A BIOFIDELIC EVALUATION OF THE HEAD IMPACT TELEMETRY SYSTEM ON MEASURING HEAD IMPACT LOCATION AND FREQUENCY THROUGH LABORATORY AND ON-FIELD ASSESSMENTS

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A dissertation submitted to the faculty at the University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Doctor of Philosophy in the Human Movement Science Curriculum in the School of Medicine.

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ABSTRACT

Kody Ryan Campbell: A Biofidelic Evaluation of the Head Impact Telemetry System on
Measuring Head Impact Location and Frequency Through Laboratory and On-Field
Assessments
(Under the Direction of Jason P. Mihalik)

Concussions are caused by excessive linear and rotational head accelerations from direct or indirect head loading. Researchers have used head impact sensors, like the Head Impact Telemetry (HIT) System, to measure football head impact frequency, magnitude, and location. The HIT System's measurement accuracy has previously been compared to gold-standard reference sensors rigidly coupled to anthropometric test device (ATD) heads. The ATD heads' skin does not mimic human skin compliance and oiliness, creating an artificial coupling surface between the ATD head and the HIT System. This dissertation's purpose was to evaluate the HIT System's impact detection and location accuracy while the system was coupled to more biofidelic conditions better replicating the system's intended use on the football playing field. Using an innovative biofidelic surrogate head testing paradigm—cadaver human head drops—the HIT System measured statistically different impact location coordinates than reference sensors except at the facemask drop location ($p>0.05$). The HIT System had low agreement with reference sensors in measuring impact location category on most drop sites. We subsequently quantified the HIT System's impact detection and head impact location measurement accuracy while high school football players wore the system during special teams plays in games. Video observed impacts and impact locations were documented and merged with HIT System impact data. The HIT System's impact filtering algorithm accurately categorized 70% of the data collection trigger events as either true head impacts or non-head

impacts. The HIT System agreed with video observations of impact location on 64% of 129 analyzed impacts. Head impact frequency may be underestimated for studies using the HIT System during special teams plays, and potentially other play types. We recommend confirming head impact locations with video analysis to ensure accurately quantifying the head loading environment to the extent possible. Understanding the relationships between impact location and injury risk can lead to protective equipment improvements, identifying athletes for technique improvement limiting head impact exposure and concussion risk, and developing data-derived rule modifications for reducing concussion risk.

To my best friend, my exit buddy, my wife:

For your unfailing love, for your continued support when I thought it was impossible to continue,
and for your sacrificial care for me and our daughter,

I dedicate this dissertation to you.

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This study required exponentially more manpower than I could contribute alone. I want to thank all of those that were involved in the Laboratory Drop Testing at Duke in 2014, and those involved in the BeMod project, which lead to the head impact biomechanics dataset used in this dissertation. A huge thank you to Josh Boone and Adrian Boltz for your dedicated analysis of high school football impacts. Thank you for your patience and attention to detail to review impact after impact after impact.

Thank you to the UNC Injury Prevention Research Center that is partly supported by an award from the Centers for Disease Control and Prevention. This award supported in part the acquisition of video and head impact data for my dissertation.

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LIST OF ABBREVIATIONS

AE	Athletic Exposures
ATD	Anthropometric Test Device
BeMod	Behavior Modification
COG	Center of Gravity
CFC	Channel Filter Class
cm	centimeter
CT	Computed Tomography
DOF	Degree(s) of Freedom
FN	False Negative
FP	False Positive
g	linear acceleration expressed relative to gravitation acceleration
GSI	Gadd Severity Index
HIC	Head Impact Criterion
High School RIO	National High School Sports-Related Injury Surveillance System
HIT System	Head Impact Telemetry System
HITsp	Head Impact Telemetry Severity Profile
Hz	Hertz
kg	kilogram
kHz	kilohertz
m	meter
MSE	Mean Spherical Error
MHz	Mega hertz
mm	millimeter
ms	millisecond

m/s	meters per second
NOCSAE	National Operating Committee for the Standards of Athletic Equipment
rad/s^2	radians per second per second
SDE	Standard Deviation Ellipse
TN	True Negative
TP	True Positive

CHAPTER 1

SPECIFIC AIMS

Sport-related concussion is a major public health concern in part due to the potential late-life neurological sequelae associated with recurrent concussions,[1,2] and the potential for neurodegenerative diseases from repetitive head impacts that do not result in any diagnosed concussion.[3] Concussions are caused by excessive linear and rotational head accelerations, and the proportion of linear and rotational accelerations the head experiences are dictated by the head impact location and direction of loading.[4,5] Football players experience more severe head impacts during special teams plays, as special teams plays involve larger closing distances between players, resulting in higher impact velocities.[6] For this reason, high school football players more likely sustain a concussion during special teams plays than during regular pass and run plays.[7] Researchers have used head impact sensors, like the Head Impact Telemetry (HIT) System, to measure the frequency, location, and magnitude of impacts football players experience during competition in an effort to measure head impact exposures to different player positions, understand clinical correlates to concussion injury biomechanics, and establish concussion injury risk curves.[8–10] The accuracy of the HIT System at measuring head impact frequency, location, and magnitude has been compared to gold-standard reference sensors rigidly coupled to anthropometric test device (ATD) heads.[11–13] The skin of ATD heads does not mimic human skin compliance and oiliness, and friction,[14,15] which creates an artificial coupling surface between the ATD head and the HIT System. Research evaluating the HIT System's outputs in a biofidelic coupling environment is crucial if we are to use this sensor to develop data-driven rule modifications to make sports safer and quantify head impact exposures to understand potential risks for late-life neurological sequelae.

Our long-term objective is to understand concussion injury risk in sport and make sport participation safer for athletes by instrumenting athletes with accurate and reliable head impact sensor technologies. ***Our objective*** for this proposal is to perform a biofidelic evaluation of the HIT System at measuring the impact frequency and location through laboratory and on-field assessments. ***Our approach*** will evaluate the HIT System's impact location accuracy through controlled laboratory drops with cadaveric human heads. We will verify the HIT System's performance at measuring impact frequency and location in an on-field setting by evaluating video of instrumented football players competing in special teams plays during games. ***Our central hypothesis*** is that the coupling level between the head and the helmet is going to determine the HIT System's accuracy at measuring impact frequency and location. This proposal's specific aims are:

Aim 1: **To evaluate the HIT System's accuracy at measuring impact location against gold standard reference sensors coupled to cadaveric human heads while undergoing laboratory controlled drops.**

Hypothesis: Impact location coordinate mean spherical error between the HIT System and the gold standard reference sensors will be higher from drops onto the facemask location as compared to drops directly onto the helmet shell.

Significance: Head impact loading in the coronal plane produces longer durations of loss of consciousness and more persistent behavior deficits in primate and porcine testing. It is important to have sensor systems accurately measuring impact location to understand directional dependent loading factors on concussion risk and neurological health.

Innovation: Using the cadaveric human heads represents biofidelic coupling between the HIT System sensor and the head mimicking the sensor's intended use in the field. The realistic coupling will assist in evaluating the accuracy of the system in measuring head impact location in a laboratory environment.

Aim 2: To verify laboratory performance of the HIT System at measuring head impact frequency and location in an on-field setting through video analysis of special teams plays during high school football games.

Hypothesis: The HIT System will detect more impacts and have a higher percentage of agreement on impact location with a video observer from impacts directed onto the helmet shell than glancing impacts to the helmet shell or impacts to the facemask location.

Significance: A higher proportion of concussions occur to high school football players during special teams plays as compared to run/pass play events. Data-driven rule modifications to protect athletes during special teams plays at the high school football level hinge on the HIT System accurately detecting impacts and measuring impact location during special teams plays.

Innovation: This aim will quantify impact loading conditions and locations that lead to the HIT System not detecting and incorrectly classifying impact locations in the field during special teams plays.

CHAPTER 2

A. SIGNIFICANCE

As many as 3.8 million individuals suffer a sport- or recreational-related concussion annually, with an estimated 1.1 to 1.6 million concussions occurring to children under 18 years old.[16,17] These sport-related concussions cost the American health care system an estimated 60 billion dollars in direct and indirect costs.[18] It is important to study the loading conditions to the brain to understand post-concussion recovery trajectories, develop data-informed rule modifications and legislation that protect athletes, and inform equipment design for better protection against concussions. Each head impact should produce a unique response to different brain tissues and structures.[10] These unique brain tissue responses may be dictated by the head impact location, force direction, and force magnitude. Impact location and force direction are important factors in injury persistence and behavioral outcomes as immature porcine brains subjected to sagittal plane head rotations display persistent axonal brain damage and behavioral deficits as compared to porcine brains subjected to axial head rotations.[19] Head injury directional dependence extends to the primate brain, with primate brains experiencing longer durations of unconsciousness as a result of coronal plane head rotation than sagittal plane head rotation for equal head rotation magnitude.[20] The influence of head impact location, force direction, and force magnitude on injury risk, severity, and outcomes have been more consistent in animal models than in humans. Part of this inconsistency in the relationship between biomechanical inputs into injury can be attributed to limitations in the way researchers measure these biomechanical determinants through video analysis and injury reconstruction, clinical observations, or by head impact sensors. The HIT System is a head impact sensor used in football to study head impact frequency, location, and magnitude determinants for post head injury outcomes for several individuals. *This research contributes knowledge related to the HIT System's accuracy at measuring impact location in biofidelic laboratory and on-field testing environments.* This contribution is significant because it will identify the HIT System's strengths and limitations at measuring head impact location. Short-term, understanding the system's accuracy allows for more confidence in using the data to make clinical suggestions based upon directional brain loading data and make football rule adaptations to better protect athletes from injurious and non-injurious impacts. Long-term, we can improve the HIT system's limitations at measuring impact location by developing new technology iterations, better coupling methods between the sensor and the head, and improvements to their algorithms at determining impacts, impact location and impact magnitude.

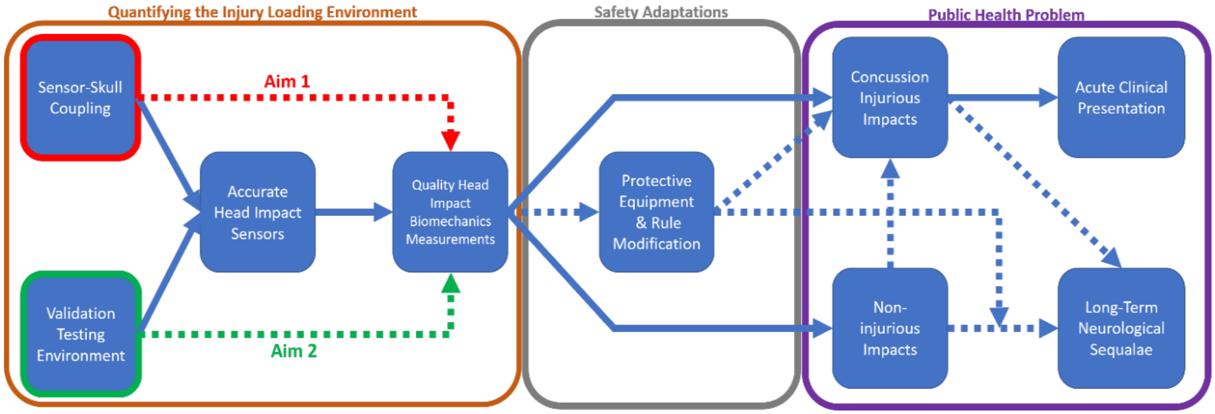


Figure 2.1: Dissertation Project Conceptual Model. The project's aims quantify the sensor-skull coupling and validation testing environment on the HIT System's impact detection and location accuracy. Accurate head impact data, collected from accurate head impact sensors, can inform protective equipment design and rule modifications, establish concussion injury risk, and quantify relationships between non-injurious impacts and long-term neurological sequalae

B. INNOVATION

American football offers an ideal environment to study concussion injury mechanisms and their relationship to injury risk[9,21–25] and potential long-term consequences,[1–3,26] because of the frequent head collisions athletes experience while playing (**Figure 2.1**). Football players are at high risk for concussions during special teams plays like kickoff/kick returns and punt/punt returns.[7,27] These special teams play types allow for larger closing distances between impacting players which results in high impact velocities.[6] To meet these unmet challenges, engineering solutions include improved helmet designs and a growing array of head impact sensors designed to measure head impact frequency, location, and magnitude from the injury loading environment. These sensors require rigid coupling to the head for accurately measuring head impacts,[28] and looser helmet fits lead to increased head impact magnitude error made by the HIT System. To date, studies evaluating the HIT System's accuracy at measuring head impact magnitude and location have commonly used the HIT System mounted on a Hybrid III ATD dummy head.[11–13,29,30] *Our proposed research is innovative because it will use human cadaveric heads, which is a more biofidelic representation of coupling between the HIT System and the head, to evaluate the HIT System's accuracy at measuring head impact location* (**Figure 2.1: Aim 1**). The Hybrid III ATD head is made of dry vinyl and does not mimic the same friction of human skin contacting the sensor, while human skin is slippery with sweat, oils, and hair.[14,15,31,32] The Hybrid III ATD head's dry vinyl skin provides a more stable coupling environment between the head and the sensor to increase head sensor coupling and increase head impact location calculation accuracy. The Hybrid III ATD head has a substantially narrow jaw and cheeks, and the bottom edge of the back of the dummy head does not extend to the bottom edge of helmets.[33] The unrepresentative geometric configuration of the Hybrid III ATD, however, could decrease head sensor coupling and decrease head impact location calculation accuracy under head loading conditions directed to the bottom of the helmet. Inaccurate head impact location measurements can alter our understanding of positional differences in head impact loading and head impact location dependent injury risk and clinical outcomes.[8,10,34–36] Our study will evaluate the HIT System's impact location accuracy in a controlled laboratory setting with cadaver human head drops. In addition, we will build on our understanding of the HIT System at measuring head impact location outside of the laboratory environment by evaluating its performance at measuring impacts and head impact location while American high school football players wear the system during special teams plays while competing in games.

The game environment offers head loading conditions that cannot be replicated in the laboratory. *Therefore, our proposed research will be the first to evaluate the HIT System's head impact detection and impact location measurement accuracy in both laboratory and on-field environments in the same study* (**Figure 2.1: Aim 2**). This innovative approach to evaluating the HIT System's impact location measurement accuracy will lead to stronger confidence in the system's head impact location measurements made from impact loading scenarios that apply more directly to its intended application on live football athletes. In addition, we will further our understanding of head loading conditions that lead to the HIT System classifying impacts as valid and relevant for consideration during special teams. Researchers can make stronger conclusions on head impact location dependent injury risk with head impact locations that are accurately measured by the HIT System during the loading environment experienced on special teams plays. Understanding head loading conditions that lead to missed impacts, misclassified impacts by the HIT System validation algorithm, and misclassified head impact location on play types associated with a high concussion risk, we hope technology improvements are made to address its limitations.

C. APPROACH

C.1. Aim 1. To evaluate the HIT System's accuracy at measuring impact location against gold standard reference sensors coupled to cadaveric human heads while undergoing laboratory-controlled drops.

C.1.1. *Introduction.* Head impact sensors need to be rigidly coupled to the skull to accurately measure head impact biomechanics and infer brain deformations. Accurately quantifying head impact biomechanics will help develop accurate concussion injury risk curves, associate directional dependent head loading injury risk and clinical outcomes, and quantify the potential role of repetitive head loading on long-term neurological health. *The objective of Aim 1* is to assess the HIT System's head impact location measurement accuracy in biofidelic laboratory testing environments. To attain this objective, *we will test the hypothesis* that the impact location coordinate mean spherical error between the HIT System and the gold standard reference sensors will be higher from drops onto the facemask location as compared to drops onto the helmet shell. We will test our hypothesis by using the *approach* of an existing dataset where we performed laboratory-controlled drops of human cadaveric heads wearing a HIT System instrumented football helmet with gold standard reference sensors coupled to the human cadaveric head. We dropped the cadaveric heads onto helmet impact locations commonly seen in American football from three separate drop heights. For this proposal, we will compare the head impact location measurements made by the HIT System to the head impact location measured by the gold standard reference sensors rigidly coupled to the cadaver head. *The rationale for Aim 1*, is that successful completion of the proposed research will quantify the strengths and limitations of one of the most used head impact sensor systems, the HIT System, at measuring head impact location while coupled to a biofidelic cadaver human head. Head impact locations accurately measured by the HIT System can provide confidence for researchers establishing impact location concussion injury risk based clinical outcome relationships. Identifying HIT System limitations can lead to refining the system's algorithms calculating head impact location or provide justification for considering other rigorously tested head impact sensors for measuring impact mechanics and locations.

C.1.2. *Preliminary Studies and Feasibility.* Measurement error can be introduced due to relative motion between the sensor and the head when head impact sensors are loosely coupled to the head.[28] Our collaboration with Siegmund et. al.[11] showed that the HIT System had less than 15% head impact kinematic measurement error at various helmet impact locations, compared against gold standard reference sensors mounted inside a Hybrid III ATD head (**Table 2.1**).[37–39] However, the Hybrid III ATD head's skin is composed of a vinyl layer that has a higher coefficient of friction than human skin.[1,2] This higher coefficient of friction provides unrealistic coupling conditions with the HIT System encoders. Therefore, the HIT System's measurement error may be increased when the encoder is in contact with a human head, which has a lower coefficient of friction than the Hybrid III ATD head. Members at Duke University as part of our collaborative research group performed human cadaveric head drops to provide a more biofidelic interaction between the HIT System encoder and the head. We evaluated the accuracy of the head impact kinematics measured by the HIT System against gold standard reference sensors. Our preliminary results in **Table 2.1** indicate the HIT System's accuracy at measuring head impact kinematics depends on the drop location, and potentially coupling rigidity to the head. In **Table 2.1** we observe that the absolute error is within 10% at the front oblique right, occipital, and vertex locations between evaluations done with a cadaveric head compared to the Hybrid III ATD head. However, the absolute errors are greater than 10% at the facemask, frontal, and parietal right locations. The HIT System's proprietary algorithms use the linear acceleration measured by six separate uniaxial linear accelerometers in the encoder to estimate impact location.[30,40] Incorrect measurements made by the accelerometers in the HIT System can lead to incorrect impact location determination. With sensor coupling a determinant for head impact biomechanics accuracy, it is important we not only quantify the accuracy of the kinematic measurements made by the widely used HIT System, but also the HIT System's impact location accuracy. Understanding the HIT System's strengths and

Table 2.1: Preliminary evaluation of the HIT System , unpublished Campbell et. al., 20XX, using a cadaveric human head shows that the absolute error of the peak resultant linear acceleration measured by HIT System compared to gold standard reference accelerometers depends on impact location, and also sensor coupling. This is shown by comparing our absolute errors to our previous collaboration with Siegmund et. al., 2016, where they used a Hybrid III dummy head. Linear acceleration measurement error made by the HIT System can propagate into incorrect impact location measurement, which we will investigate in Aim 1.

	Siegmund et. al., 2016: HIT System evaluation using a Hybrid III Dummy Head	**Campbell et. al., 20XX (unpublished): HIT System evaluation using a Cadaveric Human Head**
Impact Location	**Peak Resultant Linear Acceleration Absolute Error (%) and standard deviation**	**Peak Resultant Linear Acceleration Absolute Error (%) and standard deviation**
Facemask	137 ± 57	25 ± 16
Front Oblique Right	9 ± 8	10 ± 10
Frontal	13 ± 12	58 ± 72
Occipital	35 ± 31	31 ± 28
Parietal Right	6 ± 5	20 ± 22
Vertex	37 ± 18	30 ± 27

limitations at measuring head impact biomechanics will further our understanding of direction-dependent head loading to short and long-term neurological consequences of concussion and repeated head impacts.

C.1.3. ***Research Design.*** We will use an existing database derived from laboratory-controlled free drops with cadaveric human heads wearing a HIT System instrumented football helmet, to evaluate the head impact location accuracy measured by the HIT System compared to gold standard reference sensors rigidly coupled to the human cadaveric skull. We used three separate cadaver heads and impacted them at six locations on the helmet, from three drop heights. We will quantify the accuracy of the head impact location measurements made by the HIT System in two ways: (1) the average impact location difference in spherical coordinates of azimuth and elevation between the HIT System and the gold standard reference sensors; (2) the location category agreement between the HIT System and the intended drop location (i.e. frontal and facemask drops correspond to the front category measured by the HIT System).

C.1.4. ***Laboratory Drop Experimental Setup.*** Our research group concurrently evaluated common head impact sensors used to collect head impact biomechanics, by coupling these sensors to human cadaver heads and dropping the helmeted cadaver heads from multiple heights and onto different locations. We compared the peak head impact kinematics measured by the head impact sensor systems to the gold standard reference sensors coupled to the human cadaveric skull. The following sections will describe the data acquired for this project evaluating the HIT System's head impact location measurement accuracy.

C.1.4.1. ***Human Cadaver Heads and Instrumentation.*** Three fresh frozen male human cadaver heads were disarticulated at the atlanto-occipital joint from the rest of the neck. The mandible remained coupled to the skull and the heads were sealed at the occipital condyles with polymethylmethacrylate. Two of three heads (ID: D3, mass = 3.4 kg, circumference = 55.2 cm; ID: D5, mass = 3.6 kg, circumference = 58.0 cm) wore a large Riddell Speed helmet (Riddell, Elyria, Ohio, USA) during the drop testing. The other head (ID: D4, mass = 4.5 kg, circumference = 59.0 cm) was fitted with an extra-large Riddell Speed helmet for the drop testing. The large football helmet had a mass of 2.2 kg and the extra-large football helmet had a mass of 2.3 kg. We rigidly coupled a reference sensor block to the skull's occipital bone posterior to the foramen magnum with wood screws. The sensor block included the gold standard reference sensors: three single-axis linear accelerometers (model 7264B-2000, Endevco Corp., San Juan Capistrano, CA) and three single-axis angular rate sensors (model DTS ARS 8K, Diversified Technical System Inc., Seal Beach, CA), enabling measurement of six degrees-of-freedom (DOF) head kinematics. Voltage signals from the reference sensors were filtered with a hardware anti-alias filter at 25 kHz and sampled at 100 kHz.

C.1.4.2. ***Head Impact Telemetry (HIT) System.*** We installed a HIT System Speed MxEncoder in the available space within the crown of one large and one extra-large Riddell Speed Helmet (**Figure2.2a and 2.2b**). Our research group has used the HIT System extensively to research *in vivo* head impact biomechanics relationships to clinical outcomes,[10,35,36] establish positional, event-type, play type, and closing distance head impact biomechanics differences,[6,8] and explore visual sensory performance and functional movement quality on reducing severe head impacts.[41–43] The HIT System measured the linear acceleration of the cadaver head using six single-axis spring-loaded linear accelerometers (**Figure 2.2a**). The HIT System

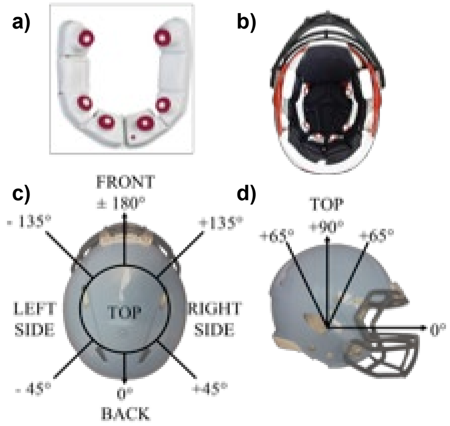


Figure 2.2: The HIT System encoder **(a)** fits into Riddell football helmets **(b)**, and uses 6 linear accelerometers and proprietary algorithms to measure head impact location in azimuth degrees around the head **(c)**, in elevation degrees **(d)**, and as an impact category. Impacts > 65° elevation are classified as Top impacts regardless of the azimuth degree calculation. Otherwise, impact location category is determined by the azimuth degree falling into one of the 4 location bins (Front, Right Side, Left Side, and Back).

collected data for 40 ms (8 ms pre-trigger, 32 ms post-trigger) at 1 kHz when any of the accelerometers detected accelerations exceeding a user-programmable threshold (14.4 g) and wirelessly transmitted the data to a laptop Sideline Response System. A proprietary algorithm then determined peak resultant linear acceleration at the head center of gravity (COG) from the raw linear acceleration signals. The algorithm determined the head impact location two ways: (1) in degrees of azimuth and elevation, and (2) as a category (**Figure 2.2c and 2.2d**).[29,30,40] All data were date and time stamped and exported from the HIT System's Redzone data cloud.

C.1.4.3. ***Drop Testing Protocol.*** We impacted the helmeted cadaver heads using a drop test methodology used before by our research group.[44,45] The helmeted heads were placed into a fine mesh net and the net was hoisted to one of three desired drop heights using a nylon line (**Figure 2.3a and 2.3b**). Before each drop, the mid-sagittal planes of the head and helmet were aligned and we positioned the top of the helmet opening 2.5 to 4 cm (1 to1.5 inches) above the cadaver's brow. The helmeted heads were positioned within the net to achieve one of the six desired drop locations (**Figure 2.3c**). The head was released into freefall by burning the nylon line. The helmeted head fell onto an aluminum plate with a tri-axial load cell (Kistler 9067, Kistler Instrument Cop., Amherst, NY) located beneath the plate sampling at 100 kHz. A 78.1 N vertical ground reaction force, measured by the load cell, triggered data acquisition of the reference sensors for 660 ms with a 100 ms pre-trigger. After every drop, the helmet was inspected to ensure the chinstrap had not moved, no hardware had come loose, and that no part of the system was damaged. The reference sensor block was examined at the end of each cadaver series to confirm that the

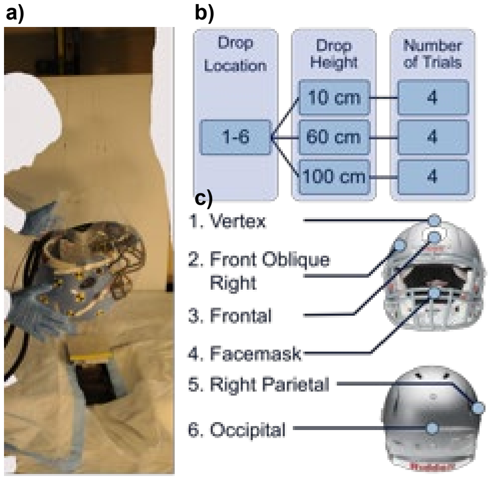


Figure 2.3: Drop Locations and Setup. Helmeted cadaver heads were inserted into fine mesh nets as part of our drop setup (a), hoisted to one of three drop heights (b), and dropped onto one of our six drop locations (c)

coupling had not changed during testing. We performed drops onto each location in blocks, and drop heights were performed in ascending order. A total of four trials were performed at each combination of three drop heights and six locations for a total of 72 drops per head, and an overall total of 216 drops.

C.1.5. *Data Processing and Reduction.* We will determine the head impact location from the gold standard reference sensors by transforming the head impact kinematics measured by the reference sensors to the cadaver head's COG. This involves demeaning linear acceleration data and rotational velocity data acquired from the reference sensors, using a four-pole Butterworth low pass digital filter with a 1650 Hz (CFC 1000) and 300 Hz (CFC 180) cutoff frequencies on the reference linear acceleration and rotational velocity data, respectively.[46,47] We will numerically differentiate rotational velocity data with a five-point stencil method to acquire reference rotational acceleration.[48] A micro-CT scanner (Nikon XT H 225 ST; Nikon Metrology Inc., Brighton, MI) will image each helmeted cadaver head, and anatomical measurements using Avizo 3D visualization software (Avizo 9.4; Thermo Fisher Scientific, Hillsboro, OR) will be made to determine the cadaver head's COG and the reference sensor's orientations. The origin of the head coordinate system will be defined as the midpoint between the porions lying in the plane made by the two external auditory meatuses and the left orbitale (i.e. Frankfurt plane; **Figure 2.4**).[49] The head coordinate system is defined with the positive x-axis emerging through the front of the head within the Frankfurt plane, the positive y-axis emerging through the left external auditory meatus in the Frankfurt plane, and the positive z-axis emerging through the top of the head normal to the Frankfurt plane. After determining the cadaver head's COG, we will transform the reference kinematic signals by rotating them to the head coordinate system and projecting the linear acceleration to the head COG using the equation for rigid body transformations.[50] Head impact location will be calculated from the transformed reference sensors block in terms of azimuth and elevation by using the x, y, and z components from the peak resultant linear acceleration vector pointing towards the head COG. We will match the time stamp for each drop from the reference sensors to the HIT System's time stamp to merge head impact location data and align azimuth, elevation degrees, and impact location according to **Figure 2.2c and 2d**. Any drops where the HIT System did not trigger will be left blank and not matched with the time stamp from the reference sensors. Mean impact directions and standard deviation ellipses for the reference sensors and the HIT System will be calculated. The azimuth and elevation direction data for drop locations about the mean should be asymmetrically distributed with a Kent distribution.[11,51,52] We will report the major and minor semi-axes for standard deviation ellipses which should contain the mean impact direction for the reference sensors and the HIT System. We will calculate mean impact directions, and standard deviation ellipses in Matlab (R2017b, MathWorks, Natick, MA) using SPAK library functions.[52]

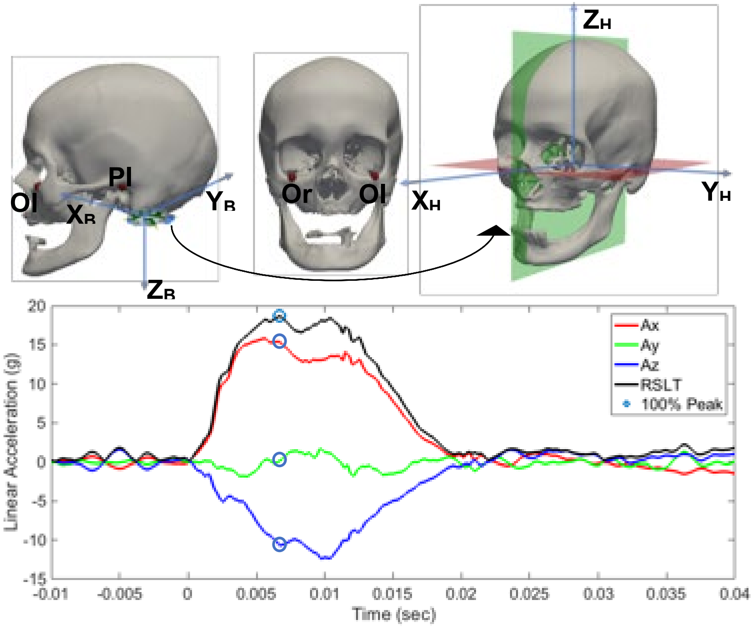


Figure 2.4: Reference block transformation to head COG process. The right and left orbitales and porions (Or, Ol, Pr, and Pl) will be measured along with the block sensor orientations (X_H, Y_H, and Z_H) to establish the Frankfurt plane and head center of gravity for our cadaver heads. Block sensor signals will be transformed to the head coordinate system (X_H, Y_H, and Z_H). The components at the peak resultant linear acceleration will be used to calculate impact location azimuth and elevation.

C.1.6. ***Statistical Analysis.*** To evaluate the head impact location accuracy measured by the HIT System against the gold standard reference sensors we will analyze the head impact location data in two ways: (1) We will assess the mean spherical error at measuring azimuth and elevation impact location coordinates between the HIT System and the reference sensors. The HIT System will not be significantly different at measuring azimuth and elevation impact location coordinates than the reference sensors, if the 95th percentile confidence ellipse of the mean spherical error at each drop location includes the origin (i.e. mean spherical error azimuth and elevation coordinates of zero).[51] (2) We will assess the percent agreement in impact location category (i.e. Front, Front Oblique Right, Top, Back, Right Side, and Left Side) between the HIT System and the reference sensors.

C.1.7. *Expected Outcomes.* We expect the HIT System will not provide similar azimuth and elevation head impact location coordinates compared to the gold standard reference sensors coupled to a cadaver head. However, we expect that a high percentage of the head impact locations measured by the HIT System will be contained within a standard deviation ellipsoid determined by the reference sensors. We expect that the HIT System will measure head impact location category in high agreement with the reference sensors for the impacts to the helmet shell, but have little agreement with the reference sensors for impacts to the facemask.[12] Head impact location is an important biomechanical input influencing loss of consciousness duration and behavioral outcomes in animal models,[19,20] and brain tissue strain in finite element models.[53] These investigations used impact location categories, which are coarser impact location measurements than azimuth and elevation coordinates, to investigate directional loading dependency on injury severity, behavioral, and brain strain outcomes. It is unknown how a ten-degree azimuth change in impact location corresponds to injury severity, behavioral outcomes, and brain tissue strain. To make stronger relationships between head impact location and clinical outcomes following a concussion in human subjects, we need a head impact sensor system like the HIT System to measure higher resolution impact location categories. For example, an impact to the front oblique location incorporates head translations and rotations along and about multiple axes that can lead to greater magnitude brain strains. In comparison, frontal or top impacts produce head translation and rotation along and about one axis leading to lesser brain strains.[53] HIT System head impact location measurement accuracy needs to be quantified using a cadaveric head surrogate because head impact sensor coupling is one important factor for measuring accurate head impact kinematics and location.[28,54] Aim 1 of our proposal will quantify impact locations that the HIT System measures using a biofidelic representation of the head. After quantifying the HIT System's impact location measurement accuracy at various helmet sites, future research can develop stronger impact location and injury risk/clinical outcome relationships. Understanding the relationships between impact location and injury risk and clinical outcomes can lead to improvements in protective equipment, identifying athletes for technique improvement to limit head impact exposure and concussion risk, and develop data derived rule modifications for reducing concussion risk.

C.1.8. ***Potential Pitfalls and Alternative Approaches.*** While we hypothesize that the HIT System will have a high agreement with the gold standard reference sensors coupled to the cadaver head with impact location category, we do expect that the azimuth and elevation impact coordinates measured by the HIT System will be significantly different than the azimuth and elevation impact coordinates measured by the reference sensors. We do not expect this to negatively impact the outcomes from this proposal. Currently, azimuth and elevation impact location coordinates are used to bin impact location into categories of front, top, back, right side, and left side impacts. The current state of research linking head impact location to concussion injury risk and clinical outcomes can largely depend on general head impact locations and likely does not require impact location resolution as measured by azimuth and elevation degrees. If we need azimuth and elevation specific impact location coordinates in order to understand impact location dependent injury risk and clinical outcomes, then we could test for any systematic differences in azimuth and elevation between the gold standard reference sensors and the HIT System. Understanding any systematic differences in azimuth and elevation between the HIT System and the references sensors can allow researchers to develop correction algorithms allowing for better coordinate measurements made by the HIT System.

Our drop test method does not allow for rotation about a fixed point, like the neck, which is observed during head impact testing using a linear impactor method and an ATD head and neck.[11,12] Our drop test method does allow us to deliver centroidal impacts through the cadaver head's COG. This represents an ideal condition for the HIT System's linear accelerometers within the encoder to contact the head and calculate head impact location. Establishing the accuracy of the HIT System at calculating head impact location in this ideal condition represents the first step before introducing more complex coupling interactions that can be created with eccentric impacts either in a laboratory setting on in an on-field setting.

C.2. Aim 2. To verify laboratory performance of the HIT System at measuring the number and location of head impacts in an on-field setting through video analysis of special teams plays during high school football games.

C.2.1. *Introduction.* The on-field football environment offers complex head loading conditions such as glancing head impacts and multiple impacts over very short durations that cannot be replicated in a laboratory environment. *The objective of aim 2* is to assess the HIT System's head impact detection and location measurement accuracy in an on-field environment during high school football special teams plays. To attain this objective, *we will test the hypothesis* that the HIT System will detect more impacts and have a higher percentage of agreement on impact location with a video observer from impacts directed onto the helmet shell than glancing impacts to the helmet shell or to the facemask location. We will test our hypothesis by using the *approach* of video confirming impact detection and location from American high school football players competing in games on special teams plays with HIT System instrumented football helmets. Impacts observed in video and measured by the HIT System will be classified according to **Table 2.2** and a Kappa agreement analysis will determine the head impact location agreement between the HIT System and a video observer. *The rationale for aim 1* is that successfully completing the proposed research will quantify head impact loading conditions

Table 2.2: Definitions used to classify impacts measured by the HIT system and impacts observed through video

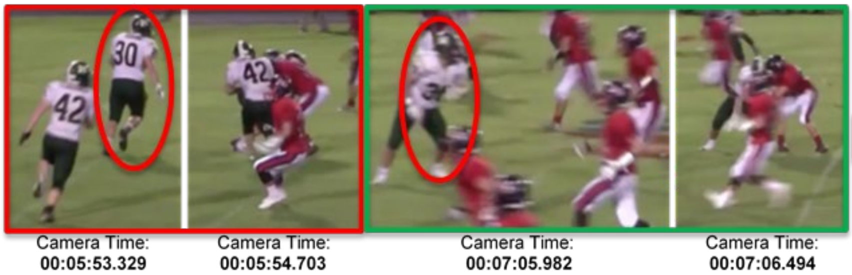
	Impact Visually identified Through Video Observation	Impact appears in HIT System Redzone Data Export
True Positive	Yes	Yes
False Positive	No	Yes
True Negative	No	No
False Negative	Yes	No

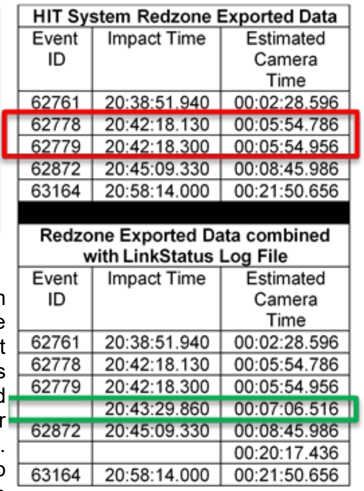
leading to undetected, invalid, and incorrect head impact locations determined by the HIT System and its proprietary algorithm on special teams plays, which are associated with increased concussion injury risk. Understanding these conditions leading to undetected head impacts and incorrect head impact location classification can lead to improvements for the HIT System's software algorithms for determining true head impacts and accurate head impact locations occurring during special teams plays. Accurate head impact frequencies and locations measured during special teams plays can provide confidence for rule modifications and legislations aimed at protecting football players during this dangerous play type.

C.2.2. ***Preliminary Studies and Feasibility.*** Our research group has used the HIT System to gather large quantities of head impact biomechanical data to understand intrinsic [41,42,55–57] and extrinsic[6,8,43,55,58–60] severe head impact exposure risk factors and to understand acute clinical concussion outcomes.[35,36,61] Using the HIT System to investigate concussion injury risk factors provides direct objective head impact biomechanical data that is free from retrospective data recall bias that may be present from medical professionals reporting to injury surveillance programs.[62,63] However, the strength of head impact exposure injury risk factors and relationships to clinical outcomes and player safety hinges on accurate head impact frequencies, locations, and magnitudes measured by the HIT System. The HIT System uses proprietary algorithms to provide users with datasets they need to answer research questions related to head impact exposures in football. The X2 Biosystem's xPatch uses similar algorithms to provide users with head impact exposure datasets. Recent work using head impact confirmation through video has shown that the xPatch's algorithm can remove true head impact loading events while keeping spurious head loading events that come from running and jumping.[64–66] Studies using the HIT System used video to correlate head impact biomechanics with play types, and closing distances, or to exclude non-head impact high magnitude events.[6,67] However, no study has evaluated the effectiveness of the HIT System's algorithm to filter out non-head impact events and retain true head impact loading events using video observation. Our research group collected head impact biomechanics from local high schools, as part of a previous study, along with high definition video. From our local high schools using the HIT System, we obtained LinkStatus Log files that tracked every data collection triggering event by a HIT System encoder before impacts were filtered by the HIT System's algorithms. The data are then made available in the HIT system's online data cloud called Redzone. We have determined that 43% of the loading events triggering data collection by the HIT System sensors were filtered from the datasets exported by Redzone. We do not know the characteristics of these impact events that are filtered from the Redzone dataset. However, some of the impacts that were filtered from the Redzone exported dataset could be relevant for understanding concussion injury mechanisms and informing rule modifications to better protect athletes **(Figure 2.5)**. With our head impact biomechanics dataset and high definition video we can better understand the HIT System's algorithms for retaining data in the Redzone exported data and the data that are filtered from the exported data during special teams plays. This video analysis approach will extend on our work in Aim 1 as we have the resources to quantify the head impact location agreement with video observers in an on-field setting.

C.2.3. ***Research Design.*** We will use head impact biomechanical data collected from high school football players wearing HIT System instrumented football helmets while competing in special teams plays during a season of games videoed by our research team. The video will be used to evaluate the head impact detection and location accuracy measured by the HIT System in an on-field setting. A video observer will verify the number and location of impacts to players during kickoff/ kick returns and punt/ punt returns with game video synchronized to the HIT

System measurements. We will compare the impact distributions of detected impacts classified as valid by the HIT System's proprietary algorithms to impacts that were detected but classified as invalid that do not appear in the data export available to HIT System users. A Kappa agreement analysis will assess the HIT System's impact location measurement accuracy against a video observer.





HIT System Redzone Exported Data		
Event ID	Impact Time	Estimated Camera Time
62761	20:38:51.940	00:02:28.596
62778	20:42:18.130	00:05:54.786
62779	20:42:18.300	00:05:54.956
62872	20:45:09.330	00:08:45.986
63164	20:58:14.000	00:21:50.656

Redzone Exported Data combined with LinkStatus Log File		
Event ID	Impact Time	Estimated Camera Time
62761	20:38:51.940	00:02:28.596
62778	20:42:18.130	00:05:54.786
62779	20:42:18.300	00:05:54.956
	20:43:29.860	00:07:06.516
62872	20:45:09.330	00:08:45.986
		00:20:17.436
63164	20:58:14.000	00:21:50.656

Figure 2.5: Example of the HIT System misclassifying a true head impact. The HIT system uses proprietary algorithms to provide users with clean datasets through their Redzone could-based software. However, impacts are removed from the Redzone exported data that could be relevant to injury risk. The player circled in red experienced a concussion in this game. The figure boxed in red shows an impact captured through video that corresponded to the HIT System Redzone exported data boxed in red on the right. Just over a minute later the same player sustained an impact (image in green) that lead to removal from the game. No data from the Redzone export corresponded to this impact with the closest time stamp occurring 3 minutes after the impact in red. Extracting data from the LinkStatus Log file showed that the HIT system triggered during the impact in green and corresponded to an impact time (green box on right) occurring a minute after the impact in red. Unfortunately, the algorithm filtered this data from the Redzone data export.

C.2.4. *Head Impact Biomechanics Data.* Our research group collected head impact biomechanics using the HIT System (see **section C.1.4.2** for details) from four local high schools in the last three years as part of a National Center for Injury Prevention and Control funded project entitled "Innovative Behavior Modification Strategies to Reduce Mild Traumatic Brain Injury Risk in High School Football Athletes" (the BeMod Study). Eligible participants required either a Riddell Revolution, Speed, or Speed Flex helmet to accommodate the HIT System sensor and were members of the high school football team. Consented and parental assented participants' helmets were instrumented with the HIT System prior to the beginning of their competitive season. Athletic trainers at each school set up the HIT System for data collection for all practices and games over three seasons. Each time the Athletic trainer created a session to collect head impact biomechanics using the HIT System, a sideline computer created a LinkStatus Log file that tracked when the session started, stopped, and any communication that occurred between the HIT System sensor in the helmets and the sideline data collection computer. During a session, such as a practice or game, the LinkStatus Log files contained a record of when a player's helmet experienced a loading event that triggered head impact kinematic data collection. Each head loading trigger event, within the LinkStatus Log File, created a date/time stamp for the trigger event and an identification number associated with that player's specific sensor. No head impact kinematic data were written into the LinkStatus Log record for a data collection triggering event. After a session, the HIT System uploaded the collected data to cloud-based software that filtered the collected data by proprietary algorithms, and these filtered data were made available to users through the online database, Redzone. From Redzone, users specified date ranges to export head impact biomechanics data for different analyses. For the purposes of this proposal, we will leverage the head impacts and head impact locations collected during special teams plays by one of our four teams, Team two, in the final data collection year (2017). Team two provides us with the highest number of instrumented participants, game impacts, and game videos to investigate the HIT System's ability at detecting impacts and correctly measuring head impact location during special teams plays (**Appendix A Table A1**). We will investigate the impact trigger events and impact locations measured during kickoff, kick return, punt, and punt return plays. The proportion of concussions high school football players experience during special teams plays are 86% higher than during run and pass plays while on offense or defense.[7] The combination of short closing distances between players on the line for punts and longer closing distances between players on kickoffs represents different helmet loading conditions that we believe will influence the HIT System's impact detection and location measurement accuracy. These impacts include Redzone exported head impact biomechanics data merged with impact trigger events stored within the LinkStatus Log files that were created for each football game. We will use a custom Matlab script to merge data exported from Redzone with data contained within the

LinkStatus Log Files. The script organizes the merged data by date and time stamps. The LinkStatus Log file contains impact trigger events that appear in the Redzone exported data. The script will remove impacts that have duplicate date/time stamps and sensor identification numbers but keep impacts that have duplicate date/time stamps and different sensor identification numbers. No head impact kinematic data is associated with trigger events from the LinkStatus Log files, but head impact kinematic data can be made available to the user through communication with Simbex, the company that created and oversees HIT System products. Simbex can validate trigger events on the backend that are discarded by the algorithm and make them available for download from Redzone. Head impact kinematic data will be available for analyses once discarded trigger events are re-validated.

