

Development of a Finite Element Model for the Pedestrian Lower Limb

– Development and Validation of a Knee Joint Model –

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ABSTRACT

A finite element model for a human lower limb in a standing position has been developed that connects long bone models of a lower limb at the knee with ligament models. The knee joint model incorporates both the articular cartilage and the meniscus, which are deemed to be important elements in the accurate prediction of ligament loading. The model has been validated against published lateral lower limb impact tests, and the results showed that the model can accurately reproduce the shearing displacement and bending angle of the knee joint. In addition, comparison of the model with test data on damage to four main knee ligaments showed that the model was able to simulate the ligamentous damage that had been observed in more than half of the test cases.

1. Introduction

The two main types of lower limb injury in car-pedestrian accidents are bone fractures and damage to knee ligaments. Consequently, investigative tools for identifying the mechanism of injury to the pedestrian lower limb need to be able to reproduce ligamentous damage of the knee in addition to bone fractures. This paper describes the development and validation of a pedestrian knee joint model constructed by combining lower limb bone models developed by the authors.

Damage to knee ligaments, unlike at the bony structure, is difficult to treat medically because ligaments have less capacity for self-repair, and this kind of damage has long been recognized as a serious injury. In addition, the knee joint does not experience the kind of constraint found in other joints, where the bones are fitted together. Instead, the

stability of the knee joint is entirely dependent on ligaments. This special characteristic of the anatomic structure means that damage to knee ligaments can seriously affect walking function and often require a long time for recovery. Given this context, attention has been focused on damage to knee ligaments from pedestrian accidents. Working Group 17 of the European Enhanced Vehicle-safety Committee (EEVC) has proposed a test procedure using a legform impactor⁽¹⁾, where the shearing displacement and bending angle of the knee are used as injury criteria for knee ligament.

Past studies involved development of simulation models that focused on the anatomic structure of the knee in order to identify the mechanism of damage to knee ligaments. Yang and Kajzer⁽²⁾ developed a knee joint model using the MADYMO program. This model represented the femoral condyle, tibial plateau, and tibial eminence with combinations of planes and ellipsoids, and modeled knee ligaments with spring elements. The damage to knee

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ligaments was compared with test results, but the model could not represent either the contact between collateral ligament and the bony structure or the deflection of the bone. As a result, the model was unable to accurately predict the load on the ligaments. Yang et al.⁽³⁾⁽⁴⁾ attempted to predict bone fractures by adding a joint element to the bone in this model. However, this required determining the site of bone fractures in advance. Although these multibody models can simulate knee response to some extent by accurately representing the anatomic structure of the knee, their accuracy is limited by the significant effect of the bone deflection on the loading of the knee ligaments. Schuster et al.⁽⁵⁾ developed a finite element model of the pedestrian lower limb. However, validation was only conducted with shearing displacement applied to the knee, and the loading rate was limited to a low speed (15 km/h).

In the present study, a lower limb model for pedestrians in a standing position was constructed by combining finite element models for bones in the lower limb that have been developed by the authors and validated for impact response. In order to accurately predict load on the knee ligaments, the articular cartilage and the meniscus were incorporated since these elements were deemed to have a significant effect on elongation of the ligaments. In addition, contact between the ligaments and bony structure was defined in order to accurately reproduce elongation of the ligaments due to knee joint deformation. The dynamic response of the knee was validated against published results of lower limb impact tests on impact force, shearing displacement, and bending angle. The injury description for ligamentous damage was also compared with the test results. Validation was carried out for a total of four conditions that combined two loading rates with two test conditions for knee shearing and bending.

2. The Structure and Material Properties of the Knee Joint

The structure of the human knee joint is shown in Fig. 1. The knee joint is the articulation between the tibia and femur, and is formed by the distal femur overlapping the flat portion of the proximal tibia. Consequently, there is no effective constraint among the bones of the knee joint, and the stability of the joint is maintained primarily by attachment of the ligaments. As shown in the figure, the knee is held together by four ligaments, the Anterior Cruciate Ligament (ACL), Posterior Cruciate Ligament (PCL), Lateral Collateral Ligament (LCL), and Medial Collateral Ligament (MCL). The cartilaginous meniscus is

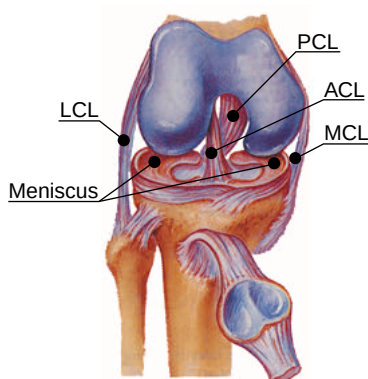


Fig. 1 Structure of knee joint

located around the tibial plateau, and the joint surfaces of the femur and tibia are covered with articular cartilage. These cartilaginous structures perform a buffer role when the knee deforms axially, suggesting that they affect the elongation of ligaments due to knee deformation by impact.

