DEVELOPMENT OF FLEXIBLE PEDESTRIAN LEGFORM IMPACTOR FE MODEL AND COMPARATIVE STUDY WITH LEG BEHAVIOR OF HUMAN FE MODEL THUMS

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ABSTRACT

The current legform impactor in pedestrian safety tests uses a steel shaft connected to metal plates to represent the femur and tibia. It evaluates leg fracture risk based on tibia acceleration, and knee ligament rupture risk based on knee bending angle and shear displacement. However, the impactor does not generate the tibia deflection that occurs when a vehicle impacts a pedestrian. The new flexible pedestrian legform impactor (Flex-PLI) currently under development is designed to simulate the impact behavior of the human leg, reproducing tibia deflection with flexible shafts and representing the knee ligaments using wires. As a result, it can be used to help assess injury based on deformation by estimating the risk of tibia fracture from the bending moment of the tibia shaft and the risk of knee ligament rupture from the elongation of the wires.

In this study, a finite element (FE) model of the Flex-PLI was developed to examine the impact test protocol for pedestrian leg injury assessment, comparing the impactor behavior and response with that of a whole human FE model. The Flex-PLI FE model was created by reverse engineering that reproduced the shape and mechanical properties of each part. The impact velocity of the impactor was set to 40 km/h based on accident data. An impact height of 75 mm above the ground has been proposed for the Flex-PLI in contrast to the current protocol, which specifies an impact height of 0 mm. The study compared results at the base impact height of 75 mm with those obtained at different heights. It also investigated the effect of adding mass to simulate the upper body of a pedestrian. Vehicle-to-pedestrian impact simulations were conducted with the Total Human Model for Safety (THUMS) to estimate the behavior and response of a human leg for comparison with the results from the impactor model. The bending moment of the tibia and the elongation of the knee ligament wires in an impact varied depending on the impact height and additional mass. Impactor behavior was closest to THUMS at a height of 0 mm, but a closer response to THUMS for bending moment and ligament elongation was obtained at 75 mm. It was also found that adding a mass of 6 kg to the upper end of the impactor in SUV impacts created a closer response to THUMS.

INTRODUCTION

In 2007, 5,744 fatalities occurred as a result of traffic accidents in Japan, roughly 30% of which were pedestrians. Pedestrians also accounted for 17% of serious injuries. 58% of the pedestrian fatalities sustained head injuries, while the lower extremities were the most frequently injured (37%) in all cases of injury.[1]

In 2003, the Japan New Car Assessment Program (JNCAP) began to assess pedestrian safety performance. Currently, only a head safety test is conducted, but another test protocol for lower leg injury assessment will be introduced in 2010. A proposal has also been drawn up to integrate a leg test into the Global Technical Regulations (GTR) that are observed as a set of international standards for vehicle safety in various countries around the world.[2] This proposal includes a new flexible pedestrian legform impactor (Flex-PLI) that is currently under development.[3] Whereas the current leg impactor uses rigid steel parts to simulate the femur and tibia, the new impactor expresses these portions with bendable flexible materials. It also uses wires to represent the ligaments in the knee joints. Development of the Flex-PLI began in 2000,[4] and a proposal for its final specifications was announced in 2008.

Studies into the conditions for the leg test have continued during the development of the Flex-PLI. It is currently proposed to collide the Flex-PLI with a vehicle at a height of 75 mm from the ground. It has been reported that the results for leg bone deflection and knee ligament elongation obtained from the impactor at a height of 75 mm are close to the response of a pedestrian's leg.[5] However, since this setting results in the initial knee joint being positioned higher than the knee of an actual pedestrian, it must be verified whether the effect of the shape of the front edge of the vehicle can be adequately evaluated.

It has also been pointed out that the mass of the pedestrian's upper body has an effect on leg behavior.[6] Behavior is also thought to be affected by the fact that legs are inclined inward from the vertical while walking.

The study focused on the following three factors using an FE model of the Flex-PLI. Comparing the impact behavior and mechanical response of this model with a human FE model, this paper discusses the optimal test conditions for predicting and assessing full-body pedestrian behavior and injury criteria.

- Impact height
- Additional mass for simulating the upper body
- Impactor inclination angle

The Flex-PLI FE model was created by reverse engineering from the actual Flex-PLI. The Total Human Model for Safety (THUMS) FE model was used for the comparison.

METHODS

Human FE Model

Outline of THUMS - THUMS is a human FE model jointly developed by Toyota Central R & D Labs., Inc. and Toyota Motor Corporation. Figure 1 shows a standing THUMS model simulating a pedestrian crossing a road. THUMS has a height of 175 cm and a mass of 77 kg to simulate a 50th percentile American male (AM50). In its walking pose, the left leg is inclined 22 degrees forward of the body and the right leg is inclined 7.2 degrees to the rear, based on the standard acetabulum angles. Both arms are hanging downward and both hands are positioned slightly in front of the torso. Parts for simulating shoes have also been added to the soles of the feet. As a result, the inferior surface of the calcaneus is positioned at a height of 29 mm from the ground. THUMS includes the major skeletal and soft tissue that form the interior of the body. The skeleton is divided into cortical and cancellous bones, which are modeled using shell and solid elements, respectively. The cortical bones are modeled with elasto-plastic properties, and bone fractures are simulated by eliminating elements where strain exceeds a threshold. The physical properties of the bones were defined in reference to the values described by Yamada et al.[7] The threshold value for bone fracture strain was assumed to be 3%, based on the study by Burstein et al.[8] The joints are modeled with bone-to-bone contacts and ligament connections, without using artificial joint elements provided in FE codes. In the same way as bone fracture, ligament rupture is simulated by eliminating elements where strain exceeds a threshold. The physical properties of ligament tissue were defined in reference to the values described by Abe et al. (1996). Kerrigan et al. reported a range of approximately 11 to 20% as the critical stretch for ligament rupture.[9] The study assumed an elongation of 15% as the threshold. Subcutaneous fat, muscle, organs, and other soft tissue were modeled with solid elements. However, neck and leg muscles are modeled with bar elements to reproduce only their resistance force when forcibly extended. THUMS contains approximately 80,000 elements and has approximately 60,000 nodes.

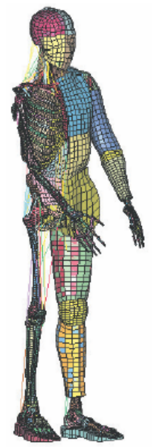


Figure 1. Pedestrian THUMS (AM50).

Validation of Leg Model - The validity of the THUMS leg was examined by comparing its mechanical response with test data reported in literature using post mortem human subjects (PMHS). Figure 2 shows a comparison between force-deflection curves calculated using the THUMS leg model and test data obtained by Yamada et al.[7] for mechanical response to static 3-point bending of the femur, tibia, and fibula bones. In the calculations, the tests were simulated by

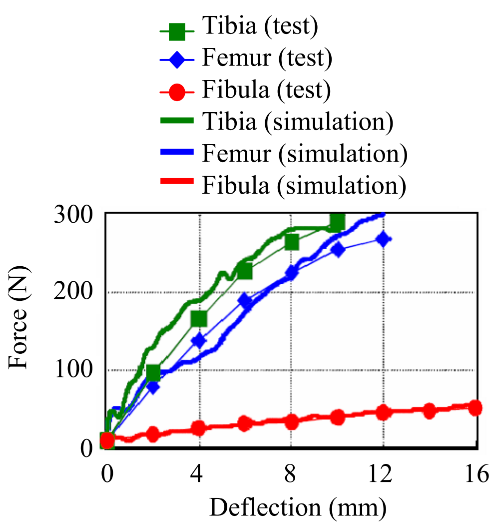


Figure 2. Comparison of force-deflection curves for THUMS response and test results.

removing each bone model from THUMS. The bones were then supported at both ends before being contacted in the center by an impactor. In the figure, the solid lines show the test results and the lines marked with symbols show the calculation results. The force curves obtained from the models correspond closely with the force curves obtained from the tests.