C.2.5. ***Video Capture.*** Our research team filmed the games of the four local high school football teams participating in the BeMod Study. We used high-definition video cameras (Canon VIXIA HF M30 & R100 video cameras; Canon Inc., Tokyo, Japan), and recorded high definition film from the side of the field at the highest vantage point in the stadium allowing for the best view of the playing field. The games were recorded at 60i (60 interlaced fields per second) with a tight angle allowing for better resolution of the player's individual actions on the field and head impacts (**Figure 2.6**). Our research team filmed all games with a continuous feed. Each time the camera began recording, filmers displayed their cell phone with the current time down to the second (example, 18:45:23). This allowed our research team to easily synchronize the film with the HIT System impact data. This synchronization process is **detailed in section C.2.6**. Games where we had two instrumented teams competing allowed for an additional camera angle, usually on the other sideline. We will standardize our video evaluation approach and only use one camera angle that provides the highest vantage point to observe impacts on the field. All research team members responsible for filming games underwent training prior to each team's first game, usually during the football team's preseason scrimmages. We had multiple members of our film team collect film across the three years for consistent game film quality. In the 2017 football season, we acquired just over 27 hours of game film for Team two, capturing a total of 10,783 head impact data collection trigger events. Only video associated with special teams plays will be used for this project.

Figure 2.6: Example of the tight camera angle for video analysis. Our filmers used high definition cameras and a tight angle to provide high resolution video seen on the left as opposed to the traditional camera angle obtained for coaching purposes seen on the right. These close-up angles will allow us to determine the head impact location through video evaluation for comparison against the impact location measured by the HIT System.

C.2.6. ***Synchronizing Video to Head Impact Biomechanics.*** Previously, our research group synchronized competition video with head impact biomechanics to examine the role of play types,[59] closing distances,[6] collision types, and anticipation level,[55,68] on head impact biomechanics in high school and collegiate football, and youth ice hockey. Using the same technique for this project, video analysis will occur in VLC media player (version 2.2.8) with the Jump to Time VLC extension (version 2.1) to acquire camera playback times with millisecond resolution. The head impact biomechanics-video synchronization procedure uses the time stamp generated from a cell phone in the camera's view recording a game. As the cell phone increments to the next whole second, the camera time associated with this event is entered into a camera time column, which is created within the HIT System LinkStatus Log file merged dataset. Camera times associated with each impact within the merged dataset will be created using simple math within excel. The time stamps from the cell phone were not synchronized with the real time clock used by the HIT System sideline computer to create the time stamps for each head impact during a session. Therefore, we improve the accuracy of the camera times associated with each impact by determining a camera time associated with a video observed impact. We adjust the camera times for each impact based on the difference between the video observed impact camera time and the phone camera time. This head impact biomechanics-video synchronization procedure allows for accurate identification of head impacts within 500 milliseconds of the camera time. The procedure incorporates steps outlined by the National Institute of Neurological Disorders and Stroke common data elements for video confirmation of biomechanical devices used in traumatic brain injury research (**See Appendix B Item 2**).

C.2.7. ***Video Evaluation on Head Impact Detection and Location.*** We will use a video assessment questionnaire, created in Qualtrics Surveys, to evaluate the HIT System's accuracy at detecting impacts and measuring impact location category compared to a video observer. This questionnaire incorporates elements from previous video assessment questionnaires used in our lab for evaluating the effect of play types,[59] closing distances,[6] collision types, and anticipation levels on head impact biomechanics.[55,68] We will use two approaches for evaluating head impact detection and impact location through video assessment. The first approach uses the video as a gold standard proxy. Video observers will document camera times when an observed player's head receives large enough contact to trigger data collection on the HIT System.[69] Observed impacts must meet all inclusion criteria outlined in **Table 2.3** before further video

Table 2.3: Trigger event inclusion and exclusion criteria for video review

Inclusion Criteria for a suspected trigger event	Exclusion Criteria for a suspected trigger event
1. Player must be on the field	1. Player is not on the field
2. Player must be within the camera view	2. Player is not within camera view
3. There must be a clear, unobstructed view of the player's helmet not blocked by another player	3. View of the player's helmet is partially or fully obstructed by another player
4. Clear evidence of helmet contact is observed – Prior to the suspected trigger event there was separation between tlshe struck and striking impact objects and a clear view of the struck and striking objects contacting one another was observed	4. No clear contact site is observed on the helmet – Contact site is away from camera view
5. A clear HIT System impact location is observed on the player's helmet and can be confidently assigned	5. A clear HIT System impact location cannot be confidently determined
6. The impact site can be clearly and confidently discriminated between facemask or helmet shell	6. The impact site cannot be discriminated between facemask or helmet shell
7. The impact centricity can be clearly and confidently determined	7. The impact centricity cannot be clear and confidently determined

analysis. Otherwise, the observed impacts are documented as excluded and not analyzed further. Video observers will log the camera times to video as a gold standard proxy data collection form. Observers will be blinded to the head impact kinematics and impact location in the combined head impact dataset **described in section C.2.4.** Details about the head impact loading conditions will be documented according to **Table 2.4.** The second approach for evaluating impact detection and location accuracy uses the head impact events recorded by the HIT System sensor as the gold standard proxy.[69] Video observers will enter in camera times associated with an impact event from the Redzone exported dataset for an observed player into VLC. Observers will then enter the impact identification number associated with the impact they are observing on film to a sensor as a gold standard data collection form and complete the details related to **Table 2.4**. Impacts analyzed with this approach must also meet all the inclusion criteria outlined in **Table 2.3** with the exception of item four. This approach maximizes the number of samples for comparing video observed impact location category against the HIT System measured impact location category. We will analyze impact trigger events occurring during kickoffs, kick returns, punts, and punt returns Our pilot work demonstrated that 13 trigger events met our inclusion criteria for unobstructed and certain impacts during special teams

Table 2.4: Head impact video verification questionnaire. List of questions from the head impact video verification questionnaire to characterize the type of loading occurring for a given impact under review.

Head Impact Verification Questionnaire	**Possible Choices**
1. What is the play type?	kickoff, kick return, punt, punt return
2. What is the player's position at the trigger event?	kicker, punter, blocker, returner, defensive player
3. What impact type caused the trigger event?	head to head, head to body, head to ground, head to object
4. Who/what did the player collide with?	teammate, opponent, ground, object, self
5. What was the player doing at the time of the trigger event?	blocking, tackling, being tackled, being blocked, recovering fumble, head hit ground, hit own head, another person
6. What was the closing distance between the observed player and the item causing the impact?	long (> 10 yards between observed player and colliding object) short (< 10 yards between observed player and colliding object)
7. Was the trigger event normal to the helmet or tangential (oblique) to the helmet	centric non-centric (Oblique)
8. Was the impact onto the helmet shell or onto the facemask?	facemask shell
9. Which impact location would the HIT System assign?	front back right left top

plays. We anticipate at least 156 impacts will meet our inclusion criteria for statistical analysis by extrapolating our pilot work over video from 12 separate games.

C.2.8. ***Video Rater Reliability.*** We will establish our video rater's accuracy and reliability at evaluating football game video during special teams plays similarly to our previous work.[6,59] Our video rater's accuracy and reliability at evaluating video will be determined for the following: impacts meeting our inclusion criteria, impacts leading to HIT System data collection triggers, video observed HIT System impact location category, shell or facemask directed impacts and impact centricity. We will use multiple evaluation sessions and multiple rater approaches. In all approaches, video raters will be blinded to the impact accelerometry and location measurements made by the HIT System to avoid rater bias.[59] We will determine the intrarater reliability on a subset of video containing 100 impact trigger events. The subset will be evaluated on two separate sessions no less than 30 days apart by the same reviewer.[6,59] An additional video reviewer will evaluate the same 100 impact trigger event video subset to determine the interrater reliability. A Kappa agreement analysis will determine the intrarater and interrater reliabilities. We expect the agreement will fall within an acceptable range ($0.60 < k < 1.0$) due to the strict definitions and instructions used for video evaluation.[6,59,70]

C.2.8. ***Data Reduction.*** For the impact detection analysis, trigger events from the combined head impact dataset will be merged with the impact details analyzed by the video as a gold standard proxy questionnaire with the trigger event camera times common to both datasets. For the impact location analysis, impacts from the Redzone dataset will be merged with the impact details analyzed by the sensor as a gold standard proxy questionnaire with an impact identification number common to both datasets. Only impact trigger events meeting all our inclusion criteria will be merged (**Table 2.3**).

C.2.9. ***Statistical Analysis.*** We will categorize observed trigger events through the video as a gold standard proxy questionnaire and impacts measured by the HIT System according to **Table 2.2, from section C.2.1.** This approach evaluates the HIT System's performance at detecting impacts in a game setting during special teams plays. The HIT System's impact detection sensitivity, specificity, positive predictive value, and accuracy will be calculated. We will describe impact condition differences (impact centricity and shell/facemask directed impacts) between trigger events classified as true positives, false positives, false negatives, and true negatives. We will use an unweighted Kappa agreement analysis to determine the head impact location agreement between a video observer and the HIT System. An unweighted Kappa statistic of one indicates perfect agreement between video observer and the HIT System. We will describe the head condition differences (impact centricity and shell/facemask directed impacts) between impacts classified with correct impact location and impacts with incorrect impact location classifications. We will set an a priori alpha level of 0.05.

C.2.10. ***Expected Outcomes.*** We expect helmet loading to the facemask and glancing impacts to the helmet shell will be missed by the HIT System or lead to impacts that are removed through the system's impact filtering algorithms. We expect that similar helmet loading conditions will lead to incorrect head impact location calculation by the HIT System as compared to video observers. It is important to understand how often these loading conditions occur that lead to inaccurate head impact detection, incorrect algorithm classification, and location measurements during special teams plays so that injurious and non-injurious data are consistently captured by improving the HIT System algorithms used to measure head impact biomechanics. Addressing limitations with the HIT System can lead to understanding head impact exposure and location relationships to injury risk and clinical outcomes. Generating further evidence for these relationships may lead to better protective equipment, identifying athletes for technique improvement to limit head impacts and concussion risk, and developing data-informed rule modifications for safer special teams play in high school football.

C.2.11. ***Potential Pitfalls and Alternative Approaches.*** We expect to characterize the estimated 156 impacts that meet our inclusion criteria proposed for Aim 2's analysis to understand the HIT System's accuracy at measuring head impact frequency and location to high school football players competing in special teams plays during games. The 156 impact trigger events sufficiently power Aim 2 outlined in **Section D**. In the event we fall short of characterizing our required impact sample size of 88 impacts, we will expand our video analysis to the 2015 and 2016 seasons on Team two. Another potential issue is that we are limited to one camera angle for 10 of the 12 games we recorded. We do not anticipate this to limit our ability to categorize and describe impacts occurring to players on the field. Our research team recorded games from the highest available point on the sideline at the center of the field. This view is preferred, over other camera angles such as those obtained from the end zone as most player movements in football are along the length of the field rather than the width.

D. OVERALL ANALYSIS. Adequate sample size for Aim 2 is estimated with an a priori power analysis. We require 88 impacts meeting our inclusion criteria in **Table 2.3** in order to have 80% power that an obtained HIT System impact detection sensitivity of 95% is different from an ideal sensitivity of 99%. All other analyses for Aim 2 should be sufficiently powered based on our estimate for powering HIT System impact detection sensitivity (**Table 2.4**).

Table 2.5: Proposed statistical analysis summary for proposal aims.

Research Question	**Variables**		**Statistical Analysis**
	Independent	**Dependent**	
Aim 1: To evaluate the HIT System's accuracy at measuring impact location against gold standard reference sensors coupled to cadaveric human heads while undergoing laboratory-controlled drops.	Reference Sensors Mean Impact Location Coordinates (degrees of azimuth and elevation)	HIT System Mean Impact Location Coordinates (degrees of azimuth and elevation)	Mean Spherical Error
	Reference Sensor Impact Location Category	HIT System Impact Location Category	Percent Agreement

Research Question	Variables		Statistical Analysis
	Independent	Dependent	
Aim 2: To verify laboratory performance of the HIT System at measuring the number and location of head impacts in an on-field setting through video analysis of special teams plays during high school football games.	**Aim 2 Impact Detection (number of impacts) Analysis**		
	Not applicable	True Positives, False Positives, False Negatives, and True Negatives (see **Table 2.2** outcome definitions)	Descriptives
	Not applicable	HIT System Impact Detection Sensitivity, Specificity, Positive Predictive Value, and Accuracy	Descriptives
	Aim 2 Impact Location Analysis		
	HIT System Impact Location Category Determined by Video Observer	HIT System Impact Location Category	Percent Agreement and Kappa Agreement Analysis

E. SUMMARY AND FUTURE DIRECTIONS With the frequency and severity of concussions in football and the widespread use of the HIT System, it is important to understand the measurement strengths and limitations of the system in a testing environment that more closely mimics the interaction of the sensor on a human head rather than an unrealistic head-sensor coupling environment provided by an ATD. Accurate data are needed to ensure that data derived rule changes are in the best interest for player safety. We will use two approaches that supply a more realistic coupling environment to evaluate HIT System measurement accuracy. First, we will perform laboratory-controlled drops to cadaver human heads wearing a HIT System instrumented football helmet. These drops will evaluate the head impact location accuracy measured by the HIT System compared to gold standard reference sensors rigidly coupled to the human cadaver heads. Next, we will build on our results in the laboratory setting and determine the HIT System capabilities in the field setting where impact conditions are more complicated. This will be accomplished by establishing and using a highly intrarater and interrater reliable video impact assessment on impacts experienced by American high school football players competing on special teams plays in games with HIT System instrumented football helmets. Achieving the aims of this project will support our hypothesis that sensor head coupling is an important determinant in the HIT System's ability at making in-field measurements. With the HIT System being the primary head impact data collection tool in football, it is important to describe the strengths and weakness of the system to solidify existing relationships related to head impact exposure and data derived rule modifications made to special teams plays. More importantly, weaknesses of the HIT System need to be addressed through design modifications or algorithm improvements. Future research can evaluate other emerging head impact sensor technologies that couple more rigidly to the human body. This in an effort to identify an optimal measurement tool for collecting head impact biomechanics in sport to allow more informed safety modifications and clinical suggestions.

F. TIMELINE. We anticipate the proposed project will take 4 months and **Table 2.5** provides a timeline for the activities involved in this proposal. April includes analyzing our existing dataset to address Aim 1, preparing data, and beginning video evaluation for data to address Aim 2. Video evaluation for Aim 2 will continue through May and June with manuscript preparation beginning for Aim 1 during this time. June will be dedicated to data analysis for addressing aim 2 and manuscript preparation. Data from this proposal will be disseminated in 2 manuscripts (Aim 1 and Aim 2) at the end of the study period.

Table 2.6: Project timeline. The proposed project will take 4 months to complete

Task	2019															
	April				**May**				**June**				**July**			
	Week				**Week**				**Week**				**Week**			
	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16
Aim 1																
Data Analysis	■	■	■	■												
Manuscript Preparation					■	■	■	■	■	■	■					
Aim 2																
Data Preparation	■	■	■	■												
Data Collection	■	■	■	■	■	■	■	■	■							
Data Analysis									■	■	■					
Manuscript Preparation									■	■	■					
Dissertation Defense														■		

CHAPTER 3

Literature Review

Introduction

Sports offer an environment for researchers to study concussions by measuring external biomechanical forces that are transmitted to the head and the effects of head impacts over a lifetime of sport participation.[4,8,71–73] These transmitted forces cause the head to undergo a combination of linear and rotational acceleration,[4,74,75] and the proportion of linear and rotational acceleration experienced by the head is dictated by the head impact location and direction of head loading.[5,76,77] Animal experiments that manipulated head directional loading causing different proportions of linear and head rotational accelerations demonstrated that head injury risk, loss of consciousness duration, and behavioral outcomes are related to head loading direction.[19,77,78] However, concussion injury risk and clinical correlates to head loading direction has not been demonstrated through human epidemiological data.[62,63]

Head impact sensors have allowed researchers to quantify head impact frequency, location, and magnitude (head impact biomechanics) in many different sports across multiple skill and age levels.[8,59,71,72,79–82] The HIT System has been used extensively in football.[8,59,79] Concussion injury acceleration risk, [23,83,84] differential head impact exposures to playing positions,[8,85,86] extrinsic and intrinsic severe head impact risk factors, [6,41,42,87,88] and the relationship between brain structural and cognitive function and head impact biomechanics have been quantified using the HIT System.[89–92] Importantly, college football players instrumented with HIT System helmets experienced larger magnitude impacts during special teams plays.[6]

This is due to most impacts experienced during special teams plays occur from larger closing distances, which allows for higher impact velocities than during run and pass plays.[6] These data helped to inform rule modifications to limit the number of kickoff return opportunities and in turn reduce concussion injury risk at the professional and collegiate levels. However, with only few high school kickers making it to the collegiate level, most kickers at the high school level lack the strength and technique to consistently kick the ball through the endzone to limit kick return opportunities and the chance for a concussion. Therefore, kick returns occur more regularly and so does concussion risk. Continued observation of kickoffs and punts at the high school level is warranted in order to modify and create rules that help reduce concussion injury risk. Head impact sensors that collect head impact biomechanics related to injurious and non-injurious impacts already informed rule changes at the professional levels.[6] However, the strength of these data derived rule modifications hinges on the HIT System's measurement precision to provide accurate data. The HIT System has been rigorously validated in a laboratory setting while on a Hybrid III ATD headform, with larger measurement error reported for testing conditions that mimic the on-field loading environment observed in football.[11,12,54,93] In addition, differences exist in the head/helmet fit and frictional interface between the Hybrid III headform and human head, which will influence the HIT System's measurement accuracy.[15,54,94,95] This warrants the need for additional evaluation of the HIT system using human surrogate head models (e.g. cadaver) that are more representative of the on-field environment.[28,96–98]

Not only do head impact sensors need to provide accurate head impact biomechanics data, but their algorithms need to provide datasets that include true head impact events experienced by the subjects wearing the sensor, with non-head impact events removed. Recent work with head impact sensors other than the HIT System demonstrated that high percentages of non-head impact events can be retained, and true head impact events can be mistakenly removed by the sensor's algorithms.[28,65,66,99–101] By cross-validating impacts measured by head impact sensors with video, research has shown that higher frequency signals within the measured kinematics are associated with non-impact trigger events and lower frequency signals are associated with true head impact events.[99–101] Currently, we do not know the characteristics of impact trigger events that are retained or removed by the HIT System algorithm. This is significant because the HIT System is the most widespread and heavily used research tool to quantify head impact biomechanics in football. If we fail to properly characterize injury risk and the relationships of head impact exposure to neurological health, then we cannot properly protect athletes with data derived rule modifications from potentially inaccurate instrumentation and software.

The purpose of this literature review is to provide a comprehensive appraisal of the content matter pertaining to the proposed project. This content includes: the rationale for studying the HIT System accuracy in a high school football setting during special teams plays; reviewing the contributions of linear and rotation acceleration on head injury risk and clinical outcomes, and how head directional loading influences these parameters; the limitations of data acquisition tools and methods outside of using recently developed head impact sensors; a thorough description of the design and function of the HIT System; a review of the results and parameters influencing HIT System measurement accuracy in laboratory validation tests; describing the fit and frictional interface differences that exist between human and ATD heads; and a review of on-field impact detection studies on head impact sensors outside of the HIT System.

Concussion Epidemiology in High School Football and Special Teams Plays

High school athletes represent one of the largest athletic cohorts in the United States with almost 8 million participants annually.[102] An estimated 1.1 to 1.6 million concussions occur to children under 18 years old, and 15% of all the sport-related injuries occurring to high school athletes are sport-related concussion.[17,103] It is important to research the health and safety of this population given the reasons previously stated in addition to ongoing neurocognitive development that occurs throughout adolescence.[104] Injury surveillance and epidemiological studies described sport-related concussion incidence in an effort to identify those at risk for the injury and how to focus concussion injury reducing interventions. Overall, high school athletes experience 2.5 to 5.1 sport-related concussions per 10,000 athletic exposures (AE), where one athletic exposure is defined as one athlete participating in one practice or competition for any amount of time.[105–108] These concussion incidence estimates are conservative because many concussions can go unreported due to injured players thinking that concussions are not serious enough to require medical attention.[109] Contact and collision sports like lacrosse, ice hockey, and soccer consistently have high concussion incidences that on average range from 1.7 to 6.5 concussions per 10,000 AE depending on the sport and the year the data were collected.[105–108] Over 1.1 million participants compete in high school lacrosse, ice hockey, and soccer annually, indicating that epidemiological studies are still required for this population to identify injury prevention and reducing interventions.[102] Football, however, has an equivalent number of annual high school participants with consistently high concussion incident rates ranging from 4.7 to 9.4 concussions per 10,000 AE over the last 15 years.[102,103,105–108,110,111] Continued efforts are needed to document concussion injury risks and trends within high school football so that injury risk reduction methods can be put in place during a critical developmental phase of life.

Football's high collision environment provides the opportunity to research concussion injury mechanisms due to players frequent exposure to head impacts and the large number of diagnosed concussions across youth, high school, collegiate, and professional populations.[8,108,112–114] Concussions occur up to seven times more frequently in games compared to practices, and player to player contact is the primary injury mechanism causing concussions across all football populations.[102,103,105–108,110,111] Reducing or eliminating contact during practices is a viable method for reducing concussion injury risk during practice scenarios but it has little effect on concussion injury risk during games.[115,116] Instead, limiting contact sessions in addition to ensuring proper football tackling and blocking techniques for player to player contact are methods that can reduce concussion risk in youth and high school football populations.[115]

In addition to regulating contact and teaching proper playing techniques, modifying existing or introducing rules that limit the potential for player collisions, and other uncontrolled high energy contacts, is another method that may lower the incidence of concussion during competition.[117,118] Special teams plays, kickoffs and punts, commonly lead to high energy player collisions due to the large closing distances between opposing players on the field.[6] The rules for special teams football plays at the professional level have undergone a number of modifications in the last 20 years to better protect players from high energy collisions that cause concussion.[119,120] From 1996 to 2007 approximately 21% and 7% of all concussions occurred on kickoff and punt plays respectively, but the kickoff and punt play type represented the highest concussion risk per 1000 plays among pass and run play types.[119,121] Similar concussion percentages of 20% and 10% are observed on kickoffs and punts, respectively, at the high school level and special teams plays represented a significantly higher risk for sustaining a concussion compared to injuries occurring during run and pass plays.[7] Some of the kickoff rule changes at the professional level included protecting defenseless players from unnecessary helmet impacts during kickoffs and punt returns, moving the kickoff from the 30 yard line to the 35 yard line to limit the number of kickoff returns, repositioning the players' setup locations on the kickoff and kick return units prior to the kick, prohibiting running starts for the kickoff unit, and limiting field locations where contact occurs between the kickoff and kick return units.[119,120] These rule modifications to the kickoff led to a 5% drop in the percentage of concussions at the professional level from the 1996 to 2001 observation period compared against recent 2016-2017 data.[27,121] Therefore, rule modifications have a role in reducing concussion injury risk especially during high energy collision events on special teams plays.

Some of these rule modifications at the professional level have made their way to the collegiate and high school football levels.[122] The kickoff location was one of the notable modifications to special teams plays because data collected from instrumented football helmets on collegiate players helped inform this modification.[6] The data demonstrated that collisions occurring on special teams plays over long closing distances were the most severe while collisions occurring on special teams and defensive plays over short closing distances resulted in the least severe impacts. Moving the kickoff location increased the number of touchbacks in professional football by 32% the year after moving the kickoff line to the 35-yard line and limited the opportunities for players to receive concussions on kickoffs.[120] Ivy league collegiate football teams experimented with moving the kickoff to the 40-yard line and this modification lead to 8.88 fewer concussions sustained for 1000 kickoff plays.[122] The football is also kicked from the 40-yard line in high school kickoffs. However, with only few high school kickers making it to the collegiate level, most kickers at the high school level lack the strength and technique to consistently kick the ball through the endzone to limit kick return opportunities and the chance for a concussion. Therefore, kick returns occur more regularly than what would have been hoped for with kickoffs from the 40-yard line. Continued observation of kickoffs and punts at the high school level is warranted in order to modify and create rules that help reduce concussion injury risk. Head impact sensors that collect head impact biomechanics related to injurious and non-injurious impacts already informed rule changes at the professional levels.[6] However, the strength of these data derived rule modifications hinge on a head impact sensor's accuracy at measuring the head impact biomechanics related to non-injurious impacts and impacts that cause concussion.

Concussion: How does it happen?

Sport-related concussions are caused by biomechanical forces that lead to brain tissue deformation causing short-lived neuronal injury or impairment, that presents with a wide range of clinical signs and symptoms.[4] Most individuals recover from their concussion symptoms and deficits spontaneously over seven to fourteen days.[4] However, a small subset experiences prolonged recovery, and suffer with symptoms and objective deficits for weeks to months after the injury.[123] Prolonged recovery can morph into post-concussion syndrome without properly resolving concussion deficits. The biomechanical forces leading to sport-related concussion are a result of direct blunt head trauma or indirect head loading where the mechanical energy from a blow to the body is transmitted through the neck to the head.[75,124–126] Regardless of direct or indirect head loading, the skull is set into motion and experiences linear and rotational accelerations in any combination of anatomical planes. The brain, however, lags behind initial skull motion because it is suspended in cerebral spinal fluid and meninges within the skull.[127–132] The lagging motion between the brain and skull interface, combined with the brain being one of the softest biological tissues in the body,[133] causes shearing stresses and strains to develop within brain tissue.[74,127,134–137] In severe traumatic brain injury and diffuse axonal injury, shearing stresses cause damage to the neuronal axons and the microtubule scaffolding of the neuronal axons.[134,135] Through animal models, compromised neuronal axons and microtubules leads to a neurometabolic cascade, upsetting the neuronal axon's ion balance and cellular homeostasis.[138] In an effort to restore the ionic balance and cellular homeostasis, an energy imbalance is created where the damaged neuron requires more energy to restore cellular homeostasis, but diminished cerebral blood flow cannot supply the energy demands to correct the ionic imbalance. Studies associating the pathophysiological changes following concussion to clinical symptom presentation still need to be conducted. However, from a speculative point of view, the potential pathophysiological changes following a concussive injury manifest immediately, or within 24 to 72 hours of the injury as observed in human studies.[4] In human's the pathophysiological changes related to concussion manifests as any combination of somatic, physical, or cognitive symptoms, postural instability, behavioral changes, cognitive impairment, or sleep/wake disturbances. On average these symptoms resolve within seven to fourteen days, with younger populations, aged five to twelve years, taking up to 28 days for clinical symptom recovery.[139]

Our current understanding of concussion injury biomechanics comes from measurements made in experiments with human cadaveric specimens,[131,132,140–142] animals,[5,143–145] video reconstruction,[24,25,146,147] and live human volunteers.[75,128,129,148] These experiments focused on the roles of linear and rotational acceleration and head loading direction on how they contribute to brain motion, injury risk, resulting pathology, or symptom presentation. The next sections will discuss how linear acceleration, rotational acceleration, and head loading direction, contribute to concussion injury risk and resulting pathology or symptom presentation.

Linear Acceleration and Concussion

Early studies examining the role of linear acceleration in concussion focused on correlating measures of peak linear acceleration of the head and intracranial pressure measurements to skull fracture risk through whole body and head only human cadaver impact experiments.[140,141] These studies showed that pressure gradients inside the skull increased with higher linear accelerations of the head. The pressures were greatest at the site of the impact while the other side of the skull, the countrecoup side, experienced a decrease in pressure. The differences in high and low intracranial pressures created a pressure gradient inside the skull. Early on, researchers hypothesized that this pressure gradient caused the brain to move in the skull and caused shear stresses to develop and injure the brain tissue.[149] This hypothesis was later rejected by using cadaver heads and examining them with high-speed x-ray to track neutral density tags inserted into cadaveric human brain tissue. The brain tissue displayed looping patterns of motion and translated less than 10 mm.[131,132] Coup and countrecoup pressure differences were created from the differential motion of the skull moving with the brain lagging behind. The looping motions are caused from the brain being tethered to the skull base by distal internal carotid arteries, optic nerves, olfactory tracts, oculomotor nerves, and the pituitary stalk.[131,132] Experiments with human cadavers were limited to quantifying skull fracture risk instead of brain injury and any resulting symptom presentation. In studies using canines, the dogs' brains were impacted/air blasted and pressure data were collected. These blasts caused concussions of varying severity, with severity determined through loss of consciousness duration and peak linear acceleration.[142] The Wayne State Tolerance Curve for Head Injury was created by combining the canine data with peak linear head acceleration data from volunteer human sled tests that resulted in no concussion or skull fracture.[75] Participants were placed on a sled and sent down a track whereby the head translation was measured when the sled came to rest. This tolerance curve was later used to generate the Gadd Severity Index (GSI) and Head Impact Criterion (HIC) by fitting the curve with exponential functions to describe skull fracture risk.[150,151] These severity indices are key in helmet protection standards to help reduce skull fracture risk. However, they have little predictive abilities in concussion risk.[22]

Early studies using human cadavers couldn't link peak linear acceleration measurements to symptomology. Studies that used animals to link peak linear acceleration to symptomology and pathology were limited to injuries resulting in loss of consciousness even though concussions can occur without a loss of consciousness.[4] Until human volunteers were instrumented while competing in contact/collision sports with a high likelihood for concussion, human volunteers were exposed to linear head accelerations that did not produce any diagnosed concussion nor symptomology presentation.[75,152] The National Football League's Mild Traumatic Brain Injury Committee established a concussion database with patient symptomology linked to head impact biomechanics.[76,147] The committee reconstructed 31 impacts in a laboratory using accelerometers rigidly coupled at the COG of ATD heads. They used the 50th percentile male Hybrid III ATD headform as a surrogate head to recreate impacts seen in the NFL to quantify the head impact biomechanics of struck and striking players that resulted in diagnosed or no diagnosed concussions. For the remainder of this document, any reference made to a Hybrid III ATD will be in reference to a 50th percentile male unless otherwise stated. The average peak linear head acceleration was 98 ± 28 g and 15 ms in duration for all struck concussed players and was significantly larger than struck players with no diagnosed concussion, 60 ± 24 g, and striking players, 56 ± 22 g.[147] While this study quantified head impact peak linear accelerations that caused concussion in professional football players, concussion risk could not be determined because of the small number of players that were struck and not diagnosed with a concussion. Linear acceleration is a useful biomechanical variable for determining concussion injury risk. However, linear acceleration magnitude and accelerational based severity indices, like the GSI and HIC indices, are related more for skull fracture risk than concussion injury risk. Linear acceleration is only one biomechanical variable to explain head motion following head loading, as rotational head kinematics occur following head loading.

Rotational Acceleration and Concussion

Isolated head rotational acceleration was hypothesized to be a more dangerous head motion than isolated head linear accelerations.[153] Isolated head rotations produce brain shearing deformation as compared to linear head translations producing brain compressive deformations. The ability of the brain to withstand shearing deformations is 10,000 times lower than compressive deformations resulting from linear directed impacts through the head COG.[133,153] However, our heads rarely experience linear and rotational accelerations in isolation because heads are attached to the body via a neck.[75,125,126] The neck serves as a rotation point for the head and in most cases, the head experiences a combination of linear and rotational accelerations following direct or indirect head loading. The initial research evaluating the effects of larger rotational acceleration proportions to linear acceleration on the brain was conducted on primates.[154] The probability of experimentally inducing concussion, which the authors defined concussion involving loss of consciousness, to Macaque monkeys increased when their cervical spines were not supported with an external brace. This produced larger linear and rotational accelerations of the head when it was impacted from a metal projectile, even at lower velocities, and increased the incidence of concussion. Researchers attributed this result to a larger amount of shearing strain and tensile strain transmitted to the cervical spinal cord of the monkey. However, when the monkeys were outfitted with a cervical collar, which reduced the rotational displacement of the head on the neck, the shearing and tensile strains to the cervical spinal cord were reduced, and the incidence of concussion was reduced even with high velocity impacts and larger linear accelerations of the head following impact. Gennarelli expanded on this work and clarified the roles of linear and rotational acceleration in producing concussion.[155] Squirrel monkeys received impacts to the head in the sagittal plane that produced predominately linear acceleration or predominately rotational acceleration of the head. All monkeys that received predominately rotational acceleration of the head received concussions, while monkeys that received predominately linear acceleration of the head did not experience concussions. The researchers proposed that rotational acceleration produced larger shearing strains and tensile strains in the cervical spinal cord and brain stem of the monkeys than did linear accelerations. From these primate experiments, researchers proposed a diffuse axonal injury threshold of 4500 rad/s^2 in sagittal rotation for angular speeds less than 30 rad/s,[156] and these proposed thresholds were increased to 8000 – 16,000 rad/s^2 in humans later by scaling primate data.[157] However, the scaled rotational acceleration thresholds for diffuse axonal injury exceeded the average peak rotational acceleration concussion threshold of 6432 ± 1813 rad/s^2 obtained from recreating concussion-inducing impacts to professional football players in the laboratory with ATD surrogates.[147] Through finite element modeling, we have observed that rotational acceleration magnitude has a larger effect on brain axonal strain than linear acceleration magnitude.[158,159] It is important to note that the accuracy of the outputs provided from finite element modelling depends on the accuracy of the material properties used in the model and the experimental datasets the finite element model is validated against.[160] There are large disparities in brain material properties, which are linked to the variations in the protocols from where the brain material properties are acquired.[133] Thus, the brain material properties used in finite element modelling should be oriented to the objectives of the modelling. A similar approach should be used for validating a finite element model against experimental data. A finite element model modelling brain tissue strain should not be validated against intracranial pressure experimental data. As long as finite element model creators consider these parameters for accurate model outputs, they should continue to serve as a powerful tool for understanding brain injury mechanics.[158,159]

The relative contributions of linear and rotational head motion influence concussion injury risk. Through animal experimental data, finite element modelling, and data acquired from laboratory recreated concussion causing impacts to professional football players, larger proportions of head rotational motion increases concussion injury risk. Therefore, head rotational acceleration should always be quantified in its relation to concussion injury risk.

Impact Location, Loading Direction and Concussion

From a mechanical perspective, brain tissue responses are dictated by the head impact location, force direction, and force magnitude.[128,129,131] Head impact force magnitude is observed through measuring head acceleration, which we discussed in the previous sections. Impact location refers to the site on the head where the impact force is applied, while impact direction refers to the angle or the line of action the impact force is applied at the impact location on the head. For example, an impact force applied to the forehead and directed along the anteroposterior anatomical axis would produce head motion in the sagittal plane, while an impact force applied to the forehead and directed along the mediolateral axis would likely produce head motion in the transverse plane. This subtlety between impact location and loading direction is important, because the brain's viscoelastic properties and stiffness, primarily white matter, is anisotropic or directional dependent.[133] Therefore, certain regions of the brain will be susceptible to deformation based on a specific loading direction as compared to other loading directions.

Humans can limit head motion in specific anatomical planes as compared to head motion in other anatomical planes.[75] Through sub-injurious sled deceleration tests, human volunteers in the testing exhibited lower head accelerations from sagittal plane oriented head loading as compared to higher head accelerations from coronal plane oriented head loading.[75]

Through cadaver head impact testing, lower magnitude head motions result in the whole brain moving as a rigid body within the skull and minimal local deformations or strain within the brain.[131,132,161] With higher magnitude head motion, the brain no longer moves as a rigid body, and higher components of strain make up a larger proportion of brain motion. Larger brain tissue strains and a higher proportion of brain tissue strain leads to higher brain injury risk.[162,163] Although it wasn't demonstrated in the sled deceleration tests, there is a potential for directional dependence in human brain injury risk. Experiments with primates, however, demonstrated directional dependence for brain injury risk and severity.[20,77,164] Primates experienced longer durations of unconsciousness from coronal plane head rotation as compared to sagittal plane head rotation for equal head rotation magnitude.[20,77,164] Although the skull, brain, and brainstem anatomy differ between bipedal primates and quadrupedal porcine; immature porcine subjected to sagittal plane head rotations demonstrated persistent behavioral deficits and more widespread axonal injury as compared to immature porcine subjected to coronal plane head rotation.[5,19] Finite element simulations of rapid rotations to immature porcine provided evidence that strain oriented along the white matter tracts were larger from sagittal and horizontal rotation as compared to smaller tensile strains experienced by white matter tracts during coronal plane rotation.[5] Humans are not only more resistant to head motion in the sagittal plane, but by extending the results from immature porcine white matter tissue to humans, our white matter tracts are also more resistant to deformation along specific head loading lines of action.

Head impact linear and rotational acceleration along with impact location and direction are biomechanical features that influence head injury risk and the resulting clinical presentation as seen through animal experimental testing.[5,19,20,77,164] Head injury risk and clinical presentation head loading directional dependence have not been observed consistently in humans.[62,63,165] It is difficult translating injury thresholds established from animal experiments to humans without properly scaling the acceleration inputs.[157] With respect to directional loading dependence, primate anatomy is more comparable to human anatomy than porcine and murine.

However, primate experimental testing is non-existent today due to ethical considerations. Porcine experimental models offer similar proportions of grey and white matter and a similar gyrencephalic brain structure to humans, but the skull, spine, and brainstem are oriented anteroposterior for the quadrupedal porcine as compared to the superior-inferiorly oriented skull, spine, and brainstem in the bipedal human. Murine experimental models are becoming frequently used in brain injury biomechanical studies due to their availability.[143,166–168] Murine, however, are quadrupedal, have lissencephalic brains, and have opposite inflammatory responses to brain injury compared to humans.[169] It would be ideal to study concussion injury risk and outcomes in humans, and collision sports, where head impacts occur frequently and a high number of concussions occur.

Head Impact Biomechanics in Football

Football is a collision sport that provides the opportunity to research concussion injury mechanisms due to players frequent exposure to head impacts and the large number of diagnosed concussions across youth, high school, collegiate, and professional populations.[8,108,112–114] Early research studying concussion injury biomechanics in football instrumented collegiate football players' helmets with bulky linear accelerometer instrumentation systems.[170,171] These systems consisted of accelerometers affixed to the head, as solidly as possible, at multiple locations with data telemetered to a receiver on the sidelines. Over 650 impacts were recorded in 30 games, with the peak linear accelerations ranging from 40 to 1000 g.[170,171] Concussions measured during competition ranged from 180 – 400 g.[170] Naunheim et al. used a triaxial accelerometer affixed to the vertex of the helmet adjacent to the football athlete's head. The peak linear accelerations measured ranged from 10 g to 120 g with an average of 29.1 g.[172] These early works proved the concept that head impact kinematics can be measured while football players compete, but these studies reported a wide range of linear head accelerations we have since learned are likely not plausible to occur in live participation. The accelerometers used in these studies were imbedded in the athlete's helmets or affixed to the head using headbands. However, the placement of the accelerometer in reference to the point of impact played a large role in the accelerations measured. The sensors would measure higher accelerations from impacts contacting the helmet closer to the accelerometer as compared to lower accelerations measured from impacts, despite a similar magnitude, located further away from the accelerometer placement.[173] As the placement of the accelerometer on the head or in the helmet played such a role in the data collected, an instrumentation system that could estimate head accelerations at the head COG would be needed to collect head impact biomechanics.

Rather than instrumenting players with accelerometers, analyzing video of impacts that caused concussions in professional football players has been used to research concussion head impact biomechanics.[27,147,174] Video observers have quantified closing distances, head impact locations, impact directions, and impact velocities of struck players receiving concussions from striking players.[147] The variables obtained through video analysis can be recreated in a laboratory with accelerometers placed at the ATD's COG. With this setup, impacts causing concussions to struck players had an average peak linear acceleration of 98 ± 28 g, an average rotational acceleration of 6432 ± 1813 rad/s^2, and were primarily located on the facemask and side locations.[24,76,147] However, these laboratory impact reconstructions based on professional football players were performed over 20 years ago. Professional football players are larger, faster, and stronger since 1996.[175] In addition to the morphological changes of the players, over 47 rule changes have been made since 2002 with the intent of reducing concussion.[174] Thus, the impact environment in the professional game has changed. Investigations recharacterized the professional football impact environment in 2010 and 2015 through video analysis.[27,174] Early results indicated that impacts located on the side of the helmet caused 47% of 547 reviewed concussion cases from 2010 to 2014, and 40% of 322 reviewed concussion cases from 2015 to 2017. In contrast, impacts to the helmet's facemask accounted for more than half of the 182 severe impacts considered for analysis 20 years ago.[76,147] Efforts are underway to quantify the head accelerations of the recently video analyzed concussion cases by using laboratory validated video analysis and multiple camera angles.[146] However, the head acceleration magnitudes determined from the video analysis will be limited to the professional game and harder to transfer to high school and youth football levels.

The National High School Sports-Related Injury Surveillance System (High School RIO) is an online injury reporting database for athletic trainers at high schools across the US. Athletic trainers submit reports on injuries of all types sustained in high school athletics. Head impact biomechanical data, specifically head impact location, can be acquired at youth and high school levels without instrumenting players with head impact sensors. This is accomplished by athletic trainers documenting concussion injury and reporting the injury head impact location to High School RIO and similar national injury surveillance programs. Using High School RIO data, impacts to the front of the helmet, which includes the facemask, caused 44% of the 2526 concussions sustained between 2008 and 2013.[62] From these epidemiological data we see that concussions sustained by high school football players differ in the head impact location as compared to professionals. Studies such as this illustrate the importance of widespread research across football levels in order to determine concussion head impact biomechanics. While researchers can collect a large amount of concussion head impact location data through injury surveillance, they are limited by the athletic trainer and the injured player accurately recalling the impact location. Their recollection of the impact location could be false. Thus, instrumenting players at many football playing levels allow researchers to collect large quantities of head impact biomechanical data. These data can be used to inform rule changes,[176] equipment enhancement,[177,178] and safer playing techniques to reduce concussion injury risk.[179–181]

Determining Head Impact Biomechanics in Football using the Head Impact Telemetry System

Engineers created the HIT System, a dedicated instrumentation system, to measure *in vivo* head impact biomechanics at the head COG of football players while they competed in practice and games. A HIT System encoder is comprised of six single axis spring-loaded linear accelerometers oriented normal to a player's head, a wireless telemetry unit, a battery, and an onboard storage unit mounted in a U-shaped encoder. The encoder can be inserted into the crown of Riddell style football helmets such as the VSR-4, Revolution, Speed, and Speed Flex helmets. Data are collected, telemetered to a sideline computer, and uploaded to a cloud-based web portal (called RedZone), where they are managed and made available for download. The spring-loading ensures that the accelerometers are in constant contact with a player's head so it measures head accelerations rather than helmet accelerations.[173] The six accelerometers continuously sample during a session until one of the single axis accelerometer channels exceeds a user programmable acceleration threshold. Typically the threshold is set to 10 g but a 14.4 g threshold has been used recently to better eliminate data capture triggers not related to a football impact, such as running or jumping.[182] Data are collected for 40 ms at 1000 Hz, and filtered with a 400 Hz onboard lowpass filter after a 0.5 Hz AC hardware filter removes any DC bias. To ensure that the entire waveform of the impact is captured, 8 ms of data are stored pre-trigger and 32 ms of data are collected post trigger with a 100 ms reset time after the impact to ensure separate impacts are collected. Data are time stamped (± 5 ms resolution) and wirelessly transmitted to a sideline receiver and laptop via a radiofrequency (903 to 927 MHz).

Impact data from the six accelerometers are processed with a proprietary algorithm that calculates the resultant linear acceleration and impact location.[93] Briefly, the proprietary algorithm uses rigid-body kinematic equations, known angles, and distance vectors from the six accelerometers to an estimated head COG based on a 50th percentile male head size. An optimization algorithm directly solves for the resultant linear acceleration magnitude and impact location in azimuth and elevation coordinates.[30,183] Rotational acceleration is calculated about two axes of rotation; rotation in the sagittal and coronal plane. This is calculated using the equations of motion for a force acting on a head from the anteroposterior and mediolateral components of the peak linear acceleration vector, and the relationship between linear and rotational acceleration acquired in the field from a 12 accelerometer HIT System capable of measure six DOF head kinematics.[9,29] Rowson et al. showed that the HIT System overestimated resultant rotational acceleration from on-field head impact data using their 12 accelerometer HIT System. This lead to the development of a correction algorithm for rotational acceleration in 2013 and any previously collected HIT System datasets can be exported from the Redzone impact database with this correction.[9,37,40] The correction algorithm accounted for the non-linear term that is part of the rigid-body kinematic equation. The non-linear term is assumed to be negligible during short duration impacts where no head rotational velocity is occurring prior to the impact. However, longer duration impacts and impacts where the head is rotating prior to the impact have a significant contribution to the non-linear term. On the football field, the head experiences some form of rotation prior to impact, necessitating this correction. The sideline laptop provides real-time data on the peak resultant linear and rotational acceleration, impact location azimuth and elevation coordinates, and three severity indexes, GSI, HIC, and the Head Impact Technology severity profile (HITsp).[83,150,151] The GSI and HIC are exponentially weighted integrals of the acceleration-duration impulse from the Wayne State Injury Tolerance Curve. The HITsp is a weighted principle component score quantifying concussion injury risk.[83] The metric uses an impact's peak linear acceleration, peak rotational acceleration, and duration as inputs and is weighted based on impact location.

