Development and Analysis of Metrics to Predict 5th Percentile Female Brain Strain Using Rotational Head Kinematics

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Abstract Female occupants have a significantly greater risk for moderate brain injuries in frontal crashes even after controlling for age, height, BMI, model year, and delta-V of the crash. Current brain injury metrics were developed and tuned to capture the brain deformation response of 50th percentile male anthropometries, but use of these male-derived metrics has not been validated for smaller female occupants. The assessed brain injury metrics (BrIC, UBrIC, and DAMAGE), with parameters tuned using a mid-sized male finite element (FE) model, overpredicted the MPS-95 response of a 5th percentile FE brain model. This study proposes three female-derived brain injury metrics (BrIC-F05, UBrIC-F05, DAMAGE-F05), which were developed using the MPS-95 from a 5th percentile female FE brain model across a range of real-world kinematics, to better predict brain deformation for a smaller occupant. The tuned DAMAGE-F05 brain injury metric had the greatest accuracy (1-NRMSE = 0.787) with a high correlation (R2 = 0.969) for a variety of impact conditions. These F05 brain injury metrics can be used with small female anthropometric test devices to best predict brain deformation for smaller occupants.

Keywords Brain biomechanics, brain injury criteria, rotational head kinematics, small female.

I. INTRODUCTION

The Centers for Disease Control documented 2.53 million traumatic brain injury (TBI) related emergency department (ED) visits and 56,800 TBI-related deaths in the United States in 2014 [1]. Additionally, there are an estimated 5.3 million people living with a TBI-related disability in the United States [2]. Motor vehicle crashes account for more than one fifth of the TBI-related hospitalisations and are the most common TBI-related cause of death for 15 to 34 year-olds and those older than 75 years old [1]. While TBIs remain a large concern for all motor vehicle occupants, recent studies have found differences in injury risk for female occupants [3,4]. In a study of frontal tow-away crashes with belted occupants, previous studies concluded that females were at a significantly greater risk for moderate (Abbreviated Injury Scale (AIS) 2-3)) brain injury with an odds ratio (OR) of 1.76, when controlling for delta-V, sex, age, height, BMI, and model year [3]. Similarly, additional studies found female occupants had a greater risk of concussion in frontal crashes while controlling for delta-V, sex, age, and model year [4]. It is not clear what variables contribute to the observed differences in TBI risk between males and females in automotive crashes, but possible factors can be physiological, e.g., sex-based differences in cellular or tissue-level properties or clinical response to TBI, biomechanical, e.g., sex-based differences in neuroanatomy, material properties, or loading conditions, and/or environmental, e.g., sex-based bias in countermeasure effectiveness resulting from the current tools and metrics used to assess safety [5-9]. Additional research is needed in each of these areas.

While there have been substantial recent advancements in the development of brain injury criteria for application to mid-sized male anthropometric test devices (ATDs) and human body models, efforts to develop complementary metrics for application to female ATDs and models have lagged. Existing brain injury metrics can be categorised as kinematic-based or tissue-level-based metrics. Kinematic-based metrics are dependent on the rigid-body motion of the head, and metrics based on a full kinematic time-history and rotational head kinematics have been identified as better predictors of brain strain (the mechanism of injury for diffuse-type brain injury) compared to linear kinematic-based metrics [10-12]. In contrast, tissue-level metrics are dependent

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on the mechanics and deformations observed with a brain finite element (FE) model subjected to impact or prescribed kinematics [11,13]. Ideally, robust kinematic-based brain injury metrics could predict tissue-based brain strains across a variety of loading conditions using the head kinematics measured in ATDs.

Recently developed brain injury metrics, such as the brain injury criterion (BrIC), the universal brain injury criterion (UBrIC), and the Diffuse Axonal Multi-Axis General Evaluation (DAMAGE), combine the kinematic- and strain-based metrics by using rotational head kinematics to predict brain strain with minimal computational cost [10-12]. For each of these three rotational kinematic brain injury metrics, maximum principal strain (MPS) values were calculated using a FE brain model based on a 50th percentile male geometry, but their validity for smaller occupants with smaller brain volumes is unknown. There are several factors that could affect the difference in brain biomechanical response and TBI risk between males and females, but one in particular, intracranial volume, has been shown to affect brain strain response in finite element brain models [14,15]. Using subject-specific male and female neuroanatomies to develop FE brain models and simulating these models using consistent input head kinematics, the resulting MPS increased linearly with increasing intracranial volume [14]. Because female brains are, on average, 12% smaller than male brains and intracranial volume has been shown to correlate with tissue-level injury metrics, it is critical to consider the effect of an FE brain model's total volume when developing brain injury metrics, e.g., BrIC, UBrIC, DAMAGE [14,16]. Additionally, it is important to consider the application of these metrics when applied to anthropometries with brain volumes different than those used in development and validation. As the use of a small female ATDs increase in regulatory vehicle testing, a complementary rotational kinematic-based brain injury metric is needed for vehicle testing and evaluation. Therefore, the goal of this research was to develop rotational kinematic-based brain injury metrics to predict maximum brain strain for a small female using the response from a 5th percentile female FE brain model.