Figure 3 shows the relationship between knee bending angle and knee bending moment when a knee ligament ruptures. The data is obtained from 3-point bending tests performed by Kajzer et al.,[10],[11] Levine et al.,[12] and Ramet et al.[13] on PHMS knees. The results are mainly distributed within the dotted line circle. Previous studies have reported that the medial collateral ligament (MCL) is likely to be ruptured in a vehicle-to-pedestrian impact. The MCL of THUMS is modeled to rupture when the bending moment around the knee joint exceeds approximately 200 Nm. This condition is close to the center of the data distribution in this figure.

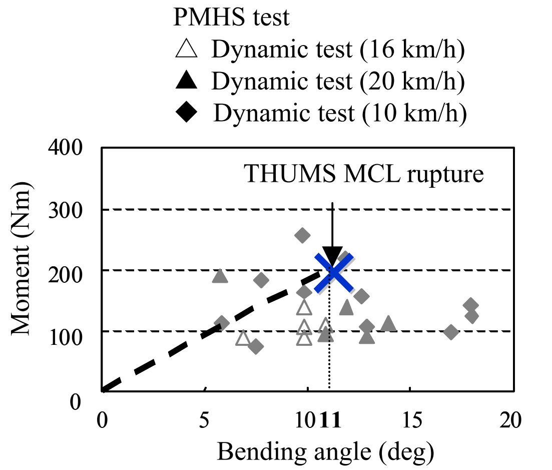


Figure 3. Comparison of knee bending moments.

The comparative verification described above shows that the mechanical response of the femur, tibia, fibula, and knee ligaments (MCL) in THUMS corresponds closely with that of the human body (PHMS).

Development of Flex-PLI FE Model

Flex-PLI - Figure 4 shows the exterior of the Flex-PLI and Table 1 lists its dimensions. The Flex-PLI was created using the values of a 50th percentile American male. It has a body portion consisting of a femur, knee joint, and tibia, and an exterior flesh portion.

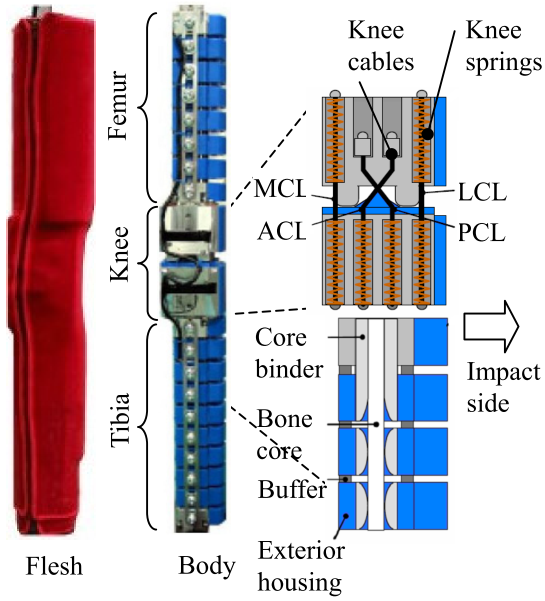


Figure 4. Flex-PLI.

Table 1.
Dimensions of AM50 and Flex-PLI

	AM50[14]	Flex-PLI
Femur length (mm)	428	434
Tibia length (mm)	493	494
CG of thigh* (mm)	218	202
CG of tibia* (mm)	233	216
Total mass (kg)	13.4	13.5
Femur mass (kg)	8.6	7.4
Tibia mass (kg)	4.8	6.1

* from the knee joint center

The femur and tibia are each constructed from multiple divided block portions, through the center of which a glass fiber reinforced plastic (GFRP) core runs from top to bottom. The tibia and femur cores are joined to the lower and upper knee joint blocks, respectively. Leg bone deflection is reproduced using the flexure of the core. To prevent breakage in excessive bending deformation, four stopper cables are inserted in the block parts.

The knee joint is constructed from the tibia- and femur-side blocks and wires connecting the blocks. The wires are connected to springs inside the knee blocks that are used to simulate the elongation of ligaments. As shown in Figure 4, the knee is provided with crossed wires to simulate the cruciate ligaments. The wires are designed to extend when the knee joint bends.

Rubber sheeting and neoprene are wrapped around the exterior of the impactor from the femur to the tibia. As the rubber sheeting is designed to simulate the shape of the femur and calf, the role of the neoprene is to hold the parts in place and alleviate the impact. The upper end of the impactor is also provided with a suspension jig that is used for performing the tests.

The Flex-PLI measures bending moment to assess the risk of leg bone fracture and injury. As shown in Figure 5, bending moment is evaluated at three locations along the femur, and four locations along the tibia. Bending moment is calculated from the output values of strain gages attached to the bone core.

The risk of knee ligament rupture is assessed based on the elongation amount of the wires representing each ligament. There are a total of four wires, representing the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), MCL, and lateral collateral ligament (LCL). The elongation of each knee ligament is output from the potentiometers provided in the Flex-PLI.

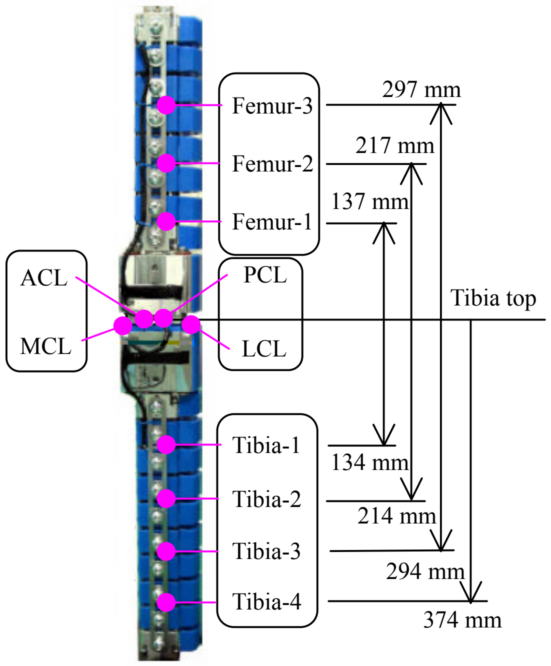


Figure 5. Measurement points.

Development of FE Model by Reverse Engineering - The FE model development process consisted of three stages: measurement of the actual geometry of the Flex-PLI, 3D reconstruction of the geometry data, and the generation of an FE mesh. Non-contact X-ray computerized tomography (CT) was used to measure the surface and internal shapes of the Flex-PLI. Using X-ray CT scans of an assembled Flex-PLI enabled the actual shapes of the component parts to be obtained, as well as positional information of these parts in an assembled state. First, an actual Flex-PLI was scanned at a pitch of 0.5 mm to express the whole of the impactor including its internal structure as point group data. The point group data was filtered while adjusting the CT values to specify and read the steel, aluminum, and non-metallic parts. This data was converted to stereolithography (STL) format 3D polygon data, which was then used as the basis to prepare the surface data for FE mesh creation. In generating the FE mesh, the element size was controlled to 2 to 3 mm so that the bone core was divided into five sections in the thickness direction.

