Responses of the Flexible Legform Impactor in Car Impacts

Yukou Takahashi , Iwao Imaizumi , Hiroyuki Asanuma , Miwako Ikeda *

Abstract The goal of this study was to clarify how well the impact responses of the human lower limb are represented by the Flexible Legform Impactor (FlexPLI) in terms of injury measure time histories in car impacts to obtain insight into the applicability and limitation of the legform. 18 simplified vehicle models were employed to run impact simulations at 40 km/h and compare injury measure time histories between human and FlexPLI FE models. Additional impact simulations were conducted in the same impact conditions to identify factors for the difference in the magnitude of the secondary peak of the tibia bending moment identified in the comparisons. The effects of the upper body, the material characteristics of the bone and the mass distribution between the bone and the flesh/skin were individually investigated. The results of the additional simulations showed that the mass distribution between the bone and the flesh/skin most significantly contributed to the difference in the secondary peak of the tibia bending moment, while the other two factors showed a smaller contribution compared to the mass distribution.

Keywords Pedestrians, FlexPLI, Human FE Model, Tibia Bending Moment, Secondary Peak

I. INTRODUCTION

The validity of the use of the Flexible Pedestrian Legform Impactor (FlexPLI) to evaluate pedestrian knee and leg injury probability has been investigated in past studies by analyzing the correlation of peak injury measures between a human finite element (FE) model and a FlexPLI FE model. The biofidelity of the second latest version of the FlexPLI (FlexPLI Type GT) was evaluated by Konosu et al. [1] by running impact simulations against 18 simplified vehicle models using an FE model of the FlexPLI Type GT and a human FE model. The results showed that the human model and FlexPLI model maximum tibia bending moments correlated well (R=0.90), suggesting the validity of the use of the FlexPLI. A similar computational study done by Japan Automobile Manufacturers Association (JAMA) and Japan Automobile Research Institute (JARI) [2] revealed that the correlation of the tibia bending moment between an FE model of the latest version of the FlexPLI called FlexPLI Type GTR and a human FE model was as good as that of the FlexPLI Type GT (R=0.90). Subsequently, Takahashi et al. [3] conducted a more thorough correlation study using the methodology similar to those employed by the previous two studies. It was found that the correlation of the maximum tibia fracture and ACL failure measures has been significantly improved with the FlexPLI compared to the conventional EEVC legform. These past studies clearly showed a good correlation between the maximum FlexPLI and human injury measures, justifying the use of the FlexPLI to evaluate pedestrian safety performance of vehicles by means of peak injury measures.

Based partly on such studies, the introduction of the FlexPLI into a Global Technical Regulation of the United Nations (UN GTR) has been actively discussed by the Informal Group on Phase-2 of UN GTR No.9 (IG GTR9-PH2) [4]. Introduction of the FlexPLI into GTR is expected to significantly contribute to the enhancement of pedestrian lower limb protection due to higher biofidelity along with the use of more relevant injury measures than those of the conventional EEVC legform. However, similar to the EEVC legform for which its mandatory application is limited to the lower bumper height less than 425 mm [5], the FlexPLI also has some limitations in representing human knee and leg impact responses due to the lack of upper body representation and some mechanical constraints. In addition to the biofidelity evaluation using peak injury measures, further investigation of the impact responses represented by the FlexPLI by means of injury measure time histories is expected to contribute to the clarification of the applicability and limitation of the legform, and to the identification of

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potential improvements needed for future expansion of its applicability.

The goal of this study was to clarify how well the impact responses of the human lower limb are represented by the legform by comparing injury measure time histories in car impacts to obtain insight into the applicability and limitation of the legform, and to investigate the factors for the difference in impact response to identify the direction of potential improvement of the legform performance. 18 simplified vehicle models were used to run impact simulations at 40 km/h and compare injury measure time histories between a human and FlexPLI FE models. Additional impact simulations were conducted using the FlexPLI FE model and the simplified vehicle models at 40 km/h to identify factors of the difference in the secondary peak of the tibia bending moment.

II. METHODS

Impact simulations were conducted at 40 km/h using simplified vehicle models and a FlexPLI FE model to compare time histories of the injury measures used for the FlexPLI. Since the results of the comparisons identified a significant difference of the secondary peak of the tibia bending moment between the human model and the FlexPLI model, the factors for the difference were investigated by running some additional impact simulations. Assuming that the factors include the upper part of the body, the material property of the bone in the unloading phase and the mass distribution within the lower limb, the effect of each factor was individually investigated by modifying the FlexPLI model. Since the performance of the FlexPLI has been optimized by setting the legform 50 mm higher than an actual pedestrian from the ground, the effect of the impact height was also investigated.

Comparison of Injury Measure Time Histories

Human FE Model : The human FE model used in the previous study by the authors [3] was also used in the current study (Figure 1). The model represents anthropometric and material characteristics close to an average male. The pelvis and the lower limb were modeled using shell and solid elements to accurately represent geometric and material characteristics of these body regions. Quasi-static and dynamic response characteristics of the pelvis and lower limb models were extensively validated against published experiments. The model validations included lateral compression of the pelvis in acetabulum and iliac loadings, 3-point bending of the thigh, femur, leg, tibia and fibula at multiple loading locations, tension of the individual knee ligament and 4-point valgus bending of an isolated knee joint. The upper part of the body was modeled using articulated rigid bodies with all of the seven cervical and five lumbar vertebrae individually modeled to represent flexibility of these regions. The full-body pedestrian model was validated against published full-scale car-pedestrian impact experiments in terms of trajectories of the head, top and middle of the thorax, and the pelvis, along with pelvis and lower limb injury prediction in collisions with a small sedan and a large SUV.

