

VALIDATION OF THE HUMAN HEAD FE MODEL AGAINST PEDESTRIAN ACCIDENT AND ITS TENTATIVE APPLICATION TO THE EXAMINATION OF THE EXISTING TOLERANCE CURVE

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ABSTRACT

The existing head injury criterion (HIC) is based on translational accelerations of the center of gravity of the head, while the importance of the rotational motion has been discussed over decades. Although most previous studies to establish injury criteria have depended on cadaveric, human volunteer or animal experiments, ever-progressing computing techniques both in software and hardware are making such studies possible using virtual experiments by FE simulation. In reality, however, such simulation models must be fully validated against a real human body before they can be used for such studies, and these validations depend on the tests using animal and human material.

In this paper, another approach to validate a human head FE model is introduced. Two cases of pedestrian accidents were selected from the accident database of the Road Accident Research Unit of the University of Adelaide and were reconstructed using a combination of physical testing and a FE model of the pedestrian/vehicle collision. The results of the FE model of the head were compared with the neuropathology of the actual victims to see if such an index as maximum principal strain was a correlate of the location and severity of injury.

After a comparison between the results of the model and the neuropathology was made, a tentative application of the model was tried. A parametric study on translational acceleration and duration time was performed and the relationship between the simulated

brain conditions and the existing head tolerance curve (WSUTC) were discussed. Finally, additional simulations where pure rotational motions were applied to the model showed the likelihood of injuries from these motions alone. From this, the need for a criterion that considers both translational and rotational motions was suggested.

INTRODUCTION

HIC has widely been used to estimate the severity of head impact among various types of experiments, representing motor vehicle collision, pedestrian impact and other accidental impacts to measure the probability of traumatic head injury in each case. HIC was first introduced by Versace[1] to represent the Wayne State Tolerance Curve. The curve was first presented by Lissner et al.[2] and then published by Gurdjian et al.[3] and Patrick et al.[4] in the form shown in Fig.1. Versace[1]'s representation was modified by NHTSA as the following expression:

$$HIC = \left\{ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right\}^{2.5} (t_2 - t_1)$$

where t_2 and t_1 are times during the acceleration pulse and $a(t)$ is a resultant linear acceleration. Since HIC was introduced in FMVSS208 by NHTSA in 1972, it has become a global standard for the criteria of head injury.

Meanwhile, head injury mechanisms were

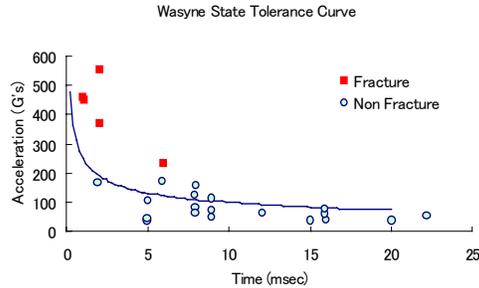


Fig.1 WSU Tolerance Curve

illustrated by Ommaya et al.[5] that both translational and rotational loading on the head caused brain injury and that especially diffuse injury was caused mainly by rotational motion. The desire to incorporate angular acceleration into criteria for head injury comes from the generally accepted hypothesis that diffuse injuries occur largely in response to the stresses and strains generated by angular loads. Diffuse injuries, such as diffuse axonal injury (DAI), are responsible for a large proportion of the mortality and morbidity from head trauma. Ommaya et al.[6] presented the tolerance curve for angular acceleration versus time duration for cerebral concussion in whiplash motion derived from primate experiments. Margulies et al.[7] proposed tolerance curve for angular acceleration versus angular velocity derived from animal experiments, which was used to estimate the influence of a newly developed side air curtain for angular motion of the head[8]. Another approach was tried by Newman et al.[9] to develop a criterion called Head Injury Power (HIP) that combined all six components of linear and angular accelerations and time into one formula. Though these researches above have contributed to the advancement in understanding the head injury mechanisms, there is not yet enough evidence to adopt a new criterion based on injuries in the living human.

Another approach to study injury mechanisms is computer simulation. This may virtually make any loading condition and in-depth analysis possible. In this study, the capability of such a technique is demonstrated through attempted validation using detailed analysis of actual accident data and the tentative use of the model for parametric analyses of brain injury mechanisms.

BACKGROUND

Since Ward et al.[10] presented a first-generation three dimensional FE model of a brain, a number of models have been developed with increasing mesh density and accuracy that has increased with advances in the power of computer hardware[11-13]. The Wayne State University Head Injury Model (WSUHIM) developed by Zhang, et al.[14] is one of the most detailed three dimensional model currently existing. It consists of about three hundred thousand elements as shown on Fig.2. This model has been validated using intracranial pressure against cadaver head impact tests conducted by Nahum et al.[15] and Trosseille et al.[16] and against recent tests conducted by Hardy et al.[17] using the displacement of the brain relative to the skull measured by the techniques using Neutral Density Target (NDT) and bi-planar high speed X-ray system.

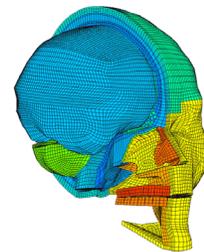


Fig.2 Wayne State University Head Injury Model (WSUHIM)

VALIDATION OF THE WSUHIM AGAINST ACTUAL ACCIDENTS

If the WSUHIM is valid, it should predict the occurrence of actual injuries in the living human produced by a head impact. In this study, two accidents were selected from the pedestrian accident database at the Road Accident Research Unit (RARU) at the University of Adelaide. RARU conducts detailed investigations of pedestrian collisions and in many fatal cases a neuropathologist examined the brain of the victim microscopically. The selected cases were reconstructed using simulation of the collision, physical tests on the same make and model of vehicle involved in the collision and finally FE simulation of the head impact.

