VALIDATION AND PARAMETER STUDY OF A MULTI-BODY MODEL FOR SIMULATION OF KNEE JOINT RESPONSES IN LATERAL IMPACTS REPRESENTING CAR-PEDESTRIAN ACCIDENTS: INFLUENCES OF LIGAMENT PROPERTIES AND BOUNDARY CONDITIONS ON MODEL RESPONSES

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ABSTRACT

This study focused on development of a multi-body model of a lower extremity for representation of shearing and bending responses at the knee joint in lateral impacts to a leg and its application to an analysis of the effect of boundary conditions and ligament properties on these responses. The model was validated against published results of previously conducted experiments using postmortem human subjects. To assure the reliability of this validation, reanalysis of the experimental results was done to understand how strongly these results were affected by the type of initial injury. A parameter study of the lower extremity model suggested that in the initial impact phase, shearing responses at the knee joint are unlikely to be influenced by the friction between the foot and ground. However, the effect of this friction may not be negligible if the analysis is focused on a time window longer than that associated with the initial injury occurrence.
Keywords: Car-Pedestrian Accidents, Lateral Loading, Leg Kinematics, Knee Joint, Shearing and Bending Responses, Multi-Body Model

REDUCTION OF LOWER EXTREMITY INJURIES in car-pedestrian accidents is one of the priority items in traffic safety strategy since such injuries are common and often result in a high risk of long-term or permanent disability. One of the elements of such strategy is to decrease the aggressiveness of the components of a car front. A commonly used method of evaluation of such aggressiveness is subsystem tests using a legform impactor. Standardized procedures for such tests have been proposed by the EEVC (1994, 1998) and ISO (1996). However, the exact response corridors for an assessment of the biofidelity of impactors used in such procedures are still under discussion.

One such response corridor has been recently proposed by Matsui et al. (1999) who analyzed the experimental data of Kajzer et al. (1997, 1999). These data were obtained using postmortem human subjects (PMHS) whose legs were impacted in a lateral direction to investigate the shearing and bending injury mechanisms at the knee joint. To represent loading conditions corresponding to these mechanisms, two distinct impact points were used by Kajzer et al. (1997, 1999): the leg was impacted either slightly below the knee joint (shearing tests) or at the ankle (bending tests).

The experiments by Kajzer et al. (1997, 1999) were conducted on relatively young PMHS (average age of around 55, fresh cadavers, nonhospitalized cases). However, the procedure for the treatment of PMHS utilized in these experiments imposed strict time constraints. Therefore, it was necessary to quickly adjust the experimental set-up for PMHS to the various heights of their bodies. For short subjects, one or two layers of foam were placed between the foot and the plywood plate that simulated the ground. On the other hand, for tall subjects, the foot was directly supported on this plate. It can be expected that such adjustment procedure could affect the leg boundary conditions by changing both the friction coefficient and force-penetration properties between the foot and the simulated ground. This leads to the question as to what extent such changes may influence the leg kinematics, which is of importance as neither EEVC (1994, 1998) nor ISO (1996) test procedures take into account foot-ground interactions. Conducting experiments on PMHS to answer this question could not be done because of their high cost and ethical constraints. For this reason, we decided to use multi-body modeling to evaluate the effects of foot-ground interaction and ligament stiffness on the leg kinematics in lateral impacts under the set-up used by Kajzer et al. (1997, 1999).

Most multi-body models that have been developed so far to represent responses of the lower extremities in a lateral impact were validated against experiments on relatively old PMHS. For instance, biofidelity of the models by Yang et al. (1992, 1995) was evaluated using the data of Kajzer et al. (1993) obtained on subjects whose average age was around 80 years. On the other hand, the current study focused on recent experiments by Kajzer et al. (1997, 1999) conducted on subjects whose average age was around 55 years. Therefore, it was necessary to develop a new lower extremity model to accurately represent these experiments. Complete validation of such a model using the corridors proposed by Matsui et al. (1999) could not be done. The reason is that, although they formulated the response corridors for time histories of the impact force, leg shearing displacement and the leg bending angle, they concluded that only that of the impact force is reliable. According to them, the general behavior of time histories of bending angle and shearing displacement was affected by type of the initial injury, and that the accuracy of measurements of these time histories could be compromised by the complexity of leg motion, *e.g.*, leg rotation around its longitudinal axis. Therefore, we complemented the results of Matsui et al. (1999) by analyzing the differences in the impact force, bending angle and shearing displacement-time histories related to type of initial injury. Based on this analysis, we attempted to formulate the corridors of these time histories that would minimize the effect of initial injury on such an evaluation.

Thus, this study focused on the following three goals:

1) Detailed analysis of the experimental results of Kajzer et al. (1997, 1999). This analysis focused on a formulation of the response corridors for validation of the multi-body leg model and a determination of the effects of the type of initial injury on the time histories of impact force, leg shearing displacement and leg bending angle;

2) Development of a multi-body model for representing kinematics of a leg in lateral impacts at various speeds;

3) A parameter study for an evaluation of the effect of foot-ground friction coefficient and force-elongation characteristics of the knee joint ligaments on the model responses.

METHODS

LOWER EXTREMITY MODEL The model was constructed based on the description of the experimental set-up of Kajzer et al. (1997, 1999), and was implemented using the MADYMO 5.3 multi-body code. The PMHS model consisted of the following body segments: foot, leg, thigh, and torso (Fig. 1). The torso was fully constrained. The dimensions and inertia properties of these segments represented the 50-percentile male subject. They were obtained by means of the Generator of BOdy Data (GEBOD) program.

The model of the experimental set-up included the impactor, ground, and two screws supporting the thigh (Fig. 1). In simulation of the shearing tests, the ground was represented using a plane attached to the global inertial coordinate system. However, in modeling of the bending tests, the plate simulating the ground was allowed to translate in the impact direction. The impactor mass m was 6.25 kg.

Positions of markers, impactor, and screws supporting the femur in the model represented their average values in the experiments by Kajzer et al. (1997, 1999).

Knee joint model The geometry of the knee joint was simplified in a way similar to that of Yang et al. (1995). The femoral condyles were represented using two ellipsoids, and the tibia articular surface was modeled as a plane (Fig. 2a). The distance between the centers of condyle ellipsoids was 50 mm. The tibial eminence was simplified by means of two planes, and its height was 5 mm. The stiffness of the contact between the femoral condyles and tibia articular surface was selected to be 500 N/mm, which is a value lower than the 750 N/mm used by Wismans (1980) and Yang and Kajzer (1992). The basis for this selection was that for an impact at a speed of 31 km/h, Yang and Kajzer

(1992) calculated the contact force between tibia and femur to be around 11 kN which is a rather high value.

In the current model of the lower extremity, ligaments of the knee joint were modeled using tension-only springs (MADYMO belt elements) in parallel with linear dampers which represented the viscosity of the ligaments. Five such elements were applied to simulate the anterior cruciate ligament ACL, posterior cruciate ligament PCL, anterior medial collateral ligament AMCL, posterior medial collateral ligament PMCL, and lateral collateral ligament LCL. The knee joint capsule CL was simplified by means of four belt elements. Attachment points of the ligaments and their mechanical properties were determined based on the literature (Noyes et al., 1974; Girgis et al., 1975; Trent et al., 1976; Wismans, 1980; Woo et al., 1991; Abdel-Rahman and Samir Hefzy, 1993; Soames, 1995). The lengths of untensed ligaments, i.e., lengths at which ligament forces are zero, utilized in the current study are summarized in Table 1. For all the ligaments, the initial elongation was assumed to be 5% of the untensed length. Rupture strain, stiffness and damping of the ligaments were adapted to obtain good correlation between the model responses and experimental results. Such adaptation resulted in selection of the rupture strain of 25% of the initial length for all the ligaments, which corresponds to the average of the data of Kennedy et al. (1976) and is close to the upper limit of the results by Trent et al. (1976). Furthermore, a stiffness of around 1000 N per 10% of elongation was chosen for all the ligaments (Fig. 2b). Viscous damping of the ligaments was calibrated based on the experimental data of Woo et al. (1997) who reported around a 30-60% increase in the ligament forces under rapid elongation in comparison to static loading. Such calibration yielded a damping coefficient of 250 Ns/m.