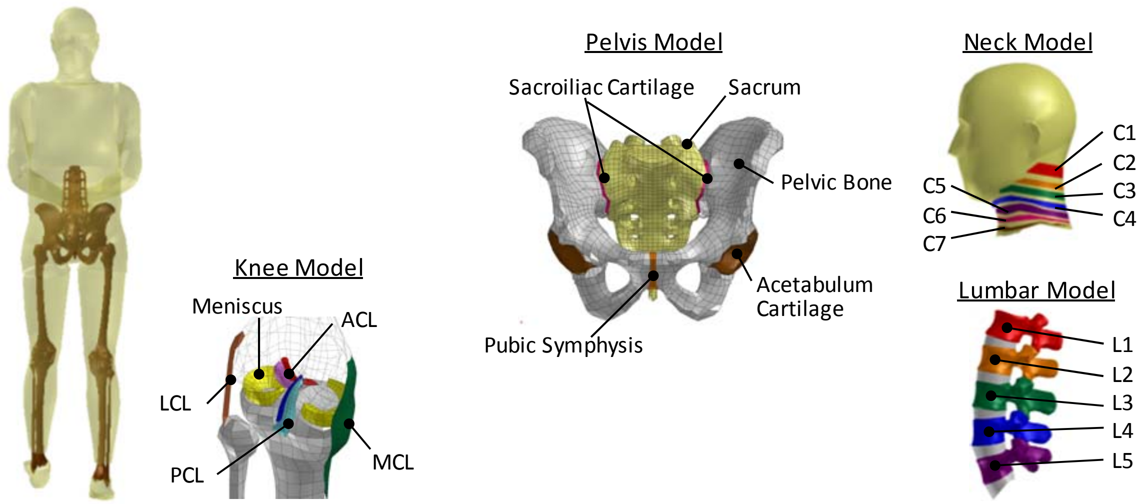

Fig. 1. Human FE model [3]

FlexPLI FE Model : In this study, the FlexPLI refers to its latest version called the FlexPLI type GTR. The FlexPLI FE model used in this study was also identical to that employed in the previous study by the authors [3]. Figure 2 shows the structure and instrumentation of the FlexPLI, along with the schematics of its FE model. The femur and tibia shafts consist of the flexible bone core made of glass fiber reinforced vinyl ester resin, the core binder, the exterior housing and the rubber buffer. Four major knee ligaments (Medial and Lateral Collateral Ligaments (MCL, LCL), Anterior and Posterior Cruciate Ligaments (ACL, PCL)) are represented by wire cables with springs on both femur and tibia sides that provide tensile stiffness of the ligaments. The strain gages affixed to the tibia bone core in four different cross-sections measure the strains that are converted to bending moments used as the tibia fracture measure. Failure of the MCL, ACL and PCL are evaluated by measuring the ligament elongation using potentiometers. The entire legform is surrounded by the Neoprene with multiple rubber layers inside. In the FE model of the FlexPLI, the bone cores, Neoprene and the rubber layer were modeled using deformable solid elements, while the core binder and exterior housing were modeled as rigid bodies. The ligament cables were modeled using bar elements that represent the compressive characteristics of the springs on both sides of the cable. The model validation included quasi-static 3-point bending of the bone core, femur, knee and tibia, assembly pendulum test, and simplified and actual vehicle impact tests, as summarized in Table I. The 3-point bending bone core certification tests are specified in the FlexPLI Technical Evaluation Group (Flex-TEG) document [6]. The 3-point bending femur, knee and tibia certification tests and the pendulum certification tests are specified in the proposal for amendment 2 of UN GTR No.9 [7].

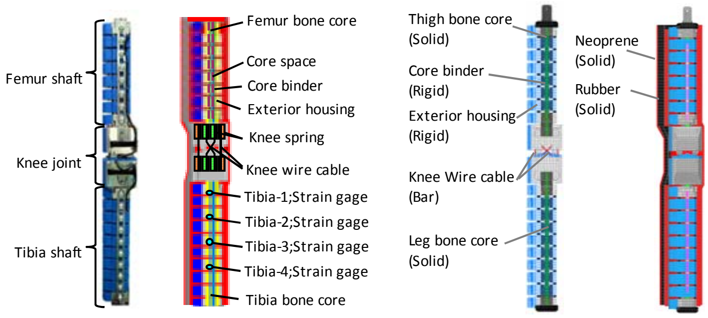

Fig. 2. Structure and instrumentation of FlexPLI and schematics of FlexPLI model [3]

TABLE I
FlexPLI MODEL VALIDATION [3]

Component	Assembly
• Quasi-static 3-point bending of bone core • 3-point bending of femur • 3-point bending of knee • 3-point bending of tibia	• Pendulum test • Simplified car test • Vehicle test

Simplified Vehicle Models : 18 simplified vehicle models used by Konosu et al. [1][8] and a subsequent study by the authors [3] were used in this study to cover various combinations of vehicle front end structure and stiffness characteristics. Figures 3 and 4 respectively show the structure of the simplified vehicle models and the definition of geometric parameters. The bonnet leading edge (BLE) was modeled using deformable shell elements representing a sheet metal structure, while the bumper (BP) and the spoiler (SP) were modeled as rigid bodies. BP, SP and the lateral ends of BLE were connected via joint elements with springs to a node with the mass of 1500 kg that represented the mass of the vehicle. The stiffness characteristics were specified by the thickness of the shell elements and the joint characteristics for BLE and BP/SP, respectively. Figure 5 shows the stiffness characteristics used for BP and SP. Table II summarizes the levels of the geometric and stiffness parameters varied among the models. Three levels were used except BLE thickness with two levels. 18 different combinations of the geometric and stiffness parameters were determined by applying L18 orthogonal array to develop vehicle models to be used to investigate the effect of each parameter with a minimal number of models. Table III summarizes the combinations of the parameters specified in the 18 models. More details of the simplified vehicle models are given by Konosu et al. [1][8].

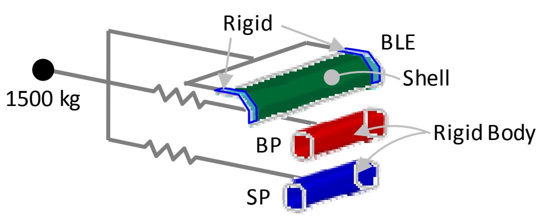

Fig. 3. Schematic of simplified vehicle model [3]

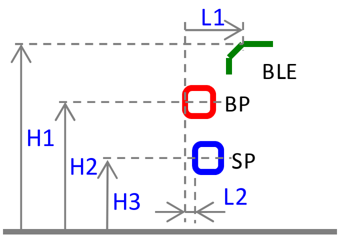

Fig. 4. Definition of geometric parameters [3]

