

AWARD NUMBER: W81XWH-17-1-0019

TITLE: Human Head Impact Dose Concussion Risk Functions and Sensor-Based Military-Specific Environmental Monitoring System

PRINCIPAL INVESTIGATOR: ADAM BARTSCH PHD PE

CONTRACTING ORGANIZATION: PREVENT BIOMETRICS

REPORT DATE: MARCH 2022

TYPE OF REPORT: FINAL

PREPARED FOR: U.S. Army Medical Research and Development Command
Fort Detrick, Maryland 21702-5012

DISTRIBUTION STATEMENT: Approved for Public Release.
Distribution Unlimited

The views, opinions and/or findings contained in this report are those of the author(s) and should not be construed as an official Department of the Army position, policy, or decision unless so designated by other documentation.

REPORT DOCUMENTATION PAGE				Form Approved OMB No. 0704-0188	
Public reporting burden for this collection of information is estimated to average 1 hour per response, including the time for reviewing instructions, searching existing data sources, gathering, and maintaining the data needed, and completing and reviewing this collection of information. Send comments regarding this burden estimate or any other aspect of this collection of information, including suggestions for reducing this burden to Department of Defense, Washington Headquarters Services, Directorate for Information Operations and Reports (0704-0188), 1215 Jefferson Davis Highway, Suite 1204, Arlington, VA 22202-4302. Respondents should be aware that notwithstanding any other provision of law, no person shall be subject to any penalty for failing to comply with a collection of information if it does not display a currently valid OMB control number. PLEASE DO NOT RETURN YOUR FORM TO THE ABOVE ADDRESS.					
1. REPORT DATE MARCH 2022		2. REPORT TYPE FINAL		3. DATES COVERED 15JAN2017 - 31OCT2021	
4. TITLE AND SUBTITLE Human Head Impact Dose Concussion Risk Functions and Sensor-Based Military-Specific Environmental Monitoring System				5a. CONTRACT NUMBER W81XWH-17-1-0019	
				5b. GRANT NUMBER	
				5c. PROGRAM ELEMENT NUMBER	
6. AUTHOR(S) ADAM BARTSCH PHD PE E-Mail: ABARTSCH@PREVENTBIOMETRICS.COM				5d. PROJECT NUMBER	
				5e. TASK NUMBER	
				5f. WORK UNIT NUMBER	
7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES) PREVENT BIOMETRICS 4600 WEST 77 TH STREET SUITE 275 MINNEAPOLIS, MN 55435				8. PERFORMING ORGANIZATION REPORT NUMBER 04	
9. SPONSORING / MONITORING AGENCY NAME(S) AND ADDRESS(ES) U.S. Army Medical Research and Development Command Fort Detrick, Maryland 21702-5012				10. SPONSOR/MONITOR'S ACRONYM(S)	
				11. SPONSOR/MONITOR'S REPORT NUMBER(S)	
12. DISTRIBUTION / AVAILABILITY STATEMENT Approved for Public Release, Distribution Unlimited					
13. SUPPLEMENTARY NOTES					
14. ABSTRACT For the Final Report, the project has completed all the specific aims in the statement of work #3, revision 2, dated 01MAY20. Aim 1 was designed to quantify head impacts across a variety of military and civilian activities. A total of 2,405 Impact Monitoring Mouthguards (IMM) were deployed during the period of performance across static and freefall parachute training, boxing, mixed martial arts and hand-to-hand combatives training, as well as a variety of helmeted and non-helmeted sporting activities. A total of n=37 impacts with concussion and/or concussion signs were observed out of 46,949 head acceleration events (HAE). The HAE range was 1g-99g peak linear acceleration and 0.1J-120J workload. Aim 2 was to quantify a dose-response relationship between HAE and clinical assessments of non-concussed individuals. SCAT3 symptom scores rose from pre-season to mid-season assessments, yet by season's end the symptoms had resolved. There was no obvious correlation over a 3-month period. Aim 3 was to correlate HAE with concussion onset. All concussions were related to a single, high-energy head impact in the top 1% by PLA, workload, or both. 77% of all HAE collected were far below the concussion ranges, on par with a pillow fight or jumping on a trampoline <10g PLA. Aim 4 laboratory calibration was finished at USAARL and other independent laboratories at Virginia Tech and Stanford University. Laboratory comparisons were made to both Hybrid III and NOCSAE headforms. The IMM fit a linear model versus Reference sensors at the headform center of gravity with a concordance correlation coefficient of 0.98 for linear and angular acceleration peaks. The MLSH headform was found to cause data noise.					
15. SUBJECT TERMS NONE LISTED					
16. SECURITY CLASSIFICATION OF:			17. LIMITATION OF ABSTRACT	18. NUMBER OF PAGES	19a. NAME OF RESPONSIBLE PERSON
a. REPORT	b. ABSTRACT	c. THIS PAGE			USAMRDC
Unclassified	Unclassified	Unclassified	Unclassified	65	19b. TELEPHONE NUMBER (include area code)

Table of Contents

	<u>Page</u>
1. Introduction.....	4
2. Keywords.....	6
3. Accomplishments.....	7
4. Impact.....	44
5. Changes/Problems.....	46
6. Products, Inventions, Patent Applications, and/or Licenses.....	47
7. Participants & Other Collaborating Organizations.....	56
8. Special Reporting Requirements.....	58
9. References.....	59
10.Appendices.....	65

1. Introduction

Concussion and mild traumatic brain injury (mTBI) are signature head injuries of the 21st Century, affecting service members, athletes and the public. Approximately 80% of military mTBI, affecting hundreds of thousands of service members, occur in non-deployed settings like combative training activities (Langlois et al. 2006). In athletics, nearly 53 million (SGMA 2010) athletes under participate yearly in mouthguard-wearing contact sports like football, ice hockey and soccer in the United States. Between 173,000 and 3.8 million of these athletes sustain concussions and mTBI (Centers for Disease Control 2011). Finally, concussion and mTBI caused by loss of consciousness is serious concern to the Departments of Defense and Transportation because it is a major impediment to egress in aviation crashes.

Military, athletic and transportation authorities have long desired a system that can monitor concussion and mTBI risk in real-time, as well as quantify head impacts that cause acute and long-term brain health decline. Despite a recent proliferation in head impact instrumentation commercialization and research, there still is no objective means to identify concussion risk in real-time from any single impact. This is a serious public health crisis as concussed persons suffer headaches, memory problems, balance dysfunction, sensitivity to light, slowed reaction time, insomnia, and significant time lost from education (Arbogast et al. 2013). Every year, persons die from second impact syndrome (Cantu et al. 1998). Cumulative sub-concussive impacts in soldiers and athletes may contribute to depression (Guskiewicz et al. 2007), suicide (Omalu et al. 2010), cognitive decline (Guskiewicz et al. 2005), chronic traumatic encephalopathy (CTE) (McKee et al. 2009), dementia (Jordan 2000), Alzheimer's (Plassman et al. 2000) and Parkinson's (Daneshvar et al. 2011) Diseases.

The lack of practical, accurate and precise head impact monitoring systems means service members and athletes sustain brain injury during activity and caregivers often cannot identify, assess or diagnose the injured; this subjective approach fails to identify at least 50% of those concussed (McCrea et al. 2004). Further, lack of real-world data means defense and transportation safety engineers continue to devise safety standards based exclusively on skull fracture risk quantified by Head Injury Criterion (HIC), failing to account for the myriad situations when sub-catastrophic head injury is endured by occupants. Finally, in fixed wing and rotorcraft crashes, concussions due to non-skull fracture loss of consciousness events cause significantly higher risk of perishing post-crash because egress time is increased.

The reasons why head impact monitoring in military and civilians remains insufficient are the following: (1) current instruments, like other mouthguards (up to 84% error), helmets (up to 580% error), and skin patches (up to 300% error) lack accurate information to quantify single impact doses (Siegmund et al. 2015, Jadischke et al. 2013, Camarillo et al. 2015), (2) estimates of single and cumulative head impact doses in American football have been reported (Rowson et al. 2009) but are not reliable due to helmet-mounted sensor measurement errors of up to 580% in any single head impact based on NFL tests, (3) reliable single head impact measurements have yet to be made in soldiers and (4) no real-world data or measurement system has previously existed to quantify loss of consciousness in aviation crashes. There is a need to develop trustworthy means to quantify acute and long-term head impact risks to devise injury assessment reference values (IARV) based on real-world data (Spiotta et al. 2011).

Hypothesis

Concussion, mTBI and loss of consciousness are functions of measurable spatial (X, Y, Z) and temporal (time duration, frequency) parameters.

Lack of accurate and precise head impact monitoring is the fundamental gap in understanding acute and long-term concussive and sub-concussive head impact effects. We are bridging this gap using an impact monitoring mouthguard system (IMM, Prevent Biometrics, Minneapolis, MN, USA), proven to be accurate and precise in single impacts (Bartsch et al. 2014). Millions of service members and civilians carry high risk for single (concussion, mTBI, loss of consciousness) and cumulative (sub-concussive) head injury, yet head impact kinematics in single events have never been quantified accurately and precisely. Without our work, caregivers will continue to struggle identifying at-risk persons requiring concussion assessment, safe limits for head impact exposure will remain un-quantified and next-generation injury risk values will remain unknown.

Solution

The most critical gap in the understanding of concussion, mTBI and loss of consciousness risk has been an inability to quantify the injury-dose response. To fill this gap this project leveraged the IMM, an accurate and precise head impact measurement instrument, achieving Level III validity per National Football League (NFL) testing protocols by producing head impact dose data within 5% of truth (Siegmund et al. 2015).

In this BAA cooperative agreement, the Study Team quantified head impact severity, video observations of concussive signs, and clinical behavioral assessments in a large-scale observational study of over 1,000 service members and athletes. Concussion injury risk thresholds and a TRL9 IMM system tailored to monitor real-time head impact injury risk in military and civilian populations were the results of this work.

FINAL REPORT PREFACE

In 2018, the Department of Defense issued the Comprehensive Strategy and Action Plan for Warfighter Brain Health (WBH, 2018). The WBH plan underscored that traumatic brain injury (TBI) is a result of blunt and/or blast forces, the effects are myriad and potentially devastating, and that “The pursuit of superior lethality must be matched by commitment to understanding and preventing TBI.”

This TBI understanding is critical for both military service members and civilians alike yet has been hamstrung by the difficult task of identifying persons in need of concussion assessment in the military training environment and athletics playing field. In military parlance, there is a desire for a ‘check engine’ warning so that personnel can be assessed by on-site medical staff as soon as is practical. The demand for a reliable ‘check engine’ warning is underscored by concussion incidence data: 82% of all DoD TBI’s are concussions (TBICoE Research 2021), civilians sustain at least 300,000 documented concussions annually (Gessel et al. 2007) and 50% to 80% of concussions are never assessed, diagnosed, or treated (Baugh et al. 2015, McCrea et al. 2004). Identifying a person for a concussion assessment is critically important, as a concussed individual who is immediately identified, assessed, and given treatment will return to baseline neurocognitive performance in half the time versus one who is removed only 15 minutes later (Elbin et al. 2016). And failure to assess, diagnose and treat concussion every year leads to fatalities from ‘Second Impact Syndrome’ (Cantu 1998).

There have been efforts over the past two decades to identify persons more precisely in need of concussion assessment. A seminal paper was published in the early 2010’s by Guskiewicz and Mihalik, about ten years after introduction of data collection using instrumented American football helmets. Their paper summarized the clinical usefulness of the technology in a dose-response relationship, declaring the *helmet sensor data* are “*clinically irrelevant*” (Guskiewicz & Mihalik 2011). Clinicians still contend head impact monitoring data is of little benefit in the clinician’s ability to assess, diagnose and treat concussion (McCrory et al. 2017, Kontos & Collins 2018).

While concussion assessment, diagnosis and treatment are clinical challenges, the laboratory-calibrated head impact monitoring mouthguard (IMM) data collected in this BAA project (W81XWH-17-1-0019) provide a potential paradigm shift in the historically confusing world of head impact dosing, and its true relationship to clinical response.

This Final Report includes civilian and military data collected under the BAA project that has been accurately monitoring blunt forces in support of the military’s need for a ‘check engine’ dosimeter. The results presented here, from 46,949 head acceleration events (HAE) should establish head impact norms and provide the necessary scientific ammunition for the DoD as to whether a ‘check engine’ warning is supported by hard data.

2. Keywords

Head Impact Monitoring, Environmental Sensors in Training (ESiT), Combatives, Paratroopers, Concussion, Impact Dose, Check Engine Impact Sensor

3. Accomplishments

This BAA cooperative agreement involved multiple DoD and civilian investigators:

Site 1 (Prime):	Prevent Biometrics
	4530 West 77th St
	Minneapolis, MN 55435
	PI: Adam Bartsch, PhD PE
Site 2:	Medical College of Wisconsin
	5000 West National Ave
	Milwaukee, WI 53295
	PI: Brian Stemper, PhD
	Co-I: Mike McCrea, PhD
	Co-I: Lindsay Nelson, PhD
Site 3:	US Army Aeromedical Research Laboratory
	PO Box 620577
	Fort Rucker, AL 36362
	PI: Valeta Carol Chancey, PhD
	Co-I: Joe McEntire, MS
	Co-I: Tyler Rooks, MS
Site 4:	US Air Force Academy
	4102 Pinion Dr
	Colorado Springs, CO 80840
	PI: Lt. Col. Gerald McGinty, DPT
	Co-I: Lt. Col. John Jackson MD
Site 5:	US Military Academy
	Army West Point Athletics
	639 Howard Rd.
	West Point, NY 10996
	Co-PI: Col. Steven Svoboda, MD
	Co-PI: Kenneth Cameron, PhD
	Co-I: Megan Houston, PhD

Data were collected at several civilian sites, as well as DoD locations with active-duty service members. The data collection sites were:

1. United States Army Aeromedical Research Laboratory (USAARL), Fort Rucker, AL
2. United States Air Force Academy (USAFA), Colorado Springs, CO
3. United States Military Academy (USMA), West Point, NY
4. 82nd Airborne Fort Bragg, NC
5. Advanced Airborne School, Fort Bragg, NC
6. US Army Special Operations Test Directorate, Fort Bragg, NC
7. US Air Force Combatives Center of Excellence, Air Force Academy, CO
8. 22nd Training Squadron, Fairchild AFB, WA
9. 421st Combat Training Squadron, Joint-Base McGuire Dix Lakehurst, NJ
10. Georgia Army Air National Guard (POMPOC), Savannah, GA
11. 371st Special Operations Combat Training Squadron, Hurlburt Field, FL
12. 351st Special Warfare Training Squadron, Kirtland AFB, NM
13. Multiple Division III Colleges in Wisconsin selected by Medical College of Wisconsin, Milwaukee, WI

MONITORING PLATFORM



Data in the study were collected using the Impact Monitoring Mouthguard System (IMM, Prevent Biometrics, Minneapolis, MN, USA).



Multiple DoD sites across a variety of training environments ranging from parachute, combatives, athletics and special warfare were involved in data collection. There were also several civilian athletic sites where data were collected.

The BAA Cooperative Agreement Specific Aims were conceived to solve some of the most pressing questions around head impact monitoring, concussion signs, and clinical outcomes in the DoD training environment. The goal of the collective aims was to define the relationship(s) between head impact kinematics and concussion outcomes for service member and athletic populations. The Study Team aimed to be able to define performance specifications for head impact monitors (e.g., ‘Environmental Sensors’) that can monitor US Military service members at risk of short- and long-term head injury in the training environment. The aims were designed to quantify kinematic inputs responsible for highest head injury risk and be able to identify environments most in need of monitoring to enhance operational readiness for line leaders. Finally, the project aimed to rapidly advance development of a user-friendly IMM system ready to monitor service members in the training environment. The four (4) Specific Aims were:

SPECIFIC AIM #1 – Quantify head impacts across a variety of contact sports and activities using an accurate and precise Impact Monitoring Mouthguard (IMM) system.

This aim, led by Dr. Bartsch (Prevent Biometrics), Mr. Rooks (USAARL) and Dr. Stemper (MCW), focused on measuring and characterizing head impact exposures in active-duty military personnel, cadets at military service academies, and civilian athletes at NCAA Division III Universities. Head impact exposures were quantified using the Prevent Biometrics (Minneapolis, MN, USA) Impact Monitoring Mouthguard during military training activities and contact sports.

SPECIFIC AIM #2 – Quantify a dose relationship between head impacts and changes in clinical assessments.

This aim, led by Dr. Stemper (MCW), attempted to correlate cumulative head impact workload in service members, cadets, and civilian athletes with changes in clinical assessments between baseline and repeated assessments obtained during and after the season or training period. Dose response analyses were completed to characterize these relationships in terms of the magnitude, number, frequency, and type of head impact workload.

SPECIFIC AIM #3 – Correlate head impact exposure measured with the IMM system with onset and severity of concussion.

This aim, led by Dr. Bartsch (Prevent Biometrics), Mr. Rooks (USAARL) and Dr. Stemper (MCW), correlated head impacts measured by the IMM with direct, video or clinically reported observations of concussion signs in service members, cadets, and civilian athletes. Injury risk functions were developed to associate increasing risk for concussion onset with the magnitude, number, frequency, and type of head impact(s).

SPECIFIC AIM #4 – Quantify laboratory performance of the IMM system under military conditions.

This aim, led by Mr. Rooks (USAARL), independently quantified the accuracy and precision of the IMM under military conditions. Controlled tests were conducted that compared IMM to Reference crash test. Military helmet(s). This aim produced military-specific performance for the IMM, which already had rigorous performance validated in over 1000 laboratory tests in civilian athletics’ environments.

SPECIFIC AIM #1 – Quantify head impacts across a variety of contact sports and activities using an accurate and precise Impact Monitoring Mouthguard (IMM) system.

This aim, led by Dr. Bartsch (Prevent Biometrics), Mr. Rooks (USAARL) and Dr. Stemper (MCW), focused on measuring and characterizing head impact exposures in active-duty military personnel, cadets at military service academies, and civilian athletes at NCAA Division III Universities. Head impact exposures were quantified using the Prevent Biometrics (Minneapolis, MN, USA) Impact Monitoring Mouthguard during military training activities and contact sports.

Informed consent was obtained from all subjects at military and civilian sites after study approval by institutional review bodies. A total of 46,949 head acceleration events (HAE) are reported here. Activities included military training (parachute, combatives, boxing, athletics) and civilian sport (helmeted and non-helmeted activities).

All HAE were measured using an impact monitoring mouthguard. The IMM system includes accelerometers and gyroscopes with sufficient range, bandwidth, and sample rate to collect bare head, helmeted and padded head impacts. When worn on the teeth as a mouthguard, the on-board accelerometer triggers data collection when the head moves, raw kinematics data at the mouth are transformed to the head center of gravity (CG), and noise is removed from the signal prior to reporting HAE kinematics and direction/location. The IMM system is the most extensively calibrated impact monitoring sensor, with thousands of laboratory and playing field collisions published amongst Independent laboratories at USAARL, Virginia Tech and Stanford University, and found to report up to 99% correlation versus reference crash dummy headform CG and human video reviewers (Bartsch et al. 2014; Bartsch et al. 2020; Hedin et al. 2016; Kieffer et al. 2020; Liu et al. 2020, Jones et al. 2022).

To screen out non-HAE data sets, video and/or mouthguard hardware were used to confirm a collision occurred and the mouthguard was on the teeth during the collision. The term head impact defines a measured HAE with 10g+ of peak linear acceleration (PLA) at the head CG. The HAE with PLA less than 10g, are bumps, taps and inertial head motions in the range of activities of daily living (ADL) like trampolining, having a pillow fight, doing physical education, or sledding. In addition to reporting PLA, this project also reports the estimated kinetic energy transfer, or Workload, in Joules, to the skull during the impact. Workload is calculated from a textbook equation (Bartsch et al. 2020), considers the linear and rotational movement of the head, as well as head mass and inertial properties during an impact. Single impact workload has been shown to correlate to impact severity observed on video, presence of clinical concussion signs and diagnosed concussion (Newman et al. 2000; Pellman et al. 2003; Clark et al. 2020).

All collisions were reviewed using modern video-verification methods to identify impacts with visible concussion signs (Davis et al. 2019). This approach to finding single impacts that cause visible concussion signs has been endorsed by on-field concussion spotters in the Australian Football League (AFL), Cricket Australia, Major League Baseball (MLB), National Football League (NFL), National Hockey League (NHL), National Rugby League (NRL), and World Rugby (WR).

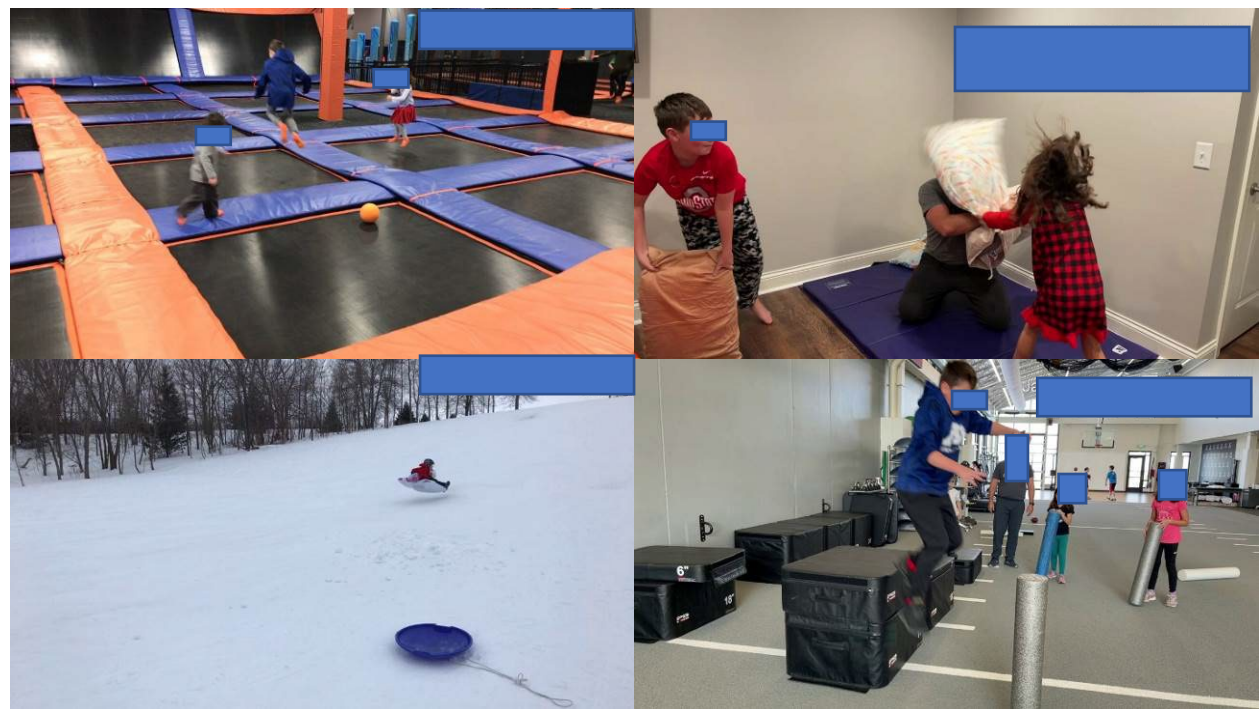
In order to align with DoD's need for a simple means to identify at-risk service members, impacts with concussions and/or visible concussion signs were referred to as 'check engine' impacts. Any 'check engine' impact had at least one visible concussion sign, was flagged, and more in-depth observations were recorded from video and/or on-site staff.

TableA1-1 summarizes the population and data. There was 36,185 (77%) of all HAE were collected below 10g PLA, in the range of ADL exposures like a pillow-fight or trampoline.

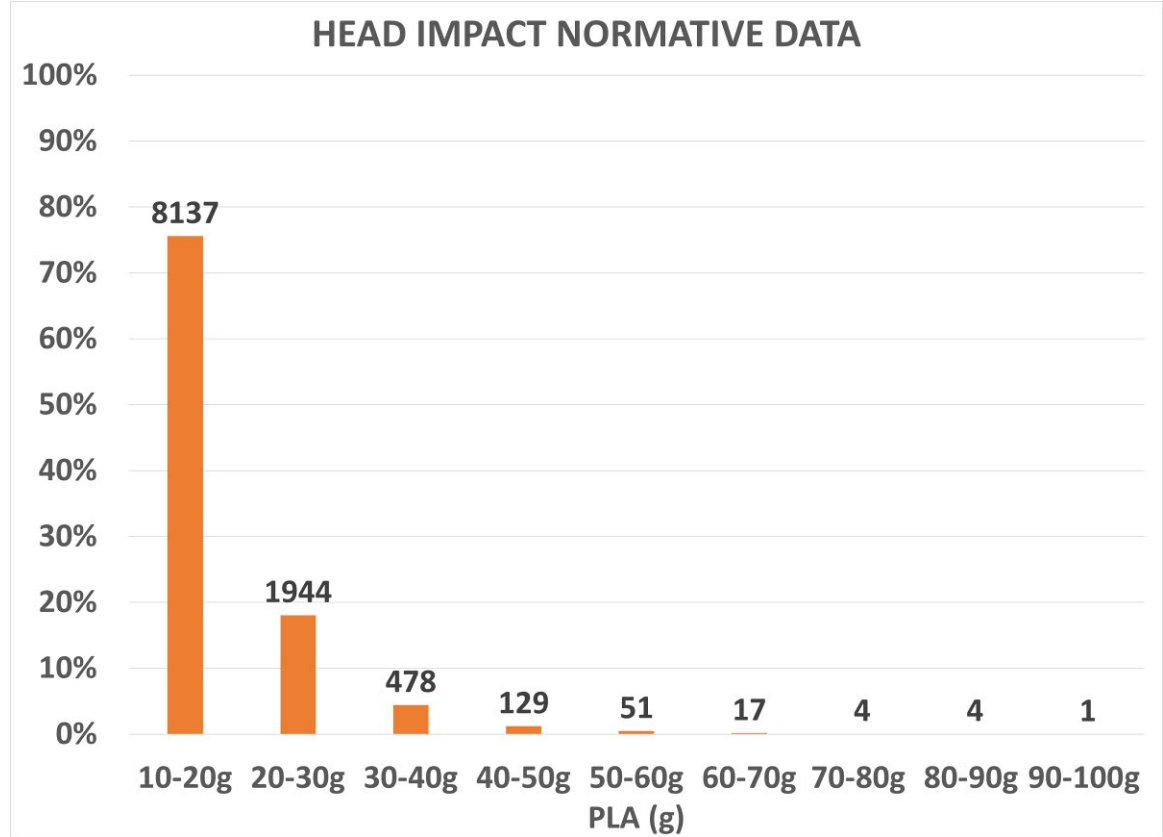
	TOTAL
IMM	2405
Subject-Days	2734
HAE	46949
ADL Magnitude, <10g	36185 (77%)
Head Impacts, >10g	10764 (23%)
'Check Engine' Impacts	37 (0.08%)

TABLEA1-1. Summary of Data. HAE=Head Acceleration Event, ADL=Activities of Daily Living. Head Impacts are HAE events >10g PLA, measured at the head center of gravity (CG). ADL events are minor impacts, bumps, and inertial motion of the head <10g PLA at the CG. **Subject-Days for civilian athletes are also known as Athlete-Exposures (AE).*

FigureA1-1 shows more detail and context around the low-g ADL events.



FIGUREA1-1. Head Acceleration Events (HAE) occur to humans every day. In activities like playing on a trampoline, having a pillow fight, sledding down a hill or doing physical education, the head will experience HAE in the range of 1-10g PLA and 0.1-2J Workload. These types of HAE are known more commonly as Activities of Daily Living (ADL). In this project, 77% of all data collected were HAE in the ADL range.

