Using a Strain-Based Computational Approach for Ice Hockey Helmet Performance Evaluation

Yanir Levy, Marco B. Gallone, Kewei Bian, Kierra McDougall, Ryan Ouckama, Haojie Mao

Abstract This study looks to develop and explore a computational approach, along with data gathered from conventional mechanical helmet testing procedures in ice hockey, in an attempt to provide new insights into how the helmet could protect an individual from concussive type impacts. In this study, five samples of six different ice hockey helmet models were tested using the methodologies set forth by The Summation of Tests for the Analysis of Risk, the STAR helmet rating protocol. Head form kinematics collected during STAR testing were used as inputs to the Global Human Body Model Consortium head finite element model, and each impact (n=672) was simulated. A 15% cumulative strain damage measure threshold was chosen as the main response variable to predict brain injury probability. The results indicate that output kinematics of rotational velocity were most correlated ($r = 0.96$, $P < 0.05$) to cumulative strain damage measure and other strain measures. Impact direction also had significant effects on the strains in the brain, with impacts to the rear, front and side showing larger statistical significance in variance to the cumulative strain damage measure than top impacts. It was also observed that specific helmets showed less deformation response in certain impact directions compared to others. This study developed a start-to-finish methodology to evaluate helmets for mild brain injury mitigation.

Keywords Concussion mitigation, cumulative strain damage measure, injury prediction pipeline, kinematic performance evaluation, mild traumatic brain injury

I. INTRODUCTION

The traumatic brain injury (TBI) has become one of the most critical issues affecting global health systems with over 69 million individuals worldwide sustaining this injury every year [1]. An estimated 80% of these injuries are considered to be mild in nature, i.e. concussions, which poses a unique challenge to the researchers, physicians and medical trainers who are tasked with diagnosing, rehabilitating and mitigating their rate of occurrence [2].

In organised sports the issue of the mTBI is rampant, especially in adolescent aged participants [3]. The competitive environment which focuses on physical contact, especially in sports such as American Football and Ice Hockey, leads to increased instances of concussive and sub-concussive impacts that accumulate and could lead to negative short- and long-term neurodegenerative disorders [4,5]. In both sports the use of a helmet is the primary method of head impact mitigation. The original purpose of a helmet was to provide its wearer protection from mechanical loading that lead to lacerations, abrasions, fractures and other forms of tissue disruptions by absorbing the energy acting on the head upon impact [6]. Helmets however, need to be improved to cushion the brain and provide protective measures for the mitigation of concussive instances.

One common pathology of mild traumatic brain injury (mTBI) is the diffuse axonal injury (DAI), which is directly correlated to injury outcomes such as unconsciousness, cognitive impairments, and if the level of injury is severe enough, death [7]. The primary mechanical mechanism in DAI is inertial forces applied to the head following impact, that cause stretching of the deep and subcortical white matter. This *twisting* effect leads to extensive deformation of the brain structure and micro-tears to the underlying axon fibre bundles [8]. The issue with DAI, and moreover mTBI, is that it is extremely difficult to quantify the extent of the damage using traditional macroscopic pathology, typically used as assessment tools, post injury [7]. This along with the perceived *randomness* associated with concussions, where no two impacts are alike and where the ability to see the

Y. Levy is a MESc student in Biomedical Engineering specialising in brain biomechanics (ylevy3@uwo.ca, +1 416 854 6041). H. Mao is Professor of Engineering in the Department of Mechanical and Materials and Biomedical Engineering at Western University in Canada.

difference in brain structure using traditional diagnosis tools, such as computed tomography (CT) and magnetic resonance imaging (MRI) scans, is difficult, spearheading the inability to properly diagnose the patients who suffer from them.

The use of physical dummy testing models has become common practice in both academia and industry to create injury criteria based on kinematics to assess injuries quantitatively. In the sport of hockey, the standards for the level of protection in helmets is governed by three different organisations; The Hockey Equipment Certification Council (HECC), The Canadian Standards Association (CSA), and the International Organization for Standardization (ISO). All three standards have very similar pass/fail criteria mainly targeted towards the reduction of the probability of sustaining catastrophic head injuries. These current testing protocols are geared towards high energy linear impacts to the head and dummy and have considerable disregard for more mild or concussive like impacts. The current issue with these organisations and the helmet standardisation and testing, is that it currently does not take into consideration (1) the effects of rotational motion on the brain, and (2) the effects that more mild or sub-concussive impacts have on the brain structure and relationship to long-term neuro-degeneration. The obvious limitation of such methods is that (1) they do not allow researchers to recreate in-vivo head impact scenarios and (2) they are not able to provide adequate representation of the complex, non-linear and anisotropic behaviour of the soft-tissue in the brain [9].

Therefore, the need for the kinematic parameters of the helmeted head are required as they provide a direct correlation to the inertial response of the brain and hence could be an invaluable tool to predict the level of injury and provide instant insight into patient diagnosis. The introduction of the Summation of Tests for the Analysis of Risk (STAR) formula and safety testing methodology allows for a novel helmet testing procedures that looks to mitigate some of these inertial effects by examining the rotational forces applied to the helmeted head in low and medium energy level impacts [6,10]. This STAR testing methodology utilises the kinematic principles of linear acceleration, rotational acceleration and head impact exposure, a metric based on male and female collegiate player's impact location and severity over several seasons [6], to provide a resource for consumers to make educated decisions on purchasing helmets which are perceived as most likely to mitigate concussive risk. The STAR Helmet rating system, in theory, should provide a conclusive rating to assess the safety of a specific helmet, acting to keep helmet manufactures truthful and innovative with their research and development into new and innovative concussion mitigation technology, benefiting consumers.

The introduction of computational head and brain models has allowed researchers and engineers to evaluate brain tissue loadings that directly link to damage. With the use of finite element (FE) head models we are now able to recreate the complicated geometries and material structure of the human head. These FE models have allowed for reliable prediction of mechanical response and an accurate description of the constitutive behaviour of the nonlinear soft tissue response to loading, e.g., [11-16]][17]. With these computational head models available, developing a computational brain injury prediction pipeline for hockey helmets will help the field to better understand the effectiveness of protection and explore new designs that can better protect the brain.

This study looks to provide details of the development of an automated injury prediction pipeline for large kinematic datasets that will be used to provide new insights into how effective current methodologies such as that of the STAR are in assessing helmet performance and determining injury likelihood. One question that we look to solve is the validity of this methodology in assessing helmet protection and whether the use of linear and rotational acceleration are the best kinematic predictors for injury to the brain structure. This study looks to combine validated computational head models along with the use of validated physical surrogate models and assess the validity of the different testing methodologies and attempt to predict the level of injury mitigation that a hockey helmet helps provide when looking at common concussion-level impacts.

