HEAD ACCELERATION ACROSS YOUTH, HIGH SCHOOL, AND COLLEGIATE SOCCER PLAYERS

by

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A dissertation submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Doctor of Philosophy in Biomechanics and Movement Science

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ABSTRACT

Soccer is the most popular sport worldwide with over 265 million active players. Soccer is unique in that you can use your head to advance the ball and purposeful heading is an integral part of the game. However, some have suggested that repeated heading of the soccer ball is associated with neurological deficits, though others have claimed that deficits are related to multiple head injuries. Still others have observed no neurological deficits. Ultimately, these studies are limited in that they often examine small, homogenous populations. With over 3 million youth soccer players and nearly 1 million high school soccer players competing across the United States each year, more research is needed to determine the risk associated with repeated purposeful heading, particularly among youth and high school athletes. Thus, the purpose of this study was to compare head acceleration during purposeful soccer heading across age and gender, determine what factors predict higher head acceleration values, and investigate acute changes in vestibular/ocular function and postural control with purposeful soccer heading. At the collegiate and high school levels, female soccer players exhibited higher head accelerations than their male counterparts, suggesting that if female soccer players experience a similar number of headers as their male counterparts, females may be exposed to greater cumulative head accelerations from repeated heading of a soccer ball over a career of soccer. Greater neck girth, head-neck segment mass, and neck strength predicted lower peak linear and rotational acceleration and may have contributed to the observed gender differences. On average, soccer players presented with higher sway velocity post-heading compared to control participants, but no other group deficits in postural control or vestibular/ocular function were observed.

Chapter 1

MINIMIZING HEAD ACCELERATION IN SOCCER: A REVIEW OF THE LITERATURE

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1.1 Abstract

Physicians and healthcare professionals are often asked for recommendations on how to keep athletes safe during contact sports such as soccer. With an increase in concussion awareness and concern with repetitive subconcussion, many parents and athletes are interested in mitigating head acceleration in soccer; so we conducted a literature review on factors that affect head acceleration in soccer. We searched electronic databases and reference lists to find studies using the keywords soccer 'OR' football 'AND' head acceleration. Because of a lack of current research in soccer heading biomechanics, this review was limited to 18 original research studies. Low head-neck segment mass predisposes athletes to high head acceleration, but head-neck-torso alignment during heading and follow-through after contact can be used to decrease head acceleration. Additionally, improvements in symmetric neck flexor and extensor strength and neuromuscular neck stiffness can decrease head acceleration. Head-to-head impacts and unanticipated ball contacts result in the highest head

acceleration. Ball contacts at high velocity may also be dangerous. The risk of concussive impacts may be lessened through the use of headgear, but headgear may also cause athletes to play more recklessly because they feel a sense of increased security. Young, but physically-capable, athletes should be taught proper heading technique in a controlled setting using a carefully planned progression of the skill.

1.2 Key Points

- Head-neck-torso alignment and exaggerated follow-through decrease head acceleration during purposeful soccer heading.
- Balanced neck flexor and extensor strength and stiffness decrease head acceleration during purposeful soccer heading.
- Headgear decreases head acceleration in high impact events, but may cause athletes to strike the ball harder and play more dangerously.

1.3 Introduction

Soccer is the most popular sport worldwide with over 265 million participants[1]. While soccer training has broad-ranging positive physiological effects, including improvements in cardiovascular health, blood lipid profile and body composition, musculoskeletal health, and functional capacity[2], soccer exposes participants to purposeful heading of a soccer ball as well as the risk of head injuries. Soccer is unique in that the ball can be directed deliberately and purposely with the head, an act referred to as "heading." Frequent heading of the soccer ball has been associated with depressed neuropsychological performance[3-11], increases in biochemical markers of brain damage[12-15], and structural changes in the brain[11, 16-18]. Others argue that these deficits are associated with unreported and undiagnosed head injuries in soccer players[19, 20]. Still others report *no* deficits in neuropsychological performance[21-28], increases in biochemical markers of brain damage[29] or structural changes in the brain[27]. Considering the estimated 265 million active soccer players, even a small percentage risk of permanent brain injury, through frequent purposeful heading of the soccer ball or via head injuries in soccer, would have serious public health implications.

Concern regarding the risk of purposeful heading and head injuries in soccer has led researchers to suggest restricting the age at which children begin heading a soccer ball[30], regulating the number of headers soccer players are allowed to perform[30], and even reducing athlete-athlete contact in soccer[31]. Recently, the U.S. Soccer Federation announced a new initiative that eliminates heading for children aged 10 and under and limits heading in practice for children between the ages of 11 and 13. However, before a more widespread approach can be implemented we must, learn more about the actual impact of the ball on the head, verify the exposure to heading across all ages, conduct longitudinal studies on soccer players focusing on exposure and injury, and determine the minimum safe age to begin heading the soccer ball[32]. While head injury biomechanics research in soccer is limited, head injury biomechanics research in American Football suggests that greater head accelerations are associated with acute head injury[33-36] and cumulative brain changes[37]. Several studies have investigated head acceleration involving soccer heading in a controlled-laboratory setting or through mathematical modeling[38-51], while only one study has investigated head acceleration during on-field play[52]. Even though these studies include small, homogenous populations, we can begin to determine ways to minimize head acceleration in soccer and thus reduce the risk of acute and cumulative head injury.

Physicians and other healthcare professionals are often asked by both parents and athletes what athletes can do to reduce the risk of head injury in soccer, particularly following return-to-play from concussion. The use of headbands in soccer and neck strengthening exercises have been suggested in popular culture media, and both have some evidence of effectiveness in the literature[39, 44-46, 49-51, 53]. The purpose of this literature review is to summarize what evidence is available for factors influencing head acceleration in soccer. The evidence provided will offer physicians and other healthcare professionals the ability to have well-informed communications with parents, athletes, and coaches with regards to the effectiveness of changes in kinematics, limits to types of contact, increases in muscle activity, changes in ball properties, and uses of headgear in decreasing head acceleration during soccer practice and competition.

1.4 Methods Used in the Review

This review identified articles for inclusion through electronic databases and reference lists of included papers. The following four databases were searched: Ovid MEDLINE, Web of Science, Scopus, and PubMed. Medical Subject Headings (MeSH) and key words including: (soccer 'OR' football) 'AND' head acceleration were used for electronic database searches. Inclusion criteria were as follows: (1) original data; (2) study sample derived from a population of soccer players; (3) studies investigating factors influencing head acceleration during soccer play. Exclusion criteria were as follows: (1) non-English articles; (2) review articles, abstracts-only, commentaries, anecdotal reports, letters, and case studies; (3) studies evaluating biomechanics of soccer using virtual reality simulations; (4) studies only reporting incidence, mechanisms, biomarkers, and neurocognitive outcomes of concussion and heading. The flow diagram of search results is shown in Figure 1. A total of 557 search results were yielded from all database searches; after removal of duplicates and screening out those that did not meet selection criteria, 14 papers were selected. An additional 4 studies were included from reviewing the references of these papers. Ultimately, 18 papers were included for review in the present study.

The studies included for review were separated into six general themes influencing head acceleration in soccer heading: kinematics; type and location of impact; muscle activity, including neck strength and stiffness; sex and anthropometric measurements; ball properties; and headgear.

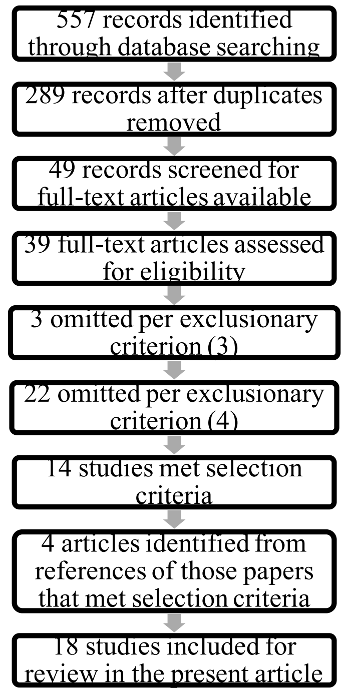

Figure 1: The results of the database search and literature review process. Exclusion criterion (3) omitted studies evaluating biomechanics of soccer using virtual reality simulations. Exclusion criterion (4) omitted studies only reporting incidence, mechanisms, biomarkers, and neurocognitive outcomes of concussion and heading.

1.5 Kinematics of Soccer Heading

Heading a soccer ball is a complex skill in which an athlete's goal is to contact the ball on the forehead at or near the hairline[32]. Athletes may have to perform this skill while standing, walking, running forward or backward, jumping, when challenged by an opponent, or even diving[32]. During a straight-on header, the trunk is first hyperextended and the chin is tucked toward the chest, then as the athlete comes to make contact with the ball the trunk is moved into flexion[32]. This technique varies when the ball is redirected in a direction other than where the ball came from. Discussion of kinematics here refers only to straight-on headers.

A 2005 study by Shewchenko et al. described three phases of soccer heading: pre-impact, ball contact, and follow-through[40]. The pre-impact phase is characterized by a split stance, with the knees bent, the torso extended, and the shoulders square with the oncoming ball. The ball contact phase is described by the torso flexed to meet the ball, and the head and shoulders flexed in-line with the torso. Finally, the follow-through phase includes the torso and head flexed, but decelerating to regain balance. This study examined different variations in head, neck, and torso alignment, specifically the effect that these variations had on head acceleration[40].

During the "normal" condition, in which subjects were asked to perform a straight-on header without any special instructions regarding technique, the rate of change in torso angle relative to the laboratory floor was relatively constant, suggesting that torso activation contributes to heading performance[40]. The rate of change in head angle relative to the laboratory floor was not as consistent as the rate of change in torso angle, suggesting the athlete did not have as much control of head angle as they did of torso angle[40].

During the "follow-through" condition, in which subjects were instructed to increase the forward flexion during deceleration, peak linear acceleration increased by 7% ("follow-through" = 181±23m/s^2, "normal" = 156±2m/s^2) (9.81m/s^2 = 1g), but peak angular acceleration decreased 13% ("follow-through" = 2.41±1.81krad/s^2, "normal" = 1.47±0.27krad/s^2) (1krad/s^2 = 1000rad/s^2) relative to the "normal" condition[40]. These changes in head acceleration and also a change in duration of head acceleration and a change in head velocity, resulted in a 33% decrease ("follow-through" = 498±302W, "normal" = 412±121W) in head impact power (HIP), which is a measure of impact severity that accounts for linear and angular accelerations, duration of acceleration, and velocity[40]. Shewchenko et al. suggested these changes in acceleration and impact severity are because of the increased duration of muscle activation of the sternocleidomastoid and trapezius throughout the ball contact phase[40].

During the “aligned” condition, in which subjects were instructed to align the ball impact with the center of gravity of the head, the ball impact with the longitudinal axis of the cervical spine, and the cervical spine with the thoracic spine, peak linear acceleration increased by 5% (“aligned” = 163±2m/s^2, “normal” = 156±2m/s^2), but peak angular acceleration did not change (“aligned” = 1.45±0.10krad/s^2, “normal” = 1.47±0.27krad/s^2) relative to the “normal” condition[40]. The change in peak linear acceleration and also a change in duration of head acceleration and a change in head velocity, resulted in a 32% decrease (“aligned” = 246±26W, “normal” = 412±121W) in HIP[40]. Shewchenko et al. suggested these changes in acceleration and impact severity were because of the more horizontal torso orientation and the more extended neck orientation when the ball made contact with the head[40].

During the “passive” condition, in which subjects were instructed to perform the act of heading without actually making contact with a ball, peak linear acceleration decreased by 77% (“passive” = 36m/s^2, “normal” = 156±2m/s^2), and peak angular acceleration decreased by 93% (“passive” = 0.12krad/s^2, “normal” = 1.47±0.27krad/s^2) relative to the “normal” condition, suggesting that ball contact had a significant effect on head acceleration[40]. The HIP was not compared because the time at which the peak HIP occurred was much later in the follow-through phase compared with ball impact conditions[40]. Shewchenko et al. also pointed out that neck muscle activity was minimal for the “passive” condition[40].

Clinically, a decrease in peak linear acceleration, peak angular acceleration, and HIP may reduce risk of acute and cumulative head injury. However, in this study the benefits of the heading techniques were opposing each other depending on which injury measure, peak linear acceleration, peak angular acceleration, or HIP, was used, such that increased peak acceleration responses were often paired with reductions in HIP[40]. Because head injury biomechanics research in soccer is limited, it is still unknown which of these head impact severity measures are most significant. For now, it seems more appropriate to view the overall picture of changes in head impact severity. The decrease in HIP during the follow-through and aligned conditions suggests that these alterations in kinematics are beneficial in reducing impact severity[40].

1.6 Type and Location of Impact

The overall goal of heading is redirection of the ball. Depending on the position of the player and the intent of the redirection, the player may apply different strategies to head the ball. For example, headers can be classified based on type (i.e. clears, passes, shots) and approach (i.e. standing, running, jumping). Although the athlete's goal is to contact the ball on the forehead at or near the hairline[32], in a game, it is often the case that the athlete makes contact with the ball in locations other than the forehead, such as on the side of the head, because of misjudgment of ball trajectory or opponent positioning. Because soccer is a contact sport, other impacts beside those from purposeful heading may occur that result in acceleration of the head. For example, two players may collide during play resulting in head-to-head contact or upper-extremity (shoulder, elbow, wrist/hand) to head contact. The type of header, location of head-contact with the ball, and type of impact may affect linear and angular acceleration of the head (Table 1).

1.6.1 Type of header

In 2001, Bauer et al. compared impact forces on the forehead and the involvement of the neck musculature associated with different types of headers in intercollegiate female soccer players[42]. The headers were defined based on three types, a clear, a shot, or a pass, and on two approaches, standing or jumping. A clearing header was defined as one that requires the ball to be projected high into the air over a longer distance[42]. A shooting header was defined as one that must have sufficient speed to elude the goalkeeper, although the ball is not projected over as large a distance[42]. A passing header was defined as one that advances the ball over a small distance towards and open area or a teammate[42]. The maximum impact force did not differ among header types (shooting = 164N, clearing = 164N, passing = 163N) or approaches (jumping = 162N, standing = 165N), although jumping headers required greater muscle activity over a longer activation time[42].

The 2005 study by Shewchenko et al. also compared different types of headers in soccer players[40]. The headers were defined based on target location. A clearing header was defined by a ball target up and away, as far as possible[40]. A controlling header was defined by a ball target 2.75m in front of the player[40]. A passing header was defined by a ball target 5.5m in front of the player[40]. The linear acceleration (clearing = 169±23 m/s^2, controlling = 194±40 m/s^2, passing = 156±2m/s^2), angular acceleration (clearing = 1.52±0.13 $krad/s^2$, controlling = 1.93±0.96 $krad/s^2$, passing = 1.47±0.27$krad/s^2$), and head impact power (HIP) (clearing = 296±82W, controlling = 565±127W, passing = 412±121W) were greater for the controlling header scenario[40]. Shewchenko et al. suggested the controlling response was highest because of the pronounced redirection of the ball down towards the ground near the player[40]. Despite the difference in head acceleration, there were no significant differences in neck muscle activity across different types of headers, though all headers were standing headers[40]. The only observed differences in muscle activity reported by Bauer et al. were between jumping and standing headers. Bauer et al. recorded maximum impact force, instead of head acceleration, and did not include controlling headers, and therefore, did not observe the differences in header type as observed by Shewchenko et al. [39,41].

In 2006, Self et al. compared head accelerations during purposeful headers off a corner kick to head accelerations during purposeful headers off a goal kick[43]. In the corner kick scenario, the athlete was asked to redirect a 12m/s ball 90 degrees from its inflight path[43]. During the goal kick scenario, the athlete was asked to head a 16m/s ball back in the direction from which it came[43]. This was the only study to observe head accelerations during strategical game scenarios, but no significant differences in peak linear acceleration (corner = 29.26g, goal = 32.64g) were observed[43]. Peak angular acceleration was not quantified.

These studies suggest that although different types of headers may result in different head accelerations, the head impact force and neck muscle activity do not vary significantly with different types of purposeful headers.

1.6.2 Location of Impact of the Head

A 2012 study by Hanlon et al. quantified head accelerations during U14 girls' soccer scrimmages[52]. The headers were described by the location of impact with the head, such as left side, right side, top, front, and back. Impacts to the front of the head (n=17) were more frequent than impacts to other locations (top, n=9; right, n=8; left, n=7; back, n=6) [52]. The peak linear acceleration was significantly greater for impacts to the side of the head (left = 27.2±14.4g, right = 28.1±20.8g) than impacts to the back of the head (11.9±5.9g) [52]. The peak angular acceleration was significantly greater for impacts to the front (1657.5±954.5rad/s^2) and side (left = 2586.6±1501.5rad/s^2, right = 3003.4±2823.9rad/s^2) of the head than impacts to the back of the head (723.2±220.3rad/s^2) [52]. However, a small number of headers (n=47) limits the generalizability of this statement[52].

In 2005, Withnall et al. compared head-to-head impacts at two different impact locations using crash test dummy heads[39]. One Hybrid III head-neck system was accelerated into a stationary Hybrid III dummy. Both front boss (that is, outer corner of the forehead at the hairline)-to-side and forehead-to-rear impacts were investigated. There were no significant differences in either linear acceleration (front boss-to-side = 35.1±0.8g, forehead-to-rear = 35.3±0.5g at 1.5m/s) or angular acceleration (front boss-to-side = 2770±24rad/s^2, forehead-to-rear = 1513±19rad/s^2 at 1.5m/s) when the contact occurred at the same speed[39].

These studies suggest that although different locations of impact may result in different head accelerations, the linear and angular acceleration within purposeful headers and within head-to-head contacts are more consistent than across impact types.

1.6.3 Type of Contact

The 2005 study by Hanlon et al. compared types of contact that resulted in head acceleration in a youth soccer population[52]. Events resulting in head acceleration were classified as headers or non-headers. Headers included purposeful contact with the ball. Non-headers included player-to-player collisions, falls, collisions with the goalpost, and unintentional strikes to the head by the ball. There were 47 headers (player-to-ball) and 20 non-headers, of which 40% were player-to-player collisions[52]. There were no significant differences in linear acceleration (header = 4.5-62.9g , non-header = 5.0-56.7g) or angular acceleration (header = 8869.1rad/s^2 (peak), non-header = 497.5-5179.5rad/s^2) between header and non-header events[52]. Unintentional strikes to the head by the ball, which was deemed a non-header event, resulted in greater linear acceleration (32.2±8.8g) and angular acceleration (1246.8±446.0rad/s^2) than purposeful headers[52]. Although the accelerations reported for both header and non-header events were similar to those reported in laboratory studies, no head-to-head contacts were observed.

In 2005, Withnall et al. compared upper extremity-to-head and head-to-head impacts through re-enacting contacts from soccer game videos in a laboratory[38]. Laboratory re-enactment of elbow-to-head and hand/wrist/forearm-to-head was accomplished by volunteer subjects striking an instrumented crash test manikin[38]. Head-to-head impacts comprised two scenarios: the front boss-to-side and forehead-to-rear[38]. There was no significant difference between elbow-to-head (linear = 21.3±10.1g, angular = 891.3±247.1rad/s^2, HIP = 1.1±0.8kW) or hand/wrist/forearm-to-head impacts (linear = 20.4±7.7g, angular = 1445.0±636.6rad/s^2, HIP = 0.6±0.4kW) [38]. Moreover, the data suggested that a linear or angular acceleration of a sufficient level to achieve a 5% risk of concussion, according to the HIP concussion risk percentage, would occur roughly three times in 100 from elbow contact, and less than once in 100 from hand/wrist/forearm contact[38]. However, head-to-head contacts resulted in significantly greater head acceleration (linear = 35.2g, angular =

$2141 rad/s^2$, HIP = 1.7kW at 1.5m/s; linear = 82.9g, angular = $5066 rad/s^2$, HIP = 6.4kW at 3.0m/s) than either upper extremity contact and, in fact, a front boss-to-side impact at 3m/s results in a 67% risk of concussion, according to the HIP concussion risk percentage[38].

These studies suggest that head-to-head contacts and unintentional head impacts result in the highest acceleration of the head, and therefore, should be minimized. Although contact during play is inevitable, we suggest that soccer players wait until they have mastered proper heading technique, including the ability to protect themselves in the air from opponents, before being allowed to head the ball during the game. Creating space, by using the arms during heading duals, may decrease the occurrence of head-to-head impacts. Although this will likely cause an increase in upper extremity contact, it appears the upper extremity contact results in significantly lower head accelerations than head-to-head contact. These studies also support head injury data, in that player-to-player contact is more likely to result in concussion than player-to-ball[54-63]. The greatest risk of player-to-ball contacts occur during unanticipated contacts. Player-to-ball contact, regardless of the type or approach, is unlikely to cause concussion[40, 42, 43], but there is little scientific evidence regarding the subconcussive cumulative effects of frequent purposeful heading in soccer.

Table 1: Relationship between type and location of impact and head acceleration.

