BMES BIOMEDICAL ENGINEERING SOCIETY

Concussions in Sports

Head Kinematics and Injury Metrics for Laboratory Hockey-Relevant Head Impact Experiments

Yanir Levy,[1] Kewei Bian,[2] Luke Patterson,[2] Ryan Ouckama,[3]
and Haojie Mao[1,2]

[1]School of Biomedical Engineering, Western University, 1151 Richmond St, London, ON N6A 3K7, Canada [2]Department of Mechanical and Materials Engineering, Faculty of Engineering, Western University, 1151 Richmond St, London, ON N6A 3K7, Canada [3]Bauer Hockey Ltd, 60 rue Jean-Paul Cayer, Blainville, Québec J7C 0N9, Canada

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

Abstract—Investigating head responses during hockey-related blunt impacts and hence understanding how to mitigate brain injury risk from such impacts still needs more exploration. This study used the recently developed hockey helmet testing methodology, known as the Hockey Summation of Tests for the Analysis of Risk (Hockey STAR), to collect 672 laboratory helmeted impacts. Brain strains were then calculated from the according 672 simulations using the detailed Global Human Body Models Consortium (GHBMC) finite element head model. Experimentally measured head kinematics and brain strains were used to calculate head/brain injury metrics including peak linear acceleration, peak rotational acceleration, peak rotational velocity, Gadd Severity Index (GSI), Head Injury Criteria (HIC$_{15}$), Generalized Acceleration Model for Brain Injury Threshold (GAMBIT), Brain Injury Criteria (BrIC), Universal Brain Injury Criterion (UBrIC), Diffuse Axonal Multi-Axis General Equation (DAMAGE), average maximum principal strain (MPS) and cumulative strain damage measure (CSDM). Correlation analysis of kinematics-based and strain-based metrics highlighted the importance of rotational velocity. Injury metrics that use rotational velocity correlated highly to average MPS and CSDM with UBrIC yielding the strongest correlation. In summary, a comprehensive analysis for kinematics-based and strain-based injury metrics was conducted through a hybrid experimental (672 impacts) and computational (672 simulations) approach. The results can provide references for adopting brain injury metrics when using the Hockey STAR approach and guide ice hockey helmet designs that help reduce brain injury risks.

Keywords—Brain, Biomechanics, Mild traumatic brain injury, Concussion, Helmet, Impact, Sports.

Address correspondence to Haojie Mao, School of Biomedical Engineering, Western University, 1151 Richmond St, London, ON N6A 3K7, Canada. Electronic mail: hmao8@uwo.ca

INTRODUCTION

The mild traumatic brain injury (mTBI) has become a critical health issue worldwide with over sixty-nine million individuals experiencing a TBI every year, of which mTBI make up a large portion of Ref. [12]. It has been estimated that there are roughly 100–300 mTBI patients per 100,000 population.[10] Typically, victims diagnosed with mTBI/concussions experience multiple symptoms, both in the short-term such as unconsciousness, headaches, cognitive impairments, dizziness, etc.) and in the long-term such as neurodegeneration, and in some cases, death.[5] Minimizing mTBI in sports is needed, especially in ice hockey, which is estimated to contribute 2 to 20% of mTBI cases.[7]

Head kinematics and kinematic-based metrics are used to predict mTBI risk. Head linear acceleration and rotational velocity were collected from hockey-related head impacts to investigate brain injury.[1,36] One of the earliest head injury tolerance curves was based on linear accelerations and durations,[21,28] which serves as the basis of several injury metrics, such as the Gadd Severity Index (GSI). The GSI is effective at quantifying severe skull fractures and severe brain injuries, but has limitations in its ability to predict the risk of mTBI or concussion.[18] Another commonly used prediction metric is the Head Injury Criterion (HIC), which is also limited to linear accelerations and durations, with the duration of 15 ms being commonly used.[11]

Brain injury prediction metrics were expanded with the inclusion of rotational motion to incorporate the

2914

deformation of the brain typically associated with common pathological injuries such as diffuse axonal injury (DAI). These new metrics were suited to provide assessments when a rotation-based injury mechanism was involved. The development of the Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) became one of the early criterions that used rotational kinematics in addition to linear kinematics.[34] The Brain Injury Criteria (BrIC) was also developed to examine the predicting potential of angular velocity.[40] Injury metrics continued to develop and encompassed more details including real world head impact data. An example of this is the Head Impact Telemetry Severity Profile (HITsp), a weighted composite score which takes into account linear acceleration, rotational acceleration, impact duration and impact location through the use of HIC and GSI.[20] Recently, the Universal Brain Injury Criterion (UBrIC)[16] and the Diffuse Axonal Multi-Axis General Equation (DAMAGE)[17] were introduced using rotational kinematics.[16,42,44] In addition to new brain injury metrics, the Summation of Tests for the Analysis of Risk (STAR) developed by Rowson et al. allowed for a novel helmet testing procedure that examined the rotational forces applied to the helmeted head during impacts.[38,39] This STAR testing methodology utilizes linear acceleration, rotational acceleration and head impact exposure based on extensive collegiate hockey player's impact data collected over several seasons.[38]

The development of computational head models allowed researchers to explore detailed human brain responses. Various intracranial parameters such as coup pressure, contre-coup pressure, von Mises stress, shear stress, and tensile strain could be assessed to predict the level of brain injury.[45] Several highly tested and validated, three-dimensional, finite element (FE) human head models have been developed with varying levels of anatomical features and complexities.[25,26,30,32,41,46,47] Meanwhile, there were efforts to use and compare various FE head models according to their strain predictions.[14] Moreover, it is accepted that strains in the brain have direct correlation to common traumatic brain injury pathologies such as DAI.[43] Hence, using a strain-based injury metric such as maximum principal strain (MPS) is one method of determining the outcome of a traumatic head impact scenario. The cumulative strain damage measure (CSDM) provides a volume-based correlation of the extent of damage that could be attributed to DAI, and this metric predicts DAI by calculating the MPS level at a volume fraction of the FE model.[40]

