RESPONSE OF TONIC LOWER LIMB FE MODEL IN VARIOUS REAL LIFE CAR-PEDESTRIAN IMPACT CONFIGURATIONS – A PARAMETRIC STUDY FOR STANDING POSTURE

ABSTRACT

This paper investigates the effect of muscle contraction on lower extremity injuries in car-pedestrian lateral impacts. Three variables, viz. height of impact, pedestrian offset with respect to car centre and impact speed, have been considered here. Full scale car-pedestrian FE simulations have been performed using the full body pedestrian model with active lower extremities (PMALE) and front structures of a car model. Two pre-impact conditions of a symmetrically standing pedestrian, representing a cadaver and an unaware pedestrian have been simulated. It is concluded that (1) with muscle contraction risk of ligament failure decreases whereas risk of bone fracture increases (2) ligament and bone strains are dependent on the impact location (3) chances of ligament injuries are higher when the impact occurs near the outer corner of the car (4) risk of bone fracture increases with speed (5) bone fracture reduces the risk of ligament failure.

Keywords: PMALE, Lower extremity model, Finite element model, Dynamic simulation, Muscle contraction, Standing posture, Car-pedestrian impact, Knee injury

INTRODUCTION

In vehicle–pedestrian crashes, lower extremities are the second most frequently injured body region (Chidester et al. 2001; Mizuno 2003). Pedestrian Crash Data Study (PCDS) (Chidester et al. 2001) reports that passenger cars have the biggest share in vehicle-pedestrian accidents. Further, the front bumper is the major source of injury to the lower extremity when injuries were caused by a vehicle structure (Mizuno 2003). This has posed a challenge for vehicle designers to design pedestrian friendly car front structures. However, to devise effective pedestrian protection systems, it is essential to understand knee injury mechanisms.

So far, lower limb injury mechanism in car-pedestrian crashes have been studied through tests on human cadaver specimens (Kajzer et al. 1990, 1993, 1997, 1999; Bhalla et al. 2005) and simulations using validated passive FE models (Schuster et al. 2000; Maeno et al. 2001; Takahashi et al. 2001; Nagasaka et al. 2003; Chawla et al. 2004). However, the major shortcoming in these existing experimental and computational studies is that they do not account for muscle action. Therefore, effects of pre-crash muscle contraction on the response of lower limbs in car-pedestrian crashes have remained unclear. Of late, Soni et al. (2007) have reported a single leg FE model with forty two active muscles (A-LEMS) and investigated the probable outcome of muscle contraction. Later, Chawla et al. (2007) have performed a study using the A-LEMS and reported that with muscle contraction the risk of knee ligament failure is likely to be lower than that predicted through the cadaver tests or simulations with the passive FE models. More recently, Soni et al. (2008) have extended the single leg model (A-LEMS) to a full body pedestrian model with active muscles in both legs (PMALE) and studied the effects of muscle contraction on the response of lower extremity in full scale car-pedestrian lateral impacts using explicit FE simulations. The results of this study have reinforced the previous findings that muscle contraction has reduced the risk of ligament failure.

Vehicle-pedestrian crash data (Chidester et al. 2001) indicates that pedestrian crashes occur in a variety of impact configurations. For example, the location of impact on the lower extremity may be at different heights as the height of bumper varies with the type of a vehicle from passenger car, to minivan, to SUV's. Besides this, pedestrian pre-crash standing position in front of the car may also vary from center to the outer most sides. Since, shape and effective stiffness of the car front varies along the width due to construction; the lower extremity injuries and kinematics are expected to differ.

The present study extends our earlier study to investigate the effects of muscle contraction on the response of lower limb in various real life car-pedestrian lateral impact configurations. Three variables, viz., height of impact, pedestrian offset with respect to the car centre and impact speed, have been considered here. Simulations have been performed using a full body model (PMALE) (Soni et al. 2008) and a car front model at four impact heights on the lower extremity (i.e. mid femur, on-knee, below knee and mid tibia), for four standing positions of a pedestrian in front of the car (central, 150 mm offset, 300 mm offset and 450 mm offset) and with five impact speeds (20, 25, 30, 35 and 40 kmph). Two sets of simulations, i.e. with deactivated muscles and with activated muscles (including reflex action) for an unaware symmetrically standing pedestrian (both legs in side by side stance) have been performed for each impact configuration. Strains in knee ligaments and VonMises stress distribution in the bones are then compared for each impact configuration to assess the effect of muscle contraction.

METHODS

DESCRIPTION OF PMALE

PMALE (Soni et al. 2008) is an assembled full body pedestrian model with active muscles in both legs. Its upper extremity and the lower extremity segments are aligned together so as to represent a symmetrically standing pedestrian with legs in side by side stance. The upper extremity in the PMALE is taken from standard H-III dummy model whereas the lower extremities are made up of A-LEMS (Soni et al. 2007) (Figure 1(a)).

