Response Corridors to Evaluate the Biofidelity of the Lower Limbs of Pedestrian Dummies

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ABSTRACT

This study focused on response corridors used to evaluate the biofidelity of the lower limbs of whole-body pedestrian dummies. Specifically, corridors for the thigh and the leg were investigated. The three-point bending tests of horizontally placed specimens of the thigh and the leg in past studies exhibited sagging flesh caused by gravity. This resulted in unrealistically thin flesh on the loaded side. This study investigated a methodology to eliminate the influence of the sagging flesh to develop more realistic corridors. The initial toe region of the force-deflection response from the experiment was eliminated from the test results to diminish the influence of unrealistic flesh thickness. Due to the difference in stiffness, the flesh would bottom out upon initiation of major bone deflection. The assumption was made that the magnitude of the force applied at the beginning of this phase is independent of flesh thickness. Dynamic three-point bending simulations were conducted using thigh and leg FE models with the flesh thickness varied to validate the assumption. In addition, existing experimental data were re-examined to determine the initiation of major bone deflection, and to calculate the magnitude of applied force at the same timing. Furthermore, full-scale car-pedestrian impact simulations were conducted using a human FE model with various flesh thicknesses to clarify the influence of the initial toe region of the force-deflection response. The corridors were then developed using the data after the applied force reached a specific value determined in this study. The results of three-point bending simulations with various flesh thicknesses showed that the magnitude of applied force at the initiation of major bone deflection was fairly constant, at approximately 2000 N and 1500 N for the thigh and the leg, respectively. These values were found to eliminate the initial toe region from the experimental data, while maintaining the stiffness of the subsequent region that is not influenced by the initial thickness of the flesh. The full-scale impact simulations showed that the peak values of the femur and tibia bending moments were not significantly influenced by the flesh thickness.

INTRODUCTION

In Japan, the number of car occupant fatalities has decreased by approximately 50% from 2005 to 2015. In contrast, pedestrian fatalities have decreased by only approximately 30% in the same term. In addition, pedestrians accounted for the largest proportion of traffic accident fatalities after 2008 [1]. In order to further reduce the number of all traffic fatalities, it is necessary to reduce the number of pedestrian fatalities. Although the safety performance for pedestrian was improved by using subsystem tests representing individual body regions of a pedestrian in new car assessment programs (NCAPs) and regulations, the whole-body of a pedestrian wraps around the vehicle surface in actual accidents. The complexity of the behavior of a pedestrian in actual car-pedestrian accidents does not allow the subsystem test procedure to clarify the mechanism of injuries to a pedestrian. In order to enlarge the cover range of pedestrian safety technologies, the detailed investigation for the mechanism of car-pedestrian accidents is needed. The investigation using a pedestrian dummy has been conducted as one of the ways to develop further understanding of real-world pedestrian accidents. In order to validate the results of such a study, the biofidelity of the dummy needs to be confirmed. To address this issue, the performance specifications for a whole-body pedestrian dummy are needed. Experiments using PMHSs (Post Mortem Human Subjects) have been conducted in order to create corridors to be used for performance requirements of a pedestrian dummy. Iversson et al. created force-deflection corridors for the thigh and the leg from the results of latero-medial dynamic three-point bending tests of horizontally placed specimens. As they conducted three-point bending tests by cutting the proximal and distal ends of the flesh, the flesh was sagged due to gravity and the dissection of the flesh at the proximal and distal ends. For this reason, the flesh thickness was thinner than that of a pedestrian involved in a real-world pedestrian accident. However, the effect of the flesh thickness on the force-deflection curve of three-point bending is unknown. In addition, the effect of the flesh thickness on the injury measures in a full-scale car-pedestrian impact is also unknown. The goal of this study is to develop new corridors of the thigh and the leg by considering the influence of the sagging flesh.

METHOD

In a three-point bending test, the force from the indentor is applied to the bone via the flesh. In this study, we assumed that the flesh thickness does not influence the magnitude of the force at the initiation of the major bone deflection.

Assumption for Development of New Thigh/leg corridors

Due to the significant difference in stiffness between the bone and the flesh, it can be assumed that the flesh would bottom out upon the initiation of major bone deflection. In this study, that the following assumptions were made:

1. Force-deflection curves of the thigh and the leg can be divided into two regions using the magnitude of the impactor force at the initiation of the major bone deflection (hereafter called F_{bone}). The stiffness of the initial toe region is primarily determined by the flesh, and the stiffness of the subsequent region is primarily determined by the combination of the bone and the bottomed out flesh.
2. F_{bone} is independent of the flesh thickness.