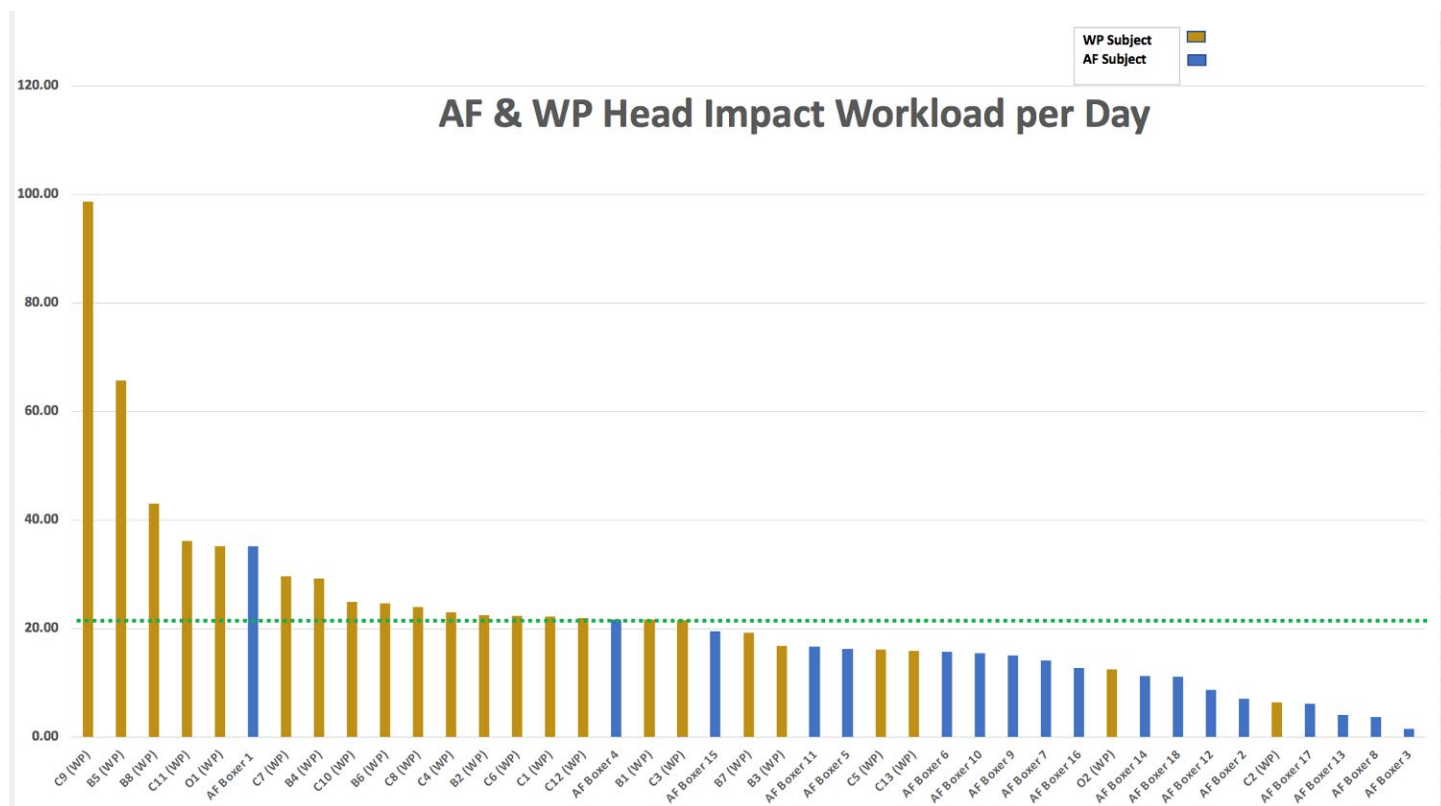


FIGUREA1-2. Summary of Head Impact Distributions for PLA >10g. Less than 1% of all head impacts were between 50g and 100g.

There were 10,764 (23%) head impacts with PLA>10g reported at the head CG. In **FigureA1-2** the left skewed head impact norms are shown for the subjects. Out of the 10,764 impacts measured, the median impact was 14g, with n=77 impacts between 50g to 100g PLA (0.2%).

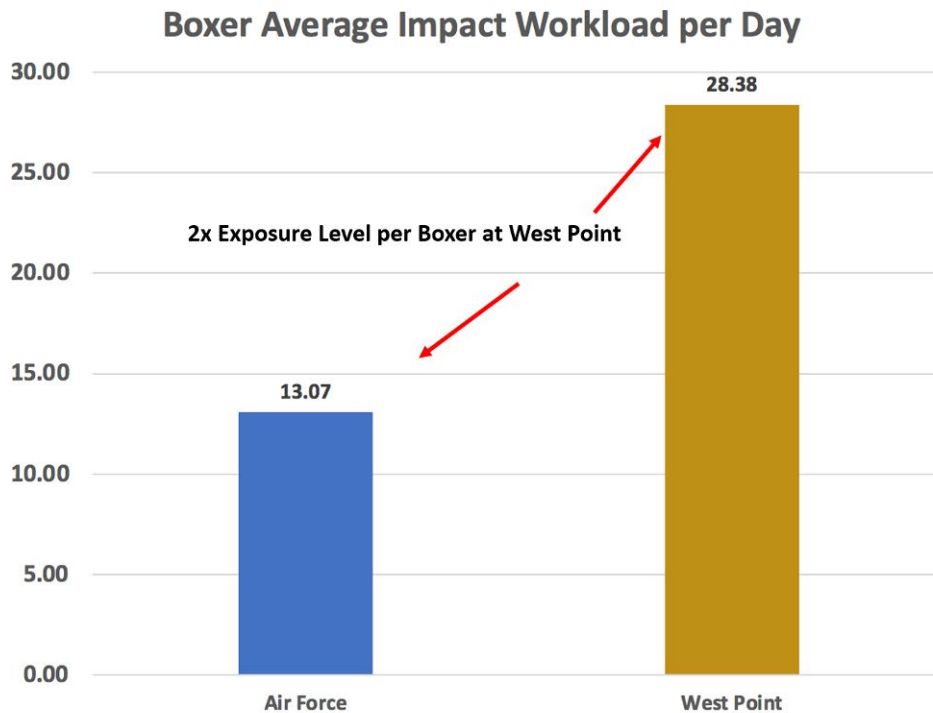
COMPARING HAE ACROSS VARIOUS SITES

Each site had unique experiences with HAE compared to other sites doing the same training. Quite a spread in head impact workload was observed, which hints at different training methods to accomplish the same training goal. It is likely in the future that data like this could help DoD identify proper training methods to avoid unnecessary HAE workload. The first comparison is between the US Military Academy and the US Air Force Academy. Both sites conducted physical education boxing training, but USMA had significantly higher workload per day and per class for students. See **FigureA1-3**.



FigureA1-3. Comparison of PE Boxing workload at US Military Academy and US Air Force Academy for individual cadets in a single day. The differences in head impact workload point to different training techniques being used to accomplish the same objective. They also point to mismatches among skill levels of cadets, even if the cadets are matched by size/strength.

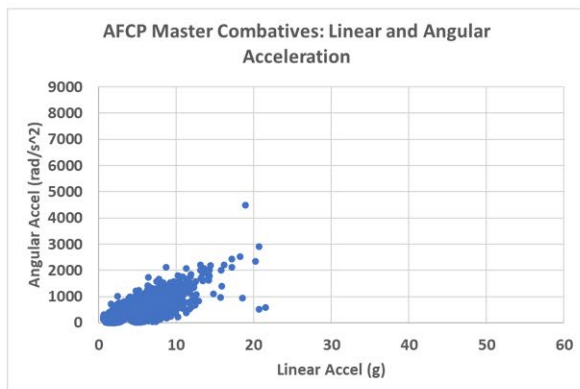
The differences were compared for the entire group of enrolled cadets at USMA and USAFA. The differences, shown in **FigureA1-4**, are quite striking. The USMA cadets, on average, sustained 200% more head impact workload than USAFA over the same period of training. This is probably due to skill mismatch, but also could be part of a rite of passage at USMA. In discussing these results with Mr. David Durnil, lead instructor at the US Air Force Combative Centers of Excellence, it was unclear what caused USMA to have such an increase in workload. But the USAFA data fell in line with the objectives of the courses and within the perception of PE boxing and combatives instructors.



FIGUREA1-4. The average head impact workload/exposure at USAFA and USMA were compared, with USMA having an average of about 200% more workload per cadet. In the future it is recommended to DoD to harmonize the training regimen so that the mission can be accomplished without subjecting cadets to unnecessary head impact workload.

To benchmark the differences at USMA and USAFA, analyses for the Air Force Master Combatives Program Master instructor Course were done. The AFCP Master Course involved some of the most skilled, and valuable, Combatives personnel in USAF. As shown in **FigureA1-5** the AFCP Master Course resembled the PE boxing experience at USAFA, as opposed to the high workload training at USMA.

AFCP Master Instructor Course



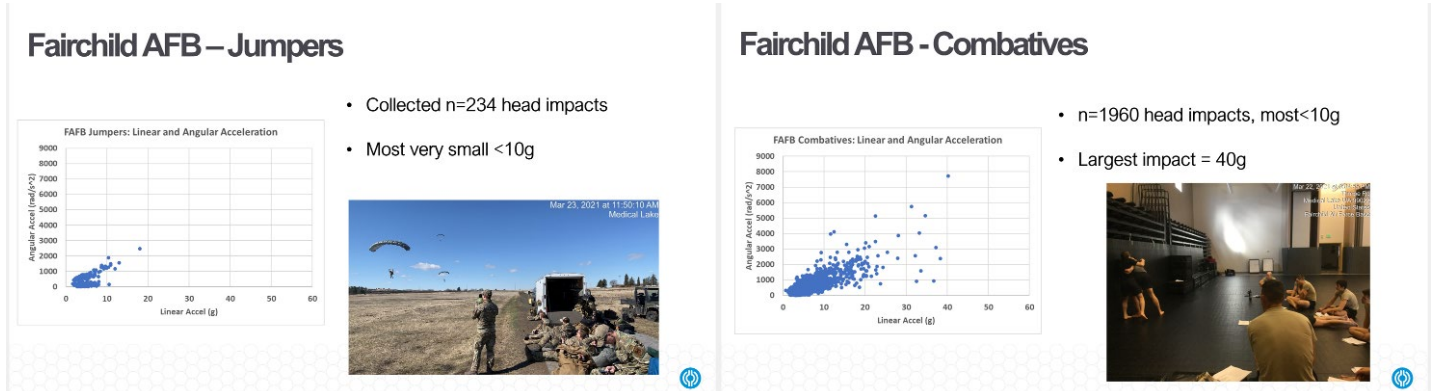
- Collected n=3231 head impacts
- Most very small <10g



FigureA1-5. The USAF Master Instructor Course head impacts were on par with trampoline (5g-10g) and the higher end of Activities of Daily Living (10g-15g). This was the intent of the course; students were supposed to

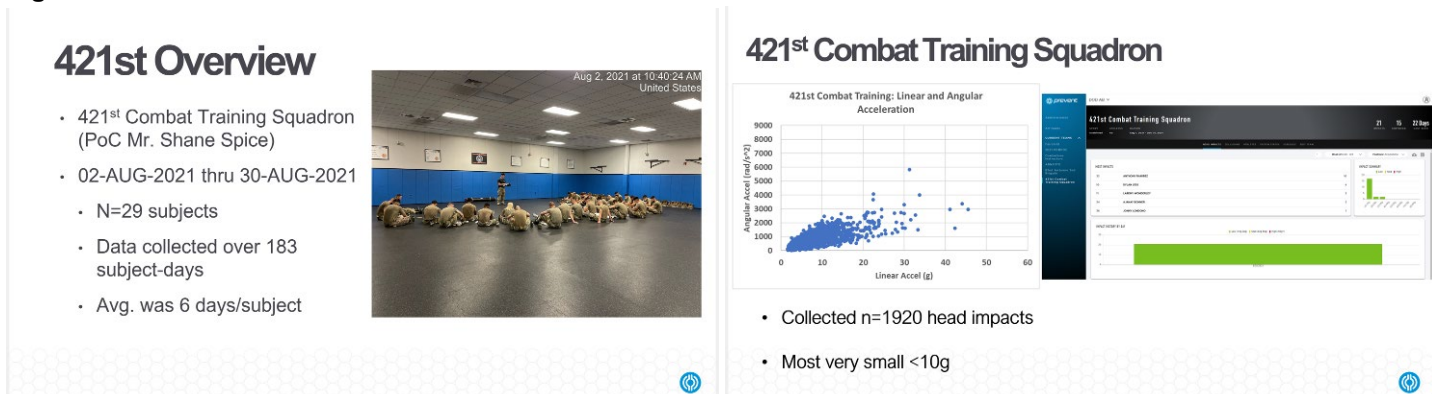
have limited head impact workload while also learning the techniques necessary to complete the training mission.

Further analysis was done in other training environments. In **FigureA1-6**, the parachute training and combatives training courses at Fairchild AFB (USAF 22nd Training Squadron) were compared for the individual head impacts.



FigureA1-6. Comparison of parachute and combatives training head impacts at Fairchild AFB (22nd Training Squadron). Notice how the parachute activities, which used the HI-5 ram air glide chute and Hollywood jump, has head impacts on par with trampoline jumping (~5-10g) while the combatives training has head impacts that approach 40g (a sledgehammer to the head is about 60g). Neither of these groups had head impact workloads of concern, but this illustrates how different training missions illicit different head impact exposure workloads, and peak head impact forces.

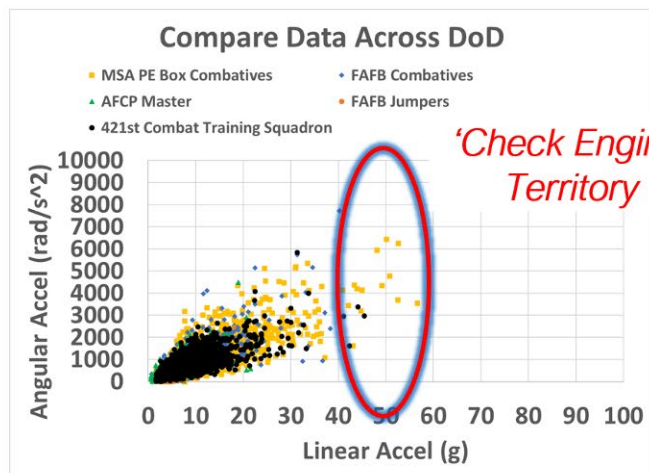
The 421st Combat Training Squadron (Joint Base McGuire-Dix Lakehurst) also conducted combatives training. Their courses operated at a higher tempo than the MSA PE boxing, USAF Master Course or Fairchild Activities. See **FigureA1-7**.



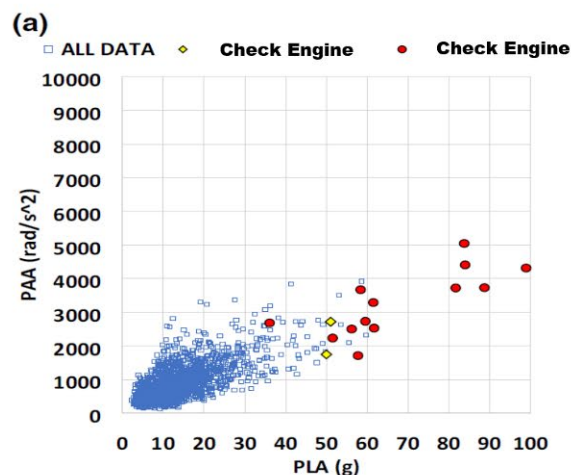
FigureA1-7. The 421st CTS at Joint Base MDL had a considerably higher tempo of combatives training, and the data showed this. The peak impact workloads were higher than those of the USAF Master Course, or the MSA PE training. The higher tempo is necessary for accomplishing the 421st mission. The data here were used by their medic to monitor for ‘check engine’ impacts, in the range of 40g-60g+, so they could immediately identify any service member who may need a concussion assessment.

When comparing training activities across the combatives and boxing sites, it was of interest to see how the USAF Master Course – used as a benchmark – contrasted with the more novice training at the Military Service Academies. **FigureA1-8** shows this comparison.

Compare to Other Data



DOD DATA



SPORTS DATA

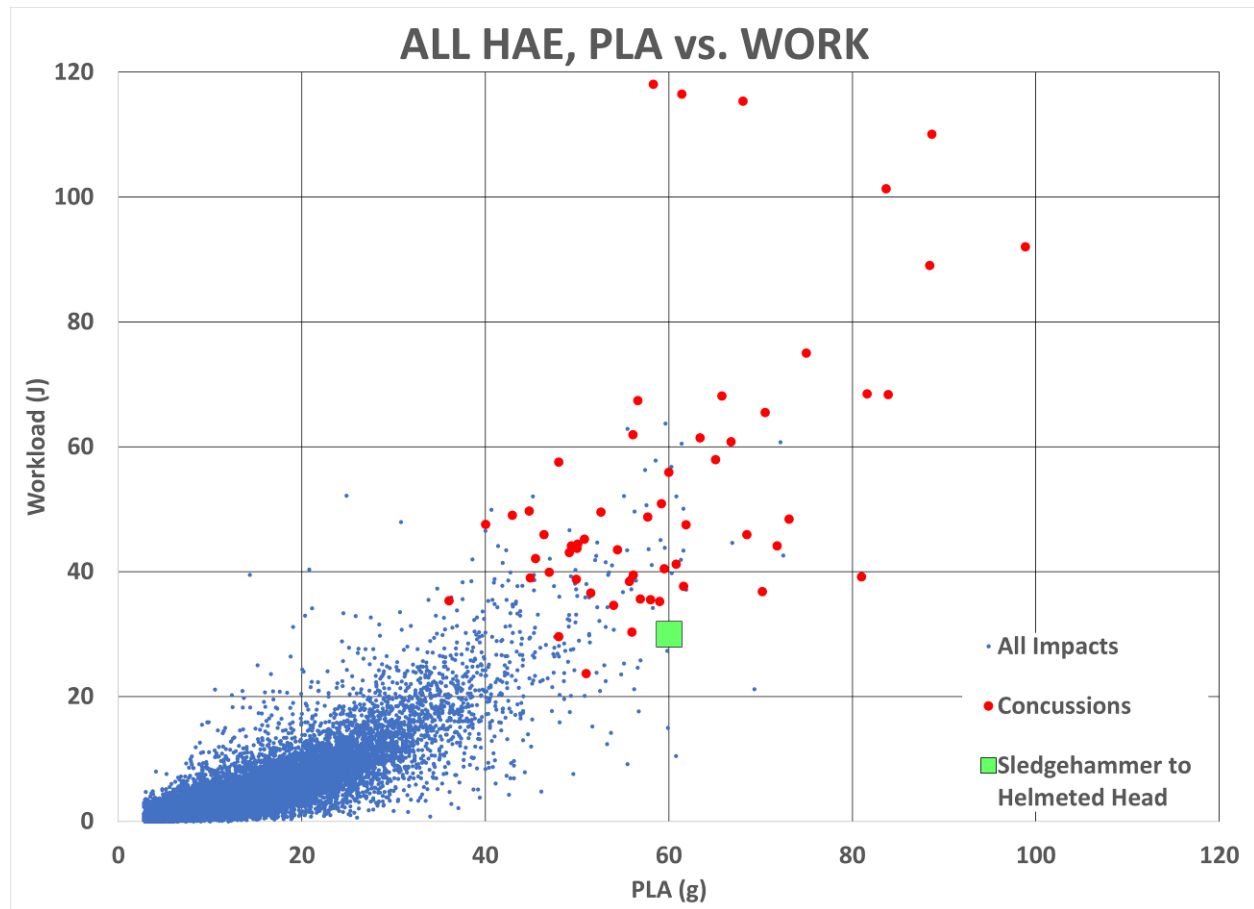
FigureA1-8. When comparing the DOD Sites (left) with published civilian sports data (right) one can see how the DoD data had a similar distribution and approached some of the 'check engine' concussion limits. Within the DoD one can also see how the various activities had different head impact workloads and peak accelerations. For example, USAF Master Instructors reduced their head impact workloads compared to the relatively novice MSA cadets. These data are extremely useful for Line Leaders who are wishing to reduce impact workloads to a low level in order to maintain mission readiness and maximum service member health.



FigureA1-9. Summary of the 82nd Airborne parachute training experience. The head impact severity in general was low, with the majority of service members executing proper PLFs during training. However, there were several instances of head impacts during T-11 training that led to 'check engine' concussion assessments, concussion and loss of consciousness. These data underscore that even experienced operators can have head impacts during training, and that identification of these impacts is important to trigger medics to conduct concussion assessment. Faster identification leads to faster assessment leads to faster return to duty with full health.

The final group to discuss is the 82nd Airborne (Fort Bragg, NC). Data collection involved service members from the US Army Advanced Airborne School, the US Army Airborne Special Operations Test Directorate and the 2nd Brigade Combat Team. Over the course of about 6 months, hundreds of jumps were conducted, and data were collected on head impacts during normal T-11 parachute opening and normal parachute landing falls (PLF). Data were also collected on more severe head impacts, including several that caused 'check engine' responses, concussion and loss of consciousness. See FigureA1-9.

All of the collected data are summarized in **FigureA1-10**. The 46,949 HAE data sets are shown by the peak PLA (g) versus Workload (J). There is also n=37 'check engine' impacts shown as the large circles. The median 'check engine' impact was 58g and 46J, and all 37 'check engine' events were in the top 1% by magnitude of PLA, Workload, or both.



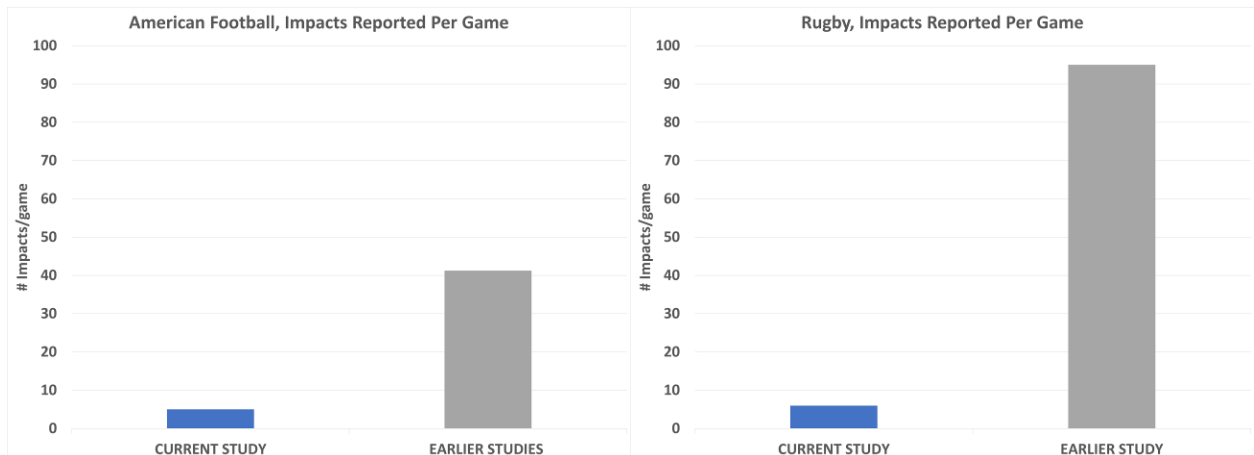
FIGUREA1-10. Summary of all 46,949 HAE data sets (small, filled circles) from the BAA project. The n=37 'check engine' impacts are shown in large, filled circles. As a frame of reference, a sledgehammer hit to a helmeted crash test dummy from the laboratory is also shown (solid square).

There were nine (9) 'check engine' impacts that resulted in concussion, and twenty-eight (28) 'check engine' impacts where the concussion assessment and diagnosis information was unavailable. See **Table 2** for details on 'check engine' impacts and diagnosed concussions.

IMPACT TYPE	MILITARY/CIVILIAN	Peak Linear Acceleration (PLA, g)	Workload (J)	Location	CONCUSSION?	CLINICAL INDICATORS
CHECK ENGINE	CIVILIAN	99	92	REAR	YES	LoC, Motionless
CHECK ENGINE	CIVILIAN	89	110	SIDE	UNKNOWN	Slow return to play, poor performance
CHECK ENGINE	CIVILIAN	84	68	SIDE	YES	Confused return to play
CHECK ENGINE	CIVILIAN	84	101	SIDE	UNKNOWN	Slow-to-rise, Team mate concern
CHECK ENGINE	CIVILIAN	82	68	SIDE	UNKNOWN	Slow-to-rise, Unsteady on feet, Avoids game participation
CHECK ENGINE	MILITARY	75	75	REAR	YES	LoC, Motionless
CHECK ENGINE	MILITARY	73	48	REAR	UNKNOWN	Avoids game participation
CHECK ENGINE	MILITARY	72	44	REAR	UNKNOWN	Slow-to-rise
CHECK ENGINE	MILITARY	71	66	REAR	UNKNOWN	Slow-to-rise
CHECK ENGINE	MILITARY	65	58	REAR	UNKNOWN	Slow-to-rise
CHECK ENGINE	MILITARY	62	48	FRONT	UNKNOWN	Slow-to-rise, Poor performance
CHECK ENGINE	CIVILIAN	62	38	FRONT	UNKNOWN	Slow-to-rise, Team mate concern
CHECK ENGINE	CIVILIAN	61	116	REAR	YES	Holds head
CHECK ENGINE	MILITARY	61	41	FRONT	YES	Slow-to-rise, Confused return to play, Player concern
CHECK ENGINE	CIVILIAN	60	56	SIDE	UNKNOWN	Remains on ground, Slow-to-rise, Unsteady on rising
CHECK ENGINE	CIVILIAN	59	40	SIDE	UNKNOWN	Unsteady on rising, Unsteady on feet, Confused return to play
CHECK ENGINE	MILITARY	59	51	SIDE	UNKNOWN	Slow-to-rise
CHECK ENGINE	CIVILIAN	58	118	SIDE	YES	LoC, Motionless
CHECK ENGINE	MILITARY	58	36	REAR	UNKNOWN	Slow-to-rise
CHECK ENGINE	CIVILIAN	58	49	SIDE	UNKNOWN	Holds head
CHECK ENGINE	CIVILIAN	56	39	REAR	YES	Remains on ground
CHECK ENGINE	MILITARY	56	30	SIDE	UNKNOWN	Slow-to-rise, Poor performance
CHECK ENGINE	MILITARY	56	38	REAR	UNKNOWN	Slow-to-rise
CHECK ENGINE	CIVILIAN	54	35	FRONT	YES	Slow-to-rise, Unsteady on feet, Holds head
CHECK ENGINE	MILITARY	53	50	SIDE	UNKNOWN	Avoids game participation
CHECK ENGINE	CIVILIAN	51	37	SIDE	UNKNOWN	Avoids game participation
CHECK ENGINE	CIVILIAN	51	24	SIDE	UNKNOWN	Motionless 3-5 secs
CHECK ENGINE	MILITARY	51	45	SIDE	UNKNOWN	Slow-to-rise
CHECK ENGINE	MILITARY	50	44	SIDE	UNKNOWN	Confused return to play
CHECK ENGINE	MILITARY	50	44	REAR	UNKNOWN	Slow-to-rise
CHECK ENGINE	CIVILIAN	50	39	SIDE	UNKNOWN	Holds head
CHECK ENGINE	MILITARY	49	44	FRONT	UNKNOWN	Slow-to-rise
CHECK ENGINE	MILITARY	48	30	REAR	UNKNOWN	Slow-to-rise
CHECK ENGINE	CIVILIAN	46	46	SIDE	YES	LoC
CHECK ENGINE	MILITARY	46	42	REAR	UNKNOWN	Slow-to-rise, Unsteady on rising, Poor performance
CHECK ENGINE	MILITARY	45	50	FRONT	UNKNOWN	Confused return to play
CHECK ENGINE	CIVILIAN	36	35	REAR	UNKNOWN	Team mate concern

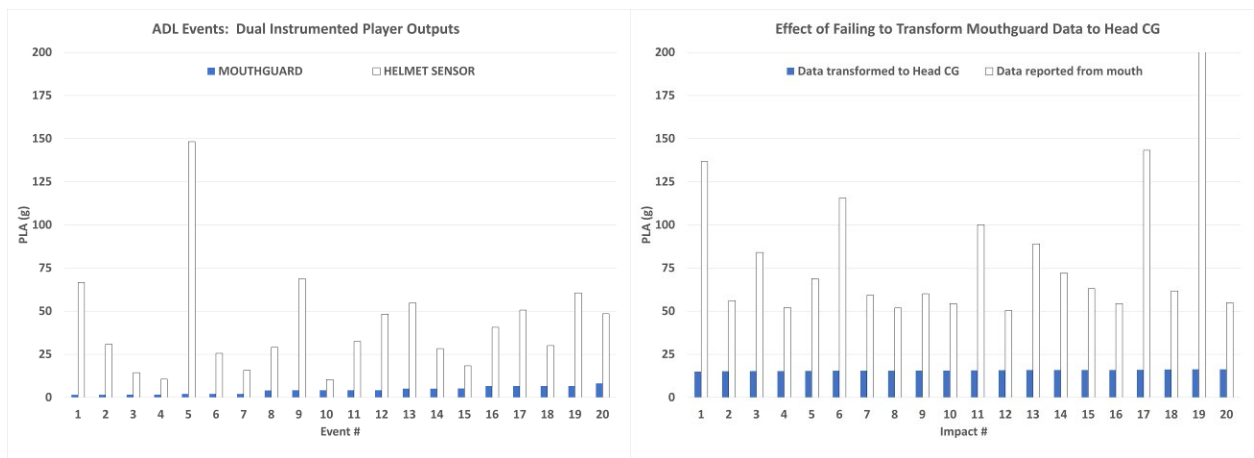
TABLE A1-2. Summary of n=37 'check engine' head impacts where the subject sustained at least one concussion sign. Of these, nine (9) were confirmed concussions and twenty-eight (28) had unknown assessment and diagnosis outcomes. Because the main purpose of the BAA project was to verify accurate head impact dosing, it was not always possible to gather full clinical details on each 'check engine' impact. In future work for DoD, the head impact dosing, 'check engine' impacts and clinical outcomes will be tied together.