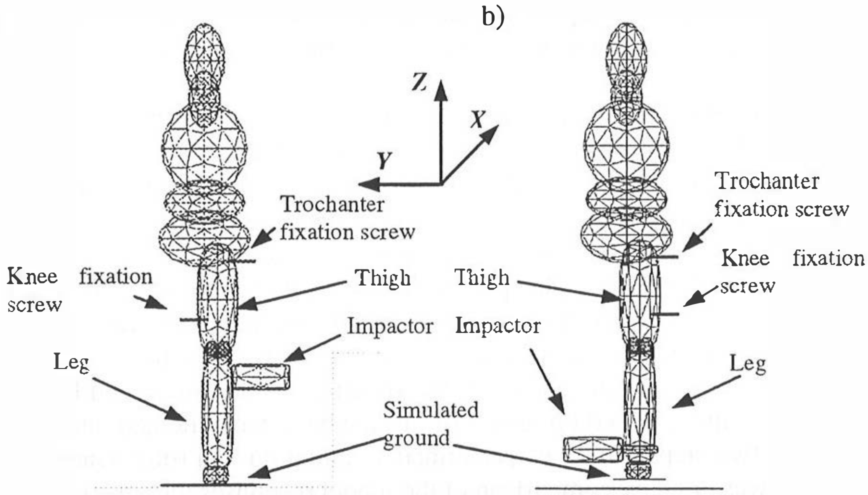

Fig. 1 Multi-body models of the lower extremity for simulating a) The shearing and b) Bending tests.

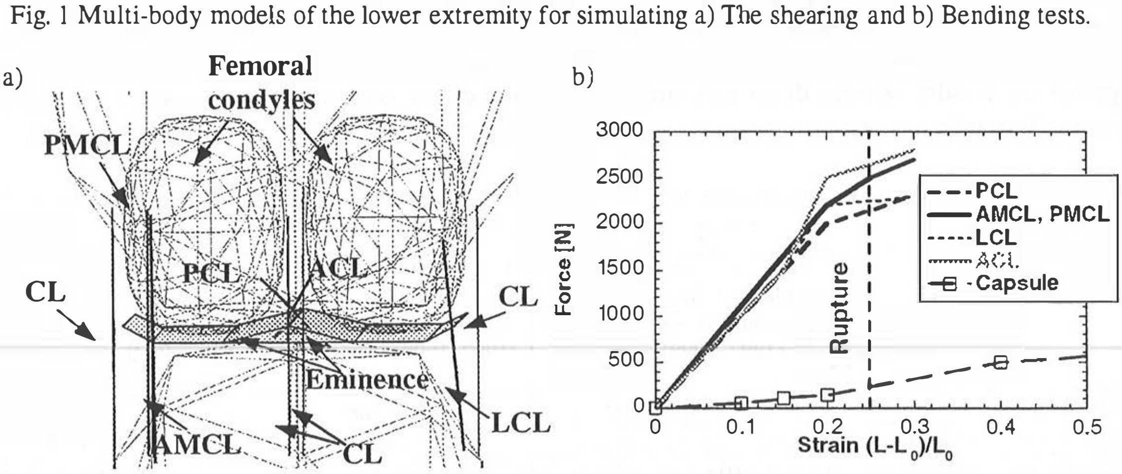

Fig. 2 a) Multi-body model of knee joint. b) Force-strain characteristics of the knee joint ligaments utilized in the current study. L and L_0 are the instantaneous and initial ligament length, respectively.

Table 1 Lengths of untensed knee joint ligaments utilized in the current study

Ligament	Length [mm]
ACL	25
PCL	35
LCL	55
AMCL	65
PMCL	65

Impactor-leg contact The force-penetration characteristic of the contact between the leg and impactor was derived based on Fig. 5 of Kajzer et al. (1997). However, an additional calibration of the characteristics presented by Kajzer et al. (1997) was necessary. The reason is that their characteristics represented the properties of the impactor padding in contact with a rigid wall, whereas in our study, the contact interactions between the impactor and leg flesh had to be modeled.

Screws supporting the femur were represented by means of two elastic springs. Their stiffness was calibrated to yield a maximum compressive force of around 1000 N in the trochanter support when simulating the shearing tests at an impactor speed of 40 km/h. This force value was experimentally obtained by Kajzer et al. (1997).

RESPONSE CORRIDORS AND MODEL VALIDATION Damages to lower extremity tissues observed in the experiments by Kajzer et al. (1997, 1999) were dependent on the impactor speed. At a speed of 40 km/h, the typical initial injury was a femur or tibia fracture, whereas at 20 km/h, ligament avulsion was the most frequent initial injury. However, the multi-body model developed in the present study was not intended to represent bone fractures. Thus, the question is how to formulate the response corridors for an evaluation of the biofidelity of this model using the experiments that resulted in such fractures.

One can derive such corridors only from those experiments in which ligament avulsion was the initial injury. However, such an approach would strongly reduce the size of the sample available for derivation, since Kajzer et al. (1997) reported that in the shearing tests at an impactor speed of 40 km/h femur fractures occurred in 7 out of 10 cases. Furthermore, the general behavior of leg response corridors by Matsui et al. (1999) suggests that variations in the time histories of leg shearing displacement and bending angle appreciably increased in the impact phase following the occurrence of the initial injury. To verify this suggestion, we performed a detailed analysis of the experimental data of Kajzer et al. (1997, 1999) and Matsui et al. (1999) with the main goal of determining differences in the time histories of the impact force, leg shearing displacement and leg bending angle related to the type of initial injury. Based on results of this analysis, we validated our multi-body model of the lower extremity in two steps. First, we performed a validation in a time window corresponding to the initial injury. This was done by comparison of the model responses with the relevant corridors obtained using results of all the experiments of Kajzer et al. (1997, 1999). Secondly, we focused on a time window 10 ms longer than that corresponding to the initial injury. Validation in this time window was done against the results of only those experiments in which either no injury occurred or ligament avulsion was the initial injury. The time windows used in the model validation are summarized in Table 2.

Table 2 Time windows used in validation of the current lower extremity model. T_A is the average time of initial injury occurrence and v is the impact speed

Test set-up	Time of occurrence of initial injury in the experiments by Kajzer et al. (1997, 1999)	Time window for validation against results of all the experiments by Kajzer et al. (1997, 1999)	Time window for validation against results of the experiments without injury or with ligament avulsion as the initial injury
Shearing; v=20 km/h	8 to 12 ms (T_A=10 ms)	15 ms	25 ms
Shearing; v=40 km/h	3 to 7 ms (T_A= 5 ms)	10 ms	20 ms
Bending; v=20 km/h	22 to 30 ms (T_A=25 ms)	25 ms	35 ms
Bending; v=40 km/h	9 to 22 ms (T_A=15 ms)	15 ms	25 ms

Leg shearing displacement and bending angle Shearing displacement of leg D was determined by means of the following formula:

$$D = Y_{P2} - Y_{P3} - g\sin(\beta), \tag{1}$$

where g is the position of marker P2 in relation to the center of the knee joint measured along the longitudinal leg axis (average g value was 50 mm), and β is the bending angle of a leg. Y_{P2} and Y_{P3} are displacements of markers P2 and P3 in the impact direction, respectively (Fig. 3). The leg bending angle was obtained as follows:

$$\beta = \arctan\left(\frac{Y_{P1} - Y_{P2}}{Z_{P2} - Z_{P1}}\right). \tag{2}$$

In the previous analysis by Matsui et al. (1999), the $g\sin(\beta)$ component was disregarded in the calculation of shearing displacement in the shearing tests.

