Research Report

Head Injury Prediction Methods Based on 6 Degree of Freedom Head Acceleration Measurements during Impact

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ABSTRACT The objectives of this study were 1) to obtain sets of three linear and three angular head accelerations using newly developed 6DOF (degree of freedom) measurement devices, 2) to apply the data to a well validated human brain FE (finite element) model, and 3) to investigate correlations between FE-based brain injury predictors and head injury criteria. One hundred cases of 6DOF head accelerations obtained from 19 college football players was applied to a detailed human head brain FE model. In addition, the 6DOF acceleration data were used to calculate head injury criteria, such as HIC_{15} (Head Injury Criterion), the maximum linear and angular acceleration, and $PRHIC_{36}$ (Power Rotational Head Injury Criterion) which is newly proposed criterion in this study with angular velocity and acceleration of the head CG (center of gravity). Analysis using human head brain FE model predicted the maximum first principal strain and CSDM (Cumulative Strain Damage Measure) which is defined as the percent volume of the brain that exceeds a specified first principal strain threshold. The maximum linear acceleration or HIC_{15} did not show any correlations with various types of strains obtained from brain model. In contrary, this study found significant correlation between $PRHIC_{36}$ and CSDM with strain thresholds of 20% (R^2 = 0.84).

KEYWORDS Safety, Acceleration, Injury, Finite Element Method (FEM), Head, Brain

1. Introduction

Head injury constitutes 72.4% of all fatal cases considering primal injured body region based on police data of Japan [(1)] in 2008. Although automotive related fatalities, which are number of victims who died within 30 days after accident, are continuously reduced to 6,023, head injury is still remarkable problem in Japan. Considering accident patterns of those fatal cases, the police data also suggest that pedestrian (29.4%) and cyclist (29.3%) were more frequent than occupant (21.3%). Feist et al.[(2)] analyzed field data from a European database and found that pedestrians sustain head angular acceleration more frequently than linear acceleration in car-to-pedestrian accidents. The HIC (Head Injury Criterion) is widely used as an injury predictor of the head. However, due to the definition of HIC, it is difficult to find a correlation between HIC and angular head accelerations. Therefore, new head injury criteria for angular head impacts are needed.

Recently, a number of isolated human brain FE (finite element) models have been proposed and used to investigate the injury predictors. Most of those were validated against several series of PMHS (post mortem human subject) tests, and some researches proposed FE-based brain injury predictors such as CSDM (Cumulative Strain Damage Measure), which is defined as the percent volume of the brain that exceeds a specified first principal strain threshold, proposed to predict DAI (diffuse axonal injury) which is one of traumatic brain injury. However, due to limited number of experimental data, it is difficult to validate FE models in angular head accelerations well.

Besides PMHS tests, the high occurrence of concussions in football provides a unique opportunity to collect biomechanical data to characterize MTBI (mild traumatic brain injury). For example, Duma et al.[(3)] presented a study to quantify head acceleration in collegiate football players by collecting over 3,000 impacts from 38 players using the HITS (Head Impact Telemetry System: Simbex, Lebanon NH) measurement devices, in which one concussive event was measured. This HITS measurement device is capable of measuring resultant linear acceleration and impact location. However, it does not measure x, y, and z axis linear acceleration. Angular acceleration is also incapable of being measured. The 6DOF (degree of freedom) measurement device (abbr. 6DOF sensor) improves the design of this measurement device to

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measure x, y, and z axis linear and angular accelerations.(4)

The objectives of this study were 1) to obtain sets of three linear and three angular head accelerations using newly developed 6DOF sensors, 2) to apply the data to a well validated human brain FE model, and 3) to investigate correlations between FE-based brain injury predictors and head injury criteria.

2. Methods

2. 1 Measurement of Head Impact Accelerations

2. 1. 1 Measurement Device

The 6DOF sensor, which utilizes twelve single-axis, high-g iMEMS accelerometers (ADXL193, Analog Devices, Norwood, MA), is designed to be integrated into Riddell Revolution football helmets. Twelve accelerometers are enclosed in the fabric padding, positioned in orthogonally oriented pairs at six different locations (**Fig. 1**). All accelerometers are orientated so that the sensing axis is tangential to the head's CG (center of gravity). The fabric pad also serves as a spring to keep the accelerometers in contact with the head (**Fig. 2**). When the helmet is impacted, the padding inside the helmet compresses and the helmet shifts the positions on the head. Therefore the measured accelerations are not helmet acceleration, but head acceleration.(5)