The A-LEMS was a single leg FE model which includes forty two active muscles modeled as 1-D bar elements, in addition to passive hard structures (such as the cortical and the spongy parts of the femur, tibia, fibula, and the patella) and passive soft structures (such as flesh, skin, and four major knee ligaments i.e. ACL, PCL, MCL and LCL). The passive version of A-LEMS has been validated for different sets of loading and boundary conditions reported in Kajzer et al. (1997, 1999) and Kerrigan et al. (2003) (Soni et al. (2007)). The passive model validates for all the test conditions and can correctly reproduce impactor forces, knee kinematics and ligament failures reported from the experiments.

SIMULATION SETUP

Figure 1 show the simulation set up used in the present study. In all thirteen impact configurations have been simulated.

Figure 1(a) shows the simulation set up of four impact configurations used to study the effect of variation in impact height. In these impact configurations (named as mid femur impact (MFI), on-knee impact (OKI), below-knee impact (BKI), and mid tibia impact (MTI)), left leg of the PMALE has been impacted at four different heights in the lateral direction by the central portion of the car front model at a speed of 25 kmph. These locations have been selected to account for variation in height of impact on pedestrian lower extremity due to the varying bumper height in different vehicles.

Figure 1(b) shows the simulation set up of other four impact configurations used to study the effect of variation in shape and effective stiffness of the car front along its width. In these impact configurations (named as central impact, 150 mm offset impact, 300 mm offset impact and 450 mm offset impact), PMALE is placed in front of the car model at four different positions so that it interacts with the front bumper at four different locations. These offset impact locations have been selected to investigate how variation in shape and effective stiffness of the car front from center to the outer most sides, mainly due to vehicle construction, effect lower extremity injuries.

FIGURE 1

In the remaining five impact configurations (Figure 1(c)) (named as 20 kmph, 25 kmph, 30 kmph, 35 kmph and 40 kmph) effect of variation in the impact speed has been studied. In these impact configurations, left leg of the PMALE has been impacted laterally at the knee level by the central portion of the car front with five different speeds.

In each simulation, PMALE is configured as standing freely with legs in a side by side stance on a rigid plate in a gravity field. Friction coefficient between the shoes and the plate has been set to 1.0. Car model with a total mass of 1158 kg (mass of the front structures is 355 kg and 803 kg is modeled as added mass to account for the remaining car structures) is propelled towards the PMALE to impact its left leg laterally in various impact configurations as shown in Figure 1.

PEDESTRIAN PRE-IMPACT CONDITIONS

Two pre-impact pedestrian conditions, i.e., with deactivated muscles and with activated muscles (including reflex action) for an unaware pedestrian have been simulated for each impact configuration in the present study. We call these conditions cadaveric and reflex conditions respectively. These conditions differ in terms of initial activation levels in muscles and whether the reflex action is enabled. By enabling the reflex action for a muscle, the activation level in that muscle rises with time during the simulation; this increases the force produced by that muscle.

In the cadaveric condition, a freely standing cadaver has been simulated. To model a cadaver in FE simulation, all the muscles in PMALE have been assigned the minimum activation level of 0.005. The reflex action is also disabled. Therefore, in this condition all the muscles function at their minimum capacity. In the reflex condition, a symmetrically standing pedestrian who is unaware of an impending crash has been simulated. To model an unaware standing pedestrian, activation levels required to maintain the stability of the standing posture (Kuo et al. 1993) are assigned in the Hill material card of each muscle. The reflex action is also enabled in this condition. Additional modeling details of these conditions have been presented in Chawla et al. (2007).

DATA ANALYSIS

Element elimination approach has been enabled to simulate the failure in the ligaments and the bones. Strain time history of each knee ligament and VonMises stress contours in bones of the impacted leg i.e. left leg of the PMALE have been recorded from the simulations. Response in cadaveric and reflex conditions has then been compared to determine the role of muscle contraction.

RESULTS

EFFECT OF MUSCLE CONTRACTION IN IMPACT AT DIFFERENT HEIGHTS

Strain in Knee Ligaments: Figure 2 compares peak strain in knee ligaments of the impacted leg (i.e. left leg of the PMALE) for four impact configurations (i.e. MFI, OKI, BKI and MTI, at different heights) in both cadaveric and reflex conditions. It is evident from Figure 2 that the strain in knee ligaments varies with the impact height. It is observed that for all four impact configurations, peak strain in the MCL has remained higher than that in other knee ligaments (i.e. ACL, PCL and LCL) whereas the LCL has remained nearly unstrained. This can be attributed to the lateral impact which forces tibia to bend medially and consequently stretches the MCL and keeps the LCL slackened.

