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ORIGINAL ARTICLE

Accuracy of Instrumented Mouthguards During Direct Jaw Impacts Seen in Boxing

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Abstract

Purpose Measuring head kinematics data is important to understand and develop methods and standards to mitigate head injuries in contact sports. Instrumented mouthguards (iMGs) have been developed to address coupling issues with previous sensors. Although validated with anthropomorphic test devices (ATDs), there is limited post-mortem human subjects (PMHS) data which provides more accurate soft tissue responses. This study evaluated two iMGs (Prevent Biometrics (PRE) and Diversified Technical Systems (DTS) in response to direct jaw impacts.

Methods Three unembalmed male cadaver heads were properly fitted with two different boil-and-bite iMGs and impacted with hook (4 m/s) and uppercut (3 m/s) punches. A reference sensor (REF) was rigidly attached to the base of the skull, impact kinematics were transformed to the head center of gravity and linear and angular kinematic data were compared to the iMGs including Peak Linear Acceleration, Peak Angular Acceleration, Peak Angular Velocity, Head Injury Criterion (HIC), HIC duration, and Brain Injury Criterion.

Results Compared to the REF sensor, the PRE iMG underpredicted most of the kinematic data with slopes of the validation regression line between 0.72 and 1.04 and the DTS overpredicted all the kinematic data with slopes of the regression line between 1.4 and 8.7.

Conclusion While the PRE iMG was closer to the REF sensor compared to the DTS iMG, the results did not support the previous findings reported with use of ATDs. Hence, our study highlights the benefits of using PMHS for validating the accuracy of iMGs since they closely mimic the human body compared to any ATD's mandible.

Keywords Instrumented mouthguards · Post-mortem human subjects · Boxing punches

Introduction

Mild traumatic brain injuries (mTBI) or concussions have been reported in many contact sports like boxing, martial arts (MMA), football, rugby, ice hockey, and soccer [1]. Approximately 20% of professional boxers experience at least one incidence of TBI in their careers and 40% of retired boxers suffer from lasting effects ranging from headaches, nausea and dizziness to Parkinson's disease, and dementia [2]. Linear acceleration (mainly causing focal brain injuries) and rotational acceleration (leading to focal and diffuse injuries) of the head are postulated as the two predominant mechanisms of head injury [3]. Measuring these quantities *in vivo* allows for the derivation of objective injury criteria which can be used to assess injury risk in both laboratory and on-field conditions. However, studying head impact biomechanics during on-field play is challenging due to difficulty achieving tight coupling of instrumentation with the head [4].

More recent advances on this front have been in the form of instrumented mouthguards (iMGs) fitted to the upper dentition. Since they are fitted on the teeth, they have better

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coupling than the other systems like the helmet and skin-mounted sensors [5–7]. In the last decade, there have been many studies that have focused on validating the efficacy of these iMGs using ATDs by simulating different conditions in sports like football and boxing [8–12], on cadavers in football helmets [10, 13] and on volunteers in boxing [14], MMA [15], and football [8, 12, 14, 15]. One ATD that has been used for the testing of instrumented mouthguards is the Mandible Load Sensing Headform (MLSH). The MLSH has steel upper and lower dentitions on a steel mandible with compliant temporomandibular joints [16]. However, given the lack of extensive data to validate the current MLSH and lack of soft tissues, the additional use of post-mortem human subjects (PMHS) for the validation of the iMGs is warranted [13, 17].

While there are several iMGs that have been developed for on-field use [11], the two that are used in this study are made by Prevent Biometrics (PRE) and Diversified Technical Systems (DTS). The mouthguard developed by Prevent Biometrics (Prevent Biometrics, Edina, MN, USA) is an "intelligent mouthguard" which can provide 6 degrees of freedom linear and rotational kinematic data [9]. Diversified Technical Systems (Diversified Technical Systems, Seal Beach, CA, USA) developed a proprietary iMG (for research) which also provides 6 degrees of freedom with both linear and rotational kinematic data.

Our group recently tested these two iMG systems under helmeted cadaver impacts representative of American football impacts [13]. While both systems failed to perform with high fidelity under some conditions, this study seeks to further examine the iMG fidelity during unhelmeted, boxing-style impacts. Video analysis of knockouts in professional boxing matches from 2014 to 2016 showed that hook and uppercut punches were the most common striking techniques causing concussion [18]. There exist two studies that used the PRE iMG during boxing matches to confirm that the head acceleration data reported matched the impact from video analysis [8] and for verification of true positive impacts [15]. Neither of these studies elaborate on the performance of the iMG during any type of jaw impact since video analysis cannot be used to validate the kinematic parameters as well as a gold-standard reference sensor rigidly attached to the skull. Therefore, this novel study aims to compare iMG accuracy in unhelmeted PMHS to a rigidly mounted reference system, during direct jaw impacts.