The correlations between various head kinematic metrics and brain responses have been extensively investigated. Compared to head linear acceleration and head rotational acceleration, head rotational velocity had the best correlation to the brain strain response.[6,27] Among 660 head impact experiments, the BrIC correlated closely to both the MPS and CSDM response with R^2 above 0.7, while the correlations between GAMBIT and MPS, as well as between GAMBIT and CSDM, were low (R^2 of 0.032 and 0.041 respectively).[15] The UBrIC demonstrated high correlation to MPS and CSDM among most of the impacts, with the R^2 values of 0.931 and 0.895, respectively.[16] DAMAGE was able to predict the MPS well, with R^2 above 0.9 for different impacts.[17]

For ice hockey, the safety standards are governed by three different organizations: The Hockey Equipment Certification Council (HECC),[33] The Canadian Standards Association (CSA),[37] and the International Organization for Standardization (ISO).[13] All three standards have similar pass/fail criteria, while a deeper understanding on how hockey helmets help reduce concussion risks is needed. The objective of this study is to use the new Hockey STAR approach, which is based on real-world hockey head impacts data and includes rotational kinematics, to collect a total of 672 laboratory frontal, side, rear and top impact experiments. Subsequently, laboratory experiments and FE simulations were combined to understand how effective the different brain injury prediction criteria are when using Hockey STAR. Seven kinematics-based injury metrics, including GSI, HIC, GAMBIT, BrIC, UBrIC, DAMAGE, and STAR Hockey mark were evaluated based on FE model predicted brain strains. Meanwhile, the effect of impact direction on brain strains was analyzed.

METHODS

Experimental Setup

A total of 672 testes were conducted on six Bauer helmet models based on the testing procedure of Hockey STAR,[38] including frontal, rear, side, and top impacts (Fig. 1(a)). The Hockey STAR equation (Equation 1) includes several unique metrics that pertain specifically to the sport of ice hockey. L represents the location of impact (rear, side, front or top) and θ represents different impact energy levels. These levels were determined in the original methodologies by the angle of the pendulum arm of the impactor. E represents exposure, the number of times a player is expected to receive an impact in a season. Finally, R is the risk of concussion as a function of linear (a) and angular (α) acceleration. Risk R values were calculated for various helmets and compared to brain strains.
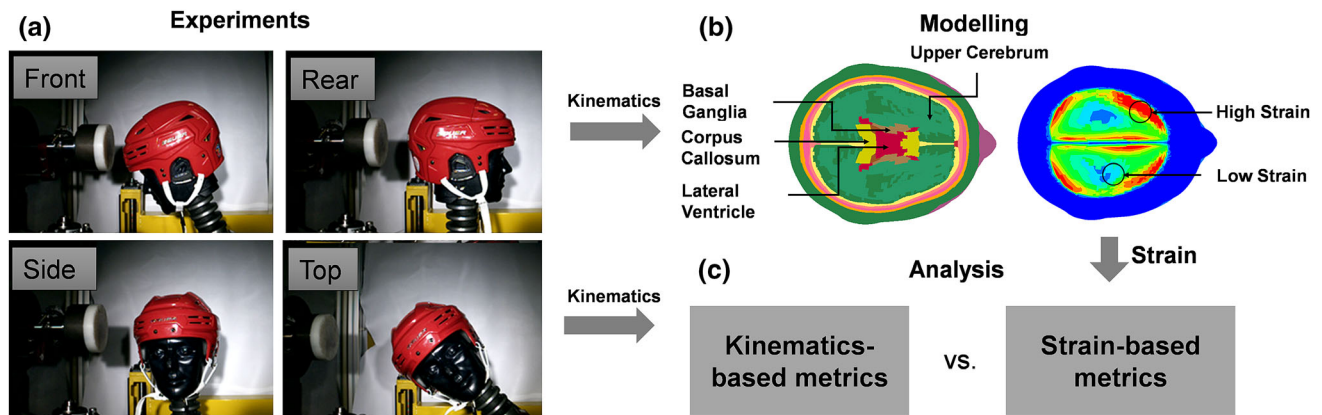
FIGURE 1. Combined experimental and computational approaches for analyzing kinematics-based and strain-based injury metrics. (a) STAR experiments, (b) Finite element brain modelling, (c) Correlation analysis. STAR: Summation of Tests for the Analysis of Risk.

$$Hockey\ STAR = \sum_{L=1}^{4} \sum_{\theta=1}^{3} E(L, \theta) * R(a, \alpha) \quad (1)$$

Rather than a pendulum as the STAR methodologies originally called for, a pneumatic impactor was used.[35] Like the original laboratory testing procedure, three impact energy levels with impact speeds of 3.0, 4.6 and 6.1 m/s respectively, and four impact locations (front, rear, side and top), were recreated to assess the viability of each helmet sample. Each helmet was hit twice with the 19.94 kg impactor per direction per impact speed per trial, with 4–5 helmet samples for each helmet model type. In this study six different helmet models were tested. In total, two helmet models went through 96 tests (4 samples per location) and four helmet models went through 120 tests (5 samples) tests, yielding a total of 672 impacts. The helmets were fitted onto a medium size NOCSAE headform mounted on a Hybrid III 50th percentile neck at the Hockey STAR biofidelic position with 3 Endevco 7264C-2KTZ-2-240 (Meggitt, Bournemouth airport, Dorset, United Kingdom) accelerometers for measuring linear acceleration, and three rotational velocity measured with the DTS6DX Pro (Diversified Technical Systems, Seal Beach, California, USA) mounted in the center of mass of the headform. The Endevco Model 136 amplifiers provided excitation voltage and signal conditioning. The kinematic data of each helmet impact, including linear acceleration, rotational acceleration, and rotational velocity, were collected at 20 kHz with a filter chain of hardware CFC1000 filter at amplifier for all channels, software CFC1000 filter on linear acceleration and software CFC155 filter on rotational velocity. A custom script was then developed to export the data into a spreadsheet including X, Y and Z axis data.

