Towards a helmet assessment metric capable of predicting diffuse brain injury while accounting for focal head injury

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ABSTRACT (191 words)

Helmets certified by today's linear acceleration based criteria are credited with all-but eliminating fatal focal injury in sports. Brain injuries may still occur where there is absence of severe focal injury leading to discussions regarding how helmet assessment methods might move towards the inclusion of impact parameters relevant to brain injury. To first understand the relationship between kinematic measures and computed brain strain, we conducted hundreds of impacts using the 50th percentile Hybrid III head-neck equipped with an ice hockey helmet. We then input the three-dimensional impact kinematics to a finite element brain model called the Simulated Injury Monitor (SIMon) to compute brain strain measures including the cumulative strain damage measure (specifically CSDM-15). Resultant change in angular velocity ($\Delta\omega_R$) was the single best kinematic predictor for CSDM-15 and better predicted strain than the current helmet certification metric, peak g. The best two-variable model included peak angular acceleration and $\Delta\omega_R$, though an efficient model for predicting CSDM-15 that included at least one linear and one angular kinematic included two variables: peak g and $\Delta\omega_R$. A preliminary metric based on peak g and $\Delta\omega_R$ was presented and a possible threshold limit was proposed.

INTRODUCTION

Helmet use is known to mitigate severe focal head injuries, though despite widespread use, sport-related traumatic brain injury (TBI) remains the second most common cause for TBI hospitalization (Gilchrist et al. n.d.). The rate of concussion in ice hockey occurs at 0.54 for high school (Marar et al. 2012), 0.41-3.1 for collegiate (Flik et al. 2005; Hootman et al. 2007) and 1.81 for professional (Wennberg and Tator 2008), per 1,000 exposures. The growing concern surrounding brain injury in sport and the role that helmets could play in mitigating brain injury has sparked discussion among standards organizations to move towards helmet certification methods to evaluate helmets relative to impact parameters related to brain injury.

Today's contemporary helmets are certified against linear acceleration magnitudes ("ASTM F1045-07: Standard Performance Specification for Ice Hockey Helmets" 2007, "CSA Z262.1-09: Standard for Ice Hockey Helmets" 2012) or functions based on acceleration ("Standard Performance Specification for Newly Manufactured Football Helmets, NOCSAE DOC (ND) 002-13m15" 2015) meant to represent impact severity. The rationale to use linear acceleration as an attenuation metric is due in part to head injury biomechanics research conducted out of Wayne State University in the 1960s (Gurdjian et al. 1964). Through consideration of animal and human exposure data, the cerebral concussion Tolerance Curve (WSTC) was developed that defined a relationship between linear acceleration and time duration and severe head injury. In doing so, the WSTC identified a limit for which the human head can tolerate a given linear acceleration magnitude. This was some of the earliest work on identifying injury thresholds and in an attempt to approximate the WSTC, severity metrics were introduced with the intention of quantifying impact severity based only on measurable kinematics.

The severity index (SI) integrates linear acceleration over time, limited by a defined threshold, and is used primarily in football helmet certification ("Standard Performance Specification for Newly Manufactured Football Helmets, NOCSAE DOC (ND) 002-13m15" 2015). The head injury criterion (HIC) also integrates linear acceleration over time, but with a set time duration of 15 msec or 36 msec. Using a simplified approach, many of today's helmet standards consider only the peak linear acceleration (peak g) to quantify impact attenuation. Applying these kinematic-based metrics during helmet certification has led to helmets that are credited with all-but eliminating fatal focal injury in contact sports (Daneshvar et al. 2011). Discussions among standards organizations are now centered on determining how best to modify certification methods to not only protect against severe focal injury, but also consider diffuse brain injury.

Studies dating back to the 1940s have established a connection between head rotation and diffuse brain injury. In 1943, Holbourn applied translational and rotational loads to a gelatin mixture intended to represent the brain and observed the occurrence of greater strains under rotational motion (Holbourn 1943). Rhesus monkeys (Yarnell and Ommaya 1969) subjected to whiplash conditions and later squirrel monkeys (Gennarelli et al. 1972) experiencing linear and angular motions were also studied to better understand the relationship between head motion and brain injury. Each of these studies confirmed the significant role that angular motion plays in causing tissue damage to the brain.