Since twelve accelerometers are redundant for the 6DOF measurement, an algorithm with an iterative optimization approach is used to solve for linear and angular acceleration.(6)

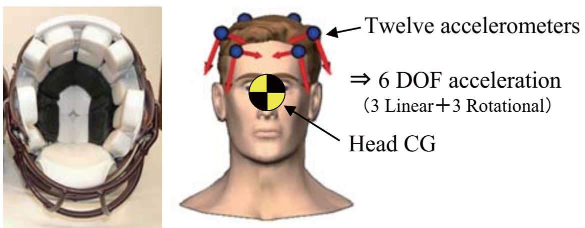

Fig. 1 6DOF sensor installed in a Riddell Revolution helmet (left) has 12 tangentially oriented accelerometers (right).

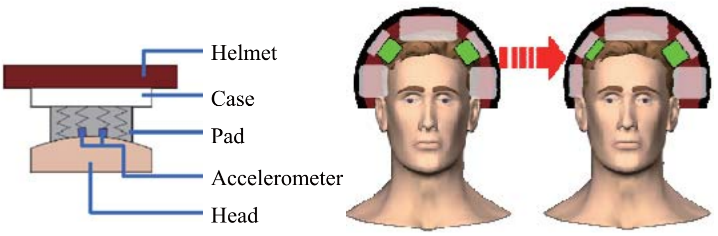

Fig. 2 Schematic of 6DOF sensor (left) and padding function which keeps accelerometers in contact with the head (right).

2. 1. 2 Validation

A total of 114 impact tests were conducted to assess the accuracy of the 6DOF sensor using an instrumented 50th percentile male Hybrid III head and neck assembly by Rowson et al. (2010).(7) The Hybrid III head was equipped with nine accelerometers (7264-2000B, Endevco, San Juan Capistrano, CA) in a 3-2-2-2 orientation; which allowed linear and angular acceleration to be calculated. The head and neck were mounted on a custom linear slide table built to NOCSAE (National Operating Committee on Standards for Athletic Equipment specification).(8) Since the linear slide table permitted five degrees of freedom, head and neck orientation could be adjusted with high repeatability.

Efforts were made to simulate a helmet fitted to a human head while using the Hybrid III. This included inserting a low-friction interface between the Hybrid III head and helmet.(7) This simulated helmet movement relative to the head that is seen in typical on-field impacts.

The helmeted Hybrid III head was struck with the pneumatic linear impactor with several combinations of impact velocities and locations (**Fig. 3**). Impact velocities were chosen so that they would simulate a range of impact severities typically experienced in tackling and blocking. The impact velocities ranged

Fig. 3 Pneumatic linear impactor and helmeted Hybrid III head mounted on the linear slide table.

from 3.0 to 9.0 m/s and were based on the NFL (National Football League) reconstruction data.[9] To account for the various ways that a helmet can be struck, seven impact locations were chosen. Each combination was tested in four trials.

2. 1. 3 Data Acquisition System

Data acquisition is triggered when any accelerometer exceeds 10 g. The threshold of 10 g was decided based on a previous experimental study in which accelerations sustained by the head during daily activities were measured.[10] Data are collected for 40 ms at 1,000 Hz, of which 8 ms are pre-trigger and 32 ms are post-trigger. After each impact is recorded, the data are sent to a computer on the sideline via a 903-927 MHz wireless transceiver. For each impact, linear and angular accelerations for each axis and impact location are measured.

2. 1. 4 Data Collection

The 6DOF sensors were installed in the helmets of 19 Virginia Tech football players throughout the 2007 and 2008 Virginia Tech college football seasons. All 19 instrumented players were either offensive or defensive linemen. Each player that participated in the study gave written informed consent with Institutional Review Board 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.

All data were up-sampled from 1,000 Hz to 10,000 Hz by linear interpolation. The data were then filtered at CFC (Channel Frequency Class) 180.

2. 2 Head Motion based Brain Injury Criteria

From investigation of retrospective researches on brain injury criteria, seven 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 three injury criteria of HIC, HIP (Head Injury Power), and PRHIC (Power Rotational Head Injury Criterion) were defined in this study as described below.

