Annals of Biomedical Engineering, Vol. 48, No. 9, September 2020 (© 2020) pp. 2310–2322
https://doi.org/10.1007/s10439-020-02504-1

BMES BIOMEDICAL ENGINEERING SOCIETY

Original Article

A Novel Testing Device to Assess the Effect of Neck Strength on Risk of Concussion

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(*Received 14 November 2019; accepted 30 March 2020; published online 6 April 2020*)

Associate Editor Joel Stitzel oversaw the review of this article.

Abstract—Concussion awareness has become more prevalent in the past decade, leading to growing calls for prevention programs such as neck strengthening. However, previous research work has shown that not all training programs have been effective, and there is a need for a reliable testing device to measure cervical strength dynamically before and after training. Therefore, this work proposes a novel Concussion Active Prevention Testing Device composed of inertial measurement units mounted on the head and a custom-designed frame to measure head kinematics during controlled sub-concussive impacts. Through an experimental study with able-bodied participants, the proposed testing device demonstrated high intra-participant repeatability between waveforms of the head acceleration and angular velocity in the sagittal plane (multiple correlation coefficient of 80%). Similarly, good and excellent intra-class correlation coefficients were obtained for head injury metrics, including range, peak, Gadd severity index, head injury criterion, and range of motion. Finally, the results showed that significantly higher head injury metrics were measured for female participants, which was in line with the findings of previous research works. We conclude that the proposed testing device can be used to measure repeatable and informative metrics for evaluating the effectiveness of athletes' neck strengthening program.

Keywords—Dynamic neck strength measurement, Head injury metrics, Head kinematics, Inertial measurement units.

ABBREVIATIONS

HIM	Head injury criterion
GSI	Gadd severity index
BIC	Brain injury criterion
IMU	Inertial measurement unit
CAPTD	Concussion active prevention testing device
CG	Head center of gravity
ROM	Head range of motion
CMC	The coefficient of multiple correlations
ICC	The intra-class correlation coefficient

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INTRODUCTION

Estimations by the Centers for Disease and Prevention show that 300,000 sport-related mild traumatic brain injuries, also known as concussions, were occurring annually in the United States,[35] and concussions have become a growing concern around the world in recent years.[16] Repetitive concussions can cause long-term neurodegenerative processes among athletes.[34] In most grading systems, the identification of concussion is based on symptoms. According to the "Evidence-Based Cantu Grading System for Concussion," the severity of the concussion can be categorized as one of the following: (1) grade 1 or mild (no loss of consciousness); (2) grade 2 or moderate (loss of consciousness lasting less than 1 min); grade 3 or severe (loss of consciousness lasting more than 1 min or posttraumatic amnesia lasting longer than 24 h).[4] Contributing factors can be categorized into (1) the intensity of the impact; (2) the number of impacts; and (3) type of injury: direct impact or acceleration-deceleration. Recent surveys showed that concussions could occur in (1) various sports, (2) at different ages, and (3) among both male and female athletes. For instance, Patel *et al.*[25] reported the occurrence of 189 concussions in the NBA from 1999 to 2018. Also, using instrumented helmets, Wilcox *et al.*[39] recorded 37,411 impacts during three seasons of collegiate ice hockey.

The concussion prevention programs designed to date can be categorized into three main groups: (1) improving helmet design and the manufacturing process; (2) devising new rules to limit the number and severity of head impacts; and (3) educating coaches, athletes, and judges of the importance of concussion prevention.[11] Although these programs have been effective in reducing the effects of concussion, they each have limitations. For example, previous research showed that certain protective equipment might prevent superficial head injury, but these items were suboptimal for concussion prevention in sport.[33] Also, through an in-lab experimental study with four American football helmets, Cournoyer *et al.*[8] showed that linear acceleration sensed over four (out of six) tested locations of the helmet increased up to 14 g after repetitive impacts. Therefore, secondary measures, such as specific physical training programs, also must be taken into consideration.

Neck strengthening can be considered the primary concussion prevention approach introduced so far.[6] For example, neck strengthening programs have been effective in increasing the neck strength in athletes and reducing neck pain in pilots.[15,31] Therefore, we hypothesized that regular, customized, self-directed neck strengthening training could reduce the risk of long-term complications of concussion, specifically in youth and female athletes.[5,14] For instance, Hislop *et al.*[14] showed that for school-aged male rugby players, those who completed the training program suffered 72% fewer overall match injuries, 72% fewer contact-related injuries, and 50% fewer days lost due to contact injuries. However, to understand the effectiveness of a neck strengthening program, injury metrics are required to quantify the risk of concussion based on measurable metrics before and after training. To this end, several injury metrics have been introduced and reported in the literature, including head injury criterion, peak translational acceleration, peak rotational velocity, Gadd severity index, and brain injury criterion.[30] Although previous research successfully showed a correlation between head injury metrics and the severity of the concussion, no attempt has been made to evaluate the effect of neck strengthening using such metrics.

Commonly, muscle strength has been measured using a hand-held dynamometer or myometer. As a cost-effective alternative tool for high schools, Collins *et al.*[6] proposed the application of a hand-held tension scale. It has also been demonstrated that the risk of concussion was directly associated with smaller neck circumference and weaker overall neck strength. Hall *et al.*[12] introduced a custom-designed frame, equipped with a digital force gauge as a standardized tool for isometric neck strength measurement. However, previous works reported inconsistent reliability for these measurement technologies due to the procedural differences like considering the mean value versus the maximum value of the contraction force or different contraction times.[12]

Moreover, Mihalik *et al.*[18] showed that, because neck strength is a protective factor during head impacts, the *static* cervical muscle strength measurements might not be able to capture the head accelerations during impacts. They suggested that future studies should measure head impact biomechanics *dynamically*. Other research has shown that, in contrast to in-lab static measurements, in-field measurement of the head kinematics during impacts was correlated with the risk of concussion. For example, Tierney *et al.*[36] showed the positive effect of isometric neck strength and girth on reducing the head acceleration, correlation coefficients of -0.48 and -0.47, respectively, caused by impacts during soccer among male and female young adults. Similarly, Bretzin *et al.*[2] showed that neck girth and strength had negative correlations with linear acceleration (correlation coefficients $= -0.60$) and rotational velocity (correlation coefficients $= -0.65$) of the head during head impacts of soccer players. However, experimental assessment of the neck response to concussive impacts in a laboratory setup is not ethical, and only three studies have used custom-designed frames to measure head kinematics during sub-concussive head impacts.[10,32,37] These studies measured head kinematics using stationary and expensive motion-capture systems only available in clinical motion analysis laboratories. Moreover, the custom-designed frames only simulated direct head impacts and not "body checks."

The objective of this study was to develop and validate an instrumented device to objectively evaluate the effect of neck strength on concussions caused by successive "body checks" that may occur during American football and ice-hockey, and similar impacts during soccer and other contact sports. For this purpose, we used a novel Concussion Active Prevention Testing Device (CAP Corp, Canada) composed of a custom-designed frame to measure neck strength during a controlled whiplash test (provisional patent[7]). We equipped this device with inertial measurement units (IMUs) to measure head kinematics during a sub-concussive chest impact in a precise and repeatable manner. Notably, the proposed testing device used IMUs to measure the effect of neck muscle strength on the kinematic response of the head during impacts applied to the chest (not directly to the head). The repeatability and sensitivity of the proposed testing device were evaluated through an experimental study by measuring five standard head injury metrics (HIMs): (1) repeatability of the testing device for HIM measurement was validated *via* the coefficient of multiple correlations (CMC) and intra-class correlation coefficient (ICC) among repeated trials; and (2) sensitivity of the measured HIMs to neck strength was validated by comparing HIMs of male and female participants. We used the HIMs previously established in the literature to assess the sensitivity and repeatability of the proposed testing device and experimental procedure in recording the head kinematics in response to controlled impact. As such, this study did not aim to evaluate the performance of any HIM.

