

# **Quantifying Postural Control, Concussion Risk, and Helmet Performance in Youth Football**

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## **Abstract**

As many as 1.9 million sports-related concussions occur annually in youth sports in the United States (U.S.). Often considered a transient injury, research has begun to relate sports-related concussions to long-term neurodegeneration. Youth athletes are considered to be more vulnerable to concussion than their adult counterparts due to myelination, increased head-body ratios, and ongoing physiologic development, among other factors. The research presented in this dissertation was aimed at informing player safety in youth football as it relates to concussion. Postural control dysfunction is often cited as one of the most common symptoms associated with a concussion. A cognitive, dual-task assessment was presented to aid clinicians in the management of sports-related concussion in youth athletes. On-field head impact data collected from youth football players wearing helmets outfitted with accelerometer arrays allowed for the characterization of the biomechanics of head impacts and concussions for this population. An injury metric was adapted from previous research to develop a youth concussion risk function, and youth concussions were observed to occur at lower biomechanical values than what has previously been reported for adult populations. The proposed testing standard for youth football helmets was assessed in the laboratory and related to on-field head impact data to determine how realistic the standard is relative to on-field head impacts in youth football. Safety standards and certifications currently operate on a pass-fail threshold that does not differentiate helmet performance. A modification of the Summation of Tests for the Analysis of Risk (STAR) evaluation system was developed for youth football helmets. Data presented in this dissertation have direct application to the development of future helmet safety standards and potentially other safety applications as well.

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## **General Audience Abstract**

As many as 1.9 million sports-related concussions occur annually in youth sports in the United States (U.S.). Often considered a short-term injury, research has begun to relate sports-related concussions to long-term breakdowns in neurological processes. Youth athletes are considered to be more vulnerable to concussion than their adult counterparts. The research presented in this dissertation was aimed at informing player safety in youth football as it relates to concussion. Abnormal balance is often cited as one of the most common symptoms associated with a concussion. Several balance assessments were assessed in order to develop a youth-specific testing protocol. An assessment involving quiet standing while being subjected to a cognitive task was presented for clinician use in the management of sports-related concussion. On-field data collected from youth football players wearing instrumented helmets allowed for characterization of the biomechanics of head impacts and concussions for this youth population. A youth concussion risk function was developed that related linear and rotational head acceleration to risk of concussion. The proposed testing standard for youth football helmets was assessed in the laboratory and observed to assess the most severe head impacts a youth player may experience during participation in football. A modification of the Summation of Tests for the Analysis of Risk (STAR) evaluation system was developed for youth football helmets in order to give consumers more information about helmet performance beyond the pass-fail criteria of the helmet standards. Data presented in this dissertation have direct application to the development of future helmet safety standards and potentially other safety applications as well.

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## ATTRIBUTION

Chapter	Title	Co-authors
2	Effects of a season of youth football on static postural control	Steven Rowson
3	Assessing static and dynamic postural control in a healthy population	Ryan A. Gellner and Steven Rowson
4	Reliability of Center of Pressure-Based Measures during Dual-Task Postural Control Testing in a Youth Population	Michael L. Madigan and Steven Rowson
5	Relationship between impact velocity and resulting head accelerations during head impacts in youth football	Ryan A. Gellner and Steven Rowson
6	Relating on-field head impacts to laboratory testing procedures	Steven Rowson
7	Development of a Concussion Risk Function for a Youth Population Using Head Linear and Rotational Acceleration	Ryan A. Gellner, Eric Smith, Srinidhi Bellamkonda, Casey Tierney, Joseph J. Crisco, Derek A. Jones, Mireille E. Kelley, Jillian E. Urban, Joel D. Stitzel, Amaris Genemaras, Jonathan G. Beckwith, Richard M. Greenwald, Arthur C. Maerlender, Per Gunnar Brolinson, Stefan M. Duma, and Steven Rowson
8	Quantifying Youth Football Helmet Performance: Assessing Linear and Rotational Head Acceleration	Ryan A. Gellner, David W. Sproule, Mark T. Begonia, and Steven Rowson

## CHAPTER 1: INTRODUCTION

As many as 1.9 million sports-related concussions occur annually in youth sports in the United States (U.S.).<sup>3</sup> It is estimated that about 50% of these injuries go untreated or unreported. Despite increasing public attention towards concussion, fewer than 30% of all youth coaches receive training in concussion care and education.<sup>28</sup> Often considered a transient condition, research has begun to relate sports-related concussions to long-term neurodegeneration.<sup>46, 47, 56</sup> Some researchers have even found that deleterious neurocognitive or neuropsychiatric effects may be related to repetitive head impact exposure and years of participation in contact sports in absence of a concussion.<sup>1, 27</sup>

Given the nature of the sport, football is associated with the highest number of concussions among team sports.<sup>8, 32</sup> Youth football players below the age of 14 represent 70% of the U.S. football-playing population, but researchers have only begun to investigate head impact exposure and the biomechanics of concussion for this vulnerable population within the last decade.<sup>4, 5, 10, 13, 14, 31, 40, 41, 58</sup> Larger scale epidemiology studies have shown that concussions in youth football occur at a rate between 1 and 2 concussions per 1000 athletic exposures.<sup>15, 34</sup> The reporting and understanding of concussion for youth football players and their caregivers is dependent on several factors, including socio-economic status and pressure from those around the athlete.<sup>35, 36</sup>

Concussions in sports are unavoidable. Incidental head impacts with other athletes or with playing surfaces can occur in all sports, with the potential to result in concussion. Further, individuals have different tolerances to injury, so a vulnerable subset of athletes will likely always exist in sports.<sup>55</sup> There are several strategies that can work to mitigate concussion incidence in high impact sports like football: rule changes, proper technique, and safer equipment.<sup>12</sup> Based on data collected from helmet-mounted accelerometers, Pop Warner issued a mandate across the nation that limited the amount of time spent in contact during practice, as well as the types of

contact that were allowed. This mandate reduced head impact exposure by nearly 50% for teams following the mandate.<sup>10, 50</sup> Some states and researchers have even sought to outlaw tackle football for youth athletes.<sup>6, 29</sup> Teaching youth football players appropriate ball carrying and tackling technique may also reduce exposure to head impacts and the severity of head impacts experienced. By reducing exposure to head impact, the risk of concussion is reduced. For those impacts that still remain, it is essential that protective equipment can mitigate energy transfer to the head and brain. All football helmets must pass a safety standard, which is certified to prevent skull fracture, but adult football helmets have been shown to vary in their ability to reduce the risk of concussion.<sup>53, 54</sup> The combination of all three strategies will work best to minimize the incidence of concussion in all sports.

## **CONCUSSION DEFINITION AND SYMPTOMS**

Sports-related concussion is defined as “a traumatic brain injury induced by biomechanical forces. It may be caused either by a direct blow to the head, face, neck or elsewhere on the body with an impulse body transmitted to the head.”<sup>38</sup> Both linear and rotational kinematics of the brain contribute towards injury, with linear kinematics associated with an induced intracranial pressure gradient during a head impact while rotational kinematics are associated with the brain’s strain response to motion.<sup>33, 48, 57</sup> The rapid acceleration or deceleration of the skull leads to a motion lag between the brain and the skull. This lagging leads to relative motion between the skull and brain which induces brain strains associated with injury.<sup>24, 25</sup>

A sports-related concussion can lead to symptoms which may affect any or all of the following domains: physical, cognitive, emotional, or sleep. Physical symptoms may include headache, nausea, neck pain, sensitivity to light or noise, or balance problems. Cognitive symptoms may include difficulty concentrating or remembering, decreased reaction time, or feelings of confusion or foggiess. Emotional symptoms may include irritability, sadness, anxiety, or lability.<sup>17, 38</sup> Sleep

symptoms may include fatigue, drowsiness, or difficulty falling asleep. These symptoms are not unique to concussion, which makes the detection of concussion difficult for individual athletes and clinicians.

## **DIFFERENCES BETWEEN YOUTH AND ADULT CONCUSSIONS**

Physical differences between youth athletes and their adult counterparts lead to heightened vulnerability for youth towards sports-related concussions. Incomplete myelination in youth brains may lead to increased strain response of the brain during head impact.<sup>9, 30, 51</sup> Youth athletes also have a higher head-body ratio than adults do, as their heads have nearly grown to 95% of full size by the age of 10.<sup>19, 45</sup> Youth athletes must support a nearly fully grown head with limited neck musculature development. It has been posited that youth athletes cannot recruit enough mass to effectively lower head acceleration, and thus concussion risk, during head impacts.<sup>11, 30, 52</sup>

Further, youth athletes who have sustained a concussion may potentially suffer from a potentially harmful social and educational effect.<sup>37</sup> The psychosocial development of children in an educational setting can be compromised by removal from school in order to recover from a sports-related concussion.<sup>30</sup> It has also been found that neurocognitive test scores of youth athletes recover to baseline scores more slowly than they do for adults.<sup>20, 22</sup> Concussive injuries in children have been shown to disrupt natural maturation of the brain.<sup>37</sup> Given the ongoing development of the brain through childhood and adolescence, it is critical to monitor the potential long-term effect of these concussive injuries and ensure that no child returns to school or to physical activity until cleared by a medical professional.

## **CLINICAL BALANCE ASSESSMENT**

One of the most commonly reported symptoms after an athlete sustains a concussion is balance dysfunction or decreased postural control. Sideline tools and clinical balance assessments have

been developed and have shown much success in differentiating concussed and healthy adult athletes. These assessments traditionally consist of quiet standing, often with a force plate. Recently, researchers have made use of dynamic postural control testing and cognitive dual-task assessments as a means of providing more sensitive testing protocols.<sup>2, 7, 16, 18, 21, 23, 26, 44, 49</sup> Similar tools, though, have not been developed for youth athletes, whom are a particularly vulnerable population due to their ongoing development through adolescence. Postural control in youth athletes is still developing, so a youth-specific assessment likely represents a more viable solution than adopting a protocol that has been validated with an adult population.

## **HELMET SAFETY STANDARDS**

All football helmets are tested and certified by the National Operating Committee on Standards for Athletic Equipment (NOCSAE). This testing standard was instrumental in effectively eliminating fatal head injuries from football. After its initial implementation in 1974, fatal head injuries were reduced by 74%.<sup>39</sup> At present, all football helmets, both adult and youth, are tested to the same impact standard that only considers linear head acceleration. Recognizing the need to consider population differences as well as the contribution of rotational head kinematics to concussion, NOCSAE has proposed 1) an updated testing standard that considers both linear and rotational head kinematics, and 2) a youth-specific helmet testing standard.<sup>42, 43</sup> The youth football helmet safety standard represents a scaled down version of the adult helmet standard, but it is unclear how representative the standard is of conditions that youth football players experience. Further, NOCSAE's testing standards assess severe head impacts and operate on pass-fail criterion that offers consumers little information regarding the relative performance of one helmet to another.

## **RESEARCH OBJECTIVES**

The research in this dissertation is multi-faceted, in that both clinical and on-field data were collected. All work was aimed at promoting player safety in youth football as it relates to concussion. The clinical management of youth athletes after sustaining a sports-related concussion is crucial. The use of population-appropriate evaluation tools during the recovery process must be considered. This work sought to find a postural control tool for clinician use with youth athletes. On-field data collected from youth football players wearing instrumented helmets allowed for characterization of the biomechanics of head impacts and concussions for this population. These on-field data served as the basis for evaluating the results of helmet testing in this research.

The specific objectives of this research are:

1. To develop a youth-specific postural control testing protocol to aid clinicians in the management of sports-related concussion.
2. To assess how representative the proposed NOCSAE youth football testing standard is of youth football head impacts.
3. To develop a youth concussion risk function based on linear and rotational head kinematics that can be used to develop more effective safety equipment.
4. To develop a youth football helmet evaluation tool based on real-world head impact data to provide consumers with information regarding helmet performance.

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## **CHAPTER 2: EFFECTS OF A SEASON OF YOUTH FOOTBALL ON STATIC POSTURAL CONTROL**

### **ABSTRACT**

Concussions occur in youth football with lower frequency than observed at higher levels of play, though the effect of repetitive subconcussive head impacts resulting from participation in youth football is unknown at this point. One measure shown to be affected by concussions is athlete postural control. The objective of this study was to compare performance on the Balance Error Scoring System (BESS) and a force plate protocol at two time points within a cohort of healthy youth football players and healthy non-contact youth track or baseball athletes. In absence of a clinically-diagnosed concussion, the hypothesis was tested that a season of youth football would affect measures of static postural control and stability. Between time points, there were no significant differences observed between either BESS scores or force plate metrics. Between athlete groups, there were no significant differences observed for either the BESS or the force plate protocol. Particularly for pediatric males, postural control is still developing and current assessments may not be sensitive enough to detect changes. Continued research is necessary to determine what postural control testing may be most viable for use within an active, pediatric population.

## INTRODUCTION

Athletes who have sustained a concussion have been shown to suffer from transient decreases in postural control.<sup>7, 13</sup> Some research has shown that even non-concussed football players may experience balance deficits.<sup>12, 14</sup> These studies have largely focused on collegiate athletes, despite the fact that youth players comprise 70% of the football-playing population. Though concussions occur less frequently for these youth athletes, potential adverse effects of subconcussive head impacts associated with playing football remain unknown at present.

Two commonly implemented tools for assessing postural control changes in instances of athlete concussion are the Balance Error Scoring System (BESS) and force plate testing. The BESS was developed as a clinical, static balance assessment for sideline use and has been shown to measure postconcussion balance changes in both youth and adult athletes.<sup>1, 5, 7</sup> Reliability of the BESS is variable, and a known practice effect exists with repeated administration.<sup>22</sup> Instrumented force plates have also been used to quantitatively assess postural control changes in athletes with and without concussion.<sup>3, 4, 15</sup> Force plate testing typically involves tracking changes in the center of pressure (COP).

The BESS and force plate protocols have seen limited use with youth athletes, with most research assessing postural control for healthy and concussed athletes.<sup>2, 15, 24</sup> The objective of this study was to compare performance on the BESS and a force plate protocol at two time points within a cohort of healthy youth football players and healthy non-contact youth baseball or track control athletes. The first time point occurred before sports participation, while the second time point occurred after the conclusion of the season, which meant completing testing after a season of head impact exposure for the football players. Postural control is still developing for pediatric males, so current balance assessments may not be sensitive enough to detect changes or suitable for use within this population.<sup>16, 20</sup> Further research investigating postural control testing

in healthy and injured youth athletes is necessary to determine what assessments may be most viable within this population.

## **METHODS**

A cohort of 9-11 year old youth football players (n=40; average age:  $9.9 \pm 0.6$  years; average mass:  $39.5 \pm 10.2$  kg) and youth track or baseball players (n=11; average age:  $9.5 \pm 0.8$  years; average mass:  $33.5 \pm 5.7$  kg) was recruited for participation in this study approved by the Virginia Tech Institutional Review Board. All youth players provided verbal assent, in addition to their guardians providing written consent. Each football player's helmet was instrumented in order to determine the number of head impacts experienced during games and practices throughout the season. The youth football players were tested at the beginning and end of their athletic season, which resulted in a time between testing that varied from 11-14 weeks. The youth track and baseball players served as age-matched, active control subjects for the youth football players and were tested at the beginning of their activity, and then again at a postseason time point that was consistent with that of the football players. Testing consisted of administration of the BESS and a force platform protocol. Each subject was tested individually by trained lab personnel. Medical history forms completed by the children's guardians were used to determine subject inclusion criteria. For the control subjects, additional information regarding contact sport history and head impact/injury history was collected. Subjects with neurological conditions like Attention Deficit/Hyperactivity Disorder or previous lower extremity injury were excluded from the pool of subjects, as these factors are known to affect postural control.<sup>19</sup>

Administration of the BESS involves testing subjects in three stances (double leg, single leg, and tandem) and on two surfaces (flat ground and a foam pad [Airex Balance Pad 81000, Power Systems, Knoxville, TN]). Subjects must try and maintain each posture for 20 seconds with their eyes closed and hands on hips. Scoring for the BESS consists of the researcher counting the

number of participant errors within each stance. Errors include opening of the eyes, removing hands from the hips, abducting or flexing the hip beyond 30°, falling, lifting toes or heels from the test surface, and failing to reset within five seconds. An overall score is determined by summing the number of errors from each individual stance. Thus, lower BESS scores are associated with better postural control.

The force platform protocol in this study consisted of standing eyes open and eyes closed trials, each of which lasted for 30 seconds. Center of pressure (COP) data were output from an IsoBALANCE®2.0 (IsoTechnology, Australia) force plate at a frequency of 10 Hz in the x (Medial-Lateral [ML]) and y (Anterior-Posterior [AP]) axes. Testing began when a subject's COP was positioned in the center of the force plate. Some measures relate to the spatial variability of the COP trajectory, while other metrics assess the temporal variability of the data. For both spatial and temporal metrics, lower values are indicative of less varied balance trajectory, and thus better postural control.

Sway in the AP and ML directions is represented by the standard deviation of movement along the respective axis. Path length represents the overall distance travelled during the test, and maximum path velocity assesses the quickest change in path during the trial. The last of the spatial metrics is the 95% confidence ellipse area, which considers the sway in the AP and ML directions, in addition to the covariance between the two axes.

Temporal variability is measured as the entropy of the COP trajectory and assesses the randomness of the data. AP and ML Entropy assess the variability of the COP trajectory along each direction. Sample entropy is determined by comparing a given data vector template from the COP trajectory to all other vectors within the trajectory and counts all those that are within a defined similarity range.<sup>9, 10, 17</sup> Other measures of entropy can be measured using graphical

representations of the COP data. Renyi and Shannon Entropy involve developing a grid that is subset based on the standard deviation in the AP and ML directions. The number of points within each subunit are then summed, with the probability of a COP coordinate residing in a particular subunit serving as the input parameter for computing both Renyi and Shannon Entropy. Detailed background on the algorithm is available.<sup>6</sup>

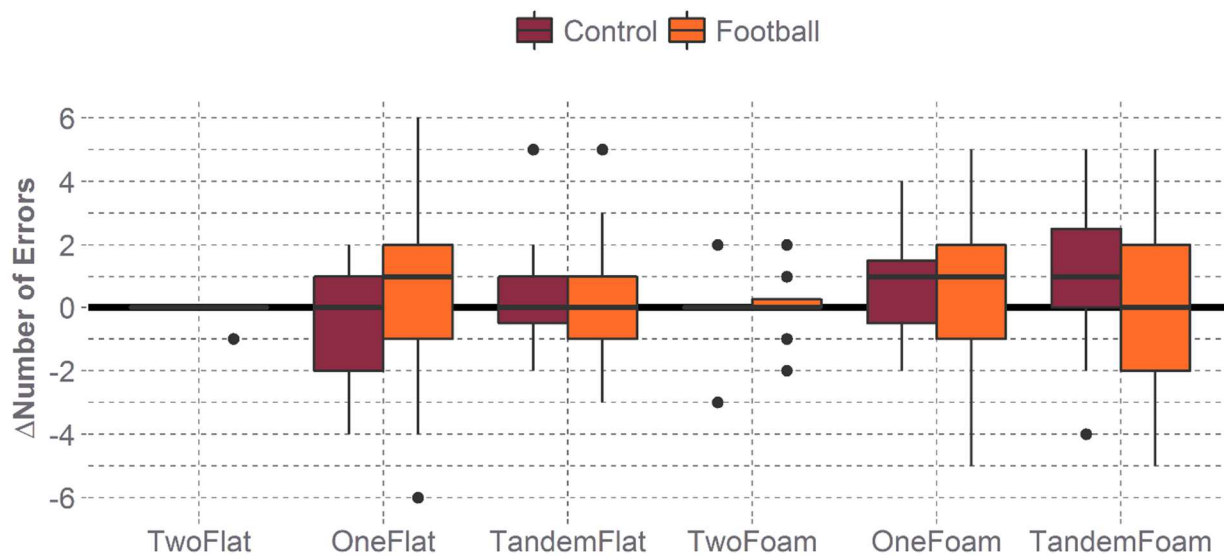
Differences between the preseason and postseason time points were assessed for the BESS and force plate data using paired t-tests for both athlete groups. Additional t-tests were run to assess differences between the eyes open and eyes closed force plate trials. Lastly, differences between the youth football players and the non-contact controls were assessed using Welch's t test. A level of significance was defined for  $\alpha < 0.05$ , though a Bonferroni correction was applied to account for multiple t-tests conducted for the data sets. For the force plate data, 3 tests were run for each metric, so a level of significance was defined for  $\alpha < 0.017$ , while the BESS data was tested twice, resulting in a level of significance for  $\alpha < 0.025$ .

## **RESULTS**

No significant difference was observed for the total BESS scores between preseason (football:  $16.5 \pm 5.4$  errors; control:  $15.3 \pm 4.2$  errors) and postseason (football:  $17.6 \pm 6.1$  errors; control:  $17.0 \pm 6.5$  errors) for either the football players ( $p = 0.41$ ) or the non-contact controls ( $p = 0.47$ ). Similarly, no significant differences between preseason and postseason data were observed for any force plate metric for both athlete groups. Individual athletes' scores still varied between the preseason and postseason.

Differences were observed between the eyes open and eyes closed trials for both control and football subjects for a variety of metrics. The two trials were found to be significantly different for AP and ML Entropy, as well as Path Length, for both control and football subjects ( $p < 0.017$ ).

Additionally, AP Sway ( $p = 0.005$ ) and Maximum Path Velocity ( $p = 0.011$ ) were significantly different between the eyes open and eyes closed trials for the youth football players.

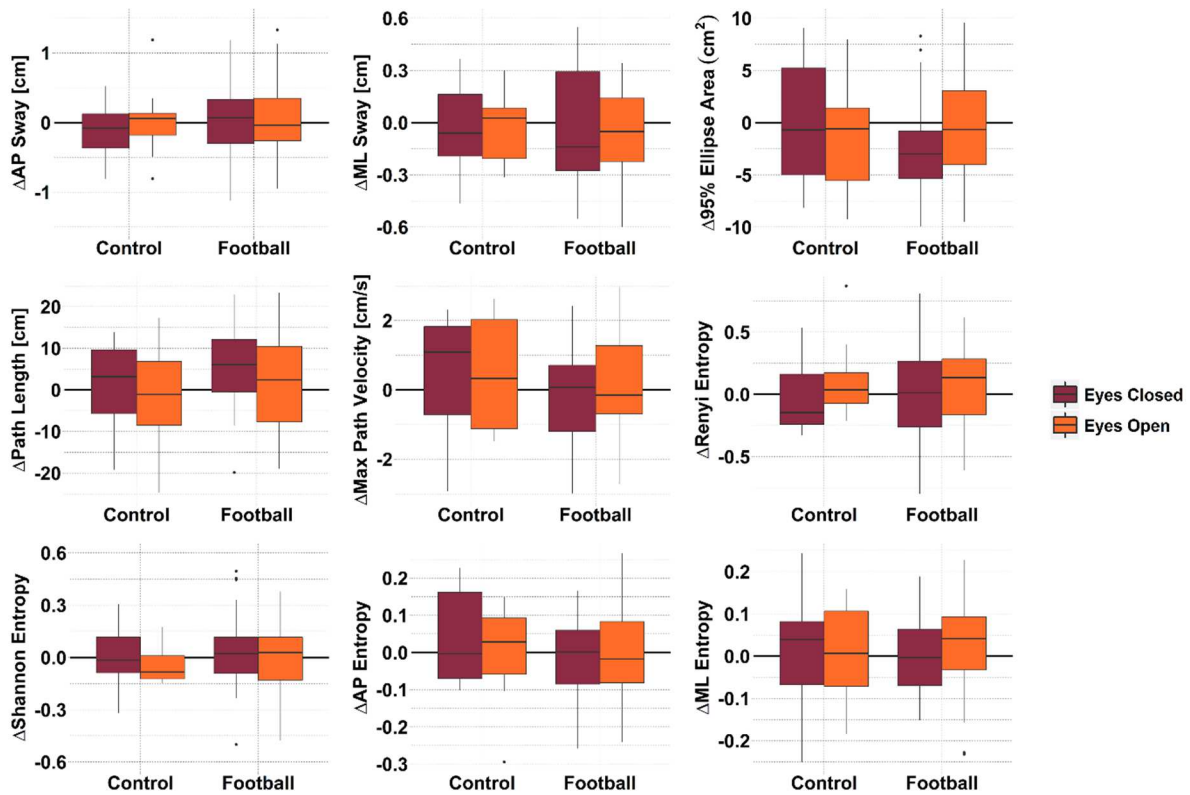


**Figure 2.1: Change in number of errors by BESS stance from the preseason to the postseason. Both control subjects and football players performed similarly for each BESS stance, with a median change of 0 for most stances. The most variance was observed to exist for the tandem stance on a foam surface and in the one foot stance on a flat surface.**

The difference from the preseason to the postseason was computed for each stance of the BESS, as well as for each force plate metric. These changes in score served as the basis of comparison between the youth football players and the non-contact control athletes. The eyes open and eyes closed trials were considered separately when comparing between the athlete groups. For the BESS, no significant difference was observed for any of the individual test stances or the overall test score between the two groups (Figure 2.1). Both of the two foot stances were associated with very few errors, while the one foot foam stance resulted in a median of six errors for both the control and youth football subjects. For the force plate metrics, no significant differences were observed between the two athlete groups for either eyes open or eyes closed metrics. In general,

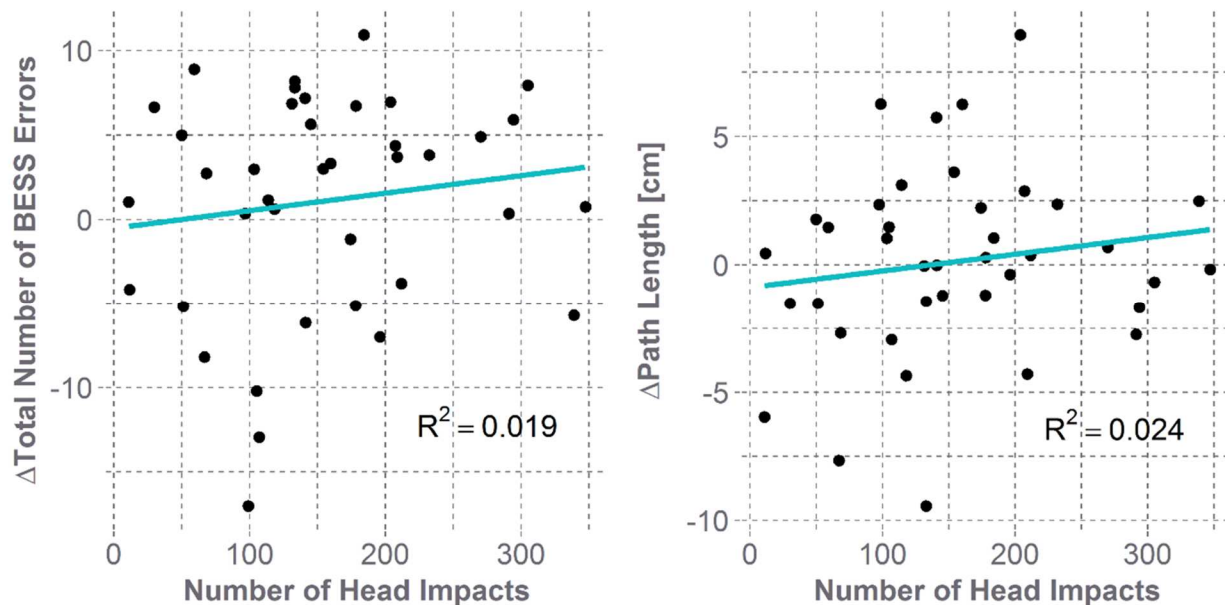


the eyes closed trials were associated with greater variability than the eyes open trials and greater median values for most metrics (Figure 2.2).



