A STUDY OF INJURY CRITERIA FOR BRAIN INJURIES IN TRAFFIC ACCIDENTS

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ABSTRACT

The goal of this study was to develop a motion-based injury criterion for brain injuries derived from the material response of the brain tissue, under the assumption that impact response of the brain tissue can be characterized by a standard linear solid. Focus was given to brain injuries that are deemed to correlate with the strain of the brain tissue, including subarachnoid hemorrhage, intracerebral hemorrhage and diffuse axonal injury. The criterion is based on rotational motion of the head because of incompressibility of the brain tissue that allows large strain primarily in rotation.

The stiffness and damping parameters of one-dimensional Kelvin model were determined for each axis of rotation of the head in such a way that scaled displacement time history matches strain time history of the brain tissue predicted by the Global Human Body Models Consortium (GHBMC) head-brain model. The convolution integral of the impulse response of the model was used to predict strain time history of the brain when an arbitrary rotational acceleration time history is applied to the head. The maximum value of the predicted strain was defined as a new brain injury criterion (Convolution of Impulse response for Brain Injury Criterion; CIBIC). Head rotational acceleration data were taken from a number of crash test data representing full frontal, oblique frontal and side impacts along with pedestrian impact simulation results to investigate correlation between the values of various brain injury criteria, including CIBIC, and the maximum principal strain from the head-brain model.

The injury criterion proposed by this study, CIBIC, resulted in a better correlation with the predicted maximum principal strain of the brain relative to those proposed by past studies in all of the four crash configurations (R^2 ranging from 0.624 to 0.864). It was also found that the coefficient of determination was smaller for the impact conditions resulting in multiple or long-duration loading than other impact configurations representing single short-duration loading.

INTRODUCTION

Head injuries account for a significant percentage of fatal injuries due to traffic accidents. The data from the National Automotive Sampling System (NASS) Crashworthiness Data System (CDS) from 2010 to 2014 and Pedestrian Crash Data Study (PCDS) from 1994 to 1998 show that the head respectively comprises 33% and 46% of all body regions sustaining Maximum Abbreviated Injury Scale (MAIS) in fatal accidents. The results present that head injury is the most frequent cause of death in both car and pedestrian crashes. Of those head injuries, brain injury accounts for 78% and 81% of the head injuries responsible for the death for the data from NASS CDS and PCDS, respectively, showing that mitigation of brain injury is crucial to further reduce the number of traffic fatalities.

Brain injury consists of a number of different damage patterns, including brain contusion, epidural hematoma, subarachnoid hemorrhage, intracranial hemorrhage, diffuse axonal injury and subdural hematoma. Based on the tissue failure and anticipated injury mechanisms, those injury types can be classified into three major categories by the primary cause of injury; pressure and/or skull fracture (brain contusion, epidural hematoma), brain strain (subarachnoid hemorrhage, intracranial hemorrhage and diffuse axonal injury) and displacement relative to the skull (subdural hematoma). Classification of the brain injury data from NASS CDS and PCDS described above into these three categories showed that brain injuries primarily due to strain in the brain are by far most frequent, accounting for 81% and 73% of all brain injuries for the datasets from NASS CDS and PCDS database, respectively. For this reason, the current study focused on predicting strain in the brain using a motion-based injury criterion to be used with crash test dummies without representation of the brain. Holbourn et al. [1] hypothesized that the shear strain in the brain primarily due to the rotational acceleration of the head is a predominant cause of brain damage due to large bulk modulus of the brain substance compared with its modulus of rigidity. In accordance with this assumption, it was decided to investigate a brain injury criterion based on the rotational motion of the head.

A number of past studies have focused on the development of a brain injury criterion based on the rotation of the head. Kimpara et al. [2] proposed a combination of Rotational Injury Criterion (RIC) and Power Rotational Head Injury Criterion (PRHIC), where RIC and PRHIC are based on the formulation of Head Injury Criterion (HIC) proposed by Versace [3] and Head Impact Power (HIP) proposed by Newman et al. [4], both of which were essentially developed using an empirical approach. Takhounts et al. [5] hypothesized that rotational velocity, not rotational acceleration, is the mechanism for anatomic brain injuries, and proposed Brain Injury Criterion (BrIC) based on statistical analyses correlating tissue-level injury criterion predicted by human FE models with the kinematics-based measure. Although those proposals were tested against vast amount of experimental data, they are not necessarily based on mechanical characteristics of brain response. Yanaoka et al. [6] represented mechanical response of the brain with a linear spring, and hypothesized that the strain in the brain is proportional to rotational acceleration to propose Rotational Velocity Change Index (RVCI). Gabler et al. [7] subsequently represented brain response with the Voigt model (parallel combination of a linear spring and a dashpot) to develop iso-strain angular acceleration-angular velocity curves to clarify the influence of these parameters. Despite the fact that these studies attempted to come up with a generalized criterion derived from fundamental mechanics rather than using an empirical curve-fit against experimental data, the mechanical models used in these studies did not represent a generalized viscoelastic material.