Fig. 2 shows a typical stress-strain curve for knee ligaments. The collagen fibers inside the ligament take a coiled form when they are not under load, and their orientation is chaotic. Application of tension, however, causes the coiled fibers to elongate until they become straight and their orientation also becomes aligned in the direction of the tension. This process corresponds to region I in Fig. 2. Under further loading, the tensile properties of the straightened collagen fibers causes them to show a more or less fixed modulus as in region II. Wismans⁽⁶⁾ showed that since region I can be approximated accurately by a quadratic function, then taking the continuity of the first order derivative into account, the average modulus of region I can be approximated by 1/2 of the modulus of region II. The present study used this relationship and approximated the stress-strain curve by a three-stage straight line as in Fig. 2. The five parameters of region II were determined by a survey of the literature⁽⁷⁾⁻⁽¹¹⁾: modulus (E), yield stress (σ_y), yield strain (ϵ_y), ultimate stress (σ_u), and ultimate strain (ϵ_u).

Fig. 3 shows the results of the survey for the ultimate stress. The horizontal axis shows the strain rate on a logarithmic scale, and data for the various ligaments are

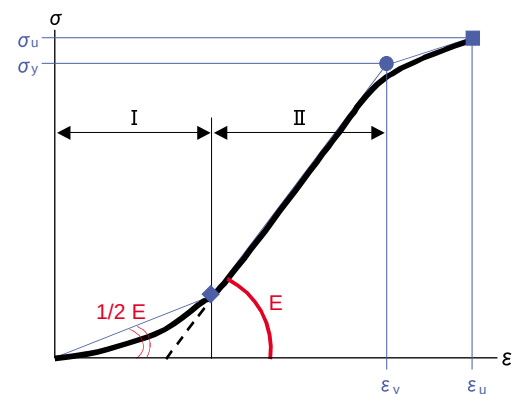


Fig. 2 Typical stress-strain curve for knee ligament

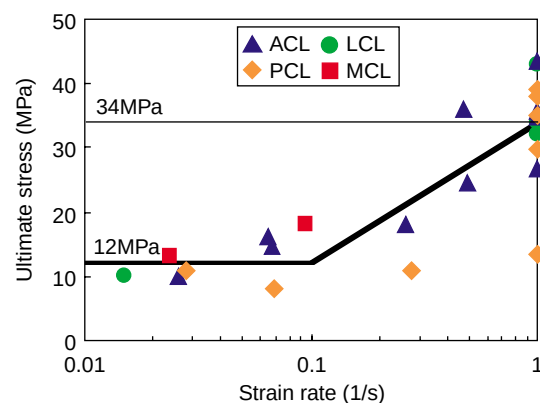


Fig. 3 Ultimate stress of knee ligaments as a function of strain rate

represented by different colors. As no significant differences in ligament properties were observed, the decision was made to assign the same properties to all ligaments. In the region where strain rate exceeds 0.1 /s, the ultimate stress was found to have a tendency to increase. Therefore, the relationship represented by a straight line in the figure was established as strain rate dependency. As no data were found for the region where the strain rate exceeds 1.0 /s, this was established by extrapolation. As shown in Fig. 4, no significant strain rate dependency was observed for the ultimate strain. Therefore, only average values were determined. As little data on the yield stress and yield strain were available in the literature, ratios between these and the ultimate stress and ultimate strain, respectively, were established as in Fig. 5. Although the modulus showed a tendency to increase with the strain rate, as shown in Fig. 6, there was a wide variety of data. It was assumed, therefore, that the strain rate dependency of the modulus was identical to that of the ultimate stress, and it was represented by a relationship depicted by the straight line shown in the graph. The skin was also assumed to have an effect on the mechanical properties of the knee joint, and its properties were established with reference to the literature.^{(12) (13)} A search of the literature with reference to the cartilage and meniscus⁽¹⁴⁾ did not yield data for the plastic region. Therefore these elements were modeled as linear elastic materials. The quasi-static stress-strain curve for the ligaments and skin determined in this way is shown in Fig. 7, and that for the cartilage and meniscus is shown in Fig. 8. As only quasi-static data could be obtained for the skin, cartilage and meniscus, it was assumed that they had a strain rate dependency identical to that of ligaments.