Methodology

First, a total human body model was validated against three PMHS tests to make sure it could reproduce the overall kinematics and particularly the head impact velocity onto the car body. Second, two fatal pedestrian accidents investigated by RARU were reconstructed to obtain head kinematics during impact. The six components of acceleration estimated from the reconstruction process were applied to the WSUHIM. Finally, the stress/strain conditions were compared with actual distribution of injuries identified microscopically.

Total Pedestrian Body Model

While Ishikawa et al.[18] obtained good agreement between a multi body dynamics model of a vehicle/pedestrian collision with PMHS tests, in the ten years since their study, progress in software and hardware for FE techniques has been remarkable. This has enabled the simulation of the collision to be made where the pedestrian body and the car can both be modeled as full FE mesh. One of the advantages of FE modeling is realistic geometry that enables realistic contact interaction. Especially in case of pedestrian collisions, the contact interaction between pedestrian and the parts of car body significantly affects the total kinematics, because external forces on the pedestrian are generated only by contact. Therefore, in this study, an FE model of the pedestrian was adopted.

The model of the total human body was developed based on H-model™ [19] developed for the dynamic FE code PAM-CRASH™[20]. Takahashi et al.[21] modified H-model™ suitable for pedestrian impact simulation modifying its lower extremities to be capable of representing bone fractures and ligament ruptures. But in their study, the upper body was constructed with connected rigid bodies because the focus was on the behavior of lower extremities. In this study, as the focus is on the behavior of head, the model was modified again. Because the injuries among lower extremities are not of concern, the model was simplified to avoid numerical errors occurring after great deformation of elements. For example, the solid elements of meniscus were replaced with the contact definition between femoral and tibial condyles. And, since it was noted by Akiyama et al.[22] that flexibility in lateral bending of upper body significantly affected

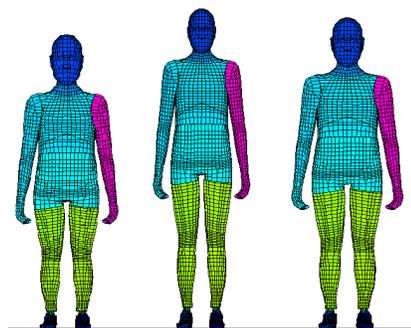
the behavior of the pedestrian dummy in lateral impact, the upper body was replaced with that having fully divided thoracic and cervical spines (H-Thorax™). The model overview is shown on Fig.3. The head was defined as a rigid body to obtain acceleration pulses at its center of gravity. The difference between rigid head and head with deformable skull and soft brain are not taken into account this time.



Fig.3 Human Pedestrian Model

Validation of the Total Pedestrian Model

Before the accident reconstruction, the model was validated against the post mortem human subject (PMHS) tests conducted by Schroeder et al.[23]. Three cases of newer model car called Y1, Y2 and Y3 were selected from the five PMHS experiments. An FE model of the car front was prepared to be a similar shape to those used in the tests and the human model



Test ID	Y1	Y2	Y3
Height (m)	1.67	1.82	1.77
Mass (kg)	68	63	84

Fig.4 Scaled Models for Three Cases

was scaled by the newly developed scaling program to be the same size as each PMHS according to its segment lengths and weight as shown on Fig.4.

Each scaled pedestrian model was positioned according to the position of the PMHS in the test and coupled with the corresponding car model like Fig.5. Hands were tied in front and legs were set back and forth according to the initial conditions of the test. Impact velocities of cars were 30 km/h for case Y1 and 40 km/h for case Y2 and Y3.

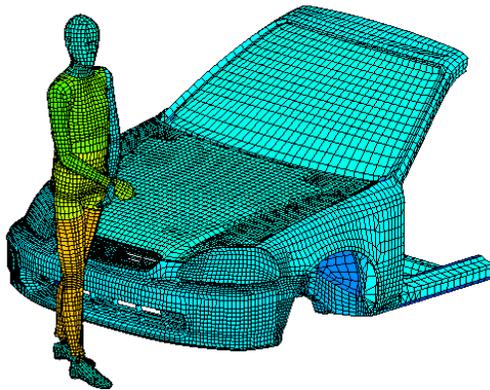
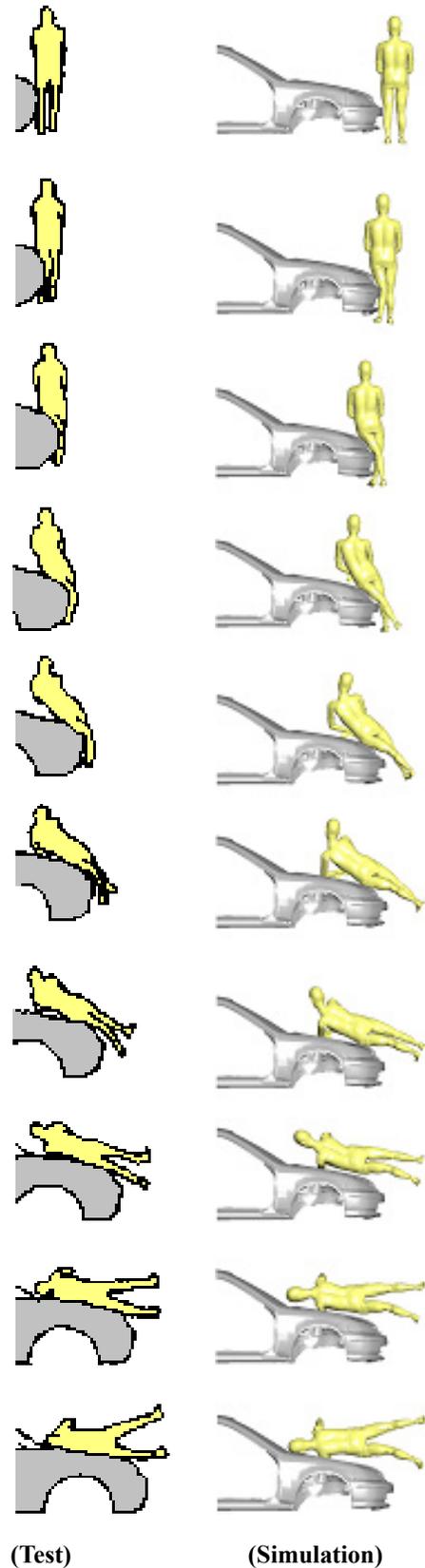