Each time a HIT System user creates a session to collect head impact biomechanics, a sideline computer creates a LinkStatus Log file that tracks when the session started, stopped, and any communication that occurs between the HIT System encoder in the helmets and the sideline data collection computer. During a session, such as a practice or game, the LinkStatus

Log files contain a record of when a player's helmet experienced a loading event that triggered head impact kinematic data collection. The LinkStatus Log File provides a chronological record of each time that the HIT System triggers but does not save the kinematic data for each trigger event. Meaning these data collected with the trigger event need to be applied to the algorithm in order to determine two things. First, the trigger event had a peak linear acceleration greater the 10 g. Prior algorithm versions provided end users trigger events that were below 10 g of peak resultant linear acceleration. End users chose to remove the trigger events less than 10 g of peak resultant linear acceleration from their datasets for analyses as these trigger events were associated with running or jumping movements and not a head impact.[35,182,184] Updated filtering algorithms now only provide trigger events that have a peak resultant linear acceleration greater than 10 g. The second criterion for a trigger even to be validated by the algorithm is the acceleration pulse had characteristics of an impact to a helmeted head based on rigid body dynamics. Once the algorithm determines that an impact meets the two specified criterion it is available in real-time on the sideline computer and later via Redzone. From Redzone, users specify date ranges to export head impact biomechanics data for different analyses. As the algorithm removes data that does not meet the criteria, the LinkStatus Log files can give an indication as to which impacts are removed and retained within a final dataset following a session. Provided the user can synchronize the date/time stamps from the LinkStatus log files with any acquired video, Simbex may be contacted in order to release or validate the kinematics that the algorithm originally deemed spurious or invalid.

The HIT System is the most heavily used head impact sensor for collecting *in vivo* head impact biomechanics in football with approximately 80 peer-reviewed published articles. As of 2014, the HIT System measured almost 1.3 million *in vivo* head accelerations in football,[178] and this number is increasing as the HIT System is used across youth,[79,87,185] high school,[59,85,186,187] and collegiate football playing levels.[8,36,43,188] Research utilizing the HIT System in the field typically falls into six categories; 1) quantifying head impact biomechanics by playing position,[8,85,86] 2) determining extrinsic factors, such as offensive schemes or closing distances between players, for high magnitude or frequent head impacts,[6,88,189] 3) determining intrinsic factors, such as visual and sensory performance, for high magnitude or frequent head impacts,[41,42,87] 4) relating clinical measures to injurious and non-injurious head impact biomechanics,[36,165,190,191] 5) constructing concussion injury risk equations based upon head impact biomechanics,[23,83,84] and 6) uncovering role of repetitive non-injurious head impacts on brain structure and function.[89–92] The accurate data needed to establish relationships in these categories rely on three fundamentals from a head impact sensor like the HIT System: 1) the head impact sensor has accurate sensor technologies for measuring head kinematics, 2) the head sensor always triggers for data collection after receiving an input exceeding its data collection threshold, and 3) the head impact sensor has an impact processing algorithm that retains true head impact events and removes non-head impact events. Research on the level of retained or removed data by the HIT System has not been published; however, other systems have been evaluated on this metric. The X2 Biosystem's xPatch is a small accelerometer affixed to the skin behind an athlete's ear using adhesive. The xPatch uses similar algorithms to provide users with head impact exposure datasets. Recent work using head impact confirmation through video has shown that the xPatch's algorithm can remove true head impact loading events while keeping spurious head loading events that come from running and jumping.[64–66] Studies using the HIT System used video to correlate head impact biomechanics with play types and closing distances or to exclude non-head impact high magnitude events.[6,67] However, no study has evaluated the effectiveness of the HIT System's algorithm to filter out non-head impact events and retain true head impact loading events using video observation. Currently, we do not know the characteristics of impact trigger events that are filtered from the Redzone dataset. As mentioned earlier, the quality of our head impact exposure relationships hinges on correctly and accurately collected data.

Validating HIT System Measurement Accuracy in Laboratory

Head impact sensor accuracy is commonly evaluated by using a human head surrogate as the physical object that the device is connected to during a controlled impact experiment. These human surrogates serve as a replacement to the living human and include ATDs (crash test dummies) and postmortem human subjects/specimens – cadavers.[11,28,37,45,66,96,97,192–195] Both surrogate types use combinations of high fidelity linear accelerometers and gyroscopes, referred to as reference sensors, rigidly coupled within or around the skull.[38,46] The reference sensors serve as the gold-standard tool for measuring head impact kinematics following a loading event and provide the measurements to compare kinematics collected by other head impact sensor systems like the HIT System. In addition to the impact testing and surrogate head methodologies influencing the HIT System's measurement accuracy, there are a number of within testing parameters, such as head impact testing location and impact energies, that can influence measurement accuracy.[11,13,54] The HIT System in different designs has been put through a number of testing methodologies evaluating its measurement accuracy in the last decade.[11,37,196–198] This section provides an overview on the validation studies quantifying the HIT System's linear and rotational acceleration and impact location measurement accuracy, starting from the proof of concept studies to more biofidelic rigorous evaluations.[11,30] The section after will focus on the experimental testing considerations influencing impact location and impact detection accuracy by the HIT System, as these are the primary outcomes of interest from the HIT System being investigated in this dissertation.

Early work validating the HIT System established that engineers could use an arrangement of arbitrarily placed single axis linear accelerometers rigidly mounted around a hemispherical object (idealization of a head) and use rigid body equations of motion to estimate head acceleration at the "head" COG.[30] The head coordinate system for the hemisphere used a right-handed coordinate system. Azimuth angles were defined from -180° to +180° with 0° at the positive x-axis and positive azimuth angles to the right of the x-axis. Elevation was defined from

0° in the horizontal plane at the base of the hemisphere to 90° at the crown of the sphere. A computational analysis showed that an array of eleven accelerometers provided within 5% error in linear acceleration of the gold-standard as compared to a six accelerometer system which performed marginally worse, but still within 7% error. While eleven accelerometers provided less error, the lower costs and less space required from using six accelerometers instead of eleven was worth 2% error tradeoff for collecting head impact linear acceleration data.[30] Further experimental testing involved embedding (rigidly attaching) the six linear accelerometer array into a metal hemisphere with a tri-axial reference linear accelerometer at the center and dropping it onto 33 different locations around the metal hemisphere. The six linear accelerometer array provided peak linear acceleration measurements up to 5% error, up to 5° azimuth error, and 15° elevation error compared to the reference tri-axial linear accelerometer.[30] The algorithm was adapted to solve for peak rotational acceleration and was validated with the six single axis linear accelerometers rigidly coupled to a Hybrid III ATD headform.[29,199] The results obtained from the rigidly mounted accelerometer array and algorithm yielded peak linear and rotational head acceleration with average relative errors of 0.2% and 2.5%, respectively. These preliminary investigations provided initial evidence that the accelerometer hardware, arrangement, and software algorithms could measure peak linear and rotational head acceleration and impact location within a reasonable error level given an ideal situation where the accelerometers were rigidly mounted to a metal surface representing the human head. A limitation of this assessment was that the validation did not consider in-the-field real-world environmental factors relevant to accelerometer coupling to the head. Additionally, this validation work used sub-components of the eventual HIT System and not a complete in-field ready system, which incorporated spring loaded accelerometers to ensure head acceleration measurement and not helmet acceleration measurement.

The six linear accelerometer array was configured for insertion into football helmets, boxing headgear, or soccer headbands.[12,188,196–198,200] These accelerometer arrays could be attached to an ATD headform through each respective headgear mounting procedure, impacted, and measurements made by the HIT System could be compared back to a 3-2-2-2 array of linear accelerometers within the ATD's skull.[199] Impact delivery methods also tried to mimic the loading conditions from each respective sport with weighted pendulums to simulate football impacts, and ball cannons to simulate soccer headers.[198,201–203] ATD headforms wearing the football and boxing HIT System received impacts through weighted pendulums and were mounted to Hybrid III ATD necks to allow for head and neck dynamics that more closely replicated a 50th percentile human head and neck.[31,204,205] The football HIT System tested with headforms, to represent a more biofidelic testing condition, had similar peak linear and rotational acceleration relative error, ±4%, as compared to the proof of concept validations, and an average impact location error of 2.45°, showing that the system was capable of capturing similar measurement accuracies relative to the idealized metal hemisphere studies.[29,30,188] The boxing HIT System had similar accuracy to the football HIT system from pendulum impacts with overall root mean squared errors of 5.6 ± 2.6 g, 595 ± 405 rad/s^2, and 9.7° ± 5.2° for peak linear acceleration, rotational acceleration, and impact location, respectively.[198] The soccer HIT System demonstrated that linear and rotational measurement accuracy depended on the impact energies it was exposed to.[197] The soccer HIT System had coefficients of determination of 0.34 and 0.57 for peak linear and rotational head acceleration from measuring impacts low energy impacts delivered from a ball through an air cannon.[197] In contrast to the soccer ball impacts, the soccer HIT System measured impacts delivered by a linear pneumatic impactor more accurately with coefficients of determination of 0.89 and 0.90 for peak linear and rotational acceleration, respectively.[197] As seen in the soccer HIT System evaluations, it is important to expose a head impact measurement tool to a range of impact energies similar to the conditions seen in its intended sport use.

Development of a pneumatically driven linear impactor allowed researchers validating HIT System accuracy the ability to more closely replicate on-field head impact loading scenarios.[200,202] In this methodology, ATD head and neck assemblies were mounted to an adjustable sliding table, which had a combined mass of the Hybrid III head neck and upper torso of 23.5 kg. The sliding table allowed for translation of the ATD headform and neck assembly. Impacts were delivered by an impactor rod, which had a rigid plastic spherical cap similar to a helmet of a striking player, and it could be accelerated to speeds up to 13 m/s.[147] In addition to using linear impactor methodology to deliver impacts to HIT System instrumented helmets for validation, researchers applied skull caps, stretched stockings, or added human hair wigs to the ATD headforms in an effort to more closely replicate the head-sensor/ head-helmet friction interface.[11,12,37,54,196] This is an important feature to replicate as the measurement error by a head impact sensor depends on how rigidly it is coupled to the head, to reduce differential motion between the sensor and the head.[12,28,206] From the studies using linear impactor methodology and manipulating the head-helmet interface, the measurement error associated with peak linear and rotational acceleration and impact location were larger[11,12,37,54,196,197] than those reported in previous validation studies with drop tower methodologies or less biofidelic HIT System couplings to the head.[29,30,188] While some level of error is acceptable in terms of calculating population derived trends in head impact exposure, measurement error needs to be minimal in measuring single impact events associated with concussion head acceleration thresholds, or relationships to acute concussion symptomology.[11] Throughout the years of validation studies the number of impact locations used have increased, impact delivery methods have changed, more variable impact speeds have been used, design changes to the system were made, different surrogate head types have been used, and the sensor-head coupling interface has become more realistic to that of a live human. The following paragraphs will describe these considerations in influencing HIT System measurement accuracy on impact detection and location.

Parameters Influencing HIT System Impact Detection and Location Accuracy

Impact Location

Impact site selection for validation testing on HIT System instrumented helmets is one parameter that influences impact location measurement accuracy. In both the computational evaluations and experimental evaluations while developing the HIT System technology and algorithms, the impact location (measured in azimuth and elevation degrees) standard deviation error increased with increasing elevation coordinate drop sites on the simulated and experimental evaluations.[30] These observations showed that future validation testing of the HIT System needed to use impact locations that covered many sites on the helmet, not just a few locations. This is especially important given that 41 of 62 helmet-to-helmet impacts to struck players and 94 of 107 impacts delivered by striking players, in a video review of severe impacts in the NFL, involved contact with helmets above the level of the facemask on the helmet shell, approximately 45° in elevation.[76] In addition, a number of struck players that suffered concussions received impacts to their facemask.[76] This shows that including the facemask impact site in HIT System validation testing is important.

While the initial HIT System proof of concept study demonstrated larger standard deviation error for azimuth and elevation impact location measurement from impact sites higher in elevation, this observation has not been observed in laboratory validation testing. The HIT System had an average error of 2.45° from validation testing using a twin-wire-guided National Operating Committee for the Standards of Athletic Equipment (NOCSAE) drop test methodology,[188] but as impact delivery methodologies changed to better replicate the loading conditions observed on the field, the average impact location error has increased.[11,12,196,198] The HIT System had an average absolute location difference between the HIT system and reference sensors of 31.2° ± 46.3° from measuring impacts directed on the facemask.[11,12] This average absolute location difference dropped from 31.2° ± 46.3° to 13.2° ± 6.3° after removing the impacts directed on the facemask,[12] and less impact location measurement variability from impacts directed to the facemask has been observed elsewhere.[11] The HIT System algorithm solves for impact location based on the temporal relationship of peak acceleration values obtained from the six accelerometers.[30] While the resultant head acceleration corresponds with the head COG, the location estimate best approximates impact location on the helmet.[12] In most cases, the impact location and acceleration direction are the same, but there are occasions where impact location and acceleration direction are not the same.[12] This is the case with facemask impacts. Incorrect identification of acceleration direction for the facemask impact site is related to the facemask and impactor surface interaction. This interaction causes the helmet to initially move downward before head acceleration-mimicking motion occurs, which typically results from an impact to the back of the head. Large impact location measurement error made by the HIT System occurs from impacts directed onto the crown.[11] The HIT System reported 55 of 64 impacts to the crown as nearly opposite to the actual impact direction in comprehensive validation using a linear impactor.[11] Test video for crown impacts did not reveal obvious differences between impacts that elicited the incorrect impact location compared to impacts that measured the correct impact location, and the true reason for these differing observations remains unclear.[11] Other studies reported HIT System impact location accuracy but used boxing and hockey versions of the systems.[196,198] It is difficult to compare the results from these versions of the HIT System to the football system because the six linear accelerometers are oriented tangentially to the head in the boxing and hockey systems as opposed to normally in the football version.[29,30,196,198] A tangential accelerometer orientation directly estimates six DOF kinematics as opposed to five DOF in the football version. Similar to the football system, however, HIT System location measurement accuracy in the boxing and hockey system depended on the impact sites used in the evaluation.[196,198]

Impacts must be detected in order to be measured. As mentioned earlier, the HIT System uses proprietary algorithms to retain true impacts that cause peak resultant linear head accelerations greater than 10 g and removes impacts that do not cause the expected helmeted head acceleration kinematics through rigid body dynamics, such as a player ripping off their helmet and dropping it on the ground.[184,196] Studies did not start reporting the number of algorithm valid and invalid impacts delivered during validation testing until 2013.[54] The football HIT System ranged in correctly retaining 75% to 96% of data that were over 10 g of peak linear acceleration during validation evaluations.[11,13,54,196] There were no impact location trends for the algorithm incorrectly removing impact data, but the system had few correctly validated impacts that were delivered at speeds greater than 9.3 m/s.[11] This observation is concerning as the average closing speed between two professional football players in impacts that caused concussion was 9.3 m/s ± 1.6 m/s and large closing distances in special teams plays affords high impact velocities.[147] This means that the HIT System may be challenged at detecting impacts that cause a concussion and on special teams play types. The influence of impact speed or energy on the HIT System's impact detection and location accuracy will be discussed below.

Impact Delivery Method and Impact Speeds/Energies

As the laboratory methods changed to replicate the head loading conditions on the field, so did the performance of the HIT System at measuring impact location and detection. Following the computational and experimental testing to develop the HIT System technology, researchers used the NOCSAE twin-wire-guided drop test system to evaluate the HIT System's head impact location measurement error.[188] The estimated impact location measured by the HIT System was repeatable within ± 2.45° or approximately ± 1.20 cm of the drop site. The twin-wire drop system had difficulty reaching drop speeds greater than 9.3 m/s.[207] Therefore, the full range of impact speeds and their influence on impact location and detection measurement accuracy were not easily assessed. The twin-wire-guided drop system also has a limited rotational response.[207] On-field impacts are described by six DOF: three DOF describe the location of the impact on the helmet according to a three-axis frame of reference, and another three DOF describe the impact direction vector.[208] However, the twin-wire-guided drop method only allows for three DOF: two DOF describe the rotational ability of the headform within the sagittal and horizontal planes and one DOF describes the drop height.[208] Currently, the twin-wire-guided drop system only simulates centric impacts, which means the impact direction vector intersects the headform COG. This head loading type would ensure optimal coupling because the spring-loaded accelerometers within the HIT System can load directly into the head, and minimize relative helmet motion.[28,188] The NOCSAE twin-wire guided drop methodology does not include drops onto the facemask even though a large percentage of impacts occur to the facemask in football.[209] However, drops onto the facemask would likely lead to the helmet decoupling from the head. The helmet decoupling from the head would produce inadequate coupling between the spring-loaded accelerometers from the HIT System and the head, which would produce large impact location measurement errors observed in evaluations using linear impactor methodology.[11,12] A similar phenomenon would occur from impacts directed from negative elevations. The helmet would translate up off of the head and the HIT System accelerometers would decouple causing measurement error.[94] The minimal impact location error validation results provided promise for the HIT System early,[188] but the limitations associated with the twin-wire-guided drop system that was used for the evaluation positioned the HIT System for impact location measurement success.

Pendulum impactor methodology permits larger ATD headform rotational responses from impacts as compared to the twin-wire-guided drop method.[207] Impact speeds achieved using pendulum impactors are limited by the length of the pendulum arm.[207] To achieve impact speeds greater than 7 m/s, the pendulum arm needs to be greater than 4 m. This presents logistical issues when finding space to accommodate a large piece of pendulum equipment. Otherwise, the pendulum’s impactor face can be increased, or elastic tensioning pieces can be used to pull down the pendulum to achieve higher impact speeds. These modifications to the pendulum impactor introduce variability in the impact response which is less ideal for laboratory evaluations.[207] Pendulum impactor methods were used in boxing HIT System validation testing.[198] This testing showed that the boxing HIT System had a root mean square error of 9.7 ± 5.2° between the measured and expected location, and an increase in the measurement error compared to using the twin-wire-guided drop method.[188,198] No football HIT System validation testing used a pendulum impactor. Instead, the linear impactor system became the popular impact delivery method for football HIT System validation testing.[11,12,37,54]

The linear impactor method uses pressurized air to accelerate a 14.3 kg impactor ram towards an ATD headform system.[201,202] The headform system comprises of an ATD head and neck assembly mounted on a sliding table allowing for the headform system to translate following the impact from the impactor ram. This impact delivery methodology became the method of choice among researchers validating head impact sensor technology because of the number of impact sites that can be achieved,[11] the high impact speeds that can be delivered to the headform system,[11,12,37,54,147,207] the ability to deliver centric and non-centric impacts,[11,210] and the setup allows for headform system rotation. NOCSAE considered implementing the linear impactor to deliver rotational impacts as part of its new helmet testing standards. The higher impact velocities reached by linear impactors can cause helmet component issues that could decouple an instrumented helmet from the headform and delay the evaluation testing.[11,207] Two football HIT System validation studies and one hockey HIT System validation study used linear impactor methods to deliver impacts to HIT System instrumented helmets and quantify impact location accuracy.[11,12,196] The football HIT System had large overall absolute impact location differences of 31°± 46° and 42° ± 33° in validation studies using linear impactor methods.[11,12] In contrast, the HIT System's impact location measurement reported using twin-wire-guided drop methodology was ± 2.45°.[188] Currently, it is unknown if larger impact location differences measured between the HIT System and reference sensors are a result of the high impact velocities the linear impactor is capable of achieving, or if it is due to the ability to deliver non-centric impacts that glance the helmet shell and decouple the helmet from the head.[11,53,211] Similarly large azimuth measurement error levels of 4.0% ± 3.3%, 9.6% ± 5.1%,and

30.5% ± 15.2% were observed on a hockey HIT System evaluation that used a linear impactor delivering impacts to the back, side, and oblique back locations respectively.[196] With the linear impactor providing larger impact velocities, recent football and hockey HIT system validation studies documented that the HIT System algorithms removed impacts from the final dataset that truly occurred during the impact testing.[11,54,196] These impacts that the algorithm incorrectly removed corresponded to impacts tests delivered at greater than 9.3 m/s.[11] It is important that impacts are detected but also retained in the final dataset for analyses. Without knowing that these impacts occurred during the validation efforts, HIT System users would lose data which could contain injury-causing impact biomechanical data.

HIT System Design

From a design perspective, there are three distinct HIT System technologies. The first is the five DOF system where the six single axis linear accelerometers are oriented normal to a user's head.[6,8,35,36,41–43,57–59,61,68,89,188,212,213] The five DOF version is the most commonly used system and used in football helmets where there is limited space in the helmet to insert the instrumentation. The six linear accelerometers and normal orientation allow for three linear acceleration components, anteroposterior, mediolateral, and superior-inferior, and two rotational acceleration components, sagittal and frontal plane, estimates. The second technology is the six DOF system of six single axis linear accelerometers that are oriented tangential to the user's head. This version of the six DOF HIT System was applied to boxing and soccer headgear,[197,198,203] and hockey helmets.[55,56,60,72,196] The padding in these head protective gear were cut away and allowed for the necessary space to tangentially orient the six linear accelerometers. This allowed for full six DOF kinematic estimates, which included the degrees of freedom from the five DOF football system and the addition of transverse plane rotational acceleration. The third HIT System version is a six DOF device comprised of six bi-axial linear accelerometers (12 total accelerometers) oriented tangential to the user's head.[37] This version was used in football helmets.[214]

Only the five DOF football HIT system, the six DOF boxing HIT system, and six DOF hockey HIT System have validation studies that report head impact location and detection accuracy.[11–13,54,196,198] Based on limited evidence the six DOF tangential accelerometer orientation in boxing and hockey systems provided more accurate, overall root mean square error of 9.7 ± 5.2°and 4.0% ± 3.3% to 30.5% ± 15.2% error,[196,198] as compared to the five DOF normal accelerometer orientation used in football with overall absolute angle differences of 31° ± 46° and 42° ± 33°.[11,12] Further testing using the six DOF football HIT System and the five DOF football HIT System would be needed to determine any influence of accelerometer orientation on impact location measurement accuracy. There is one validation study with the six DOF tangential oriented accelerometers that report the impact detection rate.[196] However, no clear difference exists between the six DOF tangential accelerometer oriented HIT System, 19% of impacts incorrectly removed by the algorithm,[196] and the five DOF normal oriented HIT system, between 4% to 25% impacts incorrectly removed by the algorithm, on impact detection.[11,13,54] In the five DOF football HIT System, data was incorrectly removed by the algorithm from impacts delivered at speeds greater than 9.3 m/s in validation testing.[11] Future research should further look into the loading characteristics, such as peak linear accelerations or frequency content,[196] to understand why the algorithm is choosing to remove true impacts delivered to a HIT System instrumented football helmet.

Surrogate Head Type Used

Surrogate heads used in validation testing for head impact sensors use ATD headforms and necks,[11,12,188,192,194] or cadaveric human specimens.[45,96–98,195] The majority of head impact validation studies use ATD headforms and necks for their surrogate head choice,[11,12,188,192,194] but using cadaveric human head specimens as a surrogate head for impact testing is occurring more recently.[45,96–98,195] There are various ATD headform and neck designs, but the 50th percentile male Hybrid III headform and neck and the medium NOCSAE headform are the two ATD head and neck combinations used in HIT System validation testing.[11–13,37,54,188,196–198]

General Motors designed the Hybrid III headform in the 1970s for automotive safety testing.[31,32] Anthropomorphic measurements, facial landmarks, and inertial properties were selected for the headform and represented the average American male.[32,215,216] The Hybrid III headform and neck exhibit impact dynamic responses for both direct and indirect head loading through the neck attachment that mimics live or cadaveric head loading.[32] Head protection researchers at around the same time, worked on an ATD headform for testing athletic helmets known as the NOCSAE headform.[217] The NOCSAE headform was designed for conducting twin-wire-guided drop tests for football helmets. Size and shape specifications were based on a cadaver head that represented an average adult football player's head. The upper part of the neck was included and served as the coupling site between the headform and the twin-wire-guided drop apparatus. The skull deflection properties conformed with human cadaver skull deflection testing.[217] Custom modifications can be made to the headform that allow for attachment to a Hybrid III neck.[177,202,218,219]

Researchers used the NOCSAE headform in twin-wire-guided drop testing for evaluating HIT System impact location accuracy.[188] The NOCSAE headform and twin-wire-guided drop method allowed 25 separate impact location evaluations. The average impact location measured by HIT System in this impact methodology was repeatable within ± 2.45°, but this was the only study to report using the NOCSAE headform.[188] Otherwise, all other validation testing using the HIT System within head protection gear used the Hybrid III headform as their surrogate head.[11–13,37,54,196–198] Overall location measurement accuracy by the HIT System ranged from 9.7° ± 5.2° to 42° ± 33°.[11,12,198] Biomechanically, both headforms were developed and validated to supply realistic head impact responses. In terms of the shape characteristics between the Hybrid III and NOCSAE headforms, little difference exists in the upper portions of the headforms.[33] Due to their similarities, we could expect minimal differences in the HIT System's impact detection and location measurement accuracy between the Hybrid III and NOCSAE headforms with impacts delivered centroidal onto that upper portion of the helmet.

The differences become greater when examining the headforms in other areas. The jaw and cheeks are narrower in the Hybrid III compared with the NOCSAE headform, and the back of the Hybrid III headform does not extend all the way to the bottom edge of helmets.[33] These differences may lead to reduced head-helmet padding surface area and likely allow for more helmet rotation relative to the headform.[33] Due to these structural differences, relative movement between the Hybrid III head and a HIT System instrumented helmet would become more pronounced from impacts delivered lower on the helmet shell, impacts directed onto the facemask, or non-centric impacts that glance the helmet.[33] These loading scenarios would introduce measurement error in both impact magnitude and location, and also make it difficult for the HIT System algorithm to determine valid and invalid impacts. Using a head surrogate that closely replicates the shape of a human head is going to offer validation results in as close to a biofidelic environment for the HIT System's intended use on the field as possible.

To summarize, there are several testing methodology parameters that will influence validation results of the HIT System at measuring impact location and detection accurately. Impact sites used in validation testing should include various sites on the helmet shell and the facemask impact site. Specifically, the HIT System inaccurately measures crown and facemask impact sites, and a large proportion of impacts occur to these sites on the football field. The linear impactor is a recently preferred impact delivery methodology for HIT System validation tests. The linear impactor methodology delivers a wide range of impact energies, with the HIT System not detecting impacts at higher impact energies related to concussion. The impacting ram and sliding table with the ATD headform and neck allow centric and non-centric impacts delivered to the HIT System instrumented headform. Reported average absolute impact location errors are higher for validation studies using linear impactor methodology as compared to evaluations using twin-wire-guided drop methodology, which delivers centric impacts and permits three DOF head response dynamics. No clear difference exists among the different HIT System derivations (five DOF vs six DOF) on detection measurement accuracy. Lastly, minimal anatomical differences exist between the Hybrid III headform and NOCSAE headform on the upper portions of the headforms. However, the Hybrid III headform has a narrower jaw and cheeks, which reduces the contact area between the helmet and the headform allowing for head-helmet relative motion during impact and measurement error. Validation testing using a NOCSAE headform demonstrated smaller impact location error as compared to testing using a Hybrid III headform. Impact testing methodologies evaluating the HIT System measurement accuracy should closely reflect the loading environment observed on the football field to provide users the clearest information of the HIT System's measurement accuracy in its intended environment.

Helmet Fit/Coupling and Head-Helmet Friction Influences on Head Impact Biomechanics and HIT System Accuracy

The HIT System uses spring-loaded accelerometers to maintain contact with the user's head. Sensors need to be rigidly coupled to the skull to provide accurate head impact kinematic and location measurements.[28,206,220] Looser fitted helmets result in measurement errors.[54] However, tightly or properly fitted helmets do not supply rigid head helmet coupling. This is an important distinction because a properly or tightly fitted HIT System instrumented helmet can still undergo relative motion with respect to a head during loading.[94]

Riddell, the only helmet brand with some models that accommodates the HIT System sensor arrays, provides guidelines to ensure a properly fitting helmet.[221] A player's head circumference determines which helmet size an athlete should wear. Riddell recommends players with up to a 51 cm head circumference wear a small helmet, a 52 to 56 cm head circumference wear a medium helmet, a 57 to 60 cm head circumference wear a large helmet, and any player with a head circumference greater than 60 cm wear an extra-large helmet.[221] Selecting the correct size helmet is only the first step in a properly fitted helmet. Riddell helmet liners have fillable air bladders to ensure a snug but comfortable fit front-to-back and side-to-side.[54,221] The helmet should sit 2.54 cm from the top of the player's brow after proper inflation and when pushing down on the helmet the forehead skin should move with the helmet with no room for twisting. Jaw pads should fit firmly against the face, and the chin strap should feel comfortable and snug.[221] Based on Riddell's helmet fitting specifications, the 50th percentile male Hybrid III and medium NOCSAE ATD headforms, the two commonly used headforms in HIT System validation testing, would need size large helmets. Early HIT System validation testing used medium sized helmets that were smaller and likely tighter than Riddell's helmet fitting I.[12,37] Some of the helmet fit parameters suggested by Riddell are quantitative, but most fit parameters rely on a player's qualitative opinion of tightness and comfort to inform optimal fit. To provide an objective helmet fit measure, researchers used a pressure sensitive cap and quantified contact pressures between the head and helmet of a team of 63 football players.[54] Peak contact pressures ranged from 20 to 200 kPa with 59% of the highest peak pressures occurring in the frontal area of the helmet.[54] Importantly, players complained of discomfort when peak contact pressures exceeded 69 kPa and these contact pressures left marks on the players' head.[54] When the same pressure sensitive cap was applied to a 50th percentile male Hybrid III headform wearing a medium sized helmet, the average contact pressures throughout the cap represented the 99th percentile for the average contact pressures throughout the cap obtained from the football players' heads.[54] The peak contact pressure, 93 kPa, exceeded the 69 kPa peak contact pressure where players began to feel discomfort.[54] The contact pressure data demonstrated that the large sized helmet represented a truer fit on the Hybrid III headform than did the medium helmet. The hypothesis, however, that a tighter fitting helmet produces a more rigid couple between the HIT System sensor and the head is inconclusive.[54,211] The HIT System validation testing using the Hybrid III head suggests that helmet fit has minimal effect on peak linear acceleration measurement error, but the effect of helmet fit on peak rotational acceleration measurement error is greater in large-sized HIT System instrumented helmets and lesser in medium-sized instrumented helmets where the properly sized helmet for the headform is a large helmet.[54] This observation is corroborated by observations from bicycle helmet evaluation testing. A reference sensor placed at the COG of an ATD headform modified with a high coefficient of friction silicone rubber skin and wearing a bicycle helmet measured greater rotational accelerations than a metal ATD headform wearing a bicycle helmet. In both headform surface conditions, linear accelerations were equivocal. The unmodified metal ATD headform allowed for relative motion between the metal ATD headform and the bicycle helmet and increased the duration over which the headform rotationally accelerated.[222] Thus, the reference sensors at the ATD's COG in the metal ATD headform condition measured lower peak rotational accelerations as compared to reference sensors at the ATD's COG in the rubber silicone skin condition. The rubber modified ATD headform reduced the relative motion between the ATD headform and the bicycle helmet, rotated more like a single rigid body, and the duration decreased for the headform-helmet system to rotationally accelerate.[222] From a speculative view point, sensors placed on the bicycle helmet would have measured higher peak rotational accelerations in the bare metal ATD headform condition compared to reference sensors placed at the ATD's COG.[94,220] This condition would mimic the higher proportion of rotational acceleration absolute errors that are greater than 15% observed when a large sized HIT System is worn during validation testing.[54] It is unclear if the hypothesis for larger proportions of absolute error greater than 15% in rotational acceleration measured by the HIT System in a large sized helmet can be applied to impact location measurement error made by the HIT System in the same sized helmet. However, a more recent HIT System validation test series used large sized helmets and observed larger overall absolute impact location measurement errors of 42° ± 33° as compared to a test series that used a medium sized helmet and reported absolute angle differences of 31° ± 46°.[11,12] Using properly fitting HIT System instrumented helmets on ATD headforms that are representative of how players on the field wear their helmets provides truer evidence to the strengths and limitations of the HIT System.

During an impact, there are two surface interactions at play. First, there is the friction interaction between the skull and the scalp.[15,95] Second, there is friction interactions between the scalp and the helmet.[15,222] Even with a correctly fitting helmet on an ATD headform, these friction interfaces would influence HIT System sensor coupling and any corresponding measurement error.[94] The Hybrid III headform has a hollow aluminum skull with a vinyl covering that does not replicate the properties of the human scalp.[14,15,32,215] The Hybrid III scalp is dry vinyl, while the human scalp is oily and compliant and may be covered with hair on the head. Similarly, the NOCSAE headform has an external polymer layer simulating scalp.[217] It is unclear how the lack of biofidelity of the Hybrid III and NOCASE headforms, specifically scalp mechanics and the scalp-helmet coefficient of friction, would influence HIT System validation outcomes. The scalp is the first tissue involved in a head impact. It is made up of five layers: Skin, dense Connective tissue, Aponeurosis, Loose connective tissue, and Periosteum.[223] The loose connective tissue allows the scalp to move on the skull. The scalp is connected anteriorly to the orbicularis oculi muscles and laterally to three sites: 1) the zygomatic bone's frontal process, 2) the superior aspect of the zygomatic arch, and 3) over the mastoid. The scalp combines with the superior nuchal line at the back of the head.[223] Scalp thickness ranges from 3 mm in children to 8 mm in adults depending on the location on head.[224] Kinetically, the scalp absorbs and distributes head impact forces, and reduces impact severity by sliding freely over the skull.[225–228] The scalp's impact force absorption and scalp-skull sliding properties reduce peak linear and rotational accelerations by increasing the impact duration at the contact area.[95,228] This impact response observed in the scalp is not replicated by the vinyl layer on the Hybrid III headform.[95] The vinyl layer is tightly coupled, with minimal relative motion, to the Hybrid III headform's aluminum skull and the coefficient of friction between the Hybrid III headform's vinyl layer and a helmet liner is 2.5 times larger than the coefficient of friction between human skin and a helmet liner.[15] The Hybrid III headform's high coefficient of friction between the vinyl skin and helmet liner reduces the head-helmet displacement during rotational impact compared to a human head.[15] Therefore, the Hybrid III headform provides a high coefficient of friction interface for the HIT System instrumented helmet to couple with during validation testing and impact location measurements.

HIT System validation testing recognized the high coefficient of friction interface between the Hybrid III headform and a HIT System instrumented helmet, so the headform surface was modified to reduce the high friction interface. Early studies did not document any friction modifications made between the headform and the headgear. These studies reported head impact location measurement errors from 2.45° to 9.7 ± 5.2°.[188,198] Later validation studies used skull caps, composed of nylon and spandex, or a double layer of nylon stockings to reduce the Hybrid III headform's coefficient of friction.[11,12,37,54,196] These studies documented that the overall average absolute location measurement error were 31° ± 46° and 42° ± 33°, respectively, and demonstrated that lowering the coefficient of friction between the head-helmet interface can lead to increased location measurement error by the HIT System.[11,12] Hair can be added to the Hybrid III headform to create a more realistic friction interaction with the helmet,[15] and results in larger relative helmet motion that introduces head impact magnitude and location measurement errors.[196,211] While using the nylon stocking and hair wig can lower the coefficient of friction between the head-helmet interface, it is unknown if these additions to a Hybrid III headform is representative of the measured coefficient of frictions from cadaveric human heads.[15] The automotive safety industry has used cadaver human specimens extensively for biofidelic loading measurements of various body parts.[44,229–234] Recently the head impact biomechanics sensor community is starting to adopt using human cadaver head specimens for validating head impact sensor performance.[45,96–98,195] Due to recognized differences between the head/helmet fit and frictional interface additional evaluation of the HIT system is warranted using human surrogate head models (e.g. cadaver) that are more representative of the on-field environment.

On-Field Head Impact Sensor Validation Studies

Laboratory validation testing methods offer controlled environments to deliver repeatable impacts to HIT System instrumented helmets on human surrogate heads to validate system accuracy. While laboratory testing environments have made advances in replicating the loading conditions seen on the field,[11,201,202] laboratory methods cannot easily replicate more complex loading conditions. These complex loading conditions can involve multiple impacts from many striking players over a short duration, impacts delivered to the body causing impulsive indirect head loading, glancing impacts, and impacts to many different surfaces including the ground, and surrounding sport equipment like benches or goal posts. While head-helmet fit and friction biofidelity in HIT System validation studies have become more realistic and representative in recent years,[11,54,196] the presence of hair,[196,211] different hairstyles (short vs. long vs. dreads),[211] sweat,[196] varying helmet fits,[54] and varying head shapes and sizes cannot be easily replicated in the laboratory to the degree of actual football players in the field. Validation testing in an on-field environment would require human volunteers to be equipped with reference sensors that were rigidly coupled to their skull and is therefore not practical or possible given ethical considerations.[28]

To get around rigidly coupling reference sensors to human volunteers' skull, head impact kinematics can be inferred through helmet impact kinematics that are determined in videogrammetry.[38,146,147,235] Videogrammetry is the science of acquiring three-dimensional measurements from two-dimensional video images.[146] Researchers used videogrammetry to determine closing speeds of concussion-causing impacts in professional football and rugby players.[147,235] With higher camera frame rates and image resolution along with model-based image matching techniques,[236,237] inferred head impact kinematics can be calculated prior, during, and after the impact.[146] This technique has the potential for *in vivo* head impact sensor systems validation beyond two dimensions,[28] but it requires at least two camera angles with one of the cameras acquiring images at over 240 frames per second to facilitate highly accurate impact kinematic measurements within 7% to 15% error.[146] Unfortunately, the cameras that provide the 240 frames per second rate necessary to capture head impact kinematics in the aforementioned error regime for videogrammetry are expensive and normally used for broadcasting professional sports. Many head impact sensor systems are currently deployed in amateur level sports that don't have access to expensive sophisticated camera systems necessary for videogrammetry.[42,57,59,79,87,189] Additionally, the same research group that reported the videogrammetry techniques for head impact analysis demonstrated that head impact kinematics cannot be inferred from helmet impact kinematics through videogrammetry.[94] In their laboratory testing, the helmet could translate up to 41 mm and rotate up to 37° downward relative to the head in certain impact conditions. The relative helmet motion lead to two to five times larger peak resultant linear acceleration measurements made by helmet sensors compared to reference sensors placed at their ATD headform's COG. Given the inability to access the camera systems required for videogrammetry in amateur sports, and the inability to infer head impact kinematics from helmet kinematics, on-field validation of kinematic magnitude measurements made by the HIT System and other head impact sensor systems are not warranted or feasible in amateur sports.

Video camera technologies used at amateur sport levels are capable of providing adequate frame rates and resolutions for validating head impact sensor detection and location measurements.[64–66,69,99,238,239] Impact video analysis evaluates the accuracy of a head impact sensor's processing algorithms at determining true and false impacts. The number of true positives, false positives, true negatives, and false negatives quantify the performance of the algorithms on determining true impacts that occur to the head during competition or false impacts where the sensor trigger is related to running or jumping and no head impact occurred (**Table 3.1**).[64,66,99] A perfect head impact sensor and its impact processing algorithm would have no false positives, and no false negatives. However, studies that have used impact video analysis to verify impact algorithm classification accuracy demonstrated that head impact sensor algorithms, such as the xPatch, incorrectly classify sensor triggers.[64–66] The false positive rate varied between 20% to 84% and false negative rate varied between 3% to 78%.[64,66] The actual false positive and false negative rates are likely closer to 84% and 3%

respectively, as over 17,865 xPatch triggering events, collected from 26 collegiate women's soccer players, were cross-referenced with video to generate these rates.[64] Regardless of the false positive and negative rates associated with a head impact sensor, it is important that data collection trigger events are confirmed as actually occurring. Without video for confirming impacts provided by head impact sensors, a higher number of data collection triggers and triggers with higher magnitudes not associated with head impacts can make their way into datasets for analysis.[65] False positive and false negative sensor measurements create an inaccurate representation of the impact environment in terms of the impact frequency, location magnitude.[65] Researchers use this data to understand head injury mechanisms both in terms of magnitude and directional loading, correlate head impact exposure to short- and long-term neurological effects, and to create or modify data derived rules for player safety. If we fail at properly characterizing the impact environment for injury risk and the relationships of head impact exposure to neurological health, then we cannot properly protect athletes and expose them to issues on the field.

Table 3.1: Definitions for impact classifications for video impact verification.

		Video	
		Head Impact Observed	**No Head Impact Observed**
Head Impact Sensor Algorithm Classification	**Valid Head Impact Classified by Algorithm**	**True Positive** (Algorithm Correct Classification)	**False Positive** (Algorithm Incorrect Classification)
	Non-Head Impact Classified by Algorithm	**False Negative** (Algorithm Incorrect Classification)	**True Negative** (Algorithm Correct Classification)

There are several methods to ensure the datasets collected by head impact sensors contain the highest quality data in terms of limiting false positives and negatives. As discussed in the prior paragraph, confirming impacts using accompanying video is one method to ensure the highest quality dataset. However, this method is time intensive and research teams utilizing head impact sensor technology may not have the personnel resources to visually confirm every data collection trigger event made by a head impact sensor through video.[65] Time filtering is a method used to ensure data collection trigger events measured outside start and end times of competition do not make their way into final datasets for analysis.[8,65] However, as high as 35% to 68% of data collection triggers within time filtered data are not associated with a head impact from investigations using the xPatch and a helmet based head impact sensor system called the GforceTracker, respectively.[65] Changing the data collection acceleration threshold is another method researchers used to ensure data collection triggers are associated with head impacts and not due to running or jumping.[64,65] This data collection acceleration threshold has typically been set at 10 g of linear acceleration and was determined by observing through video that running and jumping motion was associated with peak resultant linear accelerations below 10 g.[35,182,184] Depending on the sensor, however, running and jumping movements can result in linear accelerations greater than 10 g and trigger data collection.[64–66,69,99] A positive predictive value analysis demonstrated that a 34 g data collection acceleration threshold for the xPatch used in soccer could increase the true positive rate from 16.3% to 65%.[64] However, this would lead to a high percentage of false negatives, as a number of impacts experienced in soccer are below 30 g of linear acceleration.[172,240,241] As such, linear acceleration data collection thresholds are insufficient in differentiating head impacts from human movement.[99]

Head impact sensor developers are using additional instruments and more sophisticated processing algorithms to classify true and false impact events.[99–101] Recently, developers of a headband coupling head impact sensor, the Triax Sim-G, created algorithms that use neural networks combined with Fourier transform heuristics to classify true and false impacts measured in soccer.[100] Their pattern recognition algorithm used linear accelerations and angular velocities to classify impacts or non-impact transients. The pattern recognition algorithm had an 88% sensitivity and 47% specificity at distinguishing between real and non-impact events.[100] The Triax Sim-G's impact processing algorithm showed promise for correctly distinguishing between real and non-impact events. The datasets used to train their pattern recognition software came from collecting *in vivo* head impact kinematics from soccer players wearing the Triax Sim-G, while observers on the sideline tracked and documented head contacts. The algorithm could benefit from a larger training dataset where impacts were verified with high-speed video to increase the number of different high impact loading categories like body part to head contact. Developers of a mouthguard head impact sensor from Stanford University used a combination of infrared sensors within the mouthguard and a support vector machine classification program to correctly classify true and non-head impact events. The infrared sensor within the mouthguard quantified mouthguard coupling quality to the teeth.[45,99] Impacts collected while the infrared sensor had a low reading indicated the mouthguard was off or loosely coupled to the teeth. Therefore, the head kinematics could not be relied on as accurate.[45] The authors used low infrared readings from the mouthguard to automatically remove head impact trigger events as these were likely due to the athlete chewing on the mouthguard rather than the mouthguard being coupled to the teeth.[99] The support vector classification program used power spectrum densities and wavelet transform features, to classify true and non-head impact events. The power spectrum densities and wavelet transform features were derived from the linear acceleration and angular velocities from impacts delivered to collegiate football players while wearing the mouthguard. The authors reported greater than 87% on measures of sensitivity, specificity, precision, and accuracy when they applied their impact processing algorithm to video cross-validated impacts using a single college football player, and a team of six youth football players wearing the mouthguard head impact sensor.[99] The power spectrum density and wavelet transform features extracted from linear acceleration and angular velocity measurements showed that true impact events had higher amplitudes at lower frequencies, while non-impact events had amplitudes and oscillations at higher frequencies.[99] Developers of another mouthguard head impact sensor also demonstrated non-impact events or false positives exhibited high-frequency content within the measured kinematic signals.[101] While using linear acceleration thresholds doesn't sufficiently discriminate between true head impact and non-impact events,[64–66] the frequency content within the measured kinematic signals holds promise for correctly classifying true head impact events.[99–101]

The HIT System's impact detection processing algorithm uses two criteria to classify true and non-head impact events. First, the algorithm only retains impacts that are greater than a 10 g peak resultant linear acceleration threshold. Second, the measured acceleration pulse from the impact should contain characteristics of an impact to a helmeted head based on rigid body dynamics.[8,34,196] A number of studies using the HIT System to measure head impact biomechanics in football and hockey players also concurrently acquired video. Eight of these studies used video to confirm the injury mechanisms/characteristics for HIT System collected head impact kinematics that lead to diagnosed concussions.[9,36,83,165,186,188,242,243] Four studies used video to confirm HIT System trigger events exceeding 60-150 g of peak resultant linear acceleration depending on the study.[22,34,244,245] Six studies reviewed subsets of impacts to analyze playing behaviors in hockey,[246,247] anticipation levels for incoming impacts,[55,68] and play-type and collision closing distances in collegiate and high school football.[6,59] Two studies reported previously reviewing all collected head impacts but didn't mention any video review on recently collected head impacts.[84,248] Finally eighteen studies concurrently collected video and confirmed all head impacts collected by the HIT System.[79,85–88,189,191,249–259] A lot of credit is owed to the authors and research assistants that reviewed the head impacts to ensure the highest quality datasets because this process can take over 1000 man hours to accomplish.[65,69] However, no study has evaluated the accuracy of the HIT System's impact detection processing algorithm at correctly identifying true head impact and non-impact events on the football field in any capacity. This is important as the HIT System is a heavily used measurement tool for head impact biomechanics collection, and we do not fully understand the false positive and false negative rates. This represents a large question regarding the accuracy of the HIT System impact detection processing algorithm on the football field.

