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Bandwidth and sample rate requirements for wearable head impact sensors

Lyndia C. Wu [a,*], Kaveh Laksari [a], Calvin Kuo [b], Jason F. Luck [c], Svein Kleiven [d], Cameron R. 'Dale' Bass [c], David B. Camarillo [a,b]

[a] Department of Bioengineering, Stanford University, Stanford, CA 94305, USA
[b] Department of Mechanical Engineering, Stanford University, Stanford, CA 94305, USA
[c] Department of Biomedical Engineering, Duke University, Durham, NC 27708, USA
[d] Department of Neuronic Engineering, KTH Royal Institute of Technology, Stockholm, Sweden

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ABSTRACT

Wearable inertial sensors measure human head impact kinematics important to the on-going development and validation of head injury criteria. However, sensor specifications have not been scientifically justified in the context of the anticipated field impact dynamics. The objective of our study is to determine the minimum bandwidth and sample rate required to capture the impact frequency response relevant to injury. We used high-bandwidth head impact data as ground-truth measurements, and investigated the attenuation of various injury criteria at lower bandwidths. Given a 10% attenuation threshold, we determined the minimum bandwidths required to study injury criteria based on skull kinematics and brain deformation in three different model systems: helmeted cadaver (no neck), unhelmeted cadaver (no neck), and helmeted dummy impacts (with neck). We found that higher bandwidths are required for unhelmeted impacts in general and for studying strain rate injury criteria. Minimum gyroscope bandwidths of 300 Hz in helmeted sports and 500 Hz in unhelmeted sports are necessary to study strain rate based injury criteria. A minimum accelerometer bandwidth of 500 Hz in unhelmeted sports is necessary to study most injury criteria. Current devices typically sample at 1000 Hz, with gyroscope bandwidths below 200 Hz, which are not always sufficient according to these requirements. With hard contact test conditions, the identified requirements may be higher than most soft contacts on the field, but should be satisfied to capture the worst contact, and often higher risk, scenarios relative to the specific sport or activity. Our findings will help establish standard guidelines for sensor choice and design in traumatic brain injury research.

1. Introduction

Wearable head impact sensors measure biomechanics data for traumatic brain injury (TBI) research. Since the 1940s, a number of biomechanics-based head injury criteria have been proposed to quantify risks for both severe and mild TBI, including criteria based on head translation (Versace, 1971; Pellman et al., 2003), rotation (Ommaya and Hirsch, 1971; Margulies and Thubault, 1992), and brain deformation from finite element models (Hernandez et al., 2014; Sullivan et al., 2015). Yet to date, there is still no consensus on which injury criteria sufficiently characterize TBI, and the injury mechanism of mild TBI is especially elusive (Guskiewicz and Mihalik, 2011). To close this gap, many researchers are collecting human head impact data to compare and validate currently proposed injury criteria. Thanks to the ubiquity of micro-electro-mechanical sensors (MEMS), low-cost and low-power wearable sensors can now be mass-deployed to at-risk human subject populations.

When selecting sensors for measuring head impacts, it is important to consider sensing requirements of the anticipated head impact scenarios. One key consideration that is often overlooked is the frequency content of head impact response, which determines sensor specifications such as bandwidth and sampling rate. Voluntary human motion is usually dominated by low-frequencies, ranging from a few hertz to tens of hertz (Antonsson and Mann, 1985). However, blunt impacts to the head last only up to tens of milliseconds and may have substantial power at higher frequencies. A helmeted rat head impact model demonstrated that low-pass filter cutoffs should be 1000 Hz or more to stay within a

* Correspondence to: 443 Via Ortega, Stanford, CA, USA. fax: +650 723 8544.
E-mail address: lyndiacw@stanford.edu (L.C. Wu).

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few percent error in kinematics (Fijalkowski et al., 2009), but it is unclear how this finding translates to humans. It is also unclear what frequencies the brain would be most sensitive to, but previous research has found a brain resonance frequency of around 15 Hz (Laksari et al., 2015).

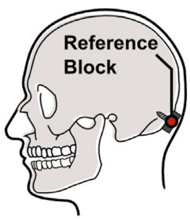
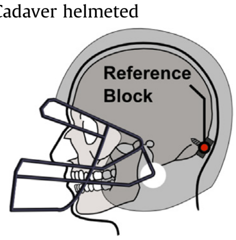
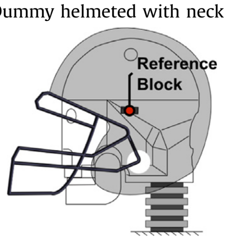
Despite this uncertainty, a variety of head impact sensors were developed, and some have been piloted in various helmeted and unhelmeted sports (Rowson et al., 2009; Urban et al., 2013, Wilcox et al., 2014; King et al., 2014; Hanlon and Bir, 2012). Only a small number of such sensors have published specifications (Table 1). The linear accelerometer bandwidths range from 300 to 2000 Hz, with sampling rates from 1000 to 4000 Hz. The gyroscope bandwidths range from 100 to 184 Hz, with sampling rates from 300 to 1000 Hz. While sensor specifications were reported, the authors did not justify the choice of sampling rate and bandwidth. The increased interest in this research area coupled with the proliferation of sensors highlights the importance of investigating whether current sensor specifications are sufficient to capture the head impact response relevant to injury. The answer will depend on the subject population and injury criteria under study, due to anticipated differences in frequency response of different types of contact and varying sensitivities to different frequency components of kinematics for each injury criterion.

The objective of this study is to determine sensor bandwidth requirements over a range of head impact model systems for studying common head injury criteria. The results of this study will help inform sensor design and choice for different applications.