II. METHODS

Experimental Procedure

To re-create an industry standard method for physical helmet evaluation, this study based its helmet testing procedure on that of Hockey STAR. This methodology of assessing the biomechanical performance of hockey helmets differs from traditional methods provided by other standardisation organisations as it primarily looks to recreate some of the rotational kinematics associated with head impacts. The Hockey STAR equation, Equation 1, includes several unique metrics that pertain specifically to the sport of Ice Hockey. The L represents the location of impact (rear, side, front or top), the θ represents different impact energy levels, these levels were determined

in the original methodologies by the angle of the pendulum arm of the impactor. The 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. One of the purposes of this study was to examine whether the variable R is a sufficient and accurate assessor of the correlation between the kinematic outputs of a traumatic impact and the true level of injury response of the brain.

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

Rather than a pendulum as the STAR methodologies originally call for, a pneumatic impactor was used as it allows for more consistent impacts transferred to the head-form and less of a safety risk in testing [18]. Like the original laboratory testing procedure; three impact energy levels (low, medium and high) with impact speeds of 2.6 m/s, 4.6 m/s and 6.0 m/s respectively, and four impact locations (front, rear, side and top), were recreated to assess the viability of each helmet sample, see Figure 1. While the front and rear impacts were directed at the centre of gravity of the NOCSAE head form, the top and side impacts were not directed at the COG of the head form and hence added an element of tangential loading. It needs to be highlighted the top impact (Figure 1D) was not a conventional impact delivered from the vertical side, but more an impact with an elevation. Each helmet was hit twice with the impactor (19.94 kg) per direction per impact speed per trial, with four to five helmet samples for each helmet model type. In this study six different helmet models were tested. In total each helmet went through an average of 112 impacts for a total of 672 impacts with corresponding kinematics. Helmet tests were analysed for repeatability by assessing the standard deviations of individual kinematic metrics of repeating trials.

The helmets were fitted onto a medium size National Operating Committee on Standards for Athletic Equipment (NOCSAE) head-form mounted on a Hybrid III 50th percentile neck with three Endevco 7264C-2KTZ-2-240 (Meggitt, Bournemouth airport, Dorset, United Kingdoms) accelerometers for linear acceleration, and three rotational velocity channels of the DTS6DX Pro (Diversified Technical Systems, Seal Beach, California, USA) mounted in the centre of mass of the head form. Two Endevco Model 136 amplifiers provided excitation voltage and signal conditioning. The kinematic data of each helmet impact; 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.

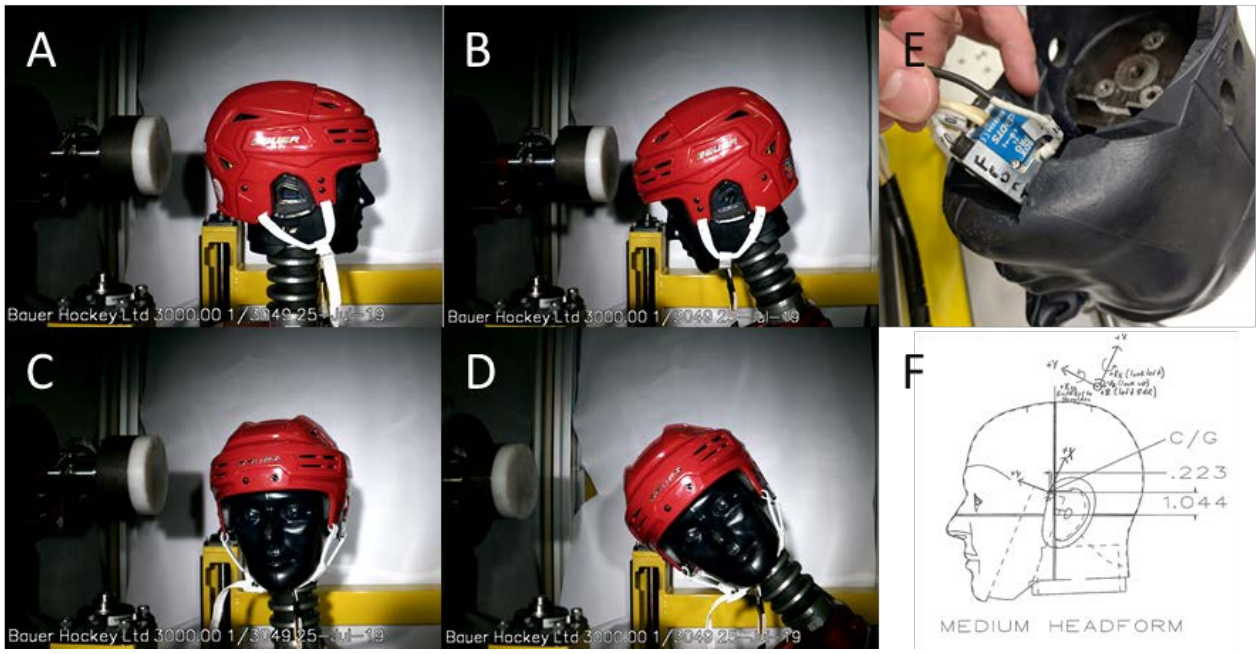

Fig. 1. The experimental setup procedure and helmet impact locations modelling that of the Virginia Tech STAR Methodology (A) Rear Impact (B) Front Impact (C) Side Impact (D) Top Impact (E) placement of Endevco accelerometer in the centre of gravity of the NOCSAE head form with Hybrid III neck and (F) the schematic of the placement of the accelerometer measured in inches for accurate recreation of kinematics in the computational model.

Computational Modelling

The finite element model used in this study to simulate the physical testing impacts was the Global Human Body Model Consortium (GHBMC) head model [11]. This validated model of the human brain and skull is based on (CT) and (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 Figure 2, allows for the quantification and visualisation of the mechanical soft-tissue material 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 grey and white matter with the skull modelled as a piecewise-linear-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 is validated against intracranial pressure and brain displacement data [19,20].

