

Mild Traumatic Brain Injury Predictors Based on Angular Accelerations During Impacts

HIDEYUKI KIMPARA and MASAMI IWAMOTO

Vehicle Safety & Biomechanics Laboratory, Toyota Central R&D Labs., Inc., 41-1 Yokomichi, Nagakute, Aichi 480-1192, Japan

(Received 3 July 2011; accepted 19 September 2011; published online 13 October 2011)

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

Abstract—Although Head Injury Criterion (HIC) is an effective criterion for head injuries caused by linear acceleration such as skull fractures, no criteria for head injuries caused by rotational kinematics has been accepted as effective so far. This study proposed two criteria based on angular accelerations for Traumatic Brain Injury (TBI), which we call Rotational Injury Criterion (RIC) and Power Rotational Head Injury Criterion (PRHIC). Concussive and non-concussive head acceleration data obtained from football head impacts were utilized to develop new injury criteria. A well-validated human brain Finite Element (FE) model was employed to find out effective injury criteria for TBI. Correlation analyses were performed between the proposed criteria and FE-based brain injury predictors such as Cumulative Strain Damage Measure (CSDM), which is defined as the percent volume of the brain that exceeds a specified first principal strain threshold, proposed to predict Diffuse Axonal Injury (DAI) which is one of TBI. The RIC was significantly correlated with the CSDMs with the strain thresholds of less than 15% ($R > 0.89$), which might predict mild TBI. In addition, PRHIC was also strongly correlated with the CSDMs with the strain thresholds equal to or greater than 20% ($R > 0.90$), which might predict more severe TBI.

Keywords—Concussion, Traumatic brain injury, Acceleration, Rotation, Injury, Injury criterion, Finite Element model, Head, Brain.

INTRODUCTION

Injury severity for the human head has been generally predicted by using the Head Injury Criterion (HIC). HIC is the only injury metric for the head used with crash dummies in vehicle safety standards, which is defined by linear acceleration and used to predict

skull fracture and brain contusion.^{9,35} On the other hand, pedestrians sustain head injuries associated with angular head acceleration as often as those associated with linear head acceleration in car-to-pedestrian accidents.⁵ In particular, the angular acceleration contributes to the generation of concussive injuries such as Diffuse Axonal Injuries (DAIs), and subdural hematomas.³³ Since HIC is only defined by resultant linear acceleration with respect to time, it is difficult to find a correlation between HIC and rotational head motion.

Historically, biomechanical researches have focused on characterizing brain injuries with rotational head motions using animal models.²² Then, some brain injury criteria with angular accelerations were proposed. However, currently the criteria were not used because the criteria were developed under limited number of specimens and head motion patterns and were not sufficiently validated for head injuries in real-world accidents. Additionally, the reason why the criteria were not used is because injury thresholds derived from animal experiments cannot be directly applied to living humans.

Recently, some researchers have developed Finite Element (FE) models of human head including the detailed brain structure. They validated their models against several cadaveric brain responses and tried to reproduce head injuries with linear and/or angular accelerations occurred in traffic accidents and football games by using their proposed FE-based brain injury predictors.^{12,13,37} In particular, FE-based injury predictors obtained from the head FE models are very useful for the detailed understanding of brain injury mechanisms under impact situations with linear and/or angular accelerations. However, those FE-based injury predictors cannot be used for direct evaluation of head injuries with angular accelerations in crash tests using

Address correspondence to Hideyuki Kimpara, Vehicle Safety & Biomechanics Laboratory, Toyota Central R&D Labs., Inc., 41-1 Yokomichi, Nagakute, Aichi 480-1192, Japan. Electronic mail: h-kimpara@mosk.tytlabs.co.jp

dummies, since FE-based injury predictors must be used with FE model which was utilized for generating predictors.

The high occurrence of concussions in contact sports, particularly American football, has gained the attention of scientists as a unique opportunity to collect biomechanical data to characterize Mild Traumatic Brain Injury (MTBI).^{1,7,27,28} In general, heads of football players are covered by helmets with face masks, so that risk of skull fractures due to football head impacts is quite lower than automotive crash cases. However, impact forces on the helmet may generate strong rotational head motion due to the large moment arm resulting from the facemask's distance away from the center of gravity (CG) of the head. Although the severity of brain injuries due to football impacts is milder than that of automotive pedestrian head injuries, data characterizing concussive head impacts in humans would be valuable in evaluating brain injury predictors with angular accelerations.

In previous study, numerous injury predictors^{6,13,15,20,22,35} based on translational or rotational head motions were proposed. On the contrary, in order to exclude any effects of translational head kinematic variables from head injury criteria, we proposed Power Rotational Head Injury Criterion (PRHIC), which is calculated as integrated power of rotational head motion, as a new predictor for head injuries associated with angular head accelerations from datasets of 6 degrees of freedom (6DOF) at CG of the heads.¹¹ Similarly, another injury predictor of Brain Injury Criterion (BRIC) based on head rotational kinematics was proposed.³⁰ These criteria have a benefit of direct injury estimation from 6DOF accelerations measured in crash dummy's head. In this study, we propose another rotational motion based HIC as one of MTBI predictors. The objectives of this study are to investigate correlations between injury criteria proposed in this study and FE-based injury predictors calculated using a human head/brain FE model and distinguish the role of these criteria for Traumatic Brain Injury (TBI).

METHODS

This study proposed two brain injury criteria with angular accelerations based on investigation and understanding of retrospective researches on brain injury criteria. A well-validated human brain FE model¹⁰ was employed to obtain FE-based injury predictors describing outputs of brain responses during head impacts. Two datasets of head accelerations due to football impacts including non-concussive and concussive data were utilized for investigating the validity

of our proposed two injury criteria and the FE-based injury predictors. Correlation analyses were conducted to find out the relationships between the proposed injury criteria and the FE-based injury predictors. In addition, a logistic analysis estimated 50% probabilities of the proposed injury criteria for MTBI using concussive football impact data. Finally, the proposed injury criteria were evaluated with another set of pedestrian head impact data which contain severe TBI of DAI.

Head Motion-Based Brain Injury Criteria

From investigation of retrospective researches on brain injury criteria, eight variables of brain injury criteria based on head motion were selected in this study. Firstly, four head motion variables of the maximum linear acceleration, maximum angular acceleration, maximum linear velocity, and maximum angular velocity were employed as basic variables of injury criteria. All acceleration and velocity variables were defined on a local coordinate system of the head. Secondly, additional four injury criteria were defined in this study as described below.

