DEVELOPMENT OF A FINITE ELEMENT MODEL OF THE TOTAL HUMAN MODEL FOR SAFETY (THUMS) AND APPLICATION TO CAR-PEDESTRIAN IMPACTS

Tomoyuki Maeno

Toyota System Research Inc., Japan

Junji Hasegawa

Toyota Motor Corporation, Japan

Paper No. 494

1. ABSTRACT

"THUMS (Total Human Model for Safety)", as shown in Figure 1, is a finite element model of a human body developed to study various injury mechanisms, and is used as a substitute for the crash test dummies used for car occupants and pedestrians. "THUMS" is designed so that the whole body can be deformed and modeled in detail up to an AM50%ile size.

In this paper, "THUMS" is used as a pedestrian model and it is validated through the verification of pedestrian's whole body kinematics and lower extremity injuries. The simulation results are in good agreement with car-pedestrian impact test results using cadavers.

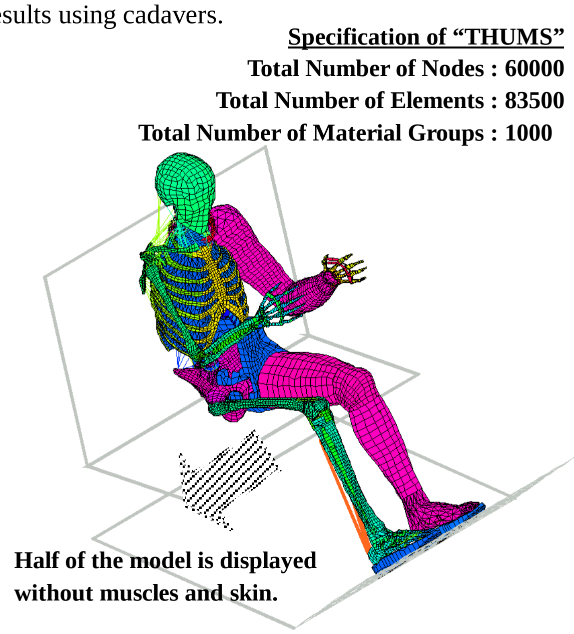


Figure 1. Outline of the "THUMS" (AM50%ile Size)

2. INTRODUCTION

Historically, various crash test dummies have been used for research purposes and in car safety tests. However, their strength and endurance characteristics are different from real humans since the dummies are intended for repeated use in crash tests. Therefore, a human body finite element model called "THUMS" is developed to study the detail injury occurrence mechanisms instead of crash test dummies.

"THUMS" has a high degree of bio-fidelity since it uses structures, shapes and material properties similar to a human body. For example, bone shape and strength are similar to real bones and soft tissues around the spine and lower extremity joints include ligaments and tendons. This paper presents an outline of "THUMS" for an AM50%ile size and examines lower extremity injuries in detail.

Next to head injuries, lower extremity injuries are the second most frequent injuries in car accidents. This paper focuses particularly on injuries to a pedestrian's lower extremity to validate "THUMS". First, modeling of lower extremity in "THUMS" is validated against cadavers for shearing and bending loads by comparing the dynamic response of a pedestrian's lower extremity due to loading from a car bumper. Secondly, "THUMS" is validated against the body kinematics of a cadaver in a car-pedestrian frontal impact test for a lower extremity impact. In both cases, the simulation results using “THUMS” are in good agreement with the cadaver tests.

3. LOWER EXTREMITY FE MODEL

The lower extremity of “THUMS”, as shown in Figure 2, consists of bones and soft tissues that include skin, muscles, ligaments and tendons. “THUMS” is developed to model a driver posture, as shown in Figure 1, since it is intended to investigate detail injury occurrence mechanisms as a substitute to using crash test dummies. For this paper, “THUMS” models a pedestrian through using a pedestrian posture. In addition, the source code of “THUMS” is now for crash analysis of PAM-CRASH.

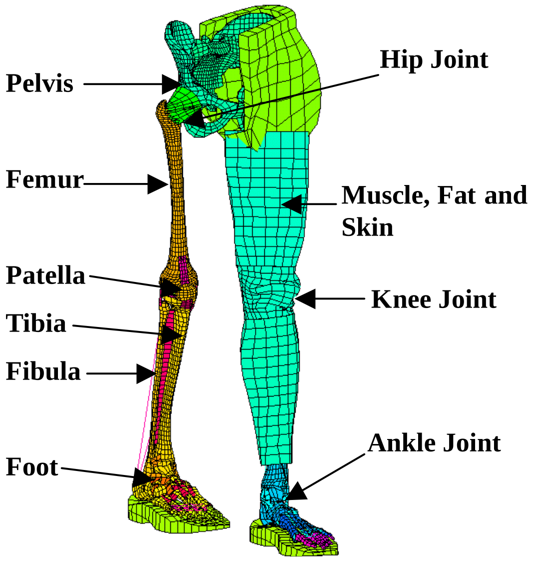


Figure 2. Lower Extremity FE Model

3-1. Lower Extremity Skeleton

The lower extremity skeleton consists of a pelvis, femur, tibia, fibula, patella and foot bones. As an example of a skeleton model, a femur FE model is shown in Figure 3. The bone is modeled by two layers, the outer hard layer called the cortical bone is modeled by shell elements, and the inner soft layer called the spongy bone is modeled by solid elements. The material property of each layer is determined on the basis of data published by Yamada (1).

