Development of a Rotational Brain Injury Criterion with Consideration of the Direction and Duration of Head Rotational Motion

Y. Miyazaki, K. Ishiwa, M. Kitagawa, M. Ueno, S. Sugimoto, R. Asahi, H. Matsumura

I. INTRODUCTION

Traumatic brain injury is the primary cause of death in traffic accidents. Brain injuries, such as concussions or diffused axonal injuries, can cause deformation of deep brain structures due to rotational movement of the head. Therefore, many injury criteria based on angular acceleration and/or velocity of the head have been proposed based on reconstructions of head impacts using finite element (FE) head models [1-3]. However, the relationship between kinematic motion of the head and strain responses of the deep brain obtained from an FE model can be affected by the modelling methods used for the complex characteristics of intracranial structures, such as skull-brain boundary conditions, and constraint due to the falx or tentorium. Our previous studies, using a three-dimensional (3D) physical model of an FE model which allows for relative motion between the skull and brain, indicated that the skull-brain relative rotation and its constraint by falx were key mechanisms for generating strain in deep brain structures [4-5]. Therefore, the purpose of this study was to develop a criterion for rotational brain injury, with consideration for the effect of the rotational direction and duration of the head, based on simulations using an FE model capable of skull-brain relative rotation, and to show preliminary results regarding a comparison of the proposed injury criterion with injury criteria that were previously reported.

II. METHODS

Finite Element Model of a Head

The head component of THUMS Ver. 4.0.2 AM50 was used as the basic FE model. In order to allow relative rotational motion between the skull and brain, the FE head model was modified to set sliding contacts between the dura and arachnoid cerebrospinal fluid (CSF) components, and between the arachnoid CSF and the pia components. This definition of sliding interfaces was validated against results from simple side impact tests using a 3D physical model of the head [4]. Shear strain at the corpus callosum (CC) in the modified FE model was 3-5 times higher than the basic model with fixed boundaries, and the maximum strain and duration of the strain response was within the range of the experimental results. Moreover, the complex arachnoid CSF layer was defined as a viscoelastic material, with the properties used by [6]. Four combinations of the brain material properties described by [7] were evaluated in the post-mortem human subject (PMHS) tests, by comparison to the relative displacement of the skull and brain [8]. From these results, the material properties were set as G_0 = 12500 [Pa], G_∞ = 6125 [Pa], and β = 0.06, with a CORA score of 0.498 in C755T2 and 0.392 in C383T1.

Investigation of Relationship between Brain Strain and Direction and Duration of Head Rotational Motion

Half sinusoidal angular velocity time histories, with 40 [rad/s] magnitude, were applied to the head model while changing its direction and duration Δt. The rotational directions were set as Coronal (X), Extension (+Y), Flexion (-Y), and Horizontal (Z) axes. The durations chosen were 24, 36, 48, 60, 72, 84, and 96 [ms]. Maximum principal strain (MPS) in the CC, cerebral white matter (CWM), and CSDM 0.10 were obtained as brain strain metrics. From the results of the simulations, a new rotational brain injury criterion (RBIC), involving peak change in angular velocity around each orthogonal axis $\Delta\omega$ and its time duration Δt and rotational direction, is proposed in the following equation. The weight functions $f(\Delta t)$ are determined by the relationship between the brain strain metrics and direction and duration of the head rotational motion obtained from the simulations.

$$\mathrm{RBIC} = \left.\sqrt{(fx(\Delta t_x)\Delta\omega_x)^2+\left(fy\left(\Delta t_y\right)\Delta\omega_y\right)^2+(fz(\Delta t_z)\Delta\omega_z)^2}\right|_{max} \quad (1)$$

Head Kinematics Data from Football Impact and Vehicle Crash Data

A total of 24 cases of 6-DOF head kinematic data of football impact/collision [9-10], with the exception of case number 69-2, from Newman [9], whose angular velocity profile did not reach the maximum value within 20 ms, were used to examine the relationship between concussion and brain strain metrics, which include MPS in the

Y.Miyazaki is an Associate Professor in the Department of Control and Systems Engineering at Tokyo Institute of Technology, Japan (+81-3-5734-3155 / y-miyazaki@sc.e.titech.ac.jp). K. Ishiwa and M.Kitagawa are MEng Students in Miyazaki Lab at Tokyo Institute of Technology, Japan. M. Ueno, S. Sugimoto, R. Asahi, H. Matsumura belong to Mazda Corporation, Japan.