Severity Index (SI) proposed by Gadd⁶ was a precursor to HIC. The SI was designed to have strong agreement with the Wayne State University Tolerance Curve (WSUTC: Fig. 1a)²³ and is shown in Eq. (1).

$$SI = \int A^n dt = TA^n \leq 1,000 \quad (1)$$

where A is either effective values of linear acceleration, force, or pressure, which is a response function producing threshold of injury, T is time duration, and n is weighting factor equal to 2.5.

An important feature of SI is a downward linear line on log-log coordinates for effective acceleration vs. time duration (Fig. 1a). Versace³⁵ represented effective acceleration " A " by the waveform average as Eq. (2). Then SI was modified as HIC, which is the current injury metric for head injury used in the Federal Motor Vehicle Safety Standards (FMVSS) 208. The equation is represented as Eq. (3).

$$A = \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t) dt \quad (2)$$

$$HIC = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t) dt \right\}^{2.5} \right]_{\max} \quad (3)$$

where $a(t)$ is resultant linear acceleration, and t_1 and t_2 represent the initial and final integral times which HIC is calculated over (t_1 and t_2 are selected to maximize

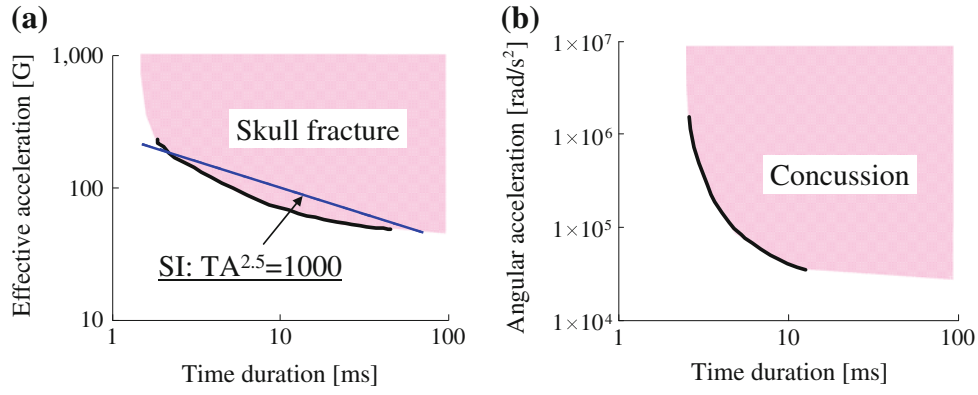


FIGURE 1. Comparison of linear and angular accelerations vs. time duration tolerance curves on log-log coordinates. (a) Effective acceleration-time tolerance curve for forehead impact to a hard, flat surface with a line of SI threshold.^{6,23} (b) Tolerance curve for angular acceleration amplitude and time duration.²²

HIC). Determining of HIC involves a computational solver which seeks the maximum value of HIC over a portion of the pulse. The maximum time duration was set as 36 ms at first, however, current standards use 15 ms. Therefore, variable term for HIC is expressed as “HIC₁₅” in this study. A HIC₁₅ of 700 was estimated as a 5% risk of Abbreviated Injury Scale (AIS) 4+ head injury.²⁵

Kleiven¹³ proposed a linear combination of HIC₃₆ and the maximum resultant angular velocity as a brain injury predictor. The Kleiven’s linear combination (denotes KLC) is represented as Eq. (4)

$$\text{KLC} = 0.004718 \cdot \omega_r + 0.000224 \cdot \text{HIC}_{36} \quad (4)$$

where ω_r is the maximum resultant angular velocity.

Ommaya *et al.*²² proposed a tolerance curve for peak angular acceleration vs. time duration using data on concussed and non-concussed monkeys. Figure 1 compares both tolerance curves of effective linear acceleration and angular acceleration with respect to time duration on log-log coordinates. Although the peak angular acceleration would differ from effective variable of angular acceleration, the tolerance curves for both linear and angular accelerations have acceleration inversely proportional to time duration. Therefore, this study postulates that the linear acceleration term of SI (Eq. 1) can be substituted with angular acceleration to approximate another injury tolerance curve for rotational head motion. In similar fashion to HIC, a new injury criterion, which we call Rotational Injury Criterion (RIC), were derived by substituting resultant angular acceleration of $\alpha(t)$ for resultant linear acceleration of $a(t)$ in Eq. (3). RIC is defined as follows:

$$\text{RIC} = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} \alpha(t) dt \right\}^{2.5} \right]_{\max} \quad (5)$$

On the other hand, Newman *et al.*²⁰ set exponent of Eq. (1) not to 2.5 but simply to 2, and found that an expression has a physical meaning of the rate of change of kinetic energy or power. Considering magnitude of the rate of change of kinetic energy, Head Injury Power (HIP) as a power expression of the human head to predict head injuries due to both linear and angular accelerations was proposed as the following equation:

$$\text{HIP} = \sum m \cdot a_i \cdot \int a_i dt + \sum I_{ii} \cdot \alpha_i \int \alpha_i dt \quad (6)$$

where m is mass of the head (kg), a_i is linear acceleration (m s^{-2}), I_{ii} is moment of inertia (MOI) (kg m^2), and α_i is angular acceleration (rad s^{-2}) when the head is assumed to be a rigid body. Considering inertial properties of mid-sized male, coefficient of mass is 4.5 kg, and those of MOI for x , y , and z directions are 0.016, 0.024, and 0.022 kg m^2 , respectively. Newman *et al.*²⁰ also determined 50% probability of HIP for MTBI as 12.8 kW.

Since this study focuses on rotational head motion, rotational components of HIP, that is, HIP_{rot} was separated from original HIP equation. In a similar manner as Eq. (5), this study substituted HIP_{rot} for resultant linear acceleration of $a(t)$ in Eq. (3), and proposed the PRHIC, which was originally introduced by Kimpara *et al.*¹¹ as the following equation:

$$\text{PRHIC} = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} \text{HIP}_{\text{rot}} dt \right\}^{2.5} \right]_{\max} \quad (7)$$

Time durations for angular acceleration obtained from football head impact data were greater than 15 ms of the maximum time duration for HIC. Therefore, the maximum integral time duration for RIC and PRHIC was set to 36 ms, which was the original time duration of HIC. Therefore, this study

proposed two new injury criteria of RIC_{36} and $PRHIC_{36}$.

In summary, in addition to four head motion variables of the maximum linear acceleration, maximum angular acceleration, maximum linear velocity, and maximum angular velocity, the four functions of HIC_{15} , HIP , RIC_{36} , and $PRHIC_{36}$ were used as injury criteria in this study.

