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Development of a Metric for Predicting Brain Strain Responses Using Head Kinematics

LEE F. GABLER, JEFF R. CRANDALL, and MATTHEW B. PANZER

Department of Mechanical and Aerospace Engineering, University of Virginia, Center for Applied Biomechanics, 4040 Lewis and Clark Drive, Charlottesville, VA 22911, USA

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Abstract—Diffuse brain injuries are caused by excessive brain deformation generated primarily by rapid rotational head motion. Metrics that describe the severity of brain injury based on head motion often do not represent the governing physics of brain deformation, rendering them ineffective over a broad range of head impact conditions. This study develops a brain injury metric based on the response of a second-order mechanical system, and relates rotational head kinematics to strain-based brain injury metrics: maximum principal strain (MPS) and cumulative strain damage measure (CSDM). This new metric, universal brain injury criterion (*UBrIC*), is applicable over a broad range of kinematics encountered in automotive crash and sports. Efficacy of *UBrIC* was demonstrated by comparing it to MPS and CSDM predicted in 1600 head impacts using two different finite element (FE) brain models. Relative to existing metrics, *UBrIC* had the highest correlation with the FE models, and performed better in most impact conditions. While *UBrIC* provides a reliable measurement for brain injury assessment in a broad range of head impact conditions, and can inform helmet and countermeasure design, an injury risk function was not incorporated into its current formulation until validated strain-based risk functions can be developed and verified against human injury data.

Keywords—Brain deformation, Finite element modeling, Rotational, Second-order system.

INTRODUCTION

Each year in the United States (US), traumatic brain injury (TBI) is a primary or secondary diagnosis in 16% of injury-related hospitalizations, and contributes to nearly one-third of injury-related deaths.[4] Although estimates vary, falls remain the leading cause of TBI (47%).[39] While MVCs are the third highest source of TBI (14%), they contribute to the largest number of TBI-related deaths among people aged (5–24) years.[39] These figures only include civilian estimates, and do not account for those receiving care at a federal facility (e.g., military personnel) nor do they include sport-related concussions, which are vastly underreported.[13] Closed-head impacts have been identified as the largest cause of TBI-diagnosed cases.[33] Thus, reliable TBI risk assessment models are needed to inform the design of safety systems that are effective at protecting against this mechanism of brain injury during head impact.

TBI risk assessments are made using criteria which consist of a biomechanical metric and an injury risk function. The metric summarizes head impact severity, and is a mathematical function of one or more biomechanical response variables. Metrics used for brain injury criteria can be categorized into two types: kinematic-based and tissue-level-based. Kinematic metrics are based on rigid-body motion parameters of the head, while tissue-level metrics are based on mechanics of the parenchyma. The risk function is a probabilistic model that relates the metric to brain injury likelihood. While the risk function is necessary for brain injury prediction, the underlying metric is responsible for representing the injury mechanism and relative severity.

Head impact kinematics have been the basis for most head injury metrics. This is likely due to the feasibility of measuring and summarizing head kinematic response, either on a dummy or a volunteer, relative to measuring brain tissue response. Head kinematics can be separated into two different types of motion: translational and rotational. Early metrics were formulated using only translational parameters of head motion, and existing safety standards used in helmet and crash testing have been based on linear head acceleration.[26,44] Although

Address correspondence to Matthew B. Panzer, Department of Mechanical and Aerospace Engineering, University of Virginia, Center for Applied Biomechanics, 4040 Lewis and Clark Drive, Charlottesville, VA 22911, USA. Electronic mail: mbp2q@virginia.edu

translational kinematics may be good indicators of head impact severity and skull fracture,[30] rotational kinematics have been shown to be better predictors of brain deformation and injury.[6,9,14,19,38] Many early brain injury criteria that were formulated using rotational head kinematics also included translational parameters. These metrics were based on a combination of either resultant, or directionally dependent linear and angular head accelerations,[18,27,28,31,36] while some included empirically derived combinations of angular velocity or acceleration, and head injury criterion (*HIC*).[10,19] As experimental and computational evidence mounted supporting rotational head motion as a brain injury mechanism, translational parameters were excluded from the mathematical form of the metric, and brain injury criteria were based solely on rotational kinematics.[16,36,38,47] Although numerous brain injury metrics have been proposed, no metric has been universally accepted as valid for a broad range of head impacts.[6]

Tissue-based metrics are desirable for assessing brain injury risk since they are a measure of the primary injury mechanism. They are based on tissue-level measurement of mechanical parameters including strain, which is believed to a primary mechanism for concussion and diffuse axonal injury (DAI).[23,38,40] Despite the appeal of strain-based metrics for brain injury assessment, direct measurement of brain strain is complicated,[11] and not currently possible with existing anthropometric test devices (ATDs). Instead, anatomically detailed finite element (FE) models of the human head are used to obtain brain strain during head impact, some examples are cited.[19,22,37] Although these measurements are simulation-based, they are used extensively in TBI-related biomechanics research.

While kinematic-based metrics are favored for their simplicity and ease of measurement, they consist of information related to head motion only. Strain-based metrics take into account the spatial and temporal response of the brain during impact; however, at the expense of higher computational cost.[7,15] Thus, the ideal metric should be able to predict brain strain response like an FE model, but with the simplicity and information of a kinematic-based metric.[5] Several studies have focused on developing brain injury criteria using correlations with FE-brain strain responses.[19,38,47] For example, the brain injury criterion (*BrIC*) was developed through correlations between a metric that uses the maximum directionally dependent head angular velocity magnitudes and two separate strain-based metrics from a FE brain model.[38] This hybrid methodology combines FE model accuracy with kinematic metric simplicity to establish risk assessment models that are more relevant for tissue-level strain prediction. Although several kinematic-based metrics have shown high correlation with strain-based responses using subsets of data, their performance over a broader range of head impacts was found to be limited.[6,7]

Limitations with existing kinematic-based metrics for predicting brain strain response have been elucidated. A recent study found that metrics based on rotational head kinematics correlated better with brain strain responses than those based on translational kinematics, over a broad range of impacts.[6] This particular finding was not surprising given that brain tissue is nearly incompressible, and shear strains are caused primarily by rotational head motion.[14] Furthermore, metrics based on angular velocity had the highest correlation out of those evaluated; however, their performance was limited in several impact conditions, namely in long duration events.[6] These findings agree with previous experimental and computational studies that have shown brain deformation dependence primarily on angular velocity.[1,11,12,19,38] Another recent study showed remarkable similarity between the deformation patterns of a single-degree-of-freedom (sDOF) model and two FE brain models.[7] This suggests that maximum brain strain under rotational head motion can be adequately represented by maximum deformation from a second-order mechanical system under excitation, a finding also supported by several experimental and computational studies.[14,23,47] Specifically, maximum brain strain depended on the magnitudes of angular velocity and angular acceleration, and their relationship to the brain's natural period.[7] While correlations between maximum brain strain and metrics based on a combination of angular velocity and angular acceleration were higher than those based on velocity or acceleration alone, they do not relate rotational head kinematics to brain deformation on a fundamental mechanics level. Thus, a new metric that represents brain deformation and includes both angular velocity and acceleration is recommended.

