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

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14. ABSTRACT For the annual reporting period, the project has completed all the specific aims in the statement of work. Aim 1 for laboratory calibration has been finished for laboratory comparison to Hybrid III ATD, with the impact monitoring mouthguard system fitting a linear model versus Reference, of the form IMM=0.97*REF+1.2, with a correlation coefficient ($R^2=0.97$). Aim 2 was designed to correlate a dose-response relationship between head impact and clinical behavioral assessments for American football players in Wisconsin Division III football. SCAT3 symptom scores rose from pre-season to mid-season assessments, yet by season's end the symptoms had resolved. The SAC cognitive tests were not correlated to head impact exposure. Aim 3 – to quantify head impacts across a variety of military and athletic populations at the US Air Force Academy, US Military Academy, at MCW and in parachutists at the POMPOC National Guard event – was successfully completed, and ongoing data collection in the coming year will serve to increase the depth and breadth of this work. Over 2,000 head impacts were independently confirmed with video recordings of activity; the parachute work was the first time in recorded history that a parachutists' head impact was accurately measured and confirmed with time-synchronized video. In Aim 4 the project correlated measured head impacts with a 'check engine' behavior seen on video. This Aim was completed and resulted in the publication of the article in the Annals of Biomedical Engineering entitled "High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video." In the final year of the study we will continue this work and use the 'check engine' thresholds to ensure any subjects who sustain these types of impacts will be made known to appropriate clinical staff regardless of whether they self-report.					
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1. Preface

Based on feedback provided from the previous year's annual report, the report format has been improved, notably:

- The introduction contains a brief narrative to describe the work completed, and the report. This introduction will be used to generate an abstract for the report which is filed with DTIC.
- This annual report is written in alignment with the SOW on file for the study, dated 01-MAY-2020.
- Embedded pdf's have been removed; the submitted report is a flat file.
- For the Reportable Outcomes (i.e. abstracts, publications, etc.) a full tabular listing which includes all relevant authors, appropriate formatted citation, presentation/abstract details (Meeting Name, Meeting Location, Meeting Date) have been included.

2. Introduction

For the annual reporting period, the project has completed all the specific aims in the statement of work. Aim 1 for laboratory calibration has been finished for laboratory comparison to Hybrid III ATD, with the impact monitoring mouthguard system fitting a linear model versus Reference, of the form $IMM=0.97*REF+1.2$, with a correlation coefficient ($R^2=0.97$). This aim continues with planned PMHS testing to be conducted at USAARL during the upcoming year. Aim 2 was designed to correlate a dose-response relationship between head impact and clinical behavioral assessments for American football players in Wisconsin Division III football. The co-investigators at Medical College of Wisconsin have completed this aim with SCAT3 and SAC testing, although the results did not demonstrate a season-long correlation. The SCAT3 symptom scores rose from pre-season to mid-season assessments, yet by season's end the symptoms had resolved. The SAC cognitive tests were not correlated to head impact exposure. Aim 3 – to quantify head impacts across a variety of military and athletic populations at the US Air Force Academy, US Military Academy, at MCW and in parachutists at the POMPOC National Guard event – was successfully completed, and ongoing data collection in the coming year will serve to increase the depth and breadth of this work. Over 2,000 head impacts were independently confirmed with video recordings of activity; the parachute work was the first time in recorded history that a parachutists' head impact was accurately measured and confirmed with time-synchronized video. In Aim 4 the project correlated measured head impacts with a 'check engine' behavior seen on video. This Aim was completed and resulted in the publication of the article in the Annals of Biomedical Engineering entitled "*High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video.*" This paper was the first time in history that an individual high-gravity head impact measured in an athlete was correlated to a time-stamped collision on video, with clinician-confirmed visible signs of concussion for side and rear head impacts in the range of 40g to 100g. In the final year of the study we will continue this work and use the 'check engine' thresholds to ensure any subjects who sustain these types of impacts will be made known to appropriate clinical staff regardless of whether they self-report. Completion of this aim is the culmination of the funded work and having an extra year to increase the depth and breadth of knowledge around the 'check engine' threshold in additional military activities is very important.

3. Keywords

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

4. Accomplishments

EXECUTIVE SUMMARY

The project is essentially finished with the aims. The Study Team will use the remaining time thru OCT-2021 to expand the depth and breadth of normative head impact data more fully in new and important populations such as Cadet Boxing, Parachute, and additional athletic events. Also, the 'check engine' capability will be utilized in the newly issued IRB. See **Figure 1** for overview:

BA150149 DELIVERABLES

- Deployed 1425 impact monitoring mouthguards in n=1011 Active Duty, Military Service Academy and Civilian Personnel
- Collected >7,500 video verified head impacts
- Correlated >30 'check engine' impacts with visible signs from video
- Created Gen 2 Impact monitor, roughly 70% smaller with same performance

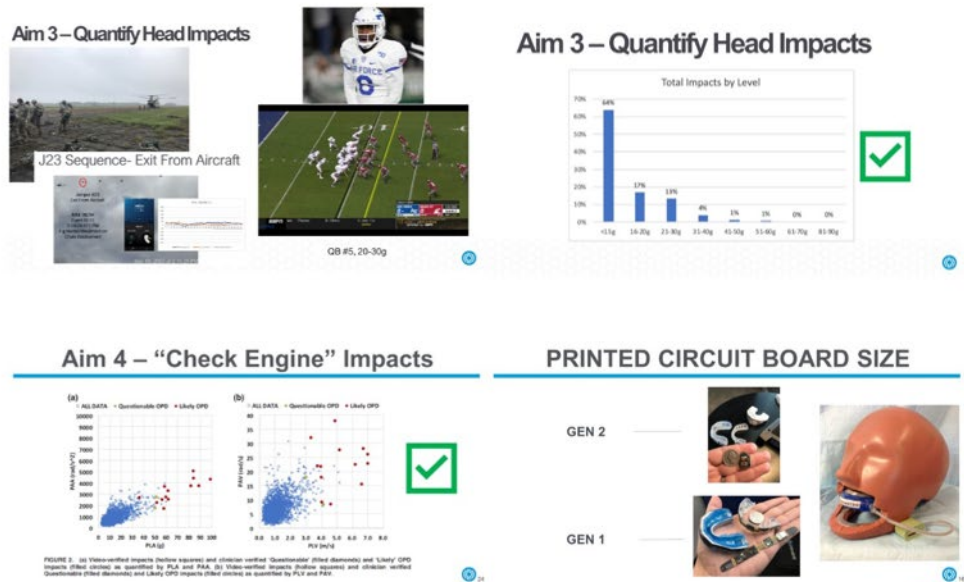


FIGURE 1. Deliverables in the past year include increasing deployment to 1011 users with 1425 impact monitoring mouthguard systems seven (7) research sites. Total impacts now exceed 7,500 with all being video verified. Further, about 30 high-gravity ‘check engine’ head impacts have been confirmed, and a new Gen 2 IMM has been developed that is about 70% smaller and has same accuracy and performance.

Despite challenges faced during the COVID-19 pandemic, the past year has seen continued progress toward successfully completing all Specific Aims within the time and budget allotted. In the past year, the following have been accomplished:

- Increased total project deployment to n=1011 users with a total of 1425 impact monitoring mouthguards at seven (7) sites: USAFA, USMA, POMPOC, Concordia College (WI), Lutheran College (WI), Carroll University (WI) and Carthage College (WI).
- Increased total collected data to 7,500 video-verified head impacts across Parachute, American football, rugby, and lacrosse.
- Identified a total of 30 large magnitude ‘check engine’ impacts where the subject has exhibited Obvious Performance Decrement (OPD)/” No-go” behavior on video.
 - These data comprise the first accurate dose-response ‘check-engine’ curve.
 - The goal for the rest of the study is to broaden this risk curve by both breadth and depth in more subjects, and more military-specific activities.
 - These data have been published in an article from SEP-2020 in the Annals of Biomedical Engineering by Bartsch et al. “High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video.”
- At POMPOC, collected first ever accurate head impact data in a Military parachute population during both chutes opening and in landing.

- Also collected first ever Military impact dose data on female military members
- Laboratory testing with crash test dummies is completed with the IMM comparing favorably to the gold standard REF head form, data fit a linear model and have an R^2 over 0.95.
- Collaborative relationships with several DoD and non-DoD sites and groups (Naval Health Research Center, Marine CQT, Canadian Special Forces, WRAIR, SOCOM, ARL, CADEX, Pennsylvania State University, POMPOC, Airborne Test Directorate) have been established. All these groups have a direct interest in the head impact data being collected and will serve to leverage the study into additional benefit for DoD research and soldier health and safety.
- First tranche of data – 2,000 impact traces – have been uploaded to FITBIR and data requests have already been submitted by Pennsylvania State University, L3-Harris, KTH Royal Institute and Worcester Polytechnic University.
 - This is the first FITBIR upload for the NINDS-DoD Common Data Elements working group from 2017.
 - Case Report forms F2689_HeadKinematicsEstimates2019 and F2699_VideoSensorConfirmation2019 were successfully approved and data uploaded with help from FITBIR staff (Dr. Olga Vovk and Ms. Mallisa Johnson).
- IRB approved at MRDC core protocol to clear path to more instrumentation of Active Duty and Military Service Academy Cadets who are at risk of sustaining head impacts. **Figure 2** shows the data inclusion and flow for BA150149 in the core protocol.
 - First target of opportunity is USAFA Cadet Boxing during Spring/2021.

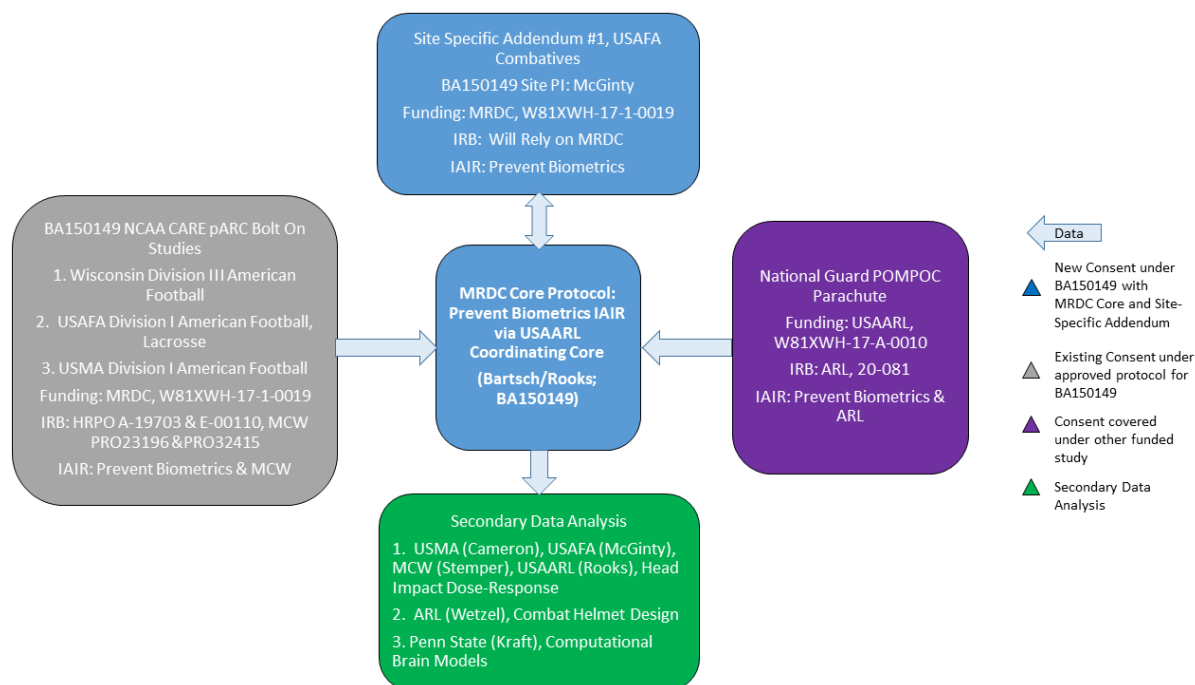


FIGURE 2. The MRDC Core IRB Delineates Data Collected by BA150149 (Blue), Data flowing into BA150149 from CARE ARC Add-on study (Gray), From the POMPOC Activity (Purple) and out to collaborators for further analysis toward other DoD projects (Green).

EXPLANATION OF SIGNIFICANT RESULTS

Please see the Appendix. The Significant Results were presented to MRDC by video conference on 09-FEB-2021. The slide deck with pertinent results, conclusions, and findings for BA150149 Specific Aims are detailed in this document.

5. Impact

We have completed the original aims of the project as awarded in FY2015 and have essentially completed the aims in the re-scope as of 2018 when the project was transferred from Cleveland Clinic to Prevent Biometrics. Along the way, we have established and strengthened working relationships amongst the Army (USAARL, ARL, USMA, Fort Gragg Airborne) as well as the Air Force (USAF) and National Guard (POMPOC, GAARG).

The capture of normative data in the National Guard paratrooper population and Military Service Academy personnel, along with confirmation of 'check engine' impacts compared with visible signs being captured on video is the first time in history that large impacts have been correlated to an impaired performance.

While the aims and objectives of the project have largely been met, we are not satisfied. In Spring/2020 the project was re-scoped to roughly double the number of subjects and to expand the depth and breadth of Active Duty and Military Service personnel. We are going to use the remaining time in 2021 to complete the study by increasing normative data collection in Military boxing, combatives and paratrooper activities. We are also going to leverage the 'check engine' capability to help identify those subjects who sustain the largest single impacts. By the end of the project period we anticipate well more than 2,000 impact monitors will have been used to collect data and we will have established normative head impact loads as well as defined where large head impacts occur most frequently for several military environments. This will lay the foundation for future efforts, to prospectively monitor the most at-risk military members, be it at the MSA's or in Military Training environments.

6. Changes/Problems

- The challenges presented by the COVID-19 pandemic caused some significant changes and re-organization to the project:
 1. The work at Medical College of Wisconsin was suspended as of FEB-2020 to present.
 2. The work at USMA was suspended as of FEB-2020 to DEC-2020.
 3. The work at USAFA was suspended from MAR-2020 to AUG-2020.
 4. Because of these changes, and the analysis of the results to date at MCW, it was decided that the MCW aim, to compute dose-response characteristics for SCAT3 and SAC testing in American football, was sufficiently complete and there was not a need to continue further data collection. This conclusion would have been reached with or without COVID-19, however, the additional time to analyze the data made the decision clearer.
 5. While USMA was unable to collect data through DEC-2020, they are eager to proceed with data collection in 2021 until the end of the project. Work on spinning up USMA cadet boxing, and USMA summer cadet training is underway. Letters of support and site-specific addenda are being procured currently.
 6. The effort at USAFA has continued relatively unabated and is accelerating. The site-specific addendum to add Cadet Boxing has been approved and is awaiting final signatures by the core protocol at MRDC IRB.
 7. After on-boarding USAFA Cadet boxing (~100-300 Spring subjects) it is anticipated that Club Boxing (n=30 men, n=10 women) and summer Cadet training will be submitted as additional SSA's.
 8. The study is actively engaged with Ft. Bragg Airborne Special Test Directorate to procure an Lol and SSA for their paratrooper population.
 9. The study is actively engaged with POMPOC through the Georgia Air National Guard for an effort with n=150 paratroopers at the end of April.
 10. While 2020 has presented challenges, the study has emerged as strong as ever and pivoted to collecting data on more Active Duty and Military Service Academy personnel engaged in Military Training. This is a logical next step for the study, given the success proven for quantifying head impacts in athletic training.

7. 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:

Type	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
Publication	Biomed Sci Instrum	Milwaukee, WI	Apr-19	Adam Bartsch PhD PE ¹ , Vincent Miele MD ² , Jay Alberts PhD ³ , Edward Benzel MD ³ , Alok Shah ⁴ , John Humm ⁴ , Brian Stemper PhD ⁴ , Michael McCrea PhD ⁴ ¹ Prevent Biometrics, ² University of Pittsburgh Medical Center, ³ Cleveland Clinic, ⁴ Medical College of Wisconsin, Zablocki VA Center & Marquette University	Estimates of High-risk Single and Cumulative Head Impact Doses in American Football [ISSN: 0067-8856 04 55 2]
Publication	ASTM Ice Hockey Summit	Denver, CO	May-19	Adam J. Bartsch, ¹ Edward C. Benzel, ² Sergey Samorezov, ³ and Vincent J. Miele ⁴ ¹ Prevent Biometrics, 4530 West 77th St., Suite 300, Minneapolis, MN 55435, USA https://orcid.org/0000-0001-7913-0067 ² Cleveland Clinic, Cleveland, OH, 44195, USA ³ Lerner Research Institute, Cleveland Clinic, Cleveland, OH, 44195, USA ⁴ Dept. of Neurosurgery, University of Pittsburgh Medical Center, Pittsburgh, PA, 15213, USA ⁵ ASTM Sixth International Symposium on Safety in Ice Hockey on May 13, 2019 in Denver, CO, USA.	Summary of accurate and precise head impacts collected on-ice in hockey players and how these data will be used to improve equipment and rules for safer play [in press]
Publication	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.	Laboratory and On-field Data Collected by a Head Impact Monitoring Mouthguard [doi: 10.1109/EMBC.2019.8856907]

				McCrea	
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 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	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
Publication	IRCOBI Conference	Florence, IT	Sep-19	Wei Zhao, Adam Bartsch, Edward Benzel, Vincent Miele, Brian D. Stemper, Songbai Ji	Regional Brain Injury Vulnerability from Two Finite Element Models of the Human Head [http://www.ircobi.org/wordpress/downloads/irc19/pdf-files/89.pdf]

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
Publication	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	Measuring Blunt Force Head Impacts in Athletes [doi: 10.1093/milmed/usz334]
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 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: 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 Academy, CO, 6 United States Army Aeromedical Research Laboratory, Fort Rucker, AL	Using an Impact Monitoring System and Video to Investigate "Obvious Performance Decrement" After Head Impacts
Publication	IRCOBI Conference	Munich, DE (Virtual)	Sep-20	Adam Bartsch, Dan Hedin, Alok Shah, Mike McCrea, Brian Stemper	Engineering Characteristics of Athlete Head Impacts that Cause "Obvious Performance Decrement" (OPD) [http://www.ircobi.org/wordpress/downloads/irc20/pdf-files/105.pdf]

Publication	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	High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video
Submitted for Publication	American Journal of Sports Medicine	Orlando, FL	Dec-20	Jay Alberts1, Sergey Samorezov1, Elizabeth Jansen1, Adam Bartsch2 1Cleveland Clinic, 2Prevent Biometrics	Characterizing Impact Exposure in Males and Females during Boxing and Mixed Martial Arts
Submitted for Publication	Journal of Biomechanics	Aberdeen, MD	Dec-20	Ryan Neice1, Alex Lurski1, Tom Plaisted1, Eric Wetzell1, Dwight Lowry2, Adam Bartsch3 1US Army Research Laboratory, 2CCDC Data and Analysis Center, 3Prevent Biometrics	An Experimental Platform Generating Simulated Blunt Impacts to the Head due to Rearward Falls
Submitted for Publication	IRCOBI Conference	Munich, DE	Sep-21	Adam Bartsch, Dan Hedin, Alok Shah, Mike McCrea, Brian Stemper	Analysis of Video Review and Impact Monitoring Data Collected from Dual-Instrumented American Football Players
Submitted for Publication	IRCOBI Conference	Munich, DE	Sep-21	Adam Bartsch, Dominic Calhoun, Lauren Hatch, Eric Kozlowski, Jon Jackson, Jerry McGinty	An Improved Impact Monitoring Mouthguard Reports Head Impacts in Elite American Football with 98% Sensitivity and 99% Specificity Versus Video

8. Participants & Other Collaborating Organizations

Name:	<i>Adam Bartsch PhD PE</i>
Project Role:	<i>PI</i>
Researcher Identifier (e.g. ORCID ID):	https://orcid.org/0000-0001-7913-0067
Nearest person month worked:	3
Contribution to Project:	<i>Dr. Bartsch is the Principal Investigator</i>
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

9. 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.

This project also requires a Quad Chart to be submitted. This Quad Chart (*QuadChart_W81XWH-17-1-0019_YEAR03*) 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



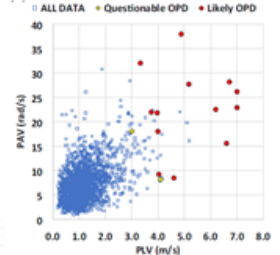
Study Aims

- 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

Approach

- The Study Team is monitoring head impacts at the US Air Force Academy (lacrosse, football), US Military Academy (rugby, football), four (4) Division III colleges in Wisconsin (football) and in Paratroopers (POMPOC). The Team is also working with USAARL to monitor head impacts in combat training activities and to conduct the laboratory testing. The total impact monitors shipped is 1425 out of 2832 allocated for the study based on current SoW (MAY-2020).

Aim 3 – Quantify Head Impacts



The project has correlated head impacts with a "check engine" behavior seen on video. This resulted in the publication of "High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video" in the 2020 ABME. This is the first time in history that a single high-gravity head impact has been successfully correlated to an observed collision on video, with clinician-confirmed visible signs of concussion for side and rear head impacts in the range of 40g to 100g.

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: (12 Feb 2021)

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)

Comments/Challenges/Issues/Concerns

- Risk of exhausting IMM supply prior to EoS 31-OCT; prioritize must-get data, populations
- Unknowns with COVID-19 pandemic; plan multiple data collection sessions in parallel with multiple sites

Budget Expenditure to Date

Projected Expenditure: \$2,443,000

Actual Expenditure: \$2,443,000 (thru Jan/2021)



BA150149, MOUTHGUARD PROJECT

Adam Bartsch PhD PE
Update with MRDC
Virtual
10-FEB-2021

OUTLINE

Outline

- Motivation
- Background
- Project Aims
- Results
- Next Steps



MOTIVATION

Motivation

- 80% of concussions go unidentified (Baugh et al., 2015 Neurotrauma)
- Military concussions occur in training → monitor DoD training impacts
 - Military Service Academies, Active-Duty
- Monitoring will identify most in need of a 'check engine' evaluation

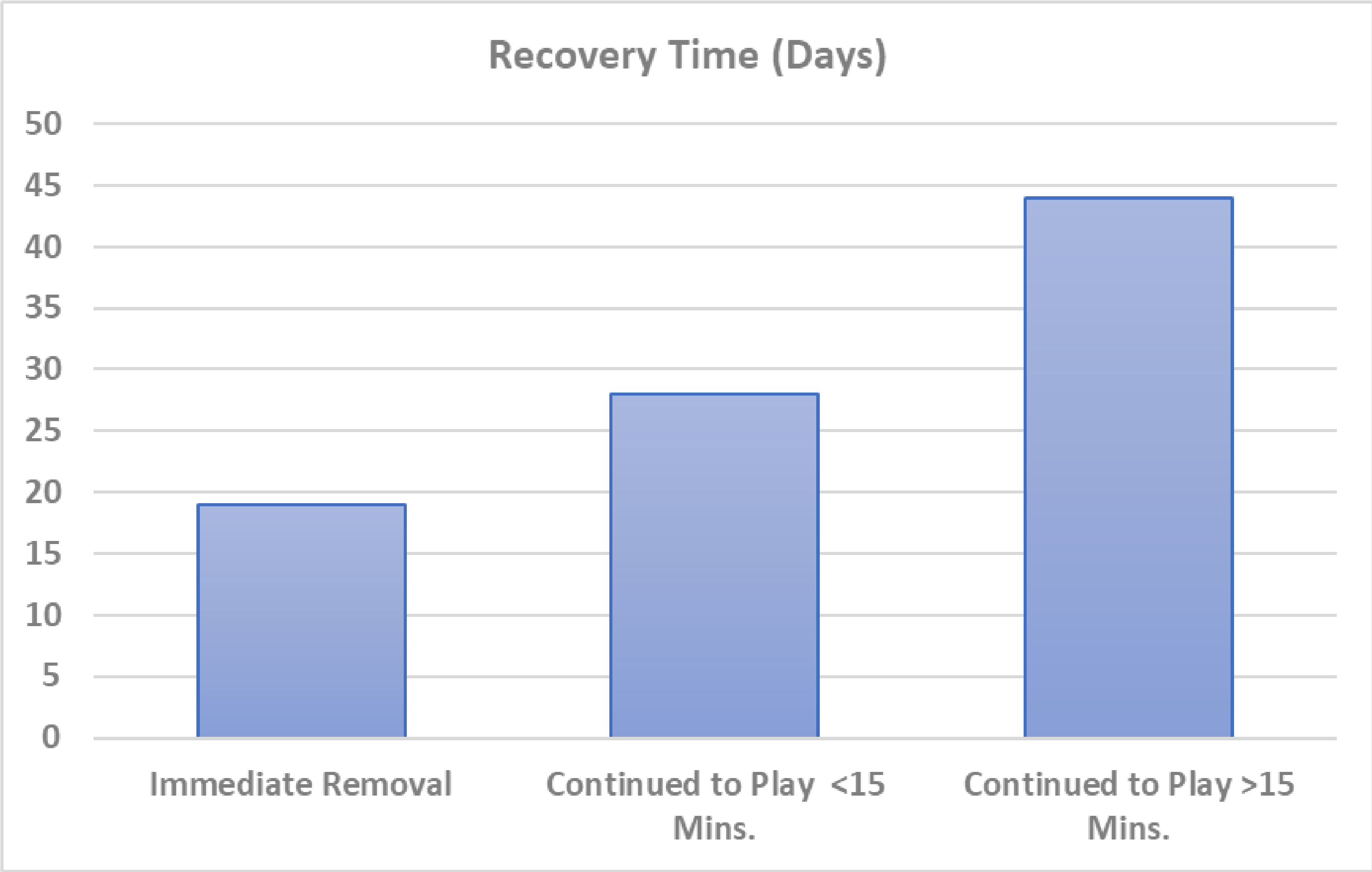


Motivation

- **DoD acquisition strategy calls for both head impact & blast sensors**
- Environmental Sensors in Training (ESiT)
 - Dr. Walter Carr (WRAIR), Dr. Gary Kamimori (USAMRMC), Tyler Rooks (USAARL)
- *“There was consensus that consolidated research on (blast and blunt force) sensors was practical... given that military service members experienced both types of threat... a combined sensor should be pursued”* – Dr. Walter Carr, Summary of 2017 Environmental Sensor Meeting, Snowbird, Utah



REMOVE FASTER, RECOVER FASTER



BACKGROUND

Background

- Awarded FY2015 to Cleveland Clinic w/USAARL
 - Develop, test, pilot a prototype impact monitor
 - N=130 subjects
- 2018 transfer to Prevent Biometrics, re-scoped
 - Added MCW, USAFA, USMA
 - N=1300 subjects, ending OCT-2020
- In 2020, re-scoped & NCE
 - More Active Duty
 - N=2800 subjects, ending OCT-2021

	YEAR		
	2015-2017	2018-2021	2022-2024
USERS	130	2800	42000
COST/USER	\$ 22,000	\$ 1,100	\$ 200



PROJECT AIMS

Aims

- ✓ 1. Laboratory accuracy
- 📈 2. Dose-response: head impacts v. clinical assessments
- ✓ 3. Quantify head impacts across military, athletics
- ✓ 4. Correlate head impacts with “check engine” response



RESULTS

IMPACT MONITORING PLATFORM

Team Charging Case

Solo Charging Case

Impact Monitoring Mouthguard



Web Data Portal

Mobile App



IMPACT MONITORING PLATFORM



Impact Monitoring Mouthguard (IMM)
Athlete Monitoring



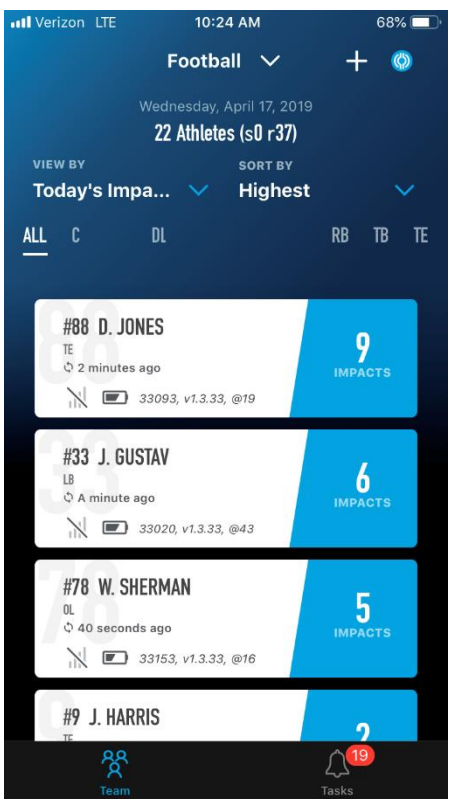
Solo case
Charging & Sanitization



Team Case
Centralized Charging for 27 IMM's



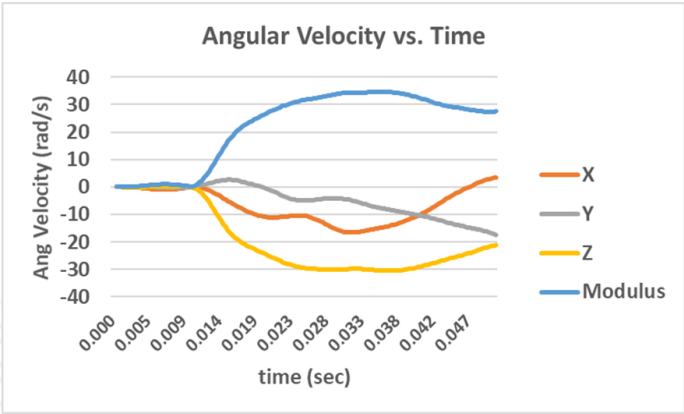
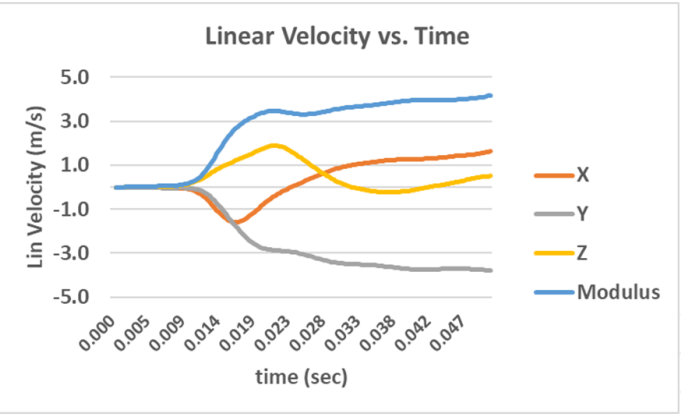
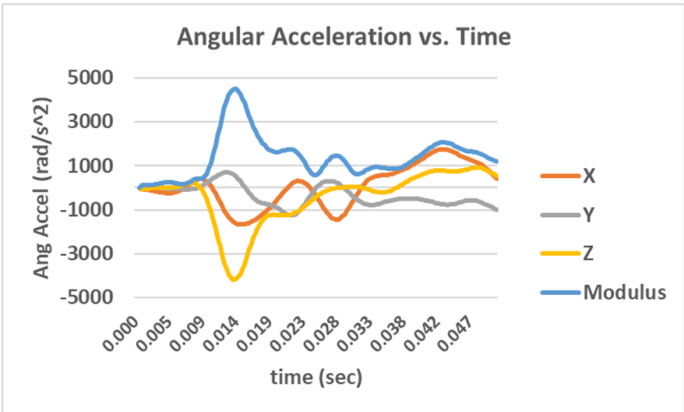
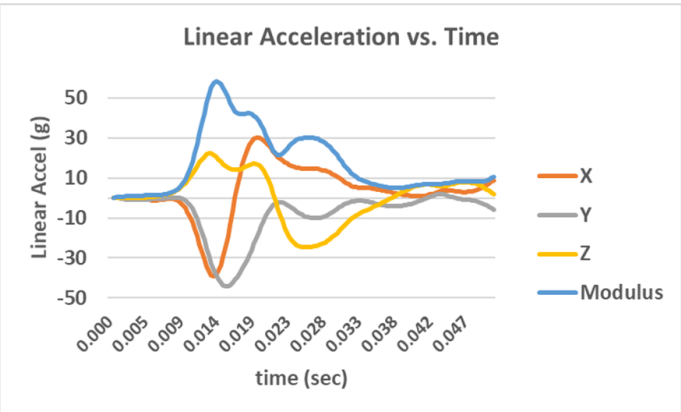
Team App
Data Sync, Team Management



