DEVELOPMENT AND VALIDATION OF A MATHEMATICAL BREAKABLE LEG MODEL

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ABSTRACT

In a previous study on impact response of the human knee joint in lateral car-pedestrian collisions a mathematical model of pedestrian lower extremity with a human-like knee joint was presented. The results showed that the model lacked adequate representation of the deformation and fracture of the leg segments since the response was shown to be strongly dependent on whether the leg was fractured or not. In this study, a mathematical model which would also represent a breakable leg was developed and implemented into the pedestrian lower extremity model.

The leg model consists of two rigid-body elements connected by a "fracturable joint". The moment-deformation characteristics of the "fracturable joint" are described by the user subroutines, which were prepared and added to the MADYMO 3D program system. The input data for the "fracturable joint" model originate from available biological specimen tests.

Computer simulations of car-pedestrian impact with this modified pedestrian model were conducted at a speed of 31 km/h in four different configurations, and compared with previously performed human leg specimen tests. Different types of bumper compliance and bumper levels were simulated. The bumper force, the accelerations, the condyle contact forces and the ligament strains were calculated during simulations. The results showed that the modified model gave a higher biofidelity than did the previous model with the undeformable representation of the leg segments. The calculated parameters such as bumper forces and accelerations corresponded with the measured parameters in tests. The impact response of the lower extremity could be well predicted by the model.

INTRODUCTION

Accidents involving pedestrians account for a significant proportion of casualties in all road traffic accidents. Therefore improvement of pedestrian safety is considered a priority in crash injury protection.

During the last couple of decades, a main objective of research on pedestrian protection has been to reduce the aggressiveness of the vehicles. Since the end of 1970's, a large number of experimental tests have been performed at the Department of Injury Prevention, Chalmers University of Technology, with biological specimens (Bunketorp, 1983; Aldman et al., 1985). In these tests, human lower extremity specimens, connected to artificial upper bodies were struck by a simulated car front with different bumper systems. A mechanical pedestrian model was also developed for experimental studies of the risk of leg injuries in car-pedestrian accidents (Kajzer, 1991). Both the biological and the mechanical model could simulate the leg and knee injury mechanisms observed in real car-pedestrian accidents.

A computer simulation technique was then developed for the modelling of pedestrian accidents, since it would largely facilitate the study of different input parameters such as geometrical configuration and stiffness of car-front structures. Good repeatability can be attained by computer modelling, which means that the effect of changes to a single parameter can be clearly and easily assessed.

Computer simulation studies on car-pedestrian impact accidents where the pedestrian model has been based on different commercial mechanical dummies have been performed by many researchers. Analysis of results from previous computer simulations of pedestrian impacts (Fowler and Harris, 1982; van Wijk et al., 1983; and Janssen et al., 1986) shows that the pedestrian model developed in the MADYMO is not an adequate representative of human body compared with biological specimen tests. The lack of adaptive human body flexibility is one of the disadvantages.

In our previous work (Kajzer, 1992), a model of lower extremity with human-like knee joint was introduced into a mathematical pedestrian model to simulate the car-pedestrian impact event. A comparison with accessible results obtained from experiments with biological material showed however, that the previously developed model of lower extremity needed some modification. The leg segment was simulated by one rigid element. It cannot model leg deformations and fractures, which are frequently observed in clinical practice and in impact experiments with human leg specimens. When bone fracture occurs in vehicle crashes, the clinical observations of the knee joint injuries, particularly to the ligament apparatus, are different from those observed in simulations with rigid elements. A new model should allow investigation of impact response of the knee joint both in cases when long bone fracture occurs as well as when fracture does not occur. To simulate this injury mechanism the model for the leg should be extended by two elements connected by a "fracturable joint". The "fracturable joint" between these elements should be situated at the contact point of impact and adjusted to the force-bending function for the tibia-fibula. The fracture resulting from impact duration of a load pulse should be defined on the "joint".

MODEL DEVELOPMENT

Model of Breakable Leg

To obtain a better biofidelity, a new model of the leg was introduced into the mathematical pedestrian model developed by Yang and Kajzer (1992). Instead of one rigid element, as in the primary model, the new leg model consists of two rigid body elements connected by a cardan universal joint. The cardan joint is located at the bumper impact level and defined as "fracturable joint" (Figure 1). The geometry and mass characteristics of the new leg model are calculated to achieve consistent properties of the one element model of the leg. The user subroutine in the MADYMO 3D (TNO, 1992) was used to describe the "fracturable joint".

