Head Exposure to Acceleration Database in Sport (HEADSport)

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Head Exposure to Acceleration Database in Sport (HEADSport): a kinematic signal processing method to enable instrumented mouthguard (iMG) field-based inter-study comparisons

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ABSTRACT

Objective Instrumented mouthguard (iMG) systems use different signal processing approaches limiting field-based inter-study comparisons, especially when artefacts are present in the signal. The objective of this study was to assess the frequency content and characteristics of head kinematic signals from head impact reconstruction laboratory and field-based environments to develop an artefact attenuation filtering method (HEADSport filter method).

Methods Laboratory impacts (n=72) on a test-dummy headform ranging from 25 to 150 g were conducted and 126 rugby union players were equipped with iMGs for 209 player-matches. Power spectral density (PSD) characteristics of the laboratory impacts and on-field head acceleration events (HAEs) (n=5694) such as the 95th percentile cumulative sum PSD frequency were used to develop the HEADSport method. The HEADSport filter method was compared with two other common filtering approaches (Butterworth-200Hz and CFC180 filter) through signal-to-noise ratio (SNR) and mixed linear effects models for laboratory and on-field events, respectively.

Results The HEADSport filter method produced marginally higher SNR than the Butterworth-200Hz and CFC180 filter and on-field peak linear acceleration (PLA) and peak angular acceleration (PAA) values within the magnitude range tested in the laboratory. Median PLA and PAA (and outlier values) were higher for the CFC180 filter than the Butterworth-200Hz and HEADSport filter method ($p<0.01$).

Conclusion The HEADSport filter method could enable iMG field-based inter-study comparisons and is openly available at https://github.com/GTBiomech/HEADSport-Filter-Method.

WHAT IS ALREADY KNOWN ON THIS TOPIC

⇒ Instrumented mouthguard (iMG) systems are a rapidly emerging technology in contact sports such as rugby union.
⇒ Peak kinematics differ considerably between different iMG systems used in similar cohorts/environments.
⇒ iMG systems use different signal processing approaches limiting field-based inter-study comparisons, especially when artefacts are present in the signal.

WHAT THIS STUDY ADDS

⇒ An openly available artefact attenuation filtering method (HEADSport filter method) for real-world/on-field iMG data has been developed based on signal power spectral density characteristics of bare-headed laboratory impacts and on-field head acceleration events (HAEs).
⇒ The HEADSport filter method is based on fundamental rigid body mechanics and signal processing principles.
⇒ The method can be applied to raw iMG kinematic signals with adequate sample rates to enable field-based inter-study comparisons.

HOW THIS STUDY MIGHT AFFECT RESEARCH, PRACTICE OR POLICY

⇒ The HEADSport filter method could enable field-based inter-study comparisons when different iMG systems are used.
⇒ The HEADSport filter method could contribute towards the development of a standard or consensus for processing sports-based head sensor kinematic signals that progresses in line with the state-of-the-art, similar to the automotive industry.

INTRODUCTION

The collision nature of many contact sports means concussion and repetitive head acceleration events (HAEs) are an issue.[1–3] HAEs can occur directly and indirectly from head and body contact on the field, respectively.[1 4] Head kinematics are associated with brain injury risk, with rotational head motion considered the primary contributor to brain deformation.[1] A recent review article indicates a growing evidence base of various biomechanical brain injury mechanisms, including those involving repetitive HAEs.[1]

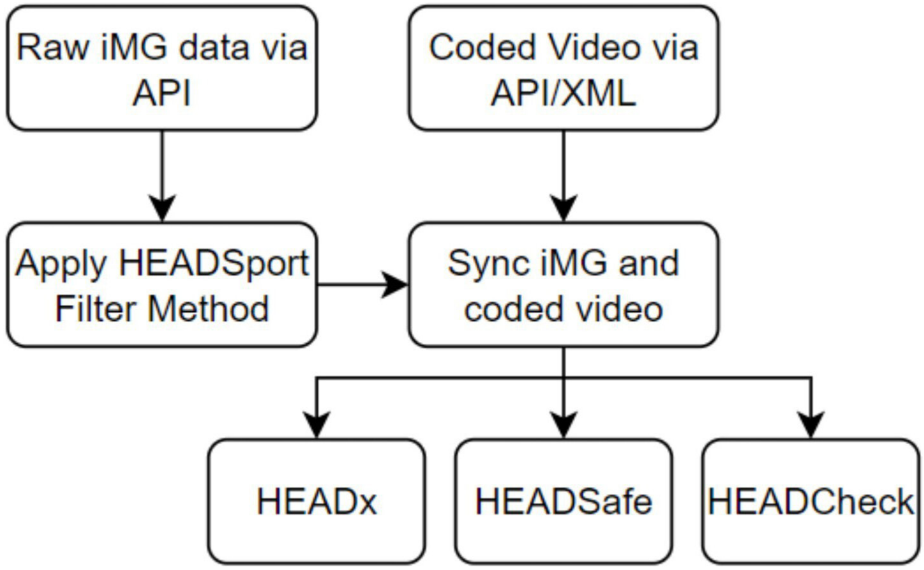

Figure 1 The HEADSport (Head Exposure to Acceleration Database in Sport) pipeline for automated analysis of instrumented mouthguard (iMG) and video data.

Repetitive HAEs will continue to be a concern in contact sports until proactive mitigation strategies are developed and this requires an understanding of the biomechanical mechanisms.