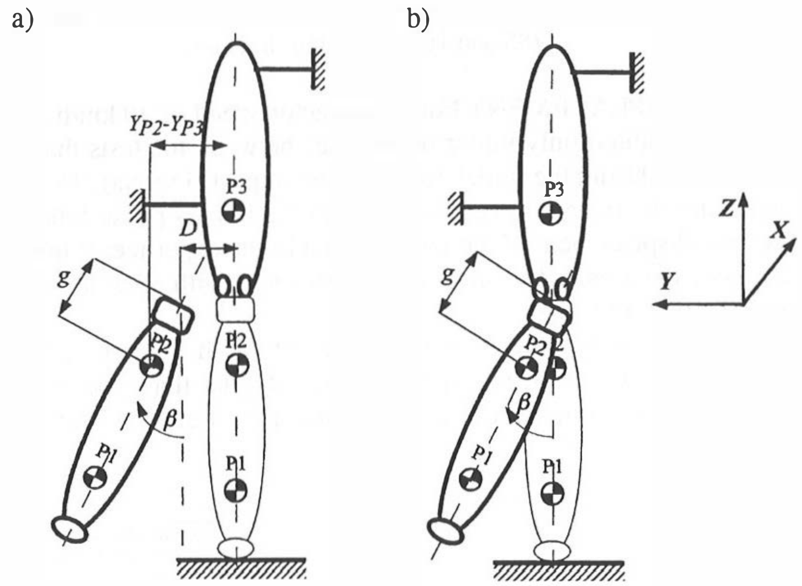

Fig. 3 Definition of a) Leg shearing displacement D; and b) Leg bending angle β.

Impact force The impact force-time histories were normalized in reference to the subjects` leg mass according to the formula based on Mertz (1984):

$$F_N = F\sqrt{\frac{M_S}{M}}, \tag{3}$$

where F_N is the normalized impact force, F is the measured impact force, M is mass of a given subject leg, and M_S is mass of the standard subject leg (i.e., 50-percentile subject).

PARAMETER STUDY OF LOWER EXTREMITY MODEL To evaluate the sensitivity of calculated leg responses to the assumed mechanical properties of the knee joint ligaments and parameters determining the boundary conditions of our model, the stiffness of the ligaments and foot-ground friction coefficient were varied from 50% to 200% of their reference values. The reference value of the friction coefficient was 0.2, and the reference force-elongation characteristics of the knee joint ligaments are summarized in Fig. 2b.

RESULTS OF ANALYSIS OF EXPERIMENTAL DATA

IMPACT FORCE Analysis of the data by Kajzer et al. (1997) suggested that in the experiments conducted using an impactor speed of 40 km/h, the time histories of impact force were independent of the type of injury during the initial 10 ms of the impact (Fig. 4). However, for time values exceeding 10 ms, these time histories differed between the experiments that resulted in initial injury caused by femur fracture and ACL avulsion (Fig. 4b). This, in turn, suggests that the impact force-time histories may vary according to injury type in the impact phase following occurrence of the initial injury.

a) b)

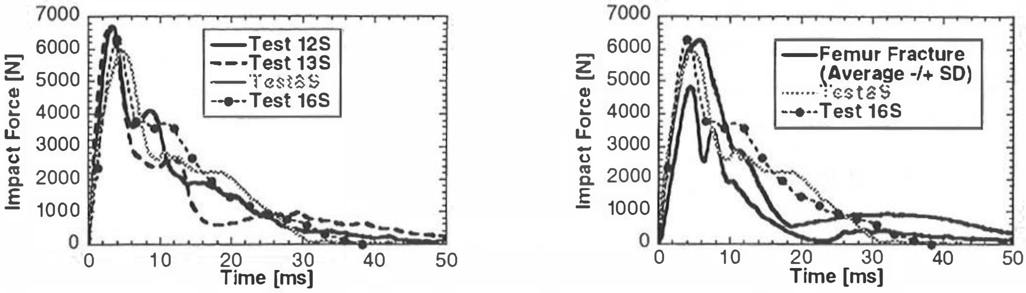

Fig. 4 Impact force-time histories in shearing tests at an impactor speed of 40 km/h. a) Initial injury caused by ligament avulsion (8S and 16S) and tibia fracture (12S and 13S). b) Initial injury caused by ligament avulsion (8S and 16S) and femur fracture.

LEG SHEARING DISPLACEMENT For an impactor speed of 40 km/h, the time histories of leg shearing displacement exhibited only minor differences between the tests that resulted in ACL/MCL avulsion and tibia fracture during the initial 10 ms of the impact (Fig. 5a). This result is similar to that obtained when analyzing the impact force. However, in the impact phase following occurrence of the initial injury, excessive displacement of the leg occurred in the experiments that yielded tibia fractures (Fig. 5a). Furthermore, the femur fractures were associated with a large variation in the shearing displacement-time histories (Fig. 5b).

In the shearing tests at an impactor speed of 20 km/h, the peak values of the shearing displacement obtained when the ACL was damaged were higher than those observed when no injury occurred (Fig. 6). This is an intuitive result as the knee joint ligaments can exert appreciable force constraining the leg motion.

a) b)

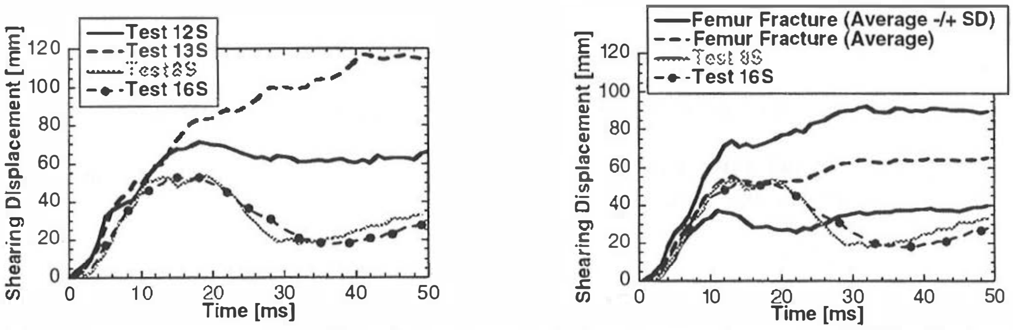

Fig. 5 Shearing displacement-time histories in shearing tests at an impactor speed of 40 km/h. a) Initial injury caused by ligament avulsion (8S and 16S) and tibia fracture (12S and 13S). b) Initial injury caused by ligament avulsion (8S and 16S) and femur fracture.

