Evaluation of Football Safety Techniques Utilizing Biomechanical Measurements

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Abstract

In recent years, concussions and the effect this injury has on the human brain has been an area of concern for many people involved in sports. And perhaps rightfully so, as between 1.6 and 3.8 million people each year sustain a sports-related concussion in the United States. In the past, concussions have been solely linked to transient symptoms; however, recent research suggests that the injury can also result in long term neurocognitive impairment. Thus, there is much needed research to better understand concussions and assist in the development of safety techniques that will reduce the occurrence of such injury. Participants of youth football are at an extreme disadvantage as very little research has been conducted on this population. The research presented in this dissertation attempts to characterize head impact exposure of a variable subgroup of youth football, middle school football, in order to better understand concussions in youth. In addition to better understanding concussions, it is imperative that correct laboratory techniques are developed to accurately simulate realistic head impacts. This dissertation also presents results from the evaluation of current testing procedures that can be used for laboratory testing of sports equipment and simulation of actual field impacts. Evaluation of these techniques will further validate their ability to act as methods for both safety and research in sports injury. Thus, the overall goal of this dissertation is to provide results that will both further understanding of concussions and evaluate the realistic performance of laboratory techniques, influencing informed decisions to reduce the risk of concussions.

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Chapter 1: Introduction to Concussion, Biomechanical Correlation, Current Research and Research Objectives

Concussions: An injury of the past, present, and future

Traumatic brain injury (TBI) is a contributing factor for one third (30.5%) of all injury-related deaths that occur in the United States [1]. This injury annually affects 1.7 million people, of which 52,000 die, 275,000 are hospitalized, and 1.365 million are treated and released from an emergency department. In addition, every year almost half a million emergency department visits for TBI are made by children aged 0 to 14 years. Of all cases of TBI, approximately 75% are considered mild traumatic brain injuries (mTBI), or concussions [2]. However, researchers estimate that concussions are severely underreported and many of these injuries go unnoticed by physicians due to patients neglecting to seek medical attention, especially in children [3-5]. One area of major contribution to this injury for both adults and children stems from participation in sports. Taking this into account, researchers have estimated that between 1.6 and 3.8 million sports-related concussions occur each year in the United States [6, 7].

In the past, concussions have been widely defined. In the research presented in this dissertation, a concussion is defined as neurological alterations resulting from biomechanical forces that directly or indirectly affect the head. These neurological alterations can come in many forms as symptoms of a concussion (Table 1). Symptoms of a concussion include, but are not limited to: headaches, confusion, nausea, loss of memory, and dizziness [8]. The diagnosis of a concussion can stem from several clinical domains: recognizing somatic symptoms (e.g. headaches), physical signs (e.g. loss of consciousness, amnesia), behavioral changes (e.g. irritability), cognitive impairment (e.g. slowed reaction time), or sleep disturbance (e.g. drowsiness) [9]. The diagnosis of a concussion also frequently involves observing neurobehavioral features of concussion (Table 2) [8]. There are several concussion assessment tools that physicians will typically utilize for assistance in diagnosing the severity of the concussion, including force plates and Sport Concussion Analysis Tool (SCAT) 3 analyses. The primary treatment for this injury at this time is to refrain from physically exerting activities and allow time for the symptoms to subside. After a concussion, an individual must gradually be reintroduced to normal activity, both in a physical and mental capacity [10]. Return to normal activities too soon, especially for athletes, may lead to further injury or what is known as Second Impact Syndrome (SIS), which is often fatal [11, 12].

Early (minutes to hours)
-Headache -Dizziness or vertigo -Lack of awareness of surroundings -Nausea and vomiting
Late (days to weeks)
-Persistent low-grade headache -Lightheadedness -Poor attention and concentration -Memory dysfunction -Easy fatigability -Irritability and low frustration tolerance -Intolerance of bright lights or difficulty focusing vision -Intolerance of loud noises or ringing in the ears -Anxiety and depressed mood -Sleep disturbance

Table 1. Symptoms of a concussion. Although divided into "early" and "late" categories, symptoms may not always be confined to a typical time course in all cases.

-Vacant stare (befuddled facial expression) -Delayed verbal and motor responses (slower to answer questions or follow instructions) -Inability to focus attention (easily distracted and unable to follow through with normal activities) -Disorientation (walking in the wrong direction, unaware of time, date, place) -Slurred or incoherent speech (making disjointed or incomprehensible statements) -Gross observable incoordination (stumbling, inability to walk tandem/straight line) -Emotionality out of proportion to circumstances (appearing distraught, crying for no apparent reason) -Memory deficits (repeatedly asking same question or inability to memorize and return 3/3 words and 3/3 objects for 5 minutes) -Any period of loss of consciousness (paralytic coma, unresponsiveness to stimuli)

Table 2. Frequently observed neurobehavioral features of concussion from Kelly et al [8].

Not only have mild traumatic brain injuries, including concussions, been associated with transient symptoms, but recent research also suggests long term consequences may be an issue as well [13-16]. Individuals in high risk populations of concussion, including athletes, have the potential of developing long-lasting or progressive symptoms due to the repetitive head injuries that may occur [17]. One particular form of progressive neurological deterioration that has been noted to occur from the repetitive brain trauma associated with boxing is chronic traumatic encephalopathy (CTE). The earliest symptoms associated with CTE include confusion, headaches, and deteriorations in attention, concentration, and memory [14]. As the brain tissue continues to deteriorate, additional symptoms, including poor judgment and dementia may arise. Severe cases of CTE exhibit a slowing of muscular movement, a staggered, propulsive gait, masked facies, impeded speech, tremors, vertigo, and deafness [18]. Some researchers have suggested that the severity of CTE is correlated with the length of time engaged in a high risk sport (e.g. boxing, football) and the number of traumatic injuries [14].

Biomechanical Testing for Research of Head Injury

Because concussions are a direct result from biomechanical forces, researchers have used this as an opportunity to correlate kinematics and biological mechanics with pathophysiological changes [20-33]. One of the primary biomechanical measurements used throughout research has been linear acceleration. Researchers will typically relate peak linear acceleration magnitude and/or pulse duration to onset of concussion or physiological alterations. Another measurement of the biomechanical forces leading to concussion that researchers have utilized is pressure. Studies correlating pressure response to head injury typically utilize pressure transducers in a cadaver or animal model to better understand how differences in pressure can damage brain tissue. And finally, the last major biomechanical measurement used to study head injury and concussions is strain. Major studies utilizing strain as a biomechanical correlation of injury have used strain gauges to identify skull fractures, and neutral density targets to assess deformation of brain tissue during impact. Researchers of head injury have exploited these biomechanical measurements, among others, to better characterize concussions and skull fractures in both past and present studies.

Research investigating head injury has been ongoing for the past century, including the early investigations of Kocher (1901), Duret (1920), and Miller (1927). In 1946, Gurdjian and Lissner published results from a study that investigated the mechanisms of head injury utilizing cadavers [34]. In the study, they utilized accelerometers, pressure transducers, strain gauges, and high speed einephotography to interpret skull fractures. Through this study, it was concluded that tissues are injured by compression, tension, and shear during an impact. Also, these modes of injury may occur simultaneously or in succession in the same accident. In addition, this study linked concussions to transient neuronal damage, and loss of consciousness to neuronal brain damage of the brain stem, posterior thalamus, and medial temporal lobes. Spurred on by this study, Lissner et al. first proposed what is now known as the Wayne State Tolerance Curve (WSTC) through dropping embalmed cadaver heads onto unyielding, flat surfaces, striking the subject on the forehead [35]. The WSTC correlates the onset of concussion with effective acceleration and pulse duration (Figure 1). For this study, skull fracture was used as the criterion for determination of concussion and the onset of brain injury. This research was then updated with results from various head injury studies, including cadaver models, animal models, human volunteers, clinical research, and injury mechanisms [36]. In this updated WSTC, skull fracture and concussion were both used as the failure criterion.

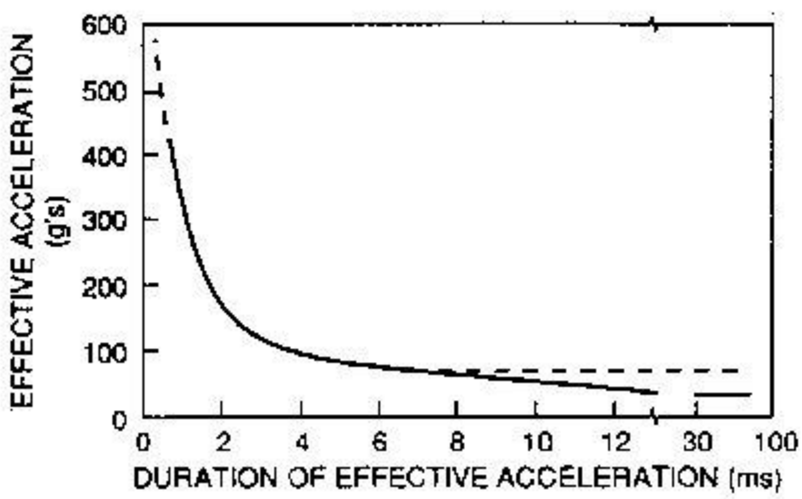

Figure 1. The Wayne State Tolerance Curve.

Utilizing the aforementioned WSTC and the associated biomechanical tests, Gadd developed the Gadd Severity Index (GSI) to be used in assessing impacts (Equation 1). The Severity Index was developed to give researchers the ability to compare different tests for relative severity of impact and for estimating whether an impact exceeds a safe maximum value. The exponent value of 2.5 was based primarily on the animal studies completed at Wayne State University and is representative of internal injury to the head from frontal blows [36]. And, at the time of creation, the maximum pulse intensity without danger to life was estimated to be 1000 for the threshold of serious internal head injury in frontal impact. A major advantage of using the GSI is that it provides a means of comparison for different tests and the hazard represented by the recorded impact pulse.

$$GSI = \int_0^T A^{2.5} dt \qquad \text{Equation 1}$$

Although the Gadd Severity Index fit the WSTC well and established a metric for easily comparing the severity of differing tests, Versace performed a study that suggested there existed incongruities in its derivation [37]. In response to this study, the National Highway Traffic Safety Administration (NHTSA) defined a new parameter to measure head injury, the Head Injury Criterion (HIC) (Equation 2). Opposed to the WSTC, the Head Injury Criterion is based on the resultant linear acceleration, rather than the frontal axis acceleration. The major advantage of HIC over the GSI was that it could handle the seemingly high tolerance levels in cases of long duration, low-level accelerations [38]. It is also important to note that the maximum time duration of HIC is typically limited to an integer between 3 and 36 ms, usually 15 or 36 ms. This time duration can fall anywhere within the acceleration pulse, and HIC would be calculated for the specific time duration that produced the maximum value.

$$HIC = \left\{ \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right]^{2.5} (t_2 - t_1) \right\}_{max} \quad \text{Equation 2}$$

Another monumental study involving the correlation of biomechanics to brain injury was conducted by Hardy et al [39]. In this study, neutral density targets were implanted in the brain and by utilizing a high-speed biplanar x-ray system, brain movement due to impact was recorded. Similar to the early studies conducted at Wayne State University, this study instrumented perfused human cadaver heads, but instead inverted the specimens on a suspension fixture for testing to simulate more realistic conditions. Test specimen were either struck at rest using a padded impactor face or stopped against an angled surface from steady-state motion. Peak resultant linear accelerations for these impacts ranged from 10 g to 150 g, and from 1000 rad/s^2 to 8000 rad/s^2 for peak resultant rotational acceleration. Through this study, the resulting brain displacement response due to skull kinematics was documented in detail. Although this study did not establish the relationship between the brain displacements and potential brain injury, the study vastly improved the correlation of accelerations measured of the skull to mechanical brain responses. The relative displacement of the brain for the low-severity tests conducted was on the order of 5 mm. One of the key findings of this research is that the brain follows a tight looping pattern when being exposed to impact and returns to its original configuration after the impact.

While the previously mentioned studies provided much insight to the accelerations involved with head impact and injury, namely skull fracture, little was still understood about the correlation of head biomechanics to concussion in humans. Unlike skull fracture, the study of mild traumatic brain injuries requires subjects capable of higher critical thinking skills. And, while there have been several animal studies completed in the past, utilizing dogs, primates, and cats, it becomes difficult to properly scale this data to a human due to anatomical differences. However, instrumenting humans with the traditional equipment to accurately record biomechanical measurements would prove to be a very invasive procedure. Moreover, exposing humans to concussive-like impacts, in terms of magnitude, would be unethical. In spite of this, there are several populations that routinely experience concussive-like impacts, namely in the realm of sports. As mentioned previously, researchers estimate that there may be at most 3.8 million sports-related concussions occurring each year in the United States, and one of the major contributors to this estimate is football [6, 7, 40, 41].

In an attempt to better understand the biomechanics involved with concussion in football, the National Football League (NFL) commissioned a study completed by Pellman et al [42-48]. This study reconstructed concussive impacts from NFL game footage in a laboratory setting using helmeted Hybrid III dummies and accelerometers. Video analysis was employed to determine impact velocity, direction, and head kinematics during impact of the players. A drop assembly with a Hybrid III neck and head attached was then used to simulate the striking player, while a Hybrid III torso, neck, and head were positioned underneath the drop assembly and simulated the struck player. Major results from this study found that concussion in NFL football involves an average impact velocity of 9.3 m/s, head velocity change of 7.2 m/s, head acceleration of 98 g, and duration of 15 milliseconds. Not only have impacts been reconstructed with dummies utilizing a drop assembly, but research studies have also used pendulums, and linear impactors [47]. In the impact reconstructions using the pendulum, Pellman et al used a pendulum with a telescoping arm to simulate impacts at 9.5 m/s and 6.7 m/s. To simulate helmet to helmet impacts, an impactor cap with a curved rigid face attached to padding was attached to the end of the pendulum arm. The pendulum arm then swung and impacted a Hybrid III head-neck assembly attached to a mounting apparatus allowing motion along the x- and y- axes. For the linear impactor tests, Pellman et al mounted a Hybrid III head-neck-torso assembly on a translating joint and adjustable table for various impact tests. Impact tests were then conducted for several locations identified from previous helmet impact reconstructions and various impact velocities (9.5 m/s, 11.2 m/s, 6.7 m/s). Results from this study provided fundamental evaluation methods and criterion for evaluating helmets for the protection against concussion.

While the extensive NFL studies provided a better understanding of realistic biomechanics associated with a concussion and also laboratory testing methods for evaluating helmets, it still had its limitations regarding the correlation of measured biomechanics to concussion. Some of these limitations could only be overcome by having actual volunteers subjected to concussive-like impacts. And, due to football players routinely being subjected to these types of impacts in terms of magnitude, Duma et al. saw this as an opportunity to better understand concussion. In 2003, Duma utilized an accelerometer array within a player's helmet to monitor real-time player accelerations as they occurred during games and practices [29]. These accelerometer arrays are a part of commercially available systems called the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH). By utilizing these sensors over the years, studies gained the ability to correlate linear and rotational acceleration, as well as other biomechanics, to concussion. As time progressed, more institutions began using this technology and expanding the dataset [29, 31, 49-65]. Another notable study using the same technology and relating biomechanics of volunteers to concussion came from Guskiewicz et al [50]. This study correlated the head impact acceleration magnitudes, as recorded by the accelerometer arrays, and related these measurements to performance in postural stability testing, neuropsychological testing, and any detectable symptoms after impact. In addition to relating impact biomechanics of football players to concussion, some studies have taken further steps and created injury risk functions, as well as evaluation methods for protective equipment. Most notably, Rowson et al developed the Summation of Tests for the Analysis of Risk (STAR) evaluation system for football helmets [59]. This evaluation system utilizes a twin-wire guided drop assembly with a helmeted National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform to evaluate linear acceleration for different impact severities. Impact severities for these tests were adjusted through the manipulation of drop heights of the helmeted headform. And, the helmets were evaluated and rated based upon their ability to reduce linear acceleration as measured through a triaxial accelerometer at the Center of Gravity (cg) of the headform. And, finally, for the computation of STAR ratings, each impact severity was weighted according to frequency and risk associated with on-field biomechanical measurement of collegiate football players. As a result, this evaluation system has served as a valuable tool to consumers and manufacturers for the education of helmet performance and led to vast improvements in helmet design. Through methods mentioned in the aforementioned studies, valuable data has been provided correlating biomechanical measurements to concussion for adults and for developing laboratory techniques to evaluate safety equipment.

Research Objectives

The research in this dissertation aims to expand upon previous studies correlating impact biomechanics to concussion by characterizing head impact exposure for a unique pediatric population. In addition, the secondary goal of this research is to evaluate aspects of traditional laboratory testing techniques used to study concussions and evaluate safety equipment that may be potential sources of variability.

The following research objectives are investigated in this dissertation:

1. To characterize the biomechanics of a unique subgroup of pediatric football in order to influence educated decisions in improving the safety of the game.
2. To investigate impact exposure differences in positional groups for pediatric football by capturing head impact accelerations from participation in youth football.
3. To evaluate aspects of current laboratory testing methods used for the evaluation of athletic safety equipment that may be sources of variability.

References

[1] Faul, M., Xu, L., Wald, M. M., and Coronado, V. G., 2010, "Traumatic Brain Injury in the United States: Emergency Department Visits, Hospitalizations, and Deaths," pp.
[2] Centers for Disease Control and Prevention (Cdc), N. C. F. I. P. a. C., 2003,
[3] Booher, M. A., Wisniewski, J., Smith, B. W., and Sigurdsson, A., 2003, "Comparison of Reporting Systems to Determine Concussion Incidence in Ncaa Division I Collegiate Football," Clin J Sport Med, 13(2), pp. 93-5.
[4] Williamson, I. J., and Goodman, D., 2006, "Converging Evidence for the under-Reporting of Concussions in Youth Ice Hockey," Br J Sports Med, 40(2), pp. 128-32; discussion 128-32.
[5] Mccrea, M., Hammeke, T., Olsen, G., Leo, P., and Guskiewicz, K., 2004, "Unreported Concussion in High School Football Players: Implications for Prevention," Clin J Sport Med, 14(1), pp. 13-7.
[6] Langlois, J. A., Rutland-Brown, W., and Wald, M. M., 2006, "The Epidemiology and Impact of Traumatic Brain Injury: A Brief Overview," J Head Trauma Rehabil, 21(5), pp. 375-8.
[7] Thurman, D. J., Alverson, C., Dunn, K. A., Guerrero, J., and Sniezek, J. E., 1999, "Traumatic Brain Injury in the United States: A Public Health Perspective," J Head Trauma Rehabil, 14(6), pp. 602-15.
[8] Kelly, J. P., and Rosenberg, J. H., 1997, "Diagnosis and Management of Concussion in Sports," Neurology, 48(3), pp. 575-80.
[9] Mccrory, P., Meeuwisse, W., Johnston, K., Dvorak, J., Aubry, M., Molloy, M., and Cantu, R., 2009, "Consensus Statement on Concussion in Sport 3rd International Conference on Concussion in Sport Held in Zurich, November 2008," Clin J Sport Med, 19(3), pp. 185-200.
[10] Guskiewicz, K. M., Bruce, S. L., Cantu, R. C., Ferrara, M. S., Kelly, J. P., Mccrea, M., Putukian, M., and Valovich Mcleod, T. C., 2004, "National Athletic Trainers' Association Position Statement: Management of Sport-Related Concussion," J Athl Train, 39(3), pp. 280-297.
[11] Cantu, R. C., 1986, "Guidelines for Return to Contact Sports after a Cerebral Concussion," Physician Sportsmed, 14(10), pp. 75-83.
[12] R. C., C., 1998, "Second-Impact Syndrome," Clinics in Sports Medicine, 17(1), pp. 37-44.
[13] Omalu, B. I., Dekosky, S. T., Minster, R. L., Kamboh, M. I., Hamilton, R. L., and Wecht, C. H., 2005, "Chronic Traumatic Encephalopathy in a National Football League Player," Neurosurgery, 57(1), pp. 128-34; discussion 128-34.
[14] Mckee, A. C., Cantu, R. C., Nowinski, C. J., Hedley-Whyte, E. T., Gavett, B. E., Budson, A. E., Santini, V. E., Lee, H. S., Kubilus, C. A., and Stern, R. A., 2009, "Chronic Traumatic Encephalopathy in Athletes: Progressive Tauopathy after Repetitive Head Injury," J Neuropathol Exp Neurol, 68(7), pp. 709-35.
[15] Daneshvar, D. H., Nowinski, C. J., Mckee, A. C., and Cantu, R. C., 2011, "The Epidemiology of Sport-Related Concussion," Clin Sports Med, 30(1), pp. 1-17, vii.
[16] Gavett, B. E., Stern, R. A., and Mckee, A. C., 2011, "Chronic Traumatic Encephalopathy: A Potential Late Effect of Sport-Related Concussive and Subconcussive Head Trauma," Clin Sports Med, 30(1), pp. 179-88, xi.
[17] Roberts, G. W., Allsop, D., and Bruton, C., 1990, "The Occult Aftermath of Boxing," J Neurol Neurosurg Psychiatry, 53(5), pp. 373-8.
[18] Millspaugh, J. A., 1937, "Dementia Pugilistica," U.S. Naval Medical Bulletin, 35(pp. 297-303.