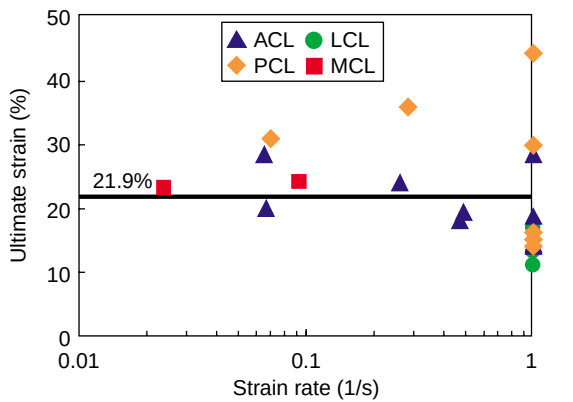


Fig. 4 Ultimate strain of knee ligaments

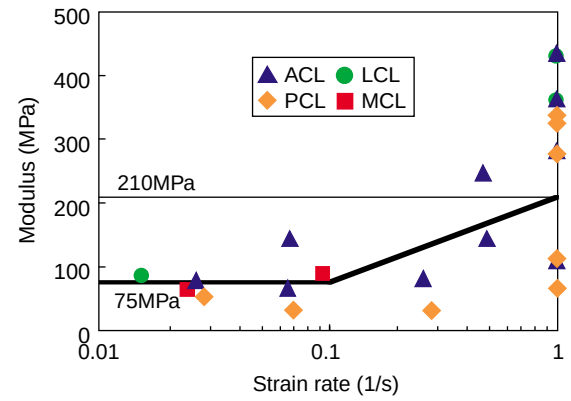
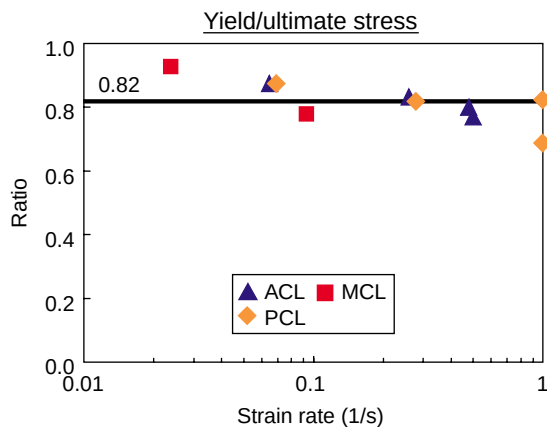


Fig. 6 Modulus of knee ligament

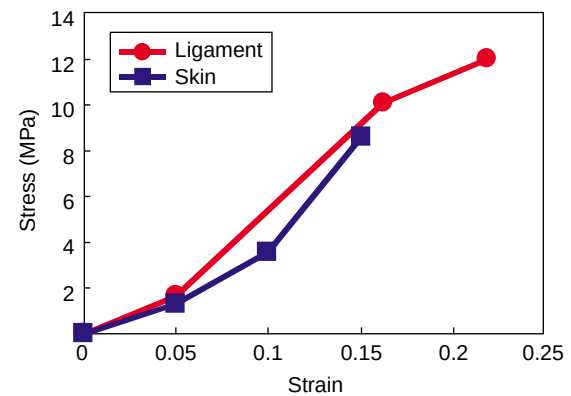


Fig. 7 Quasi-static stress-strain curve for knee ligaments and skin used in the model