The comparisons of head impact counting shown in **FigureA1-11** illustrate the differences between earlier studies and the current study for the number of head impacts per player, per game. For American football games and rugby matches, there were five to six head impacts per player per game in the current mouthguard study. The prior American football (average of 43 reported impacts per game – Schnebel et al. 2007; Guskiewicz et al. 2007; Broglio et al. 2010; Talavage et al. 2014; Stemper et al. 2019) and rugby (95 impacts reported per game – King et al. 2014) studies included 1.2 million data sets taken directly from sensor outputs but did not use video methods to confirm HAE occurred.



FIGUREA1-11. (Left) Comparison of American football head impact counts per game from the IMM in the current project contrasted with seven studies that used an instrumented American football helmet (Schnebel et al. 2007 Neurosurgery, Guskiewicz et al. 2007, Broglio et al. 2010, Crisco et al. 2010, Crisco et al. 2011, Talavage et al. 2014, Stemper et al. 2019). **(Right)** Comparison of rugby head impact counts from the IMM in the current study contrasted with a study that used a different mouthguard sensor (King et al. 2015). The large differences in impact counts are due to the lack of impact verification being conducted in the prior studies, and the inclusion of many non-HAE data sets in those studies. Accurate head impact counts are critical to DoD understanding blunt force risks and where to devote resources to monitor head impacts.

Finally, to assist DoD Stakeholders in understanding differences in reported HAE magnitudes in the current BAA Project versus prior studies, such as that funded by the DoD-NCAA CARE Consortium, **FigureA1-12, left** shows the differences in output magnitude for a subject wearing a firmly coupled impact monitor (mouthguard) and a poorly coupled sensor (helmet). **FigureA1-12 right** shows the effect of a different, yet common, error that occurs when one fails to conduct the essential mathematical transform of raw data from the mouthguard to the head CG (Higgins et al. 2007; Greybe et al. 2020; English FA 2021).



FIGUREA1-12. (Left) Outputs from a firmly coupled mouthguard sensor versus a poorly coupled helmet sensor for a single subject wearing both devices. These data show how poor coupling can turn innocuous 5g-10g ADL events into apparent 50g to 150g ‘check engine’ impacts. (Right) Even if a mouthguard sensor has good coupling to the skull, failure to properly transform kinematics from the mouth to the head CG can turn 10g-15g minor impacts into apparent 50g to 200g ‘check engine’ impacts. It has been known since the 1960’s (Gurdjian et al. 1966) that an impact in the range of 100g to 200g is life-threatening. At this high-gravity impact level, the rubber and aluminum crash test dummy will also see the head shear off the neck in testing of this severity. Reporting of random sensor outputs in the life-threatening impact range, without any clinical consequences, is pervasive in the existing literature. Most likely poor coupling, signal-noise ratio and inappropriate mathematical treatment are the culprits for these types of erroneous data. In order to provide trustworthy blunt force monitoring to the DoD, this BAA project has proven the necessity of good quality data and proper engineering approaches to gather head impact data.

Discussion

This Specific Aim aggregated 46,949 head acceleration event (HAE) data sets from 2405 impact monitoring mouthguards (IMM) deployed in military service members and civilians under BAA W81XWH-17-1-0019. The results showed that 92% of all collected HAE data sets among subjects were within the <10g PLA ranges of Activities of Daily (ADL) or in the minor head impact range of 10g to 20g.

In military subjects, the preponderance of low-gravity HAE was due to the high rate of grappling maneuvers (combatives), light head contact (boxing, combatives, American football, rugby), parachute deployment (static and freefall) and parachute landing falls (PLF) conducted in the training environments studied. In civilians, the high ratio of low-gravity HAE was due to evasive maneuvers to avoid contact, jumping/cutting during play and bumps/taps where head contact was made across all civilian sports.

There was n=37 ‘check engine’ head impacts that resulted in concussion signs observed on video or reported by staff on-site, with PLA ranging from 36g to 99g (median 58g) and Workload ranging from 24J to 118J (median 46J). These impacts were all in the top 1% by PLA magnitude, Workload magnitude, or both. And these ‘check engine’ impacts were also on par with, or more severe than, a sledgehammer hit to a helmeted crash test dummy head.

Of the ‘Check engine’ impacts, nine (9) were confirmed concussions and the other twenty-eight (28) had unknown concussion assessment and diagnosis outcomes. Unfortunately, when the IMM data were collected, the focus was to gather the head impact norms. It was not known how strongly the high-energy head impacts would correlate to ‘check engine’ concussion signs, and concussion diagnoses. Therefore, the mouthguard impact dosing data could not always be paired with clinical assessments, especially in cases where the clinical data were outside of the scope of

the mouthguard data collection protocols. In the future, all studies gathering mouthguard impact dosing will have clinical concussion outcomes as a part of the data collection protocols.

Nevertheless, the 'check engine' impact magnitudes and concussion signs, as well as the subset of diagnosed concussions that fell within the 'check engine' impacts, aligned closely with prior studies that used accurate methods to document concussion signs and estimate concussion dosing. These methods go back at least as far as the 1940's, when Dr. A.S. Holbourn from Oxford University emerged as an early proponent of velocity change being proportional to the concussion injury mechanism (A. S. Holbourn 1943). Dr. Holbourn was on to something: velocity change is correlated directly with the Workload in an impact, and Workload was found in this BAA project to correlate to 'check engine' impacts. Pioneering work continued in the 1960's with Col. John Paul Stapp MD and Dr. Larry Patrick, who, in their automotive safety research, proposed that the human concussion tolerance fell in the ranges of 40g to 80g (J.P. Stapp 1961; Patrick et al. 1965). This spans the 'check engine' ranges found in the current study. Around this same time Dr. Elisha Gurdjian, a colleague of Drs. Stapp and Patrick, published on the threat to life of 100g to 200g head impacts. This quickly became known as the Wayne State Tolerance Curve (Gurdjian et al. 1966), informing every head injury standard that exists today across transportation, military, and sporting activities.

Decades later, in a series of three studies sponsored by the NFL in the early 2000's (Newman et al. 2000; Pellman et al. 2003), a medical-engineering team estimated concussion-causing impacts to range from 50g to 140g head CG PLA and 40J to 250J Workload. The 'check engine' ranges in this study overlap substantially with these NFL lab-derived concussion ranges. In another NFL study (Zuckerman et al. 2020), documented that 74% of concussed players had at least one visible sign on video using the same video review criteria as used here. Finally, a recent paper (Clark et al. 2020) proposed concussion limits due to reconstructed equestrian falls to be in the range of 60g and 40J, which aligns closely with the median 'check engine' impact in this project.

The IMM reported impact counts of 5 to 6 per game for both American football and rugby matches. This stands in stark contrast against an average of 43 (American football) and 95 (rugby) impacts per game made in prior studies. While the current study used video verification and on-teeth sensing to document the sensor was properly coupled during a collision, these methods were almost never used in prior studies to screen out false HAE, leading to overestimates of impact counts by roughly an order of magnitude across studies with this methodologic flaw. It is hoped that the true impact counts presented in this work will help to level-set the impact counts of both military members and civilians. This may be an understated finding from the current work, as trust in the erroneous impact counts has fueled arguments in the vigorous debate around potential dose-response relationships of head impacts and degenerative brain diseases like Chronic Traumatic Encephalopathy (McKee et al. 2009).

In addition to clarifying data that quantifies true impact count and HAE magnitude distributions, this project should also help clarify why DoD clinicians have had such a hard time linking impact dose with clinical response previously. While it is conceptually simple to have any person who sustains an impact in the sledgehammer range be assessed for concussion, until now, this capability in the DoD and civilian worlds has been impaired by inaccurate sensors and methods to identify such an impact with accuracy and precision. The result has been that concussion clinicians today are violently aligned in a consensus that "*clinicians should not attempt to use impact magnitude to predict acute clinical outcomes*" (Guskiewicz & Mihalik 2011), "*the use of sensor systems to assess concussion cannot be supported at this time*" (McCroory et al. 2017), and there is "*little support for a concussive impact level*" (Kontos & Collins 2018) from sensor data.

This clinical consensus has been correct with the caveat that until now, the only available data was from head impact sensors that yielded objectively erroneous and random outputs, often paired with error-prone on-field methods where head impact video verification and time-signal engineering analyses were not used. Well-known American football helmet and skin-patch sensor studies have been widely published, with eighty-four (84) papers in PubMed.gov as of this writing.

Sensor error rates in live humans can be crippling to data quality. To amplify this point, leading engineers today, including the US Army Research Laboratory and US Army Aeromedical Research Laboratory, are only validating mouthguard-based impact monitoring systems as the gold standard (Kieffer et al. 2020; Liu et al. 2020; Jones et al. 2022). Helmet, headband, and skin-patch sensors have been dismissed due to their frequent behavior as random number generators in the laboratory and in the real-world.

While engineers have dismissed earlier systems due to inaccuracy, imprecision and randomness in the output data, there remains the issue of all the publications that contain inaccurate and imprecise impact dosing information. To help DoD Stakeholders understand why clinicians have arrived at the 'clinical irrelevance' conclusion based on earlier sensor data, a sampling of previously published head impact research is presented here.

An early helmet sensor study reported a non-sensical 2-3 impacts per game in the threat-to-life range of 120g to 200g with no reported effects (Duma et al. 2005). In another helmet sensor article, a case study of three 80g to 120g American football impacts within 16 milliseconds to one player's head was reported (Broglio et al. 2011). Rapid movement like this is physically impossible for the human head because it weighs as much as a bowling ball, and 16 milliseconds is about 1/10th of an eye-blink.

In rugby, data from a different mouthguard sensor than the current project reported approximately 5 life-threatening impacts per match up to 160g PLA (King et al. 2017). This study did not use video review or time-trace analysis to confirm accuracy of collected data. A recent study by leading rugby researchers (Jones et al. 2022) found a different mouthguard sensor system than the current study reported 5% of rugby impacts as greater than 100g, including several impacts in the life-threatening range of 150g to 300g without any ill effects to the player.

In Ice Hockey it has been reported from helmet sensors (McPherson et al. 2009) that kids under 14 years of age sustained about one impact apiece per season in the life-threatening range up to 200g without any observed clinical outcome. Using skin patch sensors in under-16 aged players, others (Smith et al. 2017) have reported about one non-injurious 100g to 200g impact per game. Both studies suffered from lack of video verification, and likely, poor quality sensor signals.

In early boxing work, sparring boxers have been outfitted with custom sensor headgear and reported 100g to 160g life-threatening head impacts with no ill effects to the fighter (Pincemaille et al. 1989). Later, it was reported from a different headgear sensor that life-threatening PLA of 180g to 190g was happening in sparring with no observed effect on the fighter (Stojisih et al. 2010). It is unlikely near-death impacts were happening in sparring, but rather, the sensor data uncertainty was enormous for these systems.

Finally, the English Football Association (FA) issued a white paper in 2021 on soccer heading impact magnitudes. They used a mouthguard sensor that did not transform data to the head CG reported the maximum PLA from a headed soccer ball in practice was 70g, akin to the sledgehammer shot to the head. Others have preceded the FA in making the same mathematical error (Higgins et al. 2007; Greybe et al. 2020). The true soccer heading range is about 5g to 20g PLA at the head CG (Kenny et al. 2022), on par with ADL events like trampolining or having a pillow fight.

While non-sensical head impact counts and magnitudes are pervasive in the published literature, the new and accurate data presented in this project could help shift that paradigm for the DoD's benefit. For clinicians who had to previously try and tease out dose-response relationships from often illogical sensor data, the problem has, until now, been intractable. While the data here do not conclusively define the oft-debated concussion threshold, they are clear about the rates of low-g HAE versus the high impact 'check engine' events.

The top 1% of impacts in this study is where all 'check engine' and concussion impacts resided, with a PLA and Workload in the ranges of 36g-99g and 24J-118J, respectively. While most of the HAE were like jumping on a trampoline or having a pillow fight, the 'check engine' impacts were at a level of a sledgehammer hit to a helmeted head, and above. If clinicians could leverage these data to know who is struck with such an intense severity, then it is likely more precise and targeted concussion assessments may reveal more precise and accurate correlation between concussion dose and response.

Armed with accurate impact monitoring data, future DoD studies need to tie in clinician assessment and diagnosis data for the top 1% by magnitude impacts, as well as any other clinically relevant impacts. By undertaking this systematic investigation and evaluation of 'check engine' impacts, the DoD may at last have viable data to help service members, and civilians will finally have useful data to assist clinicians with their concussion assessment and treatment.

SPECIFIC AIM #2 – Quantify a dose relationship between head impacts and changes in clinical assessments.

This aim, led by Dr. Stemper (MCW), attempted to correlate cumulative head impact workload in service members, cadets, and civilian athletes with changes in clinical assessments between baseline and repeated assessments obtained during and after the season or training period.

The study protocol was approved by the Institutional Review Board (IRB) at the Medical College of Wisconsin. A total of n=614 athletes from four NCAA Division III programs in southeastern Wisconsin were consented and enrolled over the course of the study.

All athletes were invited to participate in clinical assessments at three time points during the season. Each athlete was provided a unique study-specific number to prevent disclosure of personal identifiable information. The baseline clinical assessment occurred around the time of enrollment, prior to the initiation of preseason contact practices. The mid-season clinical assessment was performed approximately half-way through the season. The post-season clinical assessment was performed after the last game of the fall season.

Change of Approach Based on Impact Workload Data

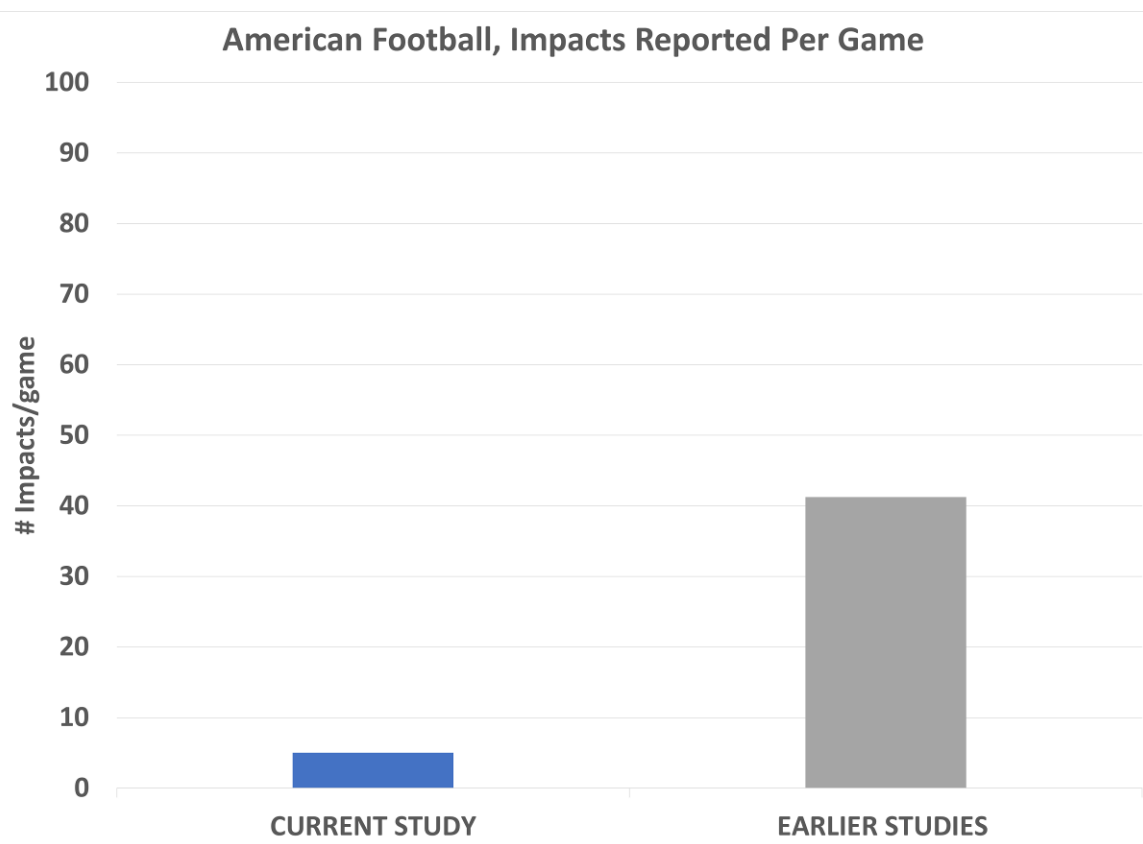
One unintended, but breakthrough finding during the study, was that head impact monitoring data were acquired from the Head Impact Telemetry System (HITS, Riddell, Chicago, IL) and the Impact Monitoring Mouthguard (IMM, Prevent Biometrics, Minneapolis, MN).

A total of n=7 American football players were dual instrumented for sixty (60) player-games with HITS and IMM. Video review was performed by two (2) ex-American football players and was time-synchronized to sensor data during each day of collection. Only gameplay data and video are included here. Data collected before game, between quarters, during halftime and after the game were removed prior to the analysis. Video reviewers classified events as Collision, No Collision (NC) or No video/Off Camera (NVOC). Estimated Collision location was reported, and collisions were qualitatively ranked as Small, Medium, or Large by the reviewers. Sensor data count was quantified for each sensor and sensor acceleration estimates for each system were compared to video. The IMM impacts presented here are a subset from previously published data (Bartsch et al. 2020).

For players who wore both systems, the data, compared to video review of head impacts, showed that the HITS system over-reported both impact count and magnitudes for all athletes on all days. The comparisons of head impact counting shown in **FigureA2-1** illustrate the differences between earlier studies and the current study for the number of head impacts per player, per game. There were five head impacts per player per game at the MCW DIII sites. Prior American football studies have averaged 43 impacts per game. Of the 1.2 million data sets taken directly from the HITS helmet, almost none of the historical data were video verified methods to confirm impacts occurred.

The large differences in impact counts are due to the lack of impact verification being conducted in the prior studies, and the inclusion of many non-head acceleration data sets in those studies.

After confirming the true head impact counting in the athletes, subsequent analysis looked at the accuracy of the data collected by each system versus video review. For this analysis, a subset of American football athletes who wore both systems simultaneously were used. A total of n=7 American football players were dual instrumented for sixty (60) player-games with the Riddell Head Impact Telemetry System (HITS) and Prevent Biometrics Impact Monitoring Mouthguard (IMM).



FIGUREA2-1. Comparison of American football head impact counts per game from the IMM in the current project contrasted with seven studies that used an instrumented American football helmet (Schnebel et al. 2007 Neurosurgery, Guskiewicz et al. 2007, Broglio et al. 2010, Crisco et al. 2010, Crisco et al. 2011, Talavage et al. 2014, Stemper et al. 2019).

Video review was performed by two (2) ex-American football players and was time-synchronized to sensor data during each day of collection. Only gameplay data and video are included here. Data collected before game, between quarters, during halftime and after the game were removed prior to the analysis.

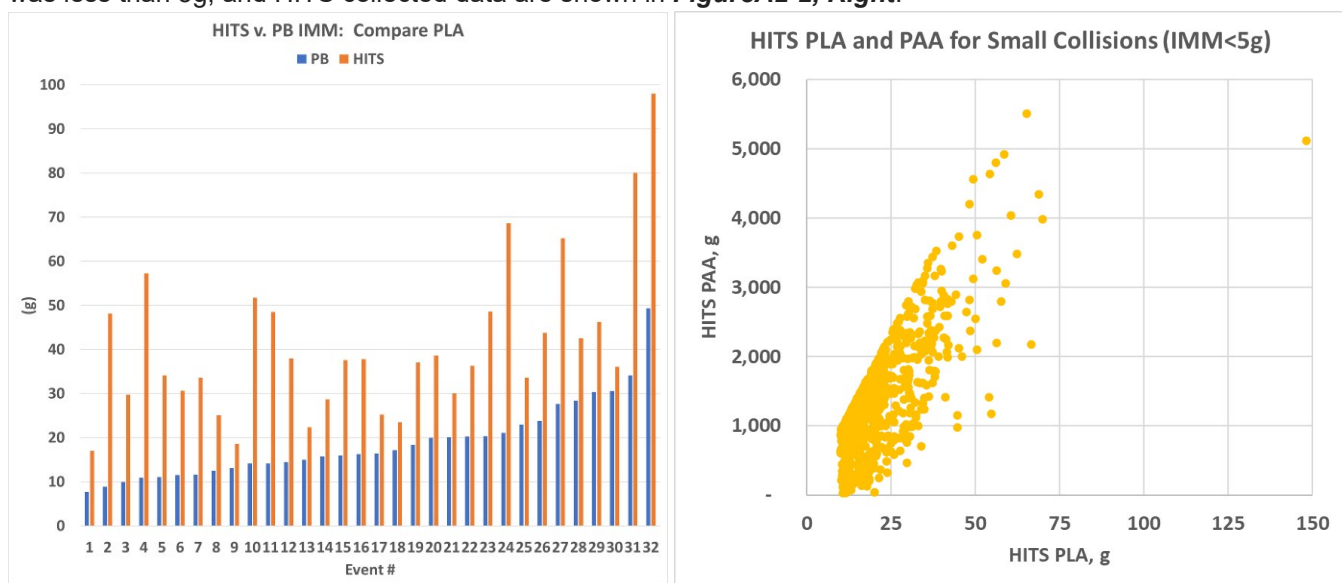
Video reviewers classified events as Collision, No Collision (NC) or No video/Off Camera (NVOC). Estimated Collision location was reported, and collisions were qualitatively ranked as Small, Medium, or Large by the reviewers. Sensor data count was quantified for each sensor and sensor acceleration estimates for each system were compared to video. The IMM impacts presented here are a subset from previously published data [Bartsch et al. 2020 Annals of Biomedical Engineering].

From the n=60 player-games with dual instrumented HITS helmets and IMM mouthguard, video reviewers observed 713 Collision, 486 NVOC, and 133 NC events. For the Collision events, Video reviewers rated 6 Large, 31 Medium and 676 Small Collisions. The IMM mouthguard collected data in 100% of the Large and Medium Collisions, as well as 83% of the Small Collisions. The HITS helmet collected data in 92% of the Large and Medium Collisions, as well as 84% of the Small Collisions, **see Table A2-1**.

Head Impact Triggering, Video Confirmed Collisions		
Collisions Magnitude, Video	IMM count [PLA range]	HITS count [PLA range]
Large (n=6)	6 [40g-60g]	6 [25g-130g]
Medium (n=31)	31 [8g-50g]	28 [15g-95g]
Small (n=676)	558 [3g-30g]	566 [10g-80g]

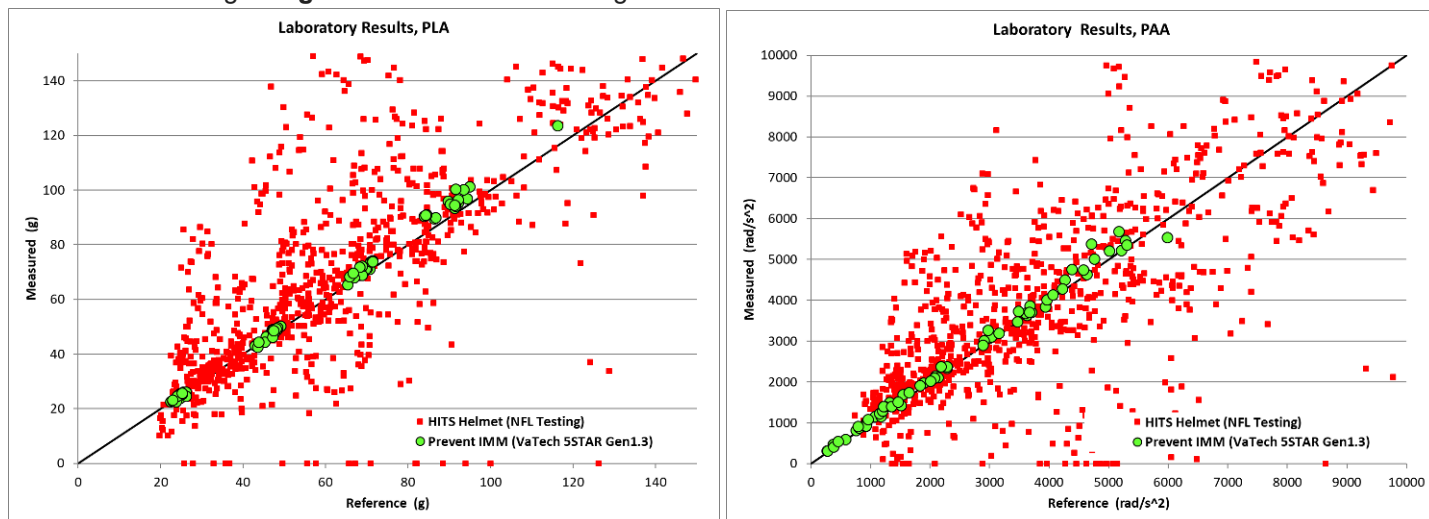
Table A2-1. Summary of HITS Helmet and Prevent Biometrics Mouthguard Sensor Triggering Versus Video

Both systems collected data simultaneously in $n=490$ data sets, and 99% of these data sets were confirmed collisions. An example from a single player is shown in **FigureA2-2, Left**. Cases where the collision magnitude was Small, the IMM was less than 5g, and HITS collected data are shown in **FigureA2-2, Right**.



FigureA2-2. (Left) Example of Dual Instrumented Player, $n=32$ Video-Verified Collisions with PLA Results from HITS Helmet (orange bars) and IMM Mouthguard (blue bars). **(Right)** HITS PLA Results from Video-Confirmed Small Collisions where the true head acceleration was less than 5g. Both plots illustrate the random errors with the HITS system versus an adequately coupled mouthguard, with errors approaching 50-75g, or more.

The finding that the HITS data were of higher magnitude than the IMM were not necessarily surprising, but what was new in this work was that the magnitude of error was as much as 50g to 150g for the smallest of head motions at 5g and below. This means that the HITS helmet reported concussion-level impacts for many cases where the true head acceleration was akin to jumping on a trampoline. A final analysis was done to confirm the on-field erroneous data. The HITS laboratory test data from an NFL study (Siegmund et al. 2015) was plotted against the IMM in the laboratory. These results are showing in **FigureA2-3** for linear and angular accelerations.



FigureA2-3. (Left) The comparison of laboratory peak linear acceleration test data for HITS versus Reference and IMM versus Reference. **(Right)** The comparison of laboratory peak angular acceleration test data for HITS versus Reference and IMM versus Reference. The lab uncertainties for HITS were up to 500% error while the IMM was roughly 5-10% different from the Reference.

After reviewing the video-based head impact counts, the head impact magnitude for dual instrumented athletes, and the laboratory test comparisons between HITS and the IMM, the errors for HITS proved to be a disqualifier from using that data in the rest of the study. This was the point where the study team elected to stop relying heavily on HITS data to correlate to clinical assessments.