IMPACT DATA

Spatial, Temporal Outputs:

- Linear Acceleration XYZ(t), Modulus(t)
- Angular Acceleration XYZ(t), Modulus(t)
- Linear Velocity XYZ(t), Modulus(t)
- Angular Velocity XYZ(t), Modulus(t)
- Impact Direction
- Impact Location



PRINTED CIRCUIT BOARD SIZE

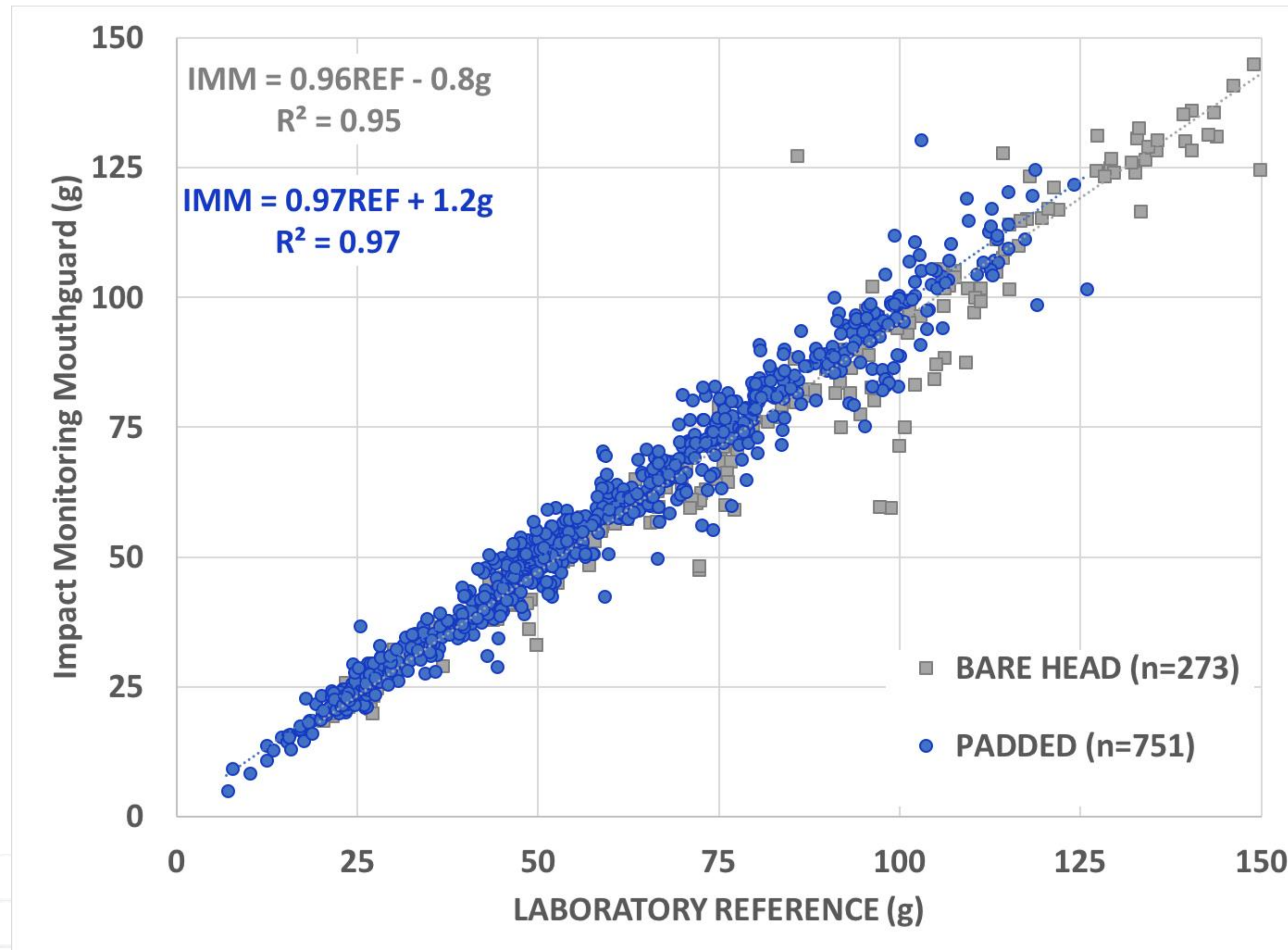
GEN 2



GEN 1

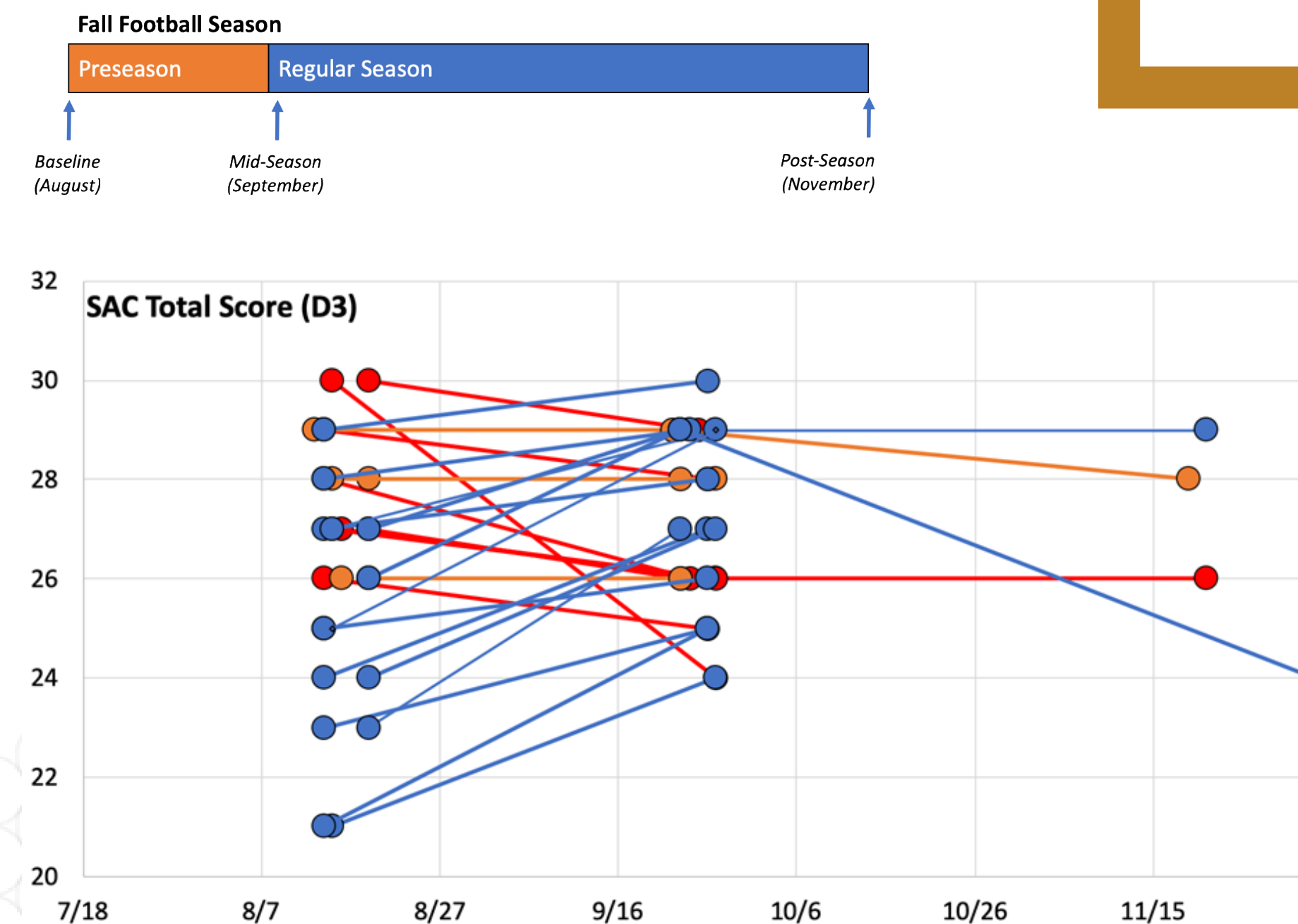
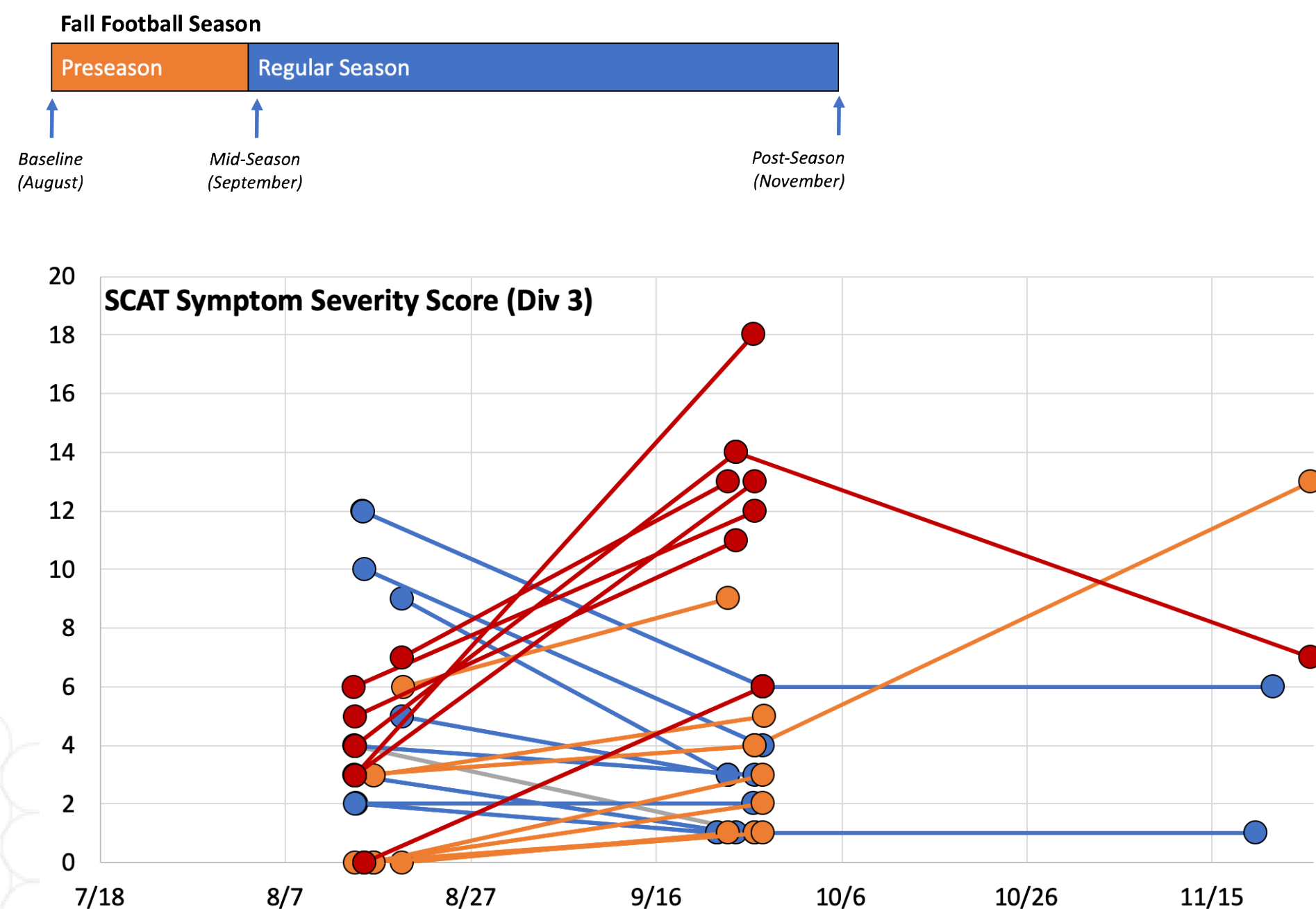
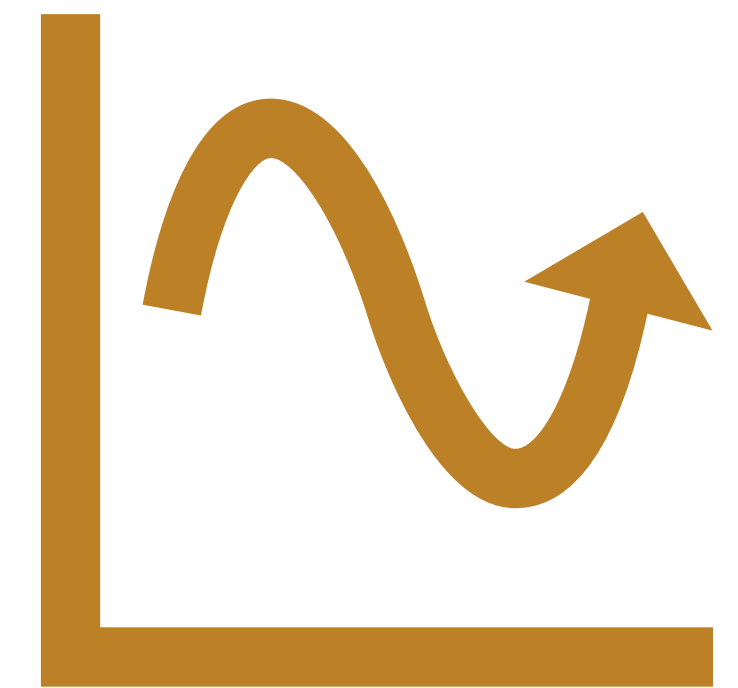


Aim 1 – Laboratory Accuracy



Aim 2 – Dose-Response

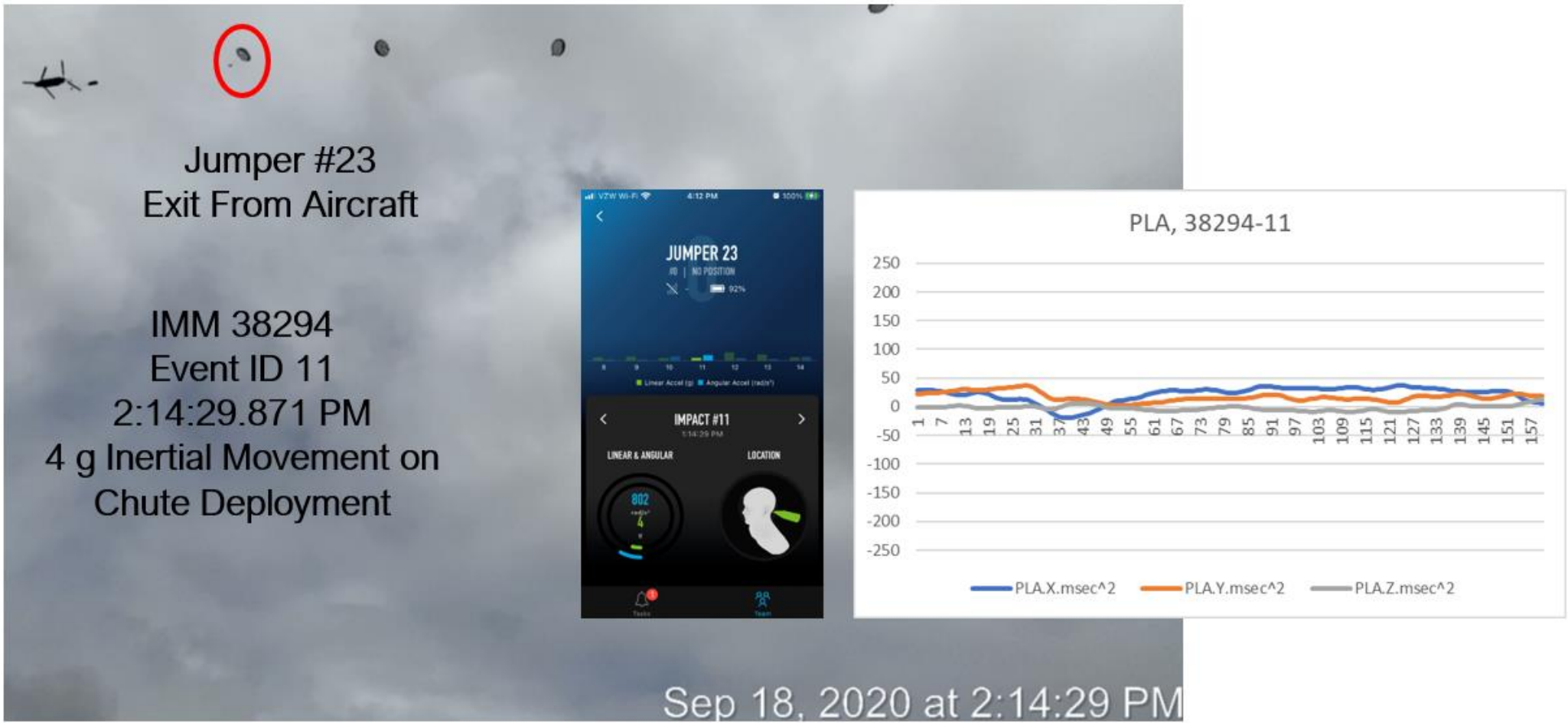
- **SCAT-3** (symptoms)
 - Midseason symptoms resolved by Post-Season
- **SAC** (cognitive)
 - Not correlated to head impact exposure



Aim 3 – Quantify Head Impacts



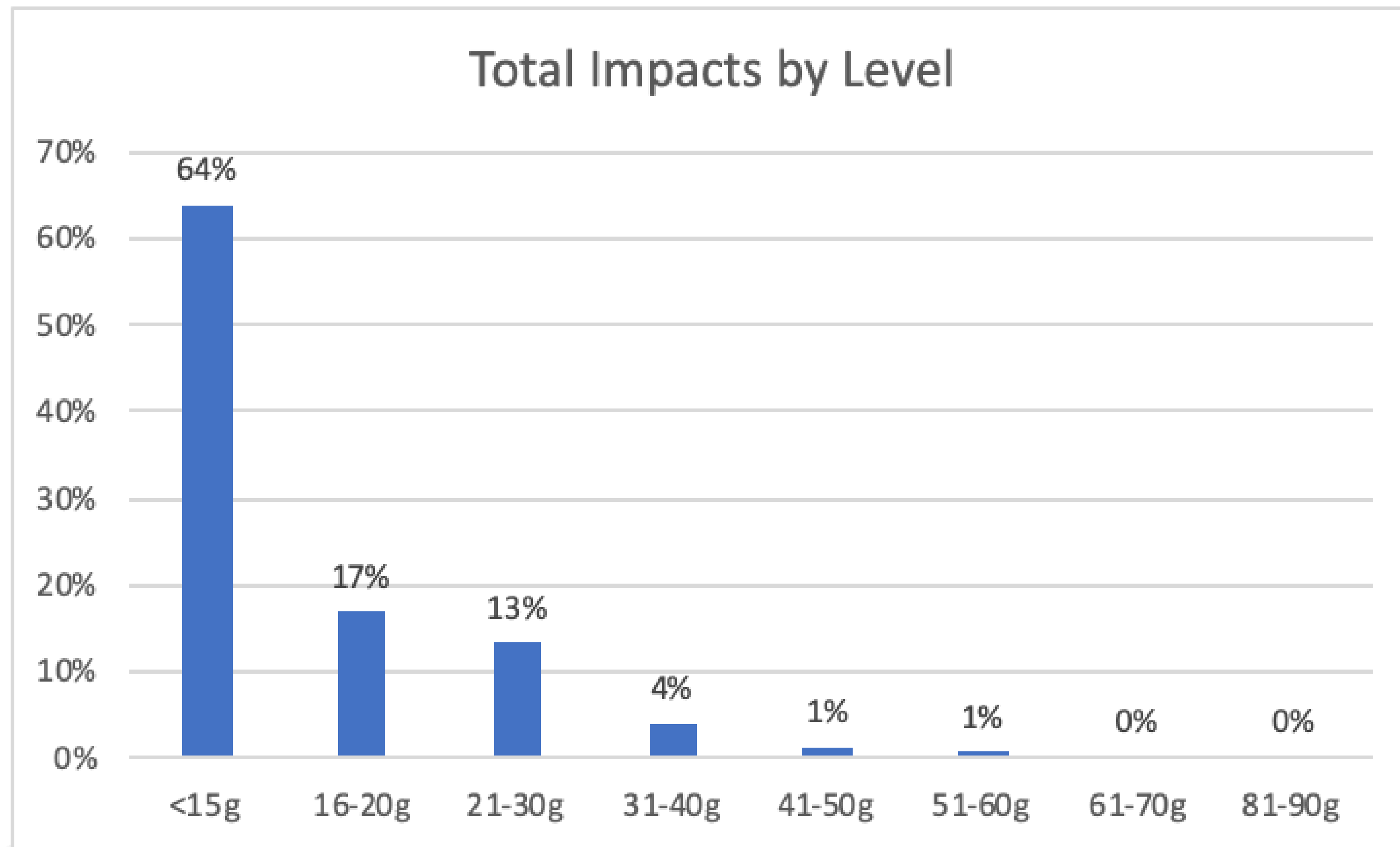
J23 Sequence- Exit From Aircraft



QB #5, 20-30g



Aim 3 – Quantify Head Impacts



Aim 4 – “Check Engine” Impacts

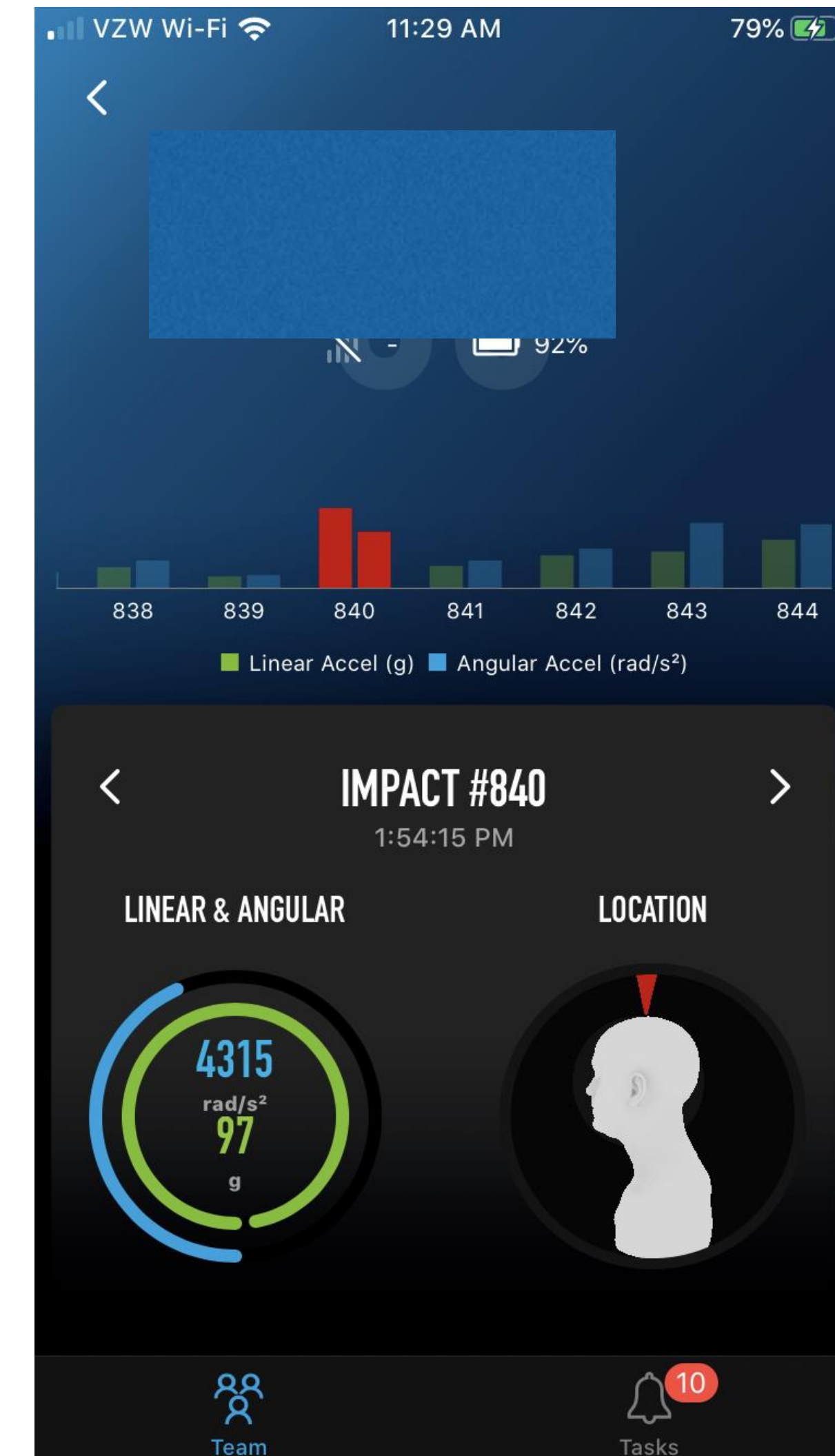
CASE STUDY

#90 IN RED

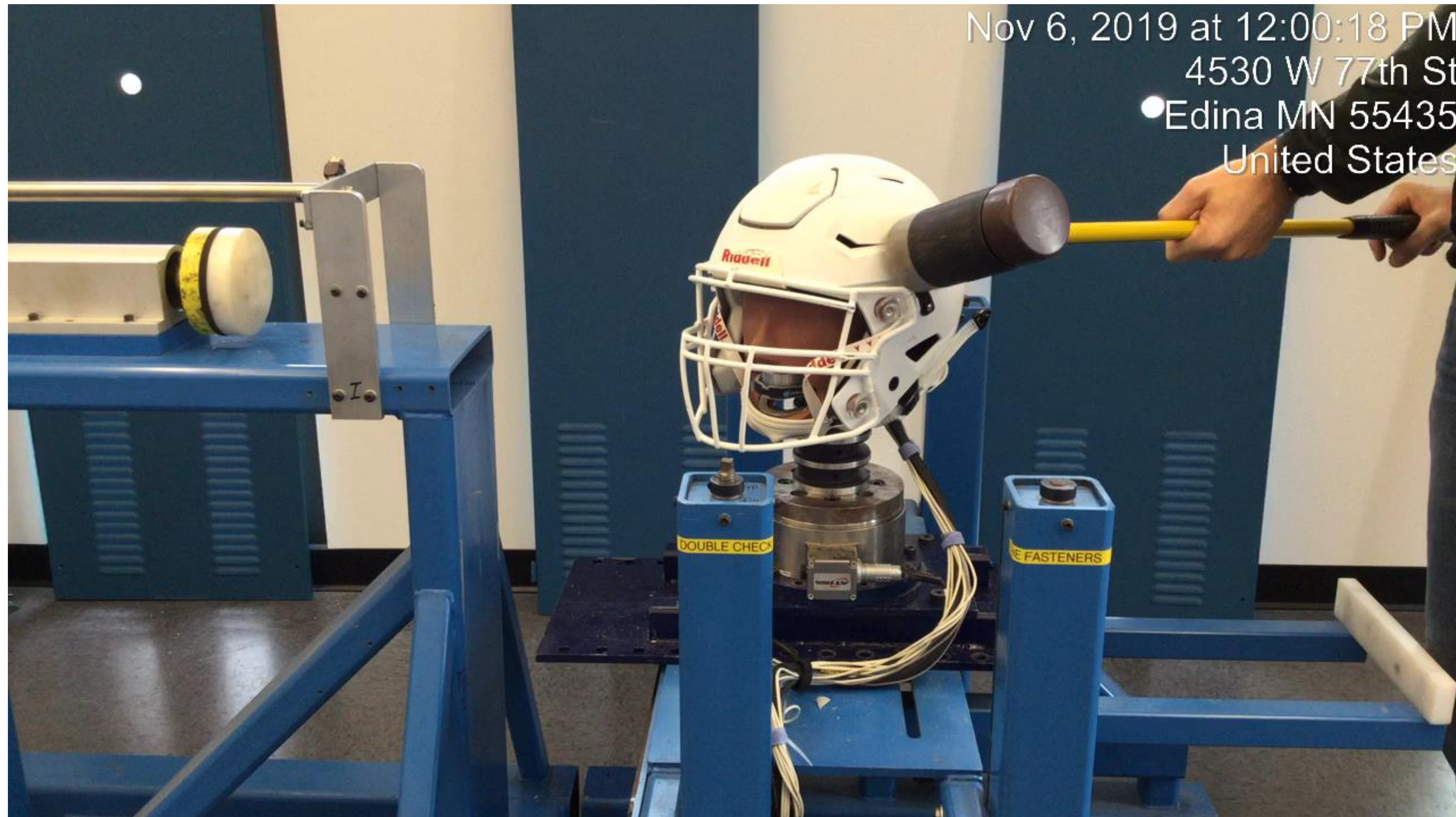
- Helmet-to-helmet with teammate
- Huge impact (54g)
- Immediately dazed/confused
- Never assessed for concussion
- Plays whole game, days later diagnosed with concussion



Aim 4 – “Check Engine” Impacts



“Check Engine” Impact, 60g



Aim 4 – “Check Engine” Impacts

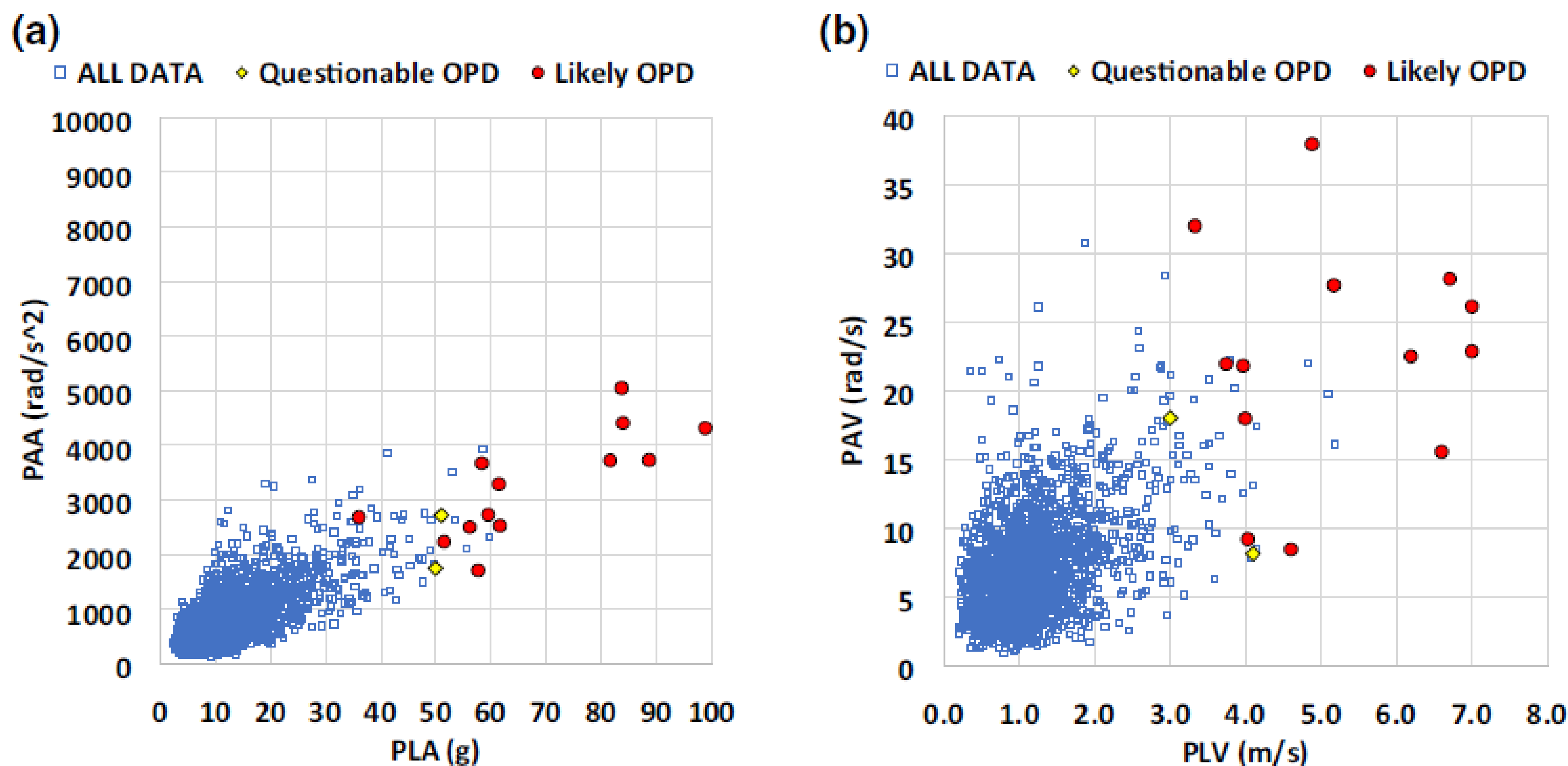


FIGURE 2. (a) Video-verified impacts (hollow squares) and clinician verified ‘Questionable’ (filled diamonds) and ‘Likely’ OPD impacts (filled circles) as quantified by PLA and PAA. (b) Video-verified impacts (hollow squares) and clinician verified Questionable (filled diamonds) and Likely OPD impacts (filled circles) as quantified by PLV and PAV.