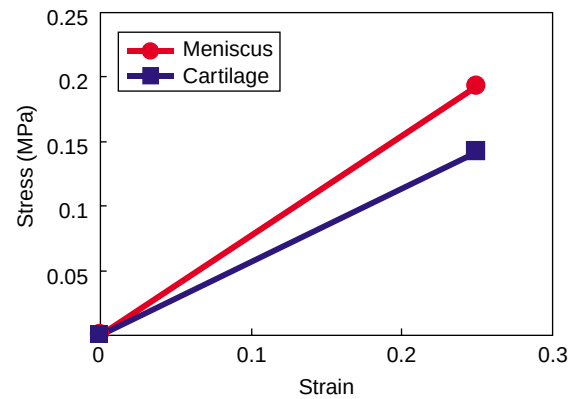


Fig. 8 Quasi-static property for meniscus and cartilage used in the model

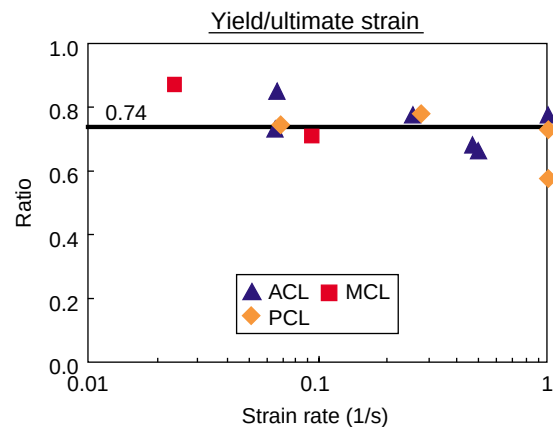


Fig. 5 Ratio between ultimate stress/strain and yield stress/strain

3. Model Description

In constructing a knee joint model by combining bone models, the lower limb was aligned with reference to the literature⁽¹⁵⁾ so that the hip joint, knee joint, and ankle joint were located on a single straight line, as shown in Fig. 9. Knee ligaments were modeled with shell elements, as shown in Fig. 10. In order to accurately reproduce the elongation of ligaments due to joint deformation, a contact between the ligaments and bones was defined. Data was obtained from the Medical School of Hannover in Germany regarding the length and cross sectional area of each ligament, and these parameters were defined as shown in Table 1 and Table 2. The two ends of the ligament model were connected to the bone model by defining a rigid body in the area of the ligament attachment so that the ligament had an effective length consistent with the data in Table 1. All ligaments were given the same width (15 mm) for simplicity, and then a different thickness was set for each ligament according to its cross sectional area provided in Table 2. Ligament rupture was reproduced using the Element Elimination option^{(16) (17)} in PAM-CRASHTM, which

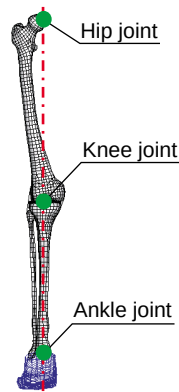


Fig. 9 Alignment of lower limb model

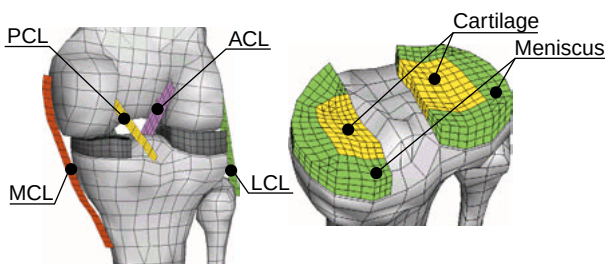


Fig. 10 Knee joint model

Table 1 Length of knee ligament used in the model

Ligament	ACL	PCL	LCL	MCL
Length (mm)	40	41	55	91

Table 2 Cross sectional area of knee ligament used in the model

Ligament	ACL	PCL	LCL	MCL
Area (mm ²)	46	77	29	37

takes the stiffness of elements that reach the threshold set for the ultimate strain and steps them down to zero. Solid element models for the cartilage and meniscus were incorporated on the tibial plateau (Fig. 10) in order to reproduce the axial deformation of the knee joint. The soft tissue around the bony structure of the thigh and leg that consists mainly of the muscle and fat was modeled with solid elements, as shown in Fig. 11. A model for this part was taken from a finite element model of the human lower limb (H-DummyTM) developed by Nihon ESI. This model was validated against the results of dynamic indentation tests conducted from every direction using human volunteers.⁽¹⁸⁾ The skin modeled with shell elements was added to the surface of this part, since it was deemed to have an effect on the knee response.

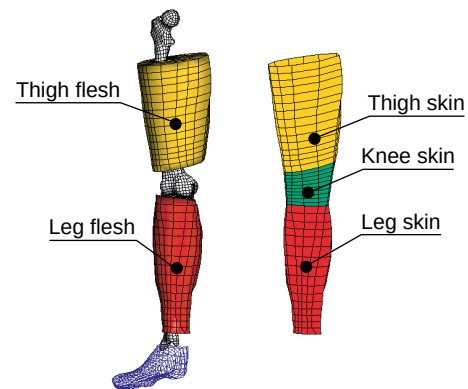


Fig. 11 Schematic diagram of lower limb model

4. Model Validation

The knee model was validated against the results of lower limb impact tests conducted by Kajzer et al.^{(19) (20)} The test conditions are shown in Fig. 12. Kajzer et al. used a jig to fix the femoral supracondyle and greater trochanter in place with a load cell interposed on each, and then propelled a 6.25 kg impactor into the leg laterally at velocities of 20 km/h and 40 km/h. The impactor was used with two impact conditions; one was a shearing setup to impact immediately below the knee joint to apply shearing deformation, and the other was a bending setup to impact against the ankle joint to apply bending deformation. A 400 N preload in the inferior direction was applied to the lower limb to simulate a walking condition, and a piece of foam was attached to the impactor head in order to prevent injury from direct impact.