When the force-defection curves of the three-point bending with several different flesh thicknesses are shifted in deflection by aligning the deflection at F_{bone}, the force-deflection curves would be similar after the initiation of major bone deflection (Figure 1). If this assumption applies, the corridors can be developed by using the region after reaching F_{bone}. The assumptions of this study were validated using the following steps:

1. F_{bone} of the thigh and the leg is calculated by means of FE simulations in dynamic three-point bending using the thigh and leg models with the flesh thickness varied to confirm that F_{bone} is not influenced by the flesh thickness.
2. F_{bone} calculated from the simulations was compared with the PMHS test results.

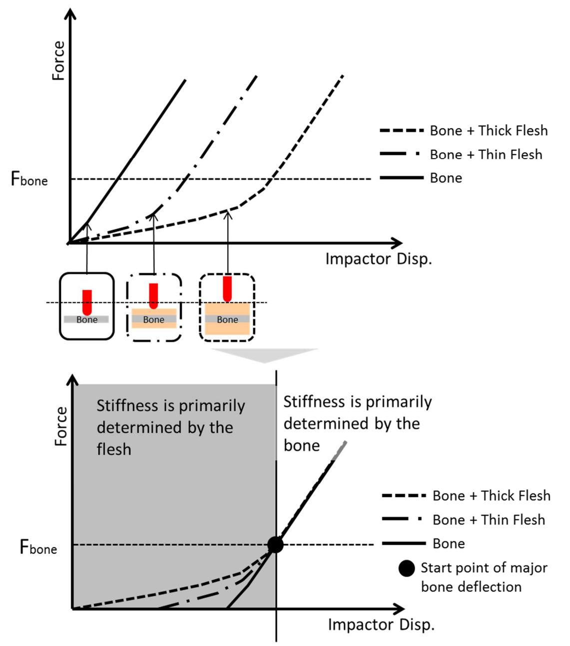


Figure 1. Assumption for Creating Thigh and Leg Corridors.

Validation of Assumption by Simulations

In order to validate the assumption shown in Figure 1, three-point bending simulations of the thigh and the leg were conducted by using the FE models for the thigh and the leg developed by Takahashi et al. [3][4]. These models were extensively validated against experimental data. The thigh and the leg model were subjected to latero-medial three-point bending at the deflection rate of 1.5 m/s. Both sides of the bones were fixed to the roller jig. The bending span length was set at 404 mm for the thigh test and 334 mm for the leg test to simulate the PMHS tests presented by Ivarsson et al. [2]. The simulations were conducted with the flesh thickness varied to investigate the influence of the flesh thickness on the stiffness before/after the initiation of the major bone deflection. The flesh thickness of the middle points of the thigh and the leg was defined as the distances from the surface of the skin to that of the femur/tibia on the impact side. In this study, two models (Component Model A and B) were developed by modifying the flesh thicknesses of the models developed by Takahashi et al. In order to investigate the effect of the flesh thickness, the material properties and the dimension of the bones were the same in the two models.

Component Model A : Flesh thicknesses of the thigh and the leg were the same as those of the original models. The flesh thicknesses of the middle points of the thigh and the leg were 46 mm and 43 mm, respectively.

Component Model B : The average thicknesses of the sagged specimens were estimated from the PMHS test results conducted by Ivarsson et al. (Thigh: 31 mm, Leg: 32 mm) [2].

F_{bone} was defined by using following steps (Figure 2):

Step 1. The flesh thickness time histories (hereafter called Curve 1) were obtained from the FE simulations. The regression lines (hereafter called Line 1) were determined by using the initial liner region of the flesh thickness time history.

Step 2. The timing of the diverging point between Curve 1 and Line 1 (hereafter called T_{bone}) was determined. The diverging point was defined as the point that the difference between Curve 1 and Line 1 in thickness direction was over the peak difference in initial linear phase.

Step 3. F_{bone} defined as the force at the timing of T_{bone} was determined from the force time histories of the three-point bending simulations.