Computational Model

The FE model used in this study to simulate the physical testing impacts was the Global Human Body Model Consortium (GHBMC) head model,[30] which was driven by head linear and rotational accelerations collected from the impact experiments (Fig. 1(b)). This validated model of the human brain and skull is based on computed tomography (CT) and magnetic resonance imaging (MRI) scans of a healthy adult male brain of average height and weight. This model allows for a biofidelic computational model to simulate and interpret the mechanical stresses and strains associated with traumatic impact. The GHBMC model, as seen in Fig. 1, allows for the quantification and visualization of the mechanical soft-tissue materials metrics in key anatomical regions such as the corpus callosum, thalamus, cerebellum, brainstem and basal ganglia. In this model, a linear visco-elastic material was used in both the gray and white matter with the skull modelled as an elastic plastic material. In total, the GHBMC head and brain model contains 62 components of bone and soft-tissue, 61 unique material properties and 270,552 total elements (beam, shell and solid), and has been validated against intracranial pressure and brain displacement data.[23,24]

An initial dataset of an impact in three different impact energy levels in a single direction based on a single helmet sample was provided to determine an optimized time of impact to allow for both analysis of the moment of maximum principal strain and efficiencies regarding computational time and resources. The kinematic curves used in this study were determined through an initial testing round. The overall time of simulation (80 ms) was based on the peak strain responses of a test impact with the impact duration of 200 ms. The simulations were then completed on a Lenovo workstation (2 X Intel Xeon

TABLE 1. Kinematics-based injury metric equation summary.[11,16–18,34,38,40]

Injury metric	Equation
1. GSI	$GSI = \int a(t)2.5dt$
2. HIC	$HIC = 0ptmax_{t_1, t_2}\left\{(t_2 - t_1)\left[\frac{1}{t_2-t_1}\int_{t_1}^{t_2}a(t)dt\right]^{2.5}\right\}$
3. GAMBIT	$GAMBIT = \left[\left(\frac{a_{max}}{a_{cr}}\right)^2 + \left(\frac{\alpha_{max}}{\alpha_{cr}}\right)^2\right]^{\frac{1}{2}}$
4. BrIC	$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2}$
5. UBrIC	$UBrIC = \left\{\sum_i\left[\omega_i^* + (\alpha_i^* - \omega_i^*)e^{\frac{x_i^*}{\omega_i^*}}\right]^r\right\}^{\frac{1}{r}}$
6. DAMAGE	$DAMAGE = \beta max_t\left\{\vec{\delta}(t)\right\}$
7. STAR (Hockey)	$HockeySTAR = \sum_{L=1}^4 \sum_{\theta=1}^3 E(L, \theta) * R(a, \alpha)$
	$R_{a,\alpha} = \dfrac{1}{1 + e^{-(-10.2+0.0433*a+0.000873*\alpha-0.000000920*a\alpha)}}$

GOLD 5118 Processor, 12 cores @ 2.3GHz, 128 GB DDR4 Memory) using LS-DYNA (LSTC ANSYS LSTC, Livermore, CA, USA) with simulation time equivalent to about 2 h per simulation using 2 central processing units (CPUs), for a total computational time of about 1344 h.

Analysis Methods

An in-house script was used to calculate common head injury prediction criteria. Each of the 672 impact scenarios was assessed for peak kinematics in the X, Y and Z directions to determine resultant peak kinematics as well as injury prediction metrics. This information is summarized in Table 1. The kinematic data was automatically processed to calculate common injury metrics such as GSI, HIC_{15}, GAMBIT, BrIC, UBrIC and DAMAGE. The peak kinematics values were also used to calculate the Hockey STAR score.

Along with analyzing the preprocessed kinematics, the post-processed strains were also analyzed using an automated extraction method. The post-processing pipeline took the simulated GHBMC model and extracted the element data output file. This process acted as a batch script to utilize a custom in-house script[29] and extracted the MPS of each element and the total volume of the brain and calculated the CSDM with a strain cut-off threshold of 0.20 (CSDM20), referred to as the reported axonal damage strain threshold.[4] Each of the peak head impact kinematics along with their associated peak resultant kinematics were assessed for

their correlation to the brain response assessment metrics as described by CSDM and average MPS. Each prediction metric was then imported into a master spreadsheet where the data for each of the 672 impact scenarios were stored. Using IBM SPSS statistics 26 (IBM, Armonk, New York), both kinematics-based metrics and strain-based metrics data were analyzed for statistically significant correlations (Fig. 1(c)). The correlation analysis between STAR score and strain metrics was not conducted. Instead, the STAR score for each helmet was compared to calculated strain metrics of the same helmet.

RESULTS

Peak Kinematics and Strain

Overall, a large range of linear and rotational kinematics were recorded through the 672 experiments (Table 2). In brief, the average resultant linear acceleration was 121.0 g ($\pm$ 80.6 g), the average rotational velocity was 28.3 rad/s, and the average acceleration was 5814.8 rad/s^2.

When examining the output strain related to metrics (Fig. 2), the most correlated input kinematic parameter was resultant peak rotational velocity (RPRV), which correlated highly with CSDM with an average r of 0.92 ($p < 0.01$) and correlated highly with $MPS_{average}$ ($r = 0.96$, $p < 0.01$). Looking at resultant peak linear acceleration (RPLA), the correlation was decreased with an r equal to 0.60 ($p < 0.01$) for CSDM and an r value of 0.66 for $MPS_{average}$. Resultant Peak Rotational Acceleration (RPRA) was the least correlated to strain metrics of the three with r equal to 0.51 ($p < 0.01$) for averaged CSDM and equal to 0.58 for $MPS_{average}$ ($p < 0.01$). Overall, average MPS and CSDM20 provided similar trends when understanding RPLA, RPRV, and RPRA.

Comparison of Different Injury Metrics

The different injury metrics (GSI, HIC_{15}, GAMBIT, BrIC and UBrIC and DAMAGE) were assessed based on the strength of their relationship to strain-based CSDM (Fig. 3). The velocity-based injury prediction criteria, BrIC ($r = 0.91$, $p < 0.001$), UBrIC $_{MPS}$ ($r = 0.92$, $p < 0.001$), and UBrIC $_{CSDM}$ ($r = 0.92$, $p < 0.01$) performed better than the linear- and rotational-acceleration based metrics GSI ($r = 0.58$, $p < 0.001$), HIC_{15} ($r = 0.70$, $p < 0.001$), GAMBIT ($r = 0.56$, $p < 0.001$). BrIC and UBrIC outperformed DAMAGE ($r = 0.88$, $p < 0.001$).

