Accepted Manuscript

Article Type: Short Communication

Title: Impact Characteristics of Two American Football Helmet Models

Authors: Brian Wallace[1]*, Kyle Petit[1], Jenna Hawk[2], Brianna Roberts[3]

Affiliations: [1]University of Wisconsin Oshkosh, Department of Kinesiology, 800 Algoma Blvd, Oshkosh, WI, USA; [2]Pacific University, School of Physical Therapy & Athletic Training, 2043 College Way, Forest Grove, OR, USA; [3]Manhattanville College, Department of Athletics and Recreation, 2900 Purchase Street, Purchase, NY, USA

Figures: 0

Tables: 2

Running Title: Helmet Impacts in American Football

Corresponding Author Email: wallaceb@uwosh.edu

Abstract

Few studies have compared the impact characteristics of modern American football helmet types. The purpose of this study was to measure and compare the collision biomechanics of the common Vicis Zero1 (2.27 kg) and Riddell Speedflex (2.09 kg) helmets. Linear drop tests were conducted from 1.52, 1.98, 2.59, and 3.05 m with a Hybrid III headform. The helmets were positioned crown-down and dropped in a manner that resulted in minimal pre-collision rotation. An 8-camera motion capture system recording at 300 Hz was used to verify that contact occurred to the region of the helmet immediately superior to the top of the facemask for each trial. Impact kinetics were recorded via a force plate sampling at 1800 Hz. Peak and mean accelerations (g) were calculated from initial contact until peak vertical ground reaction force (VGRF) was reached. Independent t-tests and effect sizes (Hedge's g) were performed. The Speedflex helmet resulted in greater peak VGRF, peak acceleration, and mean acceleration ($p=0.045$, <0.001, and <0.001, respectively). Effect sizes were medium for VGRF (0.73) and large for peak (48.07) and mean (2.28) accelerations. These data indicate that the Zero1 helmet, which has a compliant outer shell, may lead to greater athlete safety by resulting in peak forces and accelerations of less magnitude as compared to helmets with rigid shell designs.

Keywords: Riddell, VICIS, Impacts, Brain injury

Introduction

Mild traumatic brain injury (mTBI), including sport-related concussions (SRC), are a common risk of American football participation, occurring on average 6.71 times per 10,000 athlete exposures [1)]. These injuries can lead to an array of symptoms, cognitive impairments, and other health complications [2)]. SRCs are a clinical diagnosis [3)] and on average take approximately 10-14 days to recover from [2, 4)]. Sub-concussive impacts are those that occur where there is an apparent brain insult of insufficient force to result in the clinical diagnosis of SRC [5)]. Considering the difficulties associated with the diagnosis of SRCs, some clinicians have begun to utilize telemetry systems to help identify an athlete who has sustained a potentially dangerous hit to the head. Utilizing these devices, researchers have reported SRC-causing linear acceleration thresholds of 86-105 g [6, 7)]. Keeping accelerations below these thresholds through advances in equipment, technique, or competitive parameters may reduce mTBI risk. Reducing the frequency and magnitude of head impacts may help improve the long-term brain health of American football athletes [8)].

Several factors such as position, play type, and helmet design have been evaluated for their contributions to head impact characteristics in American football [9-15)]. Specifically, lineman primarily make contact with each other on the part of the helmet superior to the facemask and have been found to on average sustain the highest frequency and second highest magnitude of head impacts as compared to other position groups [8, 9)]. Additionally, special teams plays with long closing distances (i.e., kickoffs) yielded significantly greater linear and rotational acceleration forces compared to other play types. Due to these findings, American football governing bodies have made changes to kickoffs, with the aim to reduce the impact forces sustained by athletes. Although leagues and researchers have identified and addressed some of the positional and situational head impact concerns, continued efforts are needed to further reduce the impact burden sustained by American football athletes. Improvements in helmet design are another avenue for reducing the forces sustained by the head in an effort to reduce mTBI incidence.