Wearable head sensors with superior coupling to the skull provide a unique opportunity for measuring HAE kinematics in sport (eg, instrumented mouthguards (iMGs) and mouthpieces).[5] A recent study on iMG validity found that the majority of iMG devices performed highly in a head impact reconstruction laboratory.[6] However, the study illustrated that on-field head kinematics reported by iMG differed considerably, with certain systems producing much higher head kinematics (roughly 500 g and>50 krad/s^2) in comparison to others. At present, each iMG system uses different kinematic signal processing approaches (eg, filter cut-off frequencies, machine learning-based noise/artefact detection models) which limits cross-study comparison.[1 6 7] The Head Exposure to Acceleration Database in Sport (HEADSport) is a large scale iMG video analysis project and automated data pipeline that aims to (1) HEADx—quantify head acceleration exposure rates in different sports; (2) HEADSafe— understand risk factors for severe HAEs to enable mitigation strategy development; and (3) HEADCheck—aid sideline decision-making for safe player removal when a high-risk HAE is sustained (figure 1).[1] First, though, a common kinematic signal processing method is needed for iMG systems (ie, HEADSport filter method in figure 1).

Wu *et al*[8] illustrated through cadaver head impact reconstructions that low-pass filter cut-off frequencies (−3 dB) of 590 and 290 Hz were required to keep amplitude attenuation within 10% for linear acceleration and angular velocity, respectively. However, the highest −3 dB cut-off frequency used by an iMG system in Jones *et al*[6] was 300 Hz, through the use of a Channel Frequency Class 180 (CFC180) filter.

Although not representative of a real-world impact, a head impacting a rigid surface as a perfectly elastic rigid body detached from the neck can be modelled as a spring-mass system (online supplemental file A). The linear acceleration of the head during the impact is described by equation 1 and derived in online supplemental file A[9]:

$$a(t) = -v_i\sqrt{\frac{k}{m}}\sin\left(\sqrt{\frac{k}{m}}t\right) \qquad 0 \le t \le \frac{\pi}{\sqrt{\frac{k}{m}}}, \qquad 1$$

where a is the linear acceleration of the head, v_i is the head velocity at initial time of impact, k is the head stiffness, m is the head mass and t is time. Equation 1 illustrates that the acceleration of the head during an impact has (1) a characteristic pulse modelled as a half-sine wave and (2) amplitude and pulse duration influenced by the impact conditions. That is, the magnitude of the acceleration is dependent on the initial head velocity, mass and stiffness, and the acceleration pulse duration (or frequency) is dependent on the head mass and stiffness.

A head impact reconstruction laboratory provides an idealised environment for testing iMG.[1 6 7 10 11] The impact magnitudes and pulse durations are controlled to a high degree through the use of an impactor that strikes a test-dummy head-neck model, and the iMG is clamped onto a plastic upper dentition within the headform.[1 6 7 10 11] During these impacts, the processing of noise is straightforward and/or standardised.[10] Sources of noise in these reconstructions can occur from the measurement system (ie, headform sensor array), numerical differentiation (if applicable) and biomechanical transformations of these data (eg, transforming the linear acceleration signal from the sensors to the head centre of gravity (CG)).[12]

Performing well in a head impact reconstruction laboratory does not constitute a valid device for use in the field.[10] It provides a basic check that the hardware can measure impacts we expect to see on the field and the firmware/software can process and transform the data appropriately.[10]

In real-world environments such as the rugby field, there are cases where artefacts will exist in the iMG signal.[4 13] Example sources of artefacts in the context of iMG include movement of the mouthguard on the teeth due to poor fit/coupling, shouting, biting, mandible interference and direct impacts to the mouthguard.[4] Video-verification based on-field validation to quantify true-positive, false-positive and false-negative metrics can give an indication of the iMG system HAE detection algorithms.[7] However, true-positive events can have artefacts in the signal if left unattenuated. These artefacts have the potential to produce unwanted frequency content in the signal.[13] Accordingly, artefacts can result in erroneous head kinematics being reported by the iMG system which is further compounded when numerical calculations and biomechanical transformations are implemented.[12 13]

The dynamic nature of contact sports means a range of impact conditions are likely to occur on the field during HAEs.[1] Sports collisions can include different impact velocities, effective masses, complex contact characteristic (eg, non-linear stiffness and damping characteristics, influence of neck and body) as well as sources of signal artefacts.[13–15] Head impact reconstruction laboratories provide the ability to test to extreme HAE magnitudes and pulse durations seen on the sports field in a controlled environment with artefacts mitigated.[6 10] The aim of this study was to assess the frequency content and characteristics of head kinematic signals from head impact reconstruction laboratory and field-based environments. An artefact attenuation filter will be developed based on the findings and applied to an elite-level rugby union iMG dataset. The aim of the artefact attenuation filter is to enable inter-study comparisons for field-based iMG studies in sport.

MATERIALS AND METHODS

Head impact reconstruction laboratory test set-up

The dummy head-neck model configuration comprises of a medium-sized National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform affixed to a Hybrid III 50th percentile male neck, fitted on a linear slide table with 5 df.[6 10] Reference kinematics were measured at the dummy headform CG with an instrumentation package comprising three linear accelerometers (Endevco 7264b-2000; Meggitt Orange County, Irvine, California, USA) and a tri-axial angular rate sensor (DTS ARS3 Pro 18 k; Diversified Technical Systems, Seal Beach, California, USA), both recording at 20 kHz.[6 10] No filter was applied to the signals apart from the default hardware filters (−3 dB cut-off frequency of 4000 Hz). Head impact laboratory reconstruction testing was conducted using a pendulum impactor to simulate bareheaded impacts to the dummy headform. Tests captured the range of HAE magnitudes and pulse durations seen in rugby through the use of a rigid (nylon, 25 mm thickness) and padded (vinyl-nitrile foam, 40 mm thickness) impactor (127 mm diameter) to the bareheaded dummy headform. Impacts occurred at the front and front boss locations of the headform at target linear head accelerations of 25, 50, 75, 100, 125 and 150 g. Three tests were conducted at each configuration, totalling 72 impacts (6 impact magnitudes×2 impactors×2 locations×3 tests per configuration).