Fig.5 Initial Position of PMHS and Car

The kinematics of the simulation and of the tests are compared in Fig.6 to Fig.8. The simulations and tests show good agreement. Trajectories of head, thorax, lumbar and pelvis traced at the locations shown on Fig.9 are given in Fig.10 to Fig.12, which also show good agreement. Resultant head velocities from simulations and tests are compared in Fig.13 to Fig.15. Again, these results show good agreement. This shows that the total human body model was accurate enough to be used for the purpose of accident reconstruction.

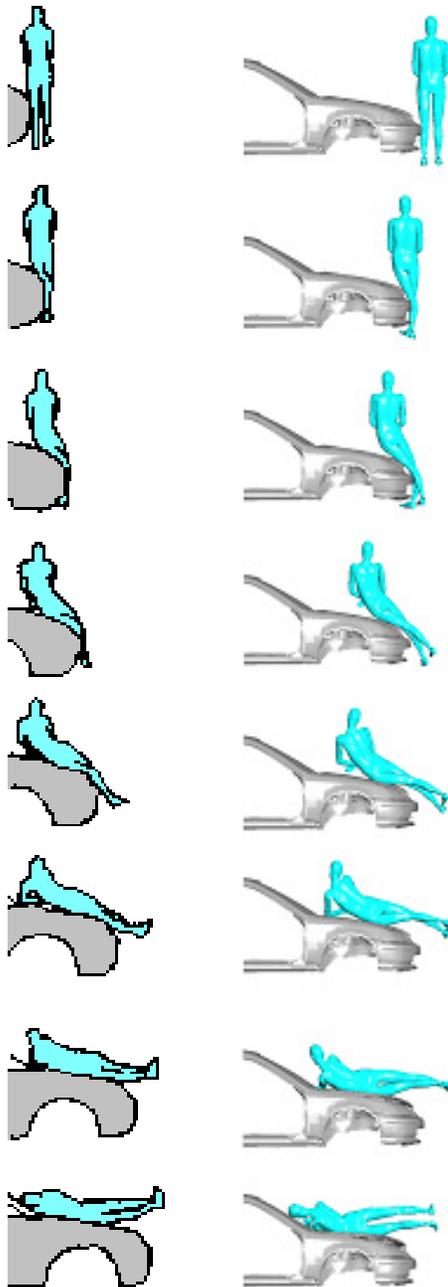
Accident Reconstruction

Two cases were selected from 200 cases in the pedestrian accident database preserved at Road Accident Research Unit (RARU) of the University of Adelaide. The selection criteria included data necessary to reconstruct the collision and existence of diffuse axonal injury (DAI) in the brain attributable to the immediate effects of the impact (survival time of 1 - 3 hours). Information on these two cases is given in Table.1.

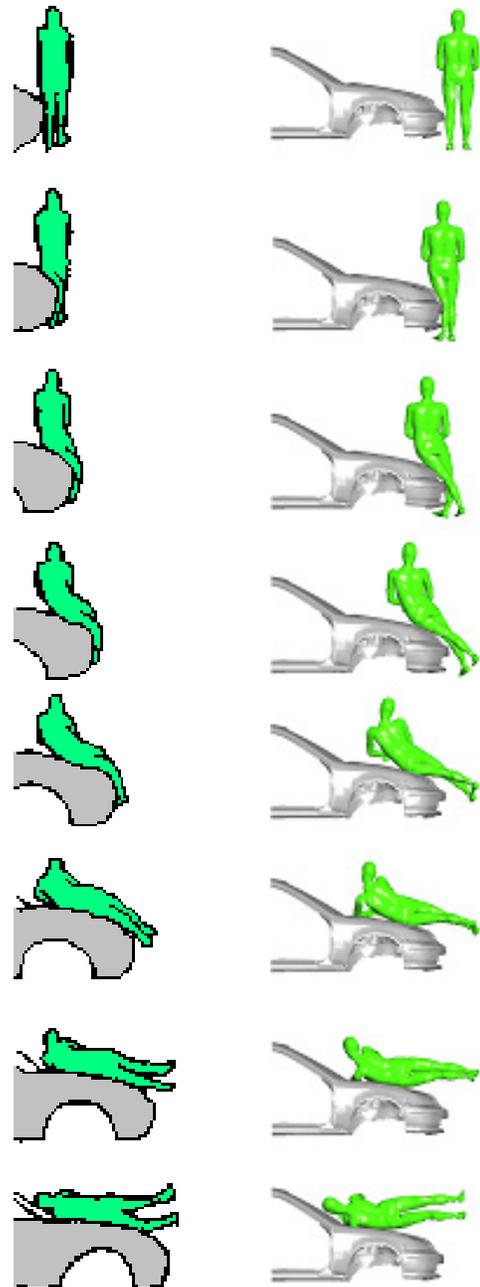


(Test) (Simulation)