Study	Contact type	Objective	Population	Methods	Results

Hanlon and Bir 2012 [52]	47 headers, 20 non-headers	To describe number of impacts and head accelerations based on location of impact and type of impact event	24 U14 girls' soccer players	Linear and angular acceleration recorded using the HITS system during 6 (30-65min) scrimmages	Impacts to the side of the head had significantly greater linear and angular acceleration than impacts to the back of the head Unintentional impacts from the ball resulted in greater head acceleration compared to impacts received from heading There was no significant difference between header and non-header events
Withnall et al. 2005 [38]	Elbow-to-head and hand/wrist/forearm-to-head Head-to-head impacts (4 configurations)	To measure head accelerations induced from upper extremity-to-head and head-to-head impacts	5 soccer players and two instrumented Hybrid III crash test manikins	Upper extremity-to-head impacts and head-to-head impacts were re-enacted in the laboratory	Elbow-to-head and hand/wrist/forearm strikes resulted in low risk of concussion Head-to-head contacts posed a high risk of concussion
Withnall et al. 2005 [39]	Head-to-head impacts (4 test speeds at 2 test sites)	To determine if soccer headgear has an effect on head impact response	Two instrumented Hybrid III head forms	One Hybrid III head-neck system was accelerated into contact with a stationary Hybrid III dummy	At 4m/s there was over a 50% risk of concussion, and at 5m/s there was virtually certain risk of concussion There was no significant difference between impact test sites

Self et al. 2006 [43]	Goal kick simulations and corner kick simulations	To determine if differences in head acceleration exist between goal kicks and corner kicks	US Air Force Academy Intercollegiate male soccer players	Ear plugs with a triaxial accelerometer and gyroscope were used during simulated practice and game events	There was no significant difference in linear acceleration between goal kicks and corner kicks
Bauer et al. 2001 [42]	Shooting, clearing, and passing using standing and jumping approaches	To determine if header type or approach influenced head impact force or neck muscle activity	15 Division-I NCAA female soccer players	Impact forces measured by a 15-sensor pressure array secured to the forehead	Maximum impact force and impulse did not differ among header types or approaches. Increased neck muscle activity was observed during jumping headers

U14 – refers to a soccer league in which all participants are 14 years and under
NCAA – National Collegiate Athletic Association
HITS – Head impact telemetry system

1.7 Muscle Activity

1.7.1 Timing and Duration of Muscle Onset

During a straight-standing header, neck flexors and neck extensors contract to brace for ball impact. The right and left sternocleidomastoid and upper trapezius muscles are often chosen to represent neck flexor and neck extensor activity, respectively, because these muscles are the most superficial in the neck region and can be studied using surface electromyography (EMG). The right and left sternocleidomastoid are activated 280-500ms before ball contact and are deactivated around the time of ball contact with peak activity occurring 90-150ms before ball contact[40, 42]. The right and left upper trapezius muscles are activated 100-300ms before ball contact and remain active throughout the follow-through[40, 42]. The time of muscle onset, the duration of muscle activity, and the level of muscle activity have all been suggested to contribute to head acceleration during purposeful soccer heading.

In their 2005 landmark study, Shewchenko et al. reported the effects of varied muscle activity during soccer heading[40]. In this study, athletes were asked to complete a straight-standing header using each of three neck muscle activity scenarios: normal, pre-tensed, and relaxed. During the normal condition, participants were not provided with any special instructions. They were asked to head the soccer ball as they would normally. During the pre-tensed condition, participants were asked to maximally tense their neck throughout the trial[40]. During the relaxed condition, participants were asked to not actively contribute to the impact[40]. Two ball speeds were used: a high speed (8m/s) and a low speed (6m/s). At high speeds, peak linear (pre-tensing = $171\pm20 m/s^2$, relaxed = $175\pm8 m/s^2$, normal = $158\pm16 m/s^2$) and peak angular (pre-tensing = $1.54\pm0.21 krad/s^2$, relaxed = $1.64\pm0.17 krad/s^2$, normal = $1.46\pm0.36 krad/s^2$) accelerations increased relative to the normal condition with both pre-tensing and with relaxation[40]. At low speeds, peak linear (pre-tensing = $153\pm0 m/s^2$, normal = $156\pm2 m/s^2$) and peak angular (pre-tensing = $1.40\pm0.12 krad/s^2$, normal = $1.47\pm0.27 krad/s^2$) acceleration decreased relative to the normal condition with pre-tensing [40]. During the pre-tensed condition, the right and left sternocleidomastoid muscles were active for a longer period of time, but the activity levels were lower than levels during the normal condition[40]. During the pre-tensed condition, additional mass is recruited from the torso, raising the effective mass of the system during head impact, and thus lowering the head acceleration. During the relaxed condition, both sternocleidomastoid and upper trapezius activity levels were lower than normal, as expected[40]. Observed increases in peak linear and peak angular accelerations at high speed suggest that during game play when ball speeds are greater, pre-tensing may not decrease head acceleration. However, observed decreases in acceleration at low speed suggest that pre-tensing maybe useful when initially teaching youth to head the soccer ball. When teaching an athlete to head the soccer ball, balls are often tossed at a low velocity from a close distance by a parent, coach, or teammate. It has even been advocated that the use of soft ("Nerf" style) balls or volleyballs be used to teach the skill to novice players. In this situation, the athlete must learn to actively head the ball, which is to *actively* strike the ball and not let the ball hit them in the head. Teaching novice soccer players to tense their neck muscles early (pre-tensing), may result in a decreased acceleration of the head during this early-learning period.

In a 2008 study by Tierney et al., muscle activity of the sternocleidomastoid and upper trapezius were reported with regards to their effect on head acceleration during purposeful soccer headers[45]. Male and female collegiate soccer players performed a series of straight-standing headers. Muscle activity was divided into preactivity (muscle activity occurring before ball contact) and reactivity (muscle activity occurring as a result of ball contact)[45]. The sternocleidomastoid preactivity was greater than reactivity in both males and females while the upper trapezius preactivity was similar to reactivity in both sexes[45]. As a result there were no sex differences in neck muscle activation strategies. However, when comparing EMG peak and area, there was a significant sex difference in muscle activity, but individual comparisons for EMG peak and area did not indicate where significant differences existed[45].

Conversely, in another study by Tierney et al., differences in EMG peak and area were reported in male and female soccer players[64]. In an attempt to simulate forces associated with purposeful soccer heading, an external force applicator was used to apply an external force to the head-neck segment. The athlete remained seated while a pulley with a weight attached caused either forced flexion or forced extension (Figure 2)[64]. Females exhibited 79% more peak muscle activity and 117% more muscle activity area compared with males, but had a 70% greater angular acceleration than males[64]. Females also exhibited muscle onset latency 29% faster than males for the sternocleidomastoid and 9% faster than males for the upper trapezius[64]. Thus, although individual comparisons for EMG peak and area did not indicate significant differences when heading a soccer ball[45], the sex differences in head acceleration are likely not attributable to sex differences in neck muscle activity; instead the EMG data presented by Tierney et al. suggested that females had significantly greater muscle activity than males and a faster onset latency time[64].

Figure 2: Forced extension trial setup with external force applicator. Reproduced from Mansell et al. [53] with permission.

1.7.2 Neck Strength

Some have suggested that it is neck strength, and not neck muscle activity that best predicts head acceleration during purposeful soccer heading. Previous research involving both adolescent[46] and collegiate[45, 64] populations demonstrates that higher neck strength predicted lower head acceleration during purposeful soccer heading. However, in another study by Mansell et al., strengthening the neck did not decrease head acceleration[53]. Mansell et al. used an 8-week traditional resistance training program, which included isotonic resistance training twice per week for 8 weeks[53]. Despite a 15% increase in neck flexor strength and a 22.5% increase in neck extensor strength across female participants, there was no decrease in head acceleration (pre = 1127.5±574.5°/s, post = 1254.1±544.7°/s in the known condition) [53]. A similar increase in neck flexor strength was observed across male participants, but there was not a significant increase in neck extensor strength across male participants[53]. The authors suggested that the isotonic resistance training may change muscle structure, but the decrease in head acceleration may be related instead to neuromuscular control[53]. This will be discussed in greater detail in Section 5.3. Neuromuscular Neck Stiffness.

Work by Dezman et al. also reported no significant relationship between pure flexor/ extensor strength and head acceleration[51]. Instead, there was a significant relationship between the neck flexor/extensor strength difference (i.e. flexion strength – extension strength) (-11.8±24.9N for men, -7.0±11.9N for women) and head acceleration[51]. The mean neck strength difference was positively correlated with head acceleration, such that a larger difference in neck flexor/extensor strength resulted in a higher head acceleration[51]. This study suggests that it is the balance of flexor and extensor strength that results in lower head acceleration[51]. In the study by Mansell et al., there was an unequal increase in neck flexor and neck extensor strength in both male and female participants following the resistance training program[53]. This may be one of the reasons that there were no observed decreases in head acceleration, despite increases in absolute neck strength. It is possible that had the investigators noticed an equal increase in both muscle groups, a decrease in head acceleration may have been observed. While neck strength is only one potential factor influencing head acceleration during purposeful soccer heading, research, although mixed, suggests that increasing neck strength may decrease head acceleration. It is still unclear what cervical resistance training programs are most beneficial in decreasing head acceleration, however.

1.7.3 Neuromuscular Neck stiffness

Increased stiffness, or resistance to movement from external perturbation, may enhance an athlete's ability to absorb external forces by allowing for a more even distribution of energy absorption[42, 65, 66]. For example, during purposeful soccer heading, neck muscle contraction may increase muscle and joint stiffness, and therefore allow for distribution of energy absorption through the head and torso[48, 67-69]. This more even distribution of energy increases the relative mass of the player, extends the time of contact with the ball, and decreases head acceleration when the ball makes contact. Previous research reports sex differences in stiffness, which were attributed to the amount of muscle tissue, such that more tissue corresponds to greater joint resistance to motion[65, 66, 70, 71]. In one study by Tierney et al., female athletes exhibited lower head-neck segment stiffness, and exhibited higher head acceleration (females = 1503.9±516.5°/s, males = 995.2±368.2°/s in the known forced flexion condition) as compared to their male counterparts, suggesting that head-neck segment stiffness was useful in predicting head acceleration[64]. Mansell et al. studied resistance training and head-neck dynamic stabilization[53]. In this investigation, a traditional isotonic resistance training program was used because the authors stated that this is the most accepted method for training the head-neck segment[53]. However, there was no significant increase in head-neck segment stiffness and no associated decrease in head acceleration following this resistance training program[53]. Therefore, the authors suggested implementing ballistic activities, such as plyometrics, which have been reported to enhance neuromuscular control and dynamic stabilization at other joints[53]. Similarly in mixed martial arts, ballistic neck strengthening activities, such as physioball neck leans and prone cobras, have been used to improve neuromuscular control[72]. These same activities may prove useful in athletes training for other sports, such as soccer, as well.

1.8 Sex and Anthropometric Measurements

The rate of concussions in high school girl's and collegiate women's soccer is nearly double that of high school boy's and collegiate men's soccer [59, 73-75]. Many laboratory heading studies have restricted participation to male-only[38-40, 43], or female-only populations[42, 46, 52]. However, a few[45, 51] have made comparisons between sexes under the same conditions. In a 2013 study by Dezman et al., both male and female soccer players were served a ball by an investigator from a distance of 3 meters mimicking a soccer heading practice scenario of low ball velocity[51]. In this study, there were no significant differences in linear (female = $101.75 \pm 64.92 m/s^2$, male = $101.70 \pm 34.30 m/s^2$) or angular (female = $901.00 \pm 459.54 rad/s^2$, male = $854.86 \pm 489.14 rad/s^2$) head accelerations between male and female participants[51]. Similarly, in a 2008 study by Tierney et al., male and female soccer players were served a ball by a JUGS soccer machine at 9.83m/s (22mph) from 11m away[45]. They too reported no significant differences in linear head acceleration between male and female participants (female = 20.16±4.12g, male = 18.25±4.48g), although head acceleration was 10% greater in the females[45]. The 9.83m/s ball velocity is still slower than that most athletes experience during games. Moreover, Dorminy et al. reported no significant difference between groups at 13.4, 17.9, or 22.4m/s, which is closer to ball velocities that athletes experience during games[29]. So, in a controlled setting, it seems that male and female athletes have similar head acceleration during purposeful soccer headers. This suggests that performing heading drills, such as when a parent, coach, or teammate tosses the ball to a soccer player, may not result in higher head accelerations regardless of sex.

Previous research demonstrates that head-neck mass is inversely correlated with head acceleration, such that a higher head-neck mass predicts lower head acceleration[45, 64]. With this in mind, a contemporary research report has suggested restricting the age at which children begin heading a soccer ball[30]. Based on Newton's second law, force = mass * acceleration, the smaller the mass, the greater the acceleration. Queen et al. designed a theoretical model based on Hertz contact theory to calculate linear and angular head acceleration during purposeful soccer heading[48]. Inputs to this model included head mass, ball size, inflation pressure and ball velocity. Variations in head mass included the 3rd percentile, 50th percentile, and 98th percentile child for age groups 6-9, 10-13, and 14-18 years. Outputs from the model included peak impact force, linear and angular acceleration of the head, and contact time. Greater head mass resulted in decreased head acceleration, but had little effect on peak impact force or contact time[48]. Therefore, children with smaller absolute head masses, may be at risk for greater head acceleration. Additionally, relative to body size, children have larger head masses than adults. This creates a theoretical bobble-head effect, when the neck strength is not great enough to control the mass of the head. Therefore, we suggest anthropometric measures, such as head mass, should be considered when determining the minimum safe age to begin heading a soccer ball.

1.9 Ball properties

The Fédération Internationale de Football Association (FIFA), through the International Football Association Board, the National Collegiate Athletic Association (NCAA), the National Federation of State High School Associations (NFHS), and the United States Youth Soccer Association define ball regulations for international, collegiate, high school, and youth game play, respectively[76-79]. According to FIFA guidelines, the ball is spherical, made of leather or other suitable material, of a circumference between 68 cm (27 in) and 70 cm (28 in), of a weight between 410 g (14 oz) and 450 g (16 oz), and of a pressure between 0.6 atm (8.8 psi) and 1.1 atm (16.2 psi)[79]. The NCAA and the NFHS use the same soccer ball regulations as defined by FIFA[77, 78]. For soccer players over 12 years of age, the US Youth Soccer Association uses the same soccer ball regulations as defined by FIFA[76]. For soccer players under 8 years of age, the US Youth Soccer Association suggests a size 3 soccer ball, of a circumference between 23 in and 24 in, of a weight between 11 oz and 12 oz, and of a pressure suggested by the manufacturer[76]. For players 8-12 years of age, the US Youth Soccer Association suggests a size 4 soccer ball, of a circumference between 25 in and 26 in, of a weight between 11 oz and 13 oz, and of a pressure suggested by the manufacturer[76].

Variations in ball size and ball pressure may affect head acceleration during purposeful soccer heading (Table 2)[41, 44, 48]. For example, in the numerical model created by Queen et al., inputs included 3, 4, and 5 soccer ball sizes and associated ball inflation pressures of 10, 12, and 14 psi respectively[48]. Greater ball size resulted in greater contact time, but had little effect on peak impact force, or linear and angular acceleration of the head[48]. Change in ball pressure had no effect on peak impact force, linear and angular head acceleration, or contact time[48]. In another study by Shewchenko et al., a numerical model, again, was used to study ball impact characteristics[41]. The inputs to this model included ball mass (290g and 350g) and ball pressure (5.8 psi, 8.7 psi, 11.6 psi, and 16.0 psi). A decrease in ball mass resulted in a decrease in head acceleration (130m/s^2, 140 m/s^2, respectively) and HIP (1.21kW, 1.30kW, respectively)[41]. Similarly, a decrease in ball pressure resulted in a decrease in head acceleration (107m/s^2, 150 m/s^2, 156 m/s^2, 170 m/s^2, respectively) and HIP (1.04kW, 1.38kW, 1.44kW, 1.53kW, respectively); however, this was only observed significant for changes in ball pressures greater than 25%[41]. Therefore, smaller changes in pressure, as examined by Queen et al., may not result in decreases in head acceleration.

FIFA, the NCAA, the NFHS, and US Youth Soccer Association set regulations for ball size, ball weight, and ball pressure. While ball size had little effect on acceleration of the head, a decrease in ball mass did result in a decrease in head acceleration. Therefore, when teaching an athlete how to properly head a soccer ball, we suggest that the athlete be encouraged to use a light-weight (soft) ball to allow improvements in technique, while minimizing head acceleration during these early-learning stages. The guidelines for inflation pressures allow for balls between 8.8 psi and 16.2 psi. In the Shewchenko et al. study, a 50% decrease in inflation pressure resulted in a 31% decrease in head acceleration[41]. Therefore, the nearly 100% increase in inflation pressure from 8.8 psi to 16.2 psi may result in a significant increase in head acceleration. Thus, FIFA, the NCAA, the NFHS, and the US Youth Soccer Association should consider narrowing this range of acceptable pressures moving forward. Additionally, coaches and officials should inflate balls to pressures on the lower end of the acceptable range during matches/practices involving younger, inexperienced soccer players.

Ball speed varies based on age, sex, and level of experience. For experienced adult male players, mean maximum ball speed has been reported to be in the range 20-30 m/s (44-67 mph)[80]. For children 8-14 years of age, mean maximum ball speed has been reported to be in the range 12-15.5 m/s (27-35 mph)[80]. In most human-subject research studies, ball speeds used during testing were far below ball speeds reported during game play[42, 43, 45, 46, 51]. However, head acceleration significantly increases with an increase in ball velocity[44, 47]. In contrast, the one study to date that investigated head acceleration at ball speeds closer to those reported during game play, showed a non-significant increase in head acceleration with ball speeds at 13.4m/s (34.7g), 17.9m/s (49.2g), 22.4m/s (50.7g)[29]. This suggests that at lower velocities, an increase in ball speed results in increased head acceleration, but this difference in head acceleration is not as pronounced among higher ball speeds that may occur during game play. So, for youth soccer players learning purposeful heading techniques, it is important to project the ball to the athlete at lower ball speeds, such as when a parent, coach, or teammate tosses the ball from a close distance.

Table 2: Characteristics of ball properties and their relationship to head acceleration.

Study	Ball velocity (m/s)	Ball size	Ball pressure (psi)	Head acceleration (g)	Results
Naunheim et al. 2003 [44]	9 12 15 9 12 15	5	6 9	15.1 18.7 28.7 15.1 21.3 30.4	Lower pressure balls have decreased head acceleration at high speeds
Funk et al. 2011 [47]	5 8.5 10 11.5	5	8	6.8 15.0 18.0 21.0	Head acceleration increases with increasing ball speed
Queen et al. 2003 [48]	0.1-30 in 0.1 increments	3, 4, 5	10, 12, 14	Variable	Increase in ball size increases contact time and HIC but has little effect on impact force and acceleration Increase in pressure has little effect on contact time, HIC, impact force, or acceleration
Shewchenko et al. 2005 [41]	6-7	3,4,5	5.8, 8.7, 11.6, 16.0	Variable	Linear acceleration decreased by 10% with a decreased ball mass of 20-32% Linear acceleration decreased by 10% with a decreased ball pressure of 50%

					Little reduction noted for ball pressure reductions less than 25%
Dorminy et al. 2015 [29]	13.4 17.9 22.4	5	8	34.7 49.2 50.7	There was a non-significant increase in head impact acceleration from the 13.4m/s to 22.4m/s ball speed

HIC – Head Injury Criterion

1.10 Headgear

The American Society for Testing and Materials (ASTM) International has adopted a product performance standard for soccer headgear[81]. However, the ASTM standard does not address head-to-ball impacts, rather head-to-hard surface impacts such as head-to-head, head-to-ground and head-to-goalpost, which are believed to be the primary mechanism involved with most soccer concussions[54-63]. There are several companies that have designed soccer headgear that meet ASTM standards, such as the Hat Trick (djOrthopedics), the Premier-A (Full90 Sports), the Forcefield FF Protective Sweatband (Forcefield FF (NA) Ltd.), and the Halo (Unequal Technologies). These headgears have an impact-absorbing foam layer, that is purported to mitigate head-to-hard surface impacts, but not head-to-ball impacts[39, 44, 50] (Table 3). During impacts with the ball, the ball deforms more than the head or the headgear, which suggests that with headgear, there is no significant decrease in head acceleration[39]. For headgear to be effective during head-to-ball impacts, the headgear would need to be at least as compliant as the ball[44]. In fact, among the female cohort, Tierney et al. described an increase in head acceleration with headgear, which the author attributes to an increase in head mass for which neck strength is not able to compensate[45]. The female cohort had longer head-neck segments compared to the male cohort, creating a "Cleopatra effect," in which the long neck coupled with decreased neck strength makes it difficult to stabilize the head. The increased mass of the headgear exaggerated the effect, and resulted in increased head acceleration[45].

Conversely, in a 2008 study by Delaney et al., youth athletes volunteered to wear headgear during practice and game play[49]. Delaney et al. suggested that soccer headgear decreased the risk of sustaining a concussion because those athletes who wore headgear experienced a lower rate of incidence of concussion[49]. However, there were limitations to the study, such as self-reported concussions from symptom checklists and a selection bias from athletes willing to wear headgear. Headgear does afford protection in direct head-to-head impacts, whereby linear acceleration and HIP are decreased approximately 33% (i.e. 81.4g barehead at 3m/s vs. 55.2g barehead at 3m/s). Ultimately, headgear may decrease acceleration for harder impacts, such as in head-to-head impacts, but a study on rugby players has suggested that headgear gives a false sense of security and causes athletes to play more recklessly[82].

Table 3: The effectiveness of soccer headgear in decreasing head acceleration.