On-field iMG data collection

A total of 126 participants (96 male and 30 female) were recruited from seven (4 male and 3 female) elite-level rugby union club/international teams in the northern hemisphere. Each participant provided written consent. Participants underwent three-dimensional dental scans and were provided with a custom-fit iMG (Prevent Biometrics, Minneapolis, Minnesota, USA). The iMGs were equipped with an accelerometer and gyroscope both sampling at 3200 Hz and a measurement range of ±200 g and ±35 rad/s, respectively.[6 16] Laboratory and on-field validity of the Prevent Biometrics iMG has been demonstrated in previous studies.[6 10 11] Most recently, a validity and feasibility study of multiple iMG systems illustrated that the Prevent Biometrics iMG had a concordance correlation coefficient value of 0.984 in laboratory-based impact testing using a crash test dummy headform.[6 7] The trigger mechanism for the iMG in this study was when 8 g was exceeded on a single axis of the iMG accelerometer, capturing 10 ms of pre-trigger data and 40 ms of post-trigger data. All linear kinematics were transformed from the iMG to the head CG and a 5 g recording threshold was applied. Peak linear acceleration (PLA) and peak angular acceleration (PAA) of the head were extracted from each HAE.

The level of noise/artefact in the kinematic signal was classified by an in-house Prevent Biometrics classification algorithm that determined whether each HAE contained minimal noise (class 0), moderate noise (class 1) or severe noise (class 2). A fourth order (2×2 pole) zero phase, low-pass Butterworth filter was applied to each signal with a cut-off frequency (−6 dB) of 200, 100 and 50 Hz for class 0, 1 and 2 HAEs, respectively (referred hereafter as the 'Prevent Biometrics processed output'). A −6 dB cut-off frequency is a consequence of not adjusting the −3 dB cut-off frequency when filtering twice (once forward and once backward), in order to prevent phase shifts.[12] The filter was applied as follows: (1) filter applied to raw linear acceleration and angular velocity; (2) filter applied to differentiated angular velocity (ie, angular acceleration); (3) filter applied to linear acceleration when transformed to head CG.[17] Data were collected from 209 player-matches (160 male and 49 female). Multiple angle, broadcast quality video footage was available for all matches. HAEs were synchronised to the video to a 40 ms resolution enabling video verification. Both the on-field data collection and head impact reconstruction laboratory impacts complied with the consensus head acceleration measurement practices (CHAMP).[4 18]

Artefact attenuation filter method development

The design of the artefact attenuation filtering method (referred hereafter as the 'HEADSport filter method') predominantly used the power spectral density (PSD) characteristics of the head impact reconstruction laboratory impacts.[19] The component (X, Y or Z) of raw linear acceleration signal with the highest peak magnitude was used for the PSD analysis. PSD was conducted only on the impact pulse.[20] The impact pulse was calculated based on the first zero-crossing timepoint before and after peak linear acceleration was reached. The frequency associated with the 95th percentile PSD (cumulative sum) was extracted from each laboratory impact.[12 19] Each pulse (on-field and laboratory) was padded using the *nextpow2* function in MATLAB (R2023a, MathWorks, Natick, Massachusetts, USA). The zeros used to pad the signal did not contribute to PSD calculations. The maximum 95th percentile PSD frequency (F_{max}) for the laboratory impacts was considered the upper limit for on-field HAEs.

The same PSD approach was conducted for on-field HAEs. Any on-field HAE with 95th percentile PSD above F_{max} was considered to have an artefact in the signal. Additionally, the first local minima frequency on the PSD spectrum, if applicable, was extracted from on-field HAEs as signal above this was considered to contain unwanted higher frequency noise due to signal artefacts.[12 21 22] A CFC filter was used for the HEADSport filter method based on impact biomechanics standards (SAE J211).[23] In brief, a CFC filter is a fourth order (2×2 pole), zero phase, low-pass Butterworth filter with frequency class (F_H) indicating the entire flat range of the frequency response (ie, just before roll-off).[23] For a CFC filter, F_H is equal to 0.6 multiplied by the −3 dB cut-off frequency (excluding CFC1000).[23]

If at least one frequency was below F_{max}, F_H was selected as the lower frequency associated with the 95th percentile PSD and local minima (if applicable). If neither frequencies were below F_{max}, the overall median CFC value for the dataset was applied. CFC60 was the lowest CFC filter applied to a kinematic signal unless a signal component exceeded 95% of the sensor range, then a CFC30 filter (−3 dB cut-off frequency of 50 Hz) was applied (<0.3% of cases). CFC30 attenuated the signal enough to not exceed the sensor range and 50 Hz is the lowest cut-off frequency previously applied by an iMG system.[6] F_H was applied to all sensor component channels. The analysis was conducted separately on both the linear and the angular acceleration pulse to mitigate attenuating signal when artefacts were more prominent in one sensor. An artificial 'raw' angular acceleration pulse was developed by filtering the raw angular velocity with a CFC filter with F_H=F_{max} and then differentiating the signal (five-point stencil method).[24] The initial filtering of the raw angular velocity data at F_{max} aided identifying a 95th percentile PSD frequency less than F_{max} after differentiation and ensured low magnitude events were less susceptible to low amplitude, high frequency noise in the signal. The HEADSport filter method code is openly available on Github.[25]

Data and statistical analysis

A continuous wavelet transform (CWT) was calculated for each laboratory impact and on-field HAE.[26] To assess the performance of the HEADSport filter method in a head impact reconstruction laboratory environment, SNR was calculated with reference to the raw signal and compared with a fourth order (2×2 pole), zero phase, low-pass Butterworth filter with −6 dB cut-off frequency of 200 Hz (Butterworth-200Hz) and a CFC180 filter applied to the raw headform linear acceleration and angular velocity signals.[27] The Butterworth-200Hz filter was selected to represent the Prevent Biometrics processed output if their in-house classification algorithm was not applied. A CFC180 filtering approach was included as it is the filter type with highest −3 dB cut-off frequency (300 Hz) used by an iMG system tested by Jones *et al*.[6]

