just below the sternal notch, tracked thorax movement (Mihalik, et al., 2008; Petschauer, et al., 2010; Toler, et al., 2010). Head movement was calculated relative to the thorax to derive head-to-thorax segment sagittal angular displacement as an estimate of cervical spine motion. Following sensor placement, each athlete stood still while anatomical landmarks were digitized, enabling the motion analysis system to recognize the head and thorax segments and orient the axes within the global coordinate system. The bridge of the nose, middle of the chin, and the occipital protuberance were used to digitize the head. Digitization points for the thorax included the T8 spinous process, L4 spinous process, xiphoid process, and C7 spinous process. We also digitized the sites of force application at anterior and posterior bracket of the head harness. Tensile force and head-neck segment displacement data were synchronized using Motion Monitor software (Innovative Sports Training, Inc., Chicago, IL).

Each participant first completed five anticipated force application trials in one of the directions (forced flexion or forced extension), followed by five unanticipated force application trials in the same direction (Mansell, et al., 2005). Forced flexion and forced extension trials were counterbalanced. During unanticipated trials, participants wore a vision blocking eye cover and noise cancelling ear-buds connected to a device playing white noise. Participants were instructed to support the preloaded weight and to resist the load from falling once they felt the tug. The mass equal to 2.5% of body mass was dropped from a 15-centimeter height by the primary investigator at a random time point following the instructions.

Sternocleidomastoid (SCM) and upper trapezius (UT) electromyography (EMG) data were collected to compute muscle onset latency using preamplified surface EMG electrodes

(Bagnoli 8 Desktop EMG System; DelSys Inc. Boston, MA) (inter-electrode distance= 10 mm; amplification factor = 10,000, 20–450 Hz; Common Mode Rejection Ratio = 60 Hz > 80 dB; input impedance > 10^{15} ohms). Muscle activity was measured on the right side only. For the SCM, the electrode was placed along the sternal head, centered at one-third of the distance between the mastoid process and the sternal notch (Almosnino, et al., 2009; Falla, et al., 2002a, 2002b). For the UT, we placed the electrode two centimeters lateral to the midpoint of the C4-C5 spinous processes and oriented along the palpated anterior border of the trapezius, in line with the direction of the muscle fibers. The reference electrode was placed on the dorsal wrist. These electrode placements were previously reported as reliable (Almosnino, et al., 2009; A. Burnett, et al., 2007). One collegiate player chose to discontinue the protocol after the first five trials of forced flexion reporting a mild headache and one high school player after five trials of forced extension declining to disclose why he chose to discontinue, but stated that he was not experiencing any pain or discomfort.

Head Impact Biomechanics

Head impact biomechanics were measured at each practice and game over the course of the 2012 football season using the Head Impact Telemetry (HIT) System technology (Riddell Corp., Elyria, OH). The HIT System consists of MxEncoder units located in the football helmets, a signal transducer, and a laptop computer that houses the Sideline Response System (Riddell Corp., Elyria, OH). MxEncoder units embedded within the helmets are comprised of six spring-loaded single-axis accelerometers, a telemetry unit, a data storage device, and a battery power source. The MxEncoders were retrofit into Riddell Revolution and Speed helmet designs (Riddell Inc., Elyria, Ohio). Each single-axis accelerometer collects data at 1 kHz for a period of 40 ms (8 ms prior to the data collection trigger and 32 ms after the trigger). Data are time-stamped, encoded, and then transmitted in real-time to the signal transducer via radiofrequency transmission at 903–927 MHz. The signal transducer is connected through a USB port to a laptop computer, which stores all head impact data. The HIT System transmits accelerometer data from distances well in excess of the length of the standard American football field.

Data Reduction

Cervical isometric strength, cervical perturbation, and raw head impact biomechanical were reduced using separate custom Matlab 7 data reduction programs (The Mathworks, Inc., Natick, MA).

Isometric Strength

Peak torque and rate of torque development were computed and averaged across the three trials. Raw torque data were zero offset and filtered with a low pass, zero lag, Butterworth filter of 10 Hz. The moment required to overcome gravity's influence on the head and neck mass was added to each torque value. Head mass was calculated by using the following regression equation: Head & Neck Mass = Body Mass * 0.0534 +2.33 (*Anthropometric Source Book Volume I: Anthropometry for Designers*, 1978). The moment arm of the head and neck's center of mass was estimated as 50.02% of the distance head-neck segment length (distance from C7 to the apex of the head measured with a clinical tape measure) (de Leva, 1996). We normalized the maximum torque (Nm) by dividing it by body mass in kilograms (Nm/kg), and averaged across the three trails. Composite peak torque was calculated by summing the normalized peak torque values across each direction (flexion, extension, right lateral flexion, and left lateral flexion).

Rate of torque development (Nm/s) was calculated by identifying the greatest slope of the torque-time curve, using a 50-millisecond sliding window from torque onset to peak torque (Almosnino, et al., 2010). Composite rate of torque development was calculated by summing rate of torque development across each direction.

Ultrasonographic Cross-Sectional Area

SCM, UT, and SSC ultrasonographic images were exported to a public domain image processing and analysis program (Image J, National Institutes of Health, USA), merged in reference to the hyper-echoic markers, and outlined to calculate cross-sectional area. The primary investigator completed all measurements by tracing the interface between the hyper-echoic fascia and the hypo-echoic muscle tissue for each muscle. Cross-sectional areas were averaged across the three images for each muscle. Composite cross-sectional area for each athlete was calculated by summing cross-sectional area of the SCM, UT, and SSC.

Cervical Stiffness

Kinematic data were zero offset and filtered with a low pass, zero lag, Butterworth filter at 10 Hz. Euler angles were used to calculate the movement of the head relative to the thorax. Orthogonal axes were defined in the order of flexion-extension (Y-axis), right and left rotation (Z-axis), and right and left lateral flexion (X-axis) (James, et al., 2004). Positive motions were flexion, left rotation, and right lateral flexion.

Stiffness was calculated as the ratio of the change in moment to the change in sagittal angular displacement of the head-neck segment relative to the thorax between peak moment and moment offset (Nm/rad). We averaged the stiffness values from trials 2-5 separately for the anticipated forced flexion and extension to eliminate a possible exaggerated neuromuscular startle response often observed during the first exposure to a transient acceleration startle response (Siegmund, et al., 2008). We observed clipping of load cell data due to tensile overloading during some trials (anticipated forced extension: 115 trials, 48.32%; anticipated forced flexion: 99 trials, 41.77%). For trials where clipping was evident, we used a regression equation to estimate peak moment derived from 30 trials where clipping did not occur. The regression equation used the participant's body mass, the last observed moment value prior to clipping, and the moment value at 50% of the peak (the value at the time point half way between the onset of force and the estimated time of peak moment assuming peak moment was reached at the midpoint of the clipped data) and predicted 96% of the variance in peak moment. We observed good reliability between the computed peak tensile load and the actual peak tensile load applied ($ICC_{3,1}$=0.92, SEM=1.19Nm). Stiffness values were normalized to each player's mass (N)*height (m). We summed stiffness values across anticipated forced extension and anticipated forced flexion conditions to compute composite stiffness for each athlete.

Peak angular displacements (rad) were calculated by identifying the absolute value of maximum displacement of the head relative to the thorax in the sagittal plane. We averaged angular displacement values from trials 2-5 separately for the anticipated forced flexion and extension. We calculated composite angular displacement by summing angular displacement across anticipated forced extension and anticipated forced flexion conditions for each athlete.

Cervical Electromyographic Measurement

The analog EMG signal was converted to a digital signal by an analog-to-digital converter card. The signal was amplified (gain 100-1000) with a single-ended amplifier and common mode rejection ratio of 130 dB at direct current. The raw digital signal was exported to a custom data reduction program in Matlab 7 where it was rectified, zero offset, filtered with a low pass, zero lag, Butterworth filter with a cutoff frequency of 10 Hz, and then smoothed using a root mean square algorithm over a 20ms-moving window. Muscle onset latency was calculated as the time in milliseconds (ms) between force application and the point at which myoelectric activity exceeded nine times the resting mean for the SCM and four times the resting mean for the UT. Force application was identified as the time point at which the load cell voltage exceeded 0.5 Nm. Resting EMG data for the SCM was very low requiring a high threshold to determine onset, however, resting EMG data for the UT data are higher because of the postural nature of the muscle (Sommerich, et al., 2000). Muscle onset latency was calculated for trials 2-5 of unanticipated forced flexion and unanticipated forced extension trials only. We excluded trials when the onset of muscle activity was ambiguous, such as when muscle activity rises briefly, but then returns to resting (unanticipated forced extension: 32 trials, 13.06%; unanticipated forced flexion: 34 trials, 13.88%). We computed composite muscle onset latency by summing the SCM onset latency measured during unanticipated forced extension trials with the UT onset latency measured during unanticipated forced flexion trials.

We split participants into a group of high and a group of low performers for each cervical characteristic using a median split. Table 4.3 includes the unit of measure and high performance category for each cervical characteristic variable.

Head Impact Biomechanics

Head impact data were exported from the Sideline Response System into Matlab 7. Consistent with previous studies, we then reduced the data to include only those impacts that register a linear acceleration greater than or equal to 10g (Guskiewicz, Mihalik, et al., 2007;

Mihalik, et al., 2007; Mihalik, Blackburn, et al., 2010; Mihalik, Greenwald, et al., 2010; Mihalik, et al., 2011; Schnebel, et al., 2007). This study focused on the three following measures of head impact magnitude: (1) peak linear acceleration (g), (2) peak rotational acceleration (rad/sec^2), and (3) Head Impact Technology severity profile - a weighted composite score including linear acceleration, rotational acceleration, impact duration, and impact location. Computation of these biomechanical measures have previously been reported (Greenwald, et al., 2008).

We categorized the linear acceleration, rotational acceleration, and HITsp of each head impacts into quartiles (Table 4.2). We also created separate categories for head impacts that occurred in the 95th and 99th percentiles. We chose to include the 95th and 99th percentile categories because collapsing head impacts of this magnitude into the 4th quartile may have limited the detail of our results regarding impacts that are considered more severe. We excluded head impacts that occurred to the top of the head because loading transmitted directly through the spinal column does not engage the large moment-generating, superficial, cervical musculature examined in this study (SCM, UT, SSC) (Banerjee, et al., 2004; Swartz, et al., 2005).

Table 4.2. Head impact biomechanics categorization cutoffs and frequencies.

	1st Quartile (1st-24th)	**2nd Quartile (25th-50th)**	**3rd Quartile (50th-74th)**	**4th Quartile (75th-94th)**	**95th Percentile (95th-98th)**	**99th Percentile (99th-100th)**
Linear Acceleration (g)	<14.8 (n=4885)	14.8 ≥ or < 20.1 (n=4927)	20.1 ≥ or < 30.1 (n=5018)	30.1 ≥ or < 56.5 (n=3953)	56.5 ≥ or < 87.4 (n=793)	≥ 87.4 (n=199)
Rotational Acceleration (rad/s^2)	<1067.6 (n=4943)	1067.6 ≥ or < 1500.9 (n=4944)	1500.9 ≥ or < 2198.1 (n=4944)	2198.1 ≥ or < 4162.2 (n=3956)	4162.2 ≥ or < 6528.5 (n=791)	≥ 6528.52 (n=197)
HITsp	<11.9 (n=4883)	11.9 ≥ or < 14.4 (n=4860)	14.4 ≥ or < 18.4 (n=5043)	18.4 ≥ or < 31.8 (n=3995)	31.8 ≥ or < 53.8 (n=796)	≥ 53.8 (n=198)

Statistical Analyses

Random intercepts, general mixed, linear, proportional odds models were used to compute odds ratios (OR) and 95% confidence intervals (CI) for each dichotomized cervical spine characteristic measure. Our predictor variables included each measure and composite measure of cervical isometric strength (5 peak torque measures, 5 rate of torque development measures), muscle size (4 cross sectional area measures), and perturbation (3 stiffness measures, 3 angular displacement measures, 3 latency measures). We computed the odds of sustaining head impacts in the 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile versus the reference category of head impacts in the 1st quartile across groups of high and low performers for each cervical characteristic for each of the following categorized measures of head impact magnitude: linear acceleration, rotational acceleration, and HITsp. For all models, 1st quartile head impacts and low performers were the reference categories. We included position group assignment (skill or line) in the model to control for differences in player types. Because we suspected that the collegiate athletes may have stronger, larger, and more stiff cervical musculature, we first analyzed group dispersions between the high school and collegiate athletes across the high and low classifications for all outcome variables using a 2 (high school, collegiate) x 2 (high, low) chi-squared goodness of fit analysis. For analyses that involved measures where dispersion was not even, we included playing level into the model as a predictor. Statistical analyses were repeated individually studying skill and line players, respectively. All statistical analyses were completed using SAS 9.3 (SAS Institute Inc., Cary, NC). Odds ratio values greater than 1 indicate an increased odds among athletes categorized into the high performance group; whereas, odds ratios below 1 indicate a reduced odds among the high performance group. Analyses were considered significant if the 95% confidence interval about the odds ratio did not include one.

Table 4.3. Cervical characteristic variable table indicating the unit of measure and high performance categories.

Cervical Characteristics	Unit of Measure	Below the Median	Above the Median
Cervical Isometric Strength			
Peak Torque (flexion, extension, right lateral flexion, left lateral flexion, composite)	Newton-meters per kilogram (Nm/kg)	Weaker	Stronger (High Performance)
Rate of Torque Development (flexion, extension, right lateral flexion, left lateral flexion, composite)	Newton-meters per second (Nm/sec)	Slower	Faster (High Performance)
Cervical Muscle Size			
Cross Sectional Area (SCM, UT, SSC, composite)	Squared centimeters (cm^2)	Smaller	Larger (High Performance)
Cervical Perturbation			
Angular Displacement (anticipated forced extension, anticipated forced flexion, composite)	Radians (rad)	Less Displacement (High Performance)	More Displacement
Stiffness (anticipated forced extension, anticipated forced flexion, composite)	Newton-meters per radian (Nm/rad)	Less Stiff	More Stiff (High Performance)
Muscle Onset Latency (unanticipated forced extension, unanticipated forced flexion, composite)	Milliseconds (ms)	Faster (High Performance)	Slower

Results

Descriptive statistics for all cervical characteristic variables are presented in Table 4.4. The high performance group performed significantly better than the low performance group for all cervical characteristics ($p<0.001$). Collegiate players were more commonly classified as high performers for the following measures: extension rate of torque development ($\chi^2=5.13$, $p=0.024$), SCM cross sectional area ($\chi^2=12.29$, $p<0.001$), SSC cross sectional area ($\chi^2=12.29$, $p<0.001$), and composite cross sectional area ($\chi^2=17.02$, $p<0.001$). Odds ratios and 95% confidence intervals for the group overall, skill players only, and linemen only, are presented in Tables 1.5, 1.6, and 1.7 (cervical isometric strength), 1.8, 1.9 and 1.10 (cervical muscle size), and 1.11, 1.12, and 1.13 (cervical perturbation).

Cervical Isometric Strength

For a majority of our analyses regarding cervical isometric strength, players had equal odds of sustaining impacts in the first quartile compared to the higher magnitude categories regardless of strength for all muscle groups. However, players with stronger right lateral flexors had higher odds of sustaining head impacts in the 2nd quartile rather than in the 1st quartile (OR: 1.29; 95%CI: 1.04, 1.60) compared to players with weaker right lateral flexors when head impact severity was measured in HITsp. Likewise, players with stronger left lateral flexors had nearly 2-fold odds of sustaining head impacts in the 99th percentile compared to the 1st (OR: 1.96; 95%CI: 1.13, 3.39) compared to players with weaker left lateral flexors, as measured by rotational acceleration.

Skill players had equal odds of sustaining impacts in the first quartile compared to all other higher magnitude categories regardless of strength for all muscle groups. However, linemen with stronger left lateral flexors had higher odds of sustaining head impacts in the 95th percentile (linear: OR: 1.84; 95%CI: 1.04, 3.25, rotational: OR: 2.15; 95%CI: 1.24, 3.72, HITsp: OR: 2.34; 95%CI: 1.12, 4.90) compared to weaker linemen (Table 4.7). Linemen with stronger left lateral flexors were also more likely to sustain head impacts in the 2nd and 4th quartiles, rather than in the 1st, compared to linemen with weaker cervical musculature. Linemen with stronger extensor muscles were also approximately 46% more likely to sustain head impacts in the 2nd and 3rd quartiles, rather than in the 1st, compared to linemen with weaker extensor cervical musculature.