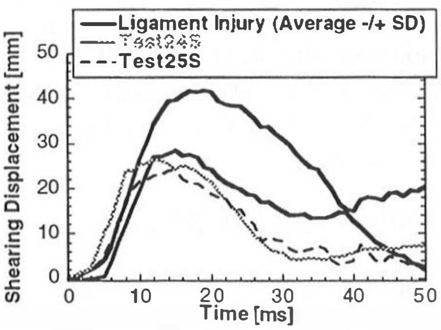

Fig. 6 Shearing displacement-time histories in shearing tests at an impactor speed of 20 km/h. Comparison of results of the tests with no injury (24S and 25S) and initial injury caused by ligament avulsion.

LEG BENDING ANGLE Shearing tests In all the experiments conducted using the shearing set-up, the leg bending angle exhibited negative values in the initial impact phase (Fig. 7, 8a). These values were up to around –5° and –10° for impactor speeds of 20 and 40 km/h, respectively. The negative bending angle could result from the rigid body rotation of the leg in the impact direction (Fig. 9).

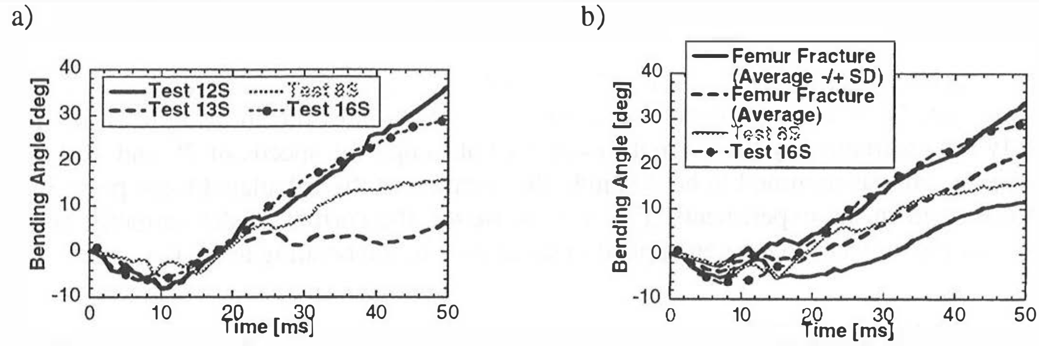

Fig. 7 Leg bending angle-time histories in shearing tests at a speed of 40 km/h. a) Initial injury caused by ligament avulsion (8S and 16S) and tibia fracture (12S and 13S). b) Initial injury caused by ligament avulsion (8S and 16S) and femur fracture.

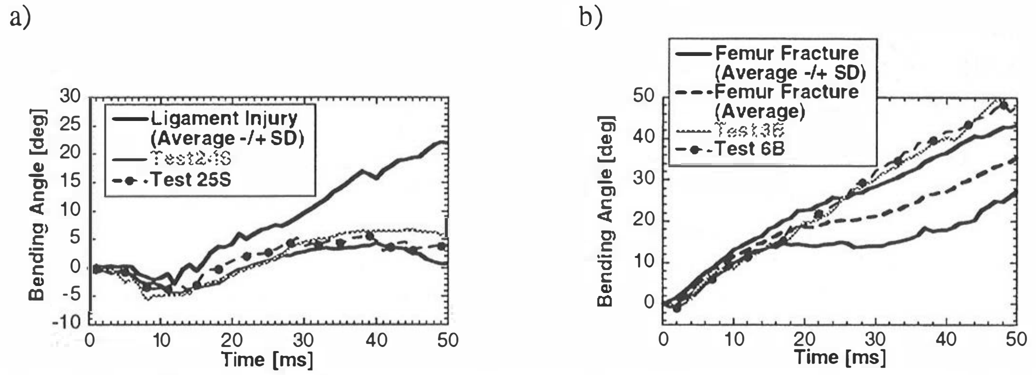

Fig. 8 a) Leg bending angle-time histories in shearing tests at a speed of 20 km/h. Comparison of results of the tests with no injury (24S and 25S) and initial injury caused by ligament avulsion (only three tests). b) Leg bending angle-time histories in bending tests at an impactor speed of 40 km/h. Comparison of results of the tests with initial injury caused by femur fracture and MCL avulsion (3B and 6B).

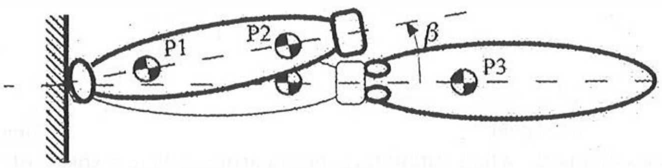

Fig. 9 Negative bending angle β.

Bending tests For majority of the experiments conducted under the bending set-up, time histories of the leg bending angle monotonically increased. These time histories seemed to be independent of injury type during the initial 15 and 20 ms at impactor speeds of 40 and 20 km/h, respectively. However, in the impact phase following the initial injury, the angle magnitude was significantly lower in the experiments that resulted in femur fractures than in those leading only to ligament avulsion (Fig. 8b).

FORMULATION OF RESPONSE CORRIDORS FOR VALIDATION OF LOWER EXTREMITY MODEL The current analysis of the results of Kajzer et al. (1997, 1999) indicated that in the majority of their experiments, the general behavior of time histories of the impact force, shearing displacement of the leg, and the leg bending angle seemed to be independent of the injury type in the initial impact phase. However, after the initial injury had occurred, these time histories exhibited a tendency to differ according to the type of injury. Thus, the current analysis of the experimental results indicates that it may not be appropriate to validate either mechanical or mathematical leg models that simulate only ligament injury against the response corridors obtained using the experiments that resulted not only in ligament but also in bone damage. Such validation requires the application of different response corridors for an evaluation of the biofidelity of the models in the impact phases preceding and following occurrence of the initial injury as proposed in the METHODS section.

MODELING RESULTS

VALIDATION OF LOWER EXTREMITY MODEL Impact force The calculated impact force-time histories nearly fit the experimental corridors. In simulation of the shearing tests, the peak force was slightly overestimated by the current model for both impactor speeds of 20 and 40 km/h. When the impactor speed was assumed to be 20 km/h, the duration of the calculated force pulse was shorter than that observed in the experiments (Fig. 11). However, the current model correctly predicted the peak values of the impact force when applied in simulation of the bending tests (Fig. 12).

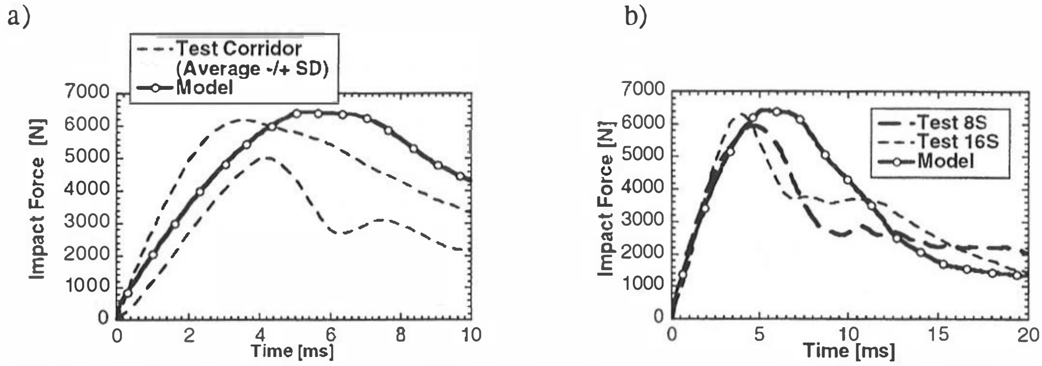

Fig. 10 Impact force calculated when simulating the shearing tests at a speed of 40 km/h. a) Comparison with the test corridor obtained using results of all the experiments by Kajzer et al. (1997); b) Comparison with results of the tests in which ligament avulsion was the initial injury (tests 8S and 16S).