Figure 6 shows the created FE model and Table 2 lists the number of nodes and elements in the model. Parts with a thickness of 1 mm or more were modeled using solid elements and parts with a thickness of less that 1 mm were modeled using shell elements. The springs, stopper cables, and ligaments in the knee blocks were modeled from beam elements and the rotatable pin joints were modeled from joint elements.

The mechanical properties of the materials of the bone core, flesh rubber, neoprene, cushion rubber, wire cables, internal knee block springs, and other parts that are thought to have a major impact on impactor response were measured, and the obtained values were input into the FE model. These input values were validated by subjecting the FE model to the same analysis as performed in materials tests. The other parts were treated as rigid bodies. The mass of the individual parts was set to the same values as the actual Flex-PLI.

It should be noted that the structure and shape of the created Flex-PLI are identical to the Flex-GT impactor announced in 2007.[15]

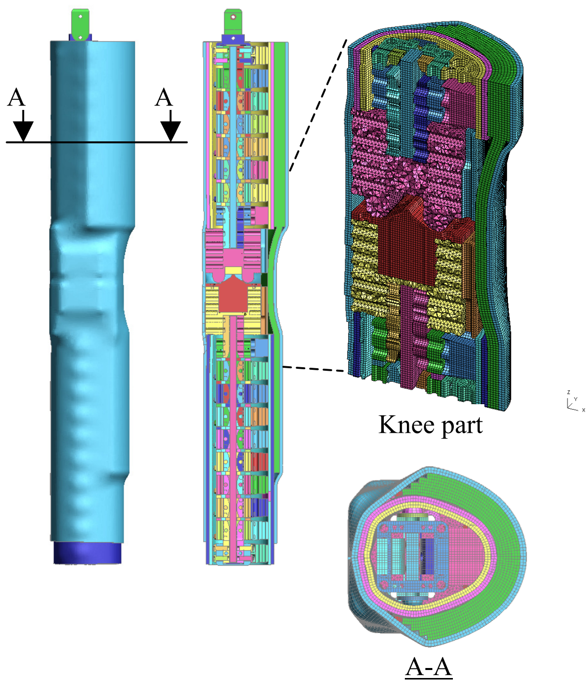


Figure 6. Flex-PLI FE model.

Table 2.
Description of Flex-PLI FE model

		Nodes	Elements
Whole model		1,330,000	1,900,000
Part	Femur	220,000	290,000
	Knee	80,000	270,000
	Tibia	260,000	360,000
	Flesh	770,000	980,000

Injury Criteria Assessment Using Flex-PLI FE Model - Strain was measured in the FE model by elements representing the strain gages in the actual Flex-PLI. Knee ligament elongation was calculated from the elongation of the beam elements simulating the wires.

The injury criteria judgment conditions were set to 299 Nm for the tibia bending moment threshold and 18 mm for the MCL elongation threshold.[16]

Model Validation

The Flex-PLI was tested using quasi-static 3-point bending tests on each of the femur, tibia, and knee joint sub-assemblies, and dynamic pendulum tests on the body assembly.[15]

First, the FE model was validated under the conditions of the quasi-static 3-point bending tests for each sub-assembly.

The tibia test was performed by fixing both ends of the tibia sub-assembly on cylindrical supports with a radius of 75 mm. The distance between the supports was 410 mm. A round block was attached on the impact side of the force application point to the outside of the bone form, and a neoprene pad with a thickness of 5 mm was placed between the block and the load cell. A force of 1.33 mm/s was applied from the impact side to the center of the supports using a 40 mm diameter cylindrical impactor.

In the FE model, the supports at both ends were simulated using rigid bodies and connected to the ends of the tibia. The force was applied at 133 mm/s. (100 times of experiment)

Figure 7 compares the actual Flex-PLI in its bent form (deflection: 26 mm) with the FE model. It also shows the graph that compares the bending moment-deflection curves of the FE model and actual Flex-PLI. The bending moment generated at the force application point (Equation 1) is plotted on the vertical axis and the deflection is plotted on the horizontal axis. The moment was calculated using Equation 1 as follows.

$$M = \frac{F}{2} \times \frac{D}{2} \qquad (1).$$

where,
M: 3-point bending moment (Nm)
F: Force (N)
D: Deflection (m)

The curve for the Flex-PLI is shown as a corridor calculated from multiple experiment results. The bending moment curve for the FE model fits inside the Flex-PLI corridor. The same comparison was performed for the femur and it was verified that the moment response of the FE model fitted inside the corridor calculated for the Flex-PLI.

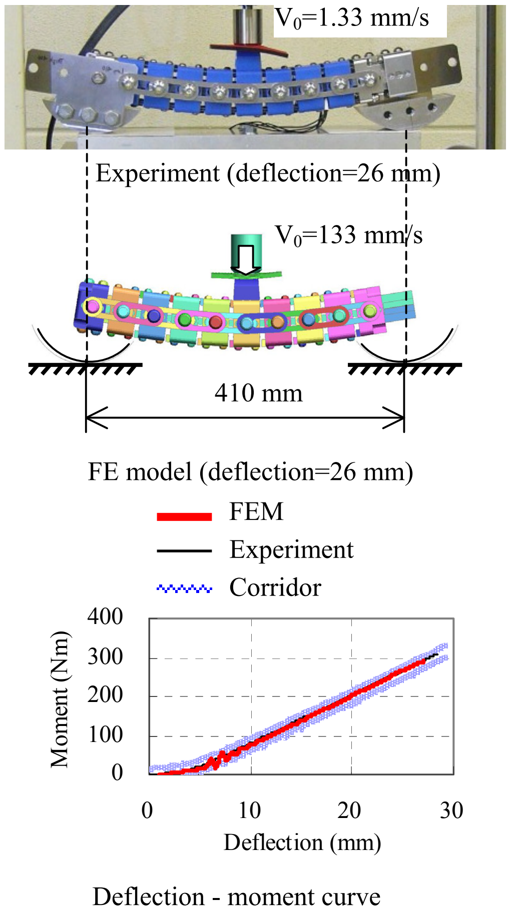


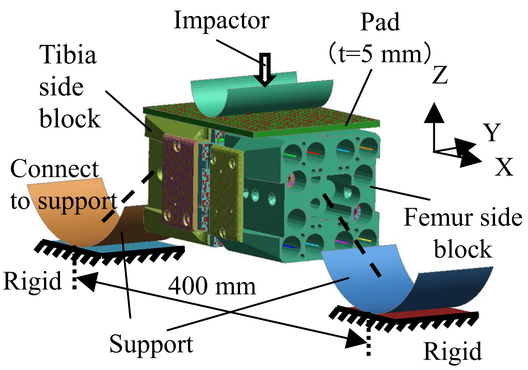
Figure 7. Quasi-static 3-point bending test of tibia sub-assembly.