NEXT STEPS

Next Steps

- Finish out current funding period thru OCT-2021
 - USAFA Boxing (Spring, Fall)
 - USMA Boxing (Spring, Summer, Fall)
 - USAFA Basic Cadet Training (Summer)
 - USMA Basic Cadet Training (Summer)
 - Ft. Bragg Airborne (Spring)
 - National Guard POMPOC (Spring)
 - USAFA, USMA sports (Fall)
- RISK: May exhaust impact monitors prior to AUG-2021

Next Steps, con't

- Continuation of BA150149, beyond OCT-2021
 - Broaden, deepen completion of specific aims over next 3 years
 - All 4 MSA's (16,000) by FY3
 - Grow Active Duty enrollment to 9,000 in FY3
 - Fully define normative head impact exposure in more military environments
 - More focus on combatives, parachute
 - Leverage “check engine” knowledge for soldier assessment
- RISK: Funding to continue needs to be in place by NOV-2021

	YEAR		
	2015-2017	2018-2021	2022-2024
USERS	130	2800	42000
COST/USER	\$ 22,000	\$ 1,100	\$ 200



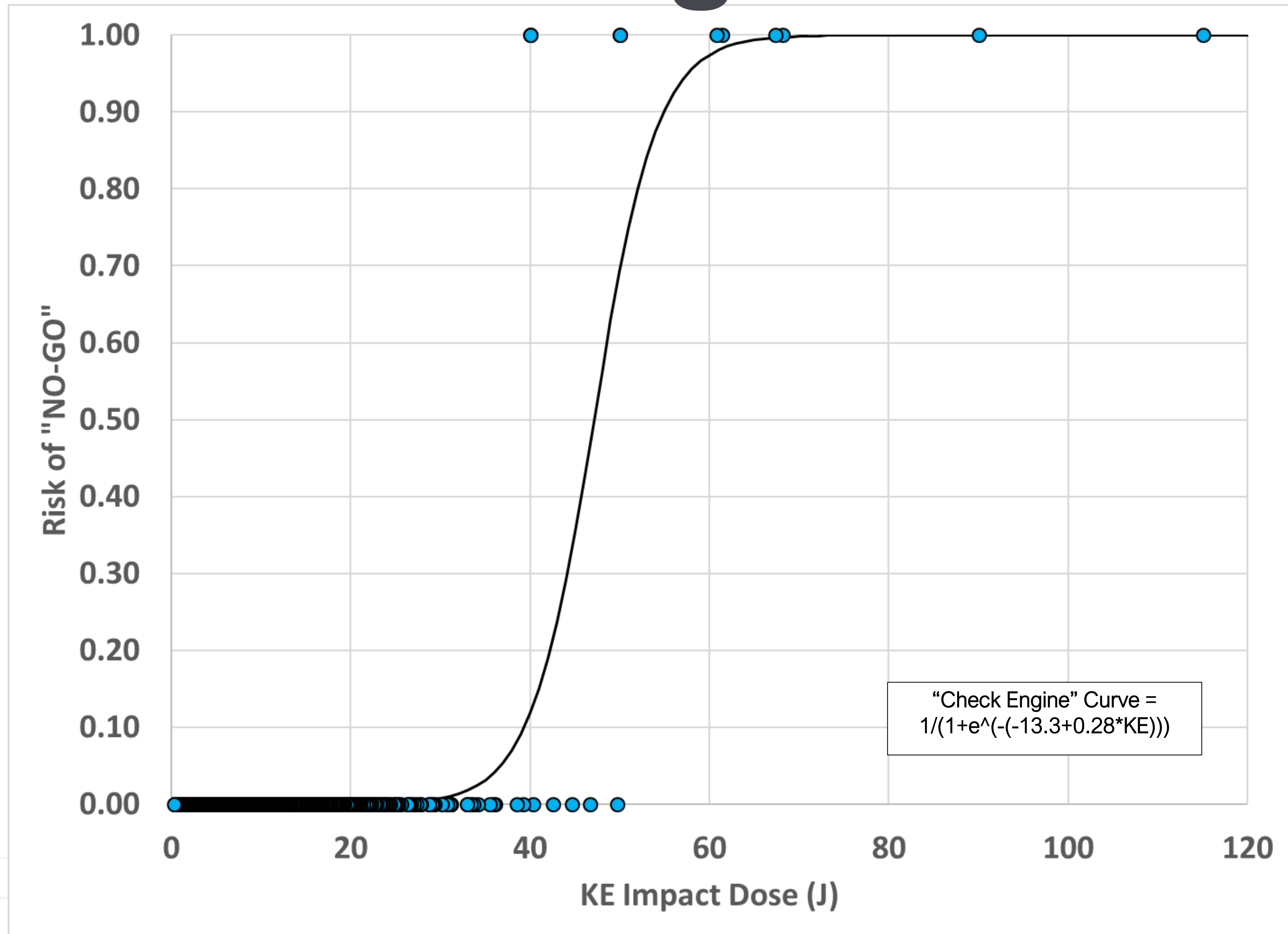
THANK YOU!



50 Cal. Sniper Rifle, Inertial Motion, 2g



“Check Engine” Curve



All “Check Engine” impacts to side and rear

n=2072
Side/Rear
Impacts



High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video

Adam J. Bartsch, Daniel Hedin, Jay Alberts, Edward C. Benzel, Jason Cruickshank, Robert S. Gray, Kenneth Cameron, Megan N. Houston, et

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High Energy Side and Rear American Football Head Impacts Cause Obvious Performance Decrement on Video

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 MEGAN N. HOUSTON,⁴ TYLER ROOKS,⁵ GERALD MCGINTY,⁶
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 VINCENT J. MIELE,⁸ J. CHRIS ASHTON,⁹ GUNTER P. SIEGMUND,¹⁰
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(Received 6 July 2020; accepted 24 September 2020)

Associate Editor Stefan M Duma oversaw the review of this article.

Abstract—The objective of this study was to compare head impact data acquired with an impact monitoring mouthguard (IMM) to the video-observed behavior of athletes' post-collision relative to their pre-collision behaviors. A total of $n = 83$ college and high school American football players wore the IMM and were video-recorded over 260 athlete-exposures. Ex-athletes and clinicians reviewed the video in a two-step process and categorized abnormal post-collision behaviors according to previously published Obvious Performance Decrement (OPD) definitions. Engineers qualitatively reviewed datasets to check head impact and non-head impact signal frequency and magnitude. The ex-athlete reviewers identified 2305 head impacts and 16 potential OPD impacts, 13 of which were separately categorized as Likely-OPD impacts by the clinical reviewers. All 13 Likely-OPD impacts were in the top 1% of impacts measured by the IMM (ranges 40–100 g, 3.3–7.0 m/s and 35–118 J) and 12 of the 13 impacts (92%) were to the side or rear of the head. These findings require confirmation in a larger data set before proposing any type of OPD impact magnitude or direction threshold exists. However, OPD cases in this study compare favorably with previously published impact monitoring studies in high school and college American football players that looked for OPD signs, impact magnitude and direction. Our OPD findings also compare well with NFL

reconstruction studies for ranges of concussion and sub-concussive impact magnitudes in side/rear collisions, as well as prior theory, analytical models and empirical research that suggest a directional sensitivity to brain injury exists for single high-energy impacts.

Keywords—Concussion, American football, Head impact sensor, Mouthguard, Video review, Obvious performance decrement.

INTRODUCTION

Millions participate yearly in contact sports in the United States. Hundreds of thousands of these athletes sustain concussions (CDC,^{12,43} yet due to subjective symptom disclosure, a lack of objective diagnostic criteria, poor athlete concussion knowledge or purposeful withholding of symptomology, approximately 50% of concussions go undiagnosed and unreported.^{35–37,48} There are increasing concerns about the potential negative consequences of acute concussion and accumulation of head impact doses on neurologic health⁶⁵; Stemper et al.^{1,10,11,13,28,29,34,51,56,57,62,68,69}) Accurate and precise (e.g. low uncertainty) head impact monitoring data from helmet-based and body-

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worn sensors provide a platform to inform our understanding of the relationship between single head impact doses, accumulation of head impact doses and resulting effects on neurologic function.

American football impact monitoring dates back to Col. John Paul Stapp in the 1970s, with later decades experimenting with a variety of instrumented bite-blocks, helmets, headbands, skin patches, mouthguards and the like.^{9,16,21,22,27,42,55,59,64,71} In the past decade, researchers have challenged sensor technologies to increase measurement trustworthiness so that new possibilities of analyzing specific head impact(s) and impact-specific decisions can be realized: “accurate measures of individual exposure will yield a direct estimate of the human tolerance”,⁵² “as more accurate sensors are designed...”,²⁰ “valid methods of measuring the direction and severity of on-field head impacts are needed”,⁶⁶ and sensors “may not consistently record head impacts or forces transmitted to the brain”.³⁰ In addition to low uncertainty kinematics estimates, qualitative video confirmation of head impacts has gained traction in the past few years as being very useful to pair with impact monitoring data,^{6,8,14,17,23–25,32,42,44,46,47,63} Bartsch *et al.*,^{4,5} Kieffer *et al.*,³⁹ Bartsch *et al.*,³ Bartsch *et al.*,^{2,18,60,73} An emerging benefit of such video reviews is that objective visible signs (VS) such as motor incoordination, lying motionless, impact seizure, among others can be documented (Bruce *et al.* BJSM⁸ NHL).

First done by the Australian Football League (AFL), clinicians have aimed to combine VS observations with head impact mechanisms to see if concussion diagnostic accuracy can be improved by more precisely identifying athletes who merit observation.^{46,47} In the past few years as a result of the new appreciation for VS video observations, professional leagues like the National Football League (NFL), National Rugby League (NRL) and National Hockey League (NHL) have followed suit by enacting measures to try and identify at-risk athletes using video review and live-game spotters.^{18,73} In American football specifically, an NFL study found that 74% of players with diagnosed concussions had at least one post-impact VS on video, though the concussion diagnostic sensitivity and specificity for VS from video alone were 73% and 65%, respectively.¹⁸

The objective of this study was to compare head impact data acquired with an impact monitoring mouthguard (IMM) to the video-observed behavior of athletes' post-collision relative to their pre-collision behaviors. Abnormal post-collision behaviors were noted according to Obvious Performance Decrement (OPD), which was based off definitions proposed by Ref. 14. We sought to quantify the range and percentile rank of the impact kinematics measured by the

IMM for the OPD impacts in comparison to the non-OPD impacts.

METHODS

Study Plan

The Study consisted of several components. First, athletes were consented and instrumented with an impact monitoring mouthguard. Then, during gameplay, instrumented athletes were video recorded. Afterwards the videos were scrutinized by two ex-Athlete Reviewers and potential collision, and head impact, timepoints identified. Athlete Reviewers also noted any timepoints where the players sustained potential OPD impact(s). After the Athlete Review was done, a panel of Clinician Reviewers examined and rated the potential OPD impacts. Engineers then reviewed kinematic traces from the IMM of the top 30 verified head impacts, inclusive of all OPD impacts, as well as all head impacts for any potential OPD player during the game, spot checks for another 1-2% of verified head impacts for non-OPD players and dozens of datasets from non-head impact events. Finally, the video-confirmed head impacts were paired by their timestamps to the engineer-reviewed kinematic data traces.

On-Field Data

The Institutional Review Board at the Medical College of Wisconsin approved all protocols. The IMM was a prototype Generation 1.3 wireless 3a3w accelerometer and gyroscope-based data acquisition system (DAQ) hermetically sealed inside an athletics mouthguard. The DAQ included kinematic sensors with sufficient range and bandwidth to estimate skull motion during impact (ADXL372, Analog Devices, Boston MA; BMG250, Bosch, Gerlingen Germany), on-board firmware for data transform from teeth to head center of gravity (CG), nonvolatile flash memory for storage, wireless rechargeable battery and wireless data offload using Bluetooth low energy. While the IMM outputs six degree of freedom kinematic time traces, only scalar peak linear acceleration change (PLA), scalar peak angular acceleration change (PAA), scalar peak linear velocity change (PLV), scalar peak angular velocity change (PAV), kinetic energy transfer based on change in linear and angular velocities (KET) and impact location are reported according to an analytical head impact model from Ref. 6. Linear peak kinematics are at the CG and the PLA has been found to be within 4% of Ref. 70.⁴⁵ Independent tests done for the Virginia Tech 5STAR Sensor Rating found

laboratory concordance correlation coefficient (CCC) to Reference headform of 0.97 and the IMM System was the best performing on-field sensor with a positive predictive value (PPV) of 96%.³⁸ Additional details on the IMM data acquisition can be found in the Appendix (Electronic supplementary material).

A total of 83 athletes wore the IMM during video-recorded American football games from Fall 2019. The athlete distribution was 1% Division I college, 63% Division III college and 36% High School. There were 260 athlete-exposures (AE), with one AE being defined as an athlete wearing the IMM in a video-recorded game. Because the IMM system was being deployed for the first time at scale, the first few months of data collection presented challenges for the Study Team. For some early season games the amount of data collected exceeded the capacity of data transmission from the IMM through a local hub to the cloud. For some games acquiring video of the entire field of play was not possible. There were also athletes who aggressively chewed on the IMM and broke the internal hardware during a game. Due to these technical issues, the large number of athletes instrumented and the frequency of data collection, some early season data was incomplete. Later in the season the system ease-of-use, and concept of operations, improved greatly.

Ex-Athlete Video Reviewers

For the initial video review, this study utilized two reviewers who were former athletes, with at least 10 years of American football training, experienced in video breakdown of dozens of American football games, knowledge about movement norms of each playing position, expected normal player behavior for each type of play and a personal understanding of what typically happens on the field during normal and high-energy plays for high school and college American football.

The athlete reviewers were first provided continuous time-stamped video of at least 30 frames per second, at least 1024 × 768 resolution and at least 720i format with synchronous time as the IMM (using a free app Timestamp Camera, available on the Apple Store), and reviewed video together to ensure they found collisions sequences likely to have head impacts for every athlete during the entire game. A collision sequence is broadly defined as any play which could have head impact(s), where an athlete appears to collide one or more times with another player or the field, such as a lineman who collides with two different players within a few seconds. Players were also observed on the sideline to quantify any out-of-play impacts. A sideline view was taken of players during on-field action, and if necessary, additional video sources such as Hudl or team-uploads to YouTube were mined to confirm collisions or to view collisions from a

different angle. The athlete reviewers first agreed whether it looked like a collision occurred during the play according to methods established by Kuo et al.,⁴² and second whether after an athlete was involved in a collision, that there was clear visual evidence of an abnormal player reaction which we called Obvious Performance Decrement (OPD). Each athlete reviewer recorded their own opinion of the OPD cases, including situations where they disagreed with each other. These comments are provided in the Appendix (Electronic supplementary material). The immediate incapacitation and altered balance and/or movement coordination OPD reactions were based on the consensus visible sign definitions from,¹⁴ and included additional football-specific atypical post-impact behaviors:

1. Immediate incapacitation

- Potential loss of consciousness, lying motionless > 2 s*
- Fencing response
- Tonic posturing*
- Decorticate primitive reflex*
- Falls to ground in unprotected manner*
- Blank/vacant look*

2. Altered balance and/or movement coordination

- Slow-to-get up*
- Gross motor instability*
- Functional disturbance - Staggering/stumbling/tripping/falling*
- Movement - slow/labored/abnormal/unstable

3. Atypical post-impact behavior

- Clutches head
- Confusion/disorientation
- Irregular sport-related actions – lines up in wrong position, incorrect playing style (runs away from contact), playing style gives opponent obvious advantage (blocks wrong team), unsure of location on field, unsure of where bench is
- Player seeks out sideline caregivers
- Teammate seeks out sideline caregivers on player's behalf

*indicates consensus visible signs defined by Ref. 14.

For games where any player exhibited potential OPD behavior the reviewers re-reviewed all collisions for the athlete during the game to confirm OPD behavior. The video review summaries are provided in the Appendix (Electronic supplementary material). After the video reviews and OPD double-checks were done, the athlete reviewers were provided time stamped IMM data and the comparison between video

review and IMM-measured impacts was initiated. Obvious IMM non-head impact and data without corresponding video were removed (e.g., before game, between plays, halftime, after game, between quarters, on sideline).

After the season was over, both athlete reviewers re-reviewed the top 1% of IMM-reported impact magnitudes (whether a potential OPD impact or not) and confirmed visual findings in a similar method to Stemper *et al.*⁶⁸

Clinician Video Review

Videos that included potential OPD player reactions were then reviewed by a panel of clinician experts—one NFL video spotter, one NFL sideline physician, two NFL unaffiliated neurological consultants and three athletic trainers with 10+ years of American football experience—to confirm whether the video reviewers' identification of potential OPD aligned with clinician findings. Clinicians developed majority consensus on whether impacts looked like No OPD, Questionable OPD and Likely OPD. The Likely OPD cases were ones that all clinicians were certain that the athlete's post-collision behavior on video fit into the OPD operational definitions. The Questionable OPD cases were ones where most clinicians thought it was plausible that video showed OPD behavior, but they could not tell for certain. The No OPD cases were ones where most clinicians felt the athlete's pre- and post-collision behavior had zero OPD behavior on video.

Engineer Time Trace Review

After video reviews were done, two engineers confirmed kinematic data quality for the top 1% of impacts based on Bartsch *et al.*⁶ To verify the thoroughness of the video reviews, spot checks were also done for another 1–2% of impact data sets selected at random across the distribution to ensure qualitative video findings agreed with IMM-reported head impact dose kinematics. Dozens of video-reported non-head impact events were also examined to confirm their lack of conformance to a head impact model (e.g. high noise, high frequency content, non-impact motion).

Data Analysis

Statistical analyses for the video-verified IMM data included summarizing percentage of impacts by location on the head, providing summary distributions by percentage of head impact severity for PLA, PAA, PLV, PAV and KET, and analyzing the top 1% and 5% of impacts for potential OPD cases.

RESULTS

Out of a total of 63,238 IMM data sets collected (roughly 240 per A–E), after combining video reviewer a total of $n = 2305$ head impacts were confirmed. There were 1716 head impacts (74%) with $PLA > 10$ g and 589 head impacts (26%) with $PLA < 10$ g. There were no cases of a video-reported collision sequence where there was no IMM data (e.g. false negative head impact measurement). About 60,000 non-head impact data sets were removed because of the video review.

The summary distribution of head impacts by estimated CG $PLA > 10$ g ($n = 1716$) is displayed in Fig. 1a. Like the PLA distribution, the PAA, PLV, PAV and KET had distributions skewed toward lower magnitudes and are shown in Figs. 1b–1e. By impact location, Fig. 1f, 67% were to the front, 20% were to the sides and 13% were to the back of the head. The top 1% of rounded head impact peaks were $PLA > 50$ g, $PLV > 4$ m/s, $PAA > 4,000$ rad/s², $PAV > 20$ rad/s and $KET > 40$ J.

All OPD impacts identified by the ex-Athlete reviewers occurred in the top 2% of impacts by PLV. For these impacts, Reviewer #1 identified 14 impacts as OPD, while Reviewer #2 identified 15 impacts as OPD. Thirteen of these OPD impacts were common to both reviewers. After final review by the clinical experts, 13 impacts were classified as Likely OPD, 2 impacts as Questionable OPD and 16 impacts as No OPD. See Table 1 for video review summary, and Table A1 (Electronic supplementary material) for complete video review details of the top 30 impacts by PLV. All potential OPD impacts flagged by non-clinician reviewers were within the top 5% by impact magnitudes.

For side and rear impacts, twelve of the top fifteen (12/15, 80%) events (PLA range 40–100 g, PAA range 1700–5000 rad/s², PLV range 3.3–7.0 m/s, PAV range 15–30 rad/s, KET range 34–118 J) resulted in clinician-classified Likely OPD, see Table 2 and Figs. 2a and 2b. There were two Likely OPD cases—one to the side and one to the rear—where the athletes appeared to lose consciousness and were lying motionless on the field. For frontal impacts, one of the top fifteen (1/15, 7%) events (PLA 62 g, PAA 2529 rad/s², PLV 4.0 m/s, PAV 9 rad/s, KET 38 J), resulted in Likely OPD.

DISCUSSION

The goal of this study was to characterize head impact kinematics associated with OPD in college and high school football players using video-verified head

High Energy Side and Rear American Football Head Impacts

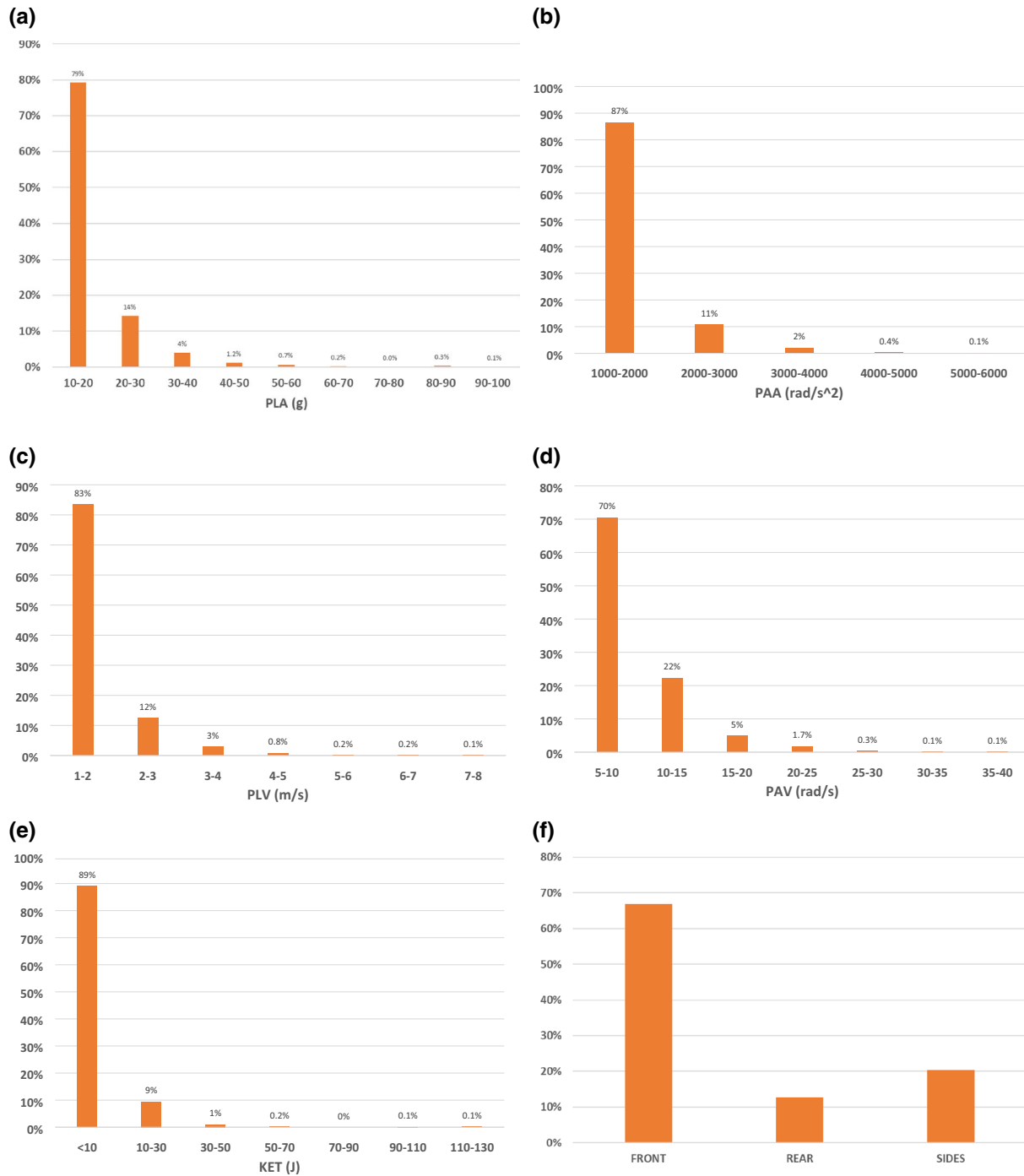


FIGURE 1. (a) Distribution of $n = 1716$ video-verified head impacts with Peak CG Linear Acceleration (PLA) > 10 g from current study. (b) Peak Angular Acceleration (PAA) head impact distribution. (c) Peak Linear Velocity (PLV) head impact distribution. (d) Peak Angular Velocity (PAV) head impact distribution. (e) Estimated Kinetic Energy Transfer (KET) head impact distribution. (f) Head impact locations for current study.

TABLE 1. Summary of top 30 head impacts by PLV with OPD observations from video.

	Reviewer #1	Reviewer #2	Clinician review
No OPD	16	15	15
Questionable OPD	N/A	N/A	2
Likely OPD	14	15	13

Non-clinician reviewers identified impacts as OPD or No OPD. An additional 2295 impacts were also classified by video reviewers as No OPD.

TABLE 2. Kinematic summary of clinician reviewed impacts.

	Clinician review	Median					Direction
		PLA	PAA	PLV	PAV	KET	
No OPD	15	47	2100	4.0	16	38	3 Side, 1 Rear, 11 Front
Questionable OPD	2	50	2200	3.5	13	32	2 Side
Likely OPD	13	61	3300	4.9	22	68	8 Side, 4 Rear, 1 Front

Twelve of thirteen (92%) likely OPD cases occurred to the side and rear of the head.

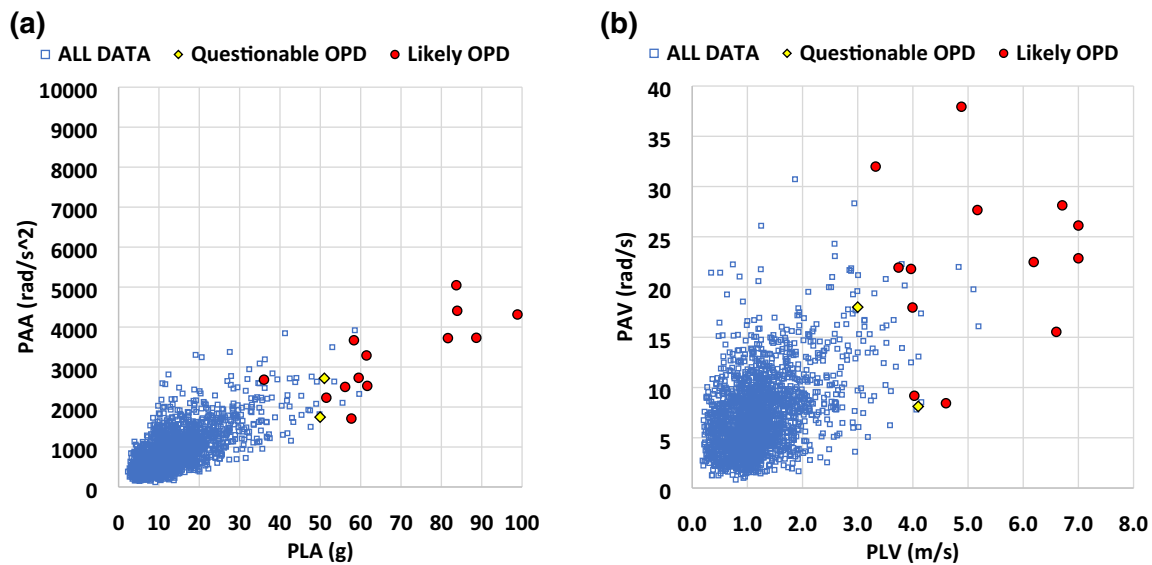


FIGURE 2. (a) Video-verified impacts (hollow squares) and clinician verified ‘Questionable’ (filled diamonds) and ‘Likely’ OPD impacts (filled circles) as quantified by PLA and PAA. (b) Video-verified impacts (hollow squares) and clinician verified Questionable (filled diamonds) and Likely OPD impacts (filled circles) as quantified by PLV and PAV.

impact data collected using an impact monitoring mouthguard

There were 15 impacts—less than 1% of the 2305 video verified impacts—flagged by non-clinician video reviewers as potential OPD. Clinician reviewers decided that 13 of these impacts were ‘Likely OPD’ and 2 were ‘Questionable OPD’. The OPD impacts in this study compared favorably to previously published data on OPD (*aka* “No-Go”) events (Bartsch *et al.*,⁴ Bartsch *et al.*,⁵ Bartsch *et al.*,³ Bartsch *et al.*² That previous work found that 100% (8/8) of the OPD cases in American football occurred to the side and rear of the head, and that these impacts were all in the top 1%

by PLA, PLV and KET impact magnitude, approximately 50–90 g, 3–7 m/s and 40–120 J, respectively.

The consistency between the OPD impact magnitudes and locations in the current study and in prior studies could inform future concussion surveillance. In particular, the NFL has for several years used clinical video reviewers and clinical gameplay observers to identify single head impacts in real-time when athletes appear affected and reported that 74% of concussed players had a visible sign.⁷³ Being able to also provide observers with reliable head impact kinematics data and a sense of how an impact’s magnitude and location relates to OPD events could be helpful. Reliable data

could also ensure that all athletes who sustain a large head impact—whether initially unnoticed by observers, seen to be OPD on video, diagnosed concussive, or with no ultimate clinical findings—be more readily identified and potentially evaluated. Reliance on video surveillance or head impact monitoring systems alone, however, should not be considered a sufficient replacement for clinical observation and evaluation.

When comparing OPD impacts observed in this study to the NFL concussion biomechanics reconstruction studies (Newman et al.,⁵⁴ Newman et al.^{53,61} Neurosurgery), the results are quite interesting. First, those authors pointed out that seminal early work by Stapp⁶⁷ and Patrick et al.⁵⁸ estimated the human tolerance to padded head impacts in the range of PLA 40–80 g. While not definitive, most of our OPD impact magnitudes fall within this PLA range. Second, the NFL reconstruction authors' estimated ranges of concussion-causing PLA, PLV and KET to the side/rear were 50–140 g, 4–10 m/s and 40–250 J, and non-concussion impacts to side/rear to be 20–80 g, 3–6 m/s and 20–80 J, respectively. The OPD impacts we collected in this study were mostly at the median or lower of the NFL concussion-causing ranges, and at the median or higher of the NFL non-concussion ranges. This makes sense: in American football collisions the NFL data likely supply an upper bound on *human tolerance* to side/rear high energy impacts.

