Investigation of Brain Trauma Biomechanics in Vehicle Traffic Accidents Using Human Body Computational Models

Jikuang Yang

Abstract This chapter aimed to study the biomechanical response and injury mechanisms of brain in passenger car-to-pedestrian collision event. The kinematics of head impact to a passenger car was reconstructed using multibody dynamics (MBD) models. The brain injury biomechanics was investigated by using an FE model of human body head (HBM-head). The HBM-head model was developed in accordance with human head anatomy. The model consists of scalp, skull, dura mater, cerebrospinal fluid, pia mater, cerebrum, cerebellum, ventricle, brain stem, falx, tentorium, etc. The existing data from cadaveric head impact tests were used to validate the head FE model. The kinematic and kinetic responses of the head were determined by using MBD model. The brain injury-related physical parameters and the distribution of the intracranial pressure were calculated from simulations of head impact to the windscreen and A-pillar by using the HBM-head model. It is proved that the head FE model has good biofidelity and can be used to study head–brain trauma and injury mechanisms in vehicle collisions.

Keywords Traffic injury · Brain trauma · Head FE model · Pedestrian MBD model · Impact biomechanics

1 Introduction

The serious and fatal brain injuries are observed frequently in vehicle traffic accidents, which is a public health issue worldwide. It resulted in a large number of social and economic problems due to head trauma-related deaths, treatment and insurance compensation. To minimize the risk of brain injury in the accident, there is a need of advanced tools to get good knowledge about the kinematics of the accidents

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and the causation of brain trauma as well as the correlation of brain injuries with the physical parameters in a vehicle crash environment.

In the past years, the head injury mechanisms and technology of injury prevention have been extensively studied by medical doctors and researchers in vehicle traffic safety field all over the world. Many studies on head–brain injury biomechanics were carried out by using physical and mathematical models. The finite element (FE) technique is an effective method for the prediction of human body injuries and analysis of injury mechanisms from vehicle traffic accidents. The finite element method was therefore developed rapidly and applied in the research field of head–brain injury biomechanics in recent two decades. These include the work of Ueno [1], Lighthall [2], Nagashima et al. [3], Chu [4], Trosseille [5], Bandak [6], Chu et al. [7], DiMasi et al. [8], Mendis et al. [9], Willinger et al. [10], Ueno et al. [11], Nishimoto and Murakami [12], Anderson et al. [13], Huang et al. [14], Willinger et al. [15], and Bradshaw et al. [16]. A number of the 3D head models have been presented and used to study human head response. For example, the head–brain FE model WSUBIM [17] was developed in Wayne State University in the USA, the ULP model [18] in Université Louis Pasteur, and the HUMOS model in EU 5th framework program (Human Models for Safety), THUMS model in Japan.

At the same time, these models have been used to study the trauma from vehicle traffic and sport accidents. The application of the validated FE models indicated that the FE models play an important role in the studies of mechanism of brain injuries by analyzing the intracranial pressure and the stress and strain of brain tissues. In order to accurately reflect the biomechanical response and injury mechanism of human head trauma in different crash accidents, it is necessary to further develop the brain FE model with the improved characteristics of human head in both the anatomy structure and the material models of biological tissues. It is also vital for researcher to evaluate the validity of the models using available biomechanical data from experimental studies. These issues have attracted an increasing attention in the simulation study on the human brain FE models.

The aim of this chapter is to investigate the mechanism of brain injury in vehicle collision by using a developed FE model of human body head (HBM-head) in accordance with human head–brain anatomy.

2 Method and Materials

The FE model of HBM-head was developed based on 3D anatomical image data [19]. The preprocessing and meshing of head brain 3D anatomy image data was carried out using Hypermesh software. The computations of brain biomechanics responses were carried out using nonlinear explicit dynamics finite element algorithm in LS-DYNA 3D code. The effectiveness of the head model was verified by comparing the results of the Nahum's impact experiment [20] using human head specimen. The sensitivity and biofidelity of the FE model for predicting brain injury were detected through parameter analysis at different impact speeds.