Materials and Methods

Specimen Preparation

Three unembalmed cadaver heads were disarticulated at the atlanto-occipital joint and prepared for this study. The average ($\pm$ SD) masses, breadth, length, and circumference of the heads were $4.4 + 0.3$ kg, $15.6 + 0.4$, $20.3 + 1.2$, and $59.5 + 1.9$ cm, respectively, which are representative of a 50th percentile male. An aluminum mounting plate was fixed near the foramen magnum on each head to serve as a base plate to attach 6 degrees of freedom (6DX PRO, Diversified Technical Systems, Seal Beach, CA, USA) reference (REF) sensor. The REF sensor measured linear acceleration ($\pm$ 2000 g) and angular velocity ($\pm$ 18,000 deg/s) at 100,000 Hz. Impressions of the upper dentition were made for all specimens and used to mold the iMGs following manufacturer instructions and pressed manually to set the iMG to the dentition. The authors qualitatively verified that there was no difference in the mouthguard fit on the PMHS specimen compared to when worn by a human. All specimens, with mounting plates and mouthguards attached, were scanned with a computed tomography (CT) machine (Nikon XTH 225 ST, Nikon Metrology Inc., Brighton, MI, USA) at a resolution of $100 \times 100 \times 100$ µm. Coordinate frame transformation matrix calculations were used to translate REF sensor and DTS iMG data to the head center of gravity (HCG) and to express kinematic data with respect to a head coordinate system (HCS).

All the specimens were scanned using a CT (computed tomography) to segregate and identify each of the voxels as soft tissue or bone. Soft tissues and bone were then assigned densities of 1 and 1.92 g/cm^3, respectively [19]. The location of the HCG was computed by spatially summing the mass distributions of bone and soft tissue from the CT scans of each specimen. The Frankfort plane was identified by the anatomic landmarks as described by Daboul et al. [20]. The HCS was defined with both the X- and Y-axes pointing anteriorly and to the cadaver's left on the Frankfort's plane, and positive Z-axis pointing superior from the HCG. Locations of the REF sensor and the DTS iMG within this HCS were identified from reconstructing the CT slices. Transformation matrices from the coordinate frame to the sensor recording frame and from the sensor recording frame to the HCS were measured from the CT data.

$$Q_{i \to S} = \begin{bmatrix} x_s \\ y_s \\ z_s \end{bmatrix} = \begin{bmatrix} x_x & x_y & x_z \\ y_x & y_y & y_z \\ z_x & z_y & z_z \end{bmatrix}$$

…where $Q_{i \to S}$ is the transformation matrix from the coordinate frame to the sensor recording frame. Angular acceleration in the HCS was calculated using equation:

$$\alpha_{HCS} = Q_{S \to HCS} x \alpha_S$$

…where α_{HCS} is the angular acceleration in the HCS is derived by multiplying the transformation matrix from the sensor frame to the HCS $\left(Q_{S \to HCS} \right)$ and the angular acceleration in the sensor frame $\left(\alpha_S \right)$. Similarly, the angular velocity

in the HCS was determined by multiplying the $Q_{S \to HCS}$ and angular velocity in the sensor frame. Lastly, linear acceleration in the HCS at the HCG was calculated using the following steps:

$$a_{HCG} = a_S + (\alpha_S xr) + \omega_S x(\omega_S xr)$$

…where a_S, α_S, and ω_S are the linear acceleration, angular acceleration, and angular velocity in the sensor frame, respectively. r is the position vector from the sensor location to the head CG in the sensor frame. Finally, linear acceleration in the HCS was calculated by multiplying the linear acceleration in the CG a_{HCG} with the transformation matrix $Q_{S \to HCS}$. These calculations were repeated for each specimen.

After transforming the linear acceleration, angular velocity, and angular acceleration to the HCG, Head Injury Criterion (HIC), Brain Injury Criterion (BrIC), and HIC duration were calculated from these signals. HIC calculates the integral of the acceleration of the head at the head center of gravity (HCG) [21]. Following claims of rotational motion giving rise to brain injuries, the BrIC was introduced utilizing angular velocity [22].

Data Processing: REF Sensor and iMGs

The 6DOF REF sensor was sampled at 100,000 Hz using the TDAS data acquisition system (DTS, Inc.) and post-processed using Diadem (National Instruments). Baseline noise was removed for linear acceleration and angular velocity data by subtracting the average value of the first 5 ms of each channel. Following similar biomechanics studies [23, 24], these signals were processed with an 8-pole, 1650-Hz, low-pass Butterworth filter to remove high-frequency noise. Angular velocity data were passed through a 300-Hz low-pass Butterworth filter as per SAEJ211-1 guidelines (SAE 2014) before transforming to the HCG. Angular acceleration was calculated from angular velocity using the 11-point central difference approximation method. First the Peak Linear Acceleration (PLA) was calculated from the maximum value of the resultant vector at the HCG. Following this, Peak Angular Acceleration (PAA) and Peak Angular Velocity (PAV) were calculated as the maximum of the respective resultant vectors within 10 ms from the PLA as in Siegmund et al. [16]. HIC, HIC duration, and BrIC were calculated at the HCG.