TABLE 2. Head kinematics measured during experiments.

|	Kinematics	Mean	Std	Minimum	Maximum
Lin.	Linear Acc. X (g)	41.0	43.2	6.8	228.7
	Linear Acc. Y (g)	63.9	62.3	2.8	326.7
	Linear Acc. Z (g)	62.4	80.0	1.3	355.2
	Resultant (g)	121.0	80.6	31.9	417.1
Rot.	Rotational Vel. X (rad/s)	4.3	3.8	0.4	19.6
	Rotational Vel. Y (rad/s)	12.6	12.6	0.3	44.4
	Rotational Vel. Z (rad/s)	18.3	14.4	1.5	47.3
	Resultant (rad/s)	28.3	8.6	11.8	47.3
	Rotational Acc. X (rad/s^2)	1556.25	1647.9	301.0	8492.5
	Rotational Acc. X (rad/s^2)	3778.7	4323.2	273.9	18813.3
	Rotational Acc. X (rad/s^2)	2944.1	2005.5	516.2	10940.6
	Resultant (rad/s^2)	5814.8	3822.7	1635.8	19321.4
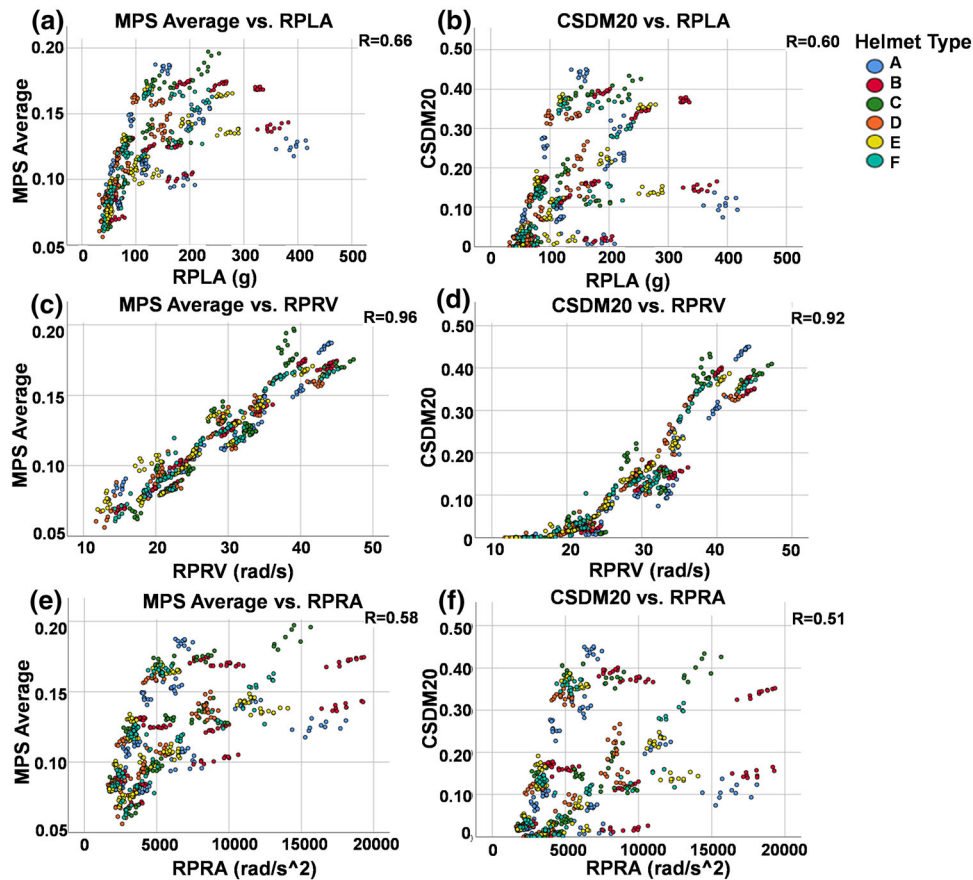
FIGURE 2. Comparison of different raw peak max resultant kinematic. The left column is compared to MPS average, whereas the right column is compared to CSDM 20, and from the top to bottom the RPLA, RPRV and RPRA comparisons are made. CSDM: cumulative strain damage; RPLA: Resultant Peak Rotational Velocity; RPLA: Resultant Peak Linear acceleration; RPRA: Resultant Peak Rotational Acceleration.

STAR VS Strain Results

The STAR score for each helmet was then compared to the strain metrics including MPS$_{average}$ and CSDM (Fig. 4). Average CSDM20 for 0-star helmets (B & C) equaled to 0.188 compared to 0.159 for a 3-star (bordered to 4-star) helmet (D), with a percentage difference of 18.2%. Average MPS for 0 star equal to 0.127 of strain compared to 0.120 for a 3-star helmet (borders to 4-star), with a percent difference of 5.8%. In general, higher-STAR-score helmets generated lower brain strains.
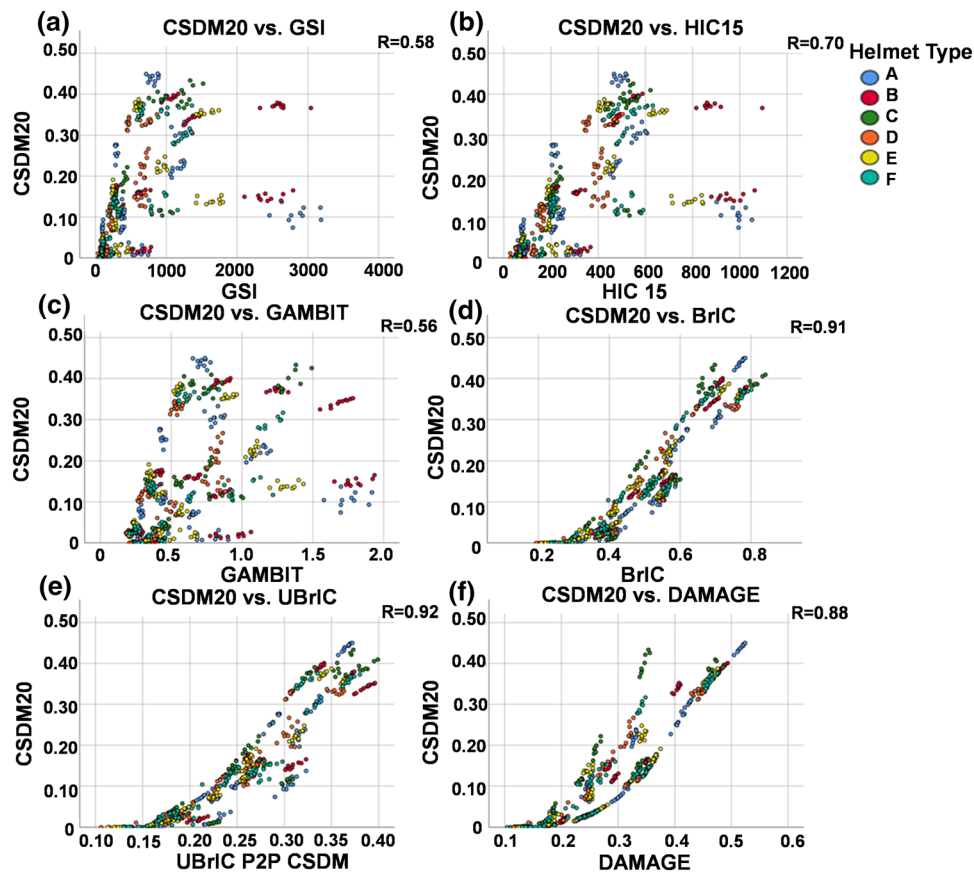
FIGURE 3. Comparison of different injury metrics to CSDM20 Colours indicate helmet models.GSI: Gadd Severity Index; HIC: Head Injury Criteria; GAMBIT: Generalized Acceleration Model for Brain Injury Threshold; BrIC: Brain Injury Criteria; UBrIC: Universal Brain Injury Criterion; DAMAGE: Diffuse Axonal Multi-Axis General Equation; MPS: Average Maximum Principal Strain; CSDM: Cumulative Strain Damage Measure.