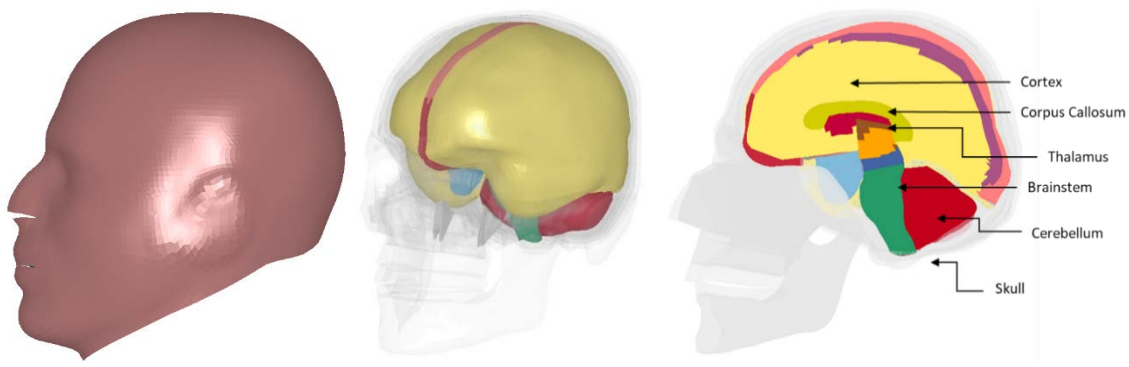

Fig. 2. Breakdown of GHBMC head and brain finite element model. From top to right, full GHBMC model with skin, isometric view of model with skull and skin transparent to view placement of brain and sagittal view of model showing the placement of different anatomical components of the brain.

When setting up the model, the direction of the kinematics was reoriented to a 23-degree offset above the horizontal Y- axis to mimic the sensor setup in the original dummy head form. 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 optimised time of impact to allow for both analysis of the moment of maximum principal strain as well as allowing for 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 (80ms) (Figure 3, left) was used based on the peak strain responses of a test impact (t = 200ms) where peak max principal strain (MPS) (Figure 3, right) was included along with subsequent inertial response. The simulations were then completed on a Lenovo workstation (2 X Intel Xeon GOLD 5118 Processor (12 cores @ 2.3GHz), 128 GB DDR4 Memory) using LS-DYNA, finite element programme, (Livermore Software Technology ANSYS LSTC, Livermore, CA, USA) with simulation time equivalent to ~2 hours per simulation at NCPU = 2, for a total computational time of ~1344 hours. Each simulation was then analysed in LS-PrePost and checked over for any logical errors.

Pipeline Logic

An in-house MATLAB script was created to orient the GHBMC head model, apply the kinematic data from the Excel file as a time history loading curve and save as the original file name into a new keyword file for the increased efficiency of setup for all 672 simulations automatically. This script also applied a rotation matrix to the original kinematics to orient peak kinematics to the GHBMC computational model's strain output and calculated the resultant linear accelerations, rotational accelerations and rotational velocities of each impact scenario. Each completed simulation was then passed through another in-house post-processing pipeline, which analysed the maximum principal strains (MPS) of all elements throughout the time history plot and, using a customised script, determined the cumulative strain damage measure (CSDM) of each impact at a pre-set strain level. CSDM which is suggested as a premier predictor of brain injury response was used as a predictor of brain responses induced by different impacts and provides the diffuse pattern of the total damage that could occur to the brain leading to a damage [21]. Based on the MPS of each element the algorithm can determine the percentage of the elements in the GHBMC model that are above a threshold specified by a user. This study examined five CSDM scenarios

ranging from mild to severe in terms of predicted brain damage (CSDM5, CSDM10, CSDM15, CSDM20 and CSDM25). Where a value of CSDM5 = 0.50 would mean that 50% volume of the elements in the GHBMC head model would experience more than 5% strain.

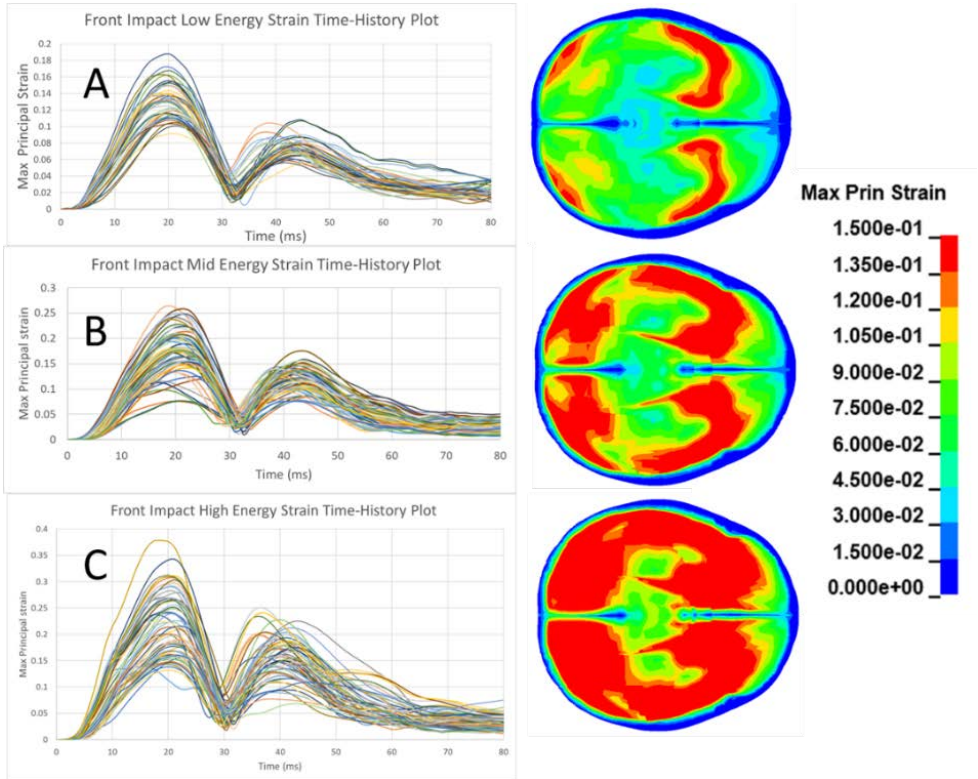

Fig. 3. Representation of the typical strain patterns in a frontal impact on helmet E at 15% max principal strain fringe level in the GHBMC model in the transverse cross sectional view. Representation of typical time-history strain patterns of (A) low 2.6m/s, (B) Mid 4.6 m/s and (C) High 6.0m/s energy impact levels. Initially 200ms plots were used as justification for 80ms simulation time on all simulations in an attempt to encompass the peak of the maximum strain to the brain elements while reducing computational time and cost.

Pre-processing Pipeline Explained

This script converted Excel kinematic data into the linear acceleration and rotational velocity time-history curves used as a boundary condition of a prescribed motion in LS-DYNA. Each kinematic impact scenario output manipulated a baseline GHBMC keyword file with new time history curves in the X, Y and Z directions for a total of six degrees of freedom, all of this being done automatically with the process pipeline which is described with associated functions below. This pipeline allows for a computational approach to convert easily reproducible XYZ data into a keyword file which is fully ready for input into the LS-DYNA solver impact simulations and dummy head form impact recreation.