The PRE iMG has a triaxial accelerometer ($\pm$ 200 g) and a triaxial gyroscope ($\pm$ 35 rad/s) that collected data at 3,200 Hz [25]. The device records data 10-ms pre-impact and 40-ms post-impact. The mouthguard has its proprietary infrared light and detection system which reflects off the enamel of the teeth to detect whether it has been fitted on the teeth properly. Upon consultation with Prevent, the threshold value for the 'on/off teeth' detection was adjusted appropriately during this study to account for any differences in enamel reflection in PMHS specimens compared to live humans. Signals were post-processed by the Prevent Biometrics backend system before display via the Prevent web portal. The PRE iMG used a proprietary algorithm to translate and re-orient recorded kinematics to an estimated HCG and HCS. For each impact, the web portal provided direction of impact, PLA, PAV, and PAA values for impacts at the estimated HCG and kinematic signals for each event aligned to the estimated HCG and HCS. Upon impact, the data from the iMG are uploaded to the portal via the phone application using Bluetooth. While Prevent does not mention the version of software used on the web portal, the phone app used to transfer the data was version 1.3.0. The software filters all data using a low-pass Butterworth filter at 200 Hz [25]. The system further classifies it into low noise (class 0), moderate noise (class 1), and high noise (class 2) filtered at 200, 100, and 50 Hz, respectively, through a proprietary noise detection algorithm [25, 26]. These filtered data were downloaded, and PLA was used as reported by PRE, PAV, and PAA values were calculated from the respective signals within 10 ms from the PLA, and HIC, HIC duration, and BrIC were also calculated. As the Prevent transformation to the HCG is proprietary, we are unable to verify the PRE HCG and DTS/REF HCG locations are coincident.

The DTS iMG has triaxial linear accelerometer (range $\pm$ 400 g) and triaxial angular accelerometer (range $\pm$ 15,000 rad/s^2) that collected data at 5500 Hz. The device records data 10.5-ms pre-trigger and 100-ms post-trigger. The DTS iMG reports untransformed data within the sensor reference frame. This data was downloaded using the software DDR Control Version 1.09.002. Data were translated and rotated to the HCG and HCS using the procedure detailed above. Data were filtered using an 8-pole, phaseless, low-pass Butterworth filter at 1650 Hz and transformed to the HCG as described. Angular velocity was calculated from angular acceleration by cumulative trapezoidal approximation. PLA, PAV and PAA, HIC, HIC duration, and BrIC were calculated at the HCG.

Statistical analysis was performed using a two-tailed unpaired t test on GraphPad Prism (V 10.2.2, GraphPad Software Inc.) to compare the mean PLA, PAV, PAA, HIC, BrIC, and HIC duration of the REF and PRE and REF and DTS iMGs. They were considered significant if $p < 0.05$. A regression line between the iMG and the REF was fit to the data with a factor of "m" being the slope (iMG = m*REF). It was forced through 0 since the iMG measured 0 when it was not impacted. For each impact, root mean square error (RMSE, Eq. 2), normalized root mean square error (NRMSE, Eq. 3) and relative error in peaks of magnitude (REPM, Eq. 1) were calculated (Table 1). REPM was calculated from the magnitudes of both the REF sensor and

Table 1 Summary statistics comparing REF sensor and iMG data (mean + SD)

	PLA g	PAV rad/s	PAA rad/s^2	HC	BrlC	HIC duration ms	Linear accl. NRMSE %	Linear accl. REPM %	Angular accl. NRMSE %	Angular accl. REPM %
Hook										
REF	55 ± 12	24 ± 3	3526 ± 147	73 ± 22	0.45 ± 0.03*	7.4 ± 2.6	9 ± 2	12 ± 17	24 ± 11	24 ± 22
PRE	51 ± 14	21 ± 5	2811 ± 1611	78 ± 26	0.394 ± 0.07	10.0 ± 6.5				
Uppercut										
REF	21 ± 10	13 ± 3*	1037 ± 239	11 ± 8	0.22 ± 0.06*	21.6 ± 11.4	27 ± 8	17 ± 11	34 ± 5	24 ± 17
PRE	19 ± 11	8 ± 2	838 ± 197	13 ± 11	0.15 ± 0.03	17.9 ± 7.1				
Hook										
REF	52 ± 23	24 ± 1*	3911 ± 1591	66 ± 44	0.49 ± 0.05*	8.9 ± 5.1*	34 ± 15	114 ± 76	20 ± 5	129 ± 118
PRE	109 ± 76	35 ± 6	9912 ± 7644	646 ± 760	0.73 ± 0.11	13.4 ± 2.3				
Uppercut										
REF	38 ± 17	6 ± 3*	903 ± 222*	32 ± 24	0.12 ± 0.06*	8.0 ± 3.7*	23 ± 6	40 ± 43	69 ± 74	221 ± 191
PRE	37 ± 7	14 ± 6	2584 ± 916	41 ± 15	0.26 ± 0.08	18.2 ± 3.1				