It is apparent from Figure 2 that ligament strains have significantly reduced in the reflex condition as compared to the cadaveric condition for each impact configuration. In ACL, peak strain value has dropped by a factor of 1.108, 1.3, 1.2 and 1.23 for MFI, OKI, BKI and MTI respectively. It is observed that reduction in peak strain is more prominent in PCL (Figure 2) than in the ACL. Peak strain in PCL has reduced by a factor of 7.3, 14, 1.67 and 11 in the reflex enabled condition as compared to the cadaveric condition for MFI, OKI, BKI and MTI respectively.

FIGURE 2

In MCL, for both MFI and BKI configurations, peak strain has reached the ligament failure limit of 15% in the cadaveric conditions whereas in the reflex conditions it has reduced to 3.77% and 14.5% respectively and hence prevented from failure. For the MFI configuration, reduction in peak MCL strain in the reflex condition can be ascribed to fracture in the femur shaft. It is found that, in the cadaveric condition the MCL failed (at around 41.4 ms) before the fracture has occurred in the femur shaft (at 46 ms); whereas, in the reflex condition, fracture in the femur occurred (around 38 ms) 8 ms earlier than in the cadaveric condition and hence relieved the strain in the MCL. For the OKI configuration, peak MCL strain has reached the ligament failure limit of 15% in both the conditions. However, failure is delayed by 5.2 ms in the reflex condition (35 ms) than that to the cadaveric condition (29.8 ms). It is found that, for the MTI configuration, peak strain in MCL has remained below 4% in both the conditions.

VonMises Stresses in Bones: Figure 3 compares the VonMises stress distribution in bones of the impacted leg (i.e. left leg of the PMALE) for MFI, OKI, BKI and MTI configurations in cadaveric and reflex conditions. It is apparent from Figure 3 that stresses in bones have increased significantly in the reflex condition as compared to the cadaveric condition for each impact configuration.

FIGURE 3

It is observed that, in MFI configuration (Figure 3 (a)) mid femur shaft has been stressed to its ultimate stress limit (123.3 MPa) in both cadaveric and reflex conditions and consequently fractured in the simulations. However, femur fracture in the reflex condition (38 ms) has occurred 8 ms earlier than that in cadaveric condition (46 ms). In both OKI and BKI configurations (Figure 3 (b) and Figure 3 (c)), the medial side of mid tibia and fibula is stressed to their ultimate stress limits (138.1 MPa) in the reflex conditions and thus fractured in the simulations at around 48 ms. Whereas, stress in tibia and fibula remained below the failure limits in the cadaveric conditions of both the impact configurations (i.e. peak stress 116 MPa in OKI and peak stress 63 MPa in BKI). In MTI configuration, stress in bones remained below the failure limits in both cadaveric and reflex conditions and hence no fracture has been observed in these simulations. However, in comparison to the cadaveric condition (peak stress 108 MPa), stress in the medial side of mid tibia and fibula has increased significantly in the reflex condition (peak stress 134 MPa) (Figure 3 (d)).

EFFECT OF MUSCLE CONTRACTION IN IMPACTS WITH VARYING OFFSETS

Strain in Knee Ligaments: Figure 4 compares peak strain in knee ligaments of the impacted leg (i.e. left leg of the PMALE) for central, 150 mm, 300 mm and 450 mm offset configurations in both cadaveric and reflex conditions.

FIGURE 4

It is observed from Figure 4 that in ACL and PCL (see Figure 4), peak strain has increased significantly in the 150 mm offset configuration as compared to the central impact configuration in both cadaveric (1.45 and 1.72 times in ACL and PCL respectively) and reflex conditions (1.34 and 9.51 times in ACL and PCL respectively). The increase in ACL and PCL peak strains is even more noticeable (see Figure 4) in the 300 mm impact configuration. In this configuration (i.e. in 300 mm) peak strain in both the ligaments (i.e. ACL and PCL) has reached to the failure limit of 15% in both cadaveric and reflex conditions. Further, in the 450 mm offset impact configuration, peak strain in ACL has reached the failure limit (i.e. 15%) in both cadaveric and reflex conditions; however, it is interesting to find that peak strain in the PCL has remained below 3% in cadaveric as well as reflex conditions. It has been noticed that in this impact configuration (i.e. 450 mm offset) curvature in the front bumper near to the outermost side has induced torsion in the impacted leg (i.e. left leg of the PMALE). As a result, the lateral impact has gradually shifted to the posterior impact. This has eventually forced the left knee joint to flex and has hence reduced the PCL strain.