Fig.6 Total Kinematics of Y1



(Test) (Simulation)
Fig.7 Total Kinematics of Y2



(Test) (Simulation)
Fig.8 Total Kinematics of Y3

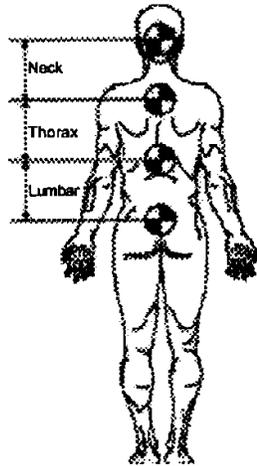


Fig.9 Fixation points of the for marks of PMHS tests [1]

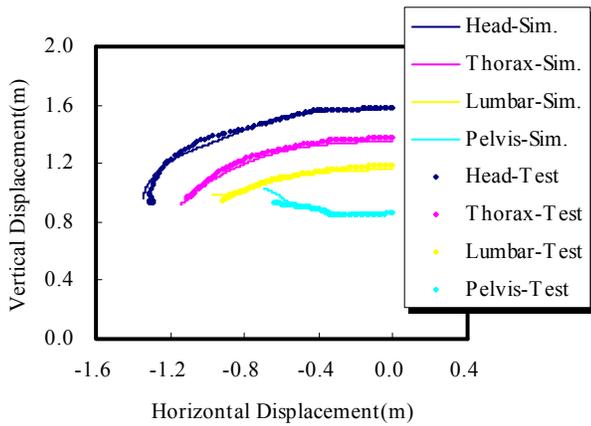


Fig.10 Trajectories of Y1

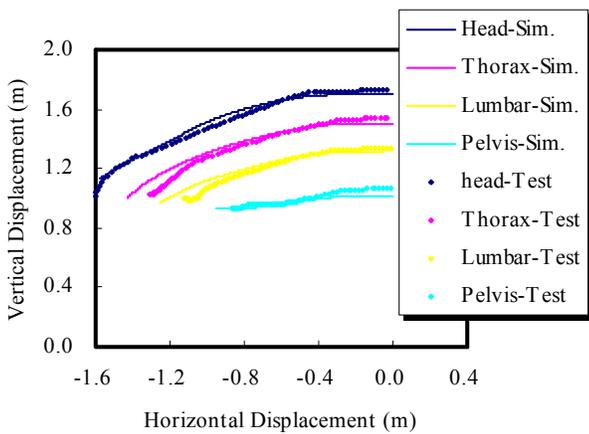


Fig.11 Trajectories of Y2

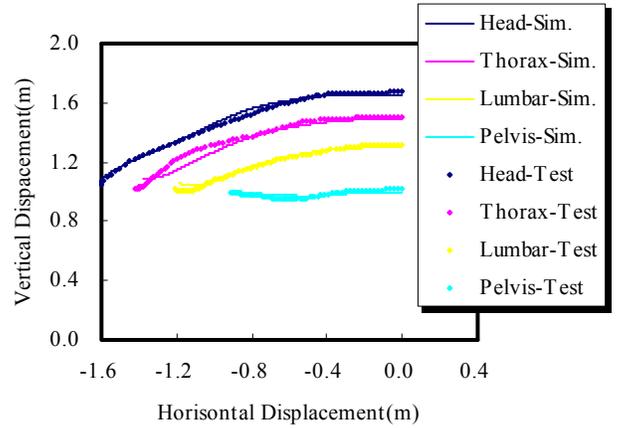


Fig.12 Trajectories of Y3

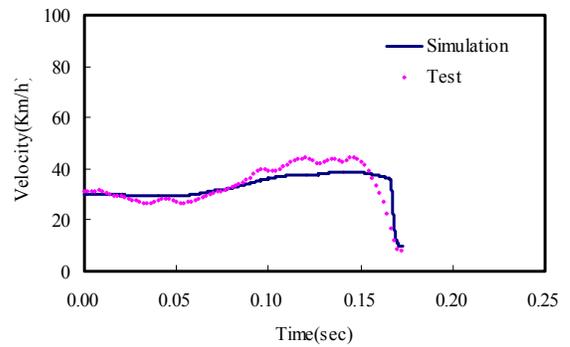


Fig.13 Head Resultant Velocity of Y1

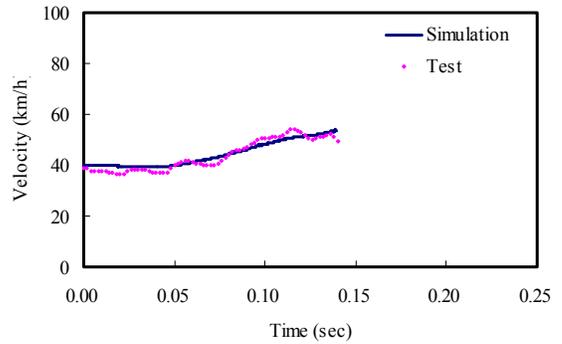


Fig.14 Head Resultant Velocity of Y2

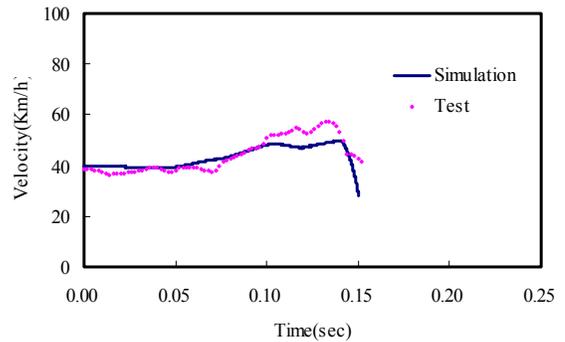


Fig.15 Head Resultant Velocity of Y3

Table.1 Case Description

Case ID	H032-86	H070-85
Sex	M	F
Age	81	14
Height (m)	1.75	1.63
Mass (kg)	75	64
Car Speed (m/sec)	11.4	16.7
Brain injury	DAI SAH	DAI SDH Contusion

DAI = Diffuse axonal injury
SAH = Subarachnoid haemorrhages
SDH = Subdural haematoma

These two cases were first reconstructed to determine the head impact condition, using a validated multi-body MADYMO model developed by RARU[24]. Later, these conditions were applied to the FEM model described previously. The model was first scaled to be the size of the victim according to the measured data recorded at autopsy (Fig.16).