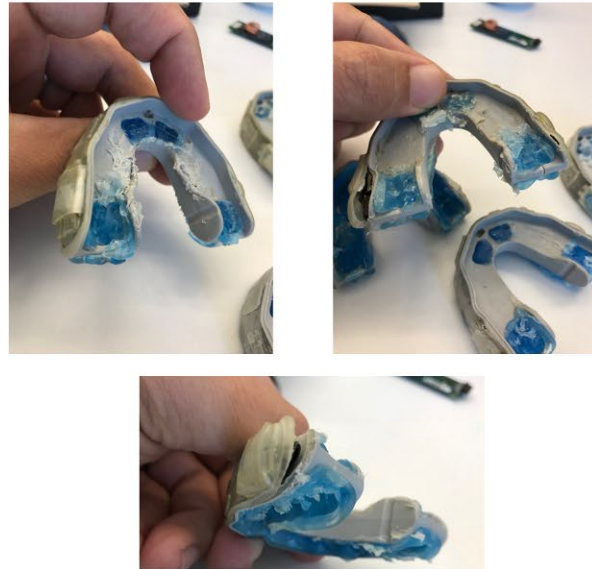
Challenges with IMM Hardware

While the HITS helmet suffered from near-random data outputs, the IMM also had its issues. When the study began the IMM was at the level of TRL6/7 and had not yet been deployed to American football players in an unrestrained manner. So while the HITS helmet had great durability and low accuracy in data, the IMM had accurate data yet suffered from mechanical failures due to a less-than-robust flexible circuit, as well as challenges to get players to avoid jamming the sensor into their facemask and/or not chewing on the sensor. The lack of durability negatively impacted the Study Team's ability to fully capture the head impact workloads for all athletes on all days.

Durability, Malice



Worn on Teeth



Chewing, Abuse

FigureA2-4. (Left) Example of properly worn IMM. (Right) Examples of IMM that were chewed, abused, jammed into helmets and broken during the study. While the Study Team did their best to have subjects wear the IMM, behavioral modifications proved tough to enact across all players.

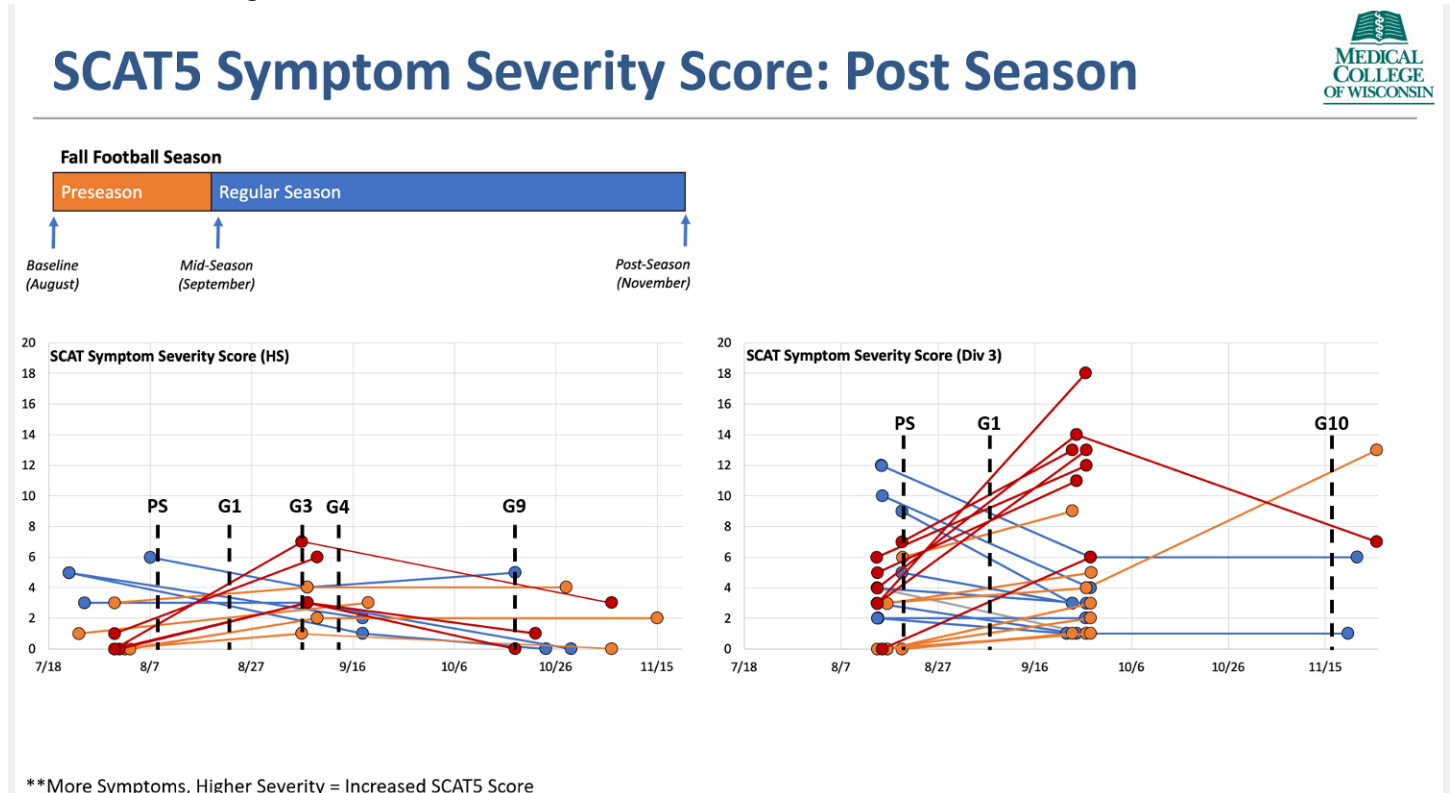
Because of the errors found in HITS data, and the challenges with acquiring the full season of data from participants who wore the IMM, the link between cumulative head impact workload and clinical data could not be made. Therefore, the results here lean heavily on the clinical findings and rely on head impact workload for only broad comparisons to athletes who participated in more or fewer athlete-exposure (AE) days.

Clinical Findings

Clinical assessments at all three time points included the Sport Concussion Assessment Tool 3 (SCAT3, McCrory et al. 2017). The administered SCAT3 assessment consisted of the symptom checklist. Clinical assessments at each of the three time points were identical with the exception of demographic data and health history information collected during the baseline assessment. The SCAT3 symptom checklist required athletes to rate the severity of 22 common post-concussive symptoms that they may have been experiencing. Each symptom was rated between 0 (none) and 6 (severe) and the cumulative score was computed between 0 (no symptoms) and 132, with higher scores indicating an increased number and higher severity of symptoms. The SCAT3 also included the Standardized Assessment of Concussion (SAC), a cognitive screening tool that quantifies higher or lower cognitive functioning.

Of the athletes consented, forty-nine (49) American football athletes were willing to undergo clinical assessments. Thirty-one athletes participated in the preseason baseline and mid-season clinical assessments. Twenty-four athletes participated in baseline and post-season clinical assessments. Preseason SCAT 3 assessments indicated that athletes had an average of 2.3 ± 3.1 symptoms at baseline with a symptom severity score of 2.8 ± 3.6 .

Mid-season assessments were obtained an average of 40 ± 3.4 days after baseline assessments. Athletes participated in 11 ± 5 sessions (range: 1-19 sessions) between baseline and mid-season assessments. Athletes had a non-significant trend of an increasing number of symptoms (3.6 ± 5.1 ; $p=0.22$) and increased symptom severity (6.0 ± 13.4 ; $p=0.20$) at the mid-season assessment compared to the baseline assessment. Athletes that participated in a greater number of sessions between baseline and mid-season assessments tended to have a greater change in concussion symptoms. For example, athletes participating in twelve or more sessions from baseline to mid-season assessment had an average increase of 2.4 ± 3.6 symptoms, with 7/17 of those athletes (41%) demonstrating an increase of three or more concussion symptoms at the mid-season assessment. 5/17 athletes had an increase of six or more concussion symptoms at the mid-season assessment. See **FigureA2-5** for details.

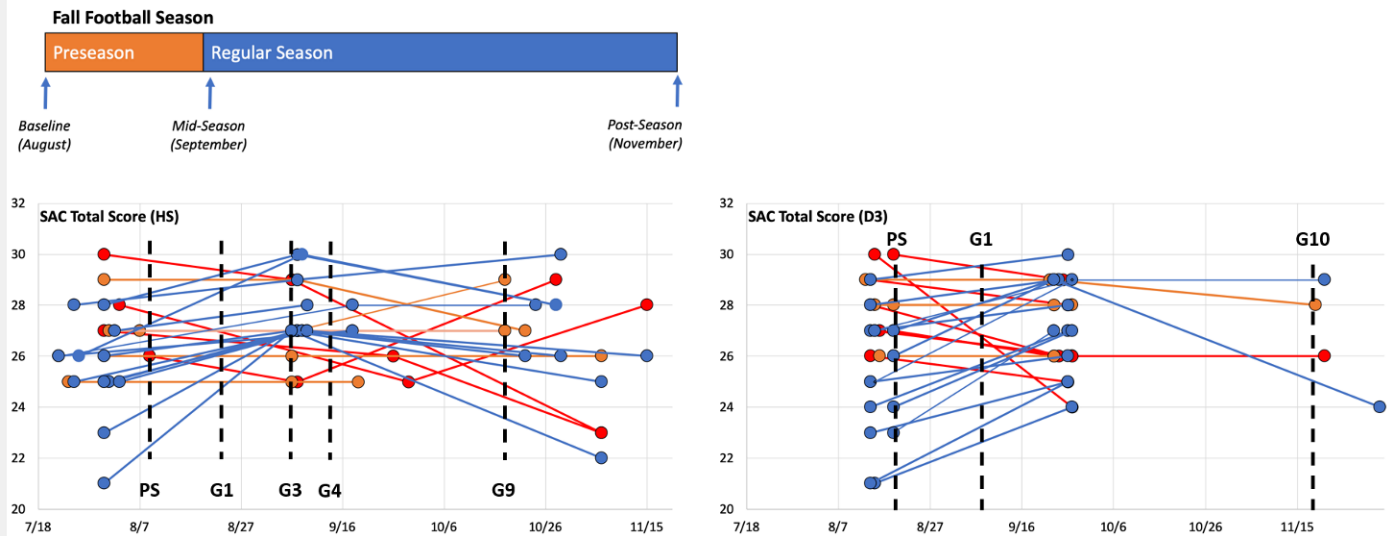


FigureA2-5. The SCAT clinical changes from pre-season, mid-season and post-season assessments did not show a clear trend for concussion symptoms for most of the athletes, and the concussion symptoms resolved by post-season.

Clinical changes identified in some athletes at the mid-season assessment may point toward progressive changes in concussion symptoms during the college football season. Of particular interest is that athletes with greater magnitude changes in the number of reported symptoms for the SCAT 3 assessments participated in a greater number of contact practices and games between the baseline and mid-season assessments. As reported above, 5/17 athletes that participated in 12+ sessions had an increase of six or more concussion symptoms from baseline to the mid-season assessment. This is noteworthy since the SCAT accounts for only 22 total symptoms. Therefore, those athletes reported a minimum increase of 27% of the symptoms in the SCAT inventory. While the magnitude of change (i.e., symptom severity score) was somewhat limited, development of 6+ symptoms during the first half of the season may provide some insight into a second mechanism of concussion associated with repetitive sub-concussive head impact exposure, as Stemper and colleagues reported previously for the CARE Consortium (Stemper et al. 2019).

Clinical scores returned to baseline levels by the post-season assessment for the most part in terms of the number of symptoms and the symptom severity. The average number of reported symptoms decreased from the mid-season to the post-season assessment to 2.4 ± 2.9 symptoms per athlete. That number was consistent with the number of symptoms reported at preseason baseline assessments (2.3 ± 3.1 symptoms per athlete). Similarly, the average symptom severity score decreased from the mid- to post-season assessment. Average post-season symptom severity score across 24 athletes was 3.2 ± 4.2 , compared to 2.8 ± 3.6 at baseline. See **FigureA2-6** for additional details on the Standardized Assessment of Concussion (SAC) testing.

SAC Symptom Severity Score: Post Season



**More Symptoms, Higher Severity = Increased SCAT3 Score

FigureA2-6. The SAC cognitive changes found that most athletes did not have cognitive changes at mid- and post-season assessment timepoints.

Decreased concussion symptoms at post-season follow-up assessments can be attributed to two possible mechanisms. Firstly, head impact exposure during preseason training camp is significantly greater than regular season head impact exposure as reported by McCrea, Stemper and colleagues for the CARE Consortium (McCrea et al. 2021; Stemper et al. 2019). This has contributed to a higher rate of concussion during the preseason, and it can be expected that athletes also develop concussion-like symptoms at a higher rate during this time period. However, decreased head impact exposure during the regular season may offer an opportunity for athletes to recover during a period with head impact frequency more in line with physiologic limits. This may explain the increased number and severity of concussion symptoms at the mid-season assessment in the current study and a reduction in both number and severity of symptoms by the post-season assessment. A possible second mechanism for this finding could be that post-season assessments were often conducted one or more weeks following the conclusion of regular season contact activities. This may point to a relatively rapid recovery that occurs for athletes with the removal of regular head impact exposure. This may also help to explain a lack of cumulative changes or concussion risk from one season to the next in college football.

SPECIFIC AIM #3 – Correlate head impact exposure measured with the IMM system with onset and severity of concussion.

This aim, led by Dr. Bartsch (Prevent Biometrics), Mr. Rooks (USAARL) and Dr. Stemper (MCW), correlated head impacts measured by the IMM with direct, video or clinically reported observations of concussion signs in service members, cadets, and civilian athletes. Injury risk functions were developed to associate increasing risk for concussion onset with the magnitude, number, frequency, and type of head impact(s).

INTRODUCTION

Over 50 million athletes in the United States participate in contact sports, and hundreds of thousands to millions sustain concussions annually. Roughly 50% of concussed athletes go unidentified (McCrea et al. Clin J Sports Med 2004) and continuing to play for only 15 minutes after a concussive blow causes recovery time to more than double (Elbin et al. Pediatrics 2016). The past decade has seen intense investigation into the accuracy and debate about clinical utility of head impact monitoring systems.

A seminal paper by Mihalik and Guskiewicz reviewed American football helmet sensor and clinical data, concluding “*a concussive injury threshold is elusive*” (Guskiewicz & Mihalik 2010). In 2014 the Institute of Medicine concluded “*there are currently inadequate data to define thresholds specifically associated with concussions*” (IOM, 2014). More recently authors have advocated for lower uncertainty data: “*accurate measures of individual exposure will yield a direct estimate of the human tolerance*” (Meaney et al. 2014), “*as more accurate sensors are designed...*” (Elliott et al. 2015), “*valid methods of measuring the direction and severity of on-field head impacts are needed*” (Siegmund et al. 2015).

In this aim we began exploring whether lower uncertainty head impact measures will reveal more clarity on the single dose-response experience in athletes. We measured and video-verified head impacts using an impact monitoring system embedded in an athletics mouthguard. Any single impact leading to concussion signs on video was noted and its magnitude was compared to previously published data (Bartsch et al. 2019; Bartsch et al. 2019; Bartsch et al. 2020).

This approach stands in contrast to almost two decades of research using helmet-mounted head impact sensors in American football has led to the acceptance by many that there is no relationship between a single, high energy head impact, and concussion. This is because researchers have never been able to reliably link high energy head impact time traces reported by the helmet sensor, qualitative video verification of magnitude, and concussion. Leading clinicians have gone so far as to declare data from helmet sensors to be ‘clinical irrelevant’ (Guskiewicz & Mihalik 2010).

While this prior work was correct in confirming the lack of a dose-response relationship for those sensors, scientists need to be careful to avoid lumping all types of sensors into the same batch of bad apples. More recent data (Filbern et al. 2021; Liu et al. 2020; Kieffer et al. 2020; Jones et al. 2022) has confirmed that impact monitoring mouthguard (IMM) sensors are capable of reporting head center of gravity (CG) kinematics within 5-10% of the true value; this is a massive improvement over earlier sensors which were shown to have upwards of 500% error in testing sponsored by the National Football League (NFL, Siegmund et al. 2015).

On the playing field, mouthguard sensors have also been shown to (1) collect head acceleration event (HAE) data when on the teeth (Bartsch 2022-submitted), (2) correlate closely with video review of collisions (Wu et al. 2015), and (3) report HAE magnitudes for direct and inertial loading in the ranges of laboratory calibrations (Jones et al. 2022). After these validation steps have been completed, the only remaining question is: will a concussion dose-response relationship be present when one is finally able to monitor HAE with accuracy and precision? This paper outlines the findings and objective data that confirm a clear dose-response relationship for head impacts in the top 1% by magnitude, versus concussion signs and concussion.

METHODS

Data were collected on 973 subjects who wore the IMM (Prevent Biometrics, Minneapolis, MN, USA) over 3449 subject-days. The IMM has been extensively calibrated in the laboratory and on the playing field. A variety of military and civilian activities were studied, and ethics approval was obtained for each activity. Military training, including parachute, hand-to-hand combat, and athletics, along with civilian American football, hand-to-hand combat, ice hockey and rugby are presented here. A total of 54,602 HAE's were analyzed using processing methods reported in-depth elsewhere. Data were persisted when any channel of the IMM raw acceleration exceeded 5g. Data were considered true HAE when the

hardware confirmed the IMM was on the teeth. Video verification, on-teeth hardware, and time trace analysis were all used to confirm HAE versus non-HAE data.

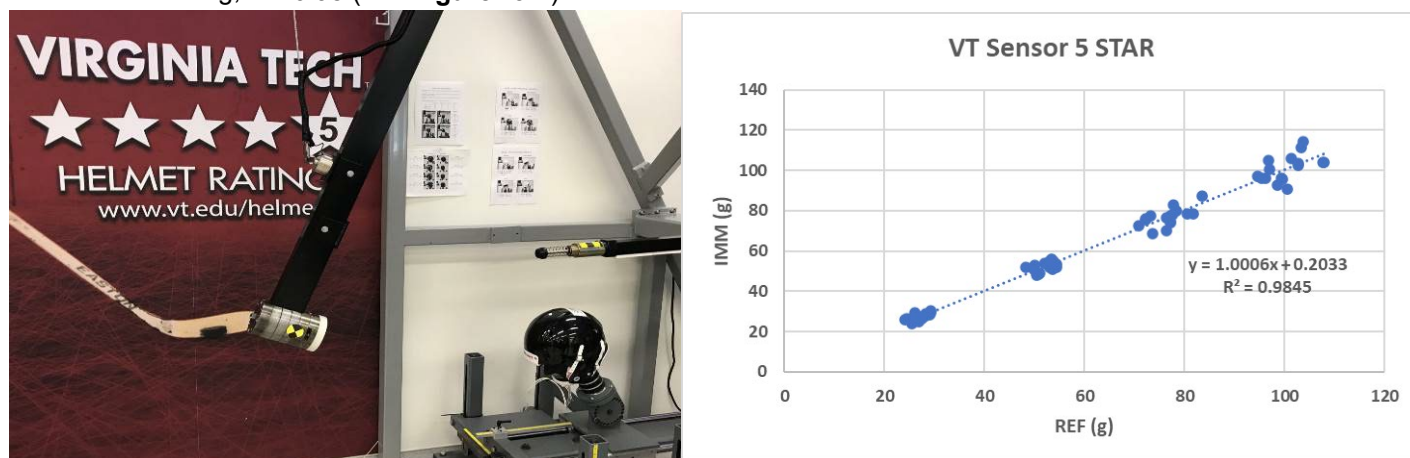
Visible concussion signs (Davis et al. 2019) were documented using the same methods as all major sporting bodies. These collisions, where at least one visible concussion sign was seen on video within 30 seconds of impact, are 'check engine' impacts. All 'check engine' impacts had their raw (at tooth) and head CG time traces examined for head impact frequency content of 20-80Hz (Bartsch et al. 2014), and the impact direction/location was confirmed against the video. The 'check engine' impact magnitude was qualitatively checked as well, to confirm whether the impact on video was Large, Medium, or Small.

Kinematic parameters of interest included peak linear acceleration (PLA) and workload. Workload, in Joules, was calculated using a textbook equation and estimated the kinetic energy transfer to the head. The SIMon finite element model (Takhounts et al. 2008) was used to estimate internal brain strains in concussion impacts. Linear and angular, velocity and acceleration, six-degree of freedom (6DOF) kinematic time traces were captured at 3200Hz over 50 milliseconds. From these time traces, peak linear acceleration (PLA), peak angular acceleration (PAA), peak linear velocity (PLV), peak angular velocity (PAV), workload/kinetic energy transfer (KET), and impact location were computed.

RESULTS

Laboratory Calibrations

To calibrate the IMM system it was tested independently at Virginia Tech University as part of the Sensor STAR Program in impacts from 25g to 100g to the front, front boss, rear boss and rear. The results fit a linear model of the form $IMM=1.00*REF+0.2g$, $R^2=0.98$ (see **FigureA3-1**).



FigureA3-1. IMM laboratory test results confirm data fit a linear model versus Reference

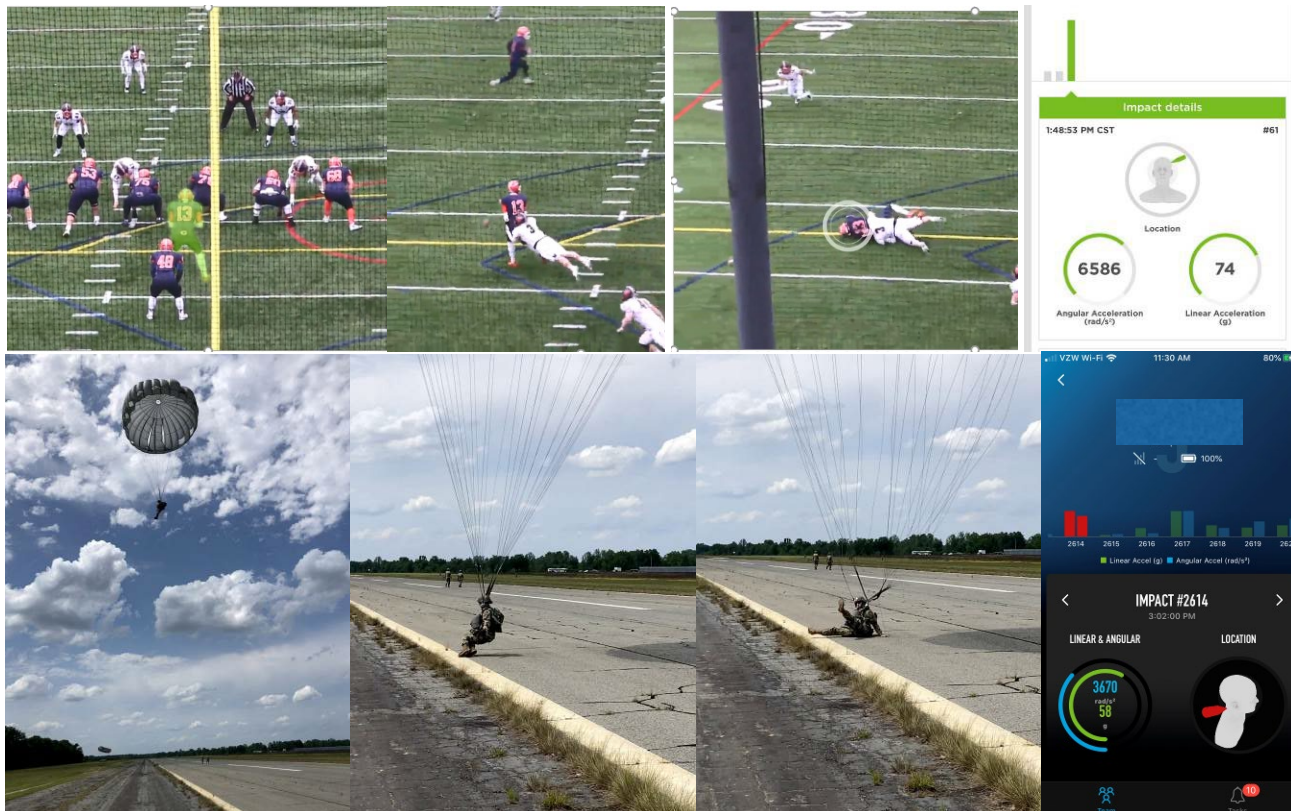
Head Impact Measurement and Concussion

Table A3-1 summarizes the subjects and data. A total of 68% of all HAE were collected below 10g Peak Linear Acceleration (PLA), in the range of activities of daily living (ADL).

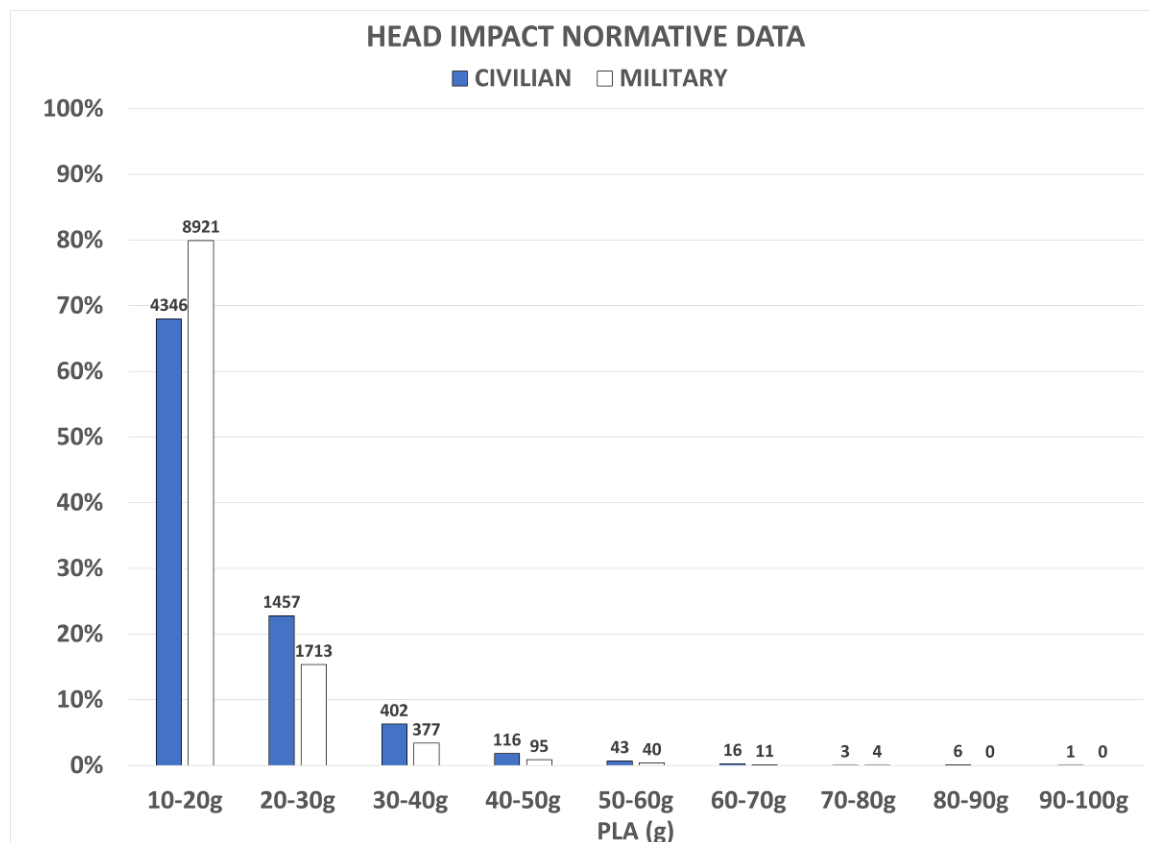
	MILITARY	CIVILIAN	TOTAL
Subjects	542	431	973
Subject-Days	2568	881	3449
HAE	44939	9663	54602
Minor HAE, <10g	33778	3273	37051
Head Impacts, >10g	11161	6390	17551
'Check Engine' Impacts	19	37	56

TABLE A3-1. There were 17551 head impacts collected with PLA>10g at the head CG. A total of 56 'check engine' impacts, where the subject had at least one concussion sign, were recorded. HAE=Head Acceleration Event.

In **FigureA3-2**, ‘check engine’ and concussion impact examples are shown. The ‘check engine’ and concussion impacts are always Large or Medium on video; often the player has difficulty getting up, has trouble balancing, and can even lose consciousness.