With the knowledge that angular motion plays a significant role in causing brain injury, assessment functions have been developed that incorporate angular kinematics. However, agreement is yet to be reached regarding which application of kinematic measures is most suitable for diffuse injury prediction. Both linear and angular kinematics are considered together in two unique functionals: the Generalized Acceleration Model for Brain Injury Tolerance (GAMBIT (Newman 1986)) and the Head Impact Power (HIP (Newman et al. 2000)). The Brain Injury Criterion (BrIC (Takhounts et al. 2013)), Rotational Injury Criterion (RIC (Kimpara and Iwamoto 2012)) and Power Rotational Injury Criterion (PRHIC (Kimpara and Iwamoto 2012)) are examples of functionals based exclusively on angular kinematics. The Hockey Summation of Tests for the Analysis of Risk (Hockey STAR) formula was developed specifically for helmet assessment and is a function of linear and angular acceleration (Rowson et al. 2015). Though assessment functions have been developed that incorporate angular kinematics, no single functional has been agreed upon for use in impact attenuation assessment for helmets.

A method for evaluating human tissue response to inertial loading with more detail than kinematic functionals, and without the need for experiments involving cadavers, comes in the form of finite element head models. The Simulated Injury Monitor (SIMon) approximates the average male skull, cerebrospinal fluid layers, bridging veins and brain (cerebrum, cerebellum and upper spinal cord) (Takhounts et al. 2008). Examples of other models currently in use include Global Human Body Modeling Consortium (GHBMC) (Takhounts et al. 2013), Wayne

State University Bead Injury Model (WSUBIM) (Zhang et al. 2001), and the University College Dublin Brain Trauma Model (UCDBTM) (Horgan and Gilchrist 2004). These models represent details beyond the major structures of the brain and skull represented by SIMon, including facial bones, scalp and in some cases a deformable skull. Tissue deformation is approximated by computing mechanical measures such as maximum principal stress, maximum principal strain and maximum pressure. The cumulative strain damage measure (CSDM) and maximum axonal strain have been proposed as measures representing diffuse brain injury risk.

Though numerical models exist for estimating brain strain, kinematic functionals continue to drive the method for assessing helmet attenuation during impact. Impact test equipment to incorporate rotational motion is yet to be determined, though it is likely that angular head kinematics will be used in future certification methods. The adoption of kinematic functionals incorporating angular head rotation will require headforms that are capable of measuring realistic head rotation and it is therefore important to understand the correlation between linear and angular kinematics and tissue strain measures. Identifying kinematics capable of predicting brain strain when used in combination with the chosen test-bed would allow a kinematic functional to be used during impact tests rather than numerical models.

The objective of this work is to identify correlations between head impact kinematics and tissue strain measures and propose a metric considering both focal and diffuse injury with a threshold strategy suitable for assessing helmet performance during impact testing. Using the HybridIII head-neck, we measured three-dimensional linear and angular head kinematics during impact, which were then input to the Simulated Injury Monitor (SIMon) head-brain finite element model to determine SIMon-computed brain tissue distress. Multiple regression techniques determined the most efficient set of kinematic variables for predicting strain measures while including linear kinematics for focal injury consideration. A metric suitable for use in certification-style helmet testing and a potential method for thresholding is proposed in this study.

METHODS

The experimental setup included a guided rail drop tower with adjustable drop gimbal, an anthropomorphic test device (ATD) head and neck (HybridIII 50th Percentile, 11 kg total mass of gimbal and head-neck) and a modular elastomer programmer surface mounted to a stationary steel impact anvil (Figure 1). This experimental arrangement is one paradigm that is currently being considered for future helmet assessment methods and is also common in head impact evaluation.