II. METHODS

Small Female Finite Element Brain Model

The Global Human Body Models Consortium (GHBMC) owned 5th percentile female (F05) detailed occupant (v5.1) brain model was used to develop the small female brain injury metrics [17,18]. The FE brain model was removed from the complete human body model and included all the major structures of the brain, e.g., cerebrum, cerebellum, brain stem, and meninges. All head kinematics were applied to the dura part through the head centre of gravity in the local anatomic coordinate system [19]. The 95th percentile maximum principal strain (MPS-95) was the strain-based brain injury metric used throughout this study; this is a commonly used strain value in brain injury metric development to avoid any potential influence of a high strain value from a single element, a possible result of numerical instability [20,21]. MPS-95 has also been demonstrated in other studies to be one of the best predictors of TBI risk [22-24]. All simulations were completed using the LS-DYNA explicit solver (mpp971R9.1.0, double precision, LSTC, Livermore, CA, USA).

Head Kinematics used for Fitting and Assessing Metrics

Two sets of head kinematics were used to fit and assess the proposed female brain injury metrics. The first database consisted of 1,747 six degree-of-freedom (DOF) real-world head kinematics collected from ATDs, post-mortem human subjects (PMHSs), and human volunteers and were recorded during a variety of impact conditions, such as crash tests, sled tests, and helmeted and un-helmeted football impacts. For additional information about the head kinematics within the real-world database, see previous studies completed by [11,12]. Each of these head impact cases were simulated using the GHBMC F05 brain model, and the MPS-95 within the cerebrum and cerebellum was calculated to fit and assess the brain injury metrics. The second database was a multiaxial parametric set of head kinematics composed of uniaxial, biaxial, and triaxial rotational pulses with varying peak angular velocities (10, 20, 30, 40, 50 rad/s) and durations (20, 40, 60 ms), which span the range of kinematics of the real-world kinematic dataset [12]. For additional information about the head kinematics within the parametric database, see previous studies completed by [12,13]. Each of these kinematics was simulated using the GHBMC F05 brain model to develop corresponding MPS-95 values within the cerebrum and cerebellum to be used in parameter fitting.

Development and Assessment of BrIC-F05

The brain injury criterion (BrIC) uses the peak magnitudes of the three head angular velocities to predict the maximum principal strain (MPS) using critical values that were fit based on the MPS response of FE brain models

[10]. BrIC combines input head kinematics experienced during an impact exposure with the tissue-based metric derived from FE brain models:

$$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{xC}}\right)^2}, \tag{1}$$

where ω_i are the maximum magnitudes of the three head angular velocities and ω_{iC} are the critical values calibrated based on the FE brain model response [10]. BrIC was developed based on the MPS results from male FE brain models, specifically the simulated injury monitor (SIMon) and the GHBMC 50th percentile male human body model, but it has not been assessed on MPS values from a smaller brain geometry [10]. For clarity throughout this study, BrIC will refer to the generic, mathematical function which relates angular velocities to MPS-95 (Eq. 1), and BrIC-M50 will refer to the tuned brain injury metric developed using the GHBMC M50 FE brain model response.

For development of a small female BrIC, referred to as BrIC-F05, the fitting and assessment procedures, established previously in [11] to refit BrIC using peak angular velocity, were repeated using the GHBMC F05 brain model [11]. In summary, a subset of the real-world head kinematics database, described above (n = 1595), was used to fit the critical values of BrIC-F05. A non-linear, least squares solver (lsqcurvefit; Matlab 2021b, The MathWorks, Natick, MA) was used to minimise the sum squared error between the predicted kinematic-based metric and the GHMBC F05 MPS-95 simulation results to optimise the critical values for BrIC-F05. The remaining real-world head kinematics (n = 152) were used as an assessment dataset for the optimised BrIC-F05.

Development and Assessment of UBrIC-F05

The universal brain injury criterion (UBrIC) uses a second-order system to establish a relationship between rotational head kinematics and brain deformation. UBrIC also leverages previous research, which concluded maximum brain deformation was primarily dependent on the magnitude of angular velocity or angular acceleration for long-duration and short-duration pulses, respectively [11,13]. To create a metric that includes this transition between velocity and acceleration, UBrIC was formulated using exponential functions and normalised, directionally dependent angular velocities and angular accelerations:

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

where ω_i^* and α_i^* are the directionally dependent maximum magnitudes of angular velocity angular acceleration, normalised using critical values ($\omega_i^* = \omega_i/\omega_{icr}$ and $\alpha_i^* = \alpha_i/\alpha_{icr}$) to calibrate the metric to the MPS result of the FE brain model [11]. Based on the highest performing UBrIC model, UBrIC is calculated with r = 2 [11]. For clarity throughout this study, UBrIC will refer to the generic, mathematical function which relates both angular velocities and accelerations to MPS-95 (Eq. 2), and UBrIC-M50 will refer to the tuned brain injury metric developed using the GHBMC M50 FE brain model response.