FE-Based Brain Injury Predictors

Isolated human brain FE model used for this study was segmented out from a commercially available human FE model, THUMS (Total HUMAN Model for Safety, Toyota Central R&D Labs., Inc. and Toyota Motor Corporation), and a few modifications were applied (Fig. 2). The model consists of 49,579 elements (24,096 solid, 25,119 shell, and 364 seatbelt elements), has a mass of 4.39 kg, and an appropriate position for the CG. The brain model consists of all hexagonal solid elements representing the cerebrum, cerebellum, brainstem with distinct white and gray matter, and cerebral spinal fluid (CSF). Additionally, solid elements were used to represent the sagittal sinus and shell elements were used to represent the dura, pia, arachnoid, meninx, falx cerebri, and tentorium. The assumed material model for brain elements was linear viscoelastic, while the dura and pia maters were represented by elastic shell elements. The model had already been validated against cadaver test data on a series of translational head impact¹⁷ and two series of rotational impacts,^{8,34} and presented high bio-fidelity. The Normalized Integral Square Error (NISE),³ which was a method for quantitative evaluation to compare time history curves of pressures and displacements

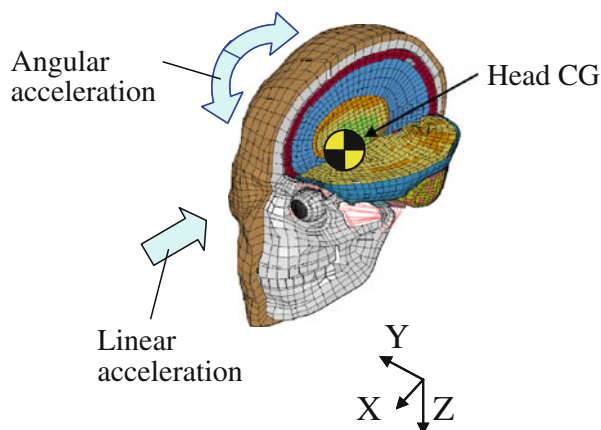


FIGURE 2. Human brain FE model and location of CG and $PRHIC_{36}$.

predicted by the model with those obtained from test data, evaluated brain responses predicted by THUMS brain FE model as excellent or good grades in 93% of assessed variables. A detailed description of THUMS head–brain model and its validation can be found in Kimpara *et al.*¹⁰ This human brain FE model was used to obtain FE-based brain injury predictors in this study.

Previously, this study showed that FE-based brain injury predictors with various types of strains were effective to predict brain responses and injury outcome.¹¹ Therefore, this study also selected 10 FE-based brain injury predictors with various types of strains obtained from whole brain elements, which included the maximum values of the first principal strain, shear strain, strain rate, product of strain, and strain rate named as “Hotspot” by Viano and Lövsund,³⁶ and Cumulative Strain Damage Measure (CSDM) utilized for evaluating DAI by Takhounts *et al.*²⁹ The CSDM was defined as the percent volume of the brain that exceeds a specified first principal strain threshold. When the threshold of the first principal strain is set to 10%, the variable term is expressed as “CSDM 10%” in this study. The CSDM used in this study included CSDM 10%, CSDM 15%, CSDM 20%, CSDM 25%, CSDM 30%, and CSDM 35% with six grades of strain thresholds to distinguish severity of damage in DAI. For example, Fig. 3 displays the difference between CSDM 10% and CSDM 20%, where the damaged volume of CSDM 10% is 49 vol.%, which is greater than that of CSDM 20% as 1.2 vol.%.

This study used the human brain FE model to predict brain injuries without any skull fractures. Therefore, the skull was modeled as rigid body and linear and angular accelerations measured in experimental studies were directly inputted as boundary conditions of the head through the skull. All simulations were conducted by a commercially available FE solver, LS-DYNA 971 Rev. 2, Shared Memory Parallel (SMP) version (LSTC, Livermore, CA) using a single core of an Intel Xeon 64 bit based computer running on a Linux operating system.

Head Acceleration Data

Head acceleration data utilized in this study represent 6DOF skull accelerations measured at CG of the head. Two datasets with football head acceleration data were used; one football dataset being non-concussive head acceleration data^{26,27} collected directly from living human subjects (referred to as 6DOF device data), another football data being concussive head acceleration data¹⁹ from the National Football League (NFL) head impacts reconstructed using Hybrid III dummies. Both the 6DOF device data and NFL data

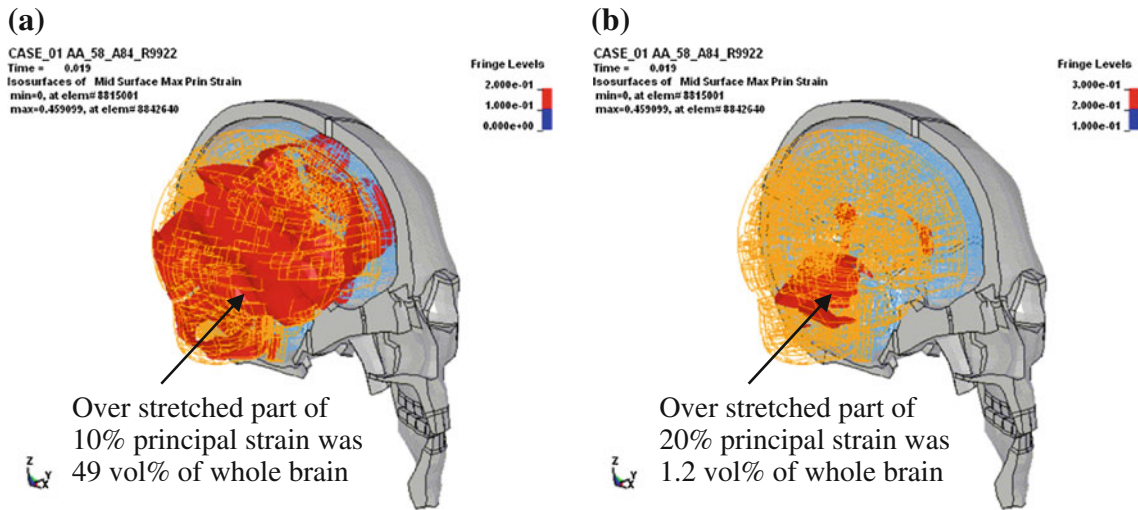


FIGURE 3. A sample of strain distribution in human brain FE model. Elements which experienced over 10% (left) and 20% (right) of the first principal strain were assumed as damaged. (a) Strain threshold 10%, (b) strain threshold 20%.

were utilized for correlation analyses between the proposed head injury criteria and the above-mentioned FE-based brain injury predictors.

The 6DOF measurement devices were installed in the helmets of Virginia Tech football players throughout the 2007 and 2008 college football seasons. All instrumented players were either offensive or defensive linemen. Each player that participated in the study gave written informed consent with Institutional Review Board (IRB) approval from both Virginia Tech and the Edward Via College of Osteopathic Medicine. Linear and angular accelerations were recorded for every impact instrumented players experienced during games and practices. A total of 4,709 impacts were recorded during practices and games for the 19 instrumented players. No instrumented player sustained MTBI in this study. In order to investigate the relationships of linear and angular head accelerations in more severe levels, this study selected 251 cases which had great linear acceleration, angular acceleration, linear velocity or angular velocity from the 6DOF device dataset.