Post-processing Pipeline Explained

The post processing pipeline looks to take the simulated GHBMC model and extracts the element data output (ELOUT) file. This process acts as a batch script to utilise a custom in-house script [22] and extract the MPS of each element and the total volume of the brain and calculate the CSDM of the brain at varying levels to provide a glimpse into the perceived level of sub-concussive and concussive injury likelihood. All files in a folder can be examined with a single click, and hence 672 individual CSDMS were examined in this experiment. A simplified schematic is provided in Figure 4. With a breakdown of the pipelines coding logic explained in Appendix A.

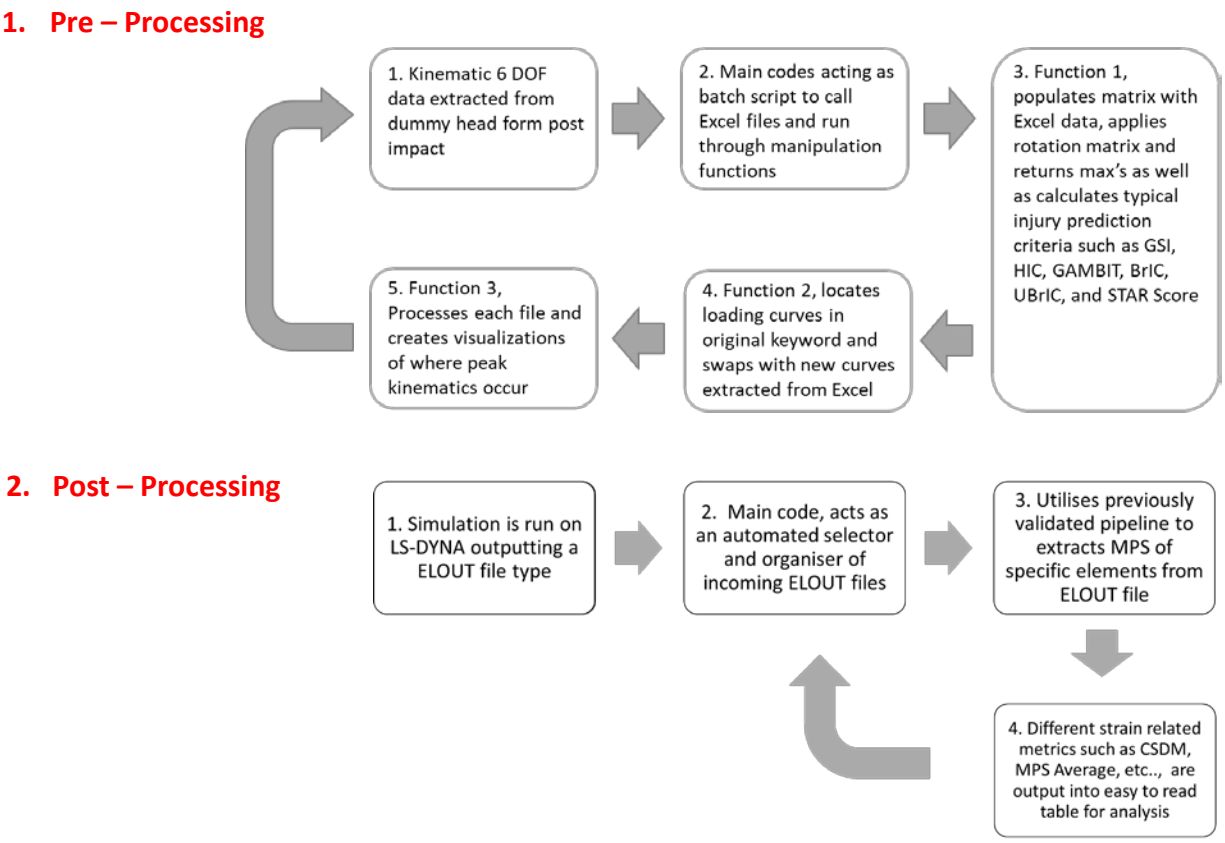

Fig. 4. A schematic of the pre and post processing pipeline in simplified terms. A full breakdown of the logic behind the kinematic injury prediction pipeline is available in the supplementary material.

Analysis

Statistical analysis tools were used to analyse the correlation between the CSDM values and the peak kinematics, along with comparing the relative safety of each helmet in terms of mitigating brain injury and reducing corresponding inertial factors. All values were analysed using IBM SPSS Statistics 26 (IBM, Armonk, New York).

III. RESULTS

Repeatability tests show that overall differences for the same impact setting were small (Table 1). For example, for the helmet A, samples broken down into its first and second trials, showing similarities in the resultant peak linear acceleration (RPLA), resultant peak rotational velocity (RPRV) and resultant peak rotational acceleration (RPRA) of each direction and at each energy level.

Table 1. Example impact showing test repeatability under same experimental conditions.

			Trail 1			Trial 2		
Helmet	Impact Energy	Direction	RPLA	RPRV	RPRA	RPLA	RPRV	RPRA
A	Low	Front	54.15	22.04	2084.66	50.26	22.37	2140.46
	Mid	Front	86.41	32.00	3604.53	85.26	32.18	3526.64
	High	Front	166.64	43.91	4407.81	181.08	43.95	4968.79
	Low	Rear	57.84	21.55	2556.78	52.99	22.07	2331.47
	Mid	Rear	77.92	29.28	3527.65	76.56	29.60	3433.27
	High	Rear	117.59	37.62	5257.20	124.24	39.12	5148.83
	Low	Side	54.71	18.59	3502.87	53.29	18.53	3612.59
	Mid	Side	107.26	26.22	6550.22	110.20	26.23	6648.86
	High	Side	206.54	34.99	11522.78	232.42	36.25	12756.79
	Low	Top	41.48	15.32	2886.14	41.99	16.51	3142.34

	Mid	Top	72.44	18.46	4827.23	79.37	22.78	5067.64
	High	Top	166.12	30.99	9589.87	197.37	33.29	11748.32

The range of linear accelerations, rotational velocities, and rotational accelerations is described in Table 2 for all 672 impact scenarios. On average, RPLA reaches 121 g's and RPRV reaches 28 rad/s.

Table 2. Breakdown of the linear accelerations, rotational velocities, and rotational accelerations.