The remainder of this paper has been organized as follows. "Materials and methods" section provides the details of the proposed testing device, the HIMs used, experimental study, calibration procedure, and repeatability/sensitivity evaluation. Repeatability of the testing device, and its ability to identify sex differences in HIMs, are presented in the "Results" section. Finally, the "Discussion" section highlights the capabilities of the proposed testing device and its comparison with the literature, and limitation/future works.

MATERIALS AND METHODS

Participants Demographic and Recruitment

The repeatability and sensitivity of the proposed testing device were evaluated through an experimental study with fourteen able-bodied participants (7 males: 35 ± 10 years old, 85 ± 14 kg, 179 ± 7 cm; 7 females: 27 ± 5 years old, 70 ± 8 kg, 167 ± 6 cm). The Research Ethics Board Committee of the University of Alberta approved the study protocol, and written consent was obtained from all participants. Participants were recruited through poster or other means of advertisement.

Concussion Active Prevention Testing Device (CAPTD)

The CAPTD was designed to simulate a controlled sub-concussive chest impact (Fig. 1a). For this purpose, a rope connected to a series of weights at one end was connected to the chest of the participant through a custom-designed harness. When the study coordinator released the lever, an impulsive impact caused by the sudden release of weights was transferred to the participant's chest *via* the rope and harness. The intensity of the impact can be increased by adding weights. Also, the delay between releasing the lever and applying the force to the participant's chest, known as the level of impact anticipation, can be controlled by the slack of the rope. Finally, the direction of the force can be tuned by changing the height of the horizontal adjustable rod.

The head kinematics were measured *via* two commercially available IMUs (MTws, Xsens Technologies, The Netherlands) attached to the participant's head (over the head and under the chin) with a head harness. Each IMU included a tri-axial accelerometer (range: ± 16 g) and a tri-axial gyroscope (range: ± 2000 deg s^{-1}) and measured the 3D linear acceleration and 3D angular velocity in a coordinate system shown in Fig. 1b. Additionally, another IMU was attached to the rope to identify the impact instant. All IMUs recorded data with a sampling frequency of 100 Hz (100 samples/second) and resolution of 16-bits/sample synchronously and transferred data wirelessly to a computer. To ensure that the sampling frequency of the IMUs was high enough to capture the head kinematics and avoid aliasing, frequency domain analysis (plotting the power spectral density versus frequency for the raw accelerometer and gyroscope readouts) was performed. The power spectral density graphs showed that almost all of the frequency contents of the signals with the heaviest weight (W3 = 52 kg) were contained in the range of 0 to 20 Hz. Therefore, in line with the Nyquist–Shannon sampling theorem, signals were low-pass filtered using a zero-phase 4^{th}-order digital Butterworth filter with the cut-off frequency of 40 Hz to remove the high-frequency noise and avoid aliasing.

Head Injury Metrics

Five HIMs were calculated using the head kinematics obtained by the head and chin IMUs to evaluate the effectiveness of a customized training program. The following HIMs were selected as they were shown to be predictive metrics of the risk of concussion among young and adult athletes of both genders.[18,27,30,38]

(1) *Range*: Difference between maximum and minimum values of a measured translational acceleration or angular velocity of the head center of gravity (CG).

(2) *Peak*: Maximum of the absolute value of the measurements above.

(3) *Gadd severity index (GSI)*: GSI was introduced by the National Operating Committee on Standards for Athletic Equipment as in Eq. (1):

$$\mathrm{GSI} = \int^{T} A_{CG,r}(t)^{2.5}\mathrm{d}t, \tag{1}$$

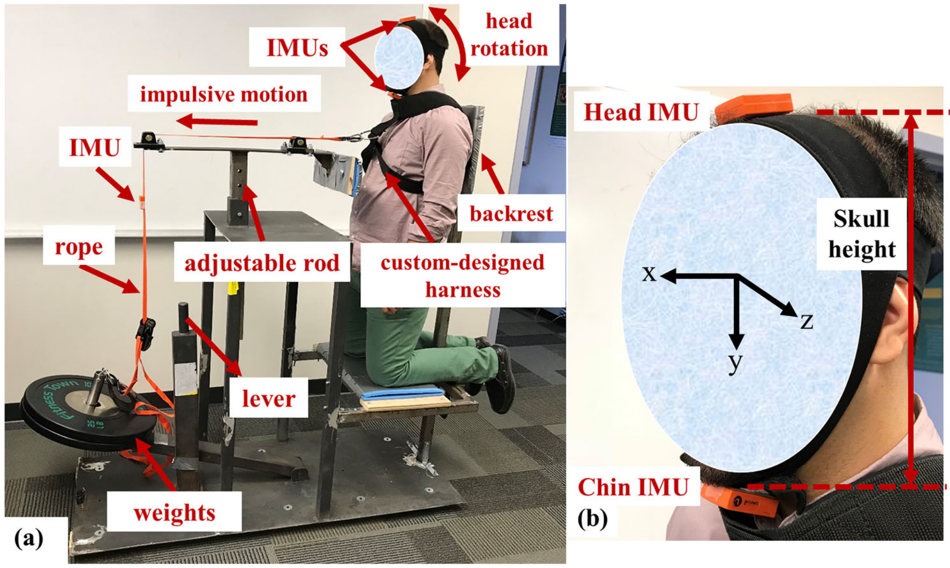

FIGURE 1. (a) The custom-designed testing device; (b) The local coordinate system defined for the head kinematics measurement using two IMUs attached over the head and under the chin.

where $A_{CG,r}(t)$ was the resultant acceleration of the head CG in m s^{-2} and T was the duration of the $A_{CG,r}$ in seconds.

(4) *Head injury criterion (HIC)*: HIC was introduced by the National Highway Traffic Safety Administration as in Eq. (2)[38]:

$$\mathrm{HIC} = \left\{ (t_2 - t_1) \left[\int_{t_1}^{t_2} A_{CG,r}(t) \mathrm{d}t / (t_2 - t_1) \right]^{2.5} \right\}_{\max} \quad (2)$$

where t_1 and t_2 were determined to maximize the HIC. In practice, to calculate HIC, we integrated $A_{CG,r}(t)$ in a window of $t_2 - t_1 = 50$ ms symmetrically around the peak of $A_{CG,r}(t)$. Both GSI and HIC were developed to assess the risk of injury in football helmet and car crash tests.

Since the head and chin IMUs measured acceleration in their local coordinate system, and not the head CG acceleration, a transformation was required to convert accelerations measured by IMUs to head CG acceleration as in Eq. (3):

$$\overrightarrow{A_{CG}} = \overrightarrow{A_{IMU}} - \vec{\omega} \times (\vec{\omega} \times \vec{r}) - \dot{\vec{\omega}} \times \vec{r} \quad (3)$$

where $\overrightarrow{A_{CG}}$ was the head CG acceleration in m s^{-2}, $\overrightarrow{A_{IMU}}$ was the measured acceleration by the head or chin IMU,$\vec{\omega}$ and $\dot{\vec{\omega}}$ were the angular velocity (in rad s^{-1}) and acceleration of the head, respectively, and $\vec{r}$ was the position vector between head CG and the head or chin IMU. We placed the head and chin IMUs in a way that the head CG was along their y-axis and thus $\vec{r}$ was along their y-axis. Also, the head CG is located approximately 40% below the vertex.[41] Thus, Eq. (3) was simplified as Eq. (4):

$$\begin{aligned} A_{CG,x} &= A_{IMU,x} - r_y(\omega_x\omega_y - \dot{\omega}_z) \\ A_{CG,y} &= A_{IMU,y} - r_y(\omega_x^2 + \omega_z^2) \\ A_{CG,z} &= A_{IMU,z} - r_y(\omega_y\omega_z + \dot{\omega}_x) \end{aligned} \quad (4)$$

where subscripts x, y, and z indicate each vector's components in the frame shown in Fig. 1. Thus, r_y was approximated by 40% of skull height (as shown in Fig. 1b). Also, $\dot{\vec{\omega}}$ calculated by a four-point central difference numerical differentiation.