Variances of Helmet Performance in Terms of Impact Direction

There were interesting differences in the performance of the different directions. Rear impact had the largest average MPS, predicted at 0.133 ± 0.031 ($n = 168$), followed by frontal impact, predicted at 0.122 ± 0.033 ($n = 168$), side impact, predicted at 0.118 ± 0.033 ($n = 168$) and finally top impact, predicted at 0.101 ± 0.026 ($n = 168$). Looking more in-depth into each direction, some helmets, such as helmet C, performed worse in side impacts, resulting in a higher strain (0.142 ± 0.038, $n = 24$) than front (0.127 ± 0.036 $n = 24$) or rear impacts (0.133 ± 0.034 $n = 24$) (Fig. 5). Helmet A performed better in side impacts, but performed worse in rear impacts (Fig. 5). For all helmets, top impacts resulted in the lowest strains (Fig. 5).

DISCUSSION

Rotational Velocity is an Effective Injury Criterion Parameter

We were able to show the advantages of rotational velocity-based injury prediction criteria such as BrIC and UBrIC. Even RPRV performed better than linear acceleration-based and rotational acceleration-based metrics such as HIC, GSI, GAMBIT, and DAMAGE in terms of predicting brain strains. Rotational velocity metrics have produced higher correlations compared to other metrics to brain strains in the literature and their inclusion into helmet assessment protocol is strongly recommended.[6] In this study, we validated the importance of rotational velocity when using the Hockey STAR approach. While our study has its limitations, including the absence of axial strain through the embedding of axonal fibers in the FE
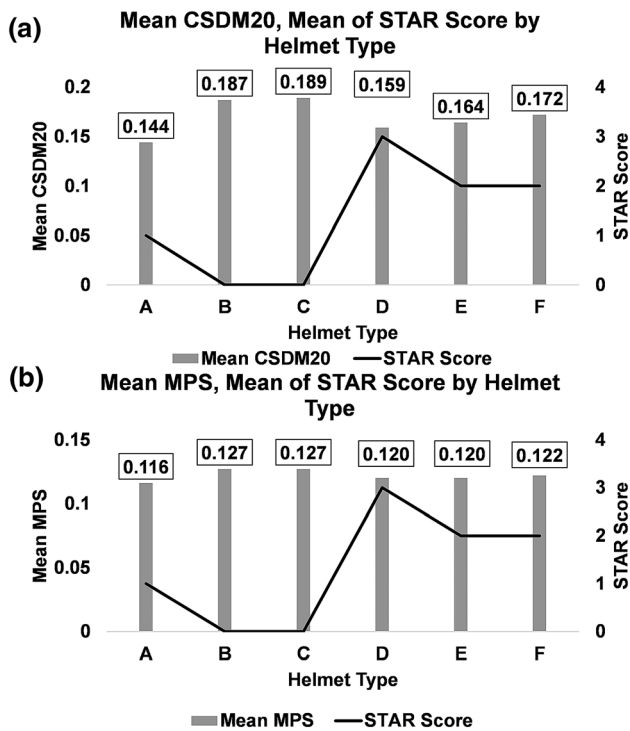
(a)

(b)

FIGURE 4. Put-together comparison of STAR score to the CSDM20 and average strain to understand two groups of results for the same helmet. Note that STAR evaluates injury risks based on real-world data while strain metrics evaluate brain strain loading simulating lab testing. STAR: Summation of Tests for the Analysis of Risk; CSDM: Cumulative Strain Damage Measure.
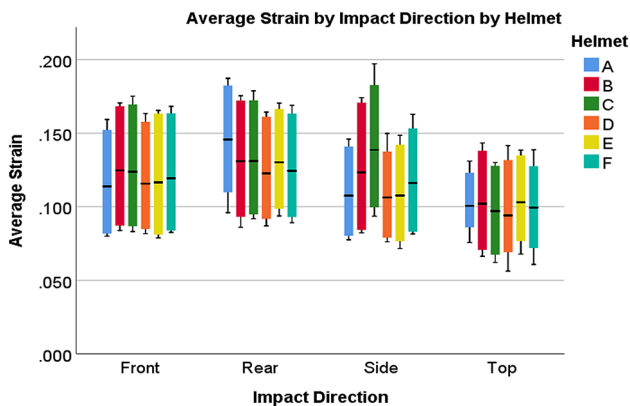

FIGURE 5. Boxplots representing the performance of each helmet, in terms of impact direction. In this example, each helmet preformed slightly differently in how they fared under each impact loading scenario. However, trends emerged such as strain being least affected by top impacts and rear impacts producing the most strain on average.

model, the study still used a widely acceptably GHBMC head model that provided a reduced computational cost for large data acquisition.