[19] Dekosky, S. T., Ikonomovic, M. D., and Gandy, S., 2010, "Traumatic Brain Injury — Football, Warfare, and Long-Term Effects," New England Journal of Medicine, 363(14), pp. 1293-1296.
[20] Reid, S. E., Tarkington, J. A., Epstein, H. M., and O'dea, T. J., 1971, "Brain Tolerance to Impact in Football," Surg Gynecol Obstet, 133(6), pp. 929-36.
[21] Moon, D. W., Beedle, C. W., and Kovacic, C. R., 1971, "Peak Head Acceleration of Athletes During Competition--Football," Med Sci Sports, 3(1), pp. 44-50.
[22] Reid, S. E., Epstein, H. M., O'dea, T. J., Louis, M. W., and Reid Jr, S. E., 1974, "Head Protection in Football," J Sports Med, 2(2), pp. 86-92.
[23] Morrison, W. E., 1983, "Calibration and Utilization of an Instrumented Football Helmet for the Monitoring of Impact Accelerations," Ph.D. thesis, Penn State University, University Park, PA.
[24] Bishop, P. J., Norman, R. W., and Kozey, J. W., 1984, "An Evaluation of Football Helmets under Impact Conditions," Am J Sports Med, 12(3), pp. 233-6.
[25] Naunheim, R. S., Standeven, J., Richter, C., and Lewis, L. M., 2000, "Comparison of Impact Data in Hockey, Football, and Soccer," J Trauma, 48(5), pp. 938-41.
[26] Mcintosh, A. S., and Mccrory, P., 2000, "Impact Energy Attenuation Performance of Football Headgear," Br J Sports Med, 34(5), pp. 337-41.
[27] Delaney, J. S., Lacroix, V. J., Leclerc, S., and Johnston, K. M., 2002, "Concussions among University Football and Soccer Players," Clin J Sport Med, 12(6), pp. 331-8.
[28] Zhang, L., Yang, K. H., and King, A. I., 2004, "A Proposed Injury Threshold for Mild Traumatic Brain Injury," J Biomech Eng, 126(2), pp. 226-36.
[29] Duma, S. M., Manoogian, S. J., Bussone, W. R., Brolinson, P. G., Goforth, M. W., Donnenwerth, J. J., Greenwald, R. M., Chu, J. J., and Crisco, J. J., 2005, "Analysis of Real-Time Head Accelerations in Collegiate Football Players," Clin J Sport Med, 15(1), pp. 3-8.
[30] Newman, J. A., Beusenberg, M. C., Shewchenko, N., Withnall, C., and Fournier, E., 2005, "Verification of Biomechanical Methods Employed in a Comprehensive Study of Mild Traumatic Brain Injury and the Effectiveness of American Football Helmets," J Biomech, 38(7), pp. 1469-81.
[31] Brolinson, P. G., Manoogian, S., Mcneely, D., Goforth, M., Greenwald, R., and Duma, S., 2006, "Analysis of Linear Head Accelerations from Collegiate Football Impacts," Curr Sports Med Rep, 5(1), pp. 23-8.
[32] Funk, J. R., Duma, S. M., Manoogian, S. J., and Rowson, S., 2007, "Biomechanical Risk Estimates for Mild Traumatic Brain Injury," Annual Proceedings of the Association for the Advancement of Automotive Medicine, 51(pp. 343-61.
[33] Kleiven, S., 2007, "Predictors for Traumatic Brain Injuries Evaluated through Accident Reconstructions," Stapp Car Crash J, 51(pp. 81-114.
[34] Gurdjian, E. S., and Lissner, H. R., 1945, "Deformation of the Skull in Head Injury; a Study with the Stresscoat Technique," Surg Gynecol Obstet, 81(pp. 679-87.
[35] Lissner, H. R., Lebow, M., and Evans, F. G., 1960, "Experimental Studies on the Relation between Acceleration and Intracranial Pressure Changes in Man," Surg Gynecol Obstet, 111(pp. 329-38.
[36] Gurdijan, E. S., Roberts, V. L., and Thomas, L. M., 1966, "Tolerance Curves of Acceleration and Intracranial Pressure and Protective Index in Experimental Head Injury.," Journal of Trauma, 6(pp. 600-604.

[37] Versace, J., 1971, "A Review of the Severity Index," SAE Technical Paper Series, SAE 710881(pp.
[38] Prasad, P., Melvin, J. W., Huelke, D. F., King, A. I., and Nyquist, G. W., 1985, "Head," Technical Report No. National Technical Information Service,
[39] Hardy, W. N., Foster, C. D., Mason, M. J., Yang, K. H., King, A. I., and Tashman, S., 2001, "Investigation of Head Injury Mechanisms Using Neutral Density Technology and High-Speed Biplanar X-Ray," Stapp Car Crash J, 45(pp. 337-68.
[40] Thurman, D. J., Branche, C. M., and Sniezek, J. E., 1998, "The Epidemiology of Sports-Related Traumatic Brain Injuries in the United States: Recent Developments," J Head Trauma Rehabil, 13(2), pp. 1-8.
[41] 2011, "Nonfatal Traumatic Brain Injuries Related to Sports and Recreation Activities among Persons Aged </=19 Years--United States, 2001-2009," MMWR Morb Mortal Wkly Rep, 60(39), pp. 1337-42.
[42] Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R., and Waeckerle, J. F., 2003, "Concussion in Professional Football: Reconstruction of Game Impacts and Injuries," Neurosurgery, 53(4), pp. 799-812; discussion 812-4.
[43] Pellman, E. J., Viano, D. C., Tucker, A. M., and Casson, I. R., 2003, "Concussion in Professional Football: Location and Direction of Helmet Impacts-Part 2," Neurosurgery, 53(6), pp. 1328-40; discussion 1340-1.
[44] Pellman, E. J., Powell, J. W., Viano, D. C., Casson, I. R., Tucker, A. M., Feuer, H., Lovell, M., Waeckerle, J. F., and Robertson, D. W., 2004, "Concussion in Professional Football: Epidemiological Features of Game Injuries and Review of the Literature--Part 3," Neurosurgery, 54(1), pp. 81-94; discussion 94-6.
[45] Pellman, E. J., Viano, D. C., Casson, I. R., Tucker, A. M., Waeckerle, J. F., Powell, J. W., and Feuer, H., 2004, "Concussion in Professional Football: Repeat Injuries--Part 4," Neurosurgery, 55(4), pp. 860-73; discussion 873-6.
[46] Viano, D. C., and Pellman, E. J., 2005, "Concussion in Professional Football: Biomechanics of the Striking Player--Part 8," Neurosurgery, 56(2), pp. 266-80; discussion 266-80.
[47] Pellman, E. J., Viano, D. C., Withnall, C., Shewchenko, N., Bir, C. A., and Halstead, P. D., 2006, "Concussion in Professional Football: Helmet Testing to Assess Impact Performance--Part 11," Neurosurgery, 58(1), pp. 78-96; discussion 78-96.
[48] Viano, D. C., Casson, I. R., and Pellman, E. J., 2007, "Concussion in Professional Football: Biomechanics of the Struck Player--Part 14," Neurosurgery, 61(2), pp. 313-27; discussion 327-8.
[49] Mihalik, J. P., Bell, D. R., Marshall, S. W., and Guskiewicz, K. M., 2007, "Measurement of Head Impacts in Collegiate Football Players: An Investigation of Positional and Event-Type Differences," Neurosurgery, 61(6), pp. 1229-35; discussion 1235.
[50] Guskiewicz, K. M., Mihalik, J. P., Shankar, V., Marshall, S. W., Crowell, D. H., Oliaro, S. M., Ciocca, M. F., and Hooker, D. N., 2007, "Measurement of Head Impacts in Collegiate Football Players: Relationship between Head Impact Biomechanics and Acute Clinical Outcome after Concussion," Neurosurgery, 61(6), pp. 1244-53.
[51] Mccaffrey, M. A., Mihalik, J. P., Crowell, D. H., Shields, E. W., and Guskiewicz, K. M., 2007, "Measurement of Head Impacts in Collegiate Football Players: Clinical Measures of Concussion after High- and Low-Magnitude Impacts," Neurosurgery, 61(6), pp. 1236-43; discussion 1243.

[52] Manoogian, S., Mcneely, D., Duma, S., Brolinson, G., and Greenwald, R., 2006, "Head Acceleration Is Less Than 10 Percent of Helmet Acceleration in Football Impacts," Biomed Sci Instrum, 42(pp. 383-8.
[53] Funk, J. R., Duma, S. M., Manoogian, S. J., and Rowson, S., 2006, "Development of Concussion Risk Curves Based on Head Impact Data from Collegiate Football Players," eds., pp.
[54] Rowson, S., Chu, J. J., Beckwith, J. G., Greenwald, R. M., Stitzel, J. D., Kimpara, H., and Duma, S. M., 2007, "Six Degree of Freedom Head Acceleration Measurements in Football Players," eds., pp.
[55] Rowson, S., and Duma, S., 2008, "Six Degree of Freedom Head Acceleration Measurements in Collegiate Football," Technical Report No. 2008-040, Virginia Tech - Wake Forest, Center for Injury Biomechanics,
[56] Rowson, S., Brolinson, G., Goforth, M., Dietter, D., and Duma, S. M., 2009, "Linear and Angular Head Acceleration Measurements in Collegiate Football," J Biomech Eng, 131(6), pp. 061016.
[57] Crisco, J. J., Fiore, R., Beckwith, J. G., Chu, J. J., Brolinson, P. G., Duma, S., Mcallister, T. W., Duhaime, A. C., and Greenwald, R. M., 2010, "Frequency and Location of Head Impact Exposures in Individual Collegiate Football Players," J Athl Train, 45(6), pp. 549-59.
[58] Kimpara, H., Nakahira, Y., Iwamoto, M., Rowson, S., and Duma, S., 2011, "Head Injury Prediction Methods Based on 6 Degree of Freedom Head Acceleration Measurements During Impact," International Journal of Automotive Engineering, 2(2), pp. 13-19.
[59] Rowson, S., and Duma, S. M., 2011, "Development of the Star Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion," Ann Biomed Eng, 39(8), pp. 2130-40.
[60] Crisco, J. J., Wilcox, B. J., Beckwith, J. G., Chu, J. J., Duhaime, A. C., Rowson, S., Duma, S. M., Maerlender, A. C., Mcallister, T. W., and Greenwald, R. M., 2011, "Head Impact Exposure in Collegiate Football Players," J Biomech, pp.
[61] Rowson, S., Duma, S. M., Beckwith, J. G., Chu, J. J., Greenwald, R. M., Crisco, J. J., Brolinson, P. G., Duhaime, A. C., Mcallister, T. W., and Maerlender, A. C., 2011, "Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion," Ann Biomed Eng, pp.
[62] Broglio, S. P., Sosnoff, J. J., Shin, S., He, X., Alcaraz, C., and Zimmerman, J., 2009, "Head Impacts During High School Football: A Biomechanical Assessment," J Athl Train, 44(4), pp. 342-9.
[63] Broglio, S. P., Schnebel, B., Sosnoff, J. J., Shin, S., Fend, X., He, X., and Zimmerman, J., 2010, "Biomechanical Properties of Concussions in High School Football," Med Sci Sports Exerc, 42(11), pp. 2064-71.
[64] Crisco, J. J., Chu, J. J., and Greenwald, R. M., 2004, "An Algorithm for Estimating Acceleration Magnitude and Impact Location Using Multiple Nonorthogonal Single-Axis Accelerometers," J Biomech Eng, 126(6), pp. 849-54.
[65] Chu, J. J., Beckwith, J. G., Crisco, J. J., and Greenwald, R., 2006, "A Novel Algorithm to Measure Linear and Rotational Head Acceleration Using Single-Axis Accelerometers," Journal of Biomechanics, 39 supplement 1(pp. S534.

Chapter 2: Head Impact Exposure in Youth Football: Middle School Ages 12 to 14 Years

Abstract

The head impact exposure experienced by football players at the college and high school levels has been well documented; however, there are limited data regarding youth football despite its dramatically larger population. The objective of this study was to investigate head impact exposure in middle school football. Impacts were monitored using a commercially available accelerometer array installed inside the helmets of 17 players aged 12 to 14 years. A total of 4678 impacts were measured, with an average (± standard deviation) of 275 ± 190 impacts per player. The average of impact distributions for each player had a median impact of 22 ± 2 g and 954 ± 122 rad/s^2, and a 95th percentile impact of 54 ± 9 g and 2525 ± 450 rad/s^2. Similar to the head impact exposure experienced by high school and collegiate players, these data show that middle school football players experience a greater number of head impacts during games than practices. There were no significant differences between median and 95th percentile head acceleration magnitudes experienced during games and practices; however, a larger number of impacts greater than 80 g occurred during games than during practices. Impacts to the front and back of the helmet were most common. Overall, these data are similar to high school and college data that have been collected using similar methods. These data have applications towards youth football helmet design, the development of strategies designed to limit head impact exposure, and child-specific brain injury criteria.

Introduction

From 2001 to 2009, the estimated number of sports and recreation–related concussion visits to emergency departments increased 62%, and it is estimated that between 1.6 and 3.8 million sports-related concussions occur each year in the United States [1-3]. Emerging research suggests that repetitive sports-related concussions may result in neurodegenerative processes, including chronic traumatic encephalopathy [4-7]. Football represents one of the sports with the highest incidence of concussion given the large participation and contact nature [8]. Understanding the head impact exposure that players experience during participation in football allows informed changes to league rules and equipment design to be made in efforts to reduce the incidence of concussion.

Overall, there are approximately 5 million athletes participating in football, with 2000 NFL players, 100,000 collegiate players, 1.3 million high school players, and 3.5 million youth players [9-11]. To date, concussion research has primarily focused on the high school, college, and professional levels of play. The National Football League (NFL) has conducted extensive concussion research, where select impacts were reconstructed using instrumented crash test dummies to further understand the impact conditions and biomechanics associated with concussions [12-17]. However, this work was mostly limited to reconstructions of open-field impacts in which one player sustained a concussion, which resulted in a biased dataset that was too heavily weighted towards concussion for appropriate risk analyses. While the concussion data are valuable, the NFL exposure data are severely lacking. More recently, researchers have instrumented collegiate and high school football players with helmet-mounted accelerometer arrays to compile a biomechanical dataset consisting of over 1.5 million head impacts [18].

These data have been used by researchers to quantify human tolerance to head impact and characterize the head impact exposure that players experience during play [19-29]. Although 70% of football athletes are competing at the youth level, this is the least studied population of football players. Compared with adults, it has been suggested that younger persons are at an increased risk for concussions with increased severity and prolonged recovery [30, 31]. A recent study which instrumented 7-8 year old football players to characterize head impact exposure found that impact magnitude, location, and number of impacts during the season in 7 to 8 year old players were considerably lower than that of head impacts experienced at the high school and collegiate levels [11]. Furthermore, it was found that higher severity impacts occurred more often in practices than in games. While valuable data were reported from this study, additional research is needed to understand the head impact exposure throughout the entire age continuum of youth football (6 to 14 years old).

The objective of this study was to quantify the head impact exposure of a middle school football team consisting of players between 12 and 14 years old. Head impact exposure is defined as the characterization of the cumulative frequency, location, and head acceleration magnitude of impacts experienced by players across a season. This was accomplished by instrumenting the helmets of players with accelerometer arrays. Head impact exposure at the middle school level is compared to that of other levels of play. These data have applications towards defining injury criteria for pediatric concussion, improving helmet design, and modifying youth football rules and structure to reduce head impact exposure.

Materials and Methods

A middle school football team consisting of children between the ages of 12 and 14 years participated in this study approved by the Virginia Tech Institutional Review Board. Each player gave assent and their parental guardians provided written informed permission. Head impact exposure was investigated in middle school football by instrumenting the helmets of players with a commercially available accelerometer array (Head Impact Telemetry (HIT) System, Simbex, Lebanon, NH). Of the 40 players on the middle school team, the helmets of 17 male players were instrumented. The 17 players had an average body mass of 62 ± 13 kg and had an average age of 13 ± 0.5 years. The average body mass of the 17 players falls at approximately the 90^{th} percentile for 13 year old boys [32]. Instrumented players had an average neck circumference of 14 ± 1.0 inches, and an average head circumference of 23 ± 0.7 inches. Both of these averages are higher than the 95^{th} percentiles for average 12.5-13.5 year old boys seen in a previous study [33]. The instrumented players were chosen due to anticipation of high participation in practices and games. Furthermore, these players wore youth medium or large sized youth Riddell Speed (Elyria, OH) helmets that were compatible with the accelerometer arrays.

The accelerometer arrays consist of 6 single-axis high-g iMEMS accelerometers and are designed to integrate into Riddell Revolution Speed football helmets. While the accelerometer arrays were originally designed for adult Speed football helmets, the device is compatible with youth helmets due to identical sizing and padding geometries between adult and youth Speed helmets. Instrumented helmets were worn by youth football players during each game and practice in which they participated. Each time an instrumented helmet had an individual accelerometer experience a linear acceleration greater than 14.4 g, data acquisition was automatically triggered. A total of 40 ms of data from each accelerometer were recorded, including 8 ms of pre-trigger data. Once data acquisition was complete, data were wirelessly transmitted to a computer on the sideline. Resultant linear acceleration, peak rotational acceleration, and impact location were estimated for each impact [28, 34, 35]. While a brief overview of the accelerometer array is presented here, detailed technical descriptions have previously been reported [36-38]. Measurement error associated with individual impacts has been shown to be 6.3% ± 15.7% for peak linear acceleration and -1.2% ± 31.7% for peak rotational acceleration [38]. Video analysis was used to verify impacts occurring in games and practices.

During this study, concussion was defined as an alteration in mental status resulting from a blow to the head, which may or may not involve loss of consciousness. Concussions were diagnosed by either player's family doctors or recommended physicians from signs, symptoms, computer-based neurocognitive testing, and clinical judgment. There are many symptoms that are associated with concussions and they may include: headache, nausea, vomiting, dizziness, balance problems, fatigue, trouble sleeping, drowsiness, sensitivity to light or noise, blurred vision, difficulty remembering, and/or difficulty concentrating. The time of concussion diagnosis can vary from immediately after the impact associated with injury to later that day or days after the injury when the athlete self-reported symptoms. By combining anecdotal observations about the injury (suspected time of injury, a description of the impact, and other comments) from the player, coaches, and trainers with video of the event and biomechanical data, the ability to associate the injury with a single head impact is gained.

All head impacts were generalized into 1 of 4 impact locations on the helmet: front, side, back, and top [35]. Impact location bins were determined from impact vector azimuth and elevation angle computations, which have been previously described and verified [34]. All impacts exceeding an elevation angle of 65 degrees were defined as impacts to the top of the helmet. All other impacts were determined based on azimuth angles as follows: -135 degrees to 135 degrees were categorized as impacts to the front of the helmet, ±45 degrees to ±135 degrees were categorized as impacts to the side of the helmet, and -45 degrees to 45 degrees were categorized as impacts to the back of the helmet [35]. Overall acceleration distributions were analyzed by impact location and session type (practice or game). Furthermore, empirical cumulative distribution functions (CDF) were computed for linear and rotational acceleration for individual players and the overall dataset. Wilcoxon paired-sample tests were conducted to compare head impact exposure differences between practices and games. Head impact exposure is presented in terms of the impact frequency, impact location, and acceleration magnitude on an individual player basis. These data are then compared to previous studies quantifying head impact exposure in a 7 to 8 year old recreational league, high school, and college football players [11, 19, 20, 27, 28].

Results

A total of 4678 impacts were recorded during practices and games for the 17 instrumented players during the middle school football season. The players participated in an average of 9 games and 29 practices over the course of the season. The distributions for linear and rotational acceleration were right-skewed, being heavily weighted toward low magnitude impacts. Cumulative distribution functions of resultant linear and rotational accelerations for individual players and the overall dataset were determined (Figure 2). Linear accelerations from recorded impacts ranged from 10 g to 149 g. The overall distribution of linear acceleration had an average peak resultant linear acceleration of 27 ± 16 g, a median value of 22 g, a 95^{th} percentile value of 60 g, and a 99^{th} percentile of 86 g. Rotational accelerations ranged from 2 rad/s^2 to 8235 rad/s^2. The overall distribution of rotational acceleration had an average value of 1166 ± 811 rad/s^2, a median value of 987 rad/s^2, a 95^{th} percentile value of 2796 rad/s^2, and a 99^{th} percentile value of 4001 rad/s^2. The impact distribution had 234 impacts that exceeded the 95^{th} percentile accelerations and 47 impacts that exceeded the 99^{th} percentile accelerations.

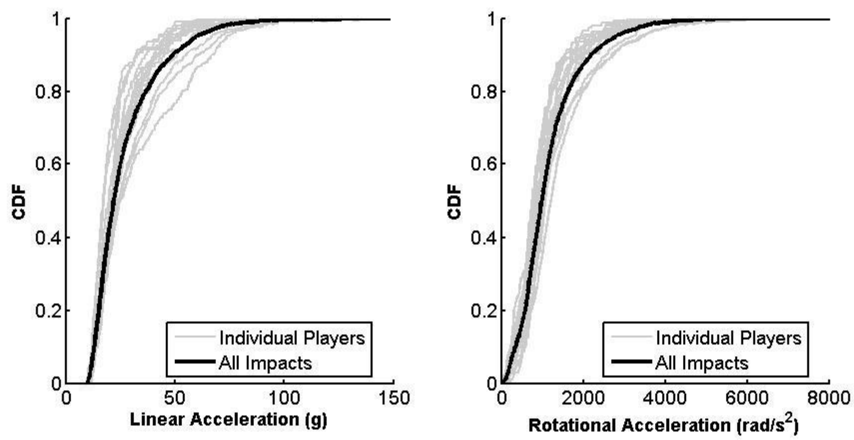

Figure 2. Cumulative distribution functions of linear and rotational acceleration for the overall dataset and individual players. Distributions are heavily weighted towards low magnitude impacts.