The NFL laboratory reconstruction authors also thought that concussion/mild TBI in their impacts occurred mainly from high energy lateral, oblique and rear head impacts and that the head might be more sensitive to lateral than to frontal impacts. Several other authors have published similar thoughts and experimental data to support brain directional sensitivity ideas over many decades (Holbourn,³³ Gennarelli et al.,^{26,40,49,72} Kleiven.^{19,41,50} Based on our current study, and the published IMM data and video review OPD findings, we may be observing in the 'living laboratory' that the brain does, in fact, have a directional sensitivity to the highest energy side and rear impacts. Between this study and the previously published IMM data (Bartsch et al.,⁴ Bartsch et al.,⁵ Bartsch et al.,³ Bartsch et al.,² 95% of OPD impacts were to the side/rear despite only 35% of all video-verified impacts occurring to the side/rear of the head.

Explanations for observed OPD directional sensitivity may be related to kinetic and structural mechanisms. One explanation may be mass/inertia dependent; when a subject is aware of an oncoming impact like in a frontal blow, they may brace and thereby engage a larger effective mass to attenuate the impact. This would result in a lower acceleration and velocity change, and lower energy transfer to the brain.

When a subject sustains a side or rear impact, they may not be able to brace in the same manner and there could be a larger acceleration/velocity change, and energy transfer to the brain. Video-verified collisions showed several athletes who sustained impacts where bracing would have been very difficult, when an unprotected head-to-ground impact occurred. Another potential mechanism may be dependent on the anatomic location for different brain function. Side and rear impacts may have a greater effect on areas of the brain which regulate vestibular-ocular, vestibular-spinal, and ocular-motor functions. Impairment of these functions would be more likely to result in OPDs observable during video review. Alternatively, frontal impacts may have a greater effect on areas of the brain which regulate functions such as emotion, language, and personality which would be less likely to result in OPDs and therefore not observed during video review.

We acknowledge the limited understanding of the mechanistic basis for OPD, impact magnitude and impact direction. Pairing OPD video observations with low-uncertainty impact monitoring data is not a clinical panacea at this point. We believe tying together impact dosing with OPD observations is an important step toward accurately quantifying head impact dose-response relationships. In the future, it is hoped that use of good quality impact monitoring data may lead to more targeted clinical efforts to oversee athlete impact doses.

Regarding the rest of the data, the two Reviewers video-verified head impacts in games in $n = 83$ American football players over 260 AE. A total of 2305 impacts were video verified, with 1716 having PLA greater than 10 g. For PLA < 10 g the Reviewers video-confirmed 589 inertial and/or head contacts that appeared 'small' on video. Based on these results, we think there were likely many hundreds—or thousands—of additional minor inertial/head contacts collected by the IMM with PLA < 10 g but were not notable to record as such by the Reviewers. Because of the immense work that would have been required to re-review the videos to extract all of the PLA < 10 g events, and the fact that head motions in this range are on par with activities of daily living, we decided it was not necessary to re-review videos for PLA < 10 g. In the future if head impact doses with PLA < 10 g demonstrate engineering or clinical importance, we may go back and re-review the videos to extract all these types of minor head motion events.

The PLA > 10 g distributions shown in Fig. 1, with the top 1% of impacts occurring above approximately 50 g, were like American football distributions published previously using a similar data collection and video reviewer methodology (Bartsch et al.,⁴ Bartsch et al.,³ While individual athlete and positional

impact distribution variances no doubt existed and were obscured by reporting of individual impacts as part of a population distribution, the results shown in Fig. 1 confirmed that the impact dosing was consistent between the current study and prior research. Also, the range of impact magnitudes, as well as impact locations confirmed that on-field data in this study fell within the ranges of calibration from the laboratory with the exception that the Virginia Tech Sensor 5 laboratory tests in the Appendix (Electronic supplementary material) did not calibrate the IMM response to facemask impacts. However, facemask impact calibrations have been published elsewhere,⁶ Hedin *et al.*,³¹ Bartsch *et al.*,⁴ Bartsch *et al.*,³ Domel *et al.*¹⁵ (submitted) and the impacts collected in this study fell within the published facemask calibration ranges.

Because of a lack of trusted reference in the 'living laboratory', video and impact monitoring results carry a degree of doubt. Video review has a qualitative and subjective nature since players can block the view, images can be blurry and the main velocity change of the head occurs in about 15 ms, while the typical video camera captures a frame every 15 to 30 ms. Head impact monitoring data is subject to difficulties of confirming high signal-noise ratio in the 'living laboratory'. Ensuring sufficient coupling to the skull is necessary, as well as engineering quality checks of data to confirm reported impact doses are physically realistic. Although video observations and impact dose estimates have some degree of doubt when each is viewed *in isolation* and without consideration of the other, their utility increases when both data sources are combined. If one can reliably determine when an impact monitor is, and is not, firmly coupled to the teeth then it becomes easier to tie impact dose data to video observations. Likewise, if one is able to check impact severities relative to what the sensor reports—e.g. a "small" sensor impact ought to look small and a "large" sensor impact should look large—on video prior to examining kinematic data traces, then some level of trust can be developed for both the video and monitoring methods, even when one of them has lower signal-noise ratio. Finally, a good check on the method is to do a laboratory reconstruction of impacts. If data and video agree, then the reconstruction should be straightforward. If not, then the reconstruction could be extremely difficult or impossible, and a re-analysis of both data and video quality are merited.

One study limitation is that the KET estimate does not reflect the true change in kinetic energy due to the impact. While the PLV and PAV reflect the absolute velocity changes, the total kinetic energy change is unknown because we do not know the initial linear or angular velocity of the athlete's head. However, the KET—and potentially the rate of energy

change—provides a simple way to quantify the combined effects of linear and rotational changes in a single metric, can be easily replicated in lab testing, and have been used by other authors to indicate the overall impact dosing (Newman *et al.*^{54,61} And the KET estimate can also provide a simple to grasp cumulative impact dose quantifier, since summing energy or power has an easier to grasp physical basis than summing peaks of acceleration or complex risk function outputs. This is one area where computational simulation may help us understand the effects of an inaccurate KET estimate on potential injury risk. Pre-impact velocity conditions can be modified virtually to quantify differences in brain stress and strain.

Another limitation of this study is that observing an OPD impact on video does not constitute a clinical diagnosis or take the place of a clinical exam for an athlete observed with concussion signs and symptoms. Additionally, on-field OPDs may be associated with very acute disturbances and/or low-level deficits that do not exceed the thresholds for clinically diagnosable concussion, or other non-concussion-related injuries. Therefore, the validity of an OPD observation with respect to a clinical concussion diagnosis remains unknown. Concussion reporting subjectivity is a well-known issue.⁷ So it is possible that historical concussion tolerance and/or concussion risk curves, most of which have not been able to tie a specific impact on video to a clinically observed sign or symptom, are based on semi-random subject reporting. This poor connection between the injury and the impact that caused the injury could reveal the real benefit of our OPD analysis—we have attempted to remove subjective self-reporting and instead provide a more objective method for identifying impacts that could result in higher risk of an injury. For these reasons, in future work we aim to notify study staff soon after a high-energy head impact is recorded and start to link the impact kinematics to quantifiable clinical findings and diagnoses for injured and uninjured athletes alike.

In conclusion, we found that the highest energy side and rear impacts to the head were associated with abnormal post-collision behaviors in 83 college and high school American football players. Athlete and clinical reviewers categorized abnormal post-collision behaviors according to Obvious Performance Decrement (OPD) definitions, finding evidence of Likely OPD in 13 of 2305 video-verified head impacts measured with an impact monitoring mouthguard system. These 13 Likely-OPD impacts were all in the top 1% of impacts when ranked by magnitude, and fell in the ranges of 40–100 g, 3.3–7.0 m/s and 35–118 J. These findings require confirmation in a larger data set before proposing any injury thresholds for OPD impacts.

ELECTRONIC SUPPLEMENTARY MATERIAL

The online version of this article (<https://doi.org/10.1007/s10439-020-02640-8>) contains supplementary material, which is available to authorized users.

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CONFLICT OF INTEREST

No benefits in any form have been or will be received from a commercial party related directly or indirectly to the subject of the manuscript. Cleveland Clinic has entered into an exclusive license agreement with Prevent Biometrics for instrumented mouthguard technology on which AB, EB and VM are inventors. AB is also an employee of Prevent Biometrics. The conflicts disclosed here are managed by the Cleveland Clinic Conflict Management Board. Complete details can be found at <https://my.clevelandclinic.org/ccf/media/files/About/COI.pdf>.

REFERENCES

¹Arbogast, K. B., A. D. McGinley, C. L. Master, M. F. Grady, R. L. Robinson, and M. R. Zonfrillo. Cognitive rest and school-based recommendations following pedi-

- atric concussion: the need for primary care support tools. *Clin Pediatr (Phila)*. 52(5):397–402, 2013.
- ²Bartsch A, Benzel E, Samorezov S, Miele V. Summary of accurate and precise head impacts collected on-ice in hockey players. In: ASTM International STP Proceedings from the 2019 International Ice Hockey Summit, Denver, Colorado, pp. 1–29, 2020.
- ³Bartsch, A., R. Dama, J. Alberts, S. Samorezov, E. Benzel, V. Miele, A. Shah, J. Humm, M. McCrea, and B. Stemper. Measuring blunt force head impacts in athletes. *Mil. Med*. 185(S1):190–196, 2020.
- ⁴Bartsch 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. In: Conf Proc IEEE Eng Med Biol Soc. 2019 Jul, pp. 2068–2072.
- ⁵Bartsch, A., V. Miele, J. Alberts, E. Benzel, A. Shah, J. Humm, B. Stemper, and M. McCrea. Estimates of high-risk single and cumulative head impact doses in American football. *Biomed. Sci. Instrum*. 55(2):224–229, 2019.
- ⁶Bartsch, A. J., S. Samorezov, E. C. Benzel, V. J. Miele, and D. Brett. Validation of an “intelligent mouthguard” single event head impact dosimeter. *Stapp Car Crash J*. 58:1–27, 2014.
- ⁷Baugh, C. M., W. P. Meehan, 3rd, E. Kroshus, T. G. McGuire, and L. A. Hatfield. College football players less likely to report concussions and other injuries with increased injury accumulation. *J. Neurotrauma* 36(13):2065–2072, 2019.
- ⁸Bruce, J. M., R. J. Echemendia, W. Meeuwisse, M. G. Hutchison, M. Aubry, and P. Comper. Development of a risk prediction model among professional hockey players with visible signs of concussion. *Br. J. Sports Med*. 52(17):1143–1148, 2018.
- ⁹Camarillo, D. B., P. B. Shull, J. Mattson, R. Shultz, and D. Garza. An instrumented mouthguard for measuring linear and angular head impact kinematics in American football. *Ann. Biomed. Eng*. 41(9):1939–1949, 2013.
- ¹⁰Cantu, R. C. Second-impact syndrome. *Clin. Sports Med*. 17:37–44, 1998.
- ¹¹Cantu, R. C., and A. D. Gean. Second-impact syndrome and a small subdural hematoma: an uncommon catastrophic result of repetitive head injury with a characteristic imaging appearance. *J. Neurotrauma* 27:1557–1564, 2010.
- ¹²Centers 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. United States Centers for Disease Control and Prevention-Morbidity and Mortality Weekly Report (MMWR); 2011. Report No.: MMWR 60(39).
- ¹³Daneshvar, D. H., D. O. Riley, C. J. Nowinski, A. C. McKee, R. A. Stern, and R. C. Cantu. Long-term consequences: effects on normal development profile after concussion. *Phys. Med. Rehabil. Clin. N. Am*. 22:683–700, 2011.
- ¹⁴Davis, G. A., M. Makdissi, P. Bloomfield, et al. International study of video review of concussion in professional sports. *Br. J. Sports Med*. 53(20):1299–1304, 2019.
- ¹⁵Domel A, Validation and Comparison of Instrumented Mouthguards in Measuring Head Kinematics and Brain Deformation in Football Impacts. ABME 2020 (submitted).
- ¹⁶Duma, S. M., S. J. Manoogian, W. R. Bussone, P. G. Brolinson, M. W. Goforth, J. J. Donnenwerth, R. M. Greenwald, J. J. Chu, and J. J. Crisco. Analysis of real-time

- head accelerations in collegiate football players. *Clin. J. Sport Med.* 15:3–8, 2005.
- ¹⁷Echemendia, R. J., J. M. Bruce, W. Meeuwisse, M. G. Hutchison, P. Comper, and M. Aubry. Can visible signs predict concussion diagnosis in the National Hockey League? *Br. J. Sports Med.* 52(17):1149–1154, 2018.
 - ¹⁸Elbin, R. J., S. L. Zuckerman, A. K. Sills, J. R. Crandall, D. J. Lessley, and G. S. Solomon. Sensitivity and specificity of on-field visible signs of concussion in the National Football League. *Neurosurgery* 87:530–537, 2020.
 - ¹⁹Elkin, B. S., L. F. Gabler, M. B. Panzer, and G. P. Siegmund. Brain tissue strains vary with head impact location: a possible explanation for increased concussion risk in struck versus striking football players. *Clin. Biomech.* 64:49–57, 2019.
 - ²⁰Elliott, M. R., S. S. Margulies, M. R. Maltese, and K. B. Arbogast. Accounting for sampling variability, injury under-reporting, and sensor error in concussion injury risk curves. *J. Biomech.* 48:3059–3065, 2015.
 - ²¹Foust D. R. Minutes from the Sixth Annual International Workshop on Human Subjects for Biomechanical Research. 23 October 1978, pp. i–xv, Ann Arbor, Michigan.
 - ²²Funk, J. R., J. M. Cormier, C. E. Bain, H. Guzman, E. Bonugli, and S. J. Manoogian. Head and neck loading in everyday and vigorous activities. *Ann. Biomed. Eng.* 39(2):766–776, 2011.
 - ²³Gardner, A. J., D. R. Howell, and G. L. Iverson. A video review of multiple concussion signs in National Rugby League match play. *Sports Med. Open.* 4(1):5, 2018.
 - ²⁴Gardner, A. J., D. R. Howell, C. R. Levi, and G. L. Iverson. Evidence of concussion signs in national rugby league match play: a video review and validation study. *SportsMed Open.* 3(1):29, 2017.
 - ²⁵Gardner, A. J., R. Kohler, W. McDonald, G. W. Fuller, R. Tucker, and M. Makdissi. The use of sideline video review to facilitate management decisions following head trauma in super rugby. *Sports Med. Open.* 4(1):20, 2018.
 - ²⁶Gennarelli T, Thibault L, Tomei G, Wiser R, et al. Directional dependence of axonal brain injury due to centroidal and non-centroidal acceleration. In: SAE Technical Paper 872197, Proceedings of the 31st Stapp Car Crash Conference, 1987.
 - ²⁷Greenwald, R. M., J. T. Gwin, J. J. Chu, and J. J. Crisco. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. *Neurosurgery* 62:789–798, 2008.
 - ²⁸Guskiewicz, K. M., S. W. Marshall, J. Bailes, M. McCrea, R. C. Cantu, C. Randolph, et al. Association between recurrent concussion and late-life cognitive impairment in retired professional football players. *Neurosurgery.* 57:719–726, 2005.
 - ²⁹Guskiewicz, K. M., J. P. Mihalik, V. Shankar, S. W. Marshall, D. H. Crowell, S. M. Oliaro, et al. Measurement of head impacts in collegiate football players: relationship between head impact biomechanics and acute clinical outcome after concussion. *Neurosurgery.* 61:1244–1252, 2007.
 - ³⁰Harmon, K. G., J. R. Clugston, K. Dec, et al. American Medical Society for Sports Medicine position statement on concussion in sport. *Br. J. Sports Med.* 53:213–225, 2019.
 - ³¹Hedin DS, Gibson PL, Bartsch AJ, Samorezov S. Development of a Head Impact Monitoring “Intelligent Mouthguard”. In: Conf Proc IEEE Eng Med Biol Soc. 2016 Aug; 2007–2009.
 - ³²Hendricks, S., S. O'Connor, M. Lambert, J. C. Brown, N. Burger, S. Mc Fie, C. Readhead, and W. Viljoen. Video analysis of concussion injury mechanism in under-18 rugby. *BMJ Open Sport Exerc. Med.* 2:e000053, 2016. <https://doi.org/10.1136/bmjsem-2015-000053>.
 - ³³Holbourn, A. S. Mechanics of head injuries. *The Lancet* 09:438, 1943.
 - ³⁴Jordan, B. D. Chronic traumatic brain injury associated with boxing. *Semin. Neurol.* 20:179–185, 2000.
 - ³⁵Kerr, Z. Y., J. K. Register-Mihalik, M. C. Kay, J. D. DeFreese, S. W. Marshall, and K. M. Guskiewicz. Concussion nondisclosure during professional career among a cohort of former National Football League athletes. *Am. J. Sports Med.* 46(1):22–29, 2018.
 - ³⁶Kerr, Z. Y., J. K. Register-Mihalik, E. Kroshus, C. M. Baugh, and S. W. Marshall. Motivations associated with nondisclosure of self-reported concussions in former collegiate athletes. *Am. J. Sports Med.* 44(1):220–225, 2016.
 - ³⁷Kerr, Z. Y., J. K. Register-Mihalik, S. W. Marshall, K. R. Evenson, J. P. Mihalik, and K. M. Guskiewicz. Disclosure and non-disclosure of concussion and concussion symptoms in athletes: review and application of the socio-ecological framework. *Brain Inj.* 28(8):1009–1021, 2014.
 - ³⁸Kieffer, E. E., M. T. Begonia, A. M. Tyson, et al. A two-phased approach to quantifying head impact sensor accuracy: in-laboratory and on-field assessments. *Ann. Biomed. Eng.* 2020. <https://doi.org/10.1007/s10439-020-02647-1>.
 - ³⁹Kieffer, E. E., G. Pierce, C. Vaillancourt, and S. Rowson. In-season concussion symptom presentation in men's and women's rugby. *Biomed. Sci. Instrum.* 55(2):236–241, 2019.
 - ⁴⁰Kleiven, S. Influence of impact direction on the human head in prediction of subdural hematoma. *J. Neurotrauma* 20(4):365–379, 2003.
 - ⁴¹Kleiven S. Influence of Direction and Duration of Impacts to the Human Head Evaluated Using the Finite Element Method. International Research Council on the Biomechanics of Impact (IRCOBI), 2005, pp. 41–57.
 - ⁴²Kuo, C., L. Wu, J. Loza, D. Senif, S. C. Anderson, and D. B. Camarillo. Comparison of video-based and sensor-based head impact exposure. *PLoS ONE* 13(6):e0199238, 2018.
 - ⁴³Langois, J. A., W. Rutland-Brown, and M. M. Wald. The epidemiology and impact of traumatic brain injury: a brief overview. *J. Head Trauma Rehabil.* 21:375–378, 2006.
 - ⁴⁴Lessley, D. J., R. W. Kent, J. R. Funk, et al. Video analysis of reported concussion events in the National Football League during the 2015–2016 and 2016–2017 seasons. *Am. J. Sports Med.* 46(14):3502–3510, 2018.
 - ⁴⁵Liu, Y., A. G. Domel, S. A. Yousefsani, et al. Validation and comparison of instrumented mouthguards for measuring head kinematics and assessing brain deformation in football impacts. *Ann. Biomed. Eng.* 2020. <https://doi.org/10.1007/s10439-020-02629-3>.
 - ⁴⁶Makdissi, M., and G. Davis. The reliability and validity of video analysis for the assessment of the clinical signs of concussion in Australian football. *J. SciMed. Sport.* 19(10):859–863, 2016.
 - ⁴⁷Makdissi, M., and G. Davis. Using video analysis for concussion surveillance in Australian football. *J. SciMed. Sport.* 19(12):958–963, 2016.
 - ⁴⁸McCrea, M., T. Hammeke, G. Olsen, P. Leo, and K. Guskiewicz. Unreported concussion in high school football players: implications for prevention. *Clin. J. Sport Med.* 14:13–17, 2004.
 - ⁴⁹McIntosh, A. S., P. McCrory, and J. Comerford. The dynamics of concussive head impacts in rugby and Australian rules football. *Med. Sci. Sports Exerc.* 32(12):1980–1984, 2000.

- ⁵⁰McIntosh, A. S., D. A. Patton, B. Fréchède, *et al.* The biomechanics of concussion in unhelmeted football players in Australia: a case-control study. *BMJ Open* 4:e005078, 2014.
- ⁵¹McKee, 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.
- ⁵²Meaney, D. F., B. Morrison, and C. D. Bass. 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.* 136:021008, 2014.
- ⁵³Newman J, Barr C, Beusenberg M, Fournier E, Shewchenko N, Welbourne E, Withnall C. A New Biomechanical Assessment of Mild Traumatic Brain Injury Part 2 – Results and Conclusions. International Research Council on Biomechanics of Injury (IRCOBI), 2000.
- ⁵⁴Newman, 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>.
- ⁵⁵O'Connor, K. L., S. Rowson, S. M. Duma, and S. P. Broglio. Head-impact-measurement devices: a systematic review. *J. Athlet. Train.* 52(3):206–227, 2017.
- ⁵⁶Omalu, B. I., S. T. DeKosky, R. L. Minster, M. I. Kamboh, R. L. Hamilton, and C. H. Wecht. Chronic traumatic encephalopathy in a National Football League player. *Neurosurgery.* 57:128–134, 2005.
- ⁵⁷Omalu, B. I., R. P. Fitzsimmons, J. Hammers, and J. Bailes. Chronic traumatic encephalopathy in a professional American wrestler. *J. Forens. Nurs.* 6:130–136, 2010.
- ⁵⁸Patrick 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.
- ⁵⁹Patton, D. A. A review of instrumented equipment to investigate head impacts in sport. *Appl. Bion. Biomech.* 2016. <https://doi.org/10.1155/2016/7049743>.
- ⁶⁰Patton, D. A., C. M. Huber, C. C. McDonald, S. S. Margulies, C. L. Master, and K. B. Arbogast. Video confirmation of head impact sensor data from high school soccer players. *Am. J. Sports Med.* 48(5):1246–1253, 2020.
- ⁶¹Pellman, 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.
- ⁶²Plassman, B. L., R. J. Havlik, D. C. Steffens, M. J. Helms, T. N. Newman, D. Drosdick, *et al.* Documented head injury in early adulthood and risk of Alzheimer's disease and other dementias. *Neurology.* 55:1158–1166, 2000.
- ⁶³Press, J. N., and S. Rowson. Quantifying head impact exposure in collegiate women's soccer. *Clin. J. Sport Med.* 27(2):104–110, 2017.
- ⁶⁴Rowson, S., and S. M. Duma. Brain injury prediction: assessing the combined probability of concussion using linear and rotational head acceleration. *Ann. Biomed. Eng.* 41(5):873–882, 2013.
- ⁶⁵Rowson, B., J. D. Stitzel, and S. M. Duma. Special issue on the NCAA-DoD CARE consortium research. *Ann. Biomed. Eng.* 47:2045–2047, 2019.
- ⁶⁶Siegmund, G. P., K. M. Guskiewicz, S. W. Marshall, A. L. DeMarco, and S. J. Bonin. Laboratory validation of two wearable sensor systems for measuring head impact severity in football players. *Ann. Biomed. Eng.* 44:1257–1274, 2016.
- ⁶⁷Stapp, 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.
- ⁶⁸Stemper, B. D., A. S. Shah, J. Harezlak, *et al.* Comparison of head impact exposure between concussed football athletes and matched controls: evidence for a possible second mechanism of sport-related concussion. *Ann. Biomed. Eng.* 47:2057–2072, 2019.
- ⁶⁹Thomas, M., T. S. Haas, J. J. Doerer, J. S. Hodges, B. O. Aicher, R. F. Garberich, *et al.* Epidemiology of sudden death in young, competitive athletes due to blunt trauma. *Pediatrics* 128:e1–e8, 2011.
- ⁷⁰Tyson, A. M., S. M. Duma, and S. Rowson. Laboratory evaluation of low-cost wearable sensors for measuring head impacts in sports. *J. Appl. Biomech.* 34(4):320–326, 2018.
- ⁷¹Wu, L. C., C. Kuo, J. Loza, M. Kurt, K. Laksari, L. Z. Yanez, D. Senif, S. C. Anderson, L. E. Miller, J. E. Urban, J. D. Stitzel, and D. B. Camarillo. Detection of American Football Head impacts using biomechanical features and support vector machine classification. *Sci. Rep.* 8:855, 2018.
- ⁷²Zhang, L., K. H. Yang, and A. I. King. Comparison of brain responses between frontal and lateral impacts by finite element modeling. *J. Neurotrauma* 18(1):21–30, 2001.
- ⁷³Zuckerman, S. L., R. J. Elbin, A. K. Sills, J. R. Crandall, D. J. Lessley, C. M. Moran, C. D. Moran, and G. S. Solomon. Concussions in the National Football League: the evolution of video review for assessing the frequency and reliability of visible signs. *Phys Sportsmed.* 2020. <https://doi.org/10.1080/00913847.2020.1731379>.

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Summary of Accurate and Precise Head Impacts Collected On-Ice in Hockey Players and How These Data Will be Used to Improve Equipment and Rules for Safer Play

Citation

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ABSTRACT

In this study, we used a laboratory-calibrated impact monitoring mouthguard (IMM) system and video review to document the presence of 77 potential blunt and inertial head impacts across 291 athlete exposures in ice hockey players aged 14 to 18. Our rate of impacts per athlete exposure was one to two orders of magnitude lower than prior studies, and we did not measure any high gravity impacts in the range 100 g to 200 g. We found four cases where the athlete sustained a "No-Go" impact per the National Football League (NFL) video review criteria. The No-Go cases were of similar impact dose severity and direction with No-Go athletes in American football and combat sport. We also measured a high-energy frontal impact dose that resulted in no visible impairment, and this also was

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on par with visibly normal athletes in American football and combat sport studied previously. The inertial and blunt impact time trace data collected in this study give a full six-degrees-of-freedom kinematic profile of true ice hockey head impacts measured in the “living laboratory.” This detailed knowledge, for the first time, permits creation of helmet tests per the on-ice exposures where designs can acutely focus on reducing impact energy dose. Assuming skull fracture risks have been sufficiently minimized, reduction of energy transfer to the brain in side and rear impacts should be the paramount design considerations. We suggest that for players to be monitored using a mouthguard-based system, coaches and referees must enforce current safety rules on mouthguard use during play. Stringent enforcement of rules against checking from behind would help further reduce instances of the already rare No-Go impacts. And while relatively few impacts occurred in our ice hockey population, it would be prudent to investigate effects of cumulative impact dosing in a broader set of athletes longitudinally.

Keywords

concussion, impact dose, velocity, energy, acceleration, mouthguard

Introduction

The original head injury biomechanical theories were developed by Dr. A. H. S. Holbourn of Oxford University in 1943.¹ Holbourn split head injuries into two categories: (1) localized brain injury due to skull distortion and fracture and (2) velocity change of the head where the skull is sufficiently rigid and remains undistorted. The second category is where athlete concussion biomechanics reside, and Holbourn made a seminal observation that, “The injury is proportional to the change of velocity of the head and not the acceleration.”

In the three decades that followed, much intense research was devoted to development of head injury risk curves, primarily to determine skull fracture tolerance via quantification of impact acceleration in blunt force loading. The reason acceleration was chosen as the primary metric was that it could be derived from force plate readings, as well as the fact that linear accelerometers were technologically advanced enough to directly mount to the skull in experiments. Systems to measure the full six-degrees-of-freedom (6DOF) spatial and temporal impact experience would arrive in the following decades.

The result of work by Gadd² and Versace³ was empirical derivation of the Wayne State Tolerance Curve (WSTC), Severity Index (SI), and Head Injury Criterion (HIC). These metrics became the means by which to estimate skull fracture risk from scalar linear acceleration, and they remain in use today in automotive crash and helmet safety testing.

Nearly six decades after Holbourn and three decades after Gadd and Versace, Newman and his co-authors^{4,5} formalized Holbourn’s velocity theory by conducting empirical analysis of professional American football collisions recreated in the laboratory through extensive video reconstructions and photogrammetry. Newman

built upon Holbourn's velocity model by including spatial and temporal parameters. He confirmed both Holbourn's theories about velocity change being most predictive of a deleterious outcome and a long-held clinical anecdote that the brain has directional sensitivity to similar impact magnitudes delivered to different impact locations on the head. For example, Newman found that the impacts of similar kinetic energy transfer and power had a more devastating effect on the players struck laterally versus in the front of the head.

In the years following Newman's findings, much work was conducted using the Head Impact Telemetry System (HITS), a first-of-its-kind system created by Simbex and commercialized by Riddell originally for use in American football helmets.⁶ The HITS is an instrument that many researchers have used to report population-based statistics and individual impact cases focusing on scalar acceleration peaks, SI, HIC, and impact direction. After reviewing American football HITS data and clinical outcomes, Guskiewicz and Mihalik concluded, in 2011, that "a concussive injury threshold is elusive and may, in fact, be irrelevant when predicting clinical outcome."⁷ This conclusion considered thirteen single concussive impacts to the top, front, side, and back of the head where scalar accelerations ranged from 60 g to 170 g and 200 rad/s² to 15,000 rad/s².

In ice hockey, a modified version of the American football HITS has been used in several studies. McPherson et al.⁸ instrumented $n = 13$ U14 players and found approximately ten head impacts per athlete-exposure (A-E). In this study, eleven impacts in the range of 100 g to 200 g were measured, and none of these high-gravity impacts resulted in a diagnosed concussion. Mihalik et al.⁹ studied $n = 52$ U14 and U16 players and found one to two head impacts per A-E. They also reported no concussions with the 95th percentile impact magnitude at 40 g to 50 g. Allison et al.¹⁰ conducted laboratory calibration tests of an HITS helmet on a Hybrid III crash dummy head. In these tests to the side, rear oblique, and rear, the single impact uncertainty was found to be 25% for scalar linear acceleration peak and 50% for scalar angular acceleration peak across the calibration ranges. Finally, Smith et al.¹¹ synthesized research on $n = 56$ ice hockey players aged 16 to 21 and found approximately seven impacts per A-E. In this study, there were 21 players with high-gravity impacts between 100 g and 200 g, and six of these players sustained a diagnosed concussion.

The more contemporary acceleration-based investigations using HITS data in American football and ice hockey, as well as the more classic velocity-based injury models of Holbourn and Newman, have generated significant knowledge and research around concussion mechanisms and measurement. Most recently, this has led to several calls to improve measurement accuracy as well as to deepen the population of low-uncertainty head kinematics data captured from the "living laboratory"¹² in the literature.

In 2014, the United States Institute of Medicine concluded that, "Available studies of head injury biomechanics have identified the importance of linear and rotational movements of the head in injury causation," and "There are currently inadequate data to define the direction- and age-related thresholds for linear and

rotational acceleration specifically associated with concussions.”¹³ Several authors have also recently advocated that “Accurate measures of individual exposure will yield a direct estimate of the human tolerance...”¹² and “as more accurate sensors are designed...”¹⁴ “valid methods of measuring the direction and severity of on-field head impacts are needed.”¹⁵

In studies of American football and combat sports athletes,^{16,17} Bartsch and coauthors found a correlation between single high-energy impact doses to the side and back of the head in the range of 40 J to 110 J and visible impairment on video as described by the National Football League’s (NFL) “No-Go” criteria.¹⁸ In these studies, single frontal impacts in the range of 50 J to 70 J did *not* cause visible impairment. These results aligned very closely with Newman’s predictions,^{4,5} Holbourn’s theory,¹ and recent laboratory experiments.¹⁹ Additionally, computational models from football impacts²⁰ predicted consistently higher principal strains in the genu, corpus callosum, and hippocampus for impacts with higher kinetic energy dosing but not necessarily higher scalar acceleration peaks.

In this study, we aimed to utilize the living laboratory as a data collection site for the full 6DOF dynamic spatial and temporal factors—inclusive of accelerations and velocities—that may lead to visible impairment in ice hockey. By directly measuring these parameters with a low-uncertainty, well-calibrated impact monitoring mouthguard (IMM), we ultimately aim to assist all stakeholders in improving athlete safety, protecting the positive benefits of ice hockey participation and advancing head impact science.