The PLA at the head CG and PAA were extracted from each on-field HAE based on the (1) Prevent Biometrics processed output; (2) HEADSport filter method; (3) Butterworth-200Hz; (4) CFC180 filter. Median, IQR and maximum values for PLA and PAA were compared for each filtering method using mixed linear effects modelling, with an alpha level of $p<0.05$ and has previously been described in detail.[28 29] The HEADSport filter method, Butterworth-200Hz and CFC filters were applied to the raw kinematic signals and at the same stages as the Prevent Biometrics processed output. Transforming the linear acceleration signal at the iMG to the head CG was validated by getting the same results as the Prevent Biometrics processed output when using 200 Hz (−6 dB cut-off frequency) for class 0 impacts.

RESULTS

The maximum 95th percentile PSD frequency for the laboratory impacts was 312 Hz and was achieved using the rigid impactor at 50–150 g. The pulse duration ranged from 3.2 to 12.3 ms for the laboratory impacts. The acceleration pulse during the laboratory impacts followed a half-sine/haversine shape with the CWT scalogram illustrating a smooth frequency spectrum and PSD exhibiting no local minima deflection points (figure 2A–C). Median SNR (and IQR) for the HEADSport filter method and CFC180 filter were similar at 19.6 dB (17.8–21.1) and 19.7 dB (17.8–21.2), respectively, for linear acceleration. Higher median SNR was achieved by the HEADSport filter method (26.1 dB; IQR 21.9–30.0) compared with the CFC180 filter (24.8 dB; IQR 21.8–30.4) for angular velocity. Both the HEADSport filter method and the CFC180 filter had greater SNR than the Butterworth-200Hz for linear acceleration (16.4 dB; IQR 11.0–19.1) and angular velocity (−0.3 dB; IQR −1.0 to −0.1).

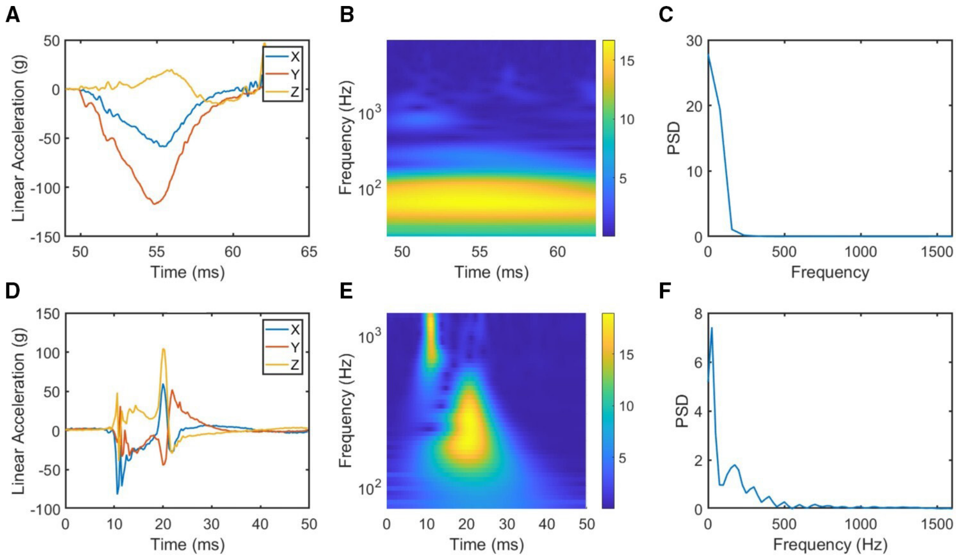

Figure 2 Linear acceleration time-series, continuous wavelet transform (CWT) scalogram (log-scale on the y-axis) and power spectral density (PSD) (impact pulse only, see the Methods section) for an example laboratory impact (A–C) and on-field HAE with localised area of high frequency and amplitude signal (D–F).

A total of 5694 video-verified HAEs with raw data signals were collected from the on-field study. The maximum 95th percentile PSD frequency for on-field HAEs was 1550 Hz and pulse duration ranged from 0.6 to >50 ms (50 ms was the iMG time window). The raw acceleration pulse during certain on-field HAEs did not follow a half-sine or haversine pulse shape with the CWT scalogram illustrating localised areas of high frequency and amplitude signal and PSD exhibiting local minima deflection points (figure 2D–F). For the HEADSport filter method, F_{max} was rounded up to 320 Hz (from 312 Hz). F_H was selected from the 95th percentile and first local minima PSD frequency more for linear than angular acceleration pulses (72.9% vs 62.7% and 21.0% vs 17.8%, respectively) with 5.8% and 19.2%, respectively, having neither within F_{max} (see figure 3). The median and IQR for both the linear and the angular F_H was 60 Hz (IQR: 60–100 Hz), with maximum values of 300 Hz. None of the 17 cases where a signal component exceeded 95% of the sensor range (figure 3) had a 95th percentile PSD frequency less

