

Development of a Finite Element Model for the Pedestrian Lower Limb

– Model validation for car-pedestrian impact –

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ABSTRACT

A finite element model for a human lower limb in a standing position has been validated against published experiments simulating impact between a pedestrian lower limb and a car front. The results showed that the model can reproduce the peak linear acceleration during the impact at both the impact level and the knee joint. The comparison of injuries to leg bones and knee ligaments between the experiment and the computer simulation showed that the model can predict different injury patterns observed under different impact conditions. Further, in addition to reproducing the response of the lower limb, the upper half of the human body was modeled with rigid segments connected by joint elements in order to simulate the kinematics of the upper body. The neck model has been validated against published experiments that used human volunteers. While validating the full body model against published full-scale car-pedestrian impact tests, it was found that the full body model can simulate both injuries to the lower limbs and the kinematics of the human body.

1. Introduction

Bone fracture and damage to knee ligaments are the two typical lower limb injuries in car-pedestrian accidents. Past experimental studies have shown that bone fractures are rarely accompanied by damage to knee ligaments.⁽¹⁾ This suggests that the leg fracture reduced the load on the knee, and conversely, the damage to knee ligaments reduced the impact load on the bony structure. The question of whether bone fractures or ligamentous damage will occur also appears to depend upon the loading conditions on the lower limb. Consequently, validation of the damage predictions by the pedestrian lower limb model will require investigation of the prediction capabilities for various injury patterns under different loading conditions. It seems reasonable to suppose that differences in lower limb response have an

effect on upper body behavior. Therefore, computer simulation models intended to be used for the enhancement of vehicle safety with regard to pedestrians should be capable of predicting the kinematics of the upper body as well as injuries to the lower limbs.

Schuster et al.⁽²⁾ developed a finite element model for the pedestrian lower limb and they validated the model against the results of lower limb impact tests. The impact loading conditions, however, were limited to a single type, and the boundary conditions for the lower limb were also different from those in actual car-pedestrian accidents. Ishikawa et al.⁽³⁾ developed a multibody model of the whole body of a pedestrian. Though this model accurately reproduced the kinematics of the whole body obtained experimentally, the prediction of lower limb injury was an issue. Likewise, pedestrian models developed in past studies have not been validated for prediction of lower limb injury

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patterns under a variety of impact conditions. In addition, these models did not succeed in predicting both the kinematics of the upper body and the injuries to the lower limb.

This paper describes the validation results for injury prediction with the finite element model of the lower limb in a standing position that was developed by the authors. The validation results for the upper body kinematics of the whole body model, obtained by combining the lower limb model with an upper body component, are also presented. The lower limb injuries predicted by the model were validated against the published results of component lower limb impact tests performed under a boundary condition similar to actual car-pedestrian accidents, as well as a variety of different impact conditions. The upper body kinematics predicted by the whole body model were validated against the published results of full-scale tests, and a comparison of lower limb injuries was also carried out.

2. Model Validation of Lower Limb Injury Prediction

Previous studies utilized the results of lower limb impact tests^{(2),(5)} performed by Kajzer et al.^{(4),(5)} to verify lower limb models. However, the tests done by Kajzer et al. had the femur fixed in place with a jig, and consequently, the boundary condition was different from an actual car-pedestrian accident. Although the response of the knee joint could be verified with the test results, this could not be used for verification of injury prediction in car-pedestrian accidents. Bunketorp et al.⁽⁶⁾ added a mass equivalent to the upper body to the isolated lower limb, and applied an impact from a sled simulating a bumper and hood edge at a velocity of 32 km/h to the lower limb in a standing position. These tests were conducted under a total of four sets of test conditions, combining two bumper and hood edge heights with two bumper stiffness settings. Bone fracture and ligamentous damage were observed in different patterns with different impact conditions. In the present study, those tests were reconstructed using the finite element model of the human lower limb in a standing position developed by the authors, and the patterns of injury to the lower limb from the tests and computer simulations were compared for each set of test conditions.