**Figure 2.2: Change in force plate metrics by population from the preseason to the postseason. No significant differences were observed within each group between the preseason and postseason time points. AP and ML Entropy were more varied between the eyes open and eyes closed trials and between the two athlete groups than either Renyi or Shannon Entropy. The eyes closed trials were generally associated with greater variability for all metrics considered in this study regardless of athlete group.**

The effect of head impact exposure on the measures of balance in this study was also investigated. The football players experienced an average of  $169 \pm 124$  head impacts, with a maximum of 720. Head impact exposure was not observed to be related to change in performance on postural control testing (Figure 2.3). None of the metrics were found to be correlated, with all  $R^2$  values less than 0.1.



**Figure 2.3: Representative correlation of changes in balance measures and number of head impacts. No measure of balance was found to have an  $R^2$  value that exceeded 0.1. One player with 720 impacts on the season was not included in the regression analysis, as that represented an extreme outlier for the number of head impacts. Positive changes in balance measures would be representative of balance worsening at the postseason testing relative to the preseason testing.**

## DISCUSSION

The two foot trials for the BESS protocol resulted in a median number of errors of 0 and have been shown previously to be associated with low variance and to not be able to differentiate healthy and concussed athletes.<sup>18, 21</sup> Further, the one foot trial on a foam surface was associated with a median of six errors for both non-contact and youth football athletes. This stance was too difficult for subjects in this study to complete, as subjects failed to maintain balance. Changes in score from the preseason to the postseason hovered around 0, with no differences observed between the youth football players and non-contact control athletes (Figure 2.1). The reliability of the BESS administration within a youth athlete population was not assessed in this study; however, it is expected that trials associated with greater levels of variance between subjects would indicate postural control differences more effectively both between and within subjects.

With half the stances associated with low variance compared to the other three stances, use of the BESS within a youth population likely warrants further investigation.

Consistent with previous research, the eyes open and eyes closed trials resulted in differences for most metrics considered in this study.<sup>4</sup> The eyes closed trials were not only associated with higher median values, which reflect the increased difficulty of the task, but also with greater variability within the data. Individual disparity between subjects was more pronounced for the eyes closed trial than the eyes open trial. The low variance associated with the eyes open trial may be indicative of a task that is not sensitive enough to detect changes in static postural control. Between the youth football players and non-contact control athletes, no force plate metric was observed to be significantly different (Figure 2.2). Measurements of entropy have been shown to differentiate postural control in youth populations and may be more viable than the spatial metrics considered in this study.<sup>4, 6, 15</sup> Some researchers have proposed assessing athlete postural control with 120 second duration trials for the eyes open and eyes closed tests rather than 30 seconds. This duration is most effective at detecting changes in entropy between concussed and non-concussed athletes. However, this duration could invalidate the use of spatially varying metrics, which may become unstable over longer duration trials as subject attention wanes.<sup>8</sup> Further, a 30 second duration has been shown to be most reliable for use in test-retest environments as was present in this study.<sup>11</sup>

The effect of head impact exposure associated with a season of youth football was assessed by relating performance changes on the balance assessments. It was found that elevated head impact exposure was not tied to decreased balance, with all  $R^2$  values below 0.1 (Figure 2.3). Some players with very few head impacts performed worse during postseason testing than during preseason testing while some players with more head impacts exhibited the opposite behavior. These individual variances may relate to the lack of reliability within the assessments in this study,

changes in balance due to fatigue or learning effects, or the ongoing development of postural control within children in this age group.<sup>14, 20, 23</sup>

Postural control differences were not observed between the football players and their non-contact counterparts in this study. All athletes in this study were considered healthy, with no history of concussion. The BESS is associated with a limited scoring sensitivity and has been shown previously to vary in reliability. Thus, the BESS may not appropriately distinguish subjects with varying levels of postural control and a youth-specific administration may be appropriate after further research. The force plate administration utilized in this study may not completely detect low frequency COP changes that manifest during longer duration trials. Assessing the reliability of these postural control tools for use within youth populations is necessary to determine their viability. Further, the development of postural control within the pediatric population, particularly males, is limited and ongoing. It is possible that variability between and within subjects and groups could stem from differences in physiologic or neurologic development rather than in postural control differences.

## **CONCLUSIONS**

A cohort of age-matched youth football players and non-contact control baseball and track players completed static postural control testing in two sessions approximately 12 weeks apart. The testing consisted of the BESS and a force plate protocol. Some of the BESS stances were associated with very low levels of variance that may not effectively measure static postural control. Differences in both spatial and temporal force plate metrics were observed within the two athlete groups, though only a single measure of entropy was shown to differ between the football players and the non-contact control athletes. No effect of subconcussive head impacts on static postural control was found in this study, though further research is necessary to determine the

effectiveness of these tools within a youth population. Consideration of dynamic postural control assessments, which perturb the center of pressure, may also be appropriate.

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## **CHAPTER 3: ASSESSING STATIC AND DYNAMIC POSTURAL CONTROL IN A HEALTHY POPULATION**

### **ABSTRACT**

Static postural control testing is often conducted by clinicians and athletic trainers for use with athletes who have sustained a concussion. Dynamic postural control involves the body's response to perturbation of the center of mass and may offer additional insight that static testing cannot capture. The objective of this study was to assess the reliability and feasibility of a balance protocol consisting of both static and dynamic postural control assessments with a healthy, adult population. Subjects stood in both unipedal and bipedal stances on a force plate to capture quantitative data regarding the center of pressure over time. Further, subjects completed the Balance Error Scoring System (BESS), a static measure, and a modified version of the Star Excursion Balance Test (SEBT), a dynamic measure. Reliability with the BESS was limited, while moderate to strong reliability was obtained for the modified SEBT. Unipedal stances were associated with a greater variance than bipedal stances for both the BESS and force plate protocol. These assessments will be applied within a pediatric populations to determine the validity of their use. Further postural control research is necessary to determine the most viable assessments for use within an active, pediatric population.

## INTRODUCTION

After mild traumatic brain injury (mTBI), it is common for postural control deficits to be observed.<sup>13, 18, 26</sup> Many post-concussion assessments now include postural control tests as an evaluative tool to determine patient health.<sup>4, 6, 12, 22</sup> Postural control represents the ability of a person to maintain balance naturally and when exposed to perturbation.<sup>10</sup> Postural control can be defined by assessing static and dynamic balance. Static balance involves an individual establishing a stable base and attempting to minimize movement while holding the particular posture. Dynamic balance, on the other hand, refers to the introduction of perturbations to this stable base of support. It can be assessed by having subjects establish a base of support and then requiring some level of movement away from that equilibrium. Static balance has been most commonly assessed in post-concussion situations, though dynamic balance assessments are gaining favor as they may involve movements similar to those experienced while playing sports.<sup>7, 11, 27</sup>

Static balance is most commonly assessed using the Balance Error Scoring System (BESS) or force plates. The BESS is an easily administered, static balance assessment for sideline use in instances of suspected concussion that asks individuals to hold different static postures while an evaluator assesses deviations from this desired posture.<sup>1, 14</sup> Instrumented force plates are used to quantitatively track the center of pressure (COP) over time during a static stance.

Dynamic balance assessments are necessarily more involved than are static assessments, and have seen less use.<sup>24</sup> One of the most commonly employed assessments is the Star Excursion Balance Test (SEBT), which tasks individuals with maintaining balance with one foot while reaching out in prescribed directions with the other foot.<sup>11</sup> By more closely aligning concussion testing assessments with physical activity, it is hypothesized that the tools will be more relevant. The SEBT is traditionally used to assess ankle instability and risk of lower extremity injury, but it



is reasonable to expect this dynamic postural control assessment to differentiate concussed athletes as well.

As pediatric populations are still developing postural control, there is a lack of research investigating either static or dynamic postural control within a youth population.<sup>7, 8, 10, 24</sup> Before assessing postural control with this population, the protocol must first be investigated in a population with fully developed balance. The objective of this study was to assess the reliability and feasibility of a balance protocol consisting of both static and dynamic postural control assessments with a healthy, adult population. Static assessments included the BESS and a force plate protocol while dynamic postural control was assessed using a modified version of the SEBT. Upon successful completion of this study, the protocol would then be studied in a cohort of youth football players.

## **METHODS**

Ten healthy, male subjects completed the balance assessments in this study approved by the Virginia Tech Institutional Review Board. These subjects had an average age of  $22.2 \pm 1.5$  years, an average body mass of  $87.6 \pm 12.2$  kg, and an average height of  $1.78 \pm 0.06$  m. Subjects completed two rounds of testing, with the sessions separated by a week. A medical history questionnaire was completed by each subject to assess relevant injuries, specifically concussion history and lower extremity injuries<sup>23</sup>. Testing was conducted by trained lab personnel and consisted of administration of the BESS, a force plate protocol, and a modified version of the SEBT.

The BESS consists of subjects holding specific stances for 20 second durations with their hands on their hips and eyes closed. The test stances (double leg, single leg, and tandem) are carried out on two surfaces (flat ground and a foam pad [Airex Balance Pad 81000, Power Systems,

Knoxville, TN]). Scoring for the BESS consists of the researcher counting the number of participant errors within each stance.<sup>21</sup> Thus, lower BESS scores are associated with better postural control.

For the force plate assessment, subjects completed four trials of 30 seconds each: bipedal eyes open, bipedal eyes closed, unipedal eyes open, and unipedal eyes closed. The unipedal stances were added to increase the difficulty of the static postural control tasks. An IsoBALANCE<sup>®</sup>2.0 (IsoTechnology, Australia) operating at a frequency of approximately 10 Hz collected COP data in the medial-lateral (ML) and anterior-posterior (AP) axes. All subjects were aligned similarly, with their feet situated along lines on the force plate and the COP centered on the force plate.

Five measures of spatial variability within the COP trajectory were computed: AP and ML sway, path length, maximum path velocity, and 95% COP area. AP and ML Sway represent the standard deviation of the COP trajectory along the respective axis of motion. Path length represents a summation of overall distance travelled by the COP during the 30 second trial, while maximum path velocity assesses the most rapid change in direction during the 30 second trial. Lastly, the 95% COP area represents the area of an ellipse that would capture the mean COP in 95% of samples. Four measures of temporal variability, or entropy, within the COP trajectories were also calculated for each trial: AP and ML Entropy, Renyi Entropy, and Shannon Entropy. Trials with lower levels of entropy, or variability, are considered representative of better static postural control. Background on the determination of these metrics may be found elsewhere.<sup>9, 15,</sup>

16

The modified version of the SEBT has been previously presented and utilizes 3 of the 8 reaching directions from the original SEBT: anterior, posterior lateral, and posterior medial.<sup>5</sup> Subjects stand on one foot with hands on hips and reach out with the most distal portion of the other foot in one

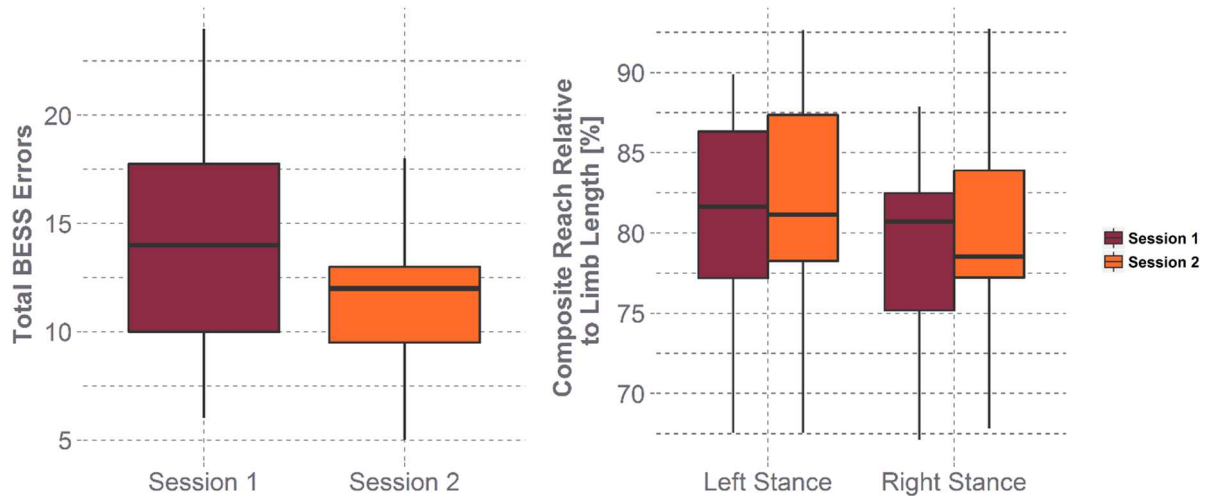
of the prescribed directions before returning to a bipedal stance. Subjects must not transfer weight away from their support foot and cannot lose balance, otherwise the trial must be repeated. Three trials with each leg are completed for the 3 directions. The three trials for each leg-direction combination are averaged and then normalized to the subject's limb length to account for differences in reach associated with varying anthropometry.

The force plate frequency of approximately 10 Hz is low compared to what is traditionally done (100 Hz). Previous research has shown that more than 95% of the signal power for bipedal stances is below 2 Hz and less than 1% of power is above 10 Hz, which would make a 10 Hz sampling frequency sufficient for measuring static balance.<sup>17, 20</sup> Research with static, unipedal stances has shown that the COP trajectory remains low frequency (95% < 2 Hz) while resulting in greater variability than bipedal stances.<sup>19, 25</sup>

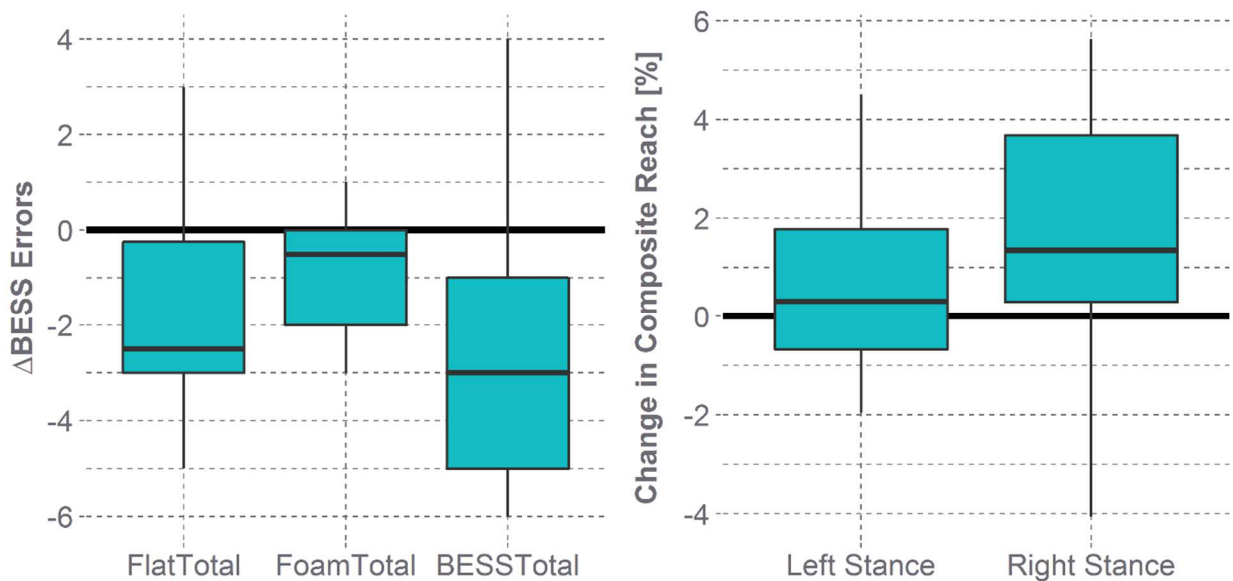
Postural control assessment test-retest reliability was assessed using intraclass correlation coefficients (ICC). ICCs with 95% confidence intervals were computed for each force plate metric for the four trials and each stance of the modified SEBT and the BESS. A reliable assessment would be one in which a healthy subject's score/performance would not vary over time. Higher correlation coefficients are thus associated with reliable assessments.

## **RESULTS**

The overall number of errors measured on the BESS decreased between the two sessions for almost all subjects (Figure 3.1). These improvements on the BESS were mostly noted for the one foot and tandem stances on a flat surface (Figure 3.2). Reliability for both two foot stances could not be calculated, as there were no errors observed in these stances. The one foot and tandem stances were associated with low reliability, though the total number of errors from the three foam stances was associated with moderate reliability (Table 3.1).



**Figure 3.1: Overall balance performance for both test sessions. Subjects experienced fewer errors with the BESS during the second session than in the first session (left). Composite reach was consistent between sessions and limbs on average, with subjects reaching 75-85% of limb length (right).**



**Figure 3.2: Matched differences between sessions for the BESS and the modified SEBT. In general, subjects performed better on the BESS during the 2nd session (left). Individual performance varied between sessions for the modified SEBT, with some subjects reaching further and others not reaching as far (right).**

**Table 3.1: ICC values for each stance of the BESS and Modified SEBT. ICC values are reported as calculated, with 95% confidence intervals shown in parentheses.**

			ICC
BESS	Flat	Two Foot	-
		One Foot	0.50 (0.00-0.84)
		Tandem	0.33 (0.00-0.75)
		Total	0.52 (0.00-0.85)
	Foam	Two Foot	-
		One Foot	0.52 (0.00-0.86)
		Tandem	0.54 (0.00-0.86)
		Total	0.81 (0.40-0.95)
Modified SEBT	Left Leg	Anterior	0.53 (0.00-0.86)
		Posterior Medial	0.93 (0.65-0.98)
		Posterior Lateral	0.77 (0.30-0.94)
		Composite	0.96 (0.86-0.99)
	Right Leg	Anterior	0.72 (0.19-0.92)
		Posterior Medial	0.87 (0.58-0.97)
		Posterior Lateral	0.87 (0.59-0.97)
		Composite	0.89 (0.62-0.97)

Performance on the modified SEBT did not vary between the two sessions nor between left and right stances (Figure 3.1). Reaches while standing on the right foot were associated with greater variation between the two sessions (Figure 3.2). Reaches in the posterior medial and posterior lateral directions were associated with high levels of test-retest reliability (ICC > 0.75). The composite values for each leg were associated with higher levels of reliability than any individual reach direction (Table 3.1).

Renyi and Shannon entropy measurements were consistent across both sessions and all test conditions for the force plate (Tables 3.2 and 3.3). ML Entropy was observed to increase for unipedal stances compared to bipedal stances. The spatially varying force plate metrics were increased for unipedal stances as well. The unipedal stance with eyes closed was associated with spatially varying parameters that were, on average, at least 2.5 times higher than that observed for the unipedal stance with eyes open. The average matched differences in force plate metrics between the two sessions were close to zero, though individual performance varied.

**Table 3.2: Summary of eyes open testing. Session 1 and 2 data are reported as mean (standard deviation). The unipedal stance with eyes open was more difficult for subjects than bipedal standing, as evidenced by the spatial varying parameters.**

	Static Bipedal			Static Unipedal		
	Session 1	Session 2	Matched Difference	Session 1	Session 2	Matched Difference
AP Sway (cm)	0.47 (0.44)	0.42 (0.15)	-0.02 (0.16)	0.68 (0.14)	0.61 (0.12)	-0.03 (0.07)
ML Sway (cm)	0.23 (0.12)	0.22 (0.09)	0.00 (0.05)	0.55 (0.11)	0.62 (0.15)	0.03 (0.07)
Path Length (cm)	17.18 (11.12)	14.36 (5.12)	-1.11 (3.80)	65.86 (24.90)	60.04 (11.54)	-2.29 (9.07)
Max Path Velocity (cm/s)	2.41 (1.09)	2.06 (0.36)	-0.14 (0.37)	7.17 (2.15)	6.35 (1.48)	-0.32 (0.89)
95% Ellipse Area (cm <sup>2</sup> )	2.75 (4.60)	1.60 (1.05)	-0.18 (0.66)	7.15 (2.63)	7.21 (2.87)	0.01 (0.55)
AP Entropy	0.37 (0.12)	0.35 (0.06)	-0.01 (0.10)	0.65 (0.05)	0.70 (0.08)	0.05 (0.06)
ML Entropy	0.30 (0.12)	0.25 (0.08)	-0.05 (0.09)	0.72 (0.06)	0.68 (0.06)	-0.04 (0.09)
Renyi Entropy	3.56 (0.28)	3.62 (0.25)	0.06 (0.39)	3.81 (0.17)	3.74 (0.23)	-0.07 (0.30)
Shannon Entropy	2.58 (0.13)	2.48 (0.14)	-0.10 (0.19)	2.73 (0.04)	2.74 (0.06)	0.01 (0.04)

**Table 3.3: Summary of eyes closed testing. Session 1 and 2 data are reported as mean (standard deviation). The unipedal stance with eyes closed was very variable, both within and between subjects, so lower levels of reliability were observed for this test. Matched differences between the two sessions were close to zero on average, with individual performance varying.**

	Static Bipedal			Static Unipedal		
	Session 1	Session 2	Matched Difference	Session 1	Session 2	Matched Difference
AP Sway (cm)	0.41 (0.11)	0.46 (0.19)	0.02 (0.06)	1.73 (0.65)	1.76 (0.76)	0.02 (0.31)
ML Sway (cm)	0.18 (0.08)	0.17 (0.10)	-0.01 (0.04)	1.46 (0.55)	1.40 (0.58)	-0.02 (0.28)
Path Length (cm)	18.58 (7.02)	16.02 (6.29)	-1.01 (1.82)	169.08 (67.52)	150.95 (71.35)	-7.14 (13.12)
Max Path Velocity (cm/s)	2.64 (0.95)	2.48 (0.58)	-0.06 (0.49)	25.19 (22.27)	20.84 (16.17)	-1.71 (10.40)
95% Ellipse Area (cm <sup>2</sup> )	1.49 (0.95)	1.57 (1.52)	0.01 (0.21)	50.53 (36.14)	51.82 (43.14)	0.20 (6.34)
AP Entropy	0.47 (0.13)	0.40 (0.10)	-0.07 (0.11)	0.34 (0.17)	0.44 (0.18)	0.10 (0.23)
ML Entropy	0.28 (0.08)	0.24 (0.10)	-0.05 (0.08)	0.47 (0.11)	0.51 (0.11)	0.03 (0.11)
Renyi Entropy	3.65 (0.23)	3.58 (0.18)	-0.07 (0.33)	3.81 (0.17)	3.82 (0.16)	0.01 (0.15)
Shannon Entropy	2.61 (0.13)	2.57 (0.16)	-0.03 (0.10)	2.63 (0.14)	2.61 (0.07)	-0.03 (0.16)

## DISCUSSION

Consistent with previous research, the BESS was observed to be associated with lower reliability than other assessments.<sup>3</sup> It was included in this balance protocol due to its ease of use and near ubiquity in the assessment of sports-related concussion. The learning effect that is often

observed in studies involving the BESS was noted in this study as well, with subjects performing better in Session 2 than in Session 1 (Figure 3.1). The stances for the BESS are generally unique to study subjects, so additional administration of the BESS allows subjects to engage compensatory mechanisms to maintain postural control (Figure 3.2). Further, the one foot stance on a foam surface may be too difficult for subjects to complete, as all of the healthy, adult subjects in this study experienced 4 or more errors for this particular stance. Pediatric subjects with limited postural control would likely struggle to complete this task as well.

The SEBT has been commonly implemented to determine whether subjects are at risk for lower extremity injury. It is a reliable measure of dynamic postural control, which is a factor that may also be comprised during instances of concussion. One limitation associated with the SEBT is that it can be difficult for a single test administrator to determine whether a subject has shifted their balance towards their reaching foot while simultaneously seeing how far a subject has reached. Further, the overall SEBT can be a time-consuming assessment, with subjects having to complete three trials in eight different directions with both legs. A modified version of the SEBT, only involving reaches in the anterior, posterior lateral, and posterior medial directions, was employed in this study and was associated with high levels of reliability (Table 3.1). Using this modified SEBT with an active, pediatric population may be viable for assessing dynamic postural control, though further research is necessary. Assessing reliability and feasibility with a healthy youth population would be required, as well as utilizing the modified SEBT with concussed youth athletes to determine its ability to discriminate between healthy and concussed youth athletes.

Performance on the force plate varied between the four test conditions, though Renyi and Shannon Entropy were largely consistent across all tests (Tables 3.2 and 3.3). Matched differences between the two sessions were close to zero, which is representative of an assessment with little within-subject variation. Between-subject variability existed for the force

plate metrics as represented by the standard deviations of computed metrics. The unipedal eyes closed test was very difficult for subjects to complete, with some having to place their second foot down at times in order to regain balance. Measures from this stance may not be representative of static postural control. The unipedal eyes open test required greater attention by subjects than the bipedal test in order to maintain postural control. This manifested itself in higher measures of all spatially varying metrics. With the frequency content of unipedal COP trajectories consisting of predominantly low frequency data (< 2 Hz), conducting this test with subject's eyes open likely represents the best opportunity to differentiate postural control. The ability of a pediatric population to complete this test requires investigation, with the possibility of utilizing a shorter duration test for unipedal stances.