It is found that in all the four impact configurations, peak strain in the MCL has reached the failure limit (i.e. 15%) whereas LCL has remained nearly unstrained in both cadaveric and reflex conditions (Figure 4). This can be attributed to the lateral impact which forces tibia to bend medially and consequently stretches the MCL and keeps the LCL slackened.

Figure 4 illustrates that in comparison to the cadaveric condition, active muscle forces in the reflex condition have significantly reduced the peak strains in ACL and PCL for the central (by a factor of 1.3 and 13.89 in ACL and PCL respectively) and the 150 mm impact configurations (by a factor of 1.4 and 2.5 in ACL and PCL respectively). However, in the remaining impact configurations (i.e. 300 mm and 450 mm), no significant effects of muscle activation on the peak strains of any knee ligament have been noticed in Figure 4. Even active muscles in the reflex condition could not prevent ACL failure in the 300 mm and the 450 mm impact configurations, PCL failure in the 300 mm impact configuration and MCL failure in any of the four impact configurations. However, it is interesting to notice that failure in ACL, PCL and MCL has been delayed (Table 1) significantly (e.g., by 10.5 ms for the ACL in the 450 mm offset case, by 2 ms for the PCL in the 300 mm offset case and by 16.3 ms for the MCL in the 150 mm offset case) in the reflex condition as compared to the cadaveric condition.

Table 1

VonMises Stresses in Bones: Figure 5 compares the VonMises stress distribution in bones of the impacted leg (i.e. left leg of the PMALE) for central, 150 mm, 300 mm and 450 mm offset impact configurations in cadaveric and reflex conditions. It is apparent from Figure 5 that stresses in bones have increased significantly in the reflex condition as compared to the cadaveric condition for each impact configuration.

In the central impact configuration (Figure 5 (a)), the medial side of mid tibia and fibula is stressed to the ultimate stress limit (138.1 MPa) in the reflex conditions. The tibia and the fibula have thus fractured in the simulation at around 48 ms. Whereas, stress in the tibia and fibula remained below the failure limits in the cadaveric conditions (peak stress 116 MPa). In 150 mm, 300 mm and 450 mm offset configurations, the bone-stress remained below the failure limits in both cadaveric and reflex conditions and hence no fracture has been observed. However, in comparison to the cadaveric condition, stresses in bones are significantly higher (in the range of 104 MPa to 138.1 MPa) in the reflex condition (peak stress values are given in Figure 5 (b-d)).

Figure 5

EFFECT OF MUSCLE CONTRACTION IN IMPACTS AT DIFFERENT SPEEDS

Strain in Knee Ligaments: Figure 6 compares peak strain in knee ligaments of the impacted leg (i.e. left leg of the PMALE) for 20 kmph, 25 kmph, 30 kmph, 35 kmph and 40 kmph impacts in both cadaveric and reflex conditions. It is apparent from Figure 6 that ligament strains have significantly dropped in the reflex condition as compared to the cadaveric condition for each impact configuration.

In ACL, peak strain value has dropped by a factor of 1.17, 1.3, 3, 1.53 and 5.06 in the reflex condition as compared to that in the cadaveric condition for 20 kmph, 25 kmph, 30 kmph, 35 kmph and 40 kmph impacts respectively. It is observed (Figure 6) that, for 30 kmph and 40 kmph configurations, peak ACL strain has reached to the failure limit of 15% in the cadaveric condition whereas in the reflex condition it is prevented from failure. This is due to the fracture in tibia shaft (at 38 ms and 26 ms respectively) (Figure 7) which has relieved the strain in the ACL.

In the PCL, reduction in peak strain is more prominent (see Figure 6) than in the ACL. For 20 kmph, 25 kmph, 30 kmph and 35 kmph impacts, peak strain in PCL has reached up to 2.40%, 4.03%, 4.38% and 4.02% respectively in the cadaveric condition whereas in the reflex condition it remained nearly 0% in the entire duration of the simulation. Similarly, for the 40 kmph impact, peak strain in the PCL is 5.33% in the cadaveric condition whereas in the reflex condition it is only 1.49%. This can be attributed to active muscle forces in the reflex condition which have pulled the tibia towards the femur and therefore; tibia displacement in inferior direction (away from the femur) has reduced. This has slackened the PCL in the reflex condition.

Figure 6

In the MCL, the peak strain has reached the failure limit of 15% for each impact configuration in the cadaveric as well as reflex condition, except in the reflex condition (12.7%) for the 20 kmph impact. However, it is noticeable that active muscles have delayed the MCL failure by 6-7 ms for each of these cases. This indicates that active muscles in the reflex condition would prevent failure in the MCL for low speed (below 20 kmph) lateral impacts, however, for impact speed above 20 kmph; the energy of impact is sufficiently high that it overpowers the resistance provided by the muscle forces in the reflex condition.