SI (Severity Index) was proposed by Gadd (1966)[11] and was a precursor to HIC. The severity index was designed to have strong agreement with the WSUTC (Wayne State University Tolerance Curve: Patrick et al., 1963 [12]) and is shown in Eq. 1.

$$SI = \int a^n dt \le 1,000 \, , \quad \cdots \cdots \cdots \cdots \cdots \cdots \quad (1)$$

where a is either of acceleration, force, or pressure, which is a response function producing threshold of injury, and n is weighting factor equal to 2.5.

Versace (1971) [13] advanced upon SI and developed HIC, which is the current injury metric for head injury used in the FMVSS (Federal Motor Vehicle Safety Standards) 208 standard. The equation is represented as Eq. 2.

$$HIC = \left[\left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t)\, dt \right\}^{2.5} (t_2 - t_1) \right]_{\max} , \quad \cdot \cdot \quad (2)$$

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 HIC). Determining 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 msec at first, however current standards use 15 msec. HIC is good injury metric for skull fracture.[14]

Since the SI and HIC equations are mostly based on the WSUTC, the variables typically only consist of resultant linear acceleration; although Gadd did not limit the variations of variables.

Newman et al.[15] described HIP as a power expression of the human head when the kinematics of the head assumed as a rigid motion.

$$HIP = \sum m \cdot a_i \cdot \int a_i dt + \sum I_{ii} \cdot \alpha_i \cdot \int \alpha_i dt \, , \quad \cdots \cdot \quad (3)$$

where m is mass of the head (kg), a_i is linear acceleration (m/s^2), I_{ii} is MOI (moment of inertia) (kg·m^2), and α_i is angular acceleration (rad/s^2). 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.

The Eq. 3 includes both of linear and angular power terms. However, this study focuses on angular acceleration, so that linear term was omitted from consideration. Therefore, the angular $HIP\_ang(t)$ represents the rate of change in angular head kinetic energy, which is described as Eq. 4.

$$\begin{aligned} HIP\_ang(t) &= \sum I_{ii} \cdot \alpha_i \cdot \int \alpha_i dt \\ &= 0.016 \cdot \alpha_x \cdot \int \alpha_x dt + 0.024 \cdot \alpha_y \cdot \int \alpha_y dt \\ &\quad + 0.022 \cdot \alpha_z \cdot \int \alpha_z dt \end{aligned} \qquad \cdots\cdots\cdots\cdots (4)$$

The plots of maximum value of $HIP\_ang(t)$ versus the time duration of HIP_ang formulate similar distributions as SI or HIC definitions. This finding is similar to previous work [16] of investigating angular kinematics using animal models. Therefore, this study substituted $HIP\_ang(t)$ for resultant linear acceleration of $a(t)$ in the Eq. 1, and proposed a new injury criterion: the PRHIC as Eq. 5. For SI or HIC definitions, the power 2.5 was determined from tolerance curve using both injurious and non-injurious data plots. However, it is currently impossible to determine new power value for PRHIC since 6DOF sensor data did not have any concussive data. Therefore we assumed the power 2.5 for PRHIC as well.

$$PRHIC = \left[\left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} HIP\_ang(t)\, dt \right\}^{2.5} (t_2 - t_1) \right]_{max} \qquad \cdots\cdots\cdots\cdots (5)$$

An ultimate goal of this research is applying new injury criterion to automotive safety. Since HIC used 15 and 36 time durations for automotive safety standards in past history, we adopted same time durations into injury criteria of the head. NFL study reported that football impact has 15 to 20 msec time duration for the acceleration peaks.[17] In addition, since brain elements consist of viscous solid materials, peak of brain responses such as the first principal strain delays approximately 15 msec from that of angular acceleration. Therefore, time duration for PRHIC which is a brain injury predictor was selected as 36 msec.