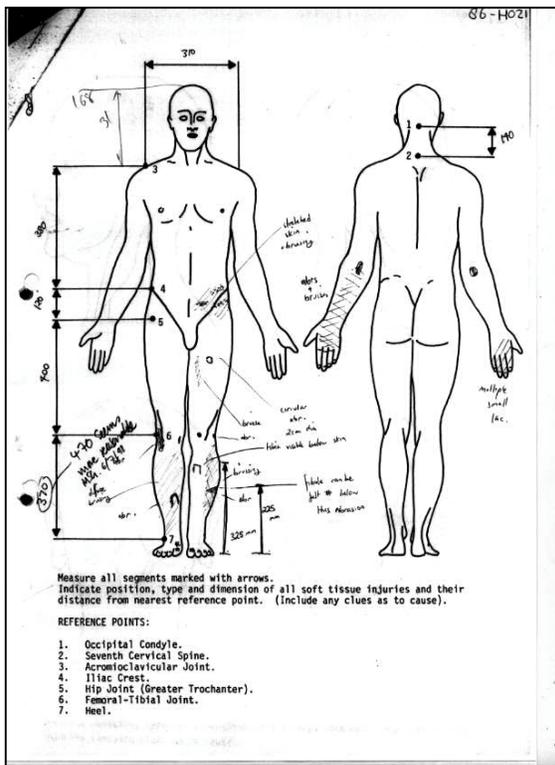


Fig.16 A Datasheet from RARU for Measured Body

For each case, 18 MADYMO simulations were carried out with different gait cycle postures and vehicle speeds to find the simulation with the best fit of contact points with the actual accident. After fitting the head contact point, head impact was physically

reconstructed using the EEVC WG10 headform using the same make and model of car. The impact data was analyzed to determine the dynamic force-deflection characteristic of the contact between head and car body. This characteristic was inserted into the MADYMO simulation to improve accuracy of the head impact response. Then, the MADYMO simulation that best reproduced the pedestrian kinematics was used to determine the posture of the pedestrian FE model in PAM-CRASH™. The contact surfaces of the car body were also converted to PAM-CRASH™ FE data. Initial conditions of the two cases generated for FE simulation are shown on Figs. 17 and 18. Car body shapes were generated by copying the geometry of the MADYMO models. The contact characteristics of the vehicle were represented not by segment-to-plane contact with force-deflection functions but by standard FE calculation with material property of mild steel. Contacts of head and windshield/dashboard were represented by segment-to-plane contact with force-deflection functions obtained from head-form impacting tests.

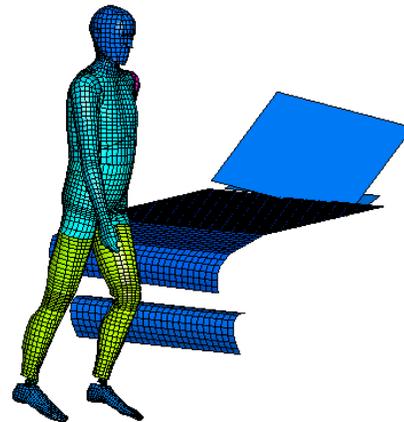


Fig.17 FE Model of case h032-86

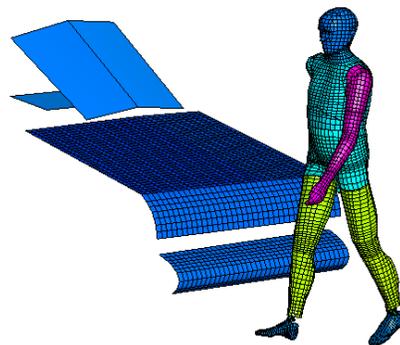
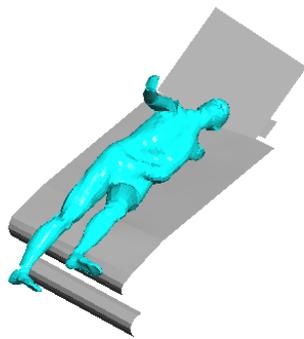


Fig.18 FE Model of case h070-85

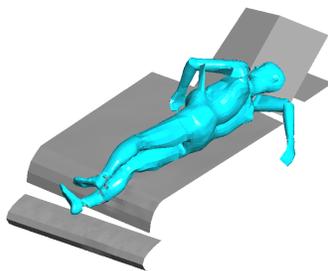


(Simulation)



(Actual Car Body)

Fig.19 Head Contact Point of h032-86



(Simulation)



(Actual Car Body)

Fig.20 Head Contact Point of h070-85

The results of the FE simulations were checked to ensure that the head contact points coincided with those of actual cases as given on Figs.19 and 20. Both cases show so good agreement that they can be recognized as accurately reconstructed cases.

The head impact acceleration data obtained from these simulations above were applied to the WSUHIM with three-dimensional forced motion. Figs. 21 and 22 show the estimate of linear and angular acceleration data of the center of gravity of the case h032-86. The acceleration between 148 and 177msec (the interval on which the HIC was calculated) was used as an input to save computing time. Figs.23 and 24 show the head impact accelerations estimated for the case h070-85. In this case, the acceleration clip was taken in the interval of 91 to 105msec. These two sets of short duration pulses of six components were applied on the center of gravity of WSUHIM, the skull of which was made rigid.