The average of impact distributions for each player had average accelerations of 26 ± 4 g and 1106 ± 151 rad/s^2, median accelerations of 22 ± 2 g and 954 ± 122 rad/s^2, and 95^{th} percentile accelerations of 54 ± 9 g and 2525 ± 450 rad/s^2. Head impact exposure varied on an individual player basis, with certain players being more prone to high severity impacts (Tables 3 and 4). A total of 1901 impacts (41%) occurred during games and 2777 impacts (59%) occurred during practices. The players experienced on average 275 ± 190 head impacts, which included 112 ±

112 impacts during games and 163 ± 93 impacts during practices. Furthermore, the players experienced on average 12 ± 11 impacts per game and 6 ± 3 impacts per practice. During games, the average of impact distributions for each player had median accelerations of 21 ± 3 g and 988 ± 140 rad/s^2 and 95th percentile accelerations of 55 ± 14 g and 2562 ± 577 rad/s^2. During practices, the average of impact distributions for each player had median accelerations of 22 ± 3 g and 928 ± 133 rad/s^2 and 95th percentile accelerations of 52 ± 9 g and 2372 ± 393 rad/s^2.

	Practices									Games								
				Linear Acceleration (g)			Rotational Acceleration (rad/s^2)						Linear Acceleration (g)			Rotational Acceleration (rad/s^2)		
Player	**Sessions**	**Hits**	**Hits/ Session**	**50th**	**95th**	**99th**	**50th**	**95th**	**99th**	**Sessions**	**Hits**	**Hits/ Session**	**50th**	**95th**	**99th**	**50th**	**95th**	**99th**
1	23	51	2	20	47	50	914	2129	2877	6	42	7	22	46	55	1183	2626	3293
2	28	111	4	17	36	57	737	1969	3253	8	28	4	19	46	92	791	2165	2624
3	32	174	5	22	51	61	820	2421	3283	10	68	7	21	41	51	871	2191	2847
4	30	151	5	22	57	65	992	2814	3624	9	62	7	21	58	73	1060	2603	3295
5	32	412	13	21	50	58	841	2454	3113	8	166	21	23	54	67	1003	2198	3383
6	34	257	8	23	64	82	874	2829	3962	11	328	30	25	77	117	1153	3705	4864
7	31	123	4	22	49	54	1075	2680	2979	11	106	10	21	53	60	1036	2390	3239
8	23	62	3	22	43	61	1012	2040	2695	5	20	4	27	54	70	1145	2919	4132
9	19	164	9	19	51	61	812	2283	3217	8	228	29	22	55	82	985	2382	4127
10	36	321	9	25	61	75	1246	3027	3976	11	399	36	25	73	106	1176	3618	5207
11	27	128	5	22	48	59	748	1954	3043	6	19	3	18	33	40	805	1694	1783
12	23	137	6	22	56	59	932	2165	3131	9	121	13	23	74	85	1085	2299	4530
13	28	111	4	22	46	62	1074	2205	3040	10	74	7	23	54	81	1073	2500	3397
14	28	95	3	21	42	60	888	1941	2958	7	14	2	16	34	44	848	1594	2121
15	27	101	4	18	45	54	845	2040	2554	9	37	4	17	46	60	787	2542	2729
16	34	215	6	29	72	87	1050	3162	3787	10	143	14	22	78	98	922	3118	3630
17	33	164	5	21	58	66	914	2213	3553	9	46	5	19	61	72	878	3006	3498
AVG	29	163	6	22	52	63	928	2372	3238	9	112	12	21	55	74	988	2562	3453
SD	5	93	3	3	9	10	133	393	416	2	112	11	3	14	22	140	576	916

Table 3. Head impact exposure for individual players during practices and games across one season.

Season Statistics												
				Linear Acceleration (g)			Rotational Acceleration (rad/s^2)			# Impacts ≥ Percentile		
Player	**Sessions**	**Impacts**	**Impacts/ Session**	**50^{th}**	**95^{th}**	**99^{th}**	**50^{th}**	**95^{th}**	**99^{th}**	**50^{th}**	**95^{th}**	**99^{th}**
1	29	93	3	21	47	51	1039	2496	3058	47	5	1
2	36	139	4	17	43	65	750	2153	3186	70	7	1
3	42	242	6	21	47	60	834	2370	3358	121	12	2
4	39	213	5	22	58	66	1017	2813	3628	107	11	2
5	40	578	14	22	50	59	890	2412	3142	289	29	6
6	45	585	13	24	71	100	1022	3474	4771	293	29	6
7	42	229	5	21	50	58	1046	2643	3129	115	11	2
8	28	82	3	23	53	72	1051	2589	3142	41	4	1
9	27	392	15	21	52	80	913	2381	3618	196	20	4
10	47	720	15	25	67	97	1210	3352	5020	360	36	7
11	33	147	4	22	47	58	770	1904	2891	74	7	1
12	32	258	8	22	58	81	1006	2272	3882	129	13	3
13	38	185	5	22	51	74	1074	2434	3348	93	9	2
14	35	109	3	20	42	59	886	1976	2890	55	5	1
15	36	138	4	17	46	57	822	2107	2729	69	7	1
16	44	358	8	25	74	93	985	3166	3747	179	18	4
17	42	210	5	20	59	70	903	2388	3710	105	11	2
AVG	37	275	7	22	54	71	954	2525	3485	138	14	3
SD	6	190	4	2	9	15	122	450	626	95	10	2

Table 4. Head impact exposure data for an entire season of middle school football on a per player basis.

Wilcoxon paired-sample tests were used to compare the number of impacts and acceleration magnitudes experienced by players between games and practices. There was a significant difference between the number of impacts per game and the number of impacts per practice experienced by players ($p = 0.0042$) (Figure 3). There were no significant differences for the average player between practices and games for median linear accelerations ($p = 0.9434$) and rotational accelerations ($p = 0.0759$). Furthermore, there were no significant differences for the average player between practices and games for 95th percentile linear accelerations ($p = 0.0759$) and rotational accelerations ($p = 0.1626$).

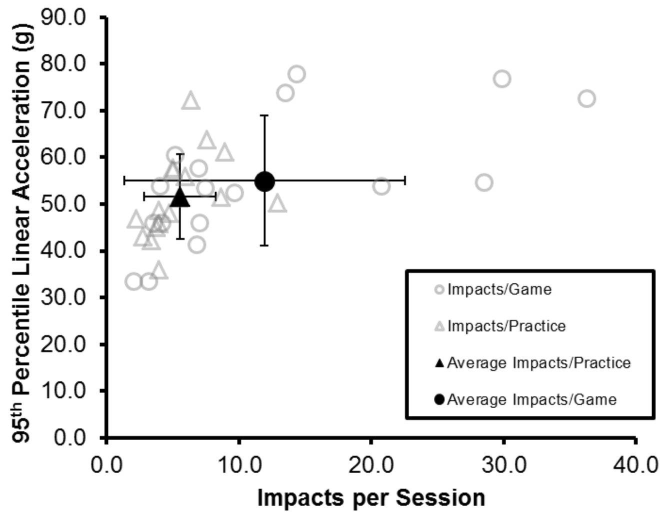

Figure 3. Comparison of number of impacts and acceleration magnitudes experienced by players for games and practices. The number of impacts sustained by players during games and practices was significantly different.

A total of 772 impacts greater than 40 g were collected, 45% of which occurred during games. Moreover, a total of 60 impacts greater than 80 g were collected, 82% of which occurred during games. Impacts resulting in linear accelerations greater than 80 g were more likely to occur during games than practices ($p < 0.001$). The players experienced on average 45 ± 48 head impacts greater than 40 g, 14 ± 21 impacts greater than 60 g, and 4 ± 6 impacts greater than 80 g.

Impacts to the back of the helmet were the most common, accounting for 40% of all impacts (Table 5). The front of the helmet accounted for 33% of all impacts. The sides and top of the helmet were impacted least frequently, accounting for 13% and 14% of all impacts, respectively.

Total Season		Linear Acceleration (g)		Rotational Acceleration (rad/s^2)	
Impact Location	Number of Impacts	Median (50%)	95%	Median (50%)	95%
Front	1543	22	57	1098	2777
Side	619	18	46	836	2041
Rear	1883	23	62	1096	3103
Top	633	24	69	494	1661
Games		**Linear Acceleration (g)**		**Rotational Acceleration (rad/s^2)**	
Impact Location	**Number of Impacts**	**Median (50%)**	**95%**	**Median (50%)**	**95%**
Front	800	24	66	1205	3384
Side	288	18	43	880	2004
Rear	592	23	72	1135	3288
Top	221	24	88	508	2025
Practices		**Linear Acceleration (g)**		**Rotational Acceleration (rad/s^2)**	
Impact Location	**Number of Impacts**	**Median (50%)**	**95%**	**Median (50%)**	**95%**
Front	743	21	47	1010	2188
Side	331	18	48	809	2170
Rear	1291	23	59	1075	3018
Top	412	24	62	488	1459

Table 5. Number of impacts and acceleration magnitudes for each impact location.

One instrumented player sustained a concussion diagnosed by physicians. Player 16 sustained a concussion associated with an impact to the rear of the helmet with acceleration magnitudes (± uncertainty due to random error in single measurements) of 95 ± 15 g and 3148 ± 998 rad/s^2.

Discussion

Head impact exposure in 12 to 14 year old football players was quantified. This age group is unique due to the greatest potential of variability between player size and ability relative to other age groups [39, 40]. Although impact distributions were weighted towards lower magnitude impacts, instrumented players still experienced high magnitude impacts (> 80 g). This level of severity is similar to some of the more severe impacts that college players experience, even though these players have less body mass and play at relatively slower speeds [41]. However, the relative frequency of high magnitude impacts is also important to take into consideration when comparing the middle school impact distribution to that of collegiate players. Collegiate players have been shown to have greater 95th and 99th percentiles for acceleration magnitude than middle school players [42]. Between the fewer impacts experienced and lower 95th and 99th percentile acceleration magnitudes, middle school players are less frequently exposed to higher magnitude impacts than college players. It is possible, but unclear, if younger players are at increased risk of injury considering that injuries still seem common while experiencing high magnitude impact less often. More work investigating concussive impacts in this population is warranted.

Of the 275 head impacts that each player sustained on average, 59% occurred during practices and 41% occurred during games. Players sustained more impacts during practices rather than games because there were more than 3 times as many practices than games, even though players experienced significantly less impacts per practice than per game. Although players sustained significantly more impacts per game than per practice, the impact magnitude distributions were similar between practices and games. This is similar to trends exhibited in high school and college football [20-22, 42], and contrasts trends exhibited by 7 and 8 year old football players where high magnitude impacts occur more frequently during practices [11, 43].

Although this study is the first to document head impact exposure at the middle school football level, research quantifying head impact exposure at other levels has been ongoing for the past decade [11, 20, 37]. When comparing the overall exposure to head impact across level of play, the number of head impacts a player sustains each season increases with increasing level of play (Table 6). As level of play increases, the number of games and practices also increase, which partially explains this trend. Average acceleration magnitudes experienced at the middle school level are very similar to those experienced at the high school and collegiate level, and notably higher than those at the 7 to 8 year old level of play. This may partially be attributed to middle school football being closer to high school football than the pee wee levels of play in terms of musculoskeletal development and game structure [34]. In regards to the game structure, at the 7-8 year old level there are no kickoffs and special team physical contact is minimized. This is in comparison to ordinary special team rules at the middle school, high school, and collegiate levels [44].

Level of Play	**Impacts per Season**	**Linear Acceleration (g)**		**Rotational Acceleration (rad/s^2)**	
		Median (50%)	**95%**	**Median (50%)**	**95%**
Youth (7-8 yrs)[11]	107	15	40	672	2347
Middle School (12-14 yrs)	275	22	60	987	2796
High School (14-18 yrs)[19-20]	520-652	21	56	903	2527
College (19-23 yrs)[22,27-28]	1000	18-20	63	904-981	2787-2975

Table 6. Comparison of overall distributions of head impact exposure by level of play on a per season basis [11, 19, 20, 22, 27, 28]. The number of impacts per season and acceleration magnitudes generally increase with increasing age.

The instrumented middle school players impacted the back of their helmets most frequently, closely followed by the number of impacts to the front of the helmet (Figure 4). This is similar to the trends exhibited by high school and collegiate players, and contrasts that of 7 to 8 year old players [11, 19, 20, 22, 41, 45]. The high number of side impacts with the 7 to 8 year old players was attributed to weak necks, relatively heavy helmets, and players impacting their heads on the ground when being tackled [11].

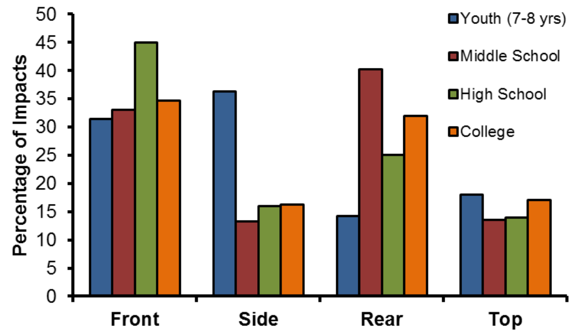

Figure 4. Comparison of impact location distributions by level of play. Middle school players had trends similar to those seen in college [27, 28] and high school football [19, 20].

The one impact associated with a diagnosed concussion resulted in acceleration magnitudes that were among the highest magnitudes for that player. Furthermore, the acceleration magnitudes are similar to acceleration magnitudes associated with concussions that have been previously reported in high school, college, and professional football [13, 19, 27, 28]. For Player 16, the impact associated with concussion resulted in a linear acceleration magnitude that was among the top 1% of magnitudes for that player.

This study has several limitations. First, it should be noted that a total of 17 players were included in this study. This is a relatively small sample size in comparison to previous studies investigating head impact exposure in high school (95 players) and college (>300 players) football [19, 28]. Second, the instrumented players ranged in age from 12 to 14 years old. More work is needed to define the entire age continuum (6 to 14 years old) of head impact exposure in youth football. Third, the HIT System used for data collection is associated with some measurement error for linear and rotational acceleration. From Beckwith et al., measurement error can be quantified in terms of systematic and random errors. Systematic errors were quantified as 6.3% for linear acceleration and -1.2% for rotational acceleration [38]. The measured data were not adjusted to account for any systematic errors so that the results of this study could be directly compared to other head impact exposure studies. Furthermore, these errors are relatively small and exhibit linearity. Random errors were quantified as ±15.7% for linear acceleration and ±31.7% for rotational acceleration [38]. While these random errors can be large for individual measurements, these errors mostly cancel out in distributions of large sample sizes. The uncertainty associated with individual player distributions varied with the number of impacts recorded for each player. However, the contribution of random error to the uncertainty associated with the descriptive statistics for the average player was on the order of 0.2 g for linear acceleration and 19 rad/s^2 for rotational acceleration with the reported sample size. The variance in the dataset is dominated by the differences between individuals, and the random error is negligible. In the context of the presented analyses, the measurement errors are acceptable when considering the challenges associated with measuring head acceleration directly from human volunteers during play.

In conclusion, valuable data describing the head impact exposure in middle school football has been presented for the first time. The helmets of 17 middle school football players were instrumented with accelerometer arrays, resulting in head acceleration data for 4678 impacts. The head impact exposure experienced by 12 to 14 year old football players was similar to that of high school and collegiate players in terms of acceleration magnitudes and impact location distributions; however, high severity impacts occur less frequently in the 12 to 14 year old players on a per season basis. Future collection of concurrent concussive data will enable the development of pediatric injury criteria through risk analyses [46-48]. With an increased understanding of the exposure and tolerance to head impact in pediatric football, better equipment can be designed to prevent head injuries at the youth level. Methods to limit head impact exposure can also be implemented in order to create a safer environment for players.

Acknowledgements

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References

[1] Thurman, D. J., Alverson, C., Dunn, K. A., Guerrero, J., and Sniezek, J. E., 1999, "Traumatic Brain Injury in the United States: A Public Health Perspective," J Head Trauma Rehabil, 14(6), pp. 602-15.
[2] Cdc, 2011, "Nonfatal Traumatic Brain Injuries Related to Sports and Recreation Activities among Persons Aged ≤19 Years," MMWR, 60(39), pp.
[3] Langlois, J. A., Rutland-Brown, W., and Wald, M. M., 2006, "The Epidemiology and Impact of Traumatic Brain Injury: A Brief Overview," J Head Trauma Rehabil, 21(5), pp. 375-8.
[4] Mckee, A. C., Cantu, R. C., Nowinski, C. J., Hedley-Whyte, E. T., Gavett, B. E., Budson, A. E., Santini, V. E., Lee, H. S., Kubilus, C. A., and Stern, R. A., 2009, "Chronic Traumatic Encephalopathy in Athletes: Progressive Tauopathy after Repetitive Head Injury," J Neuropathol Exp Neurol, 68(7), pp. 709-35.
[5] Gavett, B. E., Stern, R. A., and Mckee, A. C., 2011, "Chronic Traumatic Encephalopathy: A Potential Late Effect of Sport-Related Concussive and Subconcussive Head Trauma," Clinics in Sports Medicine, 30(1), pp. 179-188.
[6] Omalu, B. I., Dekosky, S. T., Minster, R. L., Kamboh, M. I., Hamilton, R. L., and Wecht, C. H., 2005, "Chronic Traumatic Encephalopathy in a National Football League Player," Neurosurgery, 57(1), pp. 128-34; discussion 128-34.
[7] De Beaumont, L., Lassonde, M., Leclerc, S., and Théoret, H., 2007, "Long-Term and Cumulative Effects of Sports Concussion on Motor Cortex Inhibition," Neurosurgery, 61(2), pp. 329-337 10.1227/01.NEU.0000280000.03578.B6.
[8] Thurman, D. J., Branche, C. M., and Sniezek, J. E., 1998, "The Epidemiology of Sports-Related Traumatic Brain Injuries in the United States: Recent Developments," J Head Trauma Rehabil, 13(2), pp. 1-8.
[9] Guskiewicz, K. M., Weaver, N. L., Padua, D. A., and Garrett, W. E., Jr., 2000, "Epidemiology of Concussion in Collegiate and High School Football Players," Am J Sports Med, 28(5), pp. 643-50.
[10] Powell, J. W., and Barber-Foss, K. D., 1999, "Traumatic Brain Injury in High School Athletes," JAMA: The Journal of the American Medical Association, 282(10), pp. 958-963.
[11] Daniel, R., Rowson, S., and Duma, S., 2012, "Head Impact Exposure in Youth Football," Annals of Biomedical Engineering, 40(4), pp. 976-981.
[12] Pellman, E. J., Viano, D. C., Tucker, A. M., and Casson, I. R., 2003, "Concussion in Professional Football: Location and Direction of Helmet Impacts-Part 2," Neurosurgery, 53(6), pp. 1328-40; discussion 1340-1.
[13] Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R., and Waeckerle, J. F., 2003, "Concussion in Professional Football: Reconstruction of Game Impacts and Injuries," Neurosurgery, 53(4), pp. 799-812; discussion 812-4.
[14] Viano, D., Withnall, C., and Wonnacott, M., 2012, "Football Helmet Drop Tests on Different Fields Using an Instrumented Hybrid Iii Head," Annals of Biomedical Engineering, 40(1), pp. 97-105.
[15] Viano, D., Withnall, C., and Wonnacott, M., 2012, "Effect of Mouthguards on Head Responses and Mandible Forces in Football Helmet Impacts," Annals of Biomedical Engineering, 40(1), pp. 47-69.
[16] Viano, D., Withnall, C., and Halstead, D., 2012, "Impact Performance of Modern Football Helmets," Annals of Biomedical Engineering, 40(1), pp. 160-174.