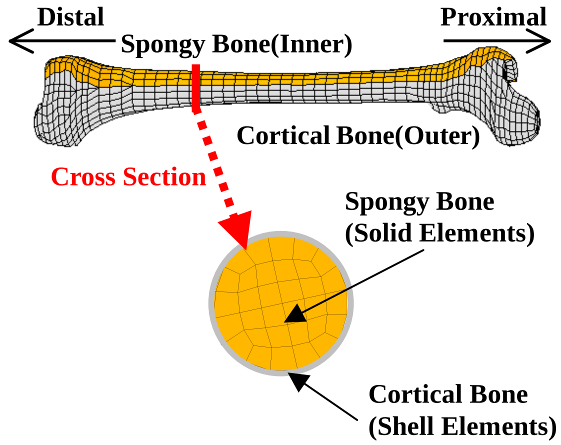


Figure 3. Femur FE Model (All View and Cross Section View)

As an example of validation of the lower extremity of “THUMS”, Figure 4 shows a simulation condition for a quasi-static three points bending simulation of the femur, the same condition presented by Yamada (1). Figure 5 shows the comparison between simulation results and test results, verified by Iwamoto et al. (2). The simulation results are clearly in good agreement with the test results for force-deflection curves obtained by Yamada (1).

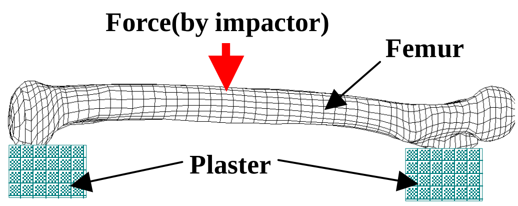


Figure 4. Quasi-Static Bending Simulation Model of Femur

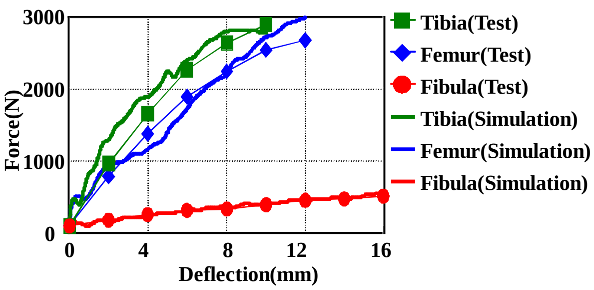


Figure 5. Comparison of Force-Deflection Curves Between Simulation Results and Test Results

3-2. Lower Extremity Joints

As an example of a joint model, a knee joint FE model is shown in Figure 6. Hip joints, knee joints, and ankle joints which have large articular movements are modeled and these bones are attached by major ligaments and tendons. The ligaments and tendons are modeled by membrane elements. The achilles tendon is modeled by bar elements.

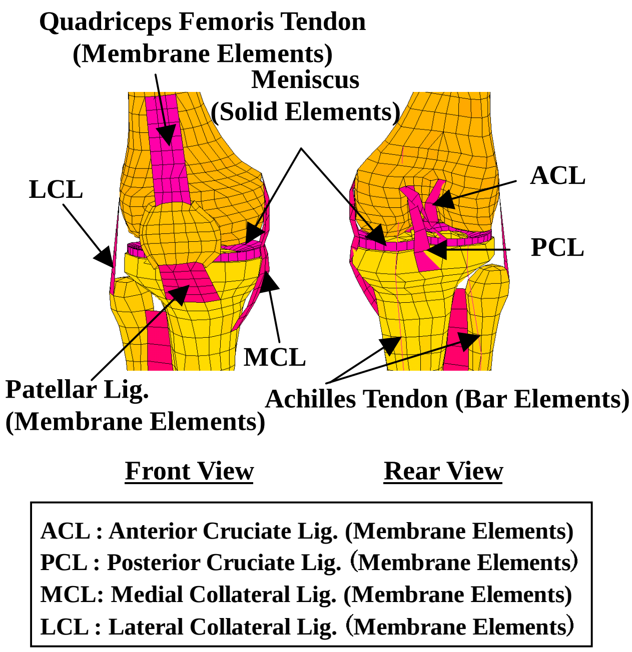


Figure 6. Right Knee Joint FE Model

As an example of validation of the lower extremity joint of "THUMS", Figure 7 shows a simulation condition for a dynamic toe impact simulation of the ankle joint, the same condition presented by Manning (3). This test was carried out to investigate tibial force and moment for a dynamic toe impact. Figure 8 shows the comparison between simulation results and test results verified by Tamura et al. (4). The simulation results are clearly in good agreement with the test results.