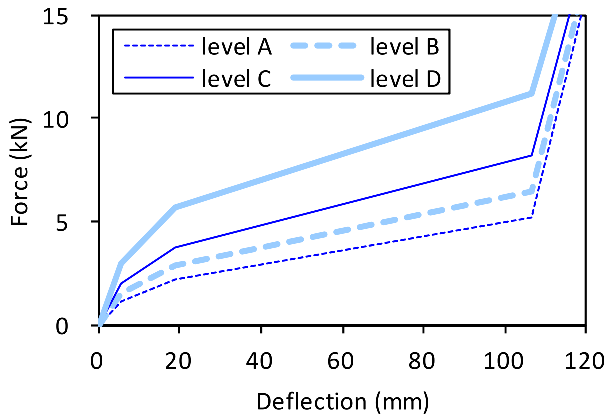

Fig. 5. Stiffness characteristics used for BP and SP [3]

TABLE II
LEVELS OF GEOMETRIC AND STIFFNESS PARAMETERS [3]

Parameter	Unit	Level 1	Level 2	Level 3
K1 (BLE thickness)	mm	0.4	0.6	
K2 (BP stiffness)	-	B	C	D
K3 (SP stiffness)	-	A	C	D
H1 (BLE height)	mm	650	700	750
H2 (BP height)	mm	450	490	530
H3 (SP height)	mm	250	270	350
L1 (BLE lead)	mm	125	200	275
L2 (SP lead)	mm	-20	0	30

TABLE III
COMBINATIONS OF GEOMETRIC AND STIFFNESS PARAMETERS [3]

ID	K1	K2	K3	H1	H2	H3	L1	L2	ID	K1	K2	K3	H1	H2	H3	L1	L2
S1	0.4	B	A	650	450	250	125	-20	S10	0.6	B	A	750	530	270	200	-20
S2	0.4	B	C	700	490	270	200	0	S11	0.6	B	C	650	450	350	275	0
S3	0.4	B	D	750	530	350	275	30	S12	0.6	B	D	700	490	250	125	30
S4	0.4	C	A	650	490	270	275	30	S13	0.6	C	A	700	530	250	275	0
S5	0.4	C	C	700	530	350	125	-20	S14	0.6	C	C	750	450	270	125	30
S6	0.4	C	D	750	450	250	200	0	S15	0.6	C	D	650	490	350	200	-20
S7	0.4	D	A	700	450	350	200	30	S16	0.6	D	A	750	490	350	125	0
S8	0.4	D	C	750	490	250	275	-20	S17	0.6	D	C	650	530	250	200	30
S9	0.4	D	D	650	530	270	125	0	S18	0.6	D	D	700	450	270	275	-20

Impact Simulations : The human FE model and the FlexPLI FE model were individually made to collide with each of the 18 simplified vehicle models at 40 km/h. Figure 6 shows the model setup. The human model was hit by the simplified vehicle model laterally from the left. For the purpose of the comparison with the FlexPLI, which is intended to be used in an upright position, the struck side (left) lower limb was positioned vertically and the contralateral lower limb was rotated about the hip joint anteriorly by 20 degrees. While the stationary human model was hit by a moving vehicle model, the FlexPLI model was propelled into the stationary vehicle model. In this case, the node representing a vehicle mass was fixed to the inertial space for all degrees of freedom. The time histories of the injury measures were compared between the human and FlexPLI models for the MCL elongation and the tibia bending moment. As shown in Figure 7, the tibia bending moment of the human model was measured at the same locations as those used for the FlexPLI.

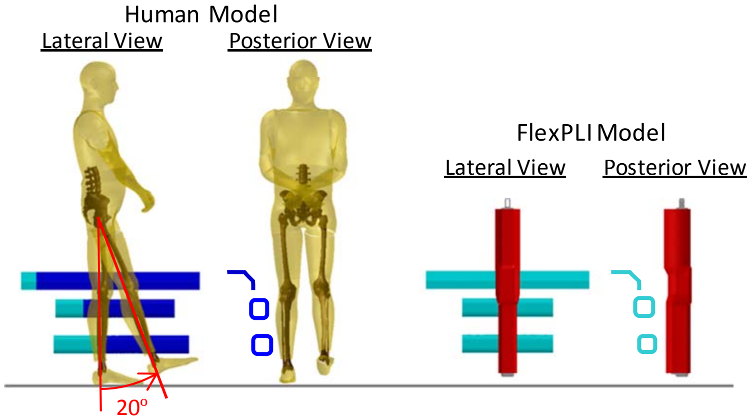

Fig. 6. Model setup

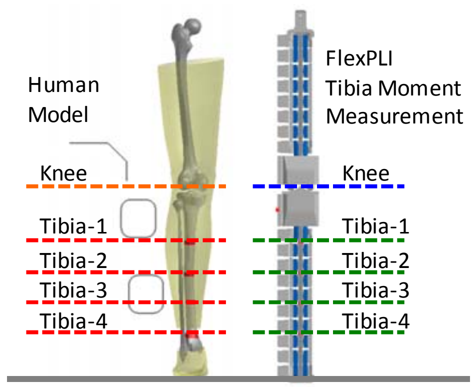

Fig. 7. Measurement locations for tibia moment [3]

Factors of Difference in Tibia Bending Moment

Since the comparisons of injury measure time histories showed that the FlexPLI model tended to overestimate the secondary peak of the tibia bending moment compared to the human model, the factors for this difference were further investigated by running some additional impact simulations to obtain an insight into potential improvements to the legform for expanding its applicability. The additional impact simulations were conducted using the following 7 out of 18 simplified vehicle models for which the FlexPLI model exhibited a significantly higher secondary peak of the tibia bending moment compared to the human model: S1, S5, S11, S14, S15, S16 and S18 (see Table III). The following three potential contributors were investigated: 1. Lack of the upper body representation, 2. Lack of hysteresis of the bone material property and 3. Difference in the mass distribution of the lower limb. Due to the lack of the representation of the upper body with the FlexPLI, the top end of the femur can move freely without any mechanical constraint, while in a human body the femoral head is engaged by the acetabulum to form a hip joint. This difference may lead to a larger bending of the femur toward the vehicle and a larger secondary peak of the tibia moment. A preliminary study on the FlexPLI bone core material property had shown its highly elastic nature with no hysteresis. In contrast, human bones are a viscoelastic material with some energy dissipation in an unloading phase. This difference could be another factor of the difference in the secondary peak of the tibia bending moment. In addition, the mass distribution between the bone and the flesh/skin was found to be significantly different between the FlexPLI and a human lower limb. Due to the limitations coming from the fundamental design of the subsystem test procedure where a stationary vehicle is impacted by a subsystem impactor, a human-like mass distribution with the much heavier flesh/skin compared to the bone may result in an uncontrollable vibration of the flesh/skin. In order to match the overall mass with a human body, the bony part of the FlexPLI is significantly heavier than that of a human lower limb. This results in a significantly different natural frequency of the leg and the thigh, which would affect the secondary peak of the tibia bending moment. In addition to the investigation of the three potential factors, the effect of the impact height was also investigated because the FlexPLI model was found in a past study [1] to exhibit a better correlation of the peak knee and tibia injury measures with a human model when the impact location of the legform is raised.