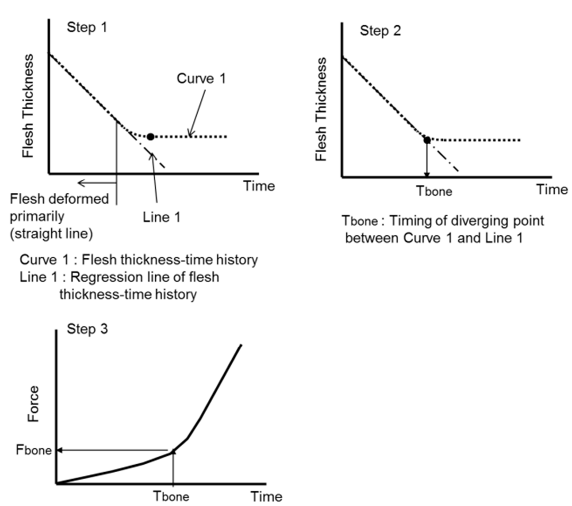


Figure 2. Method for calculation of F_{bone}

Validation of F_{bone} from the simulations against PMHS Test Results

Since F_{bone} from above was calculated by using only the results from the three-point bending simulations of the thigh and the leg, it is necessary to confirm that F_{bone} determined from the simulations is consistent with the PMHS test results conducted by Ivarsson et al. However, the flesh thickness time history in the PMHS test was unknown. If F_{bone} from the simulations is appropriate to determine the initiation of the major bone deflection, the impactor force from the PMHS tests at the timing when the flesh deflection (impactor displacement) equals to the flesh thickness should be not be greater than F_{bone}, because the major bone deflection should occur before F_{bone}. The impactor force from the PMHS tests at the timing when the flesh deflection equals the flesh thickness was compared with F_{bone} determined from the results of the three-point bending simulations.

Influence of Thigh/Leg Flesh Thickness on Injury Measure

It is necessary to investigate the influence of the flesh thickness on injury measures in full-scale car-pedestrian impact simulations in addition to component simulations in order to make sure that the corridors with the initial toe region eliminated to define requirements that ensure that the dummy is capable of predicting peak values of injury measures comparable to those of a human. Full-car models were made to collide with the full-body human FE model at 40 km/h. The human FE model developed and validated by Takahashi et al. was used. This human model represents the 50^{th} percentile male size. In addition, the lower limb of this model was extensively validated. The flesh thickness of the original model representing a pedestrian in a standing position should be the maximum without the influence from the gravity. In contrast, the flesh thickness in the condition of the three-point bending test should be the minimum due to the gravity. For this reason, three different flesh thicknesses (Whole-Body Model A, B and C) were used as shown below.

Whole-Body Model A : Flesh thicknesses of the thigh and the leg were not changed from the original model. The flesh thicknesses of the middle points of the thigh and the leg were 46 mm and 43 mm, respectively.

Whole-Body Model B : The flesh thicknesses of the thigh and the leg were the same as those of Component Model B that represented the flesh thicknesses of the specimens used in the PMHS tests. The flesh thicknesses of the middle points of the thigh and the leg were 31 mm and 32 mm, respectively.

Whole-Body Model C : The flesh thicknesses of the thigh and the leg were the averages of Whole-Body Model A and B. The flesh thicknesses of the middle points of the thigh and the leg were 39 mm and 38 mm, respectively.

The maximum values of the tibia and femur bending moments measured at multiple locations were compared (Figure 3). The dimensions of the front-end of the cars affect the injury measures, because the input conditions from a car to a pedestrian are significantly influenced by the dimension of a car. Thus, 6 car models (4 sedan, 1 SUV, 1 Minivan) were used in this study (Figure 3). The car models represented all the front-end components relevant to interaction with a pedestrian up to the peak of the femur and tibia bending moments. The total mass of the car models was adjusted with a lumped mass rigidly connected to the rear section of the car model. The full-body human FE models were collided with the center of these car models laterally from the left.

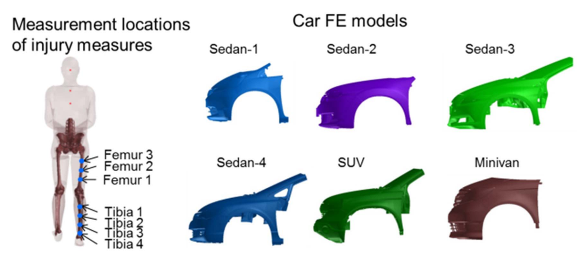


Figure 3. Measurement locations of tibia and femur bending moments and car FE models used for full-scale impact simulations.