In LCL, peak strain has remained nearly 0% for all the cases. This is because in lateral impact tibia bends medially and consequently slackens the LCL.

VonMises Stresses in Bones: Figure 7 compares the VonMises stress distribution in bones of the impacted leg (i.e. left leg of the PMALE) for the 20 kmph, 25 kmph, 30 kmph, 35 kmph and 40 kmph impacts in the cadaveric and reflex conditions. It is apparent from

Figure 7 that stresses in bones have increased significantly in the reflex condition as compared to the cadaveric condition for each impact configuration.

Figure 7

In the 20 kmph impact (Figure 7 (a)), VonMises stresses in the medial side of mid tibia and mid fibula regions are significantly higher in the reflex condition (peak stress 134 MPa) than that in the cadaveric condition (peak stress 85 MPa). However, in both cadaveric and reflex conditions, stresses in the bones have not exceeded the failure limits and hence no failure has been noticed. For the 25 kmph and 30 kmph impacts (Figure 7 (b) and (c)), stresses in the medial side of mid tibia and mid fibula regions have increased to their ultimate stress limits (138.1 MPa) in the reflex conditions. In these simulations the tibia and the fibula have consequently fractured at around 48 ms and 38 ms respectively. Whereas, in the cadaveric conditions, stresses in tibia and fibula remained below the failure limits (i.e. peak stress 116 MPa and 132 MPa in 25 kmph and 30 kmph respectively). For 35 kmph and 40 kmph impacts (Figure 7 (d) and (e)), it is observed that, in both cadaveric and reflex conditions stresses in the medial side of mid tibia and mid fibula regions have reached their ultimate stress limits. Hence, these bones have fractured in these simulations. This can be attributed to the higher energy of impact (corresponding to the impact speeds) employed in these configurations (i.e. 35 kmph and 40 kmph).

DISCUSSION

In the present study, effects of active muscle forces on the knee ligament injuries and lower-leg bone fracture for various real life car-pedestrian lateral impact configurations has been investigated using FE simulations. In all, thirteen impact configurations have been studied. For each impact configuration, two pre-impact conditions, that representing a cadaver and an unaware pedestrian, have been considered. It is observed that active muscle forces have significant effects on the lower extremity loading.

In four configurations (i.e. MFI, OKI, BKI, and MTI), effect of variation in heights has been studied. These impact locations have been selected to account for variation in height of impact due to the varying bumper height in different vehicles. It is observed that among the four impact configurations, risks of both ligament failure and bone fracture is the lowest in the MTI configuration. This suggests that chances of lower extremity injuries in car-pedestrian lateral impacts are likely to reduce if height of car front bumper is designed such that it impacts the lower extremity on or below the mid tibia level.

In four configurations with varying offsets (i.e. central, 150 mm, 300 mm and 450 mm offset); effect of pedestrian position on the muscle contribution has been investigated. It is observed that risk of lower extremities injuries have significantly increased for the offset impact configurations (i.e. 450 mm, 300 mm, and 150 mm) as compared to the central impact configuration. This is because the effective stiffness of the car front increases (due to bumper support) along the width from center to the outer most sides. As a result, more energy is imparted to the lower extremity segments, thereby altering the kinematics and hence the injury risk. This implies that additional compliance of the bumper at the outer sides could help in reducing knee injuries in pedestrian impacts.

In the remaining five impact configurations, effect of impact speeds (i.e. 20 kmph, 25 kmph, 30 kmph, 35 kmph, and 40 kmph) on muscle contribution has been studied. It is found that risk of both ligament failure and bone fracture increases with impact speed, however, for high speed impacts, chances of bone fracture are considerably higher than ligament failure. This is because; fracture in bones absorbs the impact energy which eventually unloads the knee joint and hence reduces the strain in knee ligaments. It is also observed that at all the impact speeds, stresses in tibia and fibula were consistently higher in the reflex condition as compared to the cadaveric condition. This suggests that active muscle forces in the reflex condition further increase the risk of bone fracture.

CONCLUSIONS

In the present study, effects of muscle contraction on the response of lower limb in lateral impacts at four heights (mid femur, on-knee, below knee and mid tibia) on the lower extremity, for four standing positions of a pedestrian in front of the car (central, 150 mm offset, 300 mm offset and 450 mm offset) and with five speeds (20, 25, 30, 35 and 40 kmph) have been studied. Two sets of simulations, i.e. with deactivated muscles (representing a cadaver) and with activated muscles (including reflex action) for an unaware symmetrically standing pedestrian (both legs in side by side stance) have been performed for each impact configuration. Full scale car-pedestrian lateral impact simulations have been performed using PMALE and a validated car front model. Differences in response between a cadaver and an unaware pedestrian have been then studied. To assess the effect of muscle activation, strains in knee ligaments, VonMises stress in bones of the impacted leg (i.e. left leg of the PMALE) have been compared. We conclude that:

1. For all the impact configurations, peak strains in the knee ligaments were lower in the reflex condition (with active muscles) as compared to the cadaveric condition. This reinforces our previous findings that the risk of ligament failure in real life crashes is likely to be lower than that predicted through cadaver tests or simulations.