Figure 8 shows the knee sub-assembly of the Flex-PLI FE model simulating the knee bending test and compares the MCL force-elongation curves obtained by the FE model and in the actual tests. In the testing, both ends of the knee joint were fixed on cylindrical supports with a radius of 75 mm to create a distance between the supports of 400 mm. A neoprene pad with a thickness of 5 mm was placed at the force application point and a force of

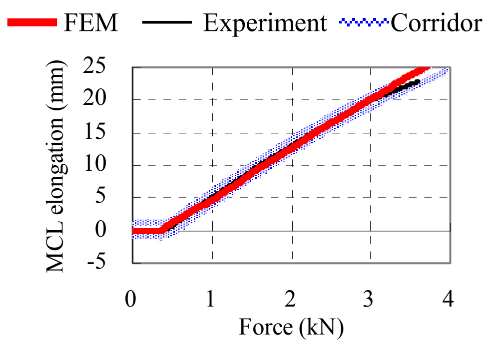
1.0 mm/s was applied from the impact side using a cylindrical impactor with a radius of 50 mm. A response corridor was generated from multiple test data.
In the FE model, the supports at both ends were simulated using rigid bodies and connected to the knee blocks. The force was applied at 500 mm/s. (500 times of experiment)
It was confirmed that the results of the FE model closely reproduced the MCL elongation characteristics and fell within the test corridor. The same comparison was performed for the elongation of other ligament wires and the bending moment of the knee joint. The elongation values of the FE model were also confirmed to be within the response corridor of the Flex-PLI.

The FE model was then validated with respect to the dynamic pendulum test applied to the body assembly. The test was conducted by suspending it by the jig at the upper end. Then, using this point as a reference, the Flex-PLI was raised 15 degrees from the horizontal and released, causing the femur-side knee block to impact the fixed stopper on the test device. A rubber and neoprene pad (width: 100 mm, height, 100 mm, thickness: 25 mm) was attached to the surface of the stopper. The bending moment at each part of the Flex-PLI and elongation of each ligament on impact were measured.
In the FE model, the support conditions of the Flex-PLI were reproduced and calculation started after applying an angular velocity from a position immediately prior to impact.

Figure 9 shows the deformed shape of the Flex-PLI at 22 ms, when the tibia bending moment was greatest. It also shows the deformed shape of the FE model at the same time, and as an example of the results, the time history of the bending moment at the tibia-1 measurement position and MCL

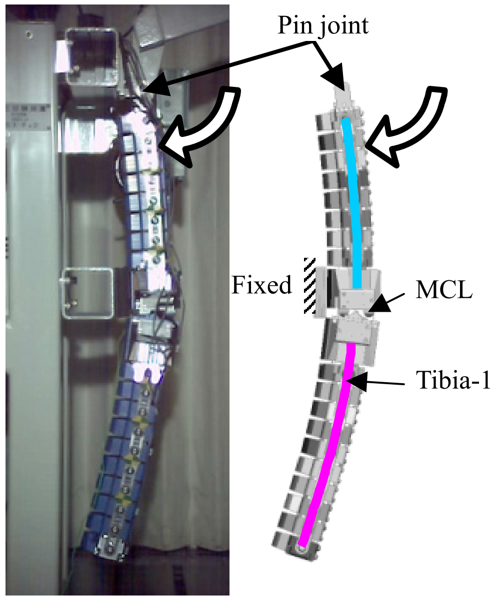


FE Model

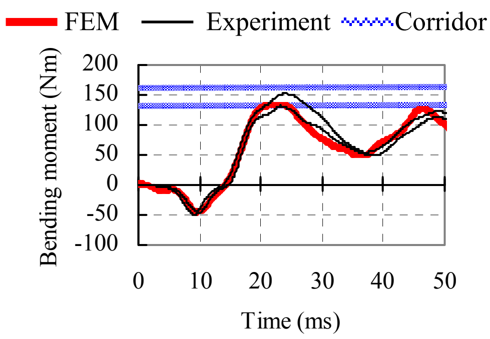


Force-elongation curve

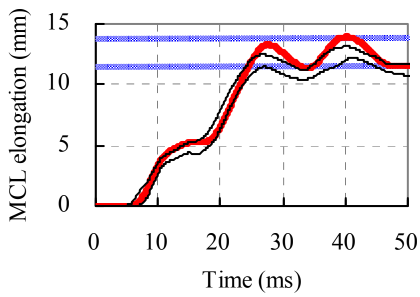
Figure 8. Quasi-static 3-point bending test of knee sub-assembly.



Experiment (22 ms) FE model (22 ms)



Bending moment (Tibia-1)



MCL elongation

Figure 9. Flex-PLI assembly dynamic pendulum test.

elongation. The way that the Flex-PLI bodies bend, the movement of the lower portion, and the waveforms for bending moment and MCL elongation indicate a close correlation between the FE model and the test results. In the test, response was deemed acceptable if the peak moment fell within the test corridor. As the peak moment of the FE model was calculated to be within the corridor, the test conditions were judged to be satisfied. The bending moment at measurement positions other than tibia-1 and the elongation for knee ligaments other than MCL also showed a close correlation with the waveforms obtained in tests. The values of the FE model were confirmed to be within the test corridors.

Vehicle Models

Two models representing the front end of a sedan and a SUV, as adopted by Yasuki et al. were used in this study.[17],[18]

The bumper cover of the sedan model is made from polypropylene (PP) and the internal structure is provided with two absorbers, one in the upper portion and the other in the lower. The upper absorber is fixed in front of the bumper reinforcement and has properties corresponding to polyurethane with an expansion ratio of 40 at thickness of 65 mm. The lower absorber is connected rigidly with the body at its rear end and has properties corresponding to polyurethane with an expansion ratio of 10 at a thickness of 150 mm. The hood is aluminum and an acrylonitrile butadiene styrene (ABS) grille is provided between the hood and the bumper cover. The vehicle mass is 1.7 tons and the model includes approximately 145,000 elements and approximately 150,000 nodes.

The bumper cover of the SUV model is PP and one internal bumper absorber is provided. This absorber is fixed in front of the bumper reinforcement and has properties corresponding to polyurethane with an expansion ratio of 40 at a maximum thickness of 65 mm. The hood is steel and the grille is made from ABS. The vehicle mass is 2.9 tons and the model includes approximately 320,000 elements and approximately 330,000 nodes. The SUV model includes drive train components such as the suspension, tires, and engine, but these parts are regarded as having only a small impact on legform impactor conditions.

Impact Simulations

Figure 10 shows the study model. The center sections of the vehicle models are displayed and the positions of the upper and lower absorbers are highlighted. The impact simulation with THUMS assumed a pedestrian-to-vehicle impact at 40 km/h where the vehicle model collided against a stationary THUMS in walking pose from the left side. Friction between the soles of the shoes and the ground was ignored. In the Flex-PLI simulation, a stationary vehicle was impacted at 40 km/h to simulate the actual assessment test. The impact point in both simulations was at the center of the vehicle front in the lateral direction. Gravitational acceleration was applied to the entire model in the vertically downward direction. Calculation was performed over 40 ms. For impactor measurement, the knee and tibia bending angles were added to the injury criteria described above. To compare the injury criteria for the Flex-PLI and THUMS, the following items were measured in THUMS: the bending moment of the tibia and femur bones at the same heights as the bending moment measurement positions of the Flex-PLI, the elongation of each knee ligament, and the knee bending angle. The impact simulations used the general-purpose finite element code LS-DYNA TM V971.

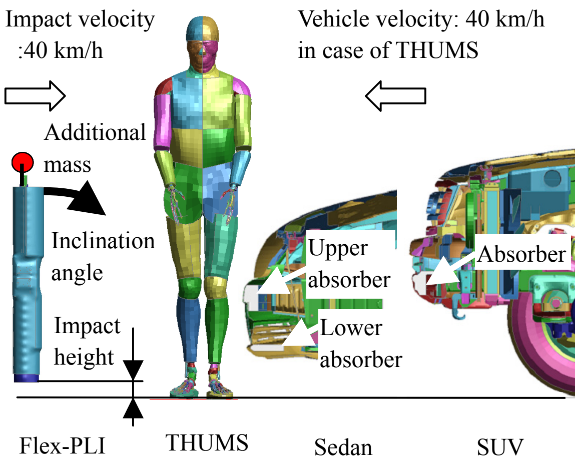


Figure 10. Study model.