Figure 1: Guided impact tower with helmeted 50th percentile H-III head and neck mounted on a custom gimbal with a purpose built velocity gate

A total of 267 impacts were conducted on 55 CSA certified helmets (Bauer 4500, size medium) including a variety of drop heights and impact locations. The experimental protocol was guided by a study by Brainard et. al. that recorded common impact sites and severities of collegiate ice hockey players. Impact distribution, sorted by impact velocity, is shown in Table 1 and impact locations referenced in this table are shown in Figure 2. The range of impact speeds included $1.2ms^{-1}$ to $5.8ms^{-1}$, encompassing speeds outlined in ice hockey helmet standards ("ASTM F1045-07: Standard Performance Specification for Ice Hockey Helmets" 2007, "CSA Z262.1-09: Standard for Ice Hockey Helmets" 2012).

Table 1: Resulting distribution of the number of impacts categorized by impact speed and impact location

	No. of Impacts				
Location	1.0-2.4 m/s	2.5-3.4 m.s	3.5-4.4 m/s	4.5-6 m/s	Total
Front	43	17	16	29	105
Back	44	9	25	10	88
Side	51	8	9	6	74
All	138	34	50	45	267

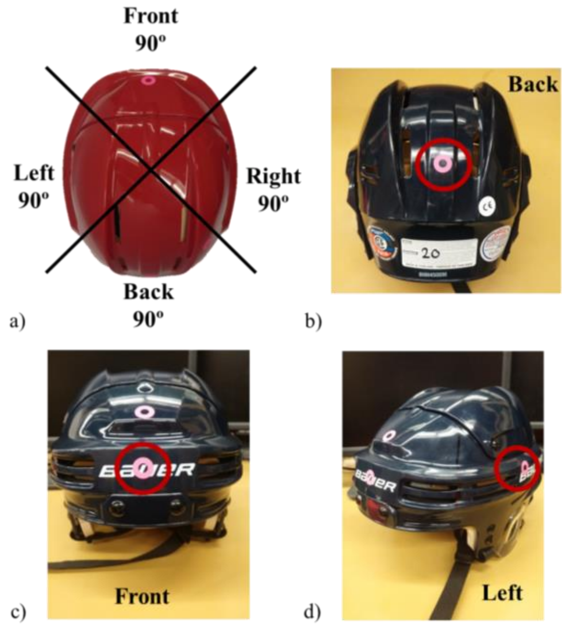


Figure 2: Bauer™ hockey helmet showing a) Top view of impact regions defined by 90-degree sections, and impact sites to the b) back, c) front and d) side

Nine uniaxial accelerometers (Measurement Specialties Inc. Hampton VA, model 64C-2000-360), arranged in a 3-2-2-2 array were mounted in the HybridIII headform. Linear acceleration measures from the 9 accelerometers were translated to linear accelerations and angular accelerations about the head center of mass (Padgaonkar et al. 1975). A forward integration function computed linear and angular velocity from linear and angular acceleration, respectively. Impact speed was measured within 40 mm of impact by a purpose-built velocity gate.

Impact acceleration data was collected and saved at 100 kHz using National Instruments hardware and software (PXI 6251 and Labview v8.5, Austin TX). Analog voltages were anti-alias filtered with cut-off frequency 4 kHz using hardware prior to post-process low-pass filtering per CFC 1000 ("SAE J211 Instrumentation for Impact Test - Part 1: Electronic Instrumentation" 2007).

HybridIII kinematics, from the 267 impacts were input to the Improved Simulated Injury Monitor (SIMon (Takhounts et al. 2008)) brain-skull FE model. The cumulative strain damage measure (specifically CSDM-15) is a mechanical measure used here to represent brain tissue deformation. CSDM-15 represents the cumulative volume fraction of the brain that reaches or exceeds a tensile strain of 15% or greater. SIMon-computed CSDM, correlates with probability of diffuse anatomic injury, based on injury data from animals and college football (Takhounts et al. 2013). CSDM-15 was determined over 80 msec, by which time CSDM-15 reached a stable maximum.