2. For all impact configurations, VonMises stresses in bones were significantly higher in the reflex condition as compared to the cadaveric condition. This leads to the conclusion that chances of bone fracture increase with muscle contraction.

3. Knee ligament strains and stresses in lower extremity bones depend on impact location. It is observed that as impact height was changed, ligament strains and bone stresses were lowest in the mid-tibia-impact configuration. This suggests that for reducing the risk of lower extremity injuries in car-pedestrian lateral impacts, bumper height should be reduced.

4. Risk of ligament injuries is significantly higher for the offset impact configurations (i.e. 450 mm, 300 mm, and 150 mm) as compared to the central impact configuration. This suggests that the effective bumper stiffness towards the ends needs to be reduced.

5. Risk of ligament failure and bone fracture increases with impact speed, and, chances of bone fracture are considerably higher than ligament failure for high speed impacts.

6. Bone fracture unloads the knee joint and hence reduces the chances of ligament failure.

ACKNOWLEDGEMENTS

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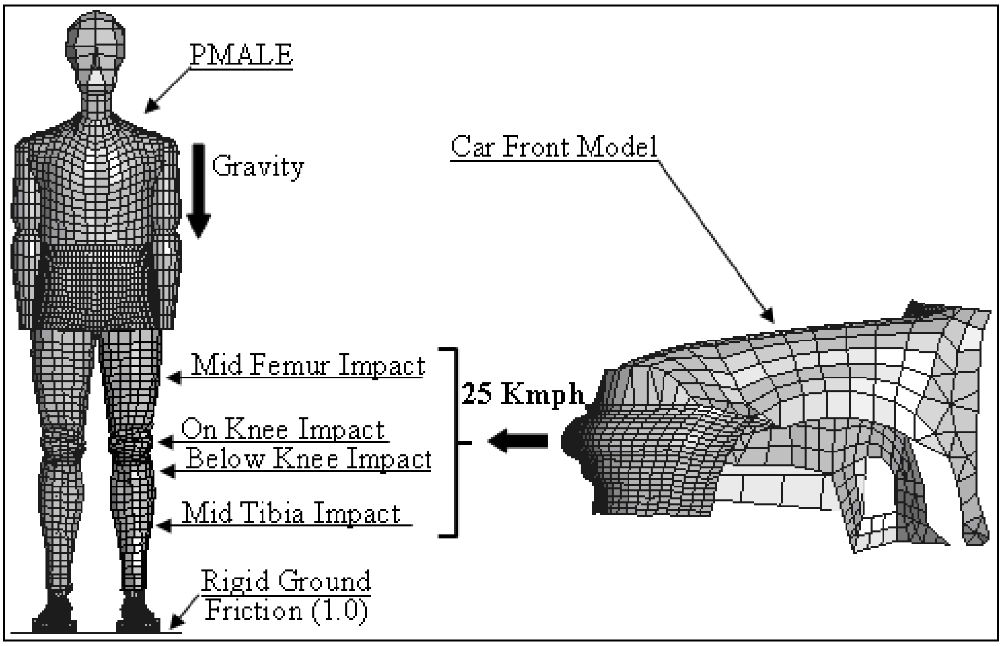
Table 1 Comparison of failure occurrence time in ACL, PCL and MCL

	ACL		PCL		MCL	
	Cadaveric	Reflex	Cadaveric	Reflex	Cadaveric	Reflex
Central	-	-	-	-	29.8 ms	35 ms
150 mm	-	-	-	-	27.1 ms	43.4 ms
300 mm	25.2 ms	32.9 ms	39.6 ms	41.6 ms	17.6 ms	22.4 ms
450 mm	16.3 ms	26.8 ms	-	-	11.3 ms	14.1 ms