In addition to $A_{CG,r}(t)$, we calculated the range, peak, GSI, and HIC metrics for head resultant angular velocity (ω_r) in the same fashion, since previous research showed that ω_r was an accurate and predictive metric for evaluating the risk of injury.[13,29] As the whiplash test was conducted in the sagittal plane of the body, we also assessed the head CG acceleration in anterior ($A_{CG,x}$) and upward ($A_{CG,y}$) directions and head angular velocity in the sagittal plane (ω_s) in addtion to $A_{CG,r}$ and ω_r.

(5) *Head range of motion (ROM)*: The range of rotational motion of the head in the sagittal plane (rotation around z-axis according to Fig. 1b) calculated using the orientation provided by the IMU.[23] However, as the duration of the test was short (less than 15 s), strap-down integration of the head angular velocity could alternatively be used for orientation estimation.

Experimental Study

To assess the repeatability and sensitivity of the testing device, the protocol of the experiment was designed as follows. After kneeling in the appropriate position according to Fig. 1a and selecting the appropriate weight (will be described later), the rope was used to connect the rod carrying weights to the chest of the participant *via* the custom-made harness. Then, the participant was asked to lean on the backrest, close their eyes, and tilt his/her head 20° backward and wait for a few seconds until the study coordinator released the lever.

To have the same experimental condition among all participants, the height of the rope was adjusted to make the rope horizontal and insert horizontal force at the very beginning of the impact. To lower the level of impact anticipation, the lever was released randomly after a few seconds. To choose the weights, several preliminary experiments were performed with multiple participants in which head kinematics was measured for various weights. The effect of increasing weight was then evaluated on repeatability and sensitivity of the HIMs. Additionally, to ensure the safety of the participants, an upper limit was considered for the selected weights; see "Limitations and future works" section for more details. According to the results of the preliminary study, the main experiments were performed with three different weights (W1 = 23 kg, W2 = 43 kg, W3 = 52 kg) to assess the effect of impact severity on the head kinematics. For each participant and each weight, three trials were performed to assess the repeatability of the testing device and experimental conditions. The impact instant was detected by applying a threshold (3.3 g, 2.3 g, and 3.2 g for W1, W2, and W3, respectively) to the resultant acceleration of the rope IMU. After impact detection, the IMU readouts were cut for 2 s, [impact-0.5 s, impact + 1.5 s] to capture the head kinematics before, during, and after impact.

IMU Local Frame Calibration

It was not practical to visually align the IMU local frame with the reference coordinate system shown in Fig. 1b. Therefore, to align the IMU local frame with the reference frame, we used a calibration procedure introduced by our research group.[21,22] To this end, before the main trials, we asked participants to kneel upright for 10 s and then perform ten consecutive lumbar flexion/extensions. Acceleration readouts during quiet sitting were then used to align the vertical axis of the IMU local frame with y-axis of the reference frame while the planar angular velocity of the IMU during lumbar flexion/extension was utilized to align the frontal and lateral axes of the IMU local frame with the x-axis and z-axis of the reference frame, respectively.[21]

Frequency Domain Analysis

The power spectrum of the IMU readouts for all trials with the highest weight showed that the sampling frequency of 100 Hz was sufficient for capturing head kinematics during the whiplash test.

Repeatability of the Testing Device

To assess the intra-participant repeatability (repeatability between the three trials for each participant and each weight), two assessments were performed: (1) the CMC[1] was calculated for $A_{CG,x}$, $A_{CG,y}$, $A_{CG,r}$, ω_s, and ω_r time-series among the three trials for each (participant, weight) pair; and (2) the ICC[17] was calculated for the HIMs mentioned in "Concussion Active Prevention Testing Device (CAPTD)" section of three trials for each (participant, weight) pair. For the ICC, the degree of absolute agreement for three independent measurements under the fixed levels of the column factor (two-way mixed model, interaction absent) was calculated. The closer the value of CMC or ICC to 1, the more repeatable the device.

Sex Differences in Head Injury Metrics

The HIMs introduced in "Concussion Active Prevention Testing Device (CAPTD)" section were used as indicators of the risk of concussion. According to the literature, the cervical muscle of females was generally weaker than males.[9,40] Therefore, we hypothesized that our device must show sex differences in the measured HIMs. To show the sensitivity and effectiveness of the proposed device in capturing head kinematics for each (sensor position, weight) pair, the HIMs were compared between male and female participants. To this end, we used the Jarque–Bera test to verify the normality of the distribution of HIMs (significance level = 5%). Then, we evaluated the equality of the variance for two data sets with normal distributions using the Bartlett test. Finally, based on the tested normality and equality of variance, two-sample t test or Wilcoxon rank-sum test (significance level = 5%) was applied to detect significant differences between HIMs of male and female participants. Inter-participant repeatability of the measured HIMs among male and female participants was compared *via* a two-sample F-test.

RESULTS

Figure 2a shows the 3D acceleration and angular velocity of the head IMU before and after IMU local frame calibration. Frequency domain analysis of impacts with the heaviest weight, W3, showed that the dominant frequency contents of the acceleration and angular velocity signals lay at frequencies lower than 20 Hz, and therefore, the sampling frequency the IMUs, 100 Hz, was enough to capture head kinematics during impacts, Fig. 3. Also, it should be noted that a higher sampling frequency will be required for capturing the head kinematics during real impacts in contact sports compared to what was used in this study for controlled sub-concussive impacts.

Figure 2a shows that after calibration, the cross-talk between IMU axes was reduced such that the gravitational acceleration was sensed only through the vertical axis of the accelerometer during the motionless period in the beginning and that the angular velocity was sensed in the sagittal plane during the impact. Figure 2b shows representative waveforms of $A_{CG,x}$, $A_{CG,r}$, ω_s, and ω_r for three trials and that the head moved in the posterior direction (negative $A_{CG,x}$ and ω_s) immediately after impact due to neck and head inertia, and then moved in the anterior direction (positive $A_{CG,x}$ and ω_s).

According to Table 1, the lowest CMC for ω_s was 0.80 ± 0.07, while very good correlations were obtained for ω_r (minimum CMC of 0.88 ± 0.10 for chin IMU). For $A_{CG,x}$, very good correlations were obtained with the head IMU (minimum CMC of 0.90 ± 0.10), while moderate and good correlations were observed among trials using the chin IMU. For $A_{CG,r}$, the highest correlation for both head and chin IMUs were obtained for W3, 0.80 ± 0.18 and 0.83 ± 0.12, respectively. Finally, the intra-participant repeatability of the head and chin IMUs were similar. However, for the $A_{CG,x}$, the Wilcoxon signed-rank test showed significantly higher ($p < 0.05$) repeatability for head IMU compared to chin IMU for all testing conditions.