Exploring DAMAGE Injury Metrics

The recent introduction of the DAMAGE injury metric provided a new injury prediction criterion which was based on strain outputs from the GHBMC brain model and relied on linear and rotational accelerations as inputs. Our results indicated that it was important to look beyond the Pearson's correlation of DAMAGE which came with $r = 0.88$ and explore how well it correlated in terms of directional performance. An interesting finding was that DAMAGE correlated well with CSDM20 in the front and rear impact directions using the Hockey STAR testing setting ($r = 0.979$ and 0.983 respectively), but correlated less strongly in the side and top impacts ($r = 0.870$ and 0.855 respectively) (Fig. 6). One possible explanation could be the tangential kinematics associated with the side and top impacts, which, as stated in the Hockey STAR methodology, are impacted not through the center of gravity like the front and rear impacts. This could provide us with more data that rotational-velocity-based metrics such as BrIC and UBrIC were more suitable to predict strain-related injuries when using the Hockey STAR approach.

Helmet Performance Variance Among Different Impact Directions

This study also examined the directional differences of helmet impacts. Other studies have shown that impact directed at a specific direction resulted in larger brain deformation, even with similar impact energies.[6,9,16,27,35] While it is presumed that the direction that causes the highest brain strains is the axial rotational direction, in this particular experimental setting, this study showed that rear and frontal impacts or impacts that would affect the flexion/extension direction produced the highest strains consistently. The exception to this was helmet C, for which the side impact made up of axial rotation and lateral bending, generated the largest strain (Fig. 5). This is interesting, especially combined with our previous analysis showing that an axial rotation produced much higher brain strains compared to flexion/extension with the same magnitude of head rotational velocity.[6] A further analysis of rotational velocity demonstrated that, under same impact speeds, frontal and rear impacts produced larger head rotational velocities. For example, under medium-speed impacts, the average rotational velocities during frontal impact, rear impact, side impact and top impact were 32.1, 29.4, 26.2, and 20.6 rad/s, respectively.
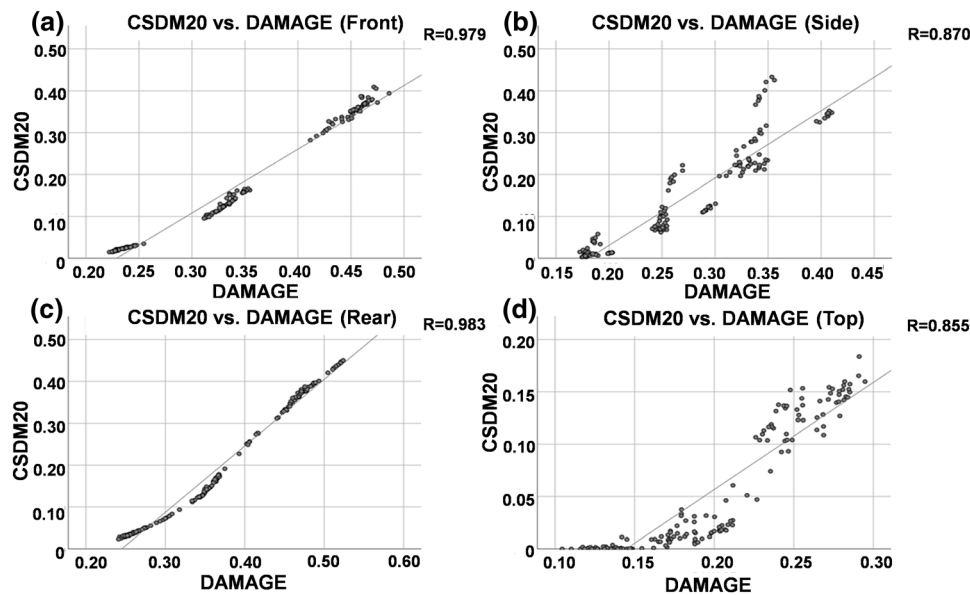
FIGURE 6. Comparison of DAMAGE correlation to CSDM20, Front, Side, Rear and Top impacts. DAMAGE is derived from strain outputs of the GHBMC model which is most likely the reason for it having good correlation to the strain results of this experiment. CSDM: Cumulative Strain Damage Measure; DAMAGE: Diffuse Axonal Multi-Axis General Equation.

Recommendation on Current Methodologies

The Hockey STAR helmet safety rating protocol provides a unique opportunity to evaluate hockey helmets based on concussion risks that are calculated through the statistical analysis of real-world head impact and injury data. For Hockey STAR, it's explained that a difference between a 12.8 star-rated-helmet and a 7.1 star-rated-helmet would equate to 44.6% less likelihood of sustaining a concussion.[38] Meanwhile, when investigating potential brain injury risks from a mechanistic approach, such as based on brain strains, similarly rated helmets such as helmet A (STAR = 8.7) and helmet B (STAR = 12.8) showed a percent difference in average strain by 9.8%. Hence, strain-based injury risk curves would be needed to explain how the 9.8% reduction of strain corresponds to the reduction of brain injury risks. Furthermore, tissue-level injury mechanisms beyond strain such as brain pressure also require attention. Still, given the strong correlation between rotational velocity and brain strain, the Hockey STAR could be further investigated in terms of rotational velocity. A recent STAR system developed for evaluating bicycle helmets adopted rotational velocity[8] and such efforts are highly recommended.

