BIOFIDELITY OF TEST DEVICES AND VALIDITY OF INJURY CRITERIA FOR EVALUATING KNEE INJURIES TO PEDESTRIANS

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ABSTRACT

In the current test procedure proposed by the European Enhanced Vehicle-safety Committee (EEVC) /WG17 for evaluating leg injuries to pedestrians, a legform impactor with a rigid bony structure is used. The risk of damages to knee ligaments is evaluated with the shearing displacement and the bending angle at the knee joint. A recent study has focused on evaluating biofidelity of the legform. However, it was not possible to obtain a local deformation at the knee joint from published experiments with Post Mortem Human Subjects (PMHSs). In addition, past PMHS experiments have suggested that the height of a bumper significantly affects the risk of ligamentous damages.

In this study, three kinds of finite element models were used in order to investigate the relationship between the bumper height and the shearing displacement / bending angle of the knee; 1) a legform impactor with a rigid bony structure, 2) a recently developed pedestrian dummy (Polar) with both a flexible tibia and a biofidelic knee joint structure, 3) a human lower limb. By utilizing the human model, a local deformation at the knee joint could be obtained. The model for the legform impactor and the pedestrian dummy have been validated against experiments with an actual car and in component level, respectively. The human lower limb model has been validated against published PMHS experiments. The result of a parameter study with these models in a range of bumper heights showed that the dynamic response of the dummy model is quite similar to that of the human model. In addition, it was found that the mass of the upper body significantly affects the bending angle of the knee. A geometric analysis of the knee joint was also performed to obtain tensile strains of four principal knee ligaments as a function of both the shearing displacement and the bending angle. The result suggested that the shearing displacement and the bending angle should be considered in combination when developing an injury criteria for knee ligaments.

INTRODUCTION

The test procedure for evaluating the risk of injuries to the leg of pedestrians proposed by EEVC/WG17 uses a legform impactor with a rigid bony structure and compliant elements at the knee joint. The shearing displacement / bending angle at the knee joint and the linear acceleration below the knee are measured to evaluate the risk of ligament failures and bone fractures, respectively [1]. The shearing displacement of 6 mm and the bending angle of 15 degrees are proposed as an injury criteria for ligamentous damages. Although the basis for these values seems to be somewhat uncertain, they are basically based on PMHS test results [2]-[4]. Consequently, the legform should ideally have complete biofidelity in order to assure a direct use of the human threshold. Otherwise, a transfer function between the human and impactor response should be developed and reflected in the criteria for the legform impactor.

Several studies have focused on the biofidelity of the legform impactor. Harris [5] stated that the effect of the upper body can be neglected because knee injuries and leg fractures occur early in an impact. Although he referred to Cesari et al. [6] as a biomechanical basis for this, there is no mention as to the timing of injuries in this paper. Sakurai et al. [7] conducted dynamic leg impact tests using the leg impactor designed by the Japan Automobile Research Institute (JARI) with and without the upper body. The test results showed that the tensile force and bending angle at the knee joint are affected by the mass of the upper body. Matsui et al. [8] conducted biofidelity legform impact tests with legform impactors designed by JARI and the Transport Research Laboratory (TRL). Their test setups were similar to those of PMHS tests conducted by Kajzer et al. [9][10]. They compared impactor test results with the response corridors obtained from the PMHS tests by Kajzer et al. and concluded that none of the currently available legform impactors meets the biofidelity requirements for the impact force and the shearing displacement / bending angle of the knee.

Recently, Honda R&D in collaboration with GESAC Inc. developed a new thigh-knee-leg complex for the pedestrian dummy called Polar [11][12]. The dummy lower limb features a human-like construction including the femoral condyles, the tibial plateau, the meniscus, the four principal knee ligaments and the flexible tibia. The dynamic response of the knee joint was validated against the response corridor for the impact force determined from the PMHS tests performed by Kajzer et al [8]-[10]. However, the shearing displacement and the bending angle of the knee have not been validated in spite of the significance of these parameters in evaluating the risk of knee ligament failures.

As has been noted, these biofidelity validations were based on the latest PMHS tests conducted by Kajzer et al. [9][10]. However, in these PMHS experiments, the shearing displacement and the bending angle were obtained from a high speed film analysis by tracking target marks screwed to the femur and the tibia. Therefore, the lateral bending of these bones could affect the results. In addition, they used an impactor with the mass comparable to that of the human leg and foot. This could result in a larger contribution of the inertia of the leg and foot rather than that of ligament forces, particularly in their bending setup where the leg was hit by the impactor at the ankle joint. In order to validate the biofidelity in the knee joint response, an alternative way that directly compares the local deformation at the knee joint is required.

The authors have developed a finite element model for a human lower limb of pedestrians. This model was presented at the STAPP Car Crash Conference in 2000 [13]. Mechanical properties of human bones and soft tissues were obtained from their extensive literature survey, and the model was validated against published PMHS tests in both quasi-static and dynamic conditions. Although again the PMHS experiments by Kajzer et al. [9][10] were used for a part of model validations, the model showed a reasonable prediction of ligament failures as compared to those observed in the experiments. In addition, the model was validated against the PMHS experiments by Bunketorp et al. [14] performed in setups that represent more realistic car-pedestrian impact situations. By utilizing this model, it is possible to estimate the dynamic local deformation at the human knee joint.

This study focused on the biofidelity evaluation of the anthropomorphic test devices in the dynamic knee joint response, and the validity of the currently proposed acceptance levels for ligament failures. Computer simulation models for the TRL legform impactor and the Polar dummy were constructed and validated against experimental results. The dynamic response of the knee joint from these models was compared with that from the human model. Based on the result of the parameter study with various bumper heights, the effect of the upper body mass, the deflection of bones and the ankle joint property will be discussed. In addition, the validity of the currently proposed injury criteria for ligamentous damages will be examined by means of geometrical analysis and computer simulations using the human model.

MODEL DESCRIPTION

Computer simulation models for the TRL legform impactor and the Polar pedestrian dummy have been constructed, and the knee joint response from these models has been validated against experiments. The finite element model for a human lower limb extensively validated in our previous study [13] has been utilized in this study.

TRL Legform Impactor

The latest version of the TRL impactor is equipped with a damper to avoid vibration in shearing. However, it was not possible to validate a model with the damper due to a lack of experimental results. Testings conducted by TRL with the damped legform showed that the damper diminished the undesirable vibration without affecting the knee shear performance [15]. Based on this, it was decided to model the TRL impactor without the damper.