Various helmet add-ons have been created to provide additional padding to the exterior of helmets, which the National Football League requires certain position groups to wear during some practices. However, the efficacy of these products is unclear based on existing research [13, 15)], with one study showing no on-field benefit [16)]. Other attempts at reducing the force sustained by an athlete's head have come with advancements in helmet design. Newer advanced helmet designs have higher impact attenuation characteristics than older helmets, which may reduce SRCs [17, 18)]. The VICIS Zero1 and Riddell Speedflex are considered advanced helmet designs [18)]. Researchers have previously compared the force-mitigation properties of some helmet types [11, 12, 14, 15, 17)], however there is a need for continuous independent comparisons of contemporary American football helmet models to ensure athletes are utilizing equipment that may best attenuate the factors associated with sporting collisions. Linear drop tests have been used to determine helmet safety as far back as the 1960s, when the National Operation Committee on Standards for Athletic Equipment was formed. Today, the effect that helmets have on reducing linear accelerations are still one of several factors used to assess a helmet's effectiveness toward reducing forces sustained by the head [17, 19-21)]. The purpose of this study was to compare the collision biomechanics of two common American football helmet models, the VICIS Zero1 helmet vs. the Riddell Speedflex helmet. We hypothesized that the Zero1 helmet would have more protective impact characteristics than the Speedflex helmet due to the compliance of its outer shell design.

Material and methods

Experimental Design

A cross-sectional study design was used to compare the impact characteristics of two helmet models commonly used by American football athletes during linear drop tests. Each helmet was dropped from four different heights onto a force platform. Vertical ground reaction forces (VGRF) were recorded for the duration of each collision., and accelerations were calculated from the force data.

Drop Test Procedures

Vertical linear drop tests for the VICIS Zero1 and Riddell Speedflex helmets (n=1 each) from heights of 1.52, 1.98, 2.59, and 3.05 meters were conducted to mimic different intensity collisions. Three measurements were performed on a standard beam scale to determine the mass of each helmet, with the three trials averaged. The order of drop heights were randomized. Five trials were recorded for each helmet at each height. The helmets were size large and in new condition. The same helmet was used for all trials per helmet type. A Hybrid III headform was used to mimic a human head [21)]. Prior to each trial the helmets were positioned upside down (e.g., the superior part of the helmet shell facing down towards the force plate) in a custom apparatus that allowed the helmet to free-fall upon being released. Contact with the force platform occurred slightly superior to the facemask.

Data collection

Kinematic data were collected for the purpose of verifying helmet impact location using eight infrared Vero 2.2 cameras (Vicon Corporation, Denver, CO, USA) at a sampling rate of 300 Hz. A custom markerset of five retro-reflective markers were affixed to the helmet to track position during the drop tests. The marker placements were right and left sides above each earhole aligned with the top of the facemask, right and left sides on the back of the helmet aligned with the top of the facemask, and right offset. Marker position data were reviewed after each trial for to verify the helmet's collision with the force plate occurred superior to the facemask, as described previously. If the point of contact was not satisfactory, the trial was omitted and repeated. Cameras were synchronized with a force platform (model OR6-7-2000, AMTI Corporation, Watertown, MA, USA) which recorded VGRFs during impact at a sampling rate of 1800 Hz. The force plate surface was level with the surrounding floor. Both kinematic and kinetic data were collected and synchronized using Nexus software (version 2.11, Vicon Corporation, Denver, CO, USA).

Data analysis

A 10 N threshold was used to determine the onset of collision with the force plate. All dependent variables were calculated from the raw vertical ground reaction force data at each time point for the collision duration of each trial. The VGRF data was not filtered prior to analysis as to not artificially attenuate the impact forces. Acceleration values were calculated by differentiating velocity, which was calculated from impulse and helmet mass per Newton's second law. Accelerations were normalized to the acceleration due to gravity. Peak VGRF values, and peak and mean acceleration values, were calculated. Initial impact and the associated momentum transfer is the most important phase of collisions related to brain trauma,[22)] therefore data was analyzed from initial contact with the force plate until the time that peak force was reached.

Statistical Analyses

Force and acceleration data for each trial were averaged within condition and used in the subsequent statistical tests. Data were collapsed between drop heights for each helmet type. Independent t-tests were performed between helmet types for peak VGRF and peak and mean accelerations. Levene's Test was used to assess equality of variances between helmets, and the corrected p-value was used where this test was significant. Effect size (ES) was assessed using Hedge's g. Statistical analyses were performed using SPSS version 29 (SPSS, Inc., Chicago, IL, USA). The alpha level was set at $p \leq 0.05$ for all analyses.