The model shown in Fig. 13 was created to reproduce these tests. Kajzer et al. conducted their tests with the whole body of a human subject. In the experiments, no significant displacement of the upper body was observed during impact to the lower limb. Therefore, for simplicity, the model only represented the lower limb, and the femoral head that is the juncture with the upper body was fixed in inertial space. A boundary condition was applied to elements in the vicinity of the fixed parts of the greater trochanter and supracondyle of the femur to fix them in inertial space. The preload was applied to the lower limb by moving the surface corresponding to the ground in superior direction prior to the impact from the impactor. Only the foam material on the impactor head was modeled with solid elements, with the remaining impactor mass being applied to a single node so that the center of gravity would

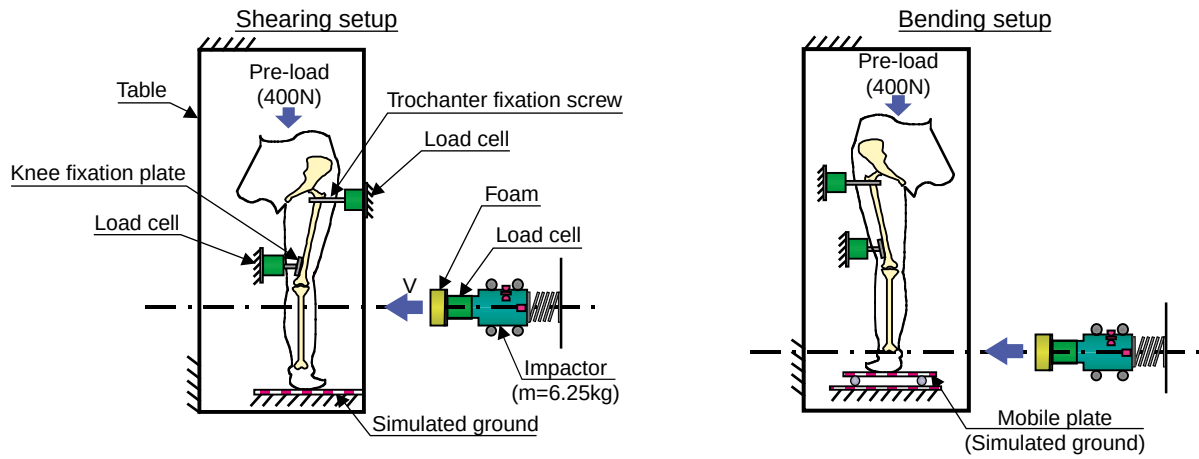


Fig. 12 Test setup for dynamic response of knee joint

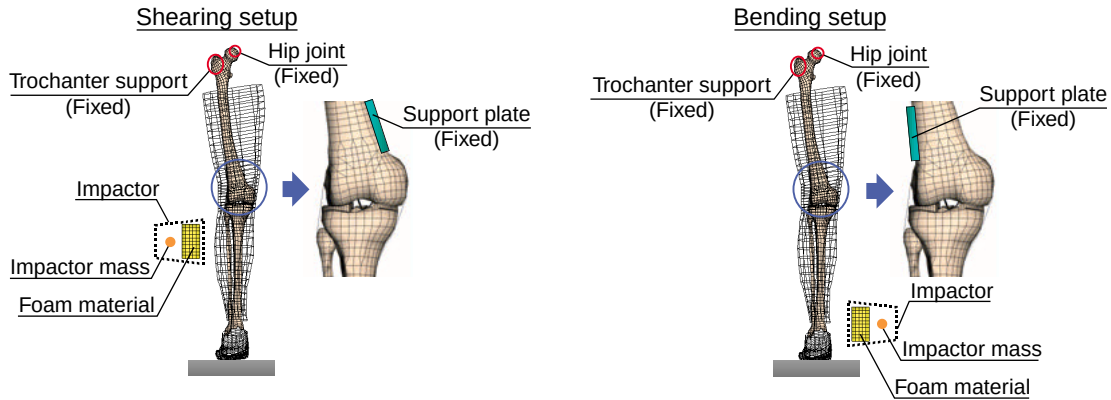


Fig. 13 Model setup for knee joint validation

correspond with that of the actual impactor. Since the foam material is the source for input to the lower limb, and therefore the accuracy of this part affects the accuracy of the knee response, validation of the foam material alone was performed in advance (Fig. 14).

Wittek et al.⁽²¹⁾ analyzed the test results from Kajzer et al. and obtained a corridor from the average $\pm$ standard deviation for the impact force to the lower limb as well as the shearing displacement and bending angle of the knee

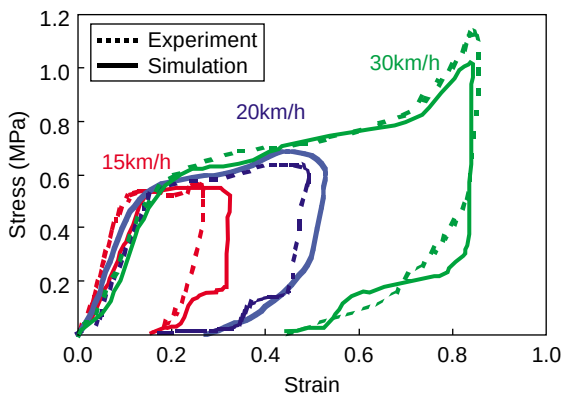


Fig. 14 Comparison of stress-strain curves of foam material between experiment and simulation (for 15, 20, 30 km/h of rate)