[17] Viano, D. C., Hamberger, A., Bolouri, H., and Saljo, A., 2012, "Evaluation of Three Animal Models for Concussion and Serious Brain Injury," Ann Biomed Eng, 40(1), pp. 213-26.
[18] Duma, S. M., and Rowson, S., 2011, "Past, Present, and Future of Head Injury Research," Exerc Sport Sci Rev, 39(1), pp. 2-3.
[19] Broglio, S., Surma, T., and Ashton-Miller, J., 2012, "High School and Collegiate Football Athlete Concussions: A Biomechanical Review," Annals of Biomedical Engineering, 40(1), pp. 37-46.
[20] Broglio, S. P., Sosnoff, J. J., Shin, S., He, X., Alcaraz, C., and Zimmerman, J., 2009, "Head Impacts During High School Football: A Biomechanical Assessment," J Athl Train, 44(4), pp. 342-9.
[21] Crisco, J. J., Fiore, R., Beckwith, J. G., Chu, J. J., Brolinson, P. G., Duma, S., Mcallister, T. W., Duhaime, A. C., and Greenwald, R. M., 2010, "Frequency and Location of Head Impact Exposures in Individual Collegiate Football Players," J Athl Train, 45(6), pp. 549-59.
[22] Crisco, J. J., Wilcox, B. J., Beckwith, J. G., Chu, J. J., Duhaime, A. C., Rowson, S., Duma, S. M., Maerlender, A. C., Mcallister, T. W., and Greenwald, R. M., 2011, "Head Impact Exposure in Collegiate Football Players," J Biomech, 44(15), pp. 2673-8.
[23] Crisco, J. J., Wilcox, B. J., Machan, J. T., Mcallister, T. W., Duhaime, A. C., Duma, S., Rowson, S., Beckwith, J. G., Chu, J. J., and Greenwald, R., 2012, "Magnitude of Head Impact Exposure in Individual Collegiate Football Players," J Appl Biomech, 28(2), pp. 174-83.
[24] Guskiewicz, K. M., and Mihalik, J. P., 2011, "Biomechanics of Sport Concussion: Quest for the Elusive Injury Threshold," Exerc Sport Sci Rev, 39(1), pp. 4-11.
[25] Guskiewicz, K. M., Mihalik, J. P., Shankar, V., Marshall, S. W., Crowell, D. H., Oliaro, S. M., Ciocca, M. F., and Hooker, D. N., 2007, "Measurement of Head Impacts in Collegiate Football Players: Relationship between Head Impact Biomechanics and Acute Clinical Outcome after Concussion," Neurosurgery, 61(6), pp. 1244-53.
[26] Rowson, S., and Duma, S., 2013, "Brain Injury Prediction: Assessing the Combined Probability of Concussion Using Linear and Rotational Head Acceleration," Ann Biomed Eng, 41(5), pp. 873-82.
[27] Rowson, S., and Duma, S. M., 2011, "Development of the Star Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion," Ann Biomed Eng, 39(8), pp. 2130-40.
[28] Rowson, S., Duma, S. M., Beckwith, J. G., Chu, J. J., Greenwald, R. M., Crisco, J. J., Brolinson, P. G., Duhaime, A. C., Mcallister, T. W., and Maerlender, A. C., 2012, "Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion," Ann Biomed Eng, 40(1), pp. 1-13.
[29] Funk, J., Rowson, S., Daniel, R., and Duma, S., 2012, "Validation of Concussion Risk Curves for Collegiate Football Players Derived from Hits Data," Annals of Biomedical Engineering, 40(1), pp. 79-89.
[30] Mccrory, P., Meeuwisse, W., Johnston, K., Dvorak, J., Aubry, M., Molloy, M., and Cantu, R., 2009, "Consensus Statement on Concussion in Sport 3rd International Conference on Concussion in Sport Held in Zurich, November 2008," Clin J Sport Med, 19(3), pp. 185-200.
[31] Field, M., Collins, M. W., Lovell, M. R., and Maroon, J., 2003, "Does Age Play a Role in Recovery from Sports-Related Concussion? A Comparison of High School and Collegiate Athletes," J Pediatr, 142(5), pp. 546-53.

[32] Kuczmarski, R. J., Ogden, C. L., Grummer-Strawn, L. M., Flegal, K. M., Guo, S. S., Wei, R., Mei, Z., Curtin, L. R., Roche, A. F., and Johnson, C. L., 2000, "Cdc Growth Charts: United States," Adv Data, 314), pp. 1-27.
[33] Snyder, Richard G., et al. Anthropometry of US infants and children. No. SAE# 750423. 1975.
[34] Crisco, J. J., Chu, J. J., and Greenwald, R. M., 2004, "An Algorithm for Estimating Acceleration Magnitude and Impact Location Using Multiple Nonorthogonal Single-Axis Accelerometers," J Biomech Eng, 126(6), pp. 849-54.
[35] Greenwald, R. M., Gwin, J. T., Chu, J. J., and Crisco, J. J., 2008, "Head Impact Severity Measures for Evaluating Mild Traumatic Brain Injury Risk Exposure," Neurosurgery, 62(4), pp. 789-98; discussion 798.
[36] Brolinson, P. G., Manoogian, S., Mcneely, D., Goforth, M., Greenwald, R., and Duma, S., 2006, "Analysis of Linear Head Accelerations from Collegiate Football Impacts," Curr Sports Med Rep, 5(1), pp. 23-8.
[37] Duma, S. M., Manoogian, S. J., Bussone, W. R., Brolinson, P. G., Goforth, M. W., Donnenwerth, J. J., Greenwald, R. M., Chu, J. J., and Crisco, J. J., 2005, "Analysis of Real-Time Head Accelerations in Collegiate Football Players," Clin J Sport Med, 15(1), pp. 3-8.
[38] Beckwith, J., Greenwald, R., and Chu, J., 2012, "Measuring Head Kinematics in Football: Correlation between the Head Impact Telemetry System and Hybrid Iii Headform," Annals of Biomedical Engineering, 40(1), pp. 237-248.
[39] Siervogel, R. M., Demerath, E. W., Schubert, C., Remsberg, K. E., Chumlea, W. C., Sun, S., Czerwinski, S. A., and Towne, B., 2003, "Puberty and Body Composition," Hormone Research in Paediatrics, 60(suppl 1)(Suppl. 1), pp. 36-45.
[40] Iuliano-Burns, S., Mirwald, R. L., and Bailey, D. A., 2001, "Timing and Magnitude of Peak Height Velocity and Peak Tissue Velocities for Early, Average, and Late Maturing Boys and Girls," American Journal of Human Biology, 13(1), pp. 1-8.
[41] Rowson, S., Brolinson, G., Goforth, M., Dietter, D., and Duma, S. M., 2009, "Linear and Angular Head Acceleration Measurements in Collegiate Football," J Biomech Eng, 131(6), pp. 061016.
[42] Schnebel, B., Gwin, J. T., Anderson, S., and Gatlin, R., 2007, "In Vivo Study of Head Impacts in Football: A Comparison of National Collegiate Athletic Association Division I Versus High School Impacts," Neurosurgery, 60(3), pp. 490-5; discussion 495-6.
[43] B. R. Cobb, J. E. Urban, E. M. Davenport, S. Rowson, S. M. Duma, J. A. Maldjian, C. T. Whitlow, A. K. Powers, and J. D. Stitzel, "Head impact exposure in youth football: elementary school ages 9-12 years and the effect of practice structure," Ann Biomed Eng, vol. 41, pp. 2463-73, Dec 2013.
[44] League, V. H. S., 2011, "Virginia High School League Handbook and Policy Manual 2011-2012,"
[45] Broglio, S. P., Schnebel, B., Sosnoff, J. J., Shin, S., Fend, X., He, X., and Zimmerman, J., 2010, "Biomechanical Properties of Concussions in High School Football," Med Sci Sports Exerc, 42(11), pp. 2064-71.
[46] Cormier, J., Manoogian, S., Bisplinghoff, J., Rowson, S., Santago, A., Mcnally, C., Duma, S., and Bolte, J. T., 2011, "The Tolerance of the Frontal Bone to Blunt Impact," J Biomech Eng, 133(2), pp. 021004.

[47] Cormier, J., Manoogian, S., Bisplinghoff, J., Rowson, S., Santago, A., Mcnally, C., Duma, S., and Bolte, J., 2011, "The Tolerance of the Maxilla to Blunt Impact," J Biomech Eng, 133(6), pp. 064501.
[48] Zhang, L., Yang, K. H., and King, A. I., 2004, "A Proposed Injury Threshold for Mild Traumatic Brain Injury," J Biomech Eng, 126(2), pp. 226-36.

Chapter 3: An Investigation of Positional Differences in Middle School Football

Abstract

There are approximately 5 million football players in the United States, with 70% of these players being youth football players, between the ages of 6 and 14 years old. The goal of this study was to investigate differences in head impact exposure between skill and line position players in a unique subgroup of youth football, middle school football. To investigate the head impact exposure differences between skill and line position players, 17 middle school football players ranging from 12 to 14 years old were selected for participation. These players were equipped with helmets instrumented with accelerometer arrays capable of measuring resultant linear and rotational acceleration. A total of 2925 impacts were collected from the 8 instrumented skill players and 1753 impacts were collected from the 9 instrumented line players. Skilled players had median linear and rotational accelerations of 22.9 ± 1.5 g and 972 ± 138 rad/s^2, and 95th percentiles of 58.2 ± 11.0 g and 2760 ± 543 rad/s^2. Line players had median linear and rotational accelerations of 20.3 ± 1.9 g and 938 ± 111 rad/s^2, and 95th percentile accelerations of 50 ± 6.0 g and 2317 ± 209 rad/s^2. Skill players sustained approximately 74% of the 772 collected impacts exceeding 40 g and 82% of the 60 impacts exceeding 80 g. Skill players also experienced significantly higher median linear acceleration magnitudes than line players ($P = 0.0061$). Similar studies at the collegiate level also found that certain skill players were more likely to sustain high magnitude impacts than linemen. By understanding the impact distribution differences between skill and line position players, methods to limit head impact exposure can be implemented to create a safer environment for players.

Introduction

Concussion has been an increasingly important issue in sports and an area of great interest among sports medicine professionals. It is estimated that between 1.6 and 3.8 million sports concussions occur each year in the United States [1, 3]. Recent research suggests that concussions not only lead to transient symptoms, but also more long term consequences as well, including motor system dysfunction and other neurodegenerative processes [4-7]. Research also suggests that younger persons are at an increased risk for concussions with increased severity and prolonged recovery [30]. One of the leading activities that individuals under the age of 19 experience concussions while participating in is football [8]. Due to the high prevalence of this injury in football, there becomes a need to better understand potential contributors. This method of understanding comes in the form of characterizing the head impact exposure for these players. Head impact exposure is defined as the characterization of the cumulative frequency, location, and head acceleration magnitude of impacts experienced by players. The ultimate goal of understanding head impact exposure is to develop further techniques to limit dangerous environments.

There are approximately 5 million football players in the United States, with the majority of these players falling in the “youth” category, 6 to 14 years old [9-11]. However, concussion research has predominantly focused on the high school, college, and professional levels of play in the past. Extensive research began at the National Football League (NFL) level with impact reconstruction utilizing crash test dummies in an attempt to better understand concussion biomechanics [12, 13, 36, 37]. In recent years, high school and collegiate athletes have been equipped with accelerometer arrays that are capable of biomechanical measurements, including linear and rotational accelerations of the head. Since the beginning of this type of data collection, over 1.5 million head impacts have been characterized and used to quantify human tolerance to head impact [18-29]. While all of the past data is useful to an adult model, very little research has been published in regards to the youth population, 6-14 years old. Although youth players are initially believed to have lower magnitude impacts, a recent study showed that even 7-8 year old players can experience high magnitude impacts (greater than 80 g) [11]. While this study shed light on the head impact exposure that some youth players are experiencing, more research is needed to understand the head impact exposure of the entire youth spectrum (6-14 years old).

One particular area in the youth football spectrum that is unique is the group of athletes playing at the middle school level (12-14 years old) due to the greatest potential of variability between player size and ability relative to other age groups. And, it is imperative that research is done to better understand the nature of head impacts this group of football players sustain on a daily basis. Thus, the goal of this study is to investigate whether or not differences in head accelerations exist between player positions and to identify which positions are more likely to sustain high magnitude impacts. Due to players at this age level playing both offensive and defensive positions, two distinguished groups were compared, skilled positions and line positions. This study also attempts to determine if differences in head accelerations exist between session types (games or practices) and location of head impact (front, back, top, and rear) for each positional group.

Methods

A total of 17 players between the ages of 12 and 14 years old from one middle school football team participated in this study (8 skilled players and 9 line players). Additionally, the sample of players was representative of many playing positions, both line positions (guards, tackles, defensive ends, and centers), and skilled positions (safety, linebackers, running backs, quarterbacks and wide receivers). Players gave their verbal assent and signed informed permission documents were attained from player parental guardians, and the study was approved by the Virginia Tech Institutional Review Board. After obtaining approval for the study and the signed permission forms from all the parents, the helmets of the participating athletes were instrumented.

This study utilized the Head Impact Telemetry System (Simbex, Lebanon, NH) accelerometer arrays and the Sideline Response System (Riddell Corp., Elyria, OH). The accelerometer arrays are comprised of six single-axis accelerometers encased in a casing designed to fit into an adult Riddell Speed football helmet. Due to the sizing and geometrical similarities between the adult and youth Speed football helmets, these arrays also have the ability to correctly be positioned into both helmets. For head acceleration to be recorded, the acceleration of any individual accelerometer must exceed 10 g. Once data collection is triggered, data will be collected at 1 kHz for 40 ms, with 8 ms deriving from pre-trigger data and 32 ms post-trigger. The data will then be time-stamped, encoded, and sent from the array to a computer stationed on the sideline of the field.

The raw head impact data was then exported from the Sideline Response System into Matlab, where the data was further filtered. Filtration included noting the time stamp to include only the impacts sustained during the limits of any particular session (game or practice). Impacts with resultant linear accelerations of 10 g or less were omitted from the study due to being negligible in respect to impact biomechanics and their relationship to head trauma. Also, because all games and practices were recorded on video and each impact was linked to a player by a unique identifier, impacts were further filtered for false impacts. Due to the unique identifiers being attached to each impact, those impacts were easily categorized based on positional information. In order to categorize the location of head impacts, the azimuth and elevation data collected from the accelerometer arrays were utilized.

The mean and standard deviations for both linear and rotational accelerations were also calculated to describe the dataset. Several one-way analysis of variance (ANOVA) tests were performed to assess whether or not differences in total accelerations existed between positions over the course of the season. One-way ANOVA were also performed to determine if differences in accelerations existed between practices and games for the different positional groups. The ANOVA tests were completed for both linear acceleration and for rotational acceleration. Finally, Kruskal-Wallis tests were performed to identify differences in location-specific (front, back, top, and side) acceleration distributions for the positional groups. Impacts were also further divided into three separate categories: high magnitude impacts (> 80 g), moderate impacts (between 40-80 g), and low magnitude impacts (< 40 g). A χ^2 test of independence was performed to identify whether or not there is an association between high magnitude impacts (>

80 g) and player position. The level of significance for all statistical analyses was set at a p-value of less than 0.05.

Results

Over the course of the 2013 middle school football season, a total of 4678 impacts were collected from the 17 instrumented players. A total of 2925 impacts were collected from the 8 instrumented skill players and 1753 impacts were collected from the 9 instrumented line players. Skill players experienced 15.2 ± 12.5 impacts per game and 6.7 ± 3.1 impacts per practice, while line players experienced 9.0 ± 8.1 impacts per game and 4.5 ± 1.8 impacts per practice. Cumulative distribution functions were created to better compare the distribution of impacts between skill and line players (Figure 5). Line players had impact distributions weighted towards low magnitude impacts more so than skill players for both linear and rotational accelerations. Skill players had median linear and rotational accelerations of 22.9 ± 1.5 g and 972 ± 138 rad/s^2, and 95th percentiles of 58.2 ± 11.0 g and 2760 ± 543 rad/s^2, respectively. Line players had median linear and rotational accelerations of 20.3 ± 1.9 g and 938 ± 111 rad/s^2, and 95th percentile accelerations of 50 ± 6.0 g and 2317 ± 209 rad/s^2.

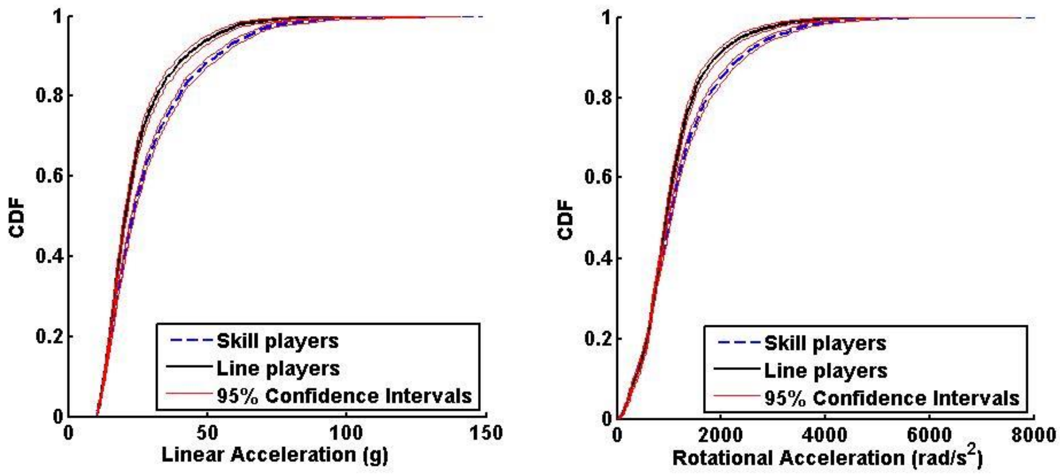

Figure 5. Cumulative distribution functions for both line and skill players with 95% confidence intervals.

Several ANOVA tests were carried out to compare the positional groups and create a better understanding of the differences that exist between them. The first group of tests completed, determined basic differences in the accelerations between the two positional groups. In terms of linear acceleration, no significant differences were found to exist between the two positional groups for median acceleration in games ($p = 0.0658$), 95^{th} percentile acceleration in games ($p = 0.3411$), 95^{th} percentile acceleration in practices ($p = 0.0598$), or 95^{th} percentile for total seasonal impacts ($p = 0.0724$). However, significant differences were found to exist between the two groups for median linear acceleration for total seasonal impacts ($p = 0.0065$) and median linear acceleration in practices ($p = 0.0112$). In terms of rotational acceleration, no significant differences were found to exist between the two positional groups for median acceleration in games ($p = 0.4430$), 95^{th} percentile acceleration in games ($p = 0.2001$), median acceleration in practices ($p = 0.5752$), or median acceleration for total seasonal impacts ($p = 0.5774$). Significant differences for rotational acceleration were found to exist between the two groups for 95^{th}

percentile accelerations for total seasonal impacts (p = 0.0380), and the 95th percentile accelerations for practices (p = 0.0275).

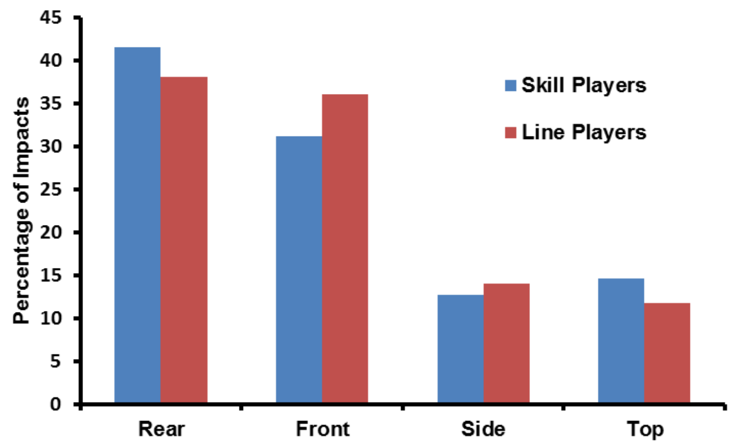

Figure 6. Impact location distributions for both Line and Skill players.

Distributions of impacts across the four generalized locations were also compared between player groups utilizing Kruskal-Wallis tests (Figure 6). When comparing the impact distributions of line and skill players in the front location, it was determined that significant differences existed for both linear and rotational acceleration (p = 5.05e-4, p = 2.14e-4 respectively). In terms of the rear location, it was determined that significant differences existed between line and skill players' impact distributions for both linear and rotational acceleration (p = 3.34e-7, p = 3.99e-5 respectively). No significant differences were found to exist between the impact distributions of the side location for the two groups in terms of both linear and rotational acceleration (p = 0.4967, 0.8997 respectively). However, significant differences were found to exist between the impact distributions of the top location for the two groups in terms of both linear and rotational acceleration (p = 1.48e-5, p =0.0099 respectively). Impacts were also divided into three groups: moderate impacts (> 40 g), high impacts (> 60 g), and severe impacts (> 80 g). Skilled players sustained approximately 74% of the 772 recorded impacts exceeding 40 g, and approximately 82% of the 60 total impacts recorded that exceed 80 g. However, utilizing a χ^2 test of independence, it was determined that there was no correlation between positional group and high magnitude impacts ($p = 0.711$).

Discussion

In the past, research has primarily focused on the adult athlete and characterizing impact distributions associated with such levels of play. However, in recent years, research has started to collect data on the vast population of youth football players, which accounts for nearly 70% of all football athletes. While impact distributions for the higher levels of play (high school, collegiate, and professional) have been characterized on a positional basis, there have yet to be any studies characterizing impact distributions at the youth level of play. However, there are several differences between the youth level and higher levels of play when looking at positional analyses. First, at the collegiate and professional levels, players generally are assigned a single position. This allows the positional analyses to be extremely detailed in the characterization of impact distributions for each individual position. While at the youth level, players tend to shift positions as they find the right fit based on preferences and team needs. A second difference between the levels of play stemming from the first is that collegiate and professional players tend to only play offense or defense, while youth players play both during games. This makes it exceptionally difficult to determine what impacts a running back would encounter during a game, when the same player also plays as a linebacker during the same game. Thus, to overcome these difficulties, two positional groups were established to determine differences among more general positions, skill players and line players.

When comparing these two groups, several distinct differences were made apparent and each difference can serve as a descriptive factor of the impact distributions for both groups. One of the first differences between the two groups that help describe the impact distributions is the median linear acceleration. ANOVA tests show that skill players have significantly higher medians when taking into consideration the entire season and during practices. The 95th percentiles and medians during games are also close to being significantly different as evidenced by the p-values. This trend suggests that skill players experience higher distributions of linear acceleration than line players, especially during practices. It is possible that this is due to differences in intensity during practices or the drills being performed by the specific groups. Rotational acceleration also displayed similar trends, although not as prevalent as those seen with linear acceleration. The two groups were seen to significantly differ in terms of 95th percentiles during practices and over the course of the entire season. This may also suggest that differences in intensity or drills during practices may play a role in the impact distribution differences.