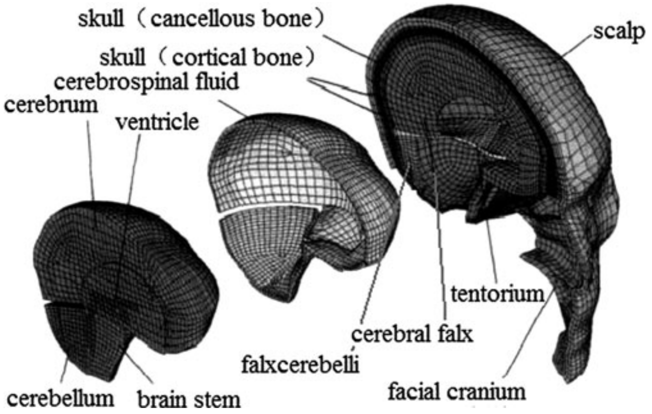

Fig. 1 An FE model of head was developed based on human head anatomy. The model consists of scalp, skull, cerebrospinal fluid (CSF), cerebrum, cerebellum, ventricle, brain stem, falx, and tentorium, etc.

2.1 Description of the HBM-Head FE Model

The HBM-head model consists of the scalp, skull, dura mater, cerebrospinal fluid (CSF), pia mater, cerebrum, cerebellum, ventricle, brain stem, falx, tentorium, etc. as shown in Fig. 1. The head is modeled using 66,624 nodes, 49,607 solid elements of eight-noded hexahedron, and 11,514 shell elements. The mass of the head model is 4.4 kg, which was based on the anthropometry size of a 50th male adult human body.

The thickness of scalp is defined as 5–7 mm [21] and it is described with two-layer solid elements. Skull was modeled with a hierarchical structure in the sandwich form of cortical bone and cancellous bone. The thickness of skull is about 5–7 mm. The two-layer solid element was used to simulate accurately the anatomical geometry of both sides of the skull. Dura mater is simulated with one layer of shell elements. The CSF of subarachnoid space is described using solid elements with a low shear modulus. The relative motion between the skull and brain is simulated by the relative sliding between the dura mater and CSF. The outer surface of CSF is defined to simulate arachnoids. The structure under the CSF is pia mater that closes the brain surface. The inner surface of CSF is defined to simulate the pia mater. The falx between the two hemispheres of the brain and the tentorium between the cerebrum and cerebellum are represented by solid elements. The overall quality of mesh was controlled in the process of modeling as shown in Table 1.

2.2 Material Parameters

Bio-tissue materials show typical viscoelastic properties related to load and speed. Viscoelastic material model is widely used to describe the material properties of

Table 1 Quality control parameters of elements

Quality control parameters	Threshold
Warpage	<35.20
Aspect ratio	<10.70
Skew	<64.00
Min. size	>0.70
Jacobian	>0.47
Min. angle quad	>16.69
Max. angle quad	<160.65

Table 2 Material definition of model components

Part	Material property	$^aE/^ak$(MPa)	Poisson ratio	G_0(kPa)	G_∞(kPa)	$\beta(s^{-1})$
Falx	Elastic	31.5	0.45			
Tentorium	Elastic	31.5	0.45			
Cortical bone	Elastic	15,000	0.21			
Cancellous bone	Elastic	4,600	0.05			
Scalp	Elastic	16.7	0.42			
Cerebellum	Viscous Elastic	2,190		10	2	80
Cerebrum	Viscous Elastic	2,190		12.5	2.5	80
CSF	Viscous Elastic	1,050		1	0.9	80
Brain stem	Viscous Elastic	2,190		22.5	4.5	80
Dura mater	Elastic	31.5	0.45			
Pia mater	Elastic	11.5	0.45			

^aE = Young's modulus, k = bulk modulus

brain tissue [17–19]. Researches have shown that water accounts for nearly 78% in brain tissue that result in incompressible characteristics. For the HBM-head model, a linear viscoelastic material was selected in this study, as the maximal strain of the head model is 0.1965 at an impact speed of 12 m/s in this study, which is based on Bathe [22]. The shear elasticity behavior of this material was obtained from the follow equation:

$$G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t}$$

where G_0: short-term shear modulus, G_∞: long-term shear modulus, β: decay constant, t: time.