Evaluation Matrix

Table 3 shows the evaluation matrix. The impact height of the Flex-PLI was adjusted to the following three levels: 0 mm, 25 mm, and 75 mm above the ground. At a height of 0 mm, the knee joint position is close to that of THUMS. The 25 mm condition simulates the thickness of shoe soles, and 75 mm is the value proposed by Matsui et al.[19] The height of the knee joint in THUMS while wearing shoes is 498 mm, whilst that of the Flex-PLI is 494 mm. Therefore, an impact height of 0 mm is the closest condition to the height of the knee.

Additional mass was set to four levels: 0 kg (no additional mass), 6 kg, 10 kg, and 14 kg. The 6 kg case was added because, although the mass of the pelvis portion of THUMS is approximately 10 kg, and approximately 14 kg in combination with the abdomen, it is possible that the effective mass that is applied to each leg individually may be less. The additional mass was positioned at the top end of the Flex-PLI and set as a mass point. The inclination angle of the Flex-PLI was set to 0 degrees from the vertical as a reference and a case with an inclination angle of 6 degrees from the external line of the THUMS leg was also performed.

Table 3.
Evaluation matrix

Case No.	Subject		Impact height (mm)			Additional mass (kg)				Inclination angle (deg)	
Sedan	THUMS	Flex-PLI	0	25	75	0	6	10	14	0	6
1	O		O								
2		O	O			O				O	
3		O	O				O			O	
4		O	O					O		O	
5		O	O						O	O	
6		O		O		O				O	
7		O			O	O				O	
8		O			O		O			O	
9		O	O						O		O
SUV	THUMS	Flex-PLI	0	25	75	0	6	10	14	0	6
10	O		O								
11		O	O			O				O	
12		O	O				O			O	
13		O	O					O		O	
14		O	O						O	O	
15		O		O		O				O	
16		O			O	O				O	
17		O			O		O			O	
18		O	O						O		O

RESULTS

This section first compares the impact behavior of THUMS and the Flex-PLI in a typical case. Figure 11 shows the behavior of the THUMS and impactor skeletons at every 10 ms in the collision with the sedan. Case 1 is shown for THUMS and case 7 for the Flex-PLI (impact height: 75 mm).

The vehicle first contacted the THUMS leg at the knee, followed by the tibia, and femur, causing bending deformation. The tibia was in contact with the bumper cover from 10 to 20 ms. After 30 ms, the lower leg rebounded. The femur contacted the hood after 30 ms. Bending deformation of the knee joint continued to increase until 40 ms. The MCL ruptured at 28 ms, but rupture did not occur in any of the other ligaments. The tibia and fibula bones did not fracture.

The lower portion of the Flex-PLI began to rebound at 20 ms after bending deformation of the tibia

a) THUMS (Case 1)

b) Flex-PLI (Case 7)

0 ms 10 ms 20 ms 30 ms

Figure 11. Behavior of a) THUMS (Case 1) and b) Flex-PLI (Case 7).

occurred. After bending briefly, bending of the knee joint decreased because the tibia rebounded upward.

Figure 12 shows the results of impacts with the SUV. Case 10 is shown for THUMS and case 16 for the Flex-PLI. The vehicle contacted THUMS close to the knee joint and comparatively less bending deformation of the femur, tibia, and fibula bones occurred than with the sedan impact. In contrast, the knee joint was bent toward the rear of the vehicle after 20 ms, and the MCL and ACL ruptured at 19 ms and 20 ms, respectively. The tibia, fibula, and femur bones did not fracture.

In the case of the Flex-PLI, the tibia was bent toward the rear of the vehicle. The knee joint flexed after 20 ms, but bending decreased after 30 ms due to the rebounding femur. Femur rebound occurred after it contacted the grille.

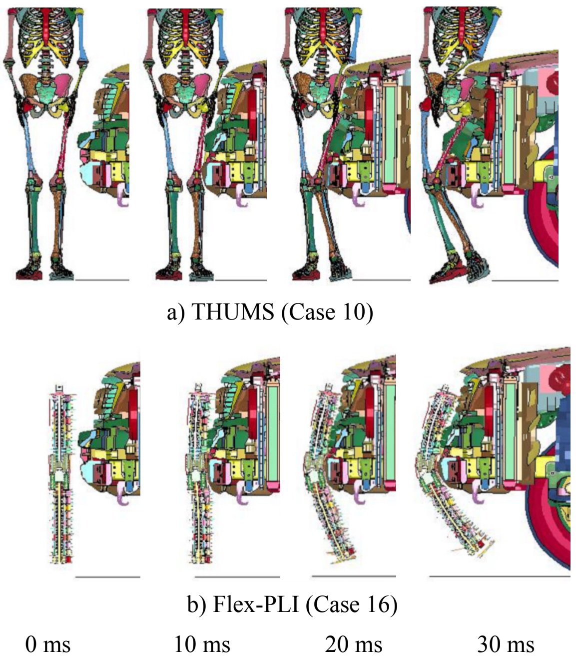


Figure 12. Behavior of a) THUMS (Case 10) and b) Flex-PLI (Case 16).

The next section discusses the comparative results for THUMS with respect to the effect of the Flex-PLI test conditions on impact behavior and response.

Effect of Impact height

Figure 13 compares the behavior of THUMS and the Flex-PLI at impact heights of 0 mm (case 2) and 75 mm (case 7) after collision with the sedan. The behavior shown in the graph occurred at 15 ms, the point at which the maximum bending moment of the tibia was generated. The distance to the knee center is plotted on the vertical axis and the X-direction displacement at each point of the THUMS tibia and impactor bone core is plotted on the horizontal axis. The origin of the graph is the center of the knee joint. The black line indicates the behavior of THUMS, the red line that of the Flex-PLI at an impact height of 0 mm, and the green line that of the Flex-PLI at an impact height of 75 mm.

Regardless of the impact height, the Flex-PLI generated a bending deformation mode corresponding to that of the THUMS leg. However, there was a discrepancy in the amount of tibia deformation. The X-direction displacement of the inferior end was approximately 50 mm from the knee center toward the impact side in THUMS. In comparison, the lower end of the Flex-PLI displaced toward the opposite side of the impact at 75 mm, but to the impact side at 0 mm. Thus, the Flex-PLI behavior at 15 ms was closer to THUMS when the impact height was adjusted to 0 mm.

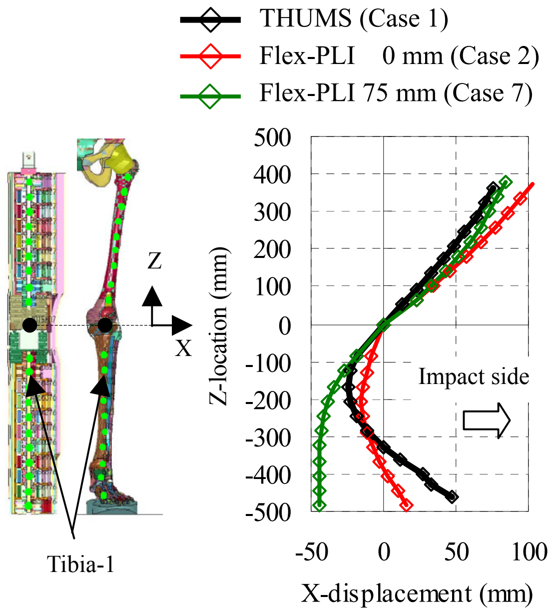


Figure 13. Effect of impact height: behavior of THUMS and Plex-PLI (sedan, 15 ms).