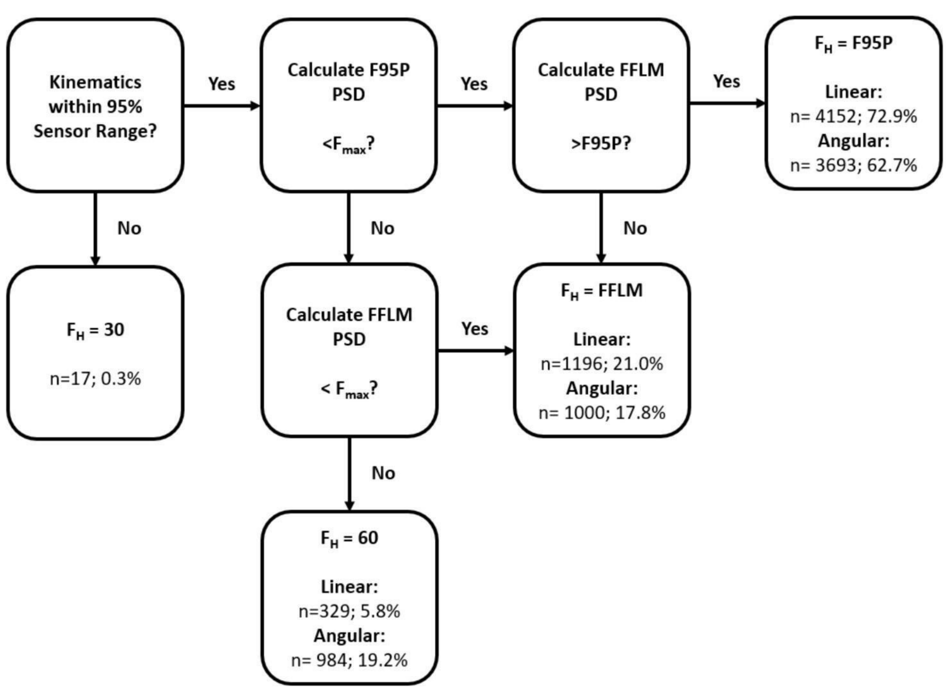

Figure 3 Flowchart illustrating the HEADSport filter method for selecting frequency class (F_H). F95P, frequency associated with 95th percentile PSD; FFLM, frequency associated with first local minima; F_{max}, maximum 95th percentile PSD frequency for the head impact reconstruction laboratory impacts; PSD, power spectral density.

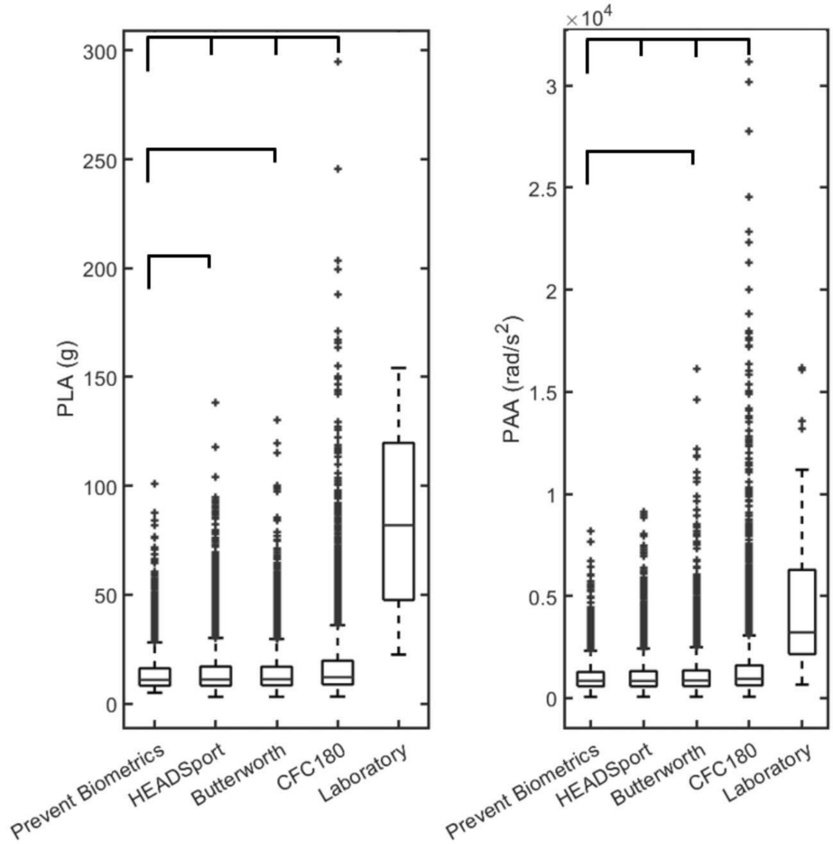

Figure 4 Peak linear acceleration (PLA) and peak angular acceleration (PAA) based on the Prevent Biometrics processed output, HEADSport filter method, Butterworth-200Hz and CFC180 filter approach. Boxplots indicate the median (box midline), IQR (box) and most extreme data points (whiskers) not considered outliers (crosses). Outliers are values greater than 1.5 times the IQR from top of the box. Horizontal lines above boxplots indicate statistical significance (p<0.05). Laboratory impacts were not included in the mixed linear effects models. Boxplots with outliers removed can be seen in online supplemental material B.

than F_{max} (cubic spline interpolation replaced the signal component datapoints that exceeded the sensor range).

Median and outlier values for PLA and PAA were higher when the CFC180 filter was applied to the raw kinematic signals in comparison to the Prevent Biometrics processed output (p<0.01), HEADSport filter method (p<0.01), Butterworth-200Hz (p<0.01) and head impact reconstruction laboratory kinematics (figure 4 and online supplemental file B). PLA and PAA values as high as 294 g and 31.2 krad/s^2 were reported using the CFC180 filter in comparison to 101 g and 8.2 krad/s^2 from the Prevent Biometrics processed output. Both the HEADSport and Butterworth-200Hz filter produced similar PLA and PAA values within the magnitude range tested in the head impact reconstruction laboratory. The HEADSport and Butterworth-200Hz filter approach produced marginally higher median PLA values than the Prevent Biometrics processed output (both p<0.01) (see online supplemental file B). The Butterworth-200Hz filter approach produced higher outlier PAA values than the HEADSport filter method. The linear and/or angular acceleration pulse did not follow a half-sine/haversine shape when the CFC180 and Butterworth-200Hz filters were applied to certain HAEs as high amplitude and frequency content were still present in the signal (figure 5).