Kinematic	Minimum	Maximum	Mean	Std. Deviation
Linear Acc. X (g)	6.81	228.65	41.01	43.23
Linear Acc. Y (g)	2.83	326.69	63.87	62.31
Linear Acc. Z (g)	1.29	355.17	62.38	79.98
RPLA (g)	31.85	417.05	121.01	80.56
Rotational Vel. X (rad/s)	0.37	19.56	4.29	3.77
Rotational Vel. Y (rad/s)	0.27	44.43	12.58	12.62
Rotational Vel. Z (rad/s)	1.51	47.28	18.31	14.43
RPRV (rad/s)	11.75	47.31	28.29	8.57
Rotational Acc. X (rad/s/s)	301.03	8492.54	1556.25	1647.92
Rotational Acc. X (rad/s/s)	273.90	18813.32	3778.66	4323.23
Rotational Acc. X (rad/s/s)	516.16	10940.55	2944.12	2005.51
RPRA (rad/s/s)	1635.78	19321.35	5814.78	3822.74

The Pipeline to Connect Head Kinematics and Brain Strains

A completed pipeline was developed and tested for all 672 impact scenarios. This pipeline reduced overall pre-processing time from approximately 20 minutes of manual keyword manipulation to approximately two minutes per scenario of automated computational manipulation. This pre-processing pipeline allowed for all 672 impacts XYZ kinematic output data to be converted into keyword files for simulation by LS-DYNA solver in approximately 22 hours of computational time compared to over 220 hours, or 10X less total time. Following the simulation of all 672 impacts, the post processing pipeline was engaged. This pipeline determined the CSDM of each simulation and organised all simulations into an Excel spreadsheet, in approximately five minutes per simulation or a total time of approximately 56 hours. The original manual extraction and manipulation of the post processed data into CSDM data was approximately 10 minutes per scenario or a total time of 112 hours. This represents a 100 percent increase in total computational time and the need for user intervention. The final results were organised into an easy to read spreadsheet that allows for data analysis.

The Correlation between Head Kinematics and Brain Strains

Assessing all 672 scenarios the peak kinematics with averages were compiled from the initial dataset and the CSDM values were all computed and related to each impact test scenario. As seen in Figure 5 all peak impact kinematics were compared with CSDM15, which was determined as a valid assessment of DAI as a threshold for the maximum strain an axon could withstand before exhibiting signs of tearing or deformation [23]. The RPLA ($R = 0.61$ $P < 0.01$) and RPRA ($R = 0.51$ $P < 0.01$) were significantly less correlated to CSDM15 than RPRV ($R = 0.96$ $P < 0.01$). This analysis was done using SPSS, and the bivariate correlation coefficient was Pearson with the test of significance being two-tailed (Figure 5). Of note, CSDM metrics correlated heavily with other similar strain based metrics such as average MPS and MPS top 1 percent and 5 percent thresholds ($R=0.99$ $P < 0.01$, $R=0.98$ $P< 0.01$ and $R=0.99$ $P<0.01$). Other widely used kinematics based injury criteria, primarily the head injury criteria (HIC_{15}) and the Brain Injury Criteria (BrIC) were also included in the analysis to better understand their correlation to the different strain metrics, with BrIC showing strong correlation to average strain ($R=0.897$ $P<0.01$) (Figure 5).

Impact Direction and Helmet Strain Effect on Brain Response

Impact direction and the effects of the helmets on reducing brain strains by way of CSDM was also analysed. Results show that impact direction has effect on relating to relative CSDM value, with rear impacts showing the largest mean CSDM values across all CSDM levels and all impact energy levels. Rear impacts were followed by Front then Side and finally by Top impacts. It is also noted that helmets exhibited varying levels of success in different directions with some helmets mitigating brain strain response in one direction more effectively than

other helmets and in other directions less effectively than other helmets. A representation of these results is shown in Figure 6 below. For impact direction relationship to CSDM15 there was a statistically significant difference between groups as determined by One-way ANOVA ($F(3, 668) = 39.846$ $p < 0.01$). A Tukey post hoc test revealed that CSDM15 was statistically significantly higher in impacts to the Front (0.337 ± 0.198 $p < 0.01$), Rear (0.382 ± 0.189 $p < 0.01$) and Side (0.295 ± 0.178 $p < 0.01$) compared to Top impacts (0.179 ± 0.195). There was no statistically significant difference between Side and Front impacts ($p = 0.139$). with the assumptions that the population was close to normal distribution, the samples independent population variance is equal and that the groups are of equal sample size (n = 168 each).