CC, CWM, and brainstem (BS), and CSDM 0.10, estimated by a receiver operating characteristic (ROC) curve. In addition to the football data, NHTSA crash test data using THOR or World SID dummy were used to examine relationships between brain strain metrics and head kinematic based metrics. The total number of cases for each test, without calculation errors, was 56, including nine frontal rigid barrier cases, nine right oblique impact cases, 20 left oblique impact cases, 12 moving deformable barrier cases, and nine pole collision cases. The HIC, BrIC, RIC, PRHIC, and RVCI were then compared with the RBIC.

III. Preliminary Results

Figure 1 demonstrates the MPS at the CC, in cases with half sinusoidal angular velocity input. In the shorter durations, the resulting rotation around the X-axis was the highest; however, the extension motion cases (Y-axis) were highest with a longer duration. These relationships between brain strain metrics and rotational direction and duration were set as weight functions for the RBIC model. Note that Δt in Figure 1 should be divided by 2 when the relationship is used as the weight function for RBIC equation, since Δt in RBIC equation denotes duration of Δω. Values for the area under the curve (AUC) in the ROC curves from the football data resulted in the following high accuracies: MPS in the CC = 0.922, MPS in the CWM = 0.933, and CSDM0.10 = 0.911. In kinematics data from a total of 80 heads, RBIC showed the best overall correlation with the MPS in the CC (Figure 2). This trend was same as other brain strain metrics, although the correlation was lower than MPS in CC.

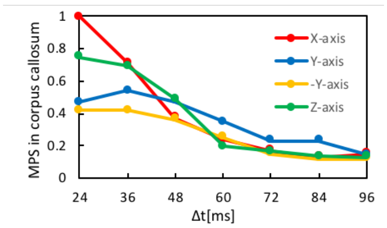

Fig. 1. MPS at CC in the cases of simple half sinusoidal angular velocity profiles.

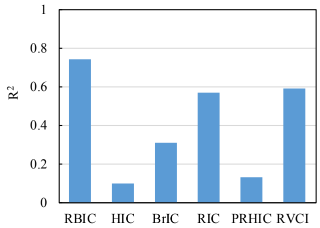

Fig. 2. Correlation between MPS in CC and kinematic based metrics in all head kinematic cases.

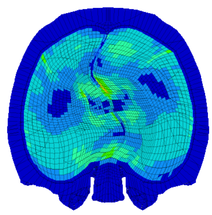

Fig. 3. MPS distribution of cerebrum on coronal plane in case of half sinusoidal angular velocity with 24 ms duration.

IV. Discussion

In the case of rotational motion around the x axis, strain distributions near the CC were affected by the sliding motion of the cerebrum near the CC occurring with falx bending (Figure 3). Therefore, because the mechanism of strain distribution near the CC is thought to depend on dynamic bending characteristics of the falx, which has a higher rigidity than the cerebrum, the MPS in rotational motion around the x axis decreased with a longer time duration compared with those around the y axis motion. Therefore, a new rotational brain injury criterion, in consideration of this strain distribution mechanism, depending on rotational direction and duration was proposed, and the RBIC showed a higher correlation than did other head kinematics-based criteria. Note that the weight function totally depends on which FE model is used. The R^2 value of the HIC, BrIC, and RVCI model results in this study tended to be lower than other studies, which used considerably more head kinematics data [11-12]. Although one explanation for this is thought to be the difference in the modelling methods for the FE models, such as the boundary conditions between the skull and brain, it should also be noted that the sample size of the head kinematics data in this study is much smaller than the previous studies. This might explain why the AUC values in this study were so high. Therefore, to confirm the predictive performance of the new criterion, RBIC, an increase in the sample size for the head kinematics data is essential.

V. References

[1] Takhounts et al., STAPP J., 2013
[2] Kimpara et al., Ann. BioMed. Eng., 2012
[3] Yanaoka et al., SAE Technical Paper, 2015
[4] Miyazaki, JSAE, 2013
[5] Miyazaki et al., JSAE, 2016
[6] Takhounts et al., STAPP J., 2008
[7] Nakahira et al., JSME Conf., 2015
[8] Hardy et al., STAPP J., 2001
[9] Newman et al., STAPP J., 2000
[10] Hernandez et al., Ann. BioMed. Eng., 2015
[11] Gabler et al., Ann. BioMed. Eng., 2016
[12] Gabler et al., J. Biomech., 2018