<u>Upper Body</u> : Similar to other legforms such as the EEVC legform, the FlexPLI only represents the lower limb of a pedestrian without representing the effect of the upper body. Takahashi et al. [9] compared the knee shear displacement and the knee bending angle as a function of the impact height using a human FE model among a full-body model, a single lower limb model with the pelvis and above removed and a single lower limb model with an additional mass attached to the femoral head in a lateral impact from a simplified bumper at 40 km/h. Using an additional mass of 28 kg that approximates half of the weight of the upper body for 50th percentile male, it was found that the single lower limb model with the additional mass showed the knee bending angle as a function of the impact height much closer to that of the full-body model than the single lower limb model without an additional mass. Based on this finding, an additional mass of 28 kg was added to a node corresponding to the location of the femoral head of a human body with the same anthropometry (Figure 8) to compare the tibia bending moment time histories with those from the baseline model.

<u>Bone Material Property</u> : The FlexPLI bone core material consists of glass fiber reinforced vinyl ester resin that shows high elasticity. In order to investigate the effect of this elasticity of the bone core, the constitutive model was changed from elastic to elastic-plastic to represent a hysteresis as shown in Figure 9. The FlexPLI measures the strain of the bone core at four locations of the tibia (see Figure 7) to be converted to the bending moment. Similarly, the FlexPLI model also measures the strain on the bone core to calculate the bending moment. However, when the hysteresis of the bone core material is applied, the tibia bending moment time histories show a plateau region corresponding to the unloading phase due to the residual plastic strain. This results in a similar plateau region of a tibia bending moment time history when a simple conversion is applied from the strain to the bending moment. In order to avoid this unrealistic prediction of the tibia bending moment time histories, the section moment (moment transmitted through a section of the model) was used instead of calculating the bending moment from the strain. For comparison purposes, this was applied to all the modified FlexPLI models used for investigating the factors of the difference in the tibia bending moment.

<u>Mass Distribution</u> : Although the FlexPLI accurately represents the total mass of the lower limb, the bony part

is much heavier and the flesh/skin is much lighter than those of a human lower limb. Since the mass of the bony part determines the natural frequency of the tibia and femur for a given stiffness of the bone core, the mass distribution between the bony part and the flesh/skin was changed to match that of a human lower limb. Table IV summarizes the ratio of the mass of the flesh/skin to the mass of the bony part for the FlexPLI model and a human model. The geometric and inertial properties of the FlexPLI leg were designed to represent the knee and below, including the foot. Due to the lack of the representation of the foot bones with the human model used in the current study, the mass of the bony part and the flesh/skin above and below the knee for a human lower limb were determined by referring to the human FE model developed by the Global Human Body Models Consortium [10]. In order to investigate the effect of the mass distribution, the mass distribution of the FlexPLI model was changed to that of a human lower limb. The Neoprene and the rubber layers were chosen for the flesh/skin and the remaining parts were defined as the bony part because the Neoprene and the rubber layers are the only parts that are not mechanically affixed to the bone core.

Impact Height : The impact height of the FlexPLI is to be set at 50 mm above the normal standing position to obtain a better correlation of peak injury measures with a human lower limb [1]. Although it was found from the results of the current study that the change of the mass distribution most significantly affects the oscillation of the tibia bending moment time history and thus provides the time history closest to that of the human model than the other two factors, the difference in the impact height would alter loading mechanisms to the lower limb. The inertia of the upper body would limit the motion of the femoral head and thus induce sliding of the lower limb over the front-end of a vehicle. Assuming that the lack of the upper body changes the wrap around motion of the lower limb, the impact height was lowered by 50 mm (to the same height as the human model) and the upper body was represented by the 28 kg additional mass set at the location corresponding to the human femoral head. These changes were applied to the FlexPLI model with the mass distribution changed to that of a human lower limb to see if the legform, with the modification of the mass distribution and the compensation for the upper body, provides a more biofidelic tibia bending moment time history with a normal standing height, which would expand its applicability.

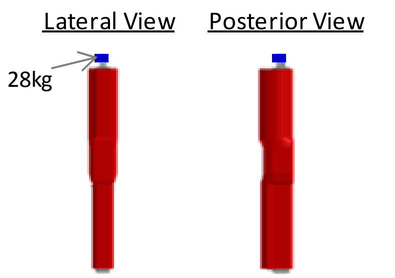

Fig. 8. Upper body representation

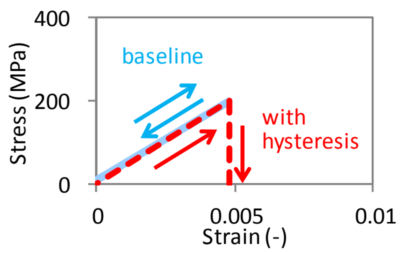

Fig. 9. Representation of hysteresis

TABLE IV
MASS RATIO ABOVE AND BELOW KNEE

Region	Mass Ratio (Flesh&Skin/Bone)	
	Human	FlexPLI
Above Knee	8.755	0.520
Below Knee	3.230	0.220

III. RESULTS

Comparison of Injury Measure Time Histories

Figures 10 and 11 show the comparisons of the MCL elongation and the tibia bending moment time histories between the human and FlexPLI models, respectively. For each of the human and FlexPLI models, the tibia bending moment time histories were compared using the results at the cross-section providing the maximum value of the four locations. Although the profiles of the MCL elongation time histories differed significantly between the two models after the peaks probably due to the lack of the representation of the hysteresis with the FlexPLI ligaments, the trend of the peak timing and values was generally similar between the two models. In contrast, the FlexPLI exhibited a more oscillatory tibia bending moment time histories compared to those from the human model. The current study focused on this difference and conducted some additional impact simulations to clarify the factors accounting for this difference. The following analyses were conducted for cases S1, S5, S11, S14, S15, S16 and S18 because for these cases both the human and FlexPLI models showed the secondary peak of the tibia moment with that of the FlexPLI model being much more exaggerated.