Fig. 1 shows the model used for the simulation. The direction of impact was set, as in the tests, so that the sled would impact the lower limb at an angle of 15 degrees anteriorly from the pure lateromedial direction. In the tests,

a block weighing 47 kg was affixed to a part of the pelvis to represent the upper body, while in the model, the block and the femoral head were connected by a spherical joint for simplicity. The joint was given the range of motion and other properties of the human hip joint. The ground was modeled with a rigid plane, and the entire model was given the downward linear acceleration field for gravity. The height of the bumper and hood edge was set to match the tests, with the bumper center at two heights, 325 mm (low) and 450 mm (high) above the ground. The tests used a sled with parts representing the bumper and hood edge attached. Information about the mass of this mechanism and the stiffness of the hood edge, however, was not available in the literature. Therefore, the model employed input force wave forms for the bumper and hood edge as they were described in the literature, and used them as prescribed external forces. For simplicity, input from the hood edge was applied as a distributed load in the area of contact between the surface of the thigh and the hood edge. However, it seemed necessary to replicate the external shape of the bumper, because the bumper was located close to the region where injury occurred. Consequently, a prescribed external force was applied to a rigid body bumper model such as shown in Fig. 1. The tests employed a total of four types of impact conditions derived from the combination of two different heights of the bumper and hood edge and two different degrees of stiffness for the bumper components (rigid, compliant). The wave forms for the input force from the bumper and hood edge that correspond to the respective test conditions are shown in Fig. 2.

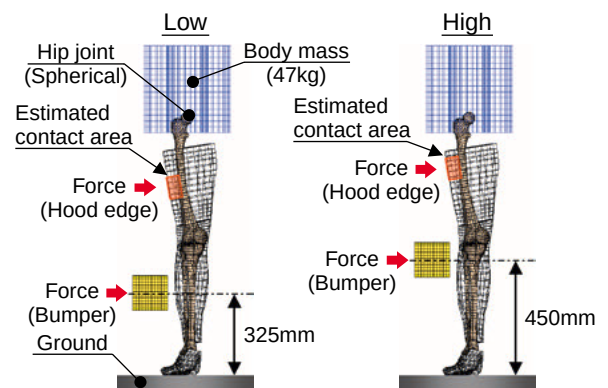


Fig. 1 Model setup for lateral impact validation

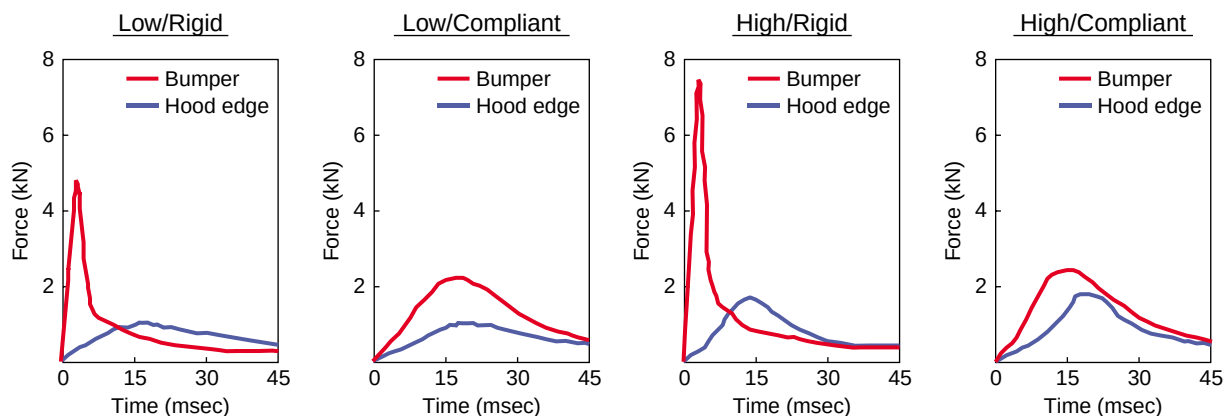


Fig. 2 External force applied to the lower limb

Simulations were performed for the four sets of test conditions described above, and comparisons were made for the peak linear acceleration in the lower limb and the patterns of damage to knee ligaments and leg fracture. Fig. 3 shows the results from comparison of the peak linear acceleration. The tests were conducted five times with each set of conditions, and the figure shows the resulting average as well as the upper and lower limits. The simulation results yielded values that were close to the test result averages for the linear acceleration at the impact level and the knee joint level. The simulation results fell almost completely between the upper and lower limits from the tests. The simulation results for linear acceleration at the ankle joint level, however, were higher than the test results. The simulation used an estimate for the coefficient of friction between the

foot and the ground because the corresponding coefficient in the tests was not available. It seems that this led to the difference in the kinematics of the foot between the tests and simulations, and thus resulted in the difference in the linear acceleration at the ankle joint level. The injuries observed in the tests, however, were damage to knee ligaments and bone fractures near the impact level, and as the linear acceleration in these two regions was accurately predicted, it was determined that this did not significantly affect the accuracy of injury prediction.