This study focuses on the development of a new kinematic-based metric called the universal *BrIC* (*UBrIC*), which is formulated based on the governing relationship between excitation and maximum deformation of a second-order system. The universal term of *UBrIC* represents the applicability of this metric to a broad range of impact conditions including automotive and sport. Efficacy of *UBrIC* for predicting strain-based responses from FE brain models was investigated relative to existing kinematic-based metrics using a broad range of head impacts. This study highlights the advantage of a physics-based metric for predicting brain strain response using head kinematics in a broad range of impact conditions, and provides a tool for assessing the safety of helmets and automotive countermeasures.

MATERIALS AND METHODS

Development of UBrIC

UBrIC is based on the assumption that maximum brain deformation under rotational head motion is analogous to deformation from a second-order system under excitation. In a previous study, a sDOF model was used to show that brain deformation in one dimension is governed by three general categories of rotational head motion, each distinguished by the pulse duration (Δt) relative to the natural period (Δt_n) of the brain–skull system: for short-duration pulses, maximum brain strain depended primarily on the magnitude of angular velocity (Fig. 1a, $\Delta t \rightarrow \Delta t_1$), for long-duration pulses, maximum brain strain depended primarily on the magnitude of angular acceleration (Fig. 1a, $\Delta t \rightarrow \Delta t_2$), and for pulses near the natural period of the brain (36–45 ms), maximum strain depended on the magnitudes of velocity and acceleration (Fig. 1a, $\Delta t \rightarrow \Delta t_n$).[7]

To generalize the transition between velocity and acceleration dependent brain strains for a one dimensional impact pulse, exponential functions were used (Fig. 1b). Adding these exponentials resulted in a function that switches between velocity and acceleration dependent deformations in a manner creating a velocity-only dependence in short-duration ($f \rightarrow \omega$ as $\Delta t \rightarrow 0$), and acceleration-only dependence in long-duration, ($f \rightarrow \alpha$ as $\Delta t \rightarrow \infty$):

$$f(\Delta t) = \omega\left(1 - e^{-\frac{1}{\Delta t}}\right) + \alpha e^{-\frac{1}{\Delta t}}, \tag{1}$$

where f is a functional that establishes brain deformation given the magnitudes of angular velocity (ω) and angular acceleration (α), and duration of a one dimensional head impact pulse. Assuming that the duration of an arbitrary impact is related to the magnitudes of angular velocity and acceleration, $\Delta t = \omega/\alpha$, and that the one dimensional deformation can be generalized for rotations about each axis of the head, the following kinematic-based metric is proposed:

$$UBrIC = \left\{ \sum_i \left[\omega_i^* + (\alpha_i^* - \omega_i^*) e^{-\frac{\alpha_i^*}{\omega_i^*}} \right]^r \right\}^{\frac{1}{r}}, \tag{2}$$

where ω_i^* and α_i^* are the directionally dependent ($i = x, y, z$) maximum magnitudes of head angular velocity and angular acceleration each normalized by a critical value (cr); $\omega_i^* = \omega_i/\omega_{icr}$ and $\alpha_i^* = \alpha_i/\alpha_{icr}$. The critical values normalize the metric to maximum brain strain from a FE model and control the transition between velocity and acceleration dependent deformations. The exponent r establishes the power at which the magnitude is evaluated; model performance was assessed for r equals one and two. Six total parameters (two critical values per direction) were used to establish the full three dimensional form of *UBrIC*.

Additional Kinematic Forms

In addition to *UBrIC*, several mathematical forms based on existing rotational metrics were assessed for predicting brain strain responses. These metrics are based on the maximum (resultant or directionally dependent) magnitudes of angular velocity and angular acceleration, and were included in the analysis to benchmark improvement using *UBrIC* with data from the current study (Table 1). Metrics based on translational kinematic parameters were not included, since they were shown to have poor correlation with strain-based metrics.[6,38]

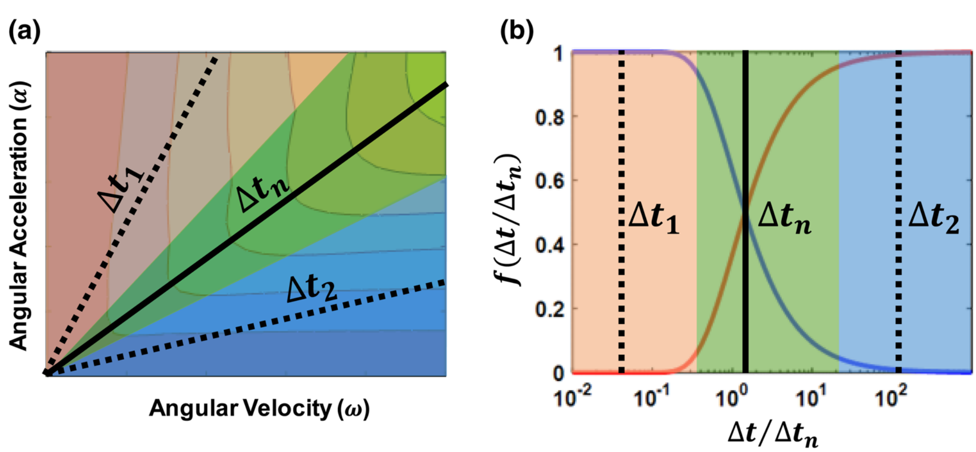

FIGURE 1. Contours of maximum deformation from the sDOF model subjected to a broad range of angular velocity and angular acceleration magnitudes (a), where $\Delta t_1 < \Delta t_n < \Delta t_2$. These contours were obtained from Gabler *et al.*,[7] and share remarkable similarity to the maximum principal strain contours from FE brain models.[7] Exponential functions that were used to generalize the deformation behavior of an arbitrary second order system (b).

Two fundamental kinematic parameters used in brain injury criteria are the maximum values of the resultant (m) angular velocity and angular acceleration:

$$z_{\mathrm{m}}^{*} = \frac{z_{\mathrm{m}}}{z_{\mathrm{mcr}}}, \tag{3}$$

where

$$z_{\mathrm{m}} = \max_{t}\{z(t)\}, \tag{4}$$

and $z(t)$ is a dummy variable for the angular velocity, (t), and angular acceleration, (t), time history vectors of the head. Linear combinations of the directionally dependent maximum magnitudes of angular velocity and angular acceleration were also included:

$$z_{i}^{*} = \left\{\sum_{i} z_{i}^{*r}\right\}^{\frac{1}{r}}, \tag{5}$$

where z indicates ω or α and performance was assessed for r equals 1 and 2. Equation (5) with $z = \omega$ and $r = 2$ represents *BrIC*, which was proposed by Takhounts *et al.*[38] In this study, Eq. (5) is referred to as *BrIC* (refit) since the critical values of *BrIC* for were refit using data from the current study to compare it equally with *UBrIC*. The original form of *BrIC* was also included in the analysis where $\omega_{\mathrm{xcr}} = 66.25$, $\omega_{\mathrm{ycr}} = 56.45$, and $\omega_{\mathrm{zcr}} = 42.87$ rad s^{-1}.[38]