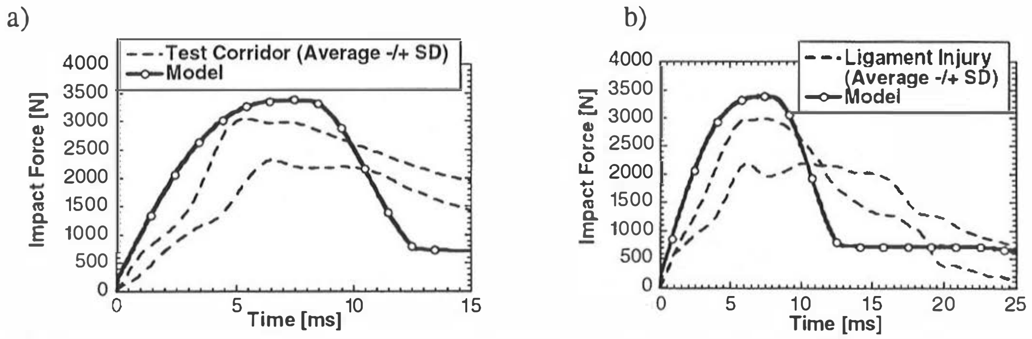

Fig. 11 Impact force calculated when simulating the shearing tests at a speed of 20 km/h. a) Comparison with the test corridor obtained using results of all the experiments by Kajzer et al. (1999); b) Comparison with the test corridor obtained using results of the experiments in which ligament avulsion was the initial injury.

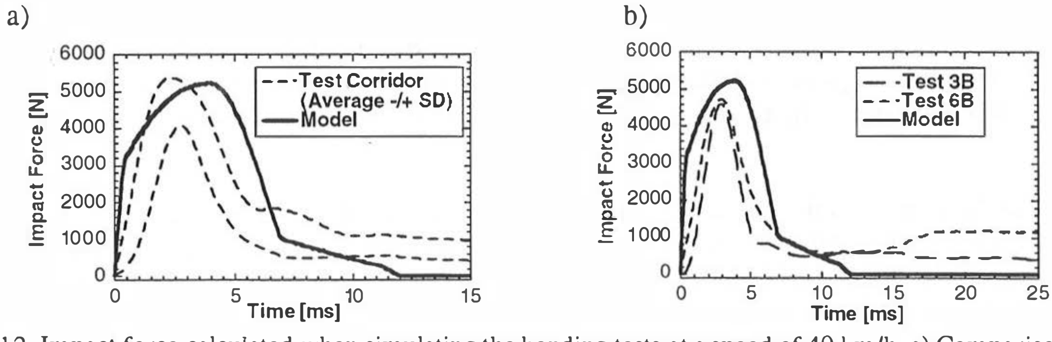

Fig. 12 Impact force calculated when simulating the bending tests at a speed of 40 km/h. a) Comparison with the test corridor obtained using results of all the experiments by Kajzer et al. (1997); b) Comparison with results of the tests in which ligament avulsion was the initial injury (tests 3B and 6B).

Leg shearing displacement Modeling the shearing tests yielded a shearing displacement of the leg inside the experimentally obtained response corridors for both impactor speeds of 20 and 40 km/h (Fig. 13 and 14).

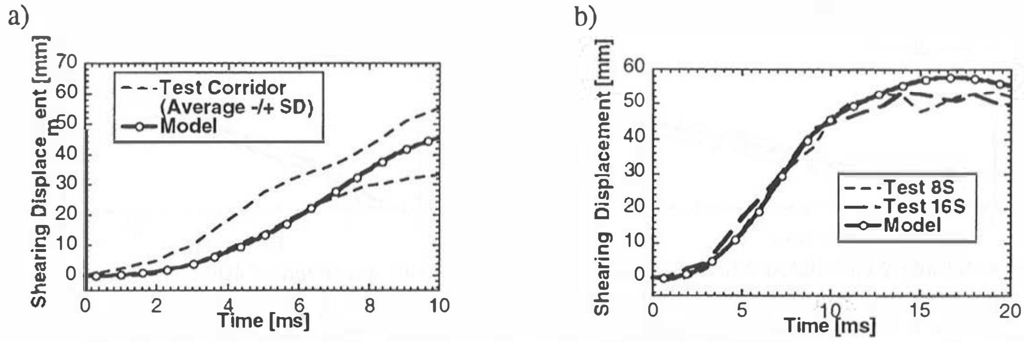

Fig. 13 Shearing displacement calculated when simulating the shearing tests at a speed of 40 km/h. a) Comparison with the test corridor obtained using results of all the experiments by Kajzer et al. (1997); b) Comparison with results of the tests in which ligament avulsion was the initial injury (tests 8S and 16S).

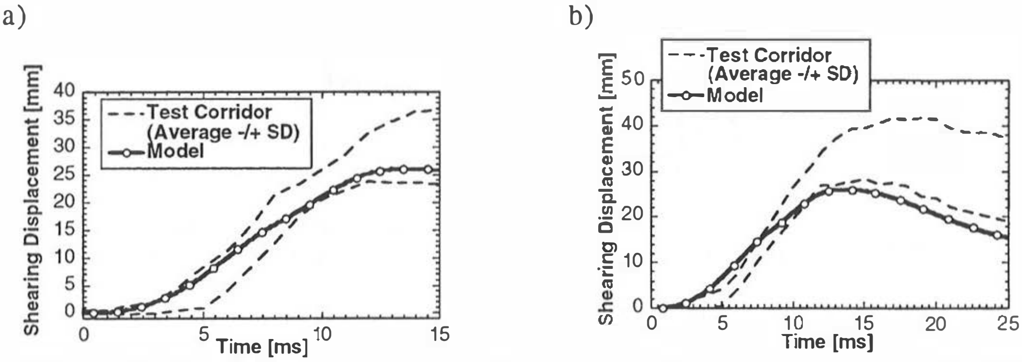

Fig. 14 Shearing displacement calculated when simulating the shearing tests at a speed of 20 km/h. a) Comparison with the test corridor obtained using results of all the experiments by Kajzer et al. (1999); b) Comparison with the test corridor obtained using results of the experiments in which ligament avulsion was the initial injury.

Leg bending angle Simulation of the shearing tests at an impactor speed of 40 km/h yielded time histories of the leg bending angle inside the experimentally obtained corridors during the initial 10 ms of impact (Fig. 15a). However, comparison of the simulation results with the bending angle-time histories determined from the experiments in which ligament avulsion was the initial injury indicated that the current model slightly overestimated the leg bending angle for time values exceeding 10 ms (Fig. 15b). When this model was computed under the assumption that the impactor speed was 20 km/h, the differences between the calculated and experimentally determined bending angle-time histories exhibited tendencies similar to those observed for the speed of 40 km/h.

Simulation of the bending tests yielded bending angle-time histories that agreed well with those experimentally determined by Kajzer et al. (1997, 1999) for both impactor speeds of 20 and 40 km/h (Fig. 16 and 17).