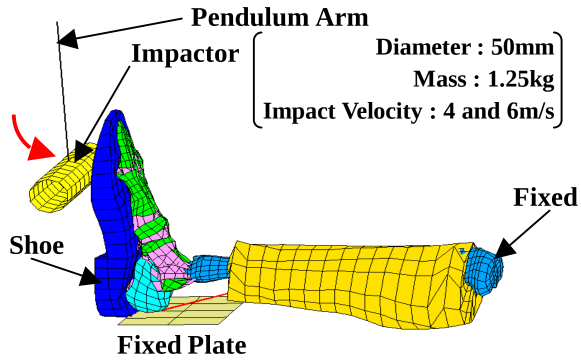


Figure 7. Dynamic Toe Impact Simulation Model

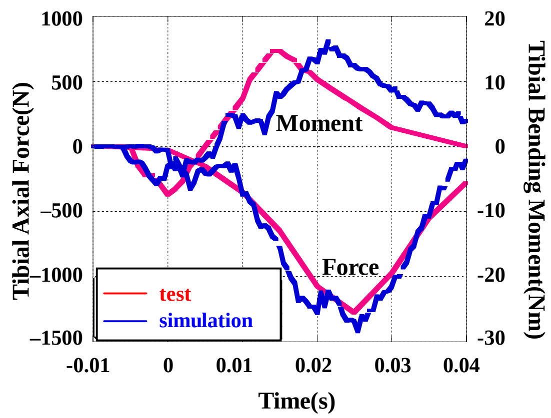


Figure 8. Comparison of Tibial Axial Force and Tibial Bending Moment Curves Between Simulation Results and Test Results

3-3. Lower Extremity Muscles and Skin

The skeleton is covered by muscles and skin. Muscles, which include fat, are modeled by solid elements. Skin, which is on the surface of the muscles, is modeled by shell elements.

As an example of validation of modeling in "THUMS", Figure 9 shows a simulation condition of a pelvic lateral impact, the same condition presented by Cesari et al. (5) and (6). This test was carried out to investigate the dynamic response in pelvic lateral impacts. Figure 10 shows the correlation between the peak force and the impact velocity. Generally, there is a similar tendency between the simulation and test results for peak forces near the upper bound of the test corridor, as obtained by Cesari et al. (5) and (6).

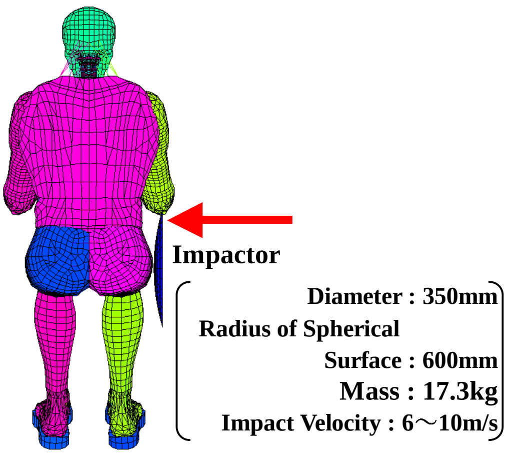


Figure 9. Pelvic Impact Simulation Model

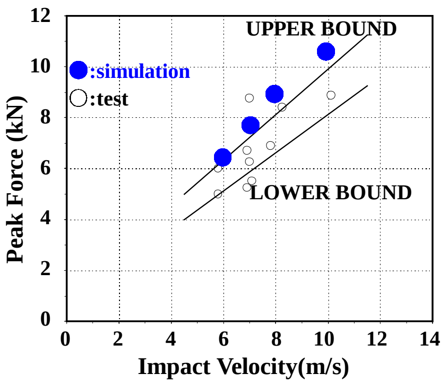


Figure 10. Response Corridor for Peak Force Versus Impact Velocity for Lateral Pelvic Impacts

4. VERIFICATION OF THE LOWER EXTREMITY UNIT OF A CADAVER

First, to investigate the validity of the lower extremity model in a car bumper impact, the lower extremity of "THUMS" is validated for the pedestrian impact test conditions used by Kajzer et al. (7) and (8). This test was carried out to investigate the damage tolerance of the extended knee joint when it is exposed to lateral impact loads in car-pedestrian accidents. In this study, it is suggested that fundamental injury mechanisms due to shearing and bending at the knee joint. Therefore, shearing and bending tests using the lower extremity of cadavers were carried out. The test condition for applying the shearing force at the knee joint, is shown in Figure 11, and the test condition for applying the bending moment at the knee joint, is shown in Figure 12.