Metrics that consisted of angular velocity and angular acceleration were also included in the current study. Mathematical forms based on a linear combination of the maximum resultant [Eq. (6)] and directionally dependent [Eq. (7)] magnitudes were investigated:

$$C_{\mathrm{m}} = \omega_{\mathrm{m}}^{*} + \alpha_{\mathrm{m}}^{*}, \tag{6}$$

and

$$C_{i} = \left\{\sum_{i} \left[\omega_{i}^{*} + \alpha_{i}^{*}\right]^{r}\right\}^{\frac{1}{r}}, \tag{7}$$

where subscripts m and i indicate resultant and directionally dependent formulations for the combination (C), respectively; Eq. (6) is based on a precursor to *BrIC*.[36]

Head Impact Data

A database of 1595 head impacts with complete 6DOF head kinematic time histories was assembled and used to fit the critical values of the metrics in Table 1. The database consists of head impacts from automotive sled and crash tests using dummies and cadavers,[3,6] impactor tests performed at multiple locations on both helmeted[42] (American football) and un-helmeted[6] dummies, and head motions from human volunteer response to sled acceleration.[32] The sled, crash, and un-helmeted impactor tests were used previously to evaluate kinematic-to-strain correlations,[6] and are based on US government crash tests. Of the more than 1800 helmeted impactor tests conducted by Viano *et al.*, 2012, a subset of 576 impacts at 5.5, 7.4, and 9.3 m s^{-1} at eight locations on 24 helmets was included in the current study.[42] Helmet impact locations and speeds were based on video analysis of professional football; where 9.3 and 7.4 m s^{-1} are the average and -1 standard deviation on-field closing speeds for concussion, respectively. The human volunteer data were obtained from a study which included 335 separate tests involving twenty volunteers subjected to low-level sled acceleration.[32] These tests provide a means of assessing kinematic metric sensitivity to longer duration head motion, which may be relevant for evaluating improved countermeasure designs. Test conditions used in the current study are provided in Table 2.

Timed 6DOF head kinematics were obtained in a local coordinate system defined by the anatomical axes of the head with an origin fixed at the CG.[45] The data were processed based on methods used previously by Gabler et *al.*[6] Angular acceleration magnitudes were calculated by taking the maximum resultant or directionally dependent magnitudes evaluated over the entire event time history [Eq. (4)]. Angular velocity

TABLE 1. Kinematic-based metrics included in the current study.

Metrics	# of critical values (k)	Directional dependence	Angular velocity type	Values for r	Total # of models
ω_{m}^{*} [Eq. (3)]	1	No	NA	NA	1
α_{m}^{*} [Eq. (3)]	1	No	NA	NA	1
C_{m} [Eq. (6)]	2	No	NA	NA	1
ω_{i}^{*} [Eq. (5)][a]	3	Yes	Peak, p2p	1, 2	4
α_{i}^{*} [Eq. (5)]	3	Yes	NA	1, 2	2
C_{i} [Eq. (7)]	6	Yes	Peak, p2p	1, 2	4
UBrIC [Eq. (2)]	6	Yes	Peak, p2p	1, 2	4

NA not applicable.
[a]*BrIC* (refit) is $r = 2$.

TABLE 2. Summary of impact conditions included in the current study.

Impact category (sample size)	Impact condition (sample size)	Test details (sample size)
Sled tests[3,6,32] (445)	Frontal (FRT) (169)	UVa driver and rear seat occupant (37)
		NBDL volunteer 0° rearward acceleration (132)
	Oblique (OBL) (128)	UVa 20° and 60° far side oblique (31)
		NBDL volunteer 45° oblique acceleration (97)
	Side (SID) (120)	UVa 90° far side pure lateral (14)
		NBDL volunteer 90° lateral acceleration (106)
	Pedestrian (28)	UVa vehicle buck laterally into pedestrian
Crash tests (381)[6]	Frontal (FRT) (238)	NHTSA NCAP driver and front/rear passenger (87)
		IIHS small and moderate overlap impact (151)
	Oblique (OBL) (54)	NHTSA R&D near side driver and rear passenger
	Side (SID) (89)	NHTSA NCAP near side driver and rear passenger (68)
		IIHS near side driver and rear passenger (21)
Impactor tests (769)[6,42]	Frontal (74)	NHTSA 0° pendulum into head CG/forehead
	Oblique (78)	NHTSA 30°/60° pendulum into head CG/forehead
	Side (41)	NHTSA 90° pendulum into head CG
	A (72)	Biokinetics oblique facemask
	B (72)	Biokinetics upper oblique facemask
	C (72)	Biokinetics side of facemask
	D (72)	Biokinetics rear boss of shell
	F (72)	Biokinetics front of shell
	R (72)	Biokinetics rear of shell
	UT (72)	Biokinetics side of shell
	AP (72)	Biokinetics lower central facemask

NHTSA impactor tests were unhelmeted[38]; biokinetics tests were helmeted.[42]
NBDL Naval Biodynamics Laboratory, *NCAP* new car assessment program, *IIHS* Insurance Institute for Highway Safety, *R&D* research and development.

TABLE 3. Descriptive statistics for maximum resultant and directionally dependent kinematic parameters.

	ω_m (rad s^{-1})	$\omega_x{}^a$ (rad s^{-1})	$\omega_y{}^a$ (rad s^{-1})	$\omega_z{}^a$ (rad s^{-1})	$\omega_x{}^b$ (rad s^{-1})	$\omega_y{}^b$ (rad s^{-1})	$\omega_z{}^b$ (rad s^{-1})	α_m (krad s^{-2})	α_x (krad s^{-2})	α_y (krad s^{-2})	α_z (krad s^{-2})	Δt_x (ms)	Δt_y (ms)	Δt_z (ms)
Median	32.1	15.2	21.1	10.9	21.1	29.6	16.3	2.83	1.14	1.65	0.988	15.2	16.3	14.5
IQR	17.8	21.8	19.3	12.6	26.7	28.1	20.2	2.58	2.11	1.95	1.36	14.5	14.8	15.5

$\Delta t_i = \omega_i/_i$ is an approximation used for calculating the directionally dependent duration of an arbitrary pulse.
IQR interquartile range.
[a]Values based on peak angular velocity [Eq. (4)].
[b]Values based on peak-to-peak angular velocity [Eq. (8)].

magnitudes were calculated two separate ways by taking: (1) the maximum magnitudes [Eq. (4), *peak* model], and (2) the difference between maximum and minimum values [Eq. (8), *peak-to-peak*; *p2p* model] in each anatomical direction over the entire event:

$$\omega_i = \max_t\{\omega_i(t)\} - \min_t\{\omega_i(t)\}. \quad (8)$$

The p2p model was included since deformation from a second-order system depends on the integral of the pulse (maximum velocity change) in short duration.[43] Descriptive statistics for kinematic parameters from the database are listed in Table 3.