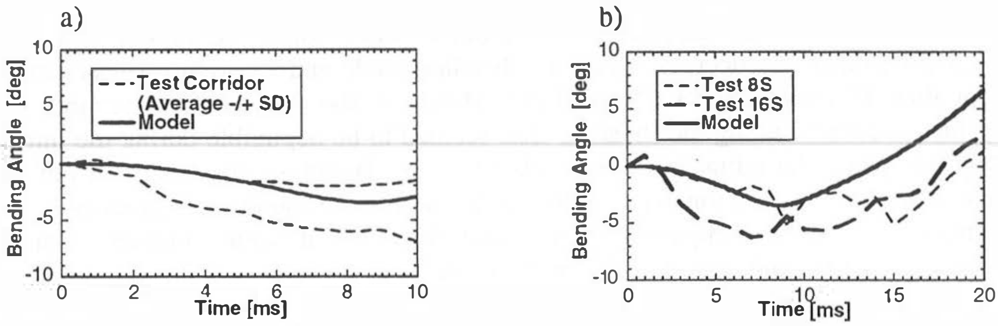

Fig. 15 Bending angle calculated when simulating the shearing tests at a speed of 40 km/h. a) Comparison with the test corridor obtained using results of all the experiments by Kajzer et al. (1997); b) Comparison with results of the tests in which ligament avulsion was the initial injury (tests 8S and 16S).

a) b)

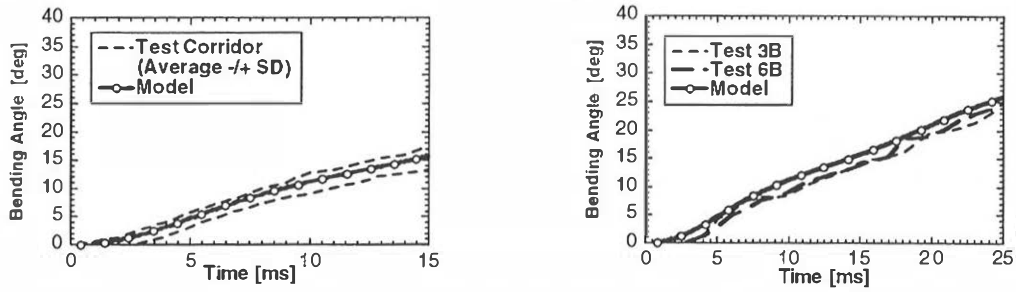

Fig. 16 Bending angle calculated when simulating the bending tests at a speed of 40 km/h. a) Comparison with the test corridor obtained using results of all the experiments by Kajzer et al. (1997); b) Comparison with results of the tests in which ligament avulsion was the initial injury (tests 3B and 6B).

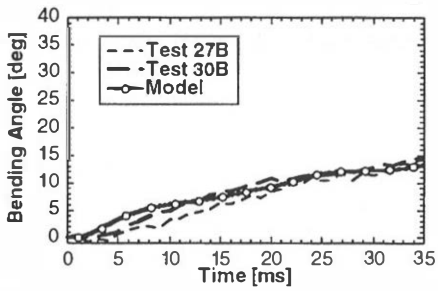

Fig. 17 Bending angle calculated in simulation of the bending tests at a speed of 20 km/h. Validation against results of the tests in which initial injury was caused by ligament avulsion (tests 27B and 30B).

Time of occurrence of initial injury The current model predicted rupture of ACL at 5 ms after the impact start when applied in simulation of the shearing tests at an impactor speed of 40 km/h. Modeling of the bending tests at this impactor speed yielded the MCL rupture at 15 ms. Thus, the times of ACL and MCL damage predicted by our model agreed well with those determined in the experiments by Kajzer et al. (1997) (Table 2). However, this model overestimated the time of occurrence of MCL damage when applied in simulation of the bending tests at an impactor speed of 20 km/h. The predicted time was 37 ms, whereas that obtained in the experiments varied between 22 and 30 ms.

PARAMETER STUDY Effect of foot-ground friction Varying the friction coefficient between the foot and simulated ground exerted a negligible effect on the results of the modeling of bending tests (Fig. 18b and 19b). The explanation is that in such tests the foot was supported on a plywood plate which was allowed to move freely in the impact direction.

In the simulation of shearing tests, the ellipsoid representing the subjects` foot interacted with the fully constrained plane connected to the global inertial coordinate system. Therefore, the calculated effect of the friction coefficient between this plane and the ellipsoid on the current model responses was stronger than that predicted when simulating the bending set-up. For the coefficient of 0.4, friction force between the foot and ground constrained the leg motion much more strongly than for the coefficient of 0.2. This resulted in an increase in the impact force (longer duration of contact between the impactor and leg) and in a decrease in the leg bending angle and shearing displacement for time values greater than 15 ms (Fig. 18a, 19a and 20). However, the effect of foot-ground friction on responses of the model simulating the shearing tests seemed to be negligible during the initial impact phase. During this phase the initial injury is likely to occur. Therefore, the results of our parameter study suggest that although friction force between the foot and ground can appreciably affect the overall kinematics of the leg impacted in a lateral direction, it seems unlikely that its exact representation is crucial for understanding the mechanisms of injury to knee joint ligaments.

a) b)

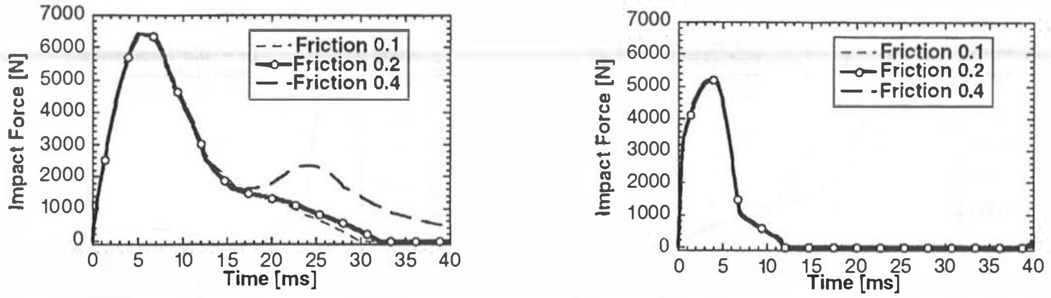

Fig. 18 Effect of the friction coefficient between the foot and simulated ground on the calculated impact forces. a) Simulation of shearing tests; and b) Simulation of bending tests.

a) b)

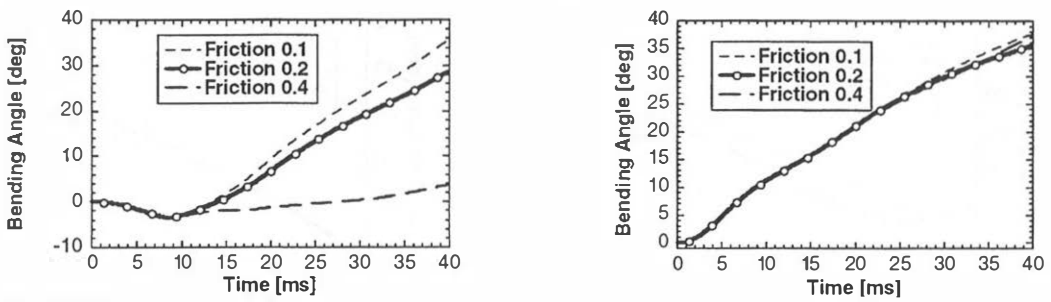

Fig. 19 Effect of the friction coefficient between the foot and simulated ground on the calculated bending angle of a leg. a) Simulation of shearing tests; and b) Simulation of bending tests.

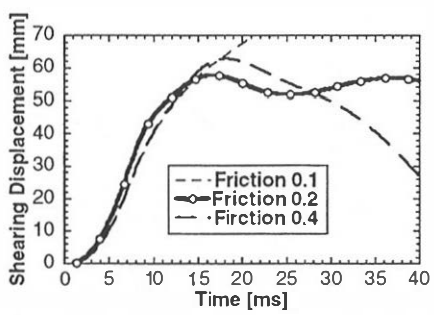

Fig. 20 Effect of the friction coefficient between the foot and simulated ground on the calculated impact shearing displacement of a leg when simulating the shearing tests.