The objective of this study was to develop a motion-based injury criterion for brain injuries derived from the material response of the brain tissue, under the assumption that impact response of the brain tissue can be characterized by a standard linear solid. The idea behind the current development of a criterion was to use a generalized analytical solution of a simplified representation of the mechanical response of the brain, rather than choosing physical parameters used in the formulation of the criterion, as has been done in past studies. A Kelvin model (generalized linear solid) analogous to rotational response of a human FE head-brain model was identified, and the convolution integral of the impulse response of the identified model was used to predict strain in the brain. The maximum value was defined as a new brain injury criterion, and correlation between the peak value of maximum principal strain (MPS) in the brain and existing rotational brain injury criteria, including the one proposed by this study, was investigated in multiple loading configurations, including full frontal, oblique frontal and side impacts of a car as well as pedestrian impacts.

METHODS

Identification of Analogous 1D Model

In this study, it was hypothesized that the peak value of MPS predicted by the Global Human Body Models Consortium (GHBMC) 3D FE head-brain model developed by Mao et al. [8] (3D model) correlates with probability of brain injuries primarily due to strain in the brain parenchyma. As 3D model represents the brain using a standard linear model, the material parameters for an analogous 1D Kelvin (standard linear) model (1D model) was identified to represent rotational response of 3D model. As the model represents a linear time-invariant system, the response to an arbitrary input is described in the form of convolution integral of the impulse response. Therefore, the material parameters for 1D model were determined such that the strain response of 3D model to the impulse of rotational acceleration is represented by 1D model.

Since the impulse (Dirac delta function) cannot be applied to 3D model, it was first assumed that a step function with a 1 ms duration well represents the impulse, and then the peak value of the response was compared against analytical solution to confirm validity of the choice of 1 ms. Figure 1 shows the schematic of these models. Due to directional dependency of the rotational response of 3D model, the material parameters for 1D model (spring coefficients k_1 and k_2, damping coefficient c) were determined for each axis of rotation. In order to obtain a certain magnitude of strain, the magnitude of the step function was set at 10,000 rad/s^2. The mass was set at 1 kg for simplification. The material parameters were determined by minimizing root mean square error of the strain time history normalized by the peak value. Due to the use of a predetermined mass, the magnitude needed to be scaled such that the value predicted by 1D model coincides with the strain from 3D model for the same duration and magnitude of the step input using the ratio of peak values between the step responses from 1D and 3D models.

Convolution of Impulse Response for Brain Injury Criterion (CIBIC)

The analytical solution of the impulse response of the Kelvin model as described in Figure 1 is given by the following formula:

$$x(t) = d_1 e^{-a_1 t} - e^{-a_2 t}\{d_1 cos(bt) + d_2 sin(bt)\} \tag{1}$$

where

$$a_1{}^3 + \frac{k_2}{c}a_1{}^2 + \frac{k_1 + k_2}{m}a_1 + \frac{k_1 k_2}{cm} = 0,\ \ a_1 \in \mathbb{R} \tag{2}$$

$$a_2 = \frac{\frac{k_2}{c} - a_1}{2} \tag{3}$$

$$b = \frac{1}{2}\sqrt{\frac{4(k_1 - k_2)}{m} - 2a_1\frac{k_2}{c} + 3a_1{}^2} \tag{4}$$

$$d_1 = \frac{a_1 - \frac{k_2}{c}}{\left\{2a_1\frac{k_2}{c} - 3a_1{}^2 + \frac{(k_1 + k_2)}{m}\right\}} \tag{5}$$

$$d_2 = \frac{a_1{}^2 + a_1\frac{k_2}{c} - a_1 a_2 + a_2\frac{k_2}{c} - \left(\frac{k_2}{c}\right)^2 + \frac{(k_1 + k_2)}{m}}{\left\{2a_1\frac{k_2}{c} - 3a_1{}^2 + \frac{(k_1 + k_2)}{m}\right\}b_2} \tag{6}$$