Materials and Methods

In this study, we used an IMM system embedded in an athlete’s mouthpiece and affixed to the upper dentition to quantify head impacts ([fig. 1](#)). The IMM system provides 6DOF spatial and temporal estimates of linear and angular accelerations and velocities, as well as impact location and direction on the head during blunt and inertial events. The IMM is an impact monitor that estimates motion of the skull during a head impact and does not provide clinical concussion assessment or diagnostic information. In this study, we used two types of IMM instruments to acquire 6DOF data: (1) a twelve-channel version that measures linear accelerations at the dentition (“12a”) and (2) a six-channel version that measures linear acceleration at the dentition and skull angular velocity (“3a3omega”). Both systems require standard rigid body kinematic transforms to compute linear kinematics at the head center of gravity (CG), and signal-noise ratio can be reduced when an impact has a frequency of 0.5 kHz+, as well as when the IMM is worn improperly.

The principal of IMM operation is that an impact event triggers data collection when an axis of acceleration exceeds a preset threshold of 5 g to 15 g; the data are processed onboard in firmware using a four-pole Butterworth low-pass filter at 200 Hz and then are transmitted wirelessly over Bluetooth low energy to a nearby mobile application, which then uploads the data to a secure cloud-based portal. The

FIG. 1 (Top) Head impact monitoring system embedded within an athletics mouthguard (Prevent Biometrics, Minneapolis, MN). The system is comprised of an impact monitoring mouthguard (IMM), a solo case for wireless charging and sanitization of a single IMM, a team case that stores and charges 27 individual IMMs, an iOS-based team application that wirelessly pulls data from IMM in real time, and a web portal where data are wirelessly transmitted and securely stored. Videos were recorded at each data collection session and time-synchronized to IMM data to confirm true impacts. (Bottom) The IMM system provides 6DOF spatial and temporal information on each head impact dose, including time traces of linear and angular accelerations and velocities as well as impact direction and location on the head.

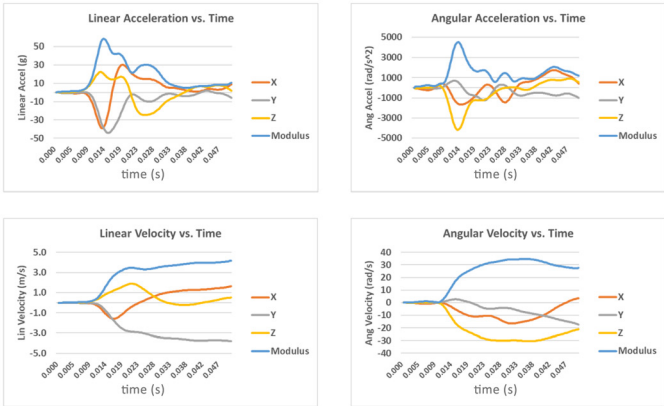
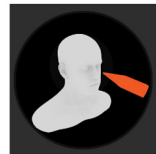
HEAD IMPACT MONITORING SYSTEM



IMPACT PARAMETERS

Spatial, Temporal Outputs:

- Linear Acceleration XYZ(t), Modulus(t)
- Angular Acceleration XYZ(t), Modulus(t)
- Linear Velocity XYZ(t), Modulus(t)
- Angular Velocity XYZ(t), Modulus(t)
- Impact Direction
- Impact Location



sample rate is 3.2 kHz; data are collected for 50 ms, and 460 impacts can be stored onboard. Onboard hardware switches confirm proper wearing of the IMM during gameplay. In hockey, the typical real-time communication distance ranges from approximately 20 m to 50 m, depending on whether the athlete is facing toward or away from the mobile application device. More extensive descriptions and background on the IMM can be found elsewhere.^{17,21–23}

LABORATORY CALIBRATION

A total of $n = 1,011$ calibration tests were performed to confirm IMM versus reference peak linear acceleration at the CG (PLA), peak angular acceleration (PAA), peak linear velocity at the CG (PLV), and peak angular velocity (PAV) data fit a linear model of the form $y = mx + b$. We used a range of laboratory test conditions that have been developed to bracket the expected ranges of spatial and temporal impacts experienced by helmeted athletes on the field of play.^{15,16,21,22,24,25}

These test conditions generate first harmonic frequency content in the range of 20 to 100 Hz—nominal pulse width of approximately 10 to 20 ms—such as what is seen on the field of play during helmeted impacts.^{16,20,21,23} Low-gravity, inertial motions, with a first harmonic in the range of 5 to 20 Hz and nominal pulse width of 50-plus ms, were not calibrated in the laboratory.

Acceleration comparisons were made by time synchronizing the IMM and REF pulses and extracting values at the time of maximum REF acceleration. Velocity estimates were computed when the primary acceleration pulse returned to nearly zero, approximately 5 ms after the peak acceleration.²¹ Based on the frequency content of output REF signals, a four-pole, low-pass Butterworth filter with a corner frequency of 200 Hz was used for REF and IMM data. Padded and helmeted impacts were delivered using a linear pneumatic impactor and Hybrid III headform with a fixed mandible ($n = 431$), a Hybrid III headform with an articulating mandible ($n = 132$), as well as using a pendulum with a National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform and fixed mandible ($n = 448$).

ATHLETE IMPACT DOSES

Quantifying an athlete head impact dose in the living laboratory is challenging and requires some extra scrutiny and verification steps. The IMM system must satisfy the following conditions to permit the output data to be deemed *trustworthy by inference*:

1. The impact dose must be in the rigid body range with a first harmonic less than 500 Hz. This condition is described more fully by the impact model in Bartsch et al.²¹
2. The IMM must have adequate coupling to the skull to generate a high signal-noise ratio measurement at the sensing location and to permit low-noise computations of kinematics at the head center of gravity. The measured and computed time traces can be investigated to confirm coupling adequacy as in Bartsch et al.²¹

3. The laboratory quantification of IMM uncertainty in a single impact under adequate coupling conditions must be known. Several authors have published on test methods to quantify uncertainty.^{15,16,21,22,24,25}
4. On the playing field, qualitative video confirmation of skull motion due to an impact or inertial event must be obtained. Kuo et al.²⁶ and Joseph et al.²⁷ describe these methods.
5. Each of the on-field head impact doses reported by the IMM system must:
 - a. Be within ranges of laboratory calibrations
 - b. Be time-synchronized to video confirmation
 - c. Qualitatively mimic video-observed impact direction, location, and severity
 - d. Confirm physically realistic motion of the rigid body skull^{21,28}
 - e. Be reported with consideration of the single dose uncertainty for the impact conditions

Of course, it is impossible to conclusively prove each of these conditions are satisfied in 100% of the impact doses collected in the real world because of the subjective nature of reviewing low-gravity collisions on the scale of 5 J (approximately 20 g) and under or when reviewing video of potential collisions where contact is obstructed by gameplay and poor quality film work. In these cases, additional information provided by the IMM system, such as hardware signals that confirm proper wearing of the instrument, and scrutiny of XYZ and scalar kinematic time traces of accelerations and velocities for physically realistic 6DOF head motion can be vital.

In this study, the IMM system was deployed in $n = 291$ athlete-exposures (281 A-E boys, 10 A-E girls) in a total of 95 athletes aged 14 to 18 and participating in bantam and high-school ice hockey games. All gameplay was reviewed on video, and potential head impact cases were identified. After video review, the potential head impact cases were compared to their time-synchronized data traces collected by the IMM. The data traces were further analyzed to confirm the estimated kinematics produced realistic motion, as well as that the impact direction and location on video was qualitatively like the data traces.

All potential head impact cases identified on video were further scrutinized by study staff per the NFL's No-Go video spotter criteria¹⁸ to document any instances of visible impairment post impact. The NFL No-Go criteria rely upon observers who watch gameplay and review individual replays to identify any player who is clearly unstable or displays any potential concussion signs following a hit to the head. While the NFL has had No-Go observers in place for several years, recently organizations such as the National Hockey League (NHL), Major League Soccer (MLS), and the National Collegiate Athletic Administration (NCAA) have adopted a similar model.

There were approximately fifty events that triggered data collection per player-game during times when the athlete was not enduring contact. These events, generally characterized by high-frequency, nonsensical kinematic profiles (see Bartsch et al.²¹ for more details) were also video confirmed, categorized as false positive impact data, and removed from the analysis.

Results

LABORATORY CALIBRATION

The scalar PLA, PAA, PLV, and PAV comparison of IMM to reference fit a linear model within the ranges of calibration for the n=1,011 laboratory tests (fig. 2). The uncertainty in a single-impact peak parameter was 7%, 17%, 5%, and 12% for

FIG. 2 Comparison of PLA, PAA, PLV, and PAV for IMM versus laboratory reference. All data fit a linear model.

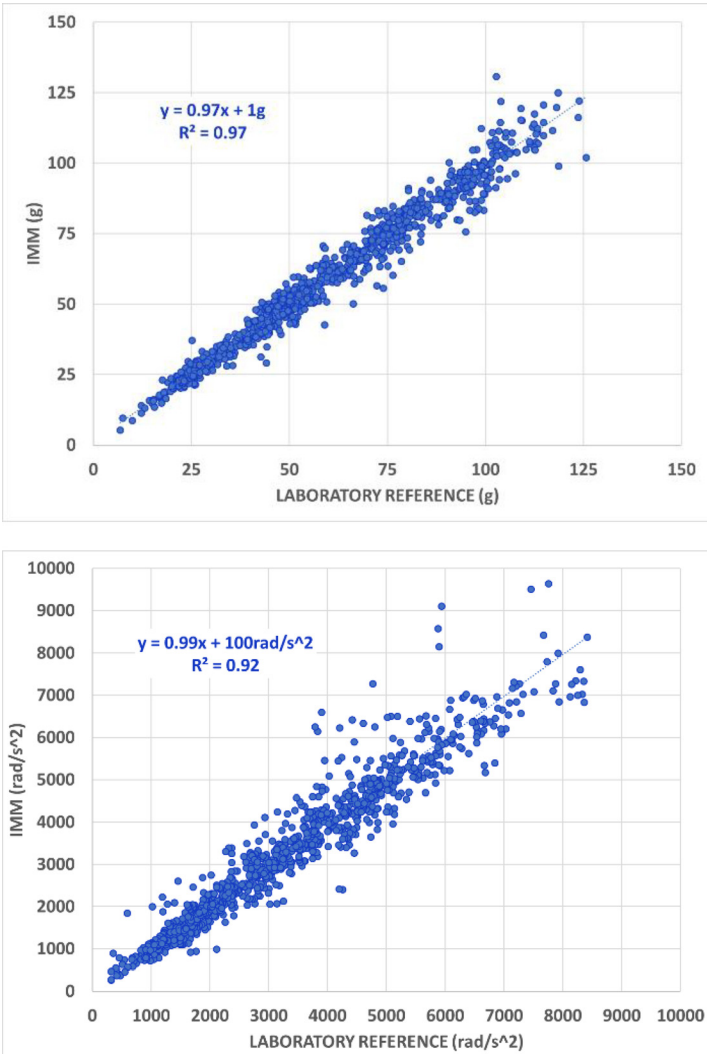
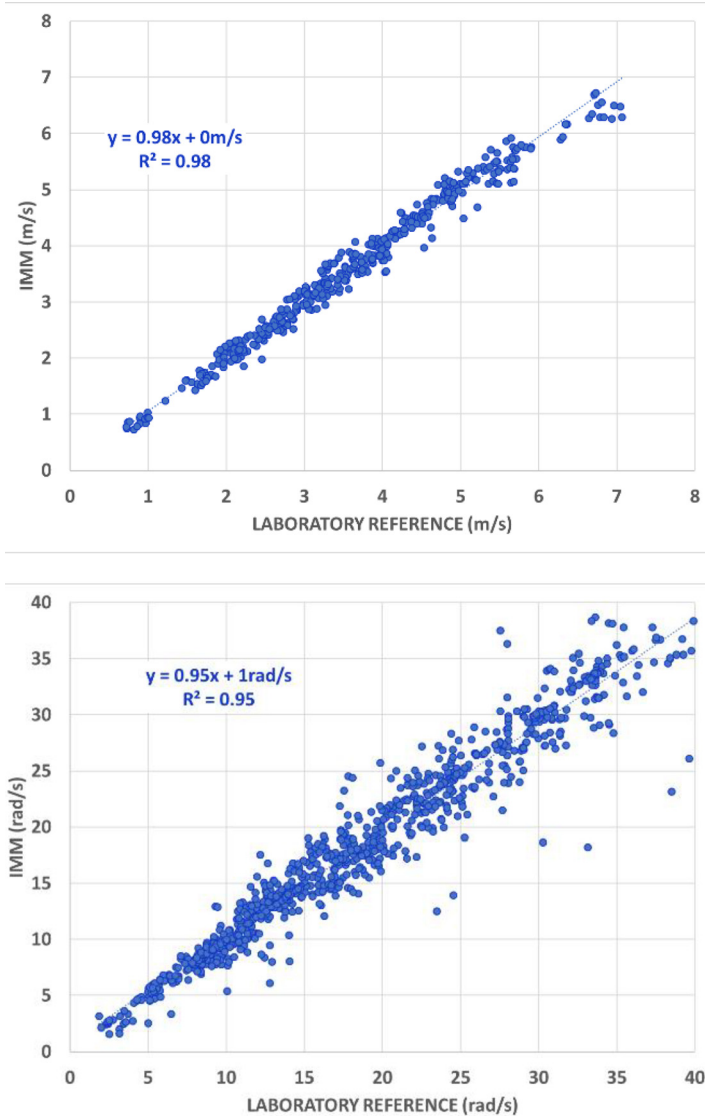


FIG. 2 (continued)

PLA, PAA, PLV, and PAV, respectively, across the ranges of calibration. The first harmonic frequency in these tests ranged from 50 Hz to 90 Hz, resulting in a typical pulse width of about 10 to 20 ms.

ATHLETE IMPACT DOSES

Out of the 291 A-E in the ice hockey athletes aged 14 to 18, we found $n = 77$ potential head impact events identified on video. The impact rate was 0.3 per A-E.

Approximately 10% of A-E resulted in one or more impact events identified on video. In 29 video-confirmed events, athlete contact occurred, but the athlete had the IMM off their teeth during the contact. Therefore, the signal-to-noise ratio was too low to reliably report the 6DOF magnitudes in these events. Examples of IMM-collected data are given in [figure 3](#) for impact and inertial events, as well as for an event created during an impact when the IMM was worn improperly.

A total of 48 video-verified true positive head impacts were captured where an athlete was properly wearing the IMM. Approximately 90% of these impacts were to the front and side of the head. The summary of the comparison of linear and angular velocities and accelerations is given in [figure 4](#) along with a comparison to previously published data from combat sports and American football.^{16,17}

FIG. 3 Examples (left) of an impact event and inertial event and (right) a time trace from an improperly worn IMM.

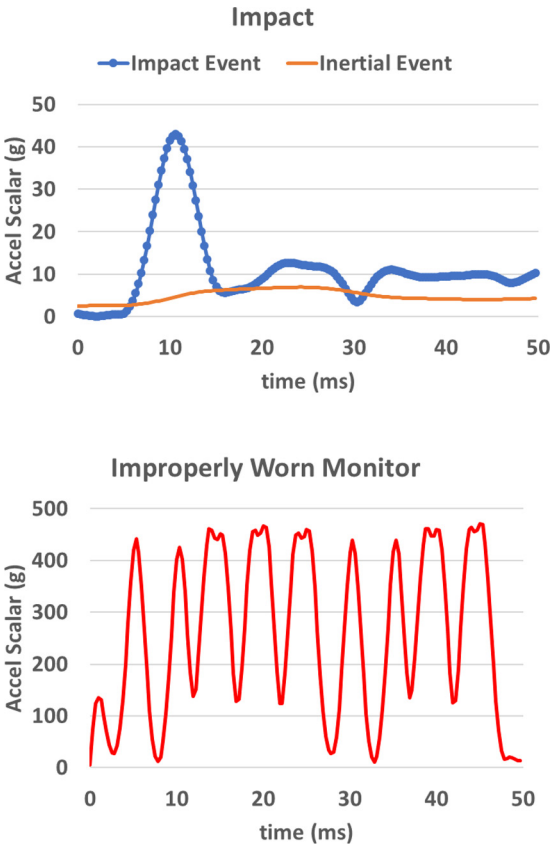
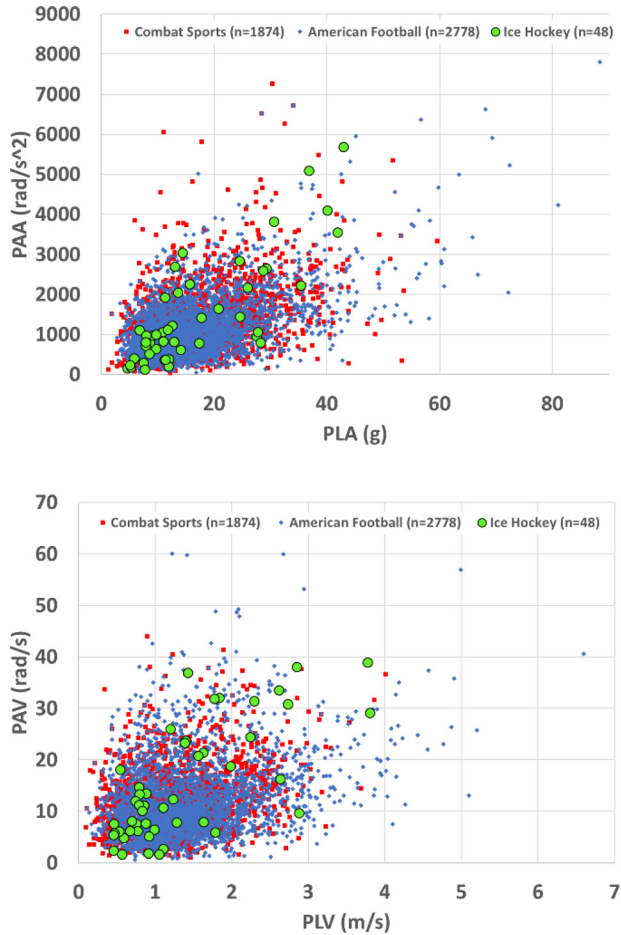


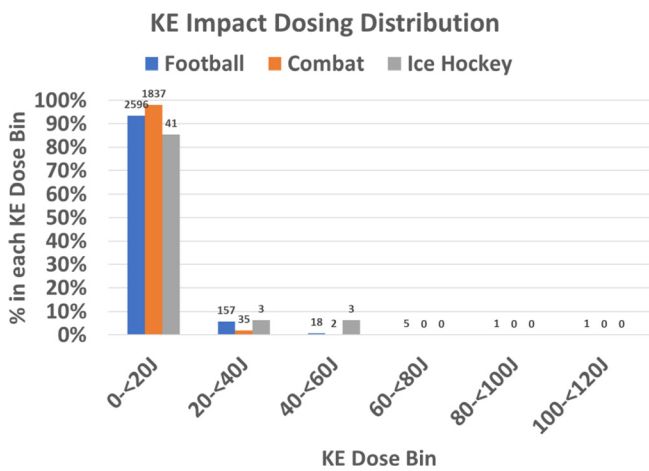
FIG. 4 Comparisons of (left) PLA versus PAA and (right) PLV versus PAV for the $n = 48$ video-verified impacts in this study, plotted alongside prior data published on video-verified impacts in combat sports and American football.^{16,17} Ice hockey is plotted as filled circles, combat sports as filled squares, and American football as filled diamonds.



The median impact dose was 5 J, and 85% of impacts were less than 20 J; this distribution is shown in [figure 5](#) and contrasted with the doses reported in combat sports and American football. The most impacts video-verified for a single player in one A-E was four absorbed by a female player.

The 48 video-verified true positive impacts were split into two categories: blunt and inertial. Blunt impacts were of the typical helmeted-impact variety, with a

FIG. 5 Kinetic energy impact dose distribution in ice hockey in the current study, contrasted with distributions previously reported in American football and combat sports.



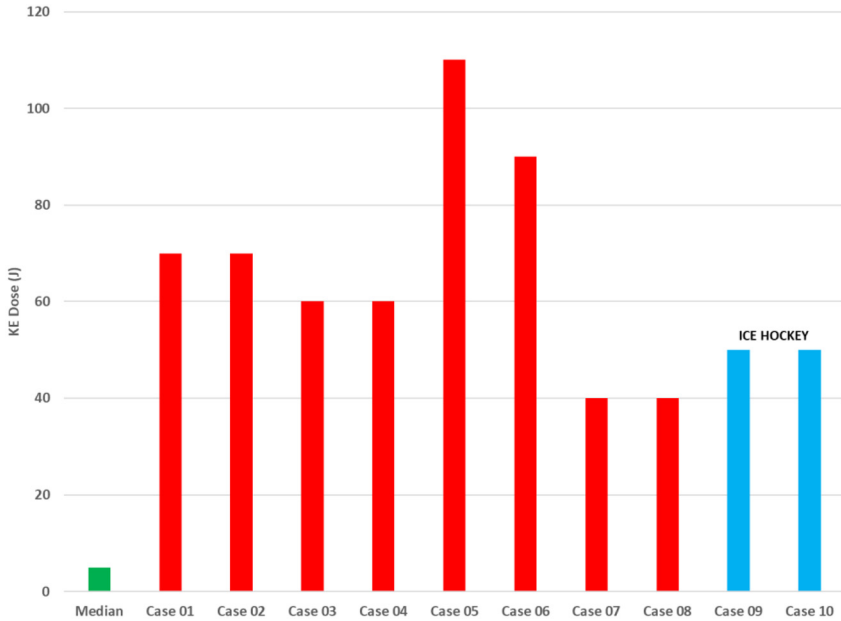
nominal pulse width of approximately 10 to 20 ms and a first harmonic less than 100 Hz. Inertial events were characterized by time durations approximately two to four times as long, with the first harmonic as low as 10 Hz to 20 Hz. There were $n = 22$ blunt and 26 inertial impacts, and these data are summarized in [table 1](#).

During the study, there were four No-Go impact events. In two of these cases, the athlete was wearing their IMM on the upper dentition and the data collected confirmed the video observations. These cases involved penalty infractions, with one athlete illegally hit from the rear into the boards by two opposing players, and the other athlete illegally hit high and from the side by an opposing player. In these No-Go players, the kinetic energy dosing was 50 J. These two players’ impact doses are compared to previously published No-Go cases from American football and combat sports in [figure 6](#).

TABLE 1 Summary of blunt and inertial ice hockey impact events captured in this study

VIDEO-VERIFIED BLUNT IMPACTS					
Count	Median (PLA)	Median (PAA)	Median (PAV)	Median (PLV)	Median Dose (J)
22	27	2200	24	1.8	17
VIDEO-VERIFIED INERTIAL EVENTS					
Count	Median (PLA)	Median (PAA)	Median (PAV)	Median (PLV)	Median Dose (J)
26	9	700	7	0.8	2

FIG. 6 No-Go cases from ice hockey in this study (Cases 09 and 10) compared with eight previously published No-Go cases in American football (Cases 01 to 08). All No-Go cases are from impact events to the side and back of the head. Combat sports had zero No-Go cases.



In two other cases, the athlete was not wearing the IMM properly and the signal-noise ratio was too low to compare 6DOF kinematics to video observations. These cases involved one player illegally struck from behind into the boards by an opponent and a player accidentally struck in the helmet by an opposing player's wrist shot.

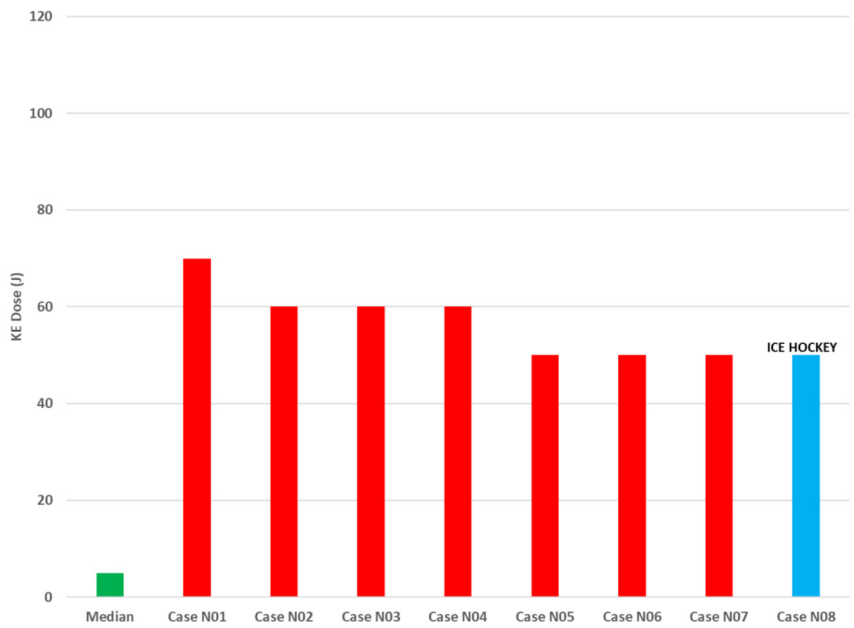
There was also one player who sustained a high-dose frontal impact of 50 J and displayed no visible impairment. This impact dose is compared to previously published high-dose frontal impact cases from American football and combat sports in [figure 7](#); in these cases, no visible impairment was observed.

Discussion

LABORATORY CALIBRATION

As can be seen in [figure 2](#), the laboratory calibration tests fit a linear model for PLA, PAA, PLV, and PAV parameters. Single-impact uncertainty in these parameters ranged from 7% to 17% across the calibration ranges. The laboratory calibrations were also confirmed to bracket the ranges of 6DOF spatial and temporal

FIG. 7 High-dose frontal impact in ice hockey where player had zero visible impairment compared with seven other high-dose frontal impact cases in American football (Case N01 through N06) and combat sports (Case N07) without visible impairment.



kinematics measured on ice, with the frequency content of 50 Hz to 90 Hz, with a nominal pulse width of 10 to 20 ms, and are comparable to the athlete-measured IMM data traces. If future data collection yields true positive impacts outside of these ranges of calibration, it may be necessary to recalibrate the IMM system across a broader range.

ATHLETE IMPACT DOSES

We video-verified 0.3 on-ice impacts per A-E and deemed these doses “trustworthy by inference” based on the verification approach described in the methods. All impact doses were within the ranges of laboratory calibration and could be assumed to have comparable uncertainty in their spatial and temporal outputs.

Compared to prior studies using similarly aged and skilled players instrumented with the HITS hockey system, we found impact exposure one order of magnitude lower than Mihalik et al.⁹ and two orders of magnitude lower than McPherson et al.⁸ and Smith et al.¹¹ Further, we did not find significant numbers of impacts in the 100 g to 200 g range like McPherson et al.⁸ and Smith et al.¹¹ The reasons for order(s) of magnitude differences in impacts per A-E and impact

magnitude for each study are likely in the style of play among the populations studied, as well as the degree to which impact dose verification methods were employed in each study.

The median impact was 5 J; more than 60% of impacts we collected were less than 10 J and 85% were less than 20 J. This means that the two No-Go cases measuring 50 J were both in the top 1% of measured data and were outliers in terms of their severity. Further, the two No-Go impacts were both (1) in the range of single-impact energy dosing and (2) had the same impact directions (rear/side) found for No-Go events in football and combat sport athletes of similar ages.

Our 50-J high-dose frontal ice hockey also aligns well with high-energy impacts that failed to cause visible impairment previously collected in American football and combat athletes of a similar age. The reason why this is a compelling finding is that historically, it has been challenging to identify a physical parameter and impact direction that causes, or fails to cause, impairment in human subjects instrumented on the field of play.

Single-impact dose directional sensitivity to side/rear impacts versus frontal impacts was predicted by Newman et al.⁴ and Newman, Shewchenko, and Welbourne,⁵ yet it has never been clearly demonstrated when comparing scalar peak acceleration data to clinical observations.^{7,29} We feel the correlation between single-impact direction, kinetic energy dosing, and visible impairment in our ice hockey players aligning with prior findings in American footballers and combat sport athletes is the most important finding of this study. And the fact that our findings in ice hockey, American football, and combat sport athletes support the injury model put forth by Holbourn and formalized by Newman and coauthors gives us assurance that the living laboratory findings are beginning to provide empirical confirmation of long-held rigid body head injury theory. Our results presented here require more extensive proving out to confirm the findings. Therefore, future studies need to compare impact direction, impact energy dose, and visible impairment to document the athlete's injury sensitivity.

Conclusions

In this study, we used a laboratory-calibrated IMM system and video review to document the presence of 77 potential head impacts across 291 A-Es in ice hockey players aged 14 to 18. We found 29 impacts where the athlete improperly wore the IMM, 26 inertial impacts, and 22 blunt impacts. We found four cases where the athlete sustained a No-Go impact per the NFL video review criteria. In two of these cases, the athlete was improperly wearing the IMM. In order for the IMM system to be used to its full potential, it is desirable to have referees, coaches, and players abide by, and reinforce, existing rules that require the athlete to wear a protective mouthguard on their teeth at all times on the ice.

In the other two cases, the athletes sustained impacts to the rear and side with kinetic energy dosing of 50 J, on par with similar doses sustained by No-Go

athletes in American football and combat sports of similar ages. We also measured a 50-J frontal impact that resulted in no visible impairment, and this also was on par with visibly normal athletes in American football and combat sports studied previously.

The inertial and blunt impact time trace data collected in this study give a full six-degrees-of-freedom kinematic profile of true ice hockey head impacts measured in the living laboratory. This detailed knowledge, for the first time, permits creation of helmet tests per the on-ice exposures where designs can acutely focus on reducing impact energy dose. Assuming skull fracture risks have been sufficiently minimized, reduction of energy transfer to the brain in side and rear impacts should be the paramount design constraints. This finding confirms energy-based head impact models developed first by Holbourn in the 1940s and formalized by Newman around the turn of the century.

Regarding ice hockey rules and safety, we suggest that for players to be monitored using a mouthguard-based system, coaches and referees must enforce current safety rules that mandate athletes always wear a mouthguard on their teeth during play. Additionally, stringent enforcement of rules against checking from behind would help reduce instances of rare No-Go impacts.

In the future, computational modeling of impact and inertial events, especially concentrating on No-Go impacts, can also help determine if brain directional sensitivity exists. And while relatively few impacts occurred in our ice hockey population, it would be prudent to investigate effects of cumulative impact dosing in a broader set of athletes longitudinally.

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AB is an employee of Prevent Biometrics and has rights to equity, stock options, and a bonus program. Cleveland Clinic has entered into an exclusive license agreement with Prevent Biometrics for Intelligent Mouthguard technology on which AB, EB, SS, and VM are inventors. Under terms of these agreements, there is potential for fee-for-service payments, royalties, or equity (or any combination thereof) arising from commercialization of head impact measurement or head impact safety product(s), or both. AB, EB, and SS receive research support from the U.S. Department of

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Time traces of the head impact data are available upon request.