Multiple regression techniques, using the equation below, compared linear regression models predicting CSDM-15. The results were used to determine the most efficient set of kinematics capable of predicting SIMon-computed brain strain. This study focuses on CSDM-15, though results were found previously considering MPS as well (Knowles and Dennison 2017).

$$CSDM15 = a_0 + \beta_1 a_1 + \beta_2 a_2 + \beta_3 a_3 + \cdots + \beta_k a_k$$

Beginning with a single predictor (k=1), as kinematic terms were added or replaced (a_k), statistical measures including weighted coefficients (β_k) and their significance ($p < 0.05$) and adjusted R^2 were computed. The F-statistic was computed to compare models with similar R^2 values and was used here to indicate model efficiency in predicting the data set and, similar to R^2, a higher F-statistic is favorable (Devore 2000).

The single kinematics to be considered individually and in combination include: peak resultant linear acceleration (peak g), impact velocity (V_i), resultant change in linear velocity (ΔV_R), peak resultant angular acceleration (α_R), resultant change in angular velocity ($\Delta\omega_R$), directional change in angular velocity ($\Delta\omega_x$, $\Delta\omega_y$, $\Delta\omega_z$), peak resultant angular velocity (ω_R) and directional peak angular velocity (ω_x, ω_y, ω_z). Maximum kinematic values were determined irrespective of the time that the values occurred.

A subset of regression models comprising linear and angular kinematics are focused on in this study. Through assessment of these regression models against established kinematic risk-tolerance curves, we propose possible methods by which threshold magnitudes can be set. Our general approach is to determine an acceptable risk of brain injury based on CSDM-15, then to determine the corresponding limiting magnitude on the assessment metric.

RESULTS

Considering all impact locations together as a single dataset, Table 2 presents regression coefficients for each variable and adjusted R^2 and F-statistic for each of the regression models considered for predicting CSDM-15. A summary of two-variable regression models used for further consideration as a new metric are shown in Table 3. Table 3 presents the variables used to create each model and the resulting adjusted R^2 and F-statistic.

Table 2: Multiple regression models for CSDM-15 with each row containing a unique set of predictor variables to form a model with the model adjusted coefficient of determination (Adj R^2) and F-statistic in the right hand columns (bold text in adj R^2 and F columns indicate maximum values). Variables included in each model are indicated by their regression coefficient displayed (each coefficient has been multiplied by 10^6; bold and italicized text indicates that it is a significant predictor with p-value < 0.05)

No. of Variables	Model No.	Peak g	Vi	ΔVR	αR	Δωx	Δωy	Δωz	ΔωR	ωx	ωy	ωz	ωR	Adj R²	F
1	1	***4064***												0.36	152
	2		***95862***											0.40	181
	3			***90104***										0.44	210
	4								***16617***					0.86	**1629**
	5												***21784***	0.82	1252
	6				***38***									0.11	33
2	7	-1590		***120345***										0.44	107
	8	***-618***							***17768***					0.86	841
	9	***722***											***20276***	0.83	658
	10			-531					***16667***					0.86	812
	11			***24986***									***19120***	0.85	731
	12	***4541***			-9									0.37	75
	13			***95844***	-4									0.46	106
	14				***-8***				***17192***					0.87	851
	15				3								***21666***	0.83	621
3	16	***-2866***		***52151***					***17003***					0.88	638
	17	***-2464***		***71045***									***19356***	0.86	544
	18									***12462***	***16076***	***6079***		0.87	593
	19					***11420***	***13590***	-2050						0.86	525
	20	***-2233***		***134006***	3									0.46	73
4	21	-245				***11575***	***13819***	-1546						0.86	394
	22	-99								***12507***	***16180***	***6493***		0.87	443
	23	***-2494***	7048	***65215***									***19394***	0.86	407
	24	***-2496***	***-93004***	***127830***					***16992***					**0.89**	525
	25	-1771	***-161161***	***268258***	5									0.49	60
5	26	***-1107***		***24095***						***12823***	***15668***	***5414***		0.87	362

Table 3: Summary of the variables included in two-variable regression models predicting CSDM-15 with the model adjusted coefficient of determination (Adj R^2) and F-statistic in the right hand columns. Each row represents a unique pair of variables. Variables included in each model are indicated (bold and italicized text indicates that it is a significant predictor with p-value < 0.05)

Impacts	Variables included in regression model			Adj R²	F
CSDM-15					
		αR	***ΔωR***	0.87	851
	Peak g		***ΔωR***	0.86	841
	Peak g	αR		0.37	75

When considering all impact locations together as one dataset, $\Delta\omega_R$ was the single best kinematic predictor for CSDM-15 with R^2 of 0.86 and F-statistic 1629 (Table 2, row 4). A single variable model containing $\Delta\omega_R$ identified as the most efficient model with the maximum F-statistic and proved a better predictor for CSDM-15 than peak g and α_R, consistent when considering all impacts together as well as evaluating impact locations individually.