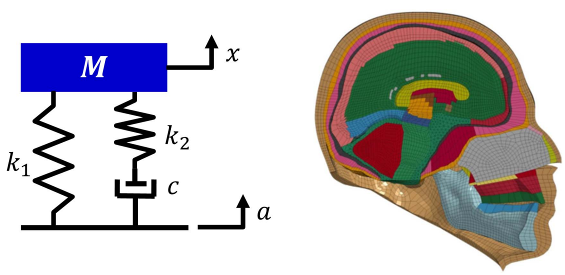

Figure 1. Schematics of 3D (GHBMC head-brain) model and 1D (standard linear solid) model.

The convolution integral of Equation (1) provides a strain time history for any given rotational acceleration time history for each of the rotational axes. Root mean square of the individual prediction for each of the three axes was used to estimate strain response to simultaneous application of rotational acceleration in three directions. As this assumes independence of the response in three directions, strain time histories from 3D model were compared between root mean square of individual response (RMS) and that obtained by applying rotational acceleration simultaneously in three directions (3D). The same step function as that used to identify 1D model (1 ms duration, 10,000 rad/s^2 magnitude) was employed. Figure 2 compares the two time histories of MPS, showing that RMS of each of the strain time histories in the three rotational directions provides reasonable approximation of the strain time histories in 3D head rotation. A new motion-based brain injury criterion proposed by this study (Convolution of Impulse response for Brain Injury Criterion; CIBIC) is given by the following formula:

$$CIBIC = \sqrt{\sum_{i=1}^{3}\left\{\int_0^t x_i(t-\tau)\alpha_i(\tau)d\tau\right\}^2}\Bigg|_{max} \tag{7}$$

where i=1,2,3 represent the x, y and z axis and α_i is rotational acceleration. This formulation simply approximates maximum strain in the brain, without choosing particular physical parameters and determining constants in the formulation by empirically correlating the measure against experimental data.

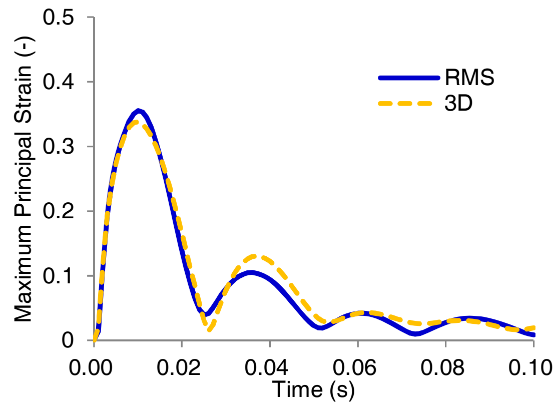

Figure 2. Comparison of time histories of MPS.

Correlation Analysis

Predictive capability of CIBIC was evaluated by analyzing correlation between the peak value of maximum principal strain (MPS) from 3D model and CIBIC. Some of the existing criteria were also compared. Head rotational acceleration data were taken from a number of crash test data representing full frontal, oblique frontal and side impact test results along with pedestrian impact simulation results. Full frontal, oblique frontal and side impact test data were obtained from the National Highway Traffic Safety Administration (NHTSA) Vehicle Crash Test Database. By eliminating the tests for which a complete set of 3 linear and 3 angular acceleration data is not available, 62 US NCAP full frontal tests at 56.3 km/h, 44 oblique impact tests at 90 km/h (15 degrees of impact angle and 35% of overlap) and 53 US NCAP moving deformable barrier (MDB) side impact tests at 61.9 km/h were used. For all of these three impact configurations, head linear and angular acceleration time histories measured by the crash test dummy mounted on a driver seat were used. Due to the lack of such a publicly available crash test database for pedestrian impact, the time histories were obtained by running MADYMO impact simulations as conducted by Yanaoka et al. [6]. In their study, a pedestrian model representing 50th percentile male was hit by a simplified car model for a sedan represented by a combination of a rigid surface and a linear spring. 36 pedestrian impact configurations were developed by applying L36 orthogonal array to the combinations of different levels of pedestrian gait cycle, pedestrian walking speed, pedestrian orientation, car speed, car deceleration and stiffness of the bumper, hood edge, hood and windshield. As the rage of the peak value of MPS obtained by the study was smaller than other impact configurations, this study developed an additional set of 36 impact configurations by following the same procedure and applying increased variability of car stiffness characteristics, and used the results of 72 pedestrian impact simulations in total. For each case, CIBIC was calculated from the head rotational acceleration time histories using Equation (1), and MPS was calculated by applying the linear and angular time histories in the x, y and z directions to 3D model. Correlation between the peak value of MPS and each of the injury criteria was evaluated using coefficient of determination (R^2). In addition to CIBIC, BrIC (Takhounts et al. [4]), RIC and PRHIC (Kimpara et al. [1]) and RVCI (Yanaoka et al. [6]) were compared.