The color map in Table 2 shows that by increasing the weight, higher ICCs were obtained for HIMs for both head and chin IMUs in general. For the range of head CG acceleration and angular velocity, excellent ICC values (minimum ICC of 0.90) were obtained with both W2 and W3, while for W1, excellent ICC values (minimum ICC of 0.91) were calculated only for ω_s, ω_r, $A_{CG,y}$. For the peak of $A_{CG,x}$, $A_{CG,r}$, ω_s, and ω_r signals, excellent ICC values were obtained with both W2 and W3, while similar to the range, excellent ICC values were calculated only for ω_s, ω_r, $A_{CG,y}$. For GSI, good and excellent ICC values were obtained among the three trials for both $A_{CG,r}$ and ω_r, while for the HIC, the three trials showed moderate ICC values (minimum ICC of 0.73) for $A_{CG,r}$ signal. Finally, the ROM of the head in the sagittal plane was measured with good and excellent repeatability for all testing conditions.

According to Table 3 and Fig. 4, significantly lower ($p < 0.05$) range and peak values were recorded for $A_{CG,x}$, ω_s, and ω_r signals for male participants using the head IMU for all testing conditions. Similarly, both GSI and HIC showed significantly lower values for $A_{CG,r}$ and ω_r in the head IMU signals, except for W3, where males only tended to have lower HIC values compared to females. Also, Fig. 4k shows that male participants had significantly lower head ROM, measured by both IMUs, in comparison to female participants. Finally, the HIMs obtained from ω_r signals of female participants had significantly higher inter-participant variability compared to their male counterparts.

DISCUSSION

Given that head inertia remains constant in adult male athletes, muscular strength and activation time of the cervical muscles would be the only contributing factor in reducing the motion of the head, and thus the risk of concussion. Therefore, there has been an immediate need for concussion prevention programs such as neck strengthening, specifically in youth and female sports, to avoid the occurrence of concussion at an epidemic level. However, as shown by Naish *et al.*,[20] not all neck training programs were successful in increasing neck strength, and the effectiveness of each training program must be evaluated separately. Moreover, Wilcox *et al.*[39] found that concussion prevention strategies need to be sport- and gender-specific, with considerations for team and session type.

Therefore, the present study proposed, for the first time, a testing device to objectively measure the effect of neck strength on concussions caused by successive "body checks." For this purpose, the proposed testing device simulated impacts applied to the chest (not directly to the head) and simultaneously measured the resulted head motion to characterize the effect of neck muscle strength on the risk of concussion. The primary outcome of this research was the introduction of a testing device with the following advantages over similar counterparts: (1) having high sensitivity and repeatability in measuring various HIMs during sub-concussive impacts; (2) being affordable and accessible for sport facilities; (3) having a simple experimental procedure without the need for an experienced operator.

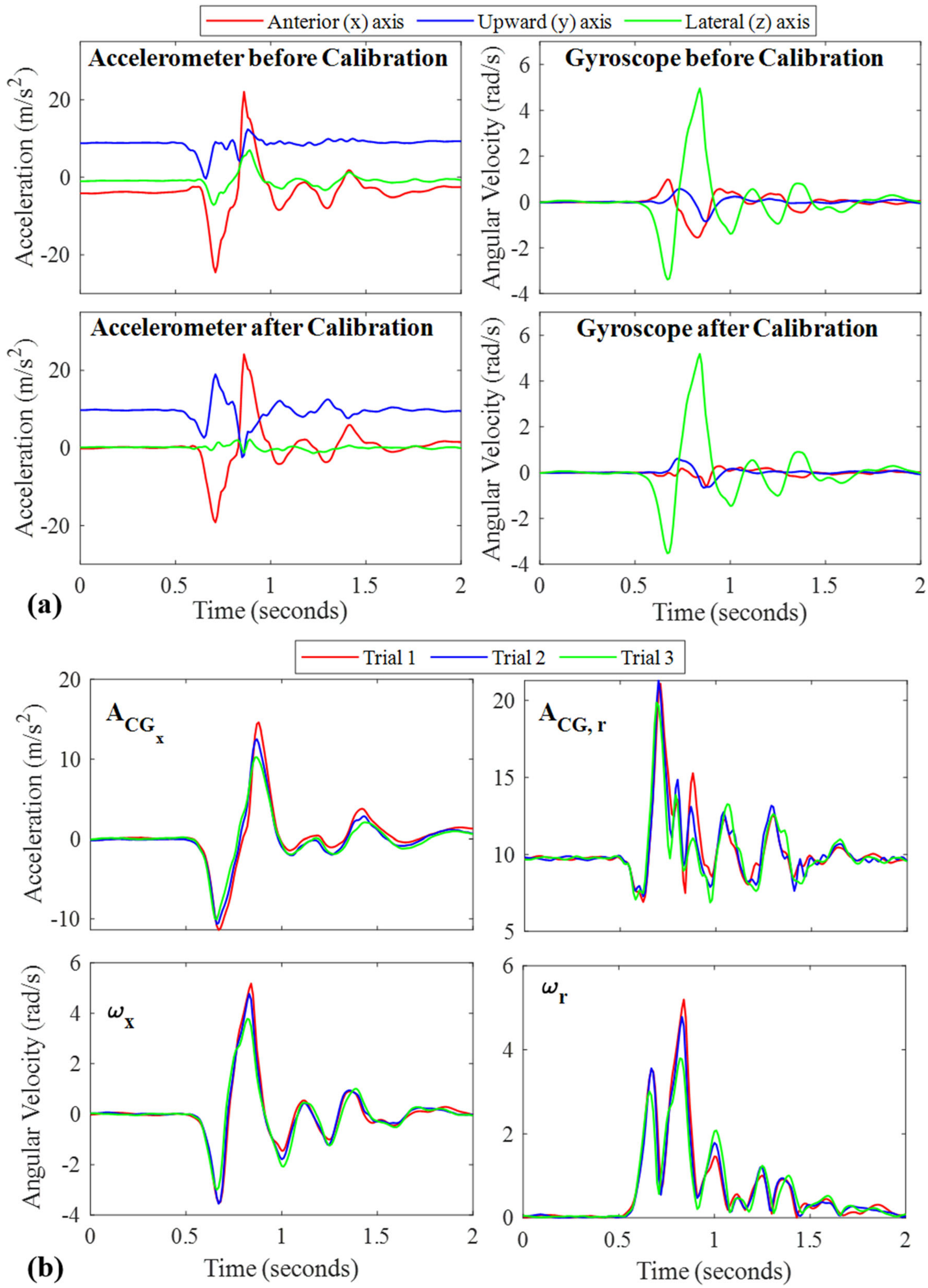

FIGURE 2. (a) 3D acceleration and angular velocity of the head IMU before and after IMU sensor frame calibration; (b) Representative waveforms of $A_{CG,x}$, $A_{CG,r}$, ω_s, and ω_r for one participant, obtained using the raw data of the head IMU for three trials with W3 = 52 kg.

Repeatability of the Testing Device

Manual measurement of neck strength may result in poor reliability due to procedural differences like considering the mean value versus the maximum value of the contraction force or different contraction times. To address this issue, the developed testing device simulated the same testing condition by automatic measurement of head kinematics during impacts using IMUs. The following parameters can be controlled in a repeatable manner.