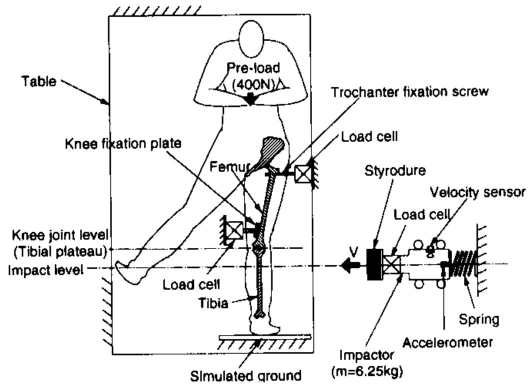


Figure 11. Test Setup for Shearing Tests

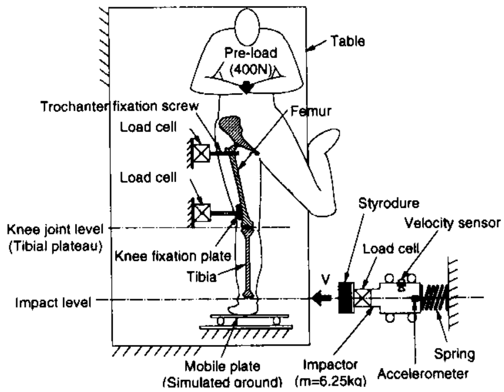


Figure 12. Test Setup for Bending Tests

The cadaver is laid down on its back, and a load of 400N, which corresponds to half the weight of the cadaver, is applied to one of the legs. The left leg of the cadaver is used in the shearing test, and the right leg of the cadaver is used in the bending test. To concentrate the load at the knee joint, the legs are fixed by two supporting points on the femur. The mass of the impactor is 6.25kg and the impact velocity is 20km/h (for low velocity) and 40km/h (for high velocity). In this section, the lower extremity is validated for low velocity test conditions, which resulted in less damage in cadavers and the results were repeatable in each of the tests. The damage levels in the low velocity tests are shown in Table 1.

Table 1.
Damage Levels from the Low Velocity Tests by Cadavers

Shearing Tests

TEST NO	Damages							
	Ligaments				Dia. or Meta.*		Epiphyses	
	ACL	PCL	MCL	LCL	emur	Tibia	emur	Tibia
21S	● 1							
24S								
25S								
28S	● 1				■			
29S	● 2							

●1:Ligament avulsion●2:Ligament stretching
■:Comminuted diaphysis fracture
***Diaphysis or Metaphysis**

Bending Tests

TEST NO	Damages							
	Ligaments				Dia. or Meta.*		Epiphyses	
	ACL	PCL	MCL	LCL	emur	Tibia	emur	Tibia
22B					■			
23B								
26B								
27B			● 1					
30B			● 2					

●1:Ligament avulsion●2:Ligament stretching
■:Metaphysis fracture
***Diaphysis or Metaphysis**

The simulation models are shown in Figure 13. Although the whole body of the cadaver is used in the tests, only the lower extremity of “THUMS” is used for the simulation since the leg is fixed by the two supports on the femur.

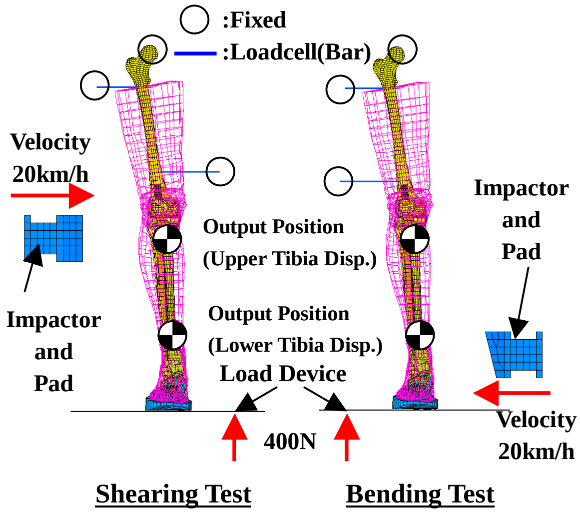


Figure 13. Shearing and Bending Test Simulation Models

The simulation results are shown in Figure 14. Test results are from TEST NO. 24S and NO. 23B shown in Table 1, since these tests used a cadaver with weight and height close to the “THUMS” models. For the shearing test condition, the peak value and the time of its occurrence for the simulation are in good agreement with the test results. The curves of simulated tibial displacement are also in good agreement with the test results qualitatively and quantitatively. For the bending test condition, the peak value and the time of its occurrence for the simulation are generally in good agreement with the test results. The curve of simulated lower tibial displacement is also generally in good agreement with the test result qualitatively and quantitatively. However, the upper tibial displacements obtained by bending test simulation tend to be larger than the test results. This difference is possibly caused by the knee joint modeling not in so detail. Therefore, for example, more detailed femur condyle and tibial plateau model are necessary.