RESULTS

Identification of Analogous 1D Model

Table 1 shows the material parameters (spring coefficients k_1 and k_2, damping coefficient c) identified for the x, y and z axis. The scaling factor for each axis is also presented. Figure 3 shows the comparison of MPS time histories from 3D and 1D models when a step function (1 ms duration, 10,000 rad/s^2 or m/s^2 magnitude) was applied. The results show that the initial peak was well represented by 1D model, while there are some discrepancies in a later phase of the response. In general, the response of 1D model tended to attenuate quicker than 3D model. When the material parameters presented in Table 1 were used, the peak strain calculated from the analytical solution of the impulse response was 0.18819, 0.17346 and 0.25841, while that estimated by 1D model was 0.18808, 0.17334 and 0.25822, in the x, y and z directions, respectively. The difference between the analytical solution and the 1D model prediction was less than 0.1% for all directions (0.056%, 0.068% and 0.071% in the x, y and z directions, respectively), showing the validity of the use of 1 ms duration step function to approximate impulse input for the 1D model used in the current study.

Table 1. Material parameters and scaling factor identified for 1D model

Direction	k_1 (kN/m)	k_2 (kN/m)	c (Ns/m)	Scaling factor (1/m)
x	12.76	22.67	129.1	0.00313
y	16.39	31.63	120.4	0.00395
z	17.04	47.52	74.40	0.00494

Correlation Analysis

Figures 4 through 7 show the correlation between the peak value of MPS from 3D model and CIBIC calculated from the rotational acceleration time histories using Equation (7) and material parameters in Table 1 for full frontal, oblique frontal, side and pedestrian impact configurations, respectively. Figure 8 shows the same plot containing all the data points. R^2 was 0.828 for all data. Figure 9 illustrates the comparison of R^2 between different injury criteria, including BrIC, RIC, PRHIC, RVCI and CIBIC. CIBIC provided the largest value of R^2 for all the impact configurations, while the value tended to be smaller for oblique frontal and pedestrian impacts, compared to full frontal and side impacts.

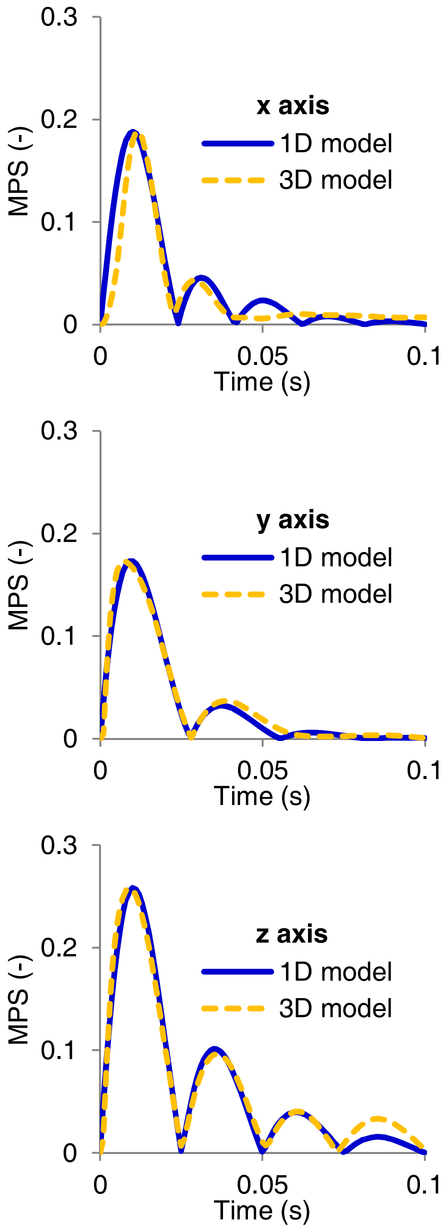

Figure 3. Comparison of MPS response to impulse input between 1D and 3D models.