Players had equal odds of sustaining impacts in the 1st quartile compared to higher quartiles regardless of how quickly they developed torque for all muscle groups. However, players who developed flexor torque more quickly, relative to those who developed torque more slowly, had reduced odds of sustaining linear head impacts in the second quartile compared to the first (OR: 0.89; 95%CI: 0.82, 0.98) (Table 4.5). Skill players and linemen had equal odds regardless of the rapidity of torque development.

Cervical Muscle Size

Much like our findings regarding cervical isometric strength, we observed that players with larger SCM, UT, SSC, and composite muscle cross sectional area, relative to those with smaller muscle size, generally had increased odds of sustaining head impacts in all of the higher percentile categories, rather than in the first quartile (Table 4.8). Similar results were observed among skill players and linemen; however, this effect was more pronounced among the skill players (Table 4.9 and Table 4.10).

Cervical Perturbation

Players with higher anticipated forced extension stiffness and composite cervical stiffness had reduced odds of sustaining head impacts in the 2nd, 3rd, and 4th quartiles, rather than in the 1st quartile (Table 4.11). Likewise, players with higher anticipated forced flexion stiffness had reduced odds of sustaining head impacts in the 2nd and 3rd quartiles, rather than in the 1st quartile. Similar results were observed among linemen, but skill players also presented with reduced odds of sustaining head impacts in the 95th percentile (Table 4.12 & Table 4.13).

Players with smaller angular displacements following perturbation had reduced odds of sustaining both head impacts in the 2^{nd}, 3^{rd}, and 4^{th} quartile, rather than the 1^{st}, compared to players with larger angular displacements (Table 4.11). Oddly, players that were high performers for anticipated forced extension stiffness (OR: 1.51; 95%CI: 1.01, 2.26) and anticipated forced extension angular displacement (OR: 1.54; 95%CI: 1.04, 2.29) had approximate 51% increased odds of sustaining head impacts in the 99^{th} percentile rather than head impacts in the 1^{st} quartile.

Generally, we did not observe any differences in odds between players for muscle onset latencies following cervical perturbation. However, linemen with quicker muscle onset latencies had increased odds of sustaining head impacts in the 2^{nd} quartile rather than the 1^{st} compared to linemen with slower muscle onset latencies (OR: 1.17; 95%CI: 1.01, 1.36) (Table 4.11).

Table 4.4. Descriptive statistics and between group comparisons for low and high performers for each cervical characteristic

	Low					High						
	n	Line†	Skill‡	Mean	SD	n	Line	Skill	Mean	SD	t	p*
Cervical Isometric Strength												
Peak Torque (Nm/kg)												
Flexion	25	14	11	0.18	0.03	24	6	18	0.30	0.06	-8.26	<0.001
Extension	25	15	10	0.43	0.08	24	5	19	0.62	0.06	-10.08	<0.001
Right Lateral Flexion	25	16	9	0.39	0.07	24	4	20	0.63	0.09	-10.10	<0.001
Left Lateral Flexion	25	15	10	0.37	0.07	24	5	19	0.60	0.09	-9.61	<0.001
Composite	25	9	16	1.41	0.19	24	4	20	2.11	0.22	-11.74	<0.001
Rate of Torque Development (Nm/sec)												
Flexion	25	12	13	51.60	13.50	24	8	16	110.33	38.56	-7.06	<0.001
Extension	25	9	16	174.03	51.24	24	11	13	360.50	122.95	-6.88	<0.001
Right Lateral Flexion	25	10	15	145.42	39.59	24	10	14	315.54	93.62	-8.34	<0.001
Left Lateral Flexion	25	13	12	147.55	42.35	24	7	17	304.84	97.89	-7.25	<0.001
Composite	25	9	16	570.99	112.22	24	11	13	1036.63	295.32	-7.35	<0.001
Cervical Muscle Size (cm^2)												
Sternocleidomastoid	25	8	17	4.65	0.60	24	12	12	7.00	0.98	-10.15	<0.001
Upper Trapezius	25	11	14	2.47	0.60	24	9	15	4.74	0.71	-12.16	<0.001
Semispinalis Capitis	25	10	15	4.70	0.63	24	10	14	7.05	0.99	-9.85	<0.001
Composite	25	10	15	12.65	1.72	24	10	14	17.91	2.18	-9.40	<0.001
Cervical Perturbation												
Stiffness (Nm/rad) -Normalized												
Anticipated Forced Extension	25	16	9	0.24	0.07	23	3	20	0.85	0.70	-4.13	<0.001
Anticipated Forced Flexion	25	14	11	0.25	0.05	23	5	18	0.60	0.32	-5.34	<0.001
Composite	24	14	10	0.52	0.11	23	4	19	1.41	0.86	-4.90	<0.001
Angular Displacement (rad)												
Anticipated Forced Extension	25	6	19	0.19	0.03	23	13	10	0.28	0.05	-8.35	<0.001
Anticipated Forced Flexion	24	10	14	0.09	0.03	24	9	15	0.16	0.02	-9.06	<0.001
Composite	24	7	17	0.29	0.06	23	11	12	0.43	0.06	-7.94	<0.001
Muscle Onset Latency (ms)												
Unanticipated Forced Extension	23	6	17	50.30	12.67	24	12	12	26.00	8.13	7.86	<0.001
Unanticipated Forced Flexion	23	9	14	43.80	15.43	23	9	15	26.32	10.91	4.50	<0.001
Composite	23	7	16	91.32	14.79	24	11	13	55.67	11.60	9.29	<0.001

† Line: defensive end, nose tackle, defensive tackle, center, guard, or offensive tackle

‡ Skill: linebacker corner, or safety, quarterback, receiver, tight end, running back, or full back

*** High and low performers were significantly different for all cervical characteristic variables.**

Table 4.5. Cervical isometric strength (Group overall): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

		df	2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Peak Torque			**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Flexion	Linear	34	0.96 (0.87-1.06)	0.97 (0.83-1.12)	0.91 (0.74-1.11)	0.98 (0.66-1.43)	0.87 (0.57-1.33)
	Rotational	34	0.97 (0.84-1.12)	0.94 (0.80-1.10)	0.92 (0.73-1.15)	1.13 (0.79-1.63)	0.95 (0.53-1.69)
	HITsp	34	1.17 (0.95-1.44)	1.06 (0.82-1.38)	1.00 (0.72-1.39)	0.86 (0.52-1.42)	1.05 (0.58-1.89)
Extension	Linear	34	1.05 (0.94-1.17)	1.04 (0.89-1.22)	0.96 (0.78-1.19)	0.96 (0.64-1.45)	1.00 (0.63-1.57)
	Rotational	34	1.13 (0.98-1.31)	1.13 (0.96-1.34)	1.03 (0.81-1.31)	1.02 (0.69-1.50)	1.12 (0.59-2.12)
	HITsp	34	1.08 (0.87-1.36)	1.08 (0.82-1.41)	0.98 (0.69-1.39)	1.02 (0.60-1.73)	0.88 (0.47-1.66)
Right Lateral Flexion	Linear	34	1.02 (0.92-1.13)	1.05 (0.90-1.24)	1.08 (0.87-1.34)	1.06 (0.70-1.58)	1.10 (0.71-1.70)
	Rotational	34	0.95 (0.82-1.10)	1.04 (0.87-1.24)	1.06 (0.83-1.35)	1.17 (0.80-1.72)	1.42 (0.77-2.59)
	HITsp	34	***1.29 (1.04-1.60) †***	1.25 (0.95-1.64)	1.32 (0.94-1.87)	1.27 (0.75-2.15)	1.59 (0.88-2.88)
Left Lateral Flexion	Linear	34	1.03 (0.93-1.14)	1.06 (0.91-1.23)	1.10 (0.90-1.35)	1.08 (0.73-1.59)	1.19 (0.78-1.82)
	Rotational	34	1.02 (0.88-1.18)	1.10 (0.94-1.30)	1.16 (0.92-1.46)	1.26 (0.88-1.82)	***1.96 (1.13-3.39) †***
	HITsp	34	1.16 (0.94-1.44)	0.91 (0.91-1.55)	1.22 (0.87-1.71)	1.15 (0.69-1.93)	1.40 (0.78-2.51)
Composite	Linear	34	0.96 (0.86-1.06)	0.95 (0.81-1.11)	0.91 (0.74-1.13)	0.96 (0.64-1.46)	1.05 (0.68-1.63)
	Rotational	34	0.99 (0.85-1.15)	1.00 (0.84-1.19)	0.97 (0.76-1.24)	1.02 (0.69-1.50)	1.41 (0.77-2.57)
	HITsp	34	1.05 (0.84-1.32)	0.99 (0.75-1.31)	0.99 (0.69-1.42)	0.93 (0.54-1.59)	1.20 (0.65-2.22)
Rate of Torque Development							
Flexion	Linear	34	***0.89 (0.82-0.98) ‡***	0.97 (0.84-1.12)	0.93 (0.77-1.13)	0.86 (0.59-1.24)	0.71 (0.48-1.07)
	Rotational	34	1.03 (0.90-1.18)	1.02 (0.87-1.20)	1.02 (0.82-1.28)	1.20 (0.84-1.70)	0.86 (0.49-1.50)
	HITsp	34	1.01 (0.82-1.24)	1.06 (0.82-1.36)	0.99 (0.72-1.37)	0.84 (0.51-1.36)	0.86 (0.49-1.51)
Extension	Linear	29*	1.03 (0.93-1.15)	1.12 (0.96-1.28)	1.10 (0.90-1.35)	1.25 (0.85-1.83)	***1.55 (1.01-2.40) †***
	Rotational	29*	1.07 (0.92-1.24)	1.13 (0.96-1.33)	1.10 (0.87-1.39)	1.28 (0.89-1.84)	1.61 (0.89-2.90)
	HITsp	29*	1.02 (0.85-1.24)	1.05 (0.82-1.33)	1.07 (0.78-1.48)	1.13 (0.69-1.84)	1.63 (0.92-2.89)
Right Lateral Flexion	Linear	34	1.01 (0.92-1.11)	1.14 (1.00-1.31)	1.17 (0.97-1.41)	1.32 (0.92-1.89)	1.20 (0.80-1.80)
	Rotational	34	0.99 (0.86-1.13)	1.03 (0.88-1.20)	1.16 (0.94-1.44)	1.19 (0.84-1.68)	1.14 (0.65-2.00)
	HITsp	34	1.13 (0.92-1.38)	1.07 (0.84-1.38)	1.19 (0.87-1.62)	1.28 (0.79-2.06)	1.28 (0.74-2.24)
Left Lateral Flexion	Linear	34	1.04 (0.94-1.14)	1.14 (0.99-1.31)	1.20 (0.99-1.45)	1.27 (0.88-1.83)	1.38 (0.93-2.05)
	Rotational	34	0.95 (0.83-1.09)	1.05 (0.90-1.23)	1.15 (0.93-1.44)	1.12 (0.78-1.60)	1.34 (0.76-2.37)
	HITsp	34	1.10 (0.89-1.35)	1.11 (0.86-1.43)	1.19 (0.87-1.65)	1.28 (0.79-2.09)	1.37 (0.78-2.42)
Composite	Linear	34	0.99 (0.90-1.09)	1.13 (0.99-1.30)	1.12 (0.93-1.35)	1.15 (0.79-1.66)	1.06 (0.70-1.60)
	Rotational	34	0.99 (0.86-1.13)	1.01 (0.87-1.19)	1.09 (0.88-1.36)	1.08 (0.76-1.54)	1.17 (0.67-2.07)
	HITsp	34	1.10 (0.90-1.35)	1.12 (0.87-1.44)	1.16 (0.85-1.59)	1.17 (0.72-1.90)	1.19 (0.68-2.09)

† Players classified as high performers had increased odds (OR greater than 1 indicate increased odds for the high group)

‡ Players classified as high performers had reduced odds (OR less than 1 indicate reduced odds for the high group)

*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance groups.**

Table 4.6. Cervical isometric strength (Skill players only): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

		df	2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Peak Torque			**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Flexion	Linear	19	0.91 (0.80-1.03)	0.90 (0.71-1.13)	0.85 (0.64-1.14)	0.85 (0.50-1.44)	0.79 (0.48-1.30)
	Rotational	19	0.85 (0.71-1.02)	0.84 (0.69-1.02)	0.78 (0.61-1.00)	0.92 (0.58-1.46)	0.71 (0.34-1.45)
	HITsp	19	1.12 (0.83-1.51)	0.95 (0.67-1.35)	0.87 (0.54-1.40)	0.68 (0.34-1.37)	1.04 (0.46-2.33)
Extension	Linear	19	1.01 (0.89-1.15)	0.98 (0.77-1.24)	0.91 (0.68-1.21)	0.89 (0.52-1.53)	1.00 (0.60-1.65)
	Rotational	19	0.97 (0.80-1.18)	0.99 (0.80-1.22)	0.91 (0.70-1.19)	0.98 (0.61-1.58)	1.42 (0.66-3.06)
	HITsp	19	0.98 (0.72-1.33)	0.89 (0.62-1.28)	0.87 (0.53-1.42)	0.81 (0.40-1.66)	1.10 (0.48-2.53)
Right Lateral Flexion	Linear	19	1.07 (0.94-1.22)	1.07 (0.85-1.36)	1.07 (0.79-1.43)	0.82 (0.47-1.41)	0.98 (0.58-1.65)
	Rotational	19	0.86 (0.72-1.03)	0.99 (0.81-1.22)	0.94 (0.72-1.23)	0.88 (0.54-1.42)	1.15 (0.54-2.48)
	HITsp	19	1.31 (0.97-1.76)	1.24 (0.87-1.77)	1.29 (0.79-2.10)	0.93 (0.45-1.92)	1.55 (0.68-3.50)
Left Lateral Flexion	Linear	19	1.03 (0.90-1.17)	0.99 (0.78-1.26)	0.96 (0.72-1.28)	0.77 (0.45-1.32)	1.09 (0.64-1.86)
	Rotational	19	0.94 (0.77-1.14)	1.03 (0.84-1.27)	0.94 (0.72-1.23)	0.86 (0.54-1.39)	1.62 (0.78-3.40)
	HITsp	19	1.03 (0.76-1.41)	1.02 (0.71-1.47)	0.97 (0.59-1.59)	0.72 (0.35-1.48)	1.12 (0.49-2.60)
Composite	Linear	19	0.96 (0.85-1.10)	0.90 (0.71-1.13)	0.82 (0.61-1.09)	0.64 (0.38-1.08)	0.85 (0.51-1.42)
	Rotational	19	0.94 (0.78-1.14)	0.95 (0.77-1.16)	0.84 (0.64-1.08)	0.73 (0.46-1.17)	1.20 (0.56-2.55)
	HITsp	19	0.95 (0.69-1.30)	0.86 (0.60-1.24)	0.83 (0.50-1.37)	0.58 (0.29-1.18)	1.01 (0.43-2.34)
Rate of Torque Development							
Flexion	Linear	19	0.89 (0.78-1.01)	1.07 (0.86-1.34)	0.97 (0.73-1.29)	0.74 (0.44-1.25)	0.69 (0.43-1.09)
	Rotational	19	1.07 (0.88-1.29)	1.08 (0.89-1.32)	0.98 (0.76-1.27)	0.98 (0.62-1.56)	0.68 (0.34-1.36)
	HITsp	19	1.00 (0.74-1.34)	1.15 (0.81-1.62)	0.97 (0.60-1.56)	0.71 (0.36-1.39)	0.81 (0.36-1.79)
Extension	Linear	14*	1.02 (0.89-1.18)	1.19 (0.94-1.51)	1.22 (0.91-1.64)	1.51 (0.88-2.59)	1.61 (0.96-2.70)
	Rotational	14*	1.00 (0.82-1.22)	1.01 (0.82-1.26)	1.09 (0.83-1.44)	1.54 (0.96-2.47)	1.93 (0.92-4.06)
	HITsp	14*	1.06 (0.79-1.43)	1.09 (0.77-1.55)	1.18 (0.72-1.94)	1.39 (0.68-2.83)	2.12 (0.99-4.52)
Right Lateral Flexion	Linear	19	1.06 (0.93-1.21)	1.22 (0.99-1.51)	1.25 (0.96-1.62)	1.22 (0.74-2.03)	1.39 (0.87-2.23)
	Rotational	19	0.92 (0.76-1.10)	0.98 (0.80-1.20)	1.12 (0.88-1.44)	1.13 (0.72-1.77)	1.36 (0.66-2.81)
	HITsp	19	1.06 (0.79-1.42)	0.94 (0.67-1.33)	1.04 (0.65-1.66)	1.04 (0.52-2.07)	1.73 (0.79-3.76)
Left Lateral Flexion	Linear	19	1.10 (0.97-1.24)	1.20 (0.96-1.50)	1.19 (0.90-1.57)	1.08 (0.64-1.82)	1.46 (0.90-2.36)
	Rotational	19	0.96 (0.79-1.15)	1.05 (0.86-1.29)	1.10 (0.86-1.43)	1.04 (0.65-1.66)	1.60 (0.77-3.34)
	HITsp	19	1.05 (0.78-1.43)	1.07 (0.75-1.51)	1.05 (0.65-1.70)	1.07 (0.53-2.15)	1.40 (0.62-3.14)
Composite	Linear	19	0.97 (0.85-1.10)	1.19 (0.96-1.47)	1.15 (0.88-1.51)	1.11 (0.67-1.86)	1.12 (0.69-1.81)
	Rotational	19	0.90 (0.75-1.09)	0.93 (0.75-1.14)	1.01 (0.78-1.30)	1.01 (0.63-1.60)	1.20 (0.58-2.47)
	HITsp	19	0.97 (0.72-1.31)	1.01 (0.71-1.42)	1.05 (0.66-1.69)	0.93 (0.46-1.86)	1.40 (0.64-3.08)