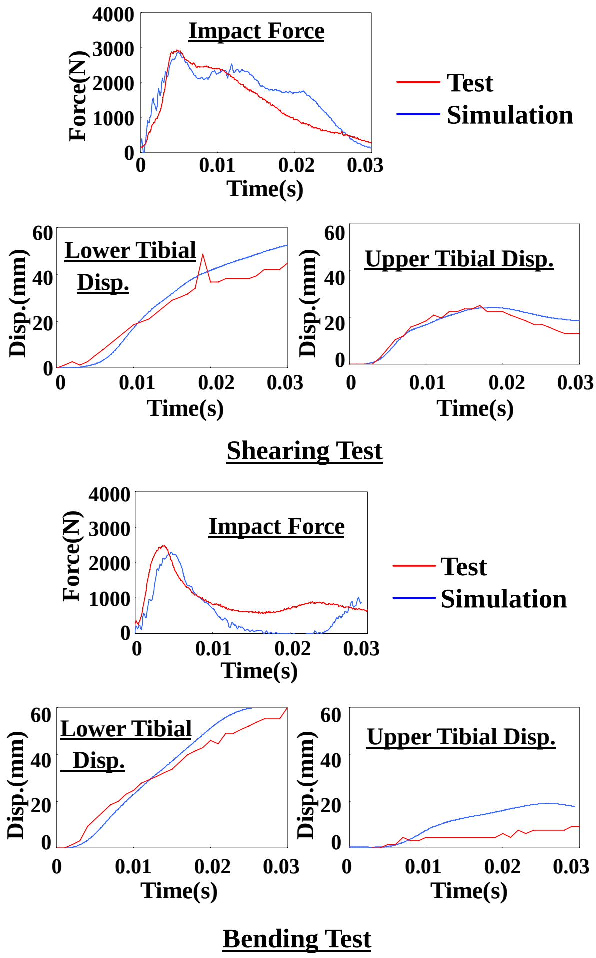


Figure 14. Comparison of Impact Force and Tibial Displacement Curves Between Simulation results and Test Results

In this section, test results are used from TEST NO. 24S and NO. 23B, shown in Table 1. These tests showed no injuries in lower extremity of cadavers. However, the avulsion and stretching of ACL occurred in other shearing tests, and the avulsion and stretching of MCL occurred in other bending tests. The knee joint ligament forces in the simulation are shown in Figure 15. There are similar load tendencies between the simulation results and the test results for the greatest ligament force occurring at ACL and PCL in the shearing test condition, and the greatest ligament force occurring at MCL and PCL in the bending test condition. The above findings validate the lower extremity model for the pedestrian low velocity impact test condition.

The knee joint ligament damage should be validated for high velocity test condition since for that condition the ligament damage occurs frequently. However, validation for the high velocity test condition is difficult since the test results were unrepeatable in each of the tests. Therefore, validation for this condition is omitted in this section.

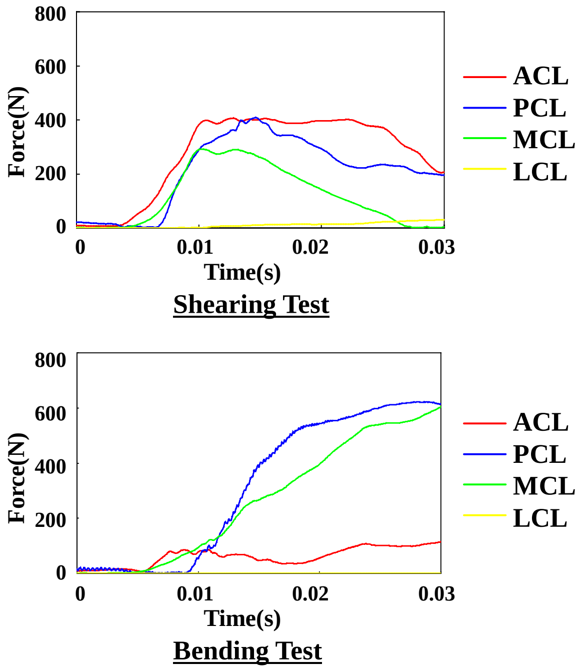


Figure 15. Force Curves of the Knee Joint Ligaments in Simulation

5. VERIFICATION OF FRONTAL CAR IMPACT TO THE PEDESTRIAN

In section 4, the lower extremity model is validated for pedestrian impact conditions by using a leg model of “THUMS”. By investigating the validity of lower extremity injuries of the pedestrian and whole body kinematics which involves the upper half of the pedestrian’s body, “THUMS” is validated for frontal car impact to the pedestrian, a test condition used by Schroeder et al. (9) and Ishikawa et al. (10). In this test, a section of a car collides with a cadaver. The test is designed to investigate the correlation between the shape of a car’s front-end and injuries to pedestrians. The test condition for this test is shown in Figure 16. A cadaver is supported by a rope so that it maintains a pedestrian posture. The car impact velocity is 40km/h.