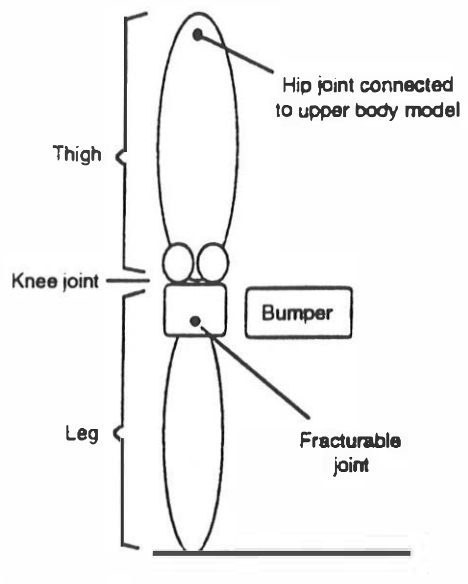


Figure 1. The lower extremity model with breakable leg.

Modelling fracturable joint by user subroutines

The characteristics of the "fracturable joint" are shown in Figures 2a and 2b, where ϕ, θ, ψ are the rotation angles in the cardan joint, T_ϕ, T_θ, T_ψ are the torques acting on the joint, and α is an undimensional coefficient. As long as the criterion of fracture initiation is not fulfilled, the coefficient $\alpha = 1$ (Figure 2b) and the "fracturable joint" will convey components of the moment according to the typical characteristic (Figure 2a), which has been introduced to the MADYMO 3D program in a standard way using the input data file.

The bone material as all biological materials has viscoelastic property. It is known from experiments (Viano, 1977; Lowne, 1982) with cadaver specimens that the force value that initiates the bone fracture process depends on the duration of a load pulse in the way shown in Figure 3, as well as on a loading rate.

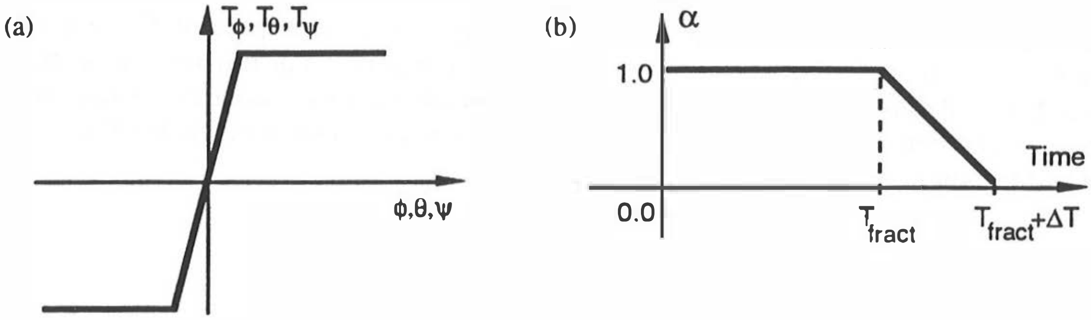


Figure 2. Components of moment-deformation characteristics of "fracturable joint".

In case of impact, the value of a load pulse changes very rapidly. Thus, to predict whether a fracture will occur, it is necessary to check the time history of the impact force between a bone and an impactor on-line during simulation. The impact force is calculated for each time step of the simulation.

A special user subroutine for the MADYMO 3D program was developed to model long bone viscoelastic properties. When the impact force is calculated step by step in integrated system equations of motion, this subroutine calculates the "time of possible fracture" T_{fract} (Figure 2b) at which fracture may occur. First it checks the leg-bumper impact force F, then, taking into account the characteristic shown in Figure 3, it calculates the time at which fracture may occur for the current value of the impact force F. Next, a comparison is made between this time and the "time of possible fracture" previously predicted in the simulation process. The lower value is saved for use in next simulation step as the new "time of possible fracture". When the duration of the current value of impact force becomes greater than T_{fract}, a fracture is initiated and the other part of the subroutine takes care of further calculations.