*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance groups**

Table 4.7. Cervical isometric strength (Linemen only): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

Cervical Isometric Strength		df	2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Peak Torque			**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Flexion	Linear	10	1.05 (0.88-1.24)	1.06 (0.85-1.31)	0.99 (0.72-1.37)	1.15 (0.60-2.18)	0.98 (0.39-2.43)
	Rotational	10	1.15 (0.90-1.47)	1.10 (0.82-1.47)	1.20 (0.75-1.90)	1.55 (0.84-2.89)	1.55 (0.57-4.27)
	HITsp	10	1.24 (0.91-1.68)	1.25 (0.82-1.91)	1.22 (0.74-1.99)	1.20 (0.52-2.77)	1.07 (0.40-2.85)
Extension	Linear	10	1.12 (0.92-1.36)	1.14 (0.89-1.45)	1.04 (0.72-1.50)	1.10 (0.54-2.24)	1.01 (0.37-2.79)
	Rotational	10	***1.47 (1.19-1.83)*** †	***1.46 (1.10-1.94)*** †	1.28 (0.79-2.09)	1.08 (0.51-2.29)	0.59 (0.15-2.30)
	HITsp	10	1.31 (0.92-1.86)	1.50 (0.96-2.34)	1.23 (0.72-2.11)	1.50 (0.60-3.75)	0.49 (0.15-1.67)
Right Lateral Flexion	Linear	10	0.94 (0.78-1.13)	1.03 (0.82-1.31)	1.09 (0.76-1.55)	1.55 (0.79-3.05)	1.48 (0.61-3.62)
	Rotational	10	1.08 (0.82-1.42)	1.09 (0.78-1.52)	1.30 (0.78-2.16)	1.85 (0.98-3.52)	1.93 (0.67-5.55)
	HITsp	10	1.26 (0.90-1.78)	1.26 (0.78-2.04)	1.35 (0.78-2.34)	2.08 (0.87-4.95)	1.65 (0.63-4.33)
Left Lateral Flexion	Linear	10	1.05 (0.89-1.23)	1.17 (0.96-1.44)	1.34 (0.99-1.82)	***1.84 (1.04-3.25) †***	1.45 (0.63-3.30)
	Rotational	10	1.13 (0.88-1.44)	1.19 (0.89-1.60)	***1.59 (1.04-2.44) †***	***2.15 (1.24-3.72) †***	2.43 (0.96-6.11)
	HITsp	10	***1.38 (1.02-1.85)*** †	1.48 (0.98-2.25)	***1.71 (1.08-2.71) †***	***2.34 (1.12-4.90) †***	1.89 (0.81-4.39)
Composite	Linear	10	0.94 (0.78-1.13)	1.03 (0.82-1.31)	1.09 (0.76-1.55)	1.55 (0.79-3.05)	1.48 (0.61-3.62)
	Rotational	10	1.08 (0.82-1.42)	1.09 (0.78-1.52)	1.30 (0.78-2.16)	1.85 (0.98-3.52)	1.93 (0.67-5.55)
	HITsp	10	1.26 (0.90-1.78)	1.26 (0.78-2.04)	1.35 (0.78-2.34)	2.08 (0.87-4.95)	1.65 (0.63-4.33)
Rate of Torque Development							
Flexion	Linear	10	0.90 (0.78-1.04)	0.87 (0.72-1.05)	0.88 (0.66-1.19)	1.00 (0.54-1.83)	0.78 (0.34-1.77)
	Rotational	10	0.99 (0.79-1.24)	0.96 (0.73-1.27)	1.14 (0.74-1.77)	1.56 (0.86-2.81)	1.20 (0.43-3.32)
	HITsp	10	1.02 (0.76-1.38)	0.94 (0.62-1.41)	1.01 (0.63-1.61)	1.02 (0.47-2.22)	0.89 (0.37-2.13)
Extension	Linear	5*	1.04 (0.87-1.24)	0.98 (0.81-1.18)	0.97 (0.70-1.34)	0.95 (0.49-1.85)	1.45 (0.54-3.86)
	Rotational	5*	1.13 (0.88-1.44)	1.28 (0.95-1.73)	1.10 (0.67-1.81)	0.98 (0.49-1.99)	1.21 (0.33-4.47)
	HITsp	5*	0.96 (0.73-1.28)	1.00 (0.66-1.50)	0.97 (0.61-1.53)	0.85 (0.36-2.01)	1.12 (0.37-3.33)
Right Lateral Flexion	Linear	10	0.97 (0.83-1.13)	1.06 (0.87-1.29)	1.09 (0.81-1.47)	1.42 (0.80-2.53)	0.96 (0.42-2.21)
	Rotational	10	1.08 (0.86-1.36)	1.11 (0.85-1.46)	1.26 (0.82-1.92)	1.25 (0.69-2.29)	0.92 (0.32-2.62)
	HITsp	10	1.22 (0.91-1.65)	1.27 (0.85-1.90)	1.37 (0.87-2.16)	1.66 (0.78-3.50)	0.91 (0.38-2.19)
Left Lateral Flexion	Linear	10	1.01 (0.86-1.19)	1.07 (0.87-1.30)	1.22 (0.91-1.65)	1.50 (0.85-2.66)	1.28 (0.57-2.87)
	Rotational	10	0.94 (0.75-1.18)	1.07 (0.81-1.41)	1.27 (0.82-1.95)	1.22 (0.67-2.26)	1.10 (0.40-3.04)
	HITsp	10	1.16 (0.85-1.58)	1.14 (0.75-1.74)	1.39 (0.87-2.22)	1.61 (0.74-3.48)	1.32 (0.55-3.14)
Composite	Linear	10	1.03 (0.87-1.21)	1.06 (0.87-1.30)	1.08 (0.80-1.45)	1.16 (0.63-2.12)	0.98 (0.43-2.26)
	Rotational	10	1.10 (0.87-1.38)	1.15 (0.87-1.51)	1.25 (0.82-1.91)	1.15 (0.62-2.14)	1.05 (0.37-2.95)
	HITsp	10	1.28 (0.95-1.72)	1.26 (0.85-1.88)	1.32 (0.84-2.08)	1.54 (0.72-3.32)	0.94 (0.39-2.27)

† Players classified as high performers had increased odds (OR greater than 1 indicate increased odds for the high group)

*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance groups.**

Table 4.8. Cervical muscle size (Group overall): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

			2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Cervical Muscle Size		**df**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Sternocleidomastoid	Linear	29*	0.92 (0.83-1.01)	1.08 (0.93-1.26)	1.15 (0.93-1.41)	1.17 (0.79-1.75)	0.98 (0.64-1.52)
	Rotational	29*	0.88 (0.76-1.02)	0.98 (0.83-1.17)	1.04 (0.81-1.33)	0.97 (0.66-1.42)	1.14 (0.62-2.10)
	HITsp	29*	1.21 (1.00-1.47)	***1.30 (1.02-1.66) †***	***1.53 (1.10-2.12) †***	1.40 (0.84-2.34)	1.51 (0.83-2.75)
Upper Trapezius	Linear	34	1.03 (0.94-1.14)	1.05 (0.91-1.22)	1.12 (0.93-1.35)	1.32 (0.92-1.88)	***1.51 (1.01-2.252) †***
	Rotational	34	0.98 (0.85-1.13)	1.00 (0.85-1.18)	0.99 (0.78-1.24)	1.04 (0.71-1.51)	0.73 (0.41-1.31)
	HITsp	34	1.15 (0.94-1.40)	1.06 (0.83-1.36)	1.21 (0.88-1.65)	1.33 (0.82-2.15)	1.57 (0.90-2.74)
Semispinalis Capitis	Linear	29*	1.00 (0.90-1.10)	1.15 (0.99-1.34)	***1.34 (1.10-1.64) †***	1.41 (0.94-2.11)	1.22 (0.78-1.90)
	Rotational	29*	0.94 (0.81-1.08)	1.18 (0.99-1.40)	1.26 (0.99-1.61)	1.37 (0.93-2.03)	1.42 (0.78-2.58)
	HITsp	29*	1.16 (0.95-1.42)	***1.32 (1.03-1.70) †***	***1.68 (1.22-2.32) †***	1.64 (0.98-2.75)	***1.88 (1.04-3.40) †***
Composite	Linear	29*	0.97 (0.87-1.09)	1.10 (0.93-1.29)	***1.26 (1.01-1.57) †***	***1.53 (1.01-2.33) †***	1.15 (0.71 -1.86)
	Rotational	29*	0.93 (0.79-1.09)	1.06 (0.88-1.28)	1.19 (0.91-1.54)	1.30 (0.87-1.94)	1.15 (0.60-2.19)
	HITsp	29*	***1.33 (1.08-1.63) †***	***1.32 (1.01-1.72) †***	***1.65 (1.17-2.32) †***	***1.87 (1.10-3.18) †***	1.53 (0.81-2.91)

† Players classified as high performers had increased odds (OR greater than 1 indicate increased odds for the high group)

*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance groups.**

Table 4.9. Cervical muscle size (Skill players only): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

			2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Cervical Muscle Size		**df**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Sternocleidomastoid	Linear	14*	0.95 (0.84-1.08)	1.24 (0.99-1.55)	***1.39 (1.07-1.82) †***	1.31 (0.78-2.22)	0.88 (0.55-1.41)
	Rotational	14*	0.90 (0.76-1.08)	1.09 (0.89-1.33)	1.24 (0.97-1.60)	1.08 (0.67-1.73)	1.01 (0.51-2.00)
	HITsp	14*	***1.44 (1.11-1.87) †***	***1.62 (1.20-2.18) †***	***2.13 (1.41-3.19) †***	1.79 (0.92-3.49)	1.53 (0.73-3.20)
Upper Trapezius	Linear	19	0.97 (0.86-1.11)	0.99 (0.79-1.24)	0.99 (0.75-1.31)	1.29 (0.77-2.16)	1.09 (0.67-1.79)
	Rotational	19	1.06 (0.88-1.27)	0.97 (0.79-1.18)	1.11 (0.86-1.43)	1.20 (0.76-1.89)	0.77 (0.38-1.59)
	HITsp	19	1.14 (0.85-1.52)	0.99 (0.70-1.39)	1.18 (0.74-1.88)	1.35 (0.68-2.68)	1.17 (0.53-2.61)
Semispinalis Capitis	Linear	14*	1.02 (0.90-1.15)	***1.25 (1.02-1.55) †***	***1.43 (1.11-1.85) †***	1.52 (0.89-2.59)	0.97 (0.60-1.56)
	Rotational	14*	0.98 (0.82-1.18)	1.21 (0.99-1.48)	1.25 (0.97-1.61)	1.33 (0.82-2.16)	0.98 (0.50-1.92)
	HITsp	14*	1.24 (0.94-1.63)	***1.41 (1.03-1.94) †***	***1.89 (1.23-2.91) †***	1.47 (0.74-2.92)	1.38 (0.65-2.90)
Composite	Linear	14*	0.96 (0.83-1.10)	1.15 (0.88-1.49)	1.32 (0.96-1.81)	***1.83 (1.02-3.28) †***	1.10 (0.62-1.94)
	Rotational	14*	0.90 (0.73-1.11)	0.97 (0.77-1.22)	1.15 (0.86-1.55)	1.44 (0.84-2.45)	0.95 (0.44-2.02)
	HITsp	14*	***1.52 (1.12-2.05) †***	1.42 (0.98-2.06)	***1.95 (1.17-3.25) †***	***2.38 (1.13-5.02) †***	1.60 (0.68-3.74)

† Players classified as high performers had increased odds (OR greater than 1 indicate increased odds for the high group)
*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance groups.**

Table 4.10. Cervical muscle size (Linemen only): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

			2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Cervical Muscle Size		**df**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Sternocleidomastoid	Linear	5*	0.88 (0.70-1.12)	0.93 (0.73-1.19)	0.88 (0.59-1.32)	1.20 (0.52-2.76)	1.10 (0.30-3.96)
	Rotational	5*	0.93 (0.68-1.28)	0.83 (0.55-1.24)	0.73 (0.39-1.36)	0.89 (0.37-2.17)	1.22 (0.23-6.42)
	HITsp	5*	0.86 (0.60-1.22)	0.83 (0.49-1.40)	0.79 (0.45-1.39)	0.87 (0.29-2.65)	1.10 (0.27-4.52)
Upper Trapezius	Linear	10	1.10 (0.95-1.28)	1.15 (0.95-1.39)	***1.35 (1.03-1.77) †***	1.36 (0.76-2.42)	***2.38 (1.13-5.01) †***
	Rotational	10	0.99 (0.79-1.24)	1.15 (0.88-1.49)	1.01 (0.66-1.57)	1.03 (0.55-1.91)	1.90 (0.73-4.97)
	HITsp	10	1.16 (0.86-1.55)	1.16 (0.78-1.73)	1.27 (0.82-1.98)	1.34 (0.62-2.90)	***2.33 (1.09-5.00) †***
Semispinalis Capitis	Linear	5*	1.01 (0.79-1.28)	1.09 (0.85-1.39)	1.49 (0.99-2.24)	1.88 (0.80-4.43)	2.18 (0.61-7.85)
	Rotational	5*	0.97 (0.70-1.36)	1.28 (0.83-1.98)	1.79 (0.90-3.54)	1.99 (0.77-5.14)	3.16 (0.63-15.98)
	HITsp	5*	1.05 (0.71-1.57)	1.22 (0.66-2.26)	1.71 (0.91-3.21)	2.54 (0.83-7.76)	3.08 (0.78-12.14)
Composite	Linear	5*	1.00 (0.81-1.24)	1.05 (0.85-1.31)	1.20 (0.84-1.71)	1.41 (0.67-2.98)	1.29 (0.42-3.99)
	Rotational	5*	0.99 (0.74-1.31)	1.22 (0.86-1.74)	1.25 (0.71-2.19)	1.19 (0.53-2.66)	1.80 (0.43-7.53)
	HITsp	5*	1.10 (0.79-1.53)	1.17 (0.72-1.90)	1.28 (0.76-2.15)	1.37 (0.51-3.65)	1.61 (0.48-5.36)

† Players classified as high performers had increased odds (OR greater than 1 indicate increased odds for the high group)

*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance groups.**

Table 4.11. Cervical perturbation (Group overall): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