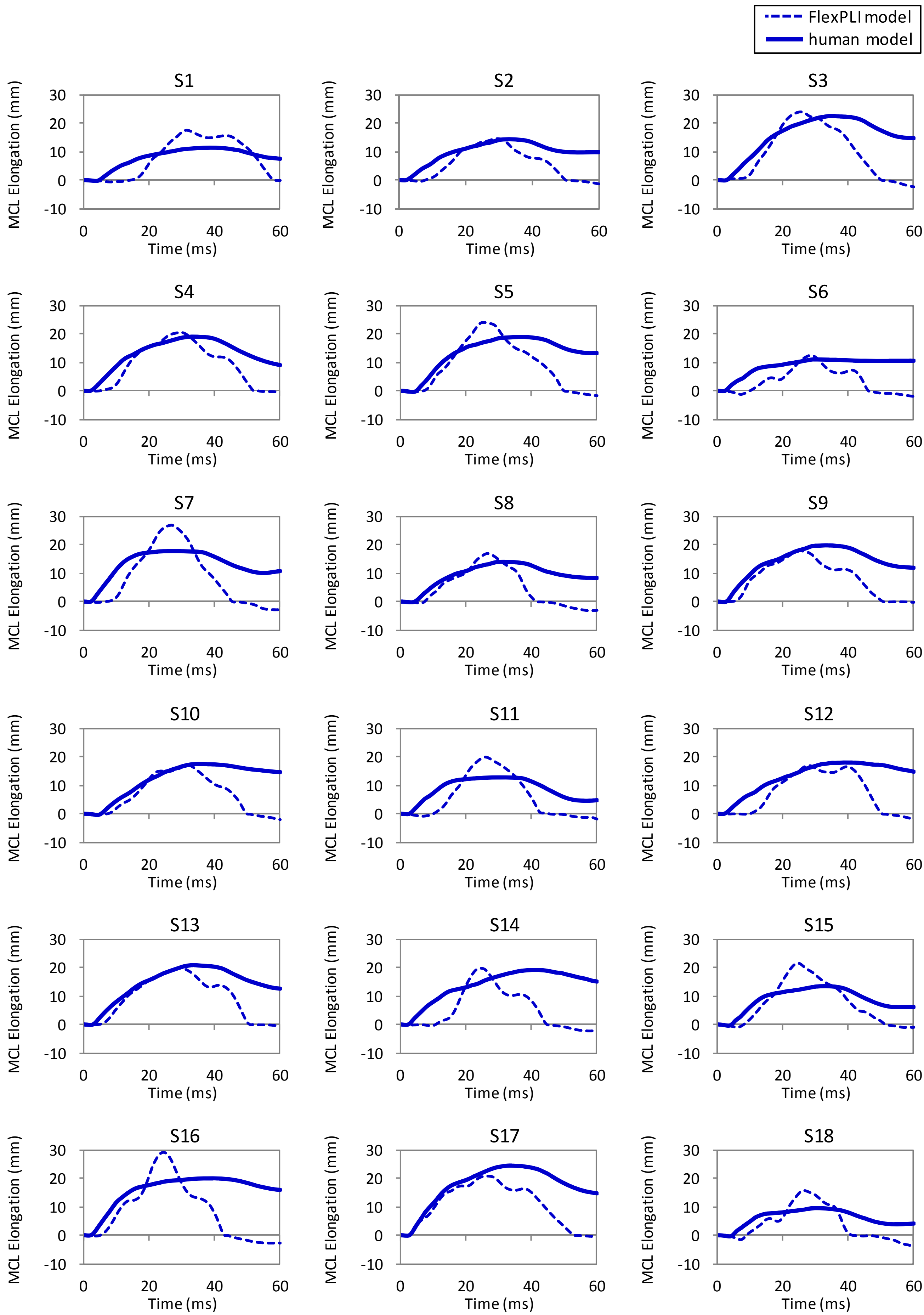

Fig. 10. Comparison of MCL elongation time histories between human and FlexPLI models