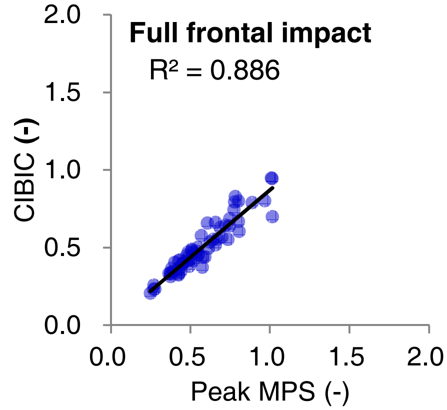

Figure 4. Correlation between peak MPS and CIBIC (full frontal impact).

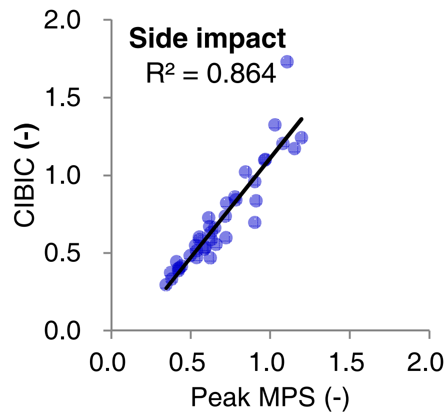

Figure 5. Correlation between peak MPS and CIBIC (side impact).

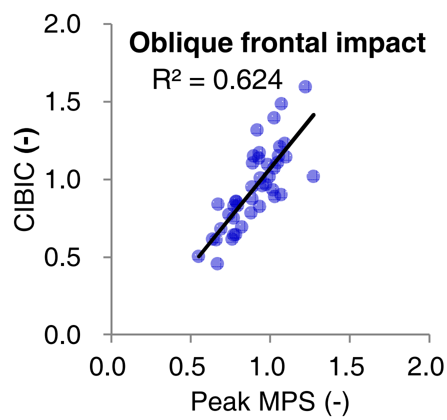

Figure 6. Correlation between peak MPS and CIBIC (oblique frontal impact).

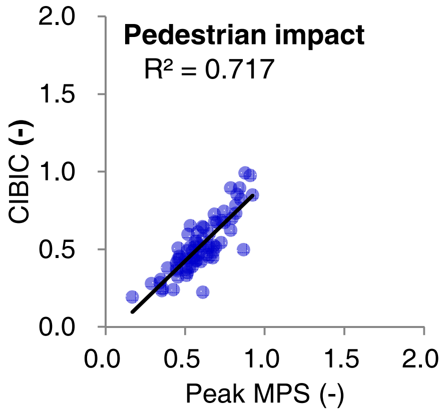

Figure 7. Correlation between peak MPS and CIBIC (pedestrian impact)

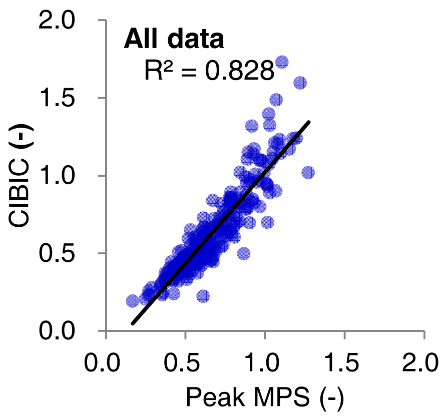

Figure 8. Correlation between peak MPS and CIBIC (all data)

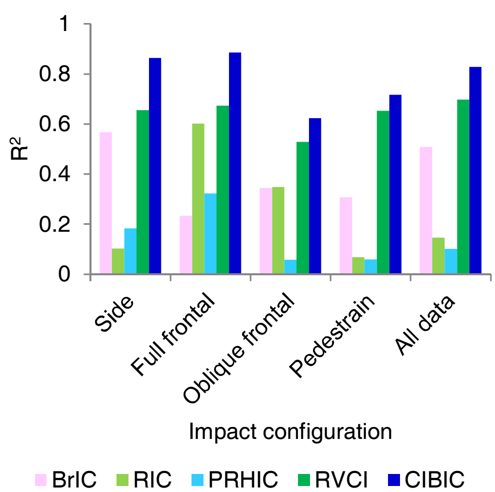

Figure 9. Comparison of coefficient of determination.