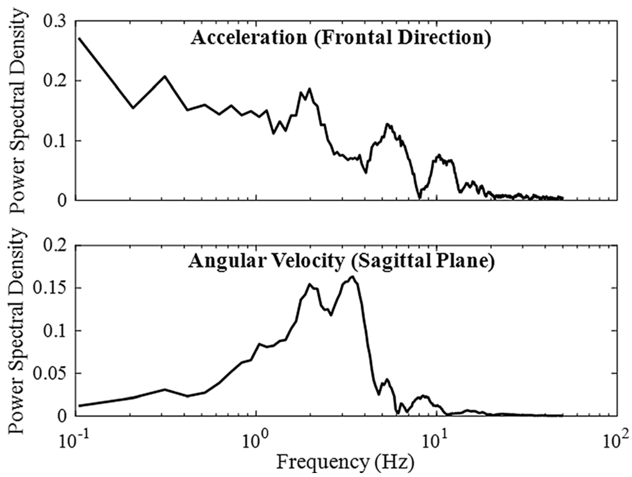

FIGURE 3. Power spectral density graph of acceleration and angular velocity of the head IMU in the anterior direction and sagittal plane, respectively, for a representative trial with W3 = 52 kg of one participant (the frequency is plotted on logarithmic scale).

1. Impact severity: by using similar weights, reducing the slack of the rope, and asking participants to lean on the backrest, the same impact severity can be simulated.
2. Impact direction: by having a constant height for the adjustable rod, the direction of the force can be kept constant.
3. IMU local frame: by applying the proposed IMU local frame calibration, the effect of IMU attachment and head initial inclination can be minimized.
4. Level of anticipation (cervical muscle onset time): by asking the participants to open/close their eyes or announcing the lever release, the level of anticipation can be controlled.

To assess the repeatability of the proposed testing device, CMC and ICC values were calculated for acceleration and angular velocity waveforms and HIMs, respectively, for the three trials associated with each (participant, weight) pair. Good and very good repeatability was obtained for ω_s and ω_r signals among the three trials (minimum CMC of 0.80 ± 0.07). While very good repeatability was obtained with head IMU for $A_{CG,x}$ for all weights, $A_{CG,y}$ and $A_{CG,r}$ signals showed moderate and good correlations. Therefore, $A_{CG,x}$, ω_s and ω_r time-series can be measured with high repeatability using the proposed testing device. Regarding HIMs, good and excellent repeatability was obtained for all HIMs and testing conditions, except for HIC obtained from $A_{CG,r}$.

TABLE 1. Mean values ± standard deviation (among all participants) of multiple correlation coefficient (CMC) obtained for three trials of each participant.

	W1 = 23 kg		W2 = 43 kg		W3 = 52 kg.	
	Head IMU	Chin IMU	Head IMU	Chin IMU	Head IMU	Chin IMU
ω_s	0.80±0.07	0.89±0.07	0.85±0.15	0.84±0.16	0.92±0.08	0.91±0.10
ω_r	0.90±0.07	0.89±0.07	0.88±0.09	0.88±0.10	0.93±0.07	0.92±0.08
$A_{CG,x}$	0.92±0.06†	0.75±0.13	0.90±0.10†	0.74±0.22	0.94±0.05†	0.80±0.19
$A_{CG,y}$	0.73±0.16	0.79±0.11†	0.71±0.23	0.68±0.32	0.80±0.17	0.78±0.22
$A_{CG,r}$	0.80±0.08	0.78±0.13	0.70±0.26	0.74±0.22	0.80±0.18	0.83±0.12

Moderate	Good	Very good	Excellent

Moderate (0.65 < CMC < 0.75), good (0.75 < CMC < 0.85), very good (0.85 < CMC < 0.95), and excellent (0.95 < CMC < 1) repeatability (defined based on[24]) are shown with color map.
Significantly higher ($p < 0.05$) CMCs among all participants obtained by the head or chin IMU for each (signal, weight) pair were shown with †
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TABLE 2. Intra-class Correlation Coefficient (ICC) and lower/upper band ICCs [LB, ICC, UP] obtained for head injury metrics including range, peak, GSI, HIC, and ROM between three trials of each participant.

	W1 = 23 kg		W2 = 43 kg		W3 = 52 kg	
	Head IMU	Chin IMU	Head IMU	Chin IMU	Head IMU	Chin IMU
Range of ω_s	[0.90,0.96,0.98]	[0.92,0.97,0.99]	[0.95,0.98,0.99]	[0.96,0.98,0.99]	[0.95,0.98,0.99]	[0.93,0.97,0.99]
Range of ω_r	[0.89,0.95,0.98]	[0.93,0.97,0.99]	[0.96,0.98,0.99]	[0.96,0.98,0.99]	[0.96,0.98,0.99]	[0.95,0.98,0.99]
Range of $A_{CG,x}$	[0.67,0.86,0.95]	[0.61,0.83,0.94]	[0.93,0.97,0.99]	[0.78,0.91,0.97]	[0.98,0.99,1.00]	[0.84,0.93,0.98]
Range of $A_{CG,y}$	[0.89,0.95,0.98]	[0.79,0.91,0.97]	[0.89,0.96,0.98]	[0.89,0.95,0.98]	[0.95,0.98,0.99]	[0.88,0.95,0.98]
Range of $A_{CG,r}$	[0.75,0.89,0.96]	[0.53,0.80,0.93]	[0.88,0.96,0.98]	[0.76,0.90,0.96]	[0.98,0.99,1.00]	[0.79,0.91,0.97]
Peak of ω_s	[0.89,0.95,0.98]	[0.93,0.97,0.99]	[0.96,0.98,0.99]	[0.96,0.98,0.99]	[0.96,0.98,0.99]	[0.95,0.98,0.99]
Peak of ω_r	[0.89,0.95,0.98]	[0.93,0.97,0.99]	[0.96,0.98,0.99]	[0.96,0.98,0.99]	[0.96,0.98,0.99]	[0.95,0.98,0.99]
Peak of $A_{CG,x}$	[0.68,0.86,0.95]	[0.54,0.80,0.93]	[0.92,0.97,0.99]	[0.77,0.90,0.96]	[0.97,0.99,1.00]	[0.85,0.94,0.98]
Peak of $A_{CG,y}$	[0.86,0.94,0.98]	[0.75,0.90,0.96]	[0.69,0.87,0.95]	[0.91,0.96,0.99]	[0.90,0.96,0.98]	[0.81,0.92,0.97]
Peak of $A_{CG,r}$	[0.63,0.84,0.94]	[0.58,0.82,0.93]	[0.87,0.95,0.98]	[0.78,0.91,0.97]	[0.97,0.99,1.00]	[0.83,0.93,0.97]
GSI of ω_r	[0.84,0.93,0.98]	[0.82,0.93,0.97]	[0.90,0.96,0.99]	[0.85,0.95,0.98]	[0.92,0.97,0.99]	[0.89,0.95,0.98]
GSI of $A_{CG,r}$	[0.74,0.89,0.96]	[0.58,0.83,0.94]	[0.85,0.94,0.98]	[0.76,0.90,0.96]	[0.92,0.97,0.99]	[0.62,0.85,0.95]
HIC of ω_r	[0.84,0.93,0.98]	[0.70,0.88,0.96]	[0.80,0.92,0.97]	[0.77,0.91,0.97]	[0.93,0.97,0.99]	[0.93,0.97,0.99]
HIC of $A_{CG,r}$	[0.55,0.81,0.93]	[0.35,0.73,0.90]	[0.59,0.83,0.94]	[0.38,0.74,0.91]	[0.35,0.73,0.90]	[0.43,0.76,0.92]
ROM (pitch)	[0.48,0.79,0.93]	[0.54,0.81,0.94]	[0.72,0.88,0.96]	[0.79,0.91,0.97]	[0.77,0.91,0.97]	[0.74,0.89,0.96]