In regards to location, several key differences have been noted to exist between the two groups. First, line players experienced a larger percentage of impacts to the front of the helmet, with significantly lower magnitude impacts than skill players to the front of the helmet. This trend is similar to previous collegiate studies noting increased frequency of lower magnitude impacts among line players [38]. Also, similar to previous studies, this study notes skill players experiencing a greater percentage of impacts to the rear and top of the helmet than line players, with significantly higher accelerations than line players [38]. So, overall, location-specific impact distributions for middle school football players are fairly similar to the impact distributions documented at the high school and collegiate levels.

In conclusion, this study has effectively documented the positional head impact exposure of a unique subsample of the youth football population, middle school football. Results from this study show that in terms of positional groups, line and skill players, the middle school football impact distributions are fairly similar to impact distributions documented for high school and collegiate players. Also, significant differences were found to exist between skill and line players for both linear and rotational accelerations, especially during practices. This may suggest that the drills that skill players specifically are required to complete during practice need to be evaluated, as these may be responsible for the higher accelerations seen by skill players. Overall, significant data has been reported in this study that increases the understanding of head impacts at the youth level. This data also has the potential to guide future modifications to youth football in order to reduce the risk of concussion.

References

[1] Thurman, D. J., Alverson, C., Dunn, K. A., Guerrero, J., and Sniezek, J. E., 1999, "Traumatic Brain Injury in the United States: A Public Health Perspective," J Head Trauma Rehabil, 14(6), pp. 602-15.

[2] Langlois, J. A., Rutland-Brown, W., and Wald, M. M., 2006, "The Epidemiology and Impact of Traumatic Brain Injury: A Brief Overview," J Head Trauma Rehabil, 21(5), pp. 375-8.

[3] Mckee, A. C., Cantu, R. C., Nowinski, C. J., Hedley-Whyte, E. T., Gavett, B. E., Budson, A. E., Santini, V. E., Lee, H. S., Kubilus, C. A., and Stern, R. A., 2009, "Chronic Traumatic Encephalopathy in Athletes: Progressive Tauopathy after Repetitive Head Injury," J Neuropathol Exp Neurol, 68(7), pp. 709-35.

[4] Gavett, B. E., Stern, R. A., and Mckee, A. C., 2011, "Chronic Traumatic Encephalopathy: A Potential Late Effect of Sport-Related Concussive and Subconcussive Head Trauma," Clinics in Sports Medicine, 30(1), pp. 179-188.

[5] De Beaumont, L., Lassonde, M., Leclerc, S., and Théoret, H., 2007, "Long-Term and Cumulative Effects of Sports Concussion on Motor Cortex Inhibition," Neurosurgery, 61(2), pp. 329-337 10.1227/01.NEU.0000280000.03578.B6.

[6] Omalu, B. I., Dekosky, S. T., Minster, R. L., Kamboh, M. I., Hamilton, R. L., and Wecht, C. H., 2005, "Chronic Traumatic Encephalopathy in a National Football League Player," Neurosurgery, 57(1), pp. 128-34; discussion 128-34.

[7] Mccrory, P., Meeuwisse, W., Johnston, K., Dvorak, J., Aubry, M., Molloy, M., and Cantu, R., 2009, "Consensus Statement on Concussion in Sport 3rd International Conference on Concussion in Sport Held in Zurich, November 2008," Clin J Sport Med, 19(3), pp. 185-200.

[8] Thurman, D. J., Branche, C. M., and Sniezek, J. E., 1998, "The Epidemiology of Sports-Related Traumatic Brain Injuries in the United States: Recent Developments," J Head Trauma Rehabil, 13(2), pp. 1-8.

[9] Guskiewicz, K. M., Weaver, N. L., Padua, D. A., and Garrett, W. E., Jr., 2000, "Epidemiology of Concussion in Collegiate and High School Football Players," Am J Sports Med, 28(5), pp. 643-50.

[10] Daniel, R., Rowson, S., and Duma, S., 2012, "Head Impact Exposure in Youth Football," Annals of Biomedical Engineering, 40(4), pp. 976-981.

[11] Powell, J. W., and Barber-Foss, K. D., 1999, "Traumatic Brain Injury in High School Athletes," JAMA: The Journal of the American Medical Association, 282(10), pp. 958-963.

[12] Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R., and Waeckerle, J. F., 2003, "Concussion in Professional Football: Reconstruction of Game Impacts and Injuries," Neurosurgery, 53(4), pp. 799-812; discussion 812-4.

[13] Pellman, E. J., Viano, D. C., Tucker, A. M., and Casson, I. R., 2003, "Concussion in Professional Football: Location and Direction of Helmet Impacts-Part 2," Neurosurgery, 53(6), pp. 1328-40; discussion 1340-1.

[14] Viano, D. C., and Pellman, E. J., 2005, "Concussion in Professional Football: Biomechanics of the Striking Player--Part 8," Neurosurgery, 56(2), pp. 266-80; discussion 266-80.

[15] Viano, D. C., Casson, I. R., and Pellman, E. J., 2007, "Concussion in Professional Football: Biomechanics of the Struck Player--Part 14," Neurosurgery, 61(2), pp. 313-27; discussion 327-8.

[16] Broglio, S., Surma, T., and Ashton-Miller, J., 2012, "High School and Collegiate Football Athlete Concussions: A Biomechanical Review," Annals of Biomedical Engineering, 40(1), pp. 37-46.
[17] Broglio, S. P., Sosnoff, J. J., Shin, S., He, X., Alcaraz, C., and Zimmerman, J., 2009, "Head Impacts During High School Football: A Biomechanical Assessment," J Athl Train, 44(4), pp. 342-9.
[18] Crisco, J. J., Fiore, R., Beckwith, J. G., Chu, J. J., Brolinson, P. G., Duma, S., Mcallister, T. W., Duhaime, A. C., and Greenwald, R. M., 2010, "Frequency and Location of Head Impact Exposures in Individual Collegiate Football Players," J Athl Train, 45(6), pp. 549-59.
[19] Crisco, J. J., Wilcox, B. J., Beckwith, J. G., Chu, J. J., Duhaime, A. C., Rowson, S., Duma, S. M., Maerlender, A. C., Mcallister, T. W., and Greenwald, R. M., 2011, "Head Impact Exposure in Collegiate Football Players," J Biomech, 44(15), pp. 2673-8.
[20] Crisco, J. J., Wilcox, B. J., Machan, J. T., Mcallister, T. W., Duhaime, A. C., Duma, S., Rowson, S., Beckwith, J. G., Chu, J. J., and Greenwald, R., 2012, "Magnitude of Head Impact Exposure in Individual Collegiate Football Players," J Appl Biomech, 28(2), pp. 174-83.
[21] Guskiewicz, K. M., and Mihalik, J. P., 2011, "Biomechanics of Sport Concussion: Quest for the Elusive Injury Threshold," Exerc Sport Sci Rev, 39(1), pp. 4-11.
[22] Guskiewicz, K. M., Mihalik, J. P., Shankar, V., Marshall, S. W., Crowell, D. H., Oliaro, S. M., Ciocca, M. F., and Hooker, D. N., 2007, "Measurement of Head Impacts in Collegiate Football Players: Relationship between Head Impact Biomechanics and Acute Clinical Outcome after Concussion," Neurosurgery, 61(6), pp. 1244-53.
[23] Rowson, S., and Duma, S., 2013, "Brain Injury Prediction: Assessing the Combined Probability of Concussion Using Linear and Rotational Head Acceleration," Ann Biomed Eng, pp.
[24] Rowson, S., and Duma, S. M., 2011, "Development of the Star Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion," Ann Biomed Eng, 39(8), pp. 2130-40.
[25] Rowson, S., Duma, S. M., Beckwith, J. G., Chu, J. J., Greenwald, R. M., Crisco, J. J., Brolinson, P. G., Duhaime, A. C., Mcallister, T. W., and Maerlender, A. C., 2012, "Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion," Ann Biomed Eng, 40(1), pp. 1-13.
[26] Funk, J., Rowson, S., Daniel, R., and Duma, S., 2012, "Validation of Concussion Risk Curves for Collegiate Football Players Derived from Hits Data," Annals of Biomedical Engineering, 40(1), pp. 79-89.
[27] Duma, S. M., and Rowson, S., 2011, "Past, Present, and Future of Head Injury Research," Exerc Sport Sci Rev, 39(1), pp. 2-3.
[28] Mihalik, J. P., Bell, D. R., Marshall, S. W., and Guskiewicz, K. M., 2007, "Measurement of Head Impacts in Collegiate Football Players: An Investigation of Positional and Event-Type Differences," Neurosurgery, 61(6), pp. 1229-35; discussion 1235.

Chapter 4: Analysis of MEP Pad Effect on Linear Acceleration During Drop Tests

Abstract

Researchers use many different tools to study the exact biomechanics associated with concussions. One of the most utilized tools in previous research studies, as well as with helmet performance evaluation, is the linear drop tower. Although the linear drop tower is an effective way to reproduce some of the same linear accelerations experienced during normal play in sports, there are several sources that may cause variability in the data. Therefore, the goal of this research is to investigate the variability and effect that one particular aspect has on drop tests. This aspect is the Modular Elastic Programmer (MEP) pad typically used in drop tests. To evaluate the effect that the MEP pad has on drop tests, various impact conditions including various impact energies and impact locations were used with and without the use of the MEP pad. A helmeted headform was instrumented with a triaxial accelerometer and linear accelerations were recorded, as well as impact velocity. Results from this testing were then statistically compared to one another and also to impact data collected from on-field impacts from previous research studies using in-helmet sensors. It was determined that results using both testing methods (with and without the use of the MEP pad) were not significantly different from one another, in terms of acceleration magnitude and impact duration. In addition, tests were determined to be fairly repeatable for both testing methods through analysis of standard deviations and coefficients of variance for repeat tests. Also, both testing methods appeared to correlate well with on-field collected data. Therefore, there appears to be minimal variability in drop tests caused by the MEP pad relative to the variability caused by the helmet.

Introduction

Traumatic Brain Injury (TBI) affects approximately 1.7 million people annually and is a contributing factor for approximately one third (30.5%) of all injury-related deaths that occur in the United States [1]. Of all cases of TBI, approximately 75% are considered mild traumatic brain injuries (mTBI), or concussions. And, one of the primary activities that can lead to concussion is participation in sports. Researchers have estimated that between 1.6 and 3.8 million sports-related concussions occur each year in the United States [2, 3]. Among all the sports that contribute to these numbers, football has been documented as one of the leading organized team activities that individuals will experience a concussion during participation [4-8]. Due to this increased risk of head injury in football, researchers have focused on improving the safety of athletic equipment for football players. However, in order to assess the ability of athletic safety equipment, it became necessary for researchers to use comparable laboratory testing methods that were representative of potentially injurious impacts.

In the past, many researchers have used animal and cadaver models to develop comparable laboratory testing methods and evaluation tools for head injury. Due to the inability to measure physiological changes in a laboratory setting, it became important to correlate concussion with measureable biomechanics. And, the primary biomechanical measurement that has been correlated to concussion through past studies is linear acceleration. One such study utilizing linear acceleration as a biomechanical correlation to concussion, and perhaps one of the most influential research studies conducted investigating head injury, introduced what has developed into the Wayne State Tolerance Curve (WSTC) [9-11]. The WSTC correlates the onset of concussion with effective acceleration and pulse duration. This curve was developed through dropping instrumented and embalmed cadaver heads onto an unyielding, flat surface, striking the subject on the forehead [9]. Following the primary introduction of the WSTC, several researchers updated the curve with results from various head injury studies, including cadaver models, animal models, human volunteers, clinical research, and injury mechanisms [12-14]. While the primary WSTC mainly correlated linear acceleration to skull fracture, the updated WSTC correlated linear acceleration to concussion and skull fracture.

Stemming from the results of the aforementioned studies, an injury metric was developed by fitting the WSTC. This injury metric was the Gadd Severity Index (GSI or SI) and was developed to assess the severity of head impacts [15]. The metric essentially involved the integration of the acceleration pulse over a period of time and designed to give researchers the ability to compare different tests for the relative severity of an impact (Equation 3). The equation used to calculate SI utilized an exponent value of 2.5, which was derived from animal studies completed at Wayne State University and its representation of internal injury to the head from frontal blows. In addition, Gadd also proposed that the injury metric could be used in the evaluation of an impact and in estimating whether an impact exceeds a safe maximum value. At the time of creation, Gadd estimated the maximum pulse intensity without danger to life to be 1000 for the threshold for serious internal head injury in frontal impact. As more research involving the correlation of linear acceleration to head injury was conducted, this threshold has shifted and changed over the years.

$$GSI = \int_0^T A^{2.5} dt \qquad \text{Equation 3}$$

Many research studies and programs have used the aforementioned Severity Index as a method of evaluating the severity of head injury, but one of the most notable efforts was completed by the National Operating Committee on Standards for Athletic Equipment (NOCSAE). Over the years, NOCSAE has made numerous contributions influencing the safety in sports. And, one of the most influential research contributions has been its evaluation and certification of football helmets [16]. Through the use of the Severity Index and other research projects correlating linear acceleration to head injury, NOCSAE developed a test protocol that evaluated a helmets ability to protect an individual from head injury. The failure criterion eventually decided on by NOCSAE was a GSI value of 1200 [17]. To reproduce similar impact conditions in the laboratory as seen on the field of play, NOCSAE proposed the use of a twin-wire guided drop tower assembly. This assembly consists of a carriage assembly, NOCSAE headform, a Modular Elastomer Programmer (MEP) pad, and a rigid anvil base. The NOCSAE test protocol is designed to include several drop heights of a helmeted NOCSAE headform onto the MEP pad attached to a metal anvil for various impact locations. Through this methodology, NOCSAE determines whether a helmet is relatively safe for the game of football. Not only have NOCSAE and helmet manufacturers adopted this testing method, but other researchers have developed safety metrics based on drop tests as well, including the STAR evaluation system [18].

Figure 7. NOCSAE thermoplastic polyurethane MEP pad.

Based upon the wide use of the drop tower and its role in determining the safety of athletic equipment, it becomes important to understand all aspects. One aspect that has the potential to produce variability within the tests is the MEP pad. The MEP pad is a ½” thick by 6” diameter pad molded from polyurethane thermoplastic elastomer (Figure 7). The main purpose of utilizing the MEP pad has postulated to modulate the energy of the impact to better simulate the impact of a helmet with elastic objects, such as a football field, whether turf or grass, or another padded player. However, the exact effects of the MEP pad on linear acceleration and acceleration pulse duration are not well documented. Thus, the goal of this study is to determine the exact effect that the NOCSAE MEP pad has in relationship to football impacts in terms of acceleration magnitude and duration.

Methodology

Instrumentation

In order to assess the effect that the MEP pad has on head accelerations during NOCSAE style drop testing, a twin-wire guided drop tower per NOCSAE protocol was utilized (Figure 8). The drop tower consists of a concrete block bolted to the floor, a metal anvil that is used to secure the MEP pad, a carriage to carry the helmeted headform along the guide wires, and a wench system to lift the carriage assembly up to the desired height. The headform was equipped with 3 single-axis accelerometers mounted at the CG of the headform. The accelerometers were oriented in such a way that the positive x-axis will come out of the forehead of the headform, the positive y axis came out the left side of the headform, and the positive z axis went through the top of the headform. The carriage assembly also has the ability to rotate the headform in a number of different ways, allowing the positioning of the headform for front, side, rear, and top impacts.

Attached to the carriage assembly was a metal card that passes through a velocity gate positioned within an inch of impact of the helmeted headform to measure impact velocity.

Figure 8. NOCSAE twin-wire guided drop tower with MEP pad, carriage assembly, and NOCSAE headform.

MEP pad evaluation

To assess the difference that the MEP pad has on linear acceleration, several drop tests were performed with and without the MEP pad utilizing the drop tower. Without the MEP pad, the helmeted headform impacted the solid steel anvil that is normally utilized to secure the MEP pad to the lower portion of the drop tower. The helmet selected for use in these drop tests was the Riddell Speed, and the same helmet was used for all testing, with and without the use of the MEP pad. The selected helmet was impacted in four locations: front, side, rear, and top. At each location, the helmeted headform was subjected to five consecutive impacts of incremental severity, made possible by increasing the drop height. The drop heights occurred in increments of 12 inches from 12 inches up to 60 inches. Measurement of the drop heights was made by using pre-cut steel bars with lengths of the desired heights to ensure minimization of human discrepancies between tests. The twenty tests were performed with the use of the MEP pad and then repeated without the pad. This full process of 40 impact tests will then be repeated two additional times to determine repeatability with the use of the MEP pad. Data was collected by utilizing standard TDAS data acquisition equipment.

Data Analysis

In order to characterize the data, several descriptive analyses were performed on the peak resultant linear accelerations for both sets, MEP pad and no MEP pad, including means and standard deviations. The Severity Index (SI) was also calculated for each testing configuration. A one-way analysis of variance (ANOVA) was performed to assess whether or not differences between the datasets exist in regards to either peak resultant linear acceleration magnitude or SI. The duration of the acceleration pulse was also analyzed and compared between datasets. The duration was described as the time period from the start of the rise in resultant linear velocity to the point which the resultant linear velocity reaches its maximum value. Resultant linear velocity was computed by integrating the acceleration pulse for each axis and then computing the resultant. A one-way ANOVA was then performed to assess differences in the datasets with respect to acceleration pulse duration.

Different impact analyses were also completed to assess the performance of drop tests utilizing an MEP pad. As mentioned previously, an analysis of the repeatability utilizing the MEP pad can be performed. To analyze the repeatability of each testing method, the coefficient of variation was calculated for each testing condition and compared due to anticipation of widely different means in linear acceleration. Furthermore, in addition to having the ability to characterize the effect of the MEP pad, the use of impacts to a rigid surface, or sans MEP pad, allows characterization of helmet response. By utilizing the linear acceleration and the impact velocity, helmet compression can be calculated for each impact. These results can then be used to define the average response of helmets to a football-like impact.

Results

To assess the effect of the MEP pad on drop tests using the linear drop tower, several statistical analyses were completed comparing the tests utilizing the MEP pad to the tests using just the metal anvil. Drop tests completed with using the MEP pad produced peak resultant linear accelerations ranging from 25.02 g to 141.04 g, and an average (± standard deviation) peak resultant linear acceleration of 69.41 ± 27.40 g, across all locations and drop heights (Table 7). This is in contrast to drop tests completed using the metal anvil producing peak resultant linear accelerations ranging from 27.68 g to 154.86 g, and an average (± standard deviation) peak resultant linear acceleration of 70.86 ± 29.22 g, across all locations and drop heights. Through ANOVA testing, it was determined that there was no significant difference between peak resultant linear accelerations produced by both testing methods ($p = 0.4685$). The Severity Index (SI) was also computed for all drop tests and compared between the two methods of testing. Tests completed using the MEP pad produced SI values ranging from 23.9 to 539.4, and had an average (± standard deviation) SI of 213.9 ± 150.4, for all testing configurations. Tests completed using the metal anvil produced SI values ranging from 27.9 to 615.3, and had an average (± standard deviation) SI of 216.1 ± 153.4, for all testing configurations. In addition, after ANOVA testing, it was determined that there was no significant difference between SI values for drop tests completed with the MEP pad and SI values for drop tests completed without the MEP pad ($p = 0.7497$).

Location	Drop Height	Linear Accel (MEP pad) (g)	Linear Accel (No MEP pad) (g)	SI (MEP pad)	SI (No MEP pad)
Front	12	27.8	28.6	26.1	28.3
Front	24	40.0	43.3	70.0	72.0
Front	36	61.8	72.8	142.2	165.0
Front	48	95.8	109.8	277.5	327.0
Front	60	136.7	149.9	512.8	579.2
Rear	12	35.2	33.7	41.8	40.0
Rear	24	52.6	49.8	102.5	95.2
Rear	36	66.8	69.8	178.6	177.9
Rear	48	78.4	79.9	266.4	270.1
Rear	60	89.9	90.5	365.6	366.9
Side	12	36.7	41.2	38.6	45.2
Side	24	57.8	61.9	102.5	115.8
Side	36	70.9	73.8	177.6	189.0
Side	48	90.1	85.7	281.4	269.4
Side	60	104.4	100.0	397.8	360.8
Top	12	39.8	36.6	54.1	48.9
Top	24	56.3	52.4	139.3	125.6
Top	36	68.9	65.9	241.2	227.2
Top	48	81.7	78.9	361.2	333.3
Top	60	96.7	92.9	500.5	484.3

Table 7. Average peak resultant linear accelerations and SI values in terms of location and drop height for the two methods used for drop testing.

In addition to acceleration magnitude and SI, impact durations were also statistically compared for both testing methods (Table 8). Impact duration was defined by the time of the rise of resultant linear velocity past 0.1 m/s to the time of maximum resultant linear velocity. Through ANOVA testing, it was determined that tests conducted with the use of the MEP pad produced significantly higher durations than drop tests conducted on the metal anvil alone ($p = <0.0001$).

Impact durations ranged from 12.3 ms to 20 ms for drop tests using the MEP pad, and from 11 ms to 18 ms for drop tests using just the metal anvil. The average (± standard deviation) impact duration for all tests completed with the MEP pad was 15.6 ± 1.5 ms, whereas the average (± standard deviation) impact duration for all tests completed without the MEP pad was 14.5 ± 2.0 ms.