2. 3 FE-based Brain Injury Predictors

An isolated human head brain FE model was used for this study (**Fig. 4**). The model consisted of 49,579 elements (24,096 solid, 25,119 shell and 364 seatbelt elements) had a mass of 4.39 kg and an appropriate position for the CG. Those inertia data of the head model were determined based on the anthropometry of AM50.[18] The skull model consisted of frontal, parietal, temporal, occipital, sphenoid, ethmoid, lacrimal, nasal, vomer, zygomatic, maxilla, mandible, and palatine bones. The skull sutures, which connected the cranial bones, were modeled with solid elements. The brain model consists of all hexagonal solid elements representing the cerebrum, cerebellum, brainstem with distinct white and gray matter, and CSF (cerebral spinal fluid). 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 model was already validated against a series of linear head impact[19] and two series of angular impacts [20, 21] (**Fig. 5**), and presented high bio-fidelity. Solid and gray lines represent the experimental data and simulation results, respectively. The origin is the

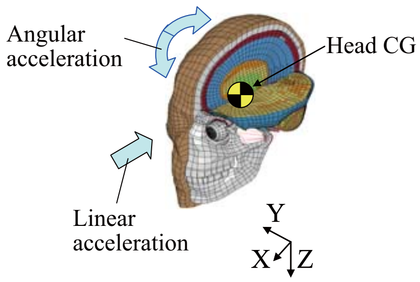

Fig. 4 Human head brain FE model and location of CG.

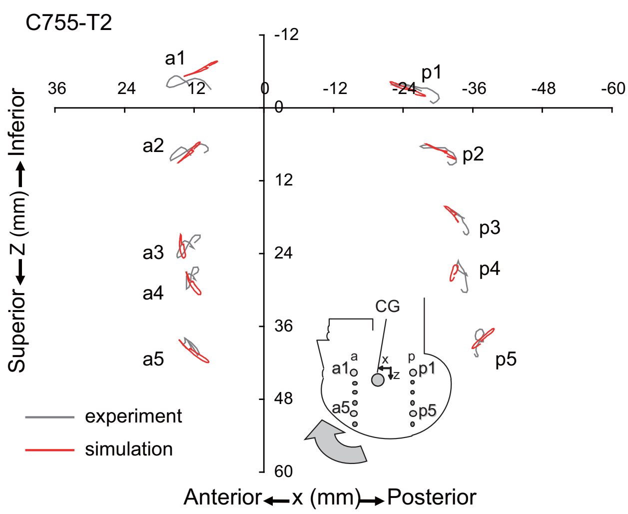

Fig. 5 Brain motions at various locations in a sagittal plane during Test C755-T2.[21]

CG of the head. The orientation of the head and the direction of angular acceleration are illustrated inside of the graph. Further details about this model are written in a previous publication.(22)

It was hypothesized that the skull deformation caused by head impacts below the threshold for fracture does not affect the brain motion. Since none of the head acceleration cases used for this study presented bone fracture, the skull of the model was assumed to be rigid. By assuming this, the measured 6DOF accelerations could be directly input as boundary conditions of the head through the skull (**Fig. 6**).

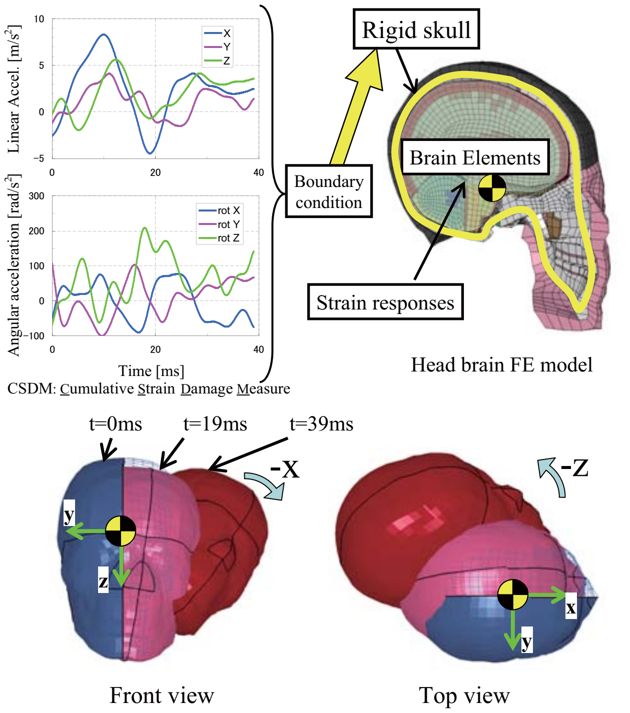

Fig. 6 FE analysis methods: 6DOF head accelerations were applied to a rigid skull model (upper). The resulting motion was described with respect to the local coordinate system for the model (lower).

All simulations were conducted by a commercially available FE solver, LS-DYNA 971 Rev. 2, SMP (Shared Memory Parallel) version (LSTC, Livermore, CA) using only single core of an Intel Xeon 64 bit based computer running on a Linux operating system.