			2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Stiffness		**df**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Anticipated Forced Extension	Linear	33	***0.88 (0.80-0.97) ‡***	***0.78 (0.67-0.90) ‡***	***0.77 (0.62-0.94) ‡***	0.76 (0.50-1.15)	0.84 (0.53-1.31)
	Rotational	33	0.89 (0.77-1.04)	***0.82 (0.69-0.97) ‡***	0.81 (0.63-1.04)	0.85 (0.57-1.26)	1.14 (0.60-2.15)
	HITsp	33	0.82 (0.66-1.03)	***0.70 (0.53-0.91) ‡***	***0.65 (0.46-0.93) ‡***	0.67 (0.39-1.16)	0.72 (0.39-1.34)
Anticipated Forced Flexion	Linear	33	1.01 (0.91-1.12)	0.98 (0.84-1.14)	1.03 (0.84-1.26)	1.23 (0.83-1.81)	***1.51 (1.01-2.26) †***
	Rotational	33	***0.84 (0.73-0.97) ‡***	***0.83 (0.71-0.98) ‡***	0.80 (0.63-1.00)	1.10 (0.76-1.60)	1.61 (0.91-2.85)
	HITsp	33	1.02 (0.82-1.27)	0.95 (0.72-1.24)	0.90 (0.64-1.27)	1.11 (0.66-1.87)	1.57 (0.88-2.80)
Composite	Linear	32	***0.88 (0.80-0.97) ‡***	***0.76 (0.66-0.88) ‡***	***0.76 (0.63-0.93) ‡***	0.78 (0.52-1.16)	1.09 (0.72-1.67)
	Rotational	32	***0.80 (0.70-0.92) ‡***	***0.76 (0.65-0.88) ‡***	***0.73 (0.59-0.92) ‡***	0.83 (0.57-1.22)	1.32 (0.72-2.41)
	HITsp	32	***0.80 (0.65-0.99) ‡***	***0.66 (0.52-0.85) ‡***	***0.61 (0.44-0.84) ‡***	0.71 (0.42-1.21)	0.96 (0.53-1.75)
Angular Displacement							
Anticipated Forced Extension	Linear	33	1.04 (0.94-1.15)	0.95 (0.82-1.12)	0.95 (0.77-1.17)	0.95 (0.64-1.42)	***1.54 (1.04-2.29) †***
	Rotational	33	0.95 (0.82-1.10)	0.92 (0.78-1.09)	0.90 (0.71-1.15)	0.87 (0.60-1.28)	1.68 (0.95-3.00)
	HITsp	33	0.91 (0.73-1.12)	0.90 (0.69-1.18)	0.78 (0.56-1.09)	0.94 (0.56-1.59)	1.15 (0.63-2.11)
Anticipated Forced Flexion	Linear	33	***0.86 (0.79-0.95) ‡***	***0.82 (0.71-0.94) ‡***	***0.81 (0.67-0.98) ‡***	0.90 (0.62-1.30)	0.96 (0.63-1.45)
	Rotational	33	***0.82 (0.73-0.94) ‡***	***0.76 (0.66-0.87) ‡***	***0.72 (0.58-0.88) ‡***	0.78 (0.55-1.11)	1.04 (0.59-1.84)
	HITsp	33	0.89 (0.73-1.09)	***0.77 (0.61-0.99) ‡***	***0.72 (0.53-0.98) ‡***	0.81 (0.49-1.32)	0.90 (0.51-1.59)
Composite	Linear	32	1.00 (0.91-1.10)	0.93 (0.80-1.07)	0.93 (0.76-1.13)	1.00 (0.69-1.46)	1.40 (0.95-2.06)
	Rotational	32	0.94 (0.82-1.08)	0.91 (0.78-1.07)	0.91 (0.72-1.13)	0.93 (0.65-1.34)	1.65 (0.96-2.84)
	HITsp	32	0.88 (0.72-1.07)	0.86 (0.67-1.11)	0.79 (0.58-1.09)	0.97 (0.59-1.60)	1.13 (0.64-2.00)
Muscle Onset Latency							
Unanticipated Forced Extension	Linear	32	0.99 (0.89-1.09)	0.98 (0.84-1.14)	1.02 (0.83-1.25)	1.01 (0.68-1.50)	1.01 (0.65-1.56)
	Rotational	32	0.98 (0.85-1.13)	1.00 (0.85-1.18)	0.99 (0.78-1.24)	1.04 (0.71-1.51)	0.73 (0.41-1.31)
	HITsp	32	0.98 (0.80-1.21)	1.05 (0.81-1.36)	1.18 (0.85-1.64)	0.98 (0.58-1.64)	0.99 (0.55-1.80)
Unanticipated Forced Flexion	Linear	32	1.08 (0.98-1.18)	1.10 (0.95-1.27)	1.08 (0.88-1.31)	1.04 (0.71-1.51)	1.17 (0.78-1.76)
	Rotational	32	1.03 (0.90-1.18)	1.05 (0.89-1.22)	1.01 (0.81-1.26)	0.88 (0.62-1.26)	1.31 (0.74-2.33)
	HITsp	32	0.91 (0.75-1.11)	1.00 (0.78-1.29)	1.02 (0.74-1.41)	0.92 (0.56-1.50)	0.89 (0.50-1.58)
Composite	Linear	32	1.04 (0.95-1.15)	1.04 (0.90-1.21)	1.06 (0.87-1.30)	1.15 (0.79-1.68)	1.23 (0.81-1.86)
	Rotational	32	0.95 (0.83-1.09)	1.03 (0.87-1.20)	1.05 (0.84-1.32)	1.11 (0.77-1.60)	1.39 (0.77-2.51)
	HITsp	32	0.97 (0.79-1.19)	1.01 (0.78-1.30)	1.17 (0.84-1.61)	1.24 (0.75-2.07)	1.24 (0.69-2.21)

† Players classified as high performers had increased odds (OR greater than 1 indicate increased odds for the high group)

‡ Players classified as high performers had reduced odds (OR less than 1 indicate reduced odds for the high group)

*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance groups.**

Table 4.12. Cervical perturbation (Skill players only): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

			2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Stiffness		**df**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Anticipated Forced Extension	Linear	19	0.94 (0.82-1.07)	0.76 (0.62-0.95)	0.71 (0.55-0.93)	0.62 (0.37-1.03)	0.90 (0.55-1.49)
	Rotational	19	0.85 (0.70-1.03)	***0.78 (0.64-0.94) ‡***	***0.67 (0.53-0.84) ‡***	***0.57 (0.37-0.87) ‡***	1.09 (0.52-2.30)
	HITsp	19	0.75 (0.56-1.01)	***0.65 (0.47-0.90) ‡***	***0.54 (0.35-0.84) ‡***	***0.47 (0.24-0.90) ‡***	0.59 (0.26-1.32)
Anticipated Forced Flexion	Linear	19	1.07 (0.95-1.22)	0.94 (0.74-1.19)	0.94 (0.70-1.25)	1.00 (0.58-1.71)	1.39 (0.85-2.28)
	Rotational	19	0.90 (0.75-1.09)	0.90 (0.74-1.11)	0.82 (0.63-1.06)	1.08 (0.67-1.73)	1.79 (0.88-3.62)
	HITsp	19	0.99 (0.73-1.35)	0.97 (0.68-1.39)	0.80 (0.49-1.31)	0.93 (0.45-1.90)	1.29 (0.56-2.97)
Composite	Linear	19	0.94 (0.83-1.07)	0.76 (0.61-0.94)	0.70 (0.54-0.90)	0.63 (0.38-1.06)	0.89 (0.54-1.46)
	Rotational	19	0.84 (0.70-1.01)	***0.77 (0.64-0.93) ‡***	***0.67 (0.53-0.84) ‡***	***0.61 (0.40-0.94) ‡***	1.09 (0.53-2.25)
	HITsp	19	0.79 (0.59-1.06)	***0.66 (0.48-0.92) ‡***	***0.53 (0.35-0.81) ‡***	0.52 (0.27-1.01)	0.62 (0.28-1.38)
Angular Displacement							
Anticipated Forced Extension	Linear	19	1.05 (0.92-1.19)	0.92 (0.73-1.15)	0.86 (0.65-1.15)	0.67 (0.40-1.13)	1.12 (0.68-1.85)
	Rotational	19	1.02 (0.84-1.24)	0.94 (0.76-1.17)	0.93 (0.71-1.21)	0.71 (0.44-1.15)	1.51 (0.73-3.13)
	HITsp	19	0.76 (0.57-1.02)	0.77 (0.54-1.08)	0.64 (0.40-1.03)	0.64 (0.32-1.28)	0.79 (0.35-1.79)
Anticipated Forced Flexion	Linear	19	0.91 (0.80-1.03)	***0.77 (0.62-0.96) ‡***	0.77 (0.59-1.01)	0.69 (0.41-1.18)	0.91 (0.55-1.50)
	Rotational	19	0.90 (0.75-1.09)	0.83 (0.68-1.00)	***0.77 (0.60-0.98) ‡***	0.76 (0.48-1.21)	1.28 (0.64-2.56)
	HITsp	19	0.88 (0.66-1.18)	0.79 (0.57-1.11)	0.65 (0.41-1.02)	0.65 (0.32-1.31)	0.79 (0.35-1.80)
Composite	Linear	19	0.98 (0.86-1.12)	0.86 (0.69-1.08)	0.82 (0.62-1.08)	0.79 (0.47-1.33)	1.15 (0.72-1.86)
	Rotational	19	1.00 (0.83-1.21)	0.90 (0.73-1.10)	0.85 (0.66-1.11)	0.79 (0.50-1.26)	1.63 (0.82-3.23)
	HITsp	19	***0.75 (0.57-0.99) ‡***	***0.69 (0.50-0.95) ‡***	***0.61 (0.39-0.96) ‡***	0.61 (0.31-1.19)	0.84 (0.38-1.86)
Muscle Onset Latency							
Unanticipated Forced Extension	Linear	19	1.02 (0.90-1.16)	1.06 (0.84-1.33)	1.08 (0.81-1.42)	1.06 (0.62-1.80)	0.99 (0.61-1.61)
	Rotational	19	1.14 (0.95-1.37)	1.17 (0.97-1.42)	1.17 (0.91-1.50)	1.28 (0.81-2.03)	0.90 (0.44-1.85)
	HITsp	19	1.06 (0.79-1.43)	1.23 (0.87-1.72)	1.42 (0.90-2.24)	1.03 (0.51-2.06)	0.92 (0.41-2.08)
Unanticipated Forced Flexion	Linear	19	1.00 (0.88-1.14)	1.05 (0.84-1.32)	1.05 (0.79-1.39)	1.11 (0.66-1.86)	1.13 (0.68-1.85)
	Rotational	19	1.01 (0.84-1.22)	1.01 (0.83-1.24)	1.00 (0.77-1.29)	1.05 (0.66-1.66)	1.64 (0.77-3.51)
	HITsp	19	0.78 (0.59-1.04)	0.82 (0.59-1.15)	0.90 (0.56-1.43)	0.76 (0.39-1.50)	0.89 (0.40-1.98)
Composite	Linear	19	1.12 (0.99-1.27)	1.21 (0.97-1.51)	1.25 (0.95-1.64)	1.42 (0.85-2.37)	1.39 (0.86-2.27)
	Rotational	19	1.01 (0.84-1.22)	1.12 (0.92-1.37)	1.16 (0.90-1.49)	1.36 (0.86-2.14)	1.84 (0.88-3.86)
	HITsp	19	1.03 (0.76-1.38)	1.10 (0.78-1.55)	1.40 (0.88-2.22)	1.34 (0.68-2.66)	1.43 (0.64-3.16)

‡ Players classified as high performers had reduced odds (OR less than 1 indicate reduced odds for the high group)

*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance groups.**

Table 4.13. Cervical perturbation (Linemen only): Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

			2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
Stiffness		**df**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Anticipated Forced Extension	Linear	9	***0.81 (0.70-0.93) ‡***	0.80 (0.63-1.00)	0.91 (0.61-1.35)	1.05 (0.47-2.35)	0.72 (0.25-2.04)
	Rotational	9	0.98 (0.73-1.31)	0.90 (0.63-1.28)	1.27 (0.71-2.28)	1.95 (0.96-3.99)	1.08 (0.29-4.04)
	HITsp	9	0.98 (0.68-1.41)	0.80 (0.46-1.37)	1.01 (0.54-1.90)	1.35 (0.47-3.89)	0.96 (0.31-2.99)
Anticipated Forced Flexion	Linear	9	0.93 (0.78-1.12)	1.05 (0.84-1.31)	1.17 (0.84-1.63)	1.55 (0.82-2.93)	1.75 (0.77-3.97)
	Rotational	9	***0.77 (0.61-0.98) ‡***	***0.73 (0.55-0.99) ‡***	0.77 (0.48-1.25)	1.16 (0.58-2.34)	1.36 (0.47-3.99)
	HITsp	9	1.07 (0.76-1.50)	0.91 (0.57-1.45)	1.10 (0.65-1.86)	1.46 (0.61-3.49)	1.99 (0.84-4.69)
Composite	Linear	8	***0.80 (0.70-0.93) ‡***	***0.77 (0.62-0.95) ‡***	0.90 (0.63-1.30)	1.01 (0.48-2.11)	1.60 (0.67-3.79)
	Rotational	8	***0.74 (0.60-0.92) ‡***	***0.74 (0.56-0.97) ‡***	0.92 (0.56-1.54)	1.59 (0.79-3.23)	1.67 (0.51-5.55)
	HITsp	8	0.81 (0.59-1.11)	0.65 (0.41-1.04)	0.81 (0.46-1.41)	1.26 (0.47-3.36)	1.93 (0.72-5.16)
Angular Displacement							
Anticipated Forced Extension	Linear	9	1.04 (0.86-1.26)	1.02 (0.80-1.29)	1.07 (0.76-1.52)	1.38 (0.70-2.71)	***2.54 (1.33-4.85) †***
	Rotational	9	0.87 (0.67-1.12)	0.91 (0.67-1.24)	0.85 (0.52-1.42)	1.27 (0.64-2.52)	1.93 (0.66-5.70)
	HITsp	9	1.20 (0.88-1.65)	1.17 (0.73-1.85)	1.09 (0.65-1.85)	1.66 (0.70-3.94)	2.18 (0.94-5.10)
Anticipated Forced Flexion	Linear	9	***0.83 (0.72-0.95) ‡***	0.87 (0.72-1.06)	0.88 (0.65-1.18)	1.08 (0.59-1.99)	0.98 (0.42-2.26)
	Rotational	9	***0.74 (0.61-0.90) ‡***	***0.68 (0.54-0.86) ‡***	0.67 (0.45-1.00)	0.84 (0.45-1.58)	0.73 (0.25-2.12)
	HITsp	9	0.89 (0.67-1.20)	0.75 (0.50-1.10)	0.82 (0.52-1.30)	1.07 (0.48-2.39)	1.03 (0.42-2.50)
Composite	Linear	8	1.02 (0.87-1.21)	1.02 (0.82-1.26)	1.11 (0.80-1.52)	1.42 (0.76-2.64)	1.89 (0.88-4.09)
	Rotational	8	0.87 (0.69-1.10)	0.94 (0.70-1.26)	1.01 (0.63-1.60)	1.22 (0.64-2.34)	1.61 (0.56-4.64)
	HITsp	8	1.11 (0.83-1.48)	1.17 (0.75-1.83)	1.16 (0.71-1.89)	1.83 (0.81-4.12)	1.76 (0.73-4.23)
Muscle Onset Latency							
Unanticipated Forced Extension	Linear	8	0.94 (0.78-1.13)	0.86 (0.69-1.08)	0.94 (0.67-1.33)	0.97 (0.48-1.97)	1.10 (0.41-2.95)
	Rotational	8	0.79 (0.61-1.01)	0.78 (0.58-1.05)	0.77 (0.48-1.23)	0.72 (0.36-1.43)	0.51 (0.16-1.60)
	HITsp	8	0.88 (0.64-1.19)	0.81 (0.51-1.28)	0.87 (0.52-1.46)	0.90 (0.36-2.25)	1.03 (0.36-2.98)
Unanticipated Forced Flexion	Linear	8	***1.17 (1.01-1.36) †***	1.17 (0.96-1.43)	1.12 (0.82-1.53)	1.05 (0.55-1.99)	1.29 (0.54-3.09)
	Rotational	8	1.05 (0.83-1.34)	1.09 (0.82-1.44)	1.03 (0.66-1.62)	0.67 (0.36-1.27)	0.93 (0.31-2.72)
	HITsp	8	1.13 (0.85-1.50)	1.34 (0.89-2.03)	1.21 (0.75-1.95)	1.19 (0.51-2.78)	1.04 (0.41-2.67)
Composite	Linear	8	0.97 (0.81-1.15)	0.87 (0.71-1.08)	0.85 (0.62-1.17)	0.95 (0.49-1.85)	1.04 (0.42-2.59)
	Rotational	8	0.88 (0.69-1.12)	0.91 (0.68-1.23)	0.92 (0.58-1.46)	0.83 (0.42-1.63)	0.76 (0.24-2.41)
	HITsp	8	0.90 (0.67-1.21)	0.88 (0.56-1.38)	0.89 (0.54-1.45)	1.11 (0.46-2.68)	0.94 (0.36-2.47)

† Players classified as high performers had increased odds (OR greater than 1 indicate increased odds for the high group)

‡ Players classified as high performers had reduced odds (OR less than 1 indicate reduced odds for the high group)

*** Playing level (high school, collegiate) was entered into the model to account for uneven distribution between high and low performance group**

Discussion

The most important finding of this study was that football players that exhibited greater cervical stiffness and less angular displacement following perturbation had reduced odds of sustaining higher magnitude head impacts. The results of this study do not support our hypothesis that players with stronger and larger cervical musculature are better able to mitigate head impact severity.