Location	Drop Height	Impact Duration (MEP pad) (ms)	Impact Duration (No MEP pad) (ms)
Front	12	17.9	16.5
Front	24	16.7	17.1
Front	36	16.6	17.3
Front	48	17.3	16.7
Front	60	17.6	16.5
Rear	12	13.3	11.7
Rear	24	15.0	11.6
Rear	36	15.2	11.4
Rear	48	15.1	11.4
Rear	60	15.1	11.3
Side	12	16.5	14.8
Side	24	15.5	15.7
Side	36	15.7	16.0
Side	48	16.1	15.6
Side	60	16.0	15.5
Top	12	14.8	14.5
Top	24	14.5	14.8
Top	36	14.2	14.5
Top	48	14.3	14.4
Top	60	14.0	13.9

Table 8. Average impact durations in terms of location and drop height for the two drop testing methods.

Differences based on impact location were also compared on a distributional basis for the two drop testing methods (Figure 9). To determine these distributional differences, Kruskal-Wallis ANOVA tests were completed. It was determined that the peak resultant linear acceleration distributions for the two testing methods were not significantly different when testing the front (p

= 0.4429), rear ($p = 0.9174$), side ($p = 0.9504$), or top ($p = 0.4186$) locations. However, the testing method did influence results on an impact location basis in terms of impact duration. In terms of impact duration, it was determined that distributions were significantly different when testing the rear location ($p = 3.01e-6$). But, when testing the front, side, and top locations, it was determined that there was no statistical difference in impact durations produced from using the two testing methods. The differences in impact duration associated with testing the rear location may be attributed to the reduction of helmet surface area interacting with the impact surface. The specific helmet type used has a small ridge that may limit the impacting surface area due to the orientation of direct linear rear impacts. This impacting surface area may be increased with tests conducted with the MEP pad, as the MEP pad compresses during tests. The increased surface area during impact may lead to increased impact durations when testing in that specific orientation. And, while the MEP pad compresses, tests conducted without the use of the MEP pad will not have the same increase in impact surface area as the test progresses. Therefore, the difference in impact surface area may lead to producing significantly different impact durations between the two testing methods when testing specifically in the rear location.

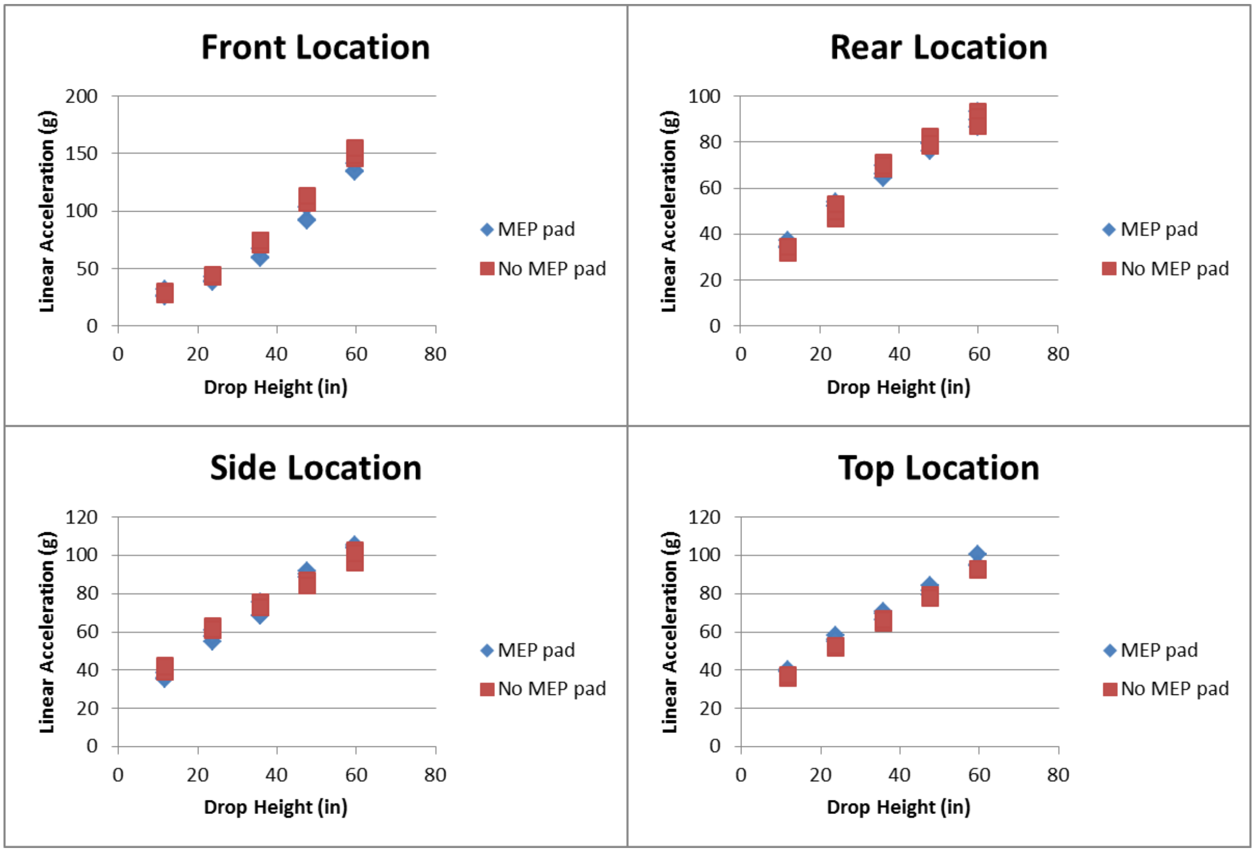

Figure 9. Comparisons of peak resultant linear acceleration distributions for the two testing methods for each impact location.

To analyze the repeatability of the drop test methods in reproducing similar peak accelerations, coefficients of variation were computed for each impact configuration (impact velocity and impact location). The coefficient of variation is unitless and used to compare distributions, regardless of large differences in means and standard deviations of each distribution. The average coefficient of variance for tests completed with the use of the MEP pad was 4.38, while the average coefficient of variance for tests completed without the use of the MEP pad was 2.77 (Table 9). Therefore, this indicates that peak accelerations are less dispersed in drop tests when conducting tests without the MEP pad.

	Linear Acceleration CV				
	Drop Height	Front	Rear	Side	Top
MEP pad	12 in	13.5	5.3	5.0	1.8
	24 in	6.7	1.9	4.9	3.1
	36 in	7.2	3.9	5.9	3.5
	48 in	7.2	2.3	1.8	2.9
	60 in	2.8	3.6	0.8	3.5
No MEP pad	12 in	4.8	4.9	4.1	2.1
	24 in	2.3	6.5	2.1	1.4
	36 in	3.0	2.0	1.9	1.8
	48 in	3.0	2.6	1.4	1.2
	60 in	3.0	3.5	3.4	0.4

Table 9. Coefficients of variation (CV) for peak linear acceleration distributions by each impact configuration.

And finally, drop tests completed without the MEP pad were utilized to compute helmet compression during impact (Table 10). By removing any variability occurring due to the use of the MEP pad, the primary source of variability becomes the helmet response. Although the compression rates are not representative of what other helmets would experience, they serve as initial approximations to the average helmet response to typical football impacts.

Drop Height (in)	Average Impact Velocity (m/s)	Average Linear Acceleration (g)	Average Compression Rate (m/s)
12	2.23	35.0	1.99
24	3.27	51.8	2.93
36	4.04	70.6	3.63
48	4.71	88.5	4.23
60	5.28	108.3	4.74

Table 10. Average compression rates of the helmet padding across the various drop heights.

Discussion

Over the past years, the linear drop tower has served as an effective tool in the evaluation of safety equipment in regards to head injury. Several researchers have developed evaluation techniques based off the performance of helmets during these drop tests. Thus, it becomes important to understand aspects of the testing procedure that may cause variability in the data. And one aspect of drop testing that has been thought to be associated with some variability is the MEP pad and its effect on linear acceleration. This study attempts to quantify any possible variability that the MEP pad causes in drop tests. By comparing impact results of tests completed with the use of the MEP pad to tests completed without using the MEP pad, the variability caused solely by the MEP pad was isolated. The variability was then quantified by running statistical tests comparing the two differing drop test methods. In addition to comparing results from test methods, researchers have utilized in-helmet sensors to capture the impact biomechanics of football players. Data derived from these studies can also be a useful comparison tool to laboratory tests to verify the replication similar impacts in terms of acceleration magnitude and duration.

Past research studies using in-helmet sensors have the ability to measure impact duration by analyzing the acceleration pulse. Several studies have been conducted and determined that the average impact duration for impacts occurring in football last approximately 14-15 ms [19-21]. These reported values are strikingly similar to drop tests conducted both with and without the use of the MEP pad. Therefore, the impact durations for typical on-field impacts are preserved in the laboratory setting. This is important as acceleration duration can be one of the factors that influence the severity of head injury [22]. In previous cadaver tests using pressure transducers within the skull, it was determined that the pressure duration was primarily related to the linear acceleration pulse [13, 14, 22-24]. Although researchers have been unable to correlate pressure within the skull to concussion directly in humans, pressure has been correlated to skull fracture in early studies using animals and cadavers.

In addition to being representative of impact durations measured in actual on-field impacts, it is important that laboratory testing is representative of impact magnitudes measured on the field of play as well. Past research studies utilizing in-helmet sensors to collect linear accelerations have estimated on-field impacts to range in magnitude from 9 g to approximately 150 g [19, 25-30]. These values are fairly similar to linear acceleration results derived from drop tests with and without the use of the MEP pad. Because there is no significant difference in terms of overall linear acceleration magnitude between the two testing methods, either can be seen as valid methods to replicate on field impacts. However, on-field data collection uses a minimum threshold of 10 g to collect biomechanical data, whereas the lowest impact condition with drop tests (12 inches) produces linear accelerations of approximately 25 g. Therefore, on-field data collection of impacts tends to be heavily weighted towards low magnitude impacts, with approximate averages of 21 g [19, 28]. This is in comparison to the completed drop tests used to simulate slightly more severe head injuries with an average of approximately 70 g. And, because the methodology used with the drop tower system is designed to test more severe impacts, comparisons to injurious data points may be more useful. Over the past decade, several institutions have utilized in-helmet sensors to collect biomechanical data associated with concussions. By combining the collection of concussive impacts between all of these groups, the average linear acceleration for concussive impacts ranges from 29 g to 178 g, and has an average of 102 g. And the majority of this range is well represented by the ranges of linear acceleration presented in this study for both drop test using the MEP pad (25 g to 141 g) and those without the MEP pad (28 g to 155 g). Although, the concussive dataset includes impacts with greater linear accelerations than those represented in laboratory testing, but only one type of helmet was used for testing and several different helmets were used in the concussive dataset. The lesser ability of older helmets to mitigate head accelerations in the concussive dataset may be the cause of the higher maximum linear acceleration in concussive impacts. However, the average linear acceleration of the concussive dataset is well contained within the range of the drop test results.

Valuable results have been presented in this study; however, there are several aspects that serve as limitations to the study. One primary limitation to the study is that only a finite number of impact locations were tested. While football players have the potential to sustain impacts from any direction, the research presented in this study only takes into account impacts occurring to the front, side, rear, and top of the helmet. However, much like previous studies and evaluation techniques, it is thought that these locations are fairly representative of the various impact conditions experienced on the field [16-18, 31]. On the same note, another limitation is this studies use of only a limited number of impact energies. Football players will also experience a vast array of impacts, in terms of magnitude, and are not just confined to experiencing energies replicated in this study. In addition to not being representative of the gamut of impacts experienced by football players on a day to day basis, the testing methods used in this paper are not representative of the rotational kinematics involved with an impact. Due to the restraints of the drop tower, only linear acceleration is correctly reproduced. Although this is the case, several studies have shown that linear acceleration is highly correlated to risk of head injury. And finally, the computation of average helmet response in football-like impacts is based solely off of the testing of one helmet. There are many different helmets currently on the market and being used in the field, and all of them may have differing energy attenuation properties. Therefore, the results in this paper detailing helmet response are only meant to serve as general estimates.

Overall, despite the aforementioned limitations, this study has successfully documented the variability and effect on drop tests associated with the use of the MEP pad. Its use has little effect on impact magnitude, and impact duration falls within reported values from the literature. Impact magnitude from drop tests using the MEP pad is also fairly representative of results from biomechanical studies of on-field impacts. All in all, it appears the MEP pad is not a significant source of variability in comparison to helmets being tested. However, the sole observational use that the MEP pad has is the possible reduction or prevention of cosmetic alterations that may occur as a result of repetitive testing. Tests utilizing the MEP pad also are fairly repeatable with an average coefficient of variance of 4.38 and an average standard deviation of 10.7 g for each testing configuration.

References

[1] Faul, M., Xu, L., Wald, M. M., and Coronado, V. G., 2010, "Traumatic Brain Injury in the United States: Emergency Department Visits, Hospitalizations, and Deaths," pp.

[2] Langlois, J. A., Rutland-Brown, W., and Wald, M. M., 2006, "The Epidemiology and Impact of Traumatic Brain Injury: A Brief Overview," J Head Trauma Rehabil, 21(5), pp. 375-8.

[3] Thurman, D. J., Branche, C. M., and Sniezek, J. E., 1998, "The Epidemiology of Sports-Related Traumatic Brain Injuries in the United States: Recent Developments," J Head Trauma Rehabil, 13(2), pp. 1-8.

[4] Cdc, 2011, "Nonfatal Traumatic Brain Injuries Related to Sports and Recreation Activities among Persons Aged ≤19 Years," MMWR, 60(39), pp.

[5] R. C., C., 1998, "Second-Impact Syndrome," Clinics in Sports Medicine, 17(1), pp. 37-44.

[6] Naunheim, R. S., Standeven, J., Richter, C., and Lewis, L. M., 2000, "Comparison of Impact Data in Hockey, Football, and Soccer," J Trauma, 48(5), pp. 938-41.

[7] Gessel, L. M., Fields, S. K., Collins, C. L., Dick, R. W., and Comstock, R. D., 2007, "Concussions among United States High School and Collegiate Athletes," J Athl Train, 42(4), pp. 495-503.

[8] Duma, S. M., and Rowson, S., 2009, "Every Newton Hertz: A Macro to Micro Approach to Investigating Brain Injury," Conf Proc IEEE Eng Med Biol Soc, 1(pp. 1123-6.

[9] Lissner, H. R., Lebow, M., and Evans, F. G., 1960, "Experimental Studies on the Relation between Acceleration and Intracranial Pressure Changes in Man," Surg Gynecol Obstet, 111(pp. 329-38.

[10] Patrick, L. M., Lissner, H. R., and Gurdijan, E. S., 1963, "Survival by Design- Head Protection," Proceedings of the 7th Stapp Car Crash Conference, pp. 483-499.

[11] Gurdijan, E. S., Lissner, H. R., Hodgson, V. R., and Patrick, L. M., 1964, "Mechanisms of Head Injury," Clin Neurosurg, 12(112-128), pp.

[12] Gurdjian, E. S., and Lissner, H. R., 1945, "Deformation of the Skull in Head Injury; a Study with the Stresscoat Technique," Surg Gynecol Obstet, 81(pp. 679-87.

[13] Gurdijan, E. S., Roberts, V. L., and Thomas, L. M., 1966, "Tolerance Curves of Acceleration and Intracranial Pressure and Protective Index in Experimental Head Injury.," Journal of Trauma, 6(pp. 600-604.

[14] L. M., P., H. R., L., and Elisha S., G., 1963, "Concussion: Mechanism and Pathology," Proceedings: American Association for Automotive Medicine Annual Conference, 7(pp. 470-482.

[15] Gadd, C. W., 1966, "Use of a Weighted-Impulse Criterion for Estimating Injury Hazard," Proceedings of the 10th Stapp Car Crash Conference, SAE 660793(pp.

[16] Hodgson, V. R., 1975, "National Operating Committee on Standards for Athletic Equipment Football Helmet Certification Program," Med Sci Sports, 7(3), pp. 225-232.

[17] Bishop, P. J., Norman, R. W., and Kozey, J. W., 1984, "An Evaluation of Football Helmets under Impact Conditions," Am J Sports Med, 12(3), pp. 233-6.

[18] Rowson, S., and Duma, S. M., 2011, "Development of the Star Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion," Ann Biomed Eng, 39(8), pp. 2130-40.

[19] Rowson, S., Brolinson, G., Goforth, M., Dietter, D., and Duma, S. M., 2009, "Linear and Angular Head Acceleration Measurements in Collegiate Football," J Biomech Eng, 131(6), pp. 061016.

[20] Pellman, E. J., Viano, D. C., Withnall, C., Shewchenko, N., Bir, C. A., and Halstead, P. D., 2006, "Concussion in Professional Football: Helmet Testing to Assess Impact Performance--Part 11," Neurosurgery, 58(1), pp. 78-96; discussion 78-96.
[21] Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R., and Waeckerle, J. F., 2003, "Concussion in Professional Football: Reconstruction of Game Impacts and Injuries," Neurosurgery, 53(4), pp. 799-812; discussion 812-4.
[22] Zhang, L., Yang, K. H., and King, A. I., 2004, "A Proposed Injury Threshold for Mild Traumatic Brain Injury," J Biomech Eng, 126(2), pp. 226-36.
[23] Hardy, W. N., Foster, C. D., Mason, M. J., Yang, K. H., King, A. I., and Tashman, S., 2001, "Investigation of Head Injury Mechanisms Using Neutral Density Technology and High-Speed Biplanar X-Ray," Stapp Car Crash J, 45(pp. 337-68.
[24] Zhang, J., Yoganandan, N., and Pintar, F. A., 2009, "Dynamic Biomechanics of the Human Head in Lateral Impacts," Ann Adv Automot Med, 53(pp. 249-56.
[25] Crisco, J. J., Wilcox, B. J., Beckwith, J. G., Chu, J. J., Duhaime, A. C., Rowson, S., Duma, S. M., Maerlender, A. C., Mcallister, T. W., and Greenwald, R. M., 2011, "Head Impact Exposure in Collegiate Football Players," J Biomech, pp.
[26] Daniel, R. W., Rowson, S., and Duma, S. M., 2012, "Head Impact Exposure in Youth Football," Ann Biomed Eng, 40(4), pp. 976-81.
[27] Young, T. J., Daniel, R. W., Rowson, S., and Duma, S. M., 9000, "Head Impact Exposure in Youth Football: Elementary School Ages 7-8 Years and the Effect of Returning Players," Clinical Journal of Sport Medicine, Publish Ahead of Print(pp. 10.1097/JSM.0000000000000055.
[28] Broglio, S. P., Sosnoff, J. J., Shin, S., He, X., Alcaraz, C., and Zimmerman, J., 2009, "Head Impacts During High School Football: A Biomechanical Assessment," J Athl Train, 44(4), pp. 342-9.
[29] Guskiewicz, K. M., Mihalik, J. P., Shankar, V., Marshall, S. W., Crowell, D. H., Oliaro, S. M., Ciocca, M. F., and Hooker, D. N., 2007, "Measurement of Head Impacts in Collegiate Football Players: Relationship between Head Impact Biomechanics and Acute Clinical Outcome after Concussion," Neurosurgery, 61(6), pp. 1244-53.
[30] Mccaffrey, M. A., Mihalik, J. P., Crowell, D. H., Shields, E. W., and Guskiewicz, K. M., 2007, "Measurement of Head Impacts in Collegiate Football Players: Clinical Measures of Concussion after High- and Low-Magnitude Impacts," Neurosurgery, 61(6), pp. 1236-43; discussion 1243.
[31] Nocsae, 2009, "Standard Test Method and Equipment Used in Evaluating the Performance Characteristics of Protective Headgear/Equipment," Technical Report No. NOCSAE DOC (ND)001-08m10, National Operating Committee on Standards for Athletic Equipment,

Chapter 5: Analysis of Impact Surface Characteristics with Tests Utilizing a Linear Impactor

Abstract

At the moment, one of the major injuries plaguing our society is concussion, and one of the primary activities leading to this injury is football. Throughout the years, researchers have attempted to better understand this injury in football through the correlation of biomechanical measurements and simulations. One method that many researchers utilize to simulate concussive impacts is the linear impactor. However, there are many aspects of testing with the linear impactor that may lead to variability in results. Thus, the goal of this study was to characterize the effects of one of these areas of variability, the impactor cap. Some research studies utilize a curved padded impactor cap when testing helmets to better represent helmet to helmet impacts, which are thought to be more severe than typical football impacts. However, little is known to the exact effect the impactor cap plays on resultant acceleration and impact duration. To characterize the linear impactor cap, several impact tests are conducted ranging in severity, with the use of both a curved padded impactor cap and a flat rigid impactor cap. The impacts are conducted using an instrumented NOCSAE headform outfitted with a Hybrid III neck, and include 4 different positions and 3 different impact velocities. The results from tests conducted with both impactor caps are then compared to one another, as well as with on-field impact biomechanics as described from research using in-helmet sensors for football players. Through testing, it was determined that the impactor caps are significantly different from one another in terms of both impact magnitude and impact duration. And, selection of which impactor cap to use for future studies is dependent on what type of impacts each study is attempting to recreate.