DISCUSSION AND IMPLICATION

An openly available (via GitHub)[25] artefact attenuation filtering method for real-world/on-field iMG data has been developed based on the PSD characteristics of bare-headed laboratory impacts and on-field HAEs. The rationale for the development of the HEADSport filter method was that peak head kinematics differed considerably between different iMG systems used in similar rugby cohorts/environments (ie, academy-level rugby league match play).[6] The HEADSport filter method produced an overall higher SNR than the Butterworth-200Hz and CFC180 filter in the head impact reconstruction laboratory environment and on-field PLA and PAA values within the magnitude range tested in the head impact reconstruction laboratory. The HEADSport filter method was designed as a fast, computationally inexpensive algorithm to be applied to any commercially available iMG system. Inter-study comparisons can be achieved through the use of the method's fundamental rigid body mechanics and signal processing principles (ie, PSD) rather than machine learning approaches that require unique on-field training datasets and could be considered 'black box'.[30] Inter-study comparisons can allow head acceleration exposure and HAE mechanisms to be compared between sports. Additionally, a common signal

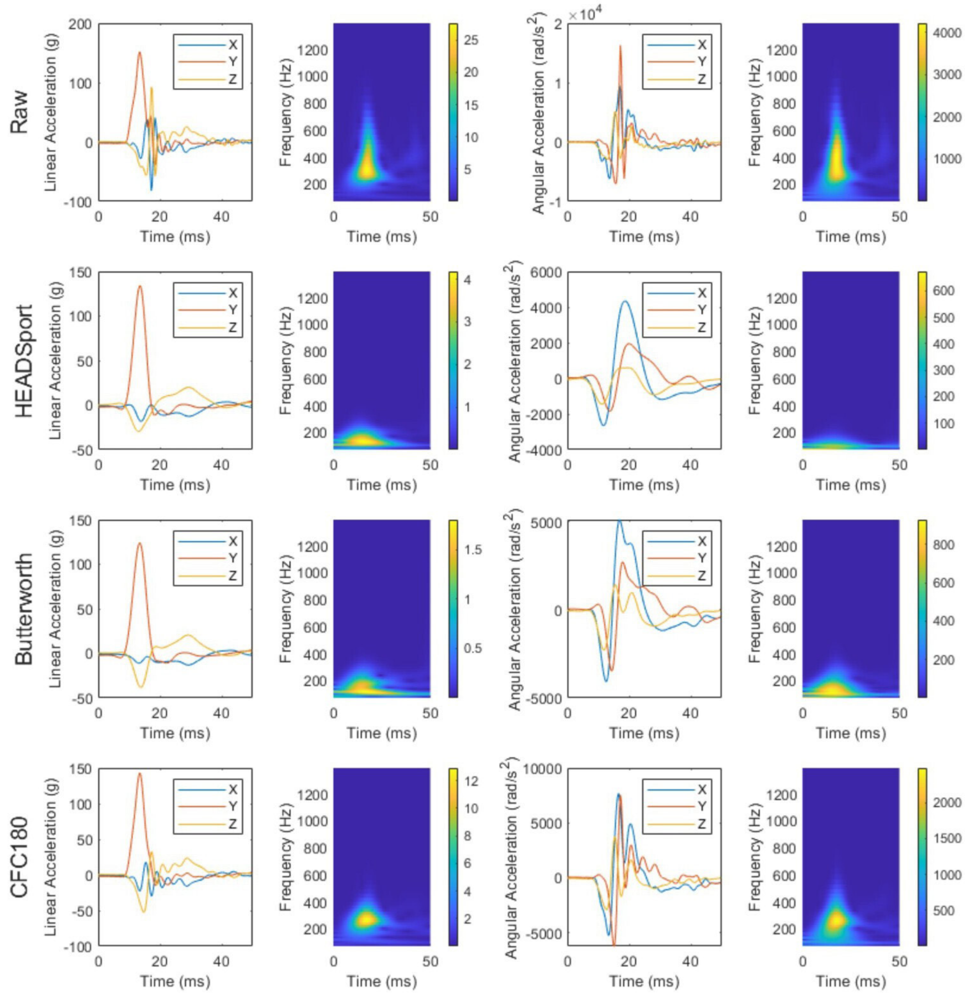

Figure 5 Linear and angular acceleration time series with corresponding continuous wavelet transform (CWT) scalogram on highest magnitude component for an example on-field HAE after the HEADSport filter method, Butterworth-200Hz and CFC180 filter approach were applied.

processing approach means that datasets can be merged for analysis. This would be particularly useful for analyses that typically lack high sample sizes (eg, kinematics of concussion injuries).[31]

Currently, the highest −3 dB cut-off frequency used by a commercially available iMG system is 300 Hz (CFC180 filter).[6] The HEADSport filter method enabled −3 dB cut-off frequencies as high as 500 Hz with the current dataset which is in line with the recommendations of Wu *et al*.[8] The cut-off frequency is also significantly below the excitation frequency of the skull, and thus complies with rigid body assumptions.[32] The median F_H value of 60 Hz (indicative of −3 dB cut-off frequency of 100 Hz) is low relative to other iMG system cut-off frequencies. However, this is likely due to the large number of indirect HAEs (ie, inertial head loading) in rugby HAE datasets which are lower frequency than direct HAEs.[16] Conducting PSD on the impact pulse enables commonality for different iMG systems that use different output time-windows (usually ranged from 50 to 200 ms).[6 11] The SAE J211 impact test standard states that sensor sample rates should be a minimum of 10 times F_H.[23] Therefore, with an F_{max} of 320 Hz, iMG sensors would need to sample at 3200 Hz as a minimum for the HEADSport filter method to measure short pulse duration HAEs. Some iMG systems sample at 1000 Hz which has previously been highlighted as a concern.[8]

The use of a CFC180 filter without any additional artefact attenuation processing resulted in the highest head kinematics for on-field data. This may explain why HAEs of roughly 500 g and >50 krad/s^2 were reported by an iMG system using a 300 Hz cut-off frequency in Jones *et al*.[6] That is, high amplitude and frequency content remained in the signal after filtering and combined with numerical calculations (eg, differentiation) and biomechanical transformations (eg, transforming linear acceleration signal to the head CG) resulted in higher head kinematics being reported.[12] The significant differences in median and outlier PLA and PAA kinematics produced by the different filtering methods (figure 4) illustrate the need for a common approach to kinematic signal processing.