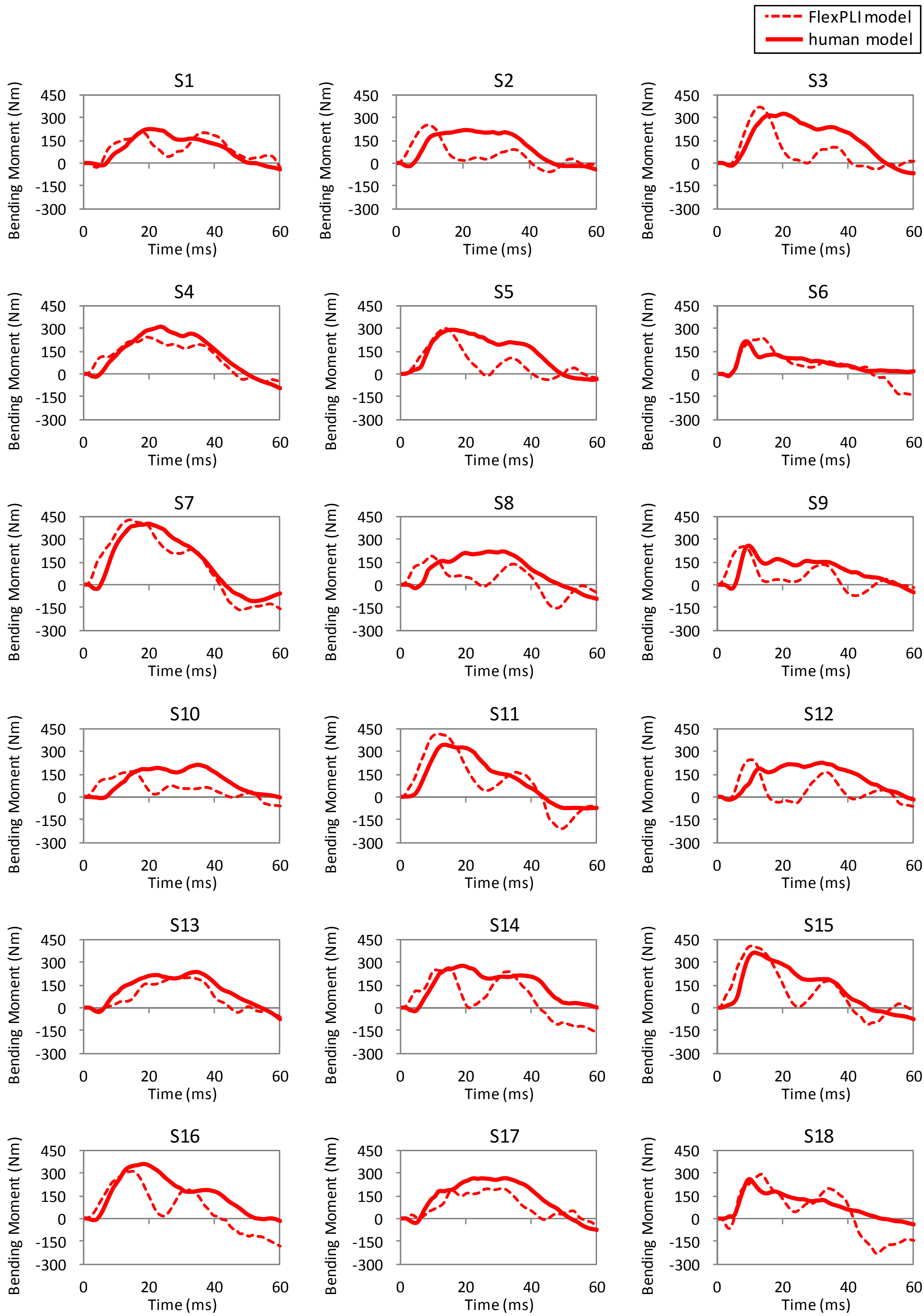

Fig. 11. Comparison of tibia bending moment time histories between human and FlexPLI models

Factors of Difference in Tibia Bending Moment

In the additional impact simulations for clarifying the factors of the difference in the tibia bending moment time histories, the tibia bending moment of the FlexPLI was directly measured by the section moment, as opposed to the conversion of the bone core strain to the bending moment to be done in actual testing. For this reason, the tibia bending moment time histories obtained in the two methods were compared for the three out of seven vehicle models chosen for the additional analyses that provided the largest difference in the peak tibia moment (Figure 12). Although the magnitude was slightly different due to the use of the calibration values based on the beam theory, the trend was exactly the same between the two methods.

Figure 13 compares the tibia bending moment time histories for the seven vehicle models between the baseline FlexPLI model and the three modified FlexPLI models addressing the three potential contributors to the difference in the secondary peak of the tibia bending moment time histories: Upper Body, Bone Material Property and Mass Distribution. The results from the human model were also superimposed. As for the FlexPLI model with the additional mass representing the upper body (Upper Body), the phase of the secondary peak was shifted and the magnitude of the secondary peak was generally diminished. However, the oscillatory nature of the wave profile was the same as the baseline. When the hysteresis was applied to the bone core material property (Bone Material Property), the secondary peak magnitude was significantly reduced for approximately half of the cases. However, the large residual plastic strain (see Figure 9) resulted in a steeper drop in the bending moment during an unloading phase, as clearly illustrated in Figure 13. The FlexPLI model with the mass distribution between the bony part and the flesh/skin changed to that of the human model (Mass Distribution) yielded no significant secondary peak of the tibia bending moment and provided the wave profile closest to that of the human model among the three potential contributors investigated.

Figure 14 compares the tibia bending moment time histories between the human model, modified FlexPLI model with the mass distribution changed (Mass Distribution) and the modified FlexPLI model with the mass distribution changed, impact height lowered by 50 mm to the height of the human model (50 mm compensation not applied) and the additional mass representing the upper body applied (Impact Height). Although in some of the cases the trend of the time histories differed after approximately 40 ms between the two modified FlexPLI models, the modified FlexPLI model (Impact Height) still exhibited similar trends of the tibia bending moment time histories to those of the human model without any significant oscillatory behavior.

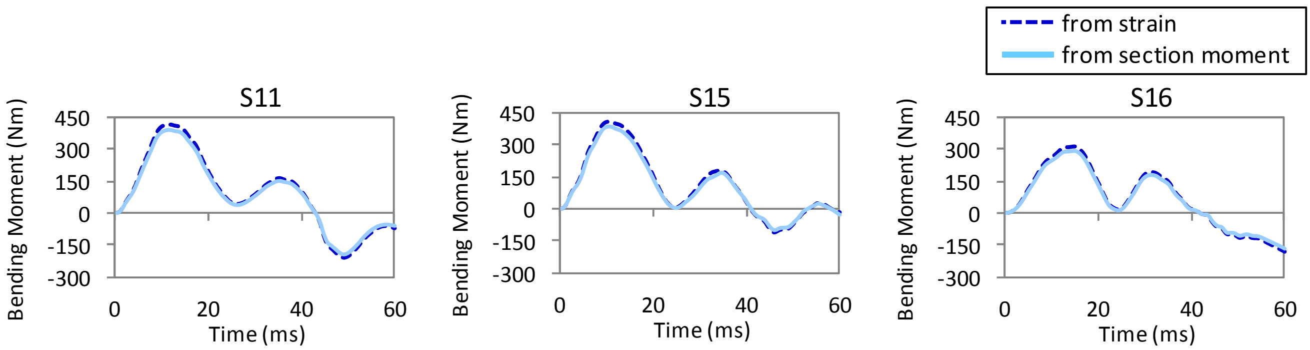

Fig. 12. Comparison of time histories of tibia bending moment from strain and section moment