For development of a small female UBrIC, referred to as UBrIC-F05, the fitting and assessment procedures, established by [11] to fit UBrIC using peak to peak angular velocity, were repeated using the GHBMC F05 brain model, where peak to peak angular velocity was defined as “the difference between maximum and minimum values in each anatomical direction over the entire event” [11]. Again, a subset of the real-world head kinematics database (n = 1595) was used to fit the critical values of UBrIC-F05. Using the same optimisation protocol as BrIC-F05, a non-linear, least squares solver was used to minimise the sum squared error between the predicted kinematic-based metric and the GHMBC F05 MPS-95 simulation results to optimise the critical values for UBrIC-F05. The remaining real-world head kinematics (n = 152) were used as an assessment dataset for the optimised UBrIC-F05.

Development and Assessment of DAMAGE-F05

The Diffuse Axonal Multi-Axis General Evaluation (DAMAGE), a brain injury metric that includes the directionally dependent angular acceleration time histories to predict the resulting brain strain:

$$DAMAGE = \beta max_t\{|\vec{\delta}(t)|\}, \tag{3}$$

where the DAMAGE metric is the product of the maximum resultant displacement ($\vec{\delta}$) of a three mass-spring-damper systems and a scale factor (β) that relates the predicted maximum resultant displacement to the MPS value from the FE brain model [12]. The equations of motion which describe the damped, coupled three mass-spring-damper system under forced excitation are:

$$\begin{bmatrix} m_x & 0 & 0 \\ 0 & m_y & 0 \\ 0 & 0 & m_z \end{bmatrix} \begin{Bmatrix} \ddot{\delta}_x \\ \ddot{\delta}_y \\ \ddot{\delta}_z \end{Bmatrix} + \begin{bmatrix} c_{xx}+c_{xy}+c_{xz} & -c_{xy} & -c_{xz} \\ -c_{xy} & c_{xy}+c_{yy}+c_{yz} & -c_{yz} \\ -c_{xz} & -c_{yz} & c_{xz}+c_{yz}+c_{zz} \end{bmatrix} \begin{Bmatrix} \dot{\delta}_x \\ \dot{\delta}_y \\ \dot{\delta}_z \end{Bmatrix} + \begin{bmatrix} k_{xx}+k_{xy}+k_{xz} & -k_{xy} & -k_{xz} \\ -k_{xy} & k_{xy}+k_{yy}+k_{yz} & -k_{yz} \\ -k_{xz} & -k_{yz} & k_{xz}+k+k_{zz} \end{bmatrix} \begin{Bmatrix} \delta_x \\ \delta_y \\ \delta_z \end{Bmatrix} = \begin{bmatrix} m_x & 0 & 0 \\ 0 & m_y & 0 \\ 0 & 0 & m_z \end{bmatrix} \begin{Bmatrix} \ddot{u}_x \\ \ddot{u}_y \\ \ddot{u}_z \end{Bmatrix} \tag{4}$$

where [m] is a diagonal matrix of masses, [c] is a symmetric matrix of damping parameters, [k] is a symmetric matrix of stiffness parameters, $\ddot{\vec{\delta}}$, $\dot{\vec{\delta}}$, and $\vec{\delta}$ are the time histories of acceleration, velocity, and displacement, and $\ddot{\vec{u}}$ is the applied acceleration time history [12]. The three mass-spring-damper system represents the brain strain dependent on rotational kinematics in each of the three anatomical head axes [12]. With tuned parameters, the solution to the equations of motion for the three mass-spring-damper system provide the displacement time histories, which are assumed to be analogous to brain deformation under rotational loading about each axis of the head. One advantage of DAMAGE, in comparison to metrics based only on peak kinematics, is the potential ability of the metric to discriminate between brain deformations resulting from multiple impacts by including the entire angular acceleration time histories of the event in the calculation of the metric [12]. For clarity throughout this study, DAMAGE will refer to the generic, mathematical function which relates angular accelerations to MPS-95 (Eq. 3), and DAMAGE-M50 will refer to the tuned brain injury metric developed using the GHBMC M50 FE brain model response.

For development of a small female DAMAGE, referred to as DAMAGE-F05, the fitting and assessment procedures established by [12] were repeated using the GHBMC F05 brain model to develop both an uncoupled and coupled model [12]. Based on the performance of the multiple DAMAGE-M50 models assessed the original development and formulation, stiffness-only proportional damping was also selected in development of DAMAGE-F05 [12]. First, the uncoupled DAMAGE-F05 was fit by optimising the uncoupled stiffness terms (k_{xx}, k_{yy}, k_{zz}), the proportional damping constant (a_1), and the scale factor (β) using a non-linear, least squares solver to minimise the sum squared error between the predicted kinematic-based metric and the GHMBC F05 MPS-95 simulation results using the uniaxial rotational head kinematics (n = 449) from the parametric dataset [12,13]. All coupled stiffness terms (k_{xy}, k_{yz}, k_{xz}) were constrained to 0 for the uncoupled metric fitting. The coupled DAMAGE-F05 metric was fit using the same methodology, but an additional coupling stiffness term (k_{xz}) was included in the optimisation, and the coupled model was fit using the multiaxial rotational head kinematics from the parametric dataset (n = 210) [12]. For the coupled model fitting, the remaining coupled stiffness terms (k_{xy}, k_{yz}) were constrained to 0, as a result of analysis completed in the development of DAMAGE where the additional coupling terms (k_{xy}, k_{yz}) were negligible, and the mass of each body (m_x, m_y, m_z) was set to 1kg. Both the uncoupled and coupled DAMAGE-F05 metrics were assessed using the real-world head kinematic database (n = 1747).