Strain-Based Metrics

Two previously developed strain-based metrics were used: maximum principal strain (MPS) and cumulative strain damage measure (CSDM). These metrics are based on FE model brain strains and have been used extensively in computational TBI research.[6,19,38] MPS is the maximum value of MPS occurring over all brain elements over the entire event time history.[37] Recently, studies have used the 95th percentile-ranked maximum element value to avoid numerical issues that may be associated with the 100th percentile element.[6,29] In this study, model fits were performed using the 95th and 50th percentile ranked brain MPS. The 50th percentile MPS was included as an average measure of global brain deformation, whereas the 95th percentile MPS may be more indicative of localized brain injury. CSDM is the cumulative volume fraction of elements that incur MPS that exceeds a predefined threshold; 25% element MPS was used for CSDM since it was

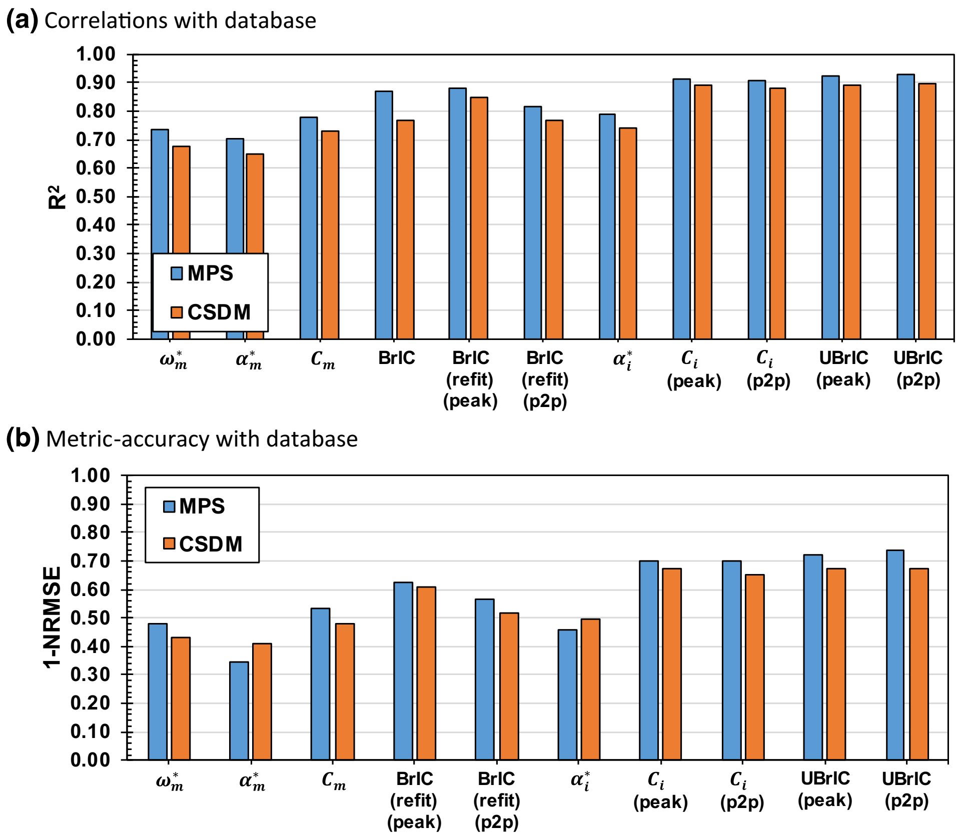

FIGURE 2. Correlations (a) and metric-accuracy (b) for kinematic-to-strain metric fits from GHBMC using the database of 1595 head impacts. Results are shown for metrics with $r = 2$ only. MPS is based on the 95th percentile value. *BrIC* was not included in the metric-accuracy assessment because its values do not correspond to MPS and CSDM from GHBMC.

best indicator of DAI based on a survival analysis using scaled animal data.[38]

FE Model Simulations

Strain-based metrics were obtained using two FE human brain models: The Global Human Body Models Consortium-owned (GHBMC) 50th percentile male (M50) detailed seated occupant (v4.3) head and the simulated injury monitor (SIMon, v4.0), a 50th percentile male human head-only model developed and distributed by the National Highway Traffic Safety Administration (NHTSA). Both FE models have been validated for intracranial responses, including relative brain–skull motion, and were recently used with a subset of the current database.[6,32] Details on the development and validation of both FE models is available in the literature.[22,37] Simulations were performed by applying the timed 6DOF head kinematics directly to the FE skull, which was rigidly attached to a local coordinate system that was consistent with the head impact data.[45] FE simulations were performed using LS-DYNA (v971 R7.1.1, double precision; LSTC, Livermore, CA).

Statistical Methods

Critical values for the kinematic-based metrics were determined through fits to strain-based metrics obtained from FE simulation of the 1595 head impacts. A non-linear, least-squares solver (lsqcurvefit; Matlab, v8.4.0, The MathWorks, Natick, MA) was used to minimize the sum squared error (*SSE*) between kinematic metric-predicted and FE model-measured MPS and CSDM. For CSDM-based fits, an intercept parameter β_o was added to Eqs. (3) and (5)–(7), and included in the fit. The intercept was used to correct for cases in which non-zero head kinematic responses resulted in zero FE model-based CSDM due to the MPS threshold; metrics were constrained to positive values only, e.g., if $UBrIC < 0$, then $UBrIC = 0$. A total of 102 model fits were performed; 17 kinematic-based metrics fit to 3 strain-based metrics from two FE brain models.

Model fits were assessed using the coefficient of determination ($\tilde{R}^2$) and the normalized root mean square error (*NRMSE*). Correlations between kinematic metric-predicted ($\hat{y}$) and FE model-measured (y) MPS and CSDM values were adjusted (R^2) for the number of critical values (k) and samples (n) used[24]:

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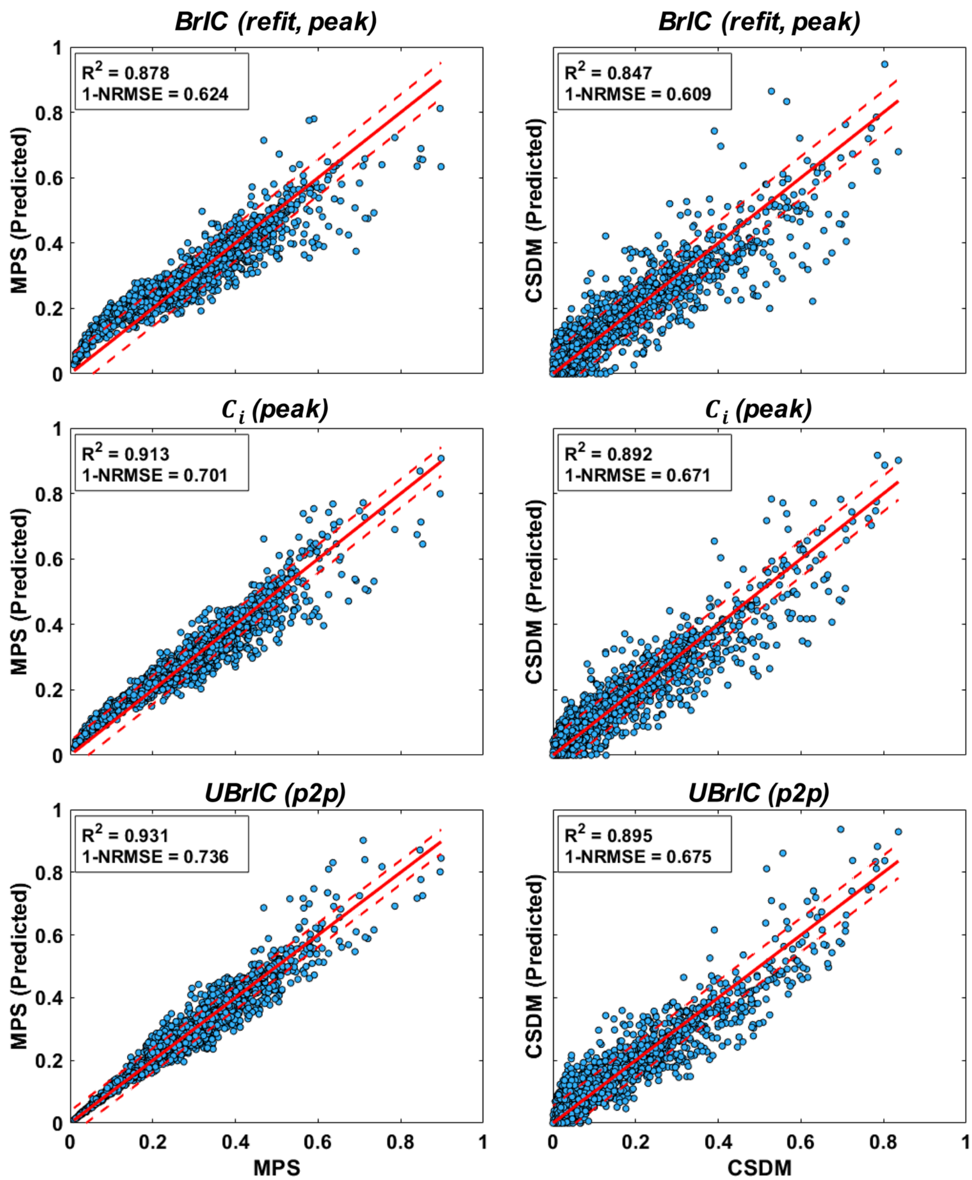

FIGURE 3. Scatter plots for top performing fitted metrics to MPS (left column) and CSDM (right column) using the database of 1595 head impacts. MPS is based on the 95th percentile value. Solid red lines indicate a one-to-one relationship, while dotted red lines are ± 1 root mean square error. Results shown are for metrics with *r* = 2.

TABLE 4. Critical values for top performing kinematic metrics determined from fits to strain-based metrics from GHBMC (n = 1595).

Metrics		β_o	ω_{xcr} (rad s^{-1})	α_{xcr} (krad s^{-2})	ω_{ycr} (rad s^{-1})	α_{ycr} (krad s^{-2})	ω_{zcr} (rad s^{-1})	α_{zcr} (krad s^{-2})	R^2	*SSE*	1 − *NRMSE*
UBrIC	MPS	0	211	20.0	171	10.3	115	7.76	0.931	2.36	0.736
(p2p) [Eq. (2)]	CSDM	− 0.275	117	17.7	119	7.03	85.8	6.45	0.895	4.66	0.675
C_i (peak) [Eq. (7)]	MPS	0	293	43.0	182	34.0	123	30.4	0.913	3.04	0.701
	CSDM	− 0.327	181	28.6	96.7	31.9	80.7	25.8	0.892	4.80	0.671
BrIC (refit, peak)	MPS	0	163	–	123	–	89.0	–	0.878	4.80	0.624
[Eq. (5)]	CSDM	− 0.358	99.9	–	71.5	–	58.9	–	0.847	6.78	0.609