In total, 31 impact events which involved 58 players and included 25 concussions from NFL data were reconstructed using Hybrid III dummies based on game video.²⁴ Since NFL data contain both of concussive and non-concussive head impacts, a logistic regression analysis was conducted for only NFL data to obtain injury risk curves of the proposed injury criteria predicting MTBI.

In addition, a set of pedestrian head acceleration data obtained from Dokko *et al.*² was used to evaluate the proposed head injury criteria and the thresholds obtained from the injury risk curves. Two car-to-pedestrian accidents with severe head impacts were

reconstructed with a multi-body model and a pedestrian FE model to obtain 6DOF head acceleration data.¹ One case (case #H032-86) includes DAI and subarachnoid hemorrhages, and another (case #H070-85) has DAI, subdural hematoma, and contusion. We could not find any descriptions on occurrences of skull fractures for both accident cases in the literature.²

Statistical Analyses and Application of Brain Injury Predictors

Correlation analyses were performed between eight variables of head motion-based brain injury criteria and 10 variables of FE-based brain injury predictors in order to investigate the relationship between the proposed injury criteria with angular accelerations and FE-based brain injury predictors, especially CSDM.

In addition, logistic regression analyses were performed to determine the predictive capabilities of the head motion-based brain injury criteria and FE-based brain injury predictors. Modified Maximum Likelihood Method (MMLM) proposed by Nakahira *et al.*,¹⁸ which is one of methods for the logistic regression analyses, was carried out using MS-Excel to determine primary injury predictors. The goodness of curve fit predicted by MMLM was evaluated using the Combined Evaluation Method (CEM). Both methods are based on two assumptions of injury probabilities, which are assumption (A) “the injury probability approaches zero when injury factors approach zero”, and (B) “obtained injury risk curves possess the maximum goodness of fit”. The CEM is the method to evaluate two assumptions of A and B. The Estimator for assumption B (EB) is defined as equal to the log likelihood. The greater EB indicates better goodness of

curve fit. Since this method has been used as an effective method in previous studies,³⁷ this study presumes injury risk probability as logistic curves determined by MMLM method. As a result, injury thresholds of a 50% probability of proposed brain injury criteria for MTBI were obtained.

In addition, the eight head motion-based brain injury criteria were also calculated using 6DOF head accelerations reconstructed from two sets of car-to-pedestrian accident data with severe head impacts² and the validity of proposed brain injury criteria was investigated.

RESULTS

Correlation Analyses Between FE-Based and Head Motion-Based Brain Injury Criteria

Firstly, the correlations between four head motion-based injury criteria and four head motion variables were investigated using the NFL data and 6DOF device data. The values of the correlation coefficients are summarized in Table 1, where results obtained from NFL and 6DOF device databases were indicated at upper and lower parts in each column, respectively. Reference marks of “*” and “**” present significance levels of 0.01 and 0.001, respectively. In addition, correlation coefficients greater than 0.89, which equals to coefficient of determination of 0.80, were indicated with bold characters. Significant correlations ($R \geq 0.90$) of HIC₁₅ and HIP with the maximum linear acceleration were observed in both datasets of 6DOF device and NFL. Although HIP was a linear combination of translational and rotational head kinetic power, HIC₁₅ and HIP were somewhat correlated with

the maximum angular acceleration in the NFL dataset ($0.63 < R < 0.75$), while the correlations of the same two criteria with the maximum angular acceleration in 6DOF device dataset were not significant ($p > 0.05$). KLC which is another linear combination of translational and rotational head kinematics, was significantly correlated with angular velocity in the NFL dataset ($R = 0.91$), while the correlation coefficients with the maximum linear acceleration and linear velocity were very minor in 6DOF device dataset ($-0.13 < R < -0.11$, $p > 0.05$). On the other hand, the injury criteria of RIC₃₆ had strong correlations ($R > 0.80$) with the maximum angular acceleration in both datasets. PRHIC₃₆ somewhat correlated with the maximum angular acceleration and maximum angular velocity ($0.47 < R < 0.77$), while correlation coefficients of PRHIC₃₆ with the maximum linear acceleration or maximum linear velocity were the smallest of all injury criteria ($R < 0.09$). In particular, since the significance levels for correlations of PRHIC₃₆ with linear acceleration or velocity were greater than the significance level of 5% ($p > 0.05$) in both datasets, PRHIC₃₆ has no correlations with translational head motions.

Secondly, the correlations between 10 FE-based predictors and eight head motion-based brain injury criteria were summarized in Table 2 using the NFL data and 6DOF device data. The maximum linear acceleration, maximum linear velocity, HIC and HIP produced weak correlations with FE-based brain injury predictors. The maximum angular acceleration, angular velocity, and KLC indicated moderate correlation with the maximum first principal strain, shear strain, and Hotspot in both datasets ($R > 0.62$). On the contrary, two proposed injury criteria of RIC₃₆ and PRHIC₃₆ produced strong correlations with FE-based injury predictors. RIC₃₆ correlated significantly with the maximum first principal strain, maximum shear strain, Hotspot which is the maximum value of product of strain and strain rate, and CSDM 10%, while PRHIC₃₆ were only correlated with the CSDMs with strain thresholds greater than 20%. Since brain elements were represented with nearly incompressible solid material, the first principal strain and shear strain were notably correlated each other. In particular, significant correlations in both the NFL data and 6DOF device data were found between RIC₃₆ and CSDM 10% ($R \geq 0.92$), and between PRHIC₃₆ and CSDM 30% ($R \geq 0.90$). Figure 4 shows their plots and estimated regression lines. Estimated regression lines for the NFL data and 6DOF device data were qualitatively matched together in correlation between RIC₃₆ and CSDM 10% in Fig. 4a. However, regression lines on relationship between PRHIC₃₆ and CSDM 30% (Fig. 4b) were not matched between the NFL data and

TABLE 1. Coefficients of correlation between head motion-based brain injury criteria and head motions based on NFL (upper) and 6DOF device data (lower).

Injury criteria	Max. linear acceleration	Max linear velocity	Max. angular acceleration	Max. angular velocity
HIC ₁₅	0.93** 0.93**	0.889** 0.68**	0.60** -0.04	0.38* -0.27**
HIP	0.89** 0.92**	0.93** 0.72**	0.72** 0.08	0.58** -0.15*
KLC	0.70** -0.13	0.82** -0.11	0.78** 0.61**	0.91** 0.71**
RIC ₃₆	0.50** -0.03	0.66** -0.09	0.81** 0.90**	0.86** 0.63**
PRHIC ₃₆	0.08 -0.04	0.31 -0.02	0.49** 0.59**	0.77** 0.56**

* $p < 0.01$, ** $p < 0.001$.