determined from high speed film analysis. This corridor was then used to compare the simulation results and test results. Fig. 15 shows the results from the shearing setup for an impact velocity of 20 km/h, and Fig. 16 shows the results from the bending setup for an impact velocity of 20 km/h. Fig. 17 and Fig. 18 show the shearing and bending setup results for an impact velocity of 40 km/h. With the shearing setup, the simulation results for the impact force were outside the corridor at both impact velocities. The simulation results for the shearing displacement fell almost smaller than the average response from the experiments. On the contrary, the simulation results for the bending angle were generally greater than the average response, and were even beyond the corridor in some parts. In tests with the shearing setup, in particular, many bone fractures due to stress concentration were observed in the vicinity of the jig fixation at the femoral supracondyle. As a result, the impact force is considered to have been lower than when there were no bone fractures. Target marks screwed in at the upper and lower part of the femur and tibia were traced in time series, and the straight line connecting the two points on each bone was used to determine the shearing displacement and bending angle. Therefore, when bone fractures of the kind noted above occurred, the shearing displacement was greater and the bending angle was smaller than when there were no bone fractures. As details of the jig fixation were not known, a simplified boundary

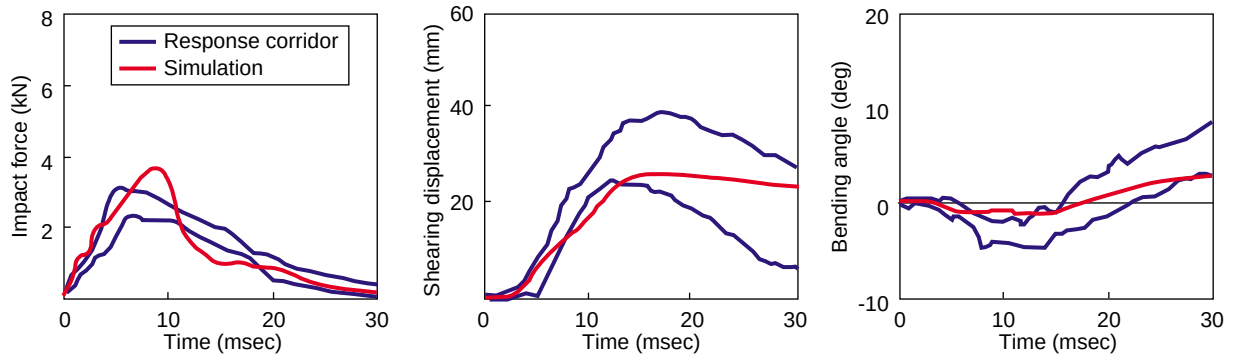


Fig. 15 Comparison between response corridor and simulation result in shearing setup at 20 km/h for impact force, shearing displacement and bending angle of the knee

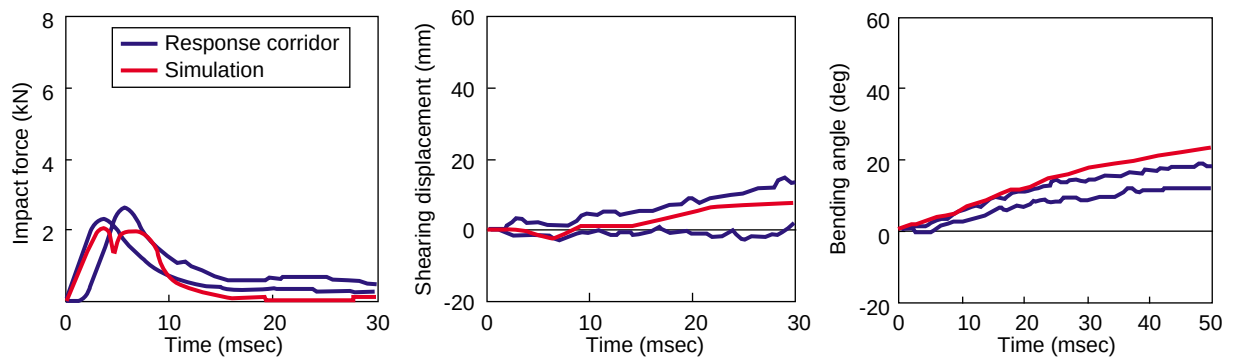


Fig. 16 Comparison between response corridor and simulation result in bending setup at 20 km/h for impact force, shearing displacement and bending angle of the knee