A good measure of balance would be one in which within-subject variability was low compared to between-subject variability. An individual subject's score should not vary considerably from test to test in absence of some disturbance to postural control. Conversely, the measure should be sensitive enough to distinguish balance between subjects. The modified SEBT exhibited this favorable ratio of between-subjects to within-subjects variability, while the BESS did not (Figures 3.1 and 3.2). Individual performance varied considerably on the BESS between sessions, while performance on the modified SEBT was more stable between the two sessions. The force plate metrics used in this study were associated with high levels of between-subjects variability relative to within-subjects variability, specifically for the unipedal eyes open test.

Though the unipedal stances on the force plate and the modified SEBT employed in this study resulted in high levels of test-retest reliability, the tasks may not be feasible to apply to a pediatric population. The unipedal stance conducted with eyes closed was difficult for the healthy adult subjects in this study to complete, with a very unstable COP trajectory. It is likely that pediatric subjects who are still developing postural control would not be able to complete this static



assessment. The anterior reaching stances of the modified SEBT resulted in lower reliability than either of the posterior reaching stances (Table 3.1). This task was also associated with the highest number of invalid trials, as subjects either transferred their weight to the reaching foot or could not maintain balance when attempting to return to their base of support. Use of this reaching stance with a pediatric population would potentially lead to lower levels of reliability and increase the total testing time. Previous research has discussed the feasibility challenges associated with using the SEBT with a pediatric group.<sup>2</sup> Using only those stances with the highest reliability from the modified SEBT would limit the overall testing time and subject fatigue, while providing the greatest opportunity for reliability with youth athletes.

## **CONCLUSIONS**

A pilot study was conducted that utilized a series of clinical static and dynamic postural control assessments, in addition to unipedal and bipedal stances on an instrumented force plate. The BESS was associated with low levels of reliability and repeatability, whereas the modified SEBT tested consistently between the two sessions in this study. For the force plate testing, unipedal stances resulted in greater levels of variability between subjects. Conducting this test with the subject's eyes closed may not be appropriate, or representative of a stable COP trajectory. This balance protocol will be further investigated in a pediatric population of youth football players. These assessments may be viable within this population, or youth-specific balance measures may be needed to appropriately assess the still developing postural control of youth athletes. Further, use of postural control testing in instances of pediatric concussion may better understanding of how this injury manifests in a younger population.

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## **CHAPTER 4: RELIABILITY OF CENTER OF PRESSURE-BASED MEASURES DURING DUAL-TASK POSTURAL CONTROL TESTING IN A YOUTH POPULATION**

### **ABSTRACT**

Dual-task assessments can identify changes in postural control during recovery from a concussion. However, the evaluation of developing postural control systems in children presents a challenge in using adult balance assessments. This study aimed to assess reliability of a dual-task postural control protocol for use within a youth population. Nine subjects (average age:  $11.6 \pm 0.5$  years) participated in this pilot study. Testing comprised nine 120 second trials of standing on a force plate collecting data at 250 Hz. Test conditions included no dual-task, counting backwards by 2, counting backwards by 3, listening, and the Stroop test. Subjects completed each test with open and closed eyes, except for the Stroop test. The force plate measured the subjects' center of pressure (COP) trajectory. Reliability was good ( $>0.6$ ) or excellent ( $>0.75$ ) for COP speed, sway, and sample entropy measures. The eyes open, no task condition produced the lowest COP measures. No differences were observed between the other dual-task conditions. Given its high measures of reliability, this dual-task protocol might detect postural control changes in concussed youth football players.

## INTRODUCTION

Up to 1.9 million sports-related concussions occur annually within the United States' youth population.<sup>2, 14</sup> One of the most common symptoms associated with a concussion is balance dysfunction during quiet standing.<sup>10, 11, 18, 19, 26</sup> These alterations are transient but serve as a key marker when evaluating patient neurologic health post-concussion.<sup>10, 11, 17</sup> The evaluation of developing postural control systems in children presents a challenge in using postural sway assessments post-concussion.<sup>25</sup> This continued development prevents the use of measures and protocols that have shown reliability in adults without first assessing their reliability in children.

Postural control assessments using a force plate traditionally consist of quiet standing. Some researchers, though, have recently begun to use cognitive dual-tasks during postural sway assessments.<sup>1, 4, 7, 12, 20</sup> The addition of a cognitive dual-task prevents the brain from allocating all of its resources towards balance and may provide a more sensitive testing protocol for assessing patient health post-concussion. In concussed athletes, for whom cognitive and postural control deficits have been noted, the effect of performing cognitive dual-tasks on quiet standing is more pronounced.<sup>3, 7, 12, 13, 20</sup>

The purpose of this study was to investigate the reliability of a cognitive dual-task postural control testing protocol among a youth sample with no history of concussion or exposure to head impacts. Several postural sway measures and cognitive dual-tasks were explored to identify a reliable combination for the potential assessment of youth football players immediately post-concussion, as well as during athlete recovery. The development of this protocol was predicated on two key outcomes: 1) test conditions must be reliable for individual subjects over time, and 2) variation in performance between subjects must be observed. This between subjects variation implies that the protocol is sensitive to differences in postural control. With these two outcomes met, the testing protocol has the potential to differentiate COP measures of concussed athletes relative to

their baseline values. Continued research with healthy and concussed youth athletes is necessary for the development of an optimal postural control assessments within this population.

## **METHODS**

Nine youth male athletes (mean  $\pm$  standard deviation age:  $11.6 \pm 0.5$  years; height:  $1.50 \pm 0.07$  m; mass:  $40.4 \pm 7.9$  kg) were recruited to participate in this pilot study. Male subjects were used because the primary application for the protocol developed in this study is youth football. The athletes were not actively involved in a contact sport during the study period. This study was approved by the Virginia Tech Institutional Review Board, and all subjects provided verbal assent while their parents and/or guardians provided written consent.

Subjects completed four identical testing sessions spaced a week apart. During each session, postural sway was measured in nine test conditions, each of which had a duration of 120 seconds. Subjects were given a rest period between trials. The nine test conditions were comprised of five dual-task conditions which subjects performed with eyes open and eyes closed. During the eyes open conditions, subjects were instructed to look at a target on a computer screen placed at eye level at a distance of 1 m. The five dual-task conditions were no task, count backwards from 100 by two, count backwards from 100 by three, count the instances of a word in a passage read to the subject by the investigator, and the Stroop test. All tasks were conducted with the subject's eyes opened and closed, with the exception of the Stroop test, which was only conducted with the subjects' eyes open. The Stroop test involved reciting the color in which a word was written.<sup>6</sup> The no task trials were conducted first each session, with the remaining trials randomized to account for possible effects due to test order. For all tests, subjects were instructed to stand facing forward with their feet together and touching, arms at their side, and to try to remain as still as possible.

During all trials, ground reaction forces and moments under the feet were sampled at 250 Hz using a force plate (AMTI, Watertown, MA). Center of pressure (COP) coordinates were then determined and used to compute eight COP-based measures of postural sway.

Four traditional sway measures were calculated. COP standard deviation in the anterior-posterior (AP) and medial-lateral (ML) directions were calculated along the respective axis. COP mean speed was calculated as the overall COP path length divided by the test duration (120 seconds). Additionally, the COP 95% confidence ellipse area was calculated as

$$2\pi F_{.05,2,N-2}(SD_{AP}^2 SD_{ML}^2 - SD_{AP-ML}^2) \quad (1)$$

where  $F_{.05,2,N-2}$  represented the value of the F statistic for a bivariate distribution of  $N$  data points at a confidence level of 95%,  $SD_{AP}$  was standard deviation in the AP direction,  $SD_{ML}$  was standard deviation in the ML direction, and  $SD_{AP-ML}$  was the covariance between the AP and ML directions.

Four entropy measures were calculated. Sample AP and ML entropy assessed the variability of the COP trajectory along each direction. Sample entropy was determined by comparing a given data vector template from the COP trajectory to all other vectors within the trajectory and counting all those that were within a defined similarity range.<sup>16</sup> Renyi Entropy and Shannon Entropy were also calculated using graphical representations of the COP data. More specifically, these involved developing a grid that is divided based on the standard deviation in the AP and ML directions.<sup>8, 22</sup> The number of points within each subunit of the grid were then summed, with the probability of a COP coordinate residing in a particular subunit serving as the input parameter for computing both Renyi and Shannon Entropy. Detailed background on the algorithm is available.<sup>15, 24</sup> In general, lower entropy values are associated with less variability in the COP trajectories.

ANOVA was used to investigate the effect of session, test condition, and their interaction on COP measures with a significance level of 0.05. Tukey's Honest Significant Difference test was used to assess specific differences in these factors. ANOVA was also used on a subset of the data to determine the effect of test order on COP measures by test condition. Since the eyes open and eyes closed, no task conditions were always conducted first and second, they were not included in this subset analysis to assess the effect of test order.

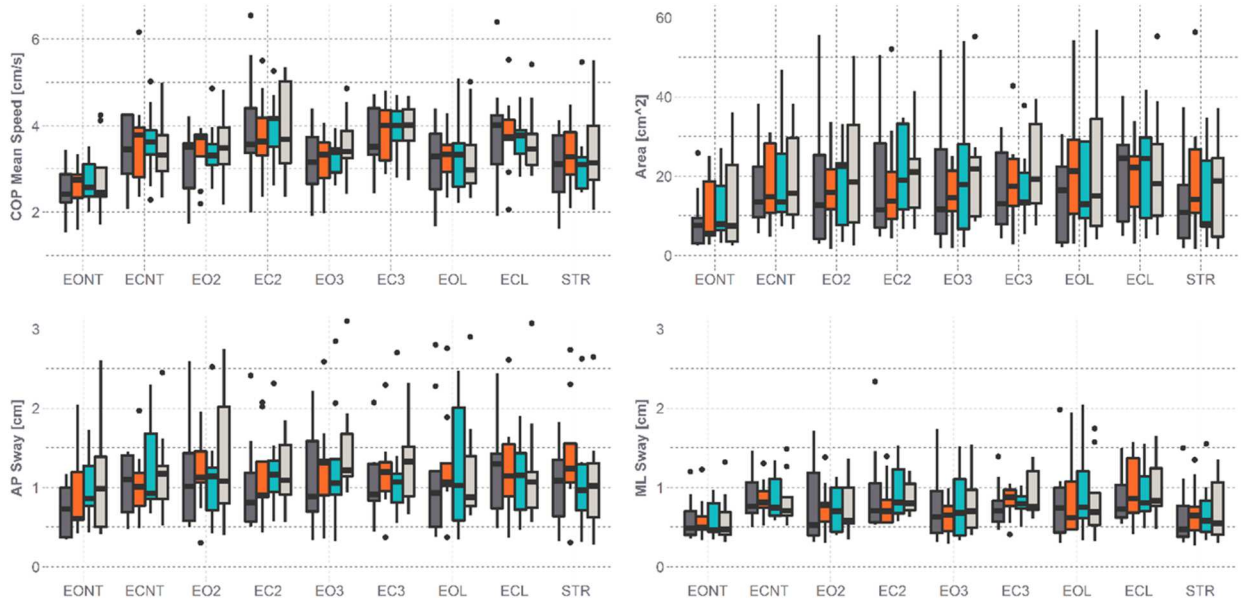
Test-retest reliability over the four weeks of testing was defined by computing the intraclass correlation coefficient [ICC(3,4k)] for each measure and test condition. ICC values were interpreted using the following criteria: 0.00-0.39 poor, 0.40-0.59 fair, 0.60-0.74 good, and 0.75-1.00 excellent.<sup>5</sup> A metric or test condition could have a high ICC value, or reliability, while failing to differentiate subjects from each other. Between-subject and within-subject variability, as well as the ratio of between-subject variability to within-subject variability, were computed to account for this. Between-subject variability was calculated as the standard deviation of mean subject performance for each measure within each test condition. Within-subject variability was calculated as the standard deviation of subject performance standard deviation for each measure within each test condition. Test conditions and measures with high ICC values, higher measures of between-subject variability and lesser levels of within-subject variability are desirable for a healthy population. This would indicate that these experimental conditions and measures are reliable indicators of intra-subject balance performance, and are sensitive to inter-subject differences.

## **RESULTS**

None of the traditional sway measures exhibited a session by test condition interaction effect ( $p = 1.000$ ) or a main effect for session ( $p > 0.3099$ ). ML standard deviation ( $p = 0.0039$ ) and COP mean speed ( $p < 0.0001$ ) exhibited a main effect for test condition, but 95% confidence ellipse



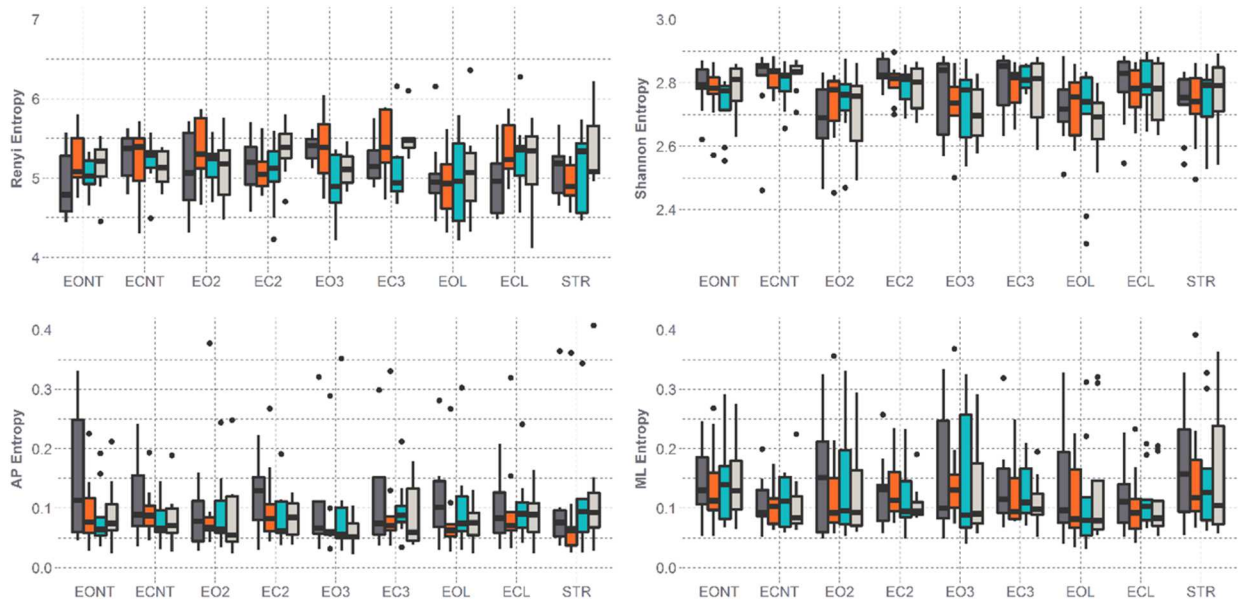
area ( $p = 0.0547$ ) and AP standard deviation ( $p = 0.3401$ ) did not (Figure 4.1). The eyes open, no task condition was associated with the lowest sway measures. In general, more sway was observed along the AP axis than the ML axis across all test conditions.



**Figure 4.1: Box plots illustrating spatially varying metrics over the four testing sessions, separated by test condition. The eyes open, no task test resulted in the lowest COP measures of all test conditions. AP standard deviation was greater than ML standard deviation. The black line in each box represents the median; boxes, the interquartile range; and black dots, the outliers. The whiskers are defined as 1.5 times the interquartile range from the first of third quartile and served as the threshold for defining outliers. EONT = eyes open, no task; ECNT = eyes closed, no task; EO2 = eyes open, count backwards by 2; EC2 = eyes closed, count backwards by 2; EO3 = eyes open, count backwards by 3; EC3 = eyes closed, count backwards by 3; EOL = eyes open, listening; ECL = eyes closed, listening; STR = Stroop**

None of the entropy measures exhibited a session by test condition interaction effect ( $p > 0.2082$ ) or a main effect for session ( $p > 0.0888$ ). Entropy measures did not differ between test conditions or session ( $p > 0.05$ ), with the exception of Shannon Entropy (*Condition:  $p < 0.0001$ ; Session:  $p = 0.858$* ). For this measure, the eyes open, no task and eyes open, listening conditions were associated with lower entropy than other test conditions (Figure 4.2). The eyes open, no task test

condition resulted in the lowest COP measures for all test conditions. Further, test order was not associated with performance ( $p > 0.05$ ) for any of the measures or test conditions.



**Figure 4.2: Entropy metrics by test condition. Performance did not vary between conditions or session for the entropy measures calculated in this study, with the exception of Shannon Entropy. The black line in each box represents the median; boxes, the interquartile range; and black dots, the outliers. The whiskers are defined as 1.5 times the interquartile range from the first of third quartile and served as the threshold for defining outliers. EONT = eyes open, no task; ECNT = eyes closed, no task; EO2 = eyes open, count backwards by 2; EC2 = eyes closed, count backwards by 2; EO3 = eyes open, count backwards by 3; EC3 = eyes closed, count backwards by 3; EOL = eyes open, listening; ECL = eyes closed, listening; STR = Stroop**

Reliability varied across measures when including all test conditions (Table 4.1). Renyi Entropy, Shannon Entropy, and 95% ellipse area exhibited poor to fair reliability ( $ICC < 0.6$ ). These measures were also associated with lower ratios of between-subject variability to within-subject variability when compared to the other measures investigated in this study. AP Entropy ( $ICC = 0.761 [0.683-0.827]$ ) and ML Entropy ( $ICC = 0.809 [0.745-0.864]$ ) were associated with excellent test-retest reliability, while AP standard deviation, ML standard deviation, and COP mean speed were observed to have good test-retest reliability (Table 4.1).

**Table 4.1: Intraclass correlation coefficients for each metric. Renyi and Shannon Entropy, as well as 95% ellipse area, were metrics associated with low levels of test-retest reliability. Subject performance, as measured by these metrics, varied considerably from week to week.**

Metric	ICC [95% Confidence Interval]
COP Mean Speed	0.707 [0.619 – 0.785]
AP Standard Deviation	0.678 [0.585 – 0.762]
ML Standard Deviation	0.626 [0.525 – 0.720]
95% Ellipse Area	0.530 [0.419 – 0.639]
Renyi Entropy	0.131 [0.032 – 0.251]
Shannon Entropy	0.533 [0.423 – 0.642]
AP Entropy	0.761 [0.683 – 0.827]
ML Entropy	0.809 [0.745 – 0.864]

Test-retest reliability for each test condition was assessed for the five measures that exhibited good or excellent reliability and was observed to vary by test condition (Table 4.2). The eyes closed, no task test condition was associated with good or excellent test-retest reliability (ICC > 0.7) for each of the five measures, as well as a ratio of between-within subject variability exceeding 3.0 for all measures. Among the measures and test conditions investigated here, AP Entropy and ML Entropy during the Stroop test exhibited the highest reliability (> 0.90) and ratio of between-within subject variance (> 8.0).

**Table 4.2: Summary of reliability and variation for each metric by test condition. ICC values exceeding 0.75 are bolded, while those between 0.6 and 0.75 are italicized. EONT = eyes open, no task; ECNT = eyes closed, no task; EO2 = eyes open, count backwards by 2; EC2 = eyes closed, count backwards by 2; EO3 = eyes open, count backwards by 3; EC3 = eyes closed, count backwards by 3; EOL = eyes open, listening; ECL = eyes closed, listening; STR = Stroop**

		EONT	ECNT	EO2	EC2	EO3	EC3	EOL	ECL	STR
COP Mean Speed [cm/s]	Between	0.23	0.42	0.23	0.34	0.22	0.23	0.31	0.39	0.31
	Within	0.09	0.13	0.09	0.15	0.10	0.07	0.09	0.12	0.10
	Ratio	2.61	3.17	2.58	2.19	2.25	3.16	3.35	3.34	3.26
	ICC	<i>0.72</i>	<b>0.80</b>	0.50	<i>0.53</i>	0.41	<i>0.60</i>	<b>0.77</b>	<b>0.75</b>	<i>0.72</i>
AP Standard Deviation [cm]	Between	0.18	0.18	0.23	0.19	0.25	0.21	0.26	0.22	0.22
	Within	0.07	0.05	0.05	0.06	0.07	0.03	0.09	0.08	0.10
	Ratio	2.60	3.31	4.32	3.02	3.89	6.30	2.84	2.59	2.08
	ICC	<i>0.66</i>	<i>0.74</i>	<i>0.72</i>	<i>0.66</i>	<i>0.73</i>	<b>0.82</b>	<i>0.69</i>	<i>0.68</i>	0.57
ML Standard Deviation [cm]	Between	0.09	0.10	0.13	0.13	0.14	0.08	0.18	0.13	0.13
	Within	0.04	0.02	0.04	0.08	0.07	0.04	0.09	0.04	0.06
	Ratio	2.15	4.62	3.08	1.61	2.00	2.31	2.08	3.07	2.23
	ICC	<i>0.62</i>	<b>0.78</b>	<i>0.69</i>	0.50	0.53	0.43	0.59	<b>0.78</b>	<i>0.67</i>
AP Entropy	Between	0.06	0.05	0.07	0.06	0.09	0.07	0.06	0.06	0.10
	Within	0.03	0.01	0.03	0.01	0.01	0.02	0.03	0.02	0.01
	Ratio	1.81	3.76	2.64	4.22	8.15	3.94	2.41	3.08	8.66
	ICC	<i>0.60</i>	<b>0.76</b>	<b>0.77</b>	<i>0.64</i>	<i>0.68</i>	<i>0.69</i>	<i>0.69</i>	<b>0.79</b>	<b>0.95</b>
ML Entropy	Between	0.06	0.04	0.08	0.05	0.10	0.06	0.08	0.05	0.10
	Within	0.02	0.01	0.02	0.01	0.02	0.02	0.03	0.01	0.01
	Ratio	3.53	5.19	3.68	7.02	4.59	3.82	2.81	6.23	8.73
	ICC	<i>0.69</i>	<b>0.80</b>	<b>0.80</b>	<b>0.89</b>	<b>0.78</b>	<b>0.83</b>	<b>0.77</b>	<b>0.89</b>	<b>0.92</b>

## DISCUSSION

This study assessed the reliability of a dual-task postural control testing protocol among a group of youth male athletes without a history of concussion or recent exposure to head impacts. Traditional COP measures and COP entropy measures were investigated during different cognitive dual-tasks to identify a reliable combination for potentially assessing concussion among youth football players. Our results indicated that selected measures and testing conditions exhibited excellent reliability (ICC > 0.75). More specifically, AP Entropy and ML Entropy exhibited excellent reliability for the highest number of test conditions, while the eyes closed, no

task and eyes closed, listening test conditions exhibited excellent reliability for the highest number of sway measures. AP Entropy and ML Entropy were observed to have the highest reliability (ICC > 0.90), as well as the highest ratio of between-subject variability to within-subject variability, for the Stroop test condition. This combination of reliability and variability suggests the capacity for this test condition to be sensitive to differences between subjects, which could potentially include concussion-related differences.

Postural control testing with youth populations has largely assessed subjects during quiet standing without the addition of a cognitive task. During eyes open and eyes closed testing with youth subjects, it has been observed that healthy and concussed populations have differing measures of postural control. Renyi Entropy values have ranged from 4.5-5.5, which was consistent with values reported here.<sup>21</sup> COP mean speed values have also been observed to vary from 1.5-3.5 cm/s in a healthy, youth population during quiet standing.<sup>23</sup> Though postural control is still developing in the youth population, previous research with this population has shown good to excellent measures of reliability for static postural control testing.<sup>9</sup> The COP and reliability values measured in the present study for the no task test conditions were in the same range as previous research utilizing quiet standing protocols.

COP measures for each test condition were observed to vary from the eyes open, no task test condition (Figures 4.1 and 4.2). The addition of a cognitive task and/or the elimination of visual feedback resulted in elevated measures for the COP measures utilized in this study. Performance was not tied to session for this subject pool. These youth subjects were not exposed to head impacts during the testing timeframe and it would not be expected for their postural control, as measured by static standing trials, to change. As subject performance did not consistently improve over time for any test conditions, no learning effect was observed in this study. Lastly, test order was determined to not be a significant factor in the measures of COP used in this study.

With nine trials lasting 120 seconds, the potential existed for fatigue to affect those test conditions towards the end of the testing protocol. By giving subjects time to rest between each test condition, the risk for fatigue was mitigated. Moreover, the future implementation of these measures and test conditions should be limited to those that exhibited good to excellent reliability, and thus reduce the number of trials required.

In addition to test-retest reliability, between-subject and within-subject variation were also computed for each COP measure and test condition combination (Table 4.2). A viable measure would be one in which within-subject variation was low, especially when compared to between-subject variation. This would mean that individual subjects' performance varied from each other while each individual subject's performance did not vary over time. If individual postural control were compromised due to concussion, we would expect the subject to have higher postural sway measures relative to a baseline because between-subject variability suggested good sensitivity. Over time, we would also expect the subject's postural sway measures to return to baseline values, which would represent within-subject variability. A protocol with specific sway measures and test conditions that maximize between-subject variability relative to within-subject variability presents the best opportunity to detect postural control changes in this population.

Several limitations of this study should be noted. This study investigated the reliability of measures and dual-task test conditions among a healthy youth sample with no history of concussion. The validity of these measures and test conditions in regards to the effects of concussion must still be evaluated. The reliability values reported in this study may not be generalizable to a different study population. As postural control is still developing in youth males, reliability of the measures and tests investigated in this study may actually be lower over longer periods of time during which further neuromusculoskeletal control development may occur.

## **CONCLUSIONS**

At present, postural control testing within youth populations largely utilizes assessments that were validated for adults. The reliability and viability of using these tools with a population still developing balance and postural control has not been addressed, which limits the generalizability of the results from these studies. Dual-task assessments have become increasingly used as part of the return-to-activity protocol in instances of sports-related concussion, as the addition of a dual-task environment increases sensitivity to either cognitive or postural deficits. A subset of the dual-task assessments utilized in this study resulted in good or excellent levels of test-retest reliability. This protocol will be applied within a group of youth football players, some of whom may experience a clinically-diagnosed concussion.