KLC a linear combination of HIC and maximum angular velocity proposed by Kleiven.¹³

TABLE 2. Coefficients of correlation between FE-based and head motion-based injury predictors based on NFL (upper) and 6DOF device data (lower).

FE-based injury predictors	Max. linear acceleration	Max. linear velocity	HIC ₁₅	HIP	KLC	Max. angular acceleration	Max. angular velocity	RIC ₃₆	PRHIC ₃₆
Max. first principal strain	0.49**	0.67**	0.51**	0.72**	0.87**	0.76**	0.87**	0.93**	0.76**
Max. shear strain	-0.09	-0.03	-0.06	0.06	0.82**	0.69**	0.79**	0.68**	0.55**
Max. strain rate	0.47**	0.65**	0.48**	0.69**	0.86**	0.76**	0.87**	0.92**	0.76**
Hotspot	-0.09	-0.04	-0.06	0.06	0.82**	0.70**	0.79**	0.70**	0.57**
Hotspot	0.08	0.03	0.07	0.02	0.01	0.06	-0.03	0.05	-0.07
Hotspot	-0.08	-0.10	-0.11	-0.09	0.59**	0.10	0.15	0.08	0.10
Hotspot	0.55**	0.62**	0.50**	0.69**	0.80**	0.85**	0.78**	0.88**	0.76**
Hotspot	0.07	0.01	0.09	0.23**	0.64**	0.83**	0.62**	0.89**	0.70**
CSDM 10%	0.36*	0.52**	0.36*	0.58**	0.75**	0.69**	0.80**	0.94**	0.68**
CSDM 10%	-0.08	-0.11	-0.08	0.08	0.48**	0.75**	0.59**	0.92**	0.84**
CSDM 15%	0.19	0.33	0.20	0.41*	0.66**	0.50**	0.77**	0.79**	0.85**
CSDM 15%	-0.01	-0.03	-0.03	0.12	0.39**	0.57**	0.46**	0.79**	0.889**
CSDM 20%	0.13	0.24	0.14	0.32	0.59**	0.42*	0.72**	0.66**	0.83**
CSDM 20%	0.03	0.01	0.00	0.15	0.35**	0.49**	0.42**	0.72**	0.92**
CSDM 25%	0.13	0.23	0.12	0.30	0.60**	0.44**	0.73**	0.65**	0.85**
CSDM 25%	0.07	0.07	0.05	0.19*	0.34**	0.49**	0.40**	0.72**	0.91**
CSDM 30%	0.16	0.33	0.17	0.39*	0.66**	0.51**	0.78**	0.76**	0.94**
CSDM 30%	0.07	0.06	0.05	0.18*	0.29**	0.42**	0.35**	0.64**	0.90**
CSDM 35%	0.09	0.26	0.09	0.33	0.59**	0.48**	0.75**	0.70**	0.95**
CSDM 35%	0.07	0.07	0.05	0.17*	0.26**	0.40**	0.33**	0.61**	0.85**

* $p < 0.01$, ** $p < 0.001$.

Hotspot Maximum value of product of strain and strain rate.

CSDM Cumulative Strain Damage Measure.

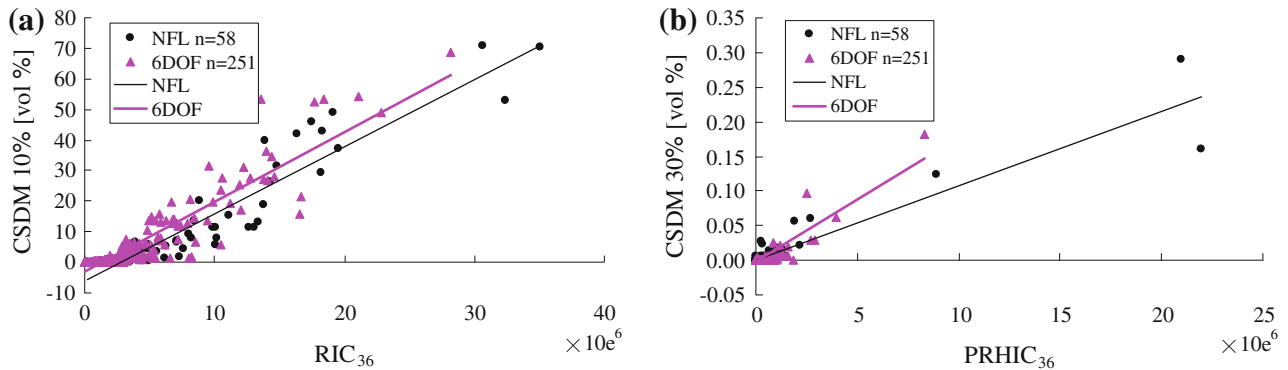


FIGURE 4. Correlations of CSDMs against RIC₃₆ and PRHIC₃₆. (a) RIC₃₆ vs. CSDM 10%, (b) PRHIC₃₆ vs. CSDM 30%.

6DOF device data due to biased data plots. This is probably because few severe impact data were included in the NFL and 6DOF device data.

Additional correlation analysis on RIC₃₆ and PRHIC₃₆ with exponent values of 1.0, 2.0, and 3.0 was carried out to investigate physical dimension of original RIC₃₆ and PRHIC₃₆. Table 3 shows correlation coefficients between injury criteria of RIC₃₆ and PRHIC₃₆ with changed exponent values and six variables, which correlated well with RIC₃₆ and PRHIC₃₆, such as the maximum angular acceleration and velocity, the first principal strain, Hotspot, CSDM 10%,

and CSDM 30%. RIC with exponent values of 1.0 and 2.0 would be related with physical dimensions of angular velocity and rotational power of head motion, respectively. The maximum angular acceleration was strongly correlated with RIC₃₆^{2.0} and RIC₃₆^{2.5} ($R > 0.90$) in 6DOF dataset, while the maximum angular velocity was correlated with PRHIC₃₆^{1.0} in both datasets of NFL and 6DOF ($R > 0.93$). Although RIC₃₆^{1.0} appeared to have an angular velocity related dimension, the correlation coefficients between RIC₃₆^{1.0} and maximum angular velocity were not so great in both datasets ($R < 0.89$). However, RIC₃₆ and PRHIC₃₆ with

TABLE 3. Coefficients of correlation for RIC₃₆ and PRHIC₃₆ with different exponents based on NFL (upper) and 6DOF device data (lower).