The bulk modulus of brain tissue was defined as 2.19 GPa, and the shear modulus was changed between 680 Pa and 268 kPa. The material parameters in literature [17–19] were used in HBM-head model as shown in Table 2. In order to obtain an accurate simulation of the relative motion between brain and skull during impact, the shear modulus and bulk modulus of CSF were lower than brain tissue. The material of cortical bone and cancellous bone were defined in reference to the skull material properties of the ULP model [18].

2.3 Contact Interface

During an event of vehicle crash, the cerebral meninges and CSF between the skull and brain will lead to a relative sliding between the skull and brain due to effect of inertial loading of the translation and rotation acceleration. There is very thin space filled with CSF between the dura mater and arachnoid preventing these structures sticking together. The CSF plays an important role for energy absorption and damping during an impact. This effect has been investigated by using finite element (FE) model of craniocerebral. There is space between arachnoid and pia mater and the space is filled with CSF. Some of blood vessel are going into the brain tissue. Some thin filament structures called arachnoid trabeculation in subarachnoid extend to pia mater from arachnoid. They play a role of fixing the pia mater to arachnoid. Based on the above knowledge of anatomy, a HBM-head FE model was developed using solid and shell elements. The dura mater is a layer attached to the inner surface of skull. The inferior of dura mater is a layer arachnoid. The pia mater is the membrane that is closely attached to the surface of brain tissue. Elements simulating pia mater are attached to the surface of brain. Outside of brain are elements representing CSF. The arachnoid is simulated by the outer surface of CSF. The contact algorithm was defined between dura mater and arachnoid in the HMB-head model. The solid elements between ventricle and brain tissue were connected using the common nodes.

2.4 Model Validation

The model was validated with the data from head impact experiments by Nahum [20]. The experimental samples were human cadavers without antiseptic treatment. The head was loaded at a certain speed using a rigid impactor with padding. The experiments were divided into two groups. The samples used in the first group have the numbers of 36–38, 41–44, and 54. The mass of impactor was defined from 5.23 to 23.09 kg. The velocity was from 4.36 to 12.95 m/s. The numbers of the second group were 46–52. These experiments were carried out at different speeds to the same sample and the speed changed from 4.42 to 8.69 m/s. And the mass of impactor was 5.23 kg. Different padding materials of impactor were used to obtain the proper impact duration.

The contact force between the impactor and head, the centroid acceleration of head and the pressure in five different positions from the Nahum experiments [20] were used in the validation of the model. The five positions were brain tissue region near the impact location on the frontal bone; the brain tissue of the side of parietal bone located in the upper of the juncture of coronal suture and squamous suture; the inferior part of the lambdoid suture of occipital bone; and the nucleus fastigii of cerebella of occipital. The detail curve data of NO 37 experiment was showed in Nahum's paper [20]. So the results of this experiment were selected to validate the simulation model. The mass of the cylinder impactor is 5.59 kg and the speed of impact is 9.94 m/s in the experiment.