## **ACKNOWLEDGMENTS**

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## **CHAPTER 5: RELATIONSHIP BETWEEN IMPACT VELOCITY AND RESULTING HEAD ACCELERATIONS DURING HEAD IMPACTS IN YOUTH FOOTBALL**

### **ABSTRACT**

Football helmet testing standards for youth players make use of the same testing protocol for adult helmets despite research showing differences in head impact exposure between these populations. The objective of this study was to pair estimated impact velocities with linear acceleration data collected from on-field head impacts in youth football to inform youth-specific helmet testing methods. A total of 49 youth football players received helmets instrumented with accelerometer arrays to measure head acceleration throughout the season. Using video recordings of games from a single camera, impact velocities were estimated for impacts with known acceleration magnitudes. On-field accelerations ranged from 40 to 85 g, while impact velocities ranged from 0.5 to 5.5 m/s. The average error associated with these velocity estimates was below 10%, and a zoomed-in camera view provided results more consistent with true velocity. Velocities estimated from direct helmet-to-helmet impacts matched more closely with linear acceleration than other kinds of impacts. These findings may be used to inform testing methods/conditions that are more representative of impacts experienced by youth football players.

## INTRODUCTION

It is estimated that nearly 70% of the participants in football are youth players below the age of 14.<sup>9,20</sup> The helmets these players must wear are certified under a safety standard established by the National Operating Committee on Standards for Athletic Equipment (NOCSAE).<sup>16</sup> This standard was developed for adult football players and only considers catastrophic head injury, such as skull fracture, rather than less severe brain injuries like concussions. NOCSAE is working to implement a youth-specific football helmet testing standard, though further information is required to address how this new standard should differ from the current standard.<sup>15</sup>

Head impact exposure research has been conducted at all levels of football and has found that increasing impact frequency and severity are observed with increasing player age (Table 5.1). Much of this research has centered on collegiate populations, though a growing body of literature exists for head impact exposure at the youth level. Research pairing on-field data with laboratory testing showed that the NOCSAE testing standard assesses the most severe impacts players may experience on the field, with limited performance differences between matched youth and adult football helmets.<sup>12, 28</sup>

As youth football players have a different head impact exposure profile from adult players<sup>2, 5, 8, 23</sup>, different anthropometry<sup>12</sup>, and a potentially lower tolerance for concussion<sup>11, 22</sup>, further knowledge surrounding impacts at the youth level is required for the development of a youth-specific testing standard. Presently, little is known regarding the biomechanics of concussion at the youth level. Previous research has investigated concussive head impact speeds for professional football using video analysis as the basis for laboratory reconstructions in order to determine the linear acceleration magnitudes associated with the concussive impacts.<sup>14, 17, 18</sup> Pairing direct acceleration measurements with on-field impact speeds has not been done previously for a youth population and can better inform NOCSAE standards test methods. By understanding the range

of speeds at which youth players experience head impacts, a more representative testing methodology may be implemented. The primary objective of this study was to use a single-camera system to estimate impact velocities from youth football games and pair these data with measured linear accelerations for those same impacts. Second, the accuracy of these velocity estimates was compared against a true measure of speed to determine the error associated with the single-camera system. This methodology may be viable for estimating impact velocity in different sports.

**Table 5.1: Youth football head impact exposure by age. As players age, they experience a greater number of impacts as well as higher magnitude head impacts.<sup>5,7,29</sup>**

Age Group	Impacts per Season	Median Impact (g)	95% Percentile Impact (g)
7 to 8 Years <sup>29</sup>	161 ± 111	16 ± 2	38 ± 13
9 to 12 Years <sup>5</sup>	236 ± 158	19 ± 2	44 ± 8
12 to 14 Years <sup>7</sup>	275 ± 190	22 ± 2	54 ± 9

## METHODS

Two teams of youth football players (average age: 12.3 ± 0.8 years; average body mass: 35.7 ± 16.2 kg) were included in this study approved by the Virginia Tech Institutional Review Board. A total of 49 players verbally assented to participation in this study while their guardians provided written consent. Each player received a Riddell Speed, Revolution, or Speed Flex helmet equipped with accelerometer arrays associated with the Head Impact Telemetry (HIT) System.

The accelerometer arrays are mounted inside the football helmets and consist of six accelerometers. The accelerometers are mounted on an elastic base so that contact is maintained with the head throughout the duration of the head impact. This ensures that head acceleration is being measured rather than helmet acceleration.<sup>13</sup> A 10 g resultant acceleration is used as a threshold in order to avoid the accelerometers falsely registering impacts for

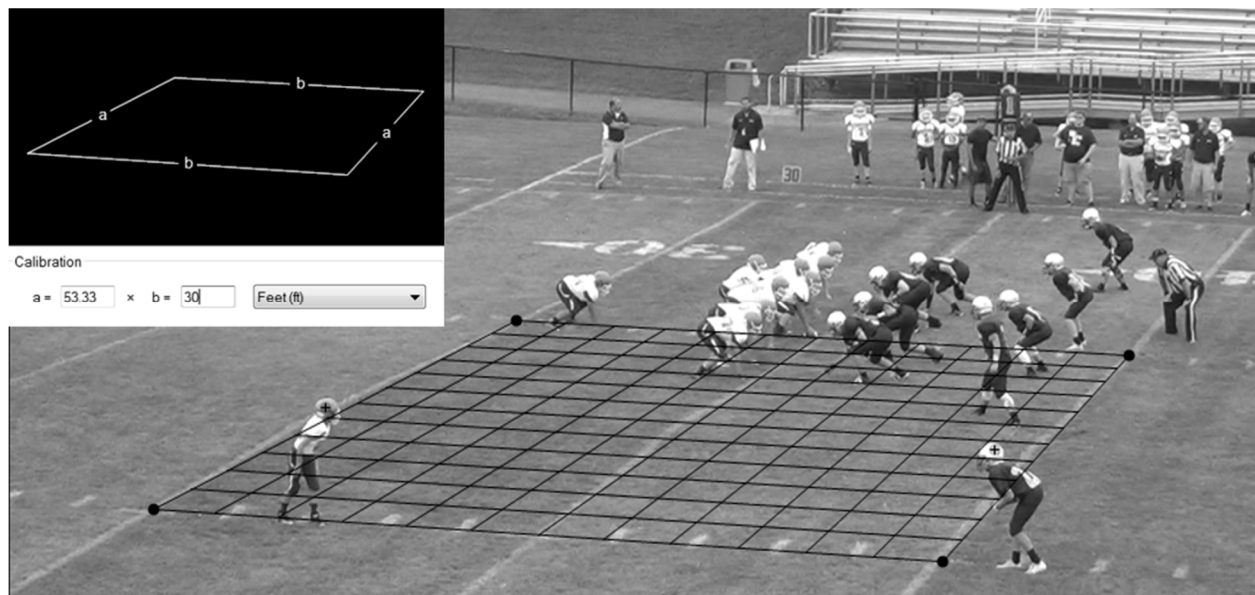
acceleration levels associated with tasks like running or jumping. All valid impact data were transmitted wirelessly from the helmets to a sideline computer to estimate linear and rotational resultant accelerations.<sup>6, 25</sup> The instrumented helmets were worn for each game and practice.

Each game was filmed using a single camera collecting video at 30 frames per second in order to visually verify head impacts. Video analysis investigated game impacts that exceeded 40 g peak linear acceleration.<sup>3, 4</sup> This study focused exclusively on those impacts that occurred during games. Practice impacts were not assessed in a similar manner, as the practice fields used by the teams in this study were not well-marked.

For the season, 336 game impacts were visually verified and resulted in a peak linear acceleration that measured greater than 40g. A subset of 50 head impacts were analyzed to evaluate this single-camera system as a means of measuring impact velocity. Head impacts were selected based on several factors. Only impacts which involved helmet-to-helmet contact between two players were chosen. Impacts in which the video footage was not stationary were excluded, as a reference grid could not be developed. Lastly, impacts in which defined field markings were not present were excluded. Football fields have consistent markings of known dimensions that help to establish a reference grid. These grids are necessary in order to compute velocity estimates.

Video footage was analyzed using open-source video analysis software (Kinovea 0.8.20, kinovea.org). For each impact, a perspective grid was developed using the field markings (Figure 5.1). The helmets of both players involved in the impact were tracked with markers over a series of video frames leading up to impact. This tracking, coupled with the dimensions of the perspective grid, allowed for determination of player displacement along the horizontal and vertical axes of the camera frame. Knowledge of the camera frame rate allowed for these displacements to be converted to average velocities for each player. The relative velocity, which

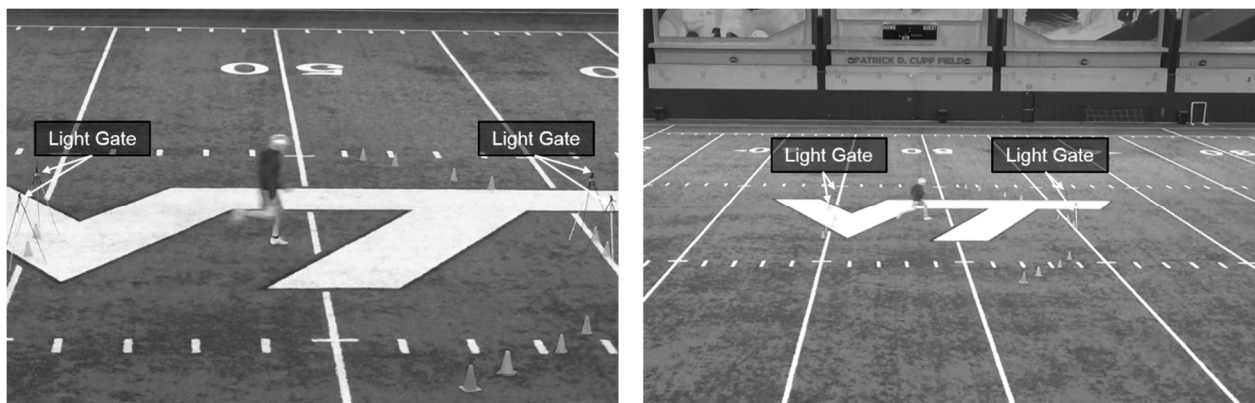
was determined by taking the difference in athlete velocity along each axis, between the two athletes represented the impact velocity. The peak linear acceleration measurement and velocity estimation were recorded, in addition to classifying each impact based on whether the head-to-head impact was the first point of contact or not. Coefficients of determination were calculated to relate estimated impact velocities to peak linear head acceleration. This was done for the overall dataset, as well as the subset of direct head-to-head impacts.



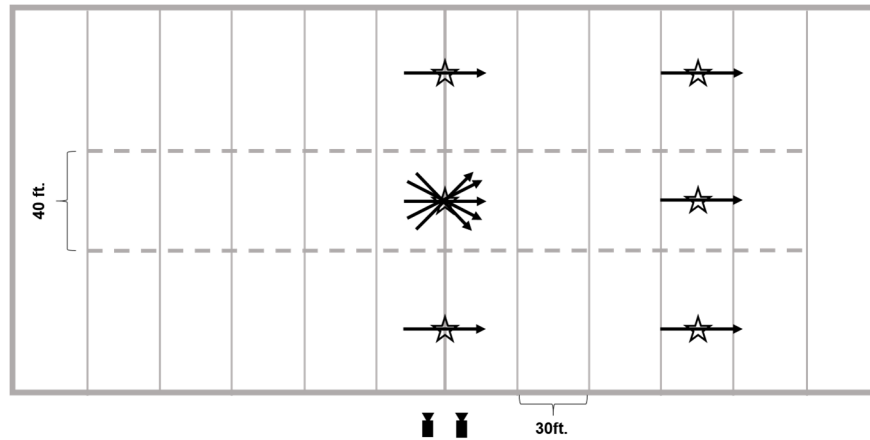
**Figure 5.1: Screenshot of perspective grid calibration in Kinovea software. Two-dimensional grids may be applied to video to estimate the kinematics of tracked objects (helmets). As only a single camera was used in this study, three-dimensional motion must be reconciled as changes in the two-dimensional grid space. Field markings of known dimensions must be in view for this method of velocity estimation. Objects are tracked using markers (seen on the helmets of the players in the foreground) over a series of frames to determine the object's change in position.**

The velocity estimates resulting from this single camera methodology were then compared against those determined using a timing system which makes use of radio frequencies (Brower TC Timing System, Brower Timing). Two cameras recording at 30 frames per second were placed at an elevated vantage point to create as close an environment to filming a football game as possible. The two cameras were positioned at the same location with differing levels of zoom in order to determine the effect of zoom on velocity estimation (Figure 5.2). Running trials were

conducted on different parts of the field to determine how the camera's distance from the view impacted velocity estimation (Figure 5.3). For all trials not at the center of the field, subjects ran 10 yards (9.14 m) in a straight line at either full speed or a self-selected lesser speed. Trials at the center of the field involved 10 yard runs at known angles of 30° or 45°, with subjects running both towards the camera for one trial and away from the camera in another trial. Subjects were instructed to take a running start of at least 5 yards in order to get up to speed prior to engaging the timing system. The timing system was set up at the beginning and end of the 10 yard running zone for each trial. A total of 25 trials were conducted for the various configurations outlined. The percent error between the single camera velocity estimation and the timing system velocity was computed for each trial for both the wide and zoomed views.



**Figure 5.2: Comparison of zoomed-in (left) and wide (right) camera views. The wide view provides a larger camera frame, while the zoomed-in view provides greater resolution for a given area of interest. The cameras were set-up next to each other and placed at an elevated point relative to the field to emulate filming from a press box at a stadium during a game.**



**Figure 5.3: Football field layout.** The field is lined evenly, with distinct markings of known dimensions. Stars represent locations on the field where running trials were conducted. Black arrows represent the running paths. The two video cameras were elevated.

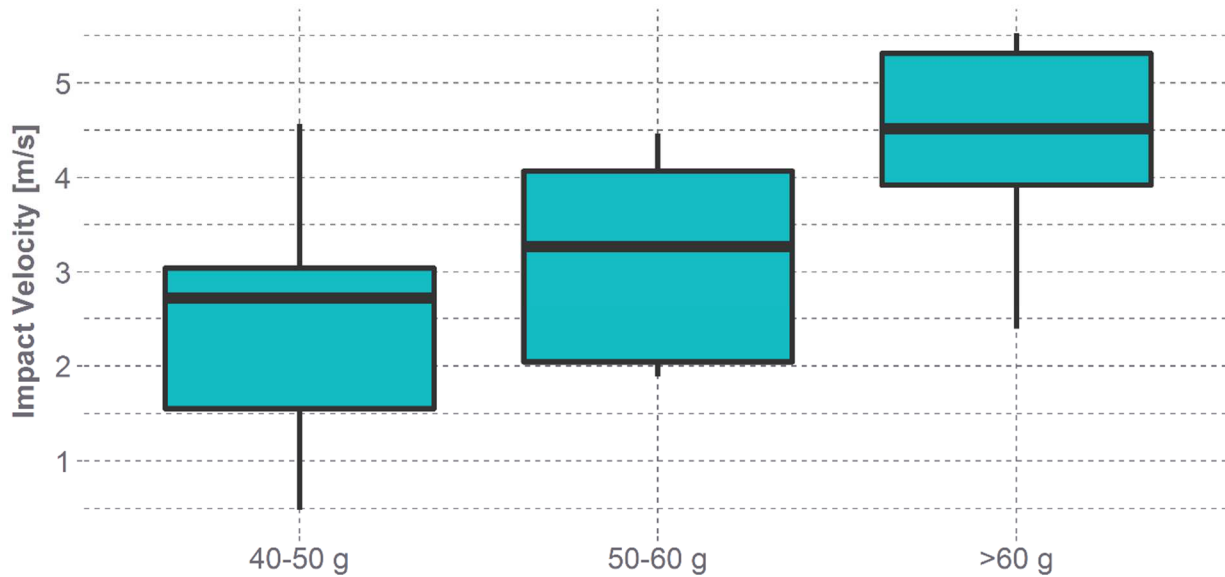
## RESULTS

The head impacts included in this study ranged in peak linear acceleration from 40-85 g. These impacts represented the top 10% of all game impacts experienced by players in this study in terms of impact magnitude. Most of the impacts were between 40 and 50 g (30), with a nearly even split for impacts between 50 and 60 g (9) and those above 60 g (11). The number of impacts in which head-to-head contact was the first point of contact (27) was similar to the number of impacts in which player-to-player contact or helmet-to-shoulder contact occurred first (23). Nearly all of the head impacts above 60 g (10 of 11) had head-to-head contact first.

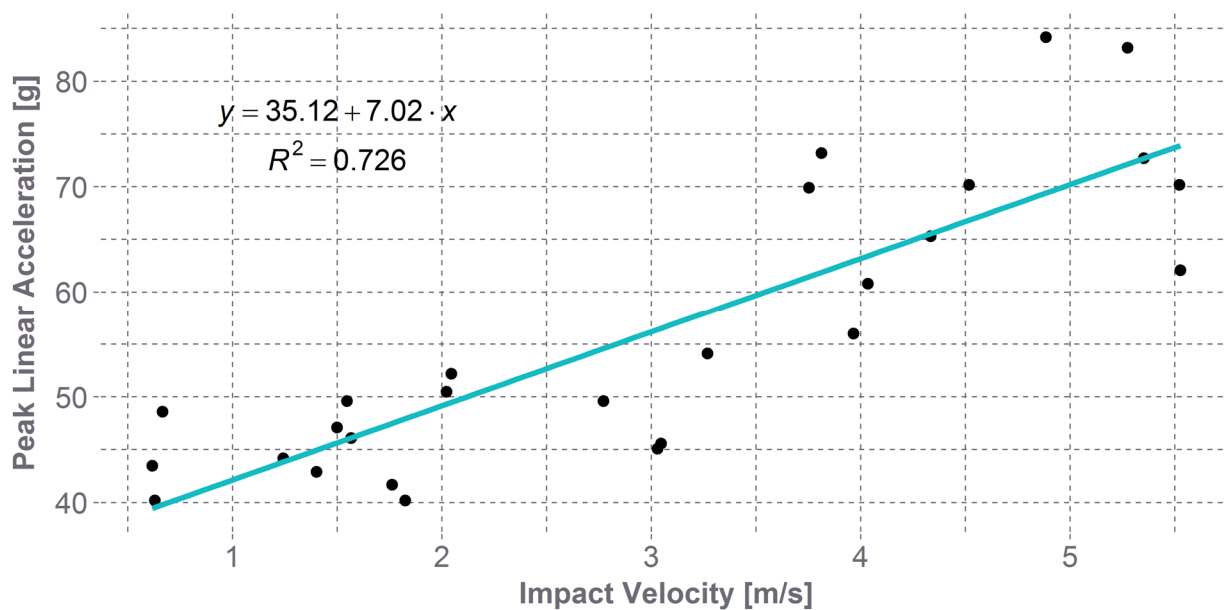
Individual player velocities varied from 0.2 to 5.4 m/s, while relative velocity varied from 0.5 to 5.5 m/s. In general, head impacts with higher peak linear accelerations were associated with higher impact velocities (Figure 5.4). Within each acceleration grouping, impact velocity varied.

For all impacts, only 37% of the variability in peak linear acceleration was found to be explained by impact velocity. By reducing the dataset to only the 27 head impacts in which head-to-head

contact was the first point of contact, a much stronger relationship between relative velocity and peak linear acceleration ( $p < 0.0001$ ;  $R^2 = 0.726$ ) was observed (Figure 5.5). The highest impact velocity estimates were found to be associated with the highest linear acceleration measurements.



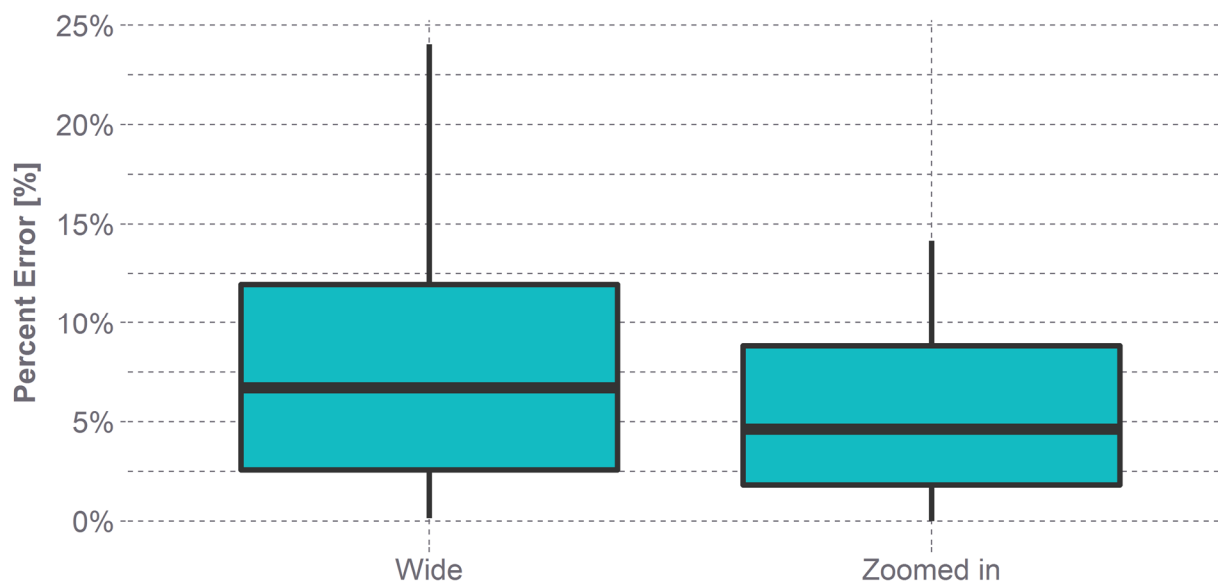
**Figure 5.4: Estimated impact velocity grouped by acceleration level. Increasing acceleration levels were associated with higher impact velocity.**



**Figure 5.5: Peak linear acceleration as a function of impact velocity for head-to-head impacts. Increasing impact velocity was associated with increased peak linear acceleration.**



For the single-camera system employed in this study, average error was observed to be less than 10%. Average error was  $8.49 \pm 7.06\%$  with the wide camera view, while the error was  $5.49 \pm 3.98\%$  for the zoomed view (Figure 5.6). Only 19 of the 25 total trials resulted in successful tracking for velocity estimation using the wide camera view. Larger errors were observed for trials where the subject moved either towards or away from the camera.



**Figure 5.6: Distribution of error in velocity estimation for the two camera views used in this study. Overall, errors in estimation were below 10%, with a few trials resulting in worse estimates. A zoomed-in view was associated with less error on average than a wide view.**

## DISCUSSION

This study paired on-field head acceleration measurements from head impact with estimated impact velocities during youth football games. The impacts represented the top 10% of all head impacts players experienced in games. It was observed that higher linear accelerations were associated with higher impact velocities (Figure 5.4). The most severe impacts generally occurred at higher velocities, while some impacts with lower acceleration levels resulted from high impact velocities. These impacts largely were the result of head-to-head contact not being the

first point of contact. Velocity estimates tracked helmet motion for the involved players up to the point of impact. Impacts where shoulder pads collided or the shoulder pad struck the helmet first would lead to decreases in true impact velocity, which would not be reflected in the velocity estimate. These impacts would result in lower measures of linear acceleration as well.

Given the limited relationship between impact velocity and measured linear acceleration for the overall dataset, the data were classified on the basis of whether head-to-head contact served as the first point of contact. As tracking markers on the helmets served to estimate impact velocity, it follows that impacts in which head-to-head contact occurred first should yield a more consistent relationship between impact velocity and linear head acceleration. Of the 50 impacts included in this study, 27 resulted in head-to-head contact as the first point of contact. The largest acceleration measurements were observed when head-to-head contact occurred first. Velocity estimates from these head-to-head impacts were better associated with linear acceleration than for the overall dataset (Figure 5.5). Nearly 75% of the variability in head acceleration was explained by impact velocity for this subset. Different sources of measurement error likely contributed towards the remainder of this variability and are considered as limitations of this work.

One of the head impacts assessed in this study resulted in a diagnosed concussion for the athlete. The impact was to the front of the helmet and had a linear acceleration of 70 g and a rotational acceleration of 3716 rad/s<sup>2</sup>. This impact was the most severe linear acceleration for that player for that game. The concussed athlete was playing defensive line and was impacted by an opposing lineman shortly after play became active. The impact velocity for this impact was estimated to be 3.75 m/s. For comparison, the average concussive impact in the National Football League was found to be  $9.3 \pm 1.9$  m/s, with linear acceleration values of  $98 \pm 28$  g.<sup>18</sup> Conducting similar analysis with other known concussive impacts at the youth level would be beneficial, though a larger number of concussive data points at the youth level is necessary in order to fully

characterize the biomechanics of concussion in youth football.

The accuracy of the velocity estimations from the single-camera system was assessed through experimental trials with a timing system. Previous research using Kinovea has found velocity estimation errors to be 5%.<sup>19, 21</sup> It was observed that average error from all the trials was below 10%, with a zoomed view offering estimations closer to those measured by the timing system (Figure 5.6). A zoomed in view would offer greater pixel resolution for player helmets than a wide view, where individual players and their helmets are smaller relatively. This is evidenced by the fact that, for six of the 25 trials, velocity estimates could not be generated for the wide camera view due to difficulty tracking the helmet. It was also found that trials in which subjects ran at an angle either towards or away from the camera were associated with larger velocity estimate error. The video cameras are typically perpendicular to player motion, so the errors associated with in-game impacts would be expected to be on the lower end of the spectrum computed in this study.

There were some difficulties associated with tracking players during live games. As with the experimental trials, wider views led to limited tracking ability. However, the use of views that are too zoomed in also poses an issue. For impacts directly after the snap or around the line of scrimmage, this method works well. The camera must follow the game action in order to capture all potential impacts, though camera motion prevents estimation of velocity using the single-camera system. Striking the right balance represents one of the key challenges of this kind of analysis, though the use of two cameras, one with a zoomed view and the other with a wider view, would likely represent an optimal solution in order to maximise the potential to capture impact velocity for head impacts in games. In football, there are a total of 22 players on the field at one time. Tracking a single player who experienced a head impact on a given play can be complicated by the potential for other players to cross in front of the player of interest. In these scenarios, the

software may lose track of the player of interest. Depending on proximity to impact, this may limit the ability to estimate velocity.