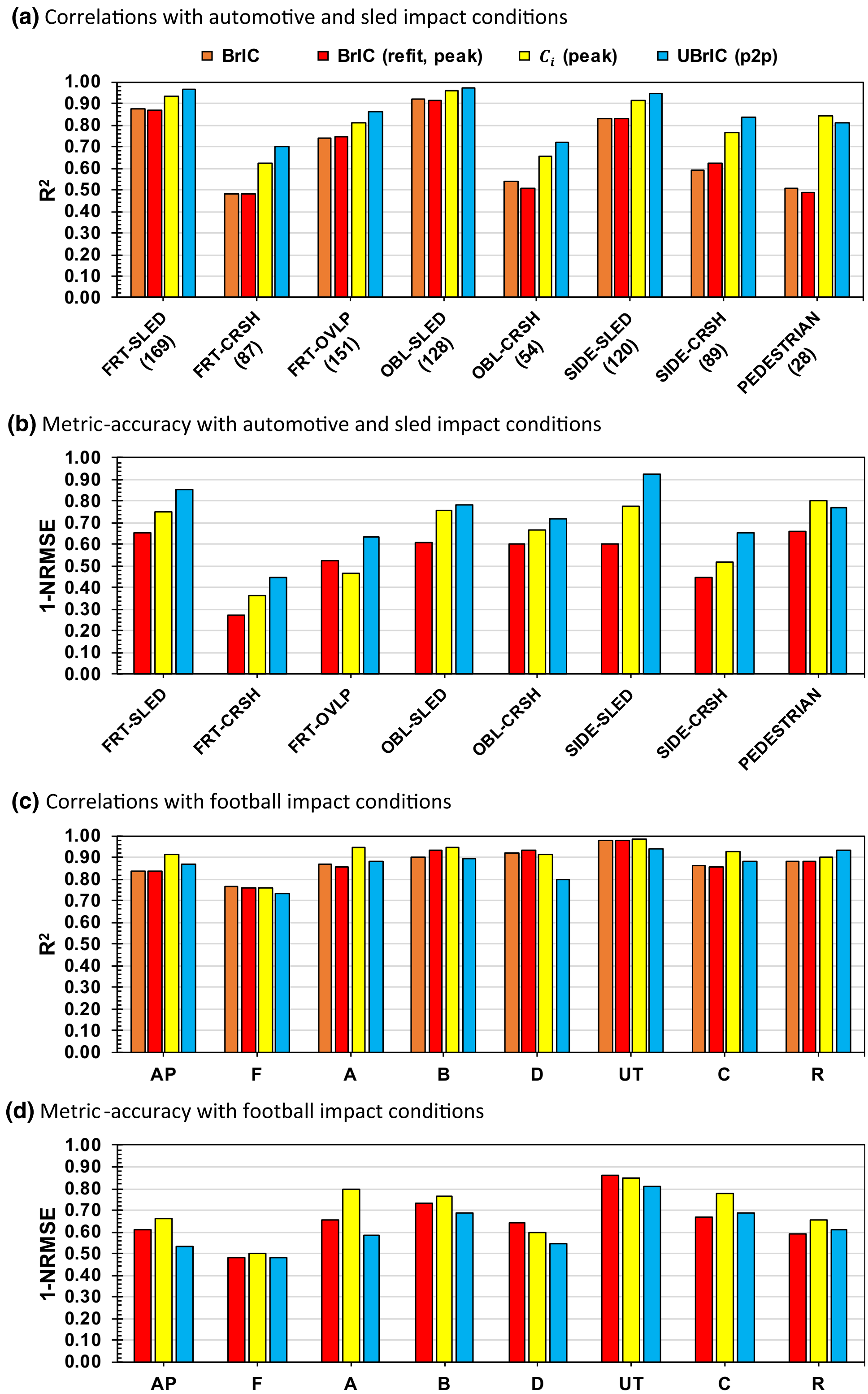

FIGURE 4. Performance of top performing fitted metrics and *BrIC* assessed by impact condition based on correlations (a, c) and metric accuracy (b, d) for automotive and sled (a, b), and football (c, d) impact conditions. Results are evaluated for metrics with $r = 2$, and relative to MPS (95th percentile) from GHBMC. Small and moderate overlap conditions (FRT-OVLP) were combined and assessed independently from the full frontal crash (FRT-CRSH) test mode. *BrIC* was not included in the metric-accuracy assessment because its values do not directly correspond to MPS and CSDM from GHBMC. Sample sizes for the sled and crash modes are shown in parenthesis (a, b), there were 72 tests for each football impact locations (c, d).

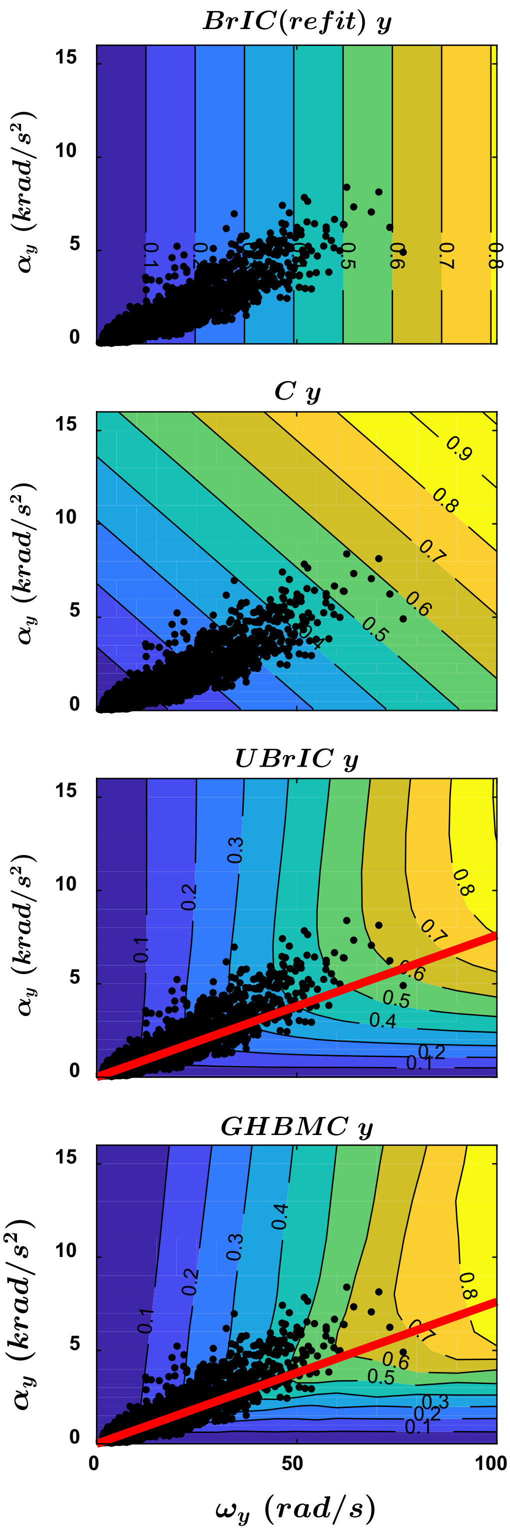

FIGURE 5. Comparison of MPS contours predicted by each of the top performing kinematic-based metrics (top three graphs) relative to MPS from the GHBMC FE model (bottom graph). Contour lines represent constant levels of MPS. Results shown are for rotational motion in the sagittal plane (*y* direction); the maximum magnitude of angular velocity (peak) and angular acceleration about the *y* axis from the database of 1595 head impacts is overlaid atop the contours (black circles). FE model results were obtained from Gabler *et al.*,[7] and reproduced in the current study. Solid red lines indicate the ratio of $\alpha_{ycr}/\omega_{ycr}$ which is inversely related to the effective one dimensional critical pulse duration (Δt_{ycr}); this line was obtained using *UBrIC* and overlaid onto the FE model contours.

$$R^2 = 1 - \frac{(1-\tilde{R}^2)(n-1)}{(n-k-1)}. \tag{9}$$

Since correlations can only be used to evaluate association between metrics, the $NRMSE$ was used to assess the accuracy of the kinematic metric-prediction of MPS and CSDM:

$$NRMSE = \sqrt{\frac{MSE(\hat{y}_j)}{MSE(y_t)}}, \tag{10}$$

where

$$MSE(u) = \frac{\sum_{j=1}^{n}(y_j - u)^2}{(n-k-1)} \tag{11}$$

and u is a dummy variable for the jth kinematic metric-prediction, $\hat{y}_j$, and the null model, y_t, which is the average FE model-based MPS or CSDM used in the fit. The $NRMSE$ is a statistical metric used to evaluate the predictive performance of a regression model relative to a minimum information model (null model; $k = 0$).[21] In this study, $1 - NRMSE$ values are reported for consistency with R^2, where higher values indicate better performance.

Model Performance Assessments

The relative performance of the fitted kinematic-based metrics was evaluated in multiple ways: First, an overall assessment was performed by comparing correlations (R^2) and metric-accuracy ($1 - NRMSE$) from fits using the full head impact database ($n = 1595$) and the metrics in Table 1. Based on this assessment, the top performing metrics were selected and compared using the *impact conditions* (Table 2). Next, idealized rotational head motions from Gabler *et al.*, 2017 were used to qualitatively assess response patterns between the top metrics and strain-based responses.[7] Finally, head impacts from NHTSA's moving deformable barrier (MDB) oblique ($n = 130$) and full frontal rigid barrier ($n = 22$) crash test modes were used as an independent (not used for fitting) dataset for assessing the relative performance between metrics.[34] A total of 152 head impacts including driver