Cervical Isometric Strength & Cervical Muscle Size

Contrary to common opinion, our study shows that players, especially linemen, with stronger cervical musculature are actually at increased odds of sustaining higher magnitude head impacts. Although our results regarding cervical muscle strength are only evident within the right and left lateral flexor muscle group for our sample as a whole, the odds of sustaining higher magnitude impacts was much more pronounced among linemen. Athletes with strong cervical musculature may be more inclined to use their head, rather than their shoulders, when making contact with other players because they perceive the head and neck to be the most protected part of their body. The risk compensation phenomenon theorizes that players have a desired level of risk that they are willing to accept while playing football (Hagel & Meeuwisse, 2004; Hedlund, 2000). Football players with stronger cervical musculature may perceive that their head and neck are more protected and that they have a lower risk of injury. Risk compensation theorists propose that the four following factors influence risk compensation behavior: (1) visibility (e.g. how obvious is the change produced by the safety measure?); (2) effect (e.g. how does the change effect the player physically or emotionally?); (3) motivation (e.g. what is the player's motivation level during the task?); and (4) control (e.g. can the player change their actions even if they want to?) (Hedlund, 2000). Implementation of cervical strengthening programs among football players as a potential safety measure would be highly visible, would have significant physical and possibly emotional effects, players would have a high level of motivation while playing football, and players would have control of their actions within the rules of football. Using this theoretical framework, we conclude that risk compensation is very likely among football players with stronger cervical musculature.

The results of this study imply that we should consider the potential consequences of risk compensation behaviors among players when considering implementation of cervical strengthening programs. The notion of risk compensation is well noted in sport injury literature, but mostly regards the use of helmets in contact sports such as football and ice hockey (Hagel & Meeuwisse, 2004). The introduction of hard-shelled helmets in the late 1940's was followed by a marked increase in the number of tackling drill fatalities, likely a result of athletes using the head as the initial point of contact rather than the shoulder (Mueller, 1998). Despite observing negative implications of having stronger cervical musculature, we acknowledge that there are many other unstudied factors that may interact with cervical muscle strength, such that recommending that athletes strive to have weak musculature is extremely premature, especially in light of the increased risk of neck injury (Cheng, et al., 2008),.

The odds of sustaining higher magnitude head impacts was even more pronounced among linemen that possessed greater cervical strength, whereas skill players with higher and lower strength were at equal odds. Our results indicate that cervical strength training may not be detrimental among skill players, but could have a risk inducing effect among linemen. The Newtonian theory of reduced head acceleration as a result of increased effective mass following contraction of the cervical musculature may not apply when players serve as the striking player (Viano, et al., 2007). By virtue of their position, linemen should expect collisions to occur during every single play. Typically, once the ball is snapped a single offensive lineman collides with a single defensive lineman after taking approximately one to two steps. The combination of fully anticipating collision, serving as the striking player, and possessing stronger cervical musculature may increase a lineman's odds of sustaining high magnitude impacts. Further research is needed to determine the preventative nature of the cervical musculature under conditions where the player is being struck versus conditions where the player is striking. Differentiating cervical strength and conditioning programs for skill and line players separately may be possible at the collegiate level, but is not likely possible at the high school level. Generally, collegiate athletes have oversight from highly trained certified strength and conditioning experts, but this is not always the case at the high school level.

Generally, we did not observe many significant results regarding cervical rate of torque development. During real play an athlete may not reach maximal muscle force before head impact. It is thought that rate of torque development may provide a better estimate of the cervical musculature's short-term damping capacity(Almosnino, et al., 2009; Almosnino, et al., 2010). We observed that players had reduced odds of sustaining head impacts in the 2nd quartile compared to the 1st if they presented with higher rate of torque development among the flexor muscle group, however, this result may be spurious given that we did not observe this effect in any other muscle groups or any other percentile grouping.

Similar results between muscle strength and cross-sectional area are not surprising, since strength increases linearly with increases in physiological cross-sectional area among the cervical musculature (Mayoux-Benhamou, et al., 1989; Rezasoltani, et al., 2002). For every squared centimeter increase in cross-sectional area, the force output of the cervical musculature increases by approximately 10N (Mayoux-Benhamou, et al., 1989).

Cervical Perturbation

Our results indicate that players with greater cervical stiffness and less angular displacement following perturbation had reduced odds of sustaining higher magnitude head impacts, particularly skill players. Players with increased stiffness are better able to engage their cervical musculature following head perturbation, and therefore acutely resist head displacement. Cervical stiffness may play a larger role than cervical strength in mitigating head impact severity because football players rarely use maximum strength when decelerating their head after impact. Stiffness of the cervical region is proportional to both muscle activity and force generated through muscular contraction (Granata, Wilson, & Padua, 2002; Morgan, 1977). Muscle stiffness is increased acutely via myoelectric activity. As a football player prepares for an impending collision, he/she increases muscle activity to generate a counter force to the load they expect to be applied to their head. Resistance to deformation following head impact may be dependent on a player’s ability to quickly reach a high level of muscle activity.

Long lasting improvements in stiffness may be obtainable through neuromuscular training of the cervical musculature (Hurd, Chmielewski, & Snyder-Mackler, 2006; Kubo, et al., 2007). Therefore, reduction in head impact magnitudes while playing football may be possible through cervical stiffness enhancement. Very few studies have examined exercises aimed at increasing cervical muscle stiffness. Mansell *et al.* (Mansell, et al., 2005) found that an eight-week traditional cervical resistance training program changed muscle structure and increased strength, but failed to improve neuromuscular plasticity. The authors concluded that neuromuscular exercises, such as plyometrics, might be needed to evoke changes in cervical dynamic stabilization. Neuromuscular training has been shown to improve cervical muscle activation in patients experiencing neck pain, but these studies use very simple exercises among a pathological population that might not be appropriate for healthy athletes (Falla, O'Leary, Farina, & Jull, 2012; Uemura, Tanaka, & Kawazoe, 2008). More research is necessary to determine whether neuromuscular training programs have the potential to reduce the odds of sustaining higher magnitude head impacts.

In addition to possessing greater cervical strength, males also exhibit greater stiffness and capacity to store elastic energy compared to females (McGill, et al., 1994). Collegiate football players included in this study had 40% greater stiffness compared to the high school cohort. We measured the odds of sustaining head impacts of higher magnitudes, but it seems possible that reduced stiffness may explain why female and adolescent athletes have a reduced resistance to concussion (Castile, et al., 2011; Gessel, et al., 2007; Marar, et al., 2012). Further research is needed regarding the potential link between reduced cervical stiffness and concussion risk.

We observed that linemen with quicker muscle onset latencies had increased odds of sustaining head impacts in the 2nd quartile, rather than the 1st. Although we observed that linemen in the high performance group had a 17% increase in odds of sustaining head impacts in the 2nd quartile compared to the 1st, these results are likely clinically inconsequential as the 2nd quartile contains head impacts ranging in linear magnitudes of 15.3g to 21.0g. The upper cutoff of this head impact category approaches the mean linear magnitude previously reported for both college and high school football players (Broglio, et al., 2009; Mihalik, et al., 2007), which suggests that although these head impacts are more serious than 1st quartile head impacts, they remain mild (Zhang, et al., 2004). This single result contradicts a majority of our other findings and the upper limit of the confidence interval approaches equal risk between groups.

Modeling the head–neck segment as a rigid segment during perturbations with a center of rotation about C7-T1 is a known limitation of this study (Portero, Quaine, Cahouet, Thoumie, & Portero, 2013). We recognize that head motion resulting from application of an external force is more complex than a simple rotation around a fixed center of rotation; however, our calculation of stiffness provides an estimate of each player's ability to resist cervical perturbation. We used a protocol that involved perturbation in the sagittal plane only. More research is needed to determine the role of cervical muscle stiffness and reduced head angular displacement following perturbations in the frontal and transverse planes.

We examined isometric cervical strength. Future research should seek to determine whether cervical strength during dynamic tasks plays a role in mitigating head impact severity. Although we examined muscle size of the highest moment generating cervical muscles, we recommend that future studies consider the possible role of smaller cervical muscles that serve to stabilize the spine (Vasavada, et al., 1998). We investigated the odds of sustaining higher magnitude head impacts, and while it is generally accepted that players who sustain high magnitude impacts are at an increased risk of sustaining concussive injuries, we did not study the risk of sustaining concussions (Guskiewicz & Mihalik, 2011). Future research is necessary to determine the risk of concussion among players with strong and weak cervical musculature.

Conclusions

Few studies have investigated the influence of the cervical musculature on head impact biomechanical measures. This study suggests that cervical strength and muscle size increases an athlete’s odds, particularly among linemen, while cervical stiffness and angular displacement following perturbation reduces an athlete’s odds of sustaining higher magnitude impacts. Because this is the first study of its kind, we do not recommend that cervical strengthening programs be prohibited at this time, but we urge sports medicine professionals and strength and conditioning experts to consider the possible deleterious effects of implementing these safety measures. Neuromuscular training may be a more suitable and effective approach to reducing the odds of sustaining high magnitude head impacts among football athletes. More research is needed to fully understand how certain cervical characteristics influence an athlete’s odds of sustaining higher magnitude head impacts.

Chapter 5

MANUSCRIPT II

Does Visual Performance Influence Head Impact Severity Among High School Football Athletes?

Introduction

Context: Athletes with diminished visual performance may be less likely to see an oncoming collision, leaving them unable to protect themselves and possibly more prone to injury. Further research is needed to determine if head impact biomechanical measures are influenced by visual performance. **Objective**: To compare the odds of sustaining higher magnitude head impacts between high school athletes with high and low visual performance. **Design:** Prospective quasi-experimental. **Setting**: Clinical-Research Center/On-field. **Patients or Other Participants**: Thirty-seven high school varsity football players. **Interventions**: Athletes completed the Nike SPARQ Sensory Station visual assessment prior to the season. Head impact biomechanics were captured for each player using the Head Impact Telemetry System. **Main Outcome Measures**: Each player was classified as either a high or low performer using a median split for each of the following visual performance measures: visual clarity, contrast sensitivity, depth perception, near-far quickness, target capture, perception span, eye-hand coordination, go/no go, and reaction time. We computed the odds of sustaining head impacts in 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile, rather than head impacts in the 1st quartile between players that were high performers relative to those that were low performers for each of the visual performance measures. **Results**: Players that were better able to switch between near and far stimuli (near-far quickness), were better able to quickly identify a target in the periphery (target capture), and players with quicker reaction times (reaction time) had increased odds of sustaining higher magnitude head impacts. High and low performers were at equal odds on all other measures. **Conclusions**: Our study suggests that visual performance plays less of a role than expected for protecting against higher magnitude head impacts among high school football players. Future research is needed to determine if visual performance influences concussion risk.

Sports are second only to motor vehicle crashes as the leading cause of traumatic brain injury among young people ages 15-24 years (Sosin, et al., 1996). Very little research is available addressing the modifiable factors that could help mitigate the severity of head impacts that result during contact sports, leaving sports medicine professionals with limited options for preventing concussion (McCrory, et al., 2013). Many sports involve quick and unpredictable movements of objects, competitors, teammates, and the athlete themselves. These movements often occur simultaneously, and place athletes at risk for injuries such as concussion. It seems possible that contact sport athletes with diminished visual performance may be less likely to see an oncoming collision, leaving them unable to anticipate and prepare, and more prone to injury.

Numerous studies have identified that athletes demonstrate better visual abilities than non-athletes, and that elite athletes have visual abilities that are superior to novice and less successful athletes (Hitzeman & Beckerman, 1993; Stine, et al., 1982; Uchida, et al., 2012).

Although the importance of visual performance in sport is widely accepted, detailed assessments are not often completed in the athletic setting. Most sports medicine clinicians ensure that athletes are able to see at 20/20 to determine if visual correction is necessary. Visual clarity, as measured with the Snellen chart alone, does not represent all of the visual components and demands placed on athletes. Athletes may require visual function that far exceeds minimal standards to adequately respond to the visual demands placed on them during sport (Laby, et al., 1996). In addition to requiring good visual clarity, most sports require athletes to scan and interpret visual information at differing contrast levels, utilize visual information at varying depths, switch between stimuli that are at near and far distances, identify stimuli in their peripheral vision, memorize and recognize patterns of movement, execute proper eye-hand and eye-foot coordination, and respond quickly while also being able to execute response inhibition.

The link between visual performance and injury prevention has not been established. Rugby ball carriers that are struck from outside of their peripheral visual field are at higher risk of general injury because they lack visual information about the impending collision (King, Hume, & Clark, 2012), suggesting that a total lack of visual information increases injury risk. When vision is eliminated completely in resistance trained individuals, lower extremity muscle power declines (Killebrew, Petrella, Jung, & Hensarling, 2013). Likewise, air assault soldiers demonstrate more dangerous landing biomechanics when their vision is occluded (Chu, et al., 2012). Although not directly related to head trauma, these studies suggest that vision is vital for athletic and sport performance. Possessing superior visual capabilities may allow athletes to gather more visual information in a shorter period of time. The athlete can then use the visual information to either avoid collision all together or to reduce collision severity by evoking head protection strategies, such as leaning, using the arms to block the face, and recoiling their head by elevating their shoulders (Metoyer, et al., 2008). Although visual performance has been linked with athletic skill, how better visual performance relates to concussion prevention is not yet known (Zimmerman, et al., 2011). Further research is needed to determine if head impact biomechanics are influenced by visual performance. The purpose of this study was to determine if high school football athletes with higher visual performance are at a reduced risk of sustaining higher magnitude head impacts, relative to athletes with lower visual performance.

Methods

Study Participants

Thirty-seven varsity football players from a single local high school participated in this study. Athletes and legal guardians signed informed consent forms approved by the Institutional Review Board. Demographic information is presented in Table 5.1.

Table 5.1. Demographic Information

	Mean	SD
Age (yrs)	16.59	0.89
Height (cm)	180.35	6.39
Mass (kg)	87.18	19.03
Year (Athletic)		
Freshmen	0 (0%)	
Sophomore	10 (27%)	
Junior	11 (30%)	
Senior	16 (43%)	
Position Group		
Skill	22	
Offense	8 (36%)	
Defense	14 (64%)	
Line	15	
Offense	10 (67%)	
Defense	5 (33%)	

Measurements & Instrumentation

Visual Performance Assessment

Visual performance was evaluated using the Nike SPARQ Sensory Station (Nike, Inc., Beaverton, Oregon). The visual performance assessment included the following subtests: visual clarity, contrast sensitivity, depth perception, near-far quickness, target capture, perception span, eye-hand coordination, go/no go, and reaction time. The Nike SPARQ Sensory Station consists of two high-resolution liquid crystal display monitors (a single 22-inch display and a single 42-inch touch-sensitive display) controlled by a single computer. A wirelessly connected Apple iPod touch (Apple Corporation, Cupertino, California) is used in several assessments (described in Table 5.2). Custom software controls the displays, input acquisition, and test procedures based on athlete responses. The validity of the Nike SPARQ Sensory Station has not yet been determined.