Figure 1 shows a schematic diagram of the TRL legform impactor without the damper. A cantilever inside the thigh segment is primarily responsible for the shearing deformation of the knee, and deformable steel ligaments provide the bending stiffness. The computer simulation model for the TRL legform developed in this study is illustrated in Figure 2. Three segments have been modeled as rigid bodies; rigid parts of the thigh and leg segment, and the cantilever inside the

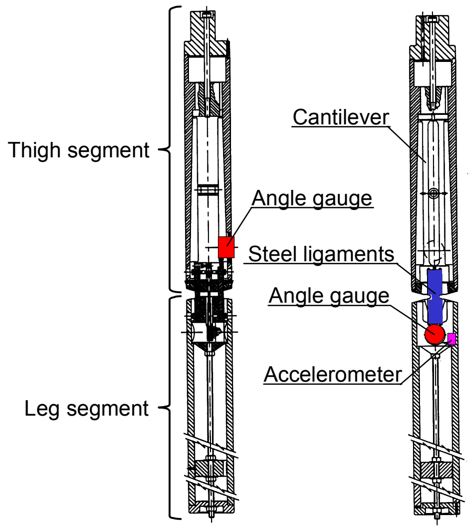

Figure 1. Schematic diagram of TRL legform impactor

thigh segment. These segments have been connected to each other using joint elements. The shearing and bending properties obtained from certification tests have been applied to the upper and lower (knee) joint elements. The foam flesh around bony structures have been modeled with solid elements.

Figure 3 shows the model setup used for validation. The setup followed the EEVC test procedure [3] and the legform was propelled into a stationary vehicle at 40 km/h. The experiment with the same setup was also performed. Figure 4 shows the validation results of this model. The shearing displacement and the bending angle of the knee as well as the linear acceleration below the knee were compared between the experiment and the computer simulation. Generally good correlation could be obtained for all of the three parameters.

Polar Pedestrian Dummy

Figure 5 illustrates the lower limb of the Polar dummy. The knee joint structure of the dummy accurately represents the human anatomy including the femoral condyles, the tibial plateau, the meniscus and the four ligaments. Spring-wire system has been used to provide tensile properties of human knee ligaments. These features have been precisely modeled as shown in Figure 6. The wires for knee ligaments have been modeled with bar elements in combination with slip rings. The meniscus has been modeled with solid elements to simulate the axial compressive property of the knee joint. The hip and ankle joint have been modeled with spherical joints, while no mechanical joint has been specified for the knee joint. Bracket joints have been applied at the centers of load cells in order to provide output for forces and moments. The femur and the epiphyses of the tibia including load cells have been modeled as rigid, and the flexible diaphysis of the tibia has been modeled with solid elements. The foam flesh around bony structures has been modeled with solid elements, and the surface skin layer has been modeled with shell elements.

The model for the flexible tibia has been validated in both quasi-static and dynamic 3-point bending. Experimental results were obtained from Artis et al.

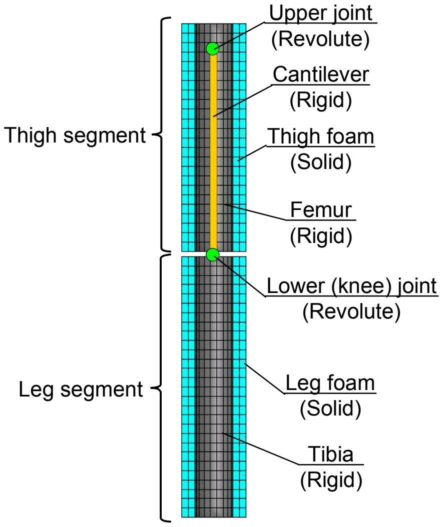

Figure 2. Structure of legform model

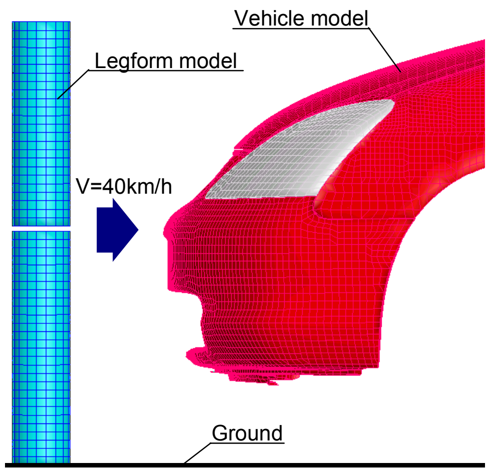

Figure 3. Setup for legform model validation

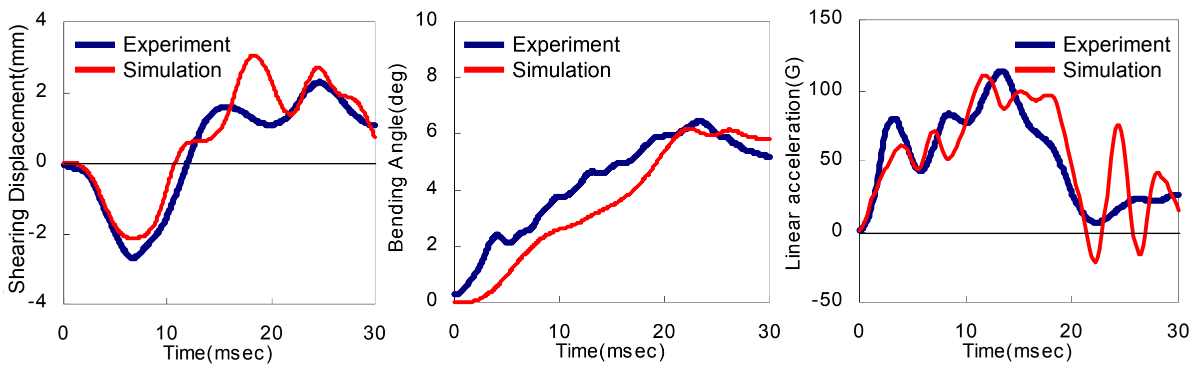

Figure 4. Comparison of shearing displacement, bending angle and linear acceleration between experiment and simulation

[11]. The model setups are summarized in Figure 7. In dynamic condition, the tibia was wrapped with the foam flesh in order to simulate a realistic interface. Figure 8 plots the force-deflection curves obtained from the experiment and the computer simulation in both quasi-static and dynamic conditions. Although the simulation result shows a slight difference as compared to the test result particularly in dynamic condition, good agreement could be obtained for peak force level.

The model has also been validated in thigh-leg-foot assembly level. Artis et al. [11] conducted knee response tests with setups similar to those used in the PMHS experiments by Kajzer et al. [9][10]. Simulation models with similar setups have been constructed as shown in Figure 9. In the tests performed by Artis et al., the lateral displacement of both the distal femur and the proximal tibia were measured. Although ideally the femur should be fixed rigidly, the lateral displacement