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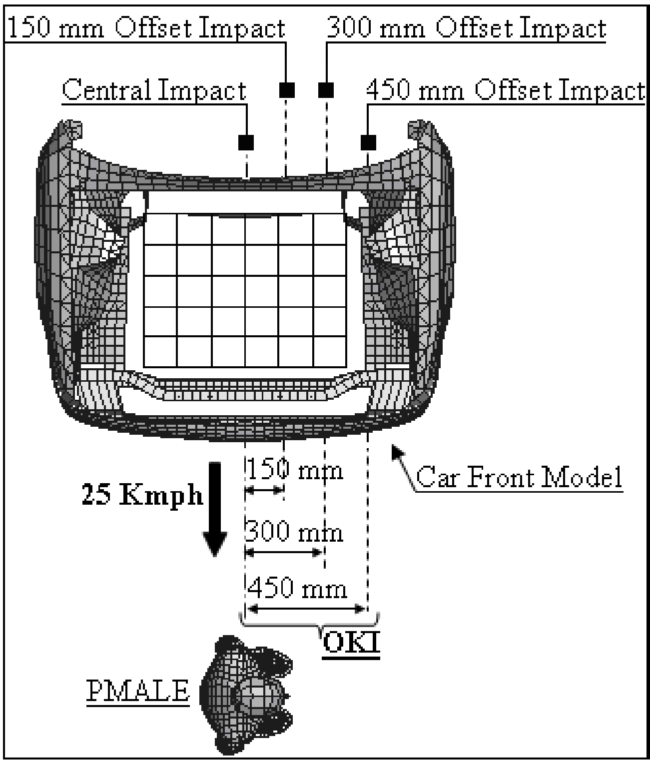

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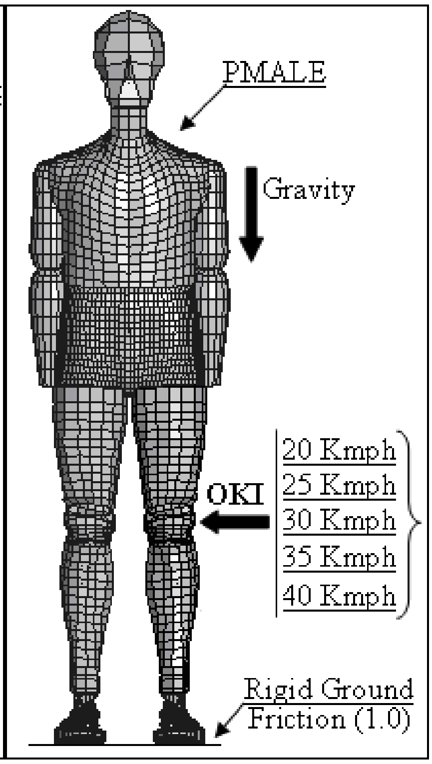

(c)

Figure 1 Simulation setup to study the effects of muscle forces in a symmetrically standing pedestrian for (a) impact at four heights (b) four standing positions in front of car and (c) five impact speeds

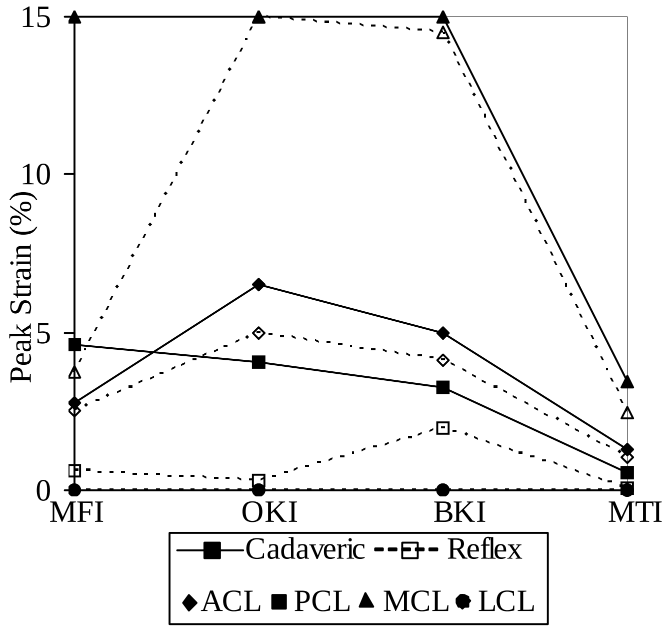

Figure 2 Comparison in peak ligament strains for MFI, OKI, BKI, and MTI configurations in both cadaveric and reflex conditions