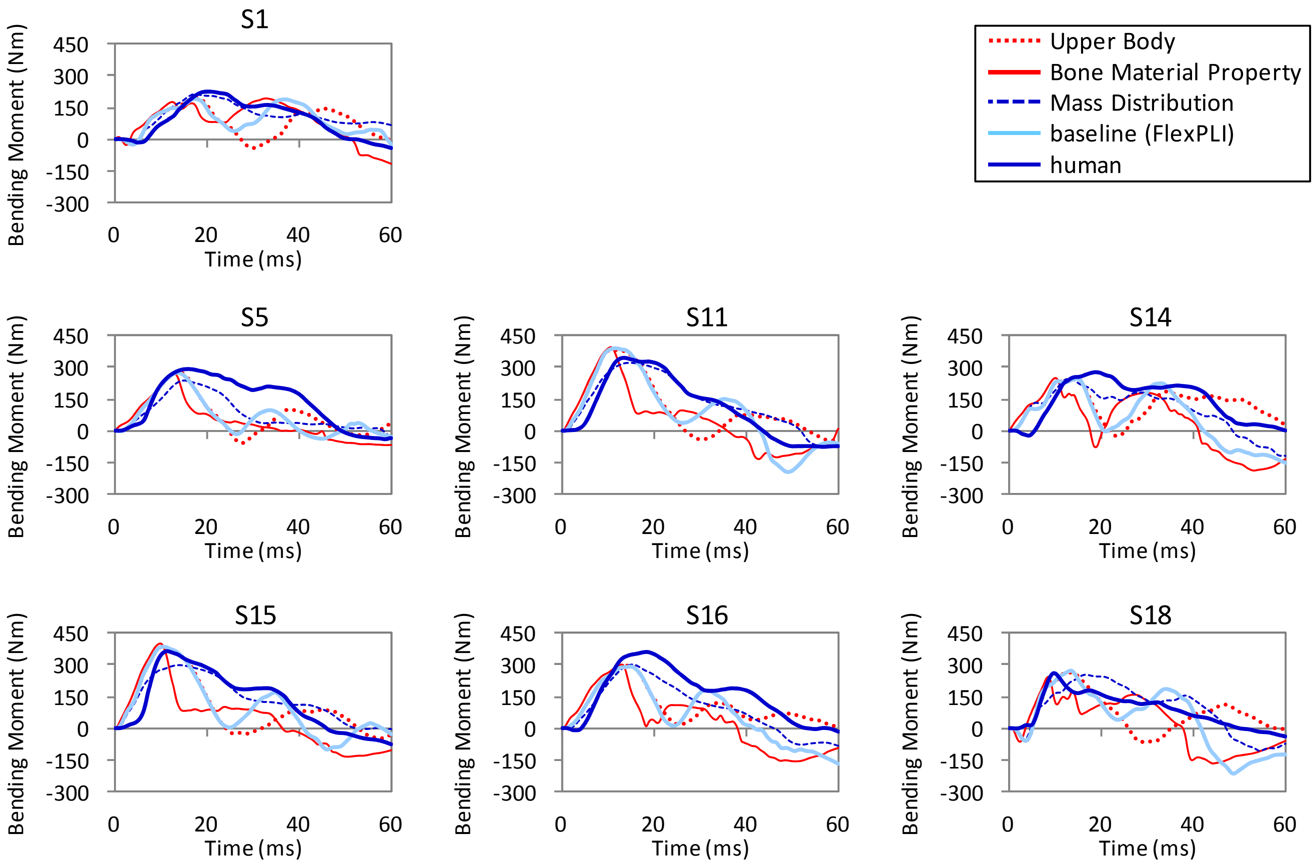

Fig. 13. Comparison of time histories of tibia bending moment between baseline FlexPLI model, human model and modified FlexPLI models (cases for Upper Body, Bone Material Property and Mass Distribution)

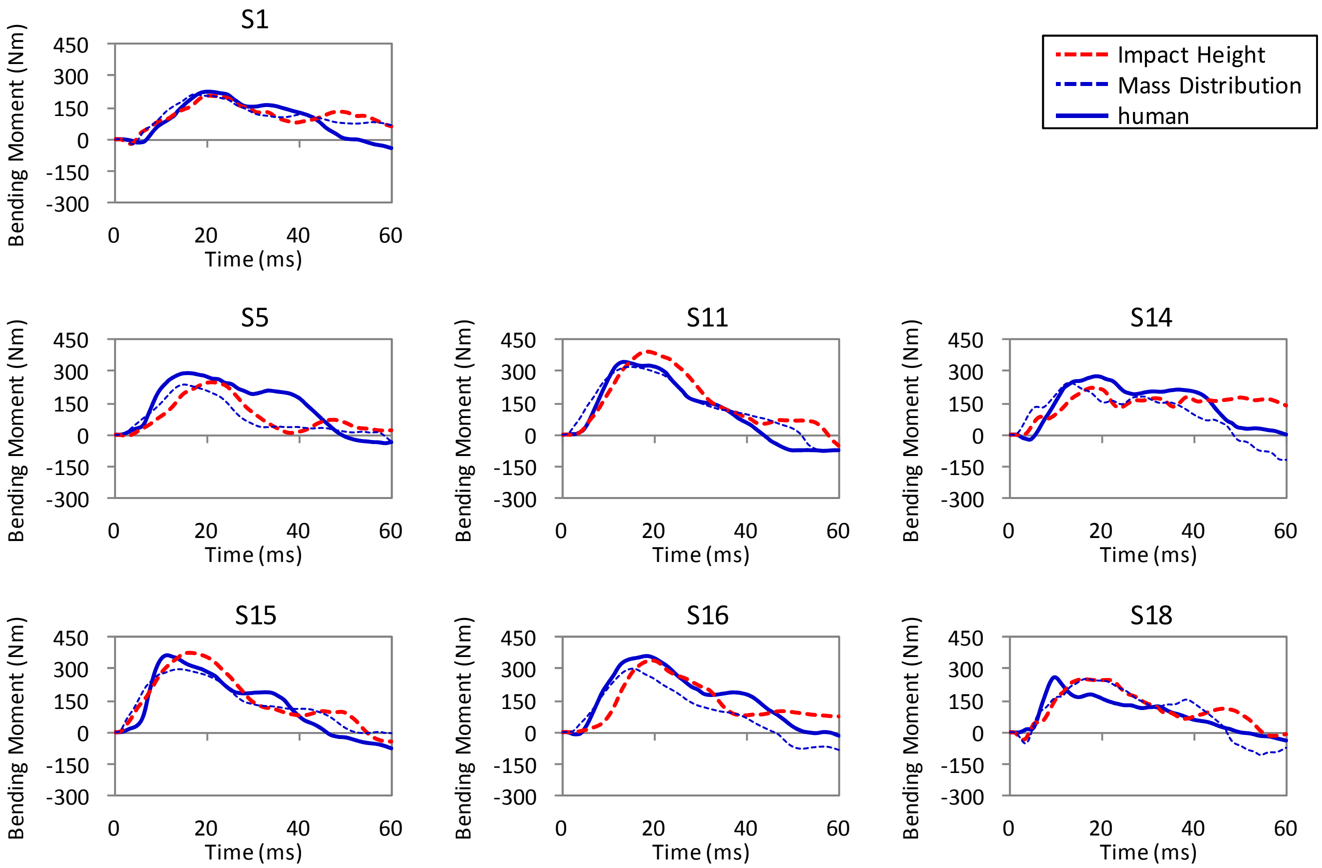

Fig. 14. Comparison of time histories of tibia bending moment between human model and modified FlexPLI models (cases for Impact Height and Mass Distribution)