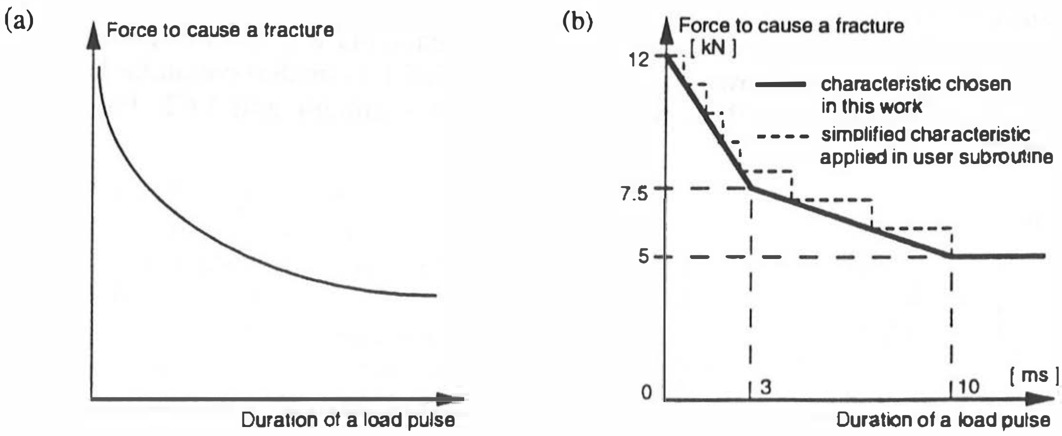
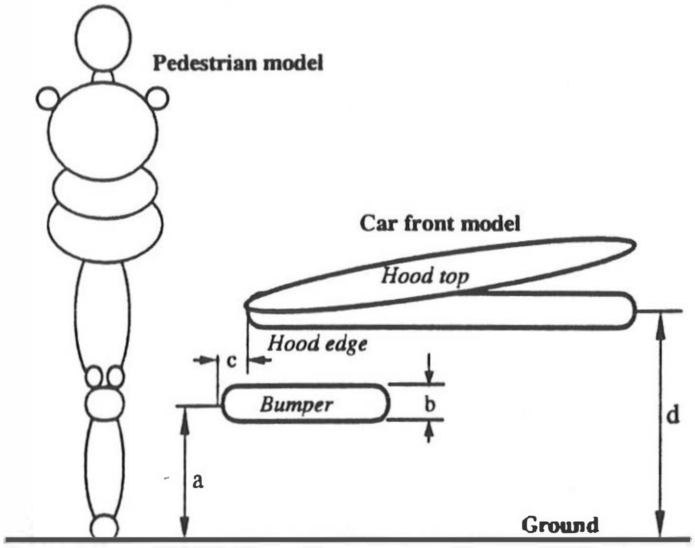


Figure 3. Dependence between the duration of the load pulse and the force value effected in fracture of a bone: (a) general case (based on Viano, 1977; Lowne, 1982), (b) characteristic used in the presented study.

The process from the beginning of the fracture to the total fracture of the bone is characterised by a sudden decrease in rigidity in the "fracturable joint". In the adopted model it is assumed that the decrease is a linear function of time (Figure 2b). The whole process ends after time interval ΔT, when the value of moment conveyed by that artificial joint decreases to the level in which only damping and friction is remaining.

Model Set-up of Car-Pedestrian Impact

The simulation set-up for car-pedestrian collision is shown in Figure 4. A pedestrian model with the human-like knee joint and with the breakable leg model was used in this study. Two additional hyper ellipsoids were attached to the upper and lower part of the leg elements so that the contact force between bumper and each part of the leg could be adequately calculated. The computer simulation of car-pedestrian impact can be performed with a 2D or 3D model. The 3D model was used in the present simulations since the human knee joint is a non symmetrical structure and the centres of gravity of the body segments are of 3D distribution.



a = bumper (impact) level, b = bumper width,
c = bumper lead distance. d = hood edge height.

Figure 4. The computer simulation set-up for car-pedestrian collision.

The force deflection characteristics of the ligaments in the knee model originated from Wismans (1985). These seemed stiff when the parameters were used directly in the model. A reduction by 20% of the original rupture force of ligaments was used in present study. The ligament characteristics is shown in Figure 5, where MCL is medial collateral ligament, ACL anterior cruciate ligament, PCL posterior cruciate ligament and LCL lateral collateral ligament.

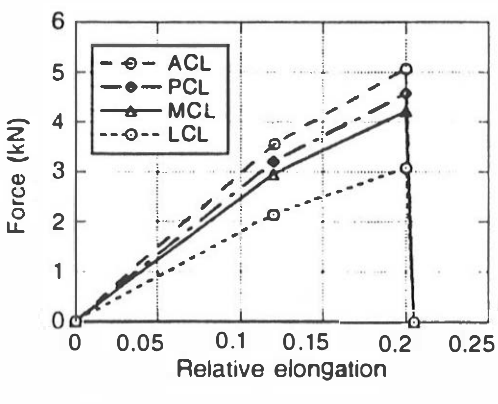


Figure 5. Force deflection characteristics of the ligaments represented by the springs in the knee model.

In this study a relation between the calculated strain of the ligament and the AIS level classification of the ligament damage observed in the tests with biological specimen was used (Table 1). This relation has been used in previous mathematical simulations of impact response of the human knee joint (Yang and Kajzer, 1992), and is based on the study performed by Butler et al. (1986). They determined that human ligaments failed at a maximum strain of 13% - 15%. In the experimental study (Bunketorp, 1983 and Aldman et al., 1985) used in our model validation, damage to the knee ligaments was classified as AIS 1 - 3. A ligament sprain was classified as AIS 2 and a ligament rupture as AIS 3. In our simulations, a calculated ligament strain of 12% was defined as the first sign of partial rupture, and 20% calculated strain was assumed to correspond to total rupture.