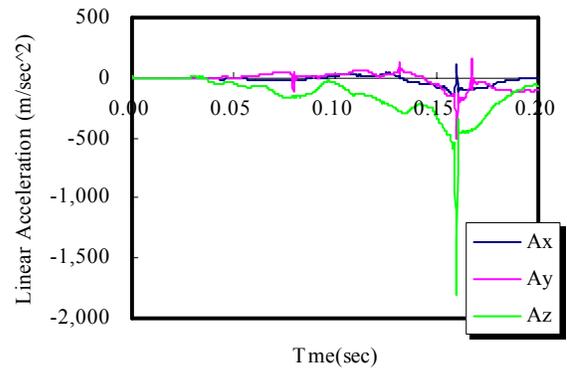


Fig.21 Linear Head Acceleration of H032-86

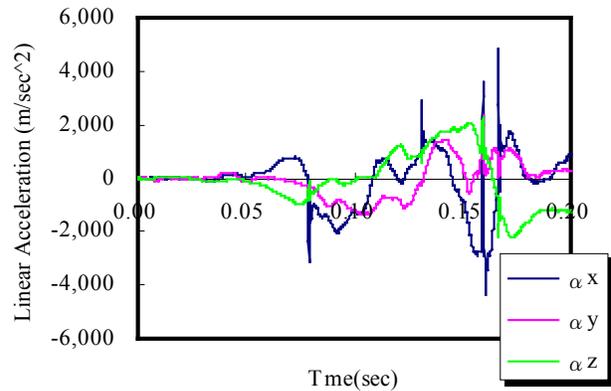


Fig.22 Angular Head Acceleration of H032-86

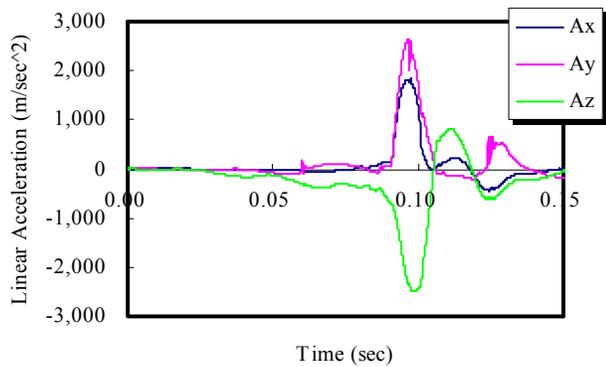


Fig.23 Linear Head Acceleration of H070-85

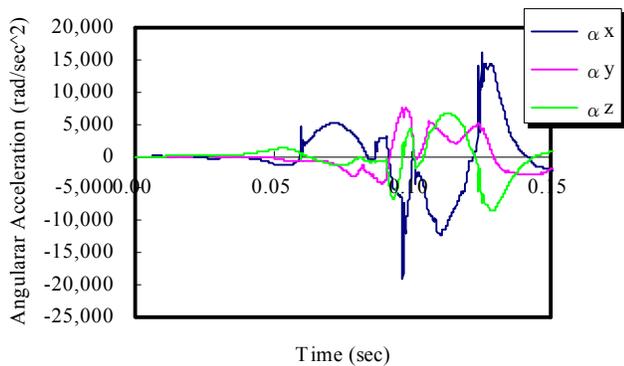


Fig.24 Angular Head Acceleration of H070-85

Maximum principal strain (MPS) was used as a predictor for DAI. In both cases, the brain was examined in detail by a neuropathologist. Sections were taken every 10 mm like Fig.25 and stained for the presence of amyloid precursor protein (APP) a marker of axonal injury. The sections were studied microscopically to detect DAI. The technique is presented in detail in Anderson [25].

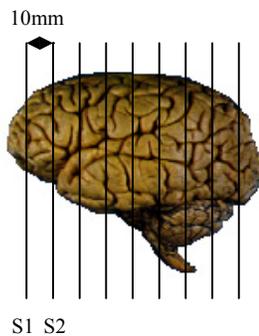


Fig.25 Locations of Coronal Section

The contours of maximum principal strain (MPS) on every section from the simulation of two cases are

compared with the maps of observed DAI for every section as shown in Figs.26 and 27. Referring to Gennarelli et al.[26], Thibault et al.[27], Ueno et al.[28] and Eppinger et al.[29], the provisional threshold of maximum principal strain for DAI was nominated to be 0.15 in this study. In the case h032-86, MPS greater than 0.15, was observed over around half a region of the cerebrum. Though this is consistent with