It needs to be emphasized that this study was focused on strain-based injury mechanism that is predominantly affected by head rotations, while other mechanisms such as brain pressure that is closely related to linear acceleration[22] also needs to be considered. Moreover, female youth hockey players experienced higher linear accelerations than male players.[31] Hence, real-world head impact data provides an invaluable opportunity to examine both linear and rotational kinematics and existing research supported the use of both kinematics to predict brain injury risks.[2,3] Further studies to collect both head kinematics and brain damage data on hockey are highly recommended. Brain responses, which directly caused neuronal tissue damage, could be correlated to brain injury risks based on these real-world hockey impacts. Meanwhile, different FE head models predicted various brain responses and the application of multiple head models could be explored.[14,19]

Conclusions

We collected 672 hockey helmeted impacts according to the Hockey STAR approach in lab and conducted 672 simulations using the GHBMC brain model. We analyzed peak kinematics and seven kinematics-based injury criteria including GSI, HIC, GAMBIT, BrIC, UBrIC, DAMAGE and STAR. Our results demonstrated that peak resultant rotational velocity correlated strongly to brain strain evaluated by both CSDM and MPS average, while peak resultant linear acceleration and peak resultant rotational acceleration did not. Accordingly, injury metrics that included rotational velocity, such as BrIC, UBrIC, and DAMAGE, correlated well to brain strain. Interestingly, DAMAGE performed well for front and rear impacts with the impactor mostly aimed at the center of gravity, but less strongly in side and top impacts

involving tangential head movements. Hockey STAR scores that are based on linear and rotational accelerations but not rotational velocities were found to be weakly related to brain strain, and hence the investigation of possibly including rotational velocity in the future is highly recommended. Our data also factors in direction, showing that Hockey STAR tests with top impacts induced the lowest strains. Together, our data provided a systematic understanding of the correlations between various head and brain injury metrics and brain strains when using the Hockey STAR approach.

ACKNOWLEDGMENTS

REFERENCES

[1]Allison, M. A., Y. S. Kang, M. R. Maltese, J. H. Bolte, and K. B. Arbogast. Measurement of Hybrid III head impact kinematics using an accelerometer and gyroscope system in ice hockey helmets. *Ann. Biomed. Eng.* 43:1896–1906, 2015.

[2]Bailey, A. M., T. L. McMurry, J. M. Cormier, J. R. Funk, J. R. Crandall, C. D. Mack, B. S. Myers, and K. B. Arbogast. Comparison of laboratory and on-field performance of American football helmets. *Ann Biomed Eng.* 48:2531–2541, 2020.

[3]Bailey, A. M., E. J. Sanchez, G. Park, L. F. Gabler, J. R. Funk, J. R. Crandall, M. Wonnacott, C. Withnall, B. S. Myers, and K. B. Arbogast. Development and evaluation of a test method for assessing the performance of American football helmets. *Ann. Biomed. Eng.* 48:2566–2579, 2020.

[4]Bain, A. C., and D. F. Meaney. Tissue-level thresholds for axonal damage in an experimental model of central nervous system white matter injury. *J. Biomech. Eng.* 122:615–622, 2000.

[5]Baugh, C. M., J. M. Stamm, D. O. Riley, B. E. Gavett, M. E. Shenton, A. Lin, C. J. Nowinski, R. C. Cantu, A. C. McKee, and R. A. Stern. Chronic traumatic encephalopathy: neurodegeneration following repetitive concussive and subconcussive brain trauma. *Brain Imaging Behav.* 6:244–254, 2012.

[6]Bian, K., and H. Mao. Mechanisms and variances of rotation-induced brain injury: a parametric investigation between head kinematics and brain strain. *Biomech. Model. Mechanobiol.* 19:2323–2341, 2020.

[7]Biasca, N., R. Agosti, Y. Tegner, H. Battaglia, and C. Gerber. Procedures after minor traumatic brain injury mTBI in ice hockey to prevent neurological sequelae. *Br. J. Sports Med.* 35:368–368, 2001.

[8]Bland, M. L., C. McNally, D. S. Zuby, B. C. Mueller, and S. Rowson. Development of the STAR evaluation system for assessing bicycle helmet protective performance. *Ann. Biomed. Eng.* 48:47–57, 2020.

[9]Camarillo, D. B., P. B. Shull, J. Mattson, R. Shultz, and D. Garza. An instrumented mouthguard for measuring linear and angular head impact kinematics in American football. *Ann. Biomed. Eng.* 41:1939–1949, 2013.

[10]Cassidy, J. D., L. Carroll, P. Peloso, J. Borg, H. Von Holst, L. Holm, J. Kraus, and V. Coronado. Incidence, risk factors and prevention of mild traumatic brain injury: results of the WHO Collaborating Centre Task Force on Mild Traumatic Brain Injury. *J. Rehabil. Med.* 36:28–60, 2004.

[11]Chou, C. C., and G. W. Nyquist. Analytical studies of the head injury criterion (HIC). *SAE Trans.* 83:398–410, 1974.

[12]Dewan, M. C., A. Rattani, S. Gupta, R. E. Baticulon, Y.-C. Hung, M. Punchak, A. Agrawal, A. O. Adeleye, M. G. Shrime, A. M. Rubiano, J. V. Rosenfeld, and K. B. Park. Estimating the global incidence of traumatic brain injury. *J. Neurosurg.* 130:1080–1097, 2019.

[13]Dixon, J. L. and I. K. Brodie. The New ISO Standards for Ice Hockey Helmets and Face Protectors: Moving Toward International Standards Harmonization and Conformity Assessment. In: *Safety in Ice Hockey: Second Volume.* ASTM International, 1993.

[14]Fahlstedt, M., F. Abayazid, M. B. Panzer, A. Trotta, W. Zhao, M. Ghajari, M. D. Gilchrist, S. Ji, S. Kleiven, X. Li, A. N. Annaidh, and P. Halldin. Ranking and rating bicycle helmet safety performance in oblique impacts using eight different brain injury models. *Ann. Biomed. Eng.* 49:1097–1109, 2021.

[15]Gabler, L. F., J. R. Crandall, and M. B. Panzer. Assessment of kinematic brain injury metrics for predicting strain responses in diverse automotive impact conditions. *Ann. Biomed. Eng.* 44:3705–3718, 2016.

[16]Gabler, L. F., J. R. Crandall, and M. B. Panzer. Development of a metric for predicting brain strain responses using head kinematics. *Ann. Biomed. Eng.* 46:972–985, 2018.

[17]Gabler, L. F., J. R. Crandall, and M. B. Panzer. Development of a second-order system for rapid estimation of maximum brain strain. *Ann. Biomed. Eng.* 47:1971–1981, 2019.

[18]Gadd, C. W. Use of a weighted-impulse criterion for estimating injury hazard. SAE Technical Paper 660793, 10th Stapp Car Crash Conference, 1966.