Summary

The HIT System is an extensively used research tool for quantifying *in vivo* head impact biomechanics to youth, high school, collegiate, and professional football players. Instrumenting college football players' helmets with the HIT System provided data that impacts experienced on special teams plays over long closing distances are severe due to the higher impact velocities achieved by the players. These data lead to modifying the kickoff rules at the professional level in order to promote more touchbacks and reduce the high concussion injury rate. Collegiate and high school levels adopted these refined kickoff rules as well. However, most kickers at the high school level lack the strength and skill to kick the ball through the back of the endzone and the strength of data derived rule modifications hinges on the HIT System's measurement precision to provide accurate data. The HIT System has been rigorously validated in a laboratory setting while on a Hybrid III headform. Unfortunately, the Hybrid III headform's vinyl skin does not mimic human skin and hair and creates an artificially high coupling interaction between the HIT System sensor and the head. The laboratory environment offers repeatability and control to investigate the HIT System's accuracy under specific loading conditions. However, the on-field environment offers a more complex loading environment, such as multiple impacts over a short duration, or glancing impacts, that can't be replicated easily in the laboratory. The HIT System's measurement accuracy has not been quantified using the closest biofidelic surrogate head and loading environments to its intended use in the field on football players' heads. Therefore, the purpose of this study is to 1) quantify the HIT System's impact location measurement accuracy in laboratory evaluations using a cadaveric human head specimen, and 2) extend the evaluation to an on-field setting by using video confirmation of impacts to quantify the HIT System's impact detection algorithm and location measurement accuracy during special teams play.

CHAPTER 4

Manuscript 1

Football HIT System Impact Location Measurement Accuracy While Using a Post Mortem Human Head Specimen as a Head Surrogate During Drop Testing

Introduction

Sports offer an environment for researchers to study concussions by measuring external biomechanical forces that are transmitted to the head and the effects of head impacts over a lifetime of sport participation.[4,8,71–73] These transmitted forces cause the head to undergo a combination of linear and rotational acceleration,[4,74,75] and the proportion of linear and rotational acceleration experienced by the head is dictated by the head impact location and direction of head loading.[5,76,77] Head impact sensors have allowed researchers to quantify head impact biomechanics—head impact frequency, location, and magnitude—in many different sports across multiple skill and age levels.[8,59,71,72,79–82] The HIT System is a head impact sensor that has been used extensively in football.[8,59,79] Differential head impact exposures to playing positions,[8,85,86] extrinsic and intrinsic severe head impact risk factors, [6,41,42,87,88] and the relationship between structural and cognitive brain function and head impact biomechanics have been quantified using the HIT System. The strength of these relationships hinges on the HIT System's ability to accurately measure head impact kinematics and locations.

During the HIT System's development, accelerometers were embedded into a metal hemisphere surrogate head.[30] Impact location measurements calculated from the embedded accelerometers were within 10° of the impact locations calculated by a gold standard reference triaxial accelerometer placed at the center of the metal hemisphere.[30] This preliminary investigation provided initial evidence that the accelerometer hardware, arrangement, and software algorithms could measure impact location within a reasonable error level in an ideal situation where the accelerometers were rigidly mounted to a metal surface representing the human head. A limitation of this assessment was that the validation did not consider in-the-field real world environmental factors relevant to accelerometer coupling to the head.

The six single axis linear accelerometers used in the HIT System are configured to fit into a U-shaped encoder inserted into the crown of football helmets. These HIT System instrumented football helmets are worn by surrogate heads such as the Hybrid III or National Operating Committee for the Standard of Athletic Equipment (NOCSAE) headforms for evaluating the HIT System impact kinematic and location accuracy. [31,32,217] Impact location measurements made by the HIT System during laboratory validation testing are compared to high fidelity gold standard reference sensors mounted inside the headforms. The HIT System had location measurement accuracies within ± 2.45° of the impact site,[188] but recent evaluations reported larger impact location measurement errors (31° ± 46° and 42° ± 33°).[11,12] These impact location measurement error differences between evaluations could be attributed to the impact delivery methods and headform choices.[86,207] The HIT System could have better helmet-headform coupling due to the low impact energies and centroidal impacts achieved through drop testing methodology.[94,207] In terms of the headform, the jaw and cheeks are narrower in the Hybrid III compared with the NOCSAE headform, and the back of the Hybrid III headform does not extend all the way to the bottom edge of helmets.[33] These differences may lead to reduced helmet- headform padding surface area interaction and likely allow for more helmet rotation relative to the headform, which introduces measurement error.[33,94] Thus, using a head surrogate with shape characteristics similar to a human head should be considered for better translating HIT System's impact kinematic and location accuracy laboratory evaluations to the HIT System's on-field application.

In addition to impact delivery and headform considerations for HIT System laboratory evaluations, researchers applied skull caps, stretched stockings, or added human hair wigs to the headforms in an effort to more closely replicate the head-sensor/ head-helmet friction interface.[11,12,37,54,196] This is an important feature to replicate as the measurement error by a head impact sensor depends on how rigidly it is coupled to the head, to reduce differential motion between the sensor and the head.[12,28,206] In addition, differences exist in the head/helmet fit and frictional interface between the Hybrid III headform and human head, which may influence the HIT System's measurement accuracy.[15,54,94,95] Due to recognized differences between the head/helmet fit and frictional interface, evaluating the HIT system using human surrogate head models (e.g., cadaver) more representative of the on-field environment is warranted.

This study quantified the HIT System's impact location measurement accuracy by dropping a biofidelic head surrogate—a cadaveric human head specimen—from various heights and onto various locations of a HIT System instrumented helmet. We hypothesized that the HIT System would have larger impact location measurement error from drops onto the facemask as compared to drops directly impacting the helmet shell.

Methods

Three fresh frozen male human cadaver heads were disarticulated at the atlanto-occipital joint from the rest of the neck. The mandible remained coupled to the skull and the heads were sealed at the occipital condyles with polymethylmethacrylate. The three cadaver heads had different masses, shapes, and wore different sized helmets (**Table 4.1**). We rigidly coupled a reference sensor block to the skull's occipital bone posterior to the foramen magnum with wood screws. The sensor block included the gold standard reference sensors: three single-axis linear accelerometers (model 7264B-2000, Endevco Corp., San Juan Capistrano, CA) and three single-axis angular rate sensors (model DTS ARS 8K, Diversified Technical System Inc., Seal Beach, CA), enabling measurement of six degrees-of-freedom (6DOF) head kinematics.

The reference sensors were filtered with a hardware anti-alias filter at 25 kHz and sampled at 100 kHz.

We installed a HIT System Speed MxEncoder in the available space within the crown of one large and one extra-large Riddell Speed Helmet (**Figure 4.1a and 4.1b**). The HIT System instrumented helmet was fit to each head according to helmet fitting procedures outlined by Riddell.[221] The HIT System measured the linear acceleration of the helmeted cadaver head using six single-axis spring-loaded linear accelerometers. The HIT System collected data for 40 ms (8 ms pre-trigger, 32 ms post-trigger) at 1 kHz when any of the accelerometers detected accelerations exceeding a 14.4 g user programmable threshold and wirelessly transmitted the data to a laptop Sideline Response System. A proprietary algorithm determined the peak resultant linear acceleration at the head center of gravity (COG) from the raw linear acceleration signals. The algorithm also determined the head impact location and reported it in two ways: (1) in degrees of azimuth and elevation, and (2) as a category (**Figure 4.1c and 4.1d**).[29,30,40] All data were date and time stamped and exported from the HIT System's Redzone data cloud.

We impacted the helmeted human cadaver head specimens using a drop test methodology.[44,45] The helmeted heads were placed into a fine mesh net and the net was hoisted to one of three desired drop heights using a nylon line (**Figure 4.2a and 4.2b**). Before each drop, the mid-sagittal planes of the head and helmet were aligned, and we positioned the top of the helmet opening 2.5 to 4 cm (1 to1.5 inches) above the cadaver's glabella. The helmeted heads were positioned within the net to achieve one of six desired drop locations (**Figure 4.2c**). The head was released into freefall by burning the nylon line. The helmeted head fell onto an aluminum plate with a tri-axial load cell (Kistler 9067, Kistler Instrument Cop., Amherst, NY) located beneath the plate sampling at 100 kHz. Data acquisition of the reference sensors for 660 ms with a 100 ms pre-trigger occurred once our trigger threshold (78.1 N vertical ground reaction force) was detected by the load cell. After every drop, the helmet was inspected to ensure the chinstrap had not moved, no hardware had come loose, and that no part of the helmet was damaged. The reference sensor block was examined at the end of each cadaver series to confirm that the coupling had not changed during testing. We performed drops onto each location in blocks, and drop heights were performed in ascending order. A total of 4 trials were performed at each combination of three drop heights and six locations for a total of 72 drops per cadaver head, and an overall total of 216 drops. Drops were repeated when the reference sensors failed to trigger (n = 2) or there were technical issues (n = 3). No drops were repeated when the HIT System failed to trigger.

We determined the head impact location from the gold standard reference sensors by transforming the head impact kinematics measured by the reference sensors to the cadaver head's COG. This involved demeaning linear acceleration data and rotational velocity data acquired from the reference sensors and using a 4-pole Butterworth low pass digital filter with 1650 Hz (CFC 1000) and 300 Hz (CFC 180) cutoff frequencies on the reference linear acceleration and rotational velocity data, respectively.[46,47] Rotational velocity data were numerically differentiated with a 5-point stencil method to acquire reference rotational acceleration.[48] A micro-CT scanner (Nikon XT H 225 ST; Nikon Metrology Inc., Brighton, MI) imaged each helmeted cadaver head. Anatomical landmarks and the reference sensors' orientations were measured from the CT images using Avizo 3D visualization software (Avizo 9.4; Thermo Fisher Scientific, Hillsboro, OR). The head coordinate system origin was defined as the midpoint between the porions lying in the plane made by the two external auditory meatuses and the left orbitale (i.e. Frankfurt plane; **Figure 4.3**).[49] We transformed the reference kinematic signals by rotating them to the head coordinate system, and projecting the linear acceleration to the head COG using the equation for rigid body transformations.[50,260] Head impact location coordinates for the reference sensors were calculated from the transformed reference signals in terms of azimuth and elevation by using the x, y, and z components from the peak resultant linear acceleration vector pointing towards the head COG. We matched the time stamp for each drop from the reference sensors to the HIT System's time stamp to merge head impact location data and aligned azimuth, elevation degrees, and impact location category according to **Figure 4.1c and 4.1d.** Any drops where the HIT System did not trigger were left blank and not matched with the time stamp from the reference sensors.

For each drop site, we calculated the mean impact location coordinates and standard deviation ellipse (SDE) for the reference sensors and the HIT System in Matlab (R2017b, MathWorks, Natick, MA) using SPAK library functions.[52] Azimuth and elevation direction data are likely to covary meaning that low variability can be observed in azimuth coordinates, but high variability can be observed in elevation coordinates and vice versa.[11,51,52,261] A Kent distribution can model asymmetries in azimuth and elevation variability as opposed to a Fisher distribution that assumes symmetry in the azimuth and elevation variability. We tested for modelling the data with a Fisher or Kent distribution according to previous studies.[11,51,52,261] For data with a Kent distribution, we reported the major (long) and minor (short) semi axes for the SDE to provide a sense of the drop site's impact location coordinate asymmetrical variability. The precision of the impact location coordinates measured by our data collection systems were evaluated by reporting the focus, where higher precision is associated with a focus closer to one.[262] For comparisons to previous HIT System evaluations, we also calculated the mean absolute angular difference between the reference sensors and the HIT System for each drop site.

Head impact location accuracy measured by the HIT System against the gold standard reference sensors were analyzed in two ways. First, we assessed the mean spherical error at measuring azimuth and elevation impact location coordinates between the HIT System and the reference sensors.[11] This was accomplished by rotating both the reference sensors and HIT System direction vectors, using the azimuth and elevation coordinates, for a drop so that the reference sensor's direction vector for that drop aligned with the pole (elevation = 90°) of a unit sphere.[261] These rotations maintained the three-dimensional angular differences between the reference sensors and HIT System impact location coordinates for a drop. The mean azimuth and elevation from the rotated HIT System direction vectors were determined and mean spherical error was calculated as the three-dimensional angular difference from the pole to that mean. A 95th percentile confidence ellipse around the mean-rotated HIT System direction vectors was calculated. If the 95th percentile confidence ellipse included the pole, then there was no statistical difference between the impact location coordinates measured by reference sensors and the HIT System for that drop location.[51,52,262] Statistical significance was set to an alpha less than 0.05.

The second way we evaluated HIT System impact location measurement accuracy was with the percent agreement in impact location category between the HIT System and the reference sensors. The azimuth and elevations measured by the reference sensors were categorized into impact location categories according to **Figure 4.1c and 4.1d**. Each drop site corresponded with a HIT System impact location per **Figure 4.1c and 4.1d** except for the front oblique right drop site. The HIT System does not provide a front oblique right impact location category. We assigned drops measured by the HIT System and the reference sensors that had elevations less than 65° and azimuths that fell between 112.5° and 157.5° as a front oblique category according to Gwin et al.[201]

Results

We preformed 221 drops of which the reference sensors collected data on 216 drops, while the HIT System collected data on 178 drops. Drops where the reference and the HIT System concurrently collected data were used for the spherical analyses of head impact location coordinates, and impact location category agreement assessments.

The reference sensors measured precise and less variable head impact location coordinates at the vertex drop location, and less precise and more variable head impact location coordinates at the front oblique right drop site (**Table 4.2**). The HIT System measured precise and low variable head impact location coordinates at the front oblique drop location, and less precise and more variable head impact location coordinates at the right parietal drop site (**Table**

4.2). The HIT System was statistically different, based on the 95th percentile confidence ellipse around the mean rotated impact location coordinate data, than the reference sensors at measuring the impact location coordinates for all drop sites except at the facemask drop site (**Figure 4.4 and Figure 4.5**). The HIT System had the lowest mean spherical error at the facemask drop site and the highest at the vertex drop site (**Table 4.2**). Mean absolute angular differences were within 10° of the mean spherical error for all drop sites except at the facemask drop site. The HIT System had large variability at measuring impact location coordinates at the facemask drop site, which contributed to the more than 10° difference between the mean spherical error and the mean absolute angular difference.

The reference sensors correctly categorized 95% of the drop sites according to the HIT System impact location categories (**Table 4.3**). One out of the 36 frontal drops was incorrectly categorized, and 10 out of the 36 front oblique drops were incorrectly categorized by the reference sensors. No trends emerged when comparing the percentage of correctly categorized drops by the reference sensors across the three different head specimens.

Overall, the HIT System correctly categorized the impact location with the intended drop site less than half of the time (**Table 4.3**) for impact location category data that were collected by the HIT System. All drops to the occipital site were correctly categorized as back by the HIT System. One out of the 31 (3%) drops were correctly categorized as top. The HIT System's overall impact location agreement with the intended drop location category was highest for the D3 cadaver head specimen. The overall impact location category agreement was lower for drops using the D4 and D5 cadaver head specimens, but the impact location category was similar between the D4 and D5 specimens

The reference sensors and HIT System agreed on the HIT System impact location category on 45% of the drops where the HIT System triggered for data collection. These impact location categories agreements were similar to the HIT System compared against the drop site location category (**Table 4.3**). The reference sensors and HIT System agreed on the impact location category for 3% of the vertex drops. There was 100% agreement on drops to the occipital site. There was a higher percent agreement on categorizing impact location from drops with the D3 specimen than with the D4 and D5 specimens.

Discussion

The HIT System provided statistically different impact location coordinates and has low agreement on impact location category compared to reference sensors while coupled to human cadaver head specimens undergoing laboratory-controlled drops. These observations were obtained using a biofidelic surrogate head testing paradigm not previously employed in the literature on the HIT system. This work extends previous studies evaluating the HIT System's impact location measurement accuracy using headform data and addresses a potential validation gap between laboratory and field studies by addressing shape characteristics and skin-helmet coefficient of friction differences between the human head and Hybrid III headforms previously employed in these studies.[11,12,15,33]

The HIT System had the lowest mean spherical error from measuring drops onto the facemask site, and these measurements did not statistically differ from the impact location coordinates measured by our reference sensors. These observations did not support our hypothesis that the HIT System would have larger impact location measurement error from drops onto the facemask than onto the helmet shell. While not statistically different, the HIT System had large variability at measuring the impact location coordinates, primarily in elevation, to the facemask drop site and contributed to a larger average absolute angular difference than mean spherical error. Previous HIT System evaluations used the Hybrid III headform and modified the helmet-skin coefficient of friction interaction with a nylon sock stretched over the headform.[11,12] These evaluations reported similar (42° ± 32°)[11] and larger (95° ± 68°)[12] mean absolute angular differences at the facemask location than what we present in our results (43° ± 50°). The choice of head surrogate, cadaveric human specimen versus Hybrid III, cannot provide an explanation of these differences because both previous evaluations used the Hybrid

III headform. One of the previous evaluations used a medium sized helmet on the Hybrid III, which provided an unrealistically tight fit and theoretically better coupling to the headform, reported larger mean absolute angular differences than our study and another HIT System evaluation.[12,54] The facemask site failed to trigger data collection on 13 out of the 36 drops in our testing. Caution is therefore warranted on interpreting our facemask site results.

The HIT System measured statistically different head impact location coordinates compared to the reference sensors at all drop sites except for the facemask location. However, the HIT System measured precise impact location coordinates from drops onto the front oblique right location, but measurements made by the HIT System from frontal oblique right drops were higher in elevation compared to the reference sensors. Based on our data, an offset may exist where helmet shell impacts are measured with higher impact elevations than the contact site and what the reference sensors measure. This offset explains why the HIT System agreed with the reference sensors on the impact location category on 14% of the drops at the front oblique right location. Otherwise the HIT System categorized 83% of the front oblique drops as top. The HIT System does not provide a front oblique right or left impact location category. We assigned this impact location category according to the azimuth falling between 112.5° and 157.5° according to Gwin et al.[201] The HIT System can provide a front oblique impact location provided more research is performed to understand a potential offset towards measuring impacts higher in elevation than the actual contact site on the helmet. It would be beneficial knowing impacts are delivered to an oblique location on the helmet. Oblique impacts incorporate head translations and rotations along and about multiple axes that can lead to greater magnitude brain tissue strains, which could increase concussion risk.[53,163] In comparison, frontal or side impacts produce head translation and rotation along and about one axis leading to lesser brain strains.[53]

Only 3% of the vertex drops were categorized by the HIT System with the correct corresponding impact location category of top. Otherwise, 97% of the vertex drops were categorized as the back location with mean impact location coordinates of 176° in azimuth and – 24° in elevation. While the HIT System measured inaccurate impact location coordinates from the vertex drops, the system measured precise location coordinates with a focus of 0.96. Siegmund et. al. had a similar observation with the HIT System measuring reflected impact location coordinates during crown impacts from a linear impactor.[11] To further explore this issue Siegmund et al. ran additional tests using a Riddell Revolution helmet, the helmet model preceding the Riddell Speed model. The HIT System within a Riddell Revolution helmet measured crown impacts with the correct corresponding top location in five tests and were not reflected to the back of the helmet. We used Riddell Speed helmets to accommodate the HIT System encoder. Between our results and those from Siegmund et al., HIT System impact location accuracy from impacts directed onto the helmet's crown (represented in our study as the vertex) may depend on the helmet model.[11] This is important and warrants further investigation as three different Riddell helmet models, Revolution, Speed, and Speed Flex, have been used since the HIT System hardware and software were initially developed for use in the Riddell VSR4. With the Riddell Speed Flex helmet now the leading Riddell model in terms of safety, according to the Virginia Tech Helmet Start Rating, it will become a popular helmet choice for football teams. The HIT System's impact location accuracy should be evaluated in the Speed Flex model to observe if the same phenomenon of reflected crown impacts to the back of the helmet is still present. Additionally, on-field evaluations of pairing video observed loading to the crown with HIT System impact location outputs should be performed to confirm or refute this phenomenon.

Our results have applications for researchers using HIT System outputs to model brain tissue deformations from head impacts using finite element methods. Most finite element head models require 6DOF head kinematics, but the football HIT System estimates 5DOF kinematics.[183,263] Algorithms are being created that estimate 6DOF kinematics from the HIT System's 5DOF outputs by generating characteristic acceleration curves that are associated with an impact region and the HIT System's sensor polarities for an impact.[264] The impact regions in their algorithm are separated by 15° elevation increments and by 15° azimuth increments at the Frankfurt plane with impact regions increasing in azimuth as the measured impact increases in elevation. With the absolute angular errors reported in this study, the impact region/location could be at least 3 regions different from what the HIT System would measure as compared to high-fidelity reference sensors coupled to the skull. It is unknown how a three-region difference would change the characteristic 6DOF acceleration curves used in a finite element model for modelling brain deformation.

Like all research, our study has limitations. We used a drop method to deliver impacts to the helmeted cadaver head specimens; therefore, the impact was delivered through the head COG. Using a pneumatic linear impactor or pendulum impactor would have allowed us to evaluate the HIT System's impact location measurement accuracy in more oblique non-centric impacts.[210] There was also no neck as part of our experimental design. A neck would have provided an endpoint for head rotation that may better represent on-field exposures; however, tests with a Hybrid III dummy with a neck showed that the head motion in the first 40 ms of an impact is unaffected by the neck compared to Hybrid III heads in freefall without a neck.[265,266] Therefore, we don't anticipate the lack of a neck on our cadaver heads influencing the impact dynamics as all of our outcomes are calculated within the first 40 ms following the helmeted head contacting the ground. The cadaver heads provided various shapes, biofidelic skin friction, and skin dynamics for the helmet and the head-sensor contact points from the HIT System encoder.[15,33,95,222] A sensor's head coupling mechanism is important in measuring accurate head impact kinematics and impact locations in the field.[28,94] This report builds off part of a larger evaluation of multiple head impact sensors. One of the aims of the larger evaluation was to determine the detection rate for the various head impact sensors. Therefore, we did not fill the complete test matrix, and this aim came at the expense of having an unequal number of drops across drop sites. Thus, the head impact locations measured by the HIT System against the reference sensors could have been biased towards only accurate or inaccurate measurements, and potentially skewed our results. We want to communicate that one of our objectives for this study and the larger study was to evaluate the head impact data collected by these sensors based on the data exports they provide as if they were used for on-field data collection. These sensor's do not get the opportunity to recollect data if they fail to trigger in the field. Acknowledging our limitations, this is still one of the first studies to report on the HIT System's impact location measurement accuracy on a more biofidelic head surrogate than the Hybrid III in terms of head shape, skin friction, and skin dynamics.

Conclusions

The HIT System provided statistically different impact location coordinates than references sensors, except for impacts directed onto the facemask, and agreed 45% of the time with the impact location category determined by the reference sensors. While the HIT System did not statistically differ from the reference sensors at measuring impact location coordinates from drops onto the facemask site, the mean absolute angular differences for the facemask site and all sites reported in this study using cadaver human head specimens as head surrogates are within the range of previous HIT System evaluations that used the Hybrid III as a head surrogate. Our preliminary data suggest that the HIT System may not provide highly accurate impact location coordinate measurements that are required for algorithms that estimate 6DOF motion from the HIT System outputs based on the impact location. However, the HIT System may provide adequate data for less refined applications that use broader impact location categories. These data will inform future studies that incorporate additional helmet impact sites and eccentric helmet loading to further explore HIT System data measurement accuracy using human cadaver head specimens. Additional on-field studies quantifying impact location category are required to confirm or refute these laboratory observations.

Tables

Table 4.1: Anthropometric summary and helmet sizes used for the three cadaver human head specimens in our drop testing

Specimen ID	Head Mass (kg)	Head Circumference (cm)	Helmet Size	Helmet Mass (kg)
D3	3.4	55.2	Large	2.2
D4	4.5	59.0	Extra Large	2.3
D5	3.6	58.0	Large	2.2

Table 4.2: Descriptive spherical statistics for the reference and Head Impact Telemetry (HIT) System at each drop site. Descriptive statistics include the number (n) of drops recorded at each drop site, the mean impact location coordinates in azimuth (Az), and elevation (El) degrees and the standard deviation ellipse (SDE) major (Maj), and minor (Min) axis lengths in degrees. Mean spherical error (MSE) on the impact location coordinates between the Reference sensors and the Head Impact Telemetry (HIT) System are presented. For comparison to previous studies, the mean absolute angular difference (Δ) and standard deviation (STD) between the reference sensors and the HIT System are provided.

	Reference Dataset				HIT System							
Drop Site	**n**	**Mean (Az°, El°)**	**SDE (Maj°, Min°)**	**Focus**	**n**	**Mean (Az°, El°)**	**SDE (Maj°, Min°)**	**Focus**	**MSE (°)**	**SDE (Maj°, Min°)**	**Mean Absolute Angular Δ (°)**	**STD (°)**
Facemask*	23	180, -8	9, 6[a]	0.98	23	-178, -18	25, 2[a]	0.65	10	(26, 4)[a]	43	50
Front Oblique Right	29	117, 30	14, 10[a]	0.96	29	141, 72	6, 4[a]	0.99	44	(15, 10)	46	12
Frontal	34	177, 35	11, 4	0.98	34	2, 69	44, 2	0.54	76	(42, 3)	84	56
Occipital	35	-5, 37	17, 4	0.95	35	3, -3	30, 2	0.85	41	(30, 4)	44	28
Right Parietal	26	78, 26	10, 5	0.98	26	44, -12	29, 11	0.84	51	(29, 10)	53	24
Vertex	31	-172, 85	9, 3	0.99	31	4, 11	11, 2	0.96	120	(14, 3)	118	20

* Denotes that the impact location coordinates collected by the HIT System were not statistically different from the impact location coordinates collected from the reference sensors at the drop site

[a] Data distributed on a sphere in a Fisher distribution (isotropic bivariate normal distribution)

Table 4.3: Percent agreement on the impact location categories derived from the azimuth and elevation coordinates measured by the reference sensors and the Head Impact Telemetry (HIT) System. Impact location category percent agreements are made by drop site for the drop contact site with the reference sensors derived category, the drop contact site with the HIT System derived category, and the reference sensors derived category with the HIT System.

	Drop Contact Site vs Reference				**Drop Contact Site vs HIT System**				**Reference vs HIT System**			
	Head Specimen ID				**Head Specimen ID**				**Head Specimen ID**			
Drop Site	**D3**	**D4**	**D5**	**Total**	**D3**	**D4**	**D5**	**Total**	**D3**	**D4**	**D5**	**Total**
Facemask	100	100	100	**100**	100	100	63	**87**	100	100	63	**87**
Front Oblique Right	50	100	67	**72**	14	0	27	**14**	14	0	27	**14**
Frontal	100	92	100	**97**	18	33	0	**18**	18	33	0	**18**
Occipital	100	100	100	**100**	100	100	100	**100**	100	100	100	**100**
Right Parietal	100	100	100	**100**	90	0	63	**54**	90	0	63	**54**
Vertex	100	100	100	**100**	13	0	0	**3**	13	0	0	**3**
Total	**92**	**99**	**94**	**95**	**56**	**41**	**40**	**45**	**56**	**41**	**40**	**45**

Figures

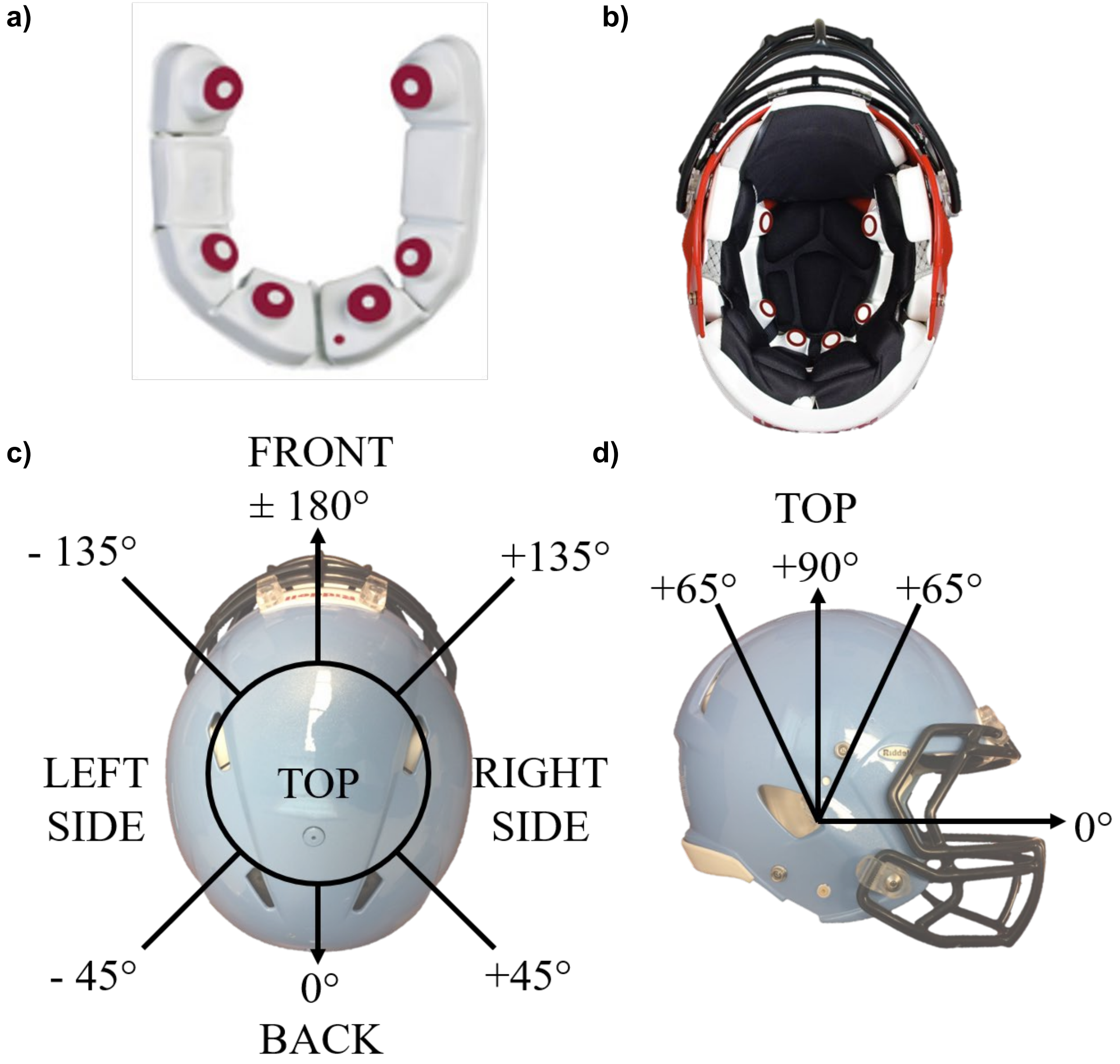


Figure 4.1: The HIT System encoder **(a)** fit into Riddell Speed football helmets **(b)**. Six linear accelerometers and proprietary algorithms measured head impact location in azimuth degrees around the head **(c)**, in elevation degrees **(d)**, and as an impact category. Impacts > 65° elevation are classified as Top impacts regardless of the azimuth degree calculation. Otherwise, impact location category was determined by the azimuth degree falling into one of the 4 location bins (Front, Right Side, Left Side, and Back).

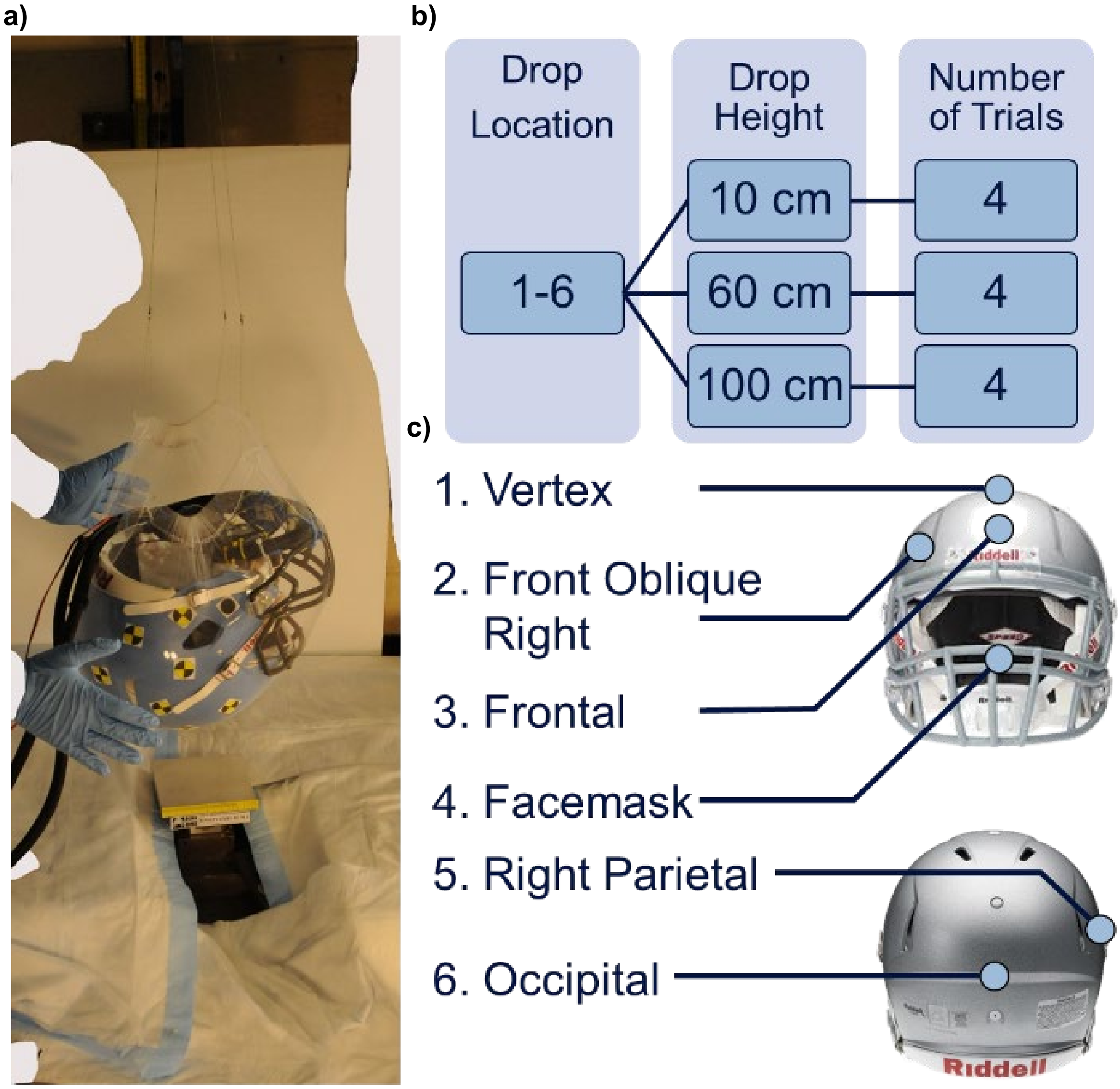


Figure 4.2: Drop Locations and Setup. Helmeted cadaver heads were inserted into fine mesh nets as part of our drop setup (a), hoisted to one of three drop heights (b), and dropped onto one of the six drop locations (c)

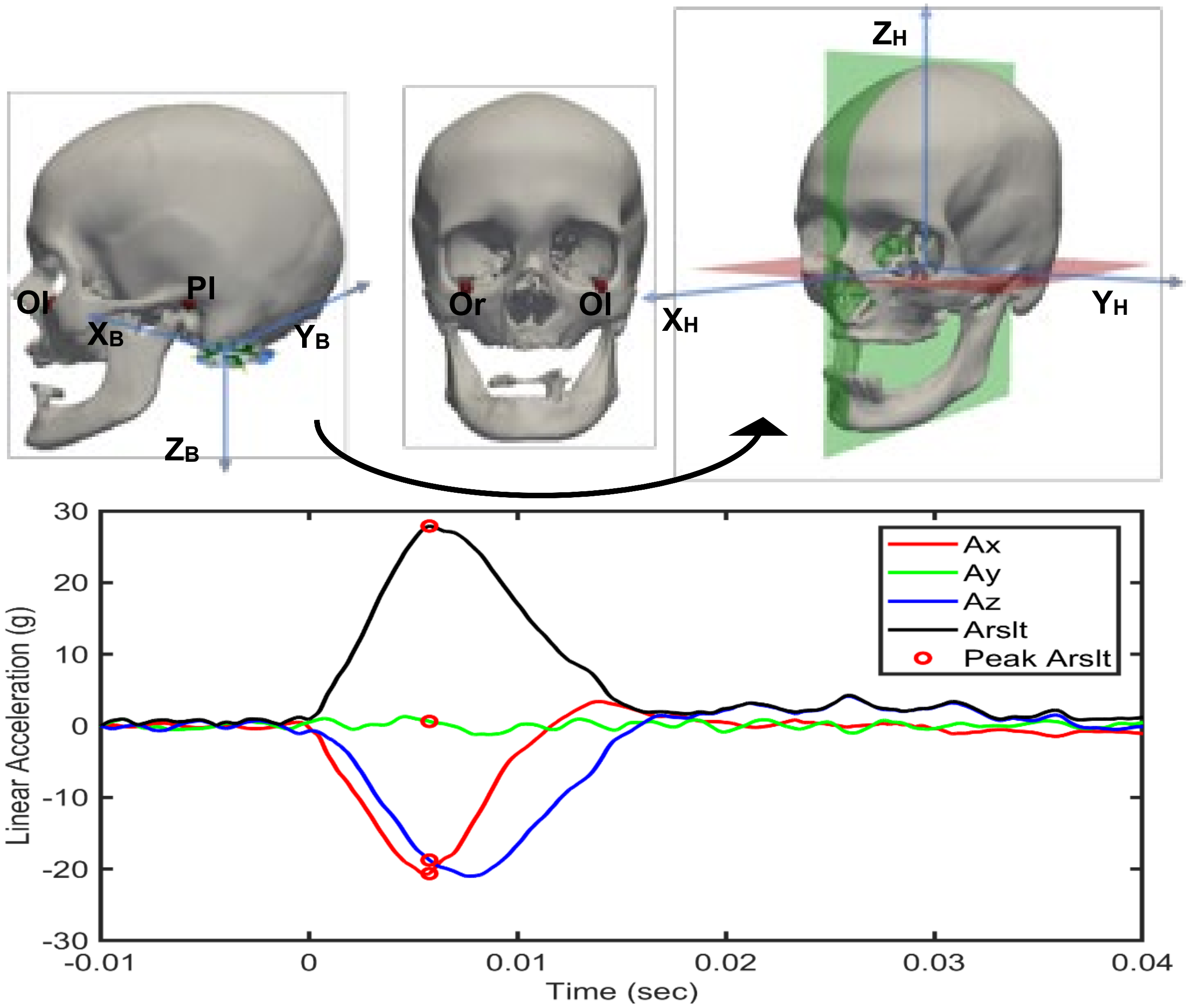


Figure 4.3: Reference block transformation to head COG process. The right and left orbitales and porions (Or, Ol, Pr, and Pl) were measured along with the block sensor orientations (X_H, Y_H, and Z_H) to establish the Frankfurt plane and head center of gravity for our three cadaver heads. Block sensor signals were transformed to the head coordinate system (X_H, Y_H, and Z_H). The components at the peak resultant linear acceleration were used to calculate impact location azimuth and elevation.

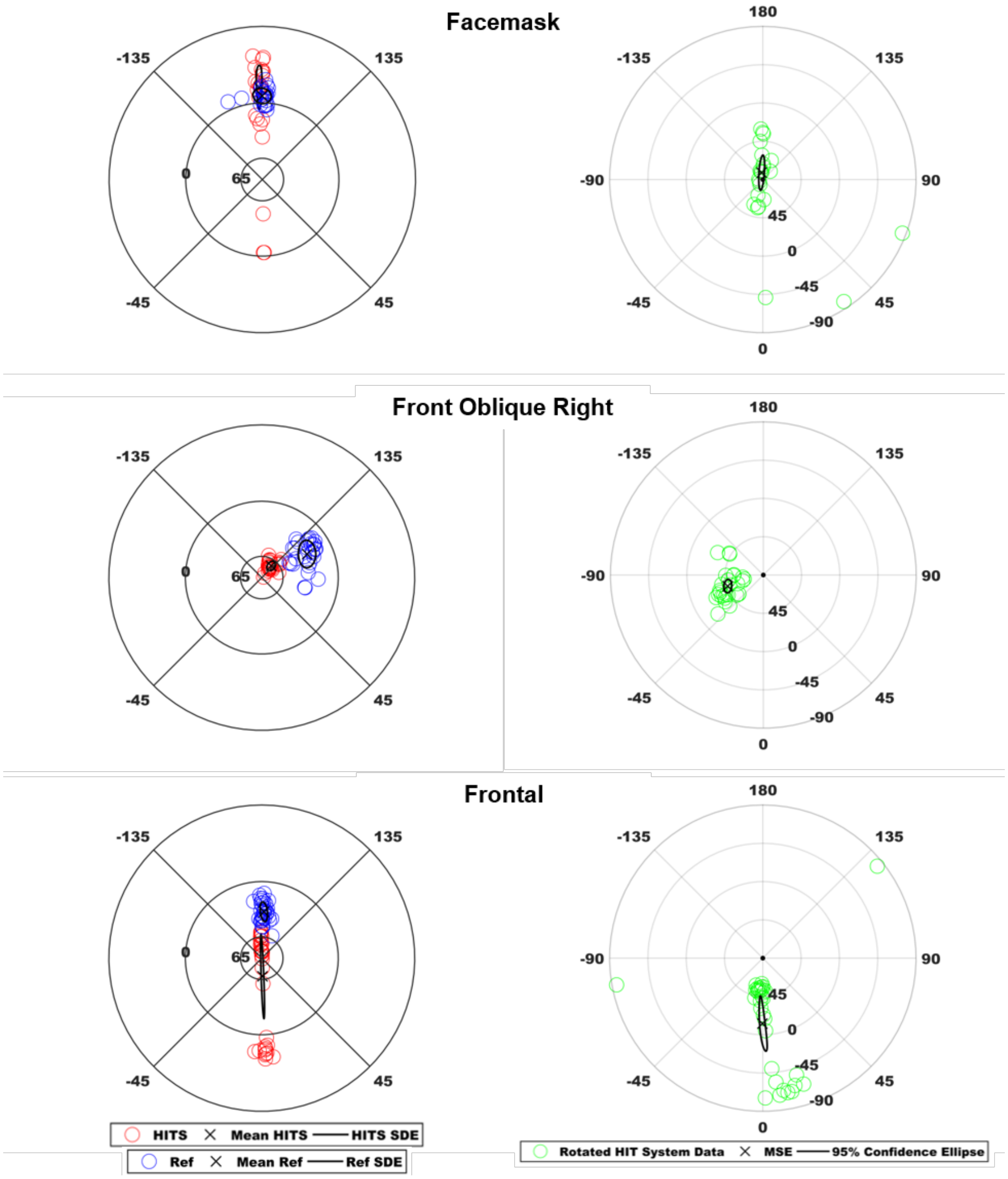


Figure 4.4: Polar plots for individual impact location coordinates (left) for Head Impact Telemetry (HIT; red) System and Reference (Ref; blue) data with corresponding rotated HIT System (green) data for spherical statistical analysis (right) for facemask, front oblique right, and frontal drop sites. Figures on the right denote azimuths and elevations similar to the HIT System coordinate convention. Data outside of the zero axis are impacts collected below the Frankfurt Plane. Figures on the right show the individual location coordinate difference between the HIT System and Reference data after rotating both datasets so that the reference data aligned with the pole (90° elevation; black dot). The mean spherical error (MSE) is contained within a 95th percentile confidence ellipse. If the 95th percentile confidence ellipse contained the pole, then the impact location coordinates collected by the HIT System were not statistically different from the reference data.