*Statistically significant $p < 0.05$

REPM relative errors in peaks of magnitude, *NRMSE* normal root mean square error

both iMGs, and RMSE and NRMSE were calculated by down-sampling the REF sensor signal to the respective iMG sampling rates and then compared across the length of the signal obtained from the iMG. Time 0 was considered when the PLA signal exceeded 5 g's similar to the study by Rich et al. [27].

$$REPM(\%) = \frac{\max(iMG) - \max(REF)}{\max(REF)} * 100 \qquad (1)$$

$$RMSE = \sqrt{\frac{\sum_i^n (iMG_i - REF_i)^2}{n}} \qquad (2)$$

$$NRMSE(\%) = \frac{RMSE}{\max(REF) - \min(REF)} * 100 \qquad (3)$$

Experimental Setup

A linear impactor (Cadex, QC, Canada) was used to simulate both a hook and uppercut to the cadaver. A "fist" was created using maple wood and a standard 10 oz boxing glove (Ringside, USA) was placed on the impactor. The wooden "fist" and glove impactor were used in accordance with USA Boxing Standards document titled "STANDARD METHOD OF IMPACT TEST AND PERFORMANCE REQUIREMENTS FOR BOXING GLOVES AND HEADGEAR" [28]. The total mass of the impactor along with the rigid fist and glove attachment was 21.9 kg. Each specimen was impacted three times at each location for both mouthguards. To represent most frequently executed punches observed in

fights, impact velocities lower than 50% of the highest possible punch velocities by Olympic boxers (hook: 11 ± 3.4 m/s and uppercuts: 6.7 ± 1.5 ms) were chosen [29]. Impacts were conducted at 3 m/s for the uppercut and 4 m/s for the hook. The specimens were suspended inside a mesh bag in the impact chamber which was allowed to translate. Since the accelerometers in the iMGs were on the left side (approximately in front of the premolar tooth), the impacts were aimed on the right side of the specimen to avoid damaging the sensor and potential erroneous results. For the hook punch, the specimen was oriented face down and impacted on the right side and for the uppercut, the specimen was oriented facing upward and impacted on the bottom of the chin (Fig. 1). The impactor for the hook punch was aimed in such a way that surface area of the boxing glove fell within the region bounded by the temporomandibular joint (TMJ) and the chin, covering the mandibular ramus. After each impact, the mouthguard was verified to be fixed to the upper dentition that there was no tissue or mandibular degeneration and the 6DOF reference block was still rigidly attached to the skull. PRE was also verified to detect the iMG as "on the teeth" in their portal after every impact.

Results

Prevent iMG

Data transformed to the HCG from the REF sensor and the PRE iMG are shown in Fig. 2, and summary statistics is provided in Table 1. Comparison of PLA, PAV, and PAA data obtained from the REF sensor and PRE iMG are shown
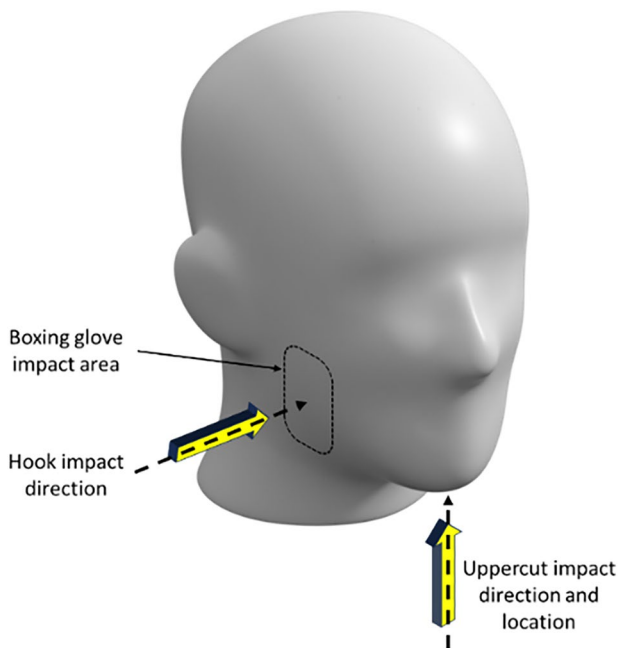
Fig. 1 Direction and impact location of the hook and uppercut punches on the PMHS surrogate. The hook is parallel to the ground on the right side and the uppercut is perpendicular to the vertical axis on the chin. Image adapted from Freepik.com

in Fig. 3. The directions of the impact were not used for the purpose of processing, but PRE detected the impact directions accurately in 7/9 of the hook impacts and only 3/9 of the uppercuts. To compare the signals for NRMSE, it was verified that the difference between the original and downsampled PLA, PAV and PAA for both hooks and uppercuts were under 0.2%, except the PLA for hook, which was 0.6%.