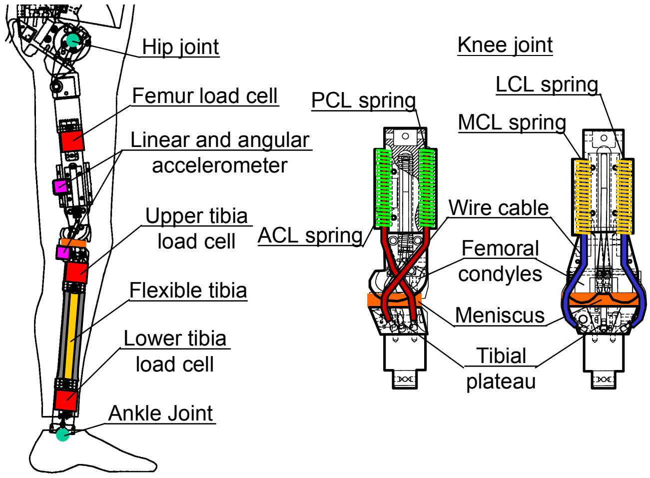

Figure 5. Schematic diagram of the Polar dummy lower limb

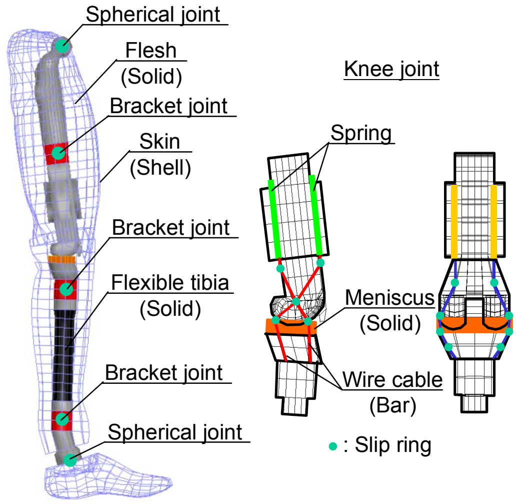

Figure 6. Structure of dummy lower limb model

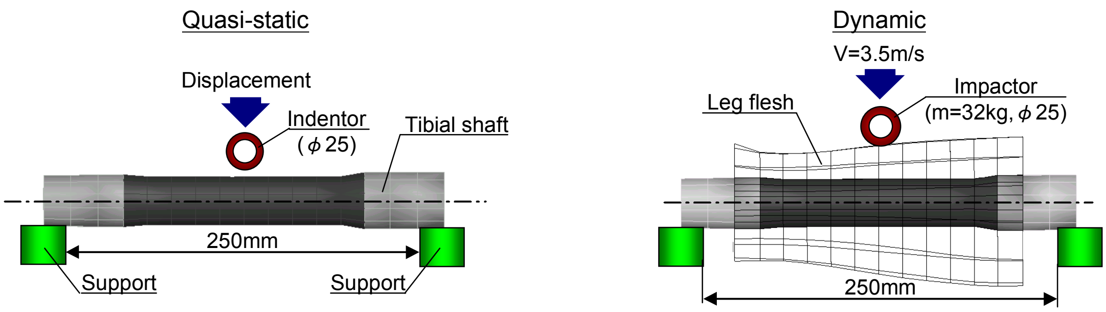

Figure 7. Model setup for quasi-static and dynamic 3-point bending of dummy flexible tibia

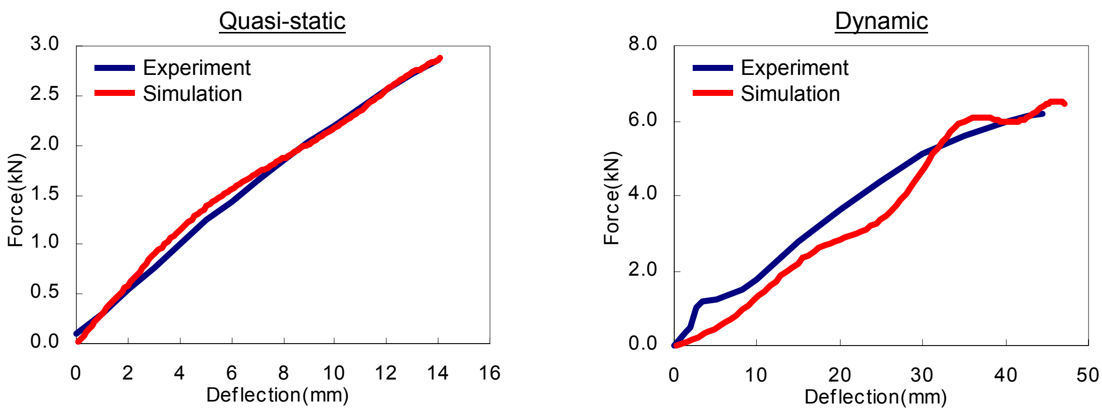

Figure 8. Comparison of force-deflection curve between experiment and simulation in both quasi-static and dynamic 3-point bending

at the distal femur was found to be comparable to that at the proximal tibia, particularly in shearing setup. Since the detail of the femur fixation was not available, it was decided to use the relative lateral displacement at the knee joint obtained by subtracting the femur displacement from that of the tibia. Both the impact force and the knee lateral displacement were compared for the impact velocities of 20 km/h and 40 km/h. Figures 10 and 11 plot the experimental and simulation results in shearing and bending setup, respectively. Due to the limited rigidity of the femur fixation, the impact force from the experiment is lower than that from the simulation in shearing setup. However, the simulation results for other parameters show good agreement with the experimental results.

Human Body

The finite element human lower limb model developed by the authors [13] is illustrated in Figure 12. The epiphysis of bones has both a surface cortical layer and an internal trabecular layer. The diaphysis consists of only a cortical layer with a medullary cavity inside. Four knee ligaments have been modeled with shell elements in order to take into account the contact between bones and ligaments. The meniscus and the articular cartilage at the knee joint have been modeled with solid elements to provide a realistic compliance in axial compression. The rupture model in PAM-CRASH™ has been applied to bones and ligaments in