Head Impact Biomechanics

Head impact biomechanics were measured at each practice and game over the course of a single season using the Head Impact Telemetry (HIT) System technology (Riddell Corp., Elyria, OH). The HIT System consists of MxEncoder units located in the football helmets, a signal transducer, and a laptop computer that houses the Sideline Response System (Riddell Corp., Elyria, OH). MxEncoder units embedded within the helmets are comprised of six spring-loaded single-axis accelerometers, a telemetry unit, a data storage device, and a battery power source. Each single-axis accelerometer collects data at 1 kHz for a period of 40 ms (8 ms prior to the data collection trigger and 32 ms after the trigger). Data are time-stamped, encoded, and then transmitted in real-time to the signal transducer via radiofrequency transmission at 903–927 MHz. A signal transducer connects through a USB port to a laptop computer, which stores all head impact data. The HIT System transmits accelerometer data from distances well in excess of the length of the standard American football field.

Procedures

Visual Performance Assessment

All participants completed the visual performance assessment prior to the season of play. The assessment was completed in quiet area sequestered by curtains in a clinical research center. Window shades and room lighting were adjusted to provide optimal lighting during examination. Prerecorded instructions played at the start of each subtest followed by a brief practice segment. Participants were instructed to take their best guess if they could not easily discriminate visual stimuli. Participants were instructed to wear the corrective lenses that they typically wore while attending school or while playing football. Several of the SPARQ subtests incorporate the use of black Landolt rings (C), with gaps at the top, bottom, left, and right, on a white background at preset acuity demands, where athletes were to swipe in the direction of the gap in the ring (Example: Ɔ - swipe finger from right to left on the iPod touch). A majority of the sensory station assessments present with good inter-session reliability, however, near-far quickness, eye-hand coordination, and go/no go are influenced by practice effects (Erickson, et al., 2011). Table 5.2 includes a protocol description and information regarding outcome measure computation for each subtest.

Head Impact Biomechanics

Participants wore a Head Impact Telemetry (HIT) System MxEncoder, embedded in their helmet to measure head impact biomechanics over the course of the preseason, regular season, and postseason. MxEncoders were retrofit into Riddell Revolution and Speed helmet designs (Riddell Inc., Elyria, Ohio). Measures of head acceleration were calculated and stored within the Sideline Response System, yielding measures of linear acceleration, rotational acceleration, Gadd Severity Index, Head Impact Technology severity profile (HITsp), and Head Injury Criterion. Proper helmet fit was ensured prior to the season using the manufacturers fitting instructions and adjustments were made as needed throughout the season.

Table 5.2. Nike SPARQ Sensory Station subtest protocol and outcome measure description.

Subtest	Protocol Description	Outcome Measure Computation
Visual Clarity **†**	A Landolt ring of 20/50 equivalent appears on the screen -Athlete instructed to swipe in the direction of the gap If correct, the ring decreases in size - this continues until the athlete does not correctly identify stimulus After an incorrect response, the stimulus increases in size until the gap direction is identified correctly Assessment continues until several reversal points are complete Performed with vision occluded in left eye, then right eye, and then binocularly	Identifies the threshold acuity between 20/8 and 20/99 using a staircase reversal algorithm - LogMar values for oculus Uterque (both eyes) used
Contrast Sensitivity **†**	4 black circles present in a diamond configuration on a light gray background 1 circle contains a pattern of concentric rings that vary in brightness from the center to the edge Athlete swipes in the direction of the circle with the contrasted pattern Assessed at 6 and then 18 cycles per degree (cpd)	Identifies the cpd threshold - Contrast sensitivity at 18 cpd was used
Depth Perception **†**	4 black circles present in a diamond configuration on a light gray background Athletes wear a pair of liquid crystal goggles that cause 1 of 4 rings to appear to float 3-dimensionally Instructed to swipe in the direction of the floating ring	Identifies the arc second threshold between 237 and 12 using a staircase reversal algorithm
Near-Far Quickness **†**	Athlete holds the iPod Touch 16-inches from the eyes, with the top edge just below the bottom of the display A black Landolt ring of 20/80-equivalent presents on the far screen -Athlete swipes in the direction of the gap If correct, a Landolt ring appears on the iPod - Continually switch focus between far and near for 30 seconds	Sum the # of correct responses within the 30 second trial
Target Capture **‡**	Athlete fixates on a central black dot A Landolt ring of 0.1 log unit > than their visual clarity threshold appears briefly in one corners of the screen Athlete swipes in the perceived direction of the gap in the ring	Identifies the millisecond threshold between 0 at 500 using a staircase reversal algorithm
Perception Scan **Ω∑**	Focused on a black dot in the center of a grid pattern composed of blank circles A pattern of turquoise dots flashes within the grid of circles simultaneously for 100 milliseconds Instructed to touch the screen to recreate the pattern - If 75% correct, the grid increases in size and # of dots The first 2 levels consist of 6 blank circles in the grid pattern with 2 and 3 dots, the next 5 levels consist of 18 blank circles with 3 to 7 dots, and the last 4 levels consist of 30 blank circles with 7 to 10 dots If not 75% correct, the level is be repeated - If failed twice, the assessment is terminated	Calculated by summing the number of correct responses minus the number of missed responses and extra guesses
Eye-Hand Coordination **Ω∑**	Athletes hold arms parallel to the ground at shoulder height viewing an 8x6 grid of equally spaced blank circles A turquoise dot appears within one blank circle of the grid Athlete is instructed to touch the dot as quickly as possible using either hand As soon as they touched the dot, another turquoise dot appears. 96 dots total	Calculated as the total time to touch all 96 dots
Go No Go **Ω∑**	Same as the eye hand coordination, except that the dot stimulus was either turquoise or red Turquoise dot: touch dot at quickly as possible / Red dot: do not to touch 96 total dots (64 turquoise, 32 red) Each dot is presented for 450 milliseconds, with no time gap between dot presentations	Calculated by summing the # of turquoise dots touched minus any red dots touched
Reaction Time **Ω**	Two annular patterns appear - place fingertips of dominant hand on the annulus on that side of the screen Center body in front of the opposite annulus and focus attention on the center of that annulus After a randomized delay of 2, 3, or 4 seconds, the test annulus turns turquoise Move hand to touch the annulus as quickly as possible	Calculated as the elapsed time between onset of the test annulus and release of the control annulus

† Athlete stands 16 feet away from the 22-inch display and respond to stimuli using iPod touch
‡ Athletes stood 16 feet away from the 42-inch display and respond to stimuli using iPod touch
Ω Athletes were positioned within arm's length of the 42-inch touch-sensitive display, with the center of the screen adjusted to their height using a ruler mounted on the right side of the Sensory Station.
∑ Dots were pseudorandomized to maintain equivalent spatial distribution within each presentation and to eliminate ''clustering'' of dots and easily recognizable patterns.

Data Reduction

Data were stored locally on the Nike SPARQ Sensory Station System computer hard drive and later transmitted to a data cloud for storage and export. We split participants into a group of high and a group of low performers for each visual performance measure using a median split. For contrast sensitivity, depth perception, near-far quickness, perception span, and go no go, high performance was indicated by performance above the median. For visual clarity, target capture, eye-hand coordination, and reaction time, high performance was indicated by performance below the median. To represent visual clarity, we chose to use the LogMar values for oculus Uterque (vision with both eyes) because football athletes are not often required to play with monocular vision (oculus dexter, oculus sinister). We also chose to use contrast sensitivity thresholds captured during trials with a contrast of 18 cycles per degree because data analysis revealed very little heterogeneity at the six cycles per degree level. One participant's reaction time scores were lost because they were not adequately stored on the cloud server prior to data export.

Head impact data were exported from the Sideline Response System into Matlab 7. Consistent with previous studies, we then reduced the data to include only those impacts that register a linear acceleration greater than or equal to 10g (Guskiewicz, Mihalik, et al., 2007; Mihalik, et al., 2007; Mihalik, Blackburn, et al., 2010; Mihalik, Greenwald, et al., 2010; Mihalik, et al., 2011; Schnebel, et al., 2007). This study focused on the three following measures of head impact magnitude: (1) peak linear acceleration (g), (2) peak rotational acceleration (rad/sec^2), and (3) Head Impact Technology severity profile, which is a weighted composite score including linear acceleration, rotational acceleration, impact duration, and impact location. Computation of these biomechanical measures have previously been reported (Greenwald, et al., 2008).

We categorized the linear acceleration, rotational acceleration, and HITsp of each head impacts into quartiles (Table 5.3). We also created separate categories for head impacts that occurred in the 95th and 99th percentiles. We chose to include the 95th and 99th percentile categories because collapsing head impacts of this magnitude into the 4th quartile may have limited the detail of our results regarding impacts that are considered more severe.

Table 5.3. Head impact biomechanics categorization cutoffs and frequencies.

	1st Quartile (1st-24th)	**2nd Quartile (25th-50th)**	**3rd Quartile (50th-74th)**	**4th Quartile (75th-94th)**	**95th Percentile (95th-98th)**	**99th Percentile (99th-100th)**
Linear Acc. (g)	<15.3 n=3592	15.3 ≥ or < 21.0 n=3864	21.0 ≥ or < 31.3 n=4224	31.3 ≥ or < 58.7 n=3473	58.7 ≥ or < 90.2 n=731	≥ 90.2 n=182
Rotational Acc (rad/sec^2)	<960.6 n=5012	960.6 ≥ or < 1409.2 n=3715	1409.2 ≥ or < 2120.3 n=3609	2120.3 ≥ or < 4093.5 n=2966	4093.5 ≥ or < 6513.9 n=605	≥ 6513.9 n=159
HITsp	<10.3 n=5016	10.3 ≥ or < 14.0 n=3586	14.0 ≥ or < 17.9 n=3695	17.9 ≥ or < 31.6 n=2983	31.6 ≥ or < 53.5 n=629	≥ 53.5 n=157

Statistical Analyses

Random intercepts, general mixed linear, proportional odds models were used to compute odds ratios (OR) and 95% confidence intervals (CI) for each of the nine dichotomized (low, high) visual performance measure. We computed the odds of sustaining head impacts in the 2nd quartile, 3rd quartile, 4th quartile, 95th percentile, or 99th percentile versus the reference category of head impacts in the 1st quartile across groups of high and low performers for each cervical characteristic for each of the following categorized measures of head impact magnitude: linear acceleration, rotational acceleration, and HITsp. For all models, 1st quartile head impacts and low performers were the reference categories.

We also included position group assignment (skill, line) as a predictor variable to control for differences across player positions. Table 5.4 includes the unit of measure and reference category for each visual performance variable.

Table 5.4. Visual performance variable table indicating the unit of measure and high performance categories.

Subtest	Unit of Measure	Below the Median	Above the Median
Visual Clarity	LogMar values for oculus Uterque	Better visual clarity (High Performance)	Worse visual clarity
Contrast Sensitivity	Cycles per degree threshold (cpd)	Worse sensitivity to contrast	Better sensitivity to contrast (High Performance)
Depth Perception	Arc second threshold (arc sec)	Worse perception of depth	Better perception of depth (High Performance)
Near Far Quickness	# of correct responses	Slower near far quickness	Faster near far quickness (High Performance)
Target Capture	millisecond threshold (ms)	Better target capture (High Performance)	Worse target capture
Perception Span	# of correct responses – incorrect responses (dots)	Worse perception span	Better perception span (High Performance)
Eye-Hand Coordination	total time to touch all 96 dots (seconds)	Better eye-hand coordination (High Performance)	Worse eye-hand coordination
Go No Go	# of turquoise dots touched minus any red dots touched (dots)	Worse decision making	Better decision making (High Performance)
Reaction Time	elapsed time between onset of the test annulus and release of the control annulus (ms)	Faster reaction time (High Performance)	Slower reaction time

All statistical analyses were completed using SAS 9.3 (SAS Institute Inc., Cary, NC). Odds ratios greater than one indicate an increased odds among athletes categorized into the high performance group, whereas odds ratios less than one indicate a reduced odds among the high performance group. Analyses were considered significant if the 95% CI about the odds ratio did not contain one.

Results

Descriptive statistics for each visual performance variable are included in Table 5.5. The high performance group performed significantly better than the low performance group for all visual performance variables ($p<0.001$).

All odds ratios and 95% confidence intervals are presented in Table 5.6. We did not observe any differences in odds between high and low performers for the following visual performance variables: visual clarity, contrast sensitivity, depth perception, perception span, eye-hand coordination, and go no go. However, contrary to our hypotheses, players that were better able to switch between near and far stimuli (near-far quickness), were better able to quickly identify a target in the periphery (target capture), and players with quicker reaction times (reaction time) had increased odds of sustaining higher magnitude head impacts. In contrast, players that were better able to switch between near and far stimuli (near-far quickness) had reduced odds of sustaining head impacts in the 2nd quartile compared to the 1st quartile.

Table 5.5. Descriptive statistics and between group comparisons for low and high performers for each visual performance variable.

	Low					High						
	n	Line†	Skill‡	Mean	SD	n	Line	Skill	Mean	SD	t	p*
Visual Performance Measure												
Visual Clarity	10	5	5	0.11	0.26	27	9	18	-0.28	0.09	4.58	<0.001
Contrast Sensitivity	30	12	18	1.57	0.32	7	2	5	2.00	0.00	-7.34	<0.001
Depth Perception	20	6	14	42.65	34.61	17	8	9	131.92	72.23	-4.66	<0.001
Near Far Quickness	21	9	12	20.00	4.10	16	5	11	27.81	2.79	-6.55	<0.001
Target Capture	20	5	15	202.50	57.30	17	9	8	423.53	113.36	-7.29	<0.001
Perception Span	19	9	10	24.74	6.19	18	5	13	41.17	8.79	-6.54	<0.001
Eye-Hand Coordination	18	12	6	66941.44	8315.33	19	2	17	56321.47	1978.86	5.28	<0.001
Go / No Go	19	10	9	8.00	4.88	18	4	14	23.39	6.65	-8.06	<0.001
Reaction Time	18	8	10	403.33	70.00	18	5	13	339.42	11.76	3.82	<0.001

† Line: defensive end, nose tackle, defensive tackle, center, guard, or offensive tackle

‡ Skill: linebacker corner, or safety, quarterback, receiver, tight end, running back, or full back

* Low and high performers were significantly different for all measures of visual performance

Table 5.6. Visual Performance: Odds ratios (OR) and 95% confidence intervals (CI) indicating the high performance group's odds of sustaining higher magnitude head impacts, rather than 1st quartile head impacts, compared to the low performance group.