Metric Performance Assessments

Each of the F05 metrics' performances were assessed using the coefficient of determination (R^2) and the normalised root mean square error (NRMSE) [11]. The coefficient of determination was used to evaluate the association between the predicted MPS-95 from the metric and the GHBMC F05 MPS-95 value. The NRMSE, or the ratio of the root mean square errors (RMSE) between the kinematic metric prediction and the average MPS-95 value from the GHBMC F05 used to fit the metric, was used to assess the accuracy of the metric-based prediction [11,25]. For consistency with previous work, 1 – NRMSE is reported, where higher values correspond to a better fit [11,12]. First, using the MPS-95 values from the GHBMC F05 brain model across the real-world kinematics database, the prediction accuracy of the male-derived metrics (BrIC-M50, UBrIC-M50, and DAMAGE-M50) were compared to the prediction accuracy of the female-derived metrics (BrIC-F05, UBrIC-F05, DAMAGE-F05). Additionally, the correlation and prediction accuracy of the female-derived metrics were assessed across the different impact conditions within the real-world head kinematics database.

III. RESULTS

Model Fits for Female-Derived Metrics

Overall, each of the female-derived metrics correlated well with the GHBMC F05 MPS-95 results in both the fitting and assessment datasets (Fig. 1, Fig. 2, Fig. 3). The correlation and metric accuracy for the coupled DAMAGE-F05 (R^2 = 0.969 and 1 – NRMSE = 0.787) was higher than both BrIC-F05 (R^2 = 0.701 and 1 – NRMSE = 0.334) and UBrIC-F05 (R^2 = 0.916 and 1 – NRMSE = 0.711) for the assessment head kinematics. Optimised parameters for BrIC-F05 and UBrIC-F05 are provided in Table I, and optimised parameters for both the uncoupled and coupled DAMAGE-F05 metrics are provided in Table II.

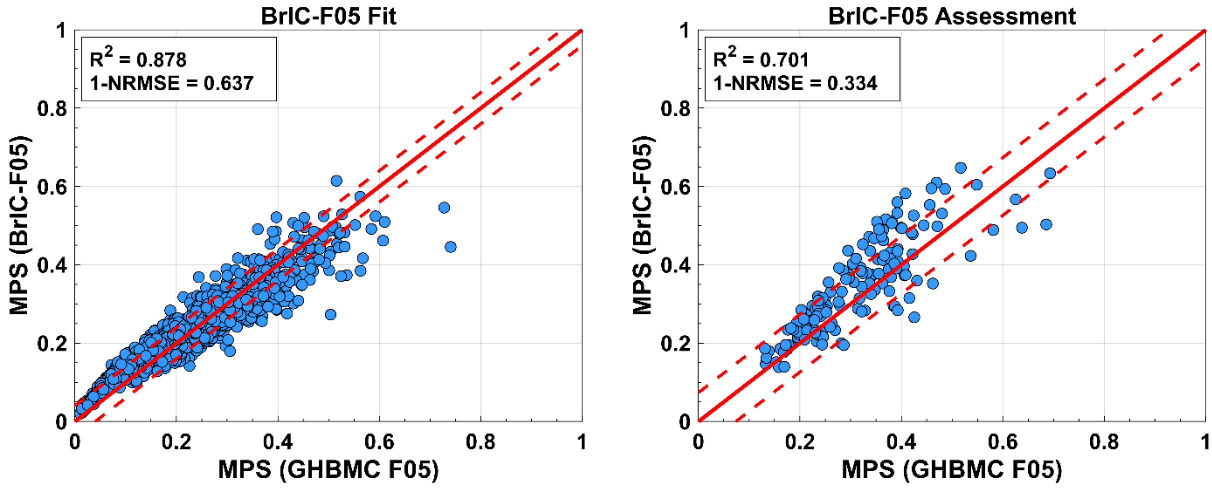

Fig. 1. (Left) Scatter plot showing BrIC-F05 predictions for the fitting dataset (n = 1595) and (right) BrIC-F05 predictions for the assessment dataset (n = 152) to the GHBMC F05 MPS-95 response. The solid line indicates a one-to-one prediction (the line y = x), and the dashed lines represent ± 1 RMSE.