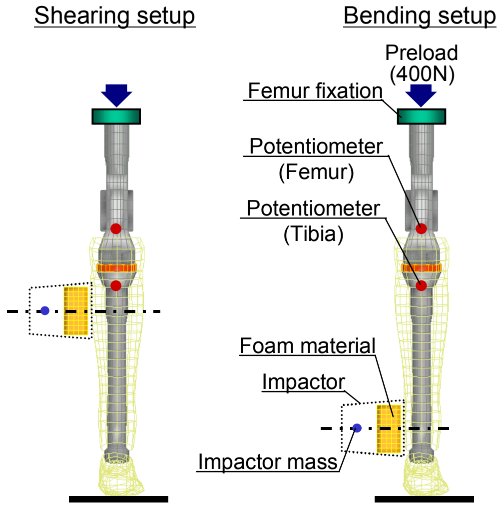

Figure 9. Dummy model setup for knee response validation

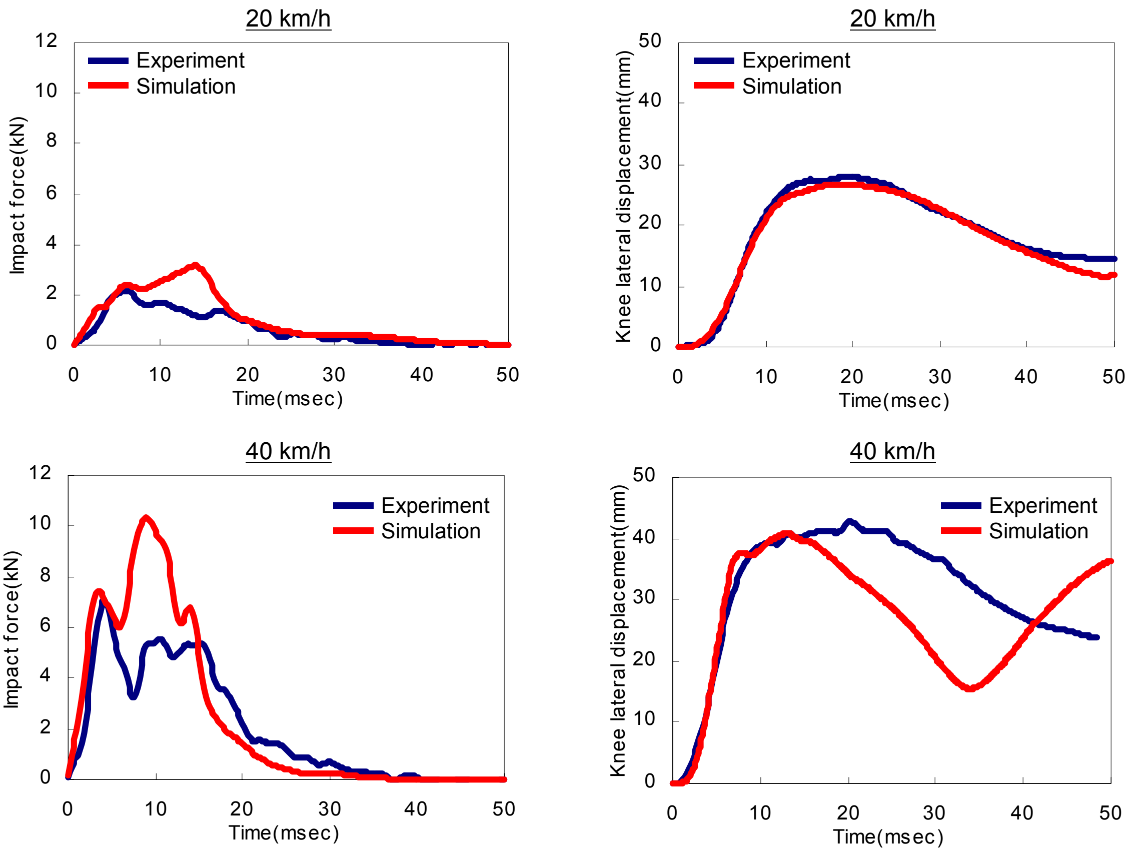

Figure 10. Comparison of impact force and knee lateral displacement between experiment and simulation for 20 km/h and 40 km/h in shearing setup

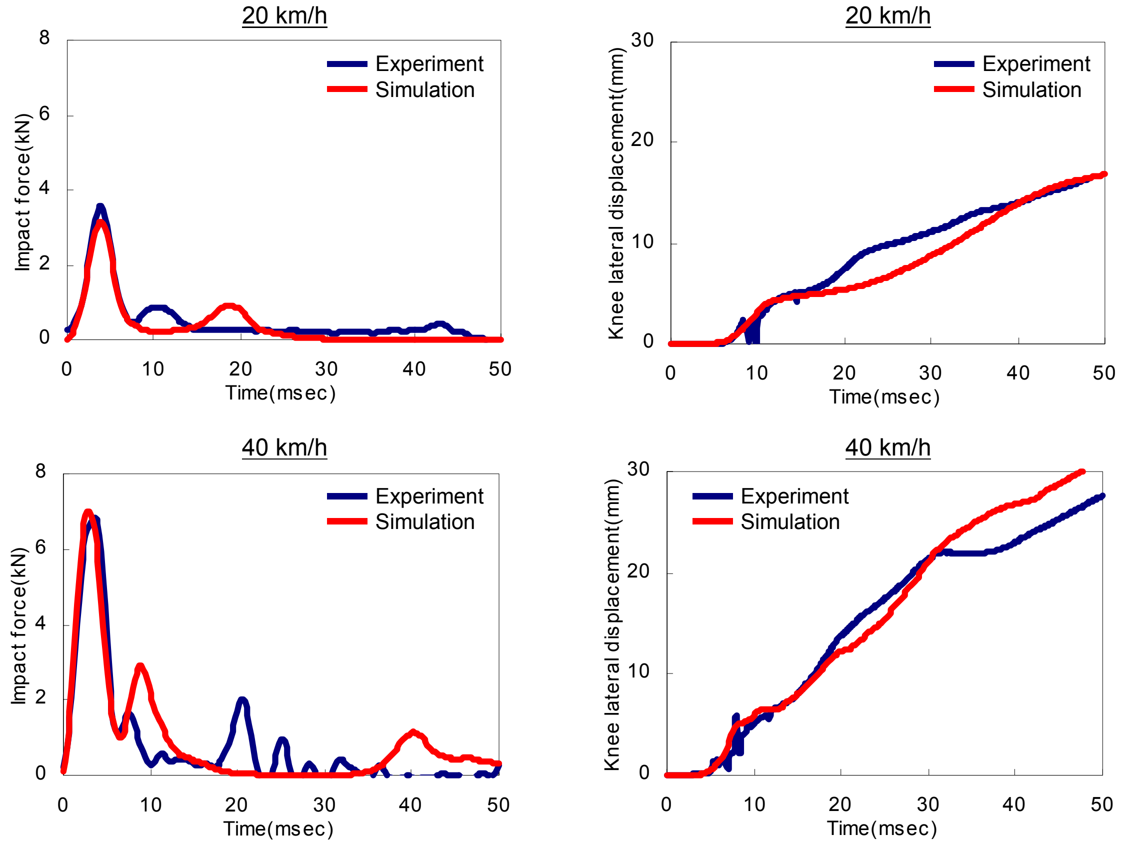

Figure 11. Comparison of impact force and knee lateral displacement between experiment and simulation for 20 km/h and 40 km/h in bending setup

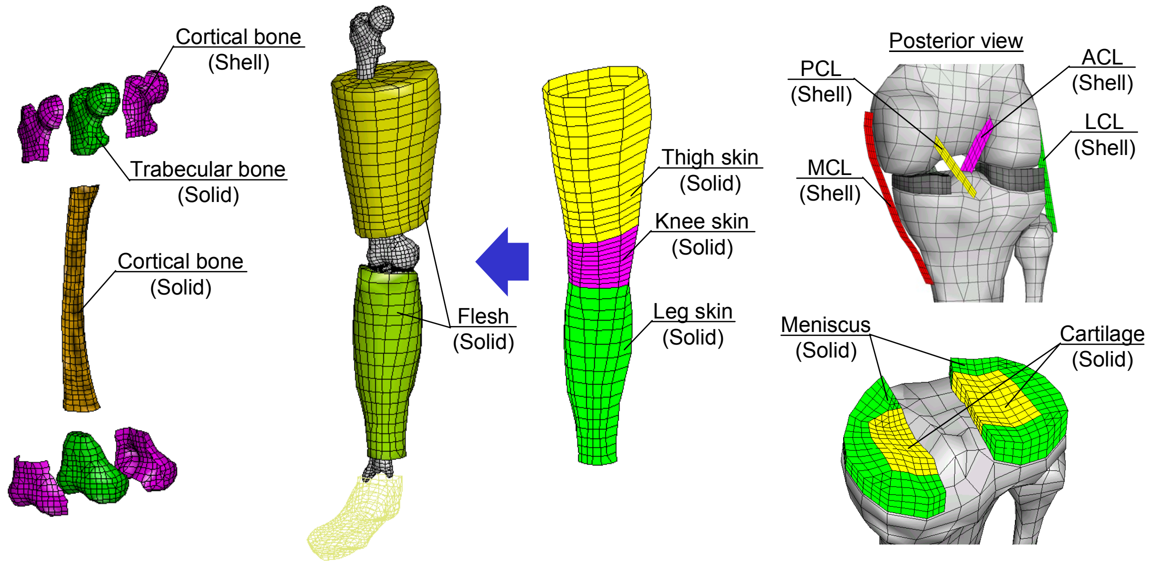

Figure 12. Schematic diagram of human lower limb model

order to simulate bone fractures and ligamentous damages. The flesh and the skin were also incorporated for precise interface definitions. The bones have been validated against published quasi-static and dynamic 3-point bending test results. The knee joint has been validated against experiments performed by Kajzer et al. [9][10] including the comparison of the injury description for knee ligaments. The lower limb model has been further validated against PMHS tests conducted by Bunketorp et al. [14] including the comparison of linear accelerations of the leg and injuries at the knee joint. For detailed information regarding the human lower limb model, see reference [13].