Several factors affected the relationship between impact velocity and resulting peak linear head acceleration. The HIT System has been shown to have individual head acceleration measurement errors that can be as large as 15%.<sup>1</sup> Impacts to the helmet facemask, which can reduce the coupling of the accelerometers and the head, are generally associated with greater error than locations on the helmet shell.<sup>10, 27</sup> The single camera system used to estimate impact velocities was found to overpredict the true velocity, though this effect was less than 10% on average. Impact velocities were computed as average velocities, rather than instantaneous velocities. For some impacts where a player was either accelerating or decelerating during the tracking process, this may result in larger measurement errors. Impacts where head-to-head contact did not represent the first point of impact would thus be associated with higher speeds relative to the resulting linear accelerations. Different locations on football helmets may offer differing levels of impact attenuation.<sup>24, 26</sup> For the same input velocity, an impact to the front of the helmet could result in lower levels of linear acceleration than the same impact to the side of the helmet. Lastly, there may be some error associated with the position of players' heads, as the software must reconcile player height in terms of the coordinate system setup by the reference grid rather than as a separate third dimension.<sup>19</sup>

## **CONCLUSION**

A methodology for estimating impact velocity from a single video camera was evaluated for youth football head impacts. On average, velocity errors associated with this system were below 10%. This single camera method will be compared against a multi-camera system in the future. These impact velocities were paired with in-helmet instrumentation measured head acceleration during impact. Increasing head acceleration measurements were observed for increasing impact

velocities, with this relationship most pronounced for direct head-to-head impacts. This combination exists for professional football, but has not been previously done at the youth level. These data have direct application for use by standards committees in developing youth-specific testing methods that are representative of real-world head impacts. The currently proposed youth football testing standard stipulates that tests are conducted at 5.2 m/s, which would be among the highest velocities observed in this study. Assessing these high severity impacts in standards testing likely represents the best opportunity to assess impacts at levels that are expected to be injurious. Further, these data may also prompt manufacturers to begin to develop youth-specific football helmets.

## **ACKNOWLEDGMENTS**

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## **CHAPTER 6: RELATING ON-FIELD HEAD IMPACTS TO LABORATORY TESTING PROCEDURES**

### **ABSTRACT**

A youth-specific football helmet testing standard has been proposed to address the physical and biomechanical differences between adult and youth football players. This study sought to relate the proposed youth standard to on-field head impacts collected from youth football players. Head impact data from 112 youth football players (ages 9-14) were collected through the use of helmet-mounted accelerometer arrays. These head impacts were filtered to only include those that resided in corridors near prescribed National Operating Committee on Standards for Athletic Equipment (NOCSAE) impact locations. Peak linear head acceleration and peak rotational head acceleration magnitudes collected from pneumatic ram impactor tests as specified by the proposed NOCSAE youth standard were compared to the distribution of on-field head impacts. All laboratory impact tests were among the top 10% of matched location head impacts experienced by youth football players. Assessing the most severe head impacts is crucial for eliminating skull fracture or serious brain injury. This testing standard may be refined to focus on mitigating concussion incidence as more becomes known regarding concussion tolerance and the biomechanics of concussion among youth athletes.



## INTRODUCTION

Sports-related concussions have received considerable attention as research has shown the potential for long-term, deleterious effects associated with these injuries.<sup>32</sup> Football has a high incidence of brain injury and has been linked to conditions like chronic traumatic encephalopathy (CTE).<sup>24, 25, 32</sup> Recently, researchers have related not only a history of concussions, but repeated head impacts in general, that may lead to these lasting effects.<sup>33</sup> Head impact exposure can be mitigated through proper teaching, rule modification, and better equipment.<sup>9, 27</sup> Proper teaching and rule modification would predominately focus on limiting an athlete's frequency of impact. Better equipment would serve to provide greater protection for head impacts that remain after other methods have been implemented.<sup>27</sup>

Currently, all football helmets are certified by the National Operating Committee on Standards for Athletic Equipment (NOCSAE).<sup>20</sup> This standard was developed to limit catastrophic head injury, like skull fracture, and has succeeded in this regard.<sup>18</sup> It has been found that the testing standard assesses only the most severe impacts that players may experience on the field and that youth and adult football helmets do not differ in terms of performance.<sup>31</sup> Youth football players are a unique population, differing from adults in terms of head impact exposure and anthropometry.<sup>2, 6, 10-14, 16, 28, 34</sup> While youth players may experience impacts as severe as adult players do, on average impacts at the youth level are associated with lower acceleration levels. Further, youth players have a head-body ratio that is considerably different from adults.<sup>16</sup> Consideration of these factors would benefit a youth-specific testing standard.<sup>19</sup>

On-field head impacts from football have been previously paired with laboratory testing for the development of the Virginia Tech STAR Ratings.<sup>26, 29</sup> NOCSAE recently proposed a youth-specific testing protocol that considers both linear and rotational head acceleration.<sup>21, 22</sup> The objective of this study was to relate the proposed youth standard to on-field head impacts. The

testing protocol should be representative, in that real-world impact velocities should be used to generate head accelerations that are realistic for that energy input. These data may confirm the proposed testing method or serve as the basis for optimizing the testing methodology to match on-field head impact kinematics.

## **METHODS**

On-field head impact data were collected from youth football players over four seasons between 2015 and 2018 in this study approved by the Virginia Tech Institutional Review Board. A total of 112 youth football players provided verbal assent to participate in this study while their guardians provided written consent. Each athlete wore either Riddell Speed (Speed) or Riddell SpeedFlex (SpeedFlex) helmets, both of which are currently commercially-available football helmets. Player ages ranged from 9 to 13 for athletes wearing the Speed helmet and from 12 to 14 for athletes wearing the SpeedFlex helmet.

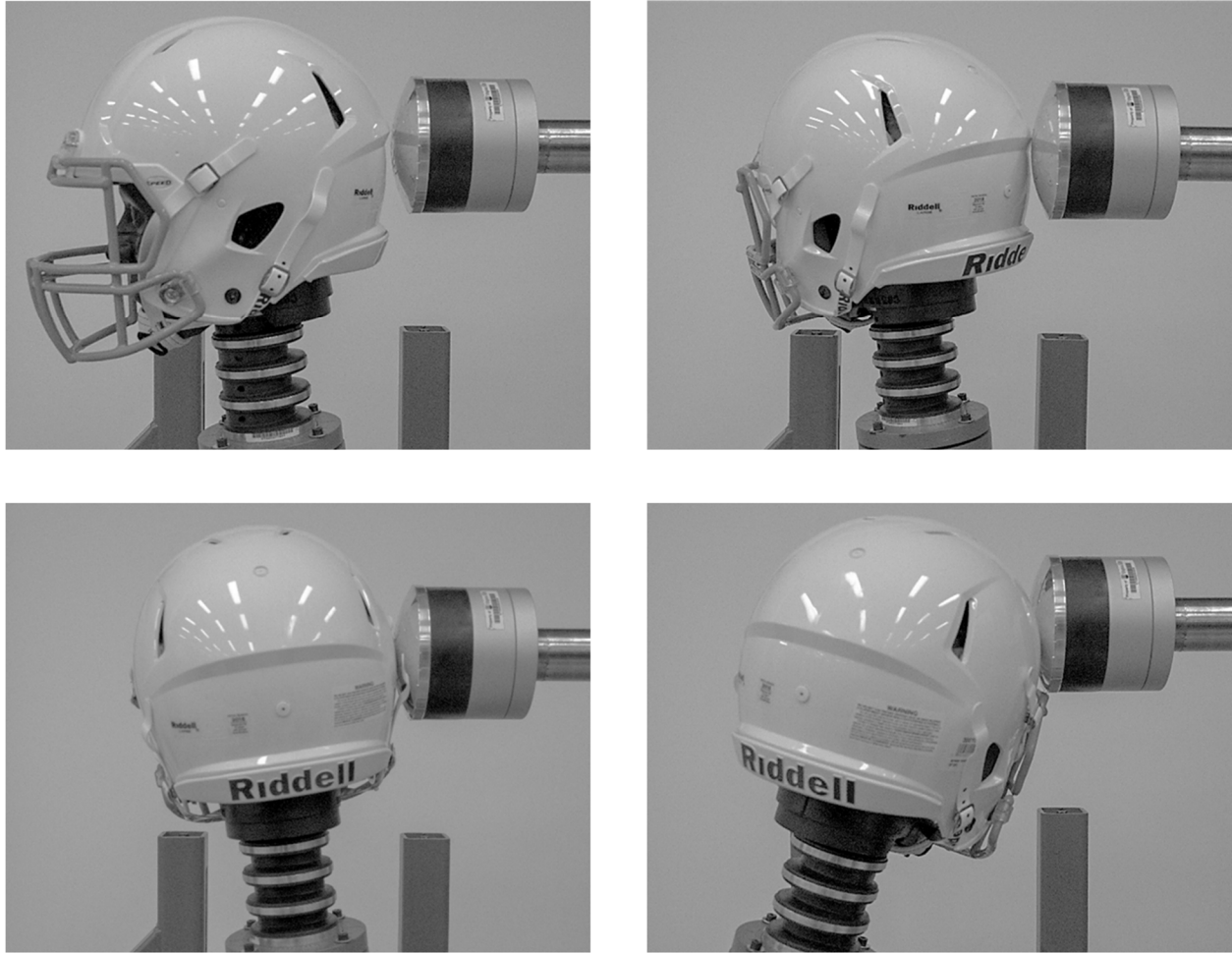
All helmets were instrumented with an accelerometer array associated with the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH USA). The array consists of six accelerometers mounted on an elastic base to maintain contact between the array and the head throughout impact. This ensures that head acceleration is measured. Previous work has shown that head acceleration represents a small percentage (<10%) of helmet acceleration.<sup>17</sup> Players wore these instrumented helmets for every game and practice. Any resultant acceleration reading below 10g was automatically filtered out of the dataset to eliminate acceleration levels associated with running or jumping rather than head impacts. Linear and rotational head accelerations for valid impacts were estimated.<sup>8,30</sup>

These on-field head impact data were filtered to include only those impacts which matched testing locations for the proposed youth standard.<sup>21,22</sup> Each impact recorded by the HIT System is also

defined by measures of azimuth and elevation angles to characterize impact location. The test locations consider all regions of the helmet, as well as centric and non-centric head impact configurations (Table 6.1, Figure 6.1). Positive rotation in the y-direction is tilting towards the ram, while positive rotation in the z-direction is clockwise. Positive displacement in the y-direction is represented by motion anterior to the coronal plane, while positive displacement in the z-direction consists of motion superior to the basic plane. The combination of rotation and translation resulted in an elevation angle of 17.8° for the side, rear boss, and rear locations, and an elevation angle of 30.7° for the front boss location. On-field head impacts within +/- 15° of azimuth and elevation of a test location were considered representative of standard testing conditions.

**Table 6.1: Pneumatic ram impact test locations. Measurements of y and z-position are relative to the basic plane of the NOCSAE headform. The proposed NOCSAE standard also stipulates testing at a non-centric rear boss location and a random location on the helmet, neither of which could reasonably be reconciled to the on-field data.<sup>22</sup>**

Impact Location	Y Rotation (deg)	Z Rotation (deg)	Y Position (mm)	Z Position (mm)
Side	7	-90	0	60
Rear Boss (Center of Gravity)	7	-135	-81	60
Rear	7	-180	0	60
Front Boss	15	-60	56	73



**Figure 6.1: Pneumatic ram impact test locations. Clockwise from top left: rear, rear boss, front boss, and side. All impacts were conducted with a medium-sized NOCSAE headform and a large youth football helmet.**

The NOCSAE testing protocol utilizes a pneumatic ram to impart energy to a head-neck-torso assembly, which is free to move upon impact. The youth-specific testing on the pneumatic ram is conducted at a velocity of 5.2 m/s ( $\pm 3\%$ ) for all test locations. For a helmet to pass the standard, the Severity Index (SI) value must not exceed 1200 and the peak rotational acceleration must not exceed 6000  $\text{rad/s}^2$ . The SI is a correlate of energy transfer to the head during an impact and is defined by acceleration ( $a$ ) and time ( $t$ ) [Equation 1]. At present, the proposed youth standard does not differentiate from the adult standard in terms of testing equipment.<sup>22</sup>

$$SI = \int a(t)^{2.5} dt \quad (1)$$



**Figure 6.2: Pneumatic ram linear impactor. A helmeted headform and neck assembly are mounted on the linear slide table. This system has the capacity to test speeds up to 11 m/s.**

A pneumatic ram linear impactor was used that is in accordance with NOCSAE testing standards (Figure 6.2). The impactor head weighs 2.3 kg and has a convex face with a radius of 127 mm.<sup>21</sup> A medium NOCSAE headform (57.6 cm circumference) modified to fit a Hybrid III 50<sup>th</sup> percentile neck was used in order to generate a more realistic helmet fit without altering the impact response.<sup>5, 7</sup> This head and neck assembly was mounted to a linear slide table (Biokinetics, Ottawa, Canada) weighing 15.75 kg. The slide table had 5 degrees of freedom, which allowed for adjustment and consistent orientation of the head and neck for each test. The NOCSAE headform was instrumented with three single-axis accelerometers (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA), as well as a tri-axial angular rate sensor (ARS3 PRO-18K, DTS, Seal Beach, CA). All instrumentation was mounted at the headform center of gravity to allow for linear and rotational head kinematics to be computed with six degrees of freedom. All data were sampled at 20,000 Hz. Linear acceleration data was filtered according to SAE J211 standards (CFC 1000), while angular rate data was filtered at a frequency of 255 Hz. For each test, peak linear resultant head acceleration, peak rotational resultant head acceleration, and SI were computed.

Two models each of the Speed and SpeedFlex helmets were tested at each of the four impact locations prescribed by the proposed youth standard that could be matched to on-field data (Table 6.1).<sup>22</sup> Each helmet-location combination was repeated three times, resulting in 48 total tests. Data from repeated trials were averaged for each helmet and location. Peak linear and rotational head accelerations from each of the conditions were mapped to the on-field data using a bivariate cumulative distribution.

Two-way ANOVA for helmet and impact location was used to determine the differences in on-field head acceleration data and the laboratory impact testing. Tukey's Honest Significant Difference test was used post-hoc to determine which factor levels differed from each other. A significance level of 0.05 was used in this study.

## **RESULTS**

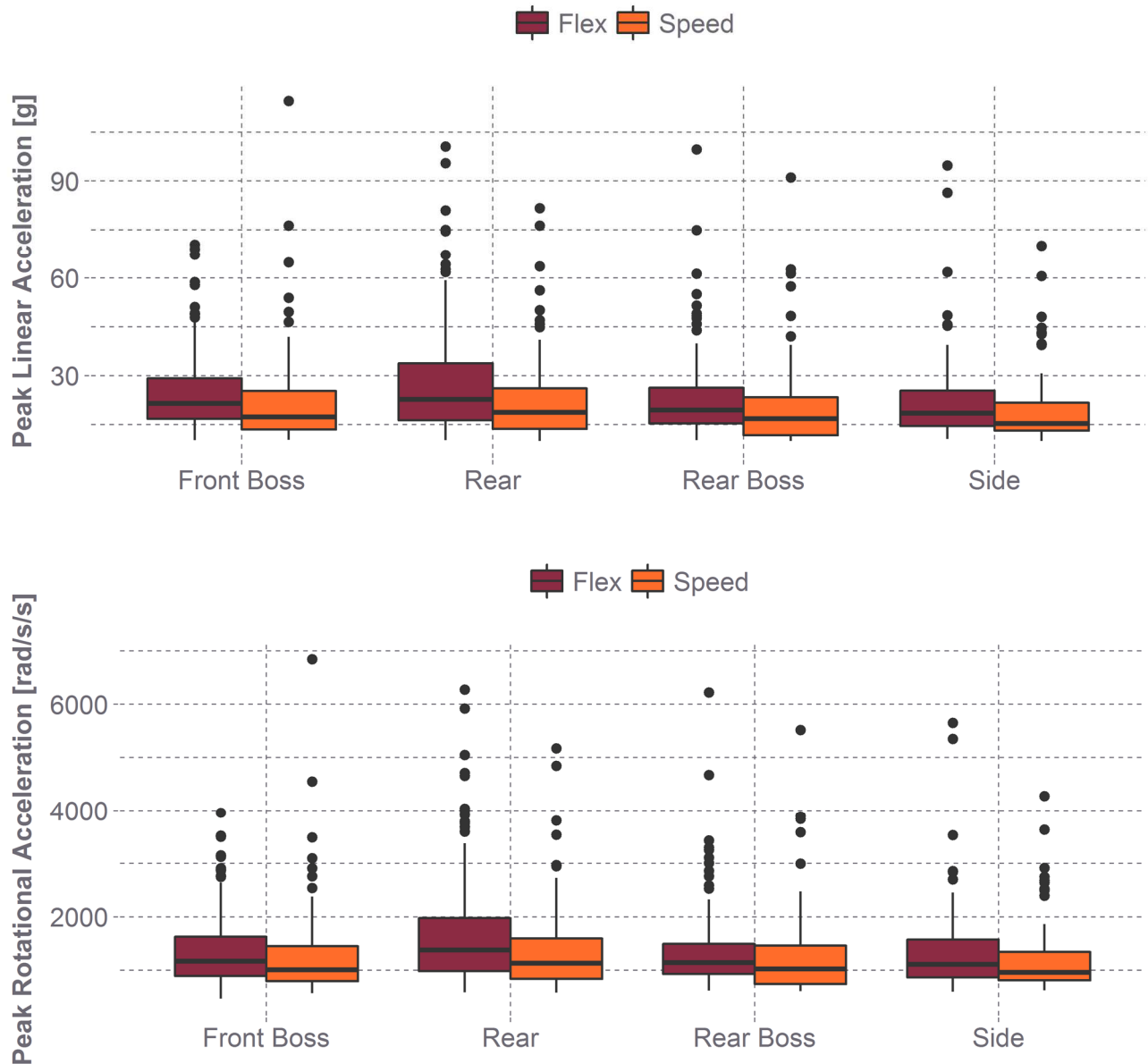
A total of 18,572 head impacts were recorded for the youth players in this study, with 10,099 occurring in the Speed helmet and 8473 in the SpeedFlex helmet. For players wearing the Speed helmet, the median and 95<sup>th</sup> percentile peak linear head acceleration values were 18.3 g (95% confidence interval (95% CI): 18.0-18.5 g) and 49.5 g (95% CI: 48.4-50.8 g), respectively. The median and 95<sup>th</sup> percentile peak rotational head acceleration values were 905 rad/s<sup>2</sup> (95% CI: 894-917 rad/s<sup>2</sup>) and 2485 rad/s<sup>2</sup> (95% CI: 2425-2544 rad/s<sup>2</sup>) for this group. Those athletes wearing the SpeedFlex helmet had median and 95<sup>th</sup> percentile peak linear head acceleration values of 19.9 g (95% CI: 19.7-20.2 g) and 44.3 g (95% CI: 42.9-45.2 g) and median and 95<sup>th</sup> percentile peak rotational head acceleration values of 1042 rad/s<sup>2</sup> (95% CI: 1028-1054 rad/s<sup>2</sup>) and 2448 rad/s<sup>2</sup> (95% CI: 2397-2513 rad/s<sup>2</sup>).

After filtering the on-field head impact data to only include those head impacts which fell within the prescribed impact locations tested, 567 head impacts remained for the Speed (5.6%) and 767 head impacts remained for the SpeedFlex (9.1%). On-field head impacts for athletes wearing the SpeedFlex helmet were associated with higher magnitudes for the rear impact location for both peak linear head acceleration ( $p = 0.001$ ) and peak rotational head acceleration ( $p = 0.003$ ) (Figure 6.3). No differences in peak linear head acceleration ( $p > 0.446$ ) or peak rotational head acceleration ( $p > 0.534$ ) were observed between locations for the Speed. For the SpeedFlex, the rear impact location was associated with higher peak linear head acceleration values than the side impact location ( $p = 0.013$ ). For the SpeedFlex, the rear impact location was associated with higher peak rotational head acceleration values than the side impact location ( $p = 0.009$ ) and the front boss impact location ( $p < 0.0001$ ).

Pneumatic ram impact tests resulted in peak resultant linear acceleration values that ranged from 49.3 to 77.0 g, peak resultant rotational accelerations values that ranged from 3096 to 6585 rad/s<sup>2</sup>, and SI values that ranged from 96 to 206. The most severe impact kinematics were observed for the Speed helmet at the front boss impact location. All of the impact tests conducted in this study resulted in impact kinematics that were within the top 10% of on-field head impacts measured for youth players for both helmets (Table 6.2).

Differences in peak linear acceleration between the Speed and SpeedFlex helmets were observed for all locations ( $p < 0.038$ ) except for the rear boss location ( $p = 0.232$ ). Differences in peak rotational acceleration were observed for all locations ( $p < 0.002$ ) except for the side location ( $p = 0.996$ ). For the SpeedFlex helmet, the rear impact location resulted in higher peak linear acceleration values than all other locations ( $p < 0.001$ ). Peak rotational acceleration was observed to vary between the side and front boss impact locations ( $p = 0.007$ ). For the Speed helmet, the front boss impact location resulted in higher peak linear acceleration values than all

other locations ( $p < 0.046$ ). Peak rotational acceleration was lowest at the rear impact location and highest at the front boss location ( $p < 0.001$ ).



**Figure 6.3: Summary of on-field data by impact location. Impacts at the rear location for athletes wearing the SpeedFlex helmet were associated with higher peak linear and rotational head acceleration values than the side impact location. Differences in on-field head impact distributions between the two helmets may be attributed to differences in player age and differences in helmet impact performance.**



**Table 6.2: Summary of laboratory impact tests mapped to on-field data. Nearly all of the impact configurations were among the top 3% of head impacts experienced by youth football players.**

	Location	PLA [g]	PRA [rad/s <sup>2</sup> ]	Percentile
Speed	Front Boss	74.0 ± 1.5	6351 ± 219	98.5
	Rear	65.8 ± 2.1	3322 ± 231	98.8
	Rear Boss	58.2 ± 2.8	4004 ± 292	98.9
	Side	69.2 ± 3.2	4124 ± 246	98.3
SpeedFlex	Front Boss	54.8 ± 3.3	4711 ± 184	100
	Rear	70.8 ± 1.6	4465 ± 218	93.4
	Rear Boss	61.9 ± 1.7	4550 ± 167	97.3
	Side	57.5 ± 3.5	4214 ± 172	98

## DISCUSSION

This study sought to investigate the proposed NOCSAE youth football helmet testing standard on the pneumatic ram as it related to on-field head impact kinematics. Peak linear head acceleration values and peak rotational head acceleration values for laboratory impacts were observed to be among the hardest head impacts youth football players would experience on the field for both helmets considered in this analysis.

On-field head impacts collected from youth football players over four seasons of play served as the point of comparison for the laboratory results in this study. The median and 95<sup>th</sup> percentile peak linear head acceleration and peak rotational head accelerations reported here are consistent with previous research with this population.<sup>4, 6, 11, 15</sup> Differences between the on-field distributions of the two helmets were noted, though it should be noted that the players in the SpeedFlex helmets were from an older population, and that only a subset of all head impacts were included in the analysis. This may explain why the median acceleration values are greater for the athletes wearing the SpeedFlex helmet, but the 95<sup>th</sup> percentile acceleration values do not exhibit the same trend.

A helmet testing standard should assess the hardest head impacts that a player may experience on the field. All pneumatic ram impact tests in this study were among the top 10% of head impacts from the on-field data set for both helmets. With the exception of the rear impacts to the SpeedFlex helmet, the impact kinematics produced in lab were in the 97<sup>th</sup> percentile or greater of the on-field head impacts (Table 6.2). The SI values in this study were much lower than the proposed threshold of 1200, while the peak rotational head acceleration values exceeded the 6000 rad/s<sup>2</sup> threshold for the front boss location for the Speed helmet. There is no prescribed rotational acceleration threshold for the other-sized NOCSAE headforms, but it is possible that different kinematic responses may result. All pneumatic ram impact tests were conducted at 5.2 m/s, as prescribed by the proposed NOCSAE youth football helmet testing standard. This velocity is on the upper end of head impact velocities that have been reported for youth football.<sup>3</sup>

Both the impact velocity and the resulting linear and rotational kinematics are at the high end of what youth football players experience on the field, which is what a testing standard should assess. Impact testing was conducted with an impact surrogate that is representative of a 50<sup>th</sup> percentile adult male, and not that of a youth male. NOCSAE's standard does require that the small-sized NOCSAE headform be used for testing with medium-sized football helmets, which is a more representative head for a youth population. However, the 50<sup>th</sup> percentile Hybrid III neck and 16 kg sliding mass are still used. Known differences in anthropometry and neck strength between children and adults may contribute towards impact response.<sup>14, 23</sup> The use of a population-specific test surrogate may be appropriate in order to optimize a youth football helmet testing standard that is representative of the youth population and the head impacts they may experience.

Several limitations of this study should be noted. Firstly, only two football helmets were considered in this analysis, and helmet performance may differ with other helmet models. The

two helmets (Riddell Speed and Riddell SpeedFlex) were used as they are the same models in which the on-field data were collected. Secondly, only the medium-sized NOCSAE headform and youth large helmets were used for impact testing. Differences in impact performance between headforms or helmet sizes were not assessed. Lastly, the helmet-mounted accelerometers used in this study have known measurement errors for individual measurements, though this error is minimized when investigating distributions of the head impact data.<sup>1</sup>

## **CONCLUSION**

The development of a youth-specific testing protocol represents a key step forward by NOCSAE. At present, the proposed standard results in impact kinematics that are consistent with the highest severity head impacts a youth player may experience. This testing standard represents a crucial first step towards considering the differences between youth and adult football players. The use of a youth-specific testing surrogate may be appropriate in order to ensure that the testing standard is truly representative of what a youth football player may experience. Should the testing standard evolve to address concussion, rather than severe head injuries which have largely been eliminated from football, further knowledge regarding concussion tolerance and the biomechanics of concussion in youth collected using on-field data will be crucial.

## **ACKNOWLEDGMENTS**

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## **CHAPTER 7: DEVELOPMENT OF A CONCUSSION RISK FUNCTION FOR A YOUTH POPULATION USING HEAD LINEAR AND ROTATIONAL ACCELERATION**

### **ABSTRACT**

Physical differences between youth and adults, which include incomplete myelination, limited neck muscle development, and a higher head-body ratio in the youth population, contribute towards the increased susceptibility of youth to concussion. Previous research efforts have considered the biomechanics of concussion for adult populations, but these known age-related differences highlight the necessity of quantifying the risk of concussion for a youth population. This study adapted an existing injury metric that combines linear and rotational head acceleration to assess the risk of concussion for a youth population. Survival analysis was used in conjunction with head impact data collected during participation in youth football to model risk between individuals who sustained medically-diagnosed concussions (n=15). Receiver operator characteristic curves were generated for several injury metrics and parameters, all of which were observed to be better injury predictors than random guessing. The youth metric was associated with an area under the curve of 0.89 (95% confidence interval: 0.82 - 0.95) when all head impacts experienced by the concussed players were considered. Concussion tolerance was observed to be lower for youth athletes, with average peak linear head acceleration of  $62.4 \pm 29.7$  g and average peak rotational head acceleration of  $2609 \pm 1591$  rad/s<sup>2</sup>, than what has been observed with adult populations. These data provide further evidence of age-related differences in concussion tolerance and may be used for the development of youth-specific protective designs.