Results

The Zero1 helmet had a greater mass as compared to the Speedflex helmet (2.27 kg vs 2.09 kg). Descriptive statistics (e.g., mean±SD) for each helmet type at each drop height are presented in Table 1. Peak VGRF and acceleration (peak and mean) data during impact for all drop heights combined are presented in Table 2. Peak VGRF and peak and mean accelerations were significantly greater in the Speedflex helmet independent of drop height ($p=0.045$, <0.001, and <0.001, respectively). The Zero1 helmet had reduced VGRF of 4.8 N, peak accelerations of 11.1 g, and mean accelerations of 9.8 g vs the Speedflex helmet. Peak VGRF was significantly greater in the Speedflex helmet (2841.01±4.97 N) vs the Zero1 helmet (2830.85±1.71 N) when values from only the two highest drop heights were combined ($p<0.001$). The ES for peak force was moderate (0.73), but very large for both peak acceleration (48.07) and mean acceleration (2.28).

Insert Tables 1 and 2 Approximately Here

Discussion

Our most meaningful finding between helmet types was that peak linear accelerations (e.g., the maximum value for how quickly helmet velocity slowed during impact) in the Zero1 helmet were of significantly less magnitude than in the Speedflex helmet, with a large ES. One recent investigation tested the attenuation of linear accelerations during collisions in several different helmets, including the Zero1 and Speedflex [17)]. Similar to our study, these authors reported that the Zero1 had preferential impact kinetics vs the Speedflex in the helmet regions superior to the facemask. Specifically, the Zero1 helmet reduced translational accelerations approximately 17-30% more than the Speedflex helmet [17)]. Our relative findings of the Zero1 helmet being more protective also align with Diekfuss and colleagues [18)], which quantified the alteration of white matter microstructure in young athletes following a season of American high school football. These authors reported that players who wore the Zero1 helmet had fewer changes in white matter structure than those who wore other helmet models, including Speedflex. These results are likely due to the Zero1 helmet having a more compliant outer shell comprised of novel internal energy redistribution structures made from thermoplastic elastometer vs the traditional polycarbonate helmet composition found in the Speedflex. This design appears to preferentially absorb impact forces better than helmets with a rigid outer shell.

Researchers have previously reported the peak linear acceleration threshold associated with SRCs as between 86 and 105 g [6, 7)]. More recently, Brennan et al. [23)] reported that the peak linear acceleration associated with SRC in wearable head sensor-based devices were 98.7 g (95% CI

82.4-115.0 g). The peak and mean linear accelerations of both helmet types in our study from initial contact until peak VGRF surpassed even the high end of this 95% CI. Repeated sub-concussive impacts resulting in head acceleration events of as little as 50 g have been shown to result in brain trauma in both American football and soccer players [24, 25)]. The portion of impact most strongly associated with brain injury is initial impact and the period of time shortly thereafter[22)]. Our mean acceleration data, which represent acceleration from initial contact until peak VGRF was reached, indicate that helmet accelerations surpass the aforementioned thresholds for the entirety of the early collision phase. Brain acceleration would likely be of less magnitude than the values obtained via our method of directly assessing the helmet due to additional protective layers between the helmet's outer shell and brain. However, the magnitude of these data indicate that reducing the impact forces sustained to the head may be important to reducing long-term brain harm in American football athletes.

The Zero1 helmet had more mass than the Speedflex helmet, however the peak VGRF were significantly greater in the Speedflex helmet, particularly at the higher drop heights which mimicked more intense collisions. This indicates that the Zero1 helmet may be more protective during collisions most likely to cause brain injury, as peak VGRF would be expected to be greater in the more massive helmet if force absorbing characteristics were similar. A player may have a more difficult time controlling a collision while wearing a more massive helmet. This is particularly true in individuals with weaker neck strength, which has been shown to be a contributing factor for concussion risk [26)]. That the Zero1 helmet resulted in lower peak VGRF even with greater mass further suggests that it's compliant outer shell design has preferential force-absorbing properties that dissipate forces associated with collision impulse more effectively. Leather helmets have a compliant outer layer and have been reported to significantly reduce the risk of developing a mTBI [27)]. Bartsch and colleagues[11)] tested accelerations and neck forces associated with rigid shell American football helmets vs mid-20th century leatherhead American football helmets in response to impacts. They reported that the leatherhead helmet protected against accelerations and neck forces just as well, and in some cases better, than the rigid shell helmets. Leatherhead helmets do not protect the skull well from fracture and therefore are unsuitable for modern sport. However, our results combined with these data suggest that helmet designs that are structurally more compliant may help reduce mTBI.