Table 1
Correlation of the ligament strain with severity classification of the ligament damage

Ligament strain ε (%)	AIS code
ε ≥ 20	3
15 < ε < 20	3 -
12 ≤ ε ≤ 15	2
ε <12	0 - 1

The simulated car-front was in accordance with the test car-front set-up. In simulations using the MADYMO 3D program, the car-front model (Figure 4) was represented by two hyper ellipsoids for the bumper and a hood edge, as well as an ellipsoid for the hood top.

Bumper stiffness characteristics are based on the force-deformation relationship determined by static tests in which an iron tube of diameter 60 mm was pressed at a right angle against the impact area of the bumper and modified to better simulate the dynamic response of the bumper. For each configuration, a compliant layer of 10 mm on the leg was considered for effect of the soft tissue. The characteristics for the bumper-leg interaction were calculated by combining stiffness of the bumper and the leg (Figure 6).

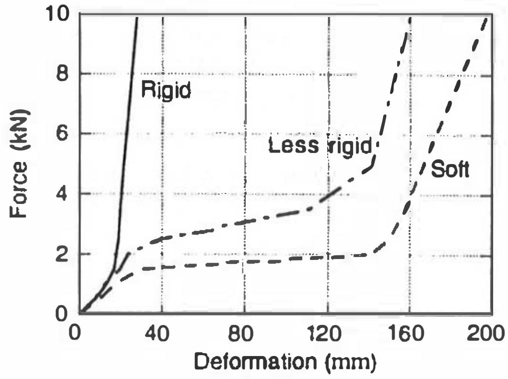


Figure 6. The resultant force deformation relationship of bumper for dynamic response.

For hood-edge characteristics, a compliant layer of 30 mm was also taken into account for the effect of thigh soft tissue. A combination of characteristics of hood-edge and thigh was used in the simulations.

MODEL VALIDATION

The model was validated by results from impact tests with human leg specimens, previously performed by Bunketorp (1983) and Aldman et al. (1985).

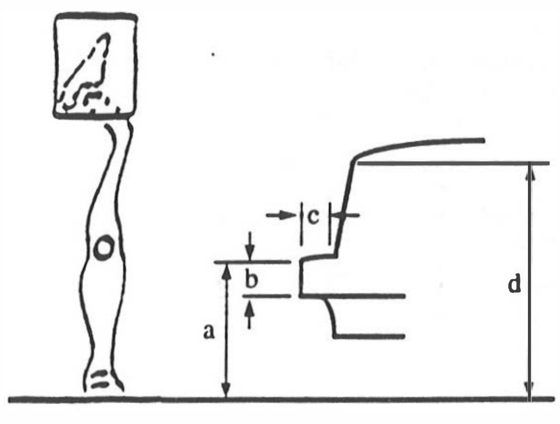


Figure 7. The test set-up for a car-front impacting human leg specimen.

A test set-up (Figure 7) was used in these impact tests. The leg specimens were amputated with a part of the pelvis. The most of the muscle of the thigh was removed. All specimens were positioned with the extended knee, balanced in an upright position and preloaded with a concentrated body mass. They were struck by a simulated car-front. Bumper type, bumper level, bumper lead distance, height of the hood-edge and impact velocity were varied in different configurations. The available information from these tests were: high speed movies and signals from force and acceleration transducers mounted on the car fronts and on the specimens.

Four configurations were chosen as input data to conduct the computer simulation study (Table 2). For each simulation, five specimen tests were available for verification.

Table 2
The configurations for the computer simulations
(based on Bunketorp, 1983; Aldman et al., 1985)

Configuration	Bumper level (mm)	Hood edge level (mm)
Rigid-high	450	800
Less rigid-high	450	800
Soft-high	450	800
Less rigid-lower	325	800

The computer simulations of car-pedestrian impact were performed at an impact speed of 31 km/h corresponding to the tests with the biological specimens.

In these tests with human leg specimens, damage to the lower extremities was mostly classified as AIS 1-3. In the knee, a ligament sprain was classified as AIS = 2 and a ligament rupture as AIS = 3. A femur fracture was classified as AIS = 3, a leg or ankle fracture as AIS = 2, or AIS = 3, if the fracture was open, displaced or comminuted. A condyle fracture was classified as AIS = 3. The damage to specimens observed in tests was documented by means of the AIS code which is also available for model verification.