BIOFIDELITY EVALUATION

Past PMHS studies [14][16] suggest that the height of the bumper significantly affects the risk of injuries to the lower limb of pedestrians. Based on this finding, it was decided to run computer simulations for biofidelity evaluation in a range of different bumper heights. The TRL legform model, the Polar dummy model and the human lower limb model were subject to the same impact from a simplified bumper model at various heights, and the shearing displacement and the bending angle were compared among the three models. For the dummy and the human model, a full body including both limbs and the upper body was used, and the right and left limb were rotated 10 degrees in flexion and extension, respectively. The mass of the bumper was set approximately 1400 kg to simulate a typical passenger car. 40 km/h for the impact velocity was selected on a basis of the EEVC test procedure. Figure 13 shows simulation setups for the three models. A simplified bumper model including a resinous bumper face and a steel bumper beam was used as an impactor. Because the height of the knee joint was slightly different among the three models, a standard height of the bumper for each model was set equal to the height of its knee joint. The bumper height was then varied from the standard position by 40 mm and 80 mm in superior (+) or inferior (-) direction. Thus, five cases of bumper heights were used in total.

Figure 14 shows the comparison of the shearing displacement of the knee as a function of the bumper height among the three models. At the standard (0 mm) and +40 mm of the bumper height, both the tibial and femoral epiphyses made contact with the bumper in the dummy and the human model due to the deformation of the bumper face. In these cases, a slight lateral protrusion of the tibial condyles relative to the femoral condyles affected the shearing displacement in the human model. Due to this unreal protrusion in the human model, the sign of the shearing displacement differs between the dummy and the human model in these two cases. This minor error in modeling may have happened when the posture of the model was changed from a seating position to a standing position. The alignment of the tibia relative to the femur in the human model should be slightly modified to obtain more realistic results. However, the result from the dummy model matched quite well with that from the human model for the rest of the cases. The shearing displacement from the legform model also showed generally good agreement with the human response except for the magnitude at the bumper height below the knee joint (-40 mm, -80 mm).

Figure 15 shows the variation in the bending angle of the knee as a function of the bumper height from the three models. The dummy response is quite similar to the human response. Although the legform impactor shows an increase in bending angle as the bumper height increases from -80 mm to the standard height (0 mm), the bending angle inversely decreases with an additional increase in the bumper height up to +80 mm. Figure 16 shows the comparison of the lower limb kinematics from the three models at the bumper height of +80 mm. It is obvious that the knee bending of the legform model is significantly smaller than that of the human and dummy model.

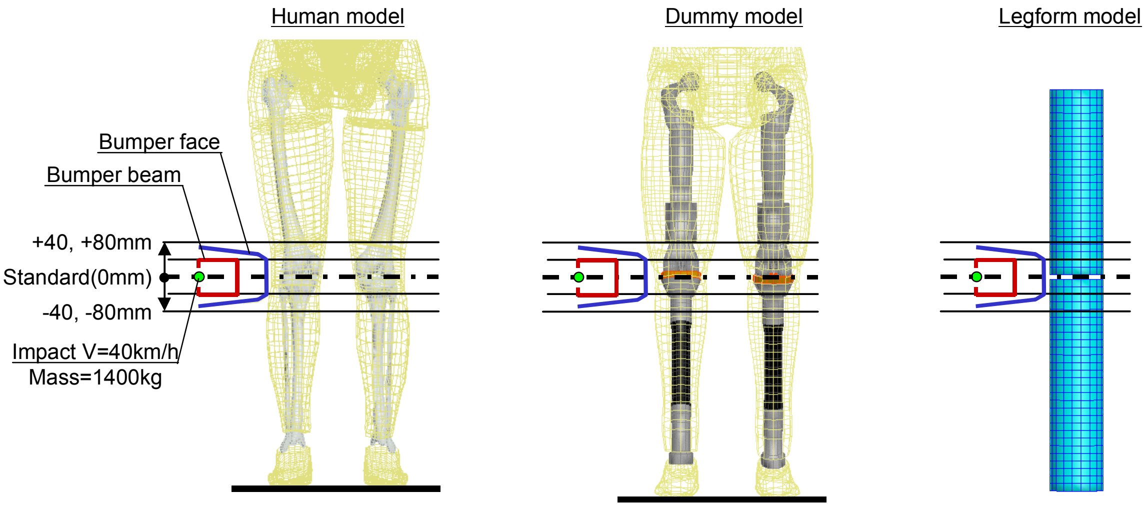

Figure 13. Simulation setup for human, dummy and legform model (posterior view)