There are some limitations to our study. First, neither the actual forces transmitted to the brain nor brain movement were measured. The currently utilized method that may best infer brain movement are mouthguard-based accelerometers, however, the validity of those at the current time is questionable [16, 28)]. The method we used is preferable to video-based methods of calculating acceleration from position data due to motion capture systems typically having a maximum frame rate of approximately 300 Hz. We also only tested two helmet designs from two manufacturers, however, the helmets chosen are widely used in American football.

Conclusion

Our results suggest that the VICIS Zero1 helmet may provide better athlete safety due to reduced acceleration values as compared to the Riddell Speedflex helmet during linear drop tests. Both helmet types, however, had peak accelerations higher than the threshold which has recently been reported to induce SRC, although differences in study methodologies may present dissipate absolute g forces. Reduced peak VGRF values in the Zero1 helmet support the use of compliant outer shell designs. Manufacturers should continue to develop helmets with the goal of reducing accelerations during the early impact phase of collisions, while concurrently working to reduce helmet mass to make the head and neck segments easier for an athlete to control during a collision.

Author Contributions: BW: conceptualization, methodology, data collection, data analysis, writing and editing. KP: data analysis, writing and editing. JH: methodology, data collection, data analysis. BR: conceptualization, methodology, data collection, data analysis. All authors approved the final version of the manuscript.

Submission Statement: All authors have read and agree with the manuscript content. The manuscript has not been published and is not under consideration for publication elsewhere.

Conflicts of Interest: The authors declare that there are no conflicts of interest.

Acknowledgements: We thank the University of Wisconsin Oshkosh football program for providing the helmets used in this study.

References

1) Zuckerman SL, Kerr ZY, Yengo-Kahn A, Wasserman E, Covassin T and Solomon GS. 2015. Epidemiology of sports-related concussion in NCAA athletes From 2009-2010 to 2013-2014: Incidence, recurrence, and mechanisms. *Am J Sports Med* 43: 2654-2662. doi:10.1177/0363546515599634.

2) McCrory P, Meeuwisse W, Dvorak J, Aubry M, Bailes J, Broglio S, Cantu RC, Cassidy D, Echemendia RJ, Castellani RJ, Davis GA, Ellenbogen R, Emery C, Engebretsen L, Feddermann-Demont N, Giza CC, Guskiewicz KM, Herring S, Iverson GL, Johnston

KM, Kissick J, Kutcher J, Leddy JJ, Maddocks D, Makdissi M, Manley GT, McCrea M, Meehan WP, Nagahiro S, Patricios J, Putukian M, Schneider KJ, Sills A, Tator CH, Turner M and Vos PE. 2017. Consensus statement on concussion in sport-the 5(th) international conference on concussion in sport held in Berlin, October 2016. *Br J Sports Med* 51: 838-847. doi:10.1136/bjsports-2017-097699.

3) Joseph JR, Swallow JS, Willsey K, Lapointe AP, Khalatbari S, Korley FK, Oppenlander ME, Park P, Szerlip NJ and Broglio SP. 2018. Elevated markers of brain injury as a result of clinically asymptomatic high-acceleration head impacts in high-school football athletes. *J Neurosurg* 1-7. doi:10.3171/2017.12.JNS172386.

4) Williams RM, Puetz TW, Giza CC and Broglio SP. 2015. Concussion recovery time among high school and collegiate athletes: a systematic review and meta-analysis. *Sports Med* 45: 893-903. doi:10.1007/s40279-015-0325-8.

5) Gavett BE, Stern RA and McKee AC. 2011. Chronic traumatic encephalopathy: a potential late effect of sport-related concussive and subconcussive head trauma. *Clin Sports Med* 30: 179-188, xi. doi:10.1016/j.csm.2010.09.007.

6) Broglio SP, Schnebel B, Sosnoff JJ, Shin S, Fend X, He X and Zimmerman J. 2010. Biomechanical properties of concussions in high school football. *Med Sci Sports Exerc* 42: 2064-2071. doi:10.1249/MSS.0b013e3181dd9156.