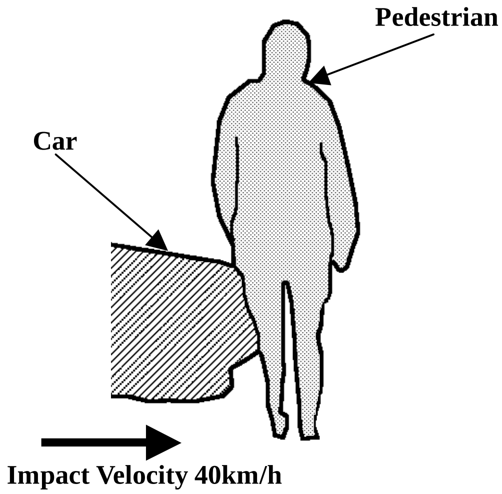


Figure 16. Test Setup for Frontal Car Impact to the Pedestrian Test

A simulation model is shown in Figure 17. The car model is made from an existing FE model by resizing the width and height. A grille similar to the test car, is added to the car model to simulate the effect on pedestrian kinematics during impact. The simulation condition is the same as the test, with an impact velocity of 40km/h assigned to the car model and gravity acceleration given to “THUMS”.

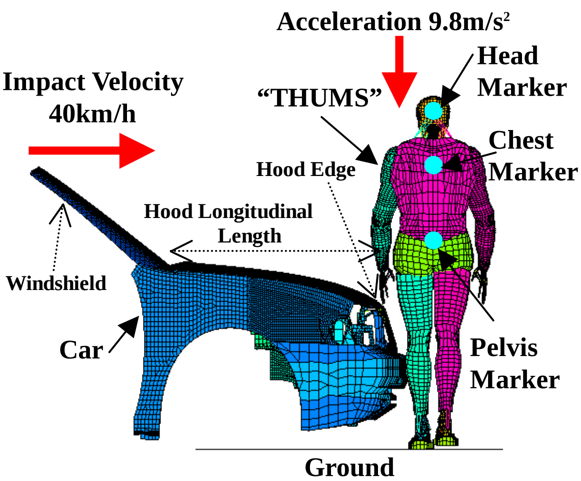


Figure 17. Frontal Car Impact to the Pedestrian Simulation Model

The simulation results are shown in Figures 18 and 19. It is found that the simulated body kinematics are generally in good agreement with test results. In Figure 19, the peak value and the time at which the peak value of the simulated left tibial acceleration occurs are in good agreement with test results. The curves of simulated horizontal displacement of each marker are also in good agreement with test results qualitatively and quantitatively. However, especially after 0.08sec, it is observed in Figure 18 that the pedestrian’s left elbow impacts the hood in test while it impacts the windshield in simulation. It is also observed that the position of the upper half of the body in simulation is higher than that in the test result. Furthermore, Figure 19 shows that the simulated vertical position of each marker is relatively higher than that in the test results. These differences are possibly caused by the differences in the hood edge height and hood longitudinal length between the simulation car model and the test car. Namely, since the hood edge of the car model is lower than that of the test car, the pelvis is apt to climb farther on the hood and the upper part of the body falls down slower in simulation. Furthermore, since the hood longitudinal length of the car model is shorter than that of the test car, the left elbow is apt to impact the windshield sooner in simulation. Based on the above findings, the whole body kinematics of “THUMS” is validated for the frontal car impact to the pedestrian test condition.