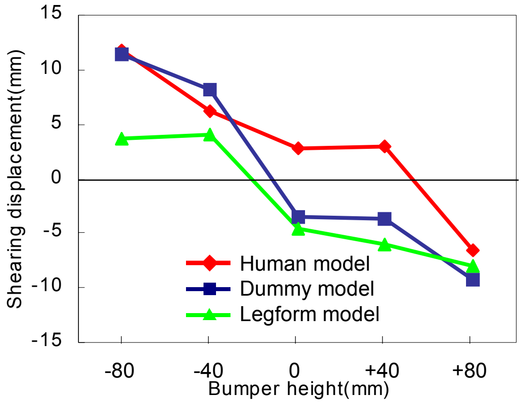

Figure 14. Comparison of shearing displacement of the knee among three models

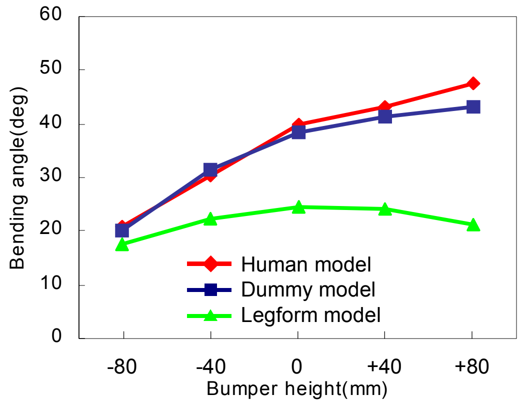

Figure 15. Comparison of bending angle of the knee among three models

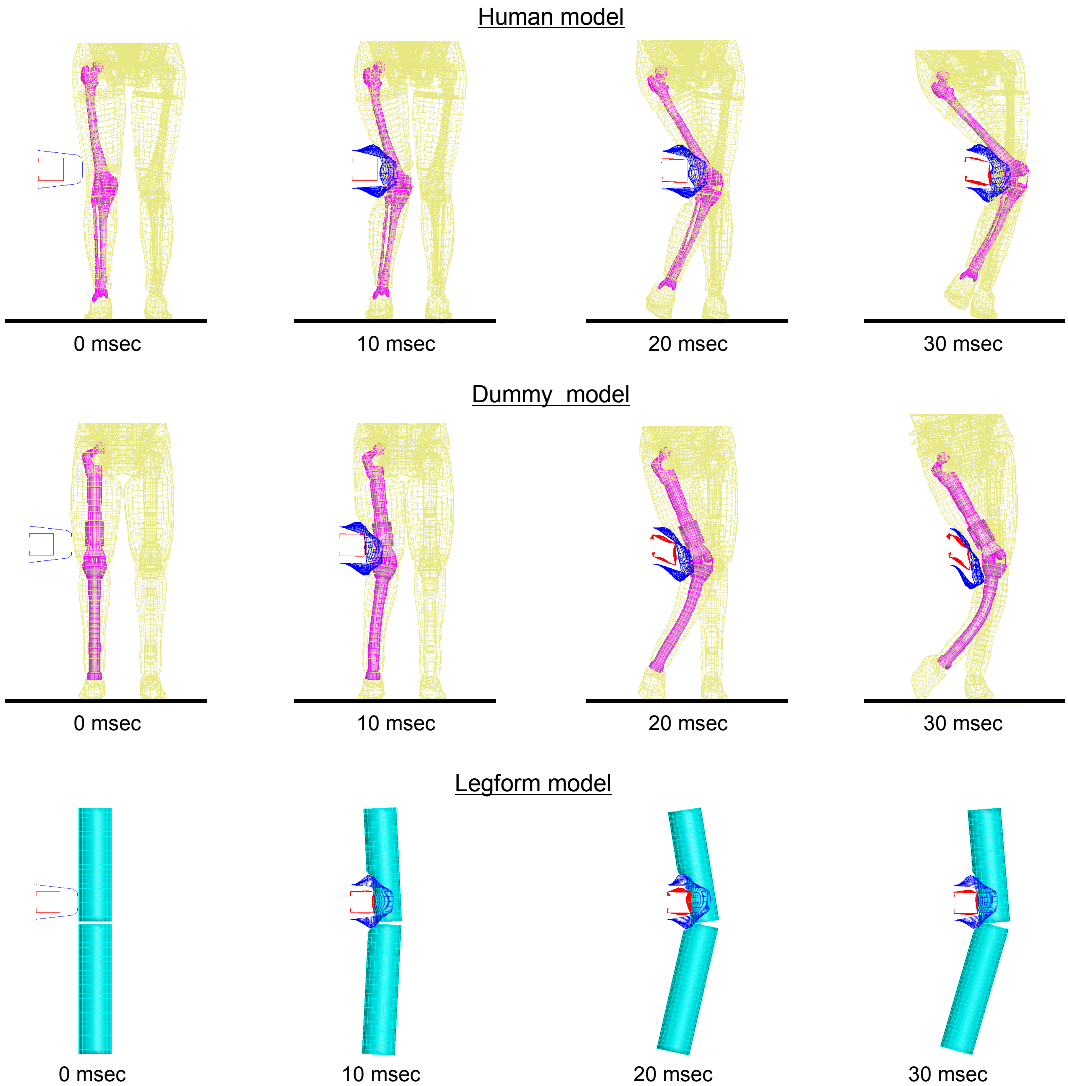

Figure 16. Comparison of knee joint kinematics among human, dummy and legform model at bumper height of +80 mm (posterior view)

INJURY CRITERIA

EEVC/WG17 proposed the shearing displacement of 6 mm and the bending angle of 15 degrees as acceptance limits for knee ligament damages. However, the validity of the proposed criteria is still under discussion. JARI and the Japan Automobile Manufacturers Association (JAMA) claimed that the acceptance level should be determined using a combination of a maximum bending angle and a maximum shearing displacement [3]. Though their idea was rejected by the working group because of the time difference between peaks of these two parameters, it can be still asserted that the shearing and bending mode can take place simultaneously and thus both parameters should be considered in combination.

One of the basis for the shearing displacement of 6 mm is that the typical injury associated with shearing mechanisms is the rupture of the anterior cruciate ligament (ACL), and a limit of 6 mm can be obtained from the length of the ligament (30 mm) multiplied by an estimated strain at rupture (20 %) [2]. The idea of using a strain of ligaments can be justified by Eppinger et al. [17]. By analyzing the properties of the human anterior longitudinal ligaments experimentally determined by Yoganandan et al., they suggested that the failure of the ligaments was only dependent on the ultimate strain and was independent of the stress. Nevertheless, apparently the geometry of the ligament attachment was ignored in estimating a limit of 6 mm for the shearing displacement. In fact, it is obvious that the ACL cannot be tensed in pure longitudinal direction when a knee was subject to shearing.

Based on the assumption that the failure of ligaments can be detected solely by their strains, a geometric analysis of the knee joint when subject to shearing and bending was performed. Figure 17 illustrates a schematic diagram of the knee joint geometry taken from our human lower limb model. 3D coordinates of both the superior and inferior attachments of four main ligaments (Anterior Cruciate Ligament ; ACL, Posterior Cruciate Ligament ; PCL, Medial Collateral Ligament ; MCL and Lateral Collateral Ligament ; LCL) were measured. Lateromedial impact on the left knee was assumed. The shearing displacement was defined positive when the tibia moves lateromedially relative to the femur. The bending angle was defined positive in valgus. By calculating the distance between two attachment points for each ligament, its strain was determined as a function of the shearing displacement and the bending angle. The center of rotation in bending was placed on the tibial plateau just below the lateral and medial femoral condyle for positive (valgus) and negative (varus) bending angles, respectively. The ultimate strain for each ligament was obtained from our human model, and the relationship between the shearing displacement and the bending angle at the ultimate strain was determined.

Figure 18 plots the results of the geometric analysis. The curves for only the ACL and the PCL were presented since the acceptance limits were found to be determined solely by these two ligaments. In order to validate the geometric results against realistic simulations, the data points obtained from the parameter study with the human model presented above were also plotted. In addition, the results from simulations representing PMHS experiments by Kajzer et al. [9][10] were also included. From these dynamic simulations, the shearing displacement and the bending angle at the time of failure were plotted (red squares). In some cases, no damage was observed to ligaments (blue circles).