Development of Modified Thigh/Leg Corridors

The corridors for the thigh and the leg were created by using the data from the three-point bending PMHS tests conducted by Ivarsson et al. and F_{bone}

determined in this study. Force-deflection curves of the individual test data were shifted in deflection so that the deflection at F_{bone} equals to zero. The leg corridor used in the current SAE J2782 was created by considering 1SD of both the force and the deflection [5]. However, in the three-point bending tests, the displacement time history of the ram was controlled by a servo-hydrodynamic test machine. Therefore, the deflection can be interpreted as an input, not a response. For this reason, it was decided to incorporate variability (1SD) of the force only when determining the upper and lower limits of the corridor.

RESULTS

Validation of Assumption by Simulations

Figure 4a shows the time histories of the impactor force and the flesh thickness from the three-point bending simulations of the thigh and the leg. As the results of the simulations of the thigh, F_{bone} of Component Model A and B were 1917 N and 2161 N, respectively. There results were rounded and F_{bone} was defined at 2000 N. In the leg simulations, F_{bone} of Component Model A and B were 1559 N and 1311 N, respectively. There results were rounded and F_{bone} was defined at 1500 N.

Validation of F_{bone} from the simulations against PMHS Test Results

Figure 4b shows the forces for the 5 thighs and 11 legs from the PMHS test results at timing when the deflection (impactor displacement) equals to the flesh thickness. As the results of the PMHS tests, the forces of the thigh and the leg from the PMHS test results at the timing when the deflection equals to the flesh thickness were not the same. The average force of the thigh from PMHS tests was not greater than 2000 N. Similarly, the test results of the leg showed that the average force of the leg from the PMHS tests were not more than 1500 N.

Influence of Thigh/Leg Flesh Thickness on Injury Measure

Figure 4c shows the femur and tibia bending moment time histories from the car-pedestrian collision simulations. Although the femur and tibia bending moments were measured at three and four locations, respectively, the results for the location that showed the maximum moment were only shown. In all car models, the femur and tibia bending moment time histories showed that the peak values and their timings were almost the same for the three models with different flesh thicknesses.