DISCUSSION

This study developed an injury criterion, CIBIC, which was found to correlate with the peak value of MPS from a human FE head-brain model in multiple impact configurations, including full and oblique frontal impact, side impact and pedestrian impact. The unique feature of the proposed criterion is that it does not depend on any specific physical parameters and an empirical curve fit against experimental data, but simply describes an analytical solution of the brain response when linear viscoelastic response characteristics are assumed. The results of the correlation analysis as presented in Figure 9 clearly show the advantage of this concept. A similar approach can be found in Bandak et al. [9], where an injury assessment tool called a Simulated Injury Monitor (SIMon) was proposed to predict probability of brain injuries. This approach applies measured crash dummy responses to a simplified human FE model of a specific body region to analyze its detailed structural response, without relying on an empirical approach. Thanks to the use of 3D human FE model, Bandak et al. [9] employed three different injury metrics (Cumulative Strain Damage Measure; CSDM, Dilatation Damage Measure; DDM, Relative Motion Damage Measure; RMDM) to address different types of brain injuries primarily induced by deformation, pressure and motion relative to the skull, respectively. Although the current study focused only on the first type of brain trauma, a simple calculation of the analytical solution as described in Equation (7) was found to provide a reasonable approximation of the strain in the brain, with the advantages of much shorter computational time and easier handling of the assessment tool.

The results of the correlation analysis presented in Figure 4 clearly showed that CIBIC works better for full frontal and side impacts compared to oblique frontal and pedestrian impacts. As shown in Figure 3, some differences of the impulse response were found between 3D and 1D models in the attenuation of the strain. As 1D model represents the same material response as that of the brain substance, this discrepancy would come from the simplification of the model, such as the lack of representation of the shape and boundary conditions of the brain, and the influence of the cerebrospinal fluid (CSF). In pedestrian impact, the head is subjected to long duration of low acceleration before it hits the surface of a car. In oblique frontal impact, the head may contact the top of the interior door trim panel, followed by contact against the roof liner. Due to the difference in the attenuation of the strain, an error in the 1D model prediction would become larger as the input tends to be low magnitude-long duration and/or multiple. Figure 10 compares the time histories of MPS from 3D model with the prediction from 1D model (CIBIC). The initial peak was found to be accurately simulated, while the second peak was much lower for the 1D model prediction, probably due to the quicker attenuation of the strain predicted by 1D model as shown in Figure 3. Although this discrepancy requires further modifications, 3D model used in this study also needs to be further validated before modifying the injury criterion, since the validity of the brain injury criterion proposed by this study solely depends on the validity of 3D model against which brain rotational response characteristics were matched. 3D model requires further validations in terms of detailed geometry of specific components, material response (constitutive model and material parameters) and structural response.

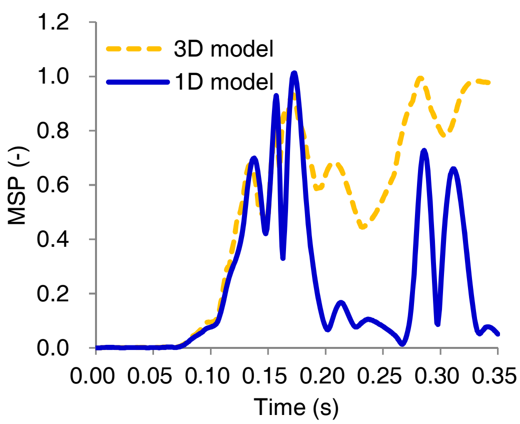

Figure 10. Comparison of time histories of MPS from 1D and 3D models.

CONCLUSIONS

A brain injury criterion was developed by assuming linear viscoelasticity of rotational response of the brain and using the convolution integral of the impulse response of the linear viscoelastic model. The criterion was found to better predict maximum principal strain of the brain predicted by an FE head-brain model in car and pedestrian impacts.

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