For all regressions models (Table 2, rows 1-26), as a new term was added to a previous model, adjusted R^2 increased or stayed the same, while the F-statistic always decreased. For example, as predictor variables were added to include one to four variables, the maximum F-statistic decreased from 1629 (Table 2, row 4) to 525 (Table 2, row 24). The modest improvement in explained variance between a model containing one variable (0.86 in row 4) and four variables (0.89 in row 24) resulted in a less efficient regression model (as indicated by low F-statistic). In nearly all cases, excluding only back impacts for predicting CSDM-15, the best two-variable model for predicting strain measures included α_R and $\Delta\omega_R$. Considering models for predicting CSDM-15 that contain at least one linear and one angular kinematic, an efficient model that maximizes F-statistic included two variables: peak g and $\Delta\omega_R$.

DISCUSSION

This study identifies optimum combinations of kinematics for predicting brain strain and determines that a single angular kinematic can predict CSDM-15. The single best kinematic predictor for CSDM-15 is $\Delta\omega_R$. Whether considering all impact locations together or considering each impact location separately, the model that included only $\Delta\omega_R$ achieved the highest F-statistic for predicting brain strain measures. Evidence that $\Delta\omega_R$ can act as a single predictor for strain is supported by work by Takhounts et. al who, considering injury in automotive impacts, found angular velocity to be a better correlate to CSDM than any other kinematic measure or functional (Takhounts et al. 2008).

$\Delta\omega_R$ better predicts CSDM-15 than the current helmet certification metric, peak g. The significance of this finding is that it confirms that predicting brain strain for the current setup requires monitoring $\Delta\omega_R$ rather than peak g alone, which is an important finding as standard organizations discuss adopting new test methods to account for diffuse injuries.

The model that achieves the greatest adjusted R^2 is not the same model that maximizes the F-statistic for predicting both CSDM-15, and therefore forces a compromise when selecting an ideal model. The model with maximum adjusted R^2 for predicting CSDM-15 includes peak g, Vi, ΔV_R, and $\Delta\omega_R$. Measuring 4 kinematics is less efficient and may not be necessary to maintain a high R^2.

To maximize the F-statistic and create an efficient model, fewer variables should be chosen. For the kinematics considered in this study, choosing a model with the highest adjusted R^2 could require measuring up to four terms, while brain strain measures can be predicted with as little as one angular variable. Provided $\Delta\omega_R$ is included in the regression model, adjusted R^2 showed a maximum improvement of 3.5% (Table 2, row 4 to row 24). This suggests a more complex, multi-variable model may not be necessary to predict brain strain measures in helmet drop tests and therefore a metric containing the least number of variables would be best.

Including $\Delta\omega_R$ accounts for diffuse injury based on its ability to predict CSDM-15 and by adding a linear term, the model could also account for focal injury. The statistical data show multi-variable options that correlate with brain strain based on both linear and angular kinematics. Aiming to maximize F while including at least one linear and one angular term, the most efficient models would contain only two predictor variables. Predicting CSDM-15, the two-variable model that achieves the maximum F includes α_R and $\Delta\omega_R$, which lacks the desired linear component. Models that comprise both a linear and angular term, and have high R^2 and F, include peak g and either α_R or $\Delta\omega_R$, achieving two goals including minimizing the number of terms and incorporating linear and angular components. Three models in preliminary functional forms can be seen in Figure 3, which identifies a functional based on peak g and $\Delta\omega_R$ as the best two-variable option for predicting CSDM-15 as it maximizes R^2 and contains both linear and angular terms. By including peak g and $\Delta\omega_R$, we can create a metric that is capable of predicting strain measures, while accounting for focal injury through the inclusion of peak g.