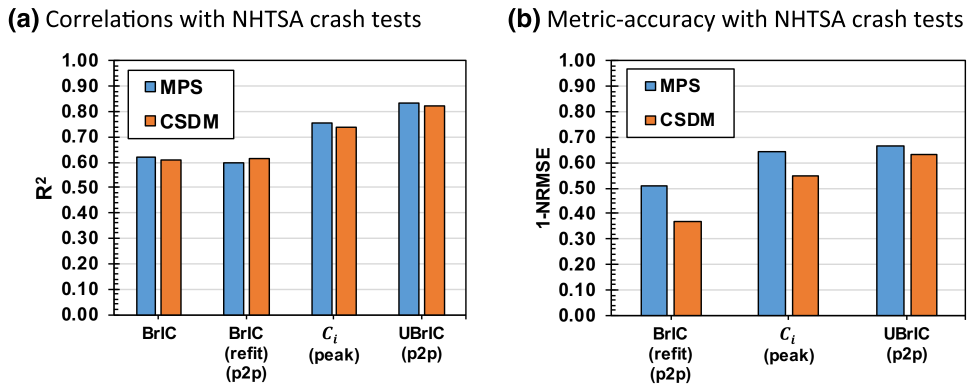

FIGURE 6. Performance of top performing fitted metrics and BrIC assessed for correlation (a) and metric accuracy (b) using NHTSA's MDB oblique and full frontal rigid barrier crash tests; 152 total head impacts were used for this assessment. Results are evaluated for metrics with $r = 2$, and relative to MPS (95th percentile) from GHBMC.

and passenger dummies were obtained from NHTSA's vehicle crash test database[46]; specific tests are listed in the Electronic Supplementary Material.

RESULTS

Overall Assessment

Relative to other metrics, *UBrIC* had the best overall performance with the database; for the p2p model with $r = 2$, $R^2 = 0.931$, 0.895 and $1 - NRMSE = 0.736$, 0.675 for MPS and CSDM, respectively were the highest reported (Fig. 2). From the additional forms considered, C_i with $r = 2$ was among the top performing metrics; R^2 and $1 - NRMSE$ were similar, but slightly lower than *UBrIC*. Fits with *BrIC* (refit) were better than the directionally dependent angular acceleration form, α_i^*, but were lower than *UBrIC* and C_i. Furthermore, *BrIC* (refit) and C_i systematically over-predicted strain-based responses for lower severity impacts (e.g., MPS; Fig. 3 top and middle left graphs). In general, metrics with $r = 2$ performed better than $r = 1$, while peak angular velocity forms performed better than those based on the p2p except for *UBrIC*. Metrics based on resultant kinematics were lower than directionally dependent forms; however, $R^2 > 0.6$ for all model fits (Fig. 2a). When compared to CSDM-based fits, correlations and metric-accuracy were higher with MPS-based metrics, while fits using the 50th percentile MPS were better than the 95th for nearly every metric; e.g., $R^2 = 0.941$ and $1 - NRMSE = 0.756$ for *UBrIC* ($r = 2$, p2p). Compared to GHBMC, model performance was slightly better using SIMon-based metrics; $R^2 > 0.65$ for all fits; however, the relative performance of the kinematics-based metrics was similar. Critical values for top performing metrics: *UBrIC*, C_i, and *BrIC* (refit) based on GHBMC are provided in Table 4. Critical values and results based on SIMon are provided in the Electronic Supplementary Material.

Assessment by Impact Condition

Top performing metrics include the p2p form of *UBrIC* and peak forms for C_i, and *BrIC* (refit), each with $r = 2$. These three metrics were compared with the original BrIC using the *impact conditions* (Table 2). Relative to *BrIC* and *BrIC* (refit), *UBrIC* and C_i performed better in the sled and crash test conditions; *UBrIC* was the top performing metric in nearly every mode (Figs. 4a and 4b). In particular, metric-accuracy in frontal (FRT), oblique (OBL), and side (SID) sled conditions were higher with *UBrIC*; $1 - NRMSE > 0.90$ for *UBrIC* in SID-SLED was the highest observed (Fig. 4b). Compared to the sled and crash test modes, performance in football impact conditions were more consistent among the metrics. While *BrIC* and *BrIC* (refit) performed better than *UBrIC* in several impact locations, C_i tended to perform better in the majority of locations (Figs. 4c and 4d). Relative model performance based on CSDM and the 50th percentile MPS were generally consistent with the 95th percentile MPS.

Assessment Using Idealized Rotational Head Motions

When compared to C_i and *BrIC* (refit), the relationship between *UBrIC*, angular velocity, and angular acceleration more closely matched the response patterns of the sDOF and FE models that were previously reported by Gabler *et al.*[7] (Fig. 5). The contours of *BrIC* and C_i were vertical and diagonal, respectively, while the contours of *UBrIC* were more akin to the FE model; vertical for shorter duration pulses and horizontal for longer duration pulses with the transition occurring near the resonance frequency of the brain–skull system.

Assessment Using NHTSA Crash Tests

Correlations and strain-based metric predictions were better using *UBrIC* with head impacts from the oblique MDB and full frontal crash test modes (Fig. 6). Using all 152 head impacts, correlations and metric accuracy were lower with *BrIC* and *BrIC* (refit) compared to *UBrIC* and C_i; $R^2 \approx 0.6$ and $1 - NRMSE \leq 0.51$ for BrIC-based metrics, which was approximately 0.2 and 0.15 lower for R^2 and $1 - NRMSE$, respectively than *UBrIC* and C_i.

DISCUSSION

Deformation is believed to be the primary mechanism for brain injury, and rotational head motion is the primary mechanism for brain deformation. However, existing kinematic-based metrics used in brain injury assessment do not represent brain strain over a broad range of head impacts. In this study, a new kinematic-based metric (*UBrIC*) was developed to predict strain-based responses (MPS and CSDM) from FE brain models using the directionally dependent magnitudes of angular velocity and acceleration from a head impact. *UBrIC* was formulated based on the governing relationship between maximum deformation and excitation from a second-order system, which was used as an analogue for brain deformation to rotational head motion. The critical values of *UBrIC* were determined through fits to strain-based metrics (MPS and CSDM) obtained from FE simulation of nearly 1600 head impacts in two different brain models. Efficacy of *UBrIC* for predicting strain-based response was assessed by comparing to fits using kinematic metrics with mathematical forms based on existing brain injury criteria. Comparisons were made through several assessments involving both real world and idealized head motions.