Fig. 4 shows an example of the kinematics of the lower limb from the simulation with a low bumper. In the case of a rigid bumper, a comparatively large deflection of the leg, particularly conspicuous at 15 msec, can be seen, and a proximal fibula fracture is observed without any damage

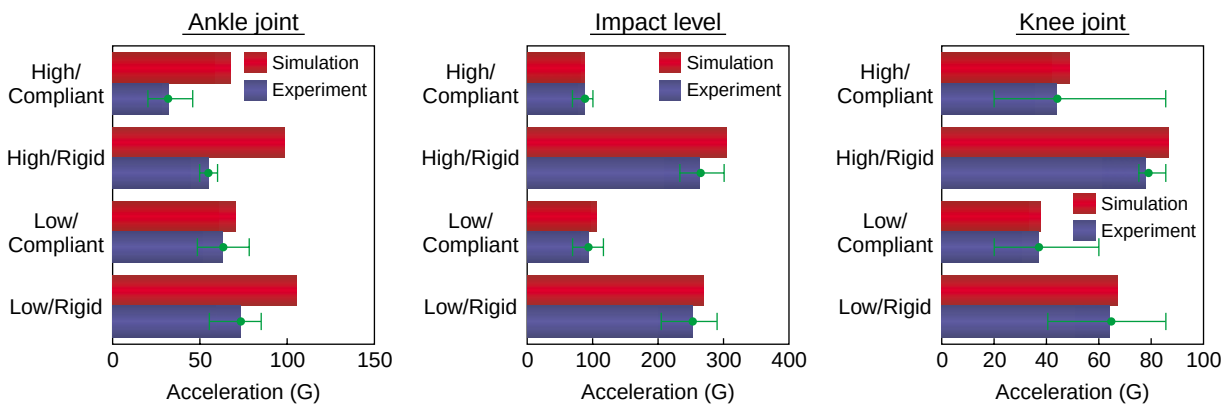


Fig. 3 Comparison of peak acceleration between experiment and simulation

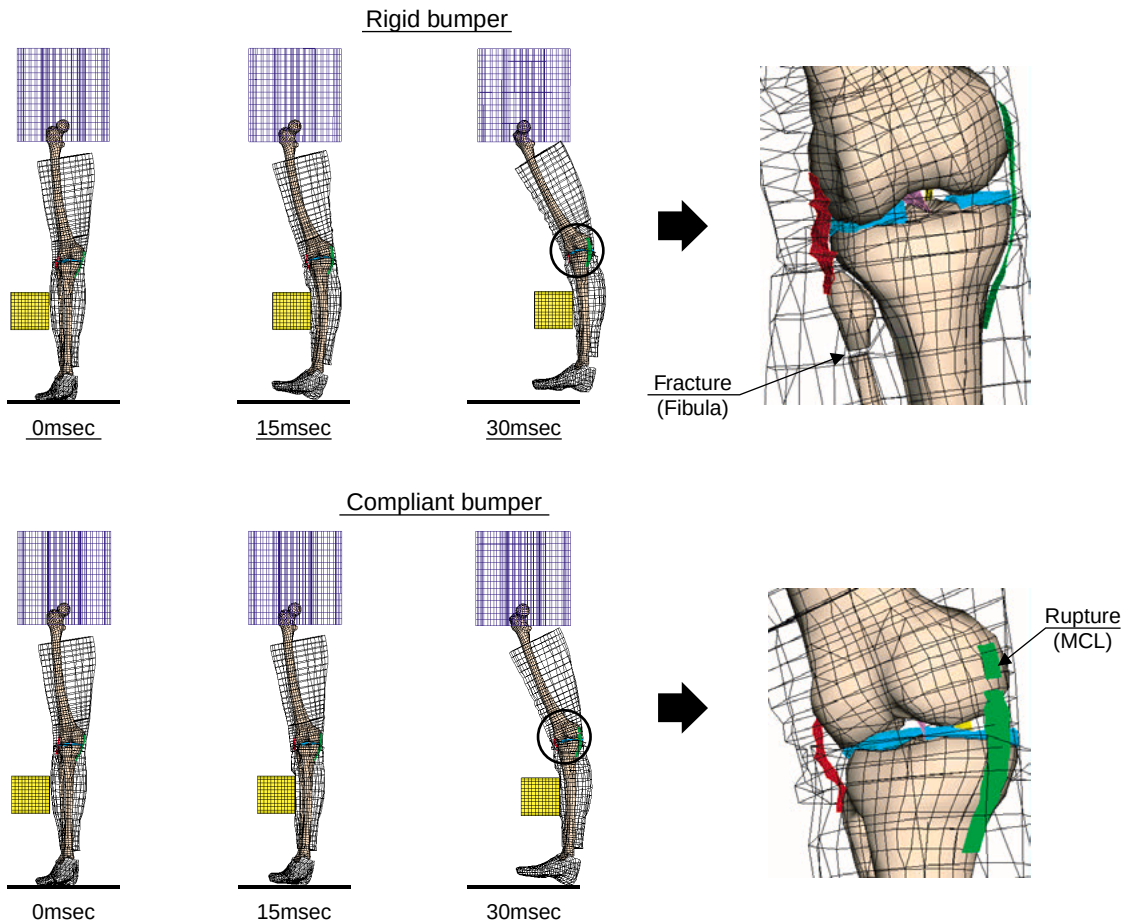


Fig. 4 Kinematics of the lower limb in low bumper configuration

to knee ligaments. Conversely, in the case of the compliant bumper, the deflection of the leg is slight, but a rupture of the medial collateral ligament is observed. Tables 1 and 2 present comparisons of the injury description including damage to knee ligaments and leg fracture for the case of the low bumper and high bumper, respectively. In the case of the low bumper (Table 1), fibula fracture was observed in all test cases where the bumper was rigid, while damage to knee ligaments occurred in only one of five cases. In cases where the bumper was compliant, rupture of the medial collateral ligament was observed in all test cases, while bone fracture occurred in only one case. In the simulation, only fibula fracture occurred with the rigid bumper, while only rupture of the medial collateral ligament occurred with the compliant bumper. It was found that the simulation predicts the difference in injury pattern in the tests due to the difference in stiffness of the bumper. Similarly, with a low bumper (Table 2), the injuries observed in more than half of the test cases were also predicted by the simulation. It was evident from these results that this model was able to predict the different injury patterns to the lower limb due to the difference in bumper height and stiffness.

3. Upper Body Modeling