Effect of stiffness of knee joint ligaments Similary to the results obtained when varying the foot-ground friction coefficient, varying the stiffness of knee joint ligaments exerted a much weaker effect on the model responses in simulation of the bending than of the shearing tests. When this stiffness was increased two times in comparison to that summarized in Fig. 2b, the peak value of the impact force virtually did not change (Fig. 21b), and the bending angle only slightly decreased (Fig. 22b). One possible explanation for the relatively low sensitivity of responses of the model representing the bending tests to the assumed ligament stiffness can be the low moment arm of ligament forces around the axis of the leg rotation in such tests.

On the other hand, results obtained when modeling the shearing tests seemed to be appreciably affected by the selection of ligament stiffness even in the initial impact phase (Fig. 21a, 22a, 23). When the ligament stiffness was assumed to be two times higher than the reference one defined in Fig. 2b, the calculated shearing displacement at 10 ms decreased by around 2.5 times in comparison to that predicted using this reference stiffness. This result can be interpreted as an indication that the appropriate representation of the force-elongation properties of knee joint ligaments is an important factor in predicting the overall kinematics of a leg subjected to a lateral impact near the knee joint.

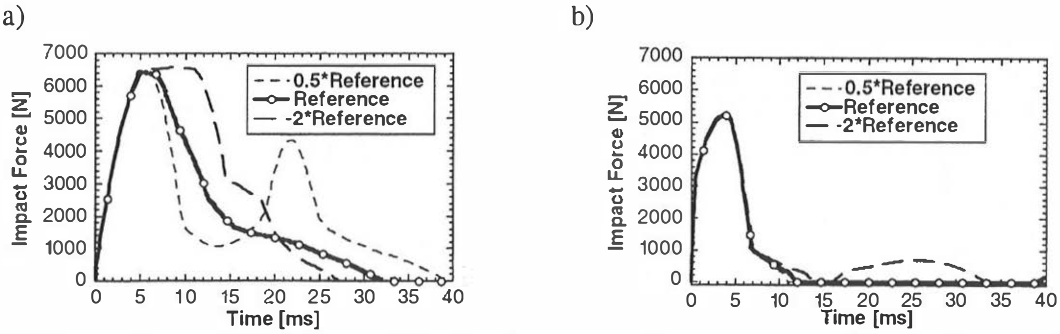

Fig. 21 Effect of the assumed ligament stiffness on the calculated impact forces. a) Simulation of shearing tests; and b) Simulation of bending tests.

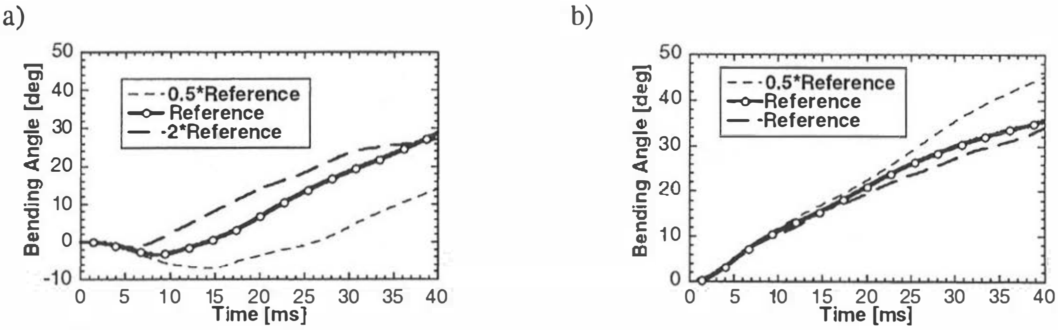

Fig. 22 Effect of the assumed ligament stiffness on the calculated bending angle of a leg. a) Simulation of shearing tests; and b) Simulation of bending tests.

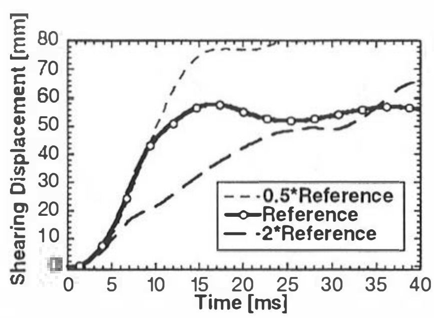

Fig. 23 Effect of the assumed ligament stiffness on the calculated shearing displacement of a leg when simulating the shearing tests.

DISCUSSION

ANALYSIS OF EXPERIMENTAL RESULTS The current analysis of the experimental results of Kajzer et al. (1997, 1999) on lateral impacts to a leg indicated two important aspects of the relation between the type of initial injury occurring in such impacts and the time histories of impact force, leg shearing displacement and leg bending angle. First, in the impact phase associated with occurrence of the initial injury, these time histories were virtually unaffected by the type of this injury. Secondly, in the impact phase following occurrence of the initial injury, variations in the time histories of leg shearing displacement and bending angle increased. For instance, in the shearing tests conducted using an impactor speed of 40 km/h, magnitude of the leg shearing displacement was clearly greater in tests that resulted in tibia fracture than in those leading only to ACL avulsion (Fig. 5).

On the other hand, the current study focused on the development and application of a multi-body model of the lower extremity, which made it possible to accurately represent only the ligament damage. For this reason, we derived corridors of time histories of the leg shearing displacement and leg bending angle for validation of this model using the methodology that minimized an effect of tibia and femur fracture on the general behavior of these corridors. Such an effect has been reported to be one of the crucial problems encountered when applying these time histories in evaluations of the biofidelity of legform impactors, *e.g.*, Matsui et al. (1999). Application of the response corridors determined by us can be one possible countermeasure against these problems.

MODEL VALIDATION The lower extremity model developed here yielded results close to those obtained in the experiments by Kajzer et al. (1997, 1999) on shearing and bending responses at the knee joint in a lateral impact to a leg. The calculated impact force-time histories nearly fit the experimental corridors (Fig. 10 to 12). However, their agreement with the experimental results was clearly better in simulation of the experiments conducted at an impactor speed of 40 than at 20 km/h. One possible explanation of this difference in the results of impact force calculations can be that in the current study multi-body modeling was used. Thus, both the leg and impactor were simplified as rigid ellipsoids, and the contact between them was represented using a force-penetration curve. For consistency, the same curve was used when modeling the experiments at impactor speeds of 20 and 40 km/h. It is rather unlikely that a single force-penetration curve can suffice for an accurate prediction of the contact force between the leg and impactor at various impact speeds.

However, simplifications in the modeling of interactions between the leg and impactor did not compromise the biofidelity of the current model in representing leg kinematics. The calculated time histories of the leg shearing displacement and bending angle were within the experimental corridor when this model was applied in simulation of both shearing and bending tests at impactor speeds of 20 and 40 km/h (Fig. 13, 14a, 15, 16, and 17).

PARAMETER STUDY As the responses of the current model of lower extremity exhibited good agreement with those determined in the experiments by Kajzer et al. (1997, 1999), it was appropriate to apply this model to an analysis of the sensitivity of the leg kinematics obtained in these experiments to the properties of knee joint ligaments and the friction coefficient between the foot and simulated ground. This analysis suggested that the conclusions on injury mechanisms to the knee joint derived from these experiments are unlikely to be affected by the elements of the experimental set-up that determined the friction coefficient between the ground and a subject's foot. The basis for this conclusion is that the effect exerted by this coefficient on both the model kinematics and the calculated impact force was negligible in the initial impact phase. On the other hand, determination of the mechanical properties of an uninjured knee joint in a lateral impact to a leg would require analysis of the leg responses in a time window longer than that associated with the occurrence of the initial injury. The current modeling results suggest that such a determination may be affected by the friction coefficient between the foot and ground, since this coefficient is likely to influence the overall leg kinematics in the experiments conducted under the shearing set-up.