2. Methods

We conducted impact tests in three different model systems (Table 2): helmeted cadaver head drop (no neck), unhelmeted cadaver head drop (no neck), and dummy head linear impact (with neck), at common impact locations and medium to high linear acceleration levels observed from football impacts on the field (Hernandez et al., 2014). With these test conditions, we can account for the following factors: biofidelity, presence of a helmet, presence of a neck, and different impact locations. From each impact, we collected high-bandwidth accelerometer and gyroscope measurements as ground truth skull kinematics. These measurements were low-pass filtered to generate signals representing lower bandwidth measurements. We then investigated the resulting attenuation in injury risk predictions for skull kinematics-based (calculated directly from sensor signals) and brain deformation-based criteria (calculated from finite element brain model simulations), which helped inform minimum bandwidth requirements.

Table 1
Review of published head impact sensor specifications.

Model system	Method	Conditions
Cadaver unhelmeted 	Drop test onto rigid platen	Frontal oblique Frontal Occipital Parietal Vertex Occipital Oblique
Cadaver helmeted 	Drop test onto rigid platen	Frontal oblique Facemask Frontal Occipital Parietal Vertex
Dummy helmeted with neck 	Linear impact with stiff elastomer impactor	Frontal oblique Facemask Frontal Occipital Parietal

2.1. Experimental data

The cadaveric free head drop tests were conducted with a male head specimen. The head was disarticulated from the cervical spine at the atlanto-occipital joint. It was dropped in either a helmeted or unhelmeted configuration, each at six different impact locations onto a platen with a 3.1 mm thick nominal Shore 40 A layer of neoprene covering an aluminum plate (Table 2). The drop height is 100 cm (head speed of 4.43 m/s) for the helmeted case and 18 cm (1.88 m/s) for the unhelmeted case. The helmeted drops resulted in linear accelerations of 100–160 g, and unhelmeted drops were from 90 to 30 g. Both had angular velocities ranging from 4 to 18 rad/s. With two conditions (helmeted and unhelmeted) and 6 impact locations, a total of 12 impacts were included.

For the linear impactor tests on a dummy head, we used a previously published anthropomorphic test device that represents a 50th percentile male human head (Camarillo et al., 2013). The head was mounted on a Hybrid III neck and impacted by a stiff elastomer impactor. We tested the same impact locations as the cadaver head drops, with the exception of the vertex location, which could not be realized with the linear impactor setup (Table 2). We used a spring combination that corresponds to an impactor speed of 6.7 m/s. These tests generated linear accelerations of 50–70 g and angular velocities of 16–28 rad/s.

2.2. Signal processing and filtering

The kinematic signals from the cadaver drops were collected using a triaxial accelerometer (Endevco 7624C-2000) and triaxial gyroscope (ARS-PRO-8k) block (3aω setup) rigidly attached to the occipital bone. Signals were sampled at 100 kHz for 600 ms, with 100 ms pre-impact and 500 ms post-impact. The accelerometer has a bandwidth of 25 kHz, limited by an analog anti-aliasing filter. The gyroscope has a datasheet-specified bandwidth of 2 kHz. We filtered linear acceleration and angular velocity at sensor bandwidths (25 kHz and 2 kHz, respectively) to obtain ground truth measurements. Angular acceleration ground truth signals were obtained through five-point stencil differentiation on ground truth angular velocity signals. Please note that even though ‘ground truth’ is used to describe these reference signals, we recognize that there could be error in these measurements.

The dummy head had a 6aω setup, which allows for calculation of angular acceleration without differentiation (Kang et al., 2011). A triaxial accelerometer (Dytran 3273A1) was mounted at the center of gravity of the head. Three single axis accelerometers (Dytran 3255A1) were mounted offset from the center of gravity (CG), on the periphery of the head. A triaxial gyroscope (ARS-PRO-18K) was also attached for angular velocity measurements. Data were collected at 100 kHz for 600 ms, with 100 ms pre-impact and 500 ms post-impact. Datasheet-specified bandwidth of the accelerometers is 10 kHz, and that of the gyroscope is 2 kHz. We filtered linear acceleration and angular velocity at sensor bandwidth (10 kHz and 2 kHz, respectively) to obtain ground truth measurements. Angular acceleration ground truth was computed without differentiation, using ground truth 6aω data filtered to the gyroscope bandwidth limit (2 kHz), according to methods detailed in Kang et al. (2011).

We then used different low-pass filter cutoffs to simulate sensor signals collected at lower bandwidths. A fourth-order Butterworth low-pass filter according to SAE Standards (SAEJ211-1, 2007) was used for the analysis. We applied the filter at 50 Hz, 100 Hz (CFC 60), 300 Hz (CFC 180), 500 Hz, 1000 Hz (CFC 600), and 1650 Hz (CFC 1000). To investigate the effects of linear acceleration and angular velocity filtering separately, we tested the following signal combinations for each impact: ground truth linear acceleration combined with each filtered angular velocity signal, and ground truth angular velocity combined with each filtered linear acceleration signal, totaling 13 cases for each impact including ground truth.

2.3. Calculating injury criteria

Using the ground truth and lower bandwidth signal combinations, we computed both skull kinematics-based and brain deformation-based injury criteria. For calculating skull kinematics-based criteria, ground truth sensor signals were projected to the center of gravity of the head and transformed to anatomical axes. Linear acceleration based injury criteria included peak linear acceleration magnitude (PLA), HIC15, and SI. Angular velocity or angular acceleration based injury criteria included peak angular acceleration magnitude (PAA), peak change in angular velocity magnitude (PAV), brain injury criteria (BrIC), rotational injury