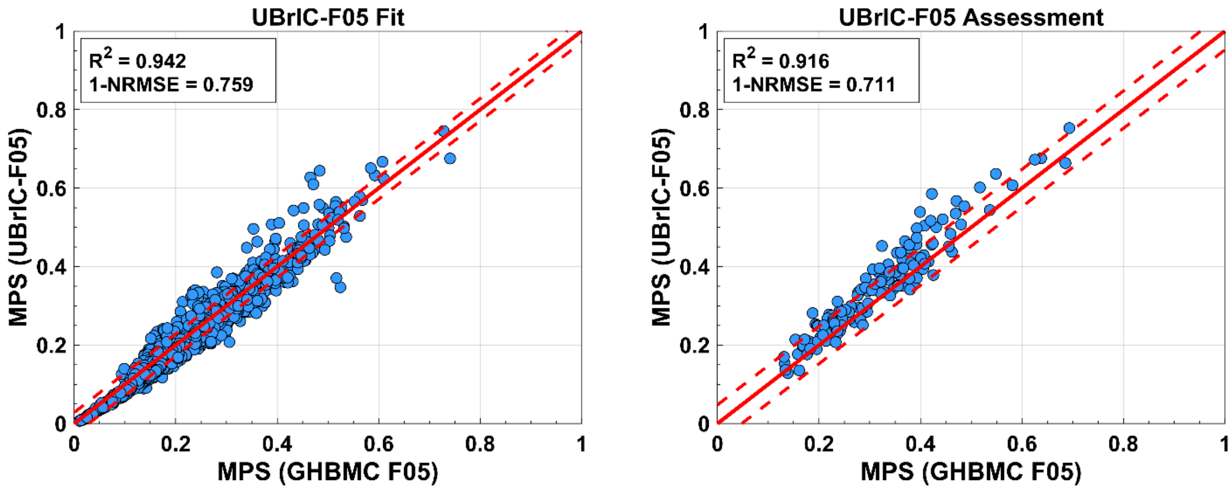

Fig. 2. (Left) Scatter plots showing UBrIC-F05 predictions for the fitting dataset (n = 1595) and (right) UBrIC-F05 predictions for the assessment dataset (n = 152) to the GHBMC F05 MPS-95 response. The solid line indicates a one-to-one prediction (the line y = x), and the dashed lines represent ± 1 RMSE.

TABLE I
FITTED MODEL PARAMETERS FOR BRIC-F05 AND UBRIC-F05.

Metric	ω_{xcr} (rad s^{-1})	α_{xcr} (krad s^{-2})	ω_{ycr} (rad s^{-2})	α_{ycr} (krad s^{-2})	ω_{zcr} (rad s^{-2})	α_{zcr} (krad s^{-2})	R^2_a	*1-NRMSE$_a$*
BrIC-F05	162	-	170	-	136	-	0.701	0.334
UBrIC-F05	202	19.1	199	18.3	149	14.2	0.916	0.711

a: Performance assessment metrics for assessment dataset

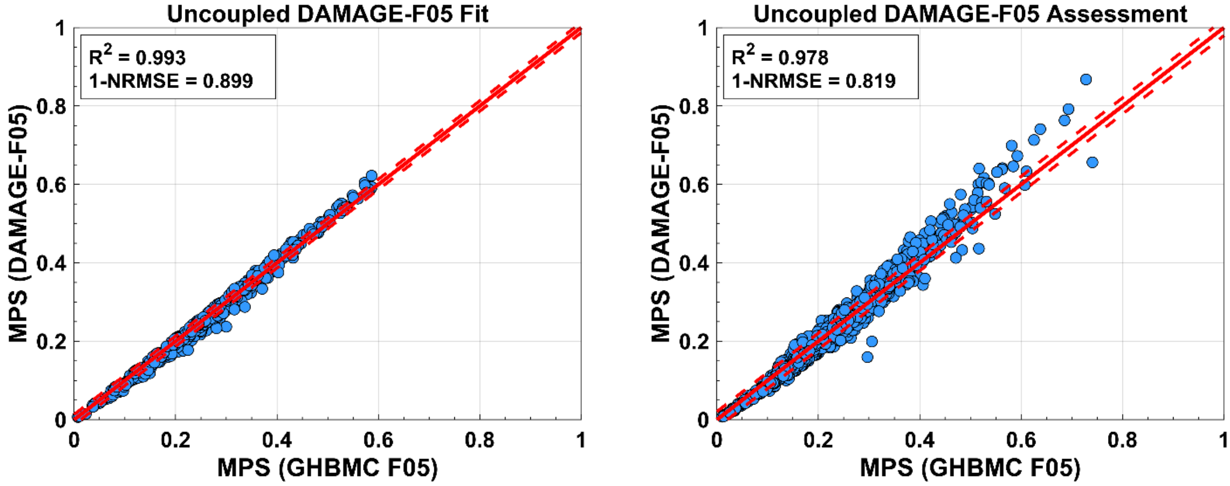

Fig. 3. (Left) Scatter plots showing uncoupled DAMAGE-F05 predictions for the fitting dataset consisting of uniaxial parametric curves (n = 449) and (right) the uncoupled DAMAGE-F05 predictions for the assessment dataset from the head impact database (n = 1747) to the GHBMC F05 MPS-95 response [12,13]. The solid line indicates a one-to-one prediction (the line y = x), and the dashed lines represent ± 1 RMSE.