RESULTS AND DISCUSSIONS

The results from the computer simulations with the "fracturable joint" model for the four configurations at an impact speed of 31 km/h are presented and compared with the results (mean values) from the biological specimen tests.

Impact forces and accelerations - The peak bumper impact forces and the peak accelerations of the leg at impact level changed with the different parameters for bumper compliance and bumper level. The impact forces varied between 2.2 kN and 7.2 kN (Table 3) for the simulation series with the breakable leg model, which was in accordance with the specimen tests. The accelerations of the leg at impact level varied from 80 g to 520 g (Table 4). The peak accelerations in the simulations were higher than the results from biological specimen tests.

Table 3
Bumper impact forces from the computer simulations and specimen tests
at a simulated car-front speed of 31 km/h
(Test results are based on Bunketorp, 1983 and Aldman et al., 1985)

configuration	Peak impact force (mean value) from tests (N=5) (kN)	Peak impact force from simulations (kN)
Rigid-high	7.4	7.2
Less rigid-high	2.4	2.5
Soft-high	2.6	2.2
Less rigid-lower	2.2	2.5

A comparison between the results of the simulations with the breakable leg model and those of earlier biological specimen tests showed that the peak values of the bumper impact force in the simulations agree with the results (mean values) from the biological specimen tests (Table 3). The input parameters of the simulation set-up were generally defined and therefore such a system cannot give a more detailed detection for each single test with a human leg specimen, but the same tendency of peak impact forces as in biological specimen tests can be predicted very well by simulations. The calculated impact force on the bumper tends to increase at the impact area with the increase in stiffness of the bumper system.

Table 4
Accelerations at impact level from the computer simulations and specimen tests at a simulated car-front speed of 31 km/h
(Test results are based on Bunketorp, 1983 and Aldman et al., 1985)

configuration	Peak acceleration (mean value) from tests (N=5) (g)	Peak acceleration from simulations (g)
Rigid-high	263	520
Less rigid-high	87	125
Soft-high	68	80
Less rigid-lower	93	120

Table 4 shows the accelerations of leg at impact level in the mathematical simulations and in the biological specimen tests. If we simply make a comparison between results from the simulation and the test of each individual configuration, an obvious difference will be found. An important reason for this is that the measurement method in the specimen tests was completely different from the calculations in the simulations. In the tests, the accelerometer strapped to the leg soft tissue at the opposite side of the impact area was unstable. The signal from the accelerometer was registered by a mechanical device that could not catch the high frequency signal. The mechanical recording device thus worked as a filter, and the high frequency signal was cut off. Therefore, it was anticipated that the results from such a measurement would be much lower than from the simulations. Another reason for this difference is probably due to variations in active mass of biological specimens in the previous tests. Even though the higher accelerations were obtained in simulations rather than in tests, a similar tendency of peak accelerations can be seen clearly for all configurations in the simulations with the breakable leg model and in biological specimen tests. It is shown that the breakable leg model is sensitive to changes in car-front parameters, such as bumper stiffness.

It is necessary to point out that an impact phenomenon lasts for a very short time (milliseconds). Thus, there is a need to apply standard filtering procedures for experimental tests as well as for computer simulations. Without such procedures any accurate comparisons between experimental and calculated results would be very difficult or even impossible, in particular if the results have been obtained by different research teams and not all the conditions of experiments are reported.

Condyle contact forces - In the computer simulations with the breakable leg model, the condyle contact forces varied between 1.9 kN and 9.8 kN for the lateral condyle, and 0.15 kN to 0.20 kN for the medial condyle (Table 5). In the human leg specimen tests, a transverse fracture below the tibial condyle at the impact level and/or a split-depression fracture of the lateral tibial condyle could be observed from tests 1 and 4 in the configuration of "soft" - high level bumper. The highest AIS for the condyle damage was 3 in the tests.

In all simulations with the high level bumper (Table 5), the calculated value of contact force for the lateral condyle ranged from 6.0 kN to 9.8 kN, and 0.15 kN to 0.20 kN for the medial condyle. In the corresponding test with the human leg specimen, the AIS 3 damage to the lateral condyle could be observed but no damage to the medial condyle. In the case of the "less rigid" - lower bumper, the condyle contact forces were lower for lateral condyle (1.9 kN). The condyle contact force from the simulations indicated that a good prediction of impact response of the knee model with the breakable leg model was achieved.