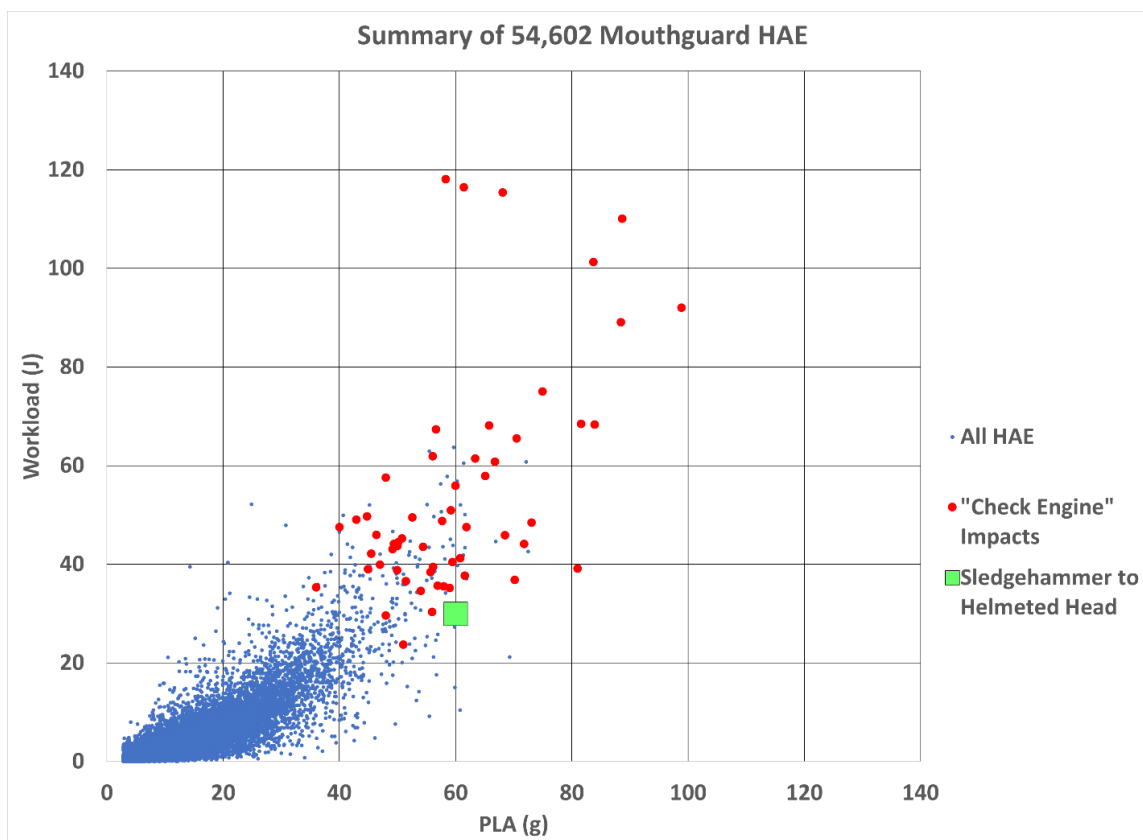


FigureA3-2. The ‘check engine’ head impacts always involved significantly higher energy and momentum transfer. Concussion signs and diagnosed concussion was only found in subjects sustaining these types of impacts, within the top 1% of all head impacts by PLA magnitude, workload magnitude, or both.



FIGUREA3-3. Head Impact Distribution for Head CG PLA >10g. 75% of the HAE>10g were in the minor range, between 10g to 20g. Less than 1% of all head impacts were between 50g and 100g.

There were 54,602 head acceleration event (HAE) data sets from 973 military service members and civilians who wore a laboratory-calibrated impact monitoring mouthguard (IMM) aggregated in this aim of the study.



FIGUREA3-4. Summary of all 54,602 HAE data sets (small, filled circles) from the current mouthguard study. The n=57 'check engine' impacts are shown in large, filled circles. A total of 18 concussions were confirmed from the 'check engine' impacts, and the other 39 impacts either had no concussion assessment or the assessment had an unknown outcome. For reference, a sledgehammer hit to a helmeted ATD is also shown (solid square).

The results showed that 92% of HAE data sets were within the <10g PLA ranges of Activities of Daily (ADL) or in the minor head impact range of 10g to 20g (see **Table A3-1**, **FigureA3-3**, and **FigureA3-4**).

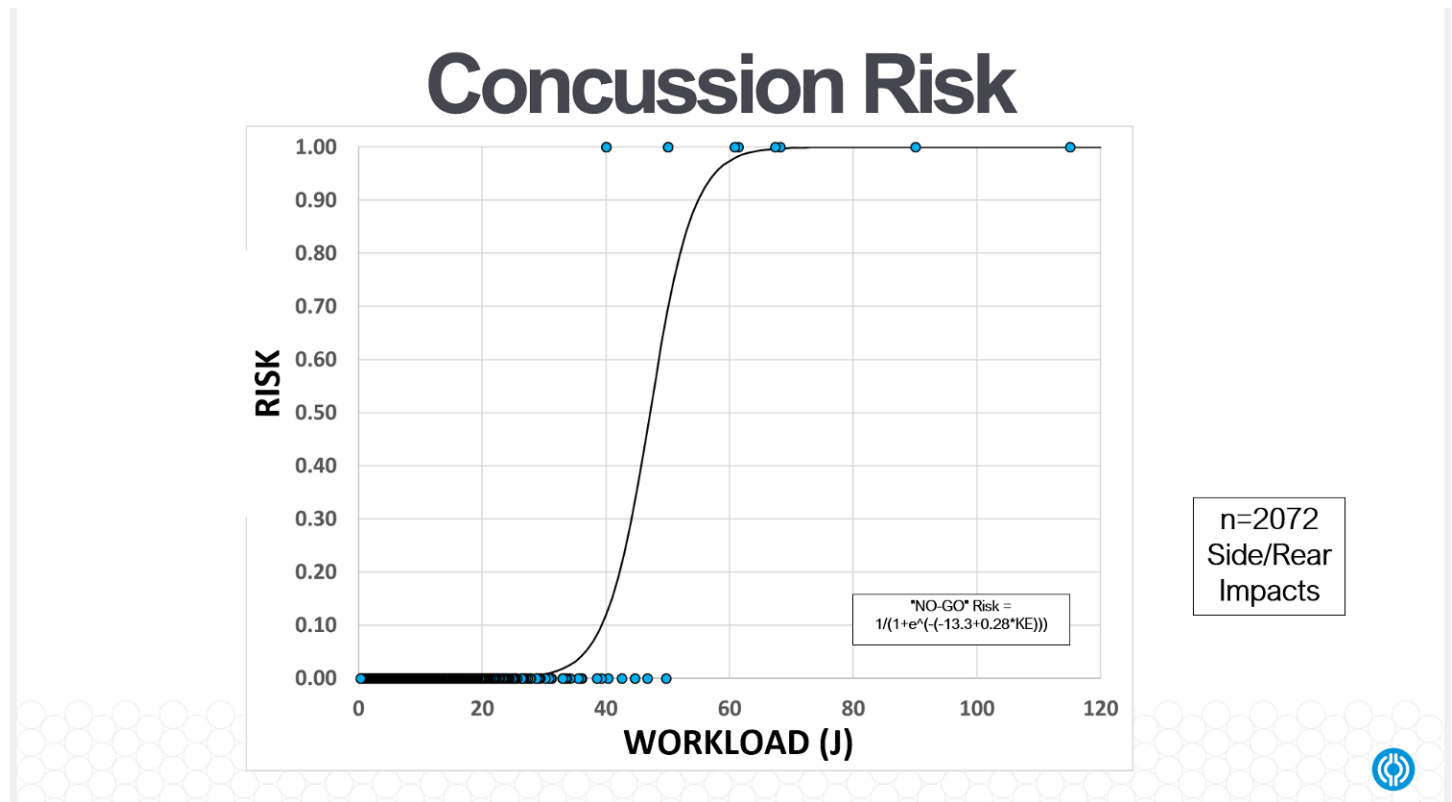
In military subjects, the preponderance of low-gravity HAE was due to the high rate of grappling maneuvers (combatives), light head contact (boxing, combatives, American football, rugby), parachute deployment (static and freefall) and parachute landing falls (PLF) conducted in training environments. In civilians, the high rate of low-gravity HAE was due to evasive maneuvers to avoid contact (boxing, mixed martial arts, karate), jumping/cutting during play (American football, ice hockey) and light bumps/taps where head contact was made. The SIMon results are shown in Table A3-2 for the n=18 'check engine' impacts with concussion.

HAE Data	PLA (g)	PAA (rad/s^2)	PLV (m/s)	PAV (rad/s)	WORKLOAD (J)
Avg	57	2733	4.6	19	54
St. Dev	16	933	1.2	8	28
SIMon Data	MPS	CSDM 0.05	CSDM 0.10	CSDM 0.15	CSDM 0.25
Avg	0.46	0.97	0.67	0.32	0.05
St. Dev	0.10	0.04	0.17	0.15	0.05

Table A3-2. Summary of HAE Data and SIMon results for n=18 concussions. The PLA, Workload, or both were in the top 1% by magnitude for all concussions. While all these players were concussed, including many who lost

consciousness, none of them had a CSDM25 above the critical level of 0.54, or above the MPS critical level of 0.87 (Takhounts et al. 2008). This could point to the need to improve computer models of concussion to correlate with what is being measured in subjects in the field.

The concussion risk function shown in **FigureA3-5** shows a clear relationship between risk and head impact workload. This curve has a very strong relationship between the energy transfer in the impact, with the 50% limit occurring at 45J, and the 10% limit occurring at 40J.



FigureA3-5. Based on data collected in this study, a concussion injury risk function was derived. The logistic regression followed a clear, and desirable, step function shape with a sharp contrast between low risk (non-concussed) and high risk (concussed) individuals. All of the concussions were from single head impacts that were in the top 1% by workload, peak linear acceleration, or both.

DISCUSSION

As shown in **FigureA3-4**, there was n=57 'check engine' head impacts that resulted in concussion signs, with PLA ranging from 36g to 99g (median 58g) and Workload ranging from 24J to 118J (median 46J). These impacts were all in the top 1% by PLA magnitude, Workload magnitude, or both. And the median 'check engine' impact PLA was on par with a sledgehammer hit to a helmeted crash test dummy head (60g), with about 50% higher workload (30J). This puts into context just how massive the 'check engine' head impacts were. The median 'check engine' impact was nearly the same as the 50% risk factor for a concussion, shown in **FigureA3-5**.

Of the 'check engine' impacts, eighteen (18) were confirmed concussions and the other thirty-nine (39) had unknown concussion assessment and diagnosis outcomes. All concussions had corresponding 'check engine' impacts, looked Large or Medium on video, and concussion signs present within 1-5 seconds post-impact. When analyzing SIMon outputs, the concussive brain strains fell well below theoretical injury limits for diffuse axonal injury despite the devastating nature of the impacts caught on video. Many of these impacts were also within ranges of laboratory concussion recreations from the National Football League (NFL, Viano et al. 2005). This paradoxical result could point to the need to more fully vet computational models versus on-field data and clinical outcomes. This will be a focus of the ongoing work.

These results may start to shift the paradigm for clinicians who had to previously try and tease out dose-response relationships from often illogical sensor outputs compared to video observations. While the data here do not conclusively

define the oft-debated concussion threshold, they do confirm the top 1% of impacts is where all 'check engine' and concussion impacts resided, with a PLA and Workload in the ranges of 36g-99g and 24J-118J, respectively. While 92% of all HAE were at or below the level of jumping on a trampoline, having a pillow fight, or heading a soccer ball, the 'check engine' impacts were at a level of a sledgehammer hit to a helmeted head, and above. By knowing who is the most severely struck, more precise and targeted concussion assessments should be feasible. And this, in turn, may reveal more precise and accurate correlation between concussion dose and response.

SPECIFIC AIM #4 – Quantify laboratory performance of the IMM system under military conditions.

This aim, led by Mr. Rooks (USAARL), independently quantified the accuracy and precision of the IMM under military conditions. Controlled tests were conducted that compared IMM to Reference crash test. Military helmet(s). This aim produced military-specific performance for the IMM, which already had rigorous performance validated in over 1000 laboratory tests in civilian athletics' environments (see Bartsch et al. 2014, Hedin et al. 2016, Bartsch et al. 2020, Liu et al. 2020, Kieffer et al. 2020, Jones et al. 2022).

INTRODUCTION

Mild traumatic brain injury (mTBI) is a concern in military environments (Defense and Veterans Brain Injury Center [DVBIC], 2020). Symptoms associated with mTBI (concussion) include loss of consciousness for up to 30 minutes, alteration of consciousness or mental state for up to 24 hours and post-traumatic amnesia for up to 24 hours (O'Neil et al. 2013). Repeated mTBI is associated with longer recovery times and reduced ability to process information (Gronwall & Wrightson, 1975). Multiple mTBIs could potentially increase the risk of neurodegenerative disorders (McAllister & McCrea, 2017). While no link has definitively been established between Alzheimer's disease and repetitive mTBI, earlier onset of Alzheimer's was observed in a study of retired football players versus the general population (Guskiewicz et al. 2005).

Timely and accurate information is important for clinical diagnosis and return to duty following a concussion (Asken, et al. 2016). Diagnoses often depend on self-reporting, which may result in missed diagnoses since individuals who suffer a concussion may not have awareness of symptoms (McCrea et al. 2004). Environmental sensors (ES) attached to a Soldier's helmet or head are a tool that may provide the ability to identify possible concussions in real time without relying on self-reporting. Environmental sensors record head accelerative exposures and give estimates of kinematics at the head center of gravity (CG). The Impact Monitoring Mouthguard (IMM) developed by Prevent Biometrics (Edina, MN) is a mouthguard-based ES that could be used to measure kinematics at the head CG resulting from exposure. The IMM is capable of recording head kinematics (linear acceleration, angular velocity) associated with potentially concussive events (Bartsch et al. 2014; Bartsch et al. 2020; Liu, et al. 2020; Jones et al. 2022; Kieffer et al. 2020).

In order for ES to be useful, they need to give reasonable estimates of accelerative exposures at the head CG. Laboratory testing using Anthropometric Test Devices (ATDs) is one way that mouthguard sensors have been validated (Bartsch et al. 2014; Camarillo, et al. 2013; Kieffer, et al. 2020; Kuo et al. 2016; Siegmund, et al. 2016). Testing has generally included direct impacts to helmeted and un-helmeted ATDs of a 50th percentile male at increasing levels of severity (Bartsch, et al. 2014; Camarillo, et al. 2013; Kieffer, et al. 2020; Kuo, et al. 2016; Siegmund, et al. 2016). Agreement levels between mouthguard and laboratory grade sensors have been excellent (Bartsch, et al. 2014; Camarillo, et al. 2013, Hedin et al. 2016, Liu et al. 2020, Kieffer et al. 2020, Jones et al. 2022). Kuo, et al. (2016) proved in a lab setting that one of the main factors in determining levels of agreement have been whether the ATD used in testing contains a "clenched mandible", "unconstrained mandible", or no mandible. These three mandible conditions were tested and used to reproduce data from previous studies (Bartsch, et al. 2014; Camarillo, et al. 2013; Siegmund, et al. 2016). One factor not tested by these studies is the accuracy of the mouthguard sensors under indirect or inertial loading.

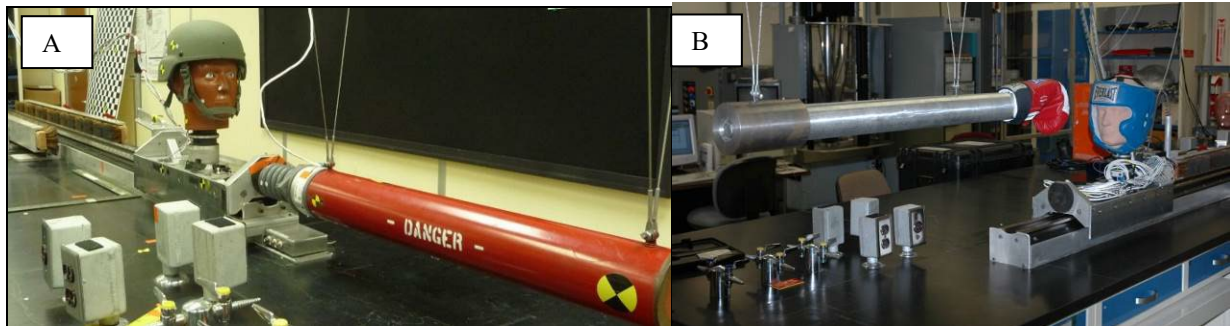
The objective of this study was to compare the kinematic response predicted by the instrumented mouthguard with that reported by a laboratory reference measured at the center of gravity of an anthropometric test device (ATD) headform. Additionally, this study evaluated the influence of the choice of ATD headform (with varying mandible designs) and indirect (e.g., inertial) versus direct (e.g., impact) loadings.

METHODS

The study was conducted at the United States Army Aeromedical Research Laboratory (USAARL) to determine the agreement between the IMM and laboratory grade sensors at the head CG of multiple ATD headforms. Testing was conducted under direct and inertial loading. The IMM was tested during frontal impacts using a pendulum and minisled setup to determine how well it reproduces kinematics measured at the head CG via laboratory grade sensors of a modified National Operating Committee on Standards for Athletic Equipment (NOCSAE) and a Mandible Load Sensing Headform (MLSH) headform. The modified NOCSAE headform incorporated a screw through the bottom of the jaw to replicate conditions of a "clenched mandible" and the MLSH was designed with a mandible that freely rotates and uses springs to simulate clamping forces in the jaw. Boxing headgear was also added to the ATD headforms for direct impacts to determine whether the headgear affects accuracy of the IMM.

MINISLED TEST DEVICE

The United States Army Aeromedical Research Laboratory (USAARL) pendulum and minisled set up consists of a horizontal rail system (3.66 meters) and a low friction sled. The sled is placed at one end of the track and a 22.7 kg rigid direct impact pendulum (**FigureA4-1**) strikes the headform directly simulating a boxer's punch. The system can also be setup for indirect impacts via a 45.4 kg indirect impact pendulum, which simulates an inertial load to the sled representing the C7-T1 junction. Palmyra brushes bring the sled to a stop approximately 1.3 meters from the impact site. The pendulum release height can be adjusted to impact the system with varying amounts of intensity and energy.

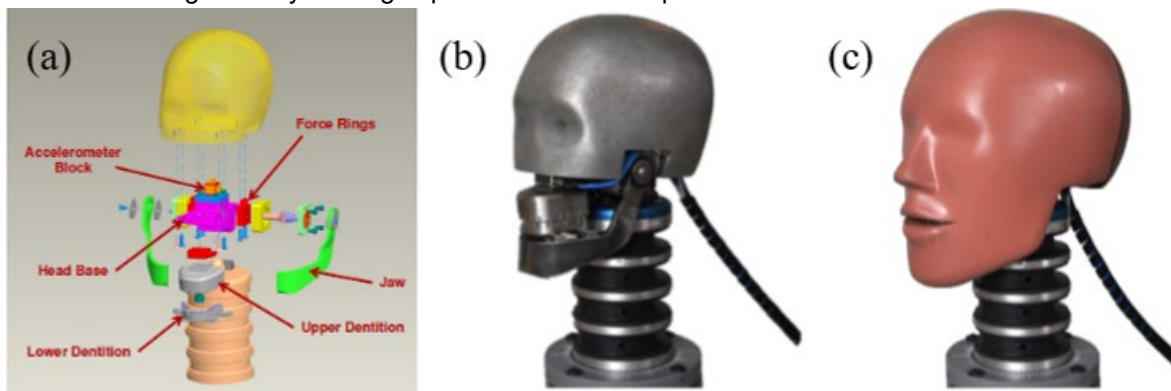


FigureA4-1. United States Army Aeromedical Research Laboratory (USAARL) Minisled System. Showing the (A) indirect, or inertial, loading setup and the (B) direct loading setup.

ANTHROPOMETRIC TEST DEVICE HEADFORMS

The MLSH (**FigureA4-2**) is a 50th percentile Hybrid-III (H-III) ATD altered to have an articulating jaw that allows testing of a mouthguard sensor. The jaw contains a steel mandible with upper and lower simulated dentition and spring-loaded temporomandibular joints (TMJ). The TMJ produces a clamp/bite force of approximately 22.3 N (Siegmund et al. 2014). The headform connects to a Hybrid III neck.

The NOCSAE headform (**FigureA4-3**) was designed for use in NOCSAE standards testing. The NOCSAE headform used in this study was altered to allow fitting of the IMM; an adapter was added to allow connection to an H-III neck. A slot in the face of the modified NOCSAE headform allows insertion of the IMM into the headform; a screw underneath the jaw allows simulation of a biting force by holding in place an aluminum plate underneath the IMM.



FigureA4-2. Mandible Load Sensing Headform (MLSH). (a) exploded view, (b) assembled and attached to Hybrid III neck, and (c) with synthetic Hybrid III skin.



FigureA4-3. Modified National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform with slot for the IMM and metal plate to hold the sensor onto the upper dentition.

INSTRUMENTATION

The MLSH headform is instrumented with a nine-accelerometer array (Endevco 7264B-2000) located at the head CG to transform linear acceleration into angular rate and angular acceleration during post-processing (Martin et al. 1998). The MLSH also includes a six degree of freedom (6-DOF) load cell (triaxial forces and triaxial moments) at the lower neck (Humanetics 7992JS1TF), 3-DOF load cells at the left and right TMJ joints (PCB Piezotronics 260A01), and a 3-DOF load cell at the upper dentition (PCB Piezotronics 260A01).

The NOCSAE headform is instrumented with three linear accelerometers (Endevco 7264C-2KTZ) and three angular rate sensors (DTS ARS PRO-8K) at the head CG. The NOCSAE headform had the same lower neck load cell as the MLSH but did not include TMJ or upper dentition load cells.

All tests included a uniaxial linear accelerometer on the sled carriage (Endevco 7264C-500). For indirect impacts, a triaxial force load cell was installed on the pendulum impactor (Humanetics 8728TF). For direct impacts, a uniaxial linear accelerometer (7264C-500TZ) was installed on the pendulum to measure linear acceleration. Linear accelerometer on the sled was used to determine repeatability of the tests.

All laboratory grade data was collected using the Synergy data acquisition system (High-Techniques, Madison, WI, USA) and sampled at 20,000 samples per second (sps). Additionally, high-speed video was captured for all tests at 1500 frames per second (fps) using a Vision Research (Wayne, NJ, USA) Phantom Miro camera. Markers were placed on the headform, neck, and sled to track movement.

TEST CONDITIONS

Tests were conducted using both inertial (indirect) loading as well as direct loading using both the modified NOCSAE and MLSH headforms. All tests were conducted with exposures in the -x direction (direction of motion is opposite the direction the ATD was facing). For inertial loading, tests were conducted at four impact energies of approximately 130 J, 198 J, 266 J, and 334 J. For each energy, three different IMM sensors were tested. Impacts for a given energy and headform were repeated three times (Appendix B1).

For the direct impacts, tests were conducted at two different impact energies simulating the punch of a Flyweight and Super heavyweight boxer (Appendix C) (Walilko et al. 2005). For each energy, three different IMM sensors were tested. Impacts for a given energy and headform were repeated three times with and without boxing headgear (Appendix B2).

THE PREVENT BIOMETRICS IMPACT MONITORING MOUTHGUARD (IMM)

The IMM is a boil-and-bite (place in boiling water 20 seconds, put on teeth for 30 seconds) or custom (made from 3D dental scan) mouthguard with embedded flexible circuitry; a tri-axial gyroscope and three linear accelerometers are attached to the circuitry inside the mouthguard. The linear accelerometers and tri-axial gyroscope allow kinematics (head angular velocity/angular acceleration and linear acceleration) to be recorded at the mouth (Bartsch et al. 2014). Linear acceleration at the mouth is transformed to the head CG using rigid body kinematics via the following equation:

$$\vec{a}_{CG/CG} = \vec{a}_{MG/CG} + \vec{\alpha}_{Head/CG} \times \vec{r}_{CG/MG} + \vec{\omega}_{Head/CG} \times (\vec{\omega}_{Head/CG} \times \vec{r}_{CG/MG}) \quad (2)$$

Where $\vec{a}_{CG/CG}$ and $\vec{a}_{MG/CG}$ are linear accelerations at the head CG and mouthguard relative to the head CG, respectively, and $\vec{r}_{CG/MG}$ is the head CG position relative to the mouthguard. Finally, $\vec{\omega}_{Head/CG}$ and $\vec{\alpha}_{Head/CG}$ are the head angular velocity and head angular acceleration relative to the head CG, respectively.

Data recorded by the headform was wirelessly transmitted to a computer for post-processing during testing. In order to confirm adequate coupling to the dentition, each IMM was placed on the dentition prior to test, and it was confirmed that it stayed in place without additional clamping.

The IMM was set in “lab mode” for all tests. The trigger threshold was 5g along any axis. Upon passing the trigger threshold, the mouthguard provides linear acceleration, angular acceleration, and angular velocity time traces for each axis as well as resultants at the head CG. Sampling rate for all data was 3200 Hz and the sensors had bandwidth >1kHz. Accelerometer and gyroscope ranges were +/- 200g and +/- 35rad/s, respectively. Time of recorded data was 50 ms with 10 ms pre- and 40ms post-trigger duration. See **Table A4-1**.

Sensor	Mount Type/Location	Primary Measurements, Battery Life, and Trigger Threshold	
Prevent Biometrics Prevent IMM	Head – Mouth (Mouthguard)	Processed linear acceleration, angular velocity, and angular acceleration along/about X, Y, and Z axes at head center of gravity (CG) in the J211 coordinate system. Summary linear acceleration, angular acceleration, and angular velocity at head CG Sampling rate: 3200 Hz (all data) Battery Life: ~10 Hours (Depends on activity level and wear of the device) Trigger: 5g in any single axis Linear accelerometer range: +/- 200g along each axis Gyroscope range: +/- 35rad/s about each axis Linear/angular acceleration and angular velocity duration and pre-trigger duration: 50 ms, 10 ms	 Source: https://preventbiometrics.com/the-system/

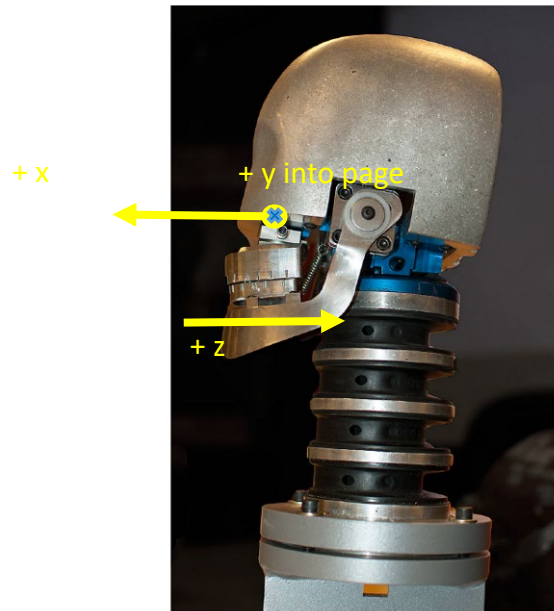
Table A4-1. The Prevent Impact Monitor Mouthguard (IMM).

DATA ANALYSIS

All linear accelerometer and angular acceleration data at the head CG was filtered at 200 Hz cutoff frequency during post-processing based on the first two harmonics in a head impact being in the range of 40-160Hz and containing 90-99% of the signal of interest (Bartsch et al. 2014)

Sled accelerometer, pendulum force, and pendulum accelerometer data were filtered at CFC 60 as defined in SAE J211 standards (SAE 1995). Coordinate system at the head CG followed SAE J1734 with +x out the front of the face, +y out the

right side of the head, and +z out the bottom of the head as defined by the right-hand rule (**FigureA4-4**). The internal coordinate system for the IMM was the same as the laboratory reference.



FigureA4-4. Head center of gravity (CG) coordinate system of MLSH without synthetic skin. Coordinate system of NOCSAE is same as MLSH.