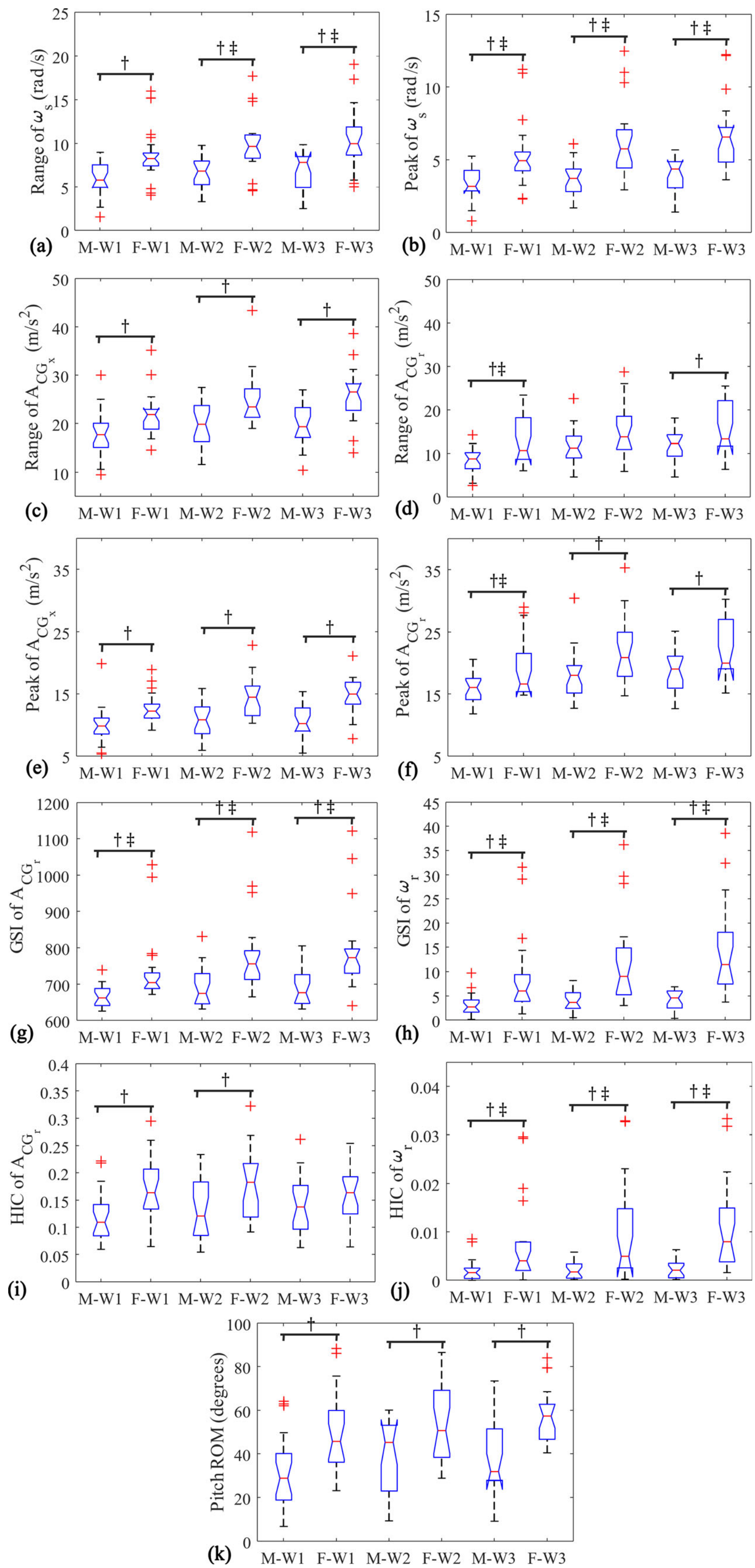
Poor	Moderate	Good	Excellent

Poor (ICC < 0.5), moderate (0.50 < ICC < 0.75), good (0.75 < ICC < 0.90), and excellent (0.90 < ICC < 1) repeatability (defined based on[17]) are shown with color map.

TABLE 3. Comparison of head injury metrics between male and female participants for each (sensor, weight) pair.

	W1 = 23 kg		W2 = 43 kg		W3 = 52 kg	
	Head IMU	Chin IMU	Head IMU	Chin IMU	Head IMU	Chin IMU
Range of ω_s	†	†	†‡	†	†‡	†
Range of ω_r	†‡	†‡	†‡	†‡	†‡	†‡
Range of $A_{CG,x}$	†	–	†	‡	†	‡
Range of $A_{CG,y}$	‡	–	‡	–	†‡	–
Range of $A_{CG,r}$	†‡	–	–	–	†	‡
Peak of ω_s	†‡	†‡	†‡	†‡	†‡	†‡
Peak of ω_r	†‡	†‡	†‡	†‡	†‡	†‡
Peak of $A_{CG,x}$	†	–	†	‡	†	‡
Peak of $A_{CG,y}$	‡	‡	–	–	–	–
Peak of $A_{CG,r}$	†‡	–	†	–	†	‡
GSI of ω_r	†‡	†‡	†‡	†‡	†‡	†‡
GSI of $A_{CG,r}$	†‡	–	†‡	–	†‡	–
HIC of ω_r	†‡	†‡	†‡	†‡	†‡	†‡
HIC of $A_{CG,r}$	†	‡	†	–	–	‡
ROM (pitch)	†	†	†	†	†	†

Significantly (significance level = 5%) stronger cervical muscle strength associated with significantly lower values of head injury metrics in male participants is indicated by †.
Significantly higher (significance level = 5%) inter-participant repeatability in measured head injury metrics in male participants in comparison to female participants is indicated by ‡.


(a)
Range of ω_s (rad/s)
(b)
Peak of ω_s (rad/s)
(c)
Range of A_{CG_x} (m/s^2)
(d)
Range of A_{CG_r} (m/s^2)
(e)
Peak of A_{CG_x} (m/s^2)
(f)
Peak of A_{CG_r} (m/s^2)
(g)
GSI of A_{CG_r}
(h)
GSI of ω_r
(i)
HIC of A_{CG_r}
(j)
HIC of ω_r
(k)
Pitch ROM (degrees)
M-W1
F-W1
M-W2
F-W2
M-W3
F-W3

◀**FIGURE 4. The head injury metrics (HIM: range, peak, GSI, HIC, and ROM) obtained by the head IMU shown as boxplot for male (M) and female (F) participants and for the weights of 23 kg (W1), 43 kg (W2), and 52 kg (W3). Similar to Table 2, significantly stronger muscle strength in male participants is indicated by † and significantly higher inter-participant repeatability in HIMs in male participants was indicated by ‡.**

Sex Differences in Head Injury Metrics

Previous research works reported lower neck strength and higher concussion rates in female athletes in comparison to their male counterparts.[9,19,40] Therefore, we assessed the ability of our testing device in differentiating between various levels of neck strength by comparing the HIMs between male and female participants. According to Table 3, head IMU was more successful than chin IMU in differentiating between male and female participants; the head IMU detected 38 significant differences out of 45 conditions (15 injury metrics × 3 weights) while the chin IMU only detected 21 significant differences. A possible explanation is that the estimated position vector between each IMU and the head CG had smaller errors for the head IMU. Therefore, we recommend placing the IMU on the top of the head for HIMs measurement with this testing device.

The measurement of the head IMU in Table 3 and Fig. 4 showed that several head kinematic parameters and HIMs were significantly greater in female participants than in male participants. In particular, female participants had significantly greater values in the following HIMs.

1. Range and peak values for the head angular velocity in the sagittal plane (ω_s), according to Figs. 4a and 4b;
2. Range and peak values for the head accelerations ($A_{CG,x}$ and $A_{CG,r}$), according to Figs. 4c to 4f;
3. GSI and HIC, according to Figs. 4g to 4j;
4. Head ROM in the sagittal plane, according to Fig. 4k.