## INTRODUCTION

As many as 3.8 million sports-related concussions occur annually in the United States, with approximately 50% occurring in youth sports.<sup>2, 20</sup> Recently, research has shown potential links between a history of concussions and long-term neurodegeneration.<sup>27, 28, 40</sup> Ongoing development of the youth brain has been suggested as a factor in the heightened vulnerability of youth towards concussion. Concussive injuries in the youth population may also result in longer recovery times or even disrupt natural maturation of the brain, which makes the clinical diagnosis and management of particular concern.<sup>17, 22</sup>

Numerous population differences, including myelination, head-body ratio, neck musculature development, likely play a role in the susceptibility of youth towards concussion. Youth brains are still developing, with myelination not complete. Unmyelinated brain fibers have been shown to recover more slowly than myelinated fibers, which lends credence to the differences in tolerance to concussion.<sup>3, 23, 31</sup> Youth heads have grown to more than 90% of full-size by the age of five and reach full-size between the ages of 10 and 16.<sup>9, 26</sup> Body development lags behind the head, resulting in an increasing head-body ratio for youths relative to adults. It is also known that children have reduced neck strength and musculature, with a limited capacity for mass recruitment to reduce resultant skull and brain acceleration.<sup>11, 15, 17, 23</sup> The unique aspects of the youth brain and its response to head impact and concussion necessitate the consideration of youth concussion as a distinct entity, and not just a scaled version of adult concussion.

Researchers have largely relied on head impact kinematic data collected from football players to assess risk of concussion.<sup>30, 33-35</sup> This population is exposed to head impacts regularly and experiences concussions at a high rate among team sports.<sup>6, 10</sup> These risk functions utilized linear and/or rotational head acceleration as predictors of concussion, as it is thought that head kinematics are related to the brain's inertial response. Reconstructions of concussive impacts in

the National Football League led to the development of three concussion risk curves, though this dataset did not consider that most head impacts in football are subconcussive.<sup>30</sup> As such, there is an overestimation of injury risk for acceleration inputs. The concussions in this dataset comprising professional football players were associated with peak head kinematics of  $98 \pm 28$  g for linear acceleration and  $6432 \pm 1813$  rad/s<sup>2</sup> for rotational acceleration.<sup>30</sup> Risk functions were also developed from on-field data generated using the Head Impact Telemetry System (HIT System, Simbex, Lebanon, NH) with collegiate football players.<sup>33-35</sup> These predictions considered estimates of concussion underreporting, as well as the relationship between concussive and subconcussive impacts. The concussions in this dataset comprising high school and collegiate football players were associated with peak head kinematics of  $105 \pm 34$  g for linear acceleration and  $3977 \pm 2272$  rad/s<sup>2</sup> for rotational acceleration.<sup>1</sup> The known differences between pediatric and adult populations preclude the use of these previously developed concussion risk functions for a youth population.

Injury data collected from head impact exposure studies provides in-depth biomechanical data on a small cohort of the youth football population. This study adapted an existing injury metric that combines linear and rotational head acceleration to assess the risk of concussion for a youth population. The predictive capacity of this injury metric was compared to previously used biomechanical parameters. We hypothesized that youth athletes would have a lower tolerance for concussion than adult athletes.

## **METHODS**

A large cohort of youth football players between the ages of 9 and 14 were instrumented with helmet-mounted accelerometer arrays. Accelerometer arrays associated with the HIT System were placed inside the youth athletes' helmets, which were worn at each game and practice. These accelerometers are mounted on an elastic base in order to maintain contact with the head



throughout impact, which allows for the measurement of head acceleration and not helmet acceleration.<sup>21</sup> In instances of suspected concussion, evaluation and diagnosis was conducted by a physician. Through conversations with the injured athletes, as well as video review of the playing session in which the injury occurred, we were able to identify specific head impacts as being the concussive impact. This study was approved by the Virginia Tech Institutional Review Board and parental consent was obtained for each athlete, with athletes providing verbal assent independently.

Underreporting of concussions is a known issue, though the youngest age group for which these data exist is the high school level. Athletic trainers reported 5% of athletes sustain a concussion, while 47% of high school players report sustaining a concussion on surveys that do not include the word “concussion”.<sup>12, 19</sup> While the underreporting rates for youth athletes are unknown, it is likely that some subset of players who were instrumented in our study sustained a concussion but failed to report it or seek medical attention. As such, only head impact data from athletes who sustained a clinically-diagnosed concussion were included in this analysis. These athletes sought medical attention, so we have no reason to believe that these subjects would not report other injuries. By not including head impact data from all instrumented athletes, the resulting analysis represented a more conservative assessment of concussion risk for a vulnerable subset of players. To increase our sample size, concussive head impact data were used from previously published work.<sup>5, 7</sup> Head impact data from 15 players who sustained concussions as a result of participation in youth football were included in this analysis. Head impact data consisted of peak linear and rotational head acceleration values. Only data from head impacts with a resultant linear acceleration exceeding 10 g were included. Each athlete’s head impact exposure history for the season in which the injury occurred was included, with head impacts being coded as concussive or non-concussive.

Rather than trying to fit a cumulative distribution function to the bivariate head impact data, the peak linear and rotational head acceleration values for each concussive head impact were combined into an aggregate measure. This aggregate measure was modeled after Generalized Acceleration Model for Brain Injury Threshold (GAMBIT)<sup>24, 25</sup>, which was developed to consider the combined effect of linear and rotational kinematics in the presentation of brain injury. GAMBIT considered more serious brain injuries than concussion, so the critical values in the original equation were modified here to be relevant to our injury severity and youth population. This modified GAMBIT equation is given by

$$YGAMBIT = \sqrt{\left(\frac{PLA}{PLA_{conc\ avg.}}\right)^2 + \left(\frac{PRA}{PRA_{conc\ avg.}}\right)^2} \quad (1)$$

where  $PLA$  and  $PRA$  are peak linear acceleration and peak rotational acceleration respectively, and  $PLA_{conc\ avg.}$  and  $PRA_{conc\ avg.}$  are critical values corresponding to the average peak head kinematics associated with concussive impacts in this study, and  $YGAMBIT$  is a modification of GAMBIT for a youth population.

A modified form of survival analysis was used to develop an injury risk curve that considered both peak linear and rotational head acceleration as predictors of concussion. Recently, it has been shown that concussion tolerance varies between individuals, and that aggregate analysis may not be the most effective way to model this injury.<sup>37</sup> Rather than modeling individual head impacts as inputs to determine risk, Kaplan-Meier curves were developed for each individual athlete's head impact history. Then, individual risk at kinematic levels associated with concussion were averaged across players to fit an overall risk distribution. This approach towards calculating risk resulted in players with different head impact histories contributing equal weighting towards the resulting risk function.

Kaplan-Meier estimators retain a 0 value for non-injurious measurement levels. For injurious levels, the estimator is calculated as having a value equal to the probability of sustaining a concussion for all head impacts sustained at the injurious level or greater.<sup>16</sup> If an athlete sustained 9 head impacts with kinematics exceeding his concussive head impact, the Kaplan-Meier estimator would have a value of 0.1, or 10%, as 1 in 10 head impacts resulted in injury at that level.

As such, 15 individual Kaplan-Meier curves were created in this analysis. Each of these curves was defined only for the range of YGAMBIT values at which the player experienced a head impact, and the point at which individual risk became non-zero was at the concussive YGAMBIT value for that specific player. The dataset consisted of 15 unique YGAMBIT values associated with clinically-diagnosed concussions. To generate a single, composite risk curve, the risk values at each of the concussive YGAMBIT values were averaged across players. Only players with impacts as severe as the concussive impact were considered in the average. For example, only 3 players in this dataset experienced an impact with linear and rotational kinematics as severe as the hardest concussion. Average risk at that severity was only computed considering those 3 players.

A log-normal distribution was then fit to the average risk values computed for magnitudes of YGAMBIT associated with concussion. The log-normal cumulative distribution function takes the form of Equation 2, with  $x$  representing YGAMBIT,  $\mu$  as the distribution mean, and  $\sigma$  as the distribution standard deviation. Though direct calculation of the probability is complicated by the presence of the error function, most software packages have built-in functionality to complete this calculation (MATLAB: *logncdf*; Microsoft Excel: *lognorm.dist*; R: *plnorm*) Log-normal parameters were estimated using a least-squares technique.

$$\ln_{cdf} = \frac{1}{2} + \frac{1}{2} \operatorname{erf} \left[ \frac{\ln(x) - \mu}{\sqrt{2}\sigma} \right] \quad (2)$$

The combined biomechanical parameter, YGAMBIT, was compared against linear and rotational head acceleration for its predictive capability.<sup>34</sup> The predictive capability was assessed by computing the area under the receiver operator characteristic (ROC) curve (AUC). For comparison, random guessing would be associated with an AUC equal to 0.5. Direct comparison of AUC for each parameter was conducted using Hanley's method.<sup>13, 14</sup> A significance level of 0.05 was used for all statistical tests.

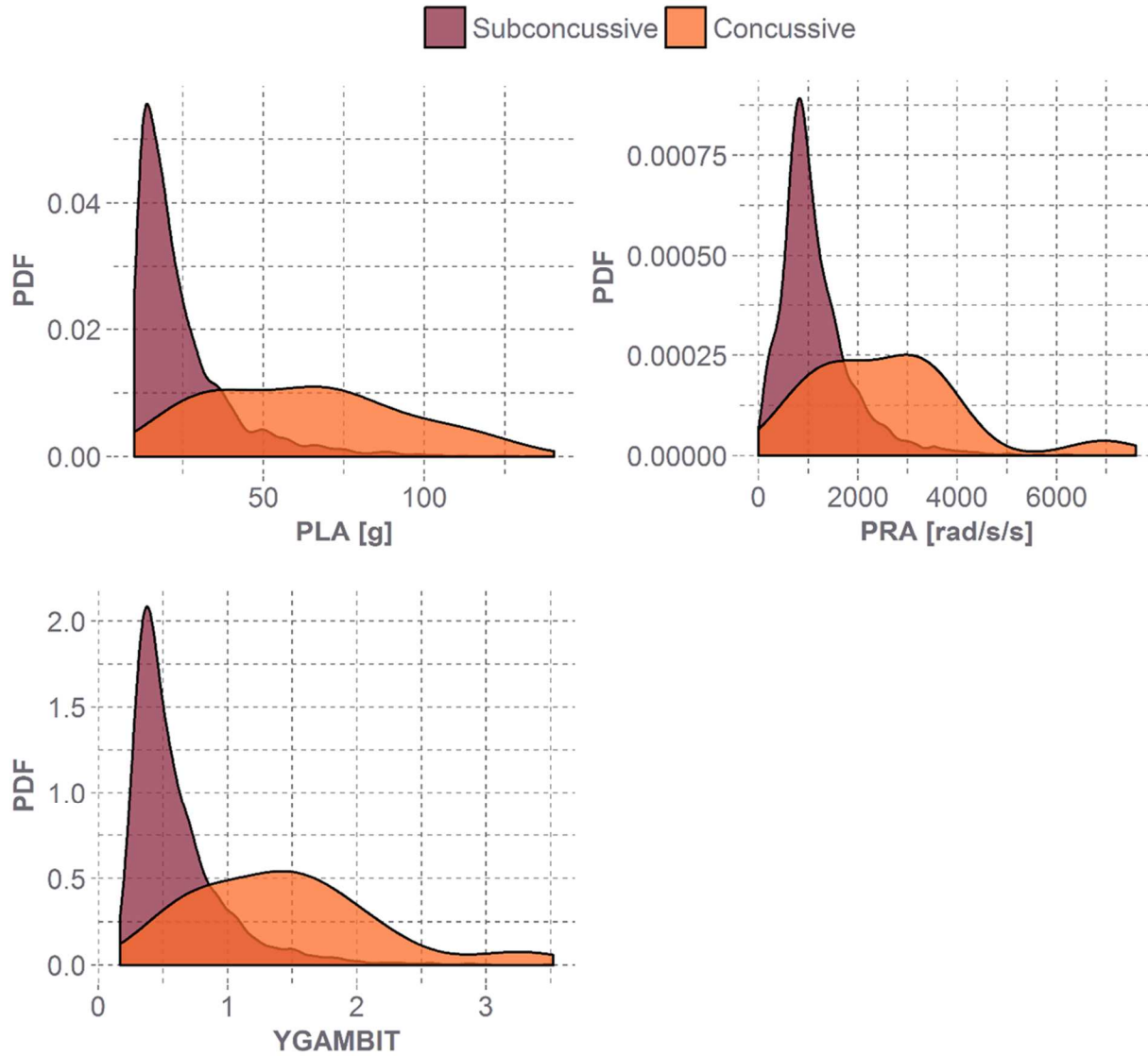
Uncertainty associated with player sampling was modeled by resampling individual Kaplan-Meier curves 10,000 times. These bootstrapped samples were then used to generate 10,000 log-normal curve fits with parameters estimated in a manner identical to the risk curve developed using the measured injury data. At each value of YGAMBIT, the 95% confidence bounds were determined by taking the 250<sup>th</sup> and 9750<sup>th</sup> ranked values.

By only including the subset of athletes who sustained a clinically-diagnosed concussion in the dataset, the potential existed that this group of players would not be representative of all youth athletes instrumented. To relate the head impact exposure profiles of the concussed and non-concussed athletes, we computed the 95<sup>th</sup> percentile YGAMBIT value and risk-weighted exposure for each athlete. The 95<sup>th</sup> percentile YGAMBIT value is a measure of the severity of an athlete's head impact profile, with a higher 95<sup>th</sup> percentile value being associated with more severe, or higher risk, head impacts. Risk-weighted exposure is an aggregate measure that combines impact frequency and magnitude. The risk of concussion was computed for each head impact sustained by an athlete, and then the individual risk values were summed together into one measure. A higher risk-weighted exposure value would be associated with an athlete who

experienced more severe head impacts. The cohort distributions were compared using a Wilcoxon Ranked Sum test, and effect size was determined using Cohen's *d*.

## **RESULTS**

The 15 players in this study experienced a total of 3757 head impacts, with a median and 95<sup>th</sup> percentile linear head acceleration of 19.5 g and 57.1 g and a median and 95<sup>th</sup> percentile rotational head acceleration of 972 rad/s<sup>2</sup> and 2593 rad/s<sup>2</sup> (Figure 7.1). Peak linear and rotational head acceleration values associated with concussion varied among the athletes in this study (Table 7.1). The average concussive head impact was associated with a peak linear head acceleration of  $62.4 \pm 29.7$  g and a peak rotational head acceleration of  $2609 \pm 1591$  rad/s<sup>2</sup>. For most athletes, the concussive head impact was among the top 10% of all head impacts experienced by that athlete (Figure 7.2).



**Figure 7.1: Distribution of subconcussive and concussive head impacts. The distribution of subconcussive head impacts was heavily right-skewed, while the distribution of concussive head impacts was less well-defined. The median peak acceleration values were 19.5 g and 970 rad/s/s for the subconcussive head impacts and 63.8 g and 2599 rad/s/s for the concussive head impacts. The median YGAMBIT value for subconcussive head impacts was 0.49 and 1.36 for concussive head impacts.**

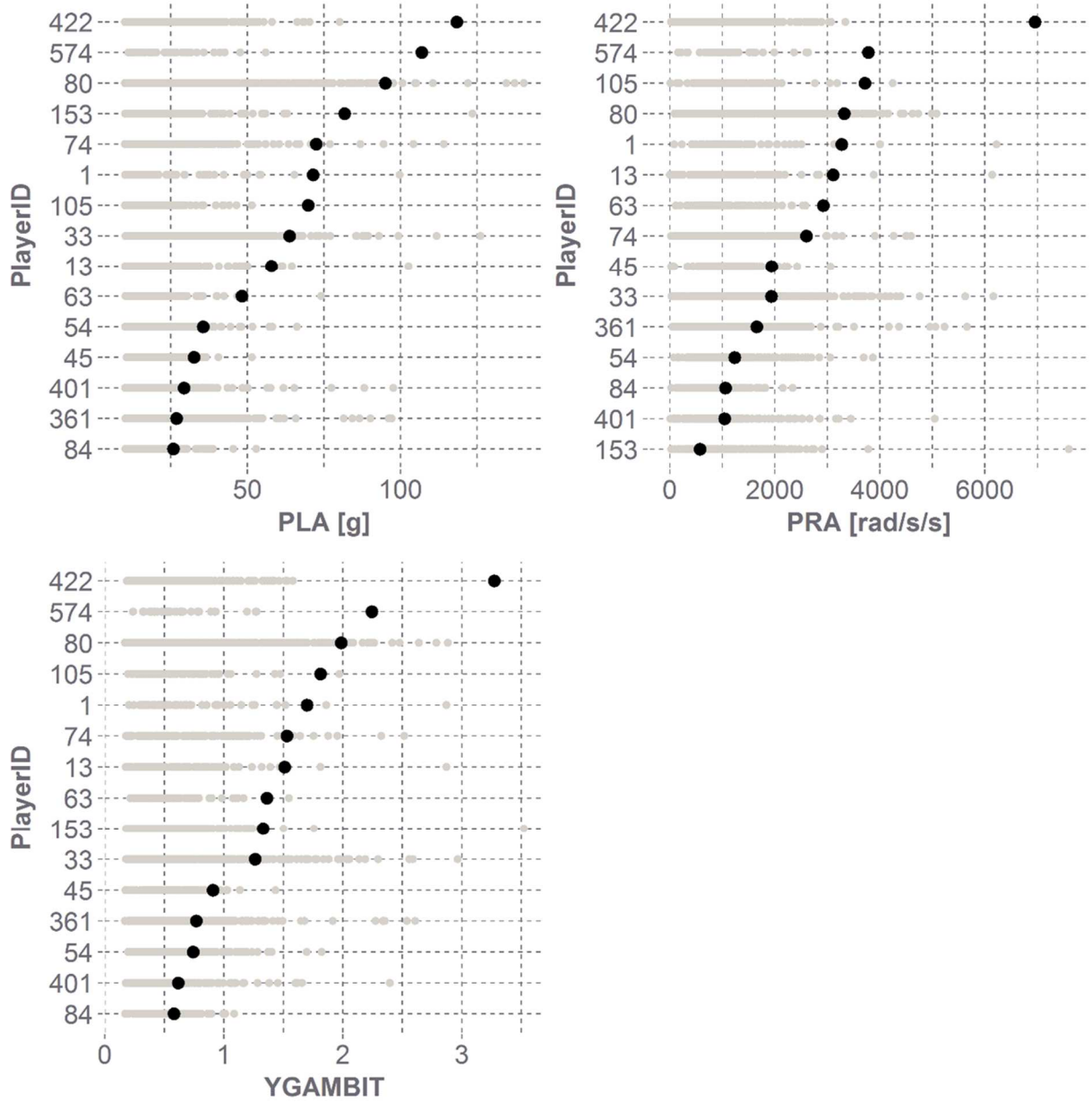


Figure 7.2: Concussion kinematics by player. 60% of concussive head impacts were among a player's top 10 hardest head impacts. All concussive head impacts were within the top quartile of a player's head impacts when considering the combination of linear and rotational kinematics.

**Table 7.1: Biomechanical summary of player concussions. While peak linear acceleration (PLA) and peak rotational acceleration (PRA) values varied for each concussion, most concussions were associated with some of the athletes' hardest head impacts.**

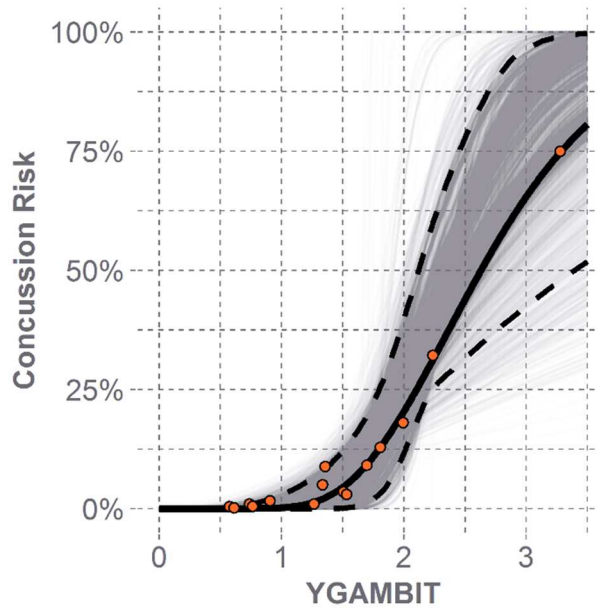
PlayerID	Impacts	PLA [g]	PRA [rad/s <sup>2</sup> ]	YGAMBIT	Rank in PLA	Rank in PRA	PLA Percentile	PRA Percentile
1	61	71.5	3272	1.70	2	3	98.3	96.7
13	159	57.9	3112	1.51	5	3	97.5	98.7
33	579	63.8	1936	1.26	26.5	113	95.6	80.7
45	163	32.6	1938	0.91	8.5	9	95.4	95.1
54	322	35.6	1238	0.74	20	86	94.1	73.6
63	113	48.2	2922	1.36	2	1	99.1	100.0
74	223	72.5	2599	1.53	7	10	97.3	96.0
80	393	95.1	3324	1.99	10	31	97.7	92.4
84	107	25.9	1061	0.58	22	30	80.4	72.9
105	93	69.9	3716	1.81	1	2	100.0	98.9
153	208	81.8	578	1.33	2	172	99.5	17.8
361	616	26.9	1658	0.77	168	92	72.9	85.2
401	228	29.3	1047	0.62	44.5	69	80.9	70.2
422	452	118.4	6955	3.27	1	1	100.0	100.0
574	40	106.9	3781	2.24	1	1	100.0	100.0

The peak linear and rotational head acceleration values for the concussive impacts were used to compute YGAMBIT. By combining these variables into a single value, a distribution could more easily be fit to the dataset. The average CDF relating YGAMBIT to risk of concussion was fit to a log-normal distribution with the following parameters:  $\mu = 0.967$  and  $\sigma = 0.331$  (Figure 7.3). These risk values were then related back to peak linear and rotational head acceleration (Figure 7.4).

$$YGAMBIT = \sqrt{\left(\frac{PLA}{62.4}\right)^2 + \left(\frac{PRA}{2609}\right)^2} \quad (3)$$

$$Concussion Risk = \frac{1}{2} + \frac{1}{2} \operatorname{erf} \left[ \frac{\ln(YGAMBIT) - 0.967}{\sqrt{2} * 0.331} \right] \quad (4)$$

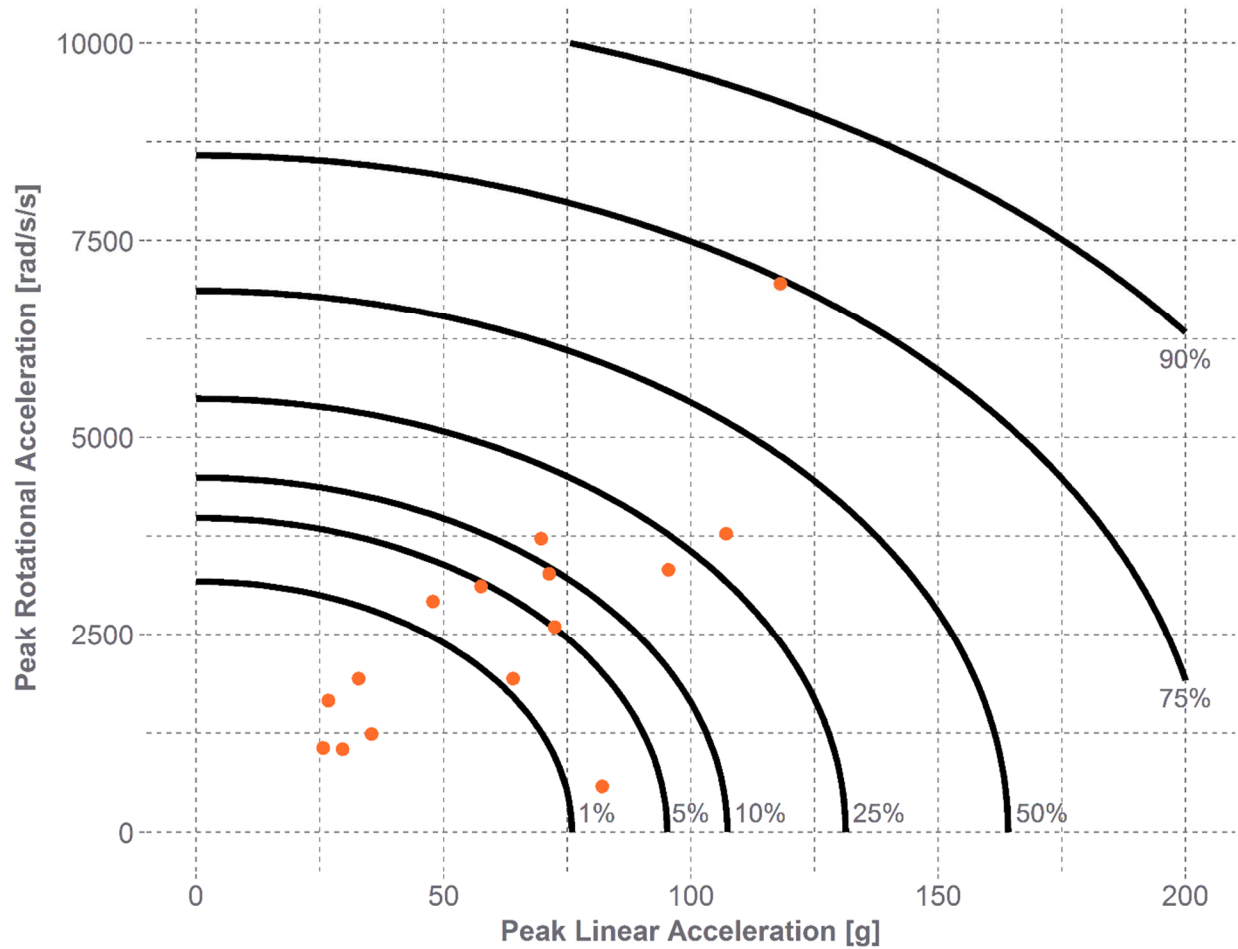




**Figure 7.3: Log-normal distribution fit to concussion data with 95% confidence bounds. Most concussive head impacts were associated with lower values of YGAMBIT. Fewer concussive head impacts were observed for higher values of YGAMBIT (> 2), so the 95% confidence bounds are much wider at these values.**

ROC curves were generated to assess the predictive capacity of YGAMBIT, peak linear acceleration, and peak rotational acceleration (Figure 7.5). All of these metrics were found to be better predictors of concussion than random guessing for this dataset ( $p < 0.05$ ). No significant difference was observed between YGAMBIT and any of the other metrics assessed in this study (Table 7.2).