Two injury predictors were calculated and analyzed in this study: the maximum first principal strain and CSDM utilized for evaluating DAI by Takhounts et al.(23) The CSDM was defined as the percent volume of the brain that exceeds a specified first principal strain threshold (**Fig. 7**). When the threshold of the first principal strain is set to 15%, the variable term is written as "CSDM 15%" in this study.

3. Results

3. 1 Validation for 6DOF Sensor

The linear relationship for resultant linear acceleration between the 6DOF sensor and Hybrid III data correlated strongly (R^2 = 0.88) (**Fig. 8**). Correlation of resultant angular acceleration was also significant (R^2 = 0.85). Error was calculated by dividing the difference in acceleration between the

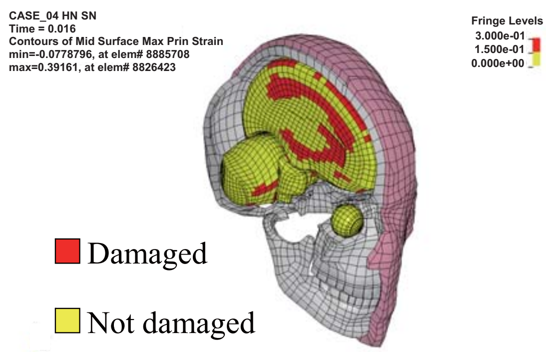

Fig. 7 A sample of strain distribution in human head brain FE model. Elements which experienced over 15% of the first principal strain were assumed as damaged.

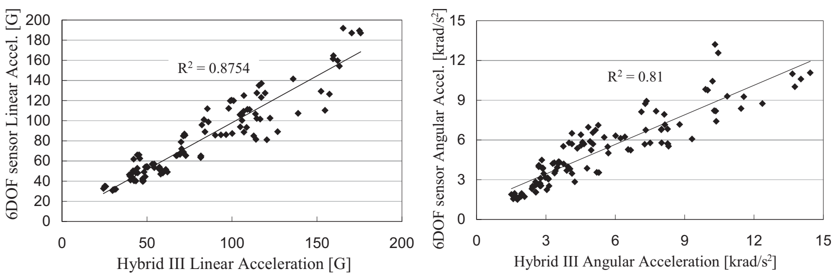

Fig. 8 Linear relationship between 6DOF sensor and Hybrid III in peak resultant linear (left) and resultant angular (right) acceleration.

6DOF sensor and the Hybrid III, by the Hybrid III acceleration. The average error between the 6DOF sensor and Hybrid III for the resultant linear accelerations was 1%. That for the resultant angular accelerations was 3%. Error was not significantly affected by impact location and direction.(7)

Referring other human subject testing devices and techniques, the NFL video analysis (24) was reported to have error as high as 15%; while the original HITS technology has an error of 8 ± 11% (mean ± SD).(25) Considering the vast amounts of data to be collected with the 6DOF sensor and the error levels of other accepted experiments, the inherent error of the 6DOF sensor is acceptable.

3. 2 Head Impact Data Collection from 6DOF Sensor

A total of 4,709 impacts were recorded during practices and games for the 19 instrumented players during the 2007 and 2008 Virginia Tech football seasons. No instrumented player sustained MTBI in this study. **Table 1** displays the frequency of impacts over specified resultant acceleration thresholds for linear and angular acceleration. For resultant linear acceleration, the majority of the impacts were under 20 g in severity. Of the 4,709 impacts, 38 were greater than 80 g, which is the nominal injury value derived by the NFL study. For resultant angular acceleration, roughly half of the impacts were less than 1,000 rad/s^2 in severity. Only 30 were greater than 6,000 rad/s^2, which is the nominal injury value derived in the NFL study. These data are described in detail by Rowson et al.(4)

Although moderate levels of both linear and angular forms of acceleration combined can often cause severe brain injuries,(26) it is generally believed that concussion mechanisms relate head impact severities and magnitudes of brain responses in impact biomechanics community. For example, NFL research group selected only severe impact events from numerious head impacts to reconstruct football head impacts.(9) Therefore this study also focused on impact severities of individual impacts without any considerations of numbers of head impacts or those accumulations which individual players earn during one-day football activity.