Table 5
Condyle contact force from simulations and condyle damage from biological specimen tests at a simulated car-front speed of 31 km/h
(Test results are based on Bunketorp, 1983 and Aldman et al., 1985)

configuration	Test No	Condyle damage* from tests MC**	Condyle damage* from tests LC**	Condyle force from simulations MC (kN)	Condyle force from simulations LC (kN)
Rigid-high	1	0	3-	0.20	6.0
	2	0	0		
	3	0	3-		
	4	0	0		
	5	0	0		
Less rigid-high	1	0	3	0.20	9.8
	2	0	0		
	3	0	3-		
	4	0	0		
	5	0	0		
Soft-high	1	0	3	0.15	9.1
	2	0	0		
	3	0	0		
	4	0	3		
	5	0	0		
Less rigid-lower	1	0	0	0.15	1.9
	2	0	0		
	3	0	0		
	4	0	0		
	5	0	0		

* AIS code
** MC = Medial condyle of tibia, LC = Lateral condyle of tibia

In a study with cadaver specimens, Powell et al. (1975) reported that the condyle fractures were observed when peak impact force varied from 7 kN to 10 kN. This test result is not directly comparable with results from our simulations due to differences in test conditions, but we noticed that a very close force level (6 kN to 10 kN) was obtained in our simulations. Our calculated condyle force seems thus to be capable of predicting the risk of condyle fracture. The condyle fractures were observed in 40% of the experiments with biological specimens, except for "less rigid" - lower bumper configuration where no condyle fracture was observed. In the mathematical simulation of this configuration the condyle contact force was low.

Ligament strains - The ligament strains varied between 6% and >20% for MCL, 3% to 17% for ACL and 2% to 11% for PCL in the simulations with the breakable leg model (Table 6). The ligament strains in these simulations indicated that the MCL experienced greater tension than did other ligaments. Ligament strains exceeding 20% mean that complete rupture of the ligament occurs. Typically, the results from tests in the "rigid bumper" - high level configuration show that MCL ligament damage was AIS 3 in all 5 tests. In test 4 for the same configuration, the impact response was also mainly connected to all ligament damage of AIS 3.

Table 6
Ligament strain from simulations and ligament damage from biological specimen tests at a simulated car-front speed of 31 km/h
(Test results are based on Bunketorp, 1983 and Aldman et al., 1985)

configuration	Test No	Ligament damage* from tests MCL	ACL	PCL	Ligament strain (%) from simulations MCL	ACL	PCL	Bone damage*
Rigid-high	1	3	0	0	>20	17	11	
	2	3	3	0				
	3	3	3	0				
	4	3	3-	3				
	5	3	3-	0				
Less rigid-high	1	0	2	2	19	6	4	(3)
	2	0	0	0				[3]
	3	0	0	0				(3)
	4	0	0	0				[3]
	5	0	0	0				
Soft-high	1	3	0	0	17	5	4	
	2	3	0	3				
	3	3	3	3				
	4	3	3-	0				
	5	3	3	3-				
Less rigid-lower	1	3-	0	0	6	3	2	
	2	3-	0	0				
	3	3	0	0				
	4	2	0	0				
	5	2	0	0				

* AIS code
(3) = Lateral condyle fracture, [3] = Tibial fracture

In all computer simulations the largest relative elongation appeared for MCL and can be seen in Table 6. The risk of damage to the MCL was high in the biological specimen tests. In these tests it corresponded to AIS 3 and AIS 2 damage of MCL. One exception is the case of the "less rigid" - high level bumper configuration. A high relative elongation (19%) for MCL in this simulation was calculated, but no damage to MCL in the corresponding test was observed. The mechanism of damage to knee joint in the test with human leg specimens is very complicated, and many structures cooperate in the energy absorption. In the mathematical simulation we cannot analyse the value of each output parameter separately. It can be noticed that in tests with biological specimens, two fractures of the lateral condyle and two fractures of tibia were observed in this configuration. These fractures may have protected the knee ligaments from damage in tests. In the "less rigid" - lower bumper configuration, the relative elongation (6%) of MCL was underestimated in the simulation.

The second largest relative elongation for all ACL can be calculated. In the simulations, the values of the ACL relative elongations varied from 3% to 17%. In the "rigid" - high level bumper configuration, the ACL relative elongation was high (17%), and a high risk of ACL damage could be seen in the tests. In the "less rigid" - lower level bumper configuration, the relative elongation of the ACL was low, and no damage to the ACL was noted in the tests.

The calculated relative elongation of all PCL was smaller than that of MCL and ACL in the same configuration. The results of PCL relative elongations from the simulations ranged from 2% to 11%. A similar tendency of relative elongations and damage for the PCL was observed in the simulations and tests. In our mathematical model the tibial and femoral condyles were represented by rigid elements. The characteristics of condyle fractures were not improved in the model. Even though the fracture of the condyle cannot be simulated, the mathematical model can adequately predict the risk of ligament damage.