In order to make a model capable of both analyzing the kinematics of the upper body and predicting lower limb injuries, a full body model for a pedestrian was constructed by incorporating a model of the pelvis and upper body into

the validated finite element model for the lower limb. Fig. 5 shows the structure of the full body model. The model of the pelvis and surrounding soft tissue was taken from the finite element human model (H-DummyTM)⁽⁷⁾ developed by Nihon ESI in order to enable the model to predict a wider range of injuries in the future including pelvic fractures. As the purpose of the upper body model was to predict kinematics, the model was constructed with rigid segments connected by joint elements so that the model would be comparatively simple and thus require less CPU time. It is crucial for the model to represent the flexibility

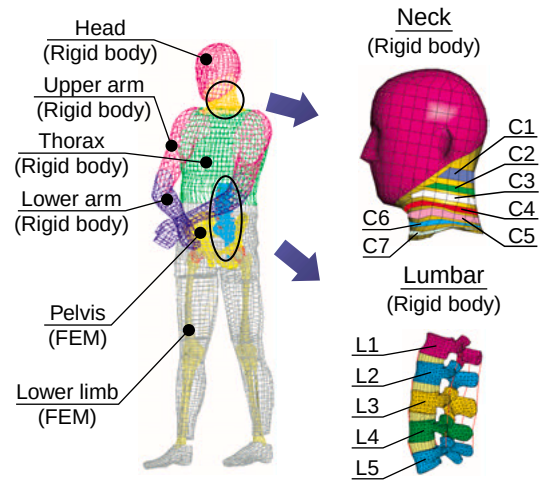


Fig. 5 Structure of full body model

Table 1 Comparison of injury description between experiment and simulation in low bumper setup

Rigid bumper							Compliant bumper						
Experiment No.	Ligament				Bone		Experiment No.	Ligament				Bone	
	ACL	PCL	LCL	MCL	Tibia	Fibula		ACL	PCL	LCL	MCL	Tibia	Fibula
Simulation						▲	Simulation				■		
11	■	■				▲	16				■		
12						▲	17				■		
13						▲	18				■		▲
14						▲	19				■		
15						▲	20				■		

* ■ : Ligament rupture, ▲ : Bone fracture, □ : Experiment, ■ : Simulation
 ACL : Anterior Cruciate Ligament, PCL : Posterior Cruciate Ligament, LCL : Lateral Collateral Ligament,
 MCL : Medial Collateral Ligament

Table 2 Comparison of injury description between experiment and simulation in high bumper setup

Rigid bumper							Compliant bumper						
Experiment No.	Ligament				Bone		Experiment No.	Ligament				Bone	
	ACL	PCL	LCL	MCL	Tibia	Fibula		ACL	PCL	LCL	MCL	Tibia	Fibula
Simulation	■			■			Simulation				■		
1	■			■			6					▲	▲
2	■			■			7						
3	■			■		▲	8		■		■		
4	■	■		■			9	■	■		■		
5	■			■			10				■		