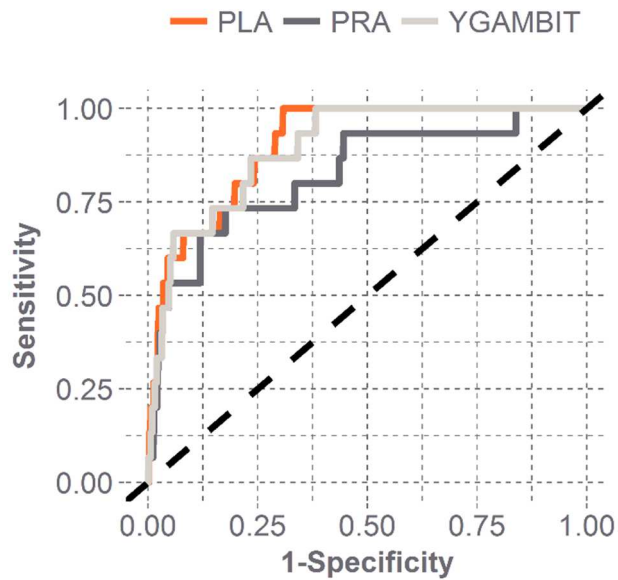
Given the paucity of concussive data points at the higher end of linear and rotational head acceleration values, there is greater uncertainty in the confidence interval for the risk function at higher biomechanical values (Figure 7.3). Nearly all (13 of 15) of the concussive head impacts occurred at YGAMBIT values below 2. Below this value, the confidence bounds were observed to be much narrower.



**Figure 7.4: Relating risk back to linear and rotational head acceleration. Most concussive head impacts were associated with average risk of concussion below 20%.**

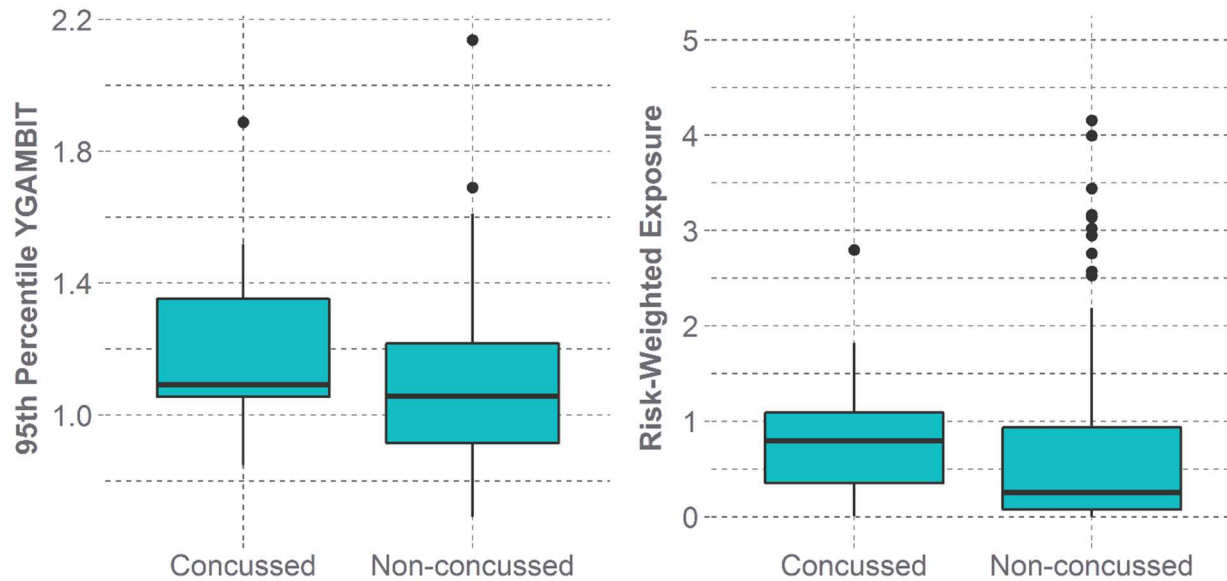
**Table 7.2: AUC for ROC curves. P-value compared to random guessing. P-value compared to YGAMBIT. All measures offer better predictive capacity than random guessing.**

Metric	AUC	95% CI	p-value RG	p-value YGAMBIT
PLA	0.904	0.842-0.951	<0.0001	0.462
PRA	0.824	0.662-0.918	<0.0001	0.267
YGAMBIT	0.894	0.818-0.947	<0.0001	-



**Figure 7.5: ROC curves for various injury metrics and parameters. All parameters were significantly better than random guessing (dashed line), with peak rotational acceleration (PRA) offering the least predictive capability among all metrics.**

The non-concussed sample used to compare to the head impact exposure profiles of the 15 athletes who sustained a clinically-diagnosed concussion in this study consisted of 152 youth athlete-seasons. There was evidence of athletes who sustained a concussion having higher 95<sup>th</sup> percentile YGAMBIT values ( $p = 0.106$  and  $d = 0.516$ ) and risk-weighted exposure ( $p = 0.052$  and  $d = 0.785$ ). The head impact exposure distribution for athletes who did not sustain a concussion was associated with a larger range of risk-weighted exposure values (Figure 7.6).



**Figure 7.6: Comparing concussed athletes to non-concussed athletes. The median values for the 95th percentile YGAMBIT and risk-weighted exposure were higher for the concussed cohort than for the non-concussed cohort.**

## DISCUSSION

This study adapted a previously-developed brain injury metric that relates peak linear and rotational head acceleration for use with a youth population to assess the probability of sustaining a concussion. The consideration of both linear and rotational kinematics stems from the fact that both likely contribute towards the development of concussion and their association with different injury mechanisms.<sup>18, 29, 41</sup> Linear kinematics are associated with an induced intracranial pressure gradient while rotational kinematics are associated with the brain's strain response to motion. This injury metric was developed from youth head impact data collecting using the HIT System. This injury metric builds on previous injury assessment efforts but was uniquely developed towards a youth population.<sup>24, 25, 34</sup> Only head impact data from youth athletes who sustained a concussion were included in the development of the present injury metric. This represented a more conservative approach to injury risk while also providing for higher confidence in the overall dataset. A subset of our instrumented athletes who did not report concussion symptoms or seek

medical attention likely sustained concussions due to participation in football. Despite a recent focus on concussion education, underreporting still remains a factor even in youth football.<sup>32</sup> The injury metric presented here was modeled after GAMBIT, though the critical values for linear and rotational head acceleration were based on the average values associated with concussion and not on more serious brain injuries which have been observed in cadaver testing.<sup>25</sup> The average values for linear and rotational head acceleration associated with concussion represented the 96<sup>th</sup> percentile of head impacts experienced by youth athletes in this dataset.

For most athletes, their concussive impact was among the top 10% most severe head impacts they experienced. When looking at all head impacts in the dataset independent of which player sustained them, some of these concussive head impacts would appear to be less severe. This way of assessing risk would be consistent with how previous concussion risk curves have been developed, in that head impacts are considered in aggregate.<sup>30, 34, 35</sup> Recently, concussion tolerance has been presented as being specific to an individual. It was observed that 90% of concussive impacts for high school and collegiate football players occurred at levels within an individual athlete's top 5 highest magnitude impacts.<sup>37</sup> By developing Kaplan-Meier curves for each athlete in this study who sustained a concussion, we retained the severity of the concussive head impact for the individual while also considering the variance between our subjects. These individual risk curves normalized each player's head impact history so that a composite risk curve could be generated using all of the injured players. This normalization process ensured that each player contributed equally towards the development of the overall risk function regardless of the number of head impacts each player experienced.

For higher values of YGAMBIT, where there are fewer concussive head impacts and fewer overall head impacts, there is much greater uncertainty in concussion risk estimated by the risk function developed here (Figure 7.3). A much narrower confidence interval is observed at lower values of

YGAMBIT, where more head impacts, both concussive and subconcussive, occurred. With a greater number of concussions, particularly at higher levels of YGAMBIT, it would be expected that the confidence in the fit would increase. Nearly all of the head impacts (97%) recorded on the field were associated with YGAMBIT values below 1.5, where there is greater confidence in the estimated risk. This risk function was developed with the target application of evaluating the relative effectiveness of youth football helmets at reducing energy transfer to the head. As helmet testing protocols should ideally be representative of the actual impact scenarios that players would experience on the field, it can be expected that the test conditions would be within this higher confidence area of the risk curve.

Peak head impact kinematics have shown to be good predictors of concussion, with increases in impact force leading to increased injury risk.<sup>1, 8, 33</sup> This was also observed for each of the predictors investigated in this study. Rotational acceleration was observed to have the lowest predictive capacity (AUC = 0.824 [95% CI: 0.662 - 0.918]) of all predictors. Peak linear acceleration (AUC = 0.904 [95% CI: 0.842 - 0.951]) and YGAMBIT (AUC = 0.894 [95% CI: 0.818 - 0.947]) were associated with similar AUCs and can predict the concussions in this dataset equally well. Most head impacts in football are similar, with impacts to the front, side, and back of the helmet having consistent relationships between linear and rotational head acceleration. Impacts to the top of the head, though, often result in very low values of rotational head acceleration despite ranges in linear acceleration values. The lower predictive capacity for rotational acceleration alone is thus expected. Peak linear acceleration, though a strong predictor of concussion, does not consider rotational kinematics. With concussion being an injury related to both linear and rotational kinematics, an approach with combined kinematics seems the most viable.

The average kinematic values associated with youth concussion (PLA:  $62.4 \pm 29.7$  g and PRA:  $2609 \pm 1591$  rad/s<sup>2</sup>) are much lower than what has been reported for high school and college athletes (PLA:  $105 \pm 34$  g and PRA:  $3977 \pm 2272$  rad/s<sup>2</sup>) and professional football players (PLA:  $98 \pm 28$  g and PRA:  $6432 \pm 1813$  rad/s<sup>2</sup>). Some of the concussive head impacts for youth athletes in this study exceeded the severity of the average concussion for the older populations considered. Based on the measured differences in concussive kinematics between the two populations and the known physical differences, data from this study support the hypothesis that the youth population has a lesser biomechanical tolerance for concussion.

The distributions between the two cohorts had considerable overlap for both 95<sup>th</sup> percentile YGAMBIT and risk-weighted exposure, though median values were greater for the concussed cohort. On average, the athletes who experienced clinically-diagnosed concussions experienced more severe head impact exposure profiles than their non-concussed counterparts, though variability between subjects cannot be understated. Athletes who sustained clinically-diagnosed concussions sat out from practices and games during their recovery process. This may partially explain why there is a higher range of risk-weighted exposure values for the athletes who did not sustain a concussion (Figure 7.6).

As helmet manufacturers continue to refine and design technologies to mitigate linear and rotational head acceleration during head impacts, the risk function developed here will be a useful tool in evaluating the effectiveness of these changes. While the evaluation of football helmets will likely be the primary application of this risk function, the potential exists to expand into other industries, such as automotive and other forms of head protection. Provided the impact profiles are similar to football head impacts, the development of this concussion risk function can be used to help the development of safer products aimed at limiting the potential for concussion.<sup>36, 38, 39</sup>

This analysis was limited by several factors. The developed injury metric was based on only data from 15 youth football players who sustained concussions. This represents the largest repository of youth injury data though, and continued data collection may lead to further refinement of this injury metric. By only using data from concussed athletes and not considering underreporting, a more conservative approach of injury risk is achieved. In the interest of evaluating the safety of football helmets and potentially other protective equipment, overestimating injury risk is more favorable than underestimating injury risk. This function considered only the effects of single impacts, as its designed application is towards the evaluation of the effectiveness of football helmets in mitigating energy transfer to the head. It is also likely that other factors, such as impact location, impact duration, exposure to repetitive head impacts, and biological factors contribute towards injury risk. These must be considered when attempting to use our injury metric to evaluate risk of concussion. The HIT System has known random measurement error uncertainties associated with single impact measurements, but these uncertainties are minimized (about 1%) when considering distributions of head impacts.<sup>4</sup> Only peak linear and rotational head acceleration were used to estimate risk of concussion, though these parameters were found to have good injury prediction capabilities.

## **CONCLUSION**

This study presents a concussion risk function for a youth population based on peak linear and rotational head acceleration from a single head impact. Concussive and subconcussive head impact data from youth football players who sustained a concussion were used to determine the injury metric with the highest predictive capability. While ROC analysis revealed that all parameters were good predictors of concussion, rotational acceleration was shown to be the least predictive. YGAMBIT is highly predictive of concussion (AUC = .894) and considers both linear and rotational head kinematics, in addition to being specific to a youth population. Concussions within the youth population were associated with lower biomechanical values than what has



previously been observed for adults. Helmet manufacturers and automotive companies may develop safer products by utilizing this risk function.

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## **CHAPTER 8: QUANTIFYING YOUTH FOOTBALL HELMET PERFORMANCE: ASSESSING LINEAR AND ROTATIONAL HEAD ACCELERATION**

### **ABSTRACT**

Youth football helmet testing standards have served to largely eliminate catastrophic head injury from the sport. These standards, though, do not presently consider concussion and do not offer consumers the capacity to differentiate the impact performance of youth football helmets. This study adapted the previously developed Summation of Tests for the Analysis of Risk (STAR) equation for youth football helmet assessment. This adaptation made use of a youth-specific testing surrogate, on-field data collected from youth football players, and a concussion risk function developed for youth athletes. Each helmet is subjected to 48 laboratory impacts across 12 impact conditions. Peak linear head acceleration and peak rotational head acceleration values from each laboratory impact are aggregated into a single STAR value that combines player exposure and risk of concussion. This single value can provide consumers with valuable information regarding the relative performance of youth football helmets.

## INTRODUCTION

It has been estimated that as many as 1.9 million sports-related concussions occur annually in the United States among youth athletes.<sup>2</sup> Football has received much public scrutiny due to its high incidence of concussion at all levels of play, with some researchers even advocating for the elimination of tackle football among youth athletes.<sup>4, 10, 40</sup> Some research has posited that exposure to repetitive subconcussive head impacts may lead to adverse neurocognitive changes.<sup>18, 19, 22</sup> Limiting exposure to head impacts represents a crucial component of reducing concussion incidence in football, but the development of improved helmet protection can also mitigate injury.<sup>9</sup> Youth athletes represent a vulnerable population as their continued development through puberty and adolescence may be affected differently by a concussion or repetitive head impact exposure than an adult.<sup>15, 20, 41, 48</sup>

Currently, all football helmets are certified by the National Operating Committee on Standards for Athletic Equipment (NOCSAE).<sup>34</sup> This standard sought to limit catastrophic head injuries, such as skull fracture, in football and has succeeded in that regard.<sup>30</sup> A youth-specific testing protocol has recently been proposed, which will assess youth football helmets separately from adult football helmets. Further, NOCSAE certification represents a pass/fail threshold, which does not provide information to consumers about the relative helmet performance. While all helmets that meet NOCSAE's performance criteria are given a seal of approval, some helmets attenuate impacts more effectively than others.<sup>13</sup>

The Summation of Tests for the Analysis of Risk (STAR) methodology was originally developed as an evaluation tool to augment the existing certification standards by providing consumers with data about relative helmet performance. All STAR methodologies are predicated on two fundamental principles.<sup>1, 43, 49</sup> The first is that helmets that reduce linear and rotational

acceleration will also decrease the risk of concussion. Although all current head injury standards are based solely on linear acceleration, all head impacts have a rotational component, and both linear and rotational kinematics should be considered in determining injury risk.<sup>32, 33, 46, 52</sup> The second principle is that the weighting of laboratory tests is based on how frequently a youth football player would experience a similar impact.

The purpose of this study was to develop a youth-specific football helmet testing protocol using data collected directly from youth football players. We accomplished this by adapting the STAR methodology for the evaluation of youth football helmets in managing concussion risk. This evaluation system considered real-world youth head impact exposure data and youth-specific concussion risk. These data consist of head impact exposure profiles and concussive injury risk. The safety certification provided by NOCSAE is meant to be augmented by the results from STAR testing to provide consumers with a complete perspective on a helmet's safety.

## METHODS

### *Youth Summation of Tests for the Analysis of Risk (Youth STAR)*

Like its predecessors, Youth STAR is predicated on relating head impact exposure and head impact risk into a single metric to differentiate the performance of football helmets (Equation 1).<sup>1, 42, 48</sup> Exposure,  $E$ , is a function of impact location,  $L$ , and velocity,  $V$ , while risk of concussion,  $R$ , is a function of peak linear head acceleration,  $a$ , and peak rotational head acceleration,  $\alpha$ . Exposure and risk are specific to each sport and population assessed. Three impact velocities were used to represent some of the more severe head impacts a youth football player may experience on the field. The desired on-field acceleration percentiles to simulate in the impact testing were the 80<sup>th</sup>, 95<sup>th</sup>, and 99<sup>th</sup> percentiles.

$$STAR = \sum_{L=1}^4 \sum_{V=1}^3 E(L, V) * R(a, \alpha) \quad (1)$$

Risk of concussion was assessed using a youth-specific risk function (ref. Youth Risk Function). This risk function relates peak linear head acceleration (PLA) and peak rotational head acceleration (PRA) into a single metric (Equation 2) that is then related to injury risk (Equation 3). Risk was mapped using a log-normal distribution, which does not have a simple solution due to the presence of the error function (erf), but it can be computed in a variety of software programs (MATLAB: *logncdf*; Microsoft Excel: *lognorm.dist*; R: *plnorm*). This risk function was developed from head impact data collected from youth football players who sustained a medically-diagnosed concussion and was designed to be used to assess the risk of single head impacts.

$$YGAMBIT = \sqrt{\left(\frac{PLA}{62.4}\right)^2 + \left(\frac{PRA}{2609}\right)^2} \quad (2)$$

$$Concussion Risk = \frac{1}{2} + \frac{1}{2} \operatorname{erf} \left[ \frac{\ln(YGAMBIT) - 0.967}{\sqrt{2} * 0.331} \right] \quad (3)$$

The STAR system tested each helmet model twice at each of the three impact velocities and four locations (front, front boss, back, and side). Two samples of each helmet model were used, for a total of 48 tests per helmet model. Peak linear and rotational head acceleration values were averaged for each impact velocity-location configuration. From this average test data, an overall STAR value may be computed for a given helmet model.

#### *Mapping of On-Field Exposure*

Bivariate empirical cumulative distribution functions (CDF) of linear and rotational acceleration were determined from on-field data collected from football players wearing youth helmets. For all games and practices, linear and rotational acceleration data were collected using helmet-mounted accelerometer arrays (Head Impact Telemetry System, Simbex, Lebanon, NH). Any impact resulting in a resultant acceleration over a 10 g threshold was included in the dataset. Data were collected from 38 youth players between the ages of 10 and 12 wearing Riddell Speed youth

helmets. STAR impact conditions were used to produce linear and rotational head accelerations for the youth surrogate equipped with a Riddell Speed youth helmet. Tests were conducted at each impact location for velocities ranging from 1.9 to 6.2 m/s. Two trials at each impact location-velocity combination were conducted and then results were averaged across impact locations. The bivariate CDFs were used to relate measured laboratory accelerations for each velocity to the desired on-field acceleration percentiles.

Exposure weightings for each impact energy were initially set based on the on-field percentiles. Head impact data from 56 youth player-seasons were used to determine the average number of head impacts experienced during a single season of youth football. These athletes were between the ages of 10 and 12 and had participated in at least 75% of their team's sessions for the season. Five percent of head impacts exceed the 95<sup>th</sup> percentile and were attributed to the highest impact energy level, 15% of head impacts occur between the 80<sup>th</sup> and 95<sup>th</sup> percentile of head impacts and were attributed to the middle impact energy level, and the remaining 80% were attributed to the lowest impact energy level. These exposure weightings were equally distributed among the four impact locations.

The final exposure weightings were determined through a combination of laboratory optimization and on-field head impact data. To avoid the potential of overweighting or underweighting specific energy-location combinations, the results from laboratory testing were averaged with the corresponding on-field data to develop the final exposure weightings. The STAR methodology is predicated on weighting impacts based on on-field data. These on-field data underpin our protocol and analysis, but we also know that the head-neck assembly responds differently for each energy-location combination, and using equivalent exposure weightings among the different locations has the potential to underweight a location. Impact tests with a vinyl-nitrile disc (40 mm thickness VN600 foam) on a bare headform were conducted to determine how each location's



exposure needed to be scaled in order to contribute 25% towards the overall STAR value. These bare headform tests represent a helmet that offers the same padding properties at all locations. Four trials at each impact location-velocity combination were conducted, with the peak linear and rotational head acceleration values averaged. These average acceleration values were used to generate risk values, and resulting STAR values using the initial exposure weightings. Each location's contribution to the overall STAR value was summed across energy levels and then normalized, so that the location would contribute 25% towards the overall STAR value. A unique normalization factor was generated for each location. These factors were applied to the exposure weightings to generate a modified exposure mapping profile, which varied by both location and velocity. Azimuth angles for each head impact were used to assign the on-field impacts to laboratory impact locations. The front impact location was defined from an azimuth angle from 0 degrees to 33.75 degrees, the front boss location from 33.75 degrees to 83.75 degrees, the side location from 83.75 degrees to 140 degrees, and the back location from 140 degrees to 180 degrees. To map the entire head, the equivalent negative angles corresponded to the same impact locations. The proportion of impacts that occurred at each impact location was then computed. This proportion, in conjunction with the average total number of head impacts and the percentiles associated with the three impact velocities, was used to determine on-field exposure weightings. The exposure values generated from these two methods were then averaged to determine the final exposure weightings.

#### *Development of Youth Surrogate*

The previously-developed adult surrogate that has been used for evaluating hockey helmets was adapted in order to have a representative youth surrogate for assessing youth football helmets.<sup>42</sup> The surrogate consists of a head-neck assembly mounted on a sliding mass which simulates the effective mass of a torso. The masses and dimensions for these surrogate components were based on a 50<sup>th</sup> percentile adult male. For this analysis, the desired anthropometry to model was

that of a 50<sup>th</sup> percentile 10-12 year old male. The head, neck, and sliding mass were modified from the adult surrogate in order to effectively represent a youth male (Table 8.1).

The human head is more than 90% of its full size by the time an individual reaches the age of five and grows gradually to full size between the ages of 10 and 16.<sup>14, 34</sup> This difference in head size is equivalent to the difference between a large and medium sized helmets. A large helmet is intended for head circumferences between 55.9 and 59.7 cm, while a medium helmet is intended for circumferences between 51.8 cm and 55.9 cm. A 50<sup>th</sup> percentile 10 year old male has a head circumference of 53.5 cm and a 50<sup>th</sup> percentile 12 year old male has a head circumference of 54.3 cm, compared to a full grown 50<sup>th</sup> percentile male head circumference of 56.6 cm.<sup>30, 41</sup> NOCSAE has different headforms, which are meant to be used with different sized helmets.<sup>33</sup> The small, red NOCSAE headform with a circumference of 53.4 cm was chosen to emulate a youth player, as a majority of youth football players wear medium sized helmets. This headform has a mass of 4.12 kg.

Youth athletes have weaker necks than their adult counterparts.<sup>16, 21, 24, 27, 35</sup> Scaling techniques have been used to develop Hybrid III necks for a 10 year old dummy and a 5<sup>th</sup> percentile female dummy.<sup>23</sup> The 10 year old neck is composed of molded butyl rubber with a center cable (mass = 0.80 kg), whereas the 5<sup>th</sup> percentile female neck is constructed of butyl rubber, segmented by aluminum discs with a center cable to limit strain on the rubber (mass = 0.91 kg). For the surrogate, the 5<sup>th</sup> percentile female neck was selected as it more closely resembles the 50<sup>th</sup> percentile neck (mass = 1.54 kg) in its design (Table 8.1).<sup>23, 28</sup>

The head-neck assembly was mounted to a sliding mass that simulates the mass of the torso.<sup>38</sup> The sliding torso mass of the adult surrogate is 16 kg and constructed of steel. Youth football players are between the ages of 6 and 14, representing a wide range of body masses. According

to Center for Disease Control (CDC) growth charts, a 50th percentile 6 year old boy has a total body mass of 21.0 kg and a 50th percentile 14 year old boy has a total body mass of 51.0 kg. If you average these two extremes you get a mass of 36.0 kg. Applying the ratio of a 50th percentile male (77.7 kg) to the adult surrogate torso (16.0 kg), an 8 kg sliding mass was selected, which corresponds to a 38.9 kg body mass. This 38.9 kg simulated total body mass is representative of a 50<sup>th</sup> percentile male child between the ages of 11 and 12 (Table 8.1).<sup>26</sup> To decrease the mass of the sliding table, the youth surrogate sliding mass was built out of aluminum rather than steel.