The PRE iMG had an R^2 value of 0.9575 with the reference sensor for PLA data (Fig. 2). The high noise (class 2) signals for the two hook punches obtained from the PRE iMG underpredicted the PLA by 26.7 and 50.4%. The PAV data of the PRE iMG showed a correlation of $R^2 = 0.9595$ with the REF sensor (Fig. 2). The PRE significantly underpredicted the corresponding REF sensor data for its uppercuts ($p = 0.02$). The PAA data of the PRE iMG showed a correlation of $R^2 = 0.9475$. The slope of the regression line for PAA was lower than the PLA and PAV at 0.73 since it underpredicted data compared to the PLA and PAV but they were not significantly different. The HIC data of the PRE iMG showed a correlation of $R^2 = 0.9368$. Similar to the PAV data, the BrIC value had an R^2 of 0.9468. Additionally, the PRE significantly underpredicted its corresponding REF data for both the hook ($p < 0.05$) and uppercuts ($p = 0.01$).
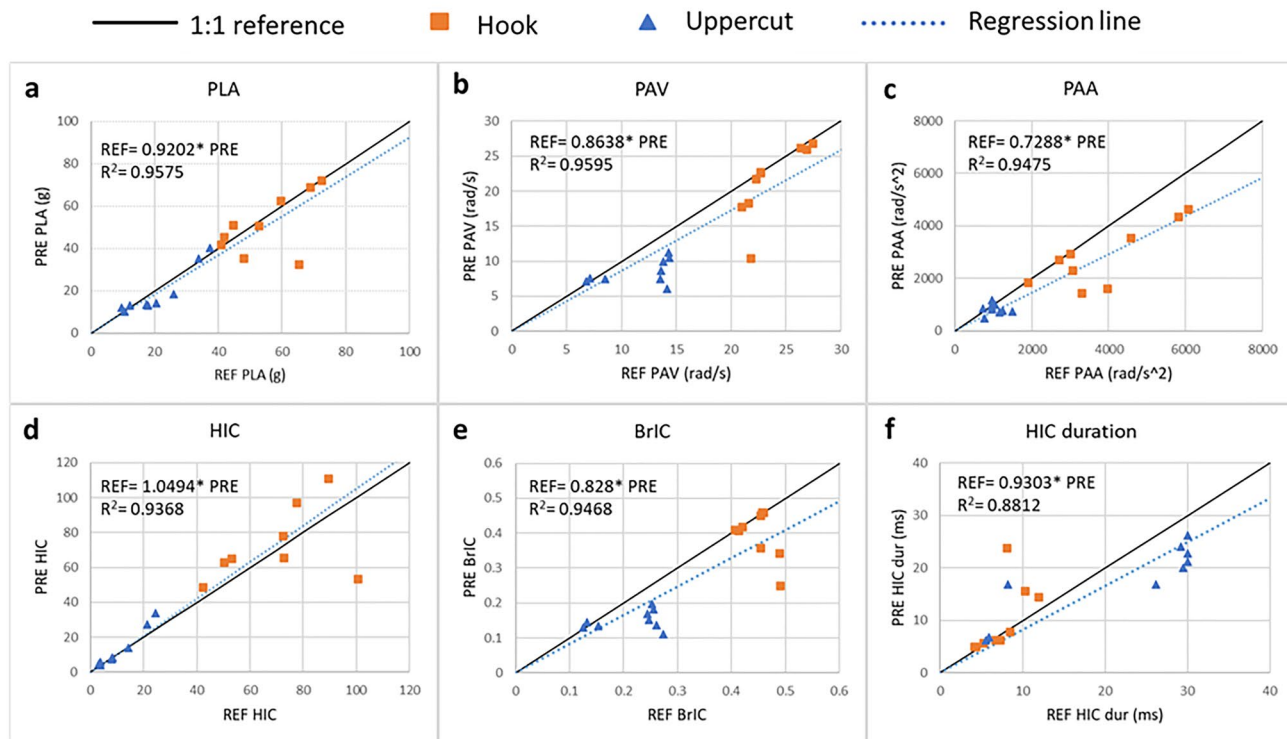
Fig. 2 Data from REF sensor and PRE iMG data. The image shows **a** PLA, **b** PAV, **c** PAA, **d** HIC, **e** BrIC, and **f** HIC duration
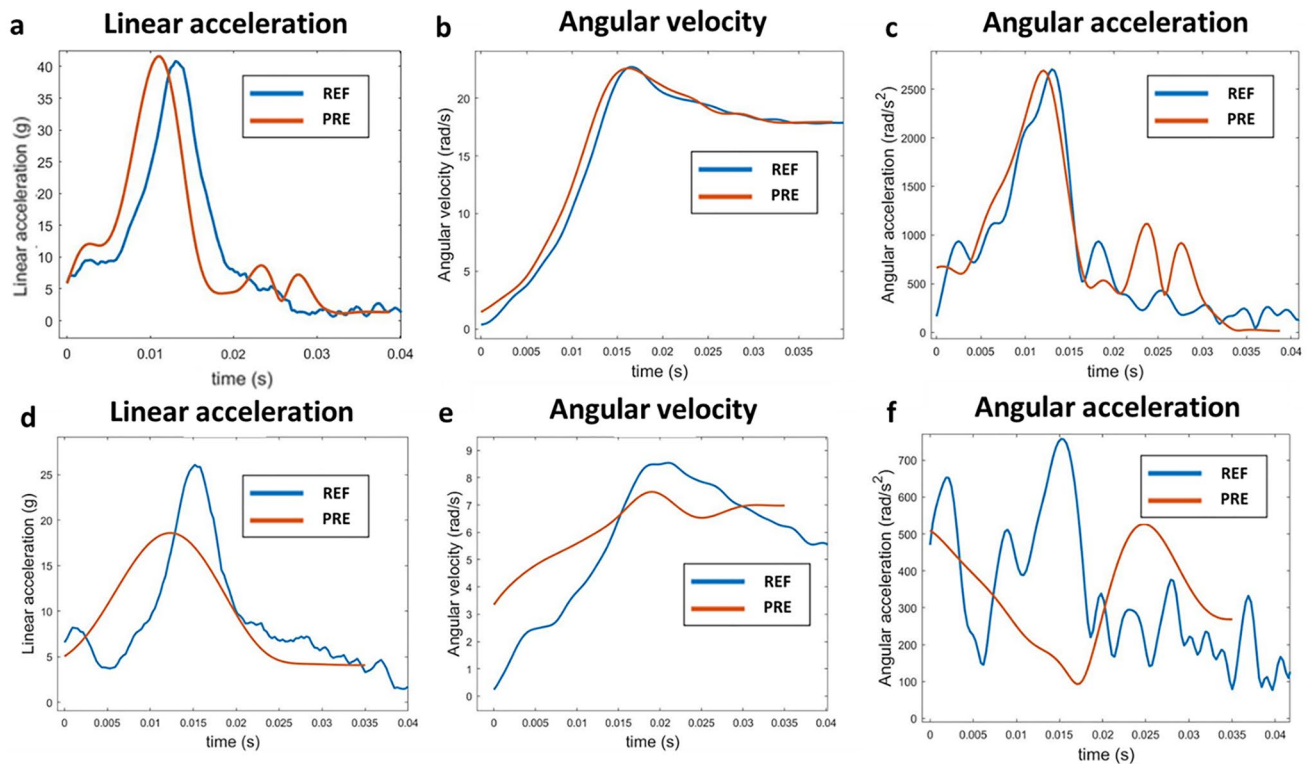
Fig. 3 Comparison of individual kinematic data for a representative class 0 (Hook: top row) and class 2 (uppercut: bottom row) impacts as detected by PRE. **a** REF PLA = 41 g, PRE PLA = 42 g, **b** REF PAV = 23 rad/s, PRE PAV = 23 rad/s, **c** REF PAA = 2705 rad/s^2, PRE PAA = 2691 rad/s^2, **d** REF PLA = 26 g, PRE PLA = 19 g, **e** REF PAV = 9 rad/s, PRE PAV = 7 rad/s, and **f** REF PAA = 758 rad/s^2, PRE PAA = 474 rad/s^2
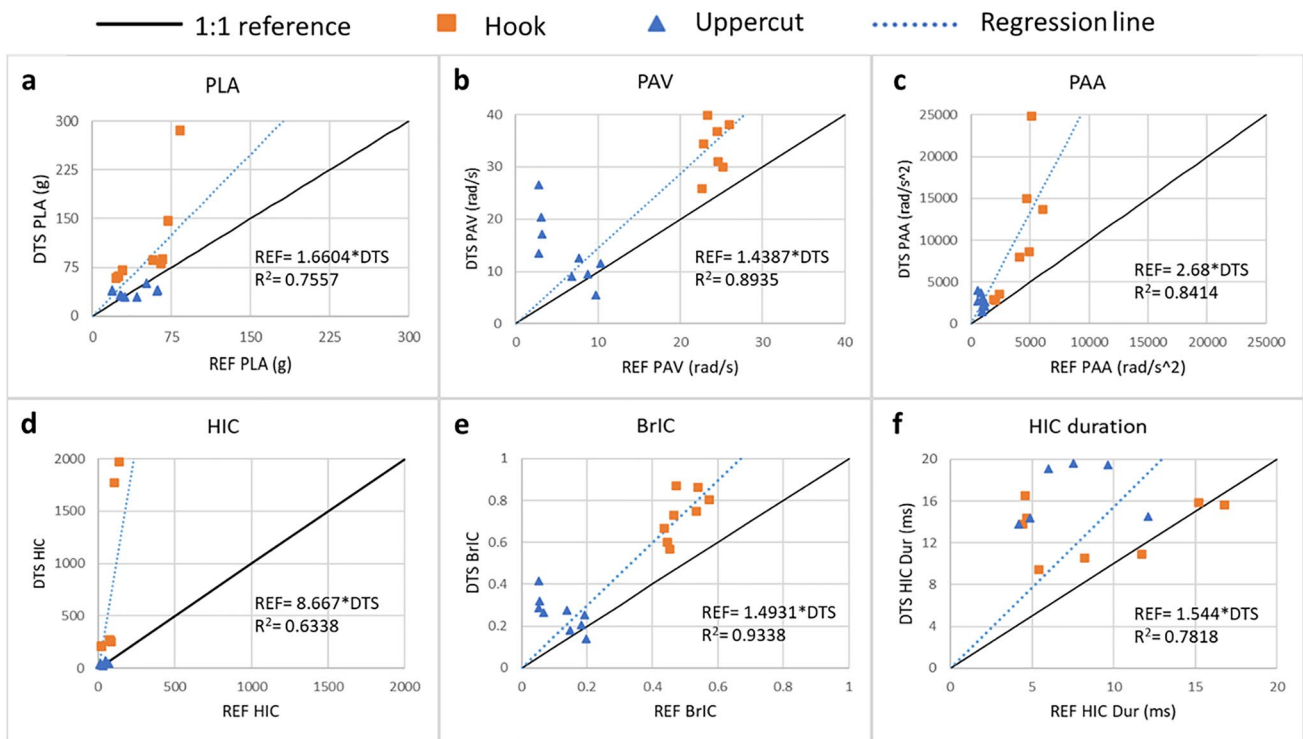
DTS iMG