Figure 4 a

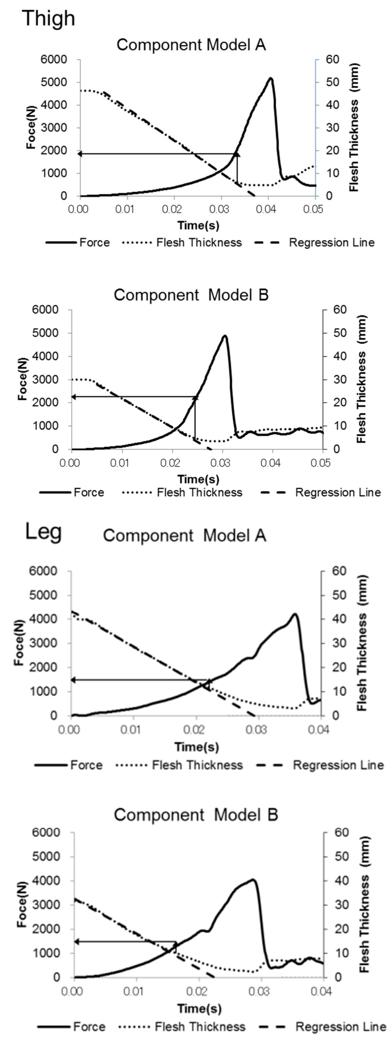


Figure 4 b

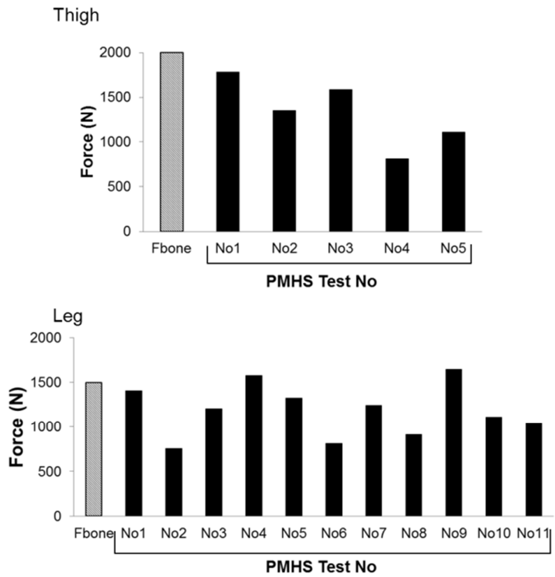
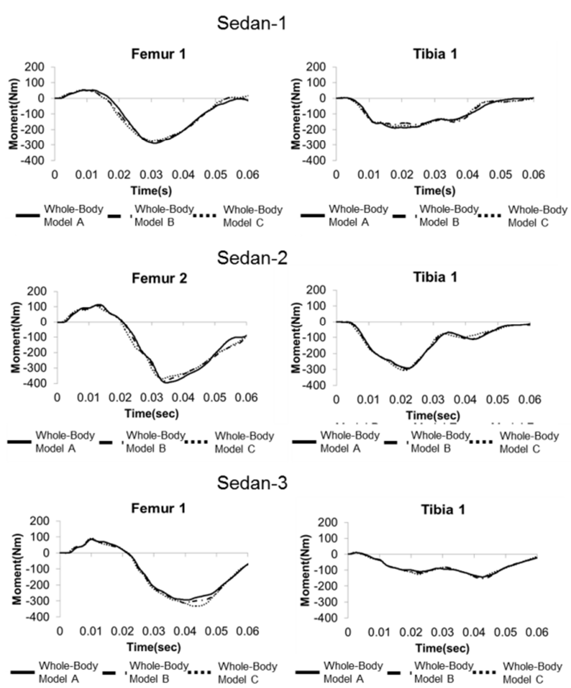


Figure 4 c



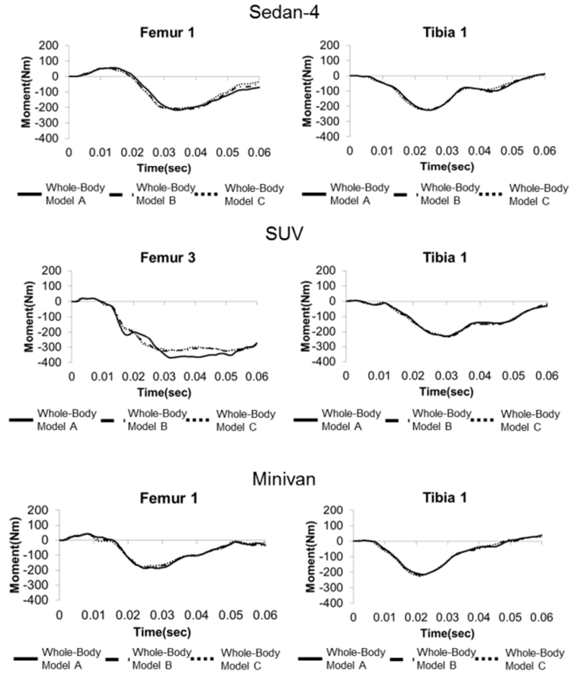


Figure 4. Results from calculation and validation of F_{bone}

Development of Modified Thigh/Leg Corridor

Considering the results from the FE simulations and the PMHS tests, the force-deflection corridors were developed from PMHS tests data by using 2000 N (thigh) and 1500 N (leg) for F_{bone}. Figure 5 shows the force-deflection corridors of the thigh and the leg in three-point bending developed in this study.

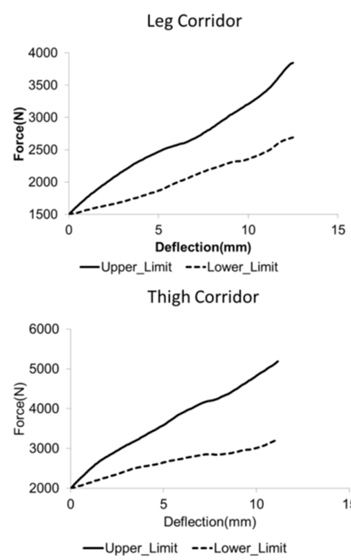


Figure 5 Force-deflection corridors of the thigh and the leg

DISCUSSION

In this study, the thigh and the leg corridors were created by eliminating the initial toe region area due to the lack of data for the flesh thickness before sagging. The average slope of the corridors for the thigh and the leg were 200 N/mm and 125 N/mm. Kerrigan et al. conducted the femur and the tibia (without flesh) three-point bending tests with the same condition as the thigh and the leg (with flesh) [6]. The stiffness of the bare bones was calculated using the results from Kerrigan et al. . As the results from bare bone tests, the stiffness of the femur and the tibia were 300 N/mm and 185 N/mm, respectively, showing that the stiffness with flesh was lower than that of the bare bone. This suggests that the stiffness of both the bare bone and the bottomed out flesh were incorporated into the corridors developed in this study.