For example, for W3, the [25th, 50th, 75th] percentiles of the peak ω_s among all male and female participants were [3.1, 4.4, 4.9] and [4.8, 6.6, 7.2] rad s^{-1}, respectively. Also, the [25th, 50th, 75th] percentiles of the peak $A_{CG,x}$ among all male and female participants were [9.1, 10.3, 12.8] and [13.4, 15.0, 16.9] m s^{-2}, respectively. Notably, since the non-parametric statistical analysis was employed for comparison, the [25th, 50th (median), 75th] percentiles of the data among male and female participants were reported here.

As such, HIMs during the controlled trunk impact conditions, in general, showed a significantly greater risk of concussion in female participants. In addition, higher inter-participant variability in the measured HIMs was observed for female participants. For instance, for W2, the interquartile range of the peak ω_s for male and female participants were 1.6 and 2.6 rad s^{-1}, respectively. This indicates a larger variation of the neck muscle activity strategies among female participants compared to their male counterparts that may recommend individual-specific training programs for female athletes. Further analysis is needed for such a recommendation.

Testing the Device Safety

The average linear acceleration associated with concussions, based on Hybrid III crash test dummies simulated using a video game, as reported by Pellman *et al.*,[26] was 98 ± 27 g. More recently, Rowson *et al.*[28] reported an average concussive linear acceleration of 105 ± 27 g collected from football players. Additionally, the latter study showed that in the most conservative case,[26] the probability of injury associated with peak head accelerations of 50 g was less than 15%. During our experiments, the average $A_{CG,r}$ values for male and female participants were recorded as 1.9 ± 0.3 g and 2.3 ± 0.5 g, respectively, which are nearly 50 times lower than the average concussive linear acceleration reported in the literature. Also, the peak $A_{CG,r}$ measured with our device was 3.1 g and 3.6 g which was nearly 25 times lower than the average concussive linear acceleration. Therefore, it can be concluded that the simulated sub-concussive impacts with the proposed device testing were safe and the risk of concussion during this experiment was nearly 1–2% based on the most conservative injury risk curve.[26] Nonetheless, all the injury risk curves present stochastic models and their precision cannot be guaranteed for all individuals.

Also, the measured peak $A_{CG,x}$ and $A_{CG,r}$ during our tests are comparable to those measured during various daily activities, such as standing, sitting, walking, running, and nodding the head, reported by Bussone *et al.*[3] (near or less than 10 and 20 m s^{-2} for $A_{CG,x}$ and $A_{CG,r}$, respectively). However, there are activities such as fast running, jumping jacks, and vertical leaping that showed higher peak linear acceleration. At the same time, by comparing the measured peak ω_s during our tests with the resultant angular velocity recorded in the same study, 10 out of 13 activities had a smaller peak of the resultant angular velocity (less than 2 rad s^{-1}), and the remaining 3 had similar values to our studies. Therefore, this device might not be recommended for very frequent testing or for those with history of back or neck pain, brain injury, or other neuromuscular impairments. Additionally, there might be a need for weight (W) adjustment based on body weight, body height, and physical condition, particularly for youth.

To further reduce the risk of concussion or even discomfort, the use of smaller testing weights should be recommended. Table 3 shows that conducting the experiments with $W1 = 23$ kg and $W3 = 52$ kg were both sensitive to neck strength differences between our male and female participants at the same level, i.e., in both cases 13 (out of 15) HIMs in Table 3 showed a significant difference between the male and female groups. At the same time, there was no significant difference between the CMC values of waveforms obtained by $W1$ and $W3$. Thus, $W1$ could be considered the testing weight to obtain responsive and repeatable results while reducing the risk of concussion or discomfort during the experiment. Nevertheless, the smallest testing weight that obtains responsive and repeatable results for each individual should be further investigated.

Limitations and Future Works

A number of factors would limit the generalization of the proposed testing device.

1. The repeatability and sensitivity of the proposed testing device were evaluated for fourteen participants only and must be further validated with more participants.
2. The ability of the testing device to differentiate between various levels of neck strength must be assessed by measuring head kinematics before and after a training program.
3. In the present work, weights were kept constant for all participants. However, by selecting weights as a percentage of the body weight, a more reliable comparison could be conducted between youth/adult or male/female.

Also, the application of the proposed testing device could be generalized in the future by modifying the frame structure to simulate impacts from other directions. Furthermore, other wearable sensors such as EMG could be added to the setup to measure other predictive measures of concussion, such as muscle activation response.

ACKNOWLEDGMENT

The authors would like to acknowledge the support of Gemini Performance Solution Inc. and Ms. Kim Adolphe and Mr. Sherwin Ganpatt, founders of CAP Corporation, for providing the testing device. The authors also thank the assistance of Mr. Sherwin Ganpatt in data collection. This research was financially supported by the Natural Sciences and Engineering Research Council of Canada (NSERC, Grant No: EGP 509196 - 17).

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 this manuscript.

REFERENCES

[1]Allison, P. D. Multiple Regression: A Primer. New York: SAGE Publications Inc, 1998.

[2]Bretzin, A. C., J. L. Mansell, R. T. Tierney, and J. K. Mcdevitt. Sex differences in anthropometrics and heading kinematics among division I soccer athletes: a pilot study. *Sport. Heal. A Multidiscip. Approach* 9:168–173, 2017.

[3]Bussone, W. R. Linear and Angular Head Accelerations in Daily Life, 2005.

[4]Cantu, R. C. Posttraumatic Retrograde and Anterograde Amnesia: pathophysiology and implications in grading and safe return to play. *J. Athl. Train*. 36:244–248, 2001.

[5]Caswell, S. V., M. York, J. P. Ambegaonkar, A. M. Caswell, and N. Cortes. Neck strengthening recommendations for concussion risk reduction in youth sport. *Int. J. Athl. Ther. Train*. 19:22–27, 2014.

[6]Collins, C. L., E. N. Fletcher, S. K. Fields, L. Kluchurosky, M. K. Rohrkemper, R. D. Comstock, and R. C. Cantu. Neck strength: a protective factor reducing risk for concussion in high school sports. *J. Prim. Prev*. 35:309–319, 2014.

[7]Corp., C. Apparatus and Method for Detecting a Subject's Susceptibility to Injury. 2017.

[8]Cournoyer, J., A. Post, P. Rousseau, and B. Hoshizaki. The ability of American Football Helmets to manage linear acceleration with repeated high-energy impacts. *J. Athl. Train*. 51:258–263, 2016.

[9]Dick, R. W., and H. Drive. Is there a gender difference in concussion incidence and outcomes? *Br. J. Sports Med*. 43:46–50, 2009.

[10]Eckner, J. T., Y. K. Oh, M. S. Joshi, J. K. Richardson, and J. A. Ashton-miller. Effect of neck muscle strength and anticipatory cervical muscle activation on the kinematic response of the head to impulsive loads. *Am. J. Sports Med*. 42:566–576, 2014.

[11]Emery, C. A., A. M. Black, A. Kolstad, G. Martinez, A. Nettel-Aguirre, L. Engebretsen, K. Johnston, J. Kissick, D. Maddocks, C. Tator, M. Aubry, J. Dvořák, S. Nagahiro, and K. Schneider. What strategies can be used to effectively reduce the risk of concussion in sport? A systematic review. *Br. J. Sports Med*. 51:978–984, 2017.

[12]Hall, T., M. P. Morissette, D. Cordingley, and J. Leiter. Reliability of a novel technique for the measurement of neck strength. *Int. J. Athl. Ther. Train*. 22:43–50, 2017.