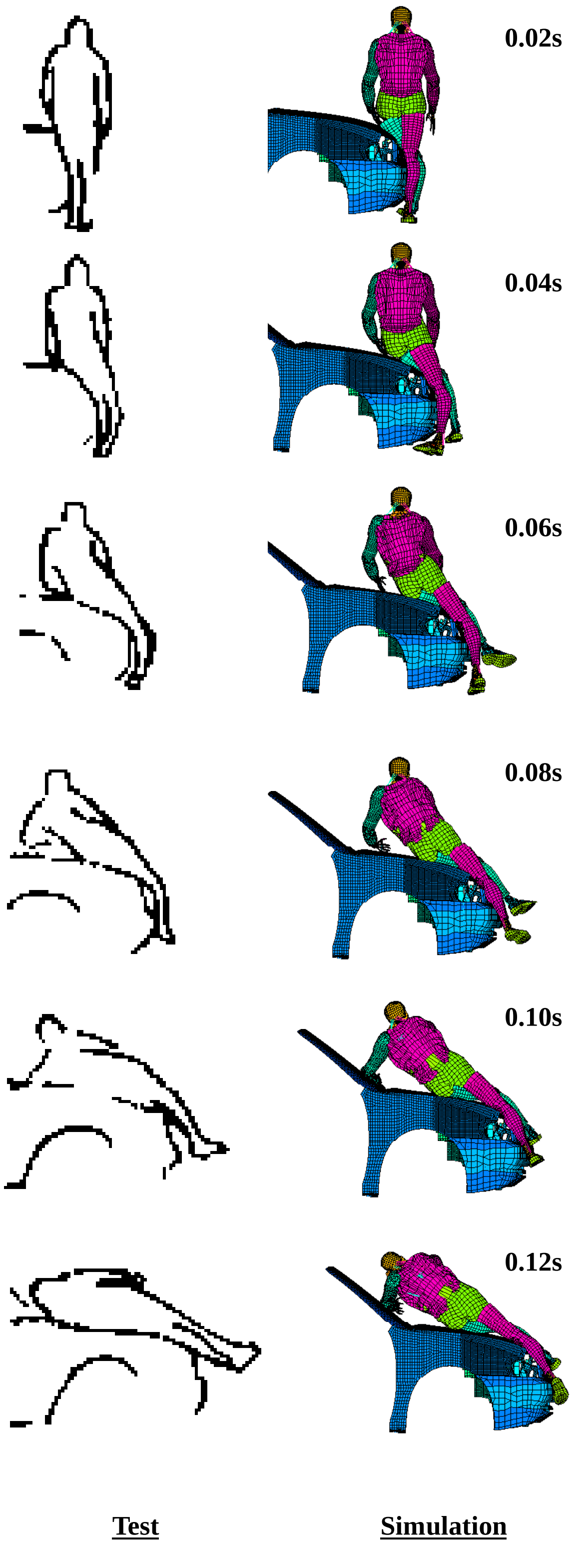


Figure 18. Comparison of the Kinematics Between Simulation Result and Test Result

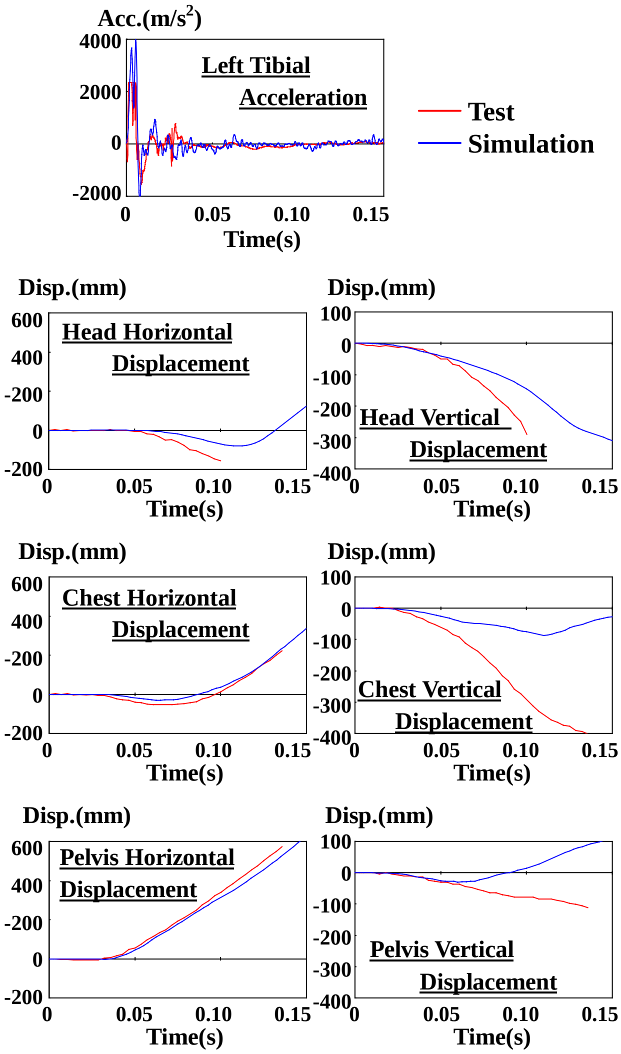


Figure 19. Comparison of Tibial Acceleration and Marker Displacement Curves Between Simulation Results and Test Results

The damage levels in the test are shown in Table 2. The lower extremity bone fracture injuries occurred in the tibia and fibula in both legs. This injury position is close to the impact position of the car bumper. The simulated stress distribution in the lower extremity cortical bones lower than the knee joint are shown in Figure 20. When the stress for bone fractures is set to about 140-150MPa as proposed by Yamada (1), it is found that the lower extremity injury predicted by "THUMS" is about the same level of stress distributed, which is in good agreement with that presented by Yamada (1).

Therefore, the lower extremity injury predicted by "THUMS" is generally in good agreement with that in the test. Based on the above discussion, it is found that the lower extremity injury predicted by "THUMS" is also validated for the frontal car impact to the pedestrian test condition.

As the future work, "THUMS" should be validated for the test condition which the ligament damage occurs frequently. Since in this paper, the lower extremity of "THUMS" is not validated for the condition when ligament damage occurs, as shown in Table 2. Furthermore, it is possible that the body kinematics are changed by bone complete fracture or ligament avulsion in real human body. In the future, the simulation of bone complete fracture and ligament avulsion will be added to "THUMS" by using the element elimination feature, namely, the element will be automatically eliminated when the stress or strain is over bone fracture and ligament avulsion level.