Introduction

Concussions have been a primary concern for medical professionals, researchers, and athletes in recent years. The recent interests in this injury may have spurned from relevant research outlining that there may be more detrimental long term effects than the normal transient symptoms. Among these long term effects are severe neurocognitive impairment, motor dysfunction, and chronic traumatic encephalopathy [1-3]. And, one major group of individuals affected by this type of injury is athletes. It is estimated that between 1.6 and 3.8 million sports concussions occur each year in the United States [4]. And, among all the sports that contribute to this large epidemic, football is one of the leading activities that individuals will experience a concussion during participation [5-8]. Because football remains an activity where many of these injuries occur, it has been a focus point for much previous research done to better understand concussions.

Many animal models have been used in the past to study the effect that football-like impacts have on concussions. Through this research, the roles of both linear and rotational accelerations were described in detail in regards to injury [9, 10]. The National Football League (NFL) has also conducted extensive research reconstructing select impacts utilizing instrumented crash test dummies to better understand the biomechanics of concussions [11-14]. This research suggested that an injury threshold of 70-75 g existed for sustaining a concussion based on linear acceleration [11]. And, throughout the years, both linear and rotational head accelerations have been thought to be the primary risk factors for concussion during an impact. These head accelerations lead to biomechanical alterations in the brain, including pressure gradients and strain patterns in the brain tissue. However, measuring these changes in the brain is extremely difficult as access to the brain is limited. In order to better connect head accelerations to come of these biomechanical alterations of the brain, Hardy et al. performed tests utilizing a bi-plane x-ray and high speed video to capture the brain changes during football-like impacts [15]. Also, the relationship of head acceleration to severe brain injury has been postulated and tested by many other researchers over the past years. In recent years, researchers have instrumented football players with accelerometer arrays to better understand the biomechanics of players during regular activities [16-18]. This type of research has been used to characterize impacts associated with on-field concussions as far as impact magnitude, and impact location.

Figure 10. Polymer caps for linear impactor. Left: Flat impactor, Right: Curved padded impactor.

Due to the past research, head accelerations remain the dominant risk factors for concussions during impact. It is this connection that has influenced laboratory tests to simulate impacts for research and safety purposes. One method that is commonly used in the laboratory for the analyses of athletic safety equipment and research purposes utilizes a linear impactor. Typical linear impactors are comprised of a thrust piston in a guide tube accelerated my accumulated compressed air. The thrust piston then impacts a helmeted headform to simulate an impact, and linear and rotational accelerations are noted. At the end of the thrust piston is a polymer cap for simulating impacts in football (Figure 10).

To simulate some of the more severe impacts, a curved padded impactor cap was designed for recreating helmet to helmet impacts often seen in football. Due to two helmets being involved in these types of collisions, both impacting objects have energy-absorbing padding. This padding in football helmets is designed to attenuate energy by prolonging the duration of contact during impact. Thus, the impactor rigid portion of the impactor cap is mounted to a pad that interfaces with the linear impactor. Also, to better simulate the exact impacting surface in a helmet to helmet impact, the impactor cap was designed to have a radius of curvature similar to the top of a football helmet. However, the exact effects of this impactor cap on linear and rotational head accelerations are not well documented. Thus, the purpose of this study is to perform characterization of a curved padded impactor cap with a linear impactor to simulate common football impacts.

Methodology

Instrumentation

In order to assess the effects of utilizing a flat polymer cap versus using a curved polymer cap, a standard linear impactor from Biokinetics was used for impact testing (Figure 11). The pneumatic impactor is comprised of a thrust piston in a guide tube that utilizes a high-pressure accumulator that stores compressed air. By using a solenoid, the set pressure acts against the piston, accelerating it to impact velocity. On the end of the piston, either a flat polymer cap or the padded curved polymer cap was utilized. The flat polymer cap was a nylon disk, with a diameter of 127 mm. And, the padded polymer cap was comprised of a nylon disk with a diameter of 127 mm and a curvature radius of 127 mm for the impacting face. A layer of padding was also used in between the curved impacting face and the thrust piston of the impactor. In order to simulate the padding of the striking player, the padding was made of DerTex VN600 vinyl nitrile with a diameter of 127 mm and a thickness of 40 mm. To account for the thickness of the foam in the padded curved impactor cap, the initial position of the headform was adjusted to strike the same location on the helmeted headform.

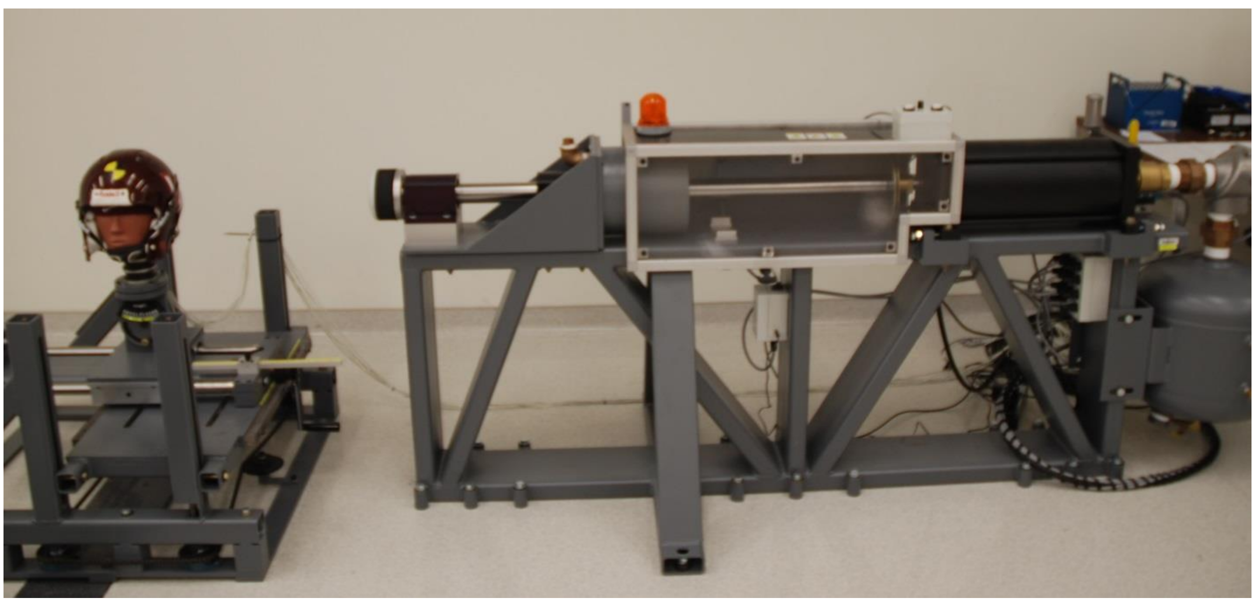

Figure 11. Biokinetics linear impactor utilizes compressed air to impact helmeted headform on mounting apparatus.

The headform that was impacted was a NOCSAE medium headform with a custom designed adapter that allowed the headform to be integrated onto a Hybrid III neck and mounting apparatus. A NOCSAE headform was chosen for this testing to simulate more realistic helmet fit with a player. The Hybrid III headform lacks a lower skull/neck area that may interface with the helmet padding during impacts. This slight difference may influence the rear impacts, and the use of the NOCSAE headform was thought to better simulate more realistic responses. Also, the Hybrid III headform has differing cheekbone structures than the NOCSAE headform. The NOCSAE headform has larger cheek areas, allowing a more realistic interaction between the helmet cheek pads and the headform. The utilized mounting apparatus has the ability to turn the headform 360 degrees in the horizontal plane and 180 degrees in the lateral or sagittal plane. The mounting apparatus also has the ability to translate the headform in any direction to align impact locations. In order to measure linear and rotational acceleration, a 6 Degrees of Freedom Sensor Package was utilized from DTS. The sensor was installed at the center of gravity (CG) of the NOCSAE headform with two bolts, and is comprised of 3 linear accelerometers and 3 angular rate sensors. Standard TDAS data acquisition hardware and software was then utilized to process and store the data on a computer.

Linear Impactor surface evaluation

The effects of the padded curved impactor cap was noted by comparing the peak linear and rotational accelerations of several impact tests on a helmeted headform with the same impacts except using a rigid flat impactor cap. The helmeted headform was impacted in 4 different locations by 3 different impact velocities. The 4 locations include the forehead area above the facemask of the helmet, the side of the helmet behind the earhole and on the left side of the NOCSAE headform (without the “ear” piece), the rear of the helmet, and the rear boss of the helmet. The 3 impact velocities were 4 m/s, 6 m/s and 8 m/s in order to simulate and view results over a range of impact severities. Impact velocities were achieved by setting the initial air pressure accordingly. This battery of testing was completed for a total of 3 iterations to assess the repeatability of the impacting surfaces.

Data Analysis

Characterization of the data was completed by performing several descriptive analyses on peak accelerations (linear and rotational), including means and standard deviations. One-way ANOVA tests were then carried out to determine if significant differences existed between the utilization of the two impactor caps. Impact durations, noted by acceleration pulse durations, were also compared between testing configurations with one-way ANOVA tests. Distributions of peak accelerations were also analyzed by impact location utilizing Kruskal-Wallis one-way ANOVA tests. A p-value of less than 0.05 was utilized for significance purposes for all statistical tests. Repeatability was then analyzed utilizing the coefficient of variation for each testing configuration and compared between impact surfaces. In addition, the relationship between the input energy and peak head acceleration was compared for the different impacting surfaces.

Results

After the completion of the laboratory tests, several statistical tests were completed to assess differences between the two impactor caps. After performing ANOVA tests, it was determined that the rigid impactor cap produced significantly higher linear ($p < 0.0001$) and rotational accelerations ($p=0.0001$) regardless of location. Peak resultant linear accelerations ranged from 45.9 g to 273.4 g for impactor tests utilizing the rigid impactor cap and from 38.2 g to 98.0 g for impactor tests utilizing the curved padded impactor cap. Peak resultant rotational accelerations ranged from 1911 rad/s^2 to 17073 rad/s^2 for all impactor tests utilizing the rigid impactor cap and from 1688 rad/s^2 to 6945 rad/s^2 for all impactor tests utilizing the curved padded impactor cap. The rigid impactor cap produced average (± standard deviation) peak resultant accelerations of 117.8 ± 61.3 g and 6643 ± 4026 rad/s2, for linear and rotational accelerations respectively, across all locations and impact velocities (Table 11). The curved padded impactor cap produced average (± standard deviation) peak resultant accelerations of 66.0 ± 17.7 g and 3724 ± 1495 rad/s2, for linear and rotational accelerations respectively, across all locations and impact velocities.

Location	Impact Velocity (m/s)	Rigid Cap (g)	Padded Cap (g)	Rigid Cap (rad/s^2)	Padded Cap (rad/s^2)
C location	4	76.3	51.6	4460	2994
C location	6	123.2	76.9	7456	4407
C location	8	181.6	93.2	10668	5861
D location	4	46.6	39.2	4217	2277
D location	6	93.3	53.2	5628	3126
D location	8	264.4	73.4	16544	4341
F location	4	63.9	45.8	1952	1726
F location	6	97.9	59.4	2516	2337
F location	8	163.5	78.2	4695	2425
R location	4	57.3	52.3	4350	3805
R location	6	94.2	76.6	6881	4826
R location	8	151.5	91.6	10345	6566

Table 11. Average linear and rotational accelerations in terms of location and impact velocity for impactor tests utilizing the two types of impactor caps.

Impact duration was defined by the time of the rise of resultant linear velocity past 0.1 m/s to the time of maximum resultant linear velocity. In terms of impact duration, the padded impactor produced significantly higher durations than the rigid impactor (p=0.0001). As expected, impact durations generally increase with increasing impact velocities for both the padded and rigid impactor caps. Impact durations ranged from 8.35 ms to 25.45 ms for all impactor tests utilizing the rigid impactor cap and from 11.10 ms to 27.65 ms for all impactor tests utilizing the curved padded impactor cap (Table 12). The average (± standard deviation) impact duration for all tests completed with the rigid impactor cap was 14.99 ± 5.51 ms, whereas the average impact duration for all tests completed with the padded impactor cap was 20.15 ± 5.11 ms.

Location	Impact Velocity (m/s)	Rigid Cap (ms)	Padded Cap (ms)
C location	4	11.38	15.33
C location	6	12.33	24.85
C location	8	23.20	27.05
D location	4	15.28	18.88
D location	6	21.90	26.38
D location	8	23.08	26.42
F location	4	9.10	12.07
F location	6	9.68	17.90
F location	8	8.45	21.57
R location	4	12.73	14.45
R location	6	14.55	16.33
R location	8	18.20	20.58

Table 12. Average impact durations in terms of location and impact velocity for impactor tests utilizing the two types of impactor caps.

For impact locations, linear and rotational acceleration distributions for the two impactor caps were compared (Figures 12 and 13). To determine differences in peak acceleration distributions by location, Kruskal-Wallis one-way ANOVA tests were completed. Through these analyses, it was determined that the peak linear acceleration distributions for the two impactor caps were significantly different when testing in impact location C ($p = 0.0243$), location D ($p = 0.0469$), and location F ($p = 0.0092$). However, significant differences in distributions were not found to exist when testing in impact location R. In terms of peak rotational acceleration for the two impactor caps, it was determined that distributions were significantly different when testing in impact location C ($p = 0.0118$) and location D ($p = 0.0031$). However, significant differences in distributions were not found to exist when testing in impact location F ($p = 0.1711$) or location R ($p = 0.1023$). The highest resultant linear accelerations occurred during impacts in the C location (side) utilizing the curved padded impactor cap and in the D location (rear boss) utilizing the rigid impactor cap. The highest resultant rotational accelerations occurred during impacts in the R location (rear) utilizing the curved padded impactor cap and in the D location (rear boss) utilizing the rigid impactor cap.

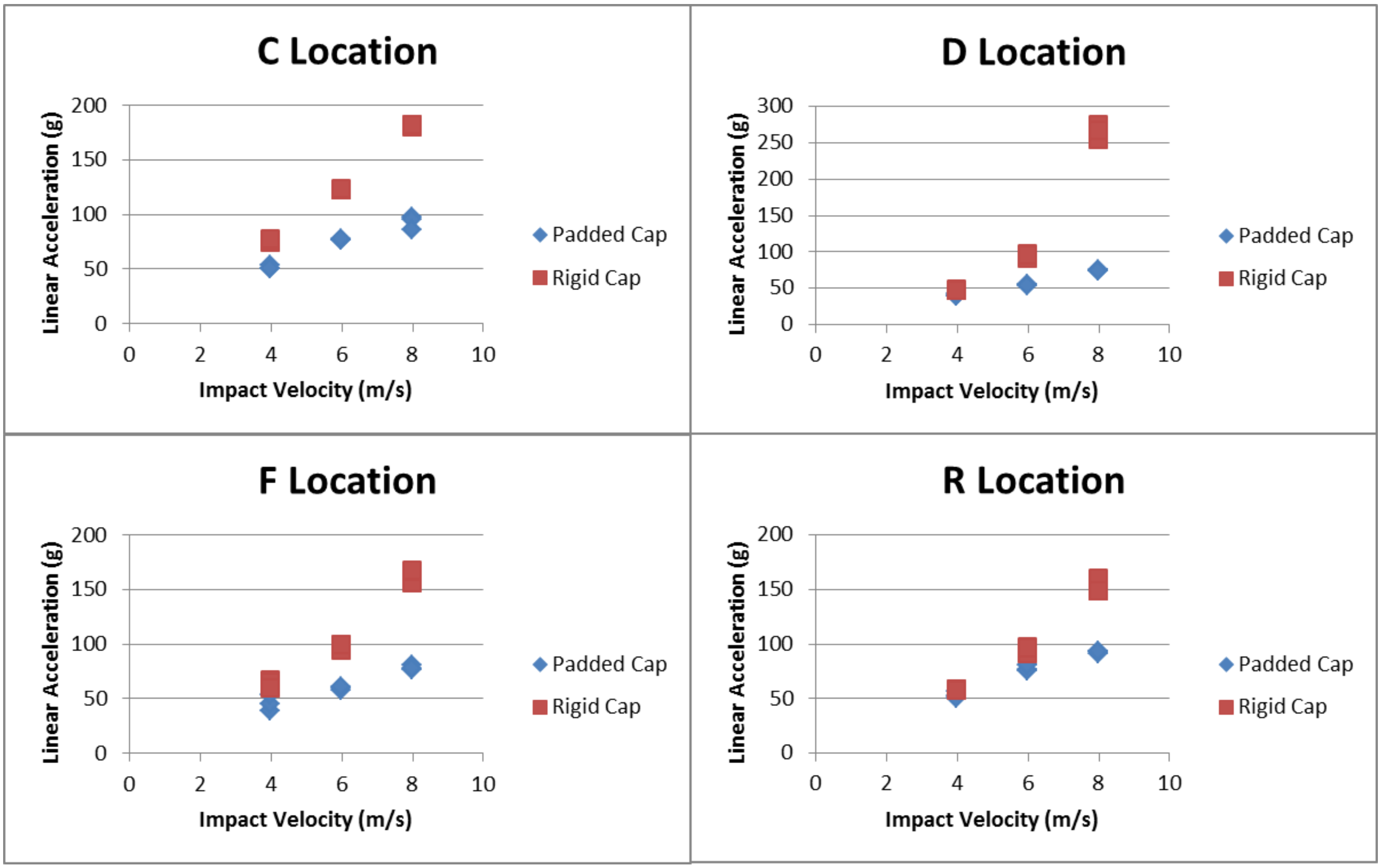

Figure 12. Comparisons of linear acceleration distributions for the two impactor caps for each impact location.

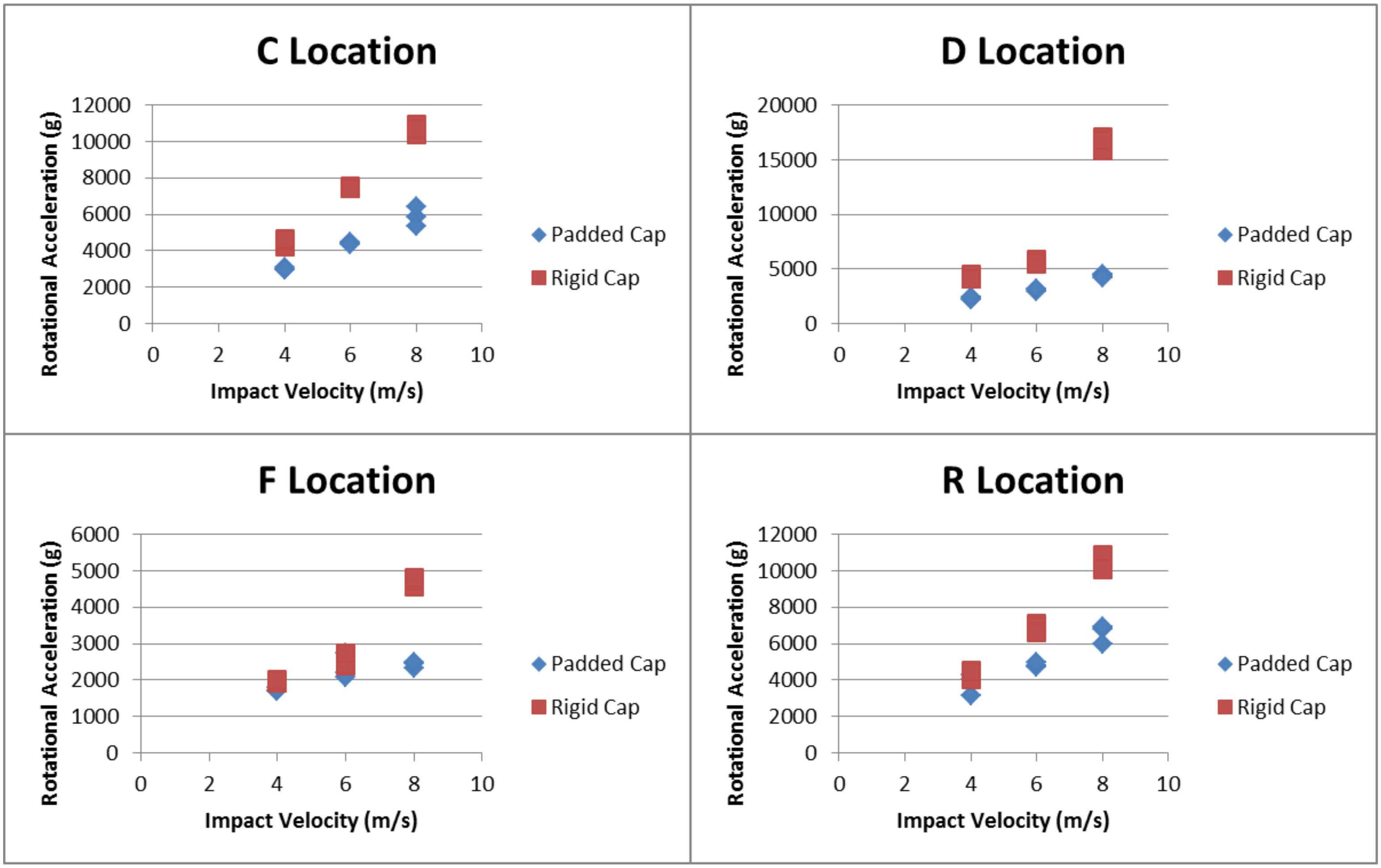

Figure 13. Comparisons of rotational acceleration distributions for the two impactor caps for each impact location.

To analyze the repeatability of the linear impactor in reproducing similar peak accelerations, coefficients of variation were computed for each impact configuration (impact velocity and impact location). The coefficient of variation is unitless and used to compare distributions, regardless of large differences in means and standard deviations of each distribution. The average coefficient of variance for the padded curved impactor cap was 5.17, while the average coefficient of variance for the rigid impactor cap was 3.33 (Table 13). Therefore, this indicates that peak accelerations are less dispersed in impactor tests when utilizing the rigid impactor cap.