[13]Hernandez, F., L. C. Wu, M. C. Yip, K. Laksari, A. R. Hoffman, J. R. Lopez, G. A. Grant, S. Kleiven, and D. B. Camarillo. Six degree-of-freedom measurements of human

mild traumatic brain injury. *Ann. Biomed. Eng.* 43:1918–1934, 2015.
[14]Hislop, M. D., K. A. Stokes, S. Williams, C. D. McKay, M. E. England, S. P. T. Kemp, and G. Trewartha. Reducing musculoskeletal injury and concussion risk in schoolboy rugby players with a pre-activity movement control exercise programme: a cluster randomised controlled trial. *Br. J. Sports Med.* 51:1140–1146, 2017.
[15]Hrysomallis, C. Neck muscular strength, training, performance and sport injury risk: a review. *Sport. Med.* 46:1111–1124, 2016.
[16]Langlois, 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.
[17]Mcgraw, K. O. Forming inferences about some intraclass correlation coefficients. *Psychol. Methods* 1:30–46, 2014.
[18]Mihalik, J. P., K. M. Guskiewicz, S. W. Marshall, R. M. Greenwald, J. T. Blackburn, and R. C. Cantu. Does cervical muscle strength in youth ice hockey players affect head impact biomechanics? *Clin. J. Sport Med.* 21:416–421, 2011.
[19]Mihalik, J. P., E. B. Wasserman, E. F. Teel, and S. W. Marshall. Head impact biomechanics differ between girls and boys youth ice hockey players. *Ann. Biomed. Eng.*, 2019.
[20]Naish, R., and A. Burnett. Can a specific neck strengthening program decrease cervical spine injuries in a men's professional Rugby Union Team? A retrospective analysis. *J. Sport. Sci. Med.* 12:542–550, 2013.
[21]Nazarahari, M., A. Noamani, N. Ahmadian, and H. Rouhani. Sensor-to-body calibration procedure for clinical motion analysis of lower limb using Magnetic and Inertial Measurement Units. *J. Biomech.* 85:224–229, 2019.
[22]Nazarahari, M., and H. Rouhani. Semi-automatic sensor-to-body calibration of inertial sensors on lower limb using gait recording. *IEEE Sens. J.* 2019. https://doi.org/10.1109/JSEN.2019.2939981.
[23]Noamani, A., M. Nazarahari, J. Lewicke, A. H. Vette, and H. Rouhani. Validity of using wearable inertial sensors for assessing the dynamics of standing balance. *Med. Eng. Phys.* 77:53–59, 2020.
[24]Palermo, E., S. Rossi, F. Marini, F. Patanè, and P. Cappa. Experimental evaluation of accuracy and repeatability of a novel body-to-sensor calibration procedure for inertial sensor-based gait analysis. *Meas. J. Int. Meas. Confed.* 52:145–155, 2014.
[25]Patel, B. H., K. R. Okoroha, T. R. Jildeh, Y. Lu, A. J. Idarraga, B. U. Nwachukwu, S. A. Shen, and B. Forsythe. Concussions in the National Basketball Association analysis of incidence, return to play, and performance from 1999 to 2018. *Orthop. J. Sport. Med.* 2019. https://doi.org/10.1177/2325967119854199.
[26]Pellman, E. J., D. C. Viano, A. M. Tucker, I. R. Casson, and J. F. Waeckerle. Concussion in professional football: reconstruction of game impacts and injuries. *Neurosurgery* 53:799–812, 2003.
[27]Rowson, S., J. G. Beckwith, J. J. Chu, D. S. Leonard, R. M. Greenwald, and S. M. Duma. A six degree of freedom head acceleration measurement device for use in football. *J. Appl. Biomech.* 27:8–14, 2011.
[28]Rowson, S., and S. M. Duma. Development of the STAR evaluation system for football helmets: Integrating player head impact exposure and risk of concussion. *Ann. Biomed. Eng.* 39:2130–2140, 2011.
[29]Rowson, S., and S. M. Duma. Brain injury prediction: assessing the combined probability of concussion using linear and rotational head acceleration. *Ann. Biomed. Eng.* 41:873–882, 2013.
[30]Rowson, S., S. M. Duma, J. G. Beckwith, J. J. Chu, R. M. Greenwald, J. J. Crisco, P. G. Brolinson, A. C. Duhaime, T. W. McAllister, and A. C. Maerlender. Rotational head kinematics in football impacts: an injury risk function for concussion. *Ann. Biomed. Eng.* 40:1–13, 2012.
[31]Salmon, D. M., M. F. Harrison, and J. P. Neary. Neck pain in military helicopter aircrew and the role of exercise therapy. *Aviat. Sp. Environ. Med.* 82:978–987, 2011.
[32]Schmidt, J. D., K. M. Guskiewicz, J. T. Blackburn, J. P. Mihalik, C. Cat, G. P. Siegmund, S. W. Marshall, C. Hill, and N. Carolina. The influence of cervical muscle characteristics on head impact biomechanics in football. *Am. J. Sports Med.* 42:2056–2066, 2014.
[33]Schneider, D. K., R. K. Grandhi, P. Bansal, G. E. Kuntz, K. E. Webster, K. Logan, K. D. Barber Foss, and G. D. Myer. Current state of concussion prevention strategies: a systematic review and meta-analysis of prospective, controlled studies. *Br. J. Sports Med.* 51:1473–1482, 2017.
[34]Sollmann, N., P. S. Echlin, V. Schultz, P. V. Viher, A. E. Lyall, Y. Tripodis, D. Kaufmann, E. Hartl, P. Kinzel, L. A. Forwell, A. M. Johnson, E. N. Skopelja, C. Lepage, S. Bouix, and O. Pasternak. Sex differences in white matter alterations following repetitive subconcussive head impacts in collegiate ice hockey players. *NeuroImage Clin* 17:642–649, 2018.
[35]Thurman, D. J., C. M. Branche, and J. E. Sniezek. The epidemiology of sports-related traumatic brain injuries in the United States: recent developments. *J. Head Trauma Rehabil.* 13:1–8, 1998.
[36]Tierney, R. T., M. Higgins, S. V. Caswell, J. Brady, K. McHardy, J. B. Driban, and K. Darvish. Sex differences in head acceleration during heading while wearing soccer headgear. *J. Athl. Train.* 43:578–584, 2008.
[37]Tierney, R. T., M. R. Sitler, C. B. Swanik, K. A. Swanik, M. Higgins, and J. Torg. Gender differences in head-neck segmnet dynamic stabilization during head accelaration. *Med. Sci. Sports Exerc.* 37:272–279, 2005.
[38]Viano, D. C., I. R. Casson, and E. J. Pellman. Concussion in professional football: Biomechanics of the struck player—part 14. *Neurosurgery* 61:313–327, 2007.
[39]Wilcox, B. J., J. G. Beckwith, R. M. Greenwald, J. J. Chu, T. W. Mcallister, L. A. Flashman, A. C. Maerlender, A. Duhaime, and J. J. Crisco. Head impact exposure in male and female collegiate ice hockey players. *J. Biomech.* 47:109–114, 2014.
[40]Wilcox, B. J., J. G. Beckwith, R. M. Greenwald, N. P. Raukar, J. J. Chu, T. W. Mcallister, L. A. Flashman, A. C. Maerlender, A. Duhaime, and J. J. Crisco. Biomechanics of head impacts associated with diagnosed concussion in female collegiate ice hockey players. *J. Biomech.* 48:2201–2204, 2015.
[41]Yoganandan, N., F. A. Pintar, J. Zhang, and J. L. Baisden. Physical properties of the human head: Mass, center of gravity and moment of inertia. *J. Biomech.* 42:1177–1192, 2009.

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