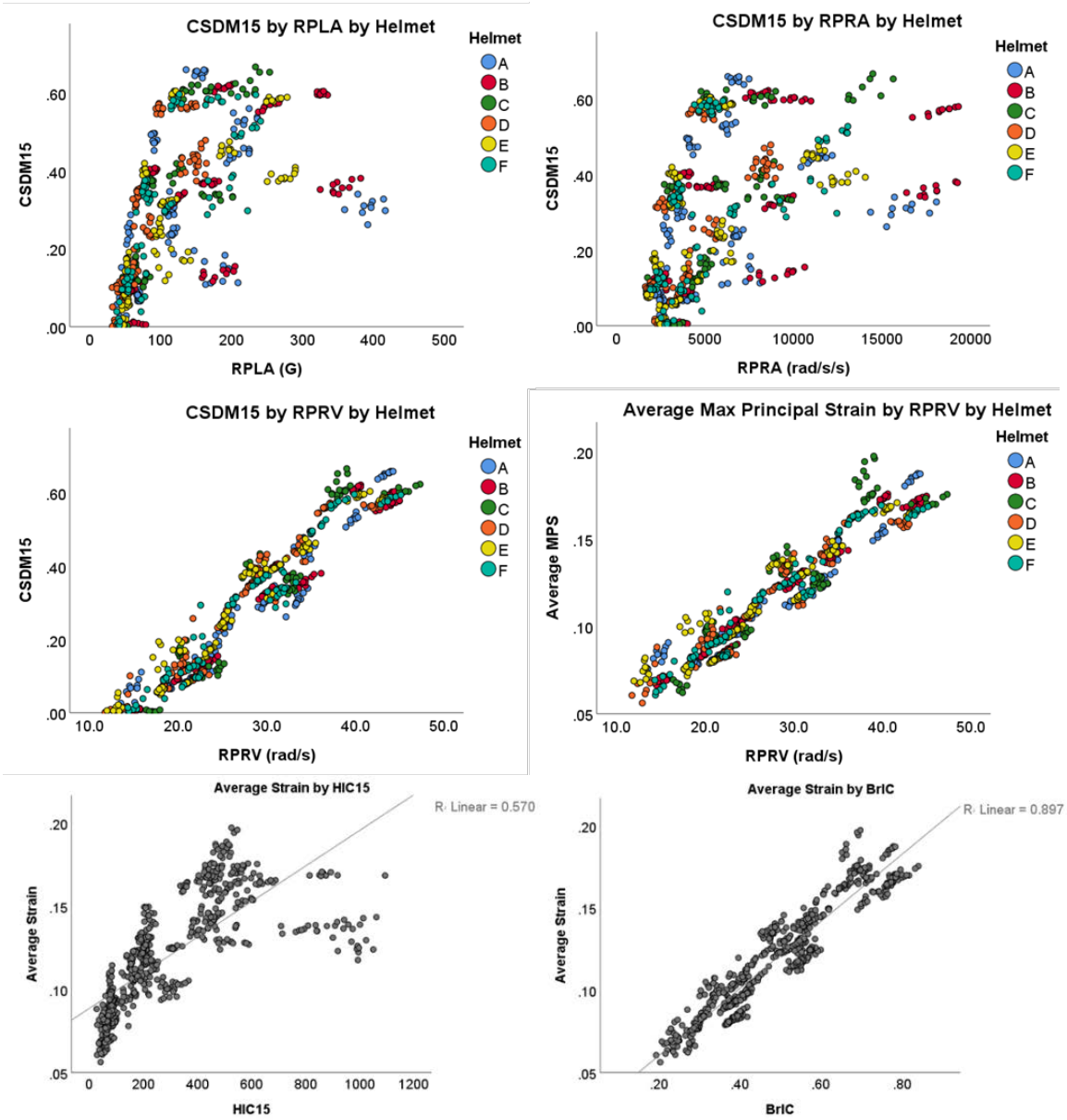

Fig. 5. These graphs represent the peak resultant kinematic of each impact scenario (n=672); Top left, resultant peak linear acceleration (g) compared to CSDM15, top right, resultant peak rotational acceleration (rad/s/s) resultant compared to CSDM15, bottom left peak rotational velocity(rad/s) compared to CSDM15 and bottom right is is RPRV compared to average max principal strain. The bottom two grayscale scatterplots, show the relationship of common injury prediction criteria HIC_{15} and BrIC.

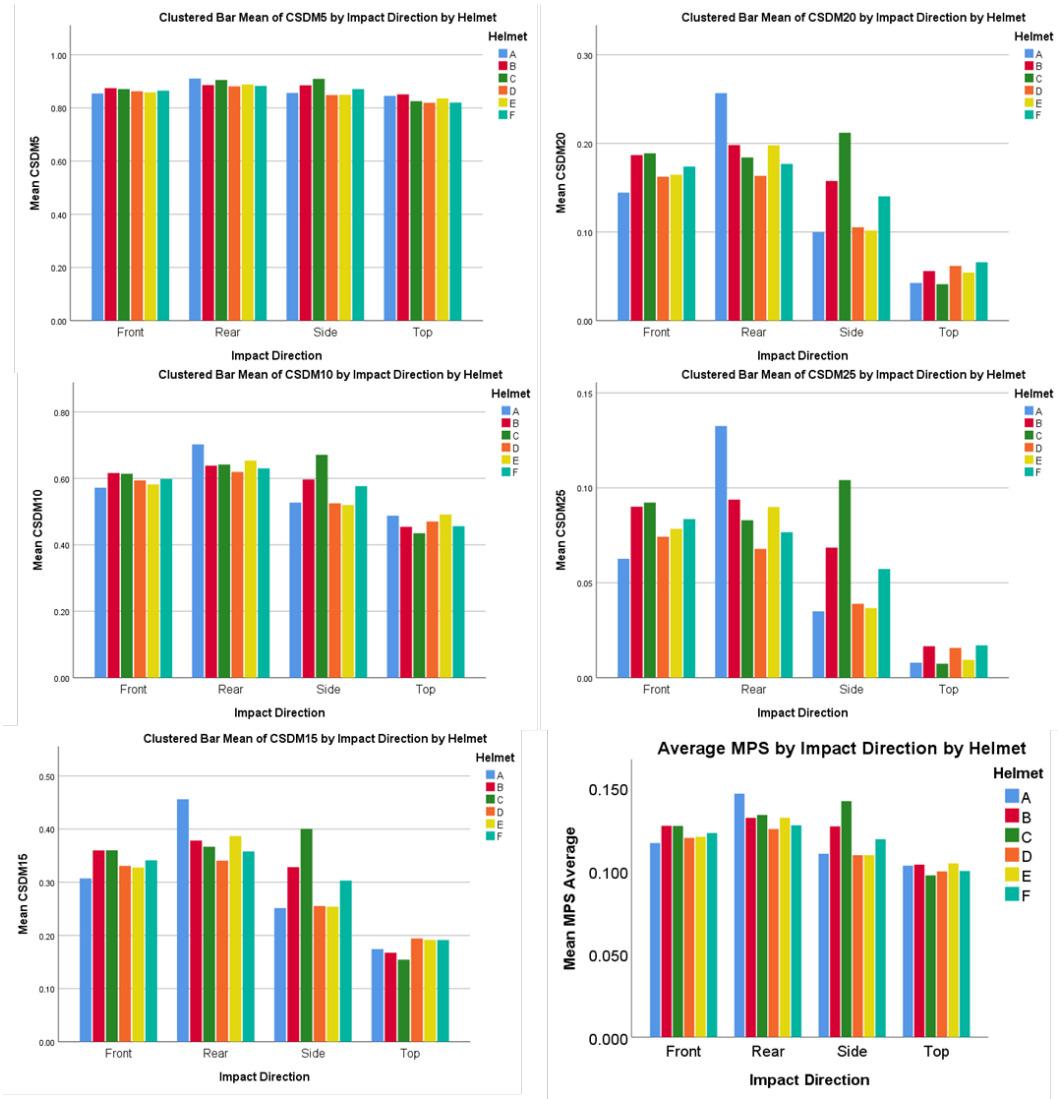

Fig.6. This graph examines the differences between impact direction and relative CSDM effect for all five CSDM levels analysed (CSDM5 to CSDM25), each direction has then been further broken down into the relative effect each helmet has in mitigating the CSDM value and for comparison average MPS.