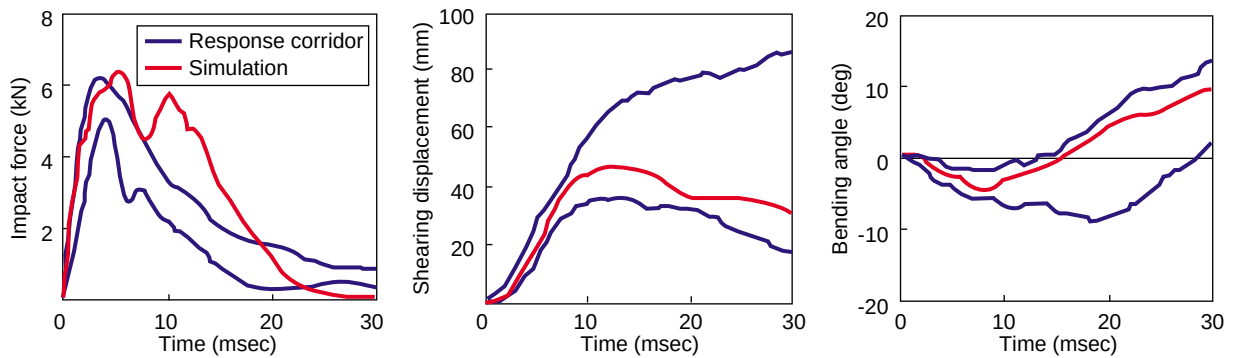


Fig. 17 Comparison between response corridor and simulation result in shearing setup at 40 km/h for impact force, shearing displacement and bending angle of the knee

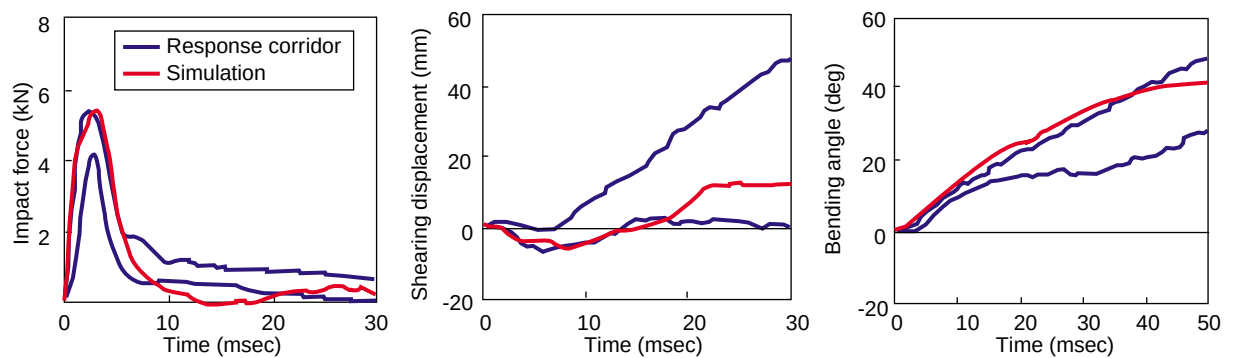


Fig. 18 Comparison between response corridor and simulation result in bending setup at 40 km/h for impact force, shearing displacement and bending angle of the knee

condition was used for the simulation, and therefore bone fractures of this kind were not reproduced. Although the simulation results for the impact force in the shearing setup, in particular, exceeded the corridor determined from the test results, when the above factors are taken into consideration, it can be concluded that the model can accurately predict the impact force and knee deformation.

The simulation results for knee deformation in the shearing and bending setups at an impact velocity of

40 km/h are shown in time series in Fig. 19 and Fig. 20, respectively. In the shearing setup, rupture of the ACL was observed approximately 6 msec after impact. In the bending setup, rupture of the MCL, ACL, and PCL takes place, in that order, during the first 20 msec after impact. The simulation results were compared with the test results for each ligament in terms of the description of damage under the various conditions (Table 3, Table 4). In order to compare the results between the experiment and simulation

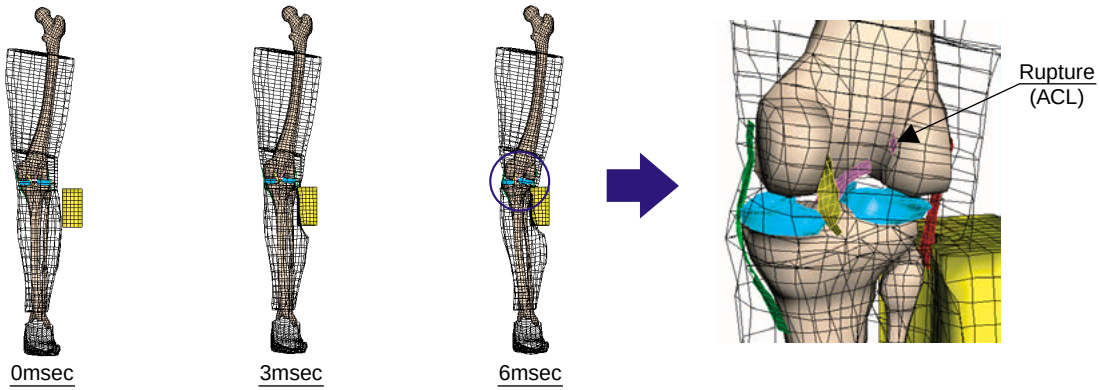


Fig. 19 Kinematics of the knee joint from FE simulation in shearing setup at 40 km/h