Time histories - A comparison of the time histories of the bumper impact forces and the leg accelerations at impact level for all configurations are illustrated in Figures 8 and 9. In the simulations, it is important to obtain an impact response corresponding to biological specimen test. The time histories of bumper impact force show that the response time of peak value from the simulation was consistent with result from corresponding specimen test. The wave forms are similar between the simulations and tests. The response time and wave forms of leg accelerations at impact level are also comparable between simulations and tests. However, in the configuration with less rigid bumper, the response time of peak value is different between simulation and test. This difference seems to be influenced by the resultant stiffness at the impact-contact area.

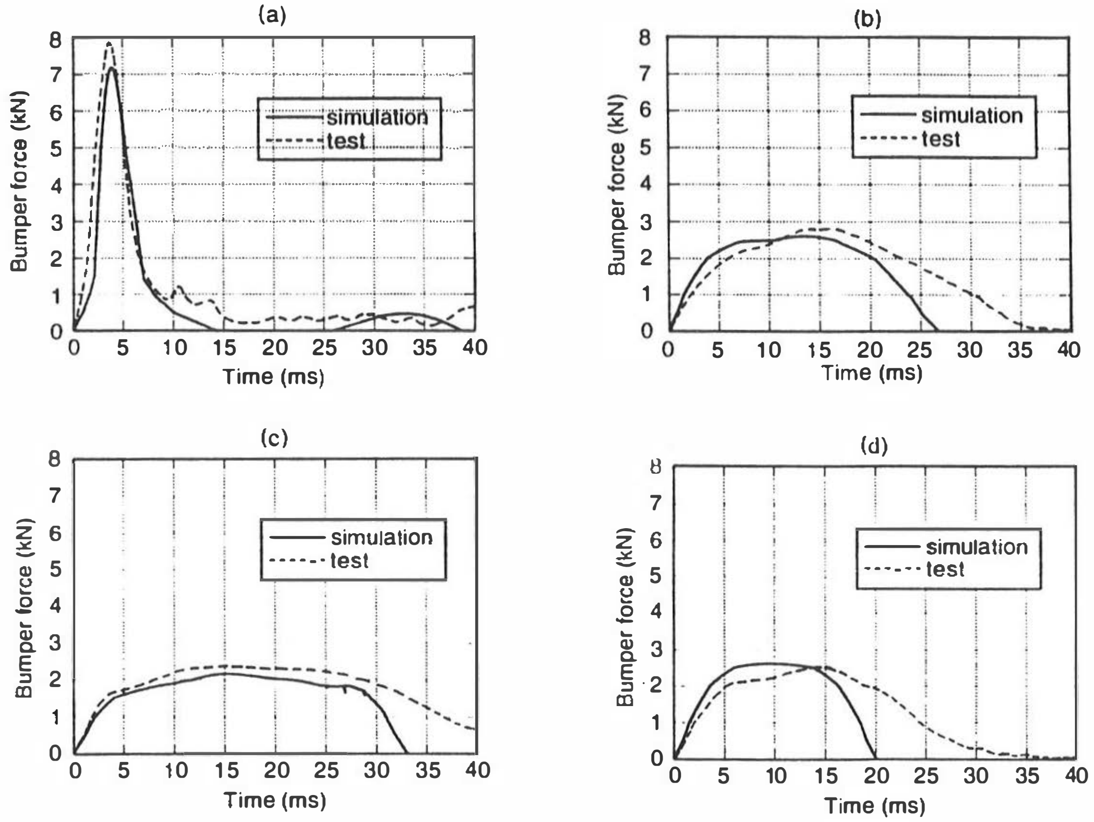


Figure 8. The time histories of bumper impact forces from the simulations and the typical tests: (a) rigid-high level bumper; (b) less rigid-high level bumper; (c) soft-high bumper; (d) less rigid-lower bumper.

In a previous simulation (Yang and Kajzer, 1992), it was shown that impact response of the knee could generally be predicted by the condyle contact force and ligament strain calculated in the model. These forces and strains might correlate to injuries to the knee observed in tests and real accidents. A conflicting result in the previous study was the outcome of the undeformable leg model. In the current study, with a breakable leg model, the condyle contact forces and the ligament strains appeared more reasonable. The model is more sensitive to changes of impact parameters.

The calculated value of impact force in simulations is less influenced by ligament strength. However, the impact response of the knee joint is significantly influenced by the ligament strength characteristics in the simulations. The lateral condyle contact forces increase and ligament elongation decreases with the increase of ligament strength. The effect of the biological material property is important to calculate the ligament force and elongation. The ligament is sensitive to the loading rate due to its viscoelasticity. In case of lateral impact to the leg, the knee joint ligaments are exposed to a high loading speed. At present the available data of ligament load-deformation are based on experiments with lower loading speeds. For instance, Kennedy et al. (1976) reported an almost 50% increase in the load to failure when the loading speed was increased fourfold from 12.5 cm/min to 50 cm/min during tensile testing of the knee joint ligaments. In the current simulations, the estimated values of ligament strength for dynamic response, are based on a study presented by Wismans (1985) (Figure 5). The results indicated that selected values for ligament strength are acceptable for this velocity range.