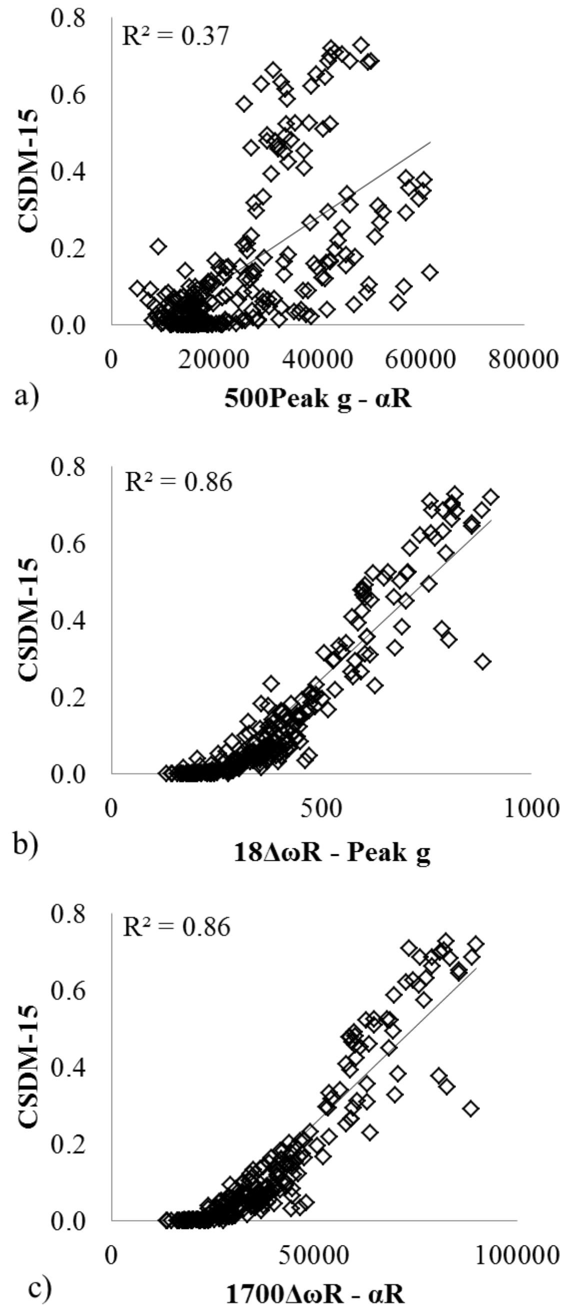
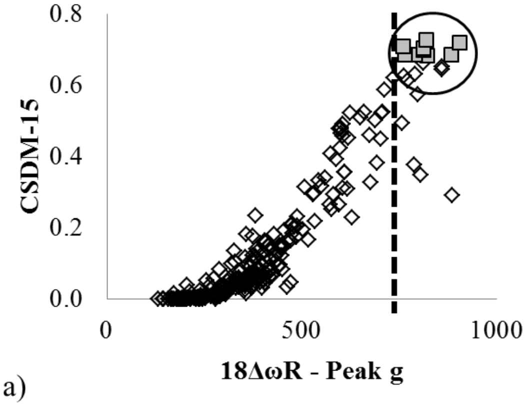


Figure 3: Preliminary considerations for an assessment metric plotted against strain measure CSDM-15 for metrics including a) peak g and α_R, b) $\Delta\omega_R$ and peak g and c) $\Delta\omega_R$ and α_R