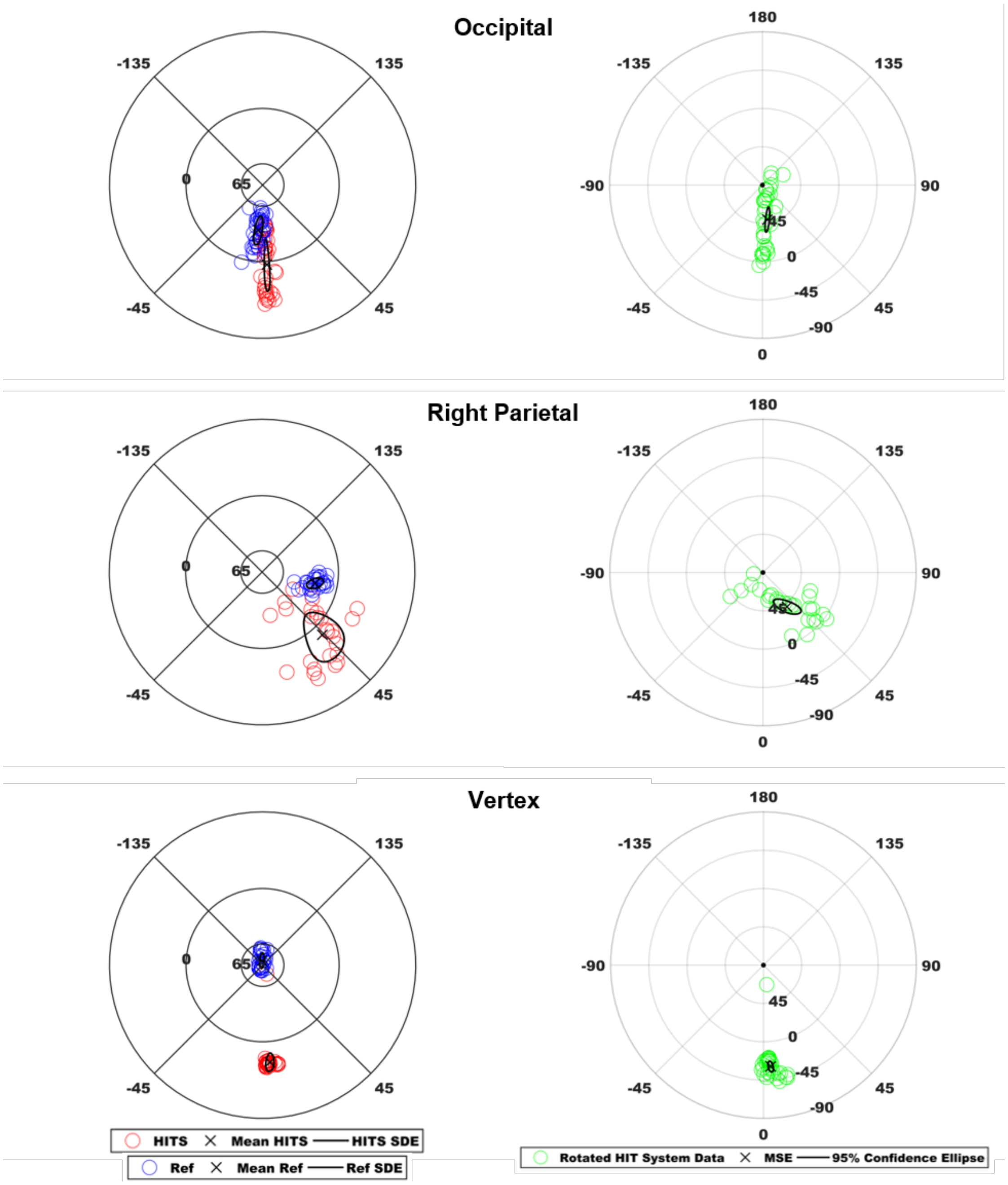


Figure 4.5: Polar plots for individual impact location coordinates (left) for Head Impact Telemetry (HIT; red) System and Reference (Ref; blue) data with corresponding rotated HIT System (green) data for spherical statistical analysis (right) for occipital, right parietal, and vertex drop sites. Figures on the right denote azimuths and elevations similar to the HIT System coordinate convention. Data outside of the zero axis are impacts collected below the Frankfurt Plane. Figures on the right show the individual location coordinate difference between the HIT System and Reference data after rotating both datasets so that the reference data aligned with the pole (90° elevation; black dot). The mean spherical error (MSE) is contained within a 95th percentile confidence ellipse. If the 95th percentile confidence ellipse contained the pole, then the impact location coordinates collected by the HIT System were not statistically different from the reference data.

CHAPTER 5

Manuscript 2

Quantifying the Head Impact Telemetry System's Impact Detection and Location Accuracy During High School Football Special Teams Plays Using Video Analysis

Introduction

High school athletes represent one of the largest athletic cohorts in the United States, with almost 8 million student-athletes participating annually.[102] Concussions account for 15% of all sport-related injuries sustained by high school athletes,[103] and may influence neurocognitive development that occurs throughout adolescence.[104] These reasons make it significant and worth studying this population's health and safety. High school football has over 1.1 million participants annually and high concussion incidence rates, ranging from 4.7 to 9.4 concussions per 10,000 athletic exposures over the last 15 years.[102,103,105–108,110,111] Football's high collision environment exposes players to frequent head impacts and thus a risk for concussions across youth, high school, college, and professional populations.[8,108,112–114] Player-to-player contact is the primary concussion injury mechanism.[108,110,267] Reducing or eliminating contact during practices and ensuring proper football tackling and blocking techniques are methods that may reduce concussion risk in youth and high school football populations.[108,110,115,116] Modifying existing rules or introducing new rules limiting the potential for player collisions and other high energy contact is another method that may lower concussion incidence, especially during competition.[117,118]

Special teams plays (i.e., kickoffs and punts) commonly lead to high energy player collisions due to the large closing distances between opposing players on the field.[6]

Kickoff/kickoff return and punt/punt return account for 10-20% of concussions in professional and high school football,[7,119,121] and, on a play by play basis, special teams plays present a 1.7 to 4.3 times higher concussion risk compared to injuries during run and pass plays in professional football.[119,121] As a result, kickoffs were moved from the 30 to the 35-yard line in professional football, which increased the number of touchbacks in professional football by 32% and limited the opportunities for players to receive potential concussions on kickoffs.[120] This rule modification was supported by data from helmet-based accelerometers in collegiate football that showed collisions occurring over long closing distances were the most severe.[6]

The Head Impact Telemetry (HIT) System is the most commonly used head impact sensor in football. As noted above, this system has been used to inform rule changes at the professional level.[6] However, the effectiveness of these data-derived rule modifications hinges on the sensor's accuracy in measuring the head impact biomechanics. In general, accurate head impact magnitude and location data from head impact sensor measurements require that the sensor (1) has sufficient bandwidth and amplitude range to measure the loading environment,[47] (2) couples securely to the head to reduce measurement error from head-sensor relative movement,[94] and (3) uses some software algorithm to retain true head impacts and remove data collection trigger events not related to a head impact.[64,99,100]

In laboratory testing using anthropometric test device (ATD) headforms, the HIT System had variable performance in measuring the impact location of impacts directed to the helmet facemask and crown.[11–13] The HIT System recorded those impacts as back of the helmet instead of what should have been the correct impact location of front or top.[11–13] However, the HIT System may measure facemask and crown loading differently when on a human head (rather than an ATD) due to head shape and head-helmet coefficient of friction differences between a human head and an ATD.[15,33,95] In addition, laboratory evaluations using the football HIT System demonstrated that between 4% to 25% of data can be missed due to data collection trigger failure or incorrect removal of data by the filtering algorithm.[11,13,54] The HIT System's impact filtering algorithm accuracy has not been quantified in an on-field setting and may be examined using video analysis.

It is important to have accurate impact data to understand head injury mechanisms in terms of impact frequency, magnitude and directional loading to create or modify data derived rules for player safety. The purpose of this study was two-fold. Our First Aim was to quantify the HIT System's head impact filtering algorithm accuracy in an on-field environment during high school football special teams plays. We used video synchronized with HIT System data to estimate the true positive, false positive, false negative, and true negative rates for impacts collected during special teams plays and processed with the HIT System's impact filtering algorithm. Additionally, our Second Aim was to quantify agreement in impact location between a video reviewer and the HIT System. This study examined the HIT System's performance in real-world settings (rather than laboratory settings). However, video review, even by a trained observer, is not without error, and thus these true positive, false positive, false negative, and true negative rates, and location agreement statistics, should be interpreted as estimates.

Because this study used video analysis as a proxy gold standard, we limited all analyses to special teams plays only. As noted above, special teams plays represent a phase of play in football with high concussion risk. From a pragmatic standpoint, special team plays are more likely to be open field plays and are therefore more amenable to video analysis, relative to impacts to linemen during a passing or running play.

Methods

Data Collection

Head impact biomechanics data used in this study were collected during the 2017 season from a single high school football team. Eligible participants for the study wore either a Riddell Revolution, Speed, or Speed Flex helmet to accommodate a HIT System encoder and were members of the high school football team. The study was approved by the University of North Carolina at Chapel Hill's Office of Human Research Ethics. Athlete consent and parental assent were required prior to study enrolment. Enrolled participants' helmets were instrumented with a HIT System encoder prior to the beginning of the competitive season and informed consent regarding the use of video filming of practices and games was obtained. An athletic trainer at the school or a research assistant initiated the HIT System for data collection for all practices and games for the 2017 season. There were 646 impact data collection trigger events from the HIT System with 371 impact data collection trigger events classified as valid by the HIT System's impact filtering algorithm. These impact trigger events were collected from 22 players participating in 218 special teams plays (58 kickoff cover plays, 54 kickoff return plays, 50 punt cover plays, and 56 punt return plays).

The head impact biomechanics data collected by the HIT System comprised of both valid and invalid impacts per HIT System's impact filtering algorithm. Six accelerometers within the HIT System encoder continuously sampled during a game until one of the accelerometer channels exceeded a 14.4 g acceleration threshold. Data were collected for 40 ms at 1 kHz and transmitted to a sideline computer where a proprietary algorithm determined the head impact kinematics, impact location, and impact algorithm validity.[29,30,184,196] Impact location category was determined according to the impact's azimuth and elevation coordinate (**Figure 5.1**). The HIT System's algorithm determined impact validity using two criteria. First, the data collection trigger event required a peak resultant linear acceleration greater than 10 g. A single accelerometer can measure an acceleration greater than the 14.4 g threshold, but the remaining five accelerometers can measure less acceleration. This can lead to peak resultant linear accelerations less than 10 g when the algorithm calculates impact kinematics. Trigger events less than 10 g of peak resultant linear acceleration are associated with running or jumping movements and not classified as a head impact.[35,182,184,268] The second criterion required that the acceleration pulse for a data collection trigger had characteristics of an impact to a helmeted head based on rigid body dynamics.[184,196] The second criterion's intended purpose removed data collection trigger events where the helmet was not on the head. Once the algorithm determined that a trigger event satisfied these two criteria, it was available in real-time on the sideline computer and later via the HIT System's cloud-based web portal (Redzone). Algorithm invalid impacts were stored locally on the sideline data collection laptop in LinkStatus Log files. LinkStatus Log files tracked when a session started, stopped, and any communication that occurred between the HIT System encoder in the helmets and the sideline data collection laptop including data collection trigger events. Each data collection trigger event, within the LinkStatus Log File, contained a date/time stamp for the trigger event and an identification number associated with that player's specific sensor. The presence of a data collection trigger event date/time stamp within the LinkStatus Log file that did not appear in the dataset downloaded from RedZone indicated that these data were deemed invalid per the HIT System's algorithm validity criteria.

Our research team filmed all the games for the high school football team participating in the study. We used video cameras (Canon VIXIA HF M30 & R100 video cameras; Canon Inc., Tokyo, Japan), and recorded high definition film from the side of the field at the highest available vantage point. Games were recorded at 60i (60 interlaced fields per second) with a tight angle allowing for better resolution of the head impacts player's sustained during competition. Our research team filmed all games with a continuous feed to easily synchronize the film with the HIT System impact data according to the procedure outlined by the National Institute of Neurological Disorders and Stroke common data elements for video confirmation of biomechanical devices used in traumatic brain injury research.[269] All research team members responsible for filming games underwent training prior to the team's first game. Our team members have collected video across multiple years and this experience was leveraged to obtain consistent game film quality. Only video for special teams plays for the instrumented high school football team in the 2017 season were evaluated for this study.

Video Review

A single reviewer analyzed on-field special teams video data using VLC media player (version 2.2.8) independent of the data generated by the HIT System including the head impact kinematics, location, and algorithm validity data. Our video reviewer for this study (KC) had ten years of football playing experience with an additional four years of reviewing impacts captured by video and the HIT System. We used video assessment questionnaires similar to previous studies (**Appendix B to H**).[6,59,270] Separate analyses were used for evaluating the HIT System's head impact filtering algorithm to detect impacts (Aim 1) and impact location accuracy (Aim 2). Both video review analyses were completed for all special teams plays over 12 games in the 2017 season.

Impact detection

The first Aim used the video-assessment of whether or not an impact occurred as the proxy gold standard with the goal of estimating the HIT system's impact detection accuracy. The video reviewer watched each instrumented player through an entire special teams play and documented camera times when an instrumented player sustained an impact capable of triggering HIT System data collection.[69] The reviewer recorded camera times when the instrumented player was no longer in view of the camera and the duration they were out of view. Video reviewed impacts had to meet all inclusion criteria outlined in **Table 5.1**. The video reviewer logged camera times of video observed head impacts to a data collection form for merging video analysis data with HIT System valid and invalid data collection impact trigger events. Only impact trigger events that satisfied all inclusion criteria were merged with HIT System data (**Table 5.1**).

Impact Location Agreement

The second aim was limited to the universe of head impact events recorded by the HIT System. The goal was to estimate the HIT system's impact location accuracy.[6,59,270] The video reviewer entered synchronized video-data collection trigger event camera times into VLC's Jump to Previous extension application for video analysis. Impacts analyzed with this approach had to meet all the inclusion criteria outlined in **Table 5.2** before the reviewer categorized the video observed impact location according to the HIT System's impact location definition (**Figure 5.1**). Video analysis data were paired with the head impact biomechanics data using a unique impact identification number. Only impact trigger events that satisfied all inclusion criteria were merged with HIT System data (**Table 5.2**).

Aim 2 used an expert adjudication process to determine impact location based on the video. Our video reviewer with the football playing and video evaluation experience (KC) identified 130 impacts that met all inclusion criteria outlined in **Table 5.2** from head impact events recorded by the HIT System. A second reviewer then analyzed the same 130 impacts for only impact location. A third video reviewer then analyzed impacts where the two initial video reviewers disagreed (n = 11). We reached a video observed impact location consensus for an impact when at least two reviewers agreed on the location. For one of the 130 impacts, all three reviewers disagreed on the impact location, and we removed this impact from the final dataset for statistical analysis because of this ambiguity in assessing impact location from video.

Inter- and Intra-Reliability Assessment

As noted above, video review of on-field play has some degree of error. To quantify the reliability of video review, the reviewer watched and reviewed a single football game twice to establish intrarater reliability for documenting impacts (Aim 1). The first and second video review sessions were separated 30 days apart. The video reviewer replicated 92% of the data collection trigger and non-trigger events between the two video review sessions and reliably applied the inclusion criteria to head impact events for later merging with HIT System data (**Table 5.1**).[59,270,271]

For the location assessment (Aim 2 – limited to the impacts recoded by the HIT system), our video reviewer watched and reviewed 60 impact data collection trigger events to establish their intrarater reliability. These 60 impacts came from multiple games evenly distributed across the four special teams play types and were reviewed on two sessions separated by 30 days. We determined our video reviewer's reliability on applying the inclusion criteria to the 60-impact data set, and reliability on analyzing the video observed impact location for impacts meeting all inclusion criteria (**Table 5.2**). The video reviewer reliably applied the inclusion criteria to the 60-impact data set (**Table 5.2**) and demonstrated a 90% agreement between the two sessions on documenting the HIT System impact location (unweighted Kappa statistic (k) = 0.82).[59,270,271]

We also assessed interrater reliability using similar methods. An additional video reviewer analyzed the same 60-impact data set for inclusion criteria and video observed head impact location. The interrater percent agreements and kappa statistics were less than the intrarater percent agreements and kappa statistics (**Table 5.2**). The two raters demonstrated moderate agreement (73%) on categorizing impact locations observed in the 60-impact data set (unweighted k = 0.52).[271]

Data reduction & statistical analyses

Head impacts observed on video that were documented with the video as a gold standard proxy and impacts measured by the HIT System were categorized according to the definitions in **Table 5.3.** In addition to estimating the true positive, false positive, false negative, and true negative rates, we estimated the HIT System's impact algorithm filter's sensitivity, specificity, positive predictive value, and accuracy during special teams plays with the equations 1-4 below:

$$\text{Estimated Sensitivity} = \frac{\text{TP}}{\text{TP+FN}} \tag{1}$$

$$\text{Estimated Specificity} = \frac{\text{TN}}{\text{TN+FP}} \tag{2}$$

$$\text{Estimated Positive Predictive Vale} = \frac{\text{TP}}{\text{TP+FP}} \tag{3}$$

$$\text{Estimated Accuracy} = \frac{\text{TP+TN}}{\text{TP+FP+FN+TN}} \tag{4}$$

where TP is the number of true positives, FN is the number of false negatives, FP is the number of false positives, and TN is the number of true negatives.

For Aim 2, we calculated the percent agreement and used an unweighted Kappa agreement analysis to determine the head impact location agreement between a video reviewer and the HIT System.[70,271] An unweighted Kappa statistic of one indicated perfect agreement between the video observer and the HIT System. We carried out our statistical analyses in SAS (Version 9.4, SAS Institute Inc.), and set an a priori alpha level of 0.05.

Results

Impact Detection Rate (Aim 1)

A total of 1500 player-plays were reviewed, and we determined 495 impacts occurring on video. Of the 495 impacts we identified on video, 316 matched with an impact trigger event from the 646 HIT System impact data collection trigger event data set. Therefore, 330 HIT System impact trigger events had no corresponding impact observed on video. The 646 HIT System impact trigger events that had the matched video observed impacts (n = 316) and trigger events with no video observed matched impacts (n = 330) were reduced to 317 trigger events after applying the inclusion criteria from **Table 5.1**.

The 317 trigger events were categorized according to **Table 5.3** and are presented**.** An estimated 70% of the 317 impact trigger events were accurately classified as true head impacts and non-head impacts by the HIT System's impact filtering algorithm. A larger proportion of estimated false negatives occurred among trigger events incorrectly classified by the HIT System's impact filtering algorithm. The HIT System impact filtering algorithm had an estimated 69% sensitivity in detecting true head impacts observed on the video, and an estimated 72% specificity in classifying non-head impact motions observed on video as non-head impact events. Finally, an estimated 88% positive predictive value indicated the likelihood that an impact classified as true head impact by HIT System's impact filtering algorithm was actually a true head impact observed on video.

Impact Location Agreement (Aim 2)

Of the 646 impact trigger events that occurred on special teams plays, our video reviewer identified 198 impact trigger events that met all inclusion criteria for further quantifying video observer impact locations (**Table 5.2**). Of the 198 included impact trigger events, 130 were classified as valid by the HIT System's impact filtering algorithm and provided impact kinematic and location information. The video from these 130 impacts underwent review by two video reviewers for determining a video observed impact location. The two video reviewers agreed on the video observed impact location on 119 impacts. A third reviewer analyzed the video for the impact location on the 11 impacts that the two initial video reviewers disagreed on. The third reviewer agreed with a video observed impact location on 10 out of the 11 impacts. Therefore, at least two video reviewers agreed on the video observed impact location for 129 out of the 130 (99%) impacts included for statistical analysis.

The impact location measured by the HIT System agreed with our impact location observed from video on 82 of the 129 (64%) reviewed impacts. The HIT System and our video observed impact location had weak agreement according to the unweighted kappa agreement statistic suggesting that 15% to 35% of the impact locations measured by the HIT System and determined by a video observer were reliable. (unweighted k = 0.43, 95% confidence interval: 0.31 – 0.54). Most of the disagreements on impact location between the HIT System and the video observed impact location differed by one location category (**Figure 5.2**). However, four impacts observed on video occurring to the front of the head were recorded as back by the HIT System, two impacts occurring to the back of the head on video were recorded as impacts to the front of the head by the HIT system, and one impact was observed occurring to the right of the head on video but the HIT System recorded it as a left impact.

Discussion

This is the first investigation to report on the HIT System's impact detection rate for special teams plays. The HIT System's impact filtering algorithm accurately categorized 70% of the video-observed head impacts as head impacts. Importantly, 23% of the video observed impacts were categorized as non-head impact events by the HIT System filtering algorithm. It is important to note our results apply to special teams plays, a phase of play with high risk of concussion that is more amenable to video analysis than other football phases at the high school level. These results indicate that studies using the HIT System as a head impact data collection system could underestimate head impact frequency during special teams plays, and possibly other play types.

This is also the first study to attempt to estimate the HIT System's impact location measurement accuracy. Almost 95% of the impact locations observed on video and measured by the HIT System either agreed on the exact or adjacent impact location region. However, caution is warranted in interpreting this result, since the regions are large. Only 52% of impacts disagreeing on impact location between the video observer and the HIT System occurred within ± 22.5° of azimuth of an impact location category boundary or within 25° of elevation along the top location category boundary.

In this study, the HIT System impact filtering algorithm's estimated sensitivity, specificity, positive predictive value, and accuracy were less than those reported for a mouthguard system used in college football players.[99] These differences could be attributed to our study only describing algorithm performance for special teams plays where the loading environment has a wide range, due to the short and long closing distances experienced across the different special teams play types.[6,270] From a technical standpoint, the mouthguard system used an infrared sensor within the mouthguard to automatically remove non-head impact triggers when the mouthguard was not on the teeth.[99] Additionally, the mouthguard system used a machine learning program to classify true and non-head impact events. The program used power spectrum densities and wavelet transform features from the linear acceleration and angular velocity data collected from impacts delivered to collegiate football players while wearing the mouthguard.[99] These spectral and wavelet transformations showed that true impact events had higher amplitudes at lower frequencies, while non-impact events had amplitudes and oscillations at higher frequencies.[99] Developers of another mouthguard head impact sensor also demonstrated non-impact events or false positives exhibited high-frequency content within the measured kinematic signals.[101] While using linear acceleration thresholds does not sufficiently discriminate between true head impact and non-impact events,[64–66] the frequency content within the measured kinematic signals holds promise for correctly classifying true head impact events.[99–101]

There were 73 impacts observed on video that were categorized as non-head impact events according to the HIT System impact filtering algorithm (**Table 5.3**). This is a higher estimated false negative rate in football than documented false negative rates in soccer on evaluations using the X2 Biosystems xPatch.[64] There are clear differences in the hardware configurations (six single axis linear accelerometers in the HIT System vs. tri-axial linear accelerometer and gyroscope in the xPatch), head coupling methods (in-helmet with the HIT System vs. head based with the xPatch), impact filtering algorithms, and loading environments (special teams high school football for the HIT System vs. collegiate soccer for the xPatch) that could reasonably explain differences in false negative rates. The studies also differed on the approach for cross-referencing impact triggers with video. We evaluated the video for head impacts independently of the HIT System trigger events.[69] This meant that when we tracked a player on a special teams play, we had no prior knowledge that the HIT System triggered for data collection. Other approaches used synchronized date-time stamps from the head impact sensor and video to jump to points in the video when the head impact sensor triggered for data collection.[64–66] Our approach removed potential bias from the video observer for determining if a true impact had occurred on video or not during the review process.

Our data indicate that the HIT System and a video observer moderately agreed on determining impact location.[271] One reason for the disagreement on impact location between a video observer and the HIT system is the acknowledged subjectivity of observing impact locations on video. This is a limitation of any video review study. The HIT System determined a quantitative impact location category based on the azimuth and elevation coordinates calculated from impact acceleration (**Figure 5.1**). Our video observer used landmarks on the helmet from **Figure 5.1** to determine a qualitative impact location category. Therefore, some error level could exist on determining the impact location through video observations. However, impacts included in the location agreement analysis between the HIT System and the video observed impact location had to meet strict inclusion criteria. These criteria included 1) unobstructed impact views, 2) observed head contact to the instrumented player, and 3) observed head impact locations that could be clearly categorized by a HIT System location. Oblique impacts directed to the boundary between two or more impact location categories were excluded to avoid increasing the chance for disagreement between the HIT System and the video observed impact location on a difficult area to judge. With these strict criteria, 130 impacts were included for impact location and statistical analysis, and at least two separate video reviewers independently agreed on 129 out of the 130 impact locations observed on video. This approach ideally decreased the subjectivity associated with categorizing impact locations according to video and allowing for stronger interpretations of our results.

The HIT System determined that 85% of the impacts were to the front and side locations, and our video observer determined a similar percentage with almost 90% of the impacts directed onto the front and side locations. A previous study showed similar impact location distributions determined from video-based methods and head impact sensor methods.[69] The HIT system could be useful for estimating population based impact location distributions for special teams plays where many impacts, on the order of thousands, are collected. Impact location for rarer outcomes, such as the impact location distribution for impacts that cause concussion, should be corroborated with video analysis by multiple video reviewers when feasible.

The results of our investigation must be framed within its methodological limitations. First, we used a single camera view to analyze potential head impacts and head impact location through video analysis. An additional camera set up in the endzone would have allowed for better impact and impact location identification for collisions where players were moving sideline-to-sideline.[69,99] A single camera view has been used in other studies that paired head impact biomechanics with video, and a camera set up on the sideline is the preferred angle because most of the player movement in football is endzone-to-endzone.[59,270] We used a tight camera angle to provide the detail required for resolving video observed impact locations. This came at the expense of not evaluating impacts occurring to players out of view of the camera. However, we captured 549 out of 646 (85%) of the impact trigger events collected by the HIT System during special teams plays. Our on-screen quantified impact percentage is higher than previous investigations quantifying video-based head impact characteristics in high school (56%) and youth football (80%) across all play types.[59,270] Documenting head impacts and head impact location from video, independent of the impacts measured by the HIT System is subjective. We used strict inclusion criteria and impact definitions in order to conservatively assess video observed impacts and impact locations. Our video reviewer had over 10 years of football playing experience (4 at the college level) and over four years of video-based impact analysis. They also conservatively applied inclusion criteria to observed impacts and impact locations to develop the data sets for statistical analysis. Fewer impacts passed the inclusion criteria for the location analysis (33/60) for our experienced video reviewer as compared to a second video reviewer who passed more impacts for the location analysis (45/60) based on the inclusion criteria during our interrater reliability assessments. Additionally, multiple reviewers assessed our conservatively identified head impact dataset of 130 impacts for impact location, and at least two reviewers independently agreed on 99% of the head impact locations observed on video. This provided further confidence that our video observed impact locations we used in our agreement analysis with the HIT System's impact location measurements were less subjective. Importantly, our results currently apply to special teams plays in high school football only. The average impact magnitude differ between special teams and run or pass plays, and at different levels of the sport.[6,270] Future research is needed to estimate the HIT System's impact filtering algorithm and location measurement accuracy on run and pass plays and in professional, collegiate, and youth settings.

In conclusion, this study provides data on the performance of the HIT System's impact filtering algorithm and location measurement accuracy for a football play type (special teams) associated with increased concussion risk. We estimate that the HIT System's impact filtering algorithm correctly categorized 222 (70%) of 317 impacts as true data collection trigger events, relative to video analysis of special teams high school football plays. Furthermore, a high proportion (23%) of the head impacts observed on video were categorized as non-head impact events by the HIT System's impact filtering algorithm. We caution that video review is a proxy gold standard and has error, therefore, we believe these findings should be treated with caution. From a research perspective, there is need for impacts and impact locations to be accurately measured in order to quantify the impact loading environment for concussion and for data-informed rule changes aimed at reducing concussion risk.

Tables

Table 5.1: Inclusion criteria used to determine if a potential head impact trigger occurred while watching video for the video gold standard proxy analysis. Intrarater agreements on determining potential impact trigger events and inclusion criteria used to include trigger events for analysis were determined by analyzing video of a single game 30 days apart

Label (Choices)	Inclusion Criteria	Intra-Rater Agreement			
		Percent Yes		Percent Agreement	Kappa (95% CI)
		Session 1	Session 2		
Potential Trigger (Yes, No)	Clear evidence of helmet contact and/or head motion	54%	47%	92%	0.82 (0.74, 0.91)
On Screen (Yes, No)	Player must be within the camera view	52%	48%	92%	0.86 (0.79, 0.93)
Unobstructed (Yes, No)	There must be a clear, unobstructed view	71%	77%	89%	0.72 (0.53, 0.90)

Table 5.2: Inclusion criteria used to determine further head impact video analysis for determining a video observed head impact location. Included are interrater and intrarater agreements on applying the inclusion criteria to a 60-impact data subset and the interrater and intrarater agreement on categorizing video observed impact locations according to the HIT System definitions.

			Interrater Agreement				Intrarater Agreement			
Label (Choices)	**Inclusion Criteria**	**N**	**Percent Yes Rater 1**	**Percent Yes Rater 2**	**Percent Agreement**	**Kappa (95% CI)**	**Percent Yes Session 1**	**Percent Yes Session 2**	**Percent Agreement**	**Kappa (95% CI)**
On Field (Yes, No)	Player must be on the field	60	92%	97%	95%	0.55 (0.11, 0.99)	92%	92%	100%	1.00 (1.00, 1.00)
On Screen (Yes, No)	Player must be within the camera view	60	80%	82%	98%	0.94 (0.84, 1.00)	80%	80%	100%	1.00 (1.00, 1.00)
Unobstructed (Yes, No)	There must be a clear, unobstructed view	48[a]	83%	96%	85%	0.32 (-0.02, 0.66)	83%	81%	98%	0.93 (0.79, 1.00)
Impact Evidence (Yes, No)	Clear evidence of helmet contact and/or head motion	48[a]	83%	94%	85%	054 (0.08, 1.00)	83%	92%	88%	0.44 (0.07, 0.80)
HIT System Location (Yes, No)	A clear HIT System impact location is observed on the player's helmet	48[a]	75%	94%	77%	0.19 (-0.10, 0.36)	69%	75%	94%	0.85 (0.68, 1.00)
HIT System Location (Front, Top, Right, Left, Back)	N/A	30	67%[c]	60%[c]	73%	0.52 (0.24, 0.80)	62%[b,c]	59%[b,c]	90%[b]	0.82[b] (0.63, 1.00)

a – Denotes that up to 12 impacts occurred off the screen and out of camera view. Therefore, only 48 impacts were assessed according to the remaining inclusion criteria

b – Denotes that 29 impacts were common to both session 1 and session 2 after a 60 impact reliability data set was evaluated for inclusion for further video analysis

c – Denotes that the front was the most common impact location observed in the interrater and intrarater assessments. Percentage of impacts categorized with the front location are within the Percent Yes columns within the table.

Table 5.3: True Positive (TP), False Positive (FP), False Negative (FN), and True Negative (TN) definitions used to categorize Head Impact Telemetry (HIT) System impact trigger events cross-referenced with video observed impacts for High School Football Special Teams Plays on one central North Carolina team, 2017.

		Video Review	
		Head Impact Observed (n = 239)	**No Head Impact Observed (n = 78)**
HIT System Impact Filtering Algorithm Classification	**Valid Head Impact Classified by HIT System Algorithm (n = 188)**	**Estimated True Positives (TP)** (HITS Algorithm Correct) **n = 166**	**Estimated False Positives (FP)** (HITS Algorithm Incorrect) **n = 22**
	Non-Head Impact Classified by HIT System Algorithm (n = 129)	**Estimated False Negatives (FN)** (HITS Algorithm Incorrect) **n = 73**	**Estimated True Negatives (TN)** (HITS Algorithm Correct) **n = 56**

Estimated Sensitivity = 166 / 239 = 69%
Estimated Specificity = 56 / 78 = 72%

Figures

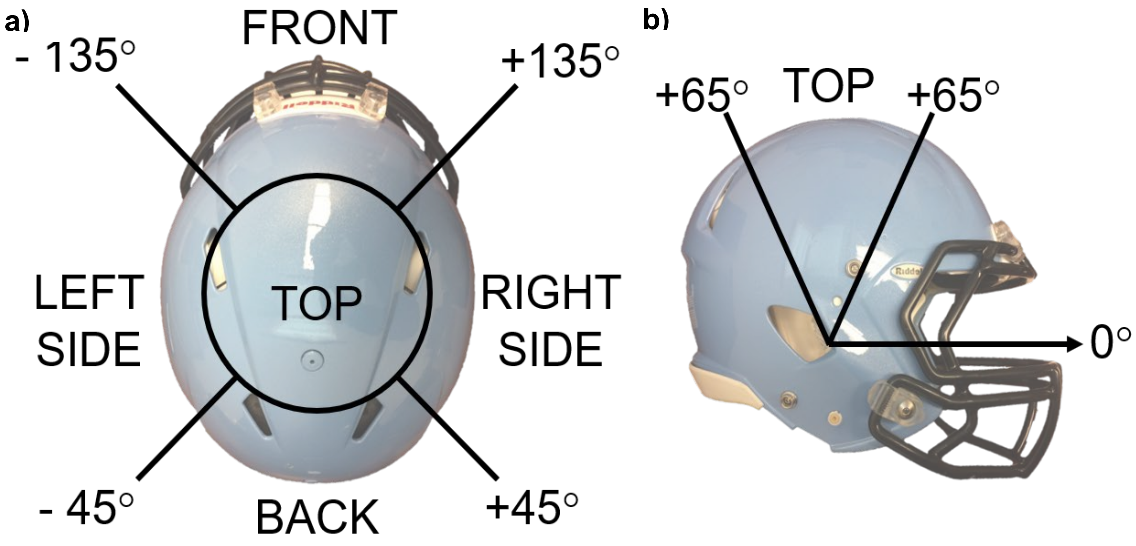


Figure 5.1: The HIT System impact location definition.. The system calculated head impact location in azimuth degrees around the head (a), in elevation degrees (b), and as an impact category. Impacts > 65° elevation are classified as Top impacts regardless of the azimuth degree calculation. Otherwise, impact location category was determined by the azimuth degree falling into one of the 4 location bins (Front, Right Side, Left Side, and Back).

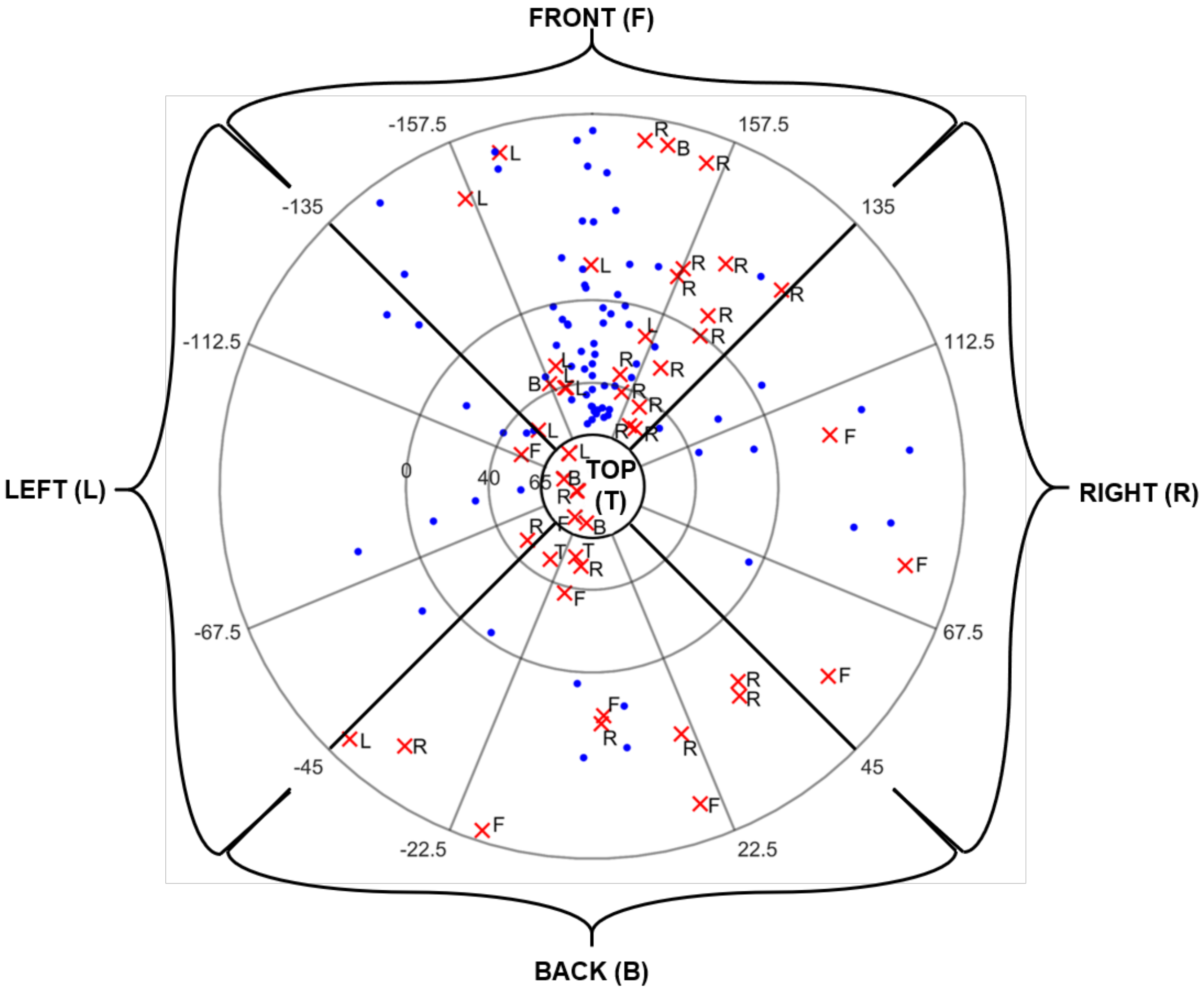


Figure 5.2: Summary of agreed and disagreed impact locations. Azimuth (numbers around figure) and elevation (numbers inside figure) coordinates for impacts where the Head Impact Telemetry (HIT) System and the video observed impact location agreed (blue dots), and disagreed (red crosses). Impact location categories at the edge of the figure describe the impact location assigned by the HIT System. Impact location for video observed impact location is shown next to the point for the disagreements.

CHAPTER 6

Dissertation Summary

Sport-related concussion is a major public health concern with just over a million concussions occurring yearly to children and adolescents under the age of 18.[17] Equally concerning is the potential late-life neurological sequelae and neurodegenerative diseases associated with recurrent concussions and repetitive head impacts.[1–3] Researchers created the Head Impact Telemetry (HIT) System, to measure the head impact frequency, location, and magnitude football players experience in competition.[30,188] This was in an effort to quantify the loading environment to understand concussion injury biomechanics and develop injury risk mitigation strategies.[6,8,57,177,188,252,257,272] Rule modifications protecting football players on special teams plays resulted from data collected with HIT System instrumented helmets.[6] The strength of HIT System data derived interventions and rule modifications to protect athletes depend on the system's measurement accuracy. The HIT System's measurement accuracy has been thoroughly evaluated with controlled impact testing in a laboratory environment but few studies have quantified the HIT System's measurement accuracy while used in an on-field setting.[11–13,54] Research evaluating the HIT System's outputs in testing environments that replicate its intended use on the field is needed if we are to use this sensor to develop data-driven rule modifications to make sports safer and quantify head impact exposures to understand potential risks for late-life neurological sequelae.

The Hybrid III anthropometric test device (ATD) has been the surrogate head choice for evaluating The HIT System's impact kinematic and location measurement accuracy.[11–13,54] One determinant for a sensor's measurement accuracy is how rigid the helmet couples to the head.[54,94] The jaw and cheeks are narrower in the Hybrid III compared with another ATD headform that has a closer approximation to a human head.[33] The frictional interface between the Hybrid III headform and a helmet is 2.5 times larger than a human head and helmet creating an unrealistically high coupling environment.[15,95] Due to these differences between the head/helmet fit and frictional interface, evaluating the HIT system using a human surrogate head model (e.g., cadaver) more representative of the on-field environment is required. Using an innovative biofidelic surrogate head testing paradigm the HIT System provided statistically different impact location coordinates and had low agreement on impact location category compared to reference sensors, with only accurately quantifying impact location from loading to the back of the head. Our data can inform concussion injury biomechanists modelling brain tissue deformations through finite element methods.[183,264] Currently, the six degree-of-freedom (DOF) head motion required for finite element head models need to be estimated from the HIT System's 5DOF motion outputs based on head impact regions.[264] Knowing the impact location coordinate error for the HIT System can help researchers that are developing impact region specific algorithms aimed at estimating 6DOF head motion from the HIT System. Additional on-field studies quantifying impact location category are required to build on our laboratory observations.

Our drop test method allowed us to deliver centroidal impacts through the cadaver head's center of gravity and represented an ideal condition for the HIT System's accelerometers within the encoder to contact the head and calculate head impact location. We built on our understanding of the HIT System at measuring head impact location outside of the laboratory environment in an on-field setting. We quantified the HIT System's impact detection and head impact location measurement accuracy while high school football players wore the system during special teams plays in games. This on-field evaluation introduced more variable head-helmet fits, frictional interfaces, and impact loading conditions that could not be easily replicated in a laboratory setting. The HIT System's impact filtering algorithm accurately categorized 70%

of the data collection trigger events as either true head impacts or non-head impacts after we cross-verified the data collection triggers with head impacts observed on video. Head impact frequency may be underestimated for studies using the HIT System during special teams plays, and potentially other play types. The HIT System agreed with video observations of impact location on 64% of 129 analyzed impacts. The overall impact location category agreement between the HIT System and our gold standard measurements (reference sensors in laboratory evaluations and video reviewer in on-field evaluations) was larger in our on-field evaluation (64%) as compared to our laboratory assessment (45%). We observed similar impact location category agreements when the HIT System measured impacts to the front in our laboratory (87%) and on-field evaluation (86%). We did not replicate our laboratory observed agreements (100%) in our on-field evaluation (55%) for impacts directed to the back of the head. Additional video review for impact location is required to further estimate the location accuracy of the HIT System for other impact locations as 50% of the 129 reviewed impacts occurred to the front.

Overall, the HIT System measured inaccurate head impact location coordinates when evaluated in our cadaver head drop testing paradigm. The HIT System had low to moderate agreement on categorizing impact location categories in both our laboratory and on-field evaluations. An important question to consider is what level of impact location accuracy is good enough from the HIT System? The answer depends on the HIT System user and their intentions with the system. Injury researchers may expect the impact location coordinate accuracy to be confined to the azimuth and elevation bounds defining an impact region if the intended use is integrating on-field biomechanical data in finite element or other injury modeling. For those researchers, coaches, or sport-safety legislatures quantifying the impact location distribution for concussions with the HIT System, then more confidence can be used with concussions measured to the front of the head as compared to concussions measured by the HIT System to the sides, back, or top. To the extent it is possible in a given research or clinical environment, we recommend that head impact locations are confirmed with video analysis to ensure accurate quantification of the head loading location environment related to concussion injury risk. Understanding the relationships between impact location and injury risk can lead to improvements in protective equipment, identifying athletes for technique improvement to limit head impact exposure and concussion risk, and develop data derived rule modifications for reducing concussion risk. Future work is required for understanding conditions that lead to missed impacts and misclassified impacts by the HIT System validation algorithm. We hope these findings lead to technology and software improvements aimed at addressing the HIT System's limitations. My research area will continue to focus on making sport participation safer for athletes by applying a mechanics based approach to understanding concussion injury risk. Continuing to develop accurate and reliable head impact sensor technologies is needed to accurately quantify the impact loading environments leading to head injury.

APPENDIX A : IMPACT DISTRIBUTION TABLE

TABLE A1: Participant distribution for proposed for impact video analysis

ID	Position	Games	Algorithm Valid Impacts	Algorithm Invalid Impacts	Total Impacts
215	OT	10	303	155	458
315	DE	8	173	196	369
515	OT	11	102	64	166
715	WR	4	32	38	70
1215	S	9	393	199	592
1515	RB/S	12	429	310	739
1715	CB	10	93	72	165
2115	OG	2	56	50	106
2215	ST	9	70	129	199
2315	QB	11	107	185	292
2415	OG	8	360	184	544
2515	C/LB	12	819	450	1269
3015	OT	12	624	371	995
3315	OG	6	66	113	179
3715	WR/ST	10	119	113	232
4415	LB	12	510	435	945
4815	CB	8	157	134	291
5315	DE	10	270	267	537
9516	RB	11	108	75	183
10016	QB	1	2	0	2
10116	RB	12	246	222	468
10216	DT	10	375	208	583
10416	DE	12	655	483	1138
10516	ST	10	16	22	38
11616	QB	12	103	71	174
12317	LB	1	20	29	49
26 Players			**6208**	**4575**	**10783**

APPENDIX B : NINDS CDE VIDEO SENSOR CONFIRMATION DATA ENTRY FORM

NINDS CDE Video Sensor Confirmation for TBI Research

[Study Name/ID pre-filled] Site Name:

***Subject ID:**

Note: The highly recommended CDEs have been listed below with asterisks (*) and bolded.