Overall, the DTS iMG had poor correlation with the REF sensor for most of the parameters measured ($0.6 < R^2 < 0.9$ for PLA, PAV, PAA, HIC, and HIC duration; only BrIC $R^2 = 0.9338$) with the data from the DTS reporting higher magnitudes. Data transformed to the HCG from the REF sensor and the DTS iMG are shown in Fig. 4, and summary statistics is shown in Table 1. After filtering the PAA with an 8-pole low-pass Butterworth filter, it was observed that the signals obtained from the DTS iMG had higher noise than the PRE data as its NRMSE was > 20%. To compare the signals for NRMSE, it was verified that the difference between the original and down-sampled PLA, PAV, and PAA for both hooks and uppercuts were under 0.2%.

The PLA values of the DTS iMG and REF sensor had a correlation of $R^2 = 0.7557$. The PAV values for the DTS and REF had a correlation of $R^2 = 0.8935$. The DTS was significantly higher than its corresponding REF sensor data for both the hook ($p < 0.001$) and uppercuts ($p = 0.01$). The DTS iMG overestimated the REF values for the PAA (m = 2.68) and had a correlation of $R^2 = 0.8414$. The DTS was significantly higher than its corresponding REF sensor data for the uppercuts ($p < 0.001$). For the HIC, the DTS iMG had

the highest overestimation (m = 8.667) to the corresponding REF sensor data and had a correlation of $R^2 = 0.6338$. The BrIC values had a correlation of $R^2 = 0.9338$ and the DTS was significantly higher than its corresponding REF sensor data for both the hook ($p < 0.05$) and uppercuts ($p < 0.001$) (Fig. 5).

Discussion

While previous research on testing several iMGs with ATDs found that the PLA, PAV, and PAA had REPM of < 20% [11], $R^2 > 0.9$, slopes of regression lines between 0.96 and 1.02 and NRMSE < 15% for linear acceleration, angular velocity, and angular acceleration data [10, 14, 27, 30], our study did not report the same results for all parameters between the iMGs and the REF sensor. The PRE and DTS iMG fixed in a rigid ATD Hybrid III and NOCSAE (National Operating Committee on Standards for Athletic Equipment) headforms were tested in multiple studies by impacting it with a pneumatic impactor which showed good fidelity [8, 9, 11, 17]. All these studies reported nearly 1:1 performance from the iMGs compared to the REF sensor and slope value close to 1. The DTS's sensor board used to validate another

Fig. 4 Data from REF sensor and DTS iMG. The image shows **a** PLA, **b** PAV, **c** PAA, **d** HIC, **e** BrIC, and **f** HIC duration
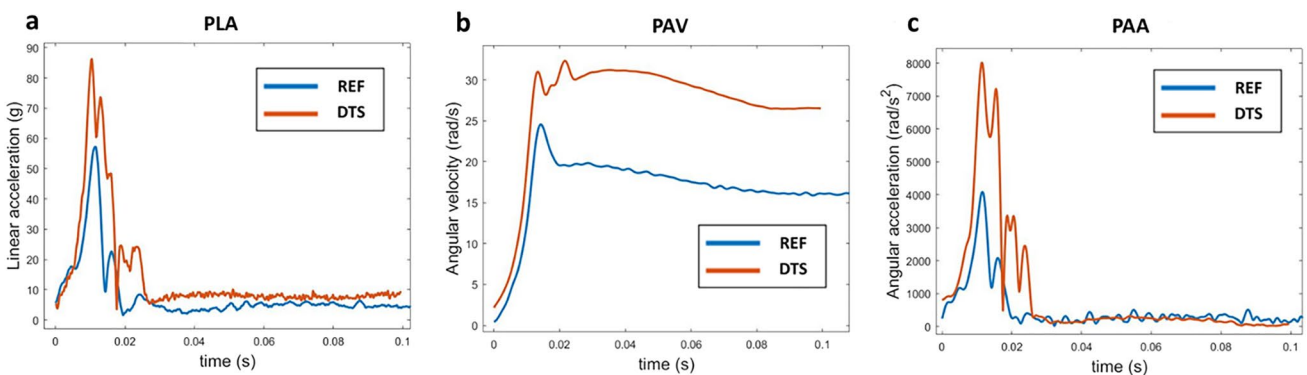
Fig. 5 Sample kinematic data for a hook punch. **a** REF PLA = 57 g, DTS PLA = 86 g, **b** REF PAV = 25 rad/s, DTS PAV = 31 rad/s, and **c** REF PAA = 4086 rad/s^2, DTS PAA = 8013 rad/s^2