Study	Headgear	Objective	Population	Methods	Results

Withnall et al. 2005 [39]	Control, Head Blast Soccer Band, Full 90 Select Performance Headguard, and Kangaroo Soccer Headgear)	To determine if soccer headgear has an effect on head impact responses	A human volunteer and a surrogate head-neck system	Linear and angular acceleration was measured during ball impacts and head-to-head impacts	Headgear did not reduce head acceleration during ball impacts, but did reduce head acceleration during head-to-head impacts
Tierney et al. 2008 [45]	Control, Head Blast Soccer Band, Full 90 Select Performance Headguard	To determine the effects of sex on head impact kinematics and dynamic stabilization during soccer heading	44 college-aged soccer players (29 women, 14 men)	Linear acceleration was measured using an accelerometer-instrumented mouth piece	Headgear effectively reduced head acceleration for the men only
Naunheim et al. 2003 [44]	Control, Soccer Docs Headgear, Kangaroo Soccer Headgear, Head Blast Soccer Band, Headers Headband	To compare the effectiveness of 4 types of headgear in reducing head acceleration	Instrumented head form	Peak linear acceleration was measured with and without the headgear at 3 speeds: 9, 12, and 15m/s	Headbands may decrease head acceleration at the highest speeds, when more forceful blows occur
Delaney et al. 2008 [49]	Various self-selected headgear brands	To compare the number of concussions experienced in a group with and a group without headgear	Youth soccer players (ages 12-17 years)	Athletes self-reported symptoms of concussion	The use of headgear may decrease the risk of concussion
Broglio et al. 2003 [50]	Head Blast Soccer Band, Headers Headband, Protectors Headband	To compare the effectiveness of three types of headgear in reducing impact from a linear blow by a soccer ball	A mounted force platform	Vertical ground reaction force was measured. Peak force, time to peak force, and impulse were calculated	All 3 headbands were effective at reducing the peak impact force

1.11 Conclusion

Physicians are often asked questions on how to minimize the risk of both acute and chronic head injury in soccer. The first step in minimizing the risk of injury is decreasing linear and angular acceleration. This review suggested several ways to decrease linear and angular acceleration, which may be especially important in a youth population or a population of athletes returning to soccer competition from head injury. These suggestions include improving head-neck-torso alignment, exaggerating follow-through, avoiding head-to-head impacts and unintentional ball-to-head impacts, increasing neck flexor and neck extensor strength equally, and enhancing neuromuscular control. Younger athletes, particularly those with a greater head-to-body ratio, should avoid heading the soccer ball altogether. Both their low absolute head mass and their lack of neck strength put these athletes at risk for brain injury. Athletes who desire to head the soccer ball, in spite of risk factors associated with their size should practice purposeful heading technique with light-weight balls, with minimal pressure, and should have someone toss them the soccer ball at a low velocity. Finally, although headgear may decrease impacts for more forceful blows (head-to-head impacts), headgear may also cause athletes to strike the ball harder and play more dangerously because they feel a sense of increased security.

This review highlighted ways to minimize head acceleration in soccer. We have presented suggestions for decreasing head acceleration during purposeful soccer heading. However, there may be unintended adverse effects from some of these recommendations (i.e. increasing neck strength, may lead to more forceful heading, which may increase head-to-head injuries because heads are colliding harder). Ultimately, a comprehensive study on head impact biomechanics across age and sex is needed. This study would allow us to compare each of the predictors described in one study to determine which has the greatest effect on minimizing head acceleration during purposeful soccer heading. From this study, we can apply changes to soccer heading recommendations, and evaluate effectiveness in these suggestions. However, until this study is complete, we recommend that young, but physically-capable, athletes be taught proper heading technique in a controlled setting using a light-weight ball.

Chapter 2

HEAD ACCELERATION ACROSS AGE AND GENDER

(Targeted for submission to *Sports Health: A Multidisciplinary Approach*)

2.1 Introduction

Soccer is the most popular sport worldwide with over 265 million active players [1], and one of the fastest growing sports in the United States, specifically among female athletes. In 2014-2015, 26,995 women played soccer across 1030 NCAA collegiate soccer teams, a 1355% increase in participation from 1981-1982 (1,855 women's soccer players across 80 teams) [83]. Similarly, participation in soccer is increasing at the youth level. In 1980-1981, 810,793 boys and girls were registered with US Youth Soccer [84]. Today, 3,055,148 boys and girls are registered with US Youth Soccer, a 277% increase [84]. Although participation in soccer has rapidly increased over the past 25 years, there has been concern about long-term effects associated with repeated heading of the soccer ball, particularly among young athletes [20, 32, 85-87]. However, the long-term effects of repeated heading of the soccer ball are still largely unknown with some studies claiming neurological deficits with frequent heading [3, 5-13, 16-18], and others reporting no impairments among soccer players [14, 21, 22, 24-28, 46, 88-93]. The first step in understanding the effects of repeated heading of the soccer ball is learning more about the actual impact of a ball on the head, for example, comparing heading kinetics across ages and genders to begin to describe differences in head impacts under the same laboratory conditions in these groups.

Groups that sustain higher magnitude repetitive head impacts may be at increased risk for neurological deficits [37, 94-97]. Greater cumulative exposure to repetitive head impacts has been associated with vestibular/ocular function deficits [94], cognitive decline [95, 96], microstructural white matter brain changes [37], and later life cognitive, behavioral, and mood impairment [97]. These impairments may not be related to cumulative exposure over one collegiate season [98, 99], but perhaps may manifest over a career. Moreover, early exposure to higher magnitude repetitive head impacts may have adverse effects [37, 100-102], although the evidence is mixed [103]. Cortical and subcortical structures reach peak levels by the age of 12, and the rate of myelination and cerebral blood flow are highest between ages 10 to 12, but neurogenesis and myelinogenesis continue well beyond the first decade of life [104-113]. Cumulative exposure to repetitive head impacts during this critical neural development period during adolescences may cause the long-term deficits observed in some individuals [37, 100-102]. Together, these studies suggest that greater cumulative linear and rotational acceleration in youth and across many years of participation may be related to structural and functional changes in the brain.

In soccer, linear head acceleration has been measured during purposeful soccer heading [40, 68, 114]. Purposeful heading involves a ~20ms impact, during which a soccer ball contacts a player's head and rebounds [68]. During ball contact, there is a peak in linear head acceleration, and during ball rebound there is also a peak in linear head acceleration [40, 114]. When resultant head accelerations are calculated, linear head accelerations appear as dual peaks [40]. In a male collegiate population, it has been reported that the second peak results in the greatest acceleration [40], suggesting that it is the forward movement during the rebound phase of ball contact that results in the greatest head acceleration [40, 114]. However, the head acceleration profile when heading a soccer ball is unknown in a youth population or in a female population who may not have the neck musculature to control the ball during impact and rebound with the same force as an adult population.

While previous studies have quantified head acceleration in collegiate and post-collegiate soccer players in a laboratory setting [40, 45, 53, 115], and others have investigated head acceleration in youth soccer scrimmages and tournaments [52, 116], it is unknown how head acceleration differs in youth, high school, and collegiate soccer players under the same conditions. Thus, the purpose of this study was to compare head acceleration across youth (12-14 years old), high school (15-18 years old), and collegiate (18-24 years old) male and female soccer players. We expected that head accelerations among youth athletes would be higher than among high school and collegiate athletes. Additionally, head accelerations among female athletes would be higher than among their male counterparts. Based on previous research [40], we hypothesized that the second peak would be higher. Greater head acceleration may suggest imperfect heading technique or size and strength differences among groups, and may indicate populations that are exposed to higher cumulative accelerations with repeated heading of a soccer ball, assuming the same number of headers.

2.2 Methods

2.2.1 Participants

One-hundred soccer players were recruited to participate in this study (Table 4). Potential participants were excluded from this investigation if they had a self-reported history of neurologic disorder, cervical spine injury, or head injury (i.e. concussion) in the past 6 months. If a participant sustained a concussion or any other injury in the past, they must have been cleared by a physician to return to full activity before participating in this study. Potential participants were included if they were currently participating in soccer within their age group and gender. Goal keepers were excluded. All participants provided written informed consent (IRB 476493-4). Children under the age of 18 years provided written informed assent, and their parent/legal guardian provided written informed parental consent.

Table 4: Subject Demographics. Mean±Standard Deviation.

Group	N	Age	Height	Weight
Collegiate Male	20	20.8±1.3[a]	181.5±5.8[a]	77.3±6.6[a]
Collegiate Female	21	19.3±1.1	166.8±4.5	59.3±3.4
High School Male	14	16.9±1.0	177.5±7.3[a]	68.9±6.7[a]
High School Female	19	16.7±0.9	163.0±7.1	58.7±6.7
Youth Male	8	13.0±0.9	166.0±14.3[a]	51.6±13.6
Youth Female	18	12.8±0.9	156.0±6.9	48.1±8.4

[a] Male cohort statistically greater than female cohort within age group ($p<0.05$).

2.2.2 Instrumentation

2.2.2.1 Motion Analysis System

The local coordinate systems of the head and the Smart Impact Monitor (SIM) (Triax Technologies Inc., Norwalk, CT) were determined using an 8-camera Motion Analysis Motion Capture System (Motion Analysis Corporation, Santa Rosa, CA). To record movement of the head and torso, 11 reflective markers (14mm) were attached to anatomic landmarks by Velcro™ tape (VELCRO USA, Manchester, NH) (Figure 3): the suprasternal notch, the xyphoid process, the 7th cervical vertebrae, the 10th thoracic vertebrae, the left and right temple, the left and right external auditory meatus (EAM), the back of the head, just below the inion (the location of the SIM), and the left inferior orbital rim. This marker set is consistent with that of Dezman et al. with additional markers on the left and right EAM and left inferior orbital rim to define local coordinate systems of the head and SIM [51, 117]. The left and right EAM and inferior orbital rim markers were used to define the local coordinate system of the head. The left and right EAM and back of the head makers were used to define the local coordinate system of the SIM. Data were recorded at 100Hz with a 1/1000 shutter speed.

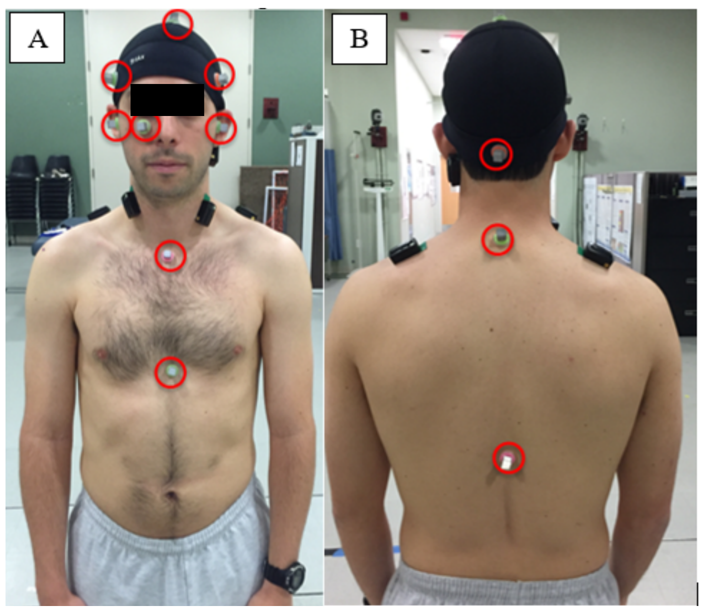

Figure 3: Marker set used to record motion of the head and torso.

2.2.2.2 Head Acceleration

The SIM was used throughout testing to quantify head acceleration. The SIM is a head accelerometer that contains a triaxial accelerometer and gyroscope. Signals from the accelerometer and gyroscope were sampled at 1000Hz. The SIM has a threshold level of 8g, meaning that when the accelerometer registered a reading of 8g or higher, information 10ms before impact and 52ms after impact were wirelessly transmitted to the computer. The SIM was secured to the back of the head, just above the greater occipital protuberance, using a custom tight-fitting elastic cap (Figure 3). Karton et al., observed no significant difference in peak linear or rotational acceleration between SIM and instrumented headform at 30g and 50g impact energy levels (Karton, 2016, in press). At 80g, there was a significant difference in peak linear and rotational acceleration between SIM and instrumented headform, such that the SIM readings showed, on average, higher peak linear and rotational accelerations than those of the headform (Karton, 2016, in press). Furthermore, correlation coefficient results showed that a strong positive relationship exists between the headform and SIM peak linear and rotational accelerations with Pearson's r>0.9 (Karton, 2016, in press).

Raw, unprocessed data from the accelerometer and the gyroscope were obtained from the manufacturer. Data in the SIM coordinate system were rotated to that of the head coordinate system using 3D motion capture data for sensor position and head position [118]. The coordinate system of the head was defined using the midpoint of the EAMs as the origin. The positive x-axis pointed anteriorly through the inferior orbital rim along the Frankfort plane. The positive z-axis pointed superiorly, perpendicular to the Frankfort plane, and the positive y-axis pointed to the left. The coordinate system of the SIM was defined using the location of the SIM as the origin. The y-axis of the triaxial accelerometer within the SIM was aligned with the positive y-axis of the head coordinate system. The positive z-axis pointed backward and downward from the origin of the head to the location of the SIM. The positive x-axis pointed backward and upward, perpendicular to both the y-axis and z-axis (Figure 4).

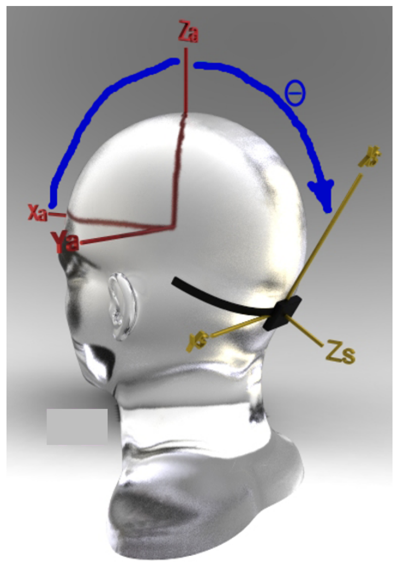

Figure 4: The coordinate systems for the head and for the SIM.

Linear accelerations of the head in the head coordinate system were transformed to the head center of gravity (CG) using standard dynamics equations applied to a rigid body

$$\overrightarrow{\alpha_{CG}} = \overrightarrow{\alpha_{SIM}} + \vec{\ddot{\theta}} x \vec{d} + \vec{\dot{\theta}} x (\vec{\dot{\theta}} x \vec{d})$$

where $\overrightarrow{\alpha_{CG}}$ was the linear acceleration of the head CG, $\overrightarrow{\alpha_{SIM}}$ was the linear acceleration of the SIM, $\vec{\ddot{\theta}}$ was the rotational acceleration of the head, $\vec{\dot{\theta}}$ was the rotational velocity of the head, and $\vec{d}$ was the distance vector from the SIM to the head CG in the head coordinate system. The CG of the head was assumed to be 8.4mm anterior and 31mm superior to the origin of the head coordinate system in the 50th

percentile male [119]. In the current study, these average values were scaled based on subject-specific head depth and height [117]. From linear acceleration data in the x, y, and z axes, a resultant linear acceleration was calculated. Two peaks were determined from the resultant linear acceleration. The first peak occurred when the ball made contact with the head; the second peak occurred when the ball rebounded from the head (Figure 5). The resultant rotational acceleration was calculated using the rotational acceleration data in the x, y, and z axes as reported by the SIM.

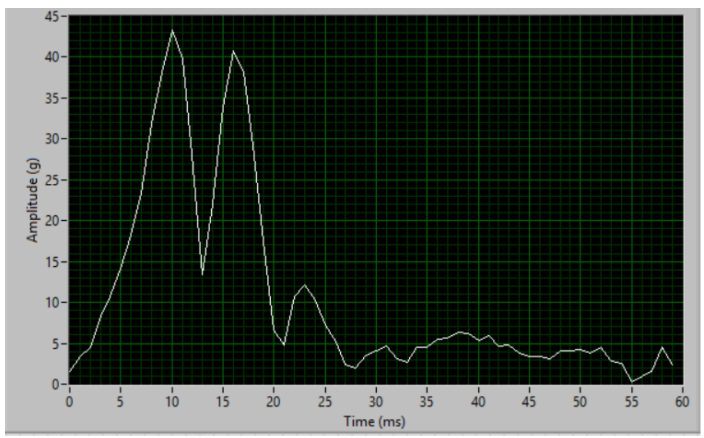

Figure 5: Example of resultant linear acceleration profile for one header. The first peak occurred when the ball made contact with the head; the second peak occurred when the ball rebounded from the head. The duration of ball contact was approximately 20ms.

2.2.3 Purposeful Soccer Heading Procedures

Participants performed a series of headers as described by Tierney et al. [45]. First, participants performed a neck warm-up consisting of 15 seconds of clockwise neck rotations, 15 seconds of counterclockwise neck rotations, 2 repetitions of stretching for 15 seconds in flexion and 15 seconds in extension, and 5-10 headers with a soft volleyball, tossed by the principal investigator. Then, the SIM was secured to the head using a custom elastic cap. Soccer balls (size 5, 450g, inflated to 9psi) were projected using a JUGS soccer machine (JUGS, Tualatin, OR). The initial velocity was 11.2m/s (25mph), the angle of projection was 40°, and the range was approximately 12m (40 ft) [120]. The participants aimed for a target (1.2x1.8m) approximately 2m in front of them, as if taking a shot on goal. Participants performed a total of 12 standing headers. The first two headers were considered warm-up headers to allow the participants to become familiar with the protocol. If a trial was not recorded due to technical error or a participant had an unusual header (i.e. he/she jumped during the header), the trial was not repeated because the pre-/post-comparison only allowed for 12 headers. There were 77/1000 (7.7%) trials excluded for one of the aforementioned reasons.

2.2.4 Data Analysis

Resultant peak linear and rotational accelerations were averaged across the 10 trials. The two peaks in linear acceleration were identified and averaged across trials.

2.2.5 Statistical Analysis

Two one-way analysis of variances (ANOVAs) were used to determine significant differences in resultant peak linear or rotational acceleration across groups. The independent variable was group, which includes six levels: youth male, youth female, high school male, high school female, collegiate male, and collegiate female. Each dependent variable (peak linear acceleration/peak rotational acceleration) was analyzed individually. The peak linear acceleration/peak rotational acceleration was the average of 10 trials.

A repeated measures ANOVA was used to determine significant differences in linear acceleration peaks across groups. The repeated measure was peak, which included the first peak and the second peak, and the independent variable was group. For all tests significance was defined by $\alpha = .05$. Accelerations from two collegiate female soccer players were not recorded because of SIM malfunction during testing.

2.3 Results

The overall ANOVAs showed a statistically significant difference between groups for both peak linear acceleration ($F = 6.797$, df [5, 92], $p < .001$) and peak rotational acceleration ($F = 5.691$, df [5,92], $p < .001$). Preliminary comparisons revealed that the homogeneity assumption underlying an ANOVA was met for both peak linear (Levene statistic = 1.825, df [5, 92], $p = .116$) and peak rotational (Levene statistic = 1.705, df [5, 92], $p = .141$) acceleration. Therefore, post hoc comparisons were done using the Tukey adjustment. Table 5 presents means and standard deviations for the six groups on the dependent variables of peak linear acceleration and peak rotational acceleration. Figure 6 presents means and standard deviations for the six groups on the dependent variables of first peak and second peak.

Table 5: Head acceleration means and standard deviations for the six groups. Mean±Standard Deviation.

Group	Peak Linear Acceleration (g)	Peak Rotational Acceleration (krad/s^2)
Collegiate Male	26.1±9.5	2.13±1.07
Collegiate Female	42.4±11.5[a,b]	3.35±0.87[a,b]
High School Male	26.7±5.9	2.18±0.40
High School Female	38.5±13.1[a,b]	3.16±1.01[a,b]
Youth Male	33.7±7.7	2.48±0.58
Youth Female	40.5±14.9[a,b]	3.20±1.24[a,b]
Overall	35.0±12.9	2.80±1.07

[a]Significantly different than collegiate male.
[b]Significantly different than high school male.

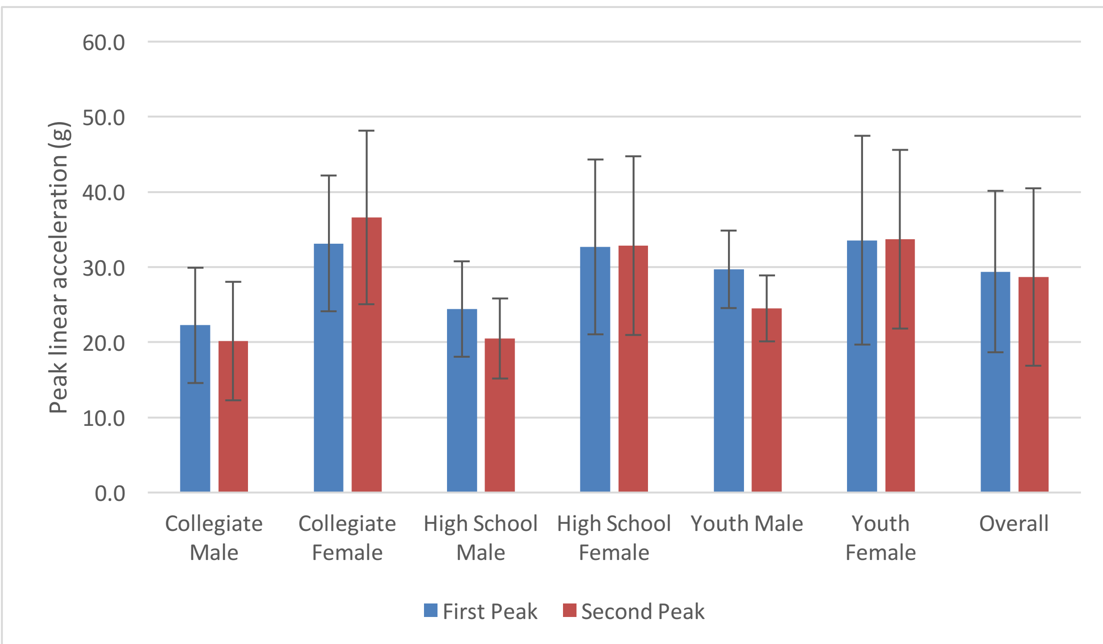

Figure 6: Head acceleration peaks means and standard deviations for the six groups.