1. ***Subject age:** (years)
2. ***Start of data collection (date / time)**
   a. **Date of first event:**
   b. **Time of first event:** ☐am ☐pm ☐24-hour clock
3. ***End of data collection (date / time)**
   a. **Date of last event:**
   b. **Time of last event:** ☐am ☐pm ☐24-hour clock
4. ***Activity (Indicate all that pertain to subject):**
5. ***Camera manufacturer:**
6. ***Camera model:**
7. ***Camera resolution:**
8. ***Camera sample rate:**
9. ***Camera positions:**
10. ***Method of timestamp creation:**
11. ***Resolution of time-synchronization between video and device: (e.g. ±1 second, ±1 millisecond)** (Note: This is different from maximum allowable DeltaT between correlated video and device exposures – see specific instructions following the questions)**:**
12. ***Method of time-synchronization between video and device:**
13. ***Method of cross-verifying video and device exposures:**

☐Device as ground truth

☐Video as ground truth

☐Only impacts/exposures verified in both device data and video are considered

☐Other, specify

14. ***Method of analysis/link to correlate video and device exposures:**

☐Maximize exposure timing correlation after identifying all video and all device impacts/exposure(s)

☐Real-time stamp matching between video and device

☐Other, specify:

NINDS CDE Video Sensor Confirmation For TBI Research

[Study Name/ID pre-filled] Site Name:

***Subject ID:**

15. ***Maximum allowable DeltaT between correlated video and device exposures:**
16. ***Number of true positive exposures:**
17. ***Number of false positive exposures:**
18. Number of false negative exposures:
19. Of the true positive exposures, number of confirmed head to head exposures:
20. Of the true positive exposures, number of confirmed head to body exposures:
21. Of the true positive exposures, number of confirmed head to ground exposures:
22. Of the true positive exposures, number of confirmed head to object exposures:
23. Of the true positive exposures, number of confirmed body exposures:
24. Number of events that were unable to be classified:

APPENDIX C : NINDS CDE VIDEO SENSOR CONFIRMATION INSTRUCTIONS

NINDS CDE Video Sensor Confirmation CRF Module Instructions

GENERAL INSTRUCTIONS

Important note: The data elements noted with an asterisk (*) on this CRF Module are classified as Supplemental-Highly Recommended (i.e., strongly recommended for Biomechanical Devices in TBI clinical studies to collect). The remaining data elements are classified as Supplemental and should only be collected if the research team considers them appropriate for their study. Please see the Data Dictionary for element classifications.

Additional considerations include adding element to link the video clip to each impact and providing guidance on what video file type should be stored, how to facilitate the sharing of video file types, and consideration on privacy, de-identification or seeking consent for release of video clips.

*** DATA ELEMENTS ARE SUPPLEMENTAL-HIGHLY RECOMMENDED. ALL OTHER DATA ELEMENTS ARE SUPPLEMENTAL.**

SPECIFIC INSTRUCTIONS

Please see the Data Dictionary for definitions for each of the data elements included in this CRF Module.

- Subject Age - Enter subject age in years.
- Start of data collection - the date/time when data acquisition started
- Date of first recorded event (mm/dd/yyyy) and Time of first recorded event (hh:mm) - Report the time and date when the first event was recorded.
- End of data collection - the date/time when data acquisition ended

- Date of last recorded event (mm/dd/yyyy) and Time of last recorded event (hh:mm) - Report the time and date when the last event was recorded.
- Activity - Please indicate all activities under study which pertain to the subject (e.g. football, soccer)
- Camera manufacturer - Please indicate manufacturer of video camera(s) used for recording event footage.
- Camera model - Please indicate model of video camera(s) used for recording event footage.
- Camera resolution - Please indicate resolution of video camera(s) used for recording event footage.
- Camera sample rate - Please indicate sample/frame rate of video camera(s) used for recording event footage.
- Camera positions - Please indicate positions of video camera(s) used for recording event footage.
- Method of timestamp creation - Please indicate method of creating timestamps on video footage. The approach used to generate a timestamp for each individual video frame used to verify impacts. The timestamp is the time of day of a given video frame which will be compared with the time of day an event is recorded via the device. The timestamp can be generated by the camera using an internal clock (e.g., a GPS synched clock, an internal software clock, etc.) or can be tracked using an external source (e.g., a digital clock in view of the camera, calibrating the frames using a digital clock shown a single time, etc.). These methods are not exclusive, and other approaches may be described.

- Resolution of time-synchronization between video and device - Resolution of time-synchronization is the amount of error allowed between the device timestamp and video timestamp. Appropriate answers for this would be ±1second, ±1millisecond, etc. The resolution will depend on the accuracy of the device clock and the accuracy of the video clock (e.g., if the device timestamp is accurate to the millisecond, and the video timestamp is accurate to the second the resolution would be ±1second). Please indicate time resolution of time-synchronization between the video and the device. For example, if both video and device has 1 second resolution for their real-time stamps, the time resolution would be 1 second.
- Method of time-synchronization between video and device - The approach used to synchronize the timestamps of the device and video. Please choose or describe the method to synchronize video and device information. The time-synchronization between the device and video can be accomplished using several approaches. For example, both device and video could be synchronized with a third source (e.g., the NIST traceable time source time.gov, or localized computer time source), or they could be synchronized by forcing events on the devices in view of the camera and documenting any offsets in time. These methods are not exclusive, and other approaches may be described. The requested input is for a description of how the time-synchronization between the device and video was accomplished.
- Method of cross-verifying video and device exposures - The accelerometry device and video recording can independently capture exposure information and can be cross-verified to increase confidence of the exposure measurement. This CDE differentiates which set of information serves as the ground truth for verification. For example, if video is served as ground truth, exposures captured on video but not measured by the device would be considered as missing (false negatives). It is also an option to only consider exposures measured by both the video and the device to be 'verified' exposures. Please

select from following options or, if another method is used, provide a detailed description of the method.

- Method of analysis/link to correlate video and device exposures - Exposures measured by the device and those observed in video need to be linked with each other for verification. For example, if a sports player was observed to sustain a head impact at 10:30:56 am on video while wearing an accelerometry device, it is expected that the accelerometry device will have a recording corresponding to this observation. The method to link the exposures could include 1) identifying the time differences between exposures in video or device and finding the time-syncing difference to maximize the correlation between the video exposure timings and device exposure timings, 2) having a timestamp for each exposure on the video or device that is synchronized with a standard real-time clock (e.g. nist.gov time) and correlating exposures via the real-time stamp. Please choose from the following options, or if another method is used, provide a detailed description of the method.
- Maximum allowable DeltaT between correlated video and device exposures – Where DeltaT is the amount of time between an identified device/video exposure(s). Due to uncertainties in real-time stamps or time offset calculations, the timing of individual exposures may not have an exact match between video and device. For example, if there is a +/- one second uncertainty in the timestamp, it is possible that a video exposure at 12:30:45 may be matched with a device exposure at 12:30:46. This CDE specifies the amount of tolerance allowed for the difference between video and device time stamps. Indicate time offset in number of seconds between video and device time stamps allowed for linking exposures.

- Number of true positive exposures - Both video and device indicate an exposure(s)- happened within the allowable time-period (Maximum Allowable DeltaT - #15 above). Head impact events in which both the video and the device indicate an exposure. Through careful review of the video, identify head impact exposures. The definition of this will vary by the sport setting studied but could include identifiable change in the head kinematics (in the case of a head impact in football for example) or an identifiable change in the ball trajectory (in the case of a head to soccer ball impact). It is highly suggested that this process be conducting by multiple coders blinded to each other's' efforts. Have a master coder reconcile any differences in exposure identification. Count the number of head impact events recorded on the device that can be confirmed via video. This may be tied to g-force level (e.g., impact as 25g+).
- Number of false positive exposures - Head impact exposures recorded on the device but unable to be verified by video within the allowable time-period (Maximum Allowable DeltaT - #15 above). Through careful review of the video, identify head impact exposures. The definition of this will vary by the sport setting studied but could include identifiable change in the head kinematics (in the case of a head impact in football for example) or an identifiable change in the ball trajectory (in the case of a head to soccer ball impact). It is highly suggested that this process be conducting by multiple coders blinded to each other's' efforts. Have a master coder reconcile any differences in exposure identification. Count the number of head impact exposures recorded on the device that cannot be confirmed via video. This should only include those events in which the player for whom the exposure is recorded is visible on the video. It should not include unverifiable exposures out of frame of the video. This may be tied to g-force level (e.g., impact as 25g+).
- Number of false negative exposures - Head impact exposures observed on video but lacking corresponding device exposure data within the allowable time-period (Maximum

Allowable DeltaT - #15 above). Through careful review of the video, identify head impact exposures. The definition of this will vary by the sport setting studied but could include identifiable change in the head kinematics (in the case of a head impact in football for example) or an identifiable change in the ball trajectory (in the case of a head to soccer ball impact). It is highly suggested that this process be conducting by multiple coders blinded to each other’s' efforts. Have a master coder reconcile any differences in exposure identification. Count the number of head impact exposures identified on video that do not have any corresponding data on the device within the allowable time-period.

- Of the true positive exposures, number of confirmed head to head exposures - The number of visually verified head impact events that resulted from head to head contact (including helmet to helmet contact).
- Of the true positive exposures, number of confirmed head to body exposures - The number of visually verified head impact events that resulted from head to body contact (e.g., head contacts the torso of another person).
- Of the true positive exposures, number of confirmed head to ground exposures - The number of visually verified head impact events that resulted from head to ground contact (e.g., while falling or diving, head contacts playing surface).
- Of the true positive exposures, number of confirmed head to object exposures -The number of visually verified head impact events that resulted from head to object contact (e.g., head contacts the ball).

- Of the true positive exposures, number of confirmed body exposures - The number of visually verified events in which the person's body comes in contact with another person, the ground, or an object that result in an “acceleration event” of the head – without direct contact to the head.

- Number of events that were unable to be classified – These could include those events where there is device data but video data is not available (e.g. player out of frame, etc).

References

Cortes N, Lincoln AE, Myer GD, Hepburn L, Higgins M, Putukian M. Video Analysis Verification of Head Impact Events Measured by Wearable Sensors. American Journal of Sports Medicine. 2017;45:2379-2387.

Kuo C, Wu LC, Loza J, Senif D, Anderson S, Camarillo DB. Comparison of video-based and sensor-based head impact exposure. bioRxiv 235432. 2017. doi: https://doi.org/10.1101/235432

APPENDIX D : VIDEO ANALYSIS PROTOCOL FOR IMPACT LOCATION

VIDEO ANALYSIS PROTOCOL – Impact Location Assessment

Application Preparation Steps

1) Open Video file for current game analysis in VLC 2.2.8

- File name example: 1.1_08-18-17_Game1_Half1

2) Open the “Jump to Previous” application in VLC

3) Open impact list for sensor eval reliability datasheet

- Filename: team2_sp_teams_impact_list_for_sensor_eval_reliability3.xlsx

4) Open Qualtrics Survey – Sensor as Gold Standard Proxy Analysis

5) Begin Video Analysis

Video Analysis Steps

1) Filter impact datasheet by the impactdate and orguniqueid columns and begin with first player ID

2) Select first camera time from datasheet and enter into Jump to previous app to analyze impact trigger event

3) Rewind and watch the video until you find the instrumented player under review

4) Review video and answer ALL Qualtrics Survey questions using the operational definitions in this protocol manual

5) Repeat steps 2 through 4 for the next impact trigger event in the list to finish player

6) Move to next player in datasheet until all players and impacts have been analyzed

7) Move to the next game until all games have been analyzed

Video Analysis Best Practices

- Move frame by frame analyzing an impact
- Back up ½ second prior to the impact to analyze impact repeatedly

- Pay attention to the reviewed player’s helmet and sensitive to any head or helmet movement
- Pay attention to the opponent, or teammate’s helmet that the reviewed player’s head/ helmet comes into contact with. Movement from the opponent or teammate’s helmet will indicate the reviewed player likely hit them with their head

OPERATIONAL DEFINITIONS and INCLUSION/EXCLUSION CRITERIA EXPLANATIONS FOR SENSOR GOLD STANDARD PROXY ANALYSIS PROTOCOL

1. Is the player on the field participating in a play?

Select **YES** if the trigger event meets all these criteria:

- The reviewed player must be participating in a special teams play on the field

Select **NO** if the trigger event meets one of these criteria:

- The reviewed player is on the sideline during a special teams play
- The trigger event occurred from the reviewed player on the field but before or after the special teams play finished
  - Example: Exclude impacts following a play - player celebratory head impacts should not be included

2. Is the player on the screen at the time of sensor trigger?

Select **YES** if the trigger event meets all these criteria:

- The reviewed player must be in the camera frame at the impact trigger time

Select **NO** if the trigger event meets any these criteria:

- The reviewed player is not in the view of the camera at the impact trigger event

3. Is there an unobstructed view of the trigger event?

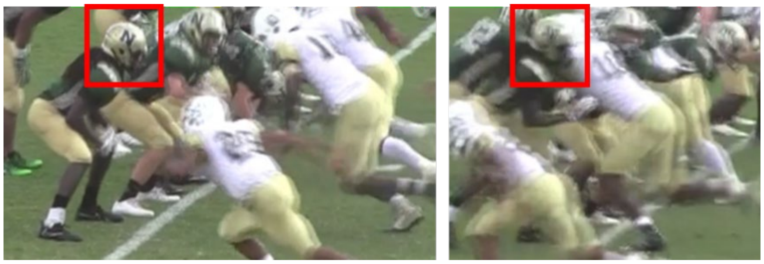
Select **YES** if the trigger event meets either of these two criteria:

- There must be a clear, unobstructed view of the reviewed player’s helmet that is not blocked by another player’s head, arm, body, etc. immediately prior to and at the impact trigger time
- The reviewed player’s head can be partially obstructed by the player they are impacting/ being impacted by at the impact trigger event AND it provides enough information to **confidently** answer the following:
    1. Is there clear evidence of helmet contact?
    2. Is a clear HIT System impact location observed (See page 9 for further information)?
    3. Can the impact be confidently assigned to the helmet shell or facemask?
    4. Can the impact be confidently defined as a centric or non-centric impact?

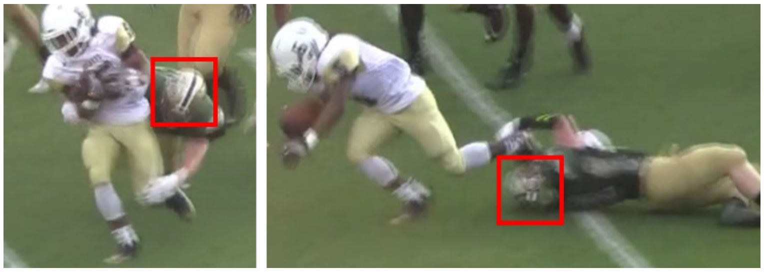
Unobstructed View Examples:

In all examples, the red boxes indicate the reviewed player’s head

Example 1 - The reviewed player’s head is completely visible and not obstructed by another player’s head/helmet, body, or body part immediately prior to and at the impact trigger event



Example 2 - The reviewed player’s head is completely visible and not obstructed by another player’s head/helmet, body, or body part immediately prior to and at the impact trigger event when hitting the ground



Example 3 - The reviewed player's head is completely visible and not obstructed by another player's head/helmet, body, or body immediately prior to and part at the impact trigger event

Partially Obstructed AND CAN Confidently Answer Succeeding Questions Examples:

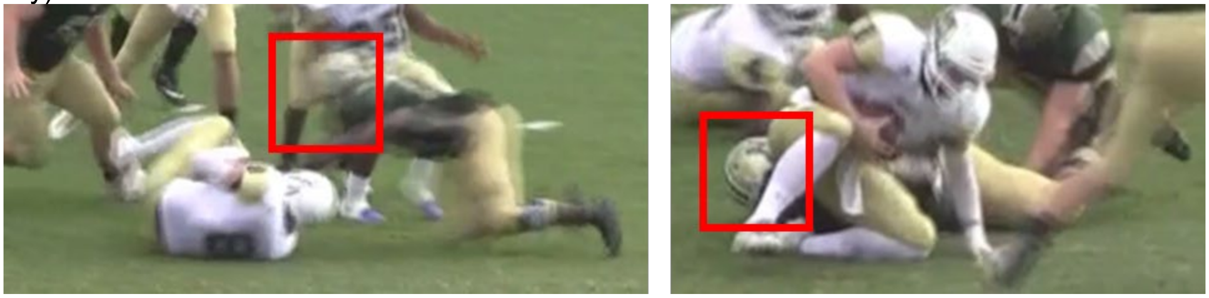
Example 1: The view of the reviewed player's head is partially obstructed by the opponent's shoulder, but we can observe an impact occurred, at the front location, on the facemask, and delivered non-centricaly

Select **YES** if the trigger event meets any these criteria:

- The reviewed player's helmet is completely covered/obstructed by another player's head, body, arm, etc.
- The reviewed player's helmet is partially covered/obstructed AND the partially obstructed view cannot provide enough information to confidently answer the following:
    1. Is there clear evidence of helmet contact?
    2. Is a clear HIT System impact location observed?
    3. Can the impact be confidently assigned to the helmet shell or facemask?
    4. Can the impact be confidently defined as a centric or non-centric impact?

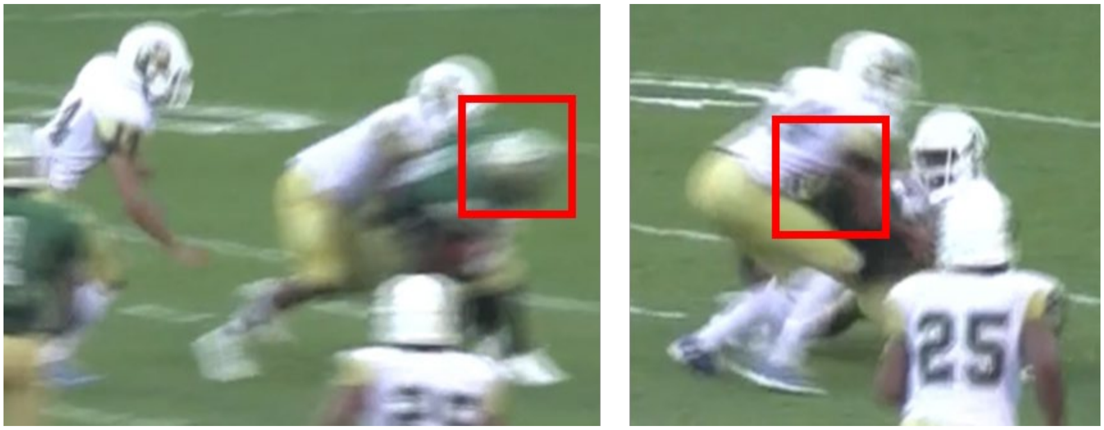
Obstructed View Examples:

Example 1 - The impact trigger event occurred when the reviewed player’s head was obstructed (for example in the bottom picture the head is obstructed because it is behind the opponent’s body)



Partially Obstructed AND CANNOT Confidently Answer Succeeding Questions Examples:

Example 1 – The reviewed player’s head is partially obstructed by the opponent’s body and leg. While we observed significant head motion to the observed player, we cannot determine the HIT System location, the shell or facemask location, or the impact centricity to any level of confidence



4. Is there clear evidence of helmet contact?

Select **YES** if the trigger event meets all these criteria:

- There is a space between the reviewed player’s head and the contacting object (player, bench, goalpost, etc.) prior to the trigger event AND is followed by helmet contact to the reviewed player
- Contact between the reviewed player’s head and the contacting object is observed on the reviewed player’s helmet at impact trigger

- To ensure contact occurred the following criteria can be observed and include either or all of the following:
    - There is **helmet motion observed to the reviewed player** following the impact trigger
    - There is **helmet motion observed to the player** being impacted by the reviewed player

Helmet Contact Examples:

See unobstructed view examples on **page 4** for helmet contact examples

Select **NO** if the trigger event did not meet any of the criteria from the YES criteria or did not meet the following criteria:

- If any of the criteria from **YES** are not satisfied
- Prior to impact trigger, there is no observable space between the reviewed player's helmet and the contacting object
- No area on the reviewed player's helmet received contact from another player or object
- There is no or a minimal amount of head motion observed on the reviewed player following the impact trigger
- There is no or a minimal amount of head motion observed on the player the reviewed player impacted following the impact trigger

No Helmet Contact Examples:

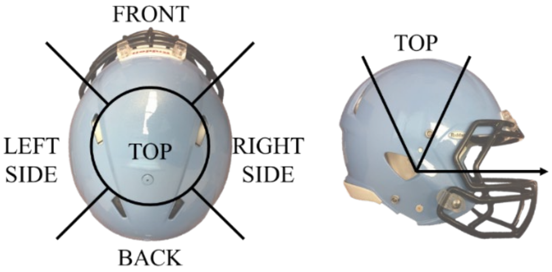
Example 1 – The reviewed player contacts the opponent, but there is no evidence of head motion from the reviewed player

Example 2 – The reviewed player received contact from the opponent, but there is no evidence of head motion from the reviewed player

5. Is a clear HIT System impact location observed?

Select **YES** if the trigger event meets these criteria:

- The trigger event for the observed player can be confidently assigned to one of the following HIT System Impact Locations according to the landmarks and naming convention below:

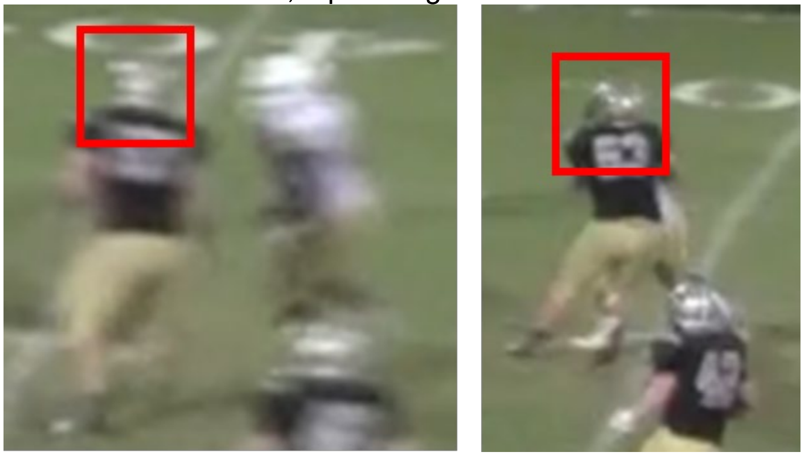


Select **NO** if the trigger event meets these criteria:

- Impact trigger event is observed at the boundary of two or three locations

Example of not Knowing HITS Location

Example 1 – The impact to the reviewed player's head occurred at the front of the head while they were looking down obliquely from the right. It is unknown where the impact came from and it is likely on the boundaries of the front, top and right HITS locations



Criteria for **80% SURE** if **NO** is Selected

- You are willing to make a highly educated and near confident guess, but don't have all the information to make it a 100% confident choice on the question
- These are cases where the impact trigger event observed on video is at the boundary line of two locations. These are commonly oblique impacts to the front or back that could be interpreted as front or side.

Criteria for **UNKNOWN** if **NO** is Selected

- There is not enough evidence to decide on the impact location the HIT System would assign to the impact trigger event under review

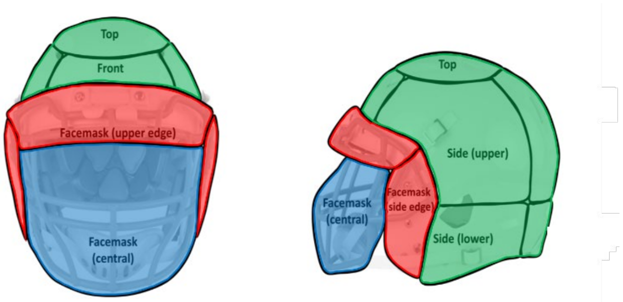
6. Can the impact be confidently assigned to the helmet shell or facemask?

Select **YES** if the trigger event meets these criteria:

- A clear contact site between the reviewed players helmet and the contacting object can be observed occurring to the facemask(blue) or to the helmet shell (green) of the reviewed player according to the figure below

Select **NO** if the trigger event meets any these criteria:

- The contact site occurs to the boundary between the facemask and the helmet shell. According to the figure below this can include the facemask's upper edge and the facemask's side edge (red)



- The contact site occurs to the reviewed player's chin
- The contact site occurs to the reviewed player's body
- There is a pixelated view of the impact
- The view is not zoomed in enough

7. Can the impact be confidently defined as a centric or non-centric impact?

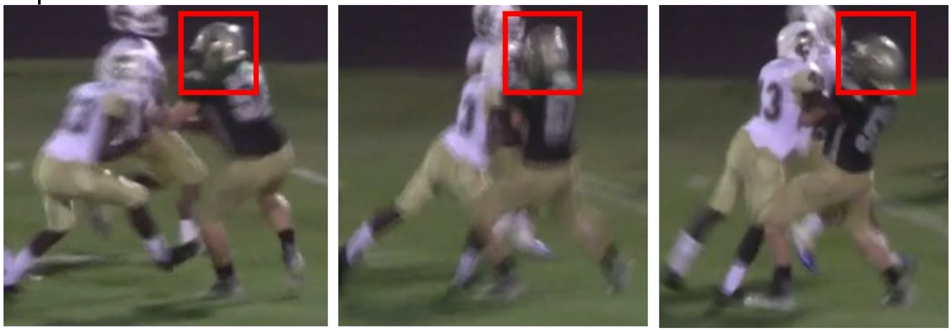
Select **YES** if the trigger event meets any of these criteria:

- A centric impact can be clearly observed occurring to the reviewed player. A centric impact is defined by the following:
    1. The striking player's line of action is directed through the center of gravity of the head of the struck reviewed player receiving the impact
    2. The striking reviewed player's line of action is directed through the center of gravity of the struck player they are impacting

3. The impact maintains contact with the helmet through the center of gravity. After, the initial impact the striking/struck player's helmet can glance or slide past the striking/struck player's helmet

Centric Impact Examples:

Example 1 – The opponent's line of action is through the reviewed player's head center of gravity. The opponent's helmet does not slide pass the reviewed player's head at impact trigger time. Not part of the criteria, but the player's head rotates about the neck, not as a result of a non-centric impact.

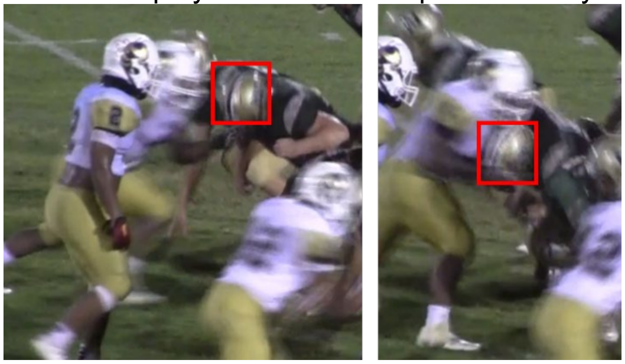


Example 2 – The reviewed player's line of action is straight through the head of the opponent at impact time and no sliding occurs between their heads. After the impact, the opponent's head rotates away about their neck.

- A non-centric/glancing impact can be clearly observed occurring to the reviewed player. A non-centric/glancing impact is defined by the following:
    1. The striking player's line of action is directed outside the center of gravity of the head of the struck reviewed player receiving the impact.
    2. The striking reviewed player's line of action is outside the center of gravity of the struck player they are impacting
    3. The striking helmet immediately slides past the helmet of the stuck player's helmet

Non - centric Impact Examples:

Example 1 – The opponent’s line of action is straight forward, but the reviewed player’s line of action is downwards. The reviewed player’s head slides past the body of the opponent



Select **NO** if the trigger event meets any of these criteria:

- There is insufficient evidence to confidently assign either a centric or non-centric impact to the reviewed impact trigger event
- There is a pixelated view of the impact
- The view is not zoomed in enough

APPENDIX E : VIDEO ANALYSIS PROTOCOL FOR IMPACT DETECTION

VIDEO ANALYSIS PROTOCOL – VIDEO GOLD STANDARD PROXY

Application Preparation Steps

1) Open Video file for current game analysis in VLC 2.2.8

- File name example: 1.1_08-18-17_Game1_Half1

2) Open time off-screen calculator excel sheet

- File name example: aim2_video_proxy_time_off_screen_calculator.xlsx

3) Open special teams plays camera times list

- Filename: Aim2SpecialTeamsTimesinVideo.xlsx

4) Open player ID list

- Filename: team2_ids_video_proxy_eval2.xlsx

5) Open Qualtrics Survey – Video as Gold Standard Proxy Analysis

6) Begin Video Analysis

Video Analysis Steps

1) Filter special teams plays camera times list for the current video/game under observation

2) Select the special teams play start time from the first play and enter into Jump to previous app to analyze potential impacts to all instrumented players on the current special teams play

3) Find an instrumented player to review from the player ID list. Watch the entire play from beginning to end. By the end of this initial review you should have identified instances of potential impacts and times the player went off screen.

- If the player went off screen, enter the camera time time they go off screen into the off-screen calculator excel sheet. Enter the camera time they come back on screen into the off-screen calculator excel sheet to determine the duration they were off screen. If they are out of view the entire play use the play start and stop times from the special teams play list as the times off screen.

4) Answer initial Qualtrics survey questions in the initial block. If no impact was observed select no and continue. Answer questions regarding position and play type.

5) Enter the camera times and durations that players go off screen from the off-screen calculator into the allocated Qualtrics question. If the player did not go off-screen, select no and continue

6) Record and insert the camera time for a potential impact into the Qualtrics analysis.

7) Review video and answer ALL Qualtrics Survey questions regarding inclusion/ exclusion criteria and impact descriptions using the operational definitions in this protocol manual

8) Repeat steps 3 through 7 for the next potential impact trigger event you observed for the same player until all potential impacts are analyzed for that player on that play

9) Move to next player observed on video that is instrumented with a HIT System helmet according to the player ID list and repeat steps 3 through 7 for impacts for this player and additional players.

10) Move to the next play until all plays for a game have been analyzed

11) Move to the next game once all special teams plays have been analyzed

Video Analysis Best Practices

- Work your way across the field from left to right/ right analyzing impacts to each instrumented player.
- Move frame by frame analyzing an impact
- Back up ½ second prior to the impact to analyze impact repeatedly
- Pay attention to the reviewed player’s helmet and sensitive to any head or helmet movement

- Pay attention to the opponent, or teammate's helmet that the reviewed player's head/ helmet comes into contact with. Movement from the opponent or teammate's helmet will indicate the reviewed player likely hit them with their head

OPERATIONAL DEFINITIONS and INCLUSION/EXCLUSION CRITERIA EXPLANATIONS FOR VIDEO GOLD STANDARD PROXY ANALYSIS PROTOCOL

1. **Is there a potential impact trigger event for the current player under review for this play?**

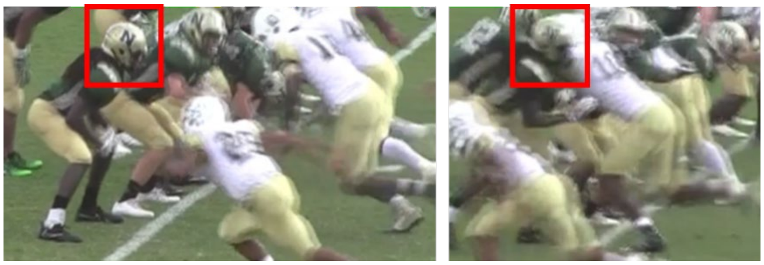
Select **YES** if the trigger event meets all these criteria:

- There is a space between the reviewed player's head and the contacting object (player, bench, goalpost, etc.) prior to the trigger event AND is followed by helmet contact to the reviewed player
- Contact between the reviewed player's head and the contacting object is observed on the reviewed player's helmet at impact trigger
- To ensure contact occurred the following criteria can be observed and include either or all of the following:
    - There is **helmet motion observed to the reviewed player** following the impact trigger
    - There is **helmet motion observed to the player** being impacted by the reviewed player

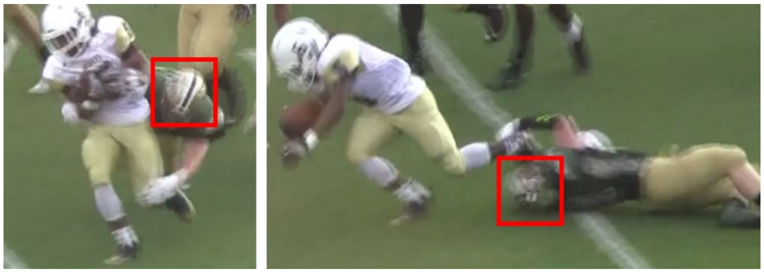
Helmet Contact Examples:

In all examples, the red boxes indicate the reviewed player's head

Example 1 - The reviewed player’s head is completely visible and not obstructed by another player’s head/helmet, body, or body part immediately prior to and at the impact trigger event



Example 2 - The reviewed player’s head is completely visible and not obstructed by another player’s head/helmet, body, or body part immediately prior to and at the impact trigger event when hitting the ground



Example 3 - The reviewed player’s head is completely visible and not obstructed by another player’s head/helmet, body, or body immediately prior to and part at the impact trigger event

Partially Obstructed AND CAN Confidently Answer Succeeding Questions Examples:

Example 1: The view of the reviewed player’s head is partially obstructed by the opponent’s shoulder, but we can observe an impact occurred, at the front location, on the facemask, and delivered non-centricaly

Select **NO** if the trigger event did not meet any of the criteria from the YES criteria or did not meet the following criteria:

- If any of the criteria from **YES** are not satisfied
- Prior to impact trigger, there is no observable space between the reviewed player’s helmet and the contacting object
- No area on the reviewed player’s helmet received contact from another player or object
- There is no or a minimal amount of head motion observed on the reviewed player following the impact trigger
- There is no or a minimal amount of head motion observed on the player the reviewed player impacted following the impact trigger

No Helmet Contact Examples:

Example 1 – The reviewed player contacts the opponent, but there is no evidence of head motion from the reviewed player

Example 2 – The reviewed player received contact from the opponent, but there is no evidence of head motion from the reviewed player

2. Is there an unobstructed view of the trigger event?

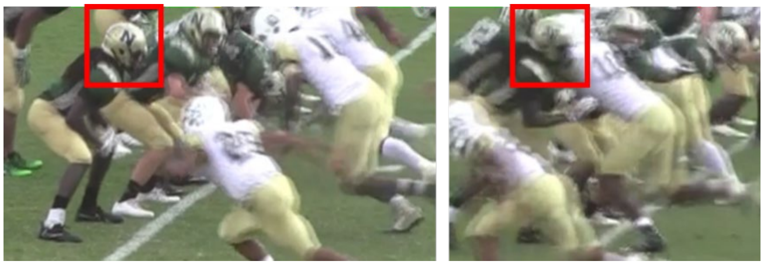
Select **YES** if the trigger event meets either of these two criteria:

- There must be a clear, unobstructed view of the reviewed player's helmet that is not blocked by another player's head, arm, body, etc. immediately prior to and at the impact trigger time
- The reviewed player's head can be partially obstructed by the player they are impacting/ being impacted by at the impact trigger event AND it provides enough information to **confidently** answer the following:
    1. Is there clear evidence of helmet contact?
    2. Is a clear HIT System impact location observed (See page 9 for further information)?
    3. Can the impact be confidently assigned to the helmet shell or facemask?
    4. Can the impact be confidently defined as a centric or non-centric impact?

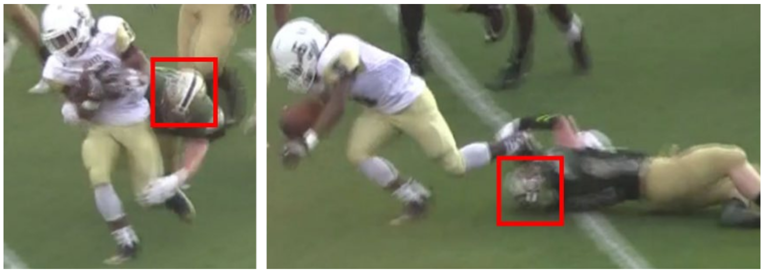
Unobstructed View Examples:

In all examples, the red boxes indicate the reviewed player's head

Example 1 - The reviewed player's head is completely visible and not obstructed by another player's head/helmet, body, or body part immediately prior to and at the impact trigger event



Example 2 - The reviewed player's head is completely visible and not obstructed by another player's head/helmet, body, or body part immediately prior to and at the impact trigger event when hitting the ground



Example 3 - The reviewed player's head is completely visible and not obstructed by another player's head/helmet, body, or body immediately prior to and part at the impact trigger event

Partially Obstructed AND CAN Confidently Answer Succeeding Questions Examples:

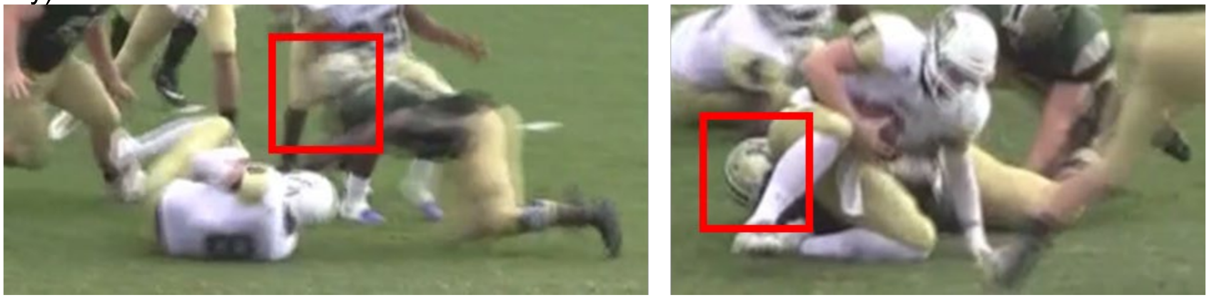
Example 1: The view of the reviewed player’s head is partially obstructed by the opponent’s shoulder, but we can observe an impact occurred, at the front location, on the facemask, and delivered non-centricaly

Select **YES** if the trigger event meets any these criteria:

- The reviewed player’s helmet is completely covered/obstructed by another player’s head, body, arm, etc.
- The reviewed player’s helmet is partially covered/obstructed AND the partially obstructed view cannot provide enough information to confidently answer the following:
    1. Is there clear evidence of helmet contact?
    2. Is a clear HIT System impact location observed?
    3. Can the impact be confidently assigned to the helmet shell or facemask?
    4. Can the impact be confidently defined as a centric or non-centric impact?

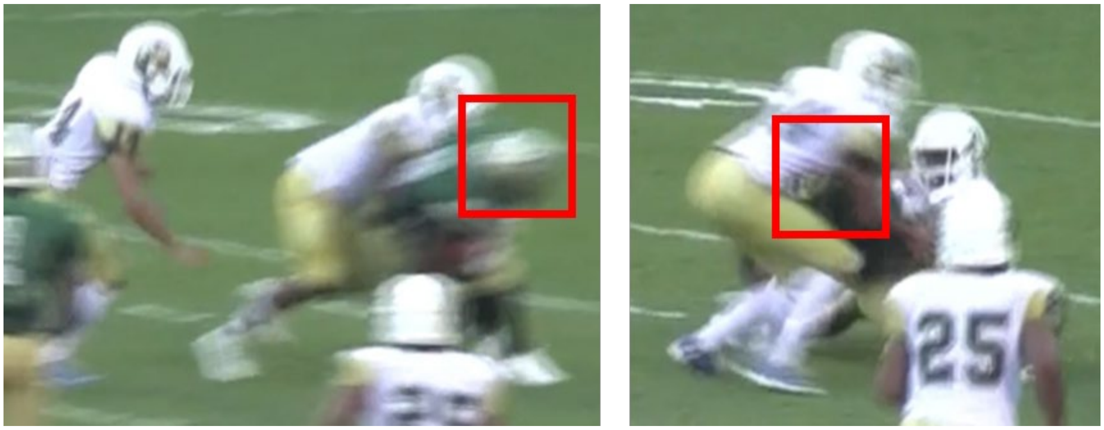
Obstructed View Examples:

Example 1 - The impact trigger event occurred when the reviewed player’s head was obstructed (for example in the bottom picture the head is obstructed because it is behind the opponent’s body)



Partially Obstructed AND CANNOT Confidently Answer Succeeding Questions Examples:

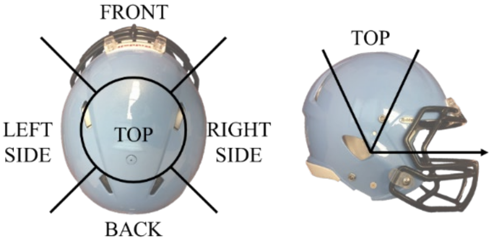
Example 1 – The reviewed player’s head is partially obstructed by the opponent’s body and leg. While we observed significant head motion to the observed player, we cannot determine the HIT System location, the shell or facemask location, or the impact centricity to any level of confidence



3. Is a clear HIT System impact location observed?

Select **YES** if the trigger event meets these criteria:

- The trigger event for the observed player can be confidently assigned to one of the following HIT System Impact Locations according to the landmarks and naming convention below:

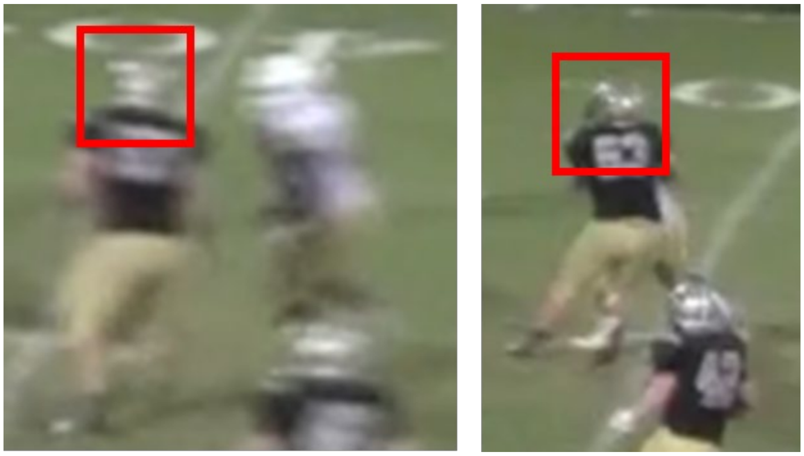


Select **NO** if the trigger event meets these criteria:

- Impact trigger event is observed at the boundary of two or three locations

Example of not Knowing HITS Location

Example 1 – The impact to the reviewed player’s head occurred at the front of the head while they were looking down obliquely from the right. It is unknown where the impact came from and it is likely on the boundaries of the front, top and right HITS locations



Criteria for **80% SURE** if **NO** is Selected

- You are willing to make a highly educated and near confident guess, but don't have all the information to make it a 100% confident choice on the question
- These are cases where the impact trigger event observed on video is at the boundary line of two locations. These are commonly oblique impacts to the front or back that could be interpreted as front or side.

Criteria for **UNKNOWN** if **NO** is Selected

- There is not enough evidence to decide on the impact location the HIT System would assign to the impact trigger event under review

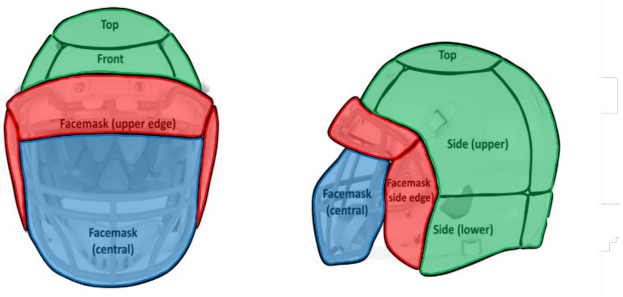
4. Can the impact be confidently assigned to the helmet shell or facemask?

Select **YES** if the trigger event meets these criteria:

- A clear contact site between the reviewed players helmet and the contacting object can be observed occurring to the facemask(blue) or to the helmet shell (green) of the reviewed player according to the figure below

Select **NO** if the trigger event meets any these criteria:

- The contact site occurs to the boundary between the facemask and the helmet shell. According to the figure below this can include the facemask's upper edge and the facemask's side edge (red)



- The contact site occurs to the reviewed player's chin
- The contact site occurs to the reviewed player's body
- There is a pixelated view of the impact
- The view is not zoomed in enough

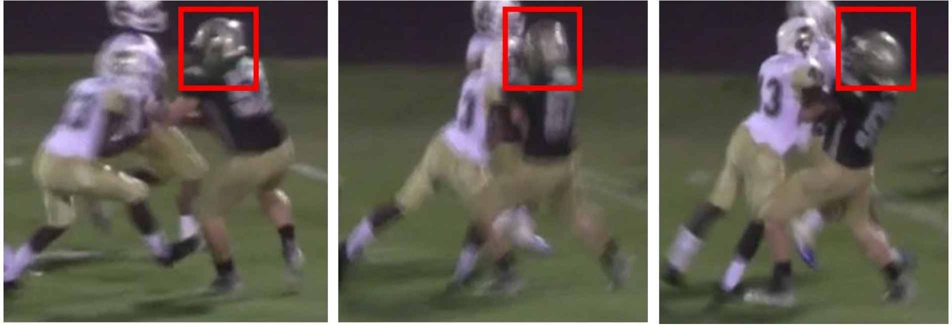
5. Can the impact be confidently defined as a centric or non-centric impact?