	Max. angular acceleration	Max. angular velocity	Max. first principal strain	Hotspot	CSDM 10%	CSDM 30%
RIC ₃₆ ^{1.0}	0.75**	0.81**	0.84**	0.71**	0.83**	0.53**
	0.77**	0.80**	0.76**	0.71**	0.66**	0.37**
RIC ₃₆ ^{2.0}	0.76**	0.83**	0.90**	0.80**	0.91**	0.67**
	0.90**	0.70**	0.73**	0.87**	0.87**	0.55**
RIC ₃₆ ^{2.5}	0.81**	0.86**	0.93**	0.88**	0.94**	0.76**
	0.90**	0.63**	0.68**	0.89**	0.92**	0.64**
RIC ₃₆ ^{3.0}	0.83**	0.86**	0.92**	0.93**	0.92**	0.82**
	0.87**	0.58**	0.65**	0.885**	0.92**	0.71**
PRHIC ₃₆ ^{1.0}	0.61**	0.93**	0.83**	0.79**	0.78**	0.94**
	0.55**	0.95**	0.74**	0.60**	0.62**	0.43**
PRHIC ₃₆ ^{2.0}	0.54**	0.83**	0.80**	0.79**	0.74**	0.96**
	0.65**	0.68**	0.64**	0.75**	0.87**	0.81**
PRHIC ₃₆ ^{2.5}	0.49**	0.77**	0.76**	0.76**	0.68**	0.94**
	0.59**	0.56**	0.55**	0.70**	0.84**	0.90**
PRHIC ₃₆ ^{3.0}	0.45**	0.73**	0.72**	0.73**	0.63**	0.91**
	0.53**	0.47**	0.47**	0.63**	0.78**	0.93**

* $p < 0.01$, ** $p < 0.001$.

Hotspot Maximum value of product of strain and strain rate.

CSDM Cumulative Strain Damage Measure.

TABLE 4. Results of logistic regression analyses of MMLM's evaluation value (EB) determined by CEM, and 50% probability of each factor for NFL data ($N = 58$).

	Injury predictor	Unit	MMLM	
			CEM (EB)	50% probability
FE-based	Hotspot	s ⁻¹	-0.491	10.5
	Max. first principal strain	%	-0.506	31.8
	Max. shear strain	%	-0.510	31.6
	CSDM 10%	vol.%	-1.289	18.2
	CSDM 30%	vol.%	-1.472	0.008
Head motion-based	RIC ₃₆	-	-0.772	1.03×10^7
	PRHIC ₃₆	-	-1.521	8.70×10^5

exponent values equal to or greater than 2.5 indicated more significant correlations with CSDM 10% and CSDM 30% in both datasets ($R > 0.89$), respectively.

Logistic Regression Analyses and Injury Risk Probabilities

Table 4 shows a summary of MMLM's evaluation value of EB determined by CEM and ranks the FE-based brain injury predictors and head motion-based brain injury criteria by EB scores. Since most of football impact data used in this study was mild impact data, the obtained EB scores indicate the possibility for predicting MTBI. The best FE-based injury predictor for MTBI was Hotspot, followed by the maximum first principal strain, shear strain, CSDM 10%, and CSDM 30%. As for the head motion-based injury criteria, the EB values of RIC₃₆ and PRHIC₃₆ were equivalent to

those of FE-based injury predictors. Figure 5 shows logistic regression injury risk curves for RIC₃₆ and PRHIC₃₆. Based on concussive NFL head impact data, a RIC₃₆ of 1.03×10^7 and a PRHIC₃₆ of 8.70×10^5 represent 50% probabilities for MTBI.

Application to Pedestrian Head Impacts with Severe TBI

Injury probabilities for TBI have been determined by the maximum angular acceleration or angular velocity so far.^{4,14,15,21} Figure 6 shows data plots of 6DOF sensors with the thresholds for DAI on a graph of angular acceleration vs. angular velocity. Case #H070-85 was already over the thresholds obtained from literatures. On the contrary, Case #H032-86 could be judged as injured based on thresholds of Lowehiell¹⁴ and Ewing,⁴ however, this same case was

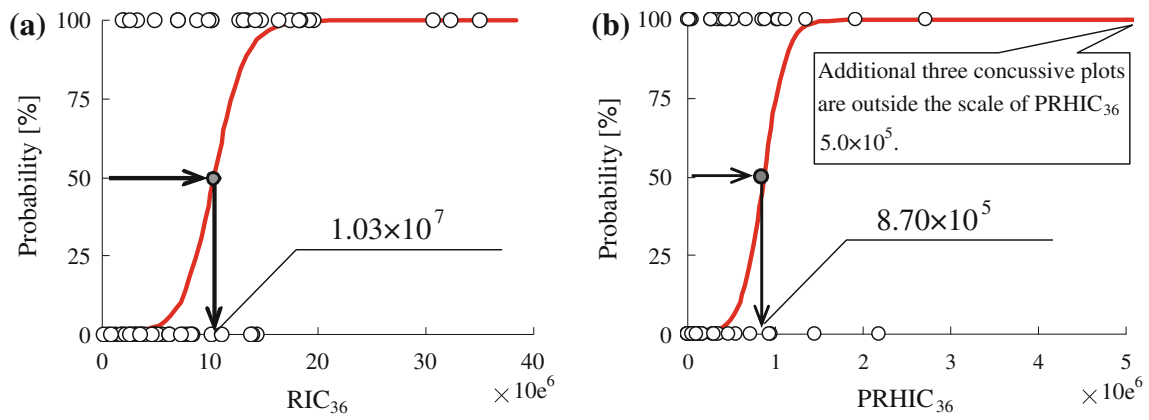


FIGURE 5. Injury risk probabilities for rotational head motion-based injury criteria determined by MMLM.

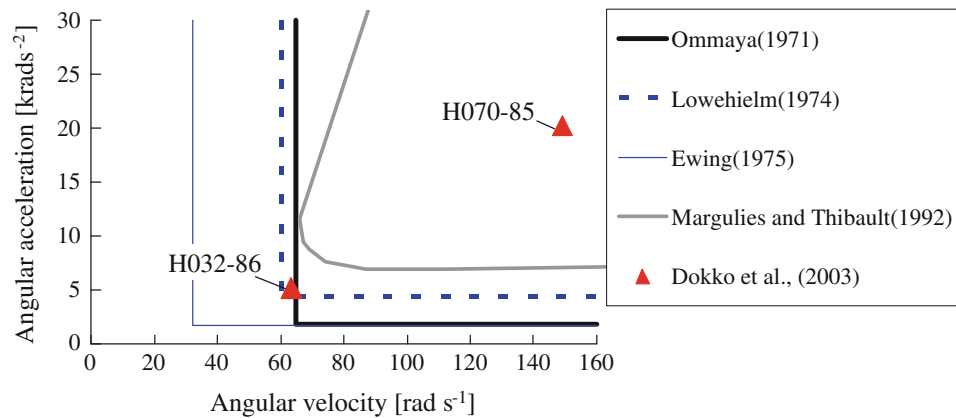


FIGURE 6. Angular acceleration vs. angular velocity of 6DOF sensor data with threshold lines obtained from literatures.