In the current SAE J2782, the force-deflection corridor of the leg is defined. The corridor was taken from Ivarsson et al. without eliminating the initial toe region of the force-deflection response. Figure 6a compares the corridor of the leg in this study with that in SAE J2782. The corridors in this study were created by eliminating the initial toe region, and the width of the corridor in this study was smaller than that in SAE J2782 (Figure 6). In the three-point bending in different flesh thicknesses, the stiffness above F_{bone} should not be the same, because the thickness of the flesh corresponding to F_{bone} is not the same due to the difference in initial thickness. In order to compare the stiffness above F_{bone} for different initial flesh thickness, the force-defection curves from the three-point bending simulations of the thigh and the leg using FE models with different initial flesh thickness were shifted in the direction of deflection so that the deflection at 2000 N/1500 N was zero (Figure 7). As shown in Figure 7, the stiffness above F_{bone} is almost the same. This suggests that the influence of the initial thickness of the flesh on stiffness is predominant in the initial toe region, while it is minimal in the region above F_{bone}. The fact that that only a small amount of difference was identified in the peak bending moment from the full-scale car-pedestrian impact simulations with different initial flesh thickness would also support this assumption.

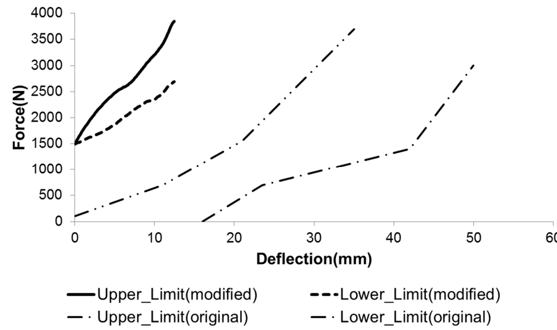


Figure 6 Comparison between original and modified leg corridors.

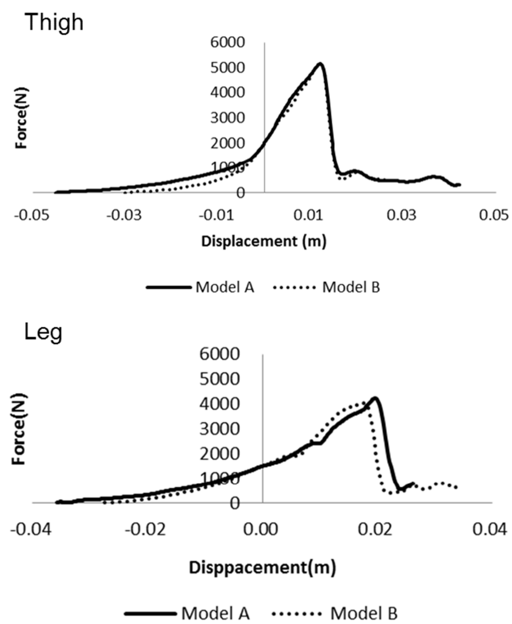


Figure 7 The force deflection curves shifted so that the deflection of the F_{bone} are equal to zero.

F_{bone} from the three-point bending simulation was larger than the forces from PMHS test results at the timing when the deflection (impactor displacement) equals to the initial sagged flesh thickness. The material property of the flesh model was defined by using the results of volunteer tests [7], not PMHS tests. It is possible that the flesh stiffness of the FE model was larger than that of PMHS due to the effects of freezing, thawing and cutting of the flesh of PMHS.Even though the influence of the initial thickness of the flesh would be limited to the initial toe region, the influence of the stiffness of the flesh would significantly affect the stiffness after the toe region..

CONCLUSIONS

In order to investigate the effect of sagged flesh of the thigh and the leg, three point bending simulations were conducted by changing the flesh thickness. As the results of simulations, it was found that the magnitude of force at the initiation of major bone deflection is not influenced by the flesh thickness (constant force magnitude at: 2000 N for the thigh, 1500 N for the leg). of the results of the full-scale car-pedestrian impact simulations with different flesh thicknesses showed that the peak values of the femur and tibia bending moments were not influenced by the initial thickness of the flesh. These results suggest that the influence of the sagged flesh can be eliminated by focusing on the stiffness after the initial toe region where the stiffness is primarily governed by the fhesh.

ACKNOWLEDGEMENTS

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