Select **YES** if the trigger event meets any of these criteria:

- A centric impact can be clearly observed occurring to the reviewed player. A centric impact is defined by the following:
    1. The striking player's line of action is directed through the center of gravity of the head of the struck reviewed player receiving the impact
    2. The striking reviewed player's line of action is directed through the center of gravity of the struck player they are impacting
    3. The impact maintains contact with the helmet through the center of gravity. After, the initial impact the striking/struck player's helmet can glance or slide past the striking/struck player's helmet

Centric Impact Examples:

Example 1 – The opponent’s line of action is through the reviewed player’s head center of gravity. The opponent’s helmet does not slide pass the reviewed player’s head at impact trigger time. Not part of the criteria, but the player’s head rotates about the neck, not as a result of a non-centric impact.



Example 2 – The reviewed player’s line of action is straight through the head of the opponent at impact time and no sliding occurs between their heads. After the impact, the opponent’s head rotates away about their neck.

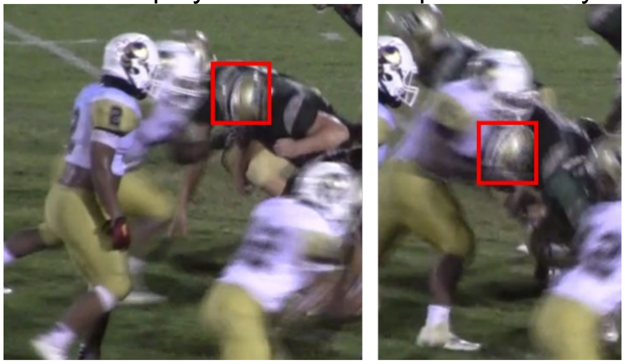
- A non-centric/glancing impact can be clearly observed occurring to the reviewed player.

  A non-centric/glancing impact is defined by the following:

  1. The striking player’s line of action is directed outside the center of gravity of the head of the struck reviewed player receiving the impact.
  2. The striking reviewed player’s line of action is outside the center of gravity of the struck player they are impacting
  3. The striking helmet immediately slides past the helmet of the stuck player’s helmet

Non - centric Impact Examples:

Example 1 – The opponent’s line of action is straight forward, but the reviewed player’s line of action is downwards. The reviewed player’s head slides past the body of the opponent



Select **NO** if the trigger event meets any of these criteria:

- There is insufficient evidence to confidently assign either a centric or non-centric impact to the reviewed impact trigger event
- There is a pixelated view of the impact

The view is not zoomed in enough

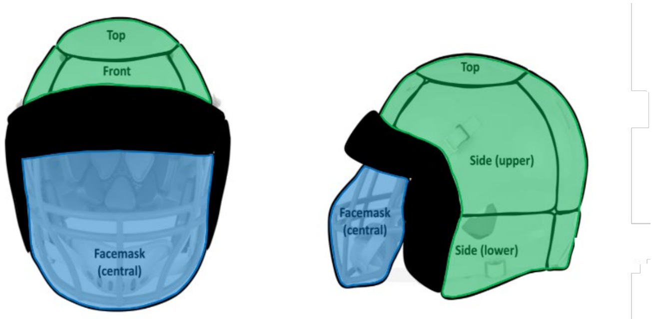
APPENDIX F : VIDEO ANALYSIS DEFINITIONS

ADDITIONAL DEFINITIONS AND EXPLANATIONS FOR ANALYSIS PROTOCOL

1. **What impact type caused the possible trigger event?**
    - helmet to helmet – The reviewed player's head contacts the teammate's/opponent's head for the impact trigger
    - helmet to body - The reviewed player's head contacts the teammate's/opponent's body for the impact trigger
    - helmet to ground - The reviewed player's head contacts the ground for the impact trigger
    - head to object - The reviewed player's head contacts an inanimate object (goal post, sideline table, yardstick marker, etc.) for the impact trigger
    - no contact to the head occurred – contact occurred to the reviewed player's body for the impact trigger
    - other – select and fill in the contact object the reviewed player's helmet hit
    - unknown – select if you cannot confidently assign one of the choices above
2. **Who did the player in question collide with?**
    - Teammate – The reviewed player contacted a teammate for the impact trigger event
    - Opponent – The reviewed player contacted an opponent for the impact trigger event
    - Ground – The reviewed player contacted the ground for the impact trigger event
    - Inanimate object – The reviewed player contacted an inanimate object (goal post, sideline table, yardstick marker, etc.) for the impact trigger event
    - Indirect hit – experienced a whiplash motion, but the players head does not connect with a person, the ground or another object
    - Self – The reviewed player hits himself in the head, celebrates with a head-butt, etc.
    - Other – select and fill in the contact object the reviewed player's helmet hit
    - Unknown – select if you cannot confidently assign one of the choices above

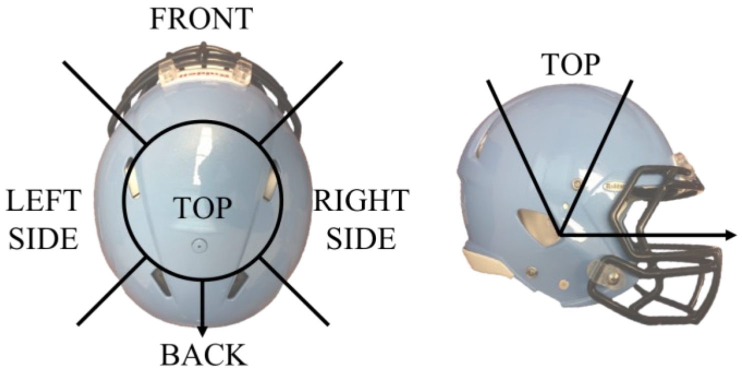
3. **Player involvement during trigger event?**
   - Making a Tackle – The reviewed player is making a tackle during the impact trigger
   - Being Tackled – The reviewed player is being tackled during the impact trigger
   - Open Field Blocking – The reviewed player is blocking on their team's kick return or their team's punt return after the ball has been kicked
   - Line Blocking – The reviewed player is blocking on the line during their team's punt cover
   - Being Blocked in Open Field – The reviewed player is being blocked during their team's kickoff cover or punt cover
   - Being Blocked on line – The reviewed player is being blocked blocking on the line during their team's punt return
   - Player was recovering a fumbled ball
   - Player's head hit the ground
   - Player did not collide anyone else – The reviewed player's body received an impact and the forces were transferred to his head causing a whiplash motion
   - Other – select and fill in the contact object the reviewed player's helmet hit
   - Unknown – select if you cannot confidently assign one of the choices above
4. **Closing Distance Type - This should be graded by how far apart the players were when the play started &/or the earliest you can see them on any of the videos.**
   - Long Distance – An Open-field collision, where the reviewed player and the impact-ing/ed source are greater than 10 yards apart before the impact occurred
   - Short Distance – A collision where the reviewed player and the impact-ing/ed source are less than 10 yards apart before the impact occurred
   - Unknown – select if you cannot confidently assign one of the choices above

5. **Was the impact centric (delivered through the head's center of gravity) or non-centric (a glancing or tangential impact)?**
    - Centric – The impact line of action is directed through the center of gravity of the head
    - Non-Centric - The impact line of action is directed outside the center of gravity of the head and causes the helmets to glance or slide past each other
    - Unknown – select if you cannot confidently assign one of the choices above
6. **Video observed Impact Location**



- Facemask – The observed impact location on video is directed on the facemask (blue)
- Shell - The observed impact location on video is directed on the helmet shell (green)
- Unknown – select if you cannot confidently assign one of the choices above

7. **HIT System Impact Location**



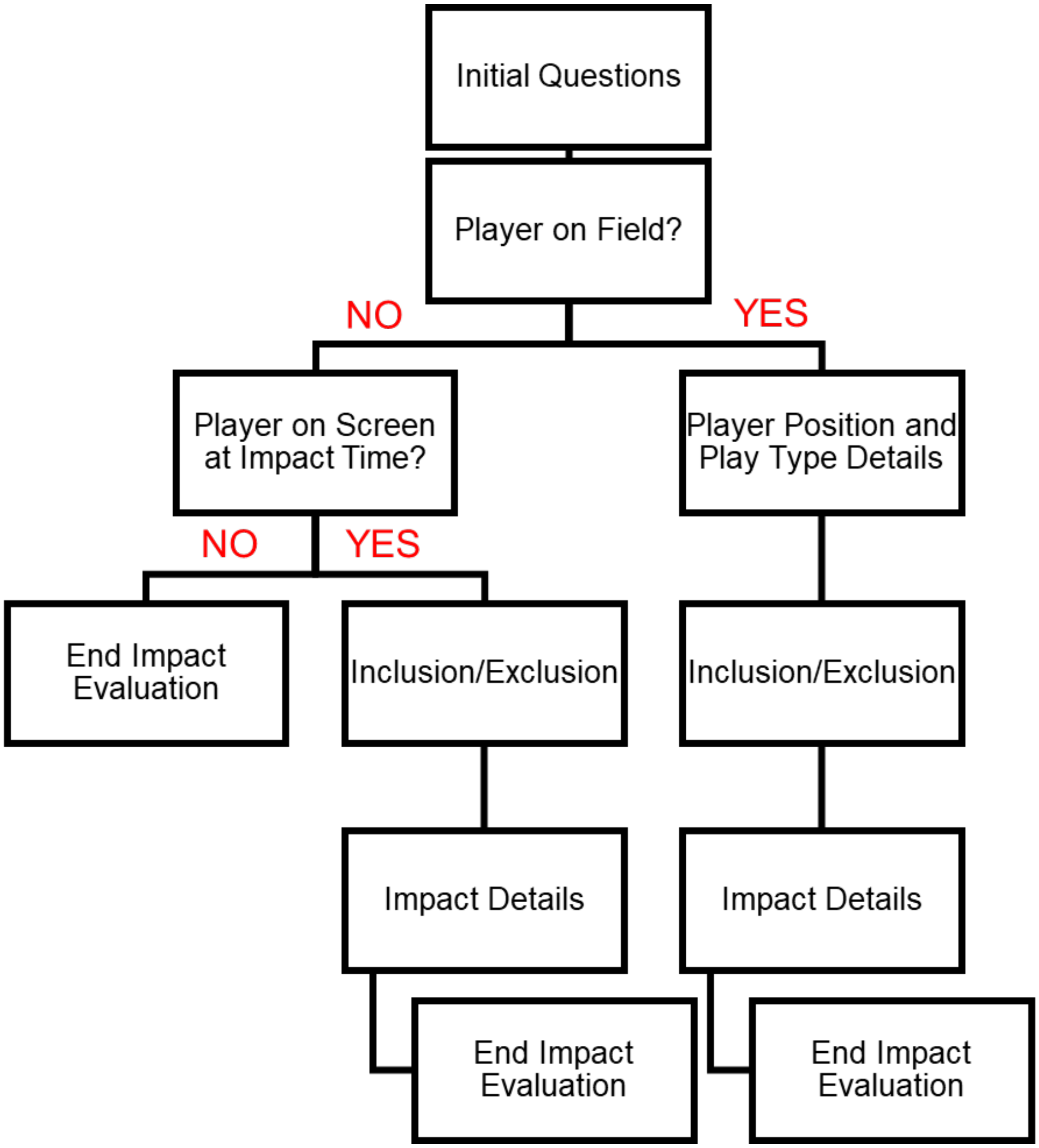
The impact location observed on film is assigned according to the HIT System impact location category convention:

- Front
- Back
- Right
- Left
- Top
- Unknown – select if you cannot confidently assign one of the choices above

APPENDIX G : VIDEO ANALYSIS DATA ENTRY FORM – IMPACT LOCATION ASSESSMENT

VIDEO ASSESSMENT DATA ENTRY FORM – IMPACT LOCATION

Survey Flow



Survey Blocks:

1. Initial Questions
2. Player on Screen
3. Inclusion/Exclusion Criteria
4. Impact Description
5. Player on Field Questions

Survey Questions:

Start of Block: Initial

raterid **Rater ID**

☐ Kody Campbell (0)

☐ Josh Boone (1)

☐ Adrian Boltz (2)

game **Game Number**

○ 1 (1)

○ 2 (2)

○ 3 (3)

○ 4 (4)

○ 5 (5)

○ 6 (6)

○ 7 (7)

○ 8 (8)

○ 9 (9)

○ 10 (10)

○ 11 (11)

○ 12 (12)

playerid **Enter Player origuniqueID below:**

____

eventmatchid **Enter the eventID below:**

____

impactcameratime **Enter the Camera Time Stamp of trigger event (format = hh:mm:ss.000)**

____

onfield **Is the player on the field participating in a football play?**

○ Yes (1)

○ No (0)

End of Block: Initial

Start of Block: Player On Screen

onscreen Is the player on the screen at the time of sensor trigger?

○ Yes (1)

○ No (0)

End of Block: Player On Screen

Start of Block: Inclusion/Exclusion Criteria

obstructedview **Is there an obstructed view of the trigger event?**

○ Yes (1)

○ No (0)

inc-exhelmetcontact **Is there clear evidence of helmet contact?**

○ Yes (1)

○ No (0)

inc-exchitloc **Is a clear HIT System impact location observed?**

○ Yes (1)

○ No (0)

Display This Question:

If Is a clear HIT System impact location observed? = No

Q86 **If there is NO clear evidence, to what extent could you choose a HIT System impact location category?**

○ 80% sure on impact location (1)

○ unknown (999)

inc-exshellfacemask **Can the impact be confidently assigned to the helmet shell or facemask?**

○ Yes (1)

○ No (0)

Display This Question:

If Can the impact be confidently assigned to the helmet shell or facemask? = No

Q88 **Why are you unable to determine the contact site on the helmet?**

○ Impact to Chin (1)

○ Impact to Body (2)

○ unknown (999)

inc-eximpcentric **Can the impact be confidently defined as a centric or non-centric impact?**

○ Yes (1)

○ No (0)

End of Block: Inclusion/Exclusion Criteria

Start of Block: Impact Description

headcontacttype **What impact type caused the possible trigger event?**

○ helmet to helmet (0)

○ helmet to body (1)

○ helmet to ground (2)

○ head to object (3)

○ no contact to the head occurred (4)

○ other (888) ____

○ unknown (999)

impactcause **Player involvement during trigger event?**

- ○ Making a Tackle (0)
- ○ Being Tackled (1)
- ○ Open Field Blocking (players on punt return and kickoff return) (2)
- ○ Line Blocking (players blocking on punt cover) (3)
- ○ Being Blocked in Open Field (players on kickoff cover and punt cover) (4)
- ○ Being Blocked on Line (players on punt return) (5)
- ○ Player was recovering a fumbled ball (6)
- ○ Player's head hit the ground (7)
- ○ Player did not collide anyone else (indirect hit) (8)
- ○ Other (888) ____
- ○ Unknown (999)

collisionwith **Who did the player in question collide with?**

- ○ Teammate (0)
- ○ Opponent (1)
- ○ Ground (2)
- ○ Inanimate object (3)
- ○ Indirect hit (experienced a whiplash motion, but the players head does not connect with a person, the ground or another object) (4)
- ○ Self (hits himself in the head, celebrates with a head-but, etc.) (5)
- ○ Other (888) ____
- ○ Unknown (999)

closing distance **Closing Distance Type - This should be graded by how far apart the players were when the play started &/or the earliest you can see them on any of the videos.**

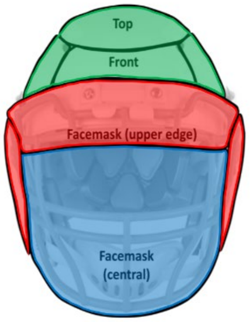
- ○ Long Distance (Open-field collision >10yds) (1)
- ○ Short Distance (Contained collision (i.e., running play, etc.) $\leq$10yds) (0)
- ○ Unknown (999)

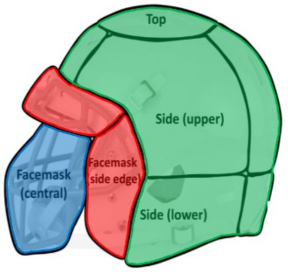
impactcentricity Was the impact centric (delivered through the head's center of gravity) or non-centric (a glancing or tangential impact)?

- ○ Centric (0)
- ○ Non-Centric (1)
- ○ Unknown (999)

videoimpactlocation **Video observed Impact Location**

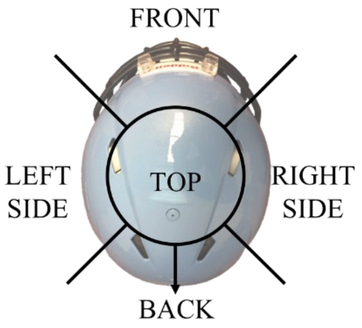
- ○ Facemask (blue) (1)
- ○ Shell (green) (0)
- ○ Unknown (999)

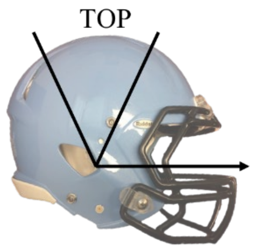




hitslocation **HIT System Impact Location**

- ○ Front (0)
- ○ Back (1)
- ○ Right (2)
- ○ Left (3)
- ○ Top (4)
- ○ Unknown (999)





addcomments Were there any mistakes made or choices you would like to flag for consideration? Please indicate any part of your analysis where a mistake was made or you would like an external review.

- ○ Yes (1) ____
- ○ No (2)

End of Block: Impact Description

Start of Block: Player on Field Questions

position **What is this player's position for the play in question?**

- ○ Kicker (0)
- ○ Punter (1)
- ○ The ball returner for kickoff return or punt return (we caught the ball) (2)
- ○ Blocking for the kickoff returner or punt returner (3)
- ○ Defensive player on kickoff and punt (we kicked the ball) (everyone but the kicker) (4)

playtype **What is the play type?**

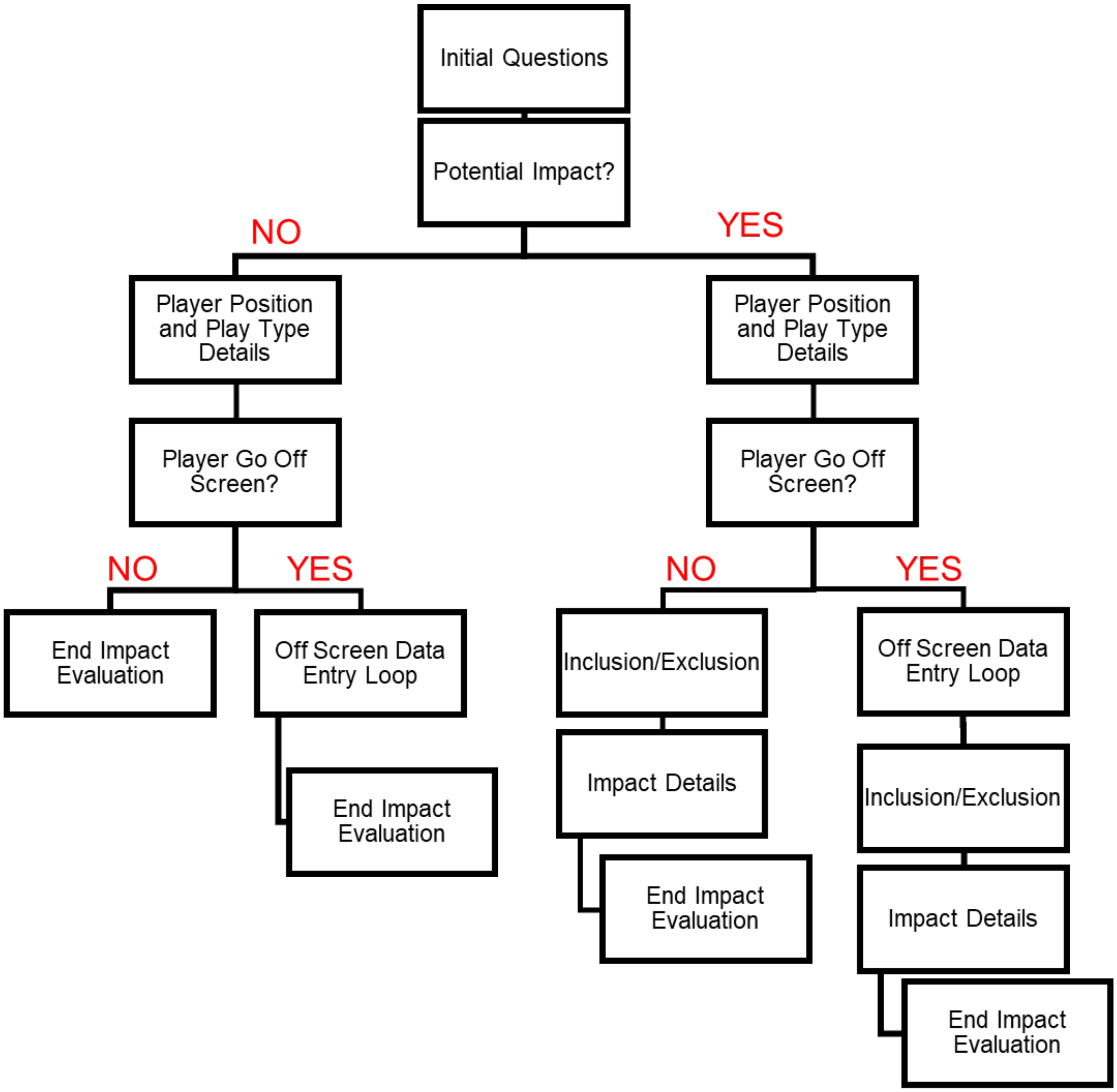
- ○ Punt Cover (0)
- ○ Punt Return (1)
- ○ Kickoff Cover (2)
- ○ Kickoff Return (3)
- ○ Punt Cover Fake (4)
- ○ Punt Return Fake (5)

End of Block: Player on Field Questions

APPENDIX H VIDEO ANALYSIS DATA ENTRY FORM – IMPACT DETECTION ASSESSMENT

VIDEO ASSESSMENT DATA ENTRY FORM – IMPACT DETECTION

Survey Flow



Survey Blocks:

1. Initial Questions
2. Player on Field Questions
3. On/Off Screen Loop
4. Inclusion/ Exclusion Criteria
5. Impact Description

Survey Questions:

Start of Block: Initial

raterid **Rater ID**

- ☐ Kody Campbell (0)
- ☐ Josh Boone (1)
- ☐ Adrian Boltz (2)

game **Game Number**

- ○ 1 (1)
- ○ 2 (2)
- ○ 3 (3)
- ○ 4 (4)
- ○ 5 (5)
- ○ 6 (6)
- ○ 7 (7)
- ○ 8 (8)
- ○ 9 (9)
- ○ 10 (10)
- ○ 11 (11)
- ○ 12 (12)

playnum **Enter the play number under review:**

____

videofile **Enter the video file you are reviewing (ex. 1.1):**

____

playerid **Enter Player ID under review below:**

____

potentialtrigger **Is there a suspected impact trigger event after watching the entire play?:**

____

Start of Block: Player on Field Questions

position **What is this player's position for the play in question?**

○ Kicker (0)

○ Punter (1)

○ The ball returner for kickoff return or punt return (we caught the ball) (2)

○ Blocking for the kickoff returner or punt returner (3)

○ Defensive player on kickoff and punt (we kicked the ball) (everyone but the kicker) (4)

playtype **What is the play type?**

○ Punt Cover (0)

○ Punt Return (1)

○ Kickoff Cover (2)

○ Kickoff Return (3)

○ Punt Cover Fake (4)

○ Punt Return Fake (5)

End of Block: Player on Field Questions

Start of Block: Player On Screen

onscreen **Did the reviewed player go off screen at any point after watching the entire play?**

○ Yes (1)

○ No (0)

End of Block: Player On Screen

Start of Block: On/Off Screen Loop

timeoffscreen **Enter the camera time when they went of screen:**

____

offscreendurationEnter **the duration they went off screen for:**

____

additionaloffscreen **Are there any additional times the reviewed player went off screen?**

- ○ Yes (1)
- ○ No (0)

End of Block: Player On Screen

Start of Block: Inclusion/Exclusion Criteria

obstructedview **Is there an obstructed view of the trigger event?**

- ○ Yes (1)
- ○ No (0)

inc-exchitloc **Is a clear HIT System impact location observed?**

- ○ Yes (1)
- ○ No (0)

Display This Question:

If Is a clear HIT System impact location observed? = No

Q86 **If there is NO clear evidence, to what extent could you choose a HIT System impact location category?**

- ○ 80% sure on impact location (1)
- ○ unknown (999)

inc-exshellfacemask **Can the impact be confidently assigned to the helmet shell or facemask?**

- ○ Yes (1)
- ○ No (0)

Display This Question:

If Can the impact be confidently assigned to the helmet shell or facemask? = No

Q88 **Why are you unable to determine the contact site on the helmet?**

- ○ Impact to Chin (1)
- ○ Impact to Body (2)
- ○ unknown (999)

inc-eximpcentric **Can the impact be confidently defined as a centric or non-centric impact?**

- ○ Yes (1)
- ○ No (0)

End of Block: Inclusion/Exclusion Criteria

Start of Block: Impact Description

headcontacttype **What impact type caused the possible trigger event?**

- ○ helmet to helmet (0)
- ○ helmet to body (1)
- ○ helmet to ground (2)
- ○ head to object (3)
- ○ no contact to the head occurred (4)
- ○ other (888) ____
- ○ unknown (999)

impactcause **Player involvement during trigger event?**

○ Making a Tackle (0)

○ Being Tackled (1)

○ Open Field Blocking (players on punt return and kickoff return) (2)

○ Line Blocking (players blocking on punt cover) (3)

○ Being Blocked in Open Field (players on kickoff cover and punt cover) (4)

○ Being Blocked on Line (players on punt return) (5)

○ Player was recovering a fumbled ball (6)

○ Player's head hit the ground (7)

○ Player did not collide anyone else (indirect hit) (8)

○ Other (888) ____

○ Unknown (999)

collisionwith **Who did the player in question collide with?**

○ Teammate (0)

○ Opponent (1)

○ Ground (2)

○ Inanimate object (3)

○ Indirect hit (experienced a whiplash motion, but the players head does not connect with a person, the ground or another object) (4)

○ Self (hits himself in the head, celebrates with a head-but, etc.) (5)

○ Other (888) ____

○ Unknown (999)

closing distance **Closing Distance Type - This should be graded by how far apart the players were when the play started &/or the earliest you can see them on any of the videos.**

○ Long Distance (Open-field collision >10yds) (1)

○ Short Distance (Contained collision (i.e., running play, etc.) ≤10yds) (0)

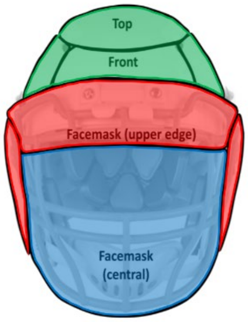
○ Unknown (999)

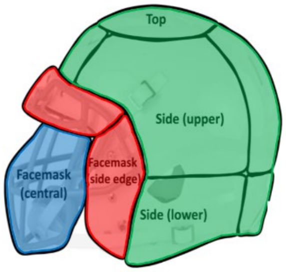
impactcentricity Was the impact centric (delivered through the head's center of gravity) or non-centric (a glancing or tangential impact)?

- ○ Centric (0)
- ○ Non-Centric (1)
- ○ Unknown (999)

videoimpactlocation **Video observed Impact Location**

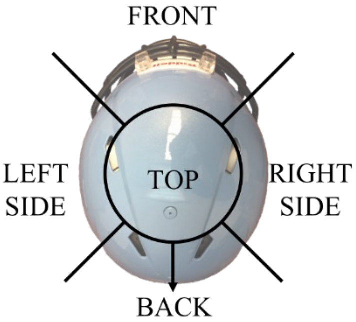
- ○ Facemask (blue) (1)
- ○ Shell (green) (0)
- ○ Unknown (999)

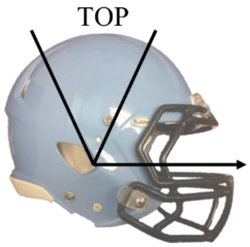




hitslocation **HIT System Impact Location**

- ○ Front (0)
- ○ Back (1)
- ○ Right (2)
- ○ Left (3)
- ○ Top (4)
- ○ Unknown (999)





REFERENCES

1. Guskiewicz KM, Marshall SW, Bailes J, et al. Recurrent concussion and risk of depression in retired professional football players. *Med Sci Sports Exerc*. 2007;39(6):903-909. doi:10.1249/mss.0b013e3180383da5

2. Guskiewicz KM, Marshall SW, Bailes J, et al. Association between Recurrent Concussion and Late-Life Cognitive Impairment in Retired Professional Football Players. *Neurosurgery*. 2005;57(4):719-726. doi:10.1227/01.neu.0000175725.75780.dd

3. Lehman EJ. Epidemiology of neurodegeneration in American-style professional football players. *Alzheimers Res Ther*. 2013;5(4):34. doi:10.1186/alzrt188

4. McCrory P, Meeuwisse W, Dvorak J, et al. Consensus statement on concussion in sport—the 5th international conference on concussion in sport held in Berlin, October 2016. *Br J Sports Med*. April 2017:bjsports-2017-097699. doi:10.1136/bjsports-2017-097699

5. Eucker SA, Smith C, Ralston J, Friess SH, Margulies SS. Physiological and histopathological responses following closed rotational head injury depend on direction of head motion. *Exp Neurol*. 2011;227(1):79-88. doi:10.1016/j.expneurol.2010.09.015

6. Ocwieja KE, Mihalik JP, Marshall SW, Schmidt JD, Trulock SC, Guskiewicz KM. The effect of play type and collision closing distance on head impact biomechanics. *Ann Biomed Eng*. 2012;40(1):90-96. doi:10.1007/s10439-011-0401-7

7. Yard EE, Comstock RD. Effects of field location, time in competition, and phase of play on injury severity in high school football. *Res Sports Med Print*. 2009;17(1):35-49. doi:10.1080/15438620802678495

8. Mihalik JP, Bell DR, Marshall SW, Guskiewicz KM. Measurement of head impacts in collegiate football players: an investigation of positional and event-type differences. *Neurosurgery*. 2007;61(6):1229-1235; discussion 1235. doi:10.1227/01.neu.0000306101.83882.c8

9. Rowson S, Duma SM, Beckwith JG, et al. Rotational head kinematics in football impacts: an injury risk function for concussion. *Ann Biomed Eng*. 2012;40(1):1-13. doi:10.1007/s10439-011-0392-4

10. Rowson S, Duma SM, Stemper BD, et al. Correlation of Concussion Symptom Profile with Head Impact Biomechanics: A Case for Individual-Specific Injury Tolerance. *J Neurotrauma*. 2017;35(4):681-690. doi:10.1089/neu.2017.5169

11. Siegmund GP, Guskiewicz KM, Marshall SW, DeMarco AL, Bonin SJ. Laboratory Validation of Two Wearable Sensor Systems for Measuring Head Impact Severity in Football Players. *Ann Biomed Eng*. 2016;44(4):1257-1274. doi:10.1007/s10439-015-1420-6

12. Beckwith JG, Greenwald RM, Chu JJ. Measuring Head Kinematics in Football: Correlation Between the Head Impact Telemetry System and Hybrid III Headform. *Ann Biomed Eng*. 2012;40(1):237-248. doi:10.1007/s10439-011-0422-2.Measuring

13. Cummiskey B, Schiffmiller D, Talavage TM, et al. Reliability and accuracy of helmet-mounted and head-mounted devices used to measure head accelerations. *Proc Inst Mech Eng Part P J Sports Eng Technol*. 2017;231(2):144-153. doi:10.1177/1754337116658395

14. Wood GW, Panzer MB, Bass CR, Myers BS. Viscoelastic Properties of Hybrid III Head Skin. *SAE Int J Mater Manuf*. 2010;3(1):186-193. doi:10.4271/2010-01-0383

15. Trotta A, Annaidh AN, Burek RO, Pelgrims B, Ivens J. Evaluation of the head-helmet sliding properties in an impact test. *J Biomech*. 2018;0(0). doi:10.1016/j.jbiomech.2018.05.003

16. Langlois JA, Rutland-Brown W, Wald MM. The epidemiology and impact of traumatic brain injury: a brief overview. *J Head Trauma Rehabil*. 2006;21(5):375-378.

17. Bryan MA, Rowhani-Rahbar A, Comstock RD, Rivara F, Collaborative on behalf of the SSCR. Sports- and Recreation-Related Concussions in US Youth. *Pediatrics*. 2016;138(1):e20154635. doi:10.1542/peds.2015-4635

18. Finkelstein EA, Corso PS, Miller TR. *The Incidence and Economic Burden of Injuries in the United States*. Oxford University Press; 2006.

19. Sullivan S, Friess SH, Ralston J, et al. Behavioral deficits and axonal injury persistence after rotational head injury are direction dependent. *J Neurotrauma*. 2013;30(7):538-545. doi:10.1089/neu.2012.2594

20. Gennarelli TA, Thibault LE, Adams JH, Graham DI, Thompson CJ, Marcincin RP. Diffuse axonal injury and traumatic coma in the primate. *Ann Neurol*. 1982;12(6):564-574. doi:10.1002/ana.410120611

21. Urban JE, Daniel RW, Cobb BR. Cumulative Exposure Risk of Concussion for Youth and High School Football Head Impacts. In: ; 2013:1-10. http://ibrc.osu.edu/wp-content/uploads/2013/12/Urban_Manuscript_2013.pdf.

22. 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;40(1):79-89. doi:10.1007/s10439-011-0400-8

23. Rowson S, Duma SM. Brain injury prediction: assessing the combined probability of concussion using linear and rotational head acceleration. *Ann Biomed Eng*. 2013;41(5):873-882. doi:10.1007/s10439-012-0731-0

24. Viano DC, Casson IR, Pellman EJ. Concussion in professional football: biomechanics of the struck player--part 14. *Neurosurgery*. 2007;61(2):313-327; discussion 327-8. doi:10.1227/01.NEU.0000279969.02685.D0

25. Viano DC, Pellman EJ. Concussion in professional football: Biomechanics of the striking player - Part 8. *Neurosurgery*. 2005;56(2):266-278. doi:10.1227/01.Neu.0000150035.54230.3c

26. McKee AC, Stern RA, Nowinski CJ, et al. The spectrum of disease in chronic traumatic encephalopathy. *Brain*. 2013;136(Pt 1):43-64. doi:10.1093/brain/aws307

27. Clark MD, Asken BM, Marshall SW, Guskiewicz KM. Descriptive characteristics of concussions in National Football League games, 2010-2011 to 2013-2014. *Am J Sports Med*. 2017;45(4):929-936.

28. Wu LC, Nangia V, Bui K, et al. In Vivo Evaluation of Wearable Head Impact Sensors. *Ann Biomed Eng*. 2016;44(4):1234-1245. doi:10.1007/s10439-015-1423-3

29. Chu JJ, Beckwith JG, Crisco JJ, Greenwald RM. A novel algorithm to measure linear and rotational head acceleration using single-axis accelerometers. *J Biomech*. 2006;39(Suppl 1):2006-2006.

30. Crisco JJ, Chu JJ, Greenwald RM. An algorithm for estimating acceleration magnitude and impact location using multiple nonorthogonal single-axis accelerometers. *J Biomech Eng*. 2004;126(6):849-854.

31. Foster JK, Kortge JO, Wolanin MJ. *Hybrid III-a Biomechanically-Based Crash Test Dummy*. SAE Technical Paper; 1977.

32. Hubbard RP, McLeod DG. *Definition and Development of a Crash Dummy Head*. SAE Technical Paper; 1974.

33. Cobb BR, MacAlister A, Young TJ, Kemper AR, Rowson S, Duma SM. Quantitative comparison of Hybrid III and National Operating Committee on Standards for Athletic Equipment headform shape characteristics and implications on football helmet fit. *Proc Inst Mech Eng Part P J Sports Eng Technol*. 2014:1754337114548245.

34. Crisco JJ, Fiore R, Beckwith JG, et al. Frequency and location of head impact exposures in individual collegiate football players. *J Athl Train*. 2010;45(6):549-559. doi:10.4085/1062-6050-45.6.549

35. McCaffrey MA, Mihalik JP, Crowell DH, Shields EW, Guskiewicz KM. Measurement of head impacts in collegiate football players: clinical measures of concussion after high- and low-magnitude impacts. *Neurosurgery*. 2007;61(6):1236-1243; discussion 1243. doi:10.1227/01.neu.0000306102.91506.8b

36. Guskiewicz KM, Mihalik JP, Shankar V, et al. Measurement of head impacts in collegiate football players: relationship between head impact biomechanics and acute clinical outcome after concussion. *Neurosurgery*. 2007;61(6):1244-1252; discussion 1252-3. doi:10.1227/01.neu.0000306103.68635.1a

37. Rowson S, Beckwith JG, Chu JJ, Leonard DS, Greenwald RM, Duma SM. A six degree of freedom head acceleration measurement device for use in football. *J Appl Biomech*. 2011;27(1):8-14.

38. Newman JA, Beusenberg MC, Shewchenko N, Withnall C, Fournier E. Verification of biomechanical methods employed in a comprehensive study of mild traumatic brain injury and the effectiveness of American football helmets. *J Biomech*. 2005;38(7):1469-1481. doi:10.1016/j.jbiomech.2004.06.025

39. Rath AL, Manoogian SJ, Duma SM, Bolton BJ, Crandall JR. An evaluation of a fiber optic based sensor for measuring chest and abdominal deflection. *SAE Trans*. 2005:769–778.

40. Franck JA, Blume J, Crisco JJ, Franck C. Extracting Time-Accurate Acceleration Vectors From Nontrivial Accelerometer Arrangements. *J Biomech Eng*. 2015;137(9):091004-091004-091011. doi:10.1115/1.4030942

41. Harpham JA, Mihalik JP, Littleton AC, Frank BS, Guskiewicz KM. The effect of visual and sensory performance on head impact biomechanics in college football players. *Ann Biomed Eng*. 2014;42(1):1-10. doi:10.1007/s10439-013-0881-8

42. Schmidt JD, Guskiewicz KM, Mihalik JP, Blackburn JT, Siegmund GP, Marshall SW. Does Visual Performance Influence Head Impact Severity Among High School Football Athletes? *Clin J Sport Med*. 2015;25(6):494-501. doi:10.1097/JSM.0000000000000143

43. Ford JM, Campbell KR, Ford CB, Boyd KE, Padua DA, Mihalik JP. Can Functional Movement Assessment Predict Football Head Impact Biomechanics? *Med Sci Sports Exerc*. 2017;Publish Ahead of Print. doi:10.1249/MSS.0000000000001538

44. Loyd AM, Nightingale RW, Song Y, et al. The response of the adult and ATD heads to impacts onto a rigid surface. *Accid Anal Prev*. 2014;72:219–229.

45. Kuo Calvin, Wu LC, Hammoor BT, et al. Effect of the mandible on mouthguard measurements of head kinematics. *J Biomech*. 2016;49(9):1845-1853. doi:10.1016/j.jbiomech.2016.04.017

46. Automotive S of, Engineers. Instrumentation for Impact Test, Part 1: Electronic Instrumentation. 2007:SAE J211/1-SAE J211/1.

47. Wu LC, Laksari K, Kuo C, et al. Bandwidth and sample rate requirements for wearable head impact sensors. *J Biomech*. 2016;49(13):2918-2924.

48. Abramowitz M, Stegun IA. *Handbook of Mathematical Functions: With Formulas, Graphs, and Mathematical Tables*. Vol 55. Courier Corporation; 1964.

49. Walker LB, Harris EH, Pontius UR. *Mass, Volume, Center of Mass, and Mass Moment of Inertia of Head and Head and Neck of Human Body*. SAE Technical Paper; 1973.

50. Camarillo DB, Shull PB, Mattson J, Shultz R, Garza D. An instrumented mouthguard for measuring linear and angular head impact kinematics in American football. *Ann Biomed Eng*. 2013;41(9):1939-1949. doi:10.1007/s10439-013-0801-y

51. Kent JT. The Fisher-Bingham Distribution on the Sphere. *J R Stat Soc Ser B Methodol*. 1982;44(1):71-80.

52. Leong P, Carlile S. Methods for spherical data analysis and visualization. *J Neurosci Methods*. 1998;80(2):191–200.

53. Elkin BS, Gabler LF, Panzer MB, Siegmund GP. Brain tissue strains vary with head impact location: A possible explanation for increased concussion risk in struck versus striking football players. *Clin Biomech*. 2018. doi:10.1016/j.clinbiomech.2018.03.021

54. Jadischke R, Viano DC, Dau N, King AI, McCarthy J. On the accuracy of the Head Impact Telemetry (HIT) System used in football helmets. *J Biomech*. 2013;46(13):2310-2315. doi:10.1016/j.jbiomech.2013.05.030

55. Mihalik JP, Blackburn JT, Greenwald RM, Cantu RC, Marshall SW, Guskiewicz KM. Collision type and player anticipation affect head impact severity among youth ice hockey players. *Pediatrics*. 2010;125(6):e1394-401. doi:10.1542/peds.2009-2849

56. Mihalik JP, Guskiewicz KM, Marshall SW, Greenwald RM, Blackburn JT, Cantu RC. Does cervical muscle strength in youth ice hockey players affect head impact biomechanics? *Clin J Sport Med*. 2011;21(5):416-421. doi:10.1097/JSM.0B013E31822C8A5C

57. Schmidt JD, Guskiewicz KM, Blackburn JT, Mihalik JP, Siegmund GP, Marshall SW. The influence of cervical muscle characteristics on head impact biomechanics in football. *Am J Sports Med*. 2014;42(9):2056-2066. doi:10.1177/0363546514536685

58. Liao S, Lynall RC, Mihalik JP. The Effect of Head Impact Location on Day of Diagnosed Concussion in College Football. *Med Sci Sports Exerc*. February 2016. doi:10.1249/mss.0000000000000896

59. Schmidt JD, Guskiewicz KM, Mihalik JP, Blackburn JT, Siegmund GP, Marshall SW. Head Impact Magnitude in American High School Football. *Pediatrics*. 2016;138(2). doi:10.1542/peds.2015-4231

60. Mihalik JP, Guskiewicz KM, Marshall SW, Blackburn JT, Cantu RC, Greenwald RM. Head impact biomechanics in youth hockey: comparisons across playing position, event types, and impact locations. *Ann Biomed Eng*. 2012;40(1):141-149. doi:10.1007/s10439-011-0405-3

61. Gysland SM, Mihalik JP, Register-Mihalik JK, Trulock SC, Shields EW, Guskiewicz KM. The relationship between subconcussive impacts and concussion history on clinical measures of neurologic function in collegiate football players. *Ann Biomed Eng*. 2012;40(1):14-22. doi:10.1007/s10439-011-0421-3

62. Kerr ZY, Collins CL, Mihalik JP, Marshall SW, Guskiewicz KM, Comstock RD. Impact locations and concussion outcomes in high school football player-to-player collisions. *Pediatrics*. 2014;134(3):489-496. doi:10.1542/peds.2014-0770

63. Kerr Z, Campbell KR, Fraser MA, et al. Head impact locations in US high school boys' and girls' soccer concussions, 2012/13-2015/16. *J Neurotrauma*. November 2017. doi:10.1089/neu.2017.5319

64. Press JN, Rowson S. Quantifying Head Impact Exposure in Collegiate Women's Soccer. *Clin J Sport Med*. March 2016. doi:10.1097/jsm.0000000000000313

65. Cortes N, Lincoln AE, Myer GD, et al. Video analysis verification of head impact events measured by wearable sensors. *Am J Sports Med*. 2017:0363546517706703.

66. Nevins D, Hildenbrand K, Kensrud J, Vasavada A, Smith L. Laboratory and field evaluation of a small form factor head impact sensor in un-helmeted play. *Proc Inst Mech Eng Part P J Sports Eng Technol*. November 2017:1754337117739458. doi:10.1177/1754337117739458

67. Beckwith JG, Greenwald RM, Chu JJ, et al. Timing of concussion diagnosis is related to head impact exposure prior to injury. *Med Sci Sports Exerc*. 2013;45(4):747-754. doi:10.1249/MSS.0b013e3182793067

68. Mihalik J, Moise K, Ocwieja K, Ocwieja K, Guskiewicz K, Register-Mihalik J. The Effects of Player Anticipation and Involvement on Head Impact Biomechanics in College Football Body Collisions. *Mech Concussion Sports*. June 2014. doi:10.1520/STP155220120108

69. Kuo C, Wu L, Loza J, Senif D, Anderson SC, Camarillo DB. Comparison of video-based and sensor-based head impact exposure. *PLOS ONE*. 2018;13(6):e0199238. doi:10.1371/journal.pone.0199238

70. Viera AJ, Garrett JM. Understanding interobserver agreement: the kappa statistic. *Fam Med*. 2005;37(5):360-363.

71. Lynall RC, Clark MD, Grand EE, et al. Head Impact Biomechanics in Women's College Soccer. *Med Sci Sports Exerc*. 2016;48(9):1772-1778. doi:10.1249/mss.0000000000000951

72. Mihalik JP, Guskiewicz KM, Jeffries J a, Greenwald RM, Marshall SW. Characteristics of head impacts sustained by youth ice hockey players. *Proc Inst Mech Eng Part P J Sports Eng Technol*. 2008;222(1):45-52. doi:10.1243/17543371JSET4

73. Manley GT, Gardner AJ, Schneider KJ, et al. A systematic review of potential long-term effects of sport-related concussion. *Br J Sports Med*. 2017:bjsports-2017-097791.