Figure 14 compares the time history curves of the tibia bending moment (measurement position: tibia-1) between THUMS and the Flex-PLI. The graph also shows the moment criterion of 299 Nm. In either case, moment began to increase from approximately 3 ms, reaching a maximum peak at around 14 ms, before falling to around zero at approximately 25 ms. Regardless of the impact height, the moment response waveforms of the Flex-PLI corresponded well with THUMS. However, the maximum moment peak for the Flex-PLI was 230 Nm at an impact height of 0 mm and 270 Nm at an impact height of 75 mm. Both values are higher than the maximum moment peak of approximately 200 Nm in THUMS.

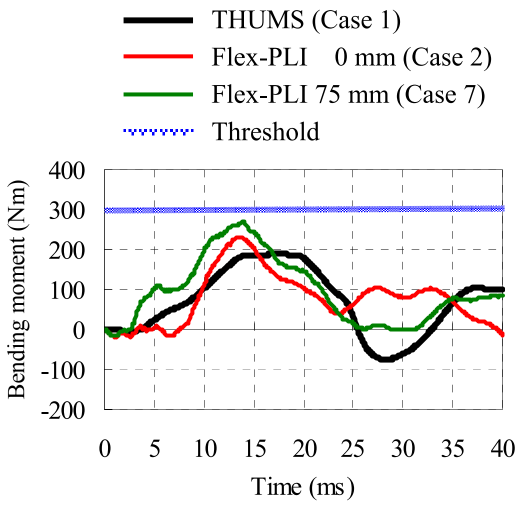


Figure 14. Effect of impact height: tibia bending moment (sedan).

Figure 15 shows the time history curves of the knee bending angle for the same cases. The knee bending angle of THUMS continued to increase, which is attributable to the MCL rupture at 10 degrees, after which the rate of increase continued to rise. Because the knee ligament wires in the Flex-PLI do not rupture, ligament elongation decreased after reaching a peak. At an impact height of 75 mm, although the start of bending was later than in THUMS, the peak value and timing was close to the point of MCL rupture in THUMS. In contrast, at an impact height of 0 mm, the knee started to bend at a timing similar to THUMS, but then increased rapidly after 12 ms. It reached a peak of 17 degrees, which is larger than the bending angle when MCL rupture occurred in THUMS. At this height, the knee bending angle decreased after 28 ms due to the rebound of the lower tibia. Therefore, the ligament rupture risk assessment is closer to THUMS when the impact height is set to 75 mm.

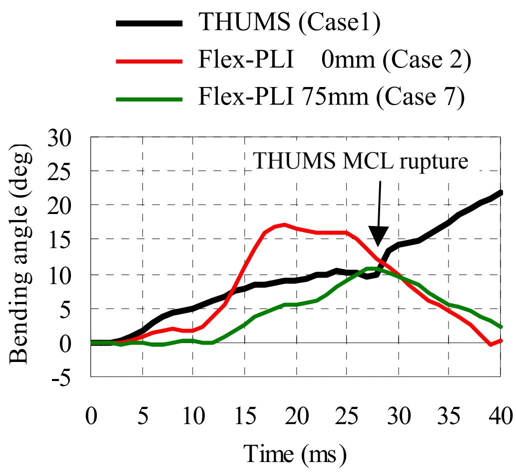


Figure 15. Effect of impact height: knee bending angle (sedan).

Effect of Additional Mass

This section compares THUMS with the results when 6 kg was added to the upper end of the Flex-PLI in the cases of SUV impact. Figure 16 compares the behavior of THUMS and the Flex-PLI at 20 ms. The condition of THUMS is equivalent to the state immediately after rupture of the MCL. The figure shows results with an impact height of 0 mm and additional mass of 0 kg (case 11) and 6 kg (case 12). The black line indicates the behavior of THUMS, the red line that of the Flex-PLI with an additional mass of 0 kg, and the brown line that of the Flex-PLI with an additional mass of 6 kg. The behavior of the femur with mass added to the Flex-PLI was closer to the behavior of THUMS. In contrast, no significant difference was observed in tibia behavior.

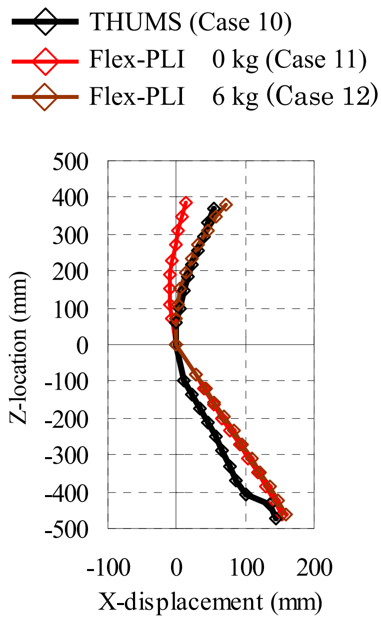


Figure 16. Effect of additional mass: behavior of THUMS and Flex-PLI (SUV).

Figure 17 shows the time history curves of the tibia bending moment (measurement position: tibia-1) between THUMS and the Flex-PLI. In THUMS, an initial negative moment was generated, which became positive moment at 15 ms after the impact. The MCL ruptured at 19 ms, after which, bending became concentrated in the knee joint due the rupture of the ACL at 20 ms, and the tibia bending moment did not increase. For the cases with the Flex-PLI, negative bending moment appeared at the beginning, but the bending moment continued to increase after 20 ms in both conditions.

Since the ligament wires in the Flex-PLI do not rupture, knee bending is restricted within a certain range. As a result, the bending moment continued to act on the knee joint side of the tibia bone. In other words, the moment generated in the Flex-PLI after 20 ms cannot be used for comparison.

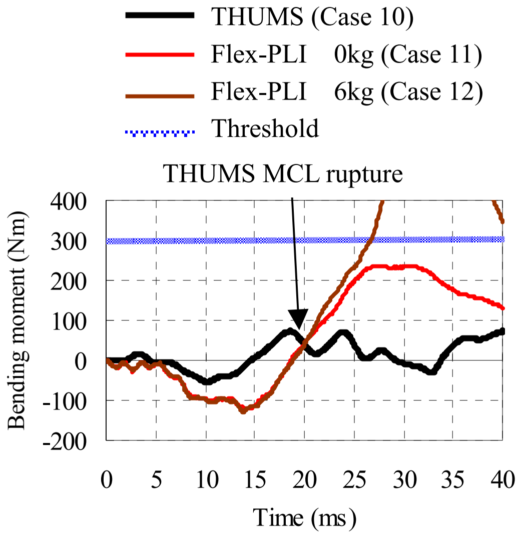


Figure 17. Effect of additional mass: tibia bending moment (SUV).