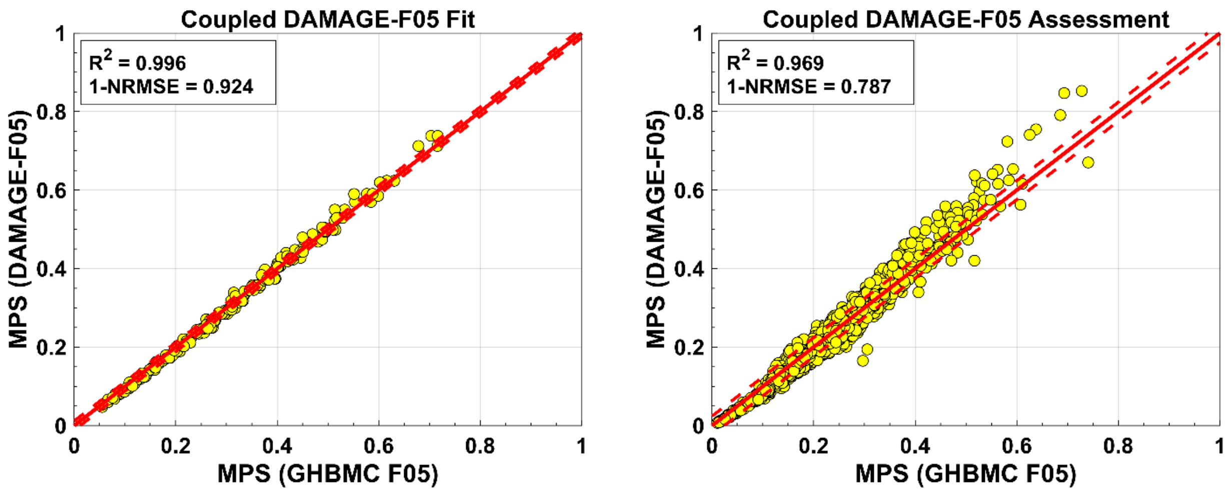

Fig. 4. (Left) Scatter plots showing coupled DAMAGE-F05 predictions for the fitting dataset consisting of multiaxial parametric curves (n = 210) and (right) the coupled DAMAGE-F05 predictions for the assessment dataset from the head impact database (n = 1747) to the GHBMC F05 MPS-95 response [12,13]. The solid line indicates a one-to-one prediction (the line y = x), and the dashed lines represent ± 1 RMSE.

TABLE II

FITTED MODEL PARAMETERS FOR UNCOUPLED AND COUPLED DAMAGE-F05 MODEL.

Model-type	k_{xx} (kN m^{-1})	k_{yy} (kN m^{-1})	k_{zz} (kN m^{-1})	k_{xz} (kN m^{-1})	a_1 (s)	β (m^{-1})	R^2_a	1-$NRMSE_a$
Uncoupled	40.4	37.1	30.7	-	9.95E-3	4.65	0.978	0.819
Coupled	34.6	31.9	25.3	0.994	8.32E-3	3.57	0.969	0.787

a: Performance assessment metrics for assessment dataset

Metric Performance Analysis

The metric accuracies of the three female-derived metrics (BrIC-F05, UBrIC-F05, DAMAGE-F05) were higher than the metric accuracy of the previously available metrics (BrIC-M50, UBrIC-M50, DAMAGE-M50), when compared to the GHBMC F05 model response (Fig. 5). To remain consistent with DAMAGE-M50, the coupled model versions of both DAMAGE-M50 and DAMAGE-F05 were used in this comparison. For each of the metrics, the F05-derived metric had greater accuracy than the M50-derived metric when comparing to the FE simulation results, and of the six metrics compared, the DAMAGE-F05 metric had the overall highest accuracy (1 – NRMSE = 0.787) across the real-world kinematics database. Instead, each of the M50-derived metrics overestimated the GHBMC F05 MPS-95 across the head kinematics simulated. As a result, the M50-derived metrics should not be used to estimate MPS-95 for F05 brain models and ATDs as the F05-derived metrics have higher accuracy across a wide range of head kinematics.