74. Ommaya AK, Rockoff SD, Baldwin M. Experimental Concussion; a First Report. *J Neurosurg*. 1964;21(4):249-265. doi:10.3171/jns.1964.21.4.0249

75. Ewing CL, Thomas DJ. *Human Head and Neck Response to Impact Acceleration.* NAVAL AEROSPACE MEDICAL RESEARCH LAB PENSACOLA FL; 1972.

76. Pellman EJ, Viano DC, Tucker AM, Casson IR, Committee on Mild Traumatic Brain Injury NFL. Concussion in professional football: location and direction of helmet impacts-Part 2. *Neurosurgery*. 2003;53(6):1328-1340; discussion 1340-1. doi:10.1227/01.NEU.0000093499.20604.21

77. Gennarelli TA, Thibault L e, Ommaya A k. Pathophysiologic responses to rotational and translational accelerations of the head. *Proc Stapp Car Crash Conf*. 1972;16. http://dx.doi.org/10.4271/720970.

78. Sullivan S, Eucker SA, Gabrieli D, et al. White matter tract-oriented deformation predicts traumatic axonal brain injury and reveals rotational direction-specific vulnerabilities. *Biomech Model Mechanobiol*. 2015;14(4):877-896. doi:10.1007/s10237-014-0643-z

79. Urban JE, Davenport EM, Golman AJ, et al. Head impact exposure in youth football: high school ages 14 to 18 years and cumulative impact analysis. *Ann Biomed Eng*. 2013;41(12):2474-2487. doi:10.1007/s10439-013-0861-z

80. King D, Hume PA, Brughelli M, Gissane C. Instrumented mouthguard acceleration analyses for head impacts in amateur rugby union players over a season of matches. *Am J Sports Med*. 2015;43(3):614-624. doi:10.1177/0363546514560876

81. Caccese JB, Lamond LC, Buckley TA, Kaminski TW. Reducing purposeful headers from goal kicks and punts may reduce cumulative exposure to head acceleration. *Res Sports Med Print*. 2016;24(4):407-415. doi:10.1080/15438627.2016.1230549

82. Reynolds BB, Patrie J, Henry EJ, et al. Quantifying Head Impacts in Collegiate Lacrosse. *Am J Sports Med*. June 2016. doi:10.1177/0363546516648442

83. Greenwald RM, Gwin JT, Chu JJ, Crisco JJ. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. *Neurosurgery*. 2008;62(4):789-798; discussion 798. doi:10.1227/01.neu.0000318162.67472.ad

84. Broglio SP, Lapointe A, O’Connor KL, McCrea M. Head Impact Density: A Model To Explain the Elusive Concussion Threshold. *J Neurotrauma*. 2017;34(19):2675-2683. doi:10.1089/neu.2016.4767

85. Broglio SP, Sosnoff JJ, Shin S, He X, Alcaraz C, Zimmerman J. Head impacts during high school football: a biomechanical assessment. *J Athl Train*. 2009;44(4):342-349. doi:10.4085/1062-6050-44.4.342

86. Cobb BR, Urban JE, Davenport EM, et al. Head impact exposure in youth football: elementary school ages 9-12 years and the effect of practice structure. *Ann Biomed Eng*. 2013;41(12):2463-2473. doi:10.1007/s10439-013-0867-6

87. Kelley ME, Urban JE, Miller LE, et al. Head Impact Exposure in Youth Football: Comparing Age- and Weight-Based Levels of Play. *J Neurotrauma*. 2017;34(11):1939-1947. doi:10.1089/neu.2016.4812

88. Martini D, Eckner J, Kutcher J, Broglio SP. Subconcussive head impact biomechanics: comparing differing offensive schemes. *Med Sci Sports Exerc*. 2013;45(4):755-761. doi:10.1249/MSS.0b013e3182798758

89. McAllister TW, Flashman LA, Maerlender A, et al. Cognitive effects of one season of head impacts in a cohort of collegiate contact sport athletes. *Neurology*. 2012;78(22):1777-1784. doi:10.1212/WNL.0b013e3182582fe7

90. Kuzminski SJ, Clark MD, Fraser MA, et al. White Matter Changes Related to Subconcussive Impact Frequency during a Single Season of High School Football. *Am J Neuroradiol*. December 2018. doi:10.3174/ajnr.A5489

91. Breedlove KM, Breedlove EL, Robinson M, et al. Detecting neurocognitive & neurophysiological changes as a result of subconcussive blows in high school football athletes. *Athl Train Sports Health Care*. 2014;6(3):119-127.

92. Rose SC, Yeates KO, Fuerst DR, Ercole PM, Nguyen JT, Pizzimenti NM. Head Impact Burden and Change in Neurocognitive Function During a Season of Youth Football. *J Head Trauma Rehabil*. 2018;Publish Ahead of Print. doi:10.1097/HTR.0000000000000441

93. Crisco JJ, Chu JJ. An Algorithm for Estimating Acceleration Magnitude and Impact Location Using Multiple Nonorthogonal Single-Axis Accelerometers. *J Biomech Eng*. 2004;126(6):849. doi:10.1115/1.1824135

94. Joodaki H, Bailey A, Lessley D, Funk J, Sherwood C, Crandall J. Relative Motion between the Helmet and Head in Football Impact Test. *J Biomech Eng*. March 2019. doi:10.1115/1.4043038

95. Trotta A, Zouzias D, Bruyne GD, Annaidh AN. The Importance of the Scalp in Head Impact Kinematics. *Ann Biomed Eng*. 2018;46(6):831-840. doi:10.1007/s10439-018-2003-0

96. Eckner JT, Conley RS, Garton HJL, et al. Comparing head impact kinematics simultaneously measured using 6 different sensors in a human cadaver model. In: *Neurology*. Vol 91. ; 2018:S2-S3. doi:10.1212/01.wnl.0000550624.74128.72

97. Campbell KR, Lynall RC, Luck JF, et al. Impact Detection Rate Analysis of Four Head Impact Sensors. In: *Journal of Athletic Training*. Vol 51. ; 2016:S206.

98. Humm J, Driesslein K, Shah A, et al. Novel experimental techniques using PMHS and CT scans to assess performance of wearable sensors for mild Traumatic Brain Injury applications for different populations. In: ; 2018.

99. Wu LC, Kuo C, Loza J, et al. Detection of American Football Head Impacts Using Biomechanical Features and Support Vector Machine Classification. *Sci Rep*. 2017;8(1):855. doi:10.1038/s41598-017-17864-3

100. Motiwale S, Eppler W, Hollingsworth D, Hollingsworth C, Morgenthau J, Kraft RH. Application of neural networks for filtering non-impact transients recorded from biomechanical sensors. In: *2016 IEEE-EMBS International Conference on Biomedical and Health Informatics (BHI)*. ; 2016:204-207. doi:10.1109/BHI.2016.7455870

101. Bartsch A, Dama R, Samorezov S, et al. Laboratory and on-field results of athlete head impact monitoring. In: *Neurology*. Vol 91. ; 2018:S1-S1. doi:10.1212/01.wnl.0000550620.51257.e3

102. National Federation of State High School Associations. National Federation of State High School Associations: 2017-2018 Participation Statistics. https://www.nfhs.org/ParticipationStatistics/ParticipationStatistics/. Published 2018 2017. Accessed April 4, 2019.

103. Meehan WP, d’Hemecourt P, Collins CL, Comstock RD. Assessment and Management of Sport-Related Concussions in United States High Schools. *Am J Sports Med*. 2011;39(11):2304-2310. doi:10.1177/0363546511423503

104. Patel DR, Greydanus DE. Neurologic considerations for adolescent athletes. *Adolesc Med Phila Pa*. 2002;13(3):569-578.

105. Rosenthal JA, Foraker RE, Collins CL, Comstock RD. National high school athlete concussion rates from 2005-2006 to 2011-2012. *Am J Sports Med*. 2014:0363546514530091.

106. Lincoln AE, Caswell SV, Almquist JL, Dunn RE, Norris JB, Hinton RY. Trends in concussion incidence in high school sports: a prospective 11-year study. *Am J Sports Med*. 2011;39(5):958-963. doi:10.1177/0363546510392326

107. Gessel LM, Fields SK, Collins CL, Dick RW, Comstock RD. Concussions among United States high school and collegiate athletes. *J Athl Train*. 2007;42(4):495-503.

108. O'Connor KL, Baker MM, Dalton SL, Dompier TP, Broglio SP, Kerr ZY. Epidemiology of Sport-Related Concussions in High School Athletes: National Athletic Treatment, Injury and Outcomes Network (NATION), 2011-2012 Through 2013-2014. *J Athl Train Dallas*. 2017;52(3):175-185. doi:http://dx.doi.org/10.4085/1062-6050-52.1.15

109. McCrea M, Hammeke T, Olsen G, Leo P, Guskiewicz K. Unreported concussion in high school football players: implications for prevention. *Clin J Sport Med*. 2004;14(1):13-17.

110. Lynall RC, Campbell KR, Wasserman EB, Dompier TP, Kerr ZY. Concussion Mechanisms and Activities in Youth, High School, and College Football. *J Neurotrauma*. 2017;34(19):2684-2690. doi:10.1089/neu.2017.5032

111. Kerr ZY, Zuckerman SL, Wasserman EB, Covassin T, Djoko A, Dompier TP. Concussion Symptoms and Return to Play Time in Youth, High School, and College American Football Athletes. *JAMA Pediatr*. 2016;170(7):647-653. doi:10.1001/jamapediatrics.2016.0073

112. Dick R, Ferrara MS, Agel J, et al. Descriptive epidemiology of collegiate men's football injuries: National Collegiate Athletic Association Injury Surveillance System, 1988-1989 through 2003-2004. *J Athl Train*. 2007;42(2):221-233.

113. Dompier TP, Kerr ZY, Marshall SW, et al. Incidence of Concussion During Practice and Games in Youth, High School, and Collegiate American Football Players. *JAMA Pediatr*. 2015;169(7):659-665. doi:10.1001/jamapediatrics.2015.0210

114. Wasserman EB, Kerr ZY, Zuckerman SL, Covassin T. Epidemiology of Sports-Related Concussions in National Collegiate Athletic Association Athletes From 2009-2010 to 2013-2014: Symptom Prevalence, Symptom Resolution Time, and Return-to-Play Time. *Am J Sports Med*. 2016;44(1):226-233. doi:10.1177/0363546515610537

115. Kerr ZY, Yeargin S, Valovich McLeod TC, et al. Comprehensive Coach Education and Practice Contact Restriction Guidelines Result in Lower Injury Rates in Youth American Football. *Orthop J Sports Med*. 2015;3(7):2325967115594578. doi:10.1177/2325967115594578

116. Kerr ZY, Dalton SL, Roos KG, Djoko A, Phelps J, Dompier TP. Comparison of Indiana High School Football Injury Rates by Inclusion of the USA Football “Heads Up Football” Player Safety Coach. *Orthop J Sports Med*. 2016;4(5):2325967116648441. doi:10.1177/2325967116648441

117. Black AM, Macpherson AK, Hagel BE, et al. Policy change eliminating body checking in non-elite ice hockey leads to a threefold reduction in injury and concussion risk in 11- and 12-year-old players. *Br J Sports Med*. 2016;50(1):55-61. doi:10.1136/bjsports-2015-095103

118. Marshall SW, Guskiewicz KM, Shankar V, McCrea M, Cantu RC. Epidemiology of sports-related concussion in seven US high school and collegiate sports. *Inj Epidemiol*. 2015;2(1). doi:10.1186/s40621-015-0045-4

119. Casson IR, Viano DC, Powell JW, Pellman EJ. Twelve Years of National Football League Concussion Data. *Sports Health*. 2010;2(6):471-483. doi:10.1177/1941738110383963

120. Ruestow PS, Duke TJ, Finley BL, Pierce JS. Effects of the NFL’s Amendments to the Free Kick Rule on Injuries during the 2010 and 2011 Seasons. *J Occup Environ Hyg*. 2015;12(12):875-882. doi:10.1080/15459624.2015.1072632

121. Pellman EJ, Powell JW, Viano DC, et al. Concussion in professional football: epidemiological features of game injuries and review of the literature—part 3. *Neurosurgery*. 2004;54(1):81–96.

122. Wiebe DJ, D’Alonzo BA, Harris R, Putukian M, Campbell-McGovern C. Association Between the Experimental Kickoff Rule and Concussion Rates in Ivy League Football. *JAMA*. 2018;320(19):2035-2036. doi:10.1001/jama.2018.14165

123. Bazarian JJ, Atabaki S. Predicting postconcussion syndrome after minor traumatic brain injury. *Acad Emerg Med Off J Soc Acad Emerg Med*. 2001;8(8):788-795.

124. Fanton M, Kuo C, Sganga J, Hernandez F, Camarillo DB. Dependency of head impact rotation on head-neck positioning and soft tissue forces. *IEEE Trans Biomed Eng*. 2018:1-1. doi:10.1109/TBME.2018.2866147

125. Kuo C, Fanton M, Wu L, Camarillo D. Spinal Constraint Modulates Head Instantaneous Center of Rotation and Dictates Head Angular Motion. *J Biomech*. 2018;0(0). doi:10.1016/j.jbiomech.2018.05.024

126. Eckersley CP, Nightingale RW, Luck JF, Bass CR. The role of cervical muscles in mitigating concussion. *J Sci Med Sport*. January 2019. doi:10.1016/j.jsams.2019.01.009

127. Badachhape AA, Okamoto RJ, Durham RS, et al. The Relationship of Three-Dimensional Human Skull Motion to Brain Tissue Deformation in Magnetic Resonance Elastography Studies. *J Biomech Eng*. 2017;139(5):051002.

128. Bayly PV, Cohen TS, Leister EP, Ajo D, Leuthardt E, Genin GM. Deformation of the human brain induced by mild acceleration. *J Neurotrauma*. 2005;22(8):845-856. doi:10.1089/neu.2005.22.845

129. Sabet AA, Christoforou E, Zatlin B, Genin GM, Bayly PV. Deformation of The Human Brain Induced By Mild Angular Head Acceleration. *J Biomech*. 2008;41(2):307-315. doi:10.1016/j.jbiomech.2007.09.016

130. Alshareef A, Giudice JS, Forman J, Salzar RS, Panzer MB. A Novel Method for Quantifying Human In Situ Whole Brain Deformation Under Rotational Loading Using Sonomicrometry. *J Neurotrauma*. November 2017. doi:10.1089/neu.2017.5362

131. Hardy WN, Mason MJ, Foster CD, et al. A study of the response of the human cadaver head to impact. *Stapp Car Crash J*. 2007;51:17-80.

132. Hardy WN, Foster CD, Mason MJ, Yang KH, King AI, Tashman S. Investigation of Head Injury Mechanisms Using Neutral Density Technology and High-Speed Biplanar X-ray. *Stapp Car Crash J*. 2001;45:337-368.

133. Chatelin S, Constantinesco A, Willinger R. Fifty years of brain tissue mechanical testing: from in vitro to in vivo investigations. *Biorheology*. 2010;47(5-6):255-276. doi:10.3233/bir-2010-0576

134. Ahmadzadeh H, Smith DH, Shenoy VB. Viscoelasticity of tau proteins leads to strain rate-dependent breaking of microtubules during axonal stretch injury: predictions from a mathematical model. *Biophys J*. 2014;106(5):1123-1133. doi:10.1016/j.bpj.2014.01.024

135. Smith DH, Meaney DF. Axonal damage in traumatic brain injury. *The neuroscientist*. 2000;6(6):483–495.

136. Unterharnscheidt F, Higgins LS. Traumatic lesions of brain and spinal cord due to nondeforming angular acceleration of the head. *Tex Rep Biol Med*. 1969;27(1):127-166.

137. Prange MT, Margulies SS. Regional, directional, and age-dependent properties of the brain undergoing large deformation. *J Biomech Eng*. 2002;124(2):244–252.

138. Giza CC, Hovda DA. The new neurometabolic cascade of concussion. *Neurosurgery*. 2014;75 Suppl 4:S24-33. doi:10.1227/NEU.0000000000000505

139. Davis GA, Anderson V, Babl FE, et al. What is the difference in concussion management in children as compared with adults? A systematic review. *Br J Sports Med*. 2017;51(12):949-957. doi:10.1136/bjsports-2016-097415

140. Gurdjian ES, Lissner HR, Evans FG, Patrick LM, Hardy WG. Intracranial pressure and acceleration accompanying head impacts in human cadavers. *Surg Gynecol Obstet*. 1961;113:185-190.

141. Lissner HR, Lebow M, Evans FG. Experimental studies on the relation between acceleration and intracranial pressure changes in man. *Surg Gynecol Obstet*. 1960;111:329-338.

142. Thomas LM, Roberts VL, Gurdjian ES. Experimental intracranial pressure gradients in the human skull. *J Neurol Neurosurg Psychiatry*. 1966;29(5):404-411.

143. Fijalkowski RJ, Stemper BD, Pintar FA, Yoganandan N, Crowe MJ, Gennarelli TA. New rat model for diffuse brain injury using coronal plane angular acceleration. *J Neurotrauma*. 2007;24(8):1387-1398.

144. Gurdjian ES, Roberts VL, Thomas LM. Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury. *J Trauma Acute Care Surg*. 1966;6(5):600–604.

145. Ommaya AK, Gennarelli TA. Cerebral concussion and traumatic unconsciousness: correlation of experimental and clinical observations on blunt head injuries. *Brain*. 1974;97(4):633–654.

146. Bailey A, Funk J, Lessley D, et al. Validation of a videogrammetry technique for analysing American football helmet kinematics. *Sports Biomech*. 2018;0(0):1-23. doi:10.1080/14763141.2018.1513059

147. Pellman EJ, Viano DC, Tucker AM, Casson IR, Committee on Mild Traumatic Brain Injury NFL. Concussion in professional football: reconstruction of game impacts and injuries. *Neurosurgery*. 2003;53(6):1328-1340; discussion 1340-1. doi:10.1227/01.NEU.0000083559.68424.3F

148. Hernandez F, Camarillo DB. Voluntary Head Rotational Velocity and Implications for Brain Injury Risk Metrics. *J Neurotrauma*. May 2018. doi:10.1089/neu.2016.4758

149. Thomas LM, Roberts VL, Gurdjian ES. Experimental intracranial pressure gradients in the human skull. *J Neurol Neurosurg Psychiatry*. 1966;29(5):404-411.

150. Gadd CW. Use of a weighted impulse criterion for estimating injruy hazard. In: Society of Automotive Engineers; 1966. http://www.snell-helmets.net/docs/articles/hic/Gadd_1966_Weighted-Impulse_Criterion_for_Estimating_Injury_H.pdf.

151. Versace J. A review of the severity index. In: ; 1971:771-796.

152. Patrick LM, Lissner HR, Gurdjian ES. Survival by design: Head protection. In: *Proceedings: American Association for Automotive Medicine Annual Conference*. Vol 7. Association for the Advancement of Automotive Medicine; 1963:483–499.

153. Hollbourn AHS. Mechanics of head injury. *Lancet*. 1943;2:438–441.

154. Ommaya AK, Rockoff SD, Baldwin M, Friauf WS. Experimental concussion: a first report. *J Neurosurg*. 1964;21(4):249-265.

155. Gennarelli TA, Thibault L e., Ommaya A k. Pathophysiologic responses to rotational and translational accelerations of the head. *Proc Stapp Car Crash Conf*. 1972;16.

156. Ommaya AK. Biomechanics of head injuries: Experimental aspects. *Biomech Trauma*. 1984.

157. Margulies SS, Thibault LE. A proposed tolerance criterion for diffuse axonal injury in man. *J Biomech*. 1992;25(8):917–923.

158. Ji S, Zhao W, Li Z, McAllister TW. Head impact accelerations for brain strain-related responses in contact sports: a model-based investigation. *Biomech Model Mechanobiol*. 2014;13(5):1121-1136. doi:10.1007/s10237-014-0562-z

159. Post A, Hoshizaki TB, Gilchrist MD, Cusimano MD. Peak linear and rotational acceleration magnitude and duration effects on maximum principal strain in the corpus callosum for sport impacts. *J Biomech*. 2017.

160. Miller LE, Urban JE, Stitzel JD. Development and validation of an atlas-based finite element brain model. *Biomech Model Mechanobiol*. January 2016. doi:10.1007/s10237-015-0754-1

161. Zou H, Schmiedeler JP, Hardy WN. Separating brain motion into rigid body displacement and deformation under low-severity impacts. *J Biomech*. 2007;40(6):1183-1191. doi:10.1016/j.jbiomech.2006.06.018

162. Takhounts EG, Craig MJ, Moorhouse K, McFadden J, Hasija V. Development of brain injury criteria (BrIC). *Stapp Car Crash J*. 2013;57:243.

163. Gabler LF, Crandall JR, Panzer MB. Assessment of kinematic brain injury metrics for predicting strain responses in diverse automotive impact conditions. *Ann Biomed Eng*. 2016;44(12):3705-3718.

164. Ommaya AK, Gennarelli TA. CORRELATION OF EXPERIMENTAL AND CLINICAL OBSERVATIONS ON BLUNT HEAD INJURIES. 1974:24.

165. Rowson S, Duma SM, Stemper BD, et al. Correlation of Concussion Symptom Profile with Head Impact Biomechanics: A Case for Individual-Specific Injury Tolerance. *J Neurotrauma*. 2017;35(4):681-690. doi:10.1089/neu.2017.5169

166. Stemper BD, Shah AS, Chiariello R, et al. Prediction of Post-Concussive Behavioral Changes in a Rodent Model Based on Head Rotational Acceleration Characteristics. *Ann Biomed Eng*. 2016;44(11):3252-3265. doi:10.1007/s10439-016-1647-x

167. Lamy M, Baumgartner D, Yoganandan N, Stemper BD, Willinger R. Experimentally validated three-dimensional finite element model of the rat for mild traumatic brain injury. *Med Biol Eng Comput*. 2013;51(3):353-365. doi:10.1007/s11517-012-1004-7

168. Stemper BD, Shah AS, Pintar FA, et al. Head rotational acceleration characteristics influence behavioral and diffusion tensor imaging outcomes following concussion. *Ann Biomed Eng*. 2015;43(5):1071-1088. doi:10.1007/s10439-014-1171-9

169. Seok J, Warren HS, Cuenca AG, et al. Genomic responses in mouse models poorly mimic human inflammatory diseases. *Proc Natl Acad Sci*. 2013;110(9):3507-3512. doi:10.1073/pnas.1222878110

170. Moon DW, Beedle CW, Kovacic CR. Peak head acceleration of athletes during competition--football. *Med Sci Sports*. 1971;3(1):44-50.

171. Reid SE, Epstein HM, O’Dea TJ, Louis MW, Reid SE Jr. Head protection in football. *J Sports Med*. 1974;2(2):86-92.

172. Naunheim RS, Standeven J, Richter C, Lewis LM. Comparison of impact data in hockey, football, and soccer. *J Trauma*. 2000;48(5):938-941.

173. Manoogian S, McNeely D, Goforth M, Brolinson G, Duma S. Head Acceleration is Less than 10 Percent of Helmet Acceleration During a Football Impact. *Biomed Sci Instrum*. 2006;42:383-388.

174. Lessley DJ, Kent RW, Funk JR, et al. Video Analysis of Reported Concussion Events in the National Football League During the 2015-2016 and 2016-2017 Seasons. *Am J Sports Med*. November 2018:363546518804498. doi:10.1177/0363546518804498

175. Anzell AR, Potteiger JA, Kraemer WJ, Otieno S. Changes in Height, Body Weight, and Body Composition in American Football Players From 1942 to 2011. *J Strength Cond Res*. 2013;27(2):277. doi:10.1519/JSC.0b013e31827f4c08

176. Ocwieja, Karen EMihalik, Jason PMarshall, Stephen WSchmidt, Julianne DTrulock, Scott CGuskiewicz, Kevin MengResearch Sup.

177. Rowson S, Duma SM. Development of the STAR evaluation system for football helmets: integrating player head impact exposure and risk of concussion. *Ann Biomed Eng*. 2011;39(8):2130-2140. doi:10.1007/s10439-011-0322-5

178. Rowson S, Duma SM, Greenwald RM, et al. Can helmet design reduce the risk of concussion in football? *J Neurosurg*. 2014;120(4):919-922. doi:10.3171/2014.1.JNS13916

179. Kerr HA, Ledet EH, Ata A, et al. Does instructional video footage improve tackle technique? *Int J Sports Sci Coach*. 2017:1747954117711867.

180. Kerr ZY, Kroshus E, Lee JG, Yeargin SW, Dompier TP. Coaches' Implementation of the USA Football "Heads Up Football" Educational Program. *Health Promot Pr*. March 2017:1524839917700398. doi:10.1177/1524839917700398

181. Kerr ZY, Yeargin SW, Valovich McLeod TC, Mensch J, Hayden R, Dompier TP. Comprehensive coach education reduces head impact exposure in American youth football. *Orthop J Sports Med*. 2015;3(10):2325967115610545.

182. Ng TP, Bussone WR, Duma SM. The effect of gender and body size on linear accelerations of the head observed during daily activities. *Biomed Sci Instrum*. 2006;42:25-30.

183. Beckwith JG, Zhao W, Ji S, et al. Estimated Brain Tissue Response Following Impacts Associated With and Without Diagnosed Concussion. *Ann Biomed Eng*. February 2018:1-12. doi:10.1007/s10439-018-1999-5

184. Crisco JJ, Wilcox BJ, Beckwith JG, et al. Head impact exposure in collegiate football players. *J Biomech*. 2011;44(15):2673-2678. doi:10.1016/j.jbiomech.2011.08.003

185. Daniel RW, Rowson S, Duma SM. Head impact exposure in youth football. *Ann Biomed Eng*. 2012;40(4):976-981. doi:10.1007/s10439-012-0530-7

186. Schnebel B, Gwin JT, Anderson S, Gatlin R. In vivo study of head impacts in football: a comparison of National Collegiate Athletic Association Division I versus high school impacts. *Neurosurgery*. 2007;60(3):490-495; discussion 495-6. doi:10.1227/01.NEU.0000249286.92255.7F

187. Broglio SP, Williams R, Rettmann A, Moore B, Eckner JT, Meehan S. No Seasonal Changes in Cognitive Functioning Among High School Football Athletes: Implementation of a Novel Electrophysiological Measure and Standard Clinical Measures. *Clin J Sport Med*. 2017.

188. Duma SM, Manoogian SJ, Bussone WR, et al. Analysis of real-time head accelerations in collegiate football players. *Clin J Sport Med*. 2005;15(1):3-8.

189. Kelley ME, Kane JM, Espeland MA, et al. Head impact exposure measured in a single youth football team during practice drills. *J Neurosurg Pediatr*. September 2017:1-9. doi:10.3171/2017.5.PEDS16627

190. Campolettano ET, Brolinson G, Rowson S. Postural Control and Head Impact Exposure in Youth Football Players: Comparison of the Balance Error Scoring System and a Force Plate Protocol. *J Appl Biomech*. November 2017:1-25. doi:10.1123/jab.2017-0066

191. Broglio SP, Eckner JT, Surma T, Kutcher JS. Post-concussion cognitive declines and symptomatology are not related to concussion biomechanics in high school football players. *J Neurotrauma*. 2011;28(10):2061-2068. doi:10.1089/neu.2011.1905

192. Tyson AM, Duma SM, Rowson S. Laboratory Evaluation of Low-Cost Wearable Sensors for Measuring Head Impacts in Sports. *J Appl Biomech*. April 2018:1-24. doi:10.1123/jab.2017-0256

193. Miller L, Kuo C, Wu LC, Urban J, Camarillo D, Stitzel JD. Validation of a Custom Instrumented Retainer Form Factor for Measuring Linear and Angular Head Impact Kinematics. *J Biomech Eng*. January 2018. doi:10.1115/1.4039165

194. Campbell KR, Warnica MJ, Levine IC, et al. Laboratory Evaluation of the gForce Tracker, a Head Impact Kinematic Measuring Device for Use in Football Helmets. *Ann Biomed Eng*. 2016;44(4):1246-1256. doi:10.1007/s10439-015-1391-7

195. Campbell KR, Lynall RC, Luck JF, et al. Peak Resultant Linear Acceleration Agreement Between Common Head Impact Sensor Devices. In: *Journal of Athletic Training*. Vol 52. ; 2017:S68.

196. Allison MA, Kang YS, Bolte JH th, Maltese MR, Arbogast KB. Validation of a helmet-based system to measure head impact biomechanics in ice hockey. *Med Sci Sports Exerc*. 2014;46(1):115-123. doi:10.1249/MSS.0b013e3182a32d0d

197. Hanlon E, Bir C. Validation of a wireless head acceleration measurement system for use in soccer play. *J Appl Biomech*. 2010;26(4):424-431.

198. Beckwith JG, Chu JJ, Greenwald RM. Validation of a noninvasive system for measuring head acceleration for use during boxing competition. *J Appl Biomech*. 2007;23(3):238-244.

199. Padgaonkar AJ, Krieger KW, King AI. Measurement of Angular Acceleration of a Rigid Body Using Linear Accelerometers. *J Appl Mech*. 1975;42(3):552. doi:10.1115/1.3423640

200. Gwin JT, Chu JJ, Greenwald RM. Head impact telemetry system^{TM} for measurement of head acceleration in ice hockey. *J Biomech*. 2006;39:S153.

201. Gwin JT, Chu JJ, Diamond SG, Halstead PD, Crisco JJ, Greenwald RM. An investigation of the NOCSAE linear impactor test method based on in vivo measures of head impact acceleration in American football. *J Biomech Eng*. 2010;132(1):011006. doi:10.1115/1.4000249

202. National Committee on Standards for Athletic E. Nocsae standard linear impactor test method and equipment used in evaluating the performance characteristics of protective headgear and faceguards. 2006;Document #:1-6.

203. Hanlon EM, Bir CA. Real-time head acceleration measurement in girls' youth soccer. *Med Sci Sports Exerc*. 2012;44(6):1102-1108. doi:10.1249/MSS.0b013e3182444d7d

204. Rousseau P. *The Influence of Neck Compliance and Head Displacement on Impact Dynamics of a Hybrid III Head*. University of Ottawa; 2008.

205. Begonia MT, Pintar FA, Yoganandan N. Comparison of NOCSAE Head Kinematics Using the Hybrid III and EuroSID-2 Necks. *J Biomech*. August 2018. doi:10.1016/j.jbiomech.2018.08.018

206. Kuo C, Wu L, Zhao W, Fanton M, Ji S, Camarillo DB. Propagation of errors from skull kinematic measurements to finite element tissue responses. *Biomech Model Mechanobiol*. August 2017. doi:10.1007/s10237-017-0957-8

207. Bina AJ, Batt GS, DesJardins JD. A review of laboratory methods and results used to evaluate protective headgear in American football. *Proc Inst Mech Eng Part P J Sports Eng Technol*. 2018;232(4):360-368. doi:10.1177/1754337118759360

208. Hernandez F, Shull PB, Camarillo DB. Evaluation of a laboratory model of human head impact biomechanics. *J Biomech*. 2015;48(12):3469-3477. doi:10.1016/j.jbiomech.2015.05.034

209. National Committee on Standards for Athletic E. Standard Test Method And Equipment Used In Evaluating The Performance Characteristics Of Headgear/Equipment. 2018;Document #:1-6.

210. Post A, Oeur A, Walsh E, Hoshizaki B, Gilchrist MD. A centric/non-centric impact protocol and finite element model methodology for the evaluation of American football helmets to evaluate risk of concussion. *Comput Methods Biomech Biomed Engin*. 2014;17(16):1785-1800. doi:10.1080/10255842.2013.766724

211. Bonin SJ, DeMarco AL, Siegmund GP. The Effect of Hair and Football Helmet Fit on Headform Kinematics. In: Vol 50. ; 2018:317-319.

212. Mihalik JP, Sumrall AZ, Yeargin SW, et al. Environmental and Physiological Factors Affect Football Head Impact Biomechanics. *Med Sci Sports Exerc*. 2017;49(10):2093–2101.

213. Guskiewicz KM, Mihalik JP. Biomechanics of sport concussion: quest for the elusive injury threshold. *Exerc Sport Sci Rev*. 2011;39(1):4-11. doi:10.1097/JES.0b013e318201f53e

214. Rowson S, Brolinson G, Goforth M, Dietter D, Duma S. Linear and angular head acceleration measurements in collegiate football. *J Biomech Eng*. 2009;131(6):061016. doi:10.1115/1.3130454

215. Hubbard RP, McLeod DG. A Basis for Crash Dummy Skull and Head Geometry. In: King WF, Mertz HJ, eds. *Human Impact Response: Measurement and Simulation*. Boston, MA: Springer US; 1973:129-152. doi:10.1007/978-1-4757-1502-6_7

216. Hertzberg HT, Daniels GS, Churchill E. *Anthropometry of Flying Personnel – 1950*. ANTIOCH COLL YELLOW SPRINGS OH; 1954. https://apps.dtic.mil/docs/citations/AD0047953. Accessed February 9, 2019.

217. Hodgson VR. National Operating Committee on Standards for Athletic Equipment football helmet certification program. *Med Sci Sports*. 1975;7(3):225-232.

218. Rowson B, Rowson S, Duma SM. Hockey STAR: A Methodology for Assessing the Biomechanical Performance of Hockey Helmets. *Ann Biomed Eng*. March 2015. doi:10.1007/s10439-015-1278-7

219. Pellman EJ, Viano DC, Withnall C, Shewchenko N, Bir CA, Halstead PD. Concussion in professional football: helmet testing to assess impact performance—part 11. *Neurosurgery*. 2006;58(1):78-96.

220. Higgins M, Halstead PD, Snyder-Mackler L, Barlow D. Measurement of impact acceleration: mouthpiece accelerometer versus helmet accelerometer. *J Athl Train*. 2007;42(1):5-10.

221. PROPER FIT | RIDDELL. https://www.riddell.com/ProperFit. Accessed February 9, 2019.

222. Ebrahimi I, Golnaraghi F, Wang GG. Factors Influencing the Oblique Impact Test of Motorcycle Helmets. *Traffic Inj Prev*. 2015;16(4):404-408. doi:10.1080/15389588.2014.937804

223. Tolhurst DE, Carstens MH, Greco RJ, Hurwitz DJ. The surgical anatomy of the scalp. *Plast Reconstr Surg*. 1991;87(4):603–12.

224. Young RW. Age changes in the thickness of the scalp in white males. *Hum Biol*. 1959;31(1):74–79.

225. Khalil TB, Goldsmith W, Sackman JL. Impact on a model head-helmet system. *Int J Mech Sci*. 1974;16(9):609–625.

226. Hardy CH, Marcal PV. Elastic analysis of a skull. *J Appl Mech*. 1973;40(4):838–842.

227. Chan HS. *Mathematical Model for Closed Head Impact*. SAE Technical Paper; 1974.

228. Falland-Cheung L, Neil Waddell J, Li KC, Tong DC, Brunton PA. Anatomical head model to measure impact force transfer through the head layers and their displacement. *J Concussion*. 2018;2:2059700218777829. doi:10.1177/2059700218777829

229. Nightingale RW, McElhaney JH, Richardson WJ, Myers BS. Dynamic responses of the head and cervical spine to axial impact loading. *J Biomech*. 1996;29(3):307-318.

230. Schmidt G, Kallieris D, Barz J, Mattern R. *Results of 49 Cadaver Tests Simulating Frontal Collision of Front Seat Passengers*. SAE Technical Paper; 1974.

231. Powell WR, Ojala SJ, Advani SH, Martin RB. *Cadaver Femur Responses to Longitudinal Impacts*. SAE Technical Paper; 1975.

232. Kleinberger M, Sun E, Eppinger R, Kuppa S, Saul R. Development of improved injury criteria for the assessment of advanced automotive restraint systems. *NHTSA Docket*. 1998;4405:9.

233. King AI, Viano DC, Mizeres N. Humanitarian benefits of cadaver research on injury prevention. *J Trauma Acute Care Surg*. 1995;38(4):564–569.

234. Dejeammes M, Tarrière C, Thomas C, Kallieris D. Exploration of biomechanical data towards a better evaluation of tolerance for children involved in automotive accidents. *SAE Trans*. 1984:742–755.

235. McIntosh AS, McCrory P, Comerford J. The dynamics of concussive head impacts in rugby and Australian rules football. *Med Sci Sports Exerc*. 2000;32(12):1980-1984.

236. Tierney GJ, Joodaki H, Krosshaug T, Forman JL, Crandall JR, Simms CK. Assessment of model-based image-matching for future reconstruction of unhelmeted sport head impact kinematics. *Sports Biomech*. 2018;17(1):33-47. doi:10.1080/14763141.2016.1271905

237. Koga H, Nakamae A, Shima Y, et al. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med*. 2010;38(11):2218-2225. doi:10.1177/0363546510373570

238. Caswell SV, Lincoln AE, Stone H, et al. Characterizing Verified Head Impacts in High School Girls' Lacrosse. *Am J Sports Med*. September 2017:0363546517724754. doi:10.1177/0363546517724754

239. Kindschi K, Higgins M, Hillman A, Penczek G, Lincoln A. Video analysis of high-magnitude head impacts in men's collegiate lacrosse. *BMJ Open Sport Exerc Med*. 2017;3.

240. Naunheim RS, Bayly PV, Standeven J, Neubauer JS, Lewis LM, Genin GM. Linear and angular head accelerations during heading of a soccer ball. *Med Sci Sports Exerc*. 2003;35(8):1406-1412. doi:10.1249/01.MSS.0000078933.84527.AE

241. Tierney RT, Higgins M, Caswell SV, et al. Sex Differences in Head Acceleration During Heading While Wearing Soccer Headgear. *J Athl Train*. 2008;43(6):578-584.

242. Brolinson PG, Manoogian S, McNeely D, Goforth M, Greenwald R, Duma S. Analysis of linear head accelerations from collegiate football impacts. *Curr Sports Med Rep*. 2006;5(1):23-28.

243. Stemper BD, Shah AS, Harezlak J, et al. Comparison of Head Impact Exposure Between Concussed Football Athletes and Matched Controls: Evidence for a Possible Second Mechanism of Sport-Related Concussion. *Ann Biomed Eng*. October 2018. doi:10.1007/s10439-018-02136-6

244. Brainard LL, Beckwith JG, Chu JJ, et al. Gender differences in head impacts sustained by collegiate ice hockey players. *Med Sci Sports Exerc*. 2012;44(2):297-304. doi:10.1249/MSS.0b013e31822b0ab4

245. Bellamkonda S, Woodward SJ, Campolettano E, et al. Head Impact Exposure in Practices Correlates with Exposure in Games for Youth Football Players. *J Appl Biomech*. 2018:1–22.

246. McPherson MN, Montelpare WJ, Keightley M, et al. An Analysis of Head Impact Profiles and Safe Hockey Behaviors in Youth Hockey Players. 2010;6(10):1-7.

247. Wilcox BJ, Machan JT, Beckwith JG, Greenwald RM, Burmeister E, Crisco JJ. Head-Impact Mechanisms in Men's and Women's Collegiate Ice Hockey. *J Athl Train*. 2014;49(4):514-520. doi:10.4085/1062-6050-49.3.19

248. O'Connor KL, Peeters T, Szymanski S, Broglio SP. Individual Impact Magnitude vs. Cumulative Magnitude for Estimating Concussion Odds. *Ann Biomed Eng*. 2017:1-8.

249. Broglio SP, Schnebel B, Sosnoff JJ, et al. Biomechanical properties of concussions in high school football. *Med Sci Sports Exerc*. 2010;42(11):2064-2071. doi:10.1249/MSS.0b013e3181dd9156

250. Broglio SP, Eckner JT, Martini D, Sosnoff JJ, Kutcher JS, Randolph C. Cumulative head impact burden in high school football. *J Neurotrauma*. 2011;28(10):2069-2078. doi:10.1089/neu.2011.1825

251. Eckner JT, Sabin M, Kutcher JS, Broglio SP. No evidence for a cumulative impact effect on concussion injury threshold. *J Neurotrauma*. 2011;28(10):2079-2090. doi:10.1089/neu.2011.1910

252. Broglio SP, Martini D, Kasper L, Eckner JT, Kutcher JS. Estimation of head impact exposure in high school football: implications for regulating contact practices. *Am J Sports Med*. 2013;41(12):2877-2884. doi:10.1177/0363546513502458

253. Daniel RW, Rowson S, Duma SM. Head Impact Exposure in Youth Football: Middle School Ages 12–14 Years. *J Biomech Eng*. 2014;136(9):094501-094501-094506. doi:10.1115/1.4027872

254. Davenport EM, Whitlow CT, Urban JE, et al. Abnormal white matter integrity related to head impact exposure in a season of high school varsity football. *J Neurotrauma*. 2014;31(19):1617-1624. doi:10.1089/neu.2013.3233

255. 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*. 2014;24(5):416-421. doi:10.1097/JSM.0000000000000055

256. Bahrami N, Sharma D, Rosenthal S, et al. Subconcussive head impact exposure and white matter tract changes over a single season of youth football. *Radiology*. 2016;281(3):919–926.

257. Campolettano ET, Rowson S, Duma SM. Drill-specific head impact exposure in youth football practice. *J Neurosurg Pediatr*. 2016;18(5):536–541.

258. Davenport EM, Apkarian K, Whitlow CT, et al. Abnormalities in Diffusional Kurtosis Metrics Related to Head Impact Exposure in a Season of High School Varsity Football. *J Neurotrauma*. May 2016. doi:10.1089/neu.2015.4267

259. Campolettano ET, Gellner RA, Rowson S. High-magnitude head impact exposure in youth football. *J Neurosurg Pediatr*. October 2017:1-9. doi:10.3171/2017.5.PEDS17185

260. Johnston ER, Beer F, Eisenberg E. *Vector Mechanics for Engineers: Statics and Dynamics.* McGraw-Hill; 2009.

261. Fice JB, Siegmund GP, Blouin J-S. Neck muscle biomechanics and neural control. *J Neurophysiol*. 2018;120(1):361-371. doi:10.1152/jn.00512.2017

262. Fisher NI, Lewis T, Embleton BJ. *Statistical Analysis of Spherical Data*. Cambridge university press; 1993.

263. Ji S, Ghadyani H, Bolander RP, et al. Parametric comparisons of intracranial mechanical responses from three validated finite element models of the human head. *Ann Biomed Eng*. 2014;42(1):11-24. doi:10.1007/s10439-013-0907-2

264. Miller LE, Urban JE, Stitzel JD. Estimation Of 6 Degree of Freedom Accelerations From Head Impact Telemetry System Outputs For Computational Modeling. In: *15th International Symposium on Computer Methods in Biomechanics and Biomedical Engineering*. ; 2018.

265. Gilchrist A, Mills NJ. Protection of the side of the head. *Accid Anal Prev*. 1996;28(4):525-535.

266. Mills NJ, Gilchrist A. Oblique impact testing of bicycle helmets. *Int J Impact Eng*. 2008;35(9):1075–1086.

267. Zuckerman SL, Kerr ZY, Yengo-Kahn A, Wasserman E, Covassin T, Solomon GS. Epidemiology of Sports-Related Concussion in NCAA Athletes From 2009-2010 to 2013-2014 Incidence, Recurrence, and Mechanisms. *Am J Sports Med*. 2015:0363546515599634.

268. O’Connor KL, Rowson S, Duma SM, Broglio SP. Head-Impact-Measurement Devices: A Systematic Review. *J Athl Train*. 2017;52(3):206-227. doi:10.4085/1062-6050.52.2.05

269. Biomechanical Devices in TBI Working Group. Traumatic Brain Injury Common Data Elements: Biomechanical Devices. NINDS Common Data Elements. https://www.commondataelements.ninds.nih.gov/TBI.aspx#tab=Data_Standards. Published 2018. Accessed April 4, 2019.

270. Urban JE, Flood WC, Zimmerman BJ, et al. Evaluation of head impact exposure measured from youth football game plays. *J Neurosurg Pediatr*. 2019;1(aop):1-10. doi:10.3171/2019.2.PEDS18558

271. McHugh ML. Interrater reliability: the kappa statistic. *Biochem Medica*. 2012;22(3):276-282.

272. Funk JR, Duma SM, Manoogian SJ, Rowson S. Biomechanical Risk Estimates for Mild Traumatic Brain Injury. *Annu Proc Assoc Adv Automot Med*. 2007;51:343-361.