Post hoc analyses demonstrated that the collegiate, high school, and youth female soccer players produced significantly higher peak linear acceleration than both the collegiate ($p < .001$, $p = .012$, $p = .002$) and high school male soccer players ($p = .002$, $p = .045$, $p = .012$). The obtained difference between the collegiate, high school, and youth female soccer players and collegiate male soccer players represented a large effect size ($d = 1.55$, $d = 1.08$, $d = 1.15$) and the difference between the collegiate, high school, and youth female soccer players and high school male soccer players also represented a large effect size ($d = 1.72$, $d = 1.16$, $d = 1.22$). Youth male soccer players were not different than collegiate, high school, or youth female soccer players ($p = .467$, $p = .920$, $p = .721$).

Collegiate, high school, and youth female soccer players produced significantly higher peak rotational acceleration than both the collegiate ($p = .002$, $p = .015$, $p = .011$) and high school male soccer players ($p = .011$, $p = .052$, $p = .040$). The obtained difference between the collegiate, high school, and youth female soccer players and collegiate male soccer players represented a large effect size ($d = 1.25$, $d = 0.99$, $d = 0.92$) and the difference between the collegiate, high school, and youth female soccer players and high school male soccer players also represented a large effect size ($d = 1.73$, $d = 1.28$, $d = 1.11$). Youth male soccer players were not different than collegiate, high school, or youth female soccer players ($p = .279$, $p = .555$, $p = .492$).

Results of the repeated measures ANOVA reveal the main effect for peak was not significant (Wilks' = .980, $F = 1.90$, df [1, 92], $p = .171$), whereby there was no difference between the first peak and the second peak. Likewise, the group-by-peak interaction was not significant (Wilks' = .904, $F = 1.95$, df [5, 92], $p = .094$).

In sum, results reveal that at the collegiate and high school levels, female soccer players had higher head accelerations than their male counterparts, but at the youth level, there were no significant differences between genders. Furthermore, within gender there were no significant differences in head accelerations across age groups. Finally, across all groups, there were no significant difference between peaks.

2.4 Discussion

The long-term effects of repeated heading of the soccer ball are largely unknown with some studies reporting deficits in neurological function [3, 5-13, 16-18], and others reporting no impairments [14, 21, 22, 24-28, 46, 88-92]. The first step in understanding these effects is learning more about the actual impact of a ball on the head. Therefore, the purpose of this study was to compare head acceleration across youth (12-14 years old), high school (15-18 years old), and collegiate (18-24 years old) male and female soccer players in a laboratory setting with all participants exposed to the same conditions (i.e. ball velocity). The primary findings were that collegiate and high school female soccer players have higher head accelerations than their male counterparts, but at the youth level, there are no significant differences between genders.

Previously, peak linear accelerations from 5.8g to 50.7g have been reported in laboratory-based soccer heading studies [29, 38, 40, 43, 45-47, 51, 121, 122]. These studies vary in populations, ball velocities, and technologies used to quantify head accelerations. Many of these studies quantify peak linear acceleration in collegiate and post-collegiate male soccer players [29, 38, 40, 43, 45, 47, 51, 121, 122]. In this population, peak linear accelerations have been observed from 10.4g (ball velocity = 9.3m/s) to 50.7g (ball velocity = 22.4m/s) [29, 38, 40, 43, 45, 47, 51, 121, 122]. Herein, we employed a ball velocity of 11.2m/s and reported a mean peak linear acceleration of 26.1±9.5g in our collegiate male cohort. Similarly, Higgins et al. used a ball speed of 11.2m/s and reported a mean peak linear acceleration of 24.6±5.0g in 9 male and 8 female collegiate soccer players, though peak linear accelerations were not compared across genders.

During laboratory-based soccer heading studies, no significant differences in peak linear acceleration were reported between male and female collegiate soccer players [45, 51]. Female soccer players exhibit peak linear accelerations from 10.4g (ball velocity = 9.3m/s) [51] to 20.2g (ball velocity = 9.83m/s) [45]. Although our reported mean peak linear acceleration of 42.4±11.5g in our collegiate female cohort is higher than previously reported in laboratory-based studies [45, 51], we employed a greater ball velocity and different methods for quantifying head acceleration. Tierney et al. used a custom-fit mouthpiece instrumented with a triaxial accelerometer [45]. While it is common in the soccer heading literature to measure peak linear accelerations using one triaxial accelerometer embedded in a mouthpiece [29, 38, 40, 45, 122], and this method has been reported to be valid for estimating head CG accelerations [123], Funk et al. suggested that these accelerations underestimate the peak linear acceleration in that they do not account for the rotation of the head, which may be quite high (i.e. whiplash) [118]. Therefore, accelerations measured at locations other than the CG of the head may not be comparable across studies because they may vary substantially depending on the test condition and sensor location [118]. Herein, we transformed peak linear accelerations to the CG of the head using linear acceleration data from a triaxial accelerometer and rotational velocity and acceleration data from a gyroscope in combination with motion capture data.

On-field soccer heading studies have described head accelerations in youth (20.4g), high school (37.6g), and collegiate (25-39.3g) female soccer players [52, 116, 124, 125]. The accelerometers (xPatch and Head Impact Telemetry System (HITS)) used in these studies transform head accelerations to the CG of the head, but apply different filtering algorithms to exclude false positive impacts and use different thresholds to identify impacts (i.e. 4.5 [52], 10g [116, 124], 20g [125]). During purposeful headers, head accelerations of 20.4g and 25±17g at the youth and collegiate level, respectively. These accelerations include all purposeful headers, some of which may occur at lower velocities than employed herein. Therefore, the variety of accelerometer technologies, thresholds and filtering algorithms, and ball velocities utilized makes these measurements difficult to compare across studies.

Research investigating the biomechanics of head injury has focused on both linear acceleration and rotational acceleration [33]. However, laboratory-based soccer heading studies have largely reported linear head acceleration only. Two studies have presented rotational acceleration during purposeful soccer heading in male and female collegiate soccer players [40, 51]. In a collegiate male population, mean peak rotational accelerations range from 1.40krad/s^2 to 2.41krad/s^2 depending on ball velocity (6m/s, 8m/s) and heading technique (exaggerated follow through, alignment, muscle tensing, etc.) [40]. These peak rotational accelerations are lower than on-field studies in high school and collegiate female soccer players, which report mean peak rotational accelerations from 4.54krad/s^2 to 7.71krad/s^2 [124, 125]. These differences may be attributed to laboratory-based versus on-field environment, and gender differences. We observed higher peak rotational accelerations in youth (3.20±1.24 krad/s^2), high school (3.16±1.01 krad/s^2), and collegiate female (3.35±0.87 krad/s^2) soccer players than in high school (2.18±0.40 krad/s^2) and collegiate (2.13±1.07 krad/s^2) male soccer players, suggesting that there is a significant difference in rotational acceleration of the head between genders at the high school and collegiate levels.

Within genders, there are no significant differences in head accelerations across age. However, at the collegiate and high school levels, female soccer players have higher head accelerations than their male counterparts, but at the youth level, there are no significant differences between genders. The similarity in head accelerations across youth groups may be a result of the pubertal cycle. Our youth groups consisted of soccer players 12-14 years old and there was no significant difference in age across the male and female youth groups. Puberty begins between the ages of 10 and 13 years old and occurs, on average, later in males than females [126]. Male musculature often develops in the late stages of puberty [126]. Although we did not take any measurements of pubertal maturation, it is reasonable to assume that our youth male participants may have been in earlier stages of development relative to their female counterparts, and therefore, may not have fully developed musculature compared to the high school and collegiate male soccer players. This difference in size and strength may contribute to the higher peak linear and rotational accelerations. Future studies should consider the effect of pubertal maturation on head acceleration during purposeful soccer heading.

Finally, across all groups, we observed no difference in the first peak linear acceleration compared to the second peak linear acceleration. In a male collegiate population of six soccer players, it has been reported that the second peak results in the greatest acceleration [40], suggesting that it is the forward movement during the rebound phase of ball contact that results in the greatest head acceleration [40, 114]. We speculate that if participants exhibited a higher first peak, they may not be able to control the ball impact and perhaps strengthening the neck musculature may help minimize the head acceleration, whereas if participants exhibited a higher second peak, they may be adding energy to the ball and therefore, strengthening the neck musculature may only increase the peak linear acceleration in that soccer players could produce an even more forceful rebound. Although the overall mean suggested the second peak was higher in male collegiate soccer players, three of the six participants observed by Shewchenko et al. presented with a higher first peak than second peak. We compared group means and found that across all groups, there were no significant differences in peaks. This suggests that there are likely within-subject and between-subject differences in peak linear acceleration profiles, and therefore individual headers may have higher peak linear acceleration during ball contact and others may have higher peak linear acceleration during ball rebound. Lack of consistency in inter-peak comparisons provides us little insight as to ways to minimize peak linear acceleration during purposeful soccer heading.

The purpose of this study was to compare head acceleration during purposeful soccer heading across age and gender. At the collegiate and high school levels, female soccer players had higher head accelerations than their male counterparts, suggesting that females may be exposed to greater cumulative head accelerations from repeated heading of a soccer ball over a career of soccer. Greater size and strength among male soccer players may result in lower peak linear and rotational acceleration and may contribute to the observed gender differences. This will be explore in Chapter 3 and more conclusions, limitations, and future directions will be presented in Chapter 5.

Chapter 3

HEAD ACCELERATION PREDICTORS

(Targeted for submission to *Clinical Biomechanics*)

3.1 Introduction

In November 2015, the U.S. Soccer Federation announced a new initiative to eliminate soccer heading for children aged 10 and under and to limit heading in practice for children between the ages of 11 and 13. Only two studies to date have described the frequency and magnitude of soccer heading in this age group [52, 116]. Chrisman et al. reported 0.93 headers per athlete per game over one weekend soccer tournament (range: 0-15) [116] and Hanlon and Bir described head accelerations from 4.5g to 62.9g during purposeful soccer heading across 6 youth soccer scrimmages [52]. These studies did not describe head accelerations during practices, which may occur more frequently as a result of heading drills, particularly at the youth level when soccer players are first learning proper heading technique. While intuitively it makes sense to limit soccer heading to avoid head injury, there may be ways to mitigate head acceleration without eliminating heading. Although minimizing head acceleration may not eradicate acute head injuries, it may reduce the effects of repeated head impacts.

Several studies have investigated soccer heading head acceleration in a controlled-laboratory setting or through mathematical modeling [38-51, 53]. These studies have investigated how various aspects of soccer heading, such as kinematics [40], muscle contraction [40, 43, 45, 46, 51, 53], anthropometric properties [45, 48], and gender [45, 51], affect head acceleration during heading. Increased forward flexion during follow-through along with increased head-torso alignment reduces head impact severity [40, 115]. Surprisingly, earlier muscle onset, longer onset duration, and greater peak amplitude relative to maximum voluntary isometric contraction do not appear reduce head acceleration [40, 64, 115]. While a few studies suggest that higher ("better") neck strength predicts lower head acceleration [45, 46]; one study that used an 8-week resistance training program to increase neck strength in a male and female collegiate population found no significant decrease in head acceleration despite increases in neck strength [53]. Another study suggested that it is not the flexor strength or the extensor strength that predicts lower head acceleration, but instead, it is the balance of flexor and extensor strength [51]. Finally, increased absolute head mass predicts a decrease in head acceleration [45, 48, 115].

Although these studies have investigated kinematics, muscle contraction, and anthropometric properties and their effects on head acceleration during purposeful soccer heading, it is unknown which of these factors are the best predictors for head acceleration during purposeful soccer heading. Moreover, soccer heading biomechanics research has largely been limited to studying collegiate male and female soccer players. Few studies have included youth and high school athletes [46, 52, 116, 125]. Because these young athletes may have lower head mass and neck strength, and perhaps poorer technique, we must determine how kinematics, muscle contraction, and anthropometric properties effect head acceleration across age and gender. Therefore, we investigated these measures simultaneously in youth, high school, and collegiate male and female soccer players to identify what factors best predict head acceleration during purposeful soccer heading.

The purpose of this study was to identify head acceleration predictors during purposeful soccer heading across youth, high school, and collegiate male and female soccer players. Consistent with previous literature, we hypothesized that head mass would be the best predictor of peak linear and rotational acceleration [45]. While many of these parameters can be improved with training, and may suggest areas to address to minimize head acceleration in soccer, head mass cannot be changed. If this was the most significant predictor, young athletes with low head mass may not be ready to begin heading the soccer ball.

3.2 Methods

3.2.1 Participants

One-hundred soccer players were recruited to participate in this study (Table 4). Potential participants were excluded from this investigation if they had a self-reported history of neurologic disorder, cervical spine injury, or head injury (i.e. concussion) in the past 6 months. If a participant sustained a concussion or any other injury in the past, they must have been cleared by a physician to return to full activity before participating in this study. Potential participants were included if they were currently participating in soccer within their age group and gender. Goal keepers were excluded. All participants provided written informed consent (IRB 476493-4). Children under the age of 18 years provided written informed assent, and their parent/legal guardian provided written informed parental consent.

3.2.2 Anthropometric Assessment

Participant height, mass, and head and neck anthropometrics were measured as previously described [53, 127, 128]. Body mass was measured using a Smart Body Analyzer (Withings Inc., Cambridge, MA). Body mass was multiplied by the sex-specific head to total body mass percentage for high school and collegiate participants (male = 8.26%, female = 8.20%) [129] or by the youth equivalent (10.1%) [130] to determine head mass [45]. Head and neck length, width, and depth were measured using an anthropometer (Lafayette Instrument Company, Lafayette, IN) [128]. Head and neck circumference were measured using a standard clinical tape measure [127]. All measurements were performed by the same principal investigator.

3.2.3 Head-Neck Isometric Strength Assessment

Isometric neck flexor, anterolateral neck flexor (sternocleidomastoid), cervical rotator, posterolateral neck extensor, and upper trapezius muscle strength were measured as described by Mihalik et al. [127]. The microFet 2 handheld dynamometer (Hoggan Health Industries, Inc., West Draper, UT) was used to quantify head-neck segment isometric muscle strength. The participants were asked to maximally contract against the dynamometer for 3 seconds. Three trials were conducted for each measurement. Participants were allowed to rest for 30 seconds between trials. Neck strength measurements were reported in kg and the peak values of each trial were averaged to determine the criterion measure. Bilateral strength measurements were averaged together into a single measure of strength for that muscle group [127]. On average, participants had greater strength on their right side than left side, although these differences were less than 1kg (Table 7). All measurements were performed by the same principal investigator (ICC > 0.9).

3.2.4 Motion Analysis System

The local coordinate systems of the head and the Smart Impact Monitor (SIM) (Triax Technologies Inc., Norwalk, CT) and heading kinematics were determined using an 8-camera Motion Analysis Motion Capture System (Motion Analysis

Corporation, Santa Rosa, CA). To record movement of the head and torso, 11 reflective markers (14mm) were attached to anatomic landmarks by Velcro™ tape (VELCRO USA, Manchester, NH) (Figure 3): the suprasternal notch, the xyphoid process, the 7th cervical vertebrae, the 10th thoracic vertebrae, the left and right temple, the left and right external auditory meatus (EAM), the back of the head, just below the inion (the location of the SIM), and the left inferior orbital rim. This marker set is consistent with that of Dezman et al. with additional markers on the left and right EAM and left inferior orbital rim to define local coordinate systems of the head and SIM [51, 117]. The left and right EAM and inferior orbital rim markers were used to define the local coordinate system of the head. The left and right EAM and back of the head makers were used to define the local coordinate system of the SIM. Data were recorded at 100Hz with a 1/1000 shutter speed. The torso angle relative to the laboratory reference frame and the relative head-to-torso angle were calculated.

3.2.5 Wireless EMG System

The Trigno™ Wireless EMG System (Delsys Inc., Natick, MA) was used to assess the EMG activity of the right and left sternocleidomastoid and upper trapezius muscles. The skin over the sternocleidomastoid and upper trapezius muscles was shaved, abraded, and cleaned with 70% alcohol. Wireless electrodes (37mm x 26mm x 15mm) were then placed over the sternocleidomastoid and upper trapezius muscles parallel to the fiber orientation using 4-slot Adhesive Skin Interfaces (Delsys Inc., Natick, MA). The sternocleidomastoid electrode was placed at 33% on the line from the sternal notch to the mastoid process in the direction of the line [131]. The upper trapezius electrode was placed at 50% on the line from the acromion to the spine on the 7th cervical vertebra in the direction of the line according the SENIAM guidelines

[132]. The raw signal (for maximum voluntary isometric contraction and trials) was sampled at a rate of 1000Hz, rectified, and smoothed using a root mean square algorithm over a 20ms moving window [53].

Peak muscle amplitude and area were recorded and normalized to peak maximal voluntary isometric contraction. Peak muscle amplitude was defined as the maximum activity at any point during the heading trial. Muscle area was defined as the area under the curve from 150ms before ball contact to 250ms after ball contact [45]. Maximal voluntary isometric contraction was assessed for each participant. Consistent with established protocols, sternocleidomastoid maximal voluntary isometric contraction was tested with the participant lying supine and flexing the neck anterolateral against resistance [133]. Upper trapezius maximal voluntary isometric contraction was tested with the participant sitting upright and elevating the shoulder girdle while extending the neck posterolateral against resistance [133]. Maximal voluntary isometric contraction was defined as the peak amplitude during each maximal voluntary isometric contraction trial. Sternocleidomastoid and upper trapezius onset time and peak time relative to ball contact were computed. Muscle onset was defined as 20% of the peak muscle amplitude [42].

3.2.6 Head Acceleration

The SIM (Triax Technologies Inc., Norwalk, CT) was used throughout testing to quantify head acceleration. The SIM is a head accelerometer that contains a triaxial accelerometer and gyroscope. Signals from the accelerometer and gyroscope were sampled at 1000Hz. The SIM has a threshold level of 8g, meaning that when the accelerometer registered a reading of 8g or higher, information 10ms before impact and 52ms after impact were wirelessly transmitted to the computer. The SIM was secured to the back of the head, just above the greater occipital protuberance, using a custom tight-fitting elastic cap (Figure 3). Karton et al., observed no significant difference in peak linear or rotational acceleration between SIM and instrumented headform at 30g and 50g impact energy levels (Karton, 2016, in press). At 80g, there was a significant difference in peak linear and rotational acceleration between SIM and instrumented headform, such that the SIM readings showed, on average, higher peak linear and rotational accelerations than those of the headform (Karton, 2016, in press). Furthermore, correlation coefficient results showed that a strong positive relationship exists between the headform and SIM peak linear and rotational accelerations with Pearson's r>0.9 (Karton, 2016, in press).

Raw, unprocessed data from the accelerometer and the gyroscope were obtained from the manufacturer. Data in the SIM coordinate system were rotated to that of the head coordinate system using 3D motion capture data for sensor position and head position [118]. The coordinate system of the head was defined using the midpoint of the EAMs as the origin. The positive x-axis pointed anteriorly through the inferior orbital rim along the Frankfort plane. The positive z-axis pointed superiorly, perpendicular to the Frankfort plane, and the positive y-axis pointed to the left. The coordinate system of the SIM was defined using the location of the SIM as the origin. The y-axis of the triaxial accelerometer within the SIM was aligned with the positive y-axis of the head coordinate system. The positive z-axis pointed backward and downward from the origin of the head to the location of the SIM. The positive x-axis pointed backward and upward, perpendicular to both the y-axis and z-axis (Figure 4).

Linear accelerations of the head in the head coordinate system were transformed to the head center of gravity (CG) using standard dynamics equations applied to a rigid body

$$\overrightarrow{\alpha_{CG}} = \overrightarrow{\alpha_{SIM}} + \ddot{\vec{\theta}} x \vec{d} + \dot{\vec{\theta}} x (\dot{\vec{\theta}} x \vec{d})$$

where $\overrightarrow{\alpha_{CG}}$ was the linear acceleration of the head CG, $\overrightarrow{\alpha_{SIM}}$ was the linear acceleration of the SIM, $\ddot{\vec{\theta}}$ was the rotational acceleration of the head, $\dot{\vec{\theta}}$ was the rotational velocity of the head, and $\vec{d}$ was the distance vector from the SIM to the head CG in the head coordinate system. The CG of the head was assumed to be 8.4mm anterior and 31mm superior to the origin of the head coordinate system in the 50th percentile male [119]. In the current study, these average values were scaled based on subject-specific head depth and height [117]. From linear acceleration data in the x, y, and z axes, a resultant linear acceleration was calculated. The resultant rotational acceleration was calculated using the rotational acceleration data in the x, y, and z axes as reported by the SIM.

3.2.7 Purposeful Soccer Heading Procedures

Participants performed a series of headers as described by Tierney et al. [45]. First, participants performed a neck warm-up consisting of 15 seconds of clockwise neck rotations, 15 seconds of counterclockwise neck rotations, 2 repetitions of stretching for 15 seconds in flexion and 15 seconds in extension, and 5-10 headers with a soft volleyball, tossed by the principal investigator. Then, the SIM was secured to the head using a custom elastic cap. Soccer balls (size 5, 450g, inflated to 9psi) were projected using a JUGS soccer machine (JUGS, Tualatin, OR). The initial velocity was 11.2m/s (25mph), the angle of projection was 40°, and the range was approximately 12m (40 ft) [120]. The participants aimed for a target (1.2x1.8m) approximately 2m in front of them, as if taking a shot on goal. Participants performed a total of 12 standing headers. The first two headers were considered warm-up headers to allow the participants to become familiar with the protocol. If a trial was not recorded due to technical error or a participant had an unusual header (i.e. he/she jumped during the header), the trial was not repeated because the pre-/post-comparison only allowed for 12 headers. There were 77/1000 (7.7%) trials excluded for one of the aforementioned reasons.