References

1. A. S. Holbourn, "Mechanics of Head Injuries," *Lancet* 242 (1943): 438–441.
2. C. W. Gadd, "Use of a Weighted-Impulse Criterion for Estimating Injury Hazard," in *Proceedings of the Tenth Stapp Car Crash Conference* (Warrendale, PA: SAE International, 1966), 164–174.
3. J. Versace, "A Review of the Severity Index," in *Proceedings of the Fifteenth Stapp Car Crash Conference* (Warrendale, PA: SAE International, 1971), 771–796.
4. J. A. Newman, M. Beusenbergh, E. Fournier, N. Shewchenko, C. Withnall, A. King, K. Yank, L. Zhang, J. McElhaney, L. Thibault, et al., "A New Biomechanical Assessment of Mild Traumatic Brain Injury. Part I: Methodology," in *Proceedings of the 1999 International IRCOBI Conference on the Biomechanics of Impact* (Stiges, Spain: IRCOBI, 1999), 17–36.
5. J. A. Newman, N. Shewchenko, and E. Welbourne, "A Proposed New Biomechanical Head Injury Assessment Function—The Maximum Power Index," *Stapp Car Crash Journal* 44 (2000): 1–35.
6. J. J. Crisco, J. J. Chu, and R. M. Greenwald, "An Algorithm for Estimating Acceleration Magnitude and Impact Location Using Multiple Nonorthogonal Single-Axis Accelerometers," *Journal of Biomechanical Engineering* 126, no. 6 (2004): 849–854.
7. K. M. Guskiewicz and J. P. Mihalik, "Biomechanics of Sport Concussion: Quest for the Elusive Injury Threshold," *Exercise and Sport Science Reviews* 39 (2011): 4–11.
8. 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>
9. J. P. Mihalik, K. M. Guskiewicz, S. W. Marshall, J. T. Blackburn, R. C. Cantu, and R. M. Greenwald, "Head Impact Biomechanics in Youth Hockey: Comparisons Across Playing Position, Event Types, and Impact Locations," *Annals of Biomedical Engineering* 40, no. 1 (2012): 141–149.
10. M. A. Allison, Y. S. Kang, J. H. Bolte, M. R. Maltese, and K. B. Arbogast, "Validation of a Helmet-Based System to Measure Head Impact Biomechanics in Ice Hockey," *Medicine & Science in Sports & Exercise* 46, no. 1 (2014): 115–123.
11. 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.
12. D. F. Meaney, B. Morrison, and C. D. Bass, "The Mechanics of Traumatic Brain Injury: A Review of What We Know and What We Need to Know for Reducing Its Societal Burden," *Journal of Biomechanical Engineering* 136, no. 2 (2014), <https://doi.org/10.1115/1.4026364>
13. 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).

14. M. R. Elliott, S. S. Margulies, M. R. Maltese, and K. B. Arbogast, "Accounting for Sampling Variability, Injury Under-Reporting, and Sensor Error in Concussion Injury Risk Curves," *Journal of Biomechanics* 48 (2015): 3059–3065.
15. G. P. Siegmund, K. M. Guskiewicz, S. W. Marshall, A. L. DeMarco, and S. J. Bonin, "Laboratory Validation of Two Wearable Sensor Systems for Measuring Head Impact Severity in Football Players," *Annals of Biomedical Engineering* 44 (2016): 1257–1274.
16. A. J. Bartsch, V. J. Miele, J. L. Alberts, E. C. Benzel, A. Shah, J. Humm, B. S. Stemper, and M. M. McCrea, "Estimates of High-Risk Single and Cumulative Head Impact Doses in American Football," *Biomedical Sciences Instrumentation* 55, no. 2 (2019): 224–229.
17. A. J. Bartsch, J. L. Alberts, E. C. Benzel, J. Cruickshank, E. Jansen, S. Samorezov, J. Humm, A. Shah, M. M. McCrea, and B. S. Stemper, "Engineering Characteristics of Single Head Impacts in Sport that Cause Functional Impairment" (paper presentation, International Research Council on the Biomechanics of Injury [IRCOBI] Europe Conference, Florence, Italy, September 11–13, 2019).
18. National Football League, *NFL Head, Neck and Spine Committee's Concussion Diagnosis and Management Protocol*, amended June 19, 2018 (New York: NFL, 2018), 1–16.
19. R. J. Fijalkowski, B. D. Stemper, F. A. Pintar, N. Yoganandan, M. J. Cowe, and T. A. Gennarelli, "New Rat Model for Diffuse Brain Injury Using Coronal Plane Angular Acceleration," *Journal of Neurotrauma* 24 (2007): 1387–1398.
20. W. Zhao, A. J. Bartsch, E. C. Benzel, V. J. Miele, B. D. Stemper, and S. Ji, "Regional Brain Injury Vulnerability from Two Finite Element Models of the Human Head" (paper presentation, International Research Council on the Biomechanics of Injury [IRCOBI] Europe Conference, Florence, Italy, September 11–13, 2019).
21. A. J. Bartsch, S. Samorezov, E. C. Benzel, V. J. Miele, and D. Brett, "Validation of an 'Intelligent Mouthguard' Single Event Head Impact Dosimeter," *STAPP Car Crash Journal* 58 (2014): 1–28.
22. D. S. Hedin, P. L. Gibson, A. J. Bartsch, and S. Samorezov, "Development of a Head Impact Monitoring 'Intelligent Mouthguard,'" *Conference Proceedings of the IEEE Engineering in Medicine and Biology Society* (2016): 2007–2009.
23. A. J. Bartsch, D. S. Hedin, P. L. Gibson, V. J. Miele, E. C. Benzel, J. L. Alberts, S. Samorezov, A. Shah, B. S. Stemper, and M. M. McCrea, "Laboratory and On-Field Data Collected by a Head Impact Monitoring Mouthguard" (paper presentation, 41st Annual International Conference of the IEEE Engineering in Medicine and Biology Society [EMBC], Berlin, Germany, July 23–27, 2019).
24. S. M. Duma and S. Rowson, "Using Sensors to Improve Sport Safety" (paper presentation, 39th Annual Meeting of the American Society of Biomechanics, Columbus, OH, August 5–8, 2015).
25. D. B. Camarillo, P. B. Shull, J. Mattson, R. Shultz, and D. Garza, "An Instrumented Mouthguard for Measuring Linear and Angular Head Impact Kinematics in American Football," *Annals of Biomedical Engineering* 41, no. 9 (2013): 1939–1949.
26. C. Kuo, L. Wu, J. Loza, D. Senif, S. C. Anderson, and D. B. Camarillo, "Comparison of Video-Based and Sensor-Based Head Impact Exposure," *PLoS ONE* 13, no. 6 (2018): e0199238.
27. K. Joseph, K. Gay, A. J. Bartsch, D. Camarillo, C. Taylor-Burds, S. Alai, J. Esterlitz, and P. Bellgowan, "Common Data Elements (CDEs) for Biomechanical Devices Used in Blunt Head Impact and Blast Exposure Dynamics: The National Institute of Neurological Disorders and Stroke (NINDS), National Institutes of Health (NIH) and Department of Defense (DOD) Version 1.0 Recommendations," *Neurology* 90, suppl. 15 (2018): P2.012.

28. L. C. Wu, K. Laksari, C. Kuo, J. F. Luck, S. Kleiven, C. R. Bass, and D. B. Camarillo, "Bandwidth and Sample Rate Requirements for Wearable Head Impact Sensors," *Journal of Biomechanics* 49, no. 13 (2016): 2918–2924.
29. B. S. Stemper, A. S. Shah, J. Harezlak, S. Rowson, J. P. Mihalik, S. M. Duma, L. D. Riggen, et al., "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 (2018): 2057–2072, <https://doi.org/10.1007/s10439-018-02136-6>

Laboratory and On-field Data Collected by a Head Impact Monitoring Mouthguard

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Abstract— Although concussion continues to be a major source of acute and chronic injury in automotive, athletic and military arenas, concussion injury mechanisms and risk functions are ill-defined. This lack of definition has hindered efforts to develop standardized concussion monitoring, safety testing and protective countermeasures. Recent research has provided evidence of the role of repetitive head impact exposure as a predisposing factor for the onset of concussion using developed instrumented helmets and mouthguards.

To overcome this knowledge gap, we have developed, tested and deployed a head impact monitoring mouthguard (IMM) system. In this study, we deployed the IMM system to gather high quality estimates of athlete head impacts in situ. And with enough longer-term data collection, potential concussive events or mild traumatic brain injuries (mTBIs) will be gathered and ideally will provide actionable risk-based threshold.

I. INTRODUCTION

Sports participation results in an estimated 1.6-3.8 million concussions annually [1] and the Centers for Disease Control and Prevention has designated concussions as a ‘silent epidemic’ [2]. Concussion incidence is as high as 20% annually in some sports [3], making athletic events ideal settings for concussion research [4–6].

Lacking real-time, trustworthy impact monitoring data, many concussions that occur during sports such as American football and soccer go unreported [6]. This is because concussions do not always result in loss and consciousness and athletes are reluctant to report symptoms so they can stay in the game. Un-assessed and undiagnosed head impacts can lead to death [7] and increase long-term risks for chronic traumatic encephalopathy (CTE) [8], Alzheimer’s and Parkinson’s [9]. Long term problems such as CTE developing later in life can have devastating results including personality changes, destroyed relationships and suicide [10]. CTE severity appears to be correlated with the number of years of playing football more so than the number of concussions reported [11]. Researchers therefore have

suggested that more sophisticated techniques than symptom assessment alone are needed to accurately assess damage due to head impact as it occurs especially when looking at long term problems [12].

II. BACKGROUND

Recent studies of football athletes, even in the absence of diagnosed concussion, have implicated repetitive head impact exposure (RHIE) in the onset of cognitive and brain structural changes [13]. As more is learned about the biomechanics of head impacts studies such as [13] suggest that accurate risk assessments can be provided which can be used to change behavior, techniques, rules, and reduce concussions. Paramount to achieving this is the development of a head impact monitoring system that is inexpensive, sport-agnostic, mass produced, reliable, easy to use, accurate, and able to reject non-head impact data.

III. METHODS

Impact Monitoring Mouthguard (IMM)

The IMM systems (Fig. 1) has been designed with two types of sensing packages. The first has twelve channels of linear acceleration (“12a”, ADXL375 or ADXL372) and the other is instrumented with three channels of linear acceleration and three channels of angular velocity (“3a3 ω ”, ST4200D, BMG250). The IMM samples at 3.2kHz, has real-time wireless data offload via BLE and stores 460 impacts on-board. Both systems have also been described elsewhere in detail [14-15]. Each IMM has an embedded flexible printed circuit board sealed between layers of FDA-grade mouthguard material. The IMM attaches to the upper dentition using slight interference fit between the mouthguard and the tooth embrasure spaces.



Figure 1. Head Impact Monitoring Mouthguard (IMM) System.

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A. Computational Approach

To evaluate scalar estimates of head impact severity, we used head impact dosimeter validity specifications, which were used in the NFL sponsored head impact dosimeter testing [16]. The IMM computations are based on a rigid body assumption. The rigid body point acceleration is described in (1), where $\mathbf{r}_p$ is a vector of constant length between a reference point O and a point of interest P, $\mathbf{a}_o$ is linear acceleration of the point O on the body, angular velocity is $\boldsymbol{\omega}$, and angular acceleration is $\dot{\boldsymbol{\omega}}$. In this case, the computations of relative acceleration and Coriolis acceleration are set to zero at each timestep.

$$\mathbf{a}_p = \mathbf{a}_o + \dot{\boldsymbol{\omega}} \times \mathbf{r}_p + \boldsymbol{\omega} \times (\boldsymbol{\omega} \times \mathbf{r}_p) \quad (1)$$

The computational method is not valid for situations when the skull is significantly deformed, or instrument-to-skull coupling is inadequate.

B. Laboratory Calibration

Laboratory calibration tests were conducted using a padded headform (n=751) or bare head (n=273), see Fig. 2.



Figure 2. Laboratory calibration tests.

Padded conditions included an American football helmet, boxing headgear or padded impactor against a bare head. Padded impacts generated first harmonic frequency content in the range of 20-100Hz, like what is seen on the field of play [14]. Bare head impacts omitted padding on the impactor and headform and simulated impacts with first harmonic in the range of 100-200Hz. Impacts were delivered using a linear pneumatic impactor and Hybrid III headform with fixed mandible (n=434 padded head and n=155 bare head, Fig. 1) or articulating mandible (n=132 padded head and n=23 bare head), as well as using a pendulum with NOCSAE headform and fixed mandible (n=185 padded head and n=95 bare head).

To confirm IMM coupling, specialized characterization tests (n=72 custom thermoformed IMM, n=72 injection molded boil-and-bite IMM) were carried out on an electrodynamic shaker that mimicked padded and bare head impact frequencies and amplitude (see Fig. 3).

C. On-field Data Collection

After calibrating the IMM system against Reference in the laboratory, n=54 amateur athletes in American football and boxing aged 11 to 22 were consented through Cleveland Clinic and Medical College of Wisconsin IRBs (CC13-899, MCW PRO00022950). Events with head center of gravity

linear acceleration (PLA) ranging 7g to 74g were collected during practices and games (see Fig. 4).

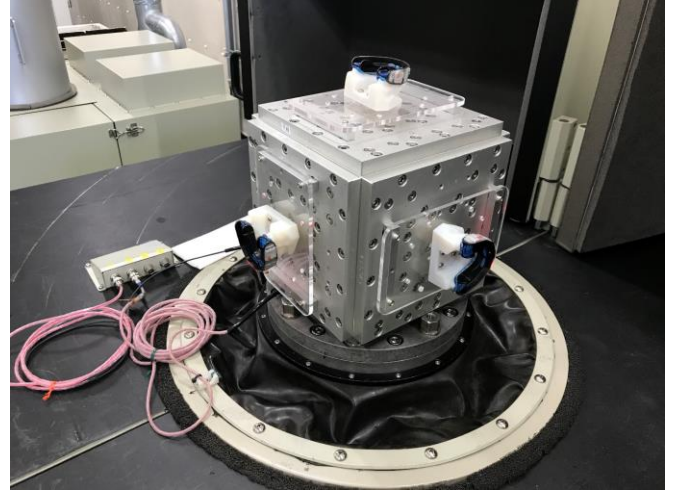


Figure 3. Electrodynamic shaker testing.



Figure 4. On-field Data Collection

Each event was time-synchronized to 30fps video collected of each activity. Any events collected when the athlete was not actively participating were considered false positives and discarded. Any event when the athlete was participating but not being impacted in the head was also considered a false positive. The remaining events were scrutinized to (a) confirm obvious head impact occurred in the video and (b) the kinematic traces computed at the head center of gravity were physically realistic and mimicked the impact observed on video [17-18].

IV. RESULTS

(a)

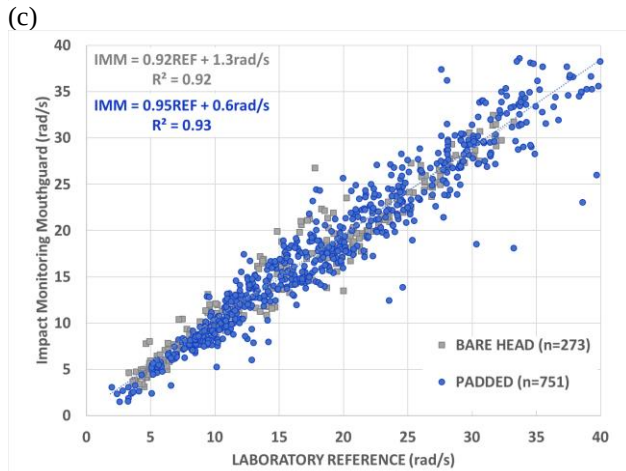
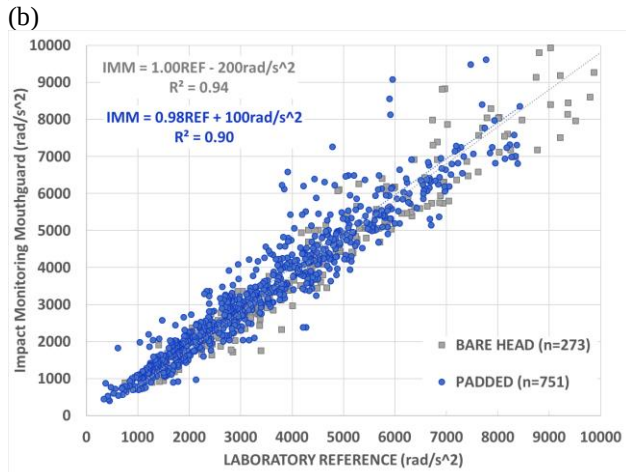
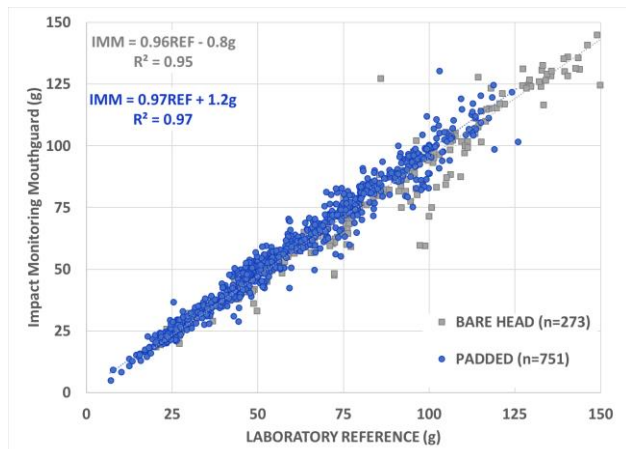


Figure 5. (a) PLA, (b) PAA, (c) PLV

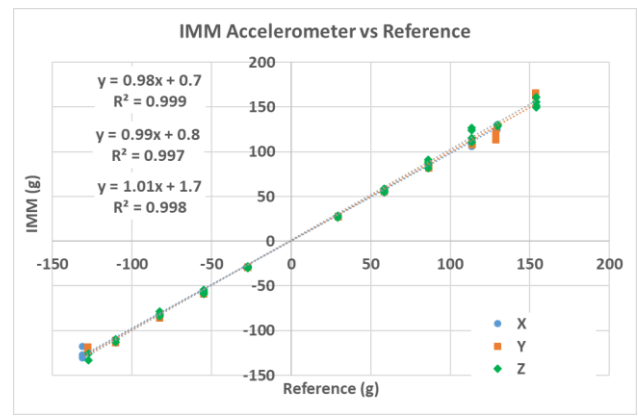


Figure 6. Shaker versus Reference

A. Laboratory

In the laboratory testing, padded and bare head impacts for IMM versus Reference fit a linear model close to the ideal linear model of form $y=x+0$, $R^2=1$ for linear acceleration (PLA), angular acceleration (PAA) and angular velocity (PAV) (Fig. 5a, 5b, 5c). Electrodynamical shaker tests confirmed (Fig. 6) that measured IMM kinematics mirrored those of the Reference for the duration of the pulse. Like the laboratory comparison to headform Reference sensors, the shaker test IMM outputs fit a linear model versus Reference. Custom IMM and Boil-and-bite IMM had similar outputs versus Reference. There was no evidence of decoupling observed in any of the shaker tests.

Adequate IMM coupling to dentition, knowledge of sensor positions and orientations with respect to Reference, and calibration of individual sensor outputs were required to achieve this level of accuracy. A non-trivial and proprietary processing algorithm was developed and validated to mitigate the space limitations – e.g. 5-10mm spread between vertically mounted accelerometers vs. 30-40mm spread horizontally – imposed inside the mouth.

B. On-field

During on-field data and video collection, we collected $n=890$ true positives confirmed after time synchronized video and time trace verification and here report peaks of the linear and angular accelerations (Fig. 7a). True Positives had first harmonic frequency in the range of 20-100Hz. "False positives" were triggered events on the IMM that contained high frequency kinematic data 200Hz+ not indicative of head motion, lower frequency kinematic data that generated nonsensical head motions or occurred during sequences where there was no head impact on video (Fig 7b).

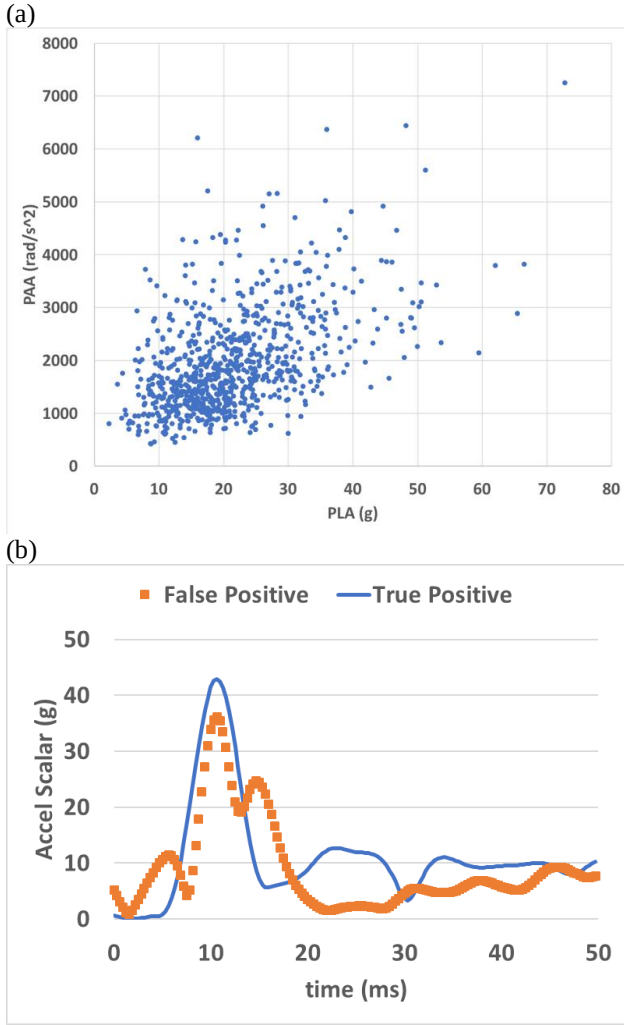


Figure 7. (a) PLA and PAA Collected in On-field Head Impacts. (b) Example of True Positive and False Positive Acceleration Traces

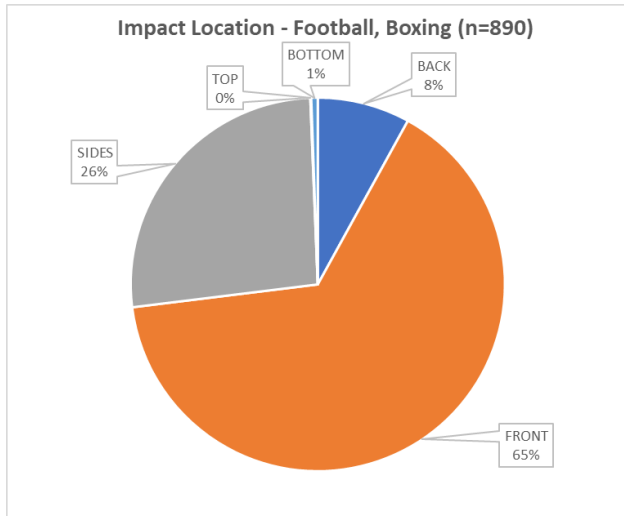


Figure 8. Impacts by Location

The median and 99th percentile of peak scalar linear acceleration (PLA) and peak angular acceleration (PAA) were 20g and 50g, and 1700rad/s² and 4600rad/s²,

respectively. Approximately 90% of impacts were to the front and sides of the head using methods of Greenwald et al. [19] (Fig. 8).

No concussions were diagnosed clinically. One of the athletes was removed from competition by athletic training staff after a single significant head impact which is an impact outlier in Fig. 7b.

V. DISCUSSION

A. Laboratory:

The results indicate that the IMM sensors have enough bandwidth and sample rate [20], suitable for measuring head impacts within the calibrated ranges and conditions of this laboratory study. The sites and impact severities of laboratory system level tests were within the ranges of published data on head impact dosimeters [14,16]. The system level test outputs also bracketed measured ranges of true positive single head impact doses measured in American football athletes [14].

B. On-field

True Positive single head impacts in this study were of an impulse type, i.e. rapid change of acceleration from zero to peak and back to zero, and then a longer time interval of a similar change in acceleration, but of opposite polarity and at lower amplitude. Such a profile is very similar to a typical data set from laboratory testing. Advancements in hardware and more sophisticated analysis methods will have to be employed to filter out more types of False Positive events with an even higher degree of confidence. A major challenge will be to examine signals with poorer signal-noise ratio due to less-than-adequate coupling.

C. IMM Considerations

Using IMM as a wireless sensing system has an advantage of being ubiquitous for several athletic and military activities. As a measurement device for kinematics-based characterization of a single head impact or severe-enough inertial motion, the IMM is subject to laboratory calibration and to verification that measured human impact dose data fall within the laboratory calibrated ranges. If prospectively collected human impact doses fall outside laboratory ranges tested here, the IMM system calibration ranges must be expanded to quantify uncertainty across the updated ranges of on-field impact doses.

VI. CONCLUSION

Based on results of over one thousand laboratory system level impact and shaker tests, we found that when compared to the Reference, IMM kinematic estimates fit a linear model with low uncertainty and high coefficient of determination for all tested impact directions/locations. Such levels of laboratory performance, in the authors' opinion, allow the use of an IMM system as a "single event head impact dosimeter", when severity of head impacts may be evaluated individually by their vector *and* scalar estimates.

The results of n=890 individually measured head impact kinematic time traces in amateur boxers and American

football players confirmed their time dependent vectors of spatial and temporal parameters were within the laboratory calibrated ranges of the IMM.

No athletes in our study sustained concussion. Further studies are required to clarify whether the uncertainty in head impact severity estimate or individual concussion tolerance variability is a cause of injury.

CONFLICT OF INTEREST STATEMENT

AJB is a full-time employee of Prevent Biometrics and has rights to employee stock options and bonus program.

Cleveland Clinic has entered into an exclusive license agreement with Prevent Biometrics for Intelligent Mouthguard technology on which AJB, ECB, VJM and SS are inventors. Under terms of these agreements there is potential for fee-for-service payments, royalties and/or equity arising from commercialization of head impact measurement and/or head impact safety product(s). The conflicts disclosed here are managed by the Cleveland Clinic Conflict Management Board. AJB, JLA, SS, ECB, DSH, PLG, MMM, AS and BSS receive research support from the United States Department of Defense, National Institutes of Health and/or Department of Transportation.

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REFERENCES

- [1] J. A. Langlois, W. Rutland-Brown, and M. M. Wald, "The epidemiology and impact of traumatic brain injury: a brief overview," *J Head Trauma Rehabil*, vol. 21, no. 5, pp. 375–378, Oct. 2006.
- [2] M. Faul, L. Xu, M. M. Wald, and V. G. Coronado, "Traumatic brain injury in the United States: emergency department visits, hospitalizations and deaths 2002–2006," *Atlanta, GA: Centers for Disease Control and Prevention, National Center for Injury Prevention and Control*, 2010.
- [3] K. M. Guskiewicz, B. L. Riemann, D. H. Perrin, and L. M. Nashner, "Alternative approaches to the assessment of mild head injury in athletes," *Medicine and science in sports and exercise*, vol. 29, pp. S213–S221, 1997.
- [4] M. McCrea, K. M. Guskiewicz, S. W. Marshall, W. Barr, C. Randolph, R. C. Cantu, J. A. Onate, J. Yang, and J. P. Kelly, "Acute effects and recovery time following concussion in collegiate football players," *JAMA: the journal of the American Medical Association*, vol. 290, no. 19, pp. 2556–2563, 2003.
- [5] E. J. Pellman, M. R. Lovell, D. C. Viano, I. R. Casson, and A. M. Tucker, "Concussion in professional football: neuropsychological testing-part 6," *Neurosurgery*, vol. 55, no. 6, pp. 1290–1305, 2004.
- [6] A. M. Spiotta, J. H. Shin, A. J. Bartsch, and E. C. Benzel, "Subconcussive impact in sports: a new era of awareness," *World neurosurgery*, vol. 75, no. 2, pp. 175–178, 2011.
- [7] R. C. Cantu and A. D. Gean, "Second-impact syndrome and a small subdural hematoma: an uncommon catastrophic result of repetitive head injury with a characteristic imaging appearance," *Journal of neurotrauma*, vol. 27, no. 9, pp. 1557–1564, 2010.
- [8] A. C. McKee, R. C. Cantu, C. J. Nowinski, E. T. Hedley-Whyte, B. E. Gavett, A. E. Budson, V. E. Santini, H.-S. Lee, C. A. Kubilus, and R. A. Stern, "Chronic traumatic encephalopathy in athletes: progressive tauopathy following repetitive head injury," *Journal of neuropathology and experimental neurology*, vol. 68, no. 7, p. 709, 2009.
- [9] D. H. Daneshvar, D. O. Riley, C. J. Nowinski, A. C. McKee, R. A. Stern, and R. C. Cantu, "Long-term consequences: effects on normal development profile after concussion," *Physical medicine and rehabilitation clinics of North America*, vol. 22, no. 4, pp. 683–700, 2011.
- [10] B. I. Omalu, J. Bailes, J. L. Hammers, and R. P. Fitzsimmons, "Chronic Traumatic Encephalopathy, Suicides and Parasuicides in Professional American Athletes: The Role of the Forensic Pathologist," *The American Journal of Forensic Medicine and Pathology*, vol. 31, no. 2, pp. 130–132, Jun. 2010.
- [11] A. C. McKee, T. D. Stein, C. J. Nowinski, R. A. Stern, D. H. Daneshvar, V. E. Alvarez, H.-S. Lee, G. Hall, S. M. Wojtowicz, C. M. Baugh, and others, "The spectrum of disease in chronic traumatic encephalopathy," *Brain*, vol. 136, no. 1, pp. 43–64, 2013.
- [12] K. Abbas, T. E. Shenk, V. N. Poole, E. L. Breedlove, L. J. Leverenz, E. A. Nauman, T. M. Talavage, and M. E. Robinson, "Alteration of default mode network in high school football athletes due to repetitive subconcussive mild traumatic brain injury: a resting-state functional magnetic resonance imaging study," *Brain connectivity*, vol. 5, no. 2, pp. 91–101, 2015.
- [13] B. D. Stemper, A. S. Shah, J. Harezlak, S. Rowson, J. P. Mihalik, S. M. Duma, L. D. Riggen, A. Brooks, K. L. Cameron, and D. Campbell, "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*, pp. 1–16, 2018.
- [14] A. J. Bartsch, S. Samorezov, E. C. Benzel, V. J. Miele, and D. Brett, "Validation of an 'Intelligent Mouthguard' Single Event Head Impact Dosimeter," *STAPP Car Crash Journal* 58 (November 2014): 1-28.
- [15] D. S. Hedin, P. L. Gibson, A. J. Bartsch, and S. Samorezov, "Development of a head impact monitoring 'Intelligent Mouthguard'," *IEEE Medicine and Biology*, 2016.
- [16] G. P. Siegmund, K. M. Guskiewicz, S. W. Marshall, A. L. DeMarco, and S. J. Bonin, "Laboratory Validation of Two Wearable Sensor Systems for Measuring Head Impact Severity in Football Players," *Ann Biomed Eng*. 2016;44:1257-74.
- [17] C. Kuo, L. Wu, J. Loza, D. Senif, S. C. Anderson, and D. B. Camarillo, "Comparison of video-based and sensor-based head impact exposure," *PloS one*, vol. 13, no. 6, p. e0199238, 2018.
- [18] K. Joseph, K. Gay, A. Bartsch, D. Camarillo, C. Taylor-Burds, S. Alai, J. Esterlitz, and P. Bellgowan, "Common Data Elements (CDEs) for Biomechanical Devices used in Blunt Head Impact and Blast Exposure Dynamics: The National Institute of Neurological Disorders and Stroke (NINDS), National Institutes of Health (NIH) and Department of Defense (DOD) Version 1.0 Recommendations (P2. 012)," *AAN Enterprises*, 2018.
- [19] R. M. Greenwald, J. T. Gwin, J. J. Chu, and J. J. Crisco, "Head impact severity measures for evaluating mild traumatic brain injury risk exposure," *Neurosurgery*, vol. 62, no. 4, p. 789, 2008.
- [20] L. C. Wu, K. Laksari, C. Kuo, J. F. Luck, S. Kleiven, R. Cameron, and D. B. Camarillo, "Bandwidth and sample rate requirements for wearable head impact sensors," *Journal of biomechanics*, vol. 49, no. 13, pp. 2918–2924, 2016.