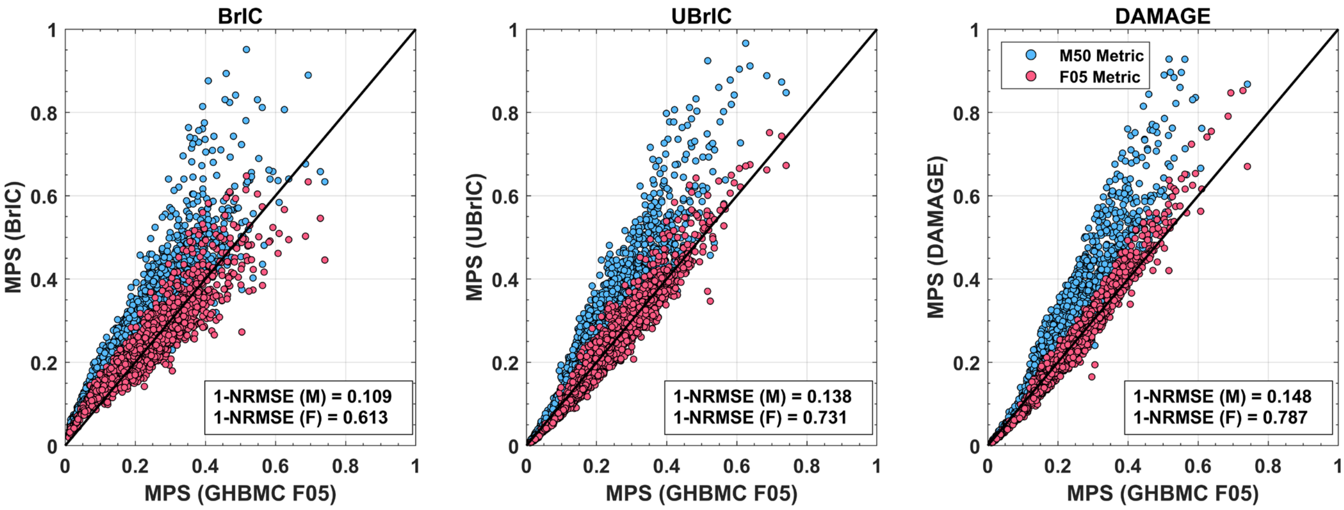

Fig. 5. Scatter plots showing the kinematic metric predicted MPS - for both M50 (blue) and F05 (pink) metrics - for BrIC (left), UBrIC (centre), and DAMAGE (right) against the MPS-95 value from the GHBMC F05 response across the entire head impact database (n = 1747). Metric accuracies (1 – NRMSE) for both male- and female-derived metrics are included. The black line indicates a one-to-one prediction.

Within the real-world head kinematics database, the three F05 brain injury metrics were compared across various impact conditions. The DAMAGE-F05 (coupled model) had the greatest R^2 for all impact conditions, as well as the highest 1 – NRMSE for all categories except un-helmeted and side sled tests (0.813 and 0.864, respectively), where UBrIC-F05 had a slightly higher 1 – NRMSE (0.825 and 0.875, respectively) (Fig. 6).

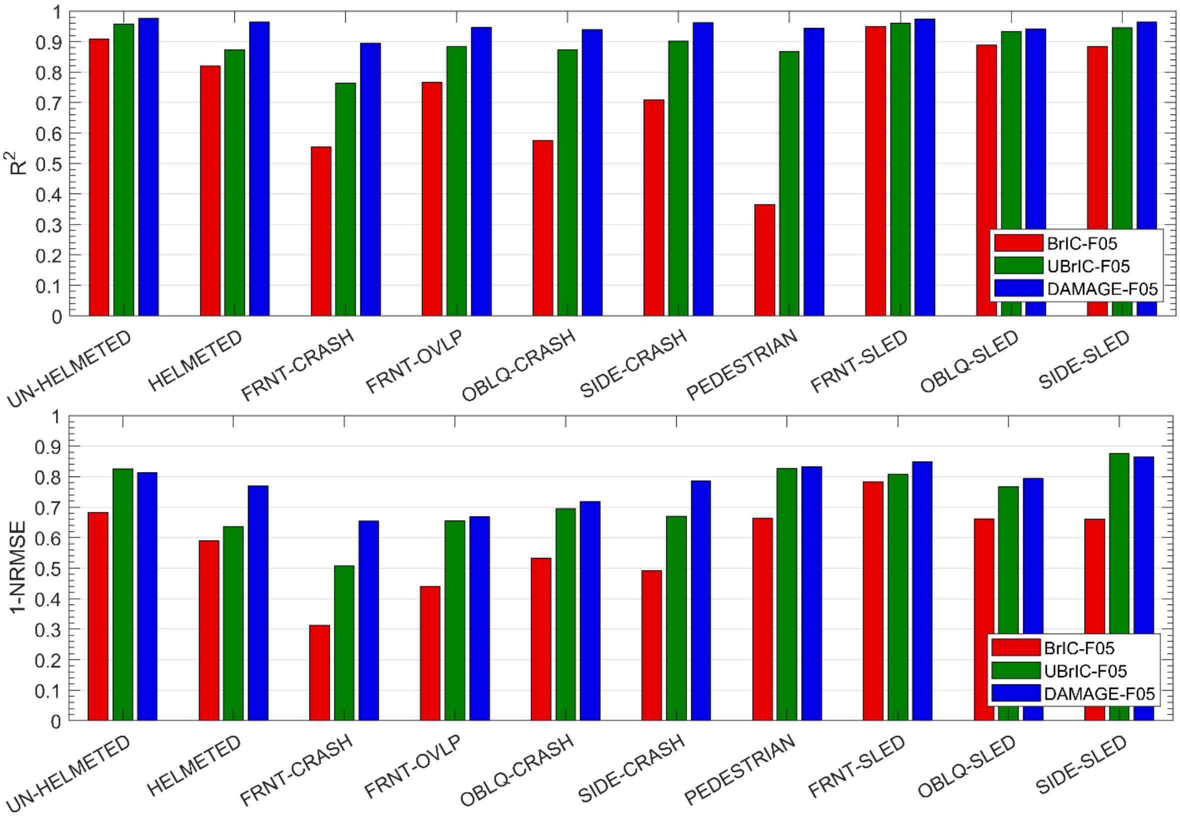

Fig. 6. Coefficients of determination (top) and metric accuracy (bottom) for each of the F05 brain injury metrics (only coupled DAMAGE-F05 included) to the FE brain model response across the different impact conditions within the head impact database (n = 1747).