Limitations

There are biofidelity issues with a NOCSAE headform in comparison to a human head for impact reconstructions.[18] However, the maximum cut-off frequency based on the 95th percentile PSD frequency (312 Hz) equates to 520 Hz which is similar to that reported for cadaver-based head impact reconstructions.[8] There was also a characteristic haversine/half-sine pulse for the laboratory head impacts which are common in impact biomechanics,[33] and aligns with equation 1. Laboratory head impacts were only conducted on the front and front boss impact locations allowing magnitudes up to 150 g to be analysed in the current study. However, the PSD analysis was also conducted on all laboratory head impacts from Jones *et al*[6] which included the front, front boss, rear boss and rear impact locations tested up to 100 g. The maximum 95th percentile PSD frequency for these additional laboratory impacts was also 312 Hz. Although a padded impactor was used as part of the testing conditions, F_{max} could be calculated for helmeted head impact reconstruction laboratory events and the HEADSport filter method assessed for helmeted contact sports (eg, American Football). The HEADSport filter method could be further assessed and adapted (eg, F_{max}) through well-designed head impact reconstructions with cadavers equipped with iMG and female-specific head-neck models.[13] Biofidelic iMG artefact reconstructions may be possible with cadaver-based head impacts.

For the on-field analysis, PSD calculations were conducted on an artificial raw angular acceleration pulse as described in the Materials and Methods section. Ideally, PSD would be conducted on true raw angular accelerometer signals. Angular accelerometers are used by certain iMG systems.[6] F_{max} was calculated based on bareheaded laboratory impacts. On-field ground truth data does not exist which makes validating the HEADSport filter method a challenge. However, the aim of the method is to allow inter-study comparison based on fundamental principles of rigid-body mechanics and signal processing. Future work should assess the HEADSport filter method on field-based datasets similar to Jones *et al*[6] where multiple iMG systems are used in the same sport and playing-level to see if comparable kinematics are produced between iMG systems. To enable field-based inter-study comparison, commonality/consensus is needed in terms of minimum sampling rates, sensor trigger and recording thresholds, head CG vector and calculation of parameters such as peak change in head angular velocity due to the high potential for non-zero head angular velocity values at the time of impact.[16] Additionally, commonality/consensus for what stages to apply the filter is important for comparable results. For laboratory data where true head kinematic signal is in the filter pass-band and noise in the stop-band, the filter only needs to be applied to the raw data before differentiation.[23] However, in real-world datasets, artefact noise may be present in the transition-band (eg, figures 2 and 5) and therefore only partially attenuated by filtering. Therefore, filtering after differentiation may be necessary as the amplitude of differentiated noise increases linearly with frequency.[12,17] Accordingly, the filter may also need to be applied after additional numerical calculations such as matrix transformations (eg, converting from iMG to SAE J211 coordinate system for angular kinematics) and this requires further investigation.

Peak kinematics differ considerably between different iMG systems used in similar cohorts/environments. An artefact attenuation filtering method (HEADSport filter method) for on-field iMG data has been developed based on the PSD characteristics of bare-headed laboratory impacts. The HEADSport filter method is based on fundamental rigid body mechanics and signal processing principles. The method can be applied to any raw iMG kinematic signals measured with adequate sensor specifications (eg, bandwidth and sample rates) to enable field-based inter-study comparisons. Ideally, the HEADSport filter method could contribute towards the development of a standard or consensus for processing sports-based head sensor kinematic signals that progresses in line with the state-of-the-art, similar to the automotive industry.

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Contributors GT conceptualised the research project and designed the study. All authors collected data for the study. GT developed the filter algorithm. All authors were responsible for analysis and interpretation of the results. GT drafted the manuscript. All authors critically reviewed and edited the manuscript prior to submission. GT is the guarantor for this work.

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Patient and public involvement Patients and/or the public were not involved in the design, or conduct, or reporting or dissemination plans of this research.

Patient consent for publication Not applicable.

Ethics approval This study involves human participants and was approved by the University Research Ethics Committee (UREC), Ulster University (#REC-21-0061). Participants gave informed consent to participate in the study before taking part.

Provenance and peer review Not commissioned; externally peer reviewed.

Data availability statement Data are available upon reasonable request. Anonymous data are available on reasonable request to the corresponding author.