			2nd v. 1st Quartile	3rd v. 1st Quartile	4th v. 1st Quartile	95th Percentile v. 1st Quartile	99th Percentile v. 1st Quartile
		df	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**	**OR (95% CI)**
Visual Clarity	Linear	22	1.02 (0.91-1.14)	0.98 (0.81-1.19)	0.95 (0.71-1.27)	1.05 (0.66-1.67)	1.35 (0.78-2.35)
	Rotational	22	0.94 (0.81-1.09)	0.90 (0.74-1.08)	0.89 (0.63-1.25)	1.04 (0.60-1.82)	1.20 (0.53-2.72)
	HITsp	22	1.09 (0.87-1.38)	0.95 (0.74-1.22)	0.94 (0.62-1.42)	0.95 (0.51-1.78)	1.36 (0.62-2.99)
Contrast Sensitivity	Linear	22	0.97 (0.86-1.09)	0.97 (0.79-1.19)	0.98 (0.71-1.35)	1.20 (0.74-1.96)	0.75 (0.42-1.35)
	Rotational	22	1.10 (0.94-1.29)	1.06 (0.86-1.30)	1.08 (0.74-1.58)	1.10 (0.60-2.01)	0.89 (0.37-2.12)
	HITsp	22	0.94 (0.73-1.21)	1.00 (0.75-1.32)	1.16 (0.74-1.82)	1.06 (0.54-2.09)	0.83 (0.36-1.91)
Depth Perception	Linear	22	1.02 (0.91-1.14)	1.00 (0.84-1.19)	0.96 (0.73-1.25)	0.89 (0.58-1.37)	0.87 (0.51-1.48)
	Rotational	22	1.14 (1.00-1.30)	1.11 (0.94-1.32)	1.13 (0.84-1.53)	1.11 (0.67-1.84)	1.21 (0.59-2.48)
	HITsp	22	1.11 (0.91-1.37)	1.23 (0.99-1.53)	1.11 (0.77-1.60)	1.02 (0.58-1.81)	0.99 (0.48-2.05)
Near-Far Quickness	Linear	22	0.97 (0.88-1.08)	1.04 (0.88-1.24)	1.11 (0.85-1.44)	1.19 (0.79-1.78)	0.82 (0.50-1.36)
	Rotational	22	1.01 (0.88-1.16)	1.06 (0.89-1.25)	1.33 (1.00-1.79)	1.14 (0.70- 1.87)	0.94 (0.45-1.91)
	HITsp	22	***1.28 (1.06-1.54) †***	***1.34 (1.09-1.65) †***	***1.51 (1.08-2.12) †***	***1.75 (1.05-2.94) †***	0.99 (0.49-2.00)
Target Capture	Linear	22	1.04 (0.93-1.17)	1.17 (0.99-1.40)	***1.44 (1.11-1.87) †***	1.35 (0.88-2.08)	1.20 (0.71-2.05)
	Rotational	22	1.07 (0.93-1.24)	***1.22 (1.03-1.45) †***	***1.22 (1.16-2.10) †***	1.18 (0.70-1.98)	1.22 (0.57-2.61)
	HITsp	22	***0.81 (0.66-0.99) ‡***	0.95 (0.75-1.21)	1.23 (0.84-1.80)	1.26 (0.69-2.27)	0.95 (0.46-1.97)
Perception Span	Linear	22	0.99 (0.89-1.10)	0.90 (0.76-1.06)	0.89 (0.68-1.16)	0.85 (0.56-1.30)	0.99 (0.60-1.65)
	Rotational	22	1.12 (0.98-1.28)	1.05 (0.88-1.25)	1.14 (0.84-1.55)	1.10 (0.66-1.82)	1.60 (0.75-3.41)
	HITsp	22	0.96 (0.77-1.19)	0.97 (0.77-1.23)	0.93 (0.64-1.35)	1.07 (0.61-1.89)	0.87 (0.43-1.77)
Eye-Hand Coordination	Linear	22	1.03 (0.91-1.17)	1.04 (0.84-1.30)	1.04 (0.75-1.45)	1.21 (0.72-2.02)	1.55 (0.84-2.85)
	Rotational	22	0.95 (0.80-1.13)	0.98 (0.79-1.22)	0.93 (0.63-1.37)	1.46 (0.78-2.71)	1.53 (0.62-3.77)
	HITsp	22	1.07 (0.82-1.39)	1.04 (0.78-1.39)	0.98 (0.62-1.56)	1.39 (0.70-2.76)	1.92 (0.81-4.57)
Go No Go	Linear	22	0.92 (0.83-1.03)	0.88 (0.74-1.04)	0.84 (0.64-1.10)	0.85 (0.56-1.31)	1.22 (0.73-2.03)
	Rotational	22	1.10 (0.96-1.26)	1.00 (0.84-1.19)	0.93 (0.67-1.27)	1.06 (0.63-1.79)	1.51 (0.73-3.13)
	HITsp	22	0.91 (0.73-1.13)	0.96 (0.75-1.22)	0.85 (0.58-1.24)	1.01 (0.57-1.81)	1.18 (0.57-2.44)
Reaction Time	Linear	21	1.07 (0.96-1.20)	1.16 (0.97-1.38)	1.30 (1.00-1.68)	1.46 (0.99-2.15)	1.18 (0.72-1.95)
	Rotational	21	1.03 (0.90-1.19)	1.15 (0.96-1.37)	***1.39 (1.05-1.85) †***	***1.73 (1.05-2.84) †***	1.36 (0.66-2.79)
	HITsp	21	1.16 (0.94-1.43)	1.16 (0.92-1.46)	***1.48 (1.04-2.12) †***	1.68 (0.97-2.89)	1.31 (0.65-2.64)

† Players classified as high performers had increased odds (OR greater than 1 indicate increased odds for the high performance group)

‡ Players classified as high performers had reduced odds (OR less than 1 indicate reduced odds for the high group)

Discussion

Overall, our results suggest that high school football players have similar odds of sustaining higher magnitude head impacts regardless of their visual abilities. These results do not support our hypotheses that players that are better able to acquire and interpret visual stimuli have reduced odds of sustaining higher magnitude head impacts. More research is needed to determine the role of visual performance in reducing the severity of head impacts sustained while playing football.

High school players within a single team often vary greatly in skill, physical strength, and athletic ability. Specifically, the high school team included in this study consisted of athletes that were very talented football players being recruited to Division I collegiate football programs and athletes that had never played organized football before. This is very rarely the case at the collegiate and professional levels. It seems likely that differences in head impact magnitude attributable to visual characteristics may be more apparent in collegiate or professional athletes because they are more homogenous in skill and experience than high school players. Head injury protection is not likely influenced by one sole attribute, such as vision. Further research is needed to determine the role of vision in mitigating head impact severity as well as other proposed preventative mechanisms, such as cervical strength, field awareness, proper tackling form, and anticipation (Mihalik, Blackburn, et al., 2010; Mihalik, et al., 2011; Viano, et al., 2007).

Visual training with the goal of improving athletic performance is supported by previous research (Hitzeman & Beckerman, 1993; Laby, et al., 1996; Stine, et al., 1982; Zimmerman, et al., 2011). Although visual training in athletes is a relatively new concept, visual exercises have been shown to improve visual performance (Maxwell, et al., 2012).

How improvements in visual performance relate to sport performance and injury prevention remains unknown. Currently, visual training interventions with the goal of concussion prevention are not feasible or appropriate at most high schools. Sports medicine professionals and strength and conditioning experts that work at the high school level are often limited in physical resources, financial resources, and time. High school athletes are often engaged in more than one sport and multiple extra curricular activities. Visual training with the goal of improving sport performance may be a more feasible recommendation at the collegiate and professional levels.

High school football players that were better able to switch between near and far stimuli and players that can more quickly identify stimuli in their peripheral vision actually had increased odds of sustaining higher magnitude head impacts. The exact mechanism behind our observations regarding near-far quickness and target capture are not fully understood, but we speculate that football players that are able to quickly shift their visual focus may alter fixations while they play, which may cause them to take their eyes off of an oncoming opponent. Athletes that are more skilled at their sport present with better overall visual performance and use visual search strategies that involve more fixations of shorter duration, allowing them to alternate their gaze more frequently (Canal-Bruland, et al., 2011; Higuchi, et al., 2011; Ida, et al., 2011; Roca, et al., 2011; Roca, Ford, McRobert, & Williams, 2013). The high performance groups in this study may have contained more skilled football players with better near far quickness and target capture performance. These players may choose to hit harder or may sustain more severe head impacts because of their role as a more skilled player on the team.

Our results regarding target capture performance were contradictory across quartiles. While we observed that players that were better able to quickly identify a target in the periphery had increased odds of sustaining head impacts in the 3^{rd} and 4^{th} quartiles, we also observed that the same players had reduced odds of sustaining head impacts in the 2^{nd} quartile compared to the 1^{st} quartile. Although we observed that the high performance group had a 19% reduction in odds of sustaining head impacts in the 2^{nd} quartile compared to the 1^{st}, these results are likely clinically inconsequential as the 2^{nd} quartile contains head impacts ranging in linear magnitudes of 15.3g to 21.0g. The upper cutoff of this head impact category approaches the mean linear magnitude previously reported for both college and high school football players (Broglio, et al., 2009; Mihalik, et al., 2007), which suggests that although these head impacts are more severe than 1^{st} quartile head impacts, they remain mild (Zhang, et al., 2004). This single result contradicts a majority of our other findings and the upper limit of the confidence interval approaches equal risk between groups.

Previous studies suggest that youth athletes tend to sustain lower magnitude impacts when they fully anticipate collisions (Mihalik, Blackburn, et al., 2010) and when they serve as the striking rather than the struck player (Viano, et al., 2007). Eckner *et al.* (Eckner, Lipps, Kim, Richardson, & Ashton-Miller, 2011) found that players with quicker reaction time have an enhanced ability to protect the head during a simulated sport activity where participants were required to block foam balls fired towards their heads. Our observed trends regarding reaction time suggest that players with quicker reaction time may be more likely to sustain higher magnitude head impacts compared to those with slower reaction time. While this may seem counterintuitive, it could be explained by the fact that football players with quicker reaction time are more skilled athletes and may be more exposed to high velocity collisions. Further research is needed to determine the potential relationship between enhanced visual performance, an athlete's ability to anticipate impending collision, and risk of concussion.

We utilized a visual performance measure that is relatively new and novel. Only one previous study has examined the reliability of the Nike SPARQ Sensory Station (Erickson, et al., 2011) and no previous studies have determined the system's validity. Although the system is thought to be a more sport specific method of measuring visual performance, more research is needed regarding this system's validity. We measured visual performance at the beginning of the season. It is possible that visual performance fluctuates throughout the season. We chose to include high school football athletes only. The results of our study may not pertain to sports with different visual demands or other levels of play, such as youth, collegiate, or professional. Football involves a wide variety of visual stimuli, which are somewhat dependent on player position. A quarterback has very different visual demands compared to a tight end and a tight end can have varying visual demands depending on the task required in a given play (i.e. blocking versus receiving). The wide range of visual demand placed on players over the course of the season may explain why we did not observe a significant link between visual performance and the odds of sustaining higher magnitude head impacts.

Conclusions

Our study does not support the notion that high school football players with improved visual performance have reduced odds of sustaining higher magnitude head impacts. At this time, we do not recommend wide spread use of visual training programs at the high school level for the purpose of reducing the risk of sustaining higher magnitude head impacts. More research is needed to determine the role of visual performance and visual training in reducing the severity of head impacts sustained while playing football.

Chapter 6

MANUSCRIPT III

Player level of anticipation prior to collision and head impact biomechanics in high school football.

Introduction

Context: Previous studies suggest that collisions that occur when an athlete has adequate time to evoke anticipatory responses may result in less severe head impact magnitudes. However, the role of anticipation in mitigating head impact severity among high school football athletes has not been studied. **Objective**: To compare head impact biomechanical measures of severity between anticipated and unanticipated collisions in high school football. **Setting**: On-field. **Patients or Other Participants**: Thirty high school American football players. **Interventions**: Head impact biomechanics were captured for each player using the Head Impact Telemetry System. We captured and analyzed video footage from 11 regular season and 2 playoff high school football games over the course of the 2012 season to determine player level of anticipation prior to collision for 2,901 head impacts. **Main Outcome Measures**: We conducted three separate random intercepts general linear mixed models to assess the differences in head impact biomechanical measures of severity (dependent variables: linear acceleration, rotational acceleration, and HITsp) between levels of anticipation (independent variables: anticipated, unanticipated) (α=0.05). **Results**: No significant differences in linear

acceleration ($F_{1,26}$=0.00, p=0.991), rotational acceleration ($F_{1,26}$=1.40, p=0.249), or HITsp ($F_{1,26}$=1.30, p=0.265), were observed between anticipated and unanticipated collisions.
<u>**Conclusions**</u>: Our results do not indicate that anticipated and unanticipated head impacts differ in severity amongst high school football players. Research utilizing more objective measures of player anticipation is needed to determine whether level of anticipation prior to collisions influences head impact severity among football players and other athletes.

Recent global conversation has focused on the dangers of participation in football. The future of football may depend on the sport's ability to address concerns regarding safety, specifically as it relates to sport-related concussion. Although the majority of attention has been paid to this issue in the National Football League, the true "concussion crisis" exists among youth and adolescent athletes who sustain concussions at higher rates, higher overall numbers, and have the least access to medical care (Gessel, et al., 2007; Marar, et al., 2012). Efforts to improve safety in football must transcend all levels of play and must be research and evidence driven. More research is needed to guide concussion prevention efforts in football.

An athlete that is able to foresee an impending collision will react with anticipatory responses, such as leaning, using the arms to block the face, and recoiling their head by elevating their shoulders (Metoyer, et al., 2008). During sport, athletes must maintain gaze fixation on a target area, such as a goal, a ball, or a teammate. Gaze fixation may limit an athlete's ability to foresee, anticipate, and prepare for impending collision (van der Kamp, 2011). Among youth ice hockey players, collisions that are unanticipated tend to result in more severe head impacts compared to anticipated collisions (Mihalik, Blackburn, et al., 2010). Studies that have reconstructed helmet-to-helmet impacts that resulted in concussion among National Football League players show that the struck players, on average, experience 98g of linear head acceleration while the striking player only experiences 58.5g (Viano, et al., 2007). Because the striking player fully anticipates the impending collision they impart much greater force on the struck player. Low-speed rear-end motor vehicle accidents that occur when the passenger is fully aware of the impending impact result in reduced acceleration of head and neck (Kumar, et al., 2000). Anticipatory responses to impending head or body collisions may help mitigate acceleration of the head, thereby reducing the potential for sustaining a brain injury and reducing the magnitude of subconcussive impacts. If level of anticipation at the time of collision serves to reduce head impact magnitude then training aimed at improving an athlete's ability to anticipate would be warranted. The purpose of this study was to compare head impact biomechanical measures of severity between anticipated and unanticipated collisions in high school football players.

Methods

Study Participants

Thirty-seven high school varsity football players from a single local high school enrolled in this study, however, no video footage was captured for 7 of the 37 participants, leaving a final sample size of 30. Demographic information is presented in Table 6.1. All data were captured at 11 regular season and 2 playoff high school football games over the course of the 2012 season. High school athletes signed informed consent forms approved by the Institutional Review Board. Legal guardians of high school athletes under the age of majority also signed informed consents forms.

Table 6.1. Demographic Information for all participants

Demographic	Mean	SD
Age (yrs)	16.71	0.92
Height (cm)	180.98	6.53
Mass (kg)	87.17	16.10
Year (Athletic)		
Freshmen	0	
Sophomore	7	
Junior	7	
Senior	16	
Position Group		
Skill (offense, defense)	17 (6, 11)	
Line (offense, defense)	13 (8, 5)	

Procedures

Head Impact Biomechanics

Head impact biomechanics were captured using the Head Impact Telemetry (HIT) System technology (Riddell Corp., Elyria, OH). The HIT System consists of MxEncoder units located in the football helmets, a signal transducer, and a laptop computer that houses the Sideline Response System (Riddell Corp., Elyria, OH). MxEncoder units embedded within the Revolution and Speed helmet designs (Riddell Inc., Elyria, Ohio) are comprised of six spring-loaded single-axis accelerometers, a telemetry unit, a data storage device, and a battery power source. Head impact biomechanical data were time-stamped, encoded, and then transmitted in real-time to the signal transducer via radiofrequency transmission at 903–927 MHz. The signal transducer was connected through a USB port to a laptop computer, which stored all head impact data. The HIT system has been described in greater detail in several previous studies (Broglio, et al., 2009; Guskiewicz & Mihalik, 2011).

Video Footage Capture

We captured video footage using a Panasonic HMC-40 (Panasonic System Communications Company of North America, Secaucus, NJ) placed above the press box (~3 stories high) at the 50-yard line. A research assistant monitored the camcorder by adjusting the zoom and field of view as plays progressed up and down the field. Every effort was made to adjust the camera to maintain adequate zoom while also maintaining a wide field of view. The camcorder and Sideline Response System were date and time synchronized prior to each game to allow for matching of observable collisions to head impact biomechanical measures recorded by the HIT System.

Data Reduction

Head Impact Biomechanics

Head impact data were exported from the Sideline Response System into Matlab 7. Consistent with previous studies, we then reduced the data to include only those impacts that register greater than or equal to 10g of linear acceleration (Guskiewicz, Mihalik, et al., 2007; Mihalik, et al., 2007; Mihalik, Blackburn, et al., 2010; Mihalik, Greenwald, et al., 2010; Mihalik, et al., 2011; Schnebel, et al., 2007). We focused primarily on two traditional measures of head impact severity (linear acceleration and rotational acceleration) and one weighted combination of several biomechanical inputs, including linear acceleration, rotational acceleration, impact duration, and impact location (HITsp) (Greenwald, et al., 2008). Once head impact biomechanical data were exported, a separate spreadsheet was generated for use during video analysis that contained the date, time, players' unique identification numbers, and a unique code assigned to each head impact. To avoid rater bias, the spreadsheet did not contain biomechanical measures of head impact severity or location.