* ■ : Ligament rupture, ▲ : Bone fracture, □ : Experiment, ■ : Simulation
 ACL : Anterior Cruciate Ligament, PCL : Posterior Cruciate Ligament, LCL : Lateral Collateral Ligament,
 MCL : Medial Collateral Ligament

of the spine in order for it to accurately predict the kinematics of the pedestrian upper body. The thoracic spine seems to be comparatively inflexible because this part of the spine, together with the ribs and sternum, forms the rib cage. However, the cervical spine and lumbar spine must be assigned the same degrees of freedom as the human body. Consequently, the thorax was modeled as a single segment for simplicity, while the cervical spine and lumbar spine were divided into seven segments and five segments, respectively, to match the number of vertebrae in each. The head, upper arm, and forearm were each modeled as a single segment.

Head injury is often the primary cause of death in fatal pedestrian accidents. Accurate prediction of the head motion is, therefore, particularly important in the prediction of the kinematics of the upper body. As the neck is considered to have a significant effect on kinematics of the head, the response of the head-neck model was subjected to validation separately from the upper body model. Thunnissen et al.⁽⁶⁾ analyzed the results from impact tests in lateral bending and flexion conducted by Ewing and Thomas⁽⁹⁾ using human volunteers to determine the corridor of the trajectory, linear acceleration, and rotational acceleration of the head relative

to the first thoracic vertebra. In this study, the wave form of the average linear acceleration of the first thoracic vertebra shown in the literature by Thunnissen et al. (Fig. 6) was applied to the joint between the thorax and the seventh cervical vertebra in the head-neck model, and the resulting trajectory, linear acceleration, and rotational acceleration of the head were compared with the corridor determined from the test results. The simulation results for the head-neck kinematics in lateral bending and flexion are shown in Fig. 7. A coordinate system as shown in the figure was used for wave form comparison. Fig. 8 shows the results of the comparison for the head trajectory. The simulation results slightly exceed the corridor in the vicinity of the peak displacement in lateral bending. Otherwise, however, the results are within the corridor, and can be considered to display a trajectory generally equal to that of the human body. Fig. 9 and Fig. 10 show the results from comparison of the linear acceleration and rotational acceleration of the head for the lateral bending and flexion, respectively. Although the simulation results are not entirely within the corridor of the test results, they did yield wave patterns and peak values that were close to the test results.