Since the current leg model was validated against the experiments obtained under set-up similar to that proposed in ISO (1996) for evaluation of legform impactors, it seems reasonable to stipulate that our results are generally valid for such impactors. However, the properties of contact between the PMHS foot and simulated ground are likely to be different from that of tube-shaped steel impactors. Therefore, it may not be appropriate to attempt to quantify an effect of the foot-ground friction coefficient on legform impactor responses based on the results obtained in the current study.

STUDY LIMITATIONS AND RECOMMENDATIONS FOR FURTHER RESEARCH In the current study multi-body modeling was used, and a leg was simplified using a single rigid ellipsoid. Such simplification did not make it possible to directly represent two phenomena observed in the leg responses to lateral impacts: 1) Bending deformation of the tibia; and 2) Movement/deformation of leg muscles caused by the impact and inertia forces. Some aspects of these phenomena were taken into account by calibrating the force-penetration characteristics for a determination of the properties of the contact between the leg and impactor ellipsoids. However, their complete representation could not be done by means of the modeling technique utilized in the current study. For instance, movements of leg muscles caused by contact and inertia forces could influence not only the properties of the leg-impactor contact but also the active mass of the leg. One possible solution to overcoming the limitations of the current study in simulation of lower extremity responses in lateral impacts might be to apply the finite element modeling technique. This technique makes it possible to accurately represent bone and muscle deformation. However, its application would lead to models with higher complexity and longer computation times than those of the multi-body model.

It should also be noted that the leg shearing displacement analyzed in the present study was determined as a relative displacement of two markers located on the femur and tibia shafts at around 50 mm from the center of the knee joint. Since it is likely that the femur and tibia were appreciably deformed in the experiments by Kajzer et al. (1997, 1999), the shearing displacement determined in this way may differ from the actual displacement between the femur condyles and tibia articular surface. Since such a displacement is considered to be one of the criteria of knee joint injury in car-pedestrian accidents, its accurate determination is of great importance. Such a determination can be made using markers placed directly on the femur condyles and the tibia articular surface as close as possible to the knee joint center.

However, it seems rather unlikely that the current conclusions on the sensitivity of leg responses in lateral impacts to the stiffness of knee joint ligaments and to the friction coefficient between the foot and ground were compromised by simplifications made in the current analysis. These conclusions, after all, were derived based on the modeling results on the overall kinematics of the leg in lateral impacts which agreed well with those obtained in the present experiments.

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REFERENCES

Abdel-Rahman, E. and Samir Hefzy, M., A two-dimensional dynamic model of the human knee joint, *Journal of Biomechanical Engienering,* 115, 1993, pp. 357-365.

EEVC, European Experimental Vehicle Committee, *Proposals for Methods to Evaluate Pedestrian Protection for Passenger Cars,* EEVC Working Group 10 Report, 1994.

EEVC, European Experimental Vehicle Committee, *Improved Test Methods to Evaluate Pedestrian Protection Afforded by Passenger Cars,* EEVC Working Group 17 Draft Report, 1998.

Girgis, F. G., Marshall, J. L. and Monajem, A. R. S., The cruciate ligaments of the knee joint, *Clinical Orthopaedics and RelatedResearch*, 106, 1975, pp. 216-231.

ISO, International Standard Organization, *Passenger Cars and Light Commercial Vehicles-Pedestrian Protection-Impact Test Methodfor Pedestrain Lower Leg and Knee*, ISO/TC22/SC10/WG2 Working Draft 11, 1996.

Kajzer, J., Cavallero, C., Bonnoit, A., Morjane, A. and Ghanouchi, S., Response of knee joint in lateral impact: Effect of bending moment, Proc. of the *International IRCOBI Conference on the Biomechanics of Impacts*, Eindhoven, The Netherlands, 1993, pp. 105-116.

Kajzer, J., Schroeder, G., Ishikawa, H., Matsui, Y. and Bosch, U. Shearing and bending effects at knee joint at high speed lateral loading, Proc. of the *IRCOBI Conference on the Biomechanics of Impacts*, Hannover, Germany, 1997, pp. 151-165.

Kajzer, J., Matsui, Y., Ishikawa, H., Schroeder, G. and Bosch, U., Shearing and bending effects at the knee joint at low speed lateral loading, Proc. of the *SAE International Congress*, Society of Automotive Engineers, Detroit, Michigan, USA, 1999, pp. 1-12.

Kennedy, J. C., Hawkins, R. J., Willis, R. B. and Danylchuk, K. D., Tension studies of human knee ligaments, *Journal of Joint and Bone Surgery,* 58-A, 1976, pp. 350-355.

Matsui, Y., Ishikawa, H., Sasaki, A., Kajzer, J. and Schroeder, G., Impact response and biofidelity of pedestrian legform impactor, Proc. of the *International IRCOBI Conference on the Biomechanics of Impacts*, Barcelona, Spain, 1999, pp. 343-354.

Mertz, H., A procedure for normalizing impact response data, SAE-Paper No. 840884, SAE Transactions Section 4, 1984, pp. 351-387.

Noyes, F. R., DeLucas, J. L. and Torvik, P. J., Biomechanics of anterior cruciate ligament failure: An analysis of strain-rate sensitivity and mechanisms of failure in primates, *Journal of Joint and Bone Surgery,* 56-A, 1974, pp. 236-253.

Soames, R. W., Skeletal system, *Gray's Anatomy*, Edited by Bannister, L. H., Berry, M. M., Collins, P., Dussek, J. E., Dyson, M., Ellis, H., Gabella, G., Salmons, S., Soames, R. W., Standring, S. M., and Williams, P. L. Churchill Livingstone, New York, 1995, pp. 425-736.

Trent, P. S., Walker, P. S. and Wolf, B., Ligament length patterns, strength, and rotational axes of the knee joint, *Clin. Orthop. andRel. Res.*, 117, 1976, pp. 263-270.
Wismans, J. A., Three Dimensional Mathematical Model of the Human Knee Joint, PhD Thesis, Eindhoven University of Technology, 1980.
Woo, S. L.-Y., Hollis, M. J., Adams, D. J., Lyon, R. M. and Takai, S., Tensile properties of the human femur-anterior cruciate ligament-tibia complex, *The American Journal of Sports Medicine,* 19, 1991, pp. 217-225.
Woo, S. L.-Y., Livesay, G. A., Runco, T. J. and Young, E. P., Structure and function of tendons and ligaments, In *Basic Orthopaedic Biomechanics*, Eds. Mow, V. C. and Hayes, C. W., Lippincott-Raven, Philadelphia, 1997, pp. 209-251.
Yang, J. K. and Kajzer, J. Computer simulation of impact response of the human knee joint in car-pedestrian accidents, Proc. of the *36th Stapp Car Crash Conference*, Seattle, USA, 1992, pp. 203-217.
Yang, J. K., Kajzer, J., Cavallero, J. and Bonnoit, J. Computer simulation of shearing and bending knee response of the knee joint to a lateral impact, Proc. of the *39th Stapp Car Crash Conference*, Coronado, California, USA, 1995, pp. 251-264.