		Linear Acceleration CV				Rotational Acceleration CV			
	Impact Velocity	Front (F)	Rear (R)	Side (C)	Rear Boss (D)	Front (F)	Rear (R)	Side (C)	Rear Boss (D)
Padded	4 m/s	15.78	5.35	3.17	3.79	2.79	15.89	3.15	7.44
	6 m/s	2.46	3.24	1.32	1.01	15.02	2.62	1.46	3.53
	8 m/s	3.25	1.27	6.4	1.31	3.03	8.08	9.06	3.7
Rigid	4 m/s	5.38	0.37	1.74	1.51	2.36	6.23	4.91	5.4
	6 m/s	3.56	3.27	0.64	4.2	7.11	3.58	0.93	3.49
	8 m/s	3.64	4.25	0.41	3.81	2.56	4.25	2.47	3.93

Table 13. Coefficients of variation (CV) for peak linear and rotational acceleration distributions by each impact configuration.

And finally, peak linear acceleration values were contrasted with input energy for each impact test (Figure 14). This was completed for both impactor caps to visually display distribution differences in energy attenuation. In general, and as mentioned previously, the rigid impactor cap produced higher peak linear accelerations than the padded curved impactor cap despite similar input energies.

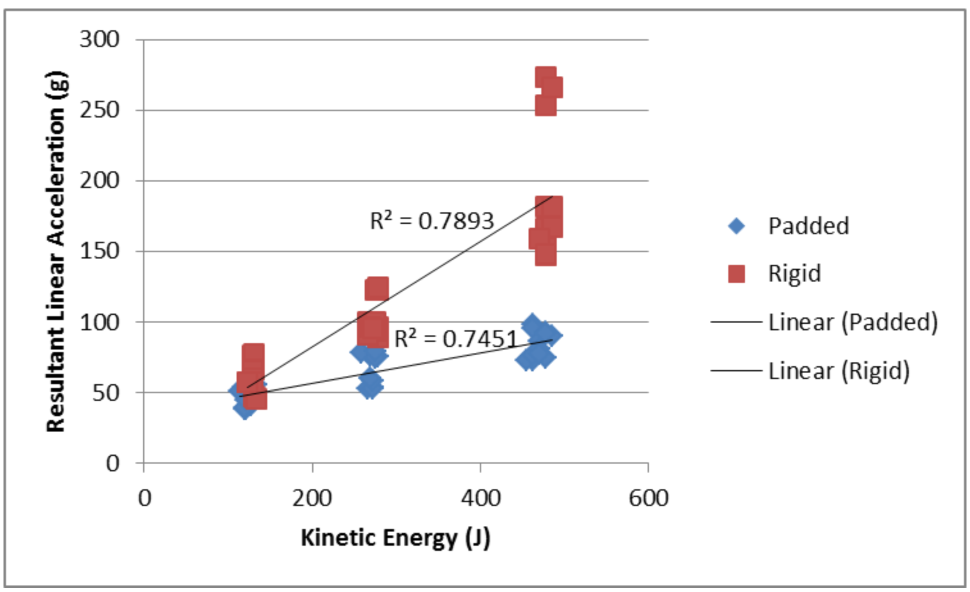

Figure 14. Input energy versus peak linear acceleration for both impactor caps.

Discussion

The linear impactor has the ability to simulate realistic impacts and has been a valuable research tool in past experiments. However, it is important that the impacts being replicated are well understood. This becomes especially true when trying to compare differing research experiments. The presented research attempts to elaborate on these differing techniques and results to better understand the role certain aspects play in experimental models. One particular aspect of the linear impactor that has not been deeply investigated is the impactor cap properties and effect on resultant accelerations. Where one methodology may prefer to use the padded curved impactor cap to better simulate helmet to helmet type impacts, other methods may prefer to simulate more general impacts with the use of the rigid impactor cap. By running matched tests in terms of impact location and impact velocity, resultant accelerations and impact durations can be computed and compared between the two impactor caps. In addition, recent studies have utilized methods to study on-field impacts during football games and practices. The data derived from such studies can then be compared to results from linear impact tests utilizing the two differing impactor caps.

One parameter derived from such datasets is the impact duration for typical impacts in collegiate football. Utilizing in-helmet sensors capable of measuring linear and rotational accelerations during an impact, it was determined that average collegiate football impacts last approximately 14 ms [19]. Compared to the presented linear impactor tests, this duration length corresponds more closely to impact durations resulting from impacts utilizing the rigid impactor cap (14.99 ms) than the padded impactor cap (20.15 ms). However, the 14 ms impact duration derived from in-helmet sensors corresponds to all types of impacts, but the curved padded impactor cap was designed specifically for helmet to helmet type impacts. Unfortunately, there is not currently a large enough collection of solely on-field helmet to helmet impacts with biomechanical data to make proper comparisons with results from the linear impactor tests utilizing the curved padded impactor cap. Note that this study is not intended to validate the use of the curved padded impactor cap, nor its ability to simulate helmet to helmet impacts as this has already been performed in past experiments.

Such validation was already completed by Pellman et al. in a detailed study commissioned by the National Football League (NFL) [20]. In this study, the use of a curved padded impactor cap and its ability to simulate helmet to helmet impacts was validated using a pendulum impactor and further tested utilizing a linear impactor. This testing was completed to assess impact performance of helmets through laboratory testing; however, this study was just part of a much larger study conducted by Pellman. One particular aspect of the overall study that has been useful for comparisons is the analysis of reconstructed concussive impacts from video of NFL games and players [11, 12]. Through video analysis, it was determined that players experienced injurious head impacts of 9.3 ± 1.8 m/s [11]. These impacts were then reconstructed in the laboratory utilizing helmeted Hybrid III dummies and identical impact velocity, direction, and head kinematics as in the game. Through this method, it was observed that the average peak linear acceleration for concussion was 98 ± 28 g and had an average impact duration of 15 ms [11]. Future studies performed logistic regression on the NFL reconstructions and produced injury thresholds of 79–82 g and 5757–5900 rad/ s^2, representing 50% risk of concussion [21, 22]. These ranges in peak resultant acceleration more closely correspond to those seen in the impactor tests utilizing the curved padded impactor cap in this current study. This is as expected, since the padded impactor cap was designed to replicate these helmet to helmet impacts derived from concussive NFL reconstructions. However, the impact durations of the reconstructions appear to more closely correspond to the impactor tests utilizing the rigid impactor cap. One key difference between these two sets of data that may explain differences in impact duration between the reconstructed dataset and padded impactor cap is the helmet being used for testing. Where the NFL reconstructed dataset utilized an older Riddell football helmet, this current study utilized a vastly improved, in terms of mitigating accelerations, Riddell Speed for testing. Impact velocities were also vastly different between the NFL reconstructions and the linear impact results presented within this report. The higher velocities used in the NFL reconstructions may have led to the full compression of the foam, therefore producing lower impact durations. Also, though the methodologies for both sets of tests were vastly different, with one utilizing a guided drop tower assembly and the other utilizing a linear impactor, it is still a useful comparison.

Although this study can be used to compare on-field impacts, the primary goal is to assess the variability associated with the use of the curved padded impactor cap. The major source of variability as displayed in the results is in relation to peak acceleration magnitude. Due to the additional compliance introduced to the testing system in the form of the padding on the cap, peak acceleration magnitude is decreased with a longer impact duration. In addition to the padding influencing peak acceleration magnitude, another aspect of the impactor cap that introduces variability into the system is the curved impactor face. The two curved surfaces of the impactor cap and helmet shell interacting with one another may produce various results. Depending on the exact impact location of the curved impactor cap relative to the impacted helmet shell, the directional vector may be altered to reflect a glancing blow rather than a head on collision. Even slight differences in impact locations with helmet positioning may lead to inconsistencies with replicating impacts. And, due to the helmet being constrained in the y-axis, the directional vectors of the impact may be further altered. However, each impact scenario was visually inspected to align the impacted helmet for a head on collision as best as possible in these series of tests. Some of these sources of variability may be reasons for the difference in coefficients of variation for both testing methods. When analyzing the coefficients of variation for both testing methods, the repeatability of the rigid impactor cap was shown to be better with a lower coefficient of variation (3.33 versus 5.17).

Valuable data and comparisons are shown through this study, but they are not without limitations. First, only four locations were investigated and compared utilizing the linear impactor. These locations were precise locations marked on the helmet and impacted for each impact and for every impact velocity. In that respect, due to repeated tests in each location, it is possible that the helmet padding may experience slight changes in energy attenuating properties. However, this change is thought to be minimal in comparison to the overall energy attenuation of the helmet. A second limitation of this study is that only one helmet was utilized for all tests. This makes comparisons to similar tests complicated should other helmet types be utilized. Different helmets will have differing energy attenuating properties and may not result in the same peak linear and rotational accelerations from this study. And, finally, only two impactor caps were evaluated. Although these two types of caps have predominantly been used in past experiments, it is possible that various other impactor caps could be utilized to simulate various impacts of varying severities. One particular impactor cap that may be of interest of future studies is the use of a flat padded impactor cap. This interface may better correlate to helmet to ground type impacts or perhaps have a better correlation to general football impacts than current impactor caps, depending on the padding properties.

Laboratory testing has always been a valuable asset to researchers; however, it becomes important to understand exactly what the laboratory tests are simulating. In the case of studying concussions from a biomechanical standpoint, it becomes important to reproduce accelerations similar to impacts experienced in a realistic setting. Reproducing these types of accelerations means recreating similarities existing in acceleration magnitude for both linear and rotational acceleration, as well as impact duration. Through this study, two different impactor interfaces have been tested, analyzed, and compared both to one another and to previously reported laboratory and volunteer testing. This was completed to determine how closely laboratory tests mimicked the biomechanics of realistic impacts. It was determined through testing that the impactor caps were significantly different from one another, in terms of resultant acceleration magnitudes and impact durations. When compared to the previously reported NFL reconstructions, it was determined that impact magnitudes resulting in a 50% risk of concussion more closely correlated to impactor tests utilizing the curved padded impactor cap at the tested impact velocities. Contrary to impact acceleration magnitudes, the impact durations corresponding to the NFL reconstructions were more similar to those produced in tests utilizing the rigid impactor cap for the tested impact velocities. This may be the case because higher velocities were used in the NFL reconstructions, causing the padding of the curved padded impactor cap to bottom out. This bottoming out of the foam may have led to lower impact durations during the high velocity NFL reconstructions. However, when describing the repeatability of the testing methods, tests conducted with the flat rigid impactor cap were determined to have good repeatability with an average coefficient of variance of 3.33.

Overall, the data presented herein shows that the choice of impactor cap varies depending on what impact properties are being replicated in the laboratory setting. Although the curved padded impactor cap can be useful for replicating helmet to helmet impacts, additional variability is introduced to the system. This variability comes in the form of two curved surfaces interacting with one another and also the compliance of the padding during impact. Due to the compliance introduced to the system by additional padding, tests conducted with the flat rigid impactor cap at lower velocities may produce similar results as tests conducted with the curved padded impactor cap at higher velocities. Tests completed with the use of the rigid impactor cap is a more repeatable testing method; however, there is not enough on-field data other than helmet to helmet impacts correlating impact velocity to peak resultant acceleration at this time. With the growth of such a dataset, better comparisons can be made in the future for correlating realistic impacts to laboratory testing. Overall, these data presented herein can be used for reference for future studies looking to use the linear impactor to simulate impacts with specific biomechanical properties.

References

[1] Guskiewicz, K. M., Marshall, S. W., Bailes, J., Mccrea, M., Cantu, R. C., Randolph, C., and Jordan, B. D., 2005, "Association between Recurrent Concussion and Late-Life Cognitive Impairment in Retired Professional Football Players," Neurosurgery, 57(4), pp. 719-26; discussion 719-26.
[2] Cantu, R. C., 2007, "Chronic Traumatic Encephalopathy in the National Football League," Neurosurgery, 61(2), pp. 223-225 10.1227/01.NEU.0000255514.73967.90.
[3] Mckee, A. C., Cantu, R. C., Nowinski, C. J., Hedley-Whyte, E. T., Gavett, B. E., Budson, A. E., Santini, V. E., Lee, H. S., Kubilus, C. A., and Stern, R. A., 2009, "Chronic Traumatic Encephalopathy in Athletes: Progressive Tauopathy after Repetitive Head Injury," J Neuropathol Exp Neurol, 68(7), pp. 709-35.
[4] Langlois, J. A., Rutland-Brown, W., and Wald, M. M., 2006, "The Epidemiology and Impact of Traumatic Brain Injury: A Brief Overview," J Head Trauma Rehabil, 21(5), pp. 375-8.
[5] Thurman, D. J., Alverson, C., Dunn, K. A., Guerrero, J., and Sniezek, J. E., 1999, "Traumatic Brain Injury in the United States: A Public Health Perspective," J Head Trauma Rehabil, 14(6), pp. 602-15.
[6] Centers for Disease Control and Prevention (Cdc), N. C. F. I. P. a. C., 2003,
[7] Hootman, J. M., Dick, R., and Agel, J., 2007, "Epidemiology of Collegiate Injuries for 15 Sports: Summary and Recommendations for Injury Prevention Initiatives," J Athl Train, 42(2), pp. 311-9.
[8] Cdc, 2011, "Nonfatal Traumatic Brain Injuries Related to Sports and Recreation Activities among Persons Aged ≤19 Years," MMWR, 60(39), pp.
[9] Ommaya, A. K., Hirsch, A. E., Flamm, E. S., and Mahone, R. H., 1966, "Cerebral Concussion in the Monkey: An Experimental Model," Science, 153(732), pp. 211-2.
[10] Ommaya, A. K., Yarnell, P., Hirsch, A. E., and Harris, E. H., 1967, "Scaling of Experimental Data on Cerebral Concussion in Sub-Human Primates to Concussion Threshold for Man," Proceedings of the 11th Stapp Car Crash Conference, SAE 670906(pp.
[11] Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R., and Waeckerle, J. F., 2003, "Concussion in Professional Football: Reconstruction of Game Impacts and Injuries," Neurosurgery, 53(4), pp. 799-812; discussion 812-4.
[12] Pellman, E. J., Viano, D. C., Tucker, A. M., and Casson, I. R., 2003, "Concussion in Professional Football: Location and Direction of Helmet Impacts-Part 2," Neurosurgery, 53(6), pp. 1328-40; discussion 1340-1.
[13] Pellman, E. J., Powell, J. W., Viano, D. C., Casson, I. R., Tucker, A. M., Feuer, H., Lovell, M., Waeckerle, J. F., and Robertson, D. W., 2004, "Concussion in Professional Football: Epidemiological Features of Game Injuries and Review of the Literature--Part 3," Neurosurgery, 54(1), pp. 81-94; discussion 94-6.
[14] Pellman, E. J., Viano, D. C., Casson, I. R., Tucker, A. M., Waeckerle, J. F., Powell, J. W., and Feuer, H., 2004, "Concussion in Professional Football: Repeat Injuries--Part 4," Neurosurgery, 55(4), pp. 860-73; discussion 873-6.
[15] Hardy, W. N., Foster, C. D., Mason, M. J., Yang, K. H., King, A. I., and Tashman, S., 2001, "Investigation of Head Injury Mechanisms Using Neutral Density Technology and High-Speed Biplanar X-Ray," Stapp Car Crash J, 45(pp. 337-68.

[16] Duma, S. M., Manoogian, S. J., Bussone, W. R., Brolinson, P. G., Goforth, M. W., Donnenwerth, J. J., Greenwald, R. M., Chu, J. J., and Crisco, J. J., 2005, "Analysis of Real-Time Head Accelerations in Collegiate Football Players," Clin J Sport Med, 15(1), pp. 3-8.
[17] Rowson, S., Chu, J. J., Beckwith, J. G., Greenwald, R. M., Stitzel, J. D., Kimpara, H., and Duma, S. M., 2007, "Six Degree of Freedom Head Acceleration Measurements in Football Players," eds., pp.
[18] Guskiewicz, K. M., Weaver, N. L., Padua, D. A., and Garrett, W. E., Jr., 2000, "Epidemiology of Concussion in Collegiate and High School Football Players," Am J Sports Med, 28(5), pp. 643-50.
[19] Rowson, S., Brolinson, G., Goforth, M., Dietter, D., and Duma, S. M., 2009, "Linear and Angular Head Acceleration Measurements in Collegiate Football," J Biomech Eng, 131(6), pp. 061016.
[20] Pellman, E. J., Viano, D. C., Withnall, C., Shewchenko, N., Bir, C. A., and Halstead, P. D., 2006, "Concussion in Professional Football: Helmet Testing to Assess Impact Performance--Part 11," Neurosurgery, 58(1), pp. 78-96; discussion 78-96.
[21] Zhang, L., Yang, K. H., and King, A. I., 2004, "A Proposed Injury Threshold for Mild Traumatic Brain Injury," J Biomech Eng, 126(2), pp. 226-36.
[22] King, A. I., Yang, K. H., Zhang, L., Hardy, W., and Viano, D. C., 2003, "Is Head Injury Caused by Linear or Angular Acceleration?," eds., Lisbon, Portugal, pp.

Chapter 6: Research Summary and Resulting Publications

Research Summary

The research presented in this dissertation aimed to characterize the head impact exposure of a unique subgroup of youth football and evaluate laboratory methods used for the research of head injury. This research has yielded an in depth analysis of head impact exposure for middle school football that can lead to informed decisions aimed at reducing concussion risk at this level of play. These decisions may come in the form of rule changes, protective equipment changes, or game and practice structure changes. In addition, the variability of major aspects of laboratory testing methods for head injury research has been quantified and the results can be used to guide future research. Depending on desired impact conditions and simulations, researchers can refer to the presented work in determining specific testing methods.

As a direct result of this dissertation, several research objectives have been addressed:

1. The head impact exposure of a widely variable subgroup of youth football, middle school football, was determined.
2. The head impact exposure of middle school football players on a positional basis, skill and line players, was determined.
3. The effects of the Modular Elastic Programmer (MEP) pad on drop tests and used in previous testing methods and helmet evaluation was determined.
4. The effect of using a curved padded linear impactor cap on testing results and its ability to replicate realistic impact conditions was determined.

Expected Publications

This research is intended to be published in several journals. Table 14 details the publications for each chapter in this dissertation.

Chapter	Title	Anticipated Journal
1	Head Impact Exposure in Youth Football: Middle School Ages 12 to 14 Years	Journal of Biomechanical Engineering
2	An Investigation of Positional Differences in Middle School Football	Clinical Journal of Sports Medicine
3	Analysis of MEP Pad Effect on Linear Acceleration During Drop Tests	Medicine & Science in Sports & Exercise
4	Analysis of Impact Surface Characteristics with Tests Utilizing a Linear Impactor	Medicine & Science in Sports & Exercise

Table 14. Expected publications from this dissertation.

Current Publication Status

Journal Publications

1. Daniel RW, Rowson S, Duma SM. Head impact exposure in youth football. Ann Biomed Eng. 2012 Apr;40(4):976-81. doi: 10.1007/s10439-012-0530-7. Epub 2012 Feb 15.
2. Young TJ, Daniel RW, Rowson S, Duma SM. Head Impact Exposure in Youth Football: Elementary School Ages 7-8 Years and the Effect of Returning Players. Clin J Sport Med. 2013 Dec 9.
3. Funk JR, Rowson S, Daniel RW, Duma SM. Validation of concussion risk curves for collegiate football players derived from HITS data. Ann Biomed Eng. 2012 Jan.
4. Rowson S, Daniel RW, Duma SM. Biomechanical performance of leather and modern football helmets. J Neurosurg. 2013 Sep.

In-preparation Publications

1. Daniel RW, Rowson S, Duma SM. Head Impact Exposure in Youth Football: Middle School Ages 12 to 14 Years. Journal of Biomechanical Engineering, 2014 (under review).
2. Daniel RW, Rowson S, Duma SM. An Investigation of Positional Differences in Middle School Football. Clinical Journal of Sports Medicine, 2014.
3. Daniel RW, Rowson S, Duma SM. Analysis of MEP Pad Effect on Linear Acceleration During Drop Tests. Medicine and Science in Sports & Exercise, 2014.
4. Daniel RW, Rowson S, Duma SM. Analysis of Impact Surface Characteristics with Tests Utilizing a Linear Impactor. Medicine and Science in Sports & Exercise, 2014.

Conference Presentations

1. Daniel RW, et al. Investigation of Effective Mass Differences in Helmet to Helmet Impacts in Football. Biomedical Engineering Society Annual Meeting. Hartford, CT, 2011.
2. Daniel RW, Rowson S, Duma SM. Linear and Angular Head Acceleration Measurements in Pediatric Football. ASME Summer Bioengineering Conference. Fajardo, PR, 2012.
3. Daniel RW, Rowson S, Duma SM. Head Impact Exposure in Youth Football. Biomedical Engineering Society Annual Meeting. Atlanta, GA, 2012.
4. Daniel RW, Rowson S, Duma SM. Helmet Evaluation and General Safety in Football. American Society of Safety Engineers, Roanoke Chapter Meeting. Roanoke, VA, 2013.
5. Daniel RW, Rowson S, Duma SM. Head Impact Exposure in Middle School Football. Biomedical Engineering Society Annual Meeting. Seattle, WA, 2013.