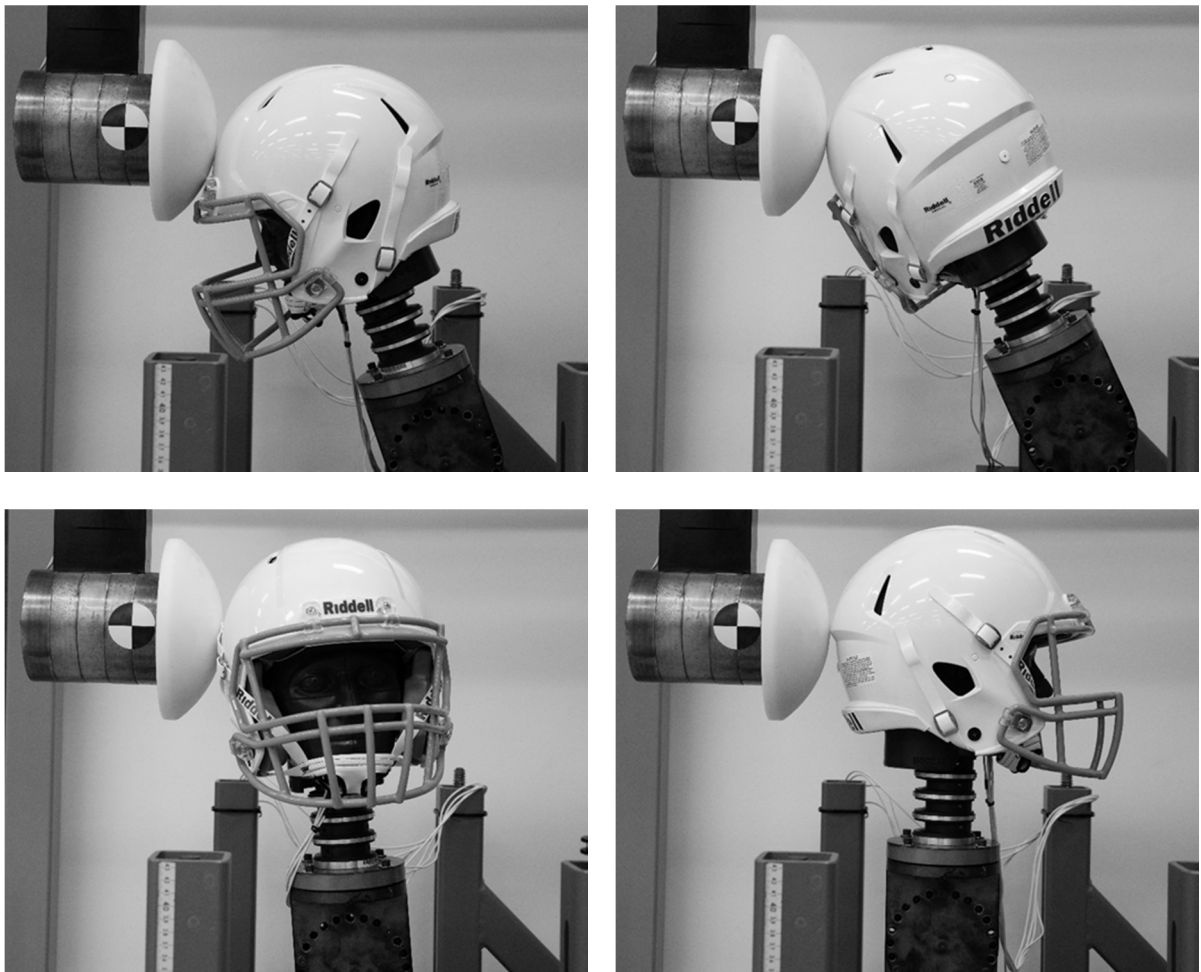
**Table 8.1: Representative masses of the adult and youth surrogates. The adult body mass is based off the 50th percentile Hybrid III dummy and the youth body mass is the equivalent of an 11 or 12 year old male. Each body mass is scaled to give a sliding mass estimate of the effective upper torso mass. The head circumference, head mass, and neck size are based off the specifications for the corresponding components of each surrogate.**

	Adult Surrogate	Youth Surrogate
Simulated Body Mass (kg)	77.7	38.1
Head Circumference (cm)	57.6	53.4
Head Mass (kg)	4.9	4.12
Neck Mass (kg)	1.54	0.91
Sliding Mass (kg)	16	8

### *Impact Testing*

An impact pendulum was used to simulate head impacts to evaluate youth football helmets. The pendulum arm, which is constructed of aluminum, is 190.5 cm in length, and has a 16.3 kg impacting mass. The pendulum arm has a total mass of 36.3 kg and mass moment of inertia of 72 kg-m<sup>2</sup>. A hemispherical nylon impactor face 20.3 cm in diameter, with a 12.7 cm radius of curvature, was used to represent a helmet shell geometry.<sup>49</sup> This impacting face is rigid, as the use of a compliant impactor face has the potential to mask differences in helmet performance.<sup>44</sup> The impact pendulum was used for impact testing as it has excellent repeatability and reproducibility compared to the pneumatic ram impactor that is traditionally used.<sup>39</sup>

The small, red NOCSAE headform was modified to fit a Hybrid III 5<sup>th</sup> percentile female neck. This head-neck assembly was then mounted on the 8 kg sliding mass. The complete youth surrogate was mounted to a linear slide table (Biokinetics, Ottawa, Ontario, Canada), which has 5 degrees of freedom to allow for control of helmet impact location. Each helmet was impacted at four impact locations: front, front boss, rear, and side (Table 8.2). These locations represented both centric and non-centric head impacts and assessed overall helmet performance (Figure 8.1). Helmet position on the headform was set with the NOCSAE nose gauge for a small headform before each test.



**Figure 8.1: Youth STAR impact locations. Impact locations clockwise from top left: front, front boss, side, and back.**

**Table 8.2: Pendulum impact testing locations. Measurements of y and z-position are relative to the basic plane of the NOCSAE headform. Positive rotation in the y-axis is away from the pendulum, and positive rotation in the z-axis is clockwise. The x-position is set such that the helmet contacts the impactor face when the pendulum arm is in a neutral vertical position for each location.**

Location	Y (cm)	Z (cm)	Ry (deg)	Rz (deg)
Front	0	+5.2	-20°	0°
Front Boss	0	+2.2	-25°	+67.5°
Side	-4	+5.7	-5°	-100°
Back	0	+3.7	0°	-180°

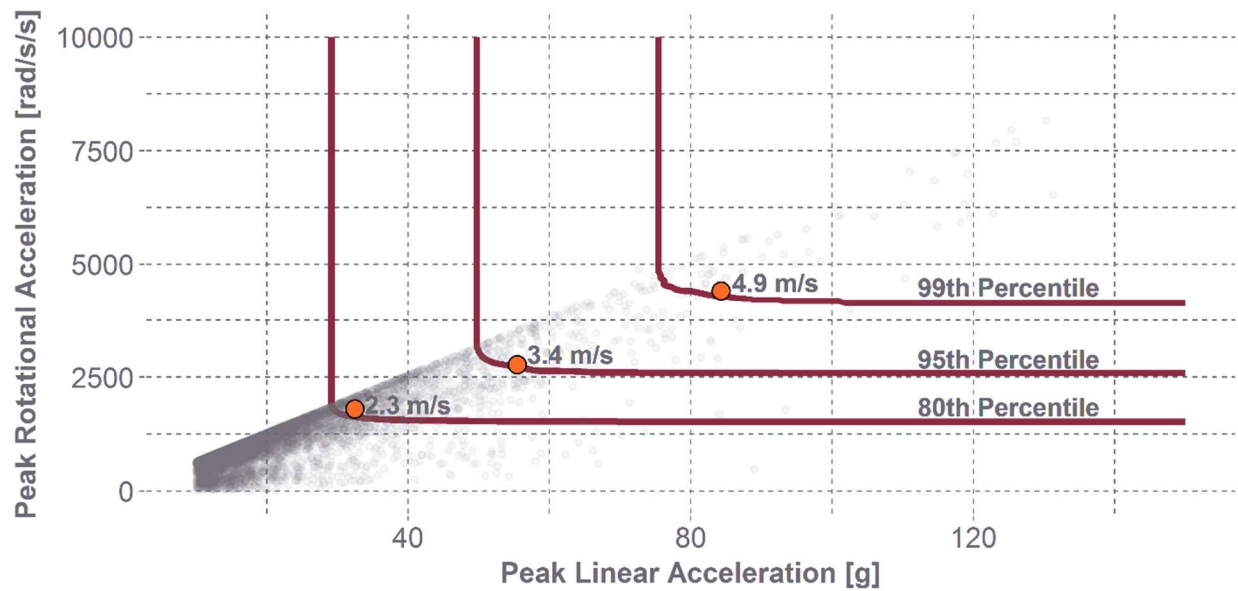
The headform was instrumented with 3 linear accelerometers (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA) and a triaxial angular rate sensor (ARS3 PRO-18K, DTS, Seal Beach, CA) at the headform center of gravity to measure linear acceleration and rotational velocity. Data were sampled at 20,000 Hz and filtered using a 4-pole phaseless Butterworth filter, with cutoff frequencies of 1650 Hz (CFC 1000) for linear data and 256 Hz (CFC 155) for angular rate data. The angular rate data were differentiated to compute measures of angular acceleration. Impact response was characterized by peak resultant linear acceleration and peak resultant rotational acceleration.

## RESULTS

### *Mapping of On-Field Exposure*

A total of 7043 head impacts were included in this dataset to determine impact velocities. Data were collected from 38 youth football players between the ages of 10 and 12. Impacts to the top of the head were not included in this analysis, as the STAR system does not assess impacts to the top of the head. The median head impact in this dataset was associated with a peak linear acceleration of  $17.9 \pm 1.6$  g and a peak rotational acceleration of  $954 \pm 103$  rad/s<sup>2</sup>, with 95<sup>th</sup> percentile accelerations of  $48.8 \pm 9.7$  g and  $2476 \pm 641$  rad/s<sup>2</sup>. Laboratory impact testing with a youth Riddell Speed helmet revealed that impact velocities of 2.3, 3.4, and 4.9 m/s were

associated with the 80<sup>th</sup>, 95<sup>th</sup>, and 99<sup>th</sup> percentile head impacts that a youth player would experience (Figure 8.2).



**Figure 8.2: Mapping of on-field data to laboratory testing.** Gray dots represent on-field head impacts experienced by youth football players. Orange dots represent the average kinematic response across all locations for laboratory pendulum impacts. These impacts represent the 80<sup>th</sup>, 95<sup>th</sup>, and 99<sup>th</sup> percentiles of on-field data and are associated with impact velocities of 2.3, 3.4, and 4.9 m/s.

The average 10-12 year old football player in our dataset experienced  $214 \pm 120$  head impacts during a single season, with a range from 65 to 637 head impacts. Using this average number of head impacts and the percentiles associated with the testing energy levels, initial exposure weightings were set as follows: 171.2 impacts for the 2.3 m/s energy level, 32.0 for the 3.4 m/s energy level, and 10.8 for the 4.9 m/s energy level. These numbers were then divided by 4 and allocated equally to each of the impact locations, so that initial weightings for each location were 42.8, 8.0, and 2.7 head impacts. The peak linear and rotational head acceleration values from the bare headform VN foam tests were used to carry out the STAR methodology to determine the exposure weightings required for each location to contribute 25% towards the STAR value. From the on-field data, it was observed that 51.5% of impacts occurred at the front location, 15.5% at the front boss, 11.2% at the side, and 21.7% at the back location. These proportions were used

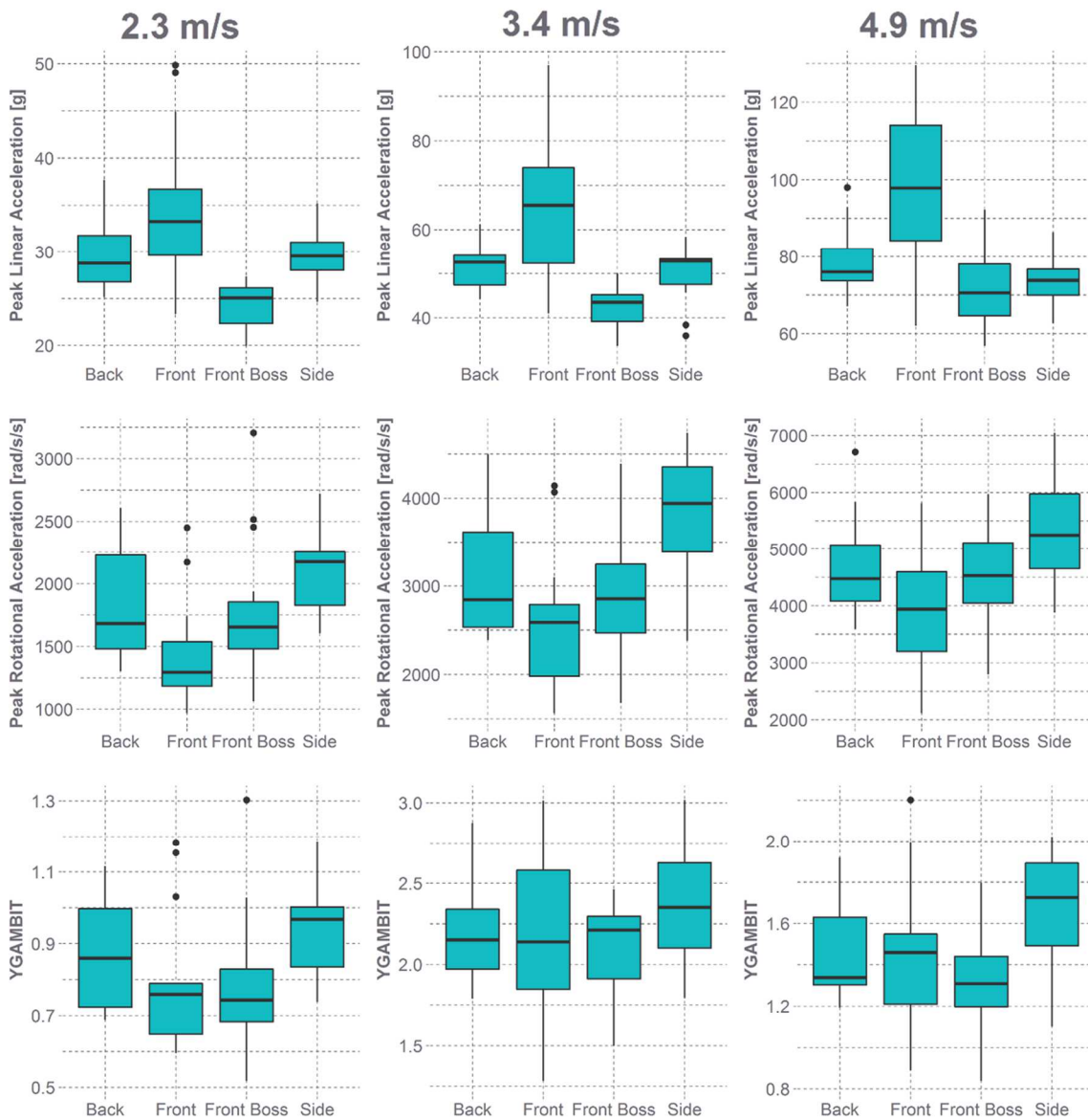
to allocate the 214 total head impacts among the four locations to determine the on-field exposure weightings. These on-field exposure weightings were then averaged with the exposure weightings generated from the laboratory VN testing to determine the final exposure weightings for each location and velocity (Table 3).

**Table 8.3: Exposure weightings used for each location and impact velocity for Youth Football STAR. The average youth football player experiences 214 head impacts per season.**

Location	2.3 m/s	3.4 m/s	4.9 m/s
Front	112.6	21.1	7.1
Front Boss	22.3	4.2	1.4
Side	13.8	2.6	0.9
Back	22.4	4.2	1.4

#### *Impact Testing Results*

Impact performance was observed to vary by impact location and velocity (Figure 8.3). For peak linear acceleration, the front location was associated with higher values than all other locations for both the 3.4 m/s ( $p < 0.0001$ ) and 4.9 m/s ( $p < 0.0001$ ) test velocities. For peak rotational acceleration, the side location was associated with higher values than all other ( $p < 0.005$ ). The front location was also associated with the lowest peak rotational acceleration of all locations ( $p < 0.042$ ). For YGAMBIT, the side location was associated with higher values than either the front or front boss location ( $p < 0.013$ ).



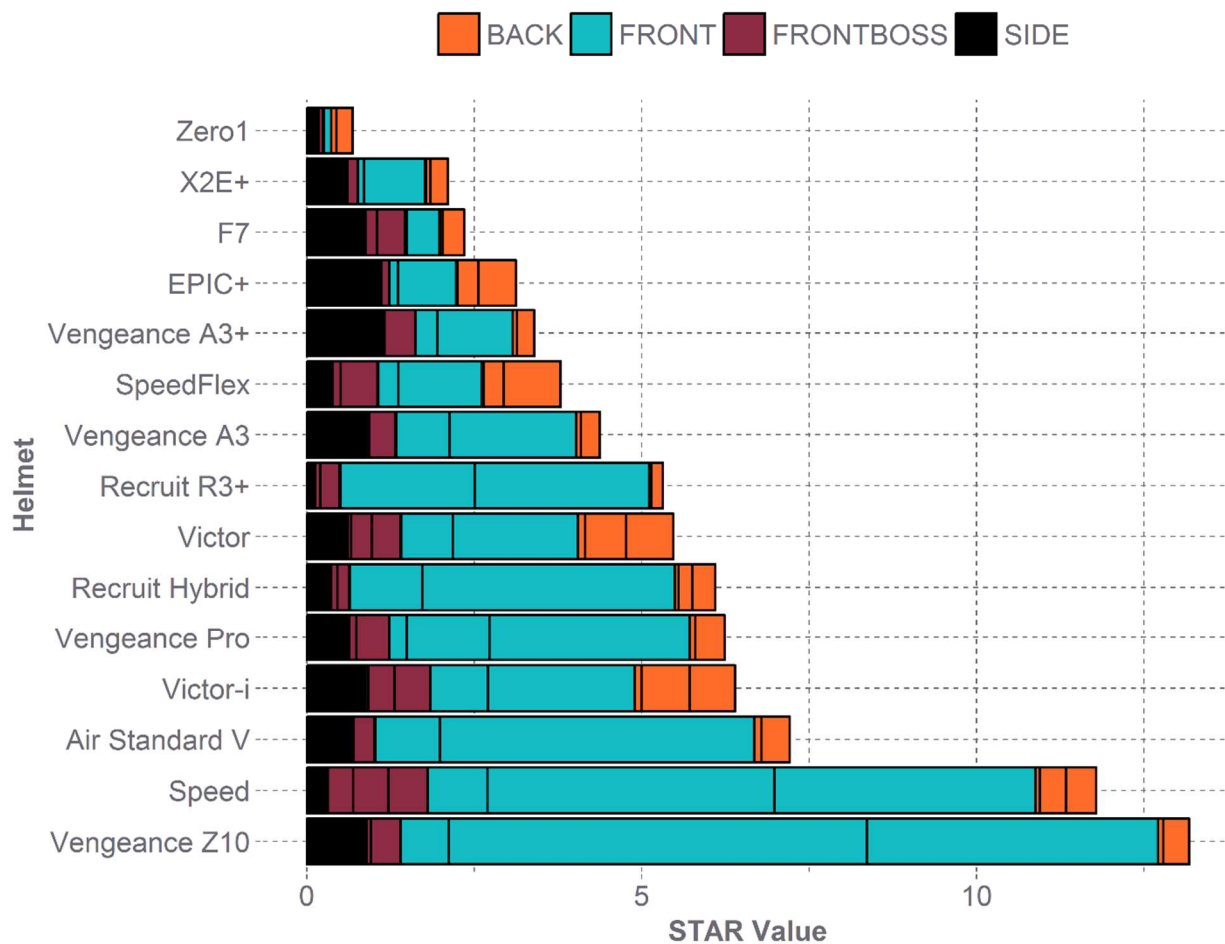
**Figure 8.3: Impact distributions by location and velocity. The front location was associated with higher peak linear acceleration values than the other locations. The side location was associated with higher peak rotational acceleration and YGAMBIT values.**

### STAR Value Assessment

A total of 15 youth football helmet models are currently commercially available and were included in this analysis. Two samples of each helmet model were tested for a total of 30 helmet samples,



each of which was tested 24 times, resulting in a total of 720 impact tests. Helmet performance varied by location and energy level across helmets. The Youth STAR equation was able to differentiate the performance of youth football helmets (Figure 8.4). A helmet with a lower STAR value would be associated with a higher number of stars in our 5-star rating system.



**Figure 8.4: Summary of STAR Values by helmet. Helmets with a STAR value below 5 are given a star rating of 5, those with a STAR value between 5 and 10 are given a star rating of 4, and those between 10 and 15 are given a STAR value of 3.**

## DISCUSSION

It should be noted that helmet technology will never eliminate all head injuries and that participation in any sporting activity carries risk. However, it has been shown that a helmet which can more effectively reduce head acceleration will also reduce concussion risk on average.<sup>37, 38,</sup>

<sup>46, 51, 52</sup> This study sought to develop the Youth STAR evaluation system for assessing the relative performance of youth football helmets. The Youth STAR evaluation system considers the combined contribution of linear and rotational kinematics towards concussion. The impact exposure and injury risk components of Youth STAR were developed from real-world head impact data collected from youth football players. The current NOCSAE standard was developed to reduce the incidence of catastrophic head injuries in football and has led to the elimination of these type of injuries from the game. Youth STAR is intended to only provide additional information and does not replace the current standard. Only those football helmets that have the NOCSAE seal will ever be tested under the Youth STAR methodology.

Youth STAR differs from the previously developed Varsity STAR by considering exposure and risk for a youth player. The average athlete considered under Varsity STAR hits his head 420 times per season, compared to the average 10-12 year old youth football player modeled here (214 head impacts per season). There are known physical differences between adults and children that contribute towards an increased susceptibility of youth to concussion.<sup>24, 28</sup> Thus, the Youth STAR methodology considers a concussion risk function that is youth-specific. For equivalent measures of peak linear acceleration and peak rotational acceleration, the youth concussion risk function results in higher probability of risk. Youth helmets are worn by those players below the age of 14, which essentially encompasses players below the high school level. Due to these population differences, there is a necessity to consider the populations separately and have different testing criteria.

The Youth STAR test surrogate was developed to model a youth player between the ages of 10 and 12. The impact durations ranged from 10 to 15 ms for tests conducted in this study. These values are similar to what has been reported in standard drop tests (8 -12 ms)<sup>45</sup>, laboratory recreations (15 ms)<sup>38</sup>, and on-field measurement (8-14 ms).<sup>7, 17</sup> Furthermore, all the peak linear and rotational

acceleration values of the impacts performed lie within the range of acceleration values observed through laboratory recreations of real world impacts<sup>51</sup> as well as on-field data.<sup>3, 7, 11, 12, 25, 50</sup> Further, the NOCSAE headform provides a more realistic helmet fit than the Hybrid III headform while providing a similar impact response.<sup>6, 8</sup>

The kinematic response to the same energy input varied by impact location (Figure 8.3). Using equal weighting between locations and not considering these response differences or the on-field data distributions would necessarily lead to underweighting or overweighting. The final exposure weightings (Table 8.3) resulted in varied performance among the youth helmet models tested (Figure 8.4). Those helmets that performed best were those that reduced kinematics at each location and velocity. For those helmets that were lower in performance, the front location was generally poor at reducing energy input into the head. By simply improving the front pad, these helmets could improve considerably. Helmet manufacturers can use these results to inform their design choices. They must also consider how those design changes might affect their performance in NOCSAE standards testing.

The Youth STAR methodology and results are based on aggregate data of head impact exposure and injury risk, and individual concussion risk may vary. It has recently been shown that concussion tolerance is individual-specific, with a subset of players who appear to be more vulnerable to concussion.<sup>47</sup> While no helmet can completely protect an athlete, those helmets with higher star ratings are those that were observed to most effectively reduce head impact kinematics. Those helmets that best reduce head acceleration should result in fewer concussions across the playing population.

There are several limitations to note with this study. The on-field head impact measurements are associated with known error for individual acceleration measurements, though the effect of these

errors is minimized when looking at distributions of head impact data.<sup>5</sup> The concussion risk function for youth athletes used in this study has a limited dataset, though it considers the individualized nature of concussion. Head impact exposure weightings were not necessarily representative of actual on-field exposure profiles that a player would experience, but they do ensure that helmets which effectively reduce head acceleration at all locations are rewarded. The current STAR value thresholds may need to be updated in the future as safer helmet models, which offer enhanced impact protection, are developed.

## **CONCLUSIONS**

An adapted version of the previously developed STAR methodology was presented for use with youth football helmets. A youth test surrogate with a smaller head, weaker neck, and scaled sliding mass were used to represent a 10-12 year old boy. Head impact exposure and concussion risk values were derived from on-field head impact data from youth football players. Each impact location was weighted to contribute equally towards the overall STAR value. The STAR values for all currently available youth football helmets were presented, with most helmets offering excellent impact protection. This information will help inform consumers purchasing football helmets by directing them towards helmets that more effectively mitigate head acceleration.

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## **CHAPTER 9: CLOSING REMARKS**

### **RESEARCH SUMMARY**

The research in this dissertation was directed towards improving the safety of youth football players. Utilizing appropriate evaluative tools for youth populations is essential in the management of concussion. Several postural control assessments were investigated to determine their efficacy for use within a youth athlete population, a small subset of whom may sustain a concussion. On-field measurements of head impacts in football served as the basis of evaluating current helmet testing methodologies and assessing risk of concussion for youth football players. These data have direct application to the development of future helmet safety standards and potentially further safety applications as well.

Specifically, the research presented in this dissertation has resulted in:

1. A youth-specific balance testing protocol that may be sensitive to sports-related concussion and useful for clinicians in the return to school and return to play processes.
2. A method for estimating head impact velocities for on-field data collection of youth football players.
3. A youth-specific concussion risk function considering both linear and rotational head acceleration.
4. A football helmet evaluation tool that will inform consumers about the relative performance of helmets.



## PUBLICATIONS

All research presented in this dissertation is expected to be published in scientific journals and/or presented at national or international biomedical engineering conferences. Table 9.1 specifies the publication for each chapter as well as the conferences at which each chapter was presented.

**Table 9.1: Publication plan for research.**

Chapter	Title	Journal/(Conference)
2	Effects of a season of youth football on static postural control	Biomedical Science Instrumentation* (Rocky Mountain Bioengineering Symposium)
3	Assessing static and dynamic postural control in a healthy population	Biomedical Science Instrumentation* (Rocky Mountain Bioengineering Symposium)
4	Reliability of Center of Pressure-Based Measures during Dual-Task Postural Control Testing in a Youth Population	Sports Biomechanics# (Biomedical Engineering Society Annual Meeting)
5	Relationship between impact velocity and resulting head accelerations during head impacts in youth football	International Research Council on Biomechanics of Injury (IRCOBI)*
6	Relating on-field head impacts to laboratory testing procedures	Journal of Sports Engineering and Technology
7	Development of a Concussion Risk Function for a Youth Population Using Head Linear and Rotational Acceleration	Annals of Biomedical Engineering
8	Quantifying Youth Football Helmet Performance: Assessing Linear and Rotational Head Acceleration	Annals of Biomedical Engineering

\*Accepted  
#Submitted