7) Duhaime AC, Beckwith JG, Maerlender AC, McAllister TW, Crisco JJ, Duma SM, Brolinson PG, Rowson S, Flashman LA, Chu JJ and Greenwald RM. 2012. Spectrum of acute clinical characteristics of diagnosed concussions in college athletes wearing instrumented helmets: clinical article. *J Neurosurg* 117: 1092-1099. doi:10.3171/2012.8.JNS112298.

8) Lee TA, Lycke RJ, Lee PJ, Cudal CM, Torolski KJ, Bucherl SE, Leiva-Molano N, Auerbach PS, Talavage TM and Nauman EA. 2021. Distribution of head acceleration events varies by position and play type in North American football. *Clin J Sport Med* 31: e245-e250. doi:10.1097/JSM.0000000000000778.

9) Broglio SP, Williams RM, O'Connor KL and Goldstick J. 2016. Football players' head-impact exposure after limiting of full-contact practices. *J Athl Train* 51: 511-518. doi:10.4085/1062-6050-51.7.04.

10) Swartz EE, Broglio SP, Cook SB, Cantu RC, Ferrara MS, Guskiewicz KM and Myers JL. 2015. Early results of a helmetless-tackling intervention to decrease head impacts in football players. *J Athl Train* 50: 1219-1222. doi:10.4085/1062-6050-51.1.06.

11) Bartsch A, Benzel E, Miele V and Prakash V. 2012. Impact test comparisons of 20th and 21st century American football helmets. *J Neurosurg* 116: 222-233. doi:10.3171/2011.9.JNS111059.

12) Bailey AM, McMurry TL, Cormier JM, Funk JR, Crandall JR, Mack CD, Myers BS and Arbogast KB. 2020. Comparison of laboratory and on-field performance of American football helmets. *Ann Biomed Eng* 48: 2531-2541. doi:10.1007/s10439-020-02627-5.

13) Bailey AM, Funk JR, Crandall JR, Myers BS and Arbogast KB. 2021. Laboratory evaluation of shell add-on products for American football helmets for professional linemen. *Ann Biomed Eng* 49: 2747-2759. doi:10.1007/s10439-021-02842-8.

14) Breedlove KM, Breedlove EL, Bowman TG and Nauman EA. 2016. Impact attenuation capabilities of football and lacrosse helmets. *J Biomech* 49: 2838-2844. doi:10.1016/j.jbiomech.2016.06.030.

15) Breedlove KM, Breedlove E, Nauman E, Bowman TG and Lininger MR. 2017. The ability of an aftermarket helmet add-on device to reduce impact-force accelerations during drop tests. *J Athl Train* 52: 802-808. doi:10.4085/1062-6050-52.6.01.

16) Cecchi NJ, Callan AA, Watson LP, Liu Y, Zhan X, Vegesna RV, Pang C, Le Flao E, Grant GA, Zeineh MM and Camarillo DB. 2023. Padded helmet shell covers in American football: A comprehensive laboratory evaluation with preliminary on-field findings. *Ann Biomed Eng* 1-14. doi:10.1007/s10439-023-03169-2.

17) McIver KG, Lee P, Bucherl S, Talavage TM, Myer GD and Nauman EA. 2023. Design considerations for the attenuation of translational and rotational accelerations in American football helmets. *J Biomech Eng* 145. doi:10.1115/1.4056653.

18) Diekfuss JA, Yuan WH, Dudley JA, DiCesare CA, Panzer MB, Talavage TM, Nauman E, Bonnette S, Slutsky-Ganesh AB, Clark J, Anand M, Altaye M, Leach JL, Lamplot JD, Galloway M, Pombo MW, Hammond KE and Myer GD. 2021. Evaluation of the effectiveness of newer helmet designs with emergent shell and padding technologies versus older helmet models for preserving white matter following a season of high school football. *Ann Biomed Eng* 49: 2863-2874. doi:10.1007/s10439-021-02863-3.

19) Huang JJ, Goya KN, Yamamoto BE and Yamamoto LG. 2023. Comparing impact and concussion risk in leatherhead and modern football and hockey helmets. *Neurosurgery* 92: 1297-1302. doi:10.1227/neu.0000000000002355.