Figure 18 shows the time history curves of the knee bending angle for the same cases. In THUMS, the knee bending angle increased rapidly after the rupture of the MCL and ACL at 20 ms, and then continued to increase. In contrast, the bending angle of the Flex-PLI started at 10 ms, which was earlier than THUMS. With an additional mass of 0 kg, the maximum bending angle of 25 degrees was reached at 32 ms, and with an additional mass of 6 kg, the maximum bending angle increased to 38 degrees.
According to Bose et al., the knee ligaments rupture at a bending angle of approximately 15 degrees.[20] The criterion for knee bending angle in Euro NCAP is also 15 degrees. In both conditions, the knee bending angle of the Flex-PLI exceeded 15 degrees, indicating the possibility of ligament rupture. It should be noted that in THUMS, the MCL ruptured

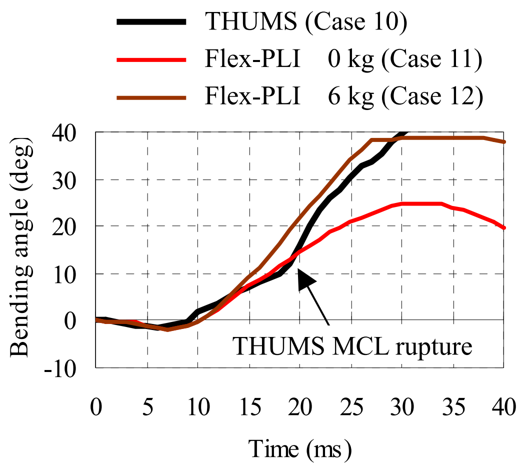


Figure 18. Effect of additional mass: knee bending angle (SUV).

at a knee bending angle of 10 degrees. Under this condition, local elongation of the MCL occurred, causing the rupture at a small angle.

These results suggest that additional mass has little effect in terms of injury assessment. However, in the case of SUV impact, impactor behavior is closer to THUMS when an additional mass is applied. A mass of 6 kg is considered to be sufficient to create such a correlation.

Effect of Impactor Inclination Angle

Figure 19 compares the behavior of THUMS and the Flex-PLI at impactor inclination angles of 0 degrees (case 5) and 6 degrees (case 9) 15 ms after impact. The impact vehicle was the sedan and an additional mass of 14 kg was applied to the upper end of the Flex-PLI. The black line indicates the behavior of THUMS, the blue line that of the Flex-PLI at an inclination angle of 0 degrees, and the green line that of the Flex-PLI at an inclination angle of 6 degrees.
It shows that femur behavior was closer to THUMS when impacted at an angle of 6 degrees. X-direction displacement of the lower tibia was also closer to THUMS when the Flex-PLI was inclined at 6 degrees.

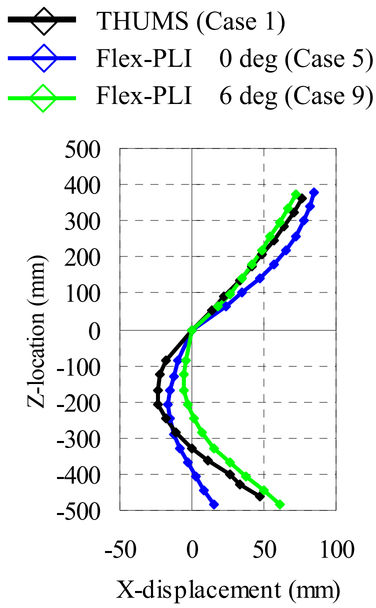


Figure 19. Effect of impactor angle: behavior of THUMS and Flex-PLI (sedan, mass: 14 kg).

Figure 20 shows the time history curves of the tibia bending moment (measurement position: tibia-1) in the above cases. Bending moment was generated at 8 ms when the inclination angle was 0 degrees, but at 3 ms at 6 degrees. This latter value was closer to THUMS. There were no major changes in maximum moment.

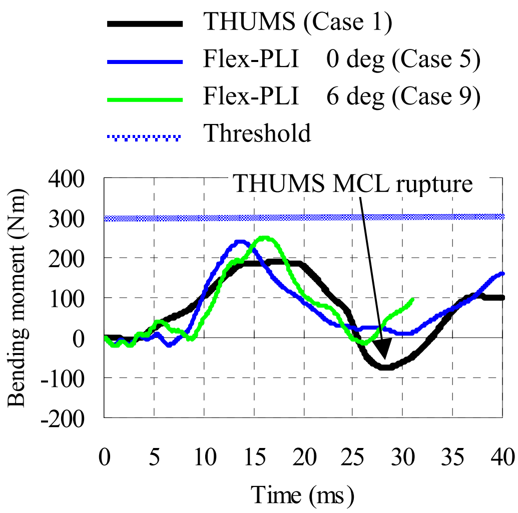


Figure 20. Effect of impactor angle: tibia bending moment (sedan, mass: 14 kg).

Figure 21 shows the time history curves of the knee bending angle for the same cases. The MCL in THUMS ruptured at approximately 10 degrees, immediately after which the bending angle increased suddenly. The bending angle continued to increase after MCL rupture. When the Flex-PLI was inclined at 6 degrees, the bending angle closely followed THUMS until 15 ms. After this point, however, the bending angle increased, greatly exceeding that of THUMS. When there was no inclination, the overall trend of the knee bending angle was closer to that of THUMS, despite the localized peak that occurred between 15 and 25 ms.

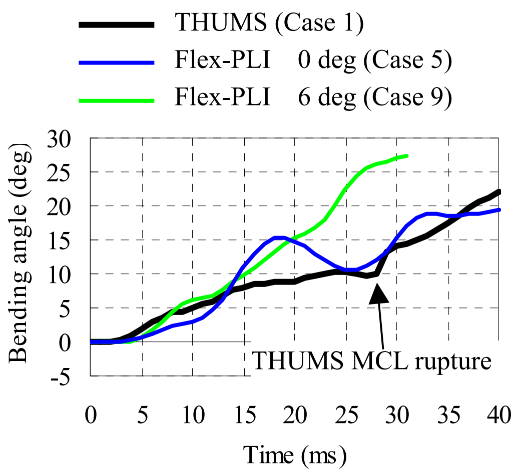
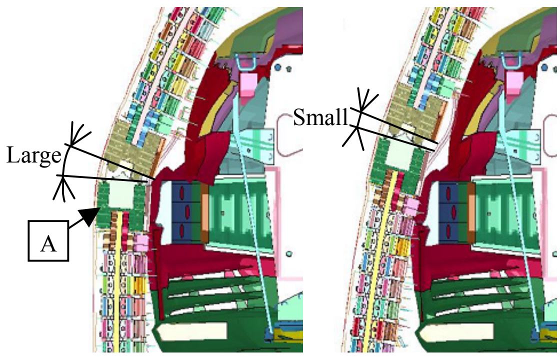


Figure 21. Effect of impactor angle: knee bending angle (sedan, mass: 14 kg).

DISCUSSION

Effect of Impact Height

Figure 22 compares the Flex-PLI behavior in the cases with the sedan at impact heights of 0 mm and 75 mm. The figure shows the degree of deformation 20 ms after the impact in each case. When the impact height was adjusted to 0 mm, the knee block on the tibia part (A) was positioned at the same height as the upper bumper absorber, meaning that the load of the vehicle bumper was mostly applied to the knee block. Since the knee block is made of highly rigid steel, bending deformation was exclusively concentrated in the knee joint and the tibia part did not deform greatly. In contrast, when the impact height was set to 75 mm, since the upper absorber contacted both the tibia and the knee block, bending deformation was generated at both the tibia part and the knee joint. In this case, bending was not concentrated only at the knee joint. Although a human tibia bone also widens proximally, in an impact with a sedan, bending deformation does not concentrate only at the knee joint. Therefore, it is preferable to set the impact height to 75 mm in order to simulate realistic bending deformation of the human tibia-knee complex, which is essential for injury assessment.



Impact height 0 mm (Case 2)

Impact height 75 mm (Case 7)

Figure 22. Deformation of Flex-PLI in impact with sedan (20 ms).