It is obvious from this plot that the shearing displacement and the bending angle do not determine the risk of ligament failures independently. The results

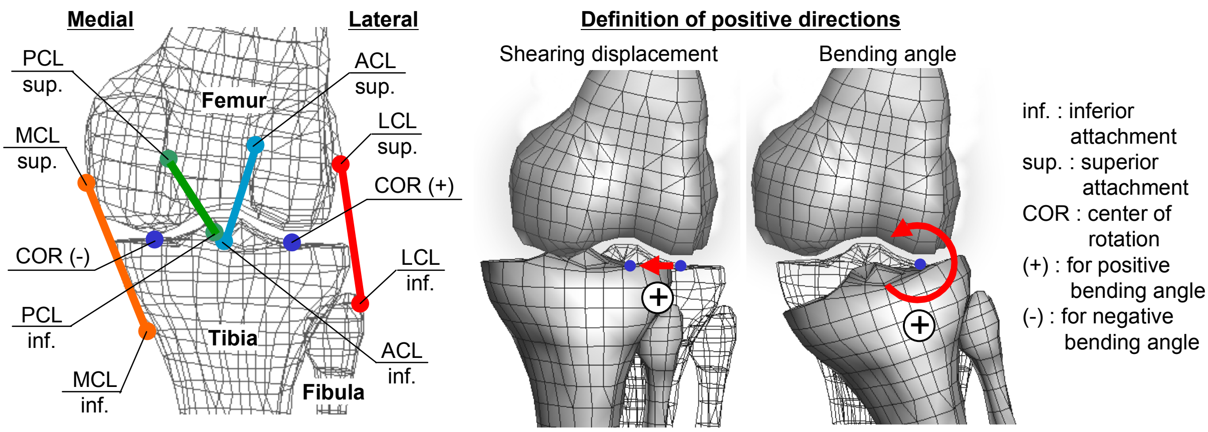

Figure 17. Knee joint geometry and definition of shearing displacement, bending angle and center of rotations

of the geometric analysis and dynamic simulations suggest that both of the two parameters should be considered in combination. In a positive shearing displacement, the most risky ligament is the ACL. However, for a negative shearing displacement, the most vulnerable ligament is the PCL. The negative shearing displacement is possible when a bumper hits the femur rather than the tibia, i.e. when the bottom of the bumper is above the knee joint. This suggests that the criteria for the PCL should be used in negative shearing displacement. It is also possible to have a negative bending angle due to the lateral bending of the tibia caused by an impact from the bumper. The dotted lines in Figure 18 show the results of the geometric analysis without consideration on the axial compression of the knee joint. It seems that the geometric result is conservative as compared to plots from dynamic simulations. This deviation between the results of geometric analysis and dynamic human model simulations can be explained by the axial compression of the knee joint due to the deflection of the meniscus and/or the articular cartilage. The axial compression at the knee joint can reduce the strain of ligaments. The solid lines in Figure 18 show the geometric results for the axial compression of 3 mm. It can be seen that the axial deflection of only 3 mm significantly affects the critical values. This suggests that a possible axial compression should be taken into consideration when developing an injury criteria. Though the results should be validated experimentally, the hatched area in Figure 18 can be suggested as an injury criteria for the knee ligaments. It should also be noted that the peak values of the shearing displacement and the bending angle do not necessarily determine the risk of ligament failures.

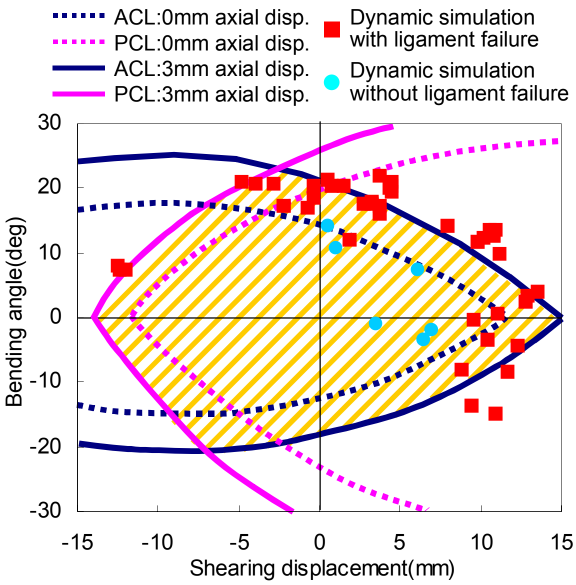

Figure 18. Injury criteria for knee ligaments

DISCUSSION

The result of the lateral impact simulations with a range of bumper heights presented above suggests that the bending response of the legform impactor differs from that of the human lower limb particularly when the bumper hits above the knee joint. This can be a major deficiency in evaluating the risk of ligamentous damages. In fact, the bending angle from the legform model decreases as the bumper height increases from the knee joint height to 80 mm above the knee, while the bending angle from the human model simply increases as the bumper height increases for the entire region. In order to further investigate the mechanism of this difference, additional simulations with the human model were run. Three factors that characterize the mechanical difference between the human lower limb and the legform were investigated. First, the effect of the upper body mass on the knee joint response has been examined since the legform completely ignores the mass of the upper body. Second, the influence of the deflection of bones has been checked because of the lack of deflection in rigid bony structures of the legform. Finally, the effect of the ankle joint property was investigated since the legform simply incorporates the mass of the foot into the leg segment without permitting any degrees of freedom between the leg and the foot.

Figure 19 shows three human model setups for evaluating the effect of the upper body mass. The full body model includes both limbs as well as an upper part of the body. The lower limb model with an additional mass contains only the hip joint and below

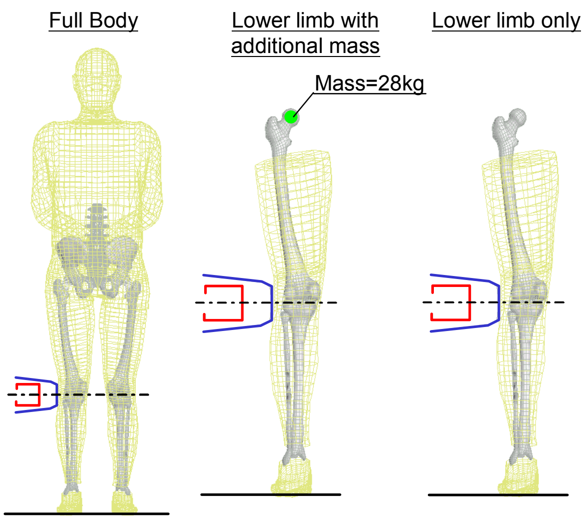

Figure 19. Simulation setup for evaluating effect of upper body mass (posterior view)

of a single limb, and a half of the upper body mass (approximately 28 kg) is added to the femoral head. The lower limb only model also includes the femur and below without the additional mass. For examining the influence of the bone deflection, the femur and/or the tibia were simply switched to rigid bodies from deformable finite elements. The effect of the ankle joint property was investigated by locking the ankle joint.