3.2.8 Data Analysis

Resultant peak linear and rotational accelerations were used to determine which of the 10 measures best predicts head acceleration through a multiple regression analysis (MRA) (Table 6).

Table 6: Ten potential predictors of head acceleration.

Predictor	Measurement
Neck girth (cm)	Circumference of the neck above the thyroid cartilage
Head mass (kg)	Percentage of total body weight
Sternocleidomastoid strength (kg)	Measured with handheld dynamometer; peak force, averaged over 3 trials
Upper trapezius strength (kg)	Measured with handheld dynamometer; peak force, averaged over 3 trials
Head-to-torso range-of-motion (degrees)	From motion analysis; change in angle from maximum extension to maximum flexion and averaged over ten trials
Torso range-of-motion (degrees)	From motion analysis; change in angle from maximum extension to maximum flexion and averaged over ten trials
Sternocleidomastoid peak amplitude (%maximum voluntary isometric contraction)	Maximum muscle activity within each trial and averaged over ten trials

Upper trapezius peak amplitude (%maximum voluntary isometric contraction)	Maximum muscle activity within each trial and averaged over ten trials
Sternocleidomastoid peak area (%maximum voluntary isometric contraction)	Area under the curve from 150ms before ball contact to 250ms after ball contact and averaged over ten trials
Upper trapezius peak area (%maximum voluntary isometric contraction)	Area under the curve from 150ms before ball contact to 250ms after ball contact and averaged over ten trials

3.2.9 Statistical Analysis

Direct-entry MRAs were used to determine which of the 10 factors best predicts linear acceleration and rotational acceleration of the head. Predictors were also grouped by technique (head relative to torso and torso relative to the laboratory floor range-of-motion, sternocleidomastoid and upper trapezius peak amplitude and peak area), strength (sternocleidomastoid and upper trapezius strength), and size (head mass, neck girth). MRAs were used to determine if technique, strength, or size predicted peak linear or rotational acceleration. Separate one-way ANOVAs were used to compare each of the ten predictors across groups. Two repeated-measures analysis of variance (ANOVAs) were used to compare soccer heading kinematics across groups. Finally, separate one-way ANOVAs were run to compare sternocleidomastoid and upper trapezius onset time and peak time relative to ball contact to determine if groups applied different muscle activation strategies. For all tests significance was defined by $p < 0.05$.

3.3 Results

3.3.1 Anthropometric Assessment

Distributional statistics (means, standard deviations) are presented in Table 7 for the anthropometric measurements.

Table 7: Anthropometric measurements by age and gender. Mean (Standard Deviation) in cm.

Group	Head Width*	Head Depth*	Head Length*	Head Girth+	Neck Width*	Neck Depth*	Neck Length	Neck Girth*
Collegiate Male	15.5 (0.5)	19.7 (0.8)	22.4 (1.7)	56.0 (1.8)	11.8 (0.4)	12.4 (0.8)	13.9 (2.3)	37.6 (1.6)
Collegiate Female	14.8 (0.5)[a]	18.7 (0.7)[a,b]	19.5 (1.1)[a,b]	54.9 (1.7)	10.4 (0.6)[a,b]	9.8 (0.5)[a,b]	13.3 (2.4)	32.7 (1.5)[a,b]
High School Male	15.4 (0.6)	19.6 (0.5)[a]	21.1 (0.9)	56.7 (1.4)	11.2 (0.7)	11.5 (0.7)[a]	13.9 (1.3)	36.3 (2.2)
High School Female	14.9 (0.7)[a]	18.8 (0.8)[a,b]	19.6 (1.2)[a,b]	55.4 (1.5)	10.2 (0.6)[a,b]	9.5 (0.5)[a,b]	14.6 (1.4)	31.8 (1.2)[a,b]
Youth Male	15.0 (0.6)	19.0 (0.6)[a]	20.7 (0.9)	55.4 (2.1)	10.4 (0.8)[a]	9.7 (1.2)[a,b]	14.2 (2.0)	32.0 (4.0)[a,b]
Youth Female	14.6 (0.6)[a,b]	18.6 (0.6)[a,b]	19.8 (1.0)[a,b]	54.9 (1.7)[b]	9.7 (0.9)[a,b,c]	9.1 (0.6)[a,b,c]	13.5 (2.2)	31.3 (2.5)[a,b]

* One-way ANOVA was significant ($p < .001$).
+One-way ANOVA was significant ($p = .031$).
[a]Collegiate male significantly greater ($p < .05$).
[b]High school male significantly greater ($p < .05$).
[c]Collegiate female significantly greater ($p < .05$).

3.3.2 Head-Neck Isometric Strength Assessment

Distributional statistics (means, standard deviations) are presented in Table 8 for the head-neck isometric strength measurements.

Table 8: Head-neck isometric strength measurements by age and gender. Mean (Standard Deviation) in kg.

Group	Flexor*	Right SCM*	Left SCM*	Right Rotator*	Left Rotator*	Right Extensor*	Left Extensor*	Right Trap*	Left Trap*
Collegiate Male	14.3 (3.3)	14.3 (2.4)	13.5 (2.5)	14.8 (3.1)	13.9 (2.6)	16.4 (5.1)	16.2 (5.7)	19.0 (3.2)	18.6 (4.1)
Collegiate Female	7.4 (2.4)[a,b]	7.9 (2.4)[a,b]	7.7 (2.2)[a,b]	8.1 (2.3)[a]	7.9 (1.9)[a,b]	9.7 (3.0)[a]	9.6 (2.6)[a]	12.6 (3.8)[a,b]	12.1 (3.8)[a]
High School Male	9.9 (2.2)[a]	11.1 (2.0)[a]	11.3 (2.3)[a]	9.7 (3.1)[a]	10.8 (2.3)[a]	10.6 (2.7)[a]	10.0 (3.1)[a]	15.9 (5.1)	15.2 (5.4)
High School Female	6.2 (1.6)[a,b]	7.3 (1.6)[a,b]	7.1 (1.8)[a,b]	7.3 (1.5)[a]	7.0 (1.7)[a,b]	7.7 (2.0)[a]	7.3 (1.9)[a]	10.3 (1.8)[a,b,d]	10.7 (2.5)[a,b,d]
Youth Male	8.5 (3.1)[a]	8.3 (1.4)[a,b]	7.3 (1.3)[a,b]	9.4 (2.2)[a]	8.5 (1.6)[a]	9.6 (2.7)[a]	9.9 (2.1)[a]	15.3 (3.1)	16.5 (3.9)
Youth Female	5.9 (1.1)[a,b]	6.8 (1.8)[a,b]	6.2 (1.6)[a,b]	6.8 (1.5)[a,b]	6.3 (1.6)[a,b]	6.9 (1.6)[a,b]	6.4 (1.5)[a,b]	9.2 (2.2)[a,b,c,d]	9.1 (3.1)[a,b,d]

SCM = Sternocleidomastoid, Trap = Upper Trapezius

* One-way ANOVA was significant ($p < .001$).

[a]Collegiate male significantly greater ($p < .05$).

[b]High school male significantly greater ($p < .05$).

[c]Collegiate female significantly greater ($p < .05$).

[d]Youth male significantly greater ($p < .05$).

3.3.3 Soccer Heading Kinematics

Distributional statistics (means, standard deviations) are presented in Figure 7 for head-to-torso kinematics across groups. Results of the repeated measures ANOVA for head-to-torso kinematics reveal the group-by-time interaction (Wilks' = .512, F = 4.61, df [15, 252], $p < .001$) was significant, and the obtained effect represented a moderate effect size (i.e., partial eta squared = .171). Four post-hoc one-way ANOVAs were run, one at each time point and demonstrated that only the starting angle was different across groups (F = 4.842, df [5,93], $p = .001$), in that the male collegiate soccer players started with their head in a more extended position than the high school male (p = .028) and youth and high school female soccer players (p = .015, p = .005).

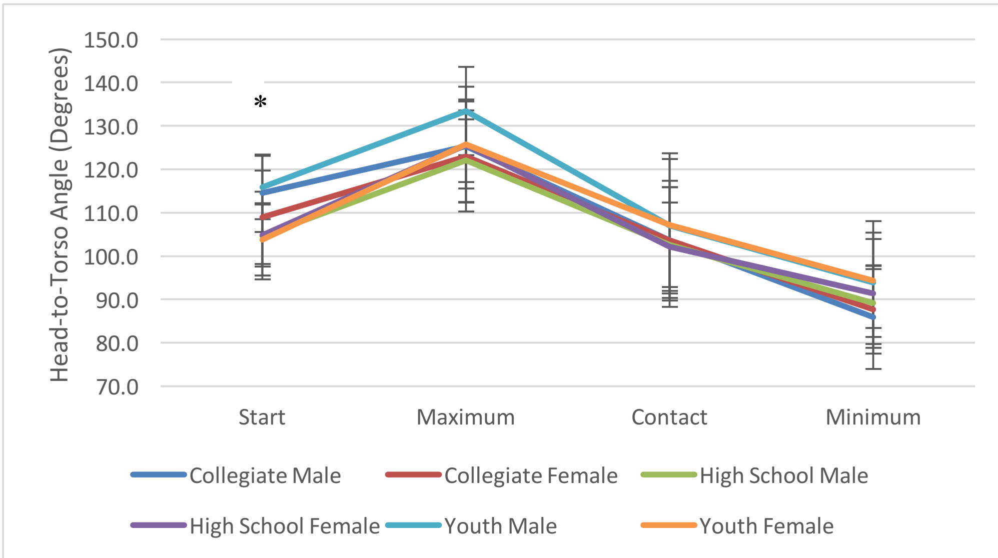

Figure 7: Head-to-torso kinematics across groups. The angle between the long axis of the torso and the horizontal plane of the head is shown. Post-hoc one-way ANOVAs demonstrated that the starting angle was different across groups, in that the male collegiate soccer players started with their head in a more extended position than the high school male (p = .028) and youth and high school female soccer players (p = .015, p = .005).

Distributional statistics (means, standard deviations) are presented in Figure 8 for torso kinematics across groups. Results of the repeated measures ANOVA for torso kinematics reveal the group-by-time interaction (Wilks' = .613, F = 3.26, df [15, 252], p < .001) was significant, and represented a moderate effect size (i.e., partial eta squared = .151). Four post-hoc one-way ANOVAs were run, one at each time point and demonstrated that the starting angle (F = 7.899, df [5,93], p < .001), the contact angle (F = 2.507, df [5,93], p = .005), and the maximum flexion angle (F = 3.578, df

[5,93], p = .036) were different across groups, in that the male collegiate soccer players started with their torso in a more flexed position across all other groups (p < 0.01). The male collegiate soccer players also differed from the female high school soccer players in contact angle and maximum flexion angle, in that the male collegiate soccer players made contact with the ball in a more flexed torso position and had a greater follow-through than the female high school soccer players.

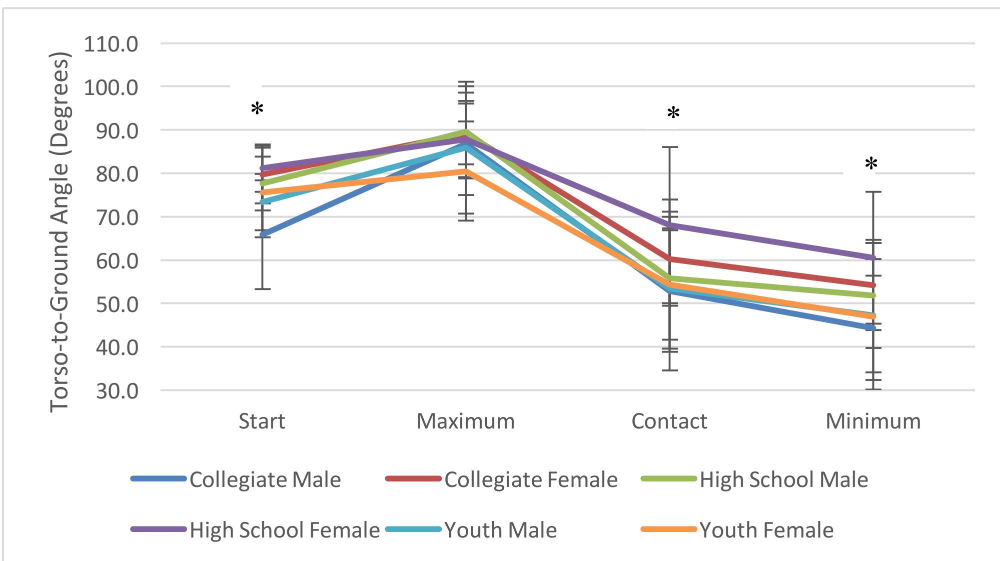

Figure 8: Torso kinematics across groups. The angle between the long axis of the torso and the horizontal plane is shown. Post-hoc one-way ANOVAs demonstrated that the starting angle, the contact angle, and the maximum flexion angle were different across groups, in that the male collegiate soccer players started with their torso in a more flexed position across all other groups (p < 0.01). The male collegiate soccer players also differed from the female high school soccer players in contact angle and maximum flexion angle, in that the male collegiate soccer players made contact with the ball in a more flexed torso position and had a greater follow-through than the female high school soccer players.

3.3.4 Muscle Activation Timing

Distributional statistics (means, standard deviations) are presented in Table 9 for muscle activation timing across groups. Results of the one-way ANOVAs for sternocleidomastoid onset ($F = 2.241$, df [5,90], $p = 0.057$) and peak ($F = .788$, df [5,90], $p = .561$) relative to ball contact and upper trapezius onset ($F = .753$, df [5,90], $p = .586$) and peak ($F = .393$, df [5,90], $p = .852$) relative to ball contact timing were not significant.

Table 9: Muscle activation timing relative to ball contact across groups. Mean time relative to ball contact in milliseconds (Standard Deviation). Results of the one-way ANOVAs for sternocleidomastoid onset and peak relative to ball contact and upper trapezius onset and peak relative to ball contact timing were not significant.

Group	Upper Trapezius Onset	Upper Trapezius Peak	Sternocleido-mastoid Onset	Sternocleido-mastoid Peak
Collegiate Male	-610 (265)	-99 (174)	-365 (170)	-98 (63)
Collegiate Female	-485 (186)	-103 (77)	-428 (215)	-97 (127)
High School Male	-683 (562)	-71 (290)	-318 (348)	-23 (300)
High School Female	-657 (388)	-45 (52)	-342 (160)	-69 (107)
Youth Male	-657 (251)	-54 (149)	-362 (47)	-59 (63)
Youth Female	-615 (416)	-81 (57)	-535 (215)	-117 (116)

3.3.5 Multiple Regression Analysis (Combined - 10 Predictors)

There were 94 of 100 total participants included in the multiple regression analysis; 6 were omitted because two were missing head acceleration measurements and four were missing EMG data because of equipment malfunction during testing.

Distributional statistics (means, standard deviations) are presented in Table 10 for the predictors and criterion. The results from the MRA suggest that the ten variables explained a statistically significant amount of variance in the dependent variable of peak linear acceleration, $R^2 = .272$, $F(10, 83) = 3.106$, $p = .002$. The ten variables also explained a statistically significant amount of variance in the dependent variable of peak rotational acceleration, $R^2 = .259$, $F(10, 83) = 2.905$, $p = .004$. The post-hoc statistical powers for both multiple regression analyses were 0.98. None of the ten head acceleration predictors made a statistically-significant, unique contribution to the prediction of peak linear or rotational acceleration (Table 11, 12).

3.3.6 Multiple Regression Analysis (Technique)

Technique variables included: head-to-torso range-of-motion, torso range-of-motion, sternocleidomastoid peak amplitude, sternocleidomastoid area, upper trapezius peak amplitude, and upper trapezius area. The results of the multiple regression analysis suggest that the technique variables *did not* explain a statistically significant amount of variance in the dependent variable of peak linear acceleration, $R^2 = .066$, $F(6, 87) = 1.029$, $p = .412$. The technique variables also *did not* explain a statistically significant amount of variance in the dependent variable of peak rotational acceleration, $R^2 = .047$, $F(6, 87) = 0.730$, $p = .627$.

3.3.7 Multiple Regression Analysis (Strength)

Strength variables included: sternocleidomastoid and upper trapezius strength. The results of the multiple regression analysis suggest that the strength variables explained a statistically significant amount of variance in the dependent variable of peak linear acceleration, $R^2 = .133$, $F(2, 91) = 7.276$, $p = .001$. The strength variables also explained a statistically significant amount of variance in the dependent variable of peak rotational acceleration, $R^2 = .172$, $F(2, 91) = 9.888$, $p < .001$. Sternocleidomastoid strength made statistically-significant, unique contribution to the prediction of peak linear and rotational acceleration (linear: $B = -1.544$, $p = .012$; rotational: $B = -.117$, $p = .018$).

3.3.8 Multiple Regression Analysis (Size)

Size variables included: head mass and neck girth. The results of the multiple regression analysis suggest that the size variables explained a statistically significant amount of variance in the dependent variable of peak linear acceleration, $R^2 = .221$, $F(2, 91) = 13.480$, $p < .001$. The size variables also explained a statistically significant amount of variance in the dependent variable of peak rotational acceleration, $R^2 = .233$, $F(2, 91) = 14.442$, $p < .001$. Head mass made statistically-significant, unique contribution to the prediction of peak rotational acceleration only ($B = -.404$, $p = .034$). There were no statistically-significant, unique contributors to the prediction of peak linear acceleration.

Table 10: Predictors and criterion. Mean (Standard Deviation). * for group measurement indicates the groups are significantly different.

Group	PLA (g)	PRA (krad/s^2)	Neck Girth* (cm)	Head Mass* (kg)	SCM Strength* (kg)	Trap Strength* (kg)	Head ROM (deg)	Torso ROM (deg)	SCM Peak (/MVIC)	SCM Area (/MVIC)	Trap Peak (/MVIC)	Trap Area (/MVIC)
Collegiate Male	26.1 (9.5)	2.13 (1.07)	37.6 (1.6)	6.38 (0.55)	13.9 (2.3)	18.8 (3.2)	39.5 (7.6)	42.5 (12.9)	3.45 (3.87)	1.08 (1.44)	1.28 (1.29)	0.23 (0.25)
Collegiate Female	42.4 (11.5) [a,b]	3.35 (0.87) [a,b]	32.7 (1.5) [a,b]	4.86 (0.27) [a,b]	7.8 (2.2) [a,b]	12.4 (3.5) [a]	35.2 (5.6)	34.7 (11.0)	5.16 (3.95)	2.15 (2.66)	2.32 (3.73)	0.58 (0.85)
High School Male	26.7 (5.9)	2.18 (0.40)	36.3 (2.2)	5.68 (0.56)	11.2 (2.0) [a]	15.6 (5.2)	32.9 (7.7)	37.8 (10.8)	1.57 (1.85)	0.97 (2.01)	0.84 (1.31)	2.81 (6.62)
High School Female	38.5 (13.1) [a,b]	3.16 (1.01) [a,b]	31.8 (1.2) [a,b]	4.82 (0.54) [a,b]	7.2 (1.6) [a,b]	10.5 (2.0) [a,b,c]	34.3 (14.7)	27.2 (10.6) [a]	2.76 (2.84)	2.88 (6.61)	1.29 (1.12)	3.57 (5.86)
Youth Male	33.7 (7.7)	2.48 (0.58)	32.0 (4.0)	5.03 (1.38)	7.8 (1.3) [a,b]	15.9 (3.1)	39.6 (9.9)	38.7 (11.9)	2.38 (2.12)	0.84 (0.96)	1.30 (0.91)	0.34 (0.18)
Youth Female	40.5 (14.9) [a,b]	3.20 (1.24) [a,b]	31.3 (2.5) [a,b]	4.86 (0.85) [a,b]	6.5 (1.7) [a,b]	9.2 (2.5) [a,b,c,d]	31.3 (5.8)	33.5 (13.1)	2.91 (3.99)	5.73 (10.14)	1.39 (1.54)	4.89 (5.93)
Overall	35.0 (12.9)	2.8 (1.1)	33.7 (3.2)	5.29 (0.89)	9.1 (3.4)	13.4 (4.8)	35.2 (9.3)	35.4 (12.6)	3.26 (3.54)	2.45 (5.55)	1.47 (2.10)	2.11 (4.62)

PLA = Peak Linear Acceleration, PRA = Peak Rotational Acceleration, SCM = Sternocleidomastoid, Trap = Upper Trapezius, ROM = Range-of-Motion

*Statistically correlated with PLA and PRA

[a] Statistically different from collegiate male

[b] Statistically different from HS male

[c] Statistically different from youth male

[d] Statistically different from collegiate female

Table 11: Coefficients for multiple regression analysis on the dependent variable of peak linear acceleration.

Model	Unstandardized Coefficients		Standardized Coefficients	t	Sig.
	B	Std. Error	Beta		
(Constant)	90.374	17.045		5.302	.000
Head-to-Torso Range of Motion	-.053	.145	-.035	-.363	.718
Torso-to-Ground Range of Motion	.047	.112	.043	.424	.673
Upper Trapezius Peak	.700	.597	.113	1.173	.244
Upper Trapezius Area	-.380	.296	-.135	-1.285	.202
Sternocleidomastoid Peak	-.430	.437	-.114	-.984	.328
Sternocleidomastoid Area	.103	.286	.044	.360	.720
Head Mass	-3.006	2.663	-.203	-1.129	.262
Neck Girth	-1.043	.706	-.257	-1.478	.143
Sternocleidomastoid Strength	-.475	.667	-.124	-.712	.478
Upper Trapezius Strength	.107	.432	.040	.249	.804

Table 12: Coefficients for multiple regression analysis on the dependent variable of peak rotational acceleration.