Table 2.
Damage Levels from Test by a Cadaver

Damage	**left leg:** **crash fracture of fibula 450mm above ground** **open wedge fracture of the tibia 335mm above ground** **right leg:** **crash fracture of tibia 340mm above ground** **crash fracture of fibula 340mm above ground** **shoulder:** **fracture of left clavicular** **crash fracture of the glenoid cavity of the left shoulder joint**

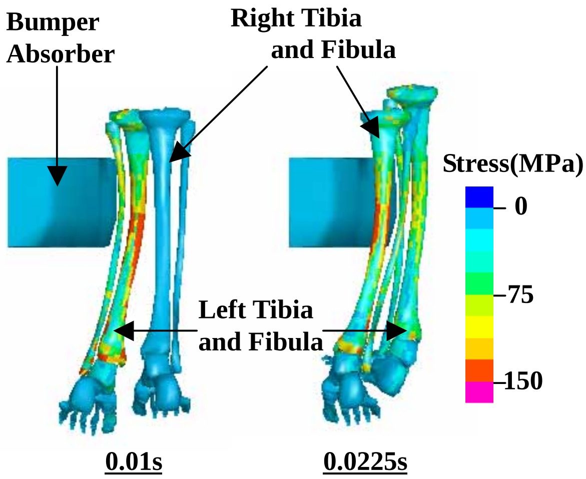


Figure 20. Distribution of Stress of Tibia and Fibula

6. CONCLUSIONS

[1] The lower extremity model of "THUMS" is validated for impact forces and tibial displacements under pedestrian impact test conditions (shearing and bending test by an impactor), since the simulation results are in good agreement with the cadaver test results.

[2] The whole body model of "THUMS" is validated for whole body kinematics and lower extremity injuries under the pedestrian impact test condition (frontal impact test by a car), since the simulation results are in good agreement with the cadaver test results.

[3] Based on [1] and [2], an effective simulation tool is developed for estimating injury and kinematics of a pedestrian impact with a car. Future works include improving the accuracy of simulation of whole body kinematics and injury, and improvements to enable reliable estimates of bone complete fracture and ligament avulsion by using the element elimination feature.

7. ACKNOWLEDGMENTS

The authors would like to express our gratitude to those concerned at Toyota Central R&D Labs., Inc., Dr. Janusz Kajzer of Kabimec Consulting, and Dr. Hirotoshi Ishikawa of Japan Automobile Research Institute for their assistance, advice, and cooperation.

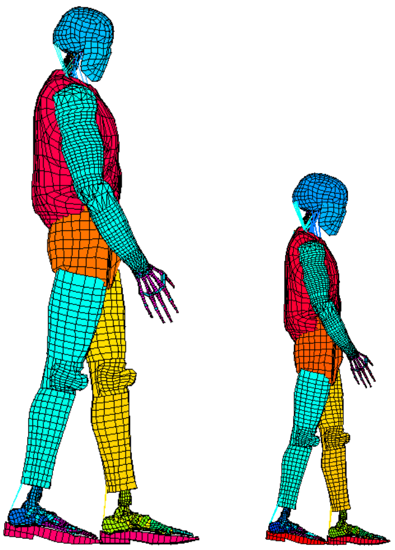
8. FUTURE DEVELOPMENTS

Future development schedules are shown below.

・Development of "THUMS" for LS-DYNA.

・Development of "THUMS" of AF05%ile size.

・Development of "THUMS" of 6-Year-Old Child size. (Shown in Figure 21)



AM50%ile Size **6-Year-Old Child Size**

Figure 21. Contrast of AM50%ile Size and 6-Year-Old Child Size (Trial Model)

9. REFERENCES

(1) Yamada H., Strength of Biological Materials, The Williams & Wilkins Co., 1970.

(2) Iwamoto M. et al., Development of a Finite Element Model of the Human Lower Extremity for Analyses of Automotive Crash Injuries, SAE 2000 World Congress 2000-01-0621, 2000.

(3) Manning P. et al., Dynamic response and injury mechanism in the human foot and ankle and an analysis of dummy biofidelity, ESV Conf., 1998.

(4) Tamura A. et al., Development of a Finite Element Model of the Human Lower Extremity for Assessing Automotive Crash Injury Potential, Human Biomechanics and Injury Prevention, Springer -Verlag Tokyo, 2000.

(5) ROAD VEHICLES-ANTHROPOMORPHIC SIDE IMPACT DUMMY-LATERAL IMPACT RESPONSE REQUIREMENTS TO ASSESS THE BIOFIDELITY OF THE DUMMY, ISO / TC22 / SC12 / WG5, Document N455, 1995.

(6) Cesari D. et al., Evaluation of Pelvic Fracture Tolerance in Side Impact, SAE Paper 801306, 1980.

(7) Kajzer J. et al., Shearing and Bending Effects at the Knee Joint at Low Speed Lateral Loading, SAE 1999 International Congress 1999-01-0712, 1999.

(8) Kajzer J. et al., Shearing and Bending Effects at the Knee Joint at High Speed Lateral Loading, SAE Paper 973326, 1997.

(9) Schroeder G. et al., Injury Mechanism of Pedestrians During a Front-End Collision with a Late Model Car, JSAE Spring Convention, 2000.

(10) Ishikawa H. et al., Computer Simulation of Impact Response of the Human Body in Car-Pedestrian Accidents, Stapp Car Crash Conf., 1993.