TABLE 5. Variables of injury criteria using reconstructed pedestrian head impact data ($N = 2$) with injury thresholds.

FE outputs	Max. linear acceleration (m s^{-2})	Max. linear velocity (m s^{-1})	HIC ₁₅ (-)	HIP (kW)	Max. angular acceleration (rad s^{-2})	Max. angular velocity (rad s^{-1})	RIC ₃₆ (-)	PRHIC ₃₆ (-)
Thresholds	-	-	700	1.28×10^1	-	-	1.03×10^7	8.70×10^5
H032-86	1839.3	30.9	396	1.57×10^2	5193.0	63.4	1.20×10^7	1.49×10^6
H070-85	3977.2	50.0	18806	5.81×10^2	20272.7	149.2	3.83×10^8	3.44×10^8

determined as non-injured with thresholds of Ommaya²¹ and Margulies and Thibault.¹⁵

Table 5 summarizes results of eight head motion-based brain injury criteria for reconstructed pedestrian head impacts. Injury thresholds of HIC₁₅, HIP, RIC₃₆, and PRHIC₃₆ were also listed in the table. A case of #H032-86 did not exceed the injury threshold of HIC₁₅, while HIP of 1.57×10^2 kW, RIC₃₆ of 1.20×10^7 , and PRHIC₃₆ of 1.49×10^6 were over the thresholds of 50% probabilities for MTBI. On the other hand, another case of #H070-85 presented that all injury criteria exceeded their thresholds for MTBI.

DISCUSSION

Rotational Head Motion Data

The characteristics of head motions between 6DOF device data and NFL data were different in correlation between linear and angular accelerations. The 6DOF data had little correlation ($R^2 = 0.25$, $p = 0$) between linear and angular accelerations,²⁷ while NFL data presented more significant linear relationship between linear and angular accelerations ($R^2 = 0.58$).²⁴ Rowson *et al.* suggested difference between NFL data and 6DOF device data.²⁷ According to their study, the reasons of correlation between linear and angular

accelerations in NFL data would be caused by biased population parameter toward concussive impacts and the neck property of Hybrid III dummy. 6DOF device dataset are more reasonable and reliable and to find out a new injury criterion of the brain because they directly measure the kinematics of the human head.

FE-Based Injury Predictors

When the first principal strains of elements in a brain FE model are assumed to be extensions of axonal filament, classification of axonal injuries defined by Maxwell *et al.*¹⁶ could be applied to the strain thresholds of CSDMs. Axonal injuries can be categorized in two severities of primary axotomy and secondary axotomy. Primary axotomy indicates severe damage of an axon filament associated with strain of more than 20%. Secondary axotomy has several stages, where focal loss of axonal transport is associated with 5–10% strain, axonal swelling is associated with 10–15% strain, and axonal bulbs are associated with 15–20% strain. Some researchers showed that strains in brain tissue level had a linear correlation with strains in axonal level based on experimental mechanical investigation using animal brain tissues.³² Therefore, we hypothesized that CSDMs with lower strain thresholds such as 10%, 15%, 20% could be related with mild TBI and CSDMs with higher strain thresholds such as 25%, 30%, 35% could be related with severe TBI in an effort to determine most appropriate strain threshold for TBI.

Selection of brain FE models would affect the relations between head accelerations inputted to the models and brain responses predicted by the models. Logistic regression analysis performed in this study estimated Hotspot as the best FE-based injury predictor for MTBI. This result had agreement with King *et al.*,¹² who suggested Hotspot in the midbrain region as the best predictors for concussion using their brain FE model with linear viscoelastic brain elements. On the contrary, Willinger and Baumgartner³⁷ concluded von Mises stress in whole stress as the best predictor of concussion using brain FE model with linear viscoelastic material including CSF with Arbitrary Lagrangian–Eulerian (ALE) method. Kleiven¹³ showed relations between local tissue brain responses of pressure, strain, product of strain and strain rate, as well as CSDM and the injuries using their brain FE models considering a variant material definition of hyperelastic and linear viscoelastic. Even though these studies analyzed the same data from the NFL reconstructions used in this study, the different FE models suggested different FE-based injury predictors. This is probably because the fact that differences in the selected material properties, mesh geometry, validation quality against experimental data, and utilized FE

codes would affect not only predicted brain response, but also the resulting injury predictors and conclusions. Therefore, further investigations on how brain responses and the damage mechanisms should be simulated by using FE models would be necessary to determine which head kinematic variables increase magnitude of strain or stress in whole brain. Although THUMS brain model presented high biofidelity with three sets of test data,¹⁰ further validation works are needed for better prediction of brain responses.

Since four injury predictors as the maximum first principal strain, maximum shear strain, maximum strain rate, and Hotspot among 10 injury predictors indicate the maximum values in whole brain elements, these injury predictors could be significantly affected by mesh size of the brain FE model. On the contrary, the CSDM with lower strain thresholds tends to present axonal damage spreading throughout brain and brainstem. The CSDM 10% causing a 50% probability for MTBI in football head impacts was 18.2 vol.% of whole brain. On the other hand, CSDM with higher strain thresholds tend to predict damaged elements at much smaller area of midbrain region. The CSDM 30% causing a 50% probability for MTBI was only 0.008 vol.% damages of whole brain. According to our hypothesis described previously, CSDM 30% could be related with severe TBI. Since the volume percent of each brain element ranges from 0.0004 to 0.027 vol.% with an average value of 0.0066 ± 0.0045 vol.%, the 0.008 vol.% of the brain may correspond to percent volumes of 1–2 solid elements in the brain model. Therefore, the maximum principal strain of 30% would be also regarded as one of the injury thresholds for severe brain injury. However, since injury risk curves predicted in this study were based on concussive football head impacts, all predicted injury thresholds indicate 50% probabilities of MTBI. Therefore, the predicted threshold for CSDM 30% might not be appropriate for severe TBI. This is also the reason why the goodness of logistic curve fit for MTBI predictor of CSDM 10% were better than that for severe TBI predictor of CSDM 30%. Although further investigations and comparison with 6DOF head acceleration data including severe head impacts and clinical observations are still needed to determine which FE-based injury predictors can precisely represent injury mechanisms of concussion and DAI, CSDMs representing a volume percent of damaged elements in whole brain elements could be better injury predictors for TBI.