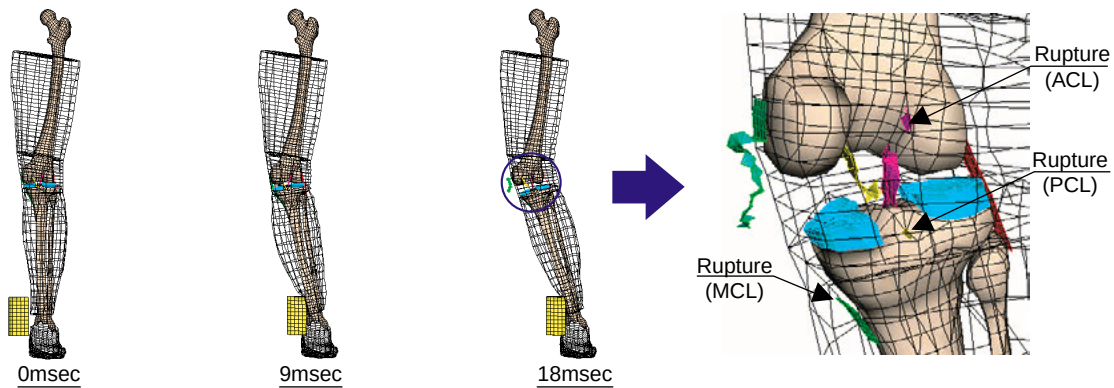


Fig. 20 Kinematics of the knee joint from FE simulation in bending setup at 40 km/h

Table 3 Injury description in shearing and bending setup at 20 km/h for both experiments and simulation

Shearing setup				
Test case	Knee ligament			
	ACL	PCL	LCL	MCL
Simulation	■			
21S	■			
24S				
25S				
29S	■			

* ■ : Ligament rupture, ■ : Experiment, ■ : Simulation

Bending setup				
Test case	Knee ligament			
	ACL	PCL	LCL	MCL
Simulation				
23B				
26B				
27B				■
30B				■

Table 4 Injury description in shearing and bending setup at 40 km/h for both experiments and simulation

Shearing setup				
Test case	Knee ligament			
	ACL	PCL	LCL	MCL
Simulation	■			
1S				
8S	■			
16S	■			

* ■ : Ligament rupture, ■ : Experiment, ■ : Simulation

Bending setup				
Test case	Knee ligament			
	ACL	PCL	LCL	MCL
Simulation	■	■		■
3B		■		■
6B	■	■		■
19B	■			■

under uniform conditions, only those test results in which there were no bone fractures due to the fixation of the femur were selected. The ligamentous damage observed in over half of the tests was also predicted by computer simulation, except for the bending setup with an impact velocity of 20 km/h. The ligaments in which no damage was observed in the tests were also undamaged in the prediction. These findings showed that this model was also highly accurate in predicting ligamentous damage of the knee.

These validation results confirmed the high accuracy of the model prediction for the deformation and ligamentous damage of the knee under impact loading. In car-pedestrian accidents, however, knee damage often include tibial plateau fractures in addition to ligamentous damage. Whereas ligamentous damage is determined only by ligament strain, bone fractures also depend upon force.⁽²²⁾ Consequently, prediction of tibial plateau fractures requires accurate prediction not only of deformation of the knee, but also of internal forces within the knee joint. Therefore, it will be necessary to validate the model against test results that reproduce tibial plateau fractures, after first validating the material properties of the ligament, cartilage, and meniscus at the high strain rate, for which extrapolation was used in the present study.

5. Conclusion

Bone models that had been validated for impact response were combined to develop a finite element model for the human knee joint in the standing position that includes the ligaments, articular cartilage, and meniscus. The model was validated through comparison with the results of lower limb impact tests, and the following conclusions were reached.

- (1) The lower limb model developed in this study was found to accurately predict test results for the impact force as well as the shearing displacement and bending angle of the knee at a loading rate of both 20 km/h and 40 km/h.
- (2) Comparison with test results in terms of the injury description for knee ligaments at a loading rate of both 20 km/h and 40 km/h showed that the simulation also generally predicted damage that occurred in one-half or more of the test cases.
- (3) Issues for the future are validation of material properties of the knee ligaments, articular cartilage, and meniscus at high strain rates, and validation of the model for tibial plateau fractures frequently observed in car-pedestrian accidents.

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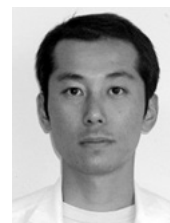
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