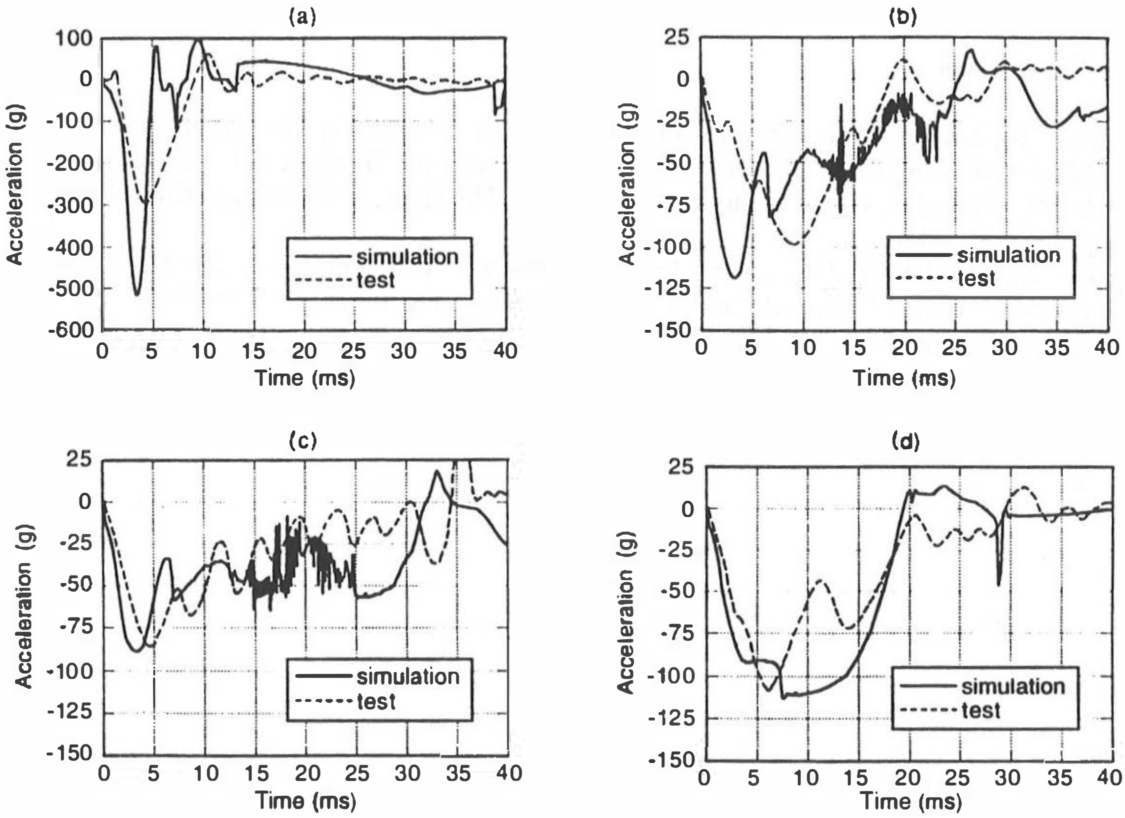


Figure 9. The time histories of leg acceleration at impact level from the simulations and the typical tests: (a) rigid-high level bumper; (b) less rigid-high level bumper; (c) soft-high bumper; (d) less rigid-lower bumper.

CONCLUSIONS

The improved pedestrian model with breakable leg elements shows much better correlation to the human leg specimen model than did the model presented in a previous study. The breakable leg model made it possible to predict the risk of ligament failure and bone fracture. The model can also be used to analyse the injury mechanism of the knee and fracture of the leg after a lateral impact to the pedestrian leg in various impact conditions.

Employing subroutines defined by the user in the MADYMO 3D system allows for free formation of input characteristics of elements and also for introducing completely new, more exact models adapted to current needs.

The method of modelling of the long bone viscoelastic properties proposed in this work can also be applied to the modelling of condyles and ligaments of the knee joint as well as other structures of the human body. But to achieve significant improvement of calculated results, it is necessary to carry out extensive experimental investigations that can provide proper dynamic characteristics both of car fronts (bumpers, hood tops, hood edges) and of human extremities (bones, ligaments).

The mathematical model of lower extremity with the human-like knee joint and breakable leg can be used to assess a subsystem impact test proposed for evaluation of car-front aggressiveness against pedestrian.

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