Percent Error between the laboratory (e.g., reference) instrumentation and the IMM was calculated to compare measurements from the IMM and kinematics at the head CG. Percent Error is given by the following:

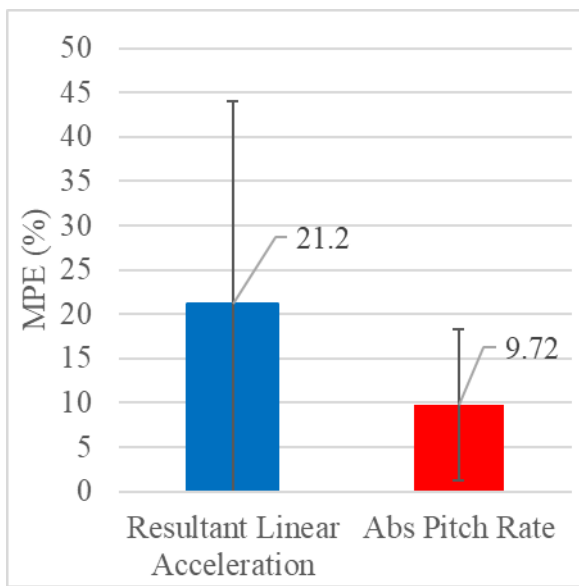
$$\text{Percent Error} = \frac{|Kinematic_{IMM} - Kinematic_{Reference}|}{Kinematic_{Reference}} * 100 \quad (1)$$

Where $Kinematic_{IMM}$ and $Kinematic_{Reference}$ are the kinematic quantity being compared of the IMM and laboratory grade sensors, respectively. Kinematics compared between the IMM, and laboratory reference instrumentation included peak resultant linear acceleration (PRLA) at the head CG and peak absolute value of pitch rate (PPR). Summary statistics of Percent Error between IMM and reference were calculated for all tests. In addition, summary statistics were calculated for each of the test conditions (e.g., direct versus indirect, exposure condition). Lines of best fit between IMM and reference peak kinematic data were constructed across all tests for each combination of loading condition and ATD headform to determine how well IMM kinematics linearly correlated with ATD kinematics.

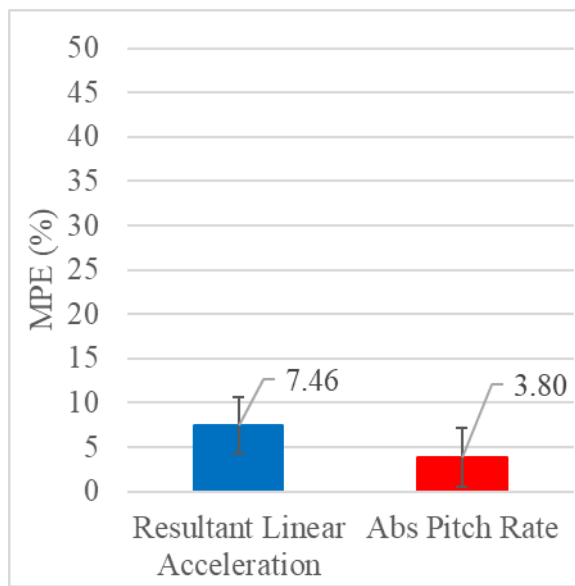
RESULTS

For the resultant linear acceleration, tests with the NOCSAE headform had smaller MPE ($21\% \pm 23\%$ MLSH versus $7\% \pm 3\%$ NOCSAE) than tests with the MLSH for indirect impacts (**FigureA4-5**). For the angular rate, tests with the NOCSAE headform also had smaller MPE ($10\% \pm 9\%$ MLSH versus $4\% \pm 3\%$ NOCSAE) than tests with the MLSH for indirect impacts (**FigureA4-5**). Similarly, the tests with the NOCSAE headform had smaller MPE in resultant linear acceleration ($27.6\% \pm 22.0\%$ MLSH versus $10.6\% \pm 9.03\%$ NOCSAE) than tests with the MLSH for direct impacts (**FigureA4-6**). Additionally, tests with the NOCSAE headform also had smaller MPE in angular rate ($14\% \pm 10\%$ MLSH versus $1\% \pm 1\%$ NOCSAE) than tests with the MLSH for direct impacts (**FigureA4-6**).

The IMM had PRLA and PPR that showed a nearly linear response compared with the NOCSAE PRLA (slope: 1.04 R^2 : 0.96) and PPR (slope: 0.96, R^2 : 0.98) values across all indirect impacts (see corresponding lines of best fit in **FigureA4-7** and **FigureA4-8**). The IMM had PPR values that were nearly linear compared with the MLSH PPR (slope: 1.1, R^2 : 0.90) values across all indirect impacts (see corresponding line of best fit in **FigureA4-8**). PRLA values from the IMM did not have a linear response compared with the MLSH PRLA values (slope: 1.23, R^2 : 0.68; **FigureA4-7**).

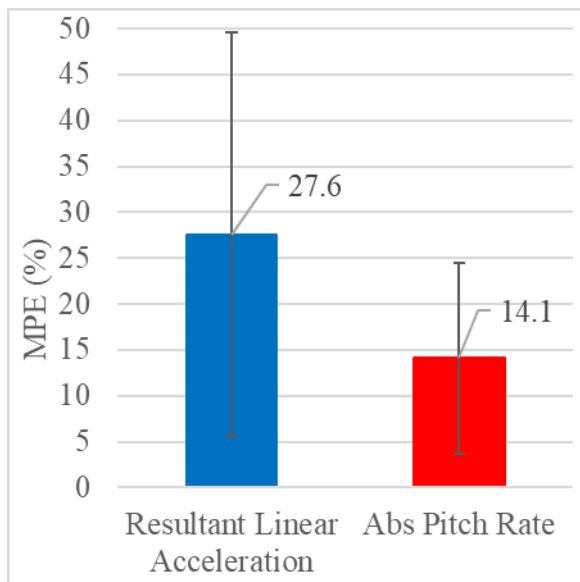


(A)

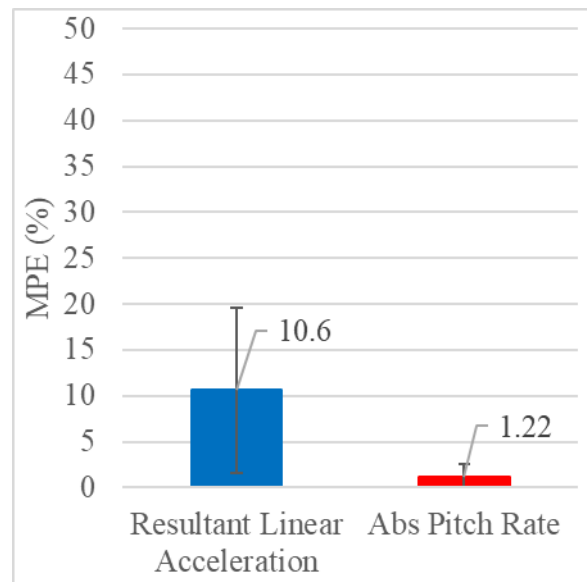


(B)

FigureA4-5. Mean percent error between anthropometric test device and instrumented mouthguard peak kinematics for (A) the Mandible Load Sensing Headform (MLSH), and (B) the National Operating Committee on Standards for Athletic Equipment (NOCSAE) during indirect impacts.

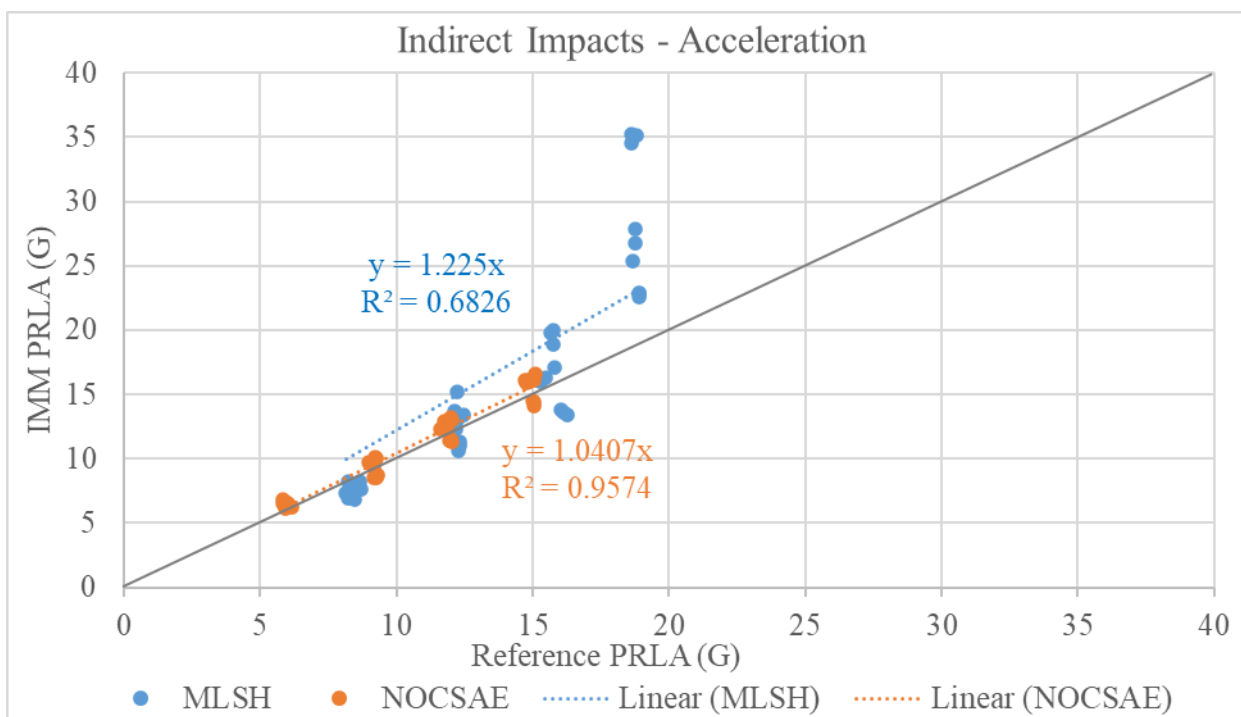


(A)

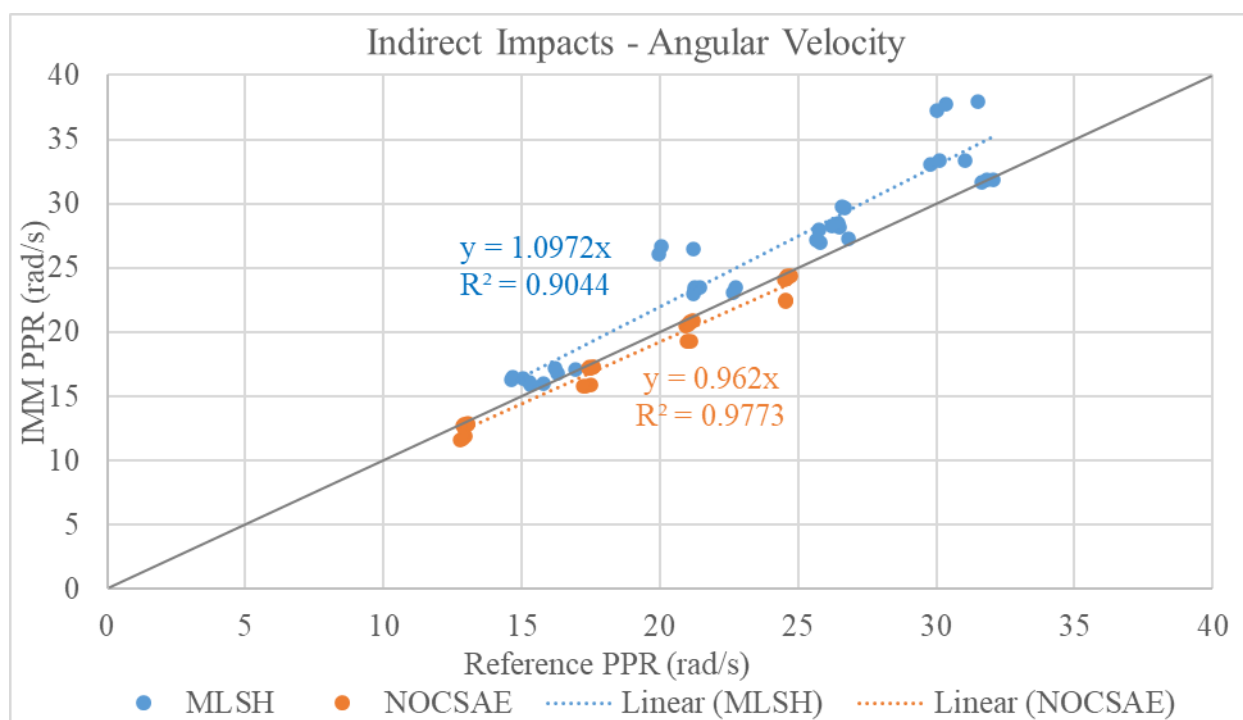


(B)

FigureA4-6. Mean percent error between anthropometric test device and instrumented mouthguard peak kinematics for (A) the Mandible Load Sensing Headform (MLSH), and (B) the National Operating Committee on Standards for Athletic Equipment (NOCSAE) during direct impacts.

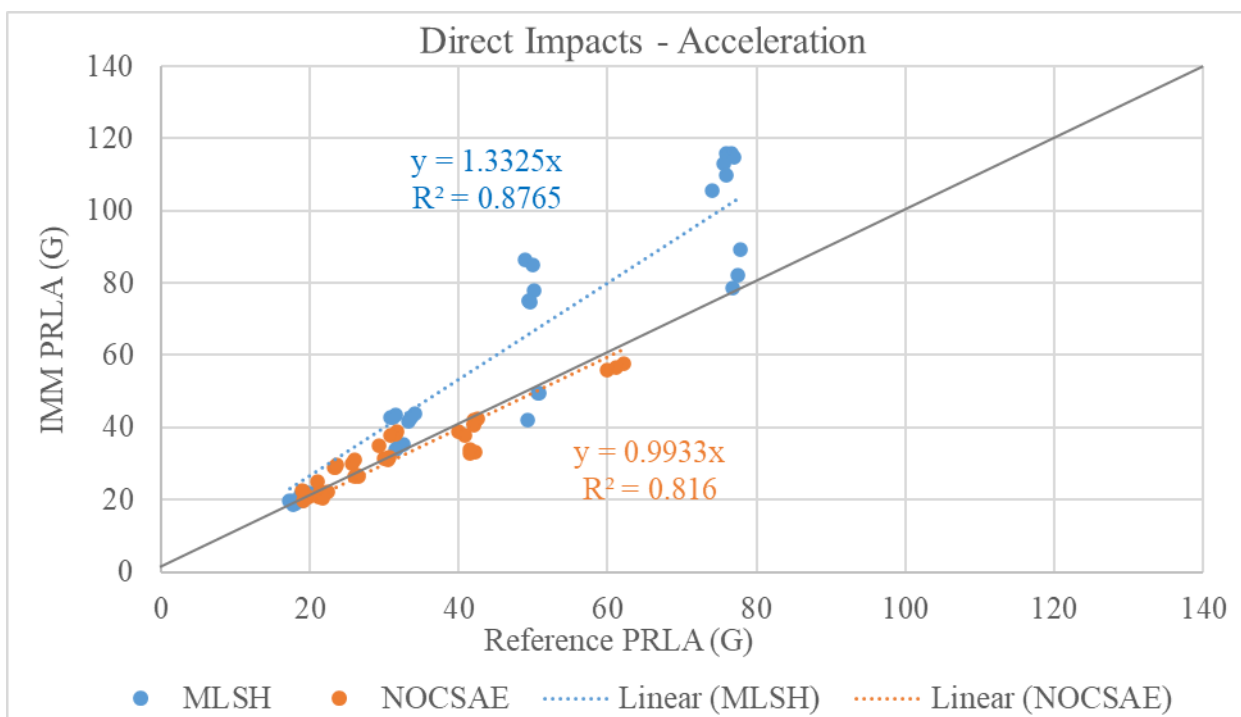


FigureA4-7. Scatter plot of anthropometric test device and instrumented mouthguard peak linear acceleration during indirect impacts.

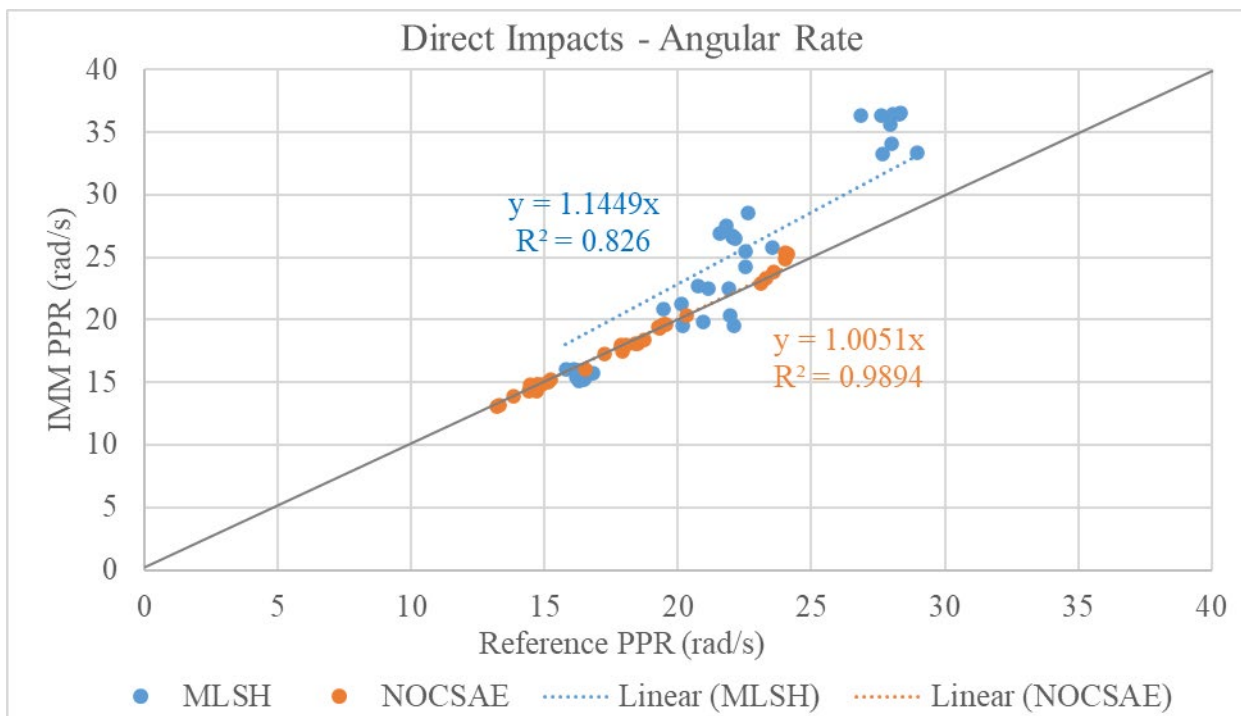


FigureA4-8. Scatter plot of anthropometric test device and instrumented mouthguard peak angular velocity during indirect impacts.

The IMM had PRLA and PPR that showed a nearly linear response compared with the NOCSAE PRLA (slope: 0.99 R^2 : 0.82) and PPR (slope: 1.01, R^2 : 0.99) values across all direct impacts (see corresponding lines of best fit in **FigureA4-9** and **FigureA4-10**). The IMM had PPR values that were nearly linear compared with the MLSH PPR (slope: 1.14, R^2 : 0.83) values across all direct impacts (see corresponding line of best fit in **FigureA4-10**). PRLA values from the IMM did not have a linear response compared with the MLSH PRLA values (slope: 1.33, R^2 : 0.88; FigureA4- 9).



FigureA4-9. Scatter plot of anthropometric test device and instrumented mouthguard peak linear acceleration during direct impacts.



FigureA4-10. Scatter plot of anthropometric test device and instrumented mouthguard peak angular velocity during direct impacts.

DISCUSSION

Agreement between IMM and ATD kinematic data was better for NOCSAE than for the MLSH (**FiguresA4-5, 6, 7, 8, 9 and 10**). This finding is in agreement with previous literature (Camarillo et al. 2013; Siegmund et al. 2014; Kuo et al. 2016; Liu et al. 2020; Kieffer et al. 2020; Jones et al. 2022). The NOCSAE headform does not have a moving mandible and represents the typical condition for on-field data (Bartsch et al. 2020, Jones et al. 2022). The MLSH mandible is not fully constrained and can rotate freely as the headform rotates. Kuo (2016) evaluated a mouthguard formed using ATD dentition molds under direct impacts for three mandible conditions including no mandible, clenched mandible, and

“unconstrained” mandible. Linear regression analyses between ATD and mouthguard peak resultant linear acceleration and peak resultant angular velocity across all impacts resulted in slopes of 0.99 and 1.01, respectively, for the clenched mandible condition; this finding is similar to what has been shown in the current study (**FiguresA4-7, 8, 9, and 10**). Similar to the MLSH results for the present study, Kuo et al. (2016) stated that for the “unconstrained” mandible condition, mouthguard peak resultant angular velocity and peak resultant linear acceleration generally over predicted corresponding reference measurements. Further, Camarillo et al. (2013) and Kieffer (2020) discuss similar findings to the present study using a clenched mandible under direct impacts (excellent agreement between sensor and ATD kinematic measurements). Finally, Siegmund et al. (2014) reported absolute errors of $12 \pm 10\%$ across multiple impact sites and severities when testing a mouthguard under direct impacts using the MLSH. It should be noted that the MLSH was not developed to test instrumented mouthguards under real-world conditions, but rather, it was developed to examine jaw condyle loading in chin impacts. A lack of biofidelic response should be expected, and our results confirm those of others who have seen a response in tests that fails to mimic the real-world response.

MPE between NOCSAE and IMM PRLA and PPR had smaller percentages and standard deviations than that of the MLSH (**FiguresA4-5 and 6**). The NOCSAE headform also had better linear correlation with IMM PRLA than that of the MLSH for both indirect and direct impacts (**FiguresA4-7, 8, 9, and 10**). The linearity between IMM and ATD highlighted the effects of the unconstrained mandible in the MLSH through increased slope compared to the expected value as well as the lower R^2 values.

The R^2 values in the present study were lower than prior work. In comparison with prior work evaluating instrumented mouthguards, the present study did not use custom molded mouthguards but instead used boil-and-bite versions that were manually fit to the ATD. The R^2 lower values were determined to be due to sub-optimal mouthguard fit. When reviewing individual IMM response compared with the ATDs, substantial differences in slope and R^2 were observed for some of the IMMs indicating differences in fit (Appendix D). As with the data considered across sensor groups, the IMM and NOCSAE headform kinematics were more similar than the MLSH and IMM kinematics. When reviewing data broken down by sensor group, it was determined that one of the sensor groups was uniquely non-correlated with the corresponding NOCSAE PRLA values during direct impacts (FigureA4-12). This finding likely contributes to the lower R^2 values when considered across sensor groups. Issues fitting the mouthguard could cause sensors to be oriented incorrectly or cause the mouthguard to slip on the teeth. Previous work has shown that the orientation of a sensor affects agreement with measurements at the head CG (Brown, et al. 2021). These findings highlight the potential variability in mouthguard response due to poor fit. Further, the findings stress the importance of ensuring a good fit both during laboratory testing and during field studies. On the plus side, these results provide a newly discovered error bound on sub-optimal mouthguard fitting that could help provide uncertainty limits for on-field data collected with poorly fitted mouthguards.

There were limitations in this study. First, only one impact direction was considered during this study. Agreement between ATD and mouthguard data could be directionally sensitive, and our results could be different if considering other directions. The main point of the study, that a clenched mandible ATD gives better agreement with IMM kinematics than a freely rotating ATD, has been shown in previous literature to be true across several impact locations and it is anticipated that this study would show the same finding if more impact locations were tested. Second, this study used one size for both the MLSH and NOCSAE headforms. Headforms of different sizes are likely to affect the ability of IMM internal algorithms to accurately predict kinematics at the head CG, although the algorithms are more sensitive to misalignment of sensitive axes' orientations.

SUMMARY OF SPECIFIC AIMS

The achievements across all four (4) Specific Aims can be summarized as follows: an accurate head impact monitoring mouthguard system was used to quantify concussive head impacts in military training environments, and that these impacts occurred in the ranges of approximately 50g to 100g, at the level of a sledgehammer hit to the head and above.

The data collected also showed that there are myriad military training environments, ranging from combatives, boxing, athletics parachute, special warfare, etc. where the IMM system was worn by active duty servicemembers and military service academy cadets. The scale of the data collected will also serve DoD decision makers well. These data are the first accurate head impact norms collected in military service members.

In order to address concussion and TBI in training environments, future work will rely upon these data to identify the riskiest environments, as well as environments where training occurs with sub-injurious head impact workloads. In order to get the best utilization and protection for the service member, DoD should define specification(s) for head impact monitoring as soon as is practical. Only when DoD defines specification(s) will technology be available for acquisition for the benefit of the most possible service members.

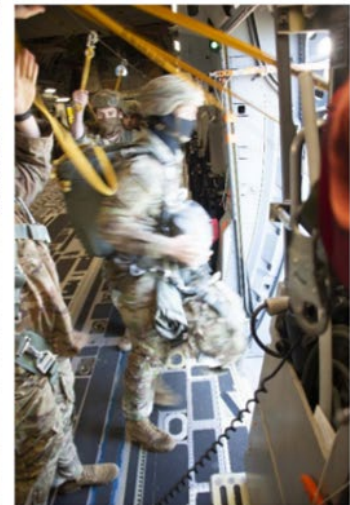
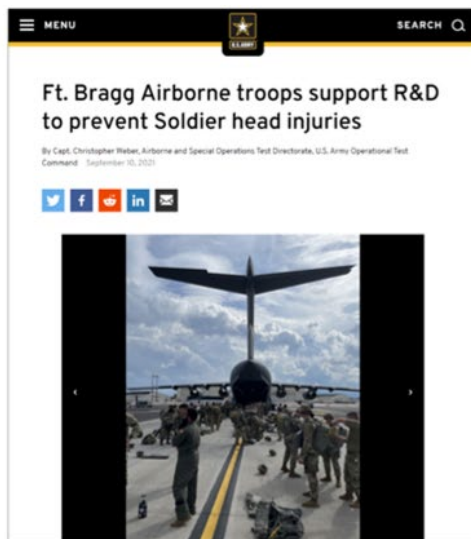
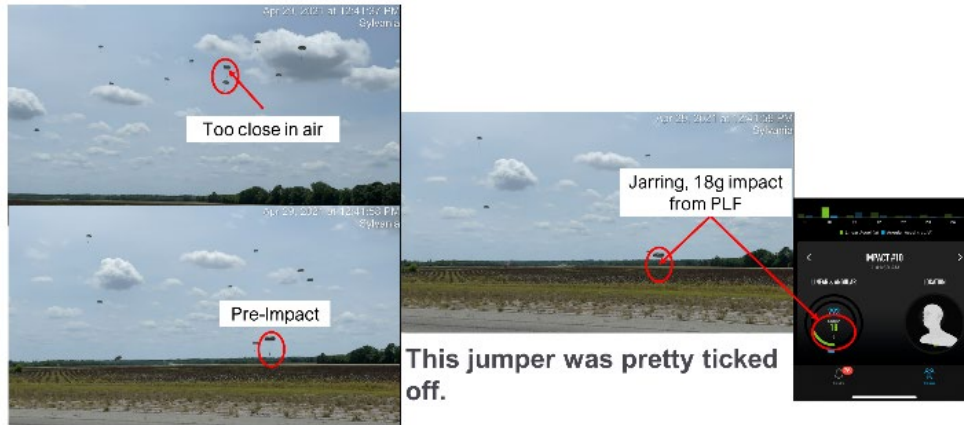
Acknowledgments

The United States Department of Defense provided funding for this study through Broad Agency Agreement W81XWH-17-1-0019. The Author appreciates assistance from Dr. Richard Shoge at the US Medical Research and Development Command (MRDC) Ft. Detrick, MD, Mr. Tyler Rooks, Ms. Jill Emerson and Ms. Stephanie Moon at USAARL, as well as data collection and prior publication help from colleagues at the Cleveland Clinic, US Military Academy, US Air Force Academy, Medical College of Wisconsin and USAARL. This work would not have been possible without the buy-in from active duty US Military sites, including the US Army 82nd Airborne 2nd Brigade Combat Team “2nd BCT FALCONS” – Fort Bragg, US Army Airborne Special Operations Test Directorate (ABNSOTD)-Fort Bragg, US Army Advanced Airborne School-Fort Bragg, USAF Combatives Center of Excellence, USAF 22nd Training Squadron “SERE”- Fairchild AFB, USAF 421st Combat Training Squadron “RAVENS”-Joint Base McGuire-Dix-Lakehurst, 371st Special Operations Combat Training Squadron “DAGRE”-Hurlburt Field, 351st Special Warfare Training Squadron “PJ SCHOOLHOUSE”-Kirtland AFB, and Georgia Army Air National Guard “POMPOC”.