Figures 20-22 demonstrate the influence of the upper body mass, deflection of bones and ankle joint property, respectively. Both the shearing displacement and the bending angle were plotted against the bumper height for each condition. The upper body mass showed no significant influence on the shearing displacement. However, for the bending angle, the lower limb only model showed the trend quite similar

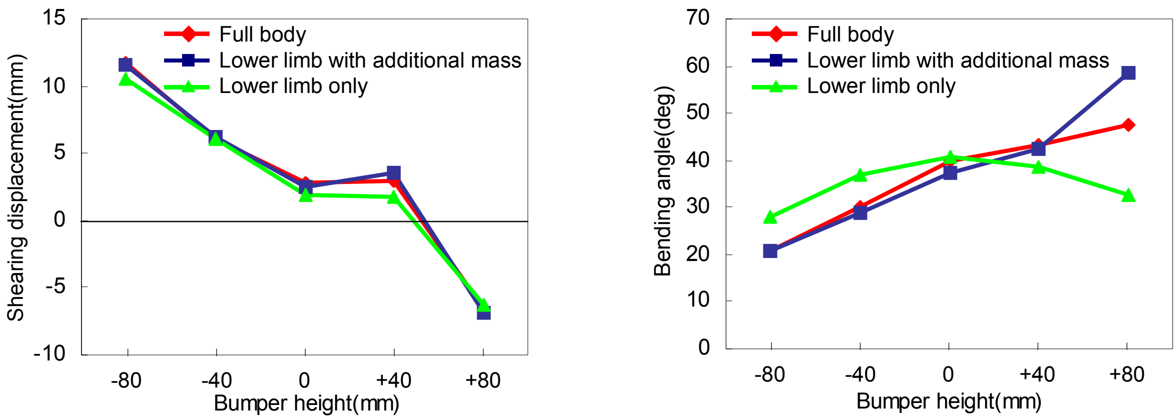

Figure 20. Effect of upper body mass on shearing displacement and bending angle

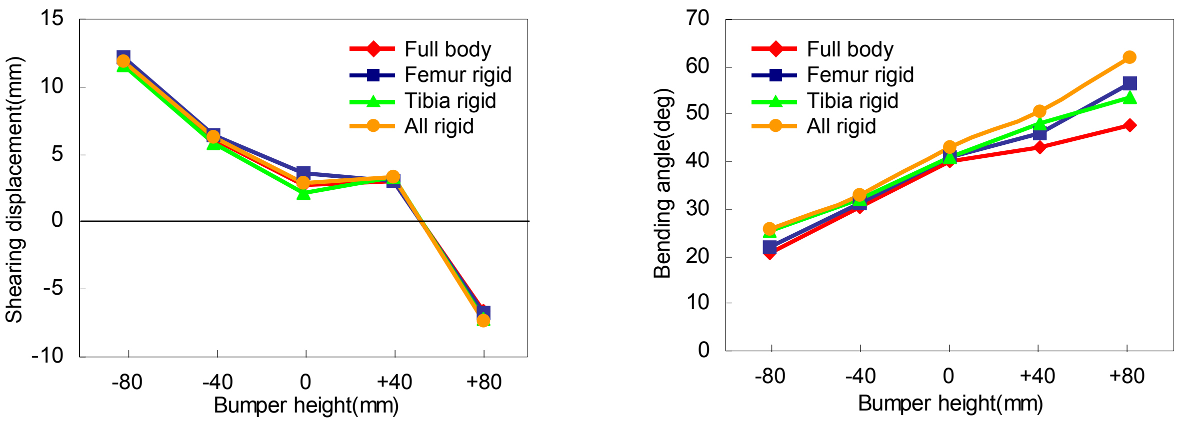

Figure 21. Effect of bone deflection on shearing displacement and bending angle

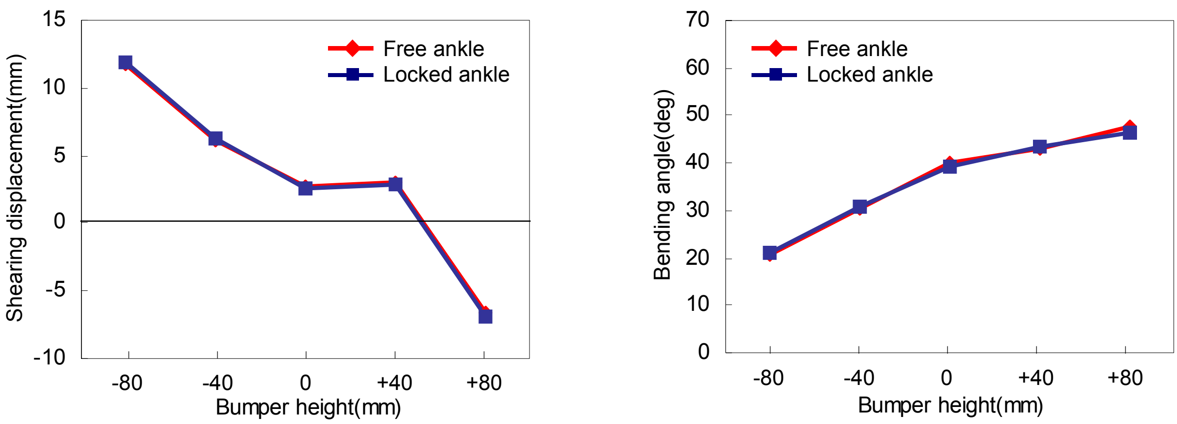

Figure 22. Effect of ankle joint property on shearing displacement and bending angle

to that of the legform impactor model presented in Figure 15, while the lower limb model with an additional mass showed the trend similar to that of the full body model. This result strongly suggests that the lack of the upper body, or the lack of the additional mass to compensate for the upper body, can be a major deficiency in evaluating the risk of ligament failures especially for the bumper height above the knee joint. In addition, it is anticipated that the difference of the bending angle in trend makes it impossible to develop a transfer function between the human and legform bending response. It may be claimed that the EEVC test procedure permits selecting the upper legform impact test instead of the legform test for the bumper height of 500 mm and higher. However, the full body simulation result shows that the bending angle increases as the bumper height increases. This suggests that the evaluation of the risk of damages to knee ligaments can be even more important for higher bumper heights. The deflection of bones has no effect on the shearing displacement, while it has a certain amount of influence on the magnitude of the bending angle, particularly at the bumper height above the knee joint. The ankle joint property showed no influence on both the shearing displacement and the bending angle.

CONCLUSION

Computer simulation models for both the EEVC legform impactor and the recently developed Polar pedestrian dummy were constructed and validated in order to evaluate the biofidelity of these anthropomorphic test devices by comparing the dynamic response of the knee joint with that from the human lower limb model previously developed and validated by the authors. Validation results for the legform and the dummy model showed generally good agreement with experimental results.

By utilizing computer simulation models for the test devices and the human body, it was possible to directly compare the dynamic local deformation at the knee joint. The results of a series of computer simulations using the three models in a range of bumper heights showed that the Polar dummy response was quite similar to that of the human model in terms of both the shearing displacement and the bending angle at the knee joint, while the legform model showed a significantly different trend in the bending angle. The most essential factor for this was found to be the lack of the mass of the upper body. The deflection of bones also has a certain amount of influence on the magnitude of the bending angle without altering a trend. The ankle joint property was found to have no significant effect on the knee joint response.

The result of the geometric analysis of the knee joint together with the data from dynamic simulations suggests that both the shearing displacement and the bending angle at the knee joint should be combined for evaluating the risk of ligamentous damages. A suggested acceptance level for knee ligament failures including both positive and negative shearing displacement / bending angle was presented. Although this suggested criteria needs to be validated experimentally, it can be pointed out that the axial deflection at the knee joint should also be taken into account in injury criteria development.

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