Model	Unstandardized Coefficients		Standardized Coefficients	t	Sig.
	B	Std. Error	Beta		
(Constant)	6.519	1.418		4.598	.000
Head-to-Torso Range of Motion	-0.001	0.012	-0.009	-0.094	0.926
Torso-to-Ground Range of Motion	0.004	0.009	0.044	0.423	0.674
Upper Trapezius Peak	0.032	0.05	0.063	0.645	0.52
Upper Trapezius Area	-0.024	0.025	-0.105	-0.989	0.326
Sternocleidomastoid Peak	-0.014	0.036	-0.046	-0.391	0.697
Sternocleidomastoid Area	-0.006	0.024	-0.029	-0.237	0.814
Head Mass	-0.329	0.221	-0.27	-1.486	0.141
Neck Girth	-0.044	0.059	-0.133	-0.754	0.453
Sternocleidomastoid Strength	-0.04	0.056	-0.128	-0.724	0.471
Upper Trapezius Strength	-0.01	0.036	-0.046	-0.282	0.779

3.3.9 Correlations

The relationships were statistically significant between head mass, neck girth, sternocleidomastoid strength, and upper trapezius strength and peak linear acceleration and rotational acceleration (Table 13). The relationships were not statistically significant between head-to-torso range-of motion, torso range-of-motion, sternocleidomastoid peak amplitude, sternocleidomastoid area, upper trapezius peak amplitude, or upper trapezius area and peak linear acceleration or peak rotational acceleration (Table 13).

3.3.10 Group Comparisons

Ten one-way ANOVAs were run to compare each of the ten predictors across groups. A Bonferroni correction was applied, and statistical significance was defined by $\alpha = 0.005$ (Table 14). Preliminary comparisons revealed that the homogeneity assumption underlying an ANOVA was met for torso range-of-motion (Levene statistic = .514, df [5, 93], $p = .765$) and sternocleidomastoid strength (Levene statistic = 1.282, df [5, 93], $p = .278$). Therefore, post-hoc comparisons were done using the Tukey adjustment for these measures. The homogeneity assumption underlying an ANOVA was not met for head mass (Levene statistic = 5.148, df [5, 93], $p < .001$), neck girth (Levene statistic = 4.220, df [5, 93], $p = .002$), or upper trapezius strength (Levene statistic = 3.299, df [5, 93], $p = .009$). Therefore, post hoc comparisons were done using the Games-Howell adjustment for these measures. Post-hoc analyses demonstrated that the collegiate male soccer players had more torso range-of-motion than high school female soccer players ($p = .001$). Other significant post-hoc analyses are presented in Table 15.

Table 13: Correlations between peak linear and rotational accelerations and ten predictors.

		Head-to-Torso ROM	**Torso-to-Ground ROM**	**Upper Trapezius Peak**	**Upper Trapezius Area**	**SCM Peak**	**SCM Area**	**Head Mass**	**Neck Girth**	**SCM Strength**	**Upper Trapezius Strength**
Pearson Correlation	Linear	-0.068	-0.098	0.160	-0.010	-0.089	-0.032	-0.447	-0.465	-0.386	-0.311
Significance	Linear	0.257	0.175	0.062	0.461	0.198	0.380	<.001*	<.001*	<.001*	0.001*
Pearson Correlation	Rotational	-0.054	-0.112	0.119	0.006	-0.058	-0.063	-0.467	-0.438	-0.410	-0.351
Significance	Rotational	0.301	0.142	0.128	0.476	0.289	0.273	<.001*	<.001*	<.001*	<.001*

*p<.005 defined as significant

SCM = Sternocleidomastoid, ROM = Range-of-Motion

Table 14: Results of the one-way ANOVAs for each of the ten predictors across groups.

	F	p
Upper Trapezius Peak	1.014	.414
Upper Trapezius Area	3.465	.007
SCM Peak	2.181	.063
SCM Area	1.936	.096
Head Mass	17.569	<.001*
Neck Girth	29.501	<.001*
SCM Strength	40.209	<.001*
Upper Trapezius Strength	21.303	<.001*
Torso-to-Ground ROM	3.626	.005*
Head-to-Torso ROM	21.008	.070

*p<.005 defined as significant
SCM = Sternocleidomastoid, ROM = Range-of-Motion

Table 15: Significant group comparisons for head mass, neck girth, sternocleidomastoid and upper trapezius strength.

Predictor	Reference Group	Comparison Group	Significance
Head Mass	Collegiate Male	Collegiate Female	<.001
		High School Male	.034
		High School Female	<.001
		Youth Male	<.001
		Youth Female	<.001
	High School Male	Collegiate Female	.006
		High School Female	.004
		Youth Female	.009
Neck Girth	Collegiate Male	Collegiate Female	<.001
		High School Female	<.001
		Youth Male	<.001
		Youth Female	<.001
		Collegiate Female	<.001

	High School Male	High School Female	<.001
		Youth Male	<.001
		Youth Female	<.001
Sternocleidomastoid Strength	Collegiate Male	Collegiate Female	<.001
		High School Male	.002
		High School Female	<.001
		Youth Male	<.001
		Youth Female	<.001
	High School Male	Collegiate Female	<.001
		High School Female	<.001
		Youth Male	.002
		Youth Female	<.001
Upper Trapezius Strength	Collegiate Male	Collegiate Female	<.001
		High School Female	<.001
		Youth Female	<.001
	Collegiate Female	Youth Female	.049
	High School Male	High School Female	.001
		Youth Female	<.001
	Youth Male	High School Female	.006
		Youth Female	<.001

The obtained difference between all comparisons represented a large effect size (i.e., d > 0.80).

In sum, results reveal that there were no statistically-significant, unique predictors of either peak linear or rotational acceleration in the combined model, although strength and size models were significant, suggesting that strength and size may predict peak linear and rotational acceleration. Furthermore, group differences in head mass, neck girth, sternocleidomastoid and upper trapezius strength, and torso range-of-motion were observed. It appears that, on average, groups apply similar kinematic and muscle activation strategies to soccer heading, though collegiate male soccer players have more torso involvement than high school female soccer players.

3.4 Discussion

Soccer heading kinematics, muscle contraction, and anthropometric combined properties may influence head acceleration during purposeful soccer heading [40, 43, 45, 46, 51, 53]. However, studies reporting these findings are often limited in population and scope. It is unknown which of these factors best predicts peak linear and rotational acceleration during soccer heading across various ages and genders. Therefore, we investigated kinematics, muscle contraction, and anthropometric properties simultaneously across youth, high school, and collegiate male and female soccer players. The purpose of this study was to determine what factors influence head acceleration during purposeful soccer heading. The primary finding of this study was that the combined regression model was significantly predictive, accounting for 27% of the variance in peak linear acceleration and 26% of the variance in peak rotational acceleration. Although there were no statistically-significant, unique predictors of either peak linear or rotational acceleration in the combined model, size (head mass and neck girth) and strength (sternocleidomastoid and upper trapezius strength) regression models were also significant, accounting for 22% and 13% of peak linear acceleration and 23% and 17% of peak rotational acceleration, respectively. These data suggest that size and strength are important parameters in minimizing head acceleration during purposeful soccer heading.

3.4.1 Anthropometric Assessment

Female participants demonstrated lower head mass compared with male participants. This is consistent with previous research [53, 64], which has observed head mass between 6.1 and 6.4kg in male collegiate soccer players and between 4.9 and 5.3kg in female collegiate soccer players. Head mass is inversely correlated with head acceleration, such that a higher head-neck mass is associated with lower head acceleration [45, 48, 64]. In our correlations, we confirmed these findings, and in addition, reported that greater size (head mass and neck girth) predicted lower peak linear and rotational acceleration.

Consistent with previous literature, we found that neck girth is inversely correlated with head acceleration, such that a greater neck girth is associated with lower head acceleration [45, 53, 64, 134]. Increased muscle tissue is related to increased neck girth, and may suggest an increase in neck stiffness [45, 53, 64, 134]. These findings suggest that soccer players with lower head mass and smaller neck girth may experience greater peak linear and rotational accelerations. Perhaps implementing a neck strengthening regimen, particularly in those athletes with smaller neck girth, could minimize head acceleration during purposeful soccer heading, although more research is needed to establish strengthening regimens that are effective in reducing head acceleration.

3.4.2 Head-Neck Isometric Strength Assessment

Neck strength in high school [135, 136] and collegiate [136] athletes and in the pediatric population [137] have been characterized in large normative databases. Across these studies, neck strength has consistently been found to be higher in men than in women. Similarly in soccer, neck strength has been found to be higher in men than women for both flexors (respectively, 10.9-15.9kg, 7.0-9.2kg) and extensors (respectively, 14.7-21.2kg, 8.9-11.4kg) [53, 64]. We chose to include sternocleidomastoid and upper trapezius strength, instead of neck flexor and extensor strength more broadly, to understand the role of strength and activity of the same muscles. We reported greater sternocleidomastoid and upper trapezius strength in males than in females.

Previous research involving both adolescent[46] and collegiate[45, 64] populations demonstrated that higher neck strength was associated with lower head acceleration during purposeful soccer heading. We found that higher neck strength among youth, high school, and collegiate male and female soccer players predicted lower peak linear and rotational acceleration. Specifically, sternocleidomastoid strength was a significant, unique predictor in the strength regression model. However, another study has suggested that isotonic resistance training to strengthen the neck musculature does not decrease head acceleration[53]. The authors suggested that head acceleration may be related to neuromuscular control instead of pure neck strength and speculated that in addition to traditional isotonic resistance training, soccer players should also implement ballistic activities, which have been reported to enhance neuromuscular control and dynamic stabilization at other joints[53]. For example, in mixed martial arts, ballistic neck strengthening activities, such as physioball neck leans and prone cobras, have been used to improve neuromuscular control[72]. These same activities may prove useful in athletes training for other sports, such as soccer, as well.

3.4.3 Soccer Heading Kinematics

In 2005, Shewchenko et al. showed that increased trunk rotation throughout the "follow-through" phase in soccer heading resulted in increased duration of muscle activation and an increase in peak linear acceleration by 7%, but a decrease in peak angular acceleration by 13% [40]. This same study reported a 5% increase in peak linear acceleration and no change in peak rotational acceleration when the participants were instructed to maintain a constant head-neck-torso alignment while heading the soccer ball [40]. These techniques would be expected to decrease head acceleration in that they are consistent with proper heading technique [40]; however, this study was limited to 7 participants who were asked to change their heading technique to meet the "follow-through" and "aligned" conditions. We found that technique variables did not predict peak linear or rotational acceleration, although they were included as inputs in the significantly predictive combined regression models. Our results, in combination with those of Shewchenko et al. [40], suggest that soccer heading technique may not be as important in minimizing head acceleration as one may believe. Though, we still suggest that soccer players learn proper heading technique because although technique may not reduce head acceleration, it may have other influences on acute head and neck injuries. Moreover, all participants in our study felt comfortable heading the ball up to 12 times. Perhaps had we included novices, who had no/little experience heading the soccer ball, we may have seen a greater effect of the technique variables.

Groups applied similar head-to-torso and torso kinematic strategies to soccer heading (Figures 6, 7). These strategies indicate that prior to ball contact, during the pre-impact phase, the torso is extended [40]. Then during the ball contact phase, the torso is flexed forward to meet the ball [40]. Finally, during the follow-through phase the torso motion continues after contact [40]. However, coefficient of variations (6-36%) within groups for kinematic measures suggests that all participants are not applying the same soccer heading strategies. Furthermore, even within subjects, different soccer heading strategies are applied from header to header.

3.4.4 Muscle Activation Timing

During soccer heading, neck flexors and neck extensors contract to brace for ball impact. We measured right and left sternocleidomastoid and upper trapezius muscles activity because these muscles are the most superficial in the neck region and can be studied using surface electromyography (EMG). First, sternocleidomastoid muscles activate 280-500ms before ball contact in preparation for a header, with peak activity occurring 90-150ms before ball contact [40, 42]. Then, the right and left upper trapezius muscles are activated 100-300ms before ball contact and remain active throughout the follow-through [40, 42]. We identified similar onset and peak timing relative to ball contact for the sternocleidomastoid muscles, but herein, participants activated upper trapezius muscles much earlier than previously reported. This may be a result of differences in definition of muscle onset, or differences in heading strategy as a result of anthropometric and ball speed variations. Ultimately, high coefficient of variations (12-1300%) within groups for muscle activation timing suggests that all participants are not applying the same soccer heading strategies despite similar group means.

The activities of the sternocleidomastoid and upper trapezius have been suggested to contribute to head acceleration during purposeful soccer heading [40]. Using an external force applicator to apply a force to the head-neck segment, Tierney et al. identified higher peak activity and EMG area, and greater angular acceleration in females than in males [64]. This may seem counterintuitive, but Tierney et al. suggested that greater peak linear acceleration should stimulate a higher percentage of mechanoreceptors in the muscular and articular tissues of the cervical spine leading to a greater activation of motor neurons and more reflexive muscle firing [64]. Tierney et al. tested physically active male and female participants who had no experience with force being applied to their heads [64]. In soccer players, however, there does not appear to be a gender difference in sternocleidomastoid or upper trapezius activity or area [45, 53], which we corroborated. Perhaps soccer heading enhances head-neck segment dynamic restraint through neuromuscular adaptations [53]. Furthermore, we found that sternocleidomastoid and upper trapezius activity and area are not associated with head acceleration, although they were used as inputs in our significantly predictive combined regression model. It has been suggested that neck muscles do not have time to react during the impact to provide additional resistance [40].

The combined multiple regression model in this investigation was significant. We reported that 27% of the variance in peak linear acceleration and 26% of the variance in peak rotational acceleration was accounted for by employing this model. Moreover, size and strength regression models suggest that greater head and neck size and neck strength predict lower peak linear and rotational acceleration. Therefore, soccer players with smaller head masses and lower neck strength, may be at risk for greater head acceleration. These athletes may experience a theoretical “bobble-head” effect, when the neck strength is not great enough to control the mass of the head. Therefore, we suggest that anthropometric and neck strength measures should be considered when determining the minimum safe age to begin heading a soccer ball. For example, measuring the circumference of the neck (neck girth) and the strength of the sternocleidomastoid muscles may allow coaches, athletic trainers, and strength and/or conditioning staff to identify athletes who may benefit from a neck strengthening regimen. More conclusions, limitations, and future directions will be presented in Chapter 5.

Chapter 4

POSTURAL CONTROL AND VESTIBULAR/OCULAR MOTOR SCREENING CHANGES WITH REPEATED HEADING OF A SOCCER BALL

(Targeted for submission to *Brain Injury*)

4.1 Introduction

Soccer players may be at risk for long-term neurocognitive deficits associated with repeated heading of the soccer ball. Several studies have suggested that players who head the ball very frequently may have depressed neuropsychological performance [3, 5-11]; though others argue that these neuropsychological deficits result from multiple concussions throughout a career [20, 32]. Still others report no deficits in neuropsychological performance [21, 22, 24-28, 46, 88-90]. Similarly, evidence is mixed regarding deficits in postural control, increases in biochemical markers of brain damage, and changes in structural imaging of the brain. Some have reported deficits in postural control [120], increases in biochemical markers of brain damage [12, 13], and abnormal white matter microstructure [11, 16-18]; yet others refute these claims [14, 29, 91-93, 138, 139]. Unfortunately, many of these studies, including those showing neurological deficits and those that suggest otherwise, include small homogenous populations or report a highly significant practice effect for both heading and comparison groups. With over 3 million athletes registered through US Youth Soccer [84] and nearly 1 million participating at the high school level [140], it is unknown how repeated heading of the soccer ball affects these athletes. Investigating neurological changes over the course of a single acute bout of soccer heading affords researchers the ability to control for other factors that may affect neurological function.

Over a single acute bout of 10 soccer headers, Haran et al. [120] reported postural control deficits; while others [138, 139] have not identified postural control deficits following an acute bout of soccer headers. This dichotomy may be attributed to the conditions and metrics used in postural control assessment. Most commonly, measures of total sway [138], 95% area, and sway velocity [139] are calculated from center of pressure (COP) data. In contrast, Haran et al. [120] used root mean squared measures to quantify postural control. Higher root mean squared measures indicate greater variability, traditionally interpreted as greater postural instability. Therefore, there may be particular changes to neurological function that can be identified through different postural control metrics.

More recently, approximate entropy (ApEn) has been used as a measure of postural control, and has identified changes in postural control when other measures, such as area and velocity, did not [141-146]. Entropy is a unitless quantity that describes how likely a given pattern is to reappear within a time series. Entropy falls between 0 (sway following a very predictable pattern, such as a sine wave) and 2 (sway following a very random pattern, such as Gaussian noise). These equations are sensitive to sampling rate and input parameters, such as the length of data points observed and the similarity threshold between observations. Therefore, data are generally re-sampled at 10Hz for entropy calculations, and standard input parameters are used [147]. Cavanaugh et al. [144] suggested that an eyes open and an eyes closed bipedal stance is sufficient to identify concussion using entropy metrics. In concussion, it has been suggested that postural control deficits are caused by a sensory integration problem, whereby athletes are unable to use information from the visual and vestibular systems effectively [148]. Maintaining postural control without visual input is a more difficult task and may identify changes in postural control that are not apparent otherwise [120, 138, 139]. Although most postural control assessments are conducted for 20-30 seconds, it has been suggested that a minimum of 90-120 seconds should be examined to identify changes in postural control [141, 149]. It is unknown how purposeful soccer heading changes entropy measures with the eyes open and eyes closed over a two-minute interval.

In addition to changes in postural control, near point of convergence (NPC) increases following an acute bout of soccer heading in collegiate soccer players [150]. Convergence involves the adduction of the eyes by contracting the medial recuts muscles, which are controlled by cranial nerve III (oculomotor) [151]. The NPC measures the closest point to which one can maintain convergence while focusing on an object before diplopia occurs [151]. Following a series of 10 controlled soccer headers, the NPC increased by 2cm and this increase in the NPC persisted 24 hours [150]. The NPC measure is one component of the Vestibular Ocular Motor Screening (VOMS) test, which was developed to assess vestibular and ocular motor impairments via patient-reported symptom provocation after each assessment [152, 153]. While the full VOMS test has never been used to study the effects of repeated heading of the soccer ball, it is commonly used in concussion assessment [88, 152-157].

The purpose of this study was to investigate changes in postural control and vestibular/ocular function following an acute bout of purposeful soccer heading. We measured both 95% area, sway velocity, and ApEn. The 95% area and sway velocity both fall within a linear model, whereby the magnitude of the 95% area or velocity represent the intensity of the perturbation or severity of disruptions in the feedback loop [147, 158]. The ApEn falls within a nonlinear model, whereby the ApEn represents the organization of the underlying control system and higher-order processing [147, 158]. In quantifying both linear and non-linear metrics, we can speculate whether postural control impairments are related to disruptions in sensory feedback (linear) or higher-order processing deficits (non-linear). Because ApEn was better able to distinguished concussed individuals from non-concussed individuals than 95% area and sway velocity, we hypothesized that after an acute bout of heading, all groups would show changes in ApEn, but not 95% area or sway velocity. Consistent with previous research, we believed all groups would show an increase in NPC.

4.2 Methods

4.2.1 Participants

One-hundred soccer players and 60 control participants of similar age and gender were recruited to participate in this study (Table 16). Potential participants were excluded from this investigation if they had a self-reported history of neurologic disorder, cervical spine injury, head injury (i.e. concussion), or lower extremity injury in the past 6 months. If a participant sustained a concussion or any other injury in the past, they must have been cleared by a physician to return to full activity before participating in this study. Potential participants were included if they were currently participating in soccer within their age group and gender. Goal keepers were excluded. All participants provided written informed consent (IRB 476493-4). Children under the age of 18 years provided written informed assent, and their parent/legal guardian provided written informed parental consent.

Table 16: Subject Demographics. Mean ± Standard Deviation. There were no significant differences in age, height, or weight between soccer players and control participants. However, collegiate male control participants reported a greater number of previous concussions than collegiate male soccer players (p = .003).

	Soccer Players				Control Participants			
Group	N	Age (years)	Height (cm)	Weight (kg)	N	Age (years)	Height (cm)	Weight (kg)
Collegiate Male	20	20.8± 1.3	181.5± 5.8	77.3± 6.6	10	21.1± 0.9	179.2± 4.9	79.0±1 0.8
Collegiate Female	21	19.3± 1.1	166.8± 4.5	59.3± 3.4	10	20.4± 0.5	166.9± 5.2	63.1± 8.4
High School Male	14	16.9± 1.0	177.5± 7.3	68.9± 6.7	10	17.8± 0.4	174.4± 7.6	71.4± 15.2
High School Female	19	16.7± 0.9	163.0± 7.1	58.7± 6.7	10	18.0± 0.0	162.8± 6.1	59.1± 12.8
Youth Male	8	13.0± 0.9	166.0± 14.3	51.6± 13.6	10	13.0± 0.7	160.2± 7.3	49.5± 13.5
Youth Female	18	12.8± 0.9	156.0± 6.9	48.1± 8.4	10	13.2± 0.6	166.4± 14.4	64.6± 28.1

4.2.2 Concussion History Questionnaire

Participants completed a custom concussion history questionnaire, which also included sport-participation history. Participants were asked to report both sport-related and non-sport related concussions. In addition to medically-diagnosed concussions, four questions attempted to capture unrecognized and undiagnosed concussions, suspected concussions, and self-reported concussions [159, 160]. These questions included:

- Did you ever suffer from a concussion and not tell anyone?
- Have you ever been knocked out?