4. Impact

The Project's impact has been demonstrated through successful completion of the aims of the project as awarded in FY2015 and those aims in the 2020 revised SOW. This project is already impacting DoD beyond the stated aims and funded objectives. Active-duty units at locations like the 421st Combat Training Squadron (USAF, Joint Base McGuire Dix Lakehurst) and the 82nd Airborne 2nd Brigade Combat Team are now using the IMM tech to maintain a mission ready force.

1. Hard PLF on grass – too close in-air



This BAA Cooperative Agreement has led to impactful work that continues beyond the project period of performance. Data from the field have confirmed about 5% of paratroopers at Ft. Bragg sustain 'check engine' impacts during jumps. These data are being used to evaluate outfitting an entire brigade to provide line leaders with information on good versus bad PLF's, as well as identify service members who may need a 'check engine' concussion assessment. Beyond that, there is a real possibility that the monitoring efforts will expand to other brigades and sites with more inexperienced service members, like Fort Benning.

In addition to the original Project Team, there were multiple collaborations that emerged during the BAA period of performance that will further cement the impact of the work into the future. This includes proposals already submitted to expand synergies between Prevent Biometrics (PI Adam Bartsch PhD PE) and blast/blunt exposure work ongoing with WRAIR (Dr. Walter Carr), USAARL (Mr. Tyler Rooks), NHRC (Dr. Karen Kelly) and ARL (Drs. Eric Wetzel & Tom Plaisted). For example, in ongoing work, the 82nd Airborne are using ARL's Rate-Activated Tether System (RATS) and Prevent Biometrics' IMM System during training missions to improve blunt force knowledge and protection. In ongoing studies led by WRAIR of the Marine Weapons Training Battalion Quantico blast/blunt exposures have been quantified. Under the direction of Dr. Kelly at NHRC, Naval Boat Crews in San Diego have been investigating blunt/blast forces during training exercises. To date, this work has determined that a 50g impact is like a sledgehammer hit to the head and merits a 'check engine' assessment (Bartsch et al. 2020 IRCOB, Bartsch et al. 2020 ABME). And the same IMM tech

that monitors 'check engine' impacts have been repurposed in related work to count blast exposures at 99% accuracy (Bartsch 2021 IFBIC Blast Meeting).

Finally, collaborations have been established that are leading into new and interesting uses of the IMM technology in the blast community. Work is underway through CRADA's with the Naval Warfare Research Center, the Walter Reed Army Institute of Research, the Canadian Defence Force and New Zealand Defence Force to monitor blunt and blast forces using the IMM. The data obtained under the BAA Project, and the collaborations developed during the Project period will continue to have a positive impact on both blunt and blast force monitoring in the US and Allied forces for years to come.

This BAA project has finally answered some of the most basic questions around how hard, how often and where DoD service members sustain blunt forces in training environments. By monitoring exposures in real-time and referring those sustaining 'check engine' head impacts each day to on-site medics/corpsmen, the service member can now have precisely targeted medical oversight. This should reduce time between injury and clinical assessment, more appropriately targeted clinical interventions, and account for service members who ought to be checked out and clinically assessed. Long-term, health and well-being of the service member should only increase.

5. Changes/Problems

1. None to report, and the Project completed its final period of performance on 31OCT21.

6. Products, Inventions, Patent Applications and/or Licenses (Appendix has flat copies of the most pertinent publications)

The results have been disseminated to communities of interest as follows:

Poster/Podium	Site/Journal Name	Location	Date	Authors	Citation/Details/Title
Poster	MHSRS Symposium	Orlando, FL	Aug-18	Adam Bartsch ¹ , Rajiv Dama ¹ , Sergey Samorezov ² , Edward Benzel ² , Vincent Miele ³ , Alok Shah ⁴ , John Humm ⁴ , Michael McCrea ⁴ , Brian Stemper ⁴ ¹ Prevent Biometrics, ² Cleveland Clinic, ³ University of Pittsburgh, ⁴ Medical College of Wisconsin & Zablocki VA Center	Laboratory and On-field Results of Athlete Head Impact Monitoring
Podium	BMES Annual Meeting	Atlanta, GA	Oct-18	Adam Bartsch ¹ , Rajiv Dama ¹ , Jay Alberts ² , Edward Benzel ² , Sergey Samorezov ² , Vincent Miele ³ , John Humm ⁴ , Alok Shah ⁴ , Michael McCrea ⁴ , Brian Stemper ⁴ ¹ Prevent Biometrics, ² Cleveland Clinic, ³ University of Pittsburgh Medical Center, ⁴ Medical College of Wisconsin & Zablocki VA	Head Impacts in American Football and Combat Sports
Poster	National Neurotrauma Society	Pittsburgh, PA	Jul-19	Adam Bartsch ¹ , Jay Alberts ² , Ed Benzel ² , Sergey Samorezov ² , Vince Miele ³ , Mike McCrea ⁴ , Alok Shah ⁴ , Brian Stemper ⁴ ¹ Prevent Biometrics, ² Cleveland Clinic, ³ University of Pittsburgh, ⁴ Medical College of Wisconsin & Zablocki VA Center	Athlete "No-Go" Head Impacts to the Side and Rear Cause Visible Deficits
Podium	SB3C Conference	Seven Springs, PA	Jun-19	Adam Bartsch PhD PE (1), Vincent Miele MD (2), Edward Benzel MD (3), Jay Alberts PhD (3), Alok Shah (4), John Humm (4), Brian Stemper PhD (4), Michael McCrea PhD (4) (1) Prevent Biometrics Minneapolis, MN, USA (2) Department of Neurosurgery University of Pittsburgh Medical Center Pittsburgh, PA, USA (3) Neurological Institute Cleveland Clinic Cleveland, OH, USA (4) Medical College of Wisconsin, Zablocki VA Center and Marquette University Milwaukee, WI, USA	Estimates of High-risk Single and Cumulative Head Impact Doses in American Football
Poster	AAN Concussion Meeting	Indianapolis, IN	Jul-19	Adam Bartsch PhD PE (1), Vincent Miele MD (2), Edward Benzel MD (3), Sergey Samorezov MS (3) (1) Prevent Biometrics (2) Department of Neurosurgery University of Pittsburgh Medical Center (3) Neurological Institute Cleveland Clinic	High Energy American Football Head Impacts to the Side and Rear are more Damaging than Impacts to the Front

Poster	ASB/ISB Meeting	Calgary, AB	Jul-19	Adam Bartsch ¹ , Vincent Miele ² , Edward Benzel ³ 1Prevent Biometrics, Minneapolis, MN, USA 2Department of Neurosurgery, University of Pittsburgh, Pittsburgh, PA, USA 3Department of Neurosurgery, Cleveland Clinic, Cleveland, OH, USA	Machine Learning can Improve Accurate Monitoring of Head Impacts in American Football
Poster	MHSRS Symposium	Orlando, FL	Aug-19	Adam Bartsch ¹ , Jay Alberts ² , Ed Benzel ² , Sergey Samorezov ² , Vince Miele ³ , Mike McCrea ⁴ , Alok Shah ⁴ , Brian Stemper ⁴ 1Prevent Biometrics, 2Cleveland Clinic, 3University of Pittsburgh, 4Medical College of Wisconsin & Zablocki VA Center	Athlete "No-Go" Head Impacts to the Side and Rear Cause Visible Deficits
Podium	BMES Annual Meeting	Philadelphia, PA	Oct-19	Adam Bartsch ¹ , Songbai Ji ² , Wei Zhao ² , Laurel Ng ³ , Jay Alberts ⁴ , Edward Benzel ⁴ , Jason Cruickshank ⁴ , Elizabeth Jansen ⁴ , Sergey Samorezov ⁴ , Vincent Miele ⁵ , John Humm ⁶ , Alok Shah ⁶ , Michael McCrea ⁶ , Brian Stemper ⁶ 1Prevent Biometrics, 2Worcester Polytechnic Institute, 3L3 Communications, 4Cleveland Clinic, 5University of Pittsburgh Medical Center, 6Medical College of Wisconsin	Athlete Spatial and Temporal Impact Characteristics Influence Brain Strain Response
Poster	BMES Annual Meeting	Philadelphia, PA	Oct-19	Adam Bartsch ¹ , Jay Alberts ² , Edward Benzel ² , Jason Cruickshank ² , Elizabeth Jansen ² , Sergey Samorezov ² , Vincent Miele ³ , John Humm ⁴ , Alok Shah ⁴ , Michael McCrea ⁴ , Brian Stemper ⁴ 1Prevent Biometrics, 2Cleveland Clinic, 3University of Pittsburgh Medical Center, 4Medical College of Wisconsin	Engineering Characteristics of Single and Cumulative Head Impacts in Sport that Cause Functional Impairment
Podium	FAA Triennial Conference	Atlantic City, NJ	Oct-19	Adam Bartsch ¹ , Jay Alberts ² , Ed Benzel ² , Sergey Samorezov ² , Vince Miele ³ 1 Prevent Biometrics, 2 Cleveland Clinic, 3 University of Pittsburgh Medical Center	High Energy Impacts to the Side/Rear Cause "No-Go" Observations
Podium	IMECE Annual Meeting	Salt Lake City, UT	Nov-19	Adam Bartsch ¹ , Jay Alberts ² , Ed Benzel ² , Sergey Samorezov ² , Vince Miele ³ 1Prevent Biometrics, 2Cleveland Clinic, 3University of Pittsburgh Medical Center	Engineering Characteristics of High Energy Head Impacts in Athletes that Cause Functional Impairment
Podium	IMECE Annual Meeting	Salt Lake City, UT	Nov-19	Reuben Kraft ¹ , Ritika Menghani ¹ , Adam Bartsch ² 1The Pennsylvania State University, 2Prevent Biometrics	Sensor Enabled Cloud Based Computational Modeling of the Brain
Podium	NCA TBI Conference	Washington, DC	Mar-20	Adam Bartsch PhD PE, Brian Stemper PhD, Alok Shah MS, Michael McCrea PhD, Daniel Hedin MS, Jerry McGinty DPT, Ken Cameron PhD, Tyler Rooks MS	Quantifying Head Impacts and Blast Exposures Using and Impact Monitoring Mouthguard

Poster	MHSRS Symposium	Orlando, FL	Aug-20	Adam Bartsch, PhD PE1, Alok Shah, MS2, Brian Stemper, PhD2, Mike McCrea, PhD2, Jon Jackson, MD3, Jerry McGinty, DPT3 1 Prevent Biometrics, Minneapolis, MN 2 Medical College of Wisconsin, Milwaukee, WI 3 United States Air Force Academy, Air Force Academy, CO	Monitoring Head Impacts in United States Air Force Academy Athletes: Lacrosse
Poster	MHSRS Symposium	Orlando, FL	Aug-20	Adam Bartsch, PhD PE1, Alok Shah, MS2, Brian Stemper, PhD2, Mike McCrea, PhD2, Jon Jackson, MD3, Jerry McGinty, DPT3 1 Prevent Biometrics, Minneapolis, MN 2 Medical College of Wisconsin, Milwaukee, WI 3 United States Air Force Academy, Air Force Academy, CO	Monitoring Head Impacts in United States Air Force Academy Athletes: American Football
Poster	MHSRS Symposium	Orlando, FL	Aug-20	Brian D. Stemper, PhD1,2, Lindsay D. Nelson, PhD1, Alok S. Shah, MS1,2, Alexa Wild, BA1, Daniel Huber, MPH1, Jason Mihalik, PhD3, Steven Rowson, PhD4, Jay Alberts, PhD5, Paul Gibson, MS6, Daniel Hedin, PhD6, Adam Bartsch, PhD7, Michael McCrea, PhD1,2 1 Medical College of Wisconsin, Milwaukee, WI, 2 Clement J. Zablocki Veterans Affairs Medical Center, Milwaukee, WI, 3 University of North Carolina, Chapel Hill, NC, 4 Virginia Tech, Blacksburg, VA, 5 Cleveland Clinic, Cleveland, OH, 6 AME Corp, Minneapolis, MN, 7 Prevent Biometrics, Minneapolis, MN	Relationship Between Subconcussive Head Impact Exposure and Clinical Changes in Non-Concussed High School and NCAA Division III Football Athletes
Poster	MHSRS Symposium	Orlando, FL	Aug-20	Adam Bartsch, PhD PE1 1 Prevent Biometrics, Minneapolis, MN	MEASURING HEAD IMPACT DOSE AND BLAST EXPOSURE USING A SINGLE BODY WORN SENSOR
Poster	MHSRS Symposium	Orlando, FL	Aug-20	Adam Bartsch, PhD PE1, Dan Hedin2, Alok Shah, MS3, Brian Stemper, PhD3, Mike McCrea, PhD3, Ken Cameron, PhD4, Megan Houston, PhD4, Jon Jackson, MD5, Jerry McGinty, DPT5, Tyler Rooks, MS6 1 Prevent Biometrics, Minneapolis, MN, 2 Advanced Medical Electronics, Maple Grove, MN, 3 Medical College of Wisconsin, Milwaukee, WI, 4 United States Military Academy, West Point, NY, 5 United States Air Force Academy, Air Force	Using an Impact Monitoring System and Video to Investigate "Obvious Performance Decrement" After Head Impacts

				Academy, CO, 6 United States Army Aeromedical Research Laboratory, Fort Rucker, AL	
Presentation	American Society of Testing and Materials (ASTM)	Virtual Meeting	Jun-21	Adam Bartsch, Prevent Biometrics	Head Impacts, Loss of Consciousness and Concussion in Helmeted and Non-Helmeted Sport
Presentation	82nd Airborne Standardization and Training Meeting	Ft Bragg, NC	May-21	Adam Bartsch, Prevent Biometrics	Head Impacts in Airborne Training
Presentation	US Air Force Combatives Center of Excellence	US Air Force Academy, CO	Jun-21	Adam Bartsch, Prevent Biometrics	Head Impacts in DoD Combatives and Boxing
Presentation	American Association of Neurological Surgeons (AANS)	Orlando, FL	Aug-21	Joseph Maroon ¹ , Adam Bartsch ² , 1UPMC 2Prevent Biometrics	The Future of Biometrics Monitoring for Concussion Management
Poster	MHSRS Symposium	Orlando, FL	Aug-21	Adam Bartsch, PhD PE ¹ , Dan Hedin ² , Alok Shah, MS ³ , Brian Stemper, PhD ³ , Mike McCrea, PhD ³ , Ken Cameron ⁴ , Megan Houston ⁴ , Jon Jackson, MD ⁵ , Jerry McGinty, DPT ⁵ , Tyler Rooks MS ⁶ 1 Prevent Biometrics 2AME 3MCW 4USMA 5USAFA 6USAARL	Using an Impact Monitoring System to Document "Check Engine" Head Impacts, MHSRS-21-03502
Poster	MHSRS Symposium	Orlando, FL	Aug-21	Adam Bartsch, PhD PE ¹ , Alok Shah, MS ² , Brian Stemper, PhD ² , Mike McCrea, PhD ² , Jon Jackson, MD ³ , Jerry McGinty, DPT ³ 1 Prevent Biometrics, Minneapolis, MN 2 Medical College of Wisconsin, Milwaukee, WI 3 United States Air Force Academy, Air Force Academy, CO	Monitoring Head Impacts in United States Air Force Academy Athletes: American Football, MHSRS-21-03583
Poster	MHSRS Symposium	Orlando, FL	Aug-21	Adam Bartsch, PhD PE ¹ , Ken Cameron ² , Megan Houston ² , Jon Jackson, MD ³ , Jerry McGinty, DPT ³ , Tyler Rooks MS ⁴ 1 Prevent Biometrics 2USMA 3USAFA 4USAARL	Normative Military Service Academy Head Impacts and 'Check Engine' Responses, MHSRS-21-03586
Presentation	5th International Forum on Blast Injury Countermeasures (IFBIC)	Tokyo, Japan (Virtual)	Sep-21	Adam Bartsch, Prevent Biometrics	Monitoring Blast and Blunt Forces Using a Single Body Worn Device

Presentation	International Olympic Committee World Conference on Prevention of Injury & Illness in Sport	Monaco	Nov-21	Adam Bartsch, Prevent Biometrics	The Sensible Use of Head-Impact Sensors in Sport, Workshop Session J, Room Poulenc 1, 1130-1230
Presentation	International Olympic Committee World Conference on Prevention of Injury & Illness in Sport	Monaco	Nov-21	Adam Bartsch, Prevent Biometrics	Making Sense of Head Impact Sensor Data, Workshop Session G, Room Bosio 2, 0930-1030
Presentation	International Olympic Committee World Conference on Prevention of Injury & Illness in Sport	Monaco	Nov-21	Adam Bartsch, Prevent Biometrics	The Top 1% of Head Impacts Cause 'Check Engine' Behavior, Free Communication Session E, Room Van Dongen, 1430-1520
Poster	International Olympic Committee World Conference on Prevention of Injury & Illness in Sport	Monaco	Nov-21	Adam Bartsch, Prevent Biometrics	The Top 1% of Head Impacts Cause a 'Check Engine' Response
Submitted	26th International Symposium on Military Aspects of Blast and Shock MABS26	Wollongong, Australia	Nov-22	Adam Bartsch, Prevent Biometrics	Measuring Head Impact Exposure and Counting Blast Exposure Using a Single Body Worn Device
Submitted	Journal of Trauma Acute Care	Orlando, FL	Jan-22	Adam Bartsch, Prevent Biometrics	Head Impacts in the Top 1% by Linear Acceleration and/or Workload Cause Concussion Signs and 'Check Engine' Responses in Civilian Athletes and Military Service Members
Presentation	6th International Forum on Blast Injury Countermeasures (IFBIC)	Washington, DC	May-22	Adam Bartsch ¹ , Maciej Skotak ² , Walter Carr ² , 1Prevent Biometrics 2WRAIR	Accurate quantification of the number and magnitude of overpressure events in highly dynamic military training environments
Poster	2022 NNS Neurotrauma Meeting	Atlanta, GA	Jun-22	Adam Bartsch, Prevent Biometrics	Concussion in civilians and military service members occurs in the top 1% of high-energy head impacts

Podium	MHSRS Symposium	Orlando, FL	Aug-22	Adam Bartsch ¹ , Tyler Rooks ² , 1Prevent Biometrics 2USAARL	Concussive and normative acceleration blunt force exposures in military training: objective data in support of a 'check engine' range in the military
Podium	MHSRS Symposium	Orlando, FL	Aug-22	Adam Bartsch ¹ , Maciej Skotak ² , Walter Carr ² , 1Prevent Biometrics 2WRAIR	Accurate quantification of the number and magnitude of overpressure events in highly dynamic military training environments
Podium	MHSRS Symposium	Orlando, FL	Aug-22	Anthony Dietz ¹ , Sean Wells-Rutherford ¹ , Adam Bartsch ² , Tyler Rooks ³ 1Paxauris, 2Prevent Biometrics, 3USAARL	Earplug Acceleration Measurement for Monitoring Impact and Blast Exposure

JOURNAL	LOCATION	DATE	AUTHORS	CITATION
Biomed Sci Instrum	Milwaukee, WI	Apr-19	Adam Bartsch PhD PE1, Vincent Miele MD2, Jay Alberts PhD3, Edward Benzel MD3, Alok Shah4, John Humm4, Brian Stemper PhD4, Michael McCrea PhD4 1Prevent Biometrics, 2University of Pittsburgh Medical Center, 3Cleveland Clinic, 4Medical College of Wisconsin, Zablocki VA Center & Marquette University	Bartsch A, Miele V, Alberts J, Benzel E, Shah A, Humm J, Stemper B, McCrea M. Estimates of High-Risk Single and Cumulative Head Impact Doses in American Football. Biomed Sci Instrum. Vol. 55(2): pp.224-229; April 2019.
ASTM Ice Hockey Summit	Denver, CO	May-19	Adam J. Bartsch,1 Edward C. Benzel,2 Sergey Samorezov,3 and Vincent J. Miele4 1Prevent Biometrics, 4530West 77th St., Suite 300, Minneapolis, MN 55435, USA 2Cleveland Clinic, Cleveland, OH, 44195, USA 3Lerner Research Institute, Cleveland Clinic, Cleveland, OH, 44195, USA 4Dept. of Neurosurgery, University of Pittsburgh Medical Center, Pittsburgh, PA, 15213, USA 5	Bartsch A, Benzel E, Samorezov S, Miele V. Summary of accurate and precise head impacts collected on-ice in hockey players. ASTM International STP Proceedings from the 2019 International Ice Hockey Summit, Denver, Colorado, pp. 1-29, 2020.
IEEE Annual Conference	Berlin, DE	Jul-19	Adam J. Bartsch, Member, IEEE, Daniel S. Hedin, Senior Member, IEEE, Paul L. Gibson, Member, IEEE, Vincent J. Miele, Edward C. Benzel, Jay L. Alberts, Sergey Samorezov, Alok Shah, Brian S. Stemper, Michael M. McCrea	Bartsch AJ, McCrea MM, Hedin DS, Gibson PL, Miele VJ, Benzel EC, Alberts JL, Samorezov S, Shah A, Stemper BS. Laboratory and On-field Data Collected by a Head Impact Monitoring Mouthguard. Annu Int Conf IEEE Eng Med Biol Soc. 2019 Jul;2019:2068-2072. doi: 10.1109/EMBC.2019.8856907. PMID: 31946308; PMCID: PMC7592723.
Military Medicine	Orlando, FL	Feb-20	Adam Bartsch, PhD, PE*; Rajiv Dama, PhD*; Jay Alberts, PhD†; Sergey Samorezov, MS‡; Edward Benzel, MD†; Vincent Miele, MD‡; Alok Shah, MS§; John Humm, MS§; Michael McCrea, PhD§; Brian Stemper, PhD§ *Prevent Biometrics †Cleveland Clinic, ‡University of Pittsburgh Medical Center, §Medical College of Wisconsin, Zablocki VA Center	Bartsch A, Dama R, Alberts J, Samorezov S, Benzel E, Miele V, Shah A, Humm J, McCrea M, Stemper B. Measuring Blunt Force Head Impacts in Athletes. Mil Med. 2020 Jan 7;185(Suppl 1):190-196. doi: 10.1093/milmed/usz334. PMID: 32074346; PMCID: PMC7029834.
IRCOBI Conference	Florence, IT	Sep-19	Wei Zhao, Adam Bartsch, Edward Benzel, Vincent Miele, Brian D. Stemper, Songbai Ji	Zhao W, Bartsch A, Benzel E, Miele V, Stemper B, Ji S. Regional Brain Injury Vulnerability from Two Finite Element Models of the Human Head. International

				Research Council on Biomechanics of Injury (IRCOBI), 2019, Florence, Italy [http://www.ircobi.org/wordpress/downloads/irc19/pdf-files/89.pdf]
IRCOBI Conference	Munich, DE (Virtual)	Sep-20	Adam Bartsch, Dan Hedin, Alok Shah, Mike McCrea, Brian Stemper	Bartsch A, Hedin D, Shah A, McCrea M, Stemper B. Engineering Characteristics of Athlete Head Impacts that Cause “Obvious Performance Decrement” (OPD). International Research Council on Biomechanics of Impact (IRCOBI), 2020. [http://www.ircobi.org/wordpress/downloads/irc20/pdf-files/105.pdf]
Annals of Biomedical Engineering	Atlanta, GA	Sep-20	ADAM J. BARTSCH ,1 DANIEL HEDIN,2 JAY ALBERTS,3 EDWARD C. BENZEL,3 JASON CRUICKSHANK,3 ROBERT S. GRAY,3 KENNETH CAMERON,4 MEGAN N. HOUSTON,4 TYLER ROOKS,5 GERALD MCGINTY,6 ERICK KOZLOWSKI,6 STEVEN ROWSON,7 JOSEPH C. MAROON,8 VINCENT J. MIELE,8 J. CHRIS ASHTON,9 GUNTER P. SIEGMUND,10 ALOK SHAH,11 MICHAEL MCCREA,11 and BRIAN STEMPER11 1Prevent Biometrics, Minneapolis, MN, USA; 2Advanced Medical Electronics, Maple Grove, MN, USA; 3Cleveland Clinic, Cleveland, OH, USA; 4Keller Army Community Hospital, West Point, NY, USA; 5United States Army Aeromedical Research Laboratory, Fort Rucker, AL, USA; 6United States Air Force Academy, Air Force Academy, CO, USA; 7Virginia Tech University, Blacksburg, VA, USA; 8University of Pittsburgh Medical Center, Pittsburgh, PA, USA; 9Twin Cities Orthopedics, Maple Grove, MN, USA; 10School of Kinesiology, University of British Columbia, Vancouver, BC, USA; and 11Medical College of Wisconsin, Wauwatosa, WI, USA	Bartsch AJ, Hedin D, Alberts J, Benzel EC, Cruickshank J, Gray RS, Cameron K, Houston MN, Rooks T, McGinty G, Kozlowski E, Rowson S, Maroon JC, Miele VJ, Ashton JC, Siegmund GP, Shah A, McCrea M, Stemper B. High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video. Ann Biomed Eng. 2020 Nov;48(11):2667-2677. doi: 10.1007/s10439-020-02640-8. Epub 2020 Oct 27. PMID: 33111969; PMCID: PMC7674260.
The Orthopaedic Journal of Sports Medicine	Cleveland, OH	Jul-21	Elizabeth Jansen1, Adam Bartsch2, Jay Alberts1, Sergey Samorezov1, 1Cleveland Clinic, 2Prevent Biometrics	Jansen AE, McGrath M, Samorezov S, Johnston J, Bartsch A, Alberts J. Characterizing Head Impact Exposure in Men and Women During Boxing and Mixed Martial Arts. Orthop J Sports Med. 2021 Dec 7;9(12):23259671211059815. doi: 10.1177/23259671211059815. PMID: 34901294; PMCID: PMC8664317.

Annals of Biomedical Engineering	Aberdeen, MD	Oct-21	Ryan Neice ¹ , Alex Lurski ¹ , Tom Plaisted ¹ , Eric Wetzel ¹ , Dwight Lowry ² , Adam Bartsch ³ 1US Army Research Laboratory, 2CCDC Data and Analysis Center, 3Prevent Biometrics	Neice RJ, Lurski AJ, Bartsch AJ, Plaisted TA, Lowry DS, Wetzel ED. An Experimental Platform Generating Simulated Blunt Impacts to the Head Due to Rearward Falls. Ann Biomed Eng. 2021 Oct;49(10):2886-2900. doi: 10.1007/s10439-021-02809-9. Epub 2021 Jun 28. PMID: 34184145.
IRCOBI Conference	Munich, DE	Sep-21	Gunter Siegmund ¹ , Ben Elkin ¹ , Stephanie Bonin ¹ , Allen Wu ¹ , Adam Bartsch ² 1MEA Forensic, 2Prevent Biometrics	Siegmund G, Elkin B, Bonin S, Wu A, Bartsch A. The Effects of Oversampling Non-Independent Data on Concussion Injury Risk Functions. International Research Council on Biomechanics of Injury (IRCOBI), 2021.