Measuring Blunt Force Head Impacts in Athletes

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ABSTRACT Introduction: Although concussion continues to be a major source of acute and chronic injuries, concussion injury mechanisms and risk functions are ill-defined. This lack of definition has hindered efforts to develop standardized concussion monitoring, safety testing, and protective countermeasures. To overcome this knowledge gap, we have developed, tested, and deployed a head impact monitoring mouthguard (IMM) system. Materials and Methods: The IMM system was first calibrated in 731 laboratory tests. Versus reference, Laboratory IMM data fit a linear model, with results close to the ideal linear model of form $y = x + 0$, $R^2 = 1$. Next, during on-field play involving $n = 54$ amateur American athletes in football and boxing, there were tens of thousands of events collected by the IMM. A total of 890 true-positive head impacts were confirmed using a combination of signal processing and National Institute of Neurological Disorders and Stroke/National Institutes of Health Common Data Elements methods. Results: The median and 99th percentile of peak scalar linear acceleration and peak angular acceleration were 20 and 50 g and 1,700 and 4,600 rad/s², respectively. No athletes were diagnosed with concussion. Conclusions: While these data are useful for preliminary human tolerance limits, a larger population must be used to quantify real-world dose response as a function of impact magnitude, direction, location, and accumulation. This work is ongoing.

INTRODUCTION

Although concussion continues to be a major source of acute and chronic injuries in military, athletic, and automotive arenas, concussion injury mechanisms and risk functions are ill-defined. This lack of definition has hindered efforts to standardize concussion assessment, laboratory safety testing, and develop protective countermeasures.

BACKGROUND

In 2011, after considering scalar peak acceleration-based thresholds, leading concussion clinicians concluded “Recent studies suggest that a concussive injury threshold is elusive, and may, in fact, be irrelevant when predicting the clinical outcome”.¹ In 2014, the Institute of Medicine concluded that “Available studies of head injury biomechanics have identified the importance of linear and rotational movements of the head in injury causation . . .” and “there are currently inadequate data to define the direction- and age-related thresholds for linear and rotational acceleration specifically associated with concussions . . .”.² The collection of trustworthy impact

monitoring data as a means to address these inadequacies has also been acknowledged by many recently “accurate measures of individual exposure will yield a direct estimate of the human tolerance,”³ “as more accurate sensors are designed . . .”,⁴ “valid methods of measuring the direction and severity of on-field head impacts are needed.”⁵

These conclusions have been informed by scalar head impact dose distributions as opposed to spatial/temporal data from individual impacts. Impact scalar distributions indicate that scalar values of peak resultant accelerations alone are insufficient descriptors of head impact dynamics. When considering the vector nature of acceleration and velocity, as well as momentum-energy-restitution balance during impact, two substantially different spatially and temporally impacts, with the same scalar acceleration magnitude, may be correlated with substantially different tissue-level injuries due to complex anatomy and physiology of a human brain. Preclinical rodent and finite-element modeling studies have also highlighted the importance of temporal characteristics of the head rotational acceleration pulse by identifying significantly different behavioral outcomes and brain strains from head accelerations with identical peak acceleration magnitudes but differing durations.^{6,7}

A next step in correlating concussion and collision severity is to quantify, with the lowest practically achievable uncertainty, single head impact doses by time-dependent kinematic vectors of skull acceleration and velocity through their spatial and temporal parameters. Impact doses can then be applied as the skull boundary conditions in a finite-element brain model to estimate tissue level injury.⁸ Thus, the single or cumulative head impact dose, described in sufficient spatial/temporal detail, can be associated with injury risk in the “passive biomechanics laboratory”³ of an athletic field or military environment.

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To help overcome the knowledge gaps in acquiring spatial and temporal aspects of head impact, we have developed, tested, and deployed a head impact monitoring mouthguard (IMM) system. The IMM system was first calibrated in $n = 1,024$ laboratory tests against Hybrid III and National Operating Committee on Standards in Athletic Equipment (NOCSAE) headforms instrumented with reference kinematic sensors. Next, during on-field play involving $n = 54$ amateur American athletes in football and boxing, there were tens of thousands of kinematic signatures collected by the IMM. A total of $n = 890$ true-positive head impacts were confirmed using a combination of signal processing and video confirmation methods.^{9,10}

METHODS

Impact Monitoring Mouthguard

Two types of IMM systems exist; one system is instrumented with 12 channels of linear acceleration measurement (“12a”), and the other system is instrumented with three channels of linear acceleration and three channels of angular velocity (“3a3 ω ”). Both systems have been described elsewhere in detail.^{11,12} Each IMM has an embedded flexible printed circuit board sealed between layers of Food and Drug Administration (FDA)-grade mouthguard material. The IMM attaches to the upper dentition using slight interference fit between the mouthguard tray and the tooth embrasure spaces. The IMM includes “12a” or “3a3 ω ” kinematics measurement channels, along with associated microprocessor, battery, and data storage.

Computational Approach

To evaluate the quality of IMM scalar and vector estimates, we compared their spatial and temporal outputs versus instrumented reference headforms. To evaluate the scalar estimates of head impact severity, we used head impact dosimeter validity specifications, which were used in the National Football League (NFL)-sponsored head impact dosimeter testing⁵. The comparison is quantified through correlation of the IMM metrics versus corresponding reference values for all tested directions/locations combined. To evaluate vector estimates, we compared synchronized time traces of kinematic components between IMM and reference.

IMM computations are based on a rigid body assumption. The rigid body point acceleration is described in Equation (1), where $\mathbf{r}_p$ is a vector of constant length between a reference point O and a point of interest P, $\mathbf{a}_0$ is a linear acceleration of the point O on the body, angular velocity is $\boldsymbol{\omega}$, and angular acceleration is $\dot{\boldsymbol{\omega}}$. In this case, the deformable body computations of relative acceleration and Coriolis acceleration are set to zero at each timestep.

$$\mathbf{a}_p = \mathbf{a}_0 + \dot{\boldsymbol{\omega}} \times \mathbf{r}_p + \boldsymbol{\omega} \times (\boldsymbol{\omega} \times \mathbf{r}_p) \quad (1)$$

The combination of rigid body assumptions and the associated data processing steps comprise the data translation algorithm (DTA) computations. The DTA is focused on accurate estimates of time-varying vectors of acceleration and velocity where accelerations are high, but velocities are still low, and requires adequate coupling of IMM to the dentition. The computational method is not valid for situations when the skull is significantly deformed, or instrument-to-skull coupling is inadequate.

Laboratory Calibration

Laboratory calibration tests were conducted using a padded headform ($n = 751$) or bare head ($n = 273$). Padded conditions included an American football helmet, boxing headgear, or padded impactor against a bare head. Padded impacts generated first harmonic frequency content in the range of 20–100 Hz, similar to what is seen on the field of play.¹¹ Bare head impacts omitted padding on the impactor and headform and simulated impacts with first harmonic in the range of 100–200 Hz. Impacts were delivered using a linear pneumatic impactor and Hybrid III headform (Supplementary Fig. S1) with fixed mandible ($n = 434$ padded head and $n = 155$ bare head) or articulating mandible ($n = 132$ padded head and $n = 23$ bare head), as well as using a pendulum with NOCSAE headform and fixed mandible ($n = 185$ padded head and $n = 95$ bare head). Reference data collection for Hybrid III and NOCSAE headforms adhered to SAE J211 specification with adequate sample rate, full scale, and frequency response within the experimental passband of 20–200 Hz. To confirm IMM coupling, specialized characterization tests ($n = 72$ custom thermoformed IMM, $n = 72$ injection molded boil-and-bite IMM) were carried out on an electrodynamic shaker that mimicked padded and bare head impact frequencies and amplitude (Supplementary Fig. S1).

On-field Data Collection

After calibrating the impact monitoring system against reference in the laboratory, $n = 54$ amateur athletes in American football and boxing aged 11–22 were consented through Cleveland Clinic and Medical College of Wisconsin IRBs (CC13–899, MCW PRO00022950). Events with head center of gravity peak scalar linear acceleration (PLA) ranging from 7 to 74 g were collected during practices and games. Each event was time synchronized to 30 fps video collected of each activity. Any events collected when the athlete was not actively participating were considered false positives and discarded. Any event when the athlete was participating but not being impacted in the head was also considered a false positive. The remaining events were scrutinized to (1) confirm obvious head impact occurred in the video and (2) the kinematic traces computed at the head center of gravity were physically realistic and mimicked the impact observed on video.^{9,10}

RESULTS

In the laboratory testing, padded and bare head impacts for IMM versus reference fit a linear model close to the ideal linear model of form $y = x + 0$, $R^2 = 1$ for PLA, angular acceleration (PAA), and angular velocity (Fig. 1). Nontrivial processing algorithm, adequate IMM coupling to dentition, knowledge of sensor positions and orientations with respect to reference, and calibration of individual sensor outputs were required to achieve this level of accuracy. Good temporal agreement between IMM and reference (Fig. 2) indicated that both IMM instrumentation and DTA computations were adequate to the task of estimating accelerations and velocities of the reference.

Electrodynamic shaker tests confirmed that measured IMM kinematics mirrored those of the reference for the duration of the pulse. Like the laboratory comparison to headform reference sensors, the shaker test IMM outputs fit a linear model versus reference. Custom IMM and Boil-and-bite IMM had similar outputs versus reference. There was no evidence of decoupling observed in any of the shaker tests (Supplementary Fig. S2).

During on-field data and video collection, there were more than 100,000 triggering events, with $n = 890$ true positives confirmed after time-synchronized video and time trace verification. True positives had first harmonic frequency in the range of 20–100 Hz. “False positives” were triggered events on the IMM that contained high-frequency kinematic data 200 Hz + not indicative of head motion, lower frequency kinematic data that generated nonsensical head motions or occurred during sequences where there was no head impact on video.

The median and 99th percentile of PLA and PAA were 20 and 50 g and 1,700 and 4,600 rad/s², respectively. Approximately, 90% of impacts were to the front and sides of the head using methods of Greenwald et al.¹³ (Fig. 3).

No concussions were diagnosed clinically. One of the athletes was removed from competition by athletic training staff after a single significant head impact.

DISCUSSION

The goal of the IMM approach is to minimize contributions of measurement uncertainty to an estimate of injury risk due to head impact, and also account for likely directional sensitivity of human brain to head impact. All 100 g impacts “are equal,” when evaluated by peak acceleration scalar magnitude. “Some are more equal than others,” when evaluated as time varying vectors in three-dimensional space.

Laboratory

The results indicate that the IMM sensors have sufficient bandwidth and sample rate,¹⁴ suitable for measuring head impacts within the calibrated ranges and conditions of this laboratory

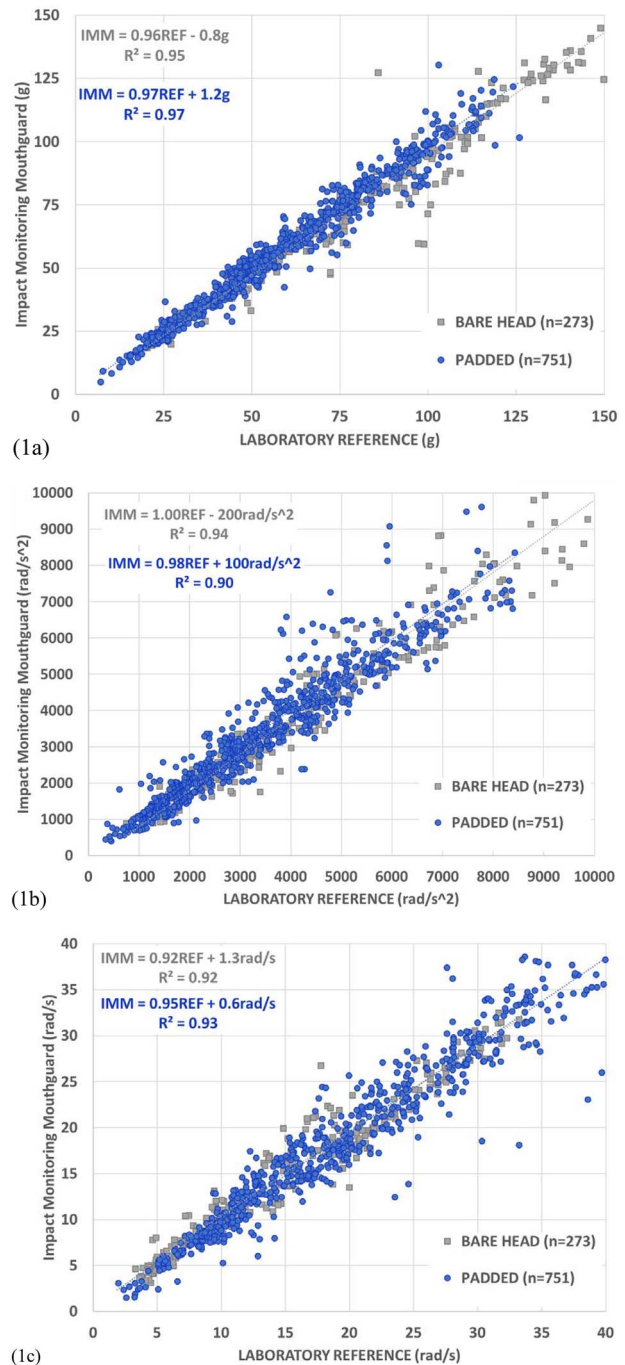


FIGURE 1. Laboratory results. Peak linear acceleration (PLA—a), peak angular acceleration (PAA—b), and peak angular velocity (PAV—c) for the IMM was compared with the laboratory reference in $n = 751$ padded and $n = 273$ bare head tests.

study. The IMM spatial and temporal impact estimates correlated well with reference for all tested impact directions and locations, based on comparison of X, Y, Z vector component and scalar peaks. The sites and impact severities of laboratory system level tests were within the ranges of published data on head impact dosimeters.^{5,11} The system-level test outputs also

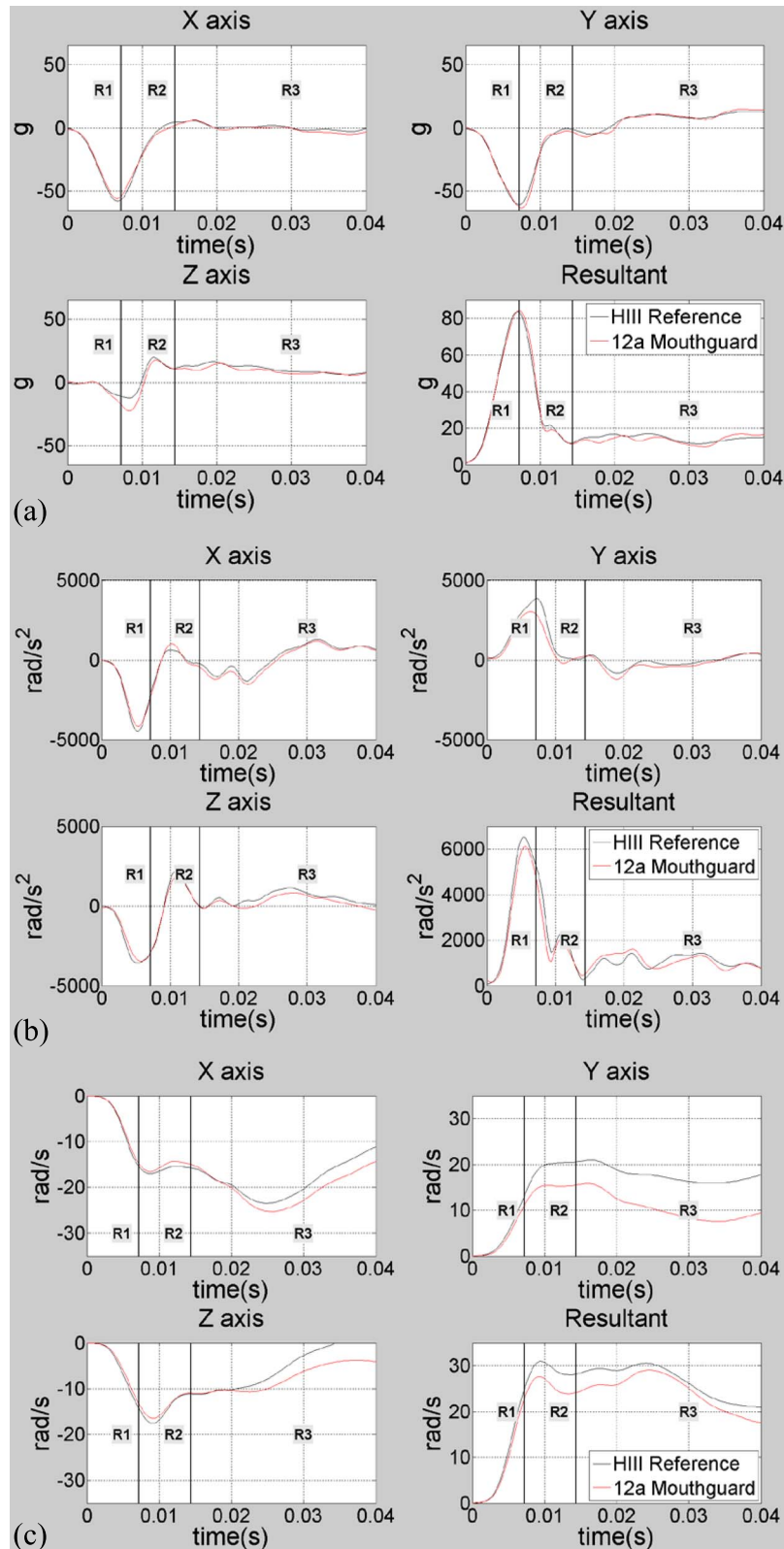
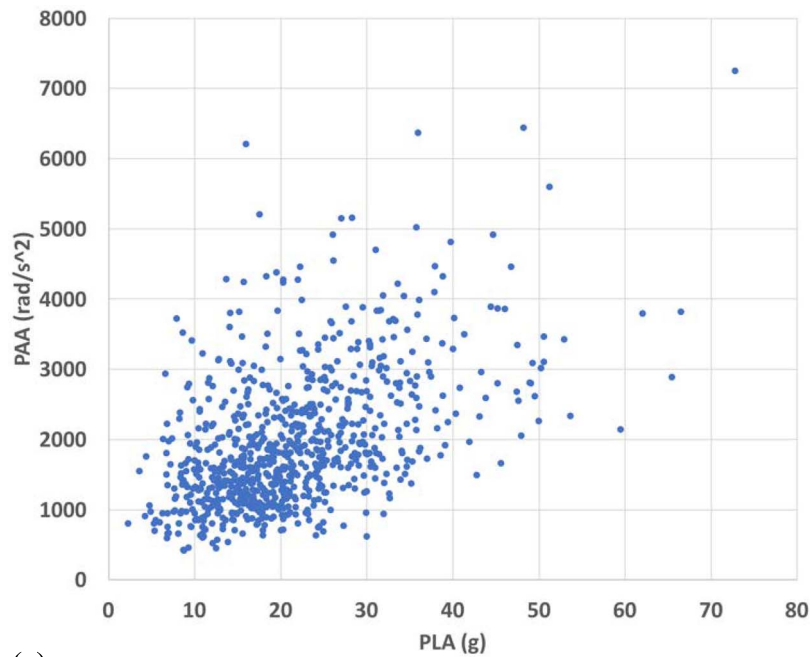
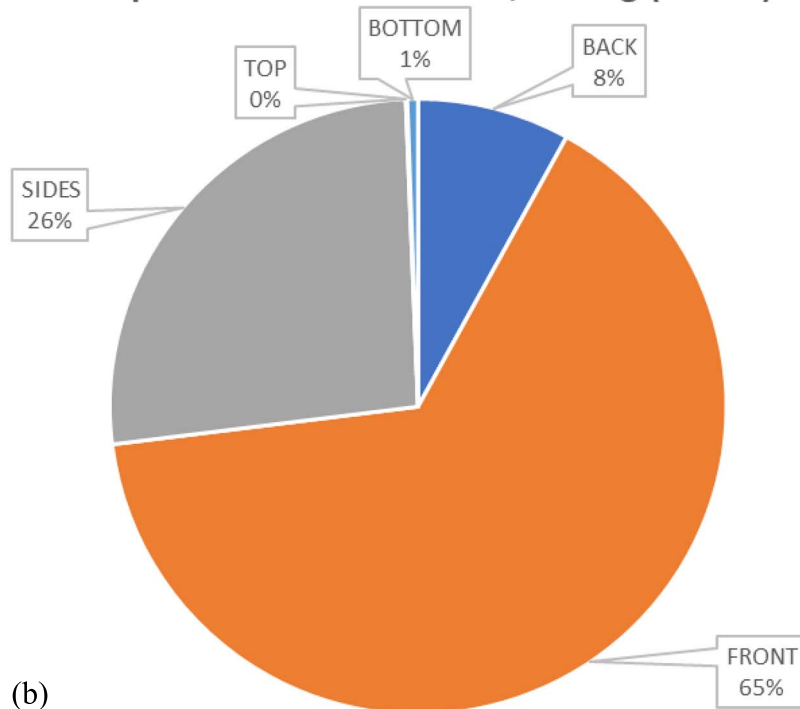


FIGURE 2. Example of linear acceleration (a), angular acceleration (b), and angular velocity (c) temporal comparison of IMM vector estimates versus reference Hybrid III measured kinematics in SAE J211 X-, Y-, Z-axes and scalar magnitude in a challenging oblique-front linear impact system test at 7.3 m/s. The assignment of regions R1, R2, and R3 are based on definitions given in Bartsch et al.¹¹.



(a)

Impact Location - Football, Boxing (n=890)



(b)

FIGURE 3. Results from $n = 890$ padded head impacts in American football and amateur boxing as reported by PLA versus PAA (a). Impact location—confirmed by video and data outputs—was 91% to the front and sides (b).

bracketed measured ranges of true-positive single head impact doses measured in American football athletes,¹¹ while being at the lower end of reported concussive head impacts estimated using instrumented helmets.^{1,13}

On-field

True-positive single head impacts in this study were of an impulse type, ie rapid change of acceleration from zero to peak (region R1) and back to zero (region R2), and then a longer

time interval of a similar change in acceleration, but of opposite polarity and at lower amplitude (region R3). Such a profile is very similar to a typical data set from laboratory testing. Approximating the acceleration in R1-R2 regions (duration T) by $(1 - \cos(2\pi \frac{t}{T}))$, one can obtain a frequency characteristic $f = \frac{1}{T}$ of the pulse. This frequency is nominally 20–100 Hz for padded impacts. And this frequency can be a criterion to provide separation of the true-positive head impact data sets from false-positive data sets. Advancements in hardware and more sophisticated analysis methods will have to be employed to filter out more types of false-positive events with an even higher degree of confidence. A major challenge will be to examine signals with poorer signal-to-noise ratio due to less-than-adequate coupling.

IMM Considerations

The low uncertainty of a single IMM head impact dose estimate comes at a cost: (1) well-fit mouthguard for adequate sensor-skull coupling, (2) sufficiently accurate knowledge of sensor positions and their sensitive axes' orientations, (3) calibration of IMM in the laboratory across on-field impact severity ranges, and (4) dealing with the fact that placing instrumentation inside the mouth imposes geometric limitations on the sensor arrangement, limits battery size, and requires protection from undesirable chemical and mechanical influences.

In spite of these costs, using IMM as a wireless sensing system has an advantage of being ubiquitous for a number of athletic and military activities. As a measurement device for kinematics-based characterization of a single head impact or severe-enough inertial motion, the IMM is subject to laboratory calibration and to verification that measured human impact dose data fall within the laboratory calibrated ranges. If prospectively collected human impact doses fall outside laboratory ranges tested here, the IMM system calibration ranges must be expanded to quantify uncertainty across the updated ranges of on-field impact doses.

CONCLUSIONS

Single or cumulative impact dose risk is a function of time-dependent **vectors**, characterized by their spatial and temporal parameters, but has previously been estimated by **scalars**. We aimed to better define impact **vectors** by monitoring single head impact doses as time-dependent vectors via an IMM with a minimal, while practical, measurement uncertainty.

Based on results of over 1,000 laboratory system-level impact and shaker tests, we found that when compared to the reference, IMM kinematic estimates fit a linear model with low uncertainty and high coefficient of determination for all tested impact directions/locations. Such levels of laboratory performance, in the authors' opinion, allow the use of an IMM system as a "single event head impact dosimeter," when severity of head impacts may be evaluated individually by their vector and scalar estimates.

The results of $n = 890$ individually measured head impact kinematic time traces in amateur boxers and American football players confirmed that their time-dependent vectors of spatial and temporal parameters were within the laboratory calibrated ranges of the IMM.

Comparison of scalar magnitudes of IMM head impact doses—leaving aside impact duration, location, and direction—with elite collegiate and professional American football concussion doses indicated that some of the head impacts in this study were at the lower end of elite American football single-dose concussive ranges as estimated in prior studies using instrumented helmets. No athletes in our study sustained concussion. Further studies are required to clarify whether the uncertainty in head impact severity estimate or individual concussion tolerance variability—or both—is the source of this discrepancy.

While these data are useful for understanding human tolerance to head impact, a larger population must be studied to quantify real-world head impact dose-response thresholds accurately and precisely as a function of impact magnitude, direction, location, and accumulation. This head impact dynamics data must also be correlated with sensitive and specific assessments of cognitive, executive, vision, and balance parameters.

SUPPLEMENTARY MATERIAL

Supplementary material is available at *MILMED* online.

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CONFLICT OF INTEREST STATEMENT

A.B. and R.D. are full-time employees of Prevent Biometrics and have rights to employee stock options and bonus program. Cleveland Clinic has entered into an exclusive license agreement with Prevent Biometrics for Intelligent Mouthguard technology on which A.J.B., E.C.B., V.J.M., and S.S. are inventors. Under terms of these agreements, there is potential for fee-for-service payments, royalties, and/or equity arising from commercialization of head impact measurement and/or head impact safety product(s). The conflicts disclosed here are managed by the Cleveland Clinic Conflict Management Board. Complete details can be found at <https://my.clevelandclinic.org/ccf/media/files/About/COI.pdf>. A.B., R.D., J.A., S.S., E.B., A.S., J.H., M.M., and B.S. receive research support from the U.S. Department of Defense, National Institutes of Health, and/or Department of Transportation.

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REFERENCES

1. Guskiewicz KM, Mihalik JP: Biomechanics of sport collision: quest for the elusive injury threshold. *Exerc Sport Sci Rev* 2011; 39: 4–11.
2. 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.
3. 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; pp. 1:14. 136
4. 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.
5. Siegmund GP, Guskiewicz KM, Marshall SW, DeMarco AL, Bonin SJ: Laboratory validation of two wearable sensor systems for measuring head impact severity in football players. *Ann Biomed Eng* 2016; 44: 1257–74.
6. Stemper BD, Shah AS, Pintar FA, et al: Head rotational acceleration characteristics influence behavioral and diffusion tensor imaging outcomes following concussion. *Ann Biomed Eng* 2015; 43(5): 1071–88. doi: [10.1007/s10439-014-1171-9](https://doi.org/10.1007/s10439-014-1171-9).
7. Lamy M, Baumgartner D, Yoganandan N, Stemper BD, Willinger R: Experimentally validated three-dimensional finite element model of the rat for mild traumatic brain injury. *Med Biol Eng Comput* 2013; 51(3): 353–65. doi: [10.1007/s11517-012-1004-7](https://doi.org/10.1007/s11517-012-1004-7).
8. McAllister TW, Ford JC, Ji S, et al: *Ann Biomed Eng* 2012; 40: 127. doi: [10.1007/s10439-011-0402-6](https://doi.org/10.1007/s10439-011-0402-6).
9. Kuo C, Wu L, Loza J, Senif D, Anderson SC, Camarillo DB: Comparison of video-based and sensor-based head impact exposure. *PLoS One* 2018; 13(6): e0199238. doi: [10.1371/journal.pone.0199238](https://doi.org/10.1371/journal.pone.0199238).
10. Gay K, Bartsch AJ, et al: Common data elements (CDEs) for biomechanical devices used in blunt head impact and blast exposure dynamics: the National Institute of Neurological Disorders and Stroke (NINDS), National Institutes of Health (NIH) and Department of Defense (DOD) version 1.0 recommendations. *Neurology* 91 (23 Supplement 1): S3.2-S3, December, 2018.
11. Bartsch AJ, Samorezov S, Benzel EC, Miele VJ, Brett D: Validation of an "Intelligent Mouthguard" single event head impact dosimeter. *Stapp Car Crash J* 2014; 58: 1–28.
12. Hedin DS, Gibson PL, Bartsch AJ, Samorezov S: Development of a head impact monitoring "Intelligent Mouthguard". *IEEE Med Biol* 2016.
13. Greenwald RM, Gwin JT, Chu JJ, Crisco JJ: Head impact severity measures for evaluating mild traumatic brain injury risk exposure. *Neurosurgery* 2008; 62: 789–98.
14. Wu LC, Laksari K, Kuo C, et al: Bandwidth and sample rate requirements for wearable head impact sensors. *J Biomech*. 2016 Sep 6; 49(13) 2918–2924. doi: [10.1016/j.jbiomech.2016.07.00](https://doi.org/10.1016/j.jbiomech.2016.07.00).

Regional Brain Injury Vulnerability in Football from Two Finite Element Models of the Human Head

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I. INTRODUCTION

Finite element (FE) models of the human head provide unparalleled rich information about brain mechanical responses upon head impact. The most frequently used injury metrics derived from FE models are either the peak magnitude of maximum principal strain or cumulative strain damage measure (CSDM). However, they do not inform how brain strains are distributed. Strain measures in specific regions of the brain have been proposed to predict injury [1-2]. Most recently, this included the use of a co-registered neuroimage atlas to analyse strains in the 50 deep white matter (WM) regions of interest (ROIs)[2]. Using the Worcester head injury model (WHIM) [3], the relative injury vulnerabilities of the deep WM ROIs were evaluated in terms of nine strain-based injury metrics [2]. Of note, the reconstructed National Football League (NFL) head impacts [4] used in that study were found to be associated with errors [5], which necessitate re-evaluations. In addition, as head injury models could have significant disparities [6], it is also necessary to investigate whether relative brain regional vulnerability depends on the head injury model used. Therefore, the purpose of this study is to reanalyse brain regional vulnerabilities using two versions of the WHIM based on a large impact dataset measured in American high school football. We also extend the evaluation of regional vulnerabilities to both the WM and gray matter (GM) ROIs.

II. METHODS

The isotropic [3] and anisotropic [7] WHIMs were used to simulate 314 head impacts measured from high school football using an Impact Monitoring Mouthguard (IMM) [8]. All of the head impacts were verified by time-synchronised video recordings. Any events collected when the athlete was not being impacted in the head were identified as false positives and were discarded. The remaining true positive events were further scrutinised based on published methods to confirm a head impact occurred in the video and that the computed motion was physically realistic and matched with the video [9]. The impacts were not necessarily associated with a diagnosed concussion, as the purpose of our study was to identify ROIs with the highest likelihood of experiencing largest strains, regardless of whether a concussion actually occurred. This may be important to provide insight in developing a cumulative injury metric based on localised tissue responses to predict concussion in the future.

For each simulated head impact, the 50 deep WM and 54 cerebral GM ROIs were localised *via* a co-registered WM atlas [10] and a separate GM atlas [11], using techniques described previously [2]. For each head impact, peak magnitude of maximum principle strain (ε_1) was obtained for each WM and GM ROI. For each ROI, tied rank values were assigned to score ε_1 in the order of their magnitudes. ROI-wise average rank values across all impact cases were then calculated to evaluate their overall relative vulnerabilities. For both WM and GM, their respective top five most vulnerable ROIs were identified that would indicate the ranks experiencing the highest strains.

To further evaluate the robustness of the ranking, a repeated random subsampling technique was adopted to generate 100k random samples by halving the sample size to 157 impacts. For each random sample, the same ranking analysis was repeated to identify the top five most vulnerable WM and GM ROIs experiencing the highest strains, respectively. From all of the random subsamples, we evaluated the consistency in ranking.

III. RESULTS

Using the isotropic WHIM, the genu (GCC) and main body (BCC) of corpus callosum, and cingulum

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(hippocampus) left (CGH-L) were consistently ranked among the top five most vulnerable WM ROIs in >99.91% of the 100k random trials. For GM, the top five most vulnerable ROIs were consistently found in >99.96% of the random trials, including right inferior frontal gyrus (IFG-R), left and right precentral gyrus (PreCG-L and PreCG-R, respectively), and left and right postcentral gyrus (PoCG-L and PoCG-R, respectively).