IV. DISCUSSION

The significant secondary peak of the tibia bending moment observed from the impact simulations with the baseline FlexPLI model can be primarily attributed to two factors : one is the reloading of the tibia in the direction of bending towards the vehicle front due to free oscillation of the tibia, and the other is bending of the femur towards the vehicle. When the additional mass representing the upper body was applied (Upper Body), femur bending towards the vehicle would be restricted by the additional mass and thus delayed, which would correspond to the delayed phase and diminished magnitude of the secondary peak of the tibia moment. However, since the additional mass has no effect on the natural frequency of the tibia, the tibia moment time history exhibited an oscillatory nature similar to that of the baseline model. When the hysteresis of the bone core material property was applied (Bone Material Property), the significant secondary peak of the tibia bending moment was not seen for cases S5, S11, S15 and S16, for which the SP height is the largest among the vehicle models used. However, the secondary peak was evident with other cases, for which the SP in a lower position resulted in an earlier rebound of the tibia and thus an early reloading of the tibia. In addition, the steeper drop of the bending moment after the peak due to the large residual plastic strain (see Figure 9) was not seen with the human model, suggesting that the application of the hysteresis to the bone core material does not necessarily provide a more biofidelic impact response. When the mass distribution between the bony part and the flesh/skin of the FlexPLI model was changed to that of the human model (Mass Distribution), no evident secondary peak of the tibia bending moment was seen and the time history was much closer to that of the human model compared to the other two cases. Since the change of the mass distribution significantly alters the natural frequency of the tibia, this suggests that the effect of the free oscillation of the tibia due to much lower natural frequency of the tibia than that of the human is much more significant than the effect of larger femur bending towards the vehicle due to the lack of the upper body.

In a car test, the FlexPLI is to be used at the impact height raised by 50 mm relative to the normal standing position to provide a better correlation of the peak injury measures with those of a human lower limb [1]. In a pedestrian impact against a car, the motion of the pelvis is limited due to the inertia of the upper body at the early stage of the impact prior to the direct contact of the pelvis with the car. This induces the sliding motion of the lower limb over the car front, while the legform without upper body representation tends to stay at the initial height and simply wraps around the car front with minimal sliding motion. The impact height of the FlexPLI can be considered as the compensation for this kinematics difference. However, the difference in the impact height results in a different impact location of a pedestrian body against a car. For example, the impact location of the knee on the car would simply differ by 50 mm in height between an actual pedestrian and the FlexPLI. Thus, it would be preferable for the FlexPLI to be used at the same impact height as the normal standing position to overcome the limitations due to the difference in the impact height. Since the raised impact height would compensate for the lack of the upper body, the additional mass representing an upper body was added to the modified FlexPLI model with the mass distribution changed (Mass Distribution), and the impact height lowered by 50 mm to the normal standing height (Impact Height). As presented in Figure 14, the extinction of the oscillatory time history of the tibia bending moment provided by the modified FlexPLI model (Mass Distribution) was also seen with the modified FlexPLI model (Impact Height) as well, showing a similar trend of the tibia moment time history to that of the human model.

The modified FlexPLI models (Mass Distribution and Impact Height) both were found to produce the tibia bending moment time history close to that of the human model with significantly diminished oscillatory components. In order to clarify the advantage of the modified FlexPLI model (Impact Height) over the modified FlexPLI model (Mass Distribution) coming from the use of the normal standing height, the correlation of the peak injury measures between the human model and the two modified FlexPLI models were investigated for the tibia bending moment and the MCL elongation (Figure 15). As a reference, the correlation for the baseline FlexPLI model was also presented. The comparison of the correlation shows that the correlation coefficient of the tibia bending moment is significantly improved from the baseline FlexPLI model for both of the two modified FlexPLI models. However, the correlation coefficient of the MCL elongation from the modified FlexPLI model (Mass Distribution) is the same as that of the baseline FlexPLI model, while the modified FlexPLI model (Impact Height) provides a significantly higher correlation coefficient. This can be interpreted as the advantage of the use of the normal standing height over the compensation of the lack of the upper body representation by raising the legform by 50 mm. These results suggest that the application of the additional mass representing the upper body, change of the mass distribution between the bony part and the flesh/skin to that of the human body and the use of the normal standing height produce the highest correlation coefficient of the FlexPLI model peak tibia bending moment and MCL elongation with those of the human model, while minimizing the oscillation of the tibia bending moment time history that yields a significantly high secondary peak of the tibia bending moment.

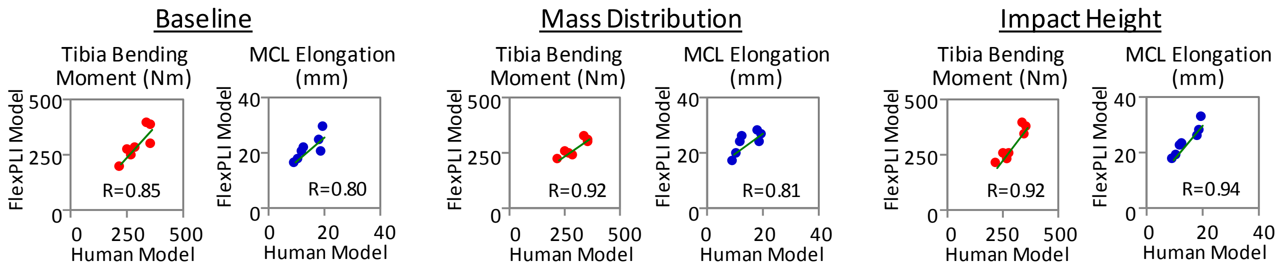

Fig. 15. Correlation of maximum values of tibia bending moment and MCL elongation between human model and modified FlexPLI models

V. CONCLUSIONS

Among the potential factors investigated in this study, the difference of the natural frequency of the FlexPLI tibia from that of the human tibia due to the difference in the mass distribution between the flesh/skin and the bony part was found to be the most significant factor of the exaggerated secondary peak of the FlexPLI tibia bending moment. The modification of the FlexPLI by means of the application of the additional mass, the change of the mass distribution of the leg to that of a human and the use of the normal standing height was found to provide a more biofidelic tibia bending moment time history, while further improving the correlation of the tibia and MCL injury measures with those of a human.

VI. REFERENCES

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