7. Participants & Other Collaborating Organizations

Name:	Adam Bartsch PhD PE
Project Role:	PI
Researcher Identifier (e.g., ORCID ID):	https://orcid.org/0000-0001-7913-0067
Nearest person month worked:	3
Contribution to Project:	Dr. Bartsch is the Principal Investigator
Funding Support:	Prevent Biometrics, NIH
Name:	Brian Stemper PhD
Project Role:	Co-Investigator
Researcher Identifier (e.g., ORCID ID):	
Nearest person month worked:	2
Contribution to Project:	Dr. Stemper is a co-Investigator at MCW
Funding Support:	MCW, NIH, DOD
Name:	Michael McCrea PhD
Project Role:	Co-Investigator
Researcher Identifier (e.g., ORCID ID):	
Nearest person month worked:	1
Contribution to Project:	Dr. McCrea is a Co-Investigator at MCW
Funding Support:	Prevent Biometrics, NIH
Name:	Jerry McGinty DPT
Project Role:	Co-Investigator
Researcher Identifier (e.g., ORCID ID):	
Nearest person month worked:	1
Contribution to Project:	Dr. McGinty is a Co- Investigator at USAFA
Funding Support:	USAFA
Name:	Jon Jackson MD
Project Role:	Co-Investigator
Researcher Identifier (e.g., ORCID ID):	
Nearest person month worked:	1
Contribution to Project:	Dr. Jackson is a Co- Investigator at USAFA

Funding Support:	USAFA
Name:	Ken Cameron PhD
Project Role:	Co-Investigator
Researcher Identifier (e.g., ORCID ID):	
Nearest person month worked:	1
Contribution to Project:	Dr. Cameron is a Co-Investigator at USMA
Funding Support:	USMA
Name:	Megan Houston PhD
Project Role:	Co-Investigator
Researcher Identifier (e.g., ORCID ID):	
Nearest person month worked:	1
Contribution to Project:	Dr. Houston is a Co-Investigator at USMA
Funding Support:	USMA
Name:	Tyler Rooks MS
Project Role:	Co-Investigator
Researcher Identifier (e.g., ORCID ID):	
Nearest person month worked:	1
Contribution to Project:	Mr. Rooks is a Co-Investigator at USAARL
Funding Support:	USAARL

8. Special Reporting Requirements

This project is required to report to FITBIR. Consent language has been included for this project and is incorporated by reference from the consents that have been collected on all subjects. As part of this process, and following the NCAA-DoD CARE model, global unique identifier (GUID) information has also been collected on subjects for de-identification and storage of data.

The FITBIR upload for Case Report Forms F2688 “VideoSensorConfirmation2019” and F2688 “HeadKinematicsEstimates2019” was completed on 01-DEC-2020. The upload was the first time FITBIR has included data for either one of these new case report forms, and a total of 2,031-time traces were uploaded.

It is anticipated that another 40,000+ time traces will be uploaded in Q1 and/or Q2 of 2022.

This project also requires a Quad Chart to be submitted. This Quad Chart (*QuadChart_W81XWH-17-1-0019_FINAL*) is shown below.

Human Head Impact Dose Measurement

Log Number: BA150149

Award Number: W81XWH-17-1-0019

PI: Bartsch, Adam

Org: Prevent Biometrics Award Amount: \$2,886,201

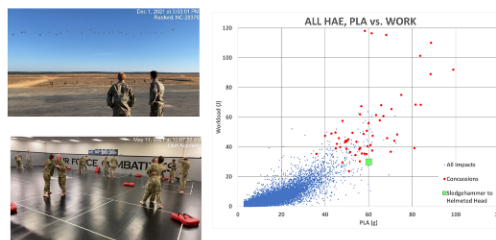


Study Aims

- Quantify head impact exposure during military training and sporting activities in military service cadets, civilian athletes, active-duty persons
- Correlate head impact exposure with onset and severity of concussion
- Correlate head impact exposure with clinical outcomes in absence of concussion
- Quantify head impact exposure during basic training and combat training activities
- Conduct laboratory testing to quantify impact monitor accuracy

Results

A total of 2,405 Impact Monitoring Mouthguards (IMM) were deployed during the period of performance across static and freefall parachute training, boxing, mixed martial arts and hand-to-hand combat training, as well as a variety of helmeted and non-helmeted sporting activities. A total of n=37 impacts with concussion and/or concussion signs were observed out of 46,949 head acceleration events (HAE). The HAE range was 1g-99g peak linear acceleration and 0.1J-120J workload.



The project has correlated head impacts with a 'check engine' behavior, concussion signs and diagnosed concussion. All 'check engine' impacts and concussions occurred in the top 1% of impacts by peak linear acceleration (PLA, g), impact energy workload (J), or both.

Timeline and Cost

Activities	CY	18	19	20	21
Head Impacts in Cadets					
Correlating to Clinical Data					
Combat Training Impacts					
Laboratory Tests USAARL					
Estimated Budget (\$K)		\$700	\$700	\$700	\$786

Updated: (25 Feb 2022)

Goals/Milestones

CY20 Goals – Ramp up Data Collection

- ☒ Usage of >1000 Impact Monitors with data collected
- ☒ Summary of ~7,500 head impacts with video
- ☒ First paratrooper data (POMPOC)

CY21 Goals – Expand Depth, Breadth of Findings

- ☒ MRDC IRB Approval for USAFA Cadet Boxing
- ☒ Increase combative enrollment to >100 users
- ☒ Increase combat training enrollees (low-g, parachute)
- ☒ Project completion

Comments/Challenges/Issues/Concerns

- None to Report

Budget Expenditure to Date

Projected Expenditure: \$2,886,201

Actual Expenditure: \$2,886,201

9. References

- Arbogast KB, McGinley AD, Master CL, Grady MF, Robinson RL, Zonfrillo MR. Cognitive rest and school-based recommendations following pediatric concussion: the need for primary care support tools. *Clin Pediatr (Phila)*. 2013 May;52(5):397-402. doi: 10.1177/0009922813478160. Epub 2013 Feb 27. PMID: 23447397.
- Asken BM, McCrea MA, Clugston JR, Snyder AR, Houck ZM, Bauer RM. "Playing Through It": Delayed Reporting and Removal From Athletic Activity After Concussion Predicts Prolonged Recovery. *J Athl Train*. 2016 Apr;51(4):329-35. doi: 10.4085/1062-6050-51.5.02. Epub 2016 Apr 25. PMID: 27111584; PMCID: PMC4874376.
- Assistant US Secretary of Defense, Warfighter Brain Health Directive 01OCT2018.
- Bartsch A, Samorezov S, Benzel E, Miele V, Brett D. Validation of an "Intelligent Mouthguard" Single Event Head Impact Dosimeter. *Stapp Car Crash J*. 2014 Nov;58:1-27. PMID: 26192948.
- Bartsch A, Miele V, Alberts J, Benzel E, Shah A, Humm J, Stemper B, McCrea M. Estimates of High-Risk Single and Cumulative Head Impact Doses in American Football. *Biomed Sci Instrum*. Vol. 55(2): pp.224-229; April 2019.
- Bartsch AJ, McCrea MM, Hedin DS, Gibson PL, Miele VJ, Benzel EC, Alberts JL, Samorezov S, Shah A, Stemper BS. Laboratory and On-field Data Collected by a Head Impact Monitoring Mouthguard. *Annu Int Conf IEEE Eng Med Biol Soc*. 2019 Jul;2019:2068-2072. doi: 10.1109/EMBC.2019.8856907. PMID: 31946308; PMCID: PMC7592723.
- Bartsch A, Benzel E, Samorezov S, Miele V. Summary of accurate and precise head impacts collected on-ice in hockey players. *ASTM International STP Proceedings from the 2019 International Ice Hockey Summit, Denver, Colorado*, pp. 1-29, 2020.
- Bartsch AJ, Hedin D, Alberts J, Benzel EC, Cruickshank J, Gray RS, Cameron K, Houston MN, Rooks T, McGinty G, Kozlowski E, Rowson S, Maroon JC, Miele VJ, Ashton JC, Siegmund GP, Shah A, McCrea M, Stemper B. High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video. *Ann Biomed Eng*. 2020 Nov;48(11):2667-2677. doi: 10.1007/s10439-020-02640-8. Epub 2020 Oct 27. PMID: 33111969; PMCID: PMC7674260.
- Bartsch A, Dama R, Alberts J, Samorezov S, Benzel E, Miele V, Shah A, Humm J, McCrea M, Stemper B. Measuring Blunt Force Head Impacts in Athletes. *Mil Med*. 2020 Jan 7;185(Suppl 1):190-196. doi: 10.1093/milmed/usz334. PMID: 32074346; PMCID: PMC7029834.
- Bartsch AJ. Head impacts in the top 1% by linear acceleration and/or workload cause concussion signs and 'check engine' responses in civilian athletes and military service members. *Journal of Trauma Acute Care*, 2022 (submitted).
- Baugh CM, Kroshus E, Daneshvar DH, Filali NA, Hiscox MJ, Glantz LH. Concussion management in United States college sports: compliance with National Collegiate Athletic Association concussion policy and areas for improvement. *Am J Sports Med*. 2015 Jan;43(1):47-56. doi: 10.1177/0363546514553090. Epub 2014 Oct 21. PMID: 25336600.
- Broglia SP, Schnebel B, Sosnoff JJ, Shin S, Fend X, He X, Zimmerman J. Biomechanical properties of concussions in high school football. *Med Sci Sports Exerc*. 2010 Nov;42(11):2064-71. doi: 10.1249/MSS.0b013e3181dd9156. PMID: 20351593; PMCID: PMC2943536.
- Broglia SP, Swartz EE, Crisco JJ, Cantu RC. In vivo biomechanical measurements of a football player's C6 spine fracture. *N Engl J Med*. 2011 Jul 21;365(3):279-81. doi: 10.1056/NEJMc1102689. PMID: 21774732.
- Brown, B. A., Daniel, R. W., Chancey, V. C., & Rooks, T. F. (2021, November). Parametric Evaluation of Head Center of Gravity Acceleration Error from Rigid Body Kinematics Assumptions Used in Environmental Sensors. In *ASME International Mechanical Engineering Congress and Exposition* (Vol. 85598, p. V005T05A074). American Society of Mechanical Engineers.
- 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 Sep;41(9):1939-49. doi: 10.1007/s10439-013-0801-y. Epub 2013 Apr 19. PMID: 23604848; PMCID: PMC3954756.
- D. B. Camarillo. Six Degree of Freedom Measurement of Human Mild Traumatic Brain Injury. *Virginia Tech Symposium on Head Acceleration Measurement Sensors*, Blacksburg, VA, USA, 2015.

Cantu RC. Second-impact syndrome. Clin Sports Med. 1998 Jan;17(1):37-44. doi: 10.1016/s0278-5919(05)70059-4. PMID: 9475969.

Centers for Disease Control and Prevention. (1997). Sports-related recurrent brain injuries--United States. *MMWR: Morbidity and mortality weekly report*, 46(10), 224-227. <https://www.cdc.gov/mmwr/preview/mmwrhtml/00046702.htm>

Centers for Disease Control. *Nonfatal sports and recreation related traumatic brain injuries among children and adolescents treated in emergency departments in the United States, 2001-2009*, MMWR 60(39), 2011.

Clark JM, Adanty K, Post A, Hoshizaki TB, Clissold J, McGoldrick A, Hill J, Annaidh AN, Gilchrist MD. Proposed injury thresholds for concussion in equestrian sports. J Sci Med Sport. 2020 Mar;23(3):222-236. doi: 10.1016/j.jsams.2019.10.006. Epub 2019 Oct 16. PMID: 31690492.

Crisco JJ, Wilcox BJ, Beckwith JG, Chu JJ, Duhaime AC, Rowson S, Duma SM, Maerlender AC, McAllister TW, Greenwald RM. Head impact exposure in collegiate football players. J Biomech. 2011 Oct 13;44(15):2673-8. doi: 10.1016/j.jbiomech.2011.08.003. Epub 2011 Aug 27. PMID: 21872862; PMCID: PMC3189296.

Daneshvar DH, Riley DO, Nowinski CJ, McKee AC, Stern RA, Cantu RC. Long-term consequences: effects on normal development profile after concussion. Phys Med Rehabil Clin N Am. 2011 Nov;22(4):683-700, ix. doi: 10.1016/j.pmr.2011.08.009. Epub 2011 Sep 23. PMID: 22050943; PMCID: PMC3208826.

Davis GA, Makdissi M, Bloomfield P, et al. international study of video review of concussion in professional sports. Br J Sports Med. 2019;53(20):1299-1304.

Defense and Veterans Brain Injury Center (DVBIC). (2020). *DOD Number for TBI 2000-2019 Q4*. Defense and Veterans Brain Injury Center (DVBIC). <https://health.mil/About-MHS/OASDHA/Defense-Health-Agency/Research-and-Development/Traumatic-Brain-Injury-Center-of-Excellence/DoD-TBI-Worldwide-Numbers>

Duma SM, Manoogian SJ, Bussone WR, Brolinson PG, Goforth MW, Donnenwerth JJ, Greenwald RM, Chu JJ, Crisco JJ. Analysis of real-time head accelerations in collegiate football players. Clin J Sport Med. 2005 Jan;15(1):3-8. doi: 10.1097/00042752-200501000-00002. PMID: 15654184.

Elbin RJ, Sufrinko A, Schatz P, French J, Henry L, Burkhardt S, Collins MW, Kontos AP. Removal from Play After Concussion and Recovery Time. Pediatrics. 2016 Sep;138(3): e20160910. doi: 10.1542/peds.2016-0910. PMID: 27573089; PMCID: PMC5005026.

Elliott MR, Margulies SS, Maltese MR, Arbogast KB. Accounting for sampling variability, injury under-reporting, and sensor error in concussion injury risk curves. J Biomech. 2015; 48:3059-65.

Filbern, T.M., Stitzel, J.D., et al. Head er Biomechanics in youth and collegiate female soccer. Journal of Biomechanics 128 (2021). <https://doi.org/10.1016/j.jbiomech.2021.110782>

Gehre, C., Gades, H., & Wernicke, P. (2009, June). Objective rating of signals using test and simulation responses. In *21st International Technical Conference on the Enhanced Safety of Vehicles Conference (ESV)* (pp. 09-0407).

Gessel LM, Fields SK, Collins CL, Dick RW, Comstock RD. Concussions among United States high school and collegiate athletes. J Athl Train. 2007 Oct-Dec;42(4):495-503. PMID: 18174937; PMCID: PMC2140075.

Greybe, D.G., Jones, C.M., Brown, M.R. et al. Comparison of head impact measurements via an instrumented mouthguard and an anthropometric testing device. Sports Eng 23, 12 (2020). <https://doi.org/10.1007/s12283-020-00324-z>

Gronwall, D., & Wrightson, P. (1975). Cumulative effect of concussion. *The Lancet*, 306(7943), 995-997. [https://doi.org/10.1016/S0140-6736\(75\)90288-3](https://doi.org/10.1016/S0140-6736(75)90288-3)

Gurdjian ES, Roberts VL, Thomas LM. Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury. J Trauma. 1966 Sep;6(5):600-4. doi: 10.1097/00005373-196609000-00005. PMID: 5928630.

Guskiewicz, K. M., Marshall, S. W., Bailes, J., McCrea, M., Cantu, R. C., Randolph, C., & Jordan, B. D. (2005). Association between recurrent concussion and late-life cognitive impairment in retired professional football players. *Neurosurgery*, 57(4), 719-726. <https://doi.org/10.1227/01.NEU.0000175725.75780.DD>

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; 61:1244-1252.

Guskiewicz KM, Mihalik JP. Biomechanics of sport concussion: quest for the elusive injury threshold. *Exerc Sport Sci Rev*. 2011 Jan;39(1):4-11. doi: 10.1097/JES.0b013e318201f53e. PMID: 21088602.

Hedin DS, Gibson PL, Bartsch AJ, Samorezov S. Development of a head impact monitoring "Intelligent Mouthguard". *Annu Int Conf IEEE Eng Med Biol Soc*. 2016 Aug;2016:2007-2009. doi: 10.1109/EMBC.2016.7591119. PMID: 28268724.

Higgins M, Halstead PD, Snyder-Mackler L, Barlow D. Measurement of impact acceleration: mouthpiece accelerometer versus helmet accelerometer. *J Athl Train*. 2007 Jan-Mar;42(1):5-10. Erratum in: *J Athl Train*. 2008 Jan-Mar;43(1):109. PMID: 17597937; PMCID: PMC1896070.

Holbourn, A. S. Mechanics of head injuries. *The Lancet* 09:438, 1943.

Institute of Medicine and National Research Council Committee on Sports Related Concussion in Youth. *Sports Related Concussions in Youth: Improving the Science, Changing the Culture*. Washington, DC: The National Academies Press, 2014.

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; 46:2310-2315.

Jones B, Tooby J, Weaving D, Till K, Own C, Begonia M, Stokes K, Rowson S, Phillips G, Hendricks S, Falvey E, Al-Dawoud M, Tierney G. Ready for impact? A validity and feasibility study of instrumented mouthguards (iMGs). Submitted to 2022 British Journal of Sports Medicine 27JAN2022, available online at <https://www.medrxiv.org/content/10.1101/2022.01.28.22270039v1.full.pdf>

Jordan BD. Chronic traumatic brain injury associated with boxing. *Semin.Neurol*. 2000; 20:179-185.

Kenny, R., Elez, M., Clansey, A. et al. Head Impact Exposure and Biomechanics in University Varsity Women's Soccer. *Ann Biomed Eng* (2022). <https://doi.org/10.1007/s10439-022-02914-3>

Kieffer, E. E., Begonia, M. T., Tyson, A. M., & Rowson, S. (2020). A two-phased approach to quantifying head impact sensor accuracy: in-laboratory and on-field assessments. *Annals of biomedical engineering*, 48(11), 2613-2625. <https://link.springer.com/article/10.1007/s10439-020-02647-1>

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 Mar;43(3):614-24. doi: 10.1177/0363546514560876. Epub 2014 Dec 22. PMID: 25535096.

Kontos AP and Collins MW. Concussion: A Clinical Profile Approach to Assessment and Treatment, pp. 28-30, Washington DC: American Psychological Association, [2018].

Kuo, C., Wu, L. C., Hammor, B. T., Luck, J. F., Cutcliffe, H. C., Lynall, R. C., ... & Camarillo, D. B. (2016). Effect of the mandible on mouthguard measurements of head kinematics. *Journal of biomechanics*, 49(9), 1845-1853. <https://doi.org/10.1016/j.jbiomech.2016.04.017>

Kuo C, Wu L, Loza J, Senif D, Anderson SC, Camarillo DB. Comparison of video-based and sensor-based head impact exposure. *PLoS ONE* 13(6): e0199238, 2018.

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

- Liu Y, Domel AG, Yousefsani SA, Kondic J, Grant G, Zeineh M, Camarillo DB. Validation and Comparison of Instrumented Mouthguards for Measuring Head Kinematics and Assessing Brain Deformation in Football Impacts. *Ann Biomed Eng.* 2020 Nov;48(11):2580-2598. doi: 10.1007/s10439-020-02629-3. Epub 2020 Sep 28. Erratum in: *Ann Biomed Eng.* 2021 Mar;49(3):1119-1120. PMID: 32989591.
- Martin, P. G., Hall, G. W., Crandall, J. R., & Pilkey, W. D. (1998). Measuring the acceleration of a rigid body. *Shock and Vibration*, 5(4), 211-224. <https://content.iospress.com/articles/shock-and-vibration/sav040>
- McAllister, T., & McCrea, M. (2017). Long-term cognitive and neuropsychiatric consequences of repetitive concussion and head-impact exposure. *Journal of athletic training*, 52(3), 309-317. <https://doi.org/10.4085/1062-6050-52.1.14>
- McCrea, M., Hammeke, T., Olsen, G., Leo, P., & Guskiewicz, K. (2004). Unreported concussion in high school football players: implications for prevention. *Clinical journal of sport medicine*, 14(1), 13-17. <https://europepmc.org/article/med/14712161>
- McCrea, M. A., Shah, A., Duma, S., Rowson, S., Harezlak, J., McAllister, T. W., Broglio, S. P., Giza, C. C., Goldman, J., Cameron, K. L., Houston, M. N., McGinty, G., Jackson, J. C., Guskiewicz, K., Mihalik, J. P., Brooks, M. A., Pasquina, P., & Stemper, B. D. (2021). Opportunities for Prevention of Concussion and Repetitive Head Impact Exposure in College Football Players. *JAMA Neurology*. <https://doi.org/10.1001/jamaneurol.2020.5193>
- 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 *British Journal of Sports Medicine* 2017;51:838-847.
- McKee, A. C., R. C. Cantu, C. J. Nowinski, E. T. Hedley- Whyte, B. E. Gavett, A. E. Budson, et al. Chronic traumatic encephalopathy in athletes: progressive tauopathy after repetitive head injury. *J. Neuropathol. Exp. Neurol.* 68:709–735, 2009.
- M. N. McPherson, W. J. Montelpare, M. Keightley, N. Reed, M. Sutherland, T. Taha, J. McAuliffe, J. R. Baker, J. Cubos, and B. Fought, “An Analysis of Head Impact Profiles and Safe Hockey Behaviors in Youth Hockey Players,” *Journal of ASTM International* 6, no. 10 (2009), <https://doi.org/10.1520/JAI101908>
- Meaney DF, Morrison B, Bass CD. The mechanics of traumatic brain injury: A review of what we know and what we need to know for reducing its societal burden. *J Biomech Eng.* 2014;136.
- Newman, J. A., N. Shewchenko, and E. Welbourne. A proposed new biomechanical head injury assessment function – the maximum power index. *Stapp Car Crash J.* 2000. <https://doi.org/10.4271/2000-01-SC16>.
- Omalu BI, Fitzsimmons RP, Hammers J, Bailes J. Chronic traumatic encephalopathy in a professional American wrestler. *J. Forensic Nurs.* 2010, ;6:130-136.
- O'Neil, M. E., Carlson, K., Storzbach, D., Brenner, L., Freeman, M., Quiñones, A., Motu'apuaka, M., Ensley, M., & Kansagara, D. (2013). *Complications of Mild Traumatic Brain Injury in Veterans and Military Personnel: A Systematic Review*. Department of Veterans Affairs (US). <https://pubmed.ncbi.nlm.nih.gov/24600749/>
- Patrick LM, Lissner HR, Gurdjian ES. Survival by design: head protection. In: *Proceedings of the 7th Stapp Car Crash Conference*. Warrendale, Society of Automotive Engineers, 1965, pp. 483–499.
- Plassman BL, Havlik RJ, Steffens DC, et al. Documented head injury in early adulthood and risk of Alzheimer's disease and other dementias. *Neurology* 2000, ;55:1158-1166.
- Pincemaille Y, Trosseille X, Mack P, Tarriere C. Some New Data Related to Human Tolerance Obtained from Volunteer Boxers. In: *Proceedings of the 33rd Stapp Car Crash Conference*, SAE Technical Paper No. 892435, November/1989.
- Pellman, E. J., D. C. Viano, A. M. Tucker, I. R. Casson, and J. F. Waeckerle. Concussion in professional football: reconstruction of game impacts and injuries. *Neurosurgery* 53:799–814, 2003.

Rowson S, Brolinson G, Goforth M, Dietter D, Duma S. Linear and angular head acceleration measurements in collegiate football. *J.Biomech.Eng.*2009, ;131:061016.

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 Mar;60(3):490-5; discussion 495-6. doi: 10.1227/01.NEU.0000249286.92255.7F. PMID: 17327793.

Siegmund, G. P., Guskiewicz, K. M., Marshall, S. W., DeMarco, A. L., & Bonin, S. J. (2014). A headform for testing helmet and mouthguard sensors that measure head impact severity in football players. *Annals of biomedical engineering*, 42(9), 1834-1845. <https://link.springer.com/article/10.1007/s10439-014-1052-2>

G. P. Siegmund, K. Guskiewicz, S. W. Marshall, A. L. DeMarco and S. J. Bonin. Validation of Wearable Sensors for Football Head Impacts. *Annals of Biomedical Engineering*, 2015.

Siegmund, G. P., Guskiewicz, K. M., Marshall, S. W., DeMarco, A. L., & Bonin, S. J. (2016). Laboratory validation of two wearable sensor systems for measuring head impact severity in football players. *Annals of biomedical engineering*, 44(4), 1257-1274. <https://pubmed.ncbi.nlm.nih.gov/26268586/>

A. M. Smith, M. J. Stuart, W. O. Roberts, D. W. Dodick, J. T. Finnoff, J. K. Jorgensen, and D. A. Krause, "Concussion in Ice Hockey: Current Gaps and Future Directions in an Objective Diagnosis," *Clinical Journal of Sport Medicine* 27, no. 5 (2017): 503–509.

Society of Automotive Engineers (SAE). (1995). Instrumentation for impact test—Part 1: Electronic instrumentation. *SAE Handbook*, 384-392.

Spiotta AM, Shin JH, Bartsch AJ, Benzel EC. Subconcussive impact in sports: A new era of awareness. *World Neurosurg.*2011, ;75:175-178.

Sporting Goods Manufacturers Association. *2010 Sports and Fitness Participation Report - The Complete Version*, 2010.

Stapp, J. P. Human tolerance to severe, abrupt deceleration. In: *Gravitational Stress in Aerospace Medicine*, edited by O. H. Gauer, and G. D. Zuidema. Boston: Little Brown and Co, 1961, pp. 165–188.

Stemper, B. D., Shah, A. S., Harezlak, J., Rowson, S., Duma, S., Mihalik, J. P., Rikken, L. D., Brooks, A., Cameron, K. L., Giza, C. C., Houston, M. N., Jackson, J., Posner, M. A., McGinty, G., DiFiori, J., Broglio, S. P., McAllister, T. W., McCrea, M., Hoy, A. M., ... Lintner, L. (2019). Repetitive Head Impact Exposure in College Football Following an NCAA Rule Change to Eliminate Two-A-Day Preseason Practices: A Study from the NCAA-DoD CARE Consortium. *Annals of Biomedical Engineering*, 47(10), 2073–2085. <https://doi.org/10.1007/s10439-019-02335-9>

Stemper, B. D., Shah, A. S., Harezlak, J., Rowson, S., Mihalik, J. P., Duma, S. M., Rikken, L. D., Brooks, A., Cameron, K. L., Campbell, D., DiFiori, J. P., Giza, C. C., Guskiewicz, K. M., Jackson, J., McGinty, G. T., Svoboda, S. J., McAllister, T. W., Broglio, S. P., & McCrea, M. (2019). Comparison of Head Impact Exposure Between Concussed Football Athletes and Matched Controls: Evidence for a Possible Second Mechanism of Sport-Related Concussion. *Annals of Biomedical Engineering*, 47(10), 2057–2072. <https://doi.org/10.1007/s10439-018-02136-6>

Stojsih S, Boitano M, Wilhelm M, Bir C. A prospective study of punch biomechanics and cognitive function for amateur boxers. *Br J Sports Med.* 2010 Aug;44(10):725-30. doi: 10.1136/bjsm.2008.052845. Epub 2008 Nov 19. PMID: 19019907.

Takhounts, E.G., Eppinger, R.H., Campbell, J.Q., Tannous, R.E., Power, E.D., Shook, L.S. (2008) On the Development of the SIMon Finite Element Head Model. *Stapp Car Crash Journal* 52: 107-133.

Talavage TM, Nauman EA, Breedlove EL, Yoruk U, Dye AE, Morigaki KE, Feuer H, Leverenz LJ. Functionally detected cognitive impairment in high school football players without clinically diagnosed concussion. *J Neurotrauma.* 2014 Feb 15;31(4):327-38. doi: 10.1089/neu.2010.1512. Epub 2013 Apr 11. PMID: 20883154; PMCID: PMC3922228.

TBICoE Research 2021, www.health.mil/Military-Health-Topics (accessed 09DEC2021).

The English Football Association. Professional Football Heading in Training Guidance, July/2021. Accessed on 01AUG021 at <https://www.thefa.com/about-football-association/field-research-study-findings>

Viano DC, Casson IR, Pellman EJ, Zhang L, King AI, Yang KH. Concussion in professional football: brain responses by finite element analysis: part 9. *Neurosurgery*. 2005 Nov;57(5):891-916; discussion 891-916. doi: 10.1227/01.neu.0000186950.54075.3b. PMID: 16284560.

Walilko TJ, Viano DC, Bir CA. Biomechanics of the head for Olympic boxer punches to the face. *Br J Sports Med*. 2005 Oct;39(10):710-9. doi: 10.1136/bjsm.2004.014126. PMID: 16183766; PMCID: PMC1725037.

Wu LC, Kuo C, Loza J, Kurt M, Laksari K, Yanez LZ, Senif D, Anderson SC, Miller LE, Urban JE, Stitzel JD, Camarillo DB. Detection of American Football Head Impacts Using Biomechanical Features and Support Vector Machine Classification. *Sci Rep*. 2017 Dec 21;8(1):855. doi: 10.1038/s41598-017-17864-3. PMID: 29321637; PMCID: PMC5762632.

Zuckerman SL, Elbin RJ, Sills AK, Crandall JR, Lessley DJ, Moran CM, Moran CD, Solomon GS. Concussions in the National Football League: the evolution of video review for assessing the frequency and reliability of visible signs. *Phys Sportsmed*. 2020 Nov;48(4):424-429. doi: 10.1080/00913847.2020.1731379. Epub 2020 Feb 23. PMID: 32067541.