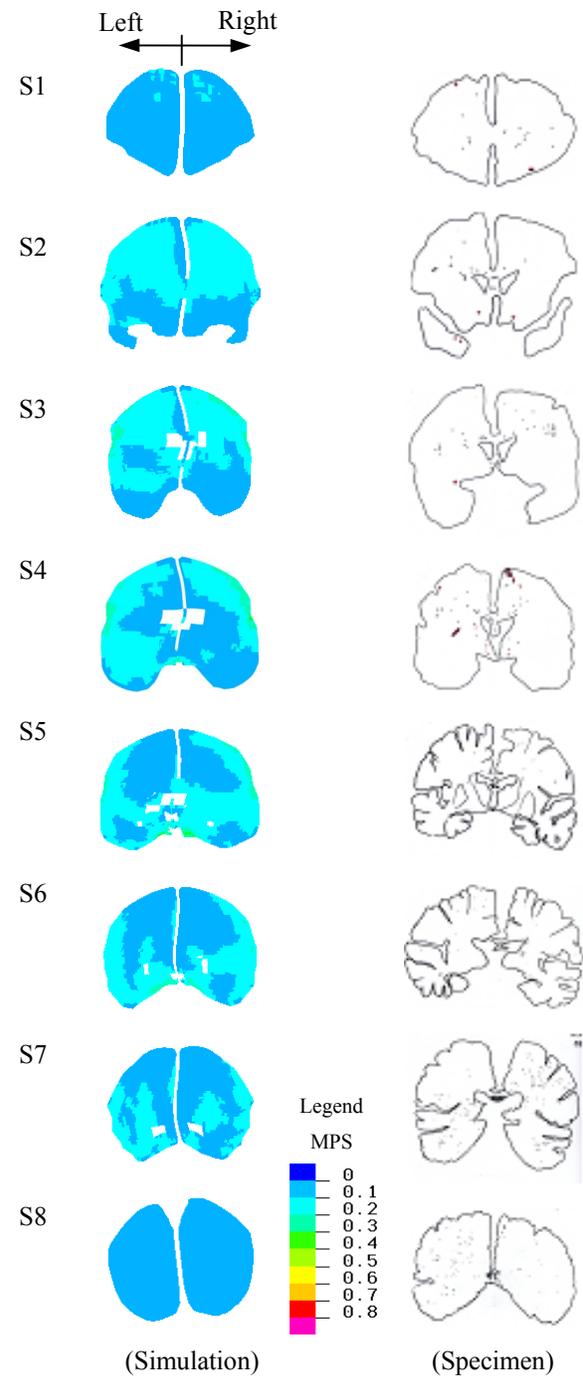


Fig.26 MPS Contour and DAI Map of H032-86

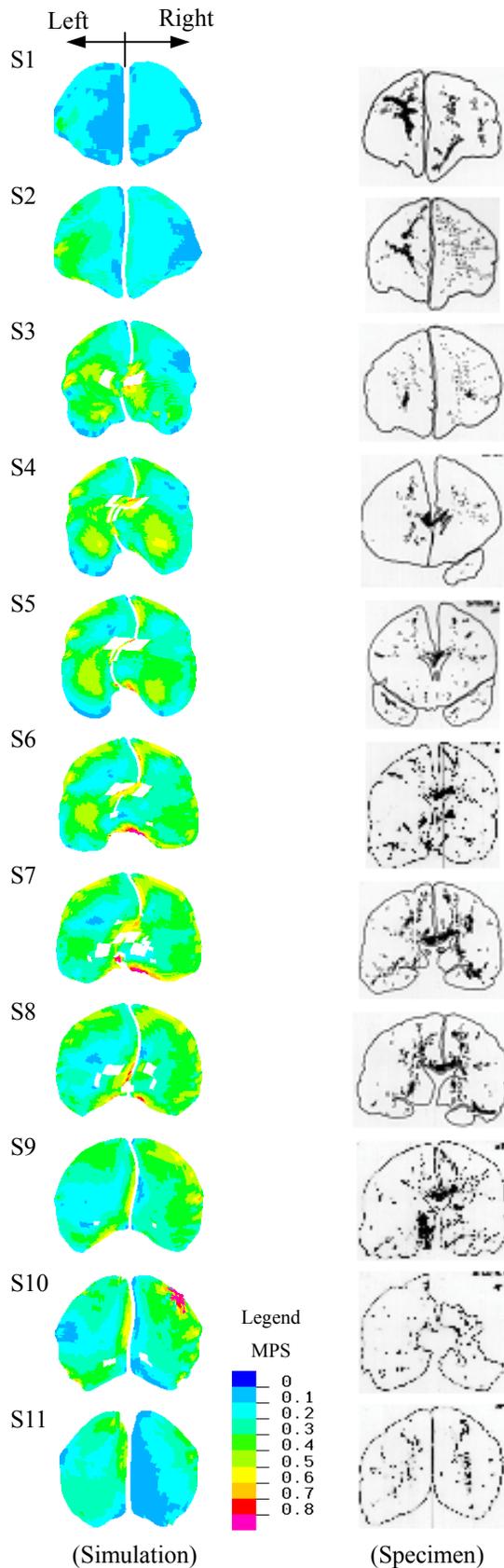


Fig.27 MPS Contour and DAI Map of H070-85

relatively low density of DAI being observed over the regions in the specimens, particular regions where axonal injuries are detected with high density don't always show high MPS on the simulation. In the case h070-85, the simulation predicted that most regions of the brain experienced MPS greater than 0.15. In this case, the neuropathology showed that DAI broadly exists over each section. But, in this case, again, it is not clear that regions of higher MPS always coincide with higher density of DAI detected.

The results show that the WSUHIM predicted the presence of DAI insofar as the model predicted that maximum principal strains greater than 0.15 were experienced. However the model used in conjunction with the reconstruction process did not accurately predict the location of axonal injury. The lack of correspondence may well be due to problems with the accuracy of the head motion obtained from the accident reconstruction rather than inaccuracies in the WSUHIM, although this could not be shown conclusively in this study. Importantly, two cases are a limited basis on which to judge the validity of the model, which highlights the intensive nature of the work required to perform such a validation as have been attempted here.

TENTATIVE APPLICATIONS OF WSUHIM

Although the model is not fully validated at this time, tentative applications are tried to demonstrate the potential of the model.

Parametric Study about WSUTC

On the curves of HIC 500, 1000 and 1500, six combinations of parameters of linear acceleration and duration time were applied to the model as depicted on Table.2. To make it simple, linear acceleration was specified as constant during the time duration. The direction of loading was anterior-posterior. A total of 18 simulations were run and maximum principal strain of each case was extracted. These values are depicted in Fig.28. Looking at Fig.28, maximum principal strain of about 0.2 corresponds to the threshold of HIC 1000. In case of shorter duration such as 5 msec and less, MPS appears relatively low. This could indicate that HIC for short duration rather predicts bone fracture than brain injury.