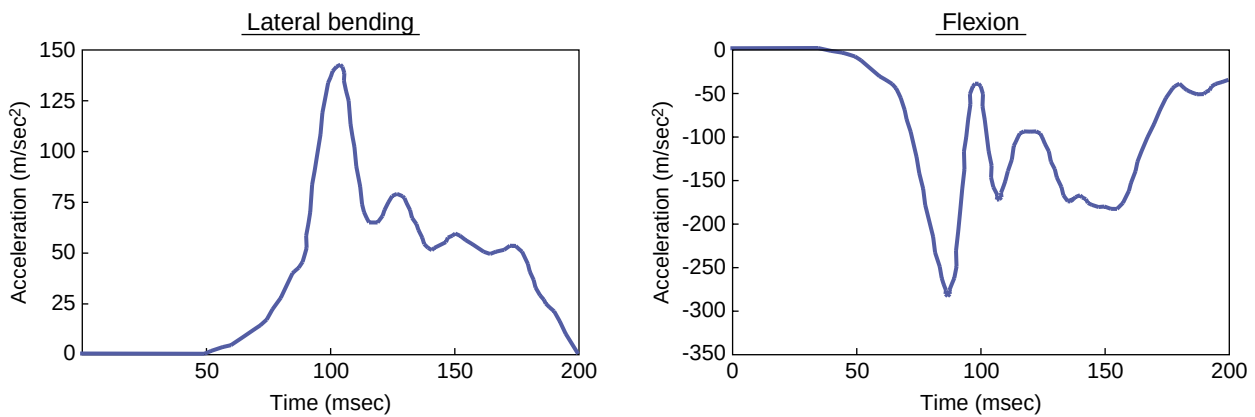


Fig. 6 Acceleration pulse applied to the first thoracic vertebra

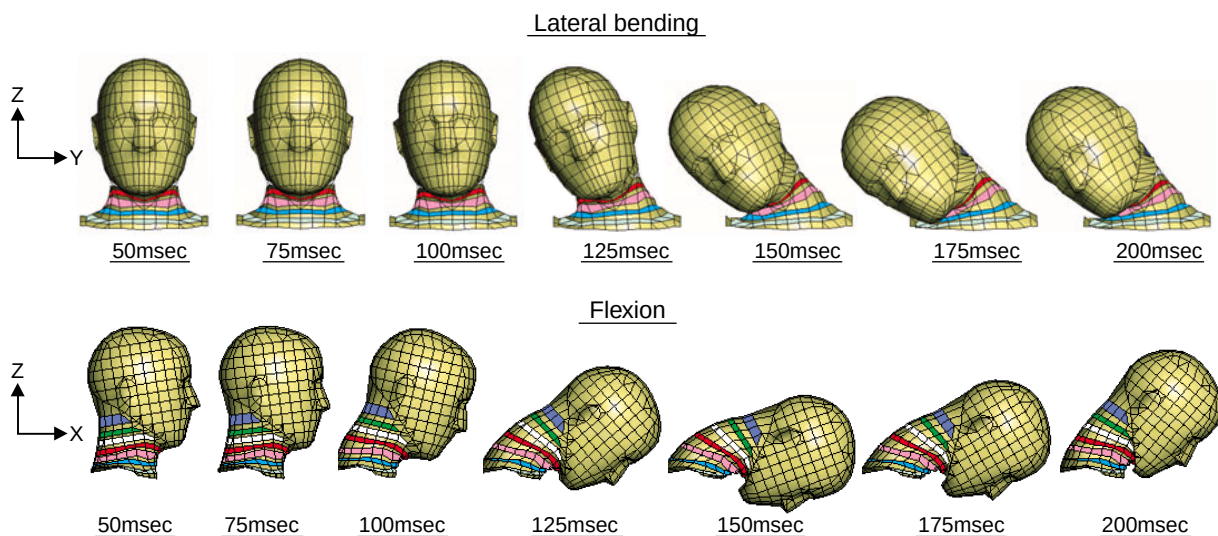


Fig. 7 Kinematics of the head-neck

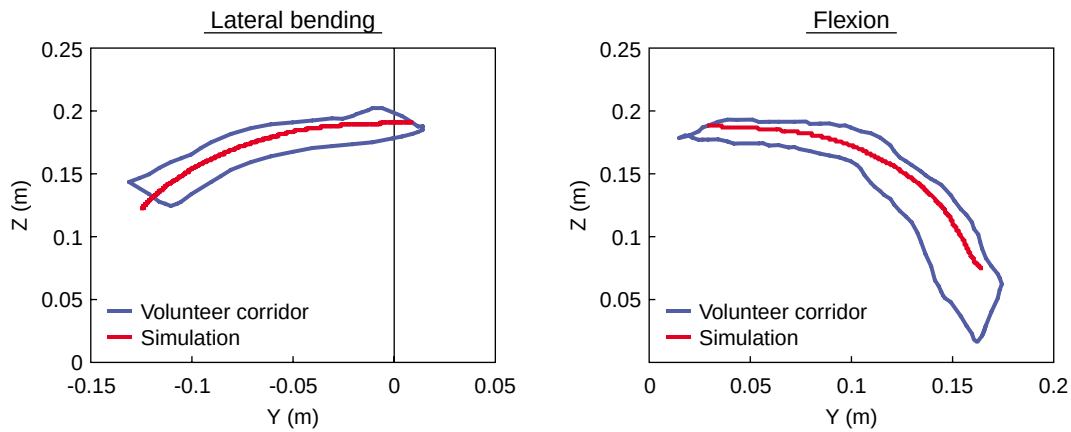


Fig. 8 Comparison of head trajectory between volunteer corridor and simulation results for lateral bending and flexion