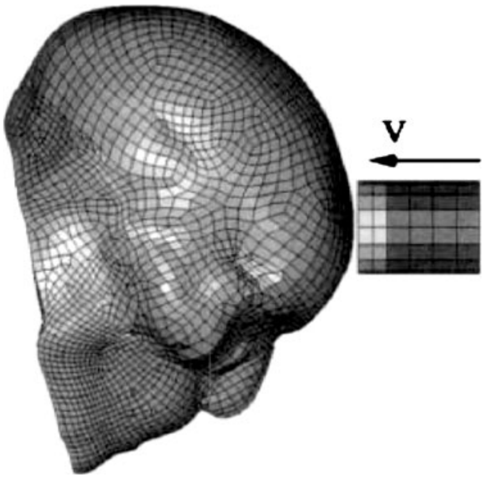

Fig. 2 Configuration for model validation

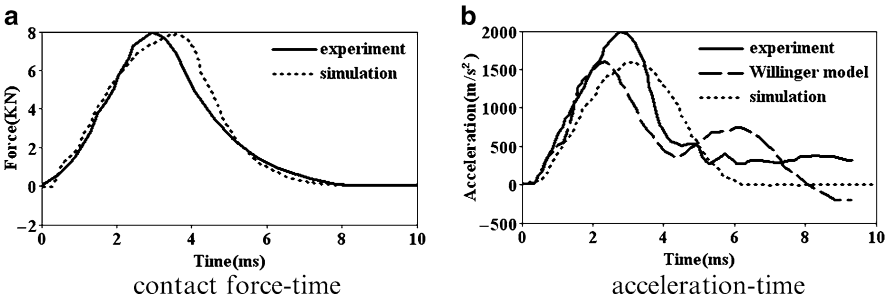

Fig. 3 The comparison of time history curve between Nahum experiment [20] and simulation

Figure 2 illustrates the configuration for simulation of the head impact test. The boundary conditions of simulation were defined based on test configurations. As the impact time was so short that the neck has little effect on head response in such short time. We assumed that the boundary condition of head is free. Referring to the Nahum experiment [20], the head was forward incline in order to make the Frankfurt plane to horizontal plane at an angle of 45°. The model of impactor with the padding materials was developed simulated by using foam material in the front end. In order to obtain the impact characteristic of padding materials, pre-analysis of different foam material were carried out with the stress–strain curves of corresponding material. According to the experiment method of Willinger et al., the mass of the cylinder impactor was set to 6.8 kg and the speeds of impact changed from 6 to 9.94 m/s.

Analysis of simulations indicated that the calculated contact force agreed well with the Nahum experiment at 6.8 m/s. The impact contact force is shown in Fig. 3.

Figure 3a illustrated a comparison of the impact forces between results from the simulation and the experiment. The accelerations of center of mass from the

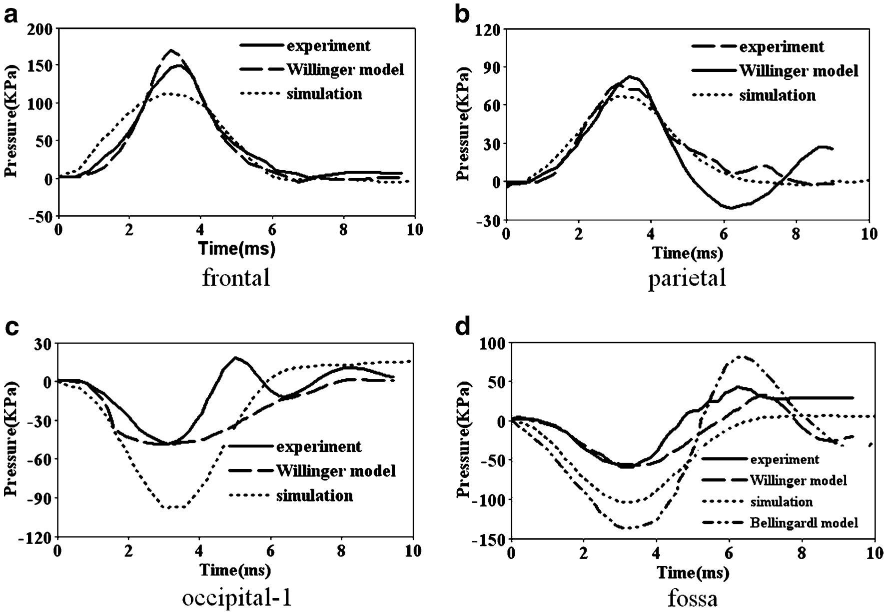

Fig. 4 The comparison of intracranial pressure time history curves between experiment and simulation

simulation, Nahum's experiment [20] and Kang's study [15] are given in Fig. 3b. The comparison among the intracranial pressure is shown in Fig. 4 in terms of results from simulation of this model and the experiment, as well as simulation curves by Kang's using ULP model and Belingardi, etc. [23].