[19]Giudice, J. S., W. Zeng, T. Wu, A. Alshareef, D. F. Shedd, and M. B. Panzer. An analytical review of the numerical methods used for finite element modeling of traumatic brain injury. *Ann. Biomed. Eng.* 47:1855–1872, 2019.

[20]Greenwald, R. M., J. T. Gwin, J. J. Chu, and J. J. Crisco. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. *Neurosurgery.* 62:789–798, 2008.

[21]Gurdjian E. S., H. Lissner, and L. Patrick. Concussion: mechanism and pathology. In: *Proceedings: American Association for Automotive Medicine Annual Conference,* Association for the Advancement of Automotive Medicine, 1963, pp. 470–482.

[22]Gurdjian, E. S., H. R. Lissner, F. R. Latimer, B. F. Haddad, and J. E. Webster. Quantitative determination of acceleration and intracranial pressure in experimental head injury; preliminary report. *Neurology.* 3:417–423, 1953.

[23]Hardy, W. N., C. D. Foster, M. J. Mason, K. H. Yang, A. I. King, and S. Tashman. Investigation of head injury mechanisms using neutral density technology and high-speed biplanar X-ray. *Stapp Car Crash J.* 45:375, 2001.

[24]Hardy, W. N., M. J. Mason, C. D. Foster, C. S. Shah, J. M. Kopacz, K. H. Yang, A. I. King, J. Bishop, M. Bey, and W. Anderst. A study of the response of the human cadaver head to impact. *Stapp Car Crash J.* 51:17, 2007.

[25]Horgan, T. J., and M. D. Gilchrist. The creation of three-dimensional finite element models for simulating head impact biomechanics. *Int. J. Crashworthiness.* 8:353–366, 2003.

[26]Kleiven, S. Finite element modeling of the human head. Doctoral thesis, KTH, 2002.

[27]Knowles, B. M., and C. R. Dennison. Predicting cumulative and maximum brain strain measures from HybridIII head kinematics: a combined laboratory study and post-hoc regression analysis. *Ann. Biomed. Eng.* 45:2146–2158, 2017.

[28]Lissner, H., M. Lebow, and F. Evans. Experimental studies on the relation between acceleration and intracranial pressure changes in man. *Surg. Gynecol. Obstetrics.* 111:329, 1960.

[29]Mao, H., K. H. Yang, A. I. King, and K. Yang. Computational neurotrauma—design, simulation, and analysis of controlled cortical impact model. *Biomech. Model.* 9:763–772, 2010.

[30]Mao, H., L. Zhang, B. Jiang, V. V. Genthikatti, X. Jin, F. Zhu, R. Makwana, A. Gill, G. Jandir, and A. Singh. Development of a finite element human head model partially validated with thirty five experimental cases. *J. Biomech. Eng.* 135:111002, 2013.

[31]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.* 48:104–111, 2020.

[32]Miller, L. E., J. E. Urban, and J. D. Stitzel. Validation performance comparison for finite element models of the human brain. *Comput. Methods Biomech. Biomed. Eng.* 20:1273–1288, 2017.

[33]Morehouse C. A. The Certification of Protective Equipment for Ice Hockey in the United States. edited by A. B. Ashare. West Conshohocken, PA: ASTM International, 2000, pp. 72–77.

[34]Newman, J. A. A generalized acceleration model for brain injury threshold (GAMBIT). In: *Proceedings of International IRCOBI Conference,* 1986.

[35]Pellman, E. J., D. C. Viano, C. Withnall, N. Shewchenko, C. A. Bir, and P. D. Halstead. Concussion in professional football: helmet testing to assess impact performance—part 11. *Neurosurgery.* 58:78–96, 2006. (**discussion 78-96**).

[36]Post, A., A. Oeur, B. Hoshizaki, and M. D. Gilchrist. Examination of the relationship between peak linear and angular accelerations to brain deformation metrics in hockey helmet impacts. *Comput. Methods Biomech. Biomed. Eng.* 16:511–519, 2013.

[37]Rousseau, P., A. Post, and T. Hoshizaki. A comparison of peak linear and angular headform accelerations using ice hockey helmets. *J. ASTM Int.* 6:1–11, 2008.

[38]Rowson, B., S. Rowson, and S. M. Duma. Hockey STAR: a methodology for assessing the biomechanical performance of hockey helmets. *Ann. Biomed. Eng.* 43:2429–2443, 2015.

[39]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.

[40]Takhounts, E. G., M. J. Craig, K. Moorhouse, J. McFadden, and V. Hasija. Development of brain injury criteria (BrIC). *Stapp Car Crash J.* 57:243–266, 2013.

[41]Takhounts, E. G., R. H. Eppinger, J. Q. Campbell, R. E. Tannous, E. D. Power, and L. S. Shook. On the Development of the SIMon Finite Element Head Model. *Stapp Car Crash J.* 47:107–133, 2003.

[42]Takhounts, E. G., S. A. Ridella, V. Hasija, R. E. Tannous, J. Q. Campbell, D. Malone, K. Danelson, J. Stitzel, S. Rowson, and S. Duma. Investigation of traumatic brain injuries using the next generation of simulated injury monitor (SIMon) finite element head model. *Stapp Car Crash J.* 52:1–31, 2008.

[43]Wright, R. M., A. Post, B. Hoshizaki, and K. T. Ramesh. A multiscale computational approach to estimating axonal damage under inertial loading of the head. *J. Neurotrauma.* 30:102–118, 2013.

[44]Yang, J. Review of injury biomechanics in car-pedestrian collisions. *Int. J. Veh. Saf.* 1:100–117, 2005.

[45]Yao, J., J. Yang, and D. Otte. Investigation of head injuries by reconstructions of real-world vehicle-versus-adult-pedestrian accidents. *Saf. Sci.* 46:1103–1114, 2008.

[46]Zhang, L., K. H. Yang, R. Dwarampudi, K. Omori, T. Li, K. Chang, W. N. Hardy, T. B. Khalil, and A. I. King. Recent advances in brain injury research: a new human head model development and validation. *Stapp Car Crash J.* 45:369–394, 2001.

[47]Zhao, W., S. Ruan, and S. Ji. Brain pressure responses in translational head impact: a dimensional analysis and a further computational study. *Biomech. Model. Mechanobiol.* 14:753–766, 2015.