Figure 23 compares the knee and tibia bending angles in each model. The tibia bending angle is defined as the difference between the inclination angle of the knee lower block from the vertical axis and the inclination angle of the lower end of the tibia.

In THUMS, although the knee bending angle showed a simple increasing trend, the tibia bending angle became extremely large before decreasing. With the Flex-PLI, both the knee and tibia bending angles decreased after reaching extremely high values. As stated above, the Flex-PLI knee bending angle is closer to THUMS at an impact height of 75 mm. The maximum impactor tibia bending angle is lower than THUMS and reached at an earlier timing at both impact heights. However, the time history curves in both cases are close to THUMS. Therefore, to assess the risk of ligament rupture related to the knee bending angle, it is preferable to set the impact height to 75 mm.

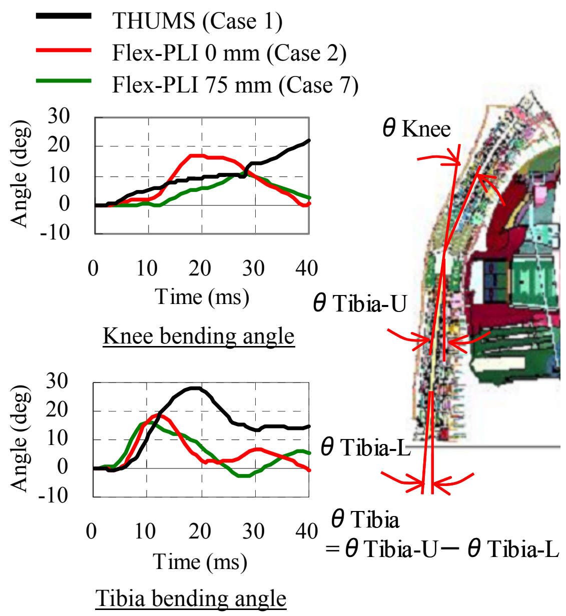


Figure 23. Tibia and knee bending angles.

Appropriate Additional Mass

Figure 24 compares the behavior of THUMS and the Flex-PLI with different additional masses applied to the top end of the Flex-PLI. The figure shows the degree of deformation at 20 ms after impact with the SUV in all cases. When no additional mass was set, X-direction displacement of the femur diverged from THUMS. In contrast, in all cases with an additional mass, X-direction displacement was closer to THUMS and there was little change based on the amount of mass.

The effect of the additional mass was considered by focusing on the leg center of gravity (CG). Figure 25 shows the CG of the THUMS leg and the Flex-PLI. In THUMS, the femur CG is located 227 mm from the knee joint. Including the femur and the pelvis, the CG of the THUMS leg becomes 310 mm from the knee joint. The femur CG of the Flex-PLI is 202 mm from the knee joint, which is closer to that of THUMS. With additional masses of 6 kg, 10 kg, and 14 kg, the CG from the knee joint is 306 mm, 358 mm, and 380 mm, respectively. Although the CG moves upward as the amount of mass is increased, the closest value to that of THUMS when the femur and pelvis are considered is with an additional mass of 6 kg. Therefore, it appears that the addition of more mass did not cause discrepancies in femur behavior because the addition of 6 kg created a CG closer to that of THUMS. Thus, an additional mass of 6 kg is considered to be sufficient for injury criteria assessment.

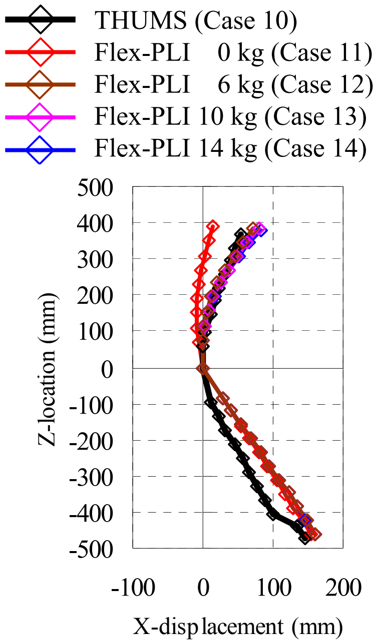


Figure 24. Effect of additional mass: behavior of THUMS and Flex-PLI (SUV, 20 ms).

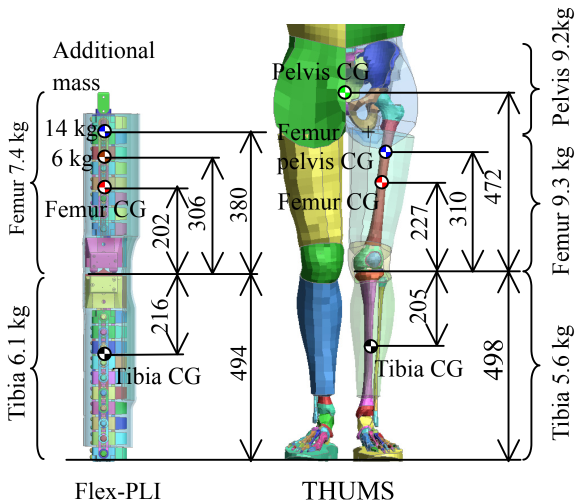


Figure 25. Centre of gravity of THUMS and Flex-PLI.

Effect of Impactor Inclination Angle

As shown in Figure 25, the skeletal structure of the human leg generally inclines inward when walking. Additionally, the external shape of the leg becomes

thinner from the base of the femur to the ankle. In a sedan impact, there is contact between the vehicle and the tibia, but the lower the part downward along the tibia, the later the contact occurs. In this study, when the Flex-PLI was inclined at 6 degrees to match the leg shape of THUMS, the trend for the timing of tibia bending moment was closer to that of THUMS. Therefore, adjusting the timing of contact with the vehicle at each leg position to reflect impact with an actual person is likely to be more effective for assessing injury criteria.
In contrast, since the results with a knee bending angle of 0 degrees were closer to THUMS than with an angle of 6 degrees, this is thought to be better for more accurate assessment.
The present results do not conclude which impactor inclination angle is better for injury criteria assessment of the leg as a whole. This is one possible area for study in the future.

CONCLUSIONS

(1) An FE model of the Flex-PLI was created by reverse engineering. X-ray CT scans were used to faithfully recreate the shape of the actual Flex-PLI, and the mechanical response of each component part was investigated before being input into the model.
(2) The Flex-PLI FE model was validated against actual impactor behavior and response in static 3-point bending tests on the femur, tibia, and knee joint, and dynamic pendulum tests in an assembled state. The results revealed that the impact behavior of the FE model closely correlated with that of the actual Flex-PLI, and that the mechanical response for moment and the like was within the test data corridors.
(3) The impact behavior and mechanical response of the Flex-PLI FE model and the THUMS full-body pedestrian FE model were compared to investigate suitable test conditions for assessing pedestrian leg injury. The following three test conditions were studied.
- Impact height above the ground
- Additional mass for simulating the upper body
- Impactor inclination angle

It was found that impact behavior at an impact height of 0 mm was closer to that of THUMS, but that 75 mm was closer in terms of injury response. For the effect of adding mass, it was found that the addition of 6 kg enabled a response closer to that of THUMS for an impact with an SUV. It was also discovered that the timing of tibia bending moment was closer to the response of THUMS at an impactor inclination angle of 6 degrees, but that the results with a knee bending angle of 0 degrees were closer.
In conclusion, these findings indicate that the recommended conditions for assessing leg safety performance with the Flex-PLI are an impact height of 75 mm above the ground, an additional mass of 6 kg for SUV impacts, and an inclination angle of 0 degrees.

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