3 Investigation of Brain Injury in Vehicle Collisions

In passenger car-to-adult pedestrian accidents, the head injuries attract particular attention due to the severe or fatal consequences. Many studies have been carried out in this area but the injury mechanisms and the tolerances of brain remain controversial. A study of the skull–brain injury mechanisms was conducted by using a MBD pedestrian model [24] and the HBM-head FE model (Fig. 5).

Head trauma accident data were selected from IVAC accident database [25] which was developed based on in-depth investigations of vehicle accidents in Changsha, China. A passenger car-to-pedestrian impact at 45 km/h was reconstructed using multibody dynamic (MBD) models to acquire the head impact conditions for the head impact velocity, head position, and head orientation. The HBM-head FE model was used for the reconstruction of skull fracture and brain injuries via a virtual test of head impact against windscreen and A-pillar (Fig. 5).

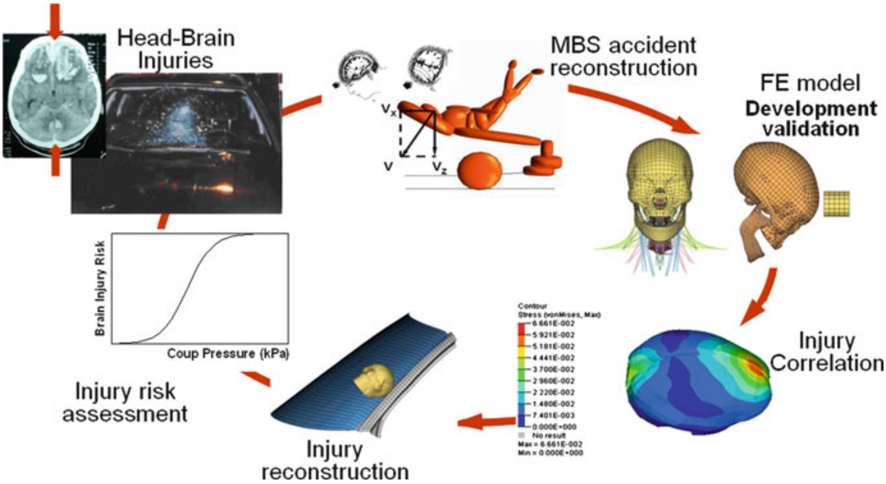

Fig. 5 A computational study of brain injury biomechanics based on accident data by using MBD and FE models

A stress analysis was conducted to determine the correlation of the stress and pressure distributions of the brain model with the injuries observed in the head–windscreen collisions.

From head–windscreen impact, the received contact force of the HBM-head model is 4.4 kN and the intracranial pressure maximum 250 kPa. From head–A-pillar impact, the received contact force of the HBM-head model is average 16 kN and the intracranial pressure maximum 815 kPa.

The skull fracture appeared in A-pillar impact, and there is no fracture in windscreen impact. The intracranial pressure maximum 250 kPa from windscreen impact could correlate with minor coup/countercoup injuries. The intracranial pressure maximum 815 kPa from A-pillar impact could correlate with severe coup/countercoup injuries.

It is necessary to point out that the approach used for calculation of the physical parameters for brain injuries in car-to-pedestrian impact will result in certain deviation. This could be due to the difference of the effective head mass between the FE head impact modeling and the head impact modeling with MBD pedestrian model in which a neck constraint force applied to the head [26].

The reconstruction results indicated that coup/countercoup pressure, Von Mises and shear stress were important physical parameters to evaluate the brain injury risk. The relationship between skull fracture and the predicted physical parameters can be determined. Thereby, we can finally obtain reasonable advices to improve safety design of car frontal structure for minimizing the risk of pedestrian head injuries.

4 Conclusions

Using both the MBD and HBM-head FE models is a valuable approach for reconstruction of vehicle-to-pedestrian collisions and analysis of the dynamic responses and injury-related physical parameters. For further study of the brain parameters, it is important to get an effective head mass in the simulations.

The brain injury-related parameters such as the head acceleration, stress, strain, contralateral intracranial pressure can be obtained, which indicate that the model can be used to study typical traffic injuries and the injury mechanism. Furthermore, the acquired knowledge can be used to improve the car safety design for protection of pedestrian head injuries.

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