Margulies and Thibault(27) described injury probability and thresholds for DAI on a graph of angular acceleration versus angular velocity. In the same manner, **Fig. 9** shows data plots of 6DOF sensors with the thresholds, which are retrospectively obtained from literature data,(16,28,29) on the angular acceleration vs. angular velocity graph. Although 6DOF dataset does not contain any injurious cases, some cases were greater than the thresholds. In order to investigate the strong angular head impact data, this study selected fifty cases with high angular velocity from each of 2007 and 2008 datasets, and the total 100 sets of 6 DOF head accelerations were applied to a human head brain FE model.

3. 3 Human Head Brain FE Model Analysis

The 6DOF acceleration data were used to calculate seven injury criteria based on head motion of HIC$_{15}$, maximum linear velocity and acceleration, maximum angular velocity and acceleration, HIP, and PRHIC$_{36}$. In addition, analysis using the human head brain FE model presented FE-based brain injury predictors of the maximum first principal strain and CSDM 10%, 15%, and 20%. Each FE analysis for 40 ms termination

Table 1 Frequency of impacts above specified resultant acceleration thresholds.

Linear acceleration	2007	2008	Angular acceleration	2007	2008
> 0 g	1712	2997	> 0 rad/s^2	1712	2997
> 20 g	684	1128	> 1000 rad/s^2	875	1303
> 40 g	172	259	> 2000 rad/s^2	339	404
> 60 g	52	70	> 3000 rad/s^2	143	141
> 80 g	11	27	> 4000 rad/s^2	57	59
> 100 g	3	14	> 5000 rad/s^2	23	33
> 120 g	1	9	> 6000 rad/s^2	12	18
> 140 g	0	2	> 7000 rad/s^2	5	12
> 160 g	0	0	> 8000 rad/s^2	4	4
> 180 g	0	0	> 9000 rad/s^2	1	1

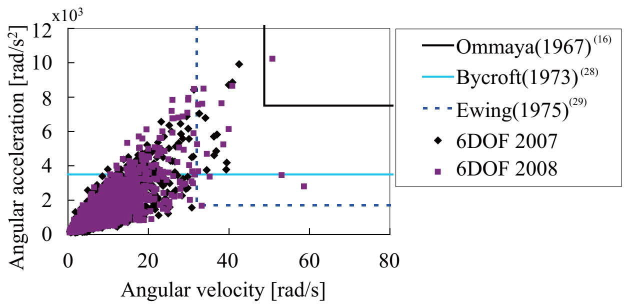

Fig. 9 Angular acceleration vs. angular velocity of 6DOF sensor data with threshold lines obtained from literatures.(16,28,29)

times took about four hours on a single core Xeon 3.0GHz CPU machine.

Table 2 shows correlations between variables of head injury criteria and FE-based brain injury predictors. The coefficient of determination (R^2) for HIC_{15} was less than 0.11 against FE-based brain injury predictors of cerebrum (**Fig. 10**). Similarly, maximum linear velocity, linear acceleration, angular velocity and HIP did not show any correlation with FE-based brain injury predictors. The maximum angular acceleration was somewhat correlated ($R^2 = 0.69$) with CSDM 10%. On the contrary, $PRHIC_{36}$, which this study newly proposed, was correlated well with CSDM values (Fig. 10). Specially, the correlation with CSDM 20% was significant ($R^2 = 0.84$).

4. Discussion

4. 1 Correlations with CSDM

The FE-based brain injury predictors did not show any correlations against HIC, linear velocity, linear acceleration, and HIP. The HIC, which is defined by integrated linear acceleration as Eq. 2, is believed as an injury predictor for skull fractures. The HIP definition includes not only angular accelerations but also linear acceleration terms (Eq. 3). Therefore, linear acceleration can be believed as insensitive to the maximum first principal strain or CSDM of the cerebrum.

On the contrary, angular acceleration shows stronger correlations with the maximum first principal strain and CSDM than linear acceleration. Definition of PRHIC contains both of angular velocity and acceleration, so that the correlations with FE-based brain injury predictors would be strong. Therefore, this study can suggest that two injury predictors are needed for head injury: one is based on linear accelerations for skull fracture, and another is based on angular accelerations for brain injury.

The CSDM was proposed as a predictor of DAI by using brain FE models.[23] In such injuries, the

Table 2 Coefficient of determinations (R^2, upper) and p-values (lower) between FE-based brain injury predictors and head injury criteria.