study [25] also found lower errors (REPM < 25% and R^2 close to 1) than the ones found in this study.

As in our previous work with helmeted cadavers, we found that the DTS iMG overestimated the reference in all kinematic measures tested, and the PRE iMG underestimated PLA, PAV, and PAA [13]. In contrast to our prior work, the PRE iMG appears to more closely fit the reference data in both PLA (PRE PLA = 0.9202* REF (current study), vs. PRE PLA = −0.003*(REF)^2 + 1.164*REF [13]) and HIC (PRE HIC = 1.0494*REF (current study), vs. PRE HIC = 0.357*REF [13]), with R^2 > 0.9 in both cases in the

current study. Notably, the impact magnitudes tested here fall within the range of impacts with good agreement for the PRE iMG during helmeted impacts [13]. This suggests that the PRE iMG may be better suited to lower magnitude exposure environments (sub-80 g) than in high impact scenarios. However, the PRE iMG underestimates both PAV and PAA as in the previous work.

One possible reason for the differences observed between the PRE and REF sensor data is PRE's proprietary noise classification and filtering algorithm. This study used filtered (and estimated to HCG) linear and angular

kinematic signals to calculate HIC and BrIC data as well. Future studies would benefit by being able to work with the raw data instead of data that is filtered and transformed (by estimation) to the HCG. The DTS consistently overestimated all kinematic data due to unknown causes. The iMG was rigidly fixed to the specimen's teeth, but it still reported REPM of over 200% and NRMSE of about 70% for its angular kinematic data. Since all transformations from the sensor to the HCG rely on angular kinematics, the linear transformations were affected as well. For example, a HIC value of 667 has a 50% probability of causing skull fractures [31], this is important in predicting brain injury and the resulting neurologic implications. While the PRE iMG predicts the HIC to about 70% of the reference value, the DTS iMG overpredicts HIC by over 8 times; both calculations would suggest a probability of injury that is much different than if calculated with the true head kinematics.

Another reason for the non-agreement between the iMG and REF sensor data would be due to the jaw that was not tightly clenched. Kieffer et al., and Liu at al., constrained the mandible of the ATDs used to prevent erroneous data due to 'jaw slap' effect [11, 12]. A study on the effect of clenched mandible that prevents jaw movement found a significant decrease in errors ($p < 0.05$) when the ATD had no mandible or a clenched mandible in comparison to an unconstrained mandible [10]. They reported NRMSE < 15% for linear and angular acceleration, compared to their unconstrained mandible condition which experienced 40–80% error. These are in line with our research which reported mean NRMSE up to 34 and 27% linear acceleration and up to 24 and 69% angular acceleration for hooks and uppercuts, respectively, in PRE and DTS iMGs. Rich et al., also reported mean NRMSE of 4 and 9% linear and angular acceleration for similar impact conditions when using 3D printed dentition fit on a NOCSAE headform [27]. Real-life impacts as those experienced in boxing may not necessarily have their jaws clenched, which is why this study was designed with an unconstrained mandible. Lastly, the DTS iMG uses an angular accelerometer which is sensitive to vibration, leading to higher PAA data [32]. While idealized boundary conditions using ATD headforms are reasonable when designing and developing novel wearable instrumentation, our findings here suggest that the increased biofidelity provided by human cadaver testing is necessary to properly validate wearable head kinematic sensors prior to field deployment.

This novel study evaluated the performance of the two iMGs with two specific boxing punches which directly impact the mandible and therefore the mouthguard. Other types of punches, as well as increasing velocities of the hook and uppercut punches were not considered, neither were conditions with a clenched jaw. As per the Consensus Head Acceleration Measurement Practices (CHAMP), our study did not test the ability of the iMG to detect impact events [33].

Declarations

Conflict of interest JFL conducts research in the head and neck injury space where he actively uses a head impact exposure device—the DASHR (earpiece system)—that was developed internally at Duke University. Duke University has a patent for the DASHR system and JFL is a co-inventor associated with this patent. The author has no financial relationship with an entity that is utilizing the DASHR system for commercial purposes, with the exception of royalties that may originate if the technology is licensed by Duke University (owner of patent). MZA additionally conducts research in the head injury space where he actively uses the DASHR system and has no financial relationship with an entity using this system for commercial purposes.

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