IV. Discussion

Adopting a Computational Brain Injury Prediction Pipeline besides Experimental Testing

This study was established behind the theory that mechanical strain is an effective tool to evaluate the brain injury response using computational head models. The use of CSDM has been an effective tool to correlate the overall strains experienced by the GHBMC model and the probability of brain injury that is to be expected following a similar impact in a real world traumatic head impact scenario [21]. Such a pipeline is justified as timely as both large quantity of experiment tests and high-quality FE human head models have been made available in the field. Such a pipeline will help to overcome the hurdle of time and difficulty of handling hundreds to thousands of head impact simulations, while allowing for quick comparisons of current injury prediction metrics and future metrics and brain strains. More datasets to be generated in the field, together with our study, will allow users to better understand the protective effects of helmets by understanding how helmets help to reduce brain strain. Where the real innovations will occur is with the compilation of such large datasets, with multiple parameters, to allow for in-depth analysis and the application of new technologies and research tools to provide a computational solution to this complex problem.

Using Peak Rotational Velocity rather than Peak Rotational Acceleration

This study was an effective indicator of the relationship between rotational velocity and the brain strain response of the brain based on 672 experimental impacts and 672 FE simulations. The substantially elevated correlation between rotational velocity and brain strain in the form of CSDM supports the usage of peak rotational velocity rather than peak rotational acceleration and peak linear acceleration. Linear acceleration as a good

indicator of intracranial pressures [12], is substantially less effective in providing insight into the brains deformation incurred through maximum principal strains, maximum shear strains and other stress-strain related metrics [24]. In previous practice, the use of helmet-mounted head impact telemetry (HIT) system [25,26] documented valuable peak head acceleration, but ignored the entire time histories of accelerations and hence missed the opportunity of calculating rotational velocity. Hence, newly developed sensors such as mouth guard sensors [27,28] that not only rigidly attached to the skull to minimise helmet-to-head sliding effect, but also provided time histories of rotational kinematics, will help to collect rotational velocities from human participants. Hence, the correlation between rotational velocity and observed concussion risks can be analysed to further improve helmet evaluation approach.

Studies that involve the use of several FE head models, head kinematics, brain responses and brain injury risks have been conducted. A study which was also focused on ice hockey and used a specific guided drop tower to collect head kinematics, reported that resultant change in angular velocity best predicted MPS and CSDM15 [29]. Beyond ice hockey, Takhounts et al. also reported strong correlation between max resultant rotational velocity and CSDM with an R^2 of 0.92 [21]. Our in-house study using various theoretical loading curves supported rotational velocity best correlated with CSDM and average MPS [30]. It was also demonstrated that the peak change in angular velocity was shown to have a better correlation with strain levels for purely rotational impulses than angular acceleration, or HIC, both of which were demonstrated in this study [31,32]. Given agreements made across research groups, injury metrics like BrIC that considers rotational velocity are recommended to predict strain-related brain damage. Together these studies highlighted that while linear acceleration is a good indicator of the intracranial pressure it lacks the ability to indicate the deformations in the brain. While rotational acceleration, which previously was believed to correlate to brain strain, is insufficient. Future helmet methodologies, particularly ones that attempt to quantify the potential risk of mTBI, are recommended to be focused on a combination of linear acceleration that correlates to brain pressure and rotational velocity that correlates to brain strain. This study reasons that injury assessment metrics, like that of Hockey STAR, can be further developed and improved by considering the inclusion of rotational velocity in combination with rotational acceleration to improve strain based prediction accuracy.

Recommended use of Brain Strains Pipeline and Limitations

This study helps to highlight the potential applications of a fully automated testing to injury prediction pipeline for the categorisation of consumer focused helmets. While this paper focused its direction on that of hockey helmets, the possibilities of this helmet process are numerous. The STAR helmet rating protocol itself encompasses more sports than just hockey and any other kinematics based helmet testing protocol that outputs X, Y and Z-direction kinematics could also be adapted to this pipeline seamlessly.

There are limitations of this study. As is the case with brain injury, the understanding of what constitutes as a definitive concussive or sub-concussive scenario is still an uncertainty and therefore the reliance on only CSDM as the singular predictor of brain injury is a limiting factor. Although CSDM is a validated DAI predictor in the GHBMC model, other predictors have been shown to be equally as effective or have exhibited different advantages in the GHBMC and other computational head and brain models. has shown that pressure is a good predictor for mTBI and that while rotational motion is effective in perpetuating strain response do very little in terms of contributing to pressure response [24]. Therefore, future studies using this dataset could include more examination and analysis of other predictor methodologies that would encompass all kinematic factors and therefore view a larger scope of the brain predictor metrics. Another limitation of this study is the inherent limitation with finite element head models that are validated against brain-skull relative motion, as it was recently determined to possibility not be fully sufficient in determining accurate strain prediction outputs [33,34]. Validation of head models against experimental brain strains in addition to brain-skull relative motion was suggested and will be further explored in future studies.

A limitation arises with the validity of this testing methodology, as while the side and top impacts have some tangential contact with the head forms, the rear and frontal impacts go through the centre of gravity and hence do no provide that tangential factor. While we acknowledge this as a limitation what this study explores is how this widely used and accepted experimental testing methodology is related and potentially lacking in its prediction of the brain's response, particularly in terms of brain strain. The Hockey STAR helmet testing protocol uses real world data as the exposure metric, validating some of the linear and rotational acceleration values, meaning that the conditions could be assumed to be realistic in hockey, and hence the evaluation of these helmets is appropriate, or at least consistent as a comparison tool.

V. CONCLUSIONS

We developed a novel computational brain injury prediction pipeline that was based on validated industry used methodologies and computational models to evaluate the concussive and sub-concussive mitigating potential of hockey helmets. The helmet industry, and in particular those of football and hockey are committed to innovating for changing consumer behaviours and provide helmets that are safer, especially in mTBI or concussive impacts. What this paper helps to provide is a stepping stone for the current limitation of some of the state of the art testing procedures and an approach into how to quickly and automatically test new metrics that could involve machine learning or artificial neural networks to predict kinematic injury thresholds to help design future helmets. This preliminary study focused on understanding and showcasing which traditional helmet testing output kinematics are most correlated to brain injury, as per the CSDM metric, and provided initial insight into the effects of direction on brain injury and how well individual helmets can mitigate these effects. Rotational velocity is a significant contributor and predictor of brain strains and future helmet testing protocols need to include this telling metric. This study has also provided some initial reporting that future innovations in helmet design need to take into account impact direction when attempting to limit brain strains. Future studies will delve deeper into understanding the underlying factors that influence the success of a particular helmet in mitigating brain strains and analyse more statistically significant relationships between particular kinematic mechanisms and the level of injury a real world subject might experience.

VI. ACKNOWLEDGEMENT

We acknowledge funding provided through the NSERC ENGAGE programme in partnership with Bauer Hockey Ltd, Blainville, Quebec, and Canada Research Chairs programme.