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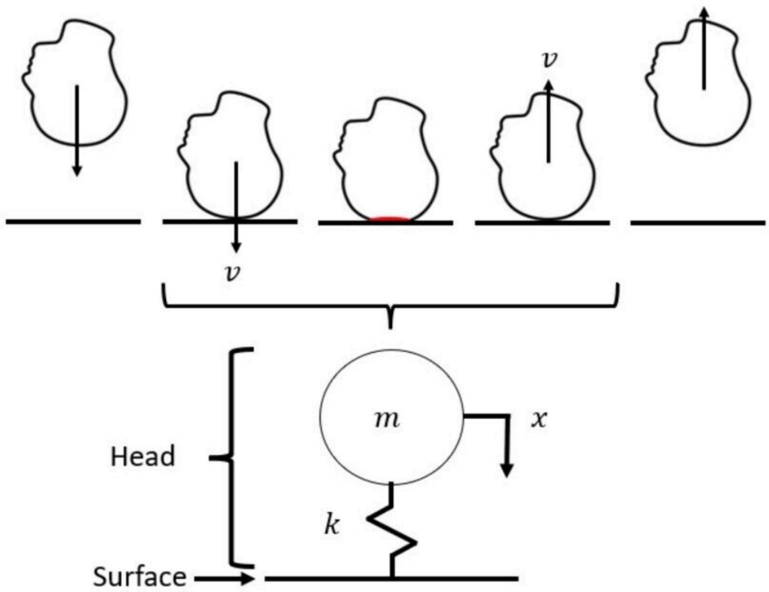


Spring mass system to represent head impact:

$$m\ddot{x}(t) + kx(t) = 0; \qquad \omega_n = \sqrt{\frac{k}{m}}$$

$$x(t) = \quad x(0)\cos(\omega_n t) \; + \; \frac{\dot{x}(0)}{\omega_n}\sin(\omega_n t)$$

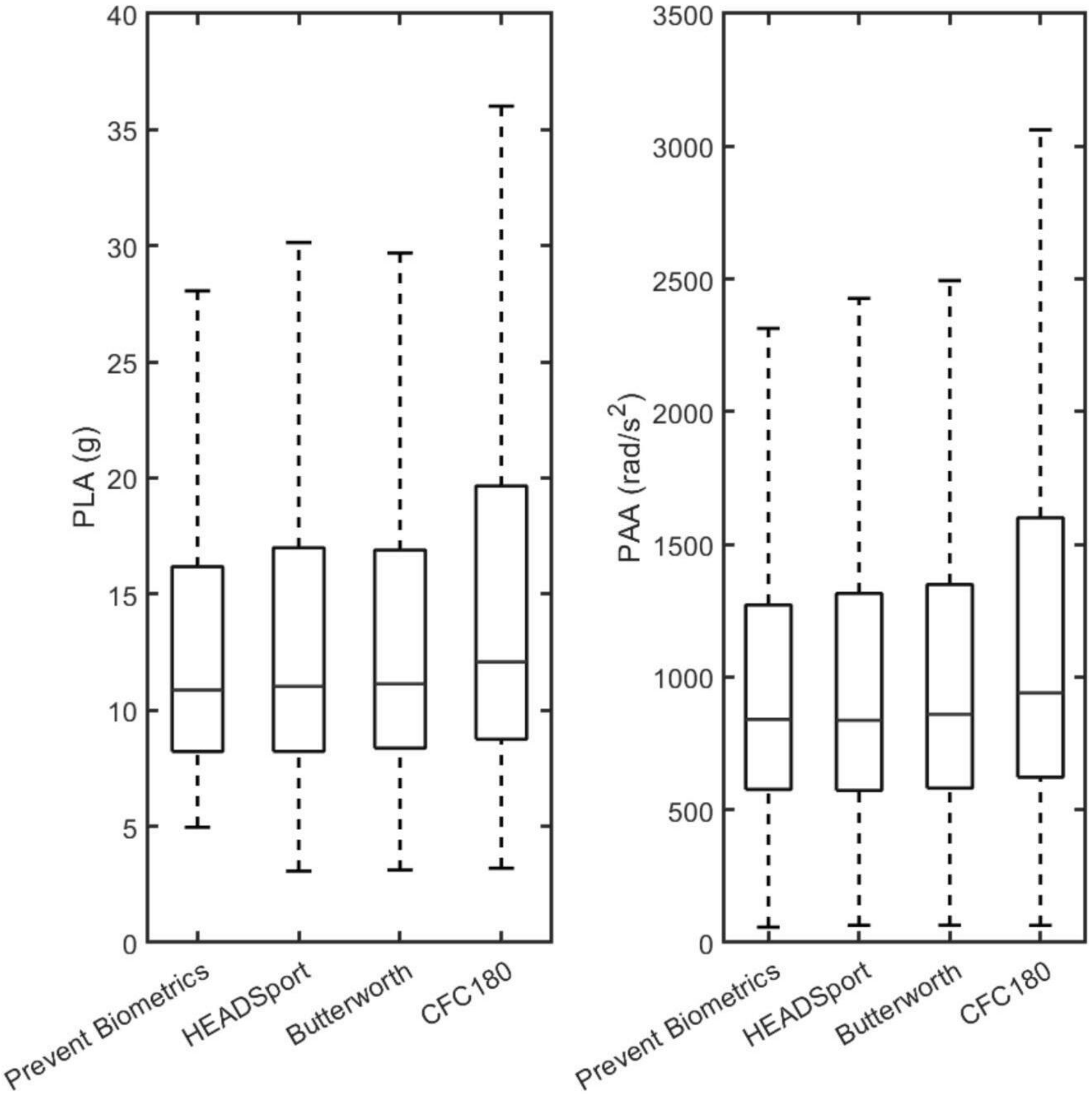
for known $x(0) = 0$ and $\dot{x}(0) = v$;

$$x(t) = \frac{v}{\omega_n}\sin\omega_n t$$

$$\dot{x}(t) = v(t) = v_i\cos\omega_n t$$

$$\ddot{x}(t) = a(t) = -v_i\,\omega_n\sin\omega_n t = -v_i\sqrt{\frac{k}{m}}\sin\omega_n t$$

Key:
m = Head Mass
k = Head Stiffness
ω_n= Head Natural Frequency
v_i = Head Initial Velocity
x = Head Displacement
v = Head Velocity
a = Head Acceleration


PLA (g)
40
35
30
25
20
15
10
5
0
Prevent Biometrics
HEADSport
Butterworth
CFC180
PAA (rad/s^2)
3500
3000
2500
2000
1500
1000
500
0
Prevent Biometrics
HEADSport
Butterworth
CFC180