20) Rowson S, Campolettano ET, Duma SM, Stemper B, Shah A, Harezlak J, Riggen L, Mihalik JP, Guskiewicz KM, Giza C, Brooks A, Cameron K, McAllister T, Broglio SP and McCrea M. 2019. Accounting for variance in concussion tolerance between

individuals: Comparing head accelerations between concussed and physically matched control subjects. *Ann Biomed Eng* 47: 2048-2056. doi:10.1007/s10439-019-02329-7.

21) Cummiskey B, Sankaran GN, McIver KG, Shyu D, Markel J, Talavage TM, Leverenz L, Meyer JJ, Adams D and Nauman EA. 2019. Quantitative evaluation of impact attenuation by football helmets using a modal impulse hammer. *Proceedings of the Institution of Mechanical Engineers Part P-Journal of Sports Engineering and Technology* 233: 301-311. doi:10.1177/1754337118823603.

22) Viano DC, Casson IR, Pellman EJ, Zhang L, King AI and Yang KH. 2005. Concussion in professional football: brain responses by finite element analysis: Part 9. *Neurosurgery* 57: 891-916. doi:10.1227/01.neu.0000186950.54075.3b.

23) Brennan JH, Mitra B, Synnot A, McKenzie J, Willmott C, McIntosh AS, Maller JJ and Rosenfeld JV. 2017. Accelerometers for the assessment of concussion in male athletes: A systematic review and meta-analysis. *Sports Med* 47: 469-478. doi:10.1007/s40279-016-0582-1.

24) Bari S, Svaldi DO, Jang I, Shenk TE, Poole VN, Lee T, Dydak U, Rispoli JV, Nauman EA and Talavage TM. 2019. Dependence on subconcussive impacts of brain metabolism in collision sport athletes: an MR spectroscopic study. *Brain Imaging Behav* 13: 735-749. doi:10.1007/s11682-018-9861-9.

25) Svaldi DO, Joshi C, McCuen EC, Music JP, Hannemann R, Leverenz LJ, Nauman EA and Talavage TM. 2020. Accumulation of high magnitude acceleration events predicts cerebrovascular reactivity changes in female high school soccer athletes. *Brain Imaging Behav* 14: 164-174. doi:10.1007/s11682-018-9983-0.

26) Farley T, Barry E, Sylvester R, Medici A and Wilson MG. 2022. Poor isometric neck extension strength as a risk factor for concussion in male professional Rugby Union players. *Br J Sports Med* 56: 616-621. doi:10.1136/bjsports-2021-104414.

27) Hollis SJ, Stevenson MR, McIntosh AS, Shores EA, Collins MW and Taylor CB. 2009. Incidence, risk, and protective factors of mild traumatic brain injury in a cohort of Australian nonprofessional male rugby players. *Am J Sports Med* 37: 2328-2333. doi:10.1177/0363546509341032.

28) Tierney G. 2021. Concussion biomechanics, head acceleration exposure and brain injury criteria in sport: a review. *Sports Biomech* 1-29. doi:10.1080/14763141.2021.2016929.

Table 1: Vertical ground reaction force (VGRF) and acceleration data at each individual height for Zero1 and Speedflex helmets. Mean±SD.

Helmet Model	Drop Height (m)	Peak VGRF (N)	Peak Acceleration (g)	Mean Acceleration (g)
Zero1	1.52	2831.90±0.10	127.17±0.00	109.12±2.01
	1.98	2832.19±0.30	127.18±0.01	107.37±1.94
	2.59	2831.09±2.67	127.13±0.12	110.14±2.13
	3.05	2830.62±1.10	127.11±0.00	104.88±7.21
Speedflex	1.52	2831.43±0.40	138.10±0.02	119.13±5.54
	1.98	2831.83±0.30	138.12±0.02	117.35±2.63
	2.59	2839.14±0.87	138.34±0.28	118.11±3.37
	3.05	2842.63±7.36	138.48±0.54	116.27±6.86

Table 2: Vertical ground reaction force (VGRF) and acceleration data collapsed between heights for Zero1 and Speedflex helmets. Mean±SD. *p≤0.05.

Helmet Model	Peak VGRF (N)	Peak Acceleration (g)	Mean Acceleration (g)
Zero1	2831.45±1.32*	127.15±0.06*	107.88±4.00*
Speedflex	2836.26±5.92	138..26±0.31	117.72±4.32