N = 100	HIC_{15}	Max. linear velocity	Max. linear acceleration	Max. angular velocity	Max. angular acceleration	HIP	$PRHIC_{36}$
Max. first principal strain	0.11 *p=0.0009*	0.11 *p=0.0004*	0.20 *p=4.0e-6*	0.23 *p=4.7e-7*	0.30 *p=4.1e-9*	0.25 *p=1.5e-7*	0.25 *p=1.4e-7*
CSDM 10%	0.04 *p=0.059*	0.001 *p=0.67*	0.10 *p=0.002*	0.25 *p=1.5e-7*	0.69 *p=5.8e-27*	0.18 *p=9.3e-6*	0.64 *p=3.1e-23*
CSDM 15%	0.01 *p=0.24*	0.003 *p=0.59*	0.06 *p=0.018*	0.28 *p=2.0e-8*	0.41 *p=7.9e-13*	0.12 *p=0.0003*	0.75 *p=0.00*
CSDM 20%	0.02 *p=0.15*	0.008 *p=0.38*	0.06 *p=0.011*	0.27 *p=3.1e-8*	0.32 *p=1.0e-9*	0.14 *p=0.0002*	0.84 *p=0.00*

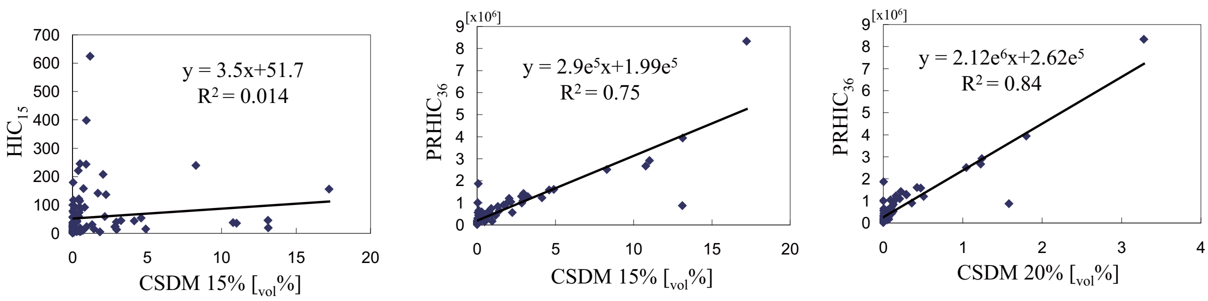

Fig. 10 Injury criteria of HIC_{15} and $PRHIC_{36}$ vs. FE-based brain injury predictors of CSDM 15% and CSDM 20%.

damaged tissue is expected to be spread throughout a large volume in the cerebrum. In FE analysis using CSDM, the smaller first principal strain level at which an element is considered damaged, the higher the predicted volume of damaged brain elements becomes. For example, Fig. 10 displays the difference of maximum abscissa values between CSDM 15% and CSDM 20%, where the maximum value of CSDM 15% is 17.2 vol%, which is greater than that of CSDM 20% as 3.2 vol%. However, an important observation was done when analyzing the statistical correlation of CSDM with other injury predictors. For high strain threshold (20%), the strongest correlation between CSDM and PRHIC was found ($R^2 = 0.84$). For such CSDM values, the volume of damaged elements is small. Since Traumatic Brain Injuries sustained by football players are usually milder than DAI, a further verification for strain thresholds is required to define MTB injuries.

4. 2 Limitation

This study obtained over four thousand impact data using 6DOF sensors. However, it is impossible to earn a threshold for brain injury using only 6DOF dataset because concussion cases were not included. In addition, 6DOF football data did not have skull fracture, while automotive crashes would contain skull fractures in head injury. Therefore, further verification studies are needed for CSDM and PRHIC using other head impact data including skull fractures.

5. Conclusions

Following four specific conclusions can be drawn for the investigation of brain response simulations for angular head impacts using a human brain FE model.

- The 6DOF sensor showed a strong linear correlation when validated against head acceleration of a Hybrid III dummy.
- Total 4,709 impacts were collected by instrumenting Virginia Tech football players' helmets with the 6DOF sensor through the 2007 and 2008 football seasons.
- $PRHIC_{36}$ which is calculated from integrated angular acceleration and angular velocity was proposed in this study.
- Although 6DOF data did not contain any concussion cases, $PRHIC_{36}$ presented strong correlation with CSDM 20%.

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