- Have you ever been “knocked silly/seen starts/bell rung/dinged” (confused/disoriented)?
- Have you ever been hit so hard you lost your memory?

4.2.3 Postural Control Assessment

The MobileMat® (Tekscan Inc., South Boston, MA) was used in postural control assessment. Data were sampled at 100Hz. Participants were asked to stand on the MobileMat® for 2 two-minute intervals with their hands on their hips and their feet together [141]. During one two-minute interval, participants had their eyes open. During the other two-minute interval, participants had their eyes closed. Using the center of pressure data, 95% area, sway velocity, and ApEn were calculated.

4.2.4 The Vestibular/Ocular Motor Screening (VOMS) Assessment

The VOMS was developed to assess vestibular/ocular motor impairments via patient-reported symptom provocation after each assessment [152, 153]. The VOMS was administered according to the description provided by Mucha et al. [152] and included tests of horizontal and vertical smooth pursuits, saccades, and vestibular ocular reflex (VOR), as well as near point of convergence, and visual motion sensitivity (VMS). After each test, patients self-reported symptoms, including: headache, dizziness, nausea, and fogginess on a scale of 0 to 10 [152]. Consistent with existing literature, the near point of convergence (NPC) assessment was done using a tongue depressor with the letter “T” in size 14-point font [152]. The participant started with an extended arm and slowly brought the tongue depressor toward the tip of their nose [152]. They were instructed to stop when they saw two distinct images or when there was an outward deviation of one eye observed by the principal investigator [152]. The NPC values were averaged across 3 trials [152].

4.2.5 Purposeful Soccer Heading Procedures

Participants performed a series of headers as described by Tierney et al. [45]. First, participants performed a neck warm-up consisting of 15 seconds of clockwise neck rotations, 15 seconds of counterclockwise neck rotations, 2 repetitions of stretching for 15 seconds in flexion and 15 seconds in extension, and 5-10 headers with a soft volleyball, tossed by the principal investigator. Soccer balls (size 5, 450g, inflated to 9psi) were projected using a JUGS soccer machine (JUGS, Tualatin, OR). The initial velocity was 11.2m/s (25mph), the angle of projection was 40°, and the range was approximately 12m (40 ft) [120]. The participants aimed for a target (1.2x1.8m) approximately 2m in front of them, as if taking a shot on goal. Participants performed a total of 12 standing headers. The first two headers were considered warm-up headers to allow the participants to become familiar with the protocol. Postural control and VOMS testing was done both before and immediately after the purposeful soccer headers. Control participants performed postural control testing before and immediately after a 15-minute wait period in which they sat quietly in the laboratory.

4.2.6 Data Analysis

We computed 95% area and sway velocity, which have been used to assess changes in postural control following an acute bout of heading [139], and M/L and A/P ApEn, which have identified postural control deficits following a concussion when other postural control metrics appeared normal [144, 145].

The 95% area is a measure of the area through which the COP passes and can be calculated by drawing an ellipse around intersecting lines that describe the maximum sway from anterior to posterior and from medial to lateral and quantifying the area of the ellipse [161]. The COP mean velocity was calculated by taking the total distance traveled (cm) and dividing it by the length of time (120s) of the trial [161].

The ApEn is a unitless quantity that describes how likely a given pattern is to reappear within a time series by determining the probability that if two series data points are similar for a length of m points, then they will remain similar at the next point [144, 145]. The COP data were re-sampled, using every third data point, at 10Hz for ApEn calculations, the length of data points, m, was set equal to 2, and the similarity between observations, r, was set equal to 0.2*standard deviation of the COP time series, which is consistent with typical ApEn analyses [144, 145]. ApEn was calculated in both the mediolateral (M/L) and anteroposterior (A/P) directions independently [144, 145].

4.2.7 Statistical Analysis

Postural control data were analyzed using multilevel linear models. The dependent variables were 95% area, sway velocity, M/L and A/P ApEn. All fixed-effects predictors were on the nominal level of measurement and were dummy coded. The fixed-effect predictors included age: youth, high school, and collegiate with collegiate coded as the base variable, gender: male or female with male coded as the base variable, group: soccer heading or control with soccer heading coded as the base variable, condition: eyes open or eyes closed with eyes open coded as the base variable, and concussion history: no history of concussion, 1 to 2 prior concussions, or

3 or more prior concussions, with no history of concussion coded as the base variable. Baseline scores were used as a covariate in the model. Regression coefficients were modeled using maximum likelihood estimation with variance components as the variance-covariance error structure.

In addition, a repeated measures ANOVA was used to compare NPC and a repeated measures multiple analysis of variance (MANOVA) was used to compare symptom scores. For all tests significance was defined by $p<0.05$.

In addition to group differences, we calculated the Reliable Change Index (RCI) for postural control and NPC measures to identify individual differences [162, 163].

Individual differences in symptom scores were also determined using clinical norms (> 2 symptoms) for any individual item in the VOMS assessment [152].

4.3 Results

4.3.1 Concussion History Questionnaire

Soccer players reported playing 11.4±4.1 years (range 1-17) and represented all field positions (33 defenders, 43 midfielders, and 24 forwards). Nearly 70% (69/100) of soccer players recorded soccer as their only sport of participation. Other sports recorded included track & field/cross country (11), basketball (9), tennis (4), lacrosse (4), volleyball (4), swimming (3), field hockey (2), football (1), softball (1), skiing (1), gymnastics (1), tae kwon do (1), and skateboarding (1). None of the soccer players reported wearing headgear when they play soccer. Eighteen soccer players reported a previous history of one or more diagnosed concussions (12, 1 concussion; 3, 2 concussions; 3, 3 or more concussions). Only 2 soccer players said that they suspected they had a concussion that they did not report; although, 17 soccer players reported loss of consciousness (2) or being knocked silly, seeing stars, having their bell rung, or being dinged (15) that were not diagnosed as a concussion (unrecognized concussions). When including unreported and unrecognized concussions, 31% of soccer players have sustained one or more concussions (17, 1 concussion; 6, 2 concussions; 8, 3 or more concussions). Table 17 presents means and standard deviations of years played and concussion history, including potentially unrecognized and unreported concussion, across groups.

Table 17: Years playing soccer and concussion history. Mean±Standard Deviation.

Group	Years Playing	Diagnosed Concussions	Potentially Unreported Concussions	Potentially Unrecognized Concussions	Total Concussions
Collegiate Male	14.6±2.0	0.3±0.7	0.0±0.0	0.2±0.5	0.5±0.8
Collegiate Female	15.0±1.2	0.5±1.3	0.0±0.2	0.3±0.5	0.8±1.8
High School Male	10.5±3.2	0.3±0.5	0.0±0.0	0.9±1.2	1.1±1.4
High School Female	9.9±3.0	0.4±0.8	0.1±0.2	0.2±0.5	0.6±0.8
Youth Male	8.8±1.0	0.1±0.4	0.0±0.0	0.1±0.4	0.3±0.5
Youth Female	6.7±2.7	0.0±0.0	0.0±0.0	0.2±0.4	0.2±0.4

4.3.2 Postural Control Assessment

Soccer players had significantly higher sway velocity post-heading than control participants during their follow up assessment after controlling for age, gender, condition, concussion history, and baseline values. Sway velocity, M/L ApEn, and

A/P ApEn was significantly higher with the eyes closed than with the eyes open. Youth and high school groups had significantly higher sway velocity than the collegiate group. Youth participants had significantly higher M/L ApEn than the collegiate participants. High school participants had significantly lower A/P ApEn than the collegiate participants. Table 18 present means and standard deviations for postural control variables (95% area, sway velocity, M/L ApEn, and A/P ApEn) for eyes open (EO) and eyes closed (EC) conditions. Table 19, 20, 21, and 22 present the results of the fixed effects from the multilevel linear models for 95% area, sway velocity, M/L entropy, and A/P entropy, respectively.

Table 18: Postural control measures across groups for the eyes open and eyes closed conditions. Means ± Standard Deviation.

	Eyes Open								**Eyes Closed**							
	Pre				Post				Pre				Post			
Group	95% Area (cm/s^2)	Velo-city (cm/s)	M/L Ap En	A/P Ap En	95% Area (cm/s^2)	Velo-city (cm/s)	M/L Ap En	A/P Ap En	95% Area (cm/s^2)	Velo-city (cm/s)	M/L Ap En	A/P Ap En	95% Area (cm/s^2)	Velo-city (cm/s)	M/L Ap En	A/P Ap En
College Soccer Male	7.3± 2.8	2.13± 0.54	.34± .10	.48± .09	8.3± 9.7	2.26± 0.57	.35± .11	.48± .11	9.9± 4.3	2.66± 0.66	.46± .13	.57± .18	11.1± 3.9	2.68± 0.62	.44± .14	.60± .14
College Soccer Female	7.9± 5.4	2.13± 0.56	.28± .10	.49± .13	9.7± 7.1	2.52± 0.94	.32± .15	.51± .16	10.8± 7.1	2.56± 0.89	.40± .15	.57± .14	9.7± 5.4	2.65± 0.67	.39± .13	.58± .18
High School Soccer Male	6.0± 2.6	3.08± 0.86	.34± .15	.49± .18	9.1± 6.9	3.23± 0.71	.39± .17	.52± .14	8.8± 4.5	3.41± 0.83	.49± .18	.54± .14	11.5± 8.3	3.50± 0.73	.47± .18	.56± .13
High School Soccer Female	7.5± 7.5	2.75± 0.76	.30± .14	.44± .17	6.1± 3.6	2.67± 0.65	.33± .18	.42± .18	8.5± 4.1	2.85± 0.66	.43± .56	.51± .17	10.1± 7.1	2.88± 0.58	.38± .17	.47± .17
Youth Soccer Male	10.5± 4.8	3.76± 1.98	.55± .14	.62± .19	9.7± 3.6	2.96± 0.82	.54± .13	.66± .20	13.1± 5.5	3.63± 1.61	.56± .14	.63± .18	10.2± 3.7	3.39± 0.91	.64± .13	.69± .21
Youth Soccer Female	8.1± 6.9	3.44± 0.76	.40± .11	.54± .13	8.3± 6.5	3.58± 0.77	.37± .12	.54± .14	10.2± 5.2	3.89± 0.74	.55± .14	.65± .12	9.5± 4.7	3.87± 0.70	.52± .14	.62± .12
College Control Male	9.2± 7.3	1.73± 0.62	.39± .23	.52± .25	10.2± 9.5	1.76± 0.81	.40± .17	.57± .21	14.2± 13.8	2.37± 0.79	.56± .31	.61± .23	11.2± 11.9	2.14± 0.75	.59± .29	.71± .26
College Control Female	6.8± 8.0	1.47± 0.41	.37± .16	.54± .14	5.6± 2.2	1.30± 0.21	.35± .14	.59± .17	8.0± 3.9	1.73± 0.37	.43± .17	.62± .14	8.9± 5.7	1.67± 0.48	.45± .16	.63± .13

High School Control Male	13.3± 6.8	2.40± 0.43	.42± .17	.52± .12	10.4± 7.7	2.31± 0.72	.33± .16	.48± .09	16.0± 11.7	2.77± 0.65	.50± .11	.56± .11	14.9± 10.4	2.96± 0.89	.46± .10	.52± .15
High School Control Female	7.4± 5.1	2.57± 1.05	.40± .11	.54± .12	8.1± 4.5	2.66± 1.11	.40± .14	.54± .13	9.7± 3.8	3.10± 1.19	.45± .11	.58± .08	11.1± 6.0	2.80± 1.19	.45± .14	.61± .09
Youth Control Male	14.4± 9.0	2.66± 0.49	.43± .08	.53± .11	15.0± 9.4	2.86± 0.48	.46± .13	.52± .11	12.2± 6.2	2.64± 0.32	.51± .11	.56± .08	14.4± 9.9	2.86± 0.56	.53± .13	.58± .10
Youth Control Female	8.7± 3.7	2.2± 0.47	.42± .13	.51± .12	12.4± 6.4	2.7± 0.53	.51± .10	.59± .08	11.2± 4.8	2.4± 0.45	.42± .14	.56± .14	9.9± 4.3	2.6± 0.50	.52± .12	.63± .11

Table 19: Fixed effects from the multilevel linear models for 95% area. The fixed effects for age, gender, group, or concussion history were not significant.

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	3.512	0.789	210.236	4.449	0.000	1.956	5.068
Youth	0.266	0.742	146.748	0.358	0.721	-1.201	1.733
High School	0.473	0.713	148.096	0.663	0.508	-0.936	1.882
Gender	-0.953	0.605	151.686	-1.574	0.118	-2.149	0.243
Group	0.057	0.630	148.581	0.090	0.928	-1.188	1.302
1-2 Concussions	-0.265	0.715	188.307	-0.371	0.711	-1.676	1.145
3+ Concussions	1.195	1.031	150.651	1.159	0.248	-0.843	3.233
Condition	0.363	0.480	156.873	0.757	0.450	-0.584	1.310
Pre	0.672	0.046	277.277	14.543	0.000*	0.581	0.763

Table 20: Fixed effects from the multilevel linear models for sway velocity. The fixed effects for both age and group were significant, whereby youth and high school groups had higher sway velocity than collegiate groups and control participants had lower sway velocity than soccer players post-heading.

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	1.128	0.144	224.502	7.814	0.000	0.844	1.413
Youth	0.414	0.119	168.829	3.478	0.001*	0.179	0.649
High School	0.239	0.109	161.887	2.194	0.030*	0.024	0.453
Gender	-0.007	0.087	163.013	-0.083	0.934	-0.179	0.165
Group	-0.291	0.097	162.559	-3.002	0.003*	-0.482	-0.100
1-2 Concussions	-0.056	0.097	244.004	-0.575	0.566	-0.246	0.135
3+ Concussions	0.056	0.150	155.806	0.370	0.712	-0.241	0.352
Condition	0.130	0.048	180.967	2.741	0.007*	0.037	0.224
Pre	0.545	0.047	262.255	11.620	0.000*	0.453	0.638

Table 21: Fixed effects from the multilevel linear models for M/L entropy. The fixed effect for youth was significant, whereby youth participants had higher M/L ApEn than collegiate participants.

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	.115	.030	224.853	3.843	.000	.056	.174
Youth	.052	.024	157.902	2.159	.032*	.004	.099
High School	-.009	.023	155.266	-.387	.700	-.053	.036
Gender	-.013	.019	160.865	-.669	.505	-.051	.025
Group	.027	.020	154.471	1.337	.183	-.013	.066
1-2 Concussions	.022	.023	195.222	.968	.334	-.023	.066
3+ Concussions	.030	.033	158.018	.924	.357	-.034	.095
Condition	.036	.016	196.010	2.236	.026*	.004	.067
Pre	.649	.059	252.186	10.960	.000*	.532	.765

Table 22: Fixed effects from the multilevel linear models for A/P entropy. The fixed effects for high school was significant, whereby high school participants had lower A/P ApEn than collegiate participants.

Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Intercept	.172	.036	214.899	4.721	.000	.100	.244
Youth	.019	.023	150.511	.823	.412	-.027	.065
High School	-.045	.022	152.728	-2.013	.046*	-.089	-.001
Gender	-.001	.019	156.311	-.058	.954	-.038	.036
Group	.031	.019	151.055	1.591	.114	-.007	.069
1-2 Concussions	.011	.022	199.560	.483	.629	-.033	.054
3+ Concussions	.046	.032	155.258	1.424	.156	-.018	.109
Condition	.030	.014	177.590	2.147	.033*	.002	.058
Pre	.675	.061	227.695	11.047	.000*	.554	.795

4.3.3 The Vestibular/Ocular Motor Screening (VOMS) Assessment

On average, NPC increased (worsened) from pre- to post- across all groups, although the 0.4cm increase from pre- to post- may be clinically insignificant. Results of the repeated measures ANOVA for NPC reveal the time by group interaction was not significant ($F = 1.162$, $p = .319$). However, both the main effect for time and the main effect for group were significant (respectively, $F = 12.260$, $p = .001$; $F = 2.676$, $p = .004$). Post-hoc analyses revealed that youth male soccer players have a higher NPC than all other groups ($p < .01$). Table 23 presents means and standard deviations for the NPC assessment.

Table 23: NPC assessment. Mean ± Standard Deviation.

Group	Pre (cm)	Post (cm)
Collegiate Soccer Male	2.5±4.2	2.4±4.0
Collegiate Soccer Female	0.6±1.4	1.0±2.7
High School Soccer Male	2.6±3.4	3.5±4.6
High School Soccer Female	1.8±1.6	2.2±2.1
Youth Soccer Male	6.6±7.8	7.9±8.6
Youth Soccer Female	0.7±1.8	1.2±2.4
Collegiate Control Male	2.8±4.3	2.9±4.8
Collegiate Control Female	2.1±3.2	2.3±3.5
High School Control Male	2.5±3.5	2.5±3.2
High School Control Female	1.6±2.0	2.1±2.7
Youth Control Male	1.9±1.7	2.1±2.2
Youth Control Female	0.2±0.4	0.2±0.4
Overall	1.9±3.4	2.3±3.8

Table 24 presents means and standard deviations for the symptom scores. Results of the repeated measures MANOVA for symptom scores reveal the time by group interaction was not significant ($F = 1.066$, $p = .335$). Neither the main effect for time nor the main effect for group were significant (main effect for time: $F = .841$, $p = .555$; main effect for group: Baseline, $p = .669$; Smooth Pursuits, $p = .177$; Horizontal

Saccades, p = .241; Vertical Saccades, p = .233; Horizontal VOR, p = .242; Vertical VOR, p = .217; VMS, p = .188).

4.3.4 Reliable Change Index

The Reliable Change Index (RCI) was calculated for each of the postural control measures and NPC. Using the RCIs presented in Table 25, we identified individuals who exhibited an increase (Table 26) or decrease (Table 27) in one of the postural control measures or NPC. An increase in 95% area, sway velocity, or NPC is traditionally interpreted as a worsening[139, 150]. Because ApEn is thought to represent an inverted U-curve, where both extreme highs and extreme lows are considered poor outcomes, we interpreted both an increase and a decrease in M/L and A/P ApEn greater than the RCI as a deficit [147, 164, 165]. Within the soccer heading group, 31% of participants presented with a deficit in one or more postural control measures or NPC (10% of male collegiate soccer players, 33% of female collegiate soccer players, 57% of male high school soccer players, 42% of female high school soccer players, 50% of male youth soccer players, 11% of female youth soccer players). Within the control group, 17% of participants presented with a deficit in one or more of the postural control measures or NPC (40% of male collegiate controls, 10% of female collegiate controls, 0% of male high school controls, 30% of female high school controls, 20% of male youth controls, 0% of female youth controls). The NPC showed the greatest number of deficits (increase in NPC), in which 10% of soccer players and 5% of control participants presented with an increase in NPC.

Table 24: Symptom scores for each VOMS item.

Group	Pre Base-line	Post Base-line	Pre Smooth Pursuit	Post Smooth Pursuit	Pre Hori-zontal Saccade	Post Hori-zontal Saccade	Pre Vertical Saccade	Post Vertical Saccade	Pre Hori-zontal VOR	Post Hori-zontal VOR	Pre Vert-ical VOR	Post Vert-ical VOR	Pre VMS	Post VMS
College Soccer Male	.3±.8	.1±.3	.3±.8	.1±.3	.4±.8	.2±.4	.4±.8	.3±.7	.8±1.3	.5±1.1	.5±1.1	.2±.5	.3±.8	.1±.3
College Soccer Female	.4±1.4	.4±1.4	.4±1.4	.4±1.4	.4±1.4	.4±1.4	.4±1.4	.4±1.4	.5±1.5	.4±1.4	.4±1.4	.4±1.4	.4±1.4	.4±1.4
High School Soccer Male	.0±.0	.3±1.1	.0±.0	.3±1.1	.0±.0	.8±2.1	.0±.0	.8±2.1	.0±.0	.6±1.4	.0±.0	.7±1.6	.0±.0	.6±1.4
High School Soccer Female	.0±.0	1.0±2.2	.0±.0	1.1±2.5	.0±.0	1.1±2.5	.0±.0	1.0±2.4	.2±.7	1.0±2.4	.2±.7	1.0±2.4	.2±.7	1.0±2.4
Youth Soccer Male	.4±1.1	.1±.4	.4±1.1	.1±.4	.4±1.1	.1±.4	.4±1.1	.1±.4	.0±.0	.1±.4	.0±.0	.1±.4	.0±.0	.1±.4
Youth Soccer Female	.0±.0	.4±1.3	.0±.0	.4±1.3	.0±.0	.5±1.5	.0±.0	.4±1.3	.0±.0	.4±1.3	.0±.0	.4±1.3	.0±.0	.4±1.3
College Control Male	.5±.8	.7±1.6	.5±.8	.7±1.6	.5±.8	.8±1.6	.6±.8	.7±1.6	.6±1.0	.9±1.6	.6±1.0	.8±1.6	.8±1.1	.7±1.6
College Control Female	.8±1.7	.9±1.9	.8±1.7	1.0±2.1	.8±1.7	.9±1.9	.8±1.7	.9±1.9	.9±1.9	1.2±2.5	.9±1.9	1.4±3.1	.8±1.7	.9±1.9

High School Control Male	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0
High School Control Female	.6±1.9	.5±1.3	2.2±5.2	1.7±3.8	2.2±5.2	1.7±3.8	2.2±5.2	1.7±3.8	2.2±5.2	1.7±3.8	2.2±5.2	1.7±3.8	2.2±5.2	1.7±3.8
Youth Control Male	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0
Youth Control Female	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0	.0±.0

Table 25: The reliability statistics and RCIs for postural control measures and NPC.