IV. Discussion

Kinematic-based brain injury metrics have historically been derived from male FE brain simulation data, and the applicability of these metrics to female data was unknown. The brain strains predicted using the M50 brain injury metrics overestimated the MPS-95 strain values of the F05 GHBMC brain model. This difference in predicted strain, due in part to a smaller brain volume, justifies the need for female-derived brain injury metrics for use with small female ATDs. This study included the development and assessment of three F05-derived rotational kinematic-based brain injury metrics (BrIC-F05, UBrIC-F05, and DAMAGE-F05) using both idealized and real-world head kinematics. Between the six brain injury metrics evaluated in this study, DAMAGE-F05, which is based on directionally dependent head angular acceleration time histories, had the highest correlation to the GHBMC F05 brain simulation results, and it had the highest accuracy across most of the head kinematic impact conditions. The DAMAGE metrics are based on a multi-body mechanical system with optimised stiffnesses to capture the deformation of the three masses, which represent the brain strain in the three anatomical axes [12]. The DAMAGE-F05 primary stiffness parameters (k_{xx}, k_{yy}, k_{zz}) were greater, relating to a stiffer spring-mass-damper system, than the DAMAGE (M50) optimised stiffnesses; the overall stiffer female metric captures the smaller predicted MPS-95 value from the GHBMC F05 brain simulations compared to the larger M50 FE strain values seen in Fig. 5 [12]. Additionally, the optimised y- and z-axis critical values for both BrIC-F05 (ω_{ycr}, ω_{zc}) and UBrIC-F05 (ω_{ycr}, α_{ycr}, ω_{zcr}, α_{zcr}) were larger than the optimised critical values for the BrIC-M50 (refit) and UBrIC-M50 to predict the smaller F05 brain strains, and the critical values in the x-axis between the F05 and M50 metrics were similar [11].

A major difference between the GHBMC M50 and F05 FE brain models is the geometry and total volume; each of the model geometries are based on an individual's medical brain scan and are based on a 50th percentile male subject and a 5th percentile female subject [26,27]. For the solid elements used in the MPS-95 calculation (cerebrum, cerebellum, brain stem, subcortical regions), the M50 FE brain has 65,631 elements for a total brain volume of 1,094cm^3; the F05 brain has 99,429 elements and a total brain volume of 877cm^3. Assuming perfect hexahedral elements, the representative element lengths for the FE models are 2.55mm and 2.07mm for the M50 and F05 models, respectively. The smaller element length of the F05 brain model could bias the calculated MPS-95 value higher than if the two models had identical representative element lengths [28]. Besides differences in geometry, total brain volume, and mesh density, the M50 and F05 brain models have differing material properties. For the brain tissue regions used in the calculation of MPS-95, modelled using a Kelvin-Maxwell viscoelastic material, the M50 model is stiffer with a higher bulk modulus, instantaneous, and quasi-static shear moduli, but uses a lower damping constant. These differences in both mesh density and material formulations could affect the difference in MPS-95 values calculated from each model, but these models are commonly used human body FE models and allow initial investigation into the effect of total volume on brain injury metric applicability and assessment. The difference in total volume between the GHBMC M50 and F05 models could suggest that for the same input conditions, the male brain experiences larger brain strains and higher injury risk. To relate the higher male brain strains to higher risk of brain injury, the assumption that male and female brains have the same injury threshold would need to be made, and further research is needed to determine the credibility of this assumption.

Each of these female-derived metrics are dependent on the FE model used to predict MPS-95 and are therefore influenced by any limitations of the chosen FE brain models. As total brain volume has been shown to influence MPS, these brain injury metrics capture the brain deformation response of a single geometry and brain volume, and additional work is needed to predict the brain deformation using these metric formulations across a variety of brain anatomies. While these metrics were developed based on a small female brain anatomy, the effect of sex-differences in material properties were not included in the development of these brain injury metrics. Healthy females have been shown to have stiffer brain tissue throughout the brain parenchyma, as well as additional studies confirming differences throughout the temporal and occipital lobes [9,29]. With these differences in stiffness and additional differences in damping ratios between male and female brains in cortical grey matter, additional research is needed to capture the effect of sex-specific material properties in FE brain models and their potential effect on brain injury metric development [8].

V. CONCLUSIONS

This study addressed the need for a female brain injury metric to predict brain strain using rotational head kinematics and a female FE brain model. The use of the male-derived brain injury metrics (BrIC-M50, UBrIC-M50, DAMAGE-M50) would overpredict the MPS-95 of the F05 FE brain model response and overpredict injury risk in small females assuming an equal MPS injury risk function between the two sexes. The DAMAGE-F05 had the greatest accuracy across most of the head impact conditions in the real-world head kinematics database. These brain injury metrics can be used with small female ATDs in development of safety countermeasures to better predict the brain deformations experienced by occupants with a smaller total brain volume.

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