References

[1] Dewan, M.C., Rattani, A., et al. Estimating the global incidence of traumatic brain injury. *Journal of Neurosurgery*, 2019. 130(4): p. 1080-1097

[2] Taylor, C.A., Bell, J.M., Breiding, M.J., and Xu, L. Traumatic Brain Injury–Related Emergency Department Visits, Hospitalizations, and Deaths — United States, 2007 and 2013. *MMWR. Surveillance Summaries*, 2017. 66(9): p. 1-16

[3] Clay, M.B., Glover, K.L., and Lowe, D.T. Epidemiology of concussion in sport: a literature review. *Journal of Chiropractic Medicine*, 2013. 12(4): p. 230-251

[4] McKee, A.C., Stein, T.D., et al. The spectrum of disease in chronic traumatic encephalopathy. *Brain*, 2013. 136(1): p. 43-64

[5] Gavett, B.E., Stern, R.A., and McKee, A.C. Chronic Traumatic Encephalopathy: A Potential Late Effect of Sport-Related Concussive and Subconcussive Head Trauma. *Clinics in Sports Medicine*, 2011. 30(1): p. 179-188

[6] Rowson, B., Rowson, S., and Duma, S.M. Hockey STAR: A Methodology for Assessing the Biomechanical Performance of Hockey Helmets. *Ann Biomed Eng*, 2015. 43(10): p. 2429-43

[7] Smith, D.H. and Meaney, D.F. Axonal Damage in Traumatic Brain Injury. 2000. 6(6): p. 483-495

[8] Wright, R.M., Post, A., Hoshizaki, B., and Ramesh, K.T. A Multiscale Computational Approach to Estimating Axonal Damage under Inertial Loading of the Head. 2013. 30(2): p. 102-118

[9] Madhukar, A. and Ostoja-Starzewski, M. Finite Element Methods in Human Head Impact Simulations: A Review. *Annals of Biomedical Engineering*, 2019

[10] Rowson, S. and Duma, S.M. Development of the STAR evaluation system for football helmets: integrating player head impact exposure and risk of concussion. *Ann Biomed Eng*, 2011. 39(8): p. 2130-40

[11] Mao, H., Zhang, L., et al. Development of a finite element human head model partially validated with thirty five experimental cases. *Journal of biomechanical engineering*, 2013. 135(11)

[12] Takhounts, E.G., Eppinger, R.H., et al. On the Development of the SIMon Finite Element Head Model. *Proceedings*.

[13] Miller, L.E., Urban, J.E., and Stitzel, J.D. Validation performance comparison for finite element models of the human brain. *Computer Methods in Biomechanics and Biomedical Engineering*, 2017. 20(12): p. 1273-1288

[14] Zhao, W., Ruan, S., and Ji, S. Brain pressure responses in translational head impact: a dimensional analysis and a further computational study. *Biomechanics and Modeling in Mechanobiology*, 2015. 14(4): p. 753-766

[15] Kleiven, S. Finite element modeling of the human head. 2002, KTH.
[16] Horgan, T.J. and Gilchrist, M.D. The creation of three-dimensional finite element models for simulating head impact biomechanics. 2003. 8(4): p. 353-366
[17] Zhang, L., Yang, K.H., et al. Recent advances in brain injury research: a new human head model development and validation. 2001, SAE Technical Paper.
[18] Pellman, E.J., Viano, D.C., et al. Concussion in professional football: helmet testing to assess impact performance--part 11. *Neurosurgery*, 2006. 58(1): p. 78-96; discussion 78-96
[19] Hardy, W.N., Foster, C.D., et al. Investigation of head injury mechanisms using neutral density technology and high-speed biplanar X-ray. 2001, SAE Technical Paper.
[20] Hardy, W.N., Mason, M.J., et al. A study of the response of the human cadaver head to impact. *Stapp car crash journal*, 2007. 51: p. 17
[21] Takhounts, E.G., Craig, M.J., Moorhouse, K., McFadden, J., and Hasija, V. Development of brain injury criteria (BrIC). 2013, SAE Technical Paper.
[22] Mao, H., Yang, K.H., King, A.I., and Yang, K. Computational neurotrauma—design, simulation, and analysis of controlled cortical impact model. 2010. 9(6): p. 763-772
[23] Bain, A.C. and Meaney, D.F. Tissue-Level Thresholds for Axonal Damage in an Experimental Model of Central Nervous System White Matter Injury. 2000. 122(6): p. 615
[24] Kleiven, S. Predictors for traumatic brain injuries evaluated through accident reconstructions. 2007, SAE Technical Paper.
[25] Rowson, S., Duma, S.M., et al. Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion. *Annals of Biomedical Engineering*, 2012. 40(1): p. 1-13
[26] Cobb, B.R., Urban, J.E., et al. Head Impact Exposure in Youth Football: Elementary School Ages 9–12 Years and the Effect of Practice Structure. *Annals of Biomedical Engineering*, 2013. 41(12): p. 2463-2473
[27] Camarillo, D.B., Shull, P.B., Mattson, J., Shultz, R., and Garza, D. An Instrumented Mouthguard for Measuring Linear and Angular Head Impact Kinematics in American Football. 2013. 41(9): p. 1939-1949
[28] Miller, L.E., Pinkerton, E.K., et al. Characterizing head impact exposure in youth female soccer with a custom-instrumented mouthpiece. *Research in Sports Medicine*, 2020. 28(1): p. 55-71
[29] Knowles, B.M. and Dennison, C.R. Predicting cumulative and maximum brain strain measures from HybridIII head kinematics: A combined laboratory study and post-hoc regression analysis. *Annals of biomedical engineering*, 2017. 45(9): p. 2146-2158
[30] Bian, K. and Mao, H. Mechanisms and variances of rotation-induced brain injury: a parametric investigation between head kinematics and brain strain. *Biomechanics and modeling in mechanobiology*, 2020: p. 1-19
[31] Kleiven, S. Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure. *International Journal of Crashworthiness*, 2006. 11(1): p. 65-79
[32] Kleiven, S. Influence of direction and duration of impacts to the human head evaluated using the finite element method. *Proceedings of International IRCOBI Conference*, 2005.
[33] Zhou, Z., Li, X., Kleiven, S., Shah, C.S., and Hardy, W.N. A reanalysis of experimental brain strain data: implication for finite element head model validation. *Proceedings of SAE 62nd Stapp Car Crash Conference, STAPP 2018; Catamaran Resort Hotel San Diego; United States; 12 November 2018 through 14 November 2018*, 2019.
[34] Giordano, C. and Kleiven, S. Development of an unbiased validation protocol to assess the biofidelity of finite element head models used in prediction of traumatic brain injury. 2016, SAE Technical Paper.