Using the anisotropic WHIM, BCC, CGH-L and CGH-R were consistently ranked among the top five WM ROIs (in >99.97% of the random trials). For GM ROIs, the same five most vulnerable ROIs were found (IFG-R, PreCG-L, PreCG-R, and PoCG-R were ranked among the top five in 100% random trials, while >96.16% for ProCG_L).

For illustration, tied rank values averaged from the entire impact dataset are reported in Fig. 1 for the isotropic WHIM. The identified top five WM and GM ROIs are also shown for the two versions of the WHIM (Fig. 2).

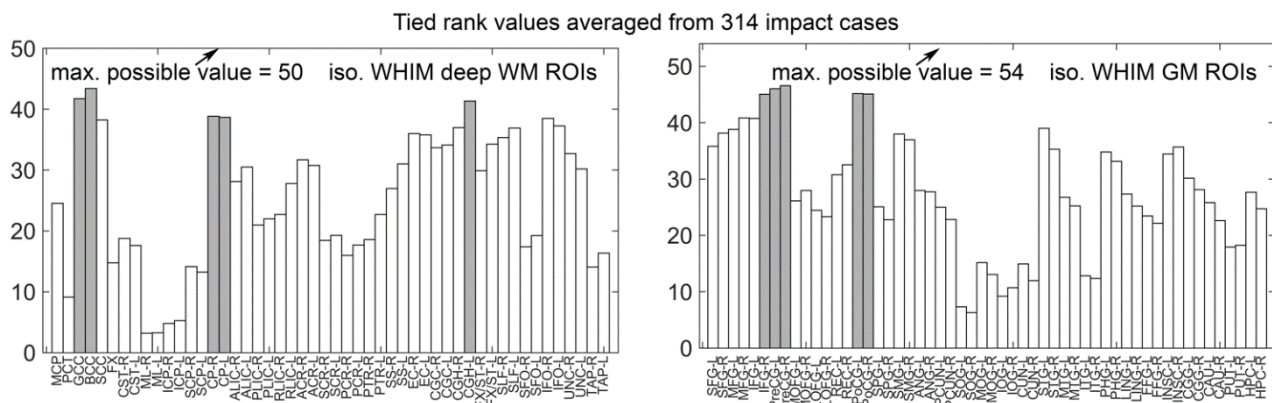


Fig. 1. Tied rank values averaged from 314 impact cases for the deep WM (left) and GM (right) ROIs with their respective top five most vulnerable ROIs highlighted using the isotropic WHIM. Abbreviated ROI labels can be found in [10-11].

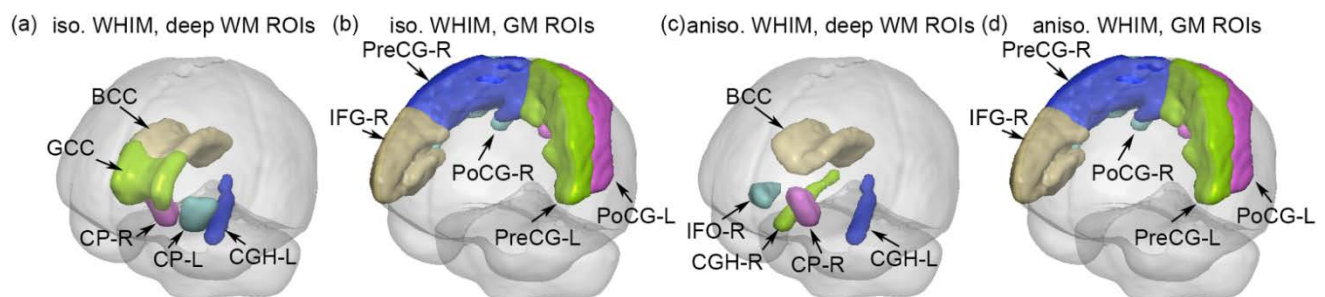


Fig. 2. The top five most vulnerable deep WM (a and c) and GM (b and d) ROIs identified using the isotropic (a and b) and anisotropic (c and d) WHIMs, respectively. For WM ROIs, three regions (BCC, CP-R and CGH-L) were identical between the two WHIMs. For GM ROIs, all of the five regions were identical between the two WHIMs.

IV. DISCUSSION

Identifying the most vulnerable regions may have important implications in understanding the mechanisms of traumatic brain injury (TBI) and in its mitigation. Applying a previously developed technique to a large impact kinematic dataset measured in American high school football using two versions of the WHIM, we found significant consistency in the brain ROIs most frequently experiencing the highest maximum principal strains. Both WHIMs identified the same top five most vulnerable GM ROIs. For deep WM ROIs, BCC, CP-R, and CGH-L were consistently ranked among the top five most vulnerable ROIs. It was interesting to note that two of them, BCC and CP-R, were also found to rank among the top five in an earlier study even though it employed erroneous reconstructed NFL impacts [2]. Further, when using the isotropic WHIM (Fig. 1), the distribution of the relative vulnerability rankings of the WM ROIs was also highly similar to the previous study based on maximum principal strain (with four identical ROIs ranked among the top five: GCC, BCC, CP-R and CP-L [2]). In general, our finding also agreed well with another study that identified the corpus callosum and midbrain regions to sustain the largest strains [12]. These results from independent datasets and head injury models may suggest certain underlying mechanisms in TBI biomechanics. Further investigation is necessary to evaluate other ROI-wise injury metrics such as fiber strain [2], and to incorporate region-dependent material properties and injury tolerances to better assess brain regional vulnerability.

V. REFERENCES

- [1] Giordano and Kleiven, Stapp Car Crash J, 2014.
- [2] Zhao et al., BMMB, 2017.
- [3] Ji et al, J. Neurotrauma, 2015.
- [4] Newman et al., J. Biomech., 2005.
- [5] Sanchez et al., Clin. Biomech., 2018.
- [6] Ji et al., ABME, 2014.
- [7] Zhao and Ji, J. Neurotrauma, 2019.
- [8] Bartsch et al., Stapp Car Crash J, 2014.
- [9] Kuo et al., PLoS One, 2018.
- [10] Mori et al., Neuroimage, 2008.
- [11] Shattuck et al., Neuroimage, 2008.
- [12] Viano et al, Neurosurgery, 2005.

Engineering Characteristics of Athlete Head Impacts that Cause “Obvious Performance Decrement” (OPD)

Adam Bartsch, Dan Hedin, Alok Shah, Mike McCrea, Brian Stemper

I. 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 [1]. Continuing to play for only 15 minutes after a concussive blow causes recovery time to more than double [2]. 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 that *“a concussive injury threshold is elusive”* [3]. In 2014 the Institute of Medicine concluded that *“there are currently inadequate data to define thresholds specifically associated with concussions”* [4]. More recently, authors have advocated for lower uncertainty data: *“... accurate measures of individual exposure will yield a direct estimate of the human tolerance”* [5], *“... as more accurate sensors are designed ...”* [6] *“... valid methods of measuring the direction and severity of on-field head impacts are needed”* [7]. In this study, we began exploring whether lower uncertainty head impact measures will deliver 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 “Obvious Performance Decrement” (OPD) on video was noted and its magnitude was compared to previously published data [8-10].

II. METHODS

Research methods were approved by a local IRB. We used a laboratory-calibrated impact monitoring mouthguard (IMM) system (Prevent Biometrics, Minneapolis, MN, USA) to measure head impacts and inertial motions in n=83 high school and college American football players over 269 athlete-days in the autumn of 2019. Head impacts meant time traces of physically realistic head kinematics in tandem with time-synchronised video verification [11]. False Positives, e.g. data traces that were triggered from non-head impact activity, were discarded using the same video review methods. Any head impact where the player was observed with loss of consciousness, dizziness, apprehensiveness, slow-to-get-up, unnatural reaction and affected speed of movement was flagged as OPD.

Linear and angular, velocity and acceleration, six degree-of-freedom (6DOF) kinematic time traces were captured at 3200 Hz over 50 ms. From these time traces, peak linear acceleration (PLA), peak angular acceleration (PAA), peak linear velocity (PLV), peak angular velocity (PAV), kinetic energy transfer (KET) and impact location were computed.

III. INITIAL FINDINGS

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 25 g to 100 g to the front, front boss, rear boss and rear. The results fit a linear model of the form $IMM = 1.00 \cdot REF + 0.2g$, $R^2 = 0.98$ (see Fig. 1).

Video-verified Impacts

A total of n=2,323 head impacts were video-verified, yielding a left-centered distribution that matched historical data [8-10] (see Fig. 2). There were zero false-negatives identified on video. The measured PLA ranged from 2 g to 100 g and included inertial and minor impacts for $PLA < 20$ g. Among impacts over 20 g, 65% were to the front, 23% were to the sides and 14% were to the back of the head, respectively. The top 1% of head impacts were in the ranges of $PLA > 50$ g, $PLV > 4$ m/s, $PAA > 4,000$ rad/s², $PAV > 20$ rad/s and $KET > 40$ J.

A. Bartsch (e-mail: abartsch@preventbiometrics.com; tel: 612.599.0591) is Chief Science Officer of Prevent Biometrics and discloses conflicts of interest in this research. D. Hedin is with Advanced Medical Electronics. B. Stemper, A. Shah and M. McCrea are from Medical College of Wisconsin, Marquette University and Zablocki VA Center.

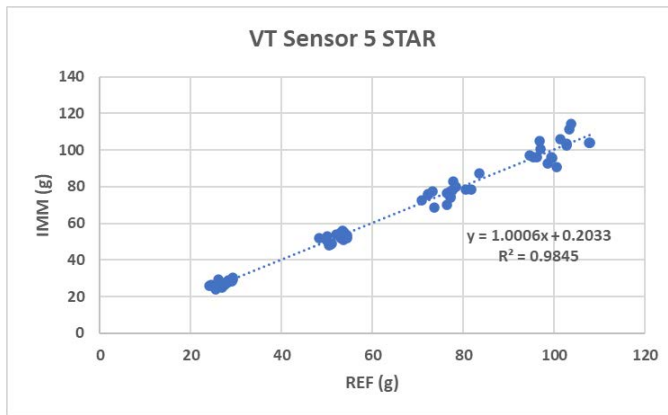


Fig. 1. IMM laboratory test results confirm data fit a linear model versus Reference.

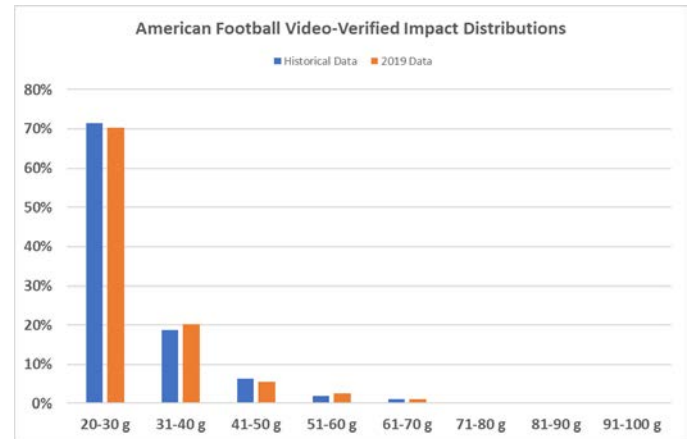


Fig. 2. IMM video-verified impacts from current study compared to published data [8-10].

OPD Impacts

Twelve of the top 15 (12/15, 80%) side- and rear-impact events (PLA range 40 g to 100 g, KET range 40 J to 120 J) resulted in OPD behaviour. One of the top 15 (1/15, 7%) frontal impact events (PLA range 35 g to 62 g, KET range 25 J to 64 J) resulted in OPD. The relationship between OPD impact magnitude and direction was comparable with previous data from [8-10] (see Fig. 3 and Fig. 4).

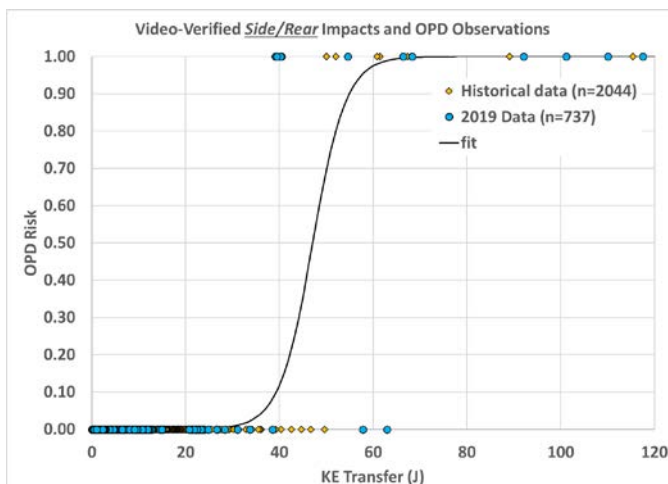


Fig. 3. Video-verified impacts for side/rear compared to published data [8-10].

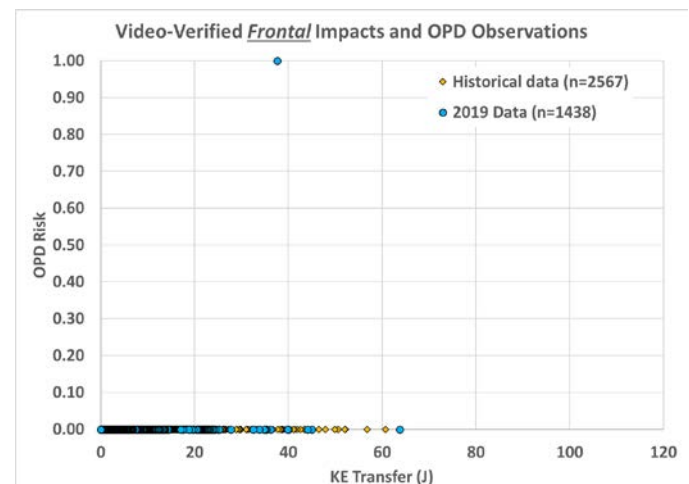


Fig. 4. Video-verified impacts for front compared to published data [8-10].

IV. DISCUSSION

Using a laboratory-calibrated IMM system, we video-verified head impacts in $n=83$ American football players over 269 athlete-days. A total of 2,323 impacts were verified, with 465 having PLA greater than 20 g. We found that head impact distributions from 2019 mimicked published data [8-10]. The KET range of previously published data on OPD events was similar to this study. We also found that OPD impacts occurred mostly to the rear and side of the head, which aligns with published data from IMM and predictions of others [12-14].

It is unclear why high-energy impacts to the side and rear have OPD behaviour more than frontal impacts. It could be that post-impact OPD behaviour possesses a directional sensitivity. We will continue adding to these data to more clearly define single-impact types that cause OPD behaviour, verify video with inter- and intra-rater testing as well as investigate cumulative impact relationships to OPD and objective clinical findings.

V. ACKNOWLEDGEMENTS

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VI. REFERENCES

- [1] McCrea, M., *Clin J Sport Med*, 2004. [2] Elbin, R., *Pediatrics*, 2016. [3] Guskiewicz, K., *Exer Sport Sci Rev*, 2011. [4] IOM, Nat Acad Press, 2014. [5] Meaney, D., *J Biomech Eng*, 2014. [6] Elliott, M., *J Biomech*, 2015. [7] Siegmund, G., ABME, 2016. [8] Bartsch, A., *Biomed Sci Instr*, 2019. [9] Bartsch, A., IEEE, 2019. [10] Bartsch, A., ASTM, 2020. [11] Kuo, K., *PLOSOne*, 2018. [12] Holbourn, A., *Lancet*, 1943. [13] Newman, J., Stapp, 2000. [14] Elkin, B., *Clin Biomech*, 2018.

ESTIMATES OF HIGH-RISK SINGLE AND CUMULATIVE HEAD IMPACT DOSES IN AMERICAN FOOTBALL

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ABSTRACT

In this study we conducted laboratory calibrations and retrospectively analyzed 2851 video-verified head impacts in American football players over 445 player-games using an impact monitoring mouthguard (IMM) system to estimate high-risk impact doses.

In 731 laboratory tests versus Reference, the IMM fit a linear model, with results close to the ideal linear model of form $IMM = 0.97 \cdot REF + 1g$, $R^2 = 0.97$.

During gameplay, the median peak scalar linear acceleration (PLA), peak angular acceleration (PAA), peak linear velocity (PLV), peak angular velocity (PAV), kinetic energy transfer (KE) and Risk-Weighted Exposure (RWE) were 21g, 1600rad/s², 12rad/s, 1.5m/s, 6J and 0.00002, respectively. Approximately 90% of impacts were to the front and sides of the head.

Notable single play (n=4 players) and cumulative full-game (n=3 players) impact doses were examined for players observed on video meeting the National Football League's "No-go" criteria. High energy doses from single play impacts to the side of the head in the coronal plane, in the range of 40J to 110J, caused players to immediately meet "No-go" criteria. High cumulative energy dosing from a game's-worth of impacts – when players were mostly struck to the front of the head in the sagittal plane, range 100J to 320J - also met the "No-go" criteria.

Future data collection will focus on monitoring head impacts in a broader set of athletes to verify these preliminary findings, and to explore single play sensitivity to sagittal impacts and cumulative dosing in the coronal plane.

Keywords: Head impact monitoring, American football, Concussion, Cumulative, Impact dose

INTRODUCTION

In 2011, after reviewing scalar on-field kinematics data leading concussion clinicians concluded *"Recent studies suggest that a concussive injury threshold is elusive, and may, in fact, be irrelevant when predicting the clinical outcome"*.^[1] In 2014 the Institute of Medicine concluded that *"Available studies of head injury biomechanics have identified the importance of linear and rotational movements of the head in injury causation..."* and *"there are currently inadequate data to define the direction- and age-related thresholds for linear and rotational acceleration specifically associated with concussions..."*.^[2] The collection of trustworthy impact monitoring data as a means to address these inadequacies has also been acknowledged by many recently *"accurate measures of individual exposure will yield a direct estimate of the human tolerance"*,^[3] *"as more accurate sensors are designed..."*,^[4] *"valid methods of measuring the direction and severity of on-field head impacts are needed"*.^[5] It is likely that higher fidelity estimates of spatial and temporal impact parameters will clarify the currently unclear impact dose-response relationship.

The aim of this study was to investigate spatial and temporal estimates of head impact doses collected with a laboratory-calibrated impact monitoring mouthguard (IMM) system in American football. We first calibrated the IMM system in n=751 laboratory American football tests against instrumented Reference headforms. Next, we analyzed time-synchronized video and IMM data collected during n=445 player-games of high school and collegiate American football. Summary statistics on all impacts were synthesized. Cases where a player sustained impacts during a single play, or during a

cumulative game's-worth of plays, and on video was demonstrably witnessed to meet the NFL's "No-go" criteria with obvious balance impairment or physical signs and symptoms, were analyzed.

METHODS

Impact Monitoring Mouthguard (IMM)

The IMM is an impact monitor that is instrumented with twelve channels of linear acceleration measurement at 3.2kHz ("12a", **Figure 1**) and has been described elsewhere. [6,7] Each IMM has an embedded flexible circuit board contained within FDA-grade mouthguard material. The IMM also includes a microprocessor, battery, wireless charging, onboard data storage and wireless data offload.



Figure 1. The Impact Monitoring Mouthguard (IMM) System.

Laboratory Calibration

To evaluate the quality of IMM impact estimates, we compared their spatial and temporal outputs versus instrumented Reference headforms in the laboratory for all tested directions/locations combined. For this comparison we used published head impact dosimeter validity specifications. [5-8]

Laboratory calibration tests (n=751) generated first harmonic frequency content in the range of 20-100Hz, like what is seen on the field of play.[6] Impacts were delivered using a linear pneumatic impactor and Hybrid III headform with fixed mandible (n=434) or articulating mandible (n=132), as well as using a pendulum with NOCSAE headform and fixed mandible (n=185).

On-field Data Collection

After calibrating the IMM system against Reference in the laboratory, a total of 2851 video-verified head impacts were retrospectively identified from 445 player-games. Players with zero head impacts during a game were not included in this analysis. Each event was time-synchronized to 30fps video collected of each game. Any events collected when the athlete was not being impacted in the head were identified as false positives and discarded. The remaining true positive events were scrutinized based on

published methods to confirm a head impact occurred in the video and the computed motion was physically realistic and matched the video. [9,10]

RESULTS

Laboratory Results

In the laboratory testing, head impacts for IMM versus Reference PLA/PAA/PAV fit a linear model close to the ideal form (**Figure 2**). Non-trivial processing algorithm, adequate IMM coupling to dentition, knowledge of sensor positions and orientations with respect to Reference, and calibration of individual sensor outputs were required to achieve this level of accuracy.

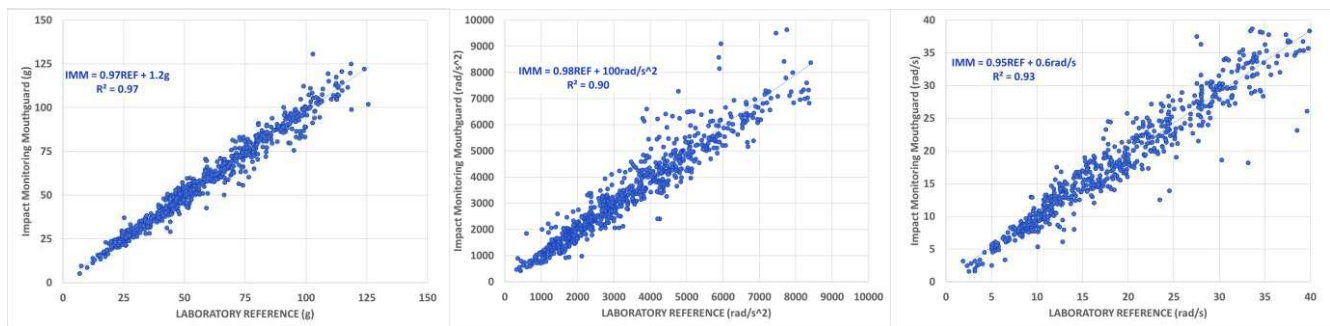


Figure 2. Comparison of IMM versus Reference in Laboratory. Left, PLA. Middle, PAA. Right, PAV.

On-field Results

During on-field data and video analysis, there were approximately 13,000 triggering events captured, with n=2851 true positives confirmed after hardware and software filters were applied, along with the time synchronized video (**Table 1**).

Summary					
# Video-verified Impacts	# Player-games	Average # Impacts/game		Impact Location ¹¹	
2851	445	6.4		58% to Front, 30% to Sides	
Median PLA (g)	Median PAA (rad/s ²)	Median PLV (m/s)	Median PAV (rad/s)	Median KE Transfer (J)	Median RWE
21	1600	1.5	12	6	0.00002

Table 1. Summary Statistics from n=2851 Video-Verified Head Impacts

Single Play Impacts

There were four (4) players with coronal plane impacts during a single play, three (3) players who had cumulative sagittal plane impacts during a game identified as “No-Go” candidates [12] (**Table 2**).

POSITION	LOCATION	PLA (g)	PAA (rad/s ²)	PLV (m/s)	PAV (rad/s)	KE (J)	RWE	OBSERVATION	PLAYER REACTION
QB	Side	68	6600	6.6	41	110	0.13	Tackled from side, could not use arms to brace, side of head hits ground.	Grabs head, lays on field.
CB	Side	88	7800	5.0	57	90	0.45	Dips head to make tackle and is struck by opposing player knee/thigh.	Visibly shaken, stumbling and unstable leaving field.
Special Teams	Side	45, 34	1700, 2000	2.9, 1.5	17, 34	40	0.00	Hit from side while running downfield, second impact immediately side of head hits ground.	Cannot get up.
LB	Side	81	4200	3.9	20	40	0.03	Dips head to make tackle and is struck by opposing player knee/thigh.	Slow to get up, stumbling, points to head.
LB	16 Impacts, 70% Front, 25% Back	6 to 72	600 to 3400	0.3 to 5.1	5 to 33	320	0.02	Using forehead to make first contact.	Vomited on sideline.
Special Teams	12 Impacts, 50% Front, 40% Side	9 to 73	500 to 6200	0.4 to 5.8	2 to 29	200	0.13	Using forehead to make first contact.	Self-reported concussion symptoms after game.
OL	13 Impacts, 100% Front	14 to 48	100 to 1500	1.0 to 3.9	1 to 14	100	0.00	Using forehead to make first contact.	Self-reported concussion symptoms after game.

Table 2. Four players sustained hard impacts during a single play, and three players sustained cumulative impacts during a game. The player observations and reactions met the NFL’s “No-go” criteria¹².

DISCUSSION

Laboratory

The PLA, PAA and PAV results indicated that the IMM system has sufficient bandwidth and sample rate, [13] suitable for measuring head impacts fitting a linear model well versus Reference in the laboratory. The IMM impact estimates correlated well with Reference for all tested impact directions and locations. The laboratory tested conditions bracketed measured ranges of true positive single head impact doses measured in this study and other published studies. [5-8]

On-field

We found an average of 6 to 7 video-verified head impacts per player-game in this study, which is in line with published video-verified studies[14] and approximately one order of magnitude lower than published studies without video verification.[15] For the four players with “No-go” single play impacts, all impacts were in the top 5% of all impacts and to the side of the players’ heads. Coronal plane impact sensitivity has been a hypothesized clinical injury mechanism. [16] and our results support that hypothesis. We also observed high PLA/PAA impacts in the sagittal plane without “No-go” observations. The coronal and sagittal plane findings disagree with other studies that have reported low-gravity impacts in coronal plane and high-gravity impacts in sagittal plane with “No-go” observations. [17]

Also observed in single play “No-go” impacts, the kinetic energy transfer – 40J to 110J – was an order of magnitude higher than the median of 6J. This makes sense, as computational models predict substantially higher brain stress and strain when kinetic energy transfer is high in an impact. [18,19] On the contrary, the RWE metric [20] was low and not correlated to severity for three of the four “No-go” single play impacts. This also makes sense as RWE values increase exponentially on a logistic regression above critical PLA and PAA values of ~120g and 6000rad/s². The PLA/PAA values in our analysis were almost always less severe than these RWE critical values.

For the three players with “No-go” cumulative impacts, 75% of cumulative impacts were to the front of the players’ heads. The total KE transfer, 100J to 320J, was 1-2 orders of magnitude higher than the median value of 6J. Although the number of subjects is modest, this result could point to a spatial and temporal – within one gameday - tolerance to head impact in the frontal direction. It was interesting that the cumulative RWE was low for the “No-go” cumulative players. This can be explained by the fact that most of the cumulative impact doses had negligible RWE since only 15% of cumulative impacts were above 40g PLA or 3000rad/s² PAA.

CONCLUSIONS

In this study we retrospectively analyzed data collected by an impact monitoring mouthguard (IMM) system and time-synchronized player video. In all, we identified 2851 video-verified head impacts in American football players over 445 player-games. Notable single play, and cumulative game impact doses were estimated when players were observed on video meeting the NFL’s “No-go” criteria.

High kinetic energy doses from single play impacts to the side of the head in the coronal plane caused n=4 players to immediately meet “No-go” criteria based on video observation. High cumulative kinetic energy dosing from a game’s-worth of impacts – when n=3 players were mostly struck to the front of the head in the sagittal plane - also met the “No-go” video observation criteria.

While the athlete populations studied here were modest, the single play and cumulative game impact dosing trends we found were very consistent. Future data collection will focus on monitoring head impacts in a broader set of athletes to verify these findings, and to study high-gravity impacts in the sagittal plane as well as cumulative impacts in the coronal plane.

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DISCLOSURES

AB is an employee of Prevent Biometrics and has rights to equity, stock options and a bonus program. Cleveland Clinic has entered into an exclusive license agreement with Prevent Biometrics for Intelligent Mouthguard technology on which AJB, ECB and VJM are inventors. Under terms of these agreements there is potential for fee-for-service payments, royalties and/or equity arising from commercialization of head impact measurement and/or head impact safety product(s). AB, JA, EB, AS, JH, MM and BS receive research support from the United States Department of Defense, National Institutes of Health and/or Department of Transportation. Cleveland Clinic, Medical College of Wisconsin and University of Pittsburgh IRBs (CC13-899, MCW PRO00022950, UPMC 10/30/2018).

REFERENCES

- [1] Guskiewicz KM, Mihalik JP: Biomechanics of sport collision: quest for the elusive Injury threshold. *Exercise and Sport Science Reviews* 2011; 39: 4-11.
- [2] 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.
- [3] 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.
- [4] 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.
- [5] Siegmund GP, Guskiewicz KM, Marshall SW, DeMarco AL, Bonin SJ: Laboratory validation of two wearable sensor systems for measuring head impact severity in football players. *Ann Biomed Eng* 2016; 44: 1257-74.
- [6] Bartsch AJ, Samorezov S, Benzel EC, Miele VJ, and Brett D: Validation of an "Intelligent Mouthguard" single event head impact dosimeter. *STAPP Car Crash Journal* 2014; 58: 1-28.
- [7] Hedin DS, Gibson, PL, Bartsch, AJ, Samorezov, S: Development of a head impact monitoring "Intelligent Mouthguard". *IEEE Medicine and Biology* 2016.
- [8] Virginia Tech Sensor STAR 2018. <https://www.helmet.beam.vt.edu/sensor-ratings.html>
- [9] Kuo C, Wu L, Loza J, Senif D, Anderson SC, Camarillo DB: Comparison of video-based and sensor-based head impact exposure. *PLoS ONE* 2018; 13(6): e0199238. <https://doi.org/10.1371/journal.pone.0199238>
- [10] Gay K, Bartsch AJ et al.: Common data elements (CDEs) for biomechanical devices used in blunt head impact and blast exposure dynamics: the national institute of neurological disorders and stroke (NINDS), national institutes of health (NIH) and department of defense (DOD) version 1.0 recommendations. *Amer Acad Neur* 2018; (in press)
- [11] Greenwald RM, Gwin JT, Chu JJ, Crisco JJ: Head impact severity measures for evaluating mild traumatic brain injury risk exposure. *Neurosurgery* 2008; 62: 789-798.
- [12] National Football League "No-go" Criteria. <https://www.playsmartplaysafe.com/wp-content/uploads/2018/08/nfl-concussion-protocol-2018.pdf>
- [13] Wu LC, Laksari K, Kuo C, et al.: Bandwidth and sample rate requirements for wearable head impact sensors. *Journal of Biomechanics* 2016; <http://dx.doi.org/10.1016/j.jbiomech.2016.07.00>
- [14] Duma SM et al. Analysis of Real-time Head Accelerations in Collegiate Football Players. *Clin J Sport Med* 2005;15:3-8.
- [15] Broglio SP et al. Biomechanical Properties of Concussions in High School Football. *Med. Sci. Sports Exerc.*, Vol. 42, No. 11, pp. 2064-2071, 2010.
- [16] Fijalkowski RJ et al. New Rat Model for Diffuse Brain Injury Using Coronal Plane Angular Acceleration. *J. Neurotrauma* 24:1387-1398 (August 2007).
- [17] Stemper BS et al. Comparison of Head Impact Exposure Between Concussed Football Athletes and Matched Controls: Evidence for a Possible Second Mechanism of Sport-related Concussion. *Annals Biomed. Engr.*, Vol. 46, 2018 epub.
- [18] McAllister TW, Ford JC, Ji S, et al.: *Ann Biomed Eng* 2012; 40: 127. <https://doi.org/10.1007/s10439-011-0402-6>
- [19] Zhang L et al. A Proposed Threshold for Mild Traumatic Brain Injury. *Transactions ASME*, Vol. 126, April 2004, pp 226-236.
- [20] Rowson S & Duma SM. Brain Injury Prediction: Assessing the Combined Probability of Concussion Using Linear and Rotational Head Acceleration. *Annals Biomed. Engr.*, Vol. 41, No. 5, May 2013 pp. 873-882.