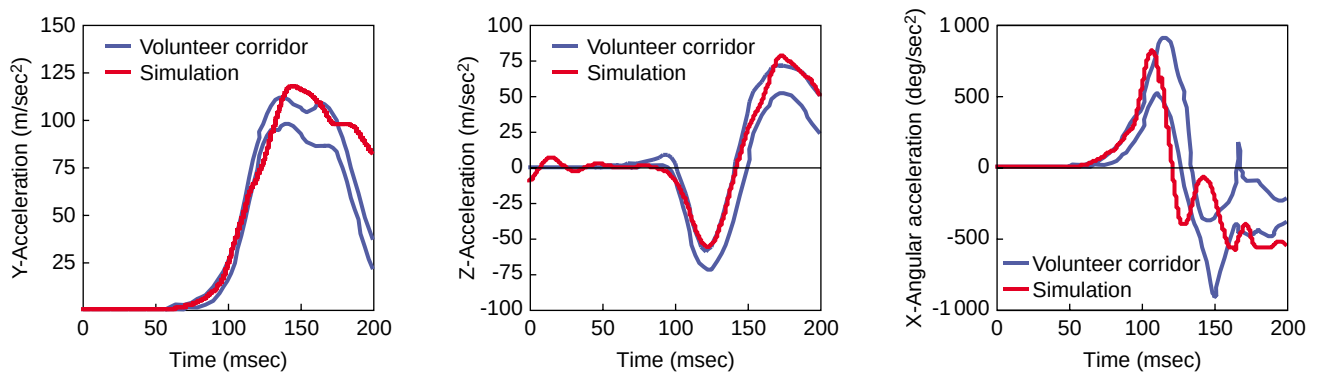


Fig. 9 Comparison of head linear and angular acceleration between volunteer corridor and simulation results for lateral bending

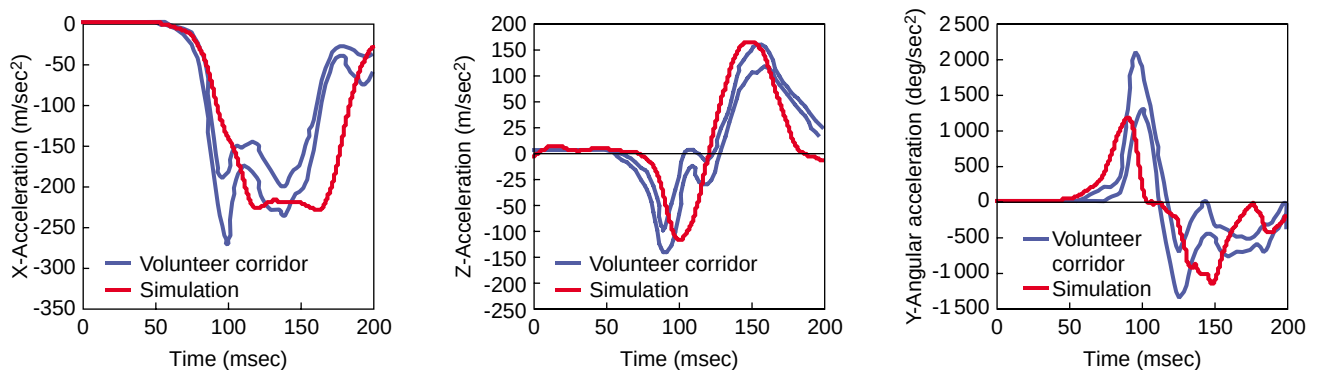


Fig. 10 Comparison of head linear and angular acceleration between volunteer corridor and simulation results for flexion

4. Model Validation for Upper Body Kinematics

Schroeder et al.⁽⁹⁾ affixed a cutaway body of a relatively recent model vehicle to a sled in order to perform full body car-pedestrian impact tests at velocities of 30 km/h and 40 km/h. Three tests were conducted with different human subjects, and since the full body model had been developed to simulate the average anthropometry of an American adult male, the test case with the subject that most closely resembled this anthropometry was chosen for computational reconstruction. Table 3 shows the geometry data for the car front and anthropometry data for the human subject used in this test case. The impact velocity was 40 km/h. The

simulation used a finite element model of a car with a front end geometry similar to this car. Fig. 11 shows the results from simulation for the kinematics of the lower limb. The lower limb injuries observed in the test were bone fractures of the tibia and fibula, and these injuries were similarly seen in the simulation. Knee ligaments were not damaged in either the test or simulation. The simulation results for the kinematics of the whole body in time series are shown in Fig. 12. Fig. 13 shows a comparison of the trajectory for the head, thorax, and pelvis. The kinematics of each region was predicted, and the full body model developed in this study was found to be very reliable with respect to the kinematics of the upper body as well as injuries to the lower limb.

Table 3 Test conditions (Schroeder et al., 2000)

Geometry of car front		Anthropometry of human subject	
Bumper height (mm)	446	Height (cm)	177
Bumper protrusion (mm)	141	Weight (kg)	84
Hood leading edge (mm)	641	Gender	Male
Hood length (mm)	1 110	Age (years)	66
Hood inclination (degree)	10		
Windshield inclination (degree)	35		