Table.2 Combinations of Duration Time and Linear Acceleration Level

Duration Time(msec)	Linear Acceleration Level for 3 HIC Values (m/sec ²)		
	(HIC)500	1000	1500
2.5	132	174	205
5	100	132	155
10	76	100	118
20	49	64	76
30	44	57	68

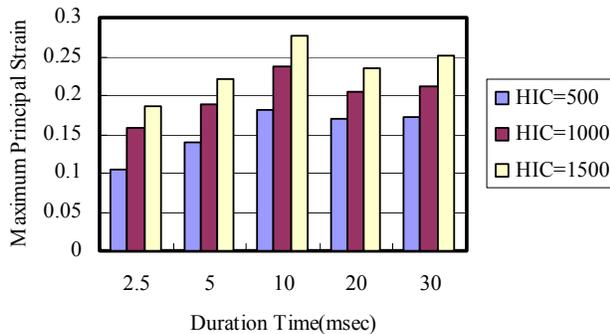


Fig.28 Maximum Principal Strain for Combinations of Parameters (Linear)

Rotational Motion

An additional parameter study was done to study the effect of rotational motion. Combinations of angular acceleration of 10,000 and 20,000 rad/sec² and duration time of 10 and 20 msec were applied to the model as a forced rotational motion around the center of gravity of the head in sagittal plane like Fig.29.

The MPS values obtained are given in Fig.30. Although the applied angular accelerations for respective duration times were so severe that all cases show higher MPS than 0.15, this does demonstrates the probability of brain injury caused by rotation.

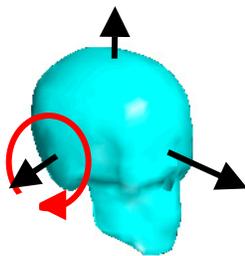


Fig.29 Sagittal Rotation of the Head

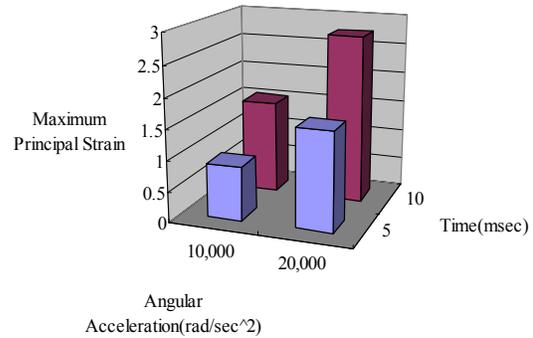


Fig.30 Maximum Principal Strain for Combinations of Parameters (Angular)

DISCUSSION

The difficulty in using accident reconstruction for the purpose of model validation is that the kinematics of the person, or any other measured pulses of movement or force must be estimated only, as obviously no direct measurements are possible. Every estimate made in the reconstruction process is made with aim of understanding the final state of the impact event (in this study, the head impact). Though it is rather easy to make the final state obtained from the simulation agree to the observed accident data in terms of the location of the head impact, it is more important to reproduce the transitional behavior from a biomechanical point of view. In this study, a simplified method was chosen to model the car body by translating a MADYMO surface to FEM, because it was hard to develop FE model of such old models of car with limited information. Therefore, while the contact points of the head agreed to those of the accidents, the transitional behavior of the model is not guaranteed in the way that the simulation of the three PMHS cases could be. Ideally, the car would be precisely modeled with all required information on geometry and material properties. There also remains uncertainty in the vehicle impact velocity and pedestrian initial conditions which also add to uncertainty.

In this study, only diffuse axonal injury in cerebrum was taken into account. Validation looking at other kinds of injuries, i.e., contusion, SDH, SAH and other vascular injuries should be conducted so that the model may be broadly used to estimate total head

injuries with corresponding criteria. In addition, other predictors for DAI could be considered as King et al.[30] have recently reported.

If the model can be fully validated and accompanied with established criteria for common brain injuries, it will be a useful tool for the investigation of traumatic brain injury mechanisms. It enables enormous number of parametrical study in the multi-dimensional space of input parameters, i.e., three-dimensional linear and angular accelerations and their time duration. Filling the space of these parameters, finally the multi-dimensional tolerance surface will be generated.

Eppinger et al.[28] are currently developing new simulation based criteria 'SIMon' in which three different criteria, CSDM, DDM and RMDM are proposed for DAI, contusion and SDH respectively. The benefit of this kind of approach is to estimate probable injuries taking all the effects of measured parameters into account individually.

In any case, in order to establish the criteria that may be broadly accepted, a detailed simulation model that can successfully predict injury is necessary. The approach described in this paper of using detailed accident investigation and reconstruction techniques to investigate injury mechanisms provides the framework for such a validation process and would supplement a limited numbers of cadaver tests.

CONCLUSIONS

- (1) The pedestrian total body FE model coupled with a scaling program showed a good correlation with PMHS tests and showed agreements in head contact points with actual accidents.
- (2) WSUHIM was used to simulate the head impacts experienced in real pedestrian accidents and demonstrated the its capability of predicting the occurrence of diffuse axonal injury by means of maximum principal strain in general.
- (3) While the study did not show that the WSUHIM was capable of predicting all the locations of diffuse axonal injury, this may have been due as much to errors in the reconstruction process as to errors in the model. Reconstruction errors include errors in vehicle speeds and contact interaction between car and head.

- (4) The WSUHIM was tentatively used to examine WSUTC parametrically and it was observed that HIC 1000 corresponds to maximum principal strain of around 0.2.
- (5) As the model showed that rotational acceleration causes large strains within the brain, it is suggested that a more inclusive injury criterion may be desirable, incorporating both the translational and rotational motions.

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