	ICC	SD_1	SEM_1	SD_2	SEM_2	SE_{DIFF}	RCI
EO 95% Area	0.667	5.852	3.377	6.265	3.615	4.947	9.696
EO Sway Velocity	0.757	0.973	0.480	0.912	0.450	0.657	1.288
EO M/L ApEn	0.696	0.143	0.079	0.148	0.082	0.113	0.222
EO A/P ApEn	0.731	0.146	0.076	0.155	0.080	0.110	0.216
NPC	0.967	3.344	0.607	3.383	0.615	0.864	1.694
EC 95% Area	0.680	6.436	3.641	6.594	3.730	5.212	10.216
EC Sway Velocity	0.709	0.953	0.514	0.883	0.476	0.701	1.374
EC M/L ApEn	0.551	0.172	0.115	0.167	0.112	0.161	0.315
EC A/P ApEn	0.502	0.147	0.104	0.158	0.111	0.152	0.289

ICC = Interclass Correlation Coefficient
SD = Standard Deviation

SEM = Standard Error of Measurement
RCI = Reliable Change Index
EO = Eyes Open
EC = Eyes Closed

Table 26: Percentage of participants within each group who experienced an increase in each of the outcome measures. An increase in 95% area, sway velocity, ApEn, and NPC can be interpreted as a deficit.

	Soccer						**Control**					
	M Collegiate	**F Collegiate**	**M HS**	**F HS**	**M Youth**	**F Youth**	**M Collegiate**	**F Collegiate**	**M HS**	**F HS**	**M Youth**	**F Youth**
EO 95% Area	5%	10%	14%	0%	0%	0%	10%	0%	0%	0%	0%	0%
EO Sway Velocity	0%	10%	7%	0%	0%	0%	0%	0%	0%	0%	0%	0%
EO M/L ApEn	0%	5%	14%	16%	0%	0%	10%	0%	0%	0%	0%	0%
EO A/P ApEn	0%	5%	14%	11%	0%	0%	10%	10%	0%	0%	0%	0%
NPC	0%	5%	21%	11%	25%	11%	10%	0%	0%	20%	0%	0%
EC 95% Area	0%	5%	14%	11%	0%	0%	0%	0%	0%	0%	0%	10%
EC Sway Velocity	0%	0%	14%	0%	13%	0%	0%	0%	0%	10%	0%	0%
EC M/L ApEn	0%	0%	0%	5%	0%	0%	10%	0%	0%	10%	0%	0%
EC A/P ApEn	5%	0%	7%	5%	0%	0%	10%	0%	0%	10%	0%	0%

M = Male, F = Female, HS = High School
EO = Eyes Open
EC = Eyes Closed

Table 27: Percentage of participants within each group who experienced a decrease in each of the outcome measures. A decrease in ApEn can be interpreted as a deficit.

	Soccer						Control					
	M Collegiate	**F Collegiate**	**M HS**	**F HS**	**M Youth**	**F Youth**	**M Collegiate**	**F Collegiate**	**M HS**	**F HS**	**M Youth**	**F Youth**
EO 95% Area	0%	0%	0%	5%	0%	0%	0%	0%	10%	0%	0%	10%
EO Sway Velocity	0%	0%	7%	0%	13%	0%	0%	0%	0%	0%	0%	0%
EO M/L ApEn	0%	0%	7%	5%	13%	0%	0%	0%	0%	0%	0%	0%
EO A/P ApEn	0%	0%	0%	5%	0%	0%	0%	0%	0%	0%	0%	0%
NPC	5%	0%	7%	5%	0%	0%	0%	0%	0%	10%	0%	0%
EC 95% Area	0%	10%	0%	0%	13%	0%	20%	0%	0%	0%	10%	0%
EC Sway Velocity	0%	5%	7%	0%	13%	0%	10%	0%	0%	0%	0%	0%
EC M/L ApEn	0%	0%	7%	0%	0%	0%	10%	0%	0%	0%	0%	0%
EC A/P ApEn	0%	0%	0%	5%	0%	0%	10%	0%	0%	0%	0%	0%

M = Male, F = Female, HS = High School
EO = Eyes Open
EC = Eyes Closed

4.3.5 Individual Differences in Symptom Scores and VOMS Items

At baseline, 7 soccer players and 5 control participants presented with abnormal symptom scores (> 2) in one or more VOMS item (soccer players: 3 collegiate males (15%), 2 collegiate females (9.5%), 1 high school female (5.3%), and 1 youth male (12.5%); control participants: 1 collegiate male (10%), 2 collegiate females (20%), 2 high school females (20%)). After soccer heading, two of the three collegiate male soccer players reported a decrease (0) in symptoms. In addition to those soccer players who reported abnormal symptom scores at baseline, 2 high school males (14.3%), 2 high school females (10.5%), and 2 youth females (11.1%) presented with abnormal symptom scores following soccer heading. There were no changes in symptom scores for control participants (i.e. no control participants with a baseline symptom score <2 reported a symptom score >2 at the time of follow up testing).

At baseline, 7 soccer players and 3 control participants presented with abnormal smooth pursuits (soccer players: 2 collegiate males (10%), 3 collegiate females (14.3%), and 2 youth females (11.1%); control participants: 1 collegiate male (10%), 1 high school male (10%), and 1 youth male (10%)). In addition to those participants who presented with abnormal smooth pursuits at baseline, 2 high school (14.3%) and 2 youth (25%) males presented with abnormal smooth pursuits after soccer heading. No other VOMS items appeared abnormal.

4.4 Discussion

The acute effects of repeated purposeful soccer heading are unclear [14, 29, 46, 91, 92, 120, 138, 139, 150]. The purpose of this study was to investigate changes in postural control and vestibular/ocular function following an acute bout of purposeful soccer heading across youth, high school, and collegiate male and female soccer players. The primary finding was that sway velocity increased among soccer players post-heading relative to the control group. There were no other observed group differences in postural control or vestibular/ocular function. Using RCIs, Individual differences were detected among both the soccer heading (31%) and control groups (17%). Our results suggest that soccer heading may impair sway velocity and there may be some individuals who experience more pronounced deficits in postural control and vestibular/ocular function immediately following an acute bout of soccer heading. It is still unknown if these changes are clinically significant, how long they persist, or if they manifest over a career of soccer.

We observed higher sway velocity post-heading in our soccer players relative to control participants. The magnitude of the sway velocity represents the intensity of the perturbation or severity of disruptions in the feedback loop and relies heavily on visual and vestibular input [147, 158, 166, 167] and vestibular dysfunction following a repeated soccer heading model has been previously established [168]. Specifically, vestibular processing is disrupted from repeated soccer heading, which causes a reduced response to galvanic vestibular stimulation (GVS) [168]. Increased sway velocity, in the absence of other postural control impairments, may be related to impairments in the vestibular system [166], whereby transient dysfunction in the vestibular system creates a greater need to control center-of-mass displacements and thus reflects higher sway velocity and poorer postural control [166].

In collegiate and professional soccer players, sway area (eyes open = 0.86-1.71cm^2, eyes closed = 0.96-1.87cm^2) and velocity (eyes open = 0.5-1.9cm/s, eyes closed = 0.6-4.1cm/s) have been reported for to 30-50s [169, 170]. Although we observed greater 95% area and sway velocity than previously reported, we also recorded postural control for a longer period of time (120s), which is consistent with previous research [141]. We identified group differences in sway velocity, whereby soccer players had higher sway velocity post-heading relative to the control group, when controlling for age, gender, concussion history, condition, and baseline sway velocities. Similarly, Haran et al. identified deficits in postural control measures following 10 controlled soccer headers [120]. In contrast, Schmitt et al. reported no differences immediately post heading in COP area or sway velocity [139]. Our study differed from that of Schmitt et al. in that we included youth, high school, and collegiate soccer players, whereas Schmitt et al. only included collegiate soccer players. Additionally, our soccer heading model involved 12 headers in 12 minutes, similar to that of Haran et al., whereas Schmitt et al. asked participants to complete 18 headers in 40 minutes [120, 139]. Perhaps longer rest time between headers may minimize the effects of a repetitive heading model.

Cavanaugh et al. [144] identified changes in ApEn following concussion, particularly among an eyes open and an eyes closed bipedal stance. During the eyes open bipedal stance, AP and ML ApEn baseline values range from .81 to .89 and from 1.01 to 1.07, respectively, and decrease within the first 48 hours following concussion [144, 145]. Similarly, during the eyes open bipedal stance, AP and ML ApEn baseline values range from .75 to .84 and from .90 to 1.03, respectively, and decrease within the first 48 hours following concussion [144, 145]. Though we reported no group differences in ApEn following an acute bout of soccer heading, we did observe some individual changes in ApEn. Within the soccer heading group, 9% and 6% experienced an increase/decrease in M/L or A/P ApEn, respectively, with the eyes open and 2% and 4%, respectively, with the eyes closed. The ApEn represents the organization of the underlying control system and higher-order processing [147, 158]. Therefore, we can speculate that soccer players, in general, do not experience a disruption in higher-order processing from a single acute bout of soccer headers.

The VOMS is commonly used in concussion assessment [88, 152-157]. Healthy athletes 9-18 years old, report 0-2 symptoms following any individual VOMS item with an average of 0.1±0.3 [152]. Following concussion, symptoms increase (average 2.1-3.7 per VOMS item) [152]. Healthy collegiate athletes experience an average of 0.35-0.41 symptoms per VOMS item with 6-8% of athletes reporting above the clinical norm (> 2 symptoms) [88]. At baseline, 7% of our soccer players reported above the clinical norm in one or more VOMS item and after soccer heading, 10% of soccer players reported above the clinical norm. This increase in symptom presentation during the VOMS assessment suggests that although group differences in symptom scores were not observed, there may be some individuals who experience symptom exacerbation and these symptoms may be elicited by testing vestibular/ocular function. Similarly, McAllister et al. identified a subgroup of athletes who experienced learning and memory deficits following repetitive subconcussive head impacts in the absence of group differences [171].

NPC is one component of the VOMS. In healthy youth, high school, and collegiate athletes, an NPC range of 0-15.3cm has been observed, with averages between 1.9 and 5.0cm [88, 94, 152, 154], which is similar to our averages (pre = 1.9±3.4cm, post = 2.3±3.8cm). NPC increases (average 5.9±7.7cm) following concussion [88, 152] and as many as 50% of athletes suffer convergence insufficiency after concussion [154]. In addition to deficits in NPC observed following concussion, NPC also appears to increase following repetitive head impacts in both football [94] and soccer [150]. In football players, only those exposed to higher peak linear and rotational accelerations experienced a worsening in NPC, and this observed deficit recovered by the end of the season [94]. We observed increases in NPC in both the soccer heading and control groups. While it is unclear why the control group may have increased NPC following a 15-minute wait period, it is possible that as control participants completed the task again, they began to experience fatigue, which has been shown to increase NPC [172]. Using RCIs, we identified a greater percentage of soccer players (10%) that experienced an increase in NPC relative to the control group (5%), suggesting that soccer heading may increase NPC. This increase in NPC may be a result of oculomotor nerve dysfunction, as the oculomotor nerve controls the medial rectus muscle, which is used in convergence [150, 151, 173]. This impairment in NPC may be associated with eyestrain, headaches, blurred vision, double vision, sleepiness, difficulty concentrating, movement of print while reading, and loss of comprehension after short periods of reading or performing close activities in the long-term [172], which would be especially problematic for student-athletes. However, NPC can be improved with targeted vision therapy, if we can recognize the athletes who may benefit from this type of treatment [173].

Although age, gender, and prior concussion history are considered potential modifying factors in concussion management [174-179], they do not appear to be significant modifying factors in post-heading postural control and vestibular/ocular function assessment. Therefore, individual difference observed may be related to a variety of factors, including: genetics, comorbidities and premorbidities (i.e. migraine history, attention deficit hyperactivity disorder (ADHD), etc.), or medication, which were not controlled for [174, 180, 181].

In conclusion, sway velocity increases post-heading relative to control participants. In addition, there are some individuals who may experience other postural control impairments and vestibular/ocular function deficits, in the absence of group differences. More conclusions, limitations, and future directions will be presented in Chapter 5.

Chapter 5

CONCLUSION

The overarching theme of this research effort was to compare head acceleration during purposeful soccer heading across age and gender, determine what factors predict higher head acceleration values, and investigate changes in vestibular/ocular function and postural control following purposeful soccer heading. At the collegiate and high school levels, female soccer players have higher head accelerations than their male counterparts, suggesting that if females are experience the same number of purposeful headers, females may be exposed to greater cumulative head accelerations from repeated heading of a soccer ball over a career of soccer. Greater head size, neck size, and neck strength predicted lower peak linear and rotational acceleration, which may contribute to the observed gender differences. Sway velocity increased post-heading relative to control participants independent of age, gender, and concussion history, and may result from vestibular dysfunction with repetitive heading.

The combined multiple regression model in this investigation was significantly predictive, including parameters of head mass, neck girth, sternocleidomastoid and upper trapezius strength, head-to-torso and torso-to-laboratory floor range-of-motion, and sternocleidomastoid and upper trapezius peak amplitude and peak area. We reported that 27% of the variance in peak linear acceleration and 26% of the variance in peak rotational acceleration was accounted for by employing this model. Although there were no statistically-significant, unique predictors of either peak linear or rotational acceleration in the combined model, size (head mass and neck girth) and strength (sternocleidomastoid and upper trapezius strength) regression models were also significant, accounting for 22% and 13% of peak linear acceleration and 23% and 17% of peak rotational acceleration, respectively. These results suggest that soccer players with smaller head masses and lower neck strength, may be at risk for greater head acceleration. These athletes may experience a theoretical "bobble-head" effect, when the neck strength is not great enough to control the mass of the head. Therefore, we suggest that anthropometric and neck strength measures should be considered when determining the minimum safe development time to begin heading a soccer ball. Perhaps a neck strength assessment could be administered before the start of the soccer season to identify those athletes with lower neck strength who may benefit from a neck strengthening regiment, although future research should identify what cervical resistance training programs are most beneficial in decreasing head acceleration, particularly among female soccer players.

Sway velocity increases post-heading relative to control participants. However, no other group differences were observed for postural control or vestibular/ocular function assessments. Increased sway velocity, in the absence of other postural control impairments, may be related to impairments in the vestibular system [166], whereby damage to the vestibular system creates a greater need to control center-of-mass displacements and thus reflects higher sway velocity and poorer postural control [166]. It is still unknown how long these deficits persist or if they manifest over the course of a career.

It is unclear why individual differences in postural control measures and NPC were observed in control participants, but on average, more soccer players (31%) presented with individual differences than control participants (17%), suggesting that there are some individuals who may experience other postural control impairments and vestibular/ocular function deficits after heading, in the absence of group differences. These individual differences may be related to a variety of factors, including: genetics, comorbidities and premorbidities, or medication, which were not controlled for [174, 180, 181].

5.1 Limitations

The conclusions presented herein are reliant on the accuracy of the sensor technology. We used the SIM to quantify peak linear and rotational acceleration. While, Wu et al. reported lower accuracy of head accelerometers embedded in tight-fitting elastic caps than head accelerometers embedded mouth guards [182], Karton et al., observed no significant difference in peak linear or rotational acceleration between SIM and instrumented headform at 30g and 50g impact energy levels (Karton, 2016, in press). At 80g, there was a significant difference in peak linear and rotational acceleration between SIM and instrumented headform, such that the SIM readings showed, on average, higher peak linear and rotational accelerations than those of the headform (Karton, 2016, in press). Furthermore, correlation coefficient results showed that a strong positive relationship exists between the headform and SIM peak linear and rotational accelerations with Pearson's $r>0.9$ (Karton, 2016, in press). These data suggest that at lower energy levels (30g and 50g), the SIM sensor is accurate at reporting peak linear and rotational accelerations, but at higher energy levels, the SIM overestimates peak linear and rotational accelerations. While this may be of concern, only 15 of our 1,000 (1.5%) observed impacts were 80g or higher and only 89 of our 1,000 (8.9%) observed impacts were greater than 60g. Because ball contact only occurs over a 15-20ms window, it was not possible to use kinematic data to quantify head acceleration.

We limited our regression model to 10 predictors. There are other measures, such as symmetrical neck strength [51], neck length [45, 53], neck stiffness [45, 53], and neck volume, which were not incorporated in our regression model. Although including all possible predictors in the model could have made the model more robust, our population of 100 participants limited our included predictors to still maintain our large statistical power and effect size.

Neck strength measurements were not randomized and this may have biased the results because neck flexor strength was always tested first, followed by anterolateral neck flexor (sternocleidomastoid), cervical rotator, posterolateral neck extensor, and upper trapezius muscle strength. Additionally, with manual muscle testing, the subject's true strength may not be captured if the subject's ability to generate force is greater than the resistance provided by the tester [183]. However, manual muscle testing does not differ from the gold standard, fixed frame dynamometry (Catenaccio, 2016, in review).

EMG muscle activity resulted in large variations across and within participants and were limited to only two muscles, sternocleidomastoid and upper trapezius. Though variable, EMG muscle activity does provide a qualitative assessment of timing and relative magnitude of activity. In the cervical region, muscles are close in proximity and relatively small. The measurement of additional muscles would provide a better understanding of the overall muscle activity, but measurement of internal muscles would be difficult without fine-wire EMG, which is not common in this region. Finally, we normalized the muscle activity to maximum voluntary isometric contraction, but soccer heading measured higher dynamic activity than observed under maximum voluntary isometric contraction, which was consistent with previous literature [40]. Thus, there may be more effective ways to normalize muscle activity in the future.

Participants only performed straight-standing headers with a pronounced re-direction at a target 2m in front of them. Heading with various ball targets [40] and approaches [42] may result in varying peak linear and rotational accelerations, kinematics, and muscle activation strategies. We chose to limit this study to straight-standing headers to control ball contact as best as possible. Although the heading scenarios were controlled in terms of the ball velocity and target location, the ball rebound velocity was likely variable. We could not quantify ball rebound speed because we were limited in our motion capture frame rate. Even under these controlled conditions, the coefficient of variation for peak linear accelerations was in the range of 22-37% and can be attributed to differences in anthropometric and neck strength variations, variations in kinematics and neck muscle activity, and potentially vibrations in the accelerometer.

We had a large sample size of 100 soccer players across 6 groups: youth, high school, and collegiate male and female. However, all collegiate and most youth and high school female soccer players came from one team. High school and collegiate male soccer players came from 3 and 2 teams, respectively. Therefore, although the sample size was large, we may not have had sufficient variation within each of the 6 groups regarding kinematics as many players within each group had the same coach and likely received the same heading training regarding technique.

Finally, postural control and vestibular/ocular function assessments were only completed immediately post soccer heading. Thus, we do not know how long increased sway velocity persisted, or if any delayed impairments manifested as a result of the potential alterations in the neurometabolic cascade following repeated subconcussive head impacts. We also limited postural control assessment to quiet stance only; future research should investigate dynamic balance, which may be more important in sport.

5.2 Clinical Implications and Future Directions

Female soccer players may be exposed to greater cumulative head accelerations from repeated heading of a soccer ball over a career of soccer. However, the frequency of heading among female soccer players relative to their male counterparts is unknown. Therefore, future research should investigate the frequency and magnitude of head acceleration among male and female soccer players across age and level of play. Additionally, incorporating neurological assessments, such as postural control, blood biomarker, neuropsychological, and imaging will allow us to understand the effects of repeated head impacts in sport over a career of participation. Greater head and neck size and neck strength predicted lower peak linear and rotational acceleration and may contribute to the observed gender differences. Future research should determine appropriate strength training regimens for athletes across age and gender. Sway velocity increases post-heading relative to control participants independent of age, gender, and concussion history. It is still unknown how long this deficit persists or if it manifests over a career, so future research should study the longitudinal effects of repetitive head impacts on both postural control and other neurological assessments. Finally, in the absence of group differences in postural control and vestibular/ocular function, individual impairments were reported. Future research should investigate modifying factors, such as genetics, in identifying those individuals who may suffer deficits with repeated heading.

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Appendix A

IRB APPROVAL

UNIVERSITY OF DELAWARE

RESEARCH OFFICE

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DATE:	July 21, 2015
TO:	Jaclyn Caccese
FROM:	University of Delaware IRB
STUDY TITLE:	[476493-4] Investigation of Linear and Rotational Acceleration During Purposeful Heading in Youth, High School, and Collegiate Female Soccer Players
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
APPROVAL DATE:	July 21, 2015
EXPIRATION DATE:	July 22, 2016
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(4)*

Thank you for your submission of Continuing Review/Progress Report materials for this research study. The University of Delaware IRB has APPROVED your submission. This approval is based on an appropriate risk/benefit ratio and a study design wherein the risks have been minimized. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on the applicable federal regulation.

Please remember that informed consent is a process beginning with a description of the study and insurance of participant understanding followed by a signed consent form. Informed consent must continue throughout the study via a dialogue between the researcher and research participant. Federal regulations require each participant receive a copy of the signed consent document.

Please note that any revision to previously approved materials must be approved by this office prior to initiation. Please use the appropriate revision forms for this procedure.

All SERIOUS and UNEXPECTED adverse events must be reported to this office. Please use the appropriate adverse event forms for this procedure. All sponsor reporting requirements should also be followed.

Please report all NON-COMPLIANCE issues or COMPLAINTS regarding this study to this office.

Please note that all research records must be retained for a minimum of three years.

- 1 -   Generated on IRBNet

Consent forms, and parental consent forms and child assent forms, which were used for participants under the age of 18, are available upon request.

Appendix B

COPYRIGHT INFORMATION FOR CHAPTER 1

Copyright information: Sports Medicine

Copyright Information

For Authors

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