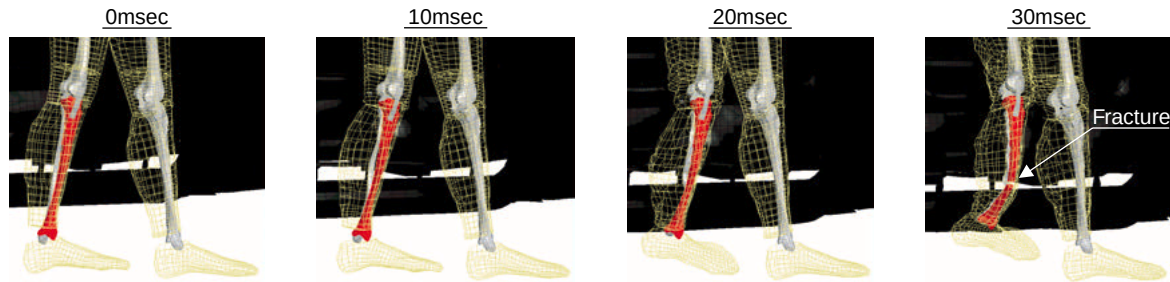


Fig. 11 Kinematics of the lower limb in car-pedestrian impact simulation

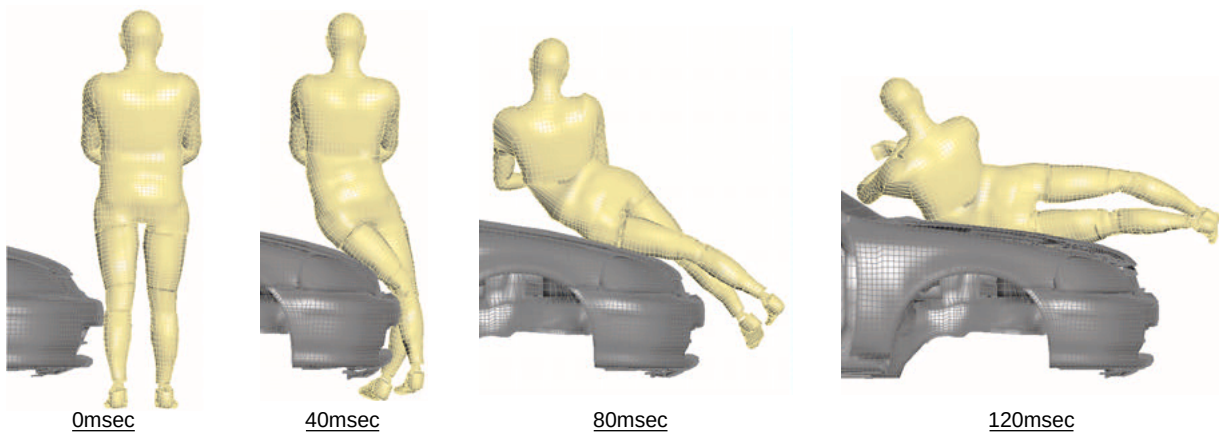


Fig. 12 Kinematics of the full body in car-pedestrian impact simulation (posterior view)

5. Conclusion

A finite element model for the lower limb that had been validated for the response of the bony structure and knee joint was used to simulate experiments conducted under conditions similar to actual car-pedestrian accidents. A full body model constructed by combining this lower limb model with an upper body model was also used to simulate a full-scale car-pedestrian impact test. Comparison of the results of these simulations with the test results yielded the following conclusions:

- (1) The lower limb model developed by the authors was found to be capable of predicting the different patterns of leg fractures and damage to knee ligaments that occurred due to the difference in impact conditions on the lower limb in situations close to an actual car-pedestrian accident.
- (2) The head-neck model with rigid segments connected by joint elements was found to be capable of reproducing the dynamic response similar to that of the human body, when each vertebra of the cervical spine was modeled as a separate segment in order to provide the same degrees of freedom as the human body.

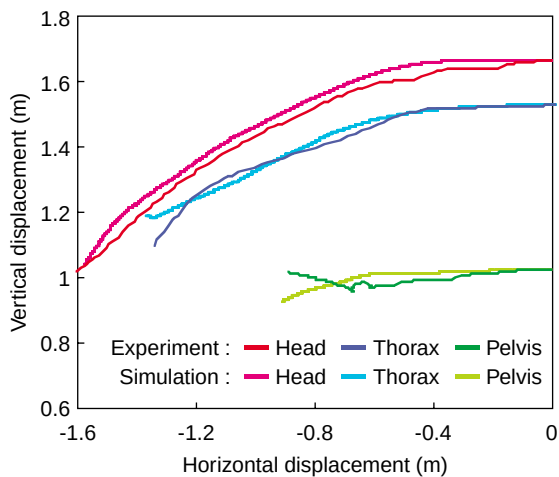


Fig. 13 Comparison of pelvis and upper body trajectories between experiment and simulation

- (3) It was found that the full body model developed in this study was capable of predicting the kinematics of the upper body as well as the injuries to the lower limb in car-pedestrian impact tests.

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