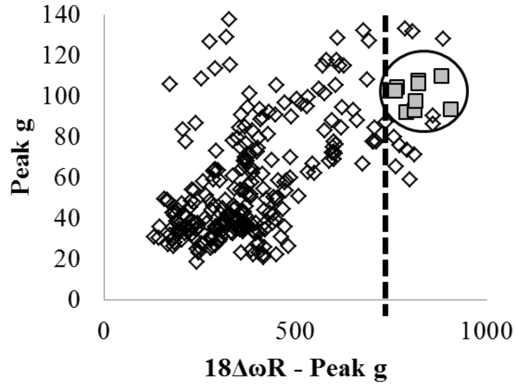
To make any metric appropriate for use in pass/ fail helmet certification standards, it is necessary to establish the type of injury and its associated acceptable risk level. Though the curve developed by Mertz et. al. is based on bare head impact and differs from our helmeted impacts, using this curve as an example suggests that a peak g value of 150 g corresponds to 1% risk of skull fracture (Mertz et al. 2003). Because all of our impacts had peak g less than 140 g, suggesting an exceedingly small risk of focal injury, we do not need to threshold based on peak-g. In this proposed thresholding approach, we focus on thresholding based on CSDM-15. Based on the risk curves developed relating CSDM and AIS injuries, a limit for CSDM-15 was selected to be 0.67 (associated with a 50% risk of severe concussion (AIS 3) (Takhounts et al. 2013)).

To determine the relationship between the preliminary metric and injury risk, the metric can be plotted against an injury measure representative of focal or diffuse injury, considered here as peak g and CSDM-15, respectively. Figure 4a plots the chosen metric ($18\Delta\omega R$ - *Peak g*) against CSDM-15. Choosing a CSDM-15 limit of 0.67 deems the circled values as unacceptable. Limiting CSDM-15 to 0.67 translates to a value for $18\Delta\omega R$ - *Peak g* of 750. In other words, the threshold magnitude for $18\Delta\omega R$ - *Peak g* is 750.

As shown in Figure 4b, plotting the metric versus peak g, and based on the presented thresholding method, this could mean peak g values of approximately 100 g could result in a failed impact test. 100 g is well below the current ice hockey helmet threshold of 275 g though it should be noted that impacts using the HybridIII head and flexible neck rarely exceeded 100 g for impact speeds at or below certification requirements. Additionally, a failed impact test for linear acceleration near 100 g only occurred in combination with a $\Delta\omega_R$ term large enough to cause the metric value to exceed 750. Alternatively, the maximum peak g reached was 138 g, as seen in Figure 4b, and could be considered an impact where the helmet would pass because angular velocity remained low enough that the combined injury risk is acceptable (i.e. metric value is less than 750). Pass/ fail thresholds by this method are driven mainly by angular velocity, however, keeping peak g in the metric allows certifiers to also assess whether a helmet reaches linear accelerations yielding skull fracture risk. The combination of peak g and $\Delta\omega_R$ creates an efficient metric for predicting CSDM-15, making it appropriate for use in testing against both focal and diffuse injuries. Setting a quantifiable limit establishes a clear failure point and allows helmet manufacturers to quickly and effectively assess helmets.



a)



b)

Figure 4: Assessment metric based on $\Delta\omega_R$ and peak g plotted against a) CSDM-15 and b) peak g. Circled data points indicate events resulting in CSDM-15 values greater than 0.67 and the dashed line indicates a metric value threshold of ~750

This work is limited by our exclusive use of SIMon. We feel the use of SIMon is appropriate for our test bed which is based on HybridIII equipment, the same equipment used by Takhounts et al. in developing strain based brain injury measures including CSDM. However, we acknowledge that use of another brain model has the potential to alter the findings in this study.

Further, we acknowledge that the results of this study and regression model coefficients are influenced by our use of the HybridIII neck and our specific experimental setup. Therefore, future work will look at helmeted impacts with the HybridIII head and no neck constraint to determine whether kinematic correlations and regression models differ from the present study as well as to provide insight for methods being considered by European standards associations (Halldin 2014).

CONCLUSIONS

This study is the first to statistically determine the best set of kinematic predictors for brain strain measures including both linear and angular terms in a certification-type helmet assessment experiment using HybridIII equipment. Statistical analysis performed on impact data of hundreds of helmeted impacts concluded that angular velocity is the single best kinematic predictor for brain strain and that a metric based on linear and angular kinematics can effectively and efficiently predict brain strain measures. The results of this study outline a method for developing a helmet assessment metric and quantifiable threshold for use in certification drops with the HybridIII head and neck.

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