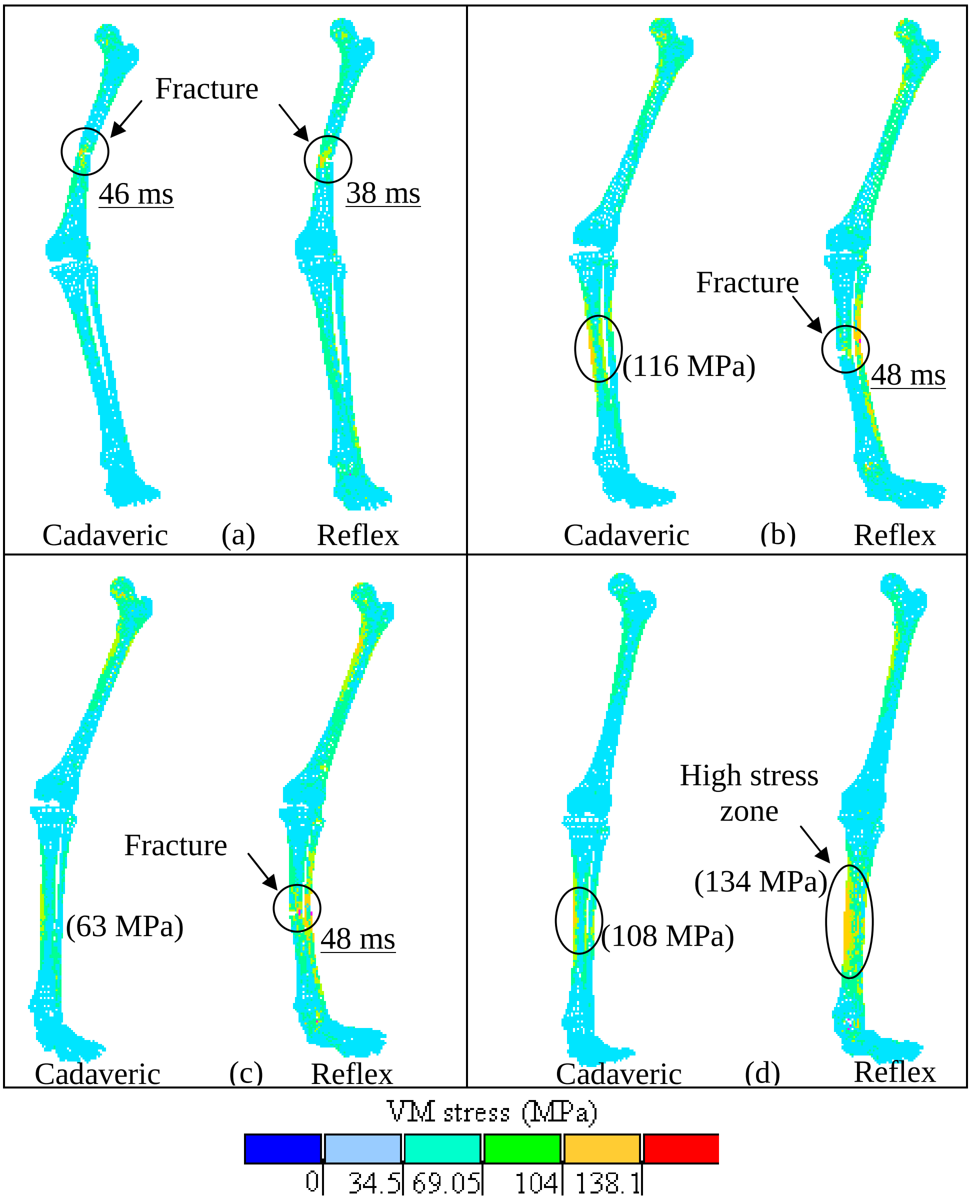

Figure 3 VonMises stress distribution in bones (peak stress values are also given) for (a) MFI (b) OKI (c) BKI and (d) MTI configurations in both cadaveric and reflex conditions

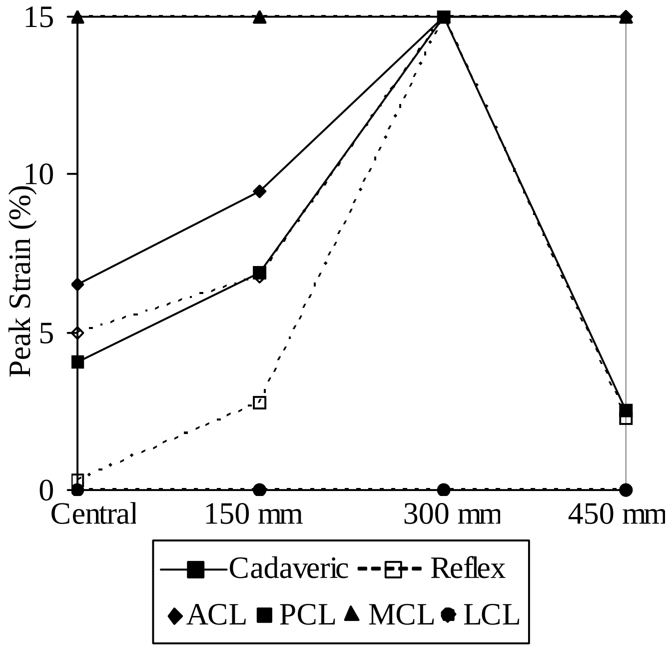

Figure 4 Comparison in peak ligament strains for central, 150 mm, 300 mm and 450 mm impact configurations in both cadaveric and reflex conditions