Video Assessment of Level of Anticipation

We analyzed video footage of on-field collisions using a modified version of the Player-to-Player Form previously used by Mihalik *et al.* (Mihalik, Blackburn, et al., 2010) to examine collision characteristics in youth ice hockey and used by Ocwieja *et al.* (Ocwieja, et al., 2012) to examine collision characteristics in collegiate football. The questions contained within the Player-to-Player Form were transferred to spreadsheet format with validated drop-down entries following the date, time, and unique ID for each head impact. We matched game head impact biomechanical measures with video using synchronized time-stamps. Head impact biomechanical data were sorted by date and by time of head impact. Raters determined the time of head impact and cued the video footage to the appropriate hour, minute, and then second. Each viewable collision was deemed as anticipated, unanticipated, or unknown using the following questions:

a. Was the player positioned to be looking in the direction of impending body collision?
b. Was the player in a general athletic readiness position (knee and trunk flexion with feet shoulder-width apart, and used their legs to drive their shoulders through the collision)?

Collisions were deemed anticipated if the impact occurred while the athlete was looking in the direction of the impending collision and was in a general athletic readiness position. Collisions were deemed unanticipated if the player was not in an athletic readiness position regardless of gaze direction. These categorizations are consistent with

Mihalik et al, (Mihalik, Blackburn, et al., 2010) with poorly anticipated and unanticipated collisions under a single category. Collisions were deemed unknown if the investigator was unable to identify the direction of gaze or the positioning of the body. Video analysis was completed over the course of four months by five different raters and the primary investigator. Each rater was instructed on proper grading by the primary investigator and completed a reliability segment of 91 head impacts. Raters were blind to assessment of their reliability. We observed moderate inter-rater reliability (kappa: 0.45-0.72, $p<0.05$) for all raters when comparing each rater to the rater that completed the most video analysis. Collisions analyzed by one rater with a kappa value of 0.11 and another with 0.25 were excluded due to poor to moderate reliability. Once all video had been analyzed, video analysis data were merged with head impact biomechanical measures of severity and location using the head impact unique IDs.

Statistical Analyses

All statistical analyses were performed in SAS (Version 9.3; SAS Institute, Inc, Cary, North Carolina) with an a priori alpha level of 0.05. Head impact biomechanical data were evaluated for skewness and natural logarithmic transformed to satisfy the normality assumptions. Descriptive statistics presented in our results are back-transformed from the natural log to display meaningful values. We excluded all head impacts that occurred outside of the camera's field of view, head impacts that did not result from a collision with another player (i.e. contact with the ground), collisions where level of anticipation could not be determined, and head impacts that resulted subsequent to an initial head impact (i.e. the player was struck twice or more during a single collision).

We conducted three separate random intercepts general linear mixed models to assess the differences in head impact biomechanical measures of severity (dependent variables: linear acceleration, rotational acceleration, and HITsp) between the levels of anticipation (independent variables: anticipated, unanticipated).

Results

We observed 6,936 game head impacts, of which, 3,866 (55.7%) were viewable on video footage. Of the viewable collisions, 258 head impacts did not result from contact with another player (e.g. ground), 313 head impacts resulted when athletic readiness and direction of gaze could not be adequately determined, and 394 head impacts were not the first impact following collision. Among the 2,901 remaining head impacts, 2,347 (75.1%) were deemed anticipated and 554 (17.7%) were deemed unanticipated.

No significant differences in linear acceleration ($F_{1,26}$=0.00, p=0.991), rotational acceleration ($F_{1,26}$=1.40, p=0.249), or HITsp ($F_{1,26}$=1.30, p=0.265), were observed between anticipated and unanticipated collisions.

Table 6.2. Descriptive and statistical results for head impacts magnitude measures between anticipated and unanticipated collisions.

	Linear Acceleration (g)					Rotational Acceleration (rad/sec^2)					HITsp				
	Mean	Lower	Upper	F	P	Mean	Lower	Upper	F	P	Mean	Lower	Upper	F	P
Anticipated	27.89	26.50	29.34	0.00	0.991	1661.21	1522.79	1812.21	1.40	0.249	17.10	16.11	18.14	1.30	0.265
Unanticipated	27.90	26.48	29.39			1745.86	1611.63	1891.45			17.69	16.66	18.79		

Discussion

Head impacts that occur as a result anticipated and unanticipated collisions did not differ in magnitude. We suspect that we did not observe a significant difference between levels of anticipation because few impacts in football are truly unanticipated. Although football is known to be a high speed, high impact sport, rules regarding striking a defenseless player may be effective in limiting the frequency and severity of unanticipated collisions.

Football plays have a very definitive start and offensive players have very planned actions. Both offensive and defensive linemen expect to make contact with an opponent during nearly every play. Offensive players, in particular, execute a planned and deliberate movement that is determined prior to the snap. This is an intrinsic difference between football and ice hockey and may explain why our results differ slightly from previously observed trends towards more severe head impacts as a result of unanticipated collisions (Mihalik, Blackburn, et al., 2010). It is possible that a football player that is not in athletic readiness position and is not looking in the direction of impending collision could still anticipate an impending collision, particularly if that player is carrying, passing, receiving or snapping the ball.

Level of anticipation prior to collision has also been investigated among collegiate football players using similar methods (Mihalik, Moise, Ocwieja, Guskiewicz, & Register-Mihalik, In Review). Much like our results, the authors did not observe a significant difference in head impact magnitudes between anticipated and unanticipated collisions; however, we observed a much higher percentage of unanticipated head impacts. Together, these studies suggest that level of anticipation does not fully explain why some players sustain higher magnitude head impacts. Other characteristics of play, such as player role (struck vs. striking), closing distance, play type, or ball possession, may better explain differences in head impact severity (Ocwieja, et al., 2012).

In rugby, ball carriers have a higher injury rate when tackled from behind their visual field compared to when tackled from within their visual field (King, et al., 2012). Although, our results do not indicate reduction of head impact magnitude when an athlete sees and prepares for an oncoming collision, we suspect that anticipation may play a larger role in other sports. Future studies should examine the influence of athlete anticipation on head protection in sports like soccer, basketball, and rugby. Sport skill and expertise may be a more important factor to examine when assessing the odds of sustaining higher magnitude head impacts.

When considering the complex task of head protection, it seems that more skilled athletes would be better able to anticipate collisions while playing football and thereby reduce head acceleration by adopting head protections strategies (Roca, et al., 2011; Roca, et al., 2013). Expert athletes present with an enhanced ability to identify subtle changes in movement patterns used by their opponent, have greater efficiency in running through narrow apertures, and are more accurate in their anticipation and decision-making judgments compared with less skilled players (Canal-Bruland, et al., 2011; Higuchi, et al., 2011; Ida, et al., 2011; Roca, et al., 2011; Roca, et al., 2013). Likewise, skilled athletes use visual search strategies that involve more fixations of shorter duration, alternating their gaze more frequently between the player in possession of the ball, the ball itself, and other areas of the field of play. In sports anticipation tasks, expert athletes show stronger neural activations than novice athletes in brain areas that are associated with visual attention and the analysis of body kinematics (Wright, et al., 2011). Novice athletes show stronger neural activation in the occipital cortex, which suggests a greater allocation of resources to low-level visual processing. Measurement of eye movements and brain activation during athletic tasks, although more methodologically challenging, may provide a better representation of player anticipation than the video analysis techniques used in this study.

We did not examine the risk of concussion between anticipated and unanticipated head impacts. Future studies should determine whether players that sustain unanticipated collisions are at an increased risk of concussion. We examined level of anticipation as a binary variable using video analysis, but it is likely that anticipation of impending collisions occurs along a spectrum and is not fully represented as a dichotomy. Future studies should identify methods for determining player anticipation more objectively. This could be done through direct identification of gaze direction using eye-tracking technology or through player self-report of their level of anticipation prior to collision. We did not analyze the influence of level of anticipation separately between conditions where players served as either the struck or striking player. Further research is necessary to determine whether level of anticipation plays a larger role when the player serves as either the struck or striking player as these two collision types possess fundamentally different collision characteristics (Viano, et al., 2007; Viano & Pellman, 2005).

Conclusions

Surprisingly, our results indicate that the severity of anticipated and unanticipated head impacts is similar among high school football players. Further research utilizing more empirical methods are needed to determine whether level of anticipation prior to collisions plays a role in head impact severity among football players. Although not directly studied, we speculate that rules regarding striking a defenseless player may be affective in protecting football athletes from encountering a high number of truly unanticipated collisions.

Chapter 7

RESEARCH QUESTION FOUR OVERVIEW

Overview I: The Relative Contributions of Cervical Characteristics, Visual Performance, and Level of Anticipation in Mitigating Head Impact Magnitude

Context: Athletes with weaker, smaller, and less stiff cervical musculature; diminished visual performance; and that do not anticipate an oncoming collision are thought to be more likely to experience rapid head acceleration following collision. Studies regarding the role of the cervical musculature, visual performance, and level of anticipation have been inconclusive. Further research is needed to determine if head impact biomechanical measures are influenced by these factors. **Objective**: To determine if cervical musculature characteristics, visual performance, and level of anticipation predict the severity of head impacts sustained by high school football players. **Design:** Prospective quasi-experimental. **Setting**: Laboratory/On-field. **Patients or Other Participants**: Twenty-seven high school football players. **Interventions**: Athletes completed the cervical testing protocol and visual performance assessment prior to the season. The cervical testing protocol consisted of measures of cervical isometric strength using an isokinetic dynamometer, ultrasonographic cross-sectional area, and dynamic cervical response to perturbation. Visual performance was measured using the Nike SPARQ Sensory Station. We reviewed video footage captured during all 13 high school football games to determine each athlete's level of anticipation at the time of collision.

Head impact biomechanics were captured for each player using the Head Impact Telemetry System. **<u>Main Outcome Measures</u>**: Predictor variables included the five following cervical characteristic measures: composite peak torque (Nm/kg), composite rate of torque development (Nm/sec), composite cross sectional area (cm^2), composite stiffness (Nm/rad), and composite muscle onset latency (ms); the following visual performance measures: visual acuity, contrast sensitivity, depth perception, near-far quickness, target capture, perception span, eye-hand coordination, go/no go, and reaction time; and the binary variable of level of anticipation (anticipated, unanticipated). Collisions where the level of anticipation could not be determined were excluded. We conducted a single random intercepts general linear mixed model for each head impact biomechanical measure of severity (criterion: linear acceleration, rotational acceleration, and HITsp) ($\alpha = 0.05$). **<u>Results</u>**: We determined level of anticipation for 2,822 head impacts. Target capture was a significant predictor of rotational acceleration ($p=0.041$). **<u>Conclusions</u>**: Our results indicate that players who can more rapidly shift their gaze to recognize of peripheral targets sustain less severe head impacts. Prevention efforts should be aimed at improving peripheral vision and the saccade efficiency. Further research is needed to determine whether cervical characteristics, other visual performance measures, and level of anticipation play a role in mitigating head impact severity.

Overview II: Do Cervical Muscle Characteristics and Visual Performance Measures Predict Biomechanical Head Impact Profiles?

Context: Cervical muscle characteristics and visual performance are thought to influence head acceleration following collision. Further research is needed to determine whether cervical strength and conditioning programs and visual performance training warrant consideration as means for concussion prevention in sport. **Objective**: To determine if cervical musculature characteristics and visual performance predict profiles of head impact severity and frequency in high school and collegiate football players. **Design:** Prospective quasi-experimental. **Setting**: Laboratory/On-field. **Patients or Other Participants**: Forty-nine American football players (34 high school, 15 collegiate) participated in this study, however nine players were excluded because they did not complete more than 50 plays throughout the season (n=40). **Interventions**: Athletes completed the cervical testing protocol and visual performance assessment prior to the 2012 football season. The cervical testing protocol consisted of measures of cervical strength, ultrasonographic cross-sectional area, and dynamic cervical response to perturbation. Visual performance was measured using the Nike SPARQ Sensory Station and included measures of visual acuity, contrast sensitivity, depth perception, near-far quickness, target capture, perception span, eye-hand coordination, go/no go, and reaction time. Head impact biomechanics were captured for each player using the Head Impact Telemetry System. The primary investigator tracked the number of plays that each player completed. **Main Outcome Measures**: Criterion variables included four separate composite profiles, which were computed by dividing the sum of the 1) linear acceleration, 2) rotational acceleration, 3) HITsp, and 4) frequency of all head impacts sustained by each player during games by the number of plays completed. Predictor variables included composite peak torque (Nm/kg), composite rate of torque development (Nm/sec), composite cross sectional area (cm^2), composite stiffness (Nm/rad), composite muscle onset latency (ms), and each athlete's composite percentile ranking for visual performance. We conducted separate multiple regression analyses for each of the four profiles using the enter method (α= 0.05). **Results**: Head impact profiles were log transformed because they were not normally distributed. Our model did not predict a significant amount of variance for any of the biomechanical profiles. **Conclusions**: Cervical characteristics and visual performance do not predict player head impact profiles for severity and frequency. Combining data obtained from head impacts sustained over the course of an entire season may not accurately reflect a player's propensity to sustain severe head impacts.

APPENDIX I: PLAYER TO PLAYER FORM

Player to Player Collision Type Evaluation Form

50848

Impact [ ][ ][ ][ ][ ][ ][ ][0]

Quarter [ ][0][0]   Game Clock [ ][ ] : [ ][ ] (Minutes) (Seconds)   Rater: [ ][0][0]

C1. Play Type

- [ ] Punt (0)
- [ ] Defense Pass (6)
- [ ] Punt Return (1)
- [ ] Defense Rush (7)
- [ ] Kickoff (2)
- [ ] Field Goal (8)
- [ ] Kickoff Return (3)
- [ ] Field Goal Block (9)
- [ ] Offense Pass (4)
- [ ] Extra Point (10)
- [ ] Offense Rush (5)
- [ ] Extra Point Block (11)

C2. Closing Distance Type

- [ ] Long Distance (0)
- [ ] Short Distance (1)
- [ ] Unknown (2)

C3. What stance did the opponent begin in

- [ ] 2pt (0)
- [ ] Unknown (3)
- [ ] 3pt (1)
- [ ] N/A (4)
- [ ] 4pt (2)

C4. What stance did UNC begin in

- [ ] 2pt (0)
- [ ] 4pt (2)
- [ ] 3pt (1)
- [ ] Unknown (3)

C5. Player involvement in body collision

- [ ] Striking player (0)
- [ ] Player struck (1)
- [ ] Unknown (2)

C6. Player looking ahead in direction of movement

- [ ] No (0)
- [ ] Yes (1)
- [ ] Unknown (2)

C7. Player appears to be looking in direction of impending body collision

- [ ] No (0)
- [ ] Yes (1)
- [ ] Unknown (2)

C8. Was the opponent stationary

- [ ] No (0)
- [ ] Unknown (2)
- [ ] Yes (1)
- [ ] N/A (3)

C9. Was the UNC player stationary

- [ ] No (0)
- [ ] Yes (1)
- [ ] Unknown (2)

C10. Did the player have possession of the ball

- [ ] No (0)
- [ ] Yes (1)
- [ ] Unknown (2)

C11. Was the player receiving/passing the ball at time of collision? (hand-off, pitch, pass, catch)

- [ ] No (0)
- [ ] Yes (1)
- [ ] Unknown (2)

C12. Was the player snapping the ball (center)

- [ ] No (0)
- [ ] Yes (1)
- [ ] N/A (2)

C13. Infraction type associated with collision

- [ ] Legal (clean) collision (0)
- [ ] Spearing (1)
- [ ] Head to head contact (2)
- [ ] Facemask/cowbow collar (3)
- [ ] Unknown (4)
- [ ] Other (5)

C14. Overall impression of body collision

- [ ] Anticipated (0)
- [ ] Unanticipated (1)
- [ ] Unknown (2)

Additional Comments:

Submit

APPENDIX II: PLAY EXPOSURE LOG

Date:__________________ Time:________ to _________ Opponent:____________

Quarter: 1st 2nd 3rd 4th

O	1											
D	2											
S	3											
	4											

O	1											
D	2											
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O	1											
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S	3											
	4											

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