Currently, *UBrIC* predicts MPS and CSDM obtained from the GHBMC and SIMon FE brain models, and has not been normalized to brain injury risk. Although brain injury criteria for MPS and CSDM were previously developed,[38] we do not recommend using these injury risk functions with *UBrIC*. The risk functions developed by Takhounts *et al.*, 2013 were fitted using the 100th percentile MPS from SIMon.[38] The current study uses the 95th percentile MPS to avoid potentially spurious values generated using the maximum element. Furthermore, laboratory tests using human volunteers and field analyses involving crash reconstructions have shown that the MPS-based risk functions over-estimate brain injury risk.[20,25,32] While the CSDM-based risk functions were more accurate when predicting non-injurious response,[32] their effectiveness at higher severity head impacts has not been assessed. Until the existing strain-based risk functions can be fully verified using human injury data, or new risk functions developed using metrics from the current study, *UBrIC* should only be used for discriminating the relative severity between head impacts. This current limitation does not reduce the utility of *UBrIC* as it can still be used to inform design similar to how the severity and *HIC* have driven improvements in helmet and automotive safety for decades.[8,41]

Two separate approaches were used for evaluating the angular velocity magnitude of a head impact signal; peak and p2p. When evaluated relative to the peak model, the p2p version of *UBrIC* had a better fit to the overall database, and was more accurate in a majority of the impact conditions. Interestingly, the performance of *BrIC* (refit) and C_i was generally worse using the p2p forms. Given that real world impacts typically have more complicated time histories that can result in non-zero impact restitution, we recommend using the p2p *UBrIC* model based on GHBMC for predicting strain-based metrics (Table 4). We also recommend using the *UBrIC* formulation with $r = 2$, since overall model performance was slightly better than $r = 1$, and to use the methods described herein for obtaining angular velocity and angular acceleration magnitudes. Use of parameters calculated in a different manner may substantially affect the prediction of strain-based metrics. Although direct measurement of head kinematics is preferred, this procedure has the advantage of obtaining angular accelerations without the high cost of deploying additional sensors in existing ATDs.

In previous studies the rotational velocity change index (*RVCI*),[47] a kinematic metric based on angular velocity change, performed well relative to existing rotational metrics,[6] and exhibited similar response patterns to the mechanical models.[7] Correlations between *RVCI* and strain-based responses from the current database were high; $R^2 = 0.821$ and 0.776 for MPS and CSDM, respectively. However, calculation of *RVCI* is more complex compared to *UBrIC*, and involves maximizing the time history integral of angular acceleration in a manner similar to *HIC*. Pre-computed atlases have also been used for estimating FE model brain strain response to impact.[15] While this technique allows for whole brain strain computation, and is a cost-effective alternative to FE simulation, pre-computed values for real-world impacts are based on interpolations that are not based on brain deformation mechanics. Thus, the accuracy of this technique should be investigated for a broader range of head impacts involving longer duration and complex pulse shapes. Furthermore, a pre-computed atlas is a black-box function, which makes it difficult for engineers to understand how to manage the trade-offs between acceleration, velocity, and duration when designing a countermeasure.

While several kinematic metrics performed well with the overall database, *UBrIC* was a better predictor of strain-based responses in most impact conditions. Compared to other metrics, *UBrIC* performed better in nearly all of the automotive and sled conditions. This finding was anticipated, since impacts from these conditions were typically longer in duration; where brain deformation becomes more dependent on angular acceleration.[7] However, improvement in the relative performance of *UBrIC* in football related impacts was not as marked. This finding was also anticipated, since direct head impacts typically result in shorter duration pulses where brain deformation response is proportional to angular velocity.[7] While *BrIC* (refit) and C_i performed well in these shorter duration conditions, the accuracy of *UBrIC* was generally better in the majority of impact conditions. Furthermore, the contours of *UBrIC* were more similar to the mechanical models when compared to the other metrics (Fig. 5). Although *BrIC* (refit) and C_i may be sufficient for predicting strain-based responses in some impact conditions, their applicability is limited to specific regimes of loading, i.e., *BrIC* can be used with shorter duration impacts, while C_i can be used with moderate duration impacts; *UBrIC* can be used with impacts of all duration.

Relative to *UBrIC*, *BrIC* (refit) systematically overestimated MPS and CSDM for low-to-moderate severity head impacts. These impacts were primarily from sled conditions, and included human volunteer and far side tests. When compared to the crash and impactor data, these cases typically had lower acceleration (higher duration), since direct head contact was either mild or did not occur. This finding suggests that the mathematical form of *BrIC* is insufficient for predicting strain-based responses for head impacts covering a broad range of durations, and thus angular acceleration should be included to improve prediction in higher velocity, lower acceleration (longer duration) events. By re-tuning the critical values of *BrIC* through fits to the current dataset the overall correlation with brain strains was improved; however, without angular acceleration, a metric based on angular velocity alone will lead to inaccurate strain predictions in certain head impact conditions.

An additional concern with the use of metrics based only on angular velocity, is their potential insensitivity to improved safety countermeasures. For example, in several of the occupant and pedestrian crash tests used in the current study high levels of head angular velocity were achieved prior to head contact with a hard surface. By eliminating head contact through improvements made to safety countermeasures, one could reduce the magnitude of head angular acceleration without substantially changing the magnitude of angular velocity. Thus, a criterion based on angular velocity alone could be insensitive to an improved countermeasure design, and could potentially inhibit innovation. With current efforts focused on improving head safety systems, the changing landscape of countermeasure technology will likely test the limitations of existing metrics by potentially pushing their use into regimes where they are less accurate, and hence may not be able to affect injury countermeasures as is intended.

Limitations

The ability of *UBrIC* to predict brain injury relies heavily on the accuracy of the FE models. Although GHBMC and SIMon have been validated for brain deformation, the head kinematics and brain strain responses for some cases used in this study fall outside the range of experimental data used to validate these models.[6] Unfortunately, data for validating brain FE model is extremely limited to due challenges associated with human testing. Thus, future studies should focus on verifying the accuracy of FE model brain deformations over a broader range of head impacts. Furthermore, strain-based metrics used for fitting the critical values of *UBrIC* are based on global measures of maximum brain deformation. Strain rate, the product of strain and strain rate, fiber-oriented and region specific strain have also been proposed as brain injury predictors.[2,17,35] While these studies are encouraging, additional work is needed to determine whether incorporating these characteristics improves brain injury prediction in humans.

Brain injuries due to skull fracture or focal bleeding were not considered in the development of *UBrIC*. These injuries occur under high acceleration, short duration head impacts,[30] which may cause highly localized strains within the head. In the current study, FE simulations were performed using a rigid skull to apply the 6DOF head kinematics. There were several pedestrian tests involving human cadavers; none of which sustained a skull fracture despite achieving the highest recorded *HIC* values.

CONCLUSIONS

A new kinematic-based brain injury metric, *UBrIC*, was developed based deformation response from a second order system, which was used as a mechanical analogue for maximum brain deformation to rotational head motion. *UBrIC* uses the directionally dependent magnitudes of head angular velocity and angular acceleration to directly calculate strain-based responses (MPS and CSDM) from FE brain models. Nearly 1600 head impacts covering a broad range of human response to impact were collected and simulated in two different FE models to obtain strain-based metrics for fitting the critical values. Relative to fits using kinematic metrics based on existing brain injury criteria, *UBrIC* was a better predictor strain-based responses in various head impact environments including those seen in automobile crashes and American football. Currently, *UBrIC* can only be used for assessing the relative severity between head impacts, since existing strain-based criteria have not been sufficiently verified using human injury datasets. By using *UBrIC*, equipment manufactures will be able to discriminate the efficacy of improved safety systems that may otherwise not be possible with existing rotational brain injury criteria.

ELECTRONIC SUPPLEMENTARY MATERIAL

The online version of this article (https://doi.org/10.1007/s10439-018-2015-9) contains supplementary material, which is available to authorized users.

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