Investigation on Physical Meanings of RIC and PRHC

Gadd stated that injury is some function of both intensity of the loading and its time duration.⁶

However, it is hard to express the physical meanings of SI variable due to unnatural physical dimension. Therefore, Newman *et al.*²⁰ exchanged exponent of 2.5 of Eq. (1) to that of 2.0 to obtain a physical meaning of the rate of change of kinetic power. Similar investigation conducted in current study (Table 3) did not show significant correlations between the maximum angular velocity and $RIC_{36}^{1.0}$ which might be related to physical dimension of angular velocity. This would be because the limited time duration of $RIC_{36}^{1.0}$ up to 36 ms could not represent the maximum angular velocity which is calculated from whole accumulation of integrated angular acceleration. On the other hand, $PRHIC_{36}^{1.0}$ was significantly correlated with the maximum angular velocity. Although $PRHIC_{36}^{1.0}$ was related to physical dimension of kinetic energy, due to the limited time duration for time integration up to 36 ms, the magnitude of $PRHIC_{36}^{1.0}$ would be affected by kinetic power rather than energy. Therefore, $PRHIC_{36}^{1.0}$ might correlate with the maximum angular velocity, which makes a part of rotational kinetic power. However, either $RIC_{36}^{1.0}$ or $PRHIC_{36}^{1.0}$ did not correlate with any injury predictors predicted by THUMS brain model. Current study indicates that RIC_{36} and $PRHIC_{36}$ with exponents greater than 2.0 had significant correlation with FE-based injury predictors. Since $PRHIC_{36}$ with exponents greater than 2.0 corresponds to uncommon physical dimension, further cautious verifications for correlations and physical meanings of original $RIC_{36}^{2.5}$ and $PRHIC_{36}^{2.5}$ would be necessary.

Injury Criteria of Rotational Head Kinematics

In previous study, Kleiven¹³ and Newman²⁰ proposed KLC and HIP which were linear combinations of translational and rotational head kinematic variables using the NFL dataset, respectively. KLC and HIP represented different relationships as in Tables 1 and 2 due to difference of weighting factors for translational and rotational head kinematic variables. However, the FE analysis showed that the translational kinematics may contribute insignificantly to the strains in the brain (Table 2). This finding agreed with FE analysis conducted by Kleiven.¹³ When brain injury was determined by the intracranial strains, injury criteria based on pure rotational head kinematics would be better than those including linear head kinematic variables for predicting TBI.

Based on hypothesis of brain injury severity with axonal stretch as mentioned above, CSDMs with a greater strain threshold of 30% may predict severe TBI, while CSDM 10% may predict milder brain injuries such as concussion due to football head impacts. Correlation analyses between FE-based brain injury predictors and head motion-based injury criteria

revealed that RIC_{36} was significantly correlated with CSDM 10% ($R > 0.92$), while $PRHIC_{36}$ was strongly correlated with CSDM 30% ($R > 0.90$). Although the maximum angular velocity showed strong correlations with the maximum principal strain ($R > 0.79$) or CSDM 10% ($R > 0.59$), correlation coefficients of RIC were greater than those of the maximum angular velocity when THUMS brain FE model was used to predict brain responses. Therefore, this study recommends RIC_{36} and $PRHIC_{36}$ as the different injury predictors for mild and severe TBI, respectively.

On the contrary, Takhounts *et al.*³¹ reported that the maximum angular velocity was correlated with CSDM 25% predicted by their SIMon brain FE model in their linear regression analyses using the same 6DOF device data used in this study. Then they proposed another brain injury criterion based on head rotational kinematics called BRIC.³⁰ The BRIC is calculated from summation of normalized maximum angular acceleration and normalized maximum angular velocity. In terms of representation of different injury severities of TBI, Takhounts *et al.*³⁰ adopted a method of severity ratios which scales up or down injury severities from injury severity of AIS 4+, while current study prepared two different criteria of RIC_{36} and $PRHIC_{36}$. Further investigation with additional head motion data including TBI cases would be necessary to evaluate the predictive capabilities of BRIC, RIC_{36} , and $PRHIC_{36}$.

Applicability to Car-to-Pedestrian Head Impacts

Two case data of pedestrian head impacts obtained from Dokko *et al.*² were used to investigate applicability of proposed injury criteria to automotive head impacts with severe TBI. The utilized pedestrian FE model was validated against three cases of PMHS whole body pedestrian impact tests and body kinematics and trajectories of whole body were duplicated, and contact points on the vehicular body showed well agreement with actual cases.² Although reconstruction of pedestrian impact and estimation of rotational head motion of pedestrian are difficult due to many uncertain physical conditions and a few evidences, simulation-based reconstruction of pedestrian head impacts would be one of the best methods in currently available techniques. Therefore, predicted 6DOF head motions would be considered as acceptable for further analysis to predict brain injuries.

Case #H070-85 of pedestrian head impact (Table 5) had extremely high HIC value, so that severe head injury could be expected without any rotational injury criteria. On the other hand, HIC value of case #H032-86 had lower than the threshold of 700. If head injury of this case was assessed with only HIC, brain injuries

could be passed over. Some of injury threshold based on angular acceleration and velocity could not judge this case as injured (Fig. 6). However, RIC₃₆ and PRHIC₃₆ exceeded the proposed injury thresholds of concussion in this case. Although another injury criterion of HIP indicates over the threshold for MTBI, correlation analysis in this study did not show high correlation coefficient between HIP and FE-based brain injury predictors.

Therefore, we propose to use all of HIC₁₅, RIC₃₆, and PRHIC₃₆ in order to investigate various head injuries. However, the proposed injury threshold with PRHIC₃₆ of 8.70×10^5 has a limitation that the threshold was derived from not severe TBI data but mild TBI data. Further studies are needed to better understand the head and brain injury mechanisms through investigation of the relationships between skull fractures and brain injuries. This may help produce more reliable head injury prediction tools for occupants and pedestrians in automotive accidents and players in football impacts.

ACKNOWLEDGMENTS

We would like to thank Dr. Steve Rowson and Dr. Stefan M. Duma of Virginia Tech—Wake Forest University, School of Biomedical Engineering and Sciences collecting the 6DOF sensor data used in this study. And we also thank Dr. Viano and Dr. Shewchenko of Biokinetics and Associates Ltd. for providing the NFL data.

CONFLICT OF INTEREST

The THUMSTM has been developed by Toyota Central R&D Labs., Inc. (TCRDL) and Toyota Motor Corporation (TMC) in conjunctions with the Wayne State University (WSU). The authors work for TCRDL which has a financial interest in the THUMSTM. However, there is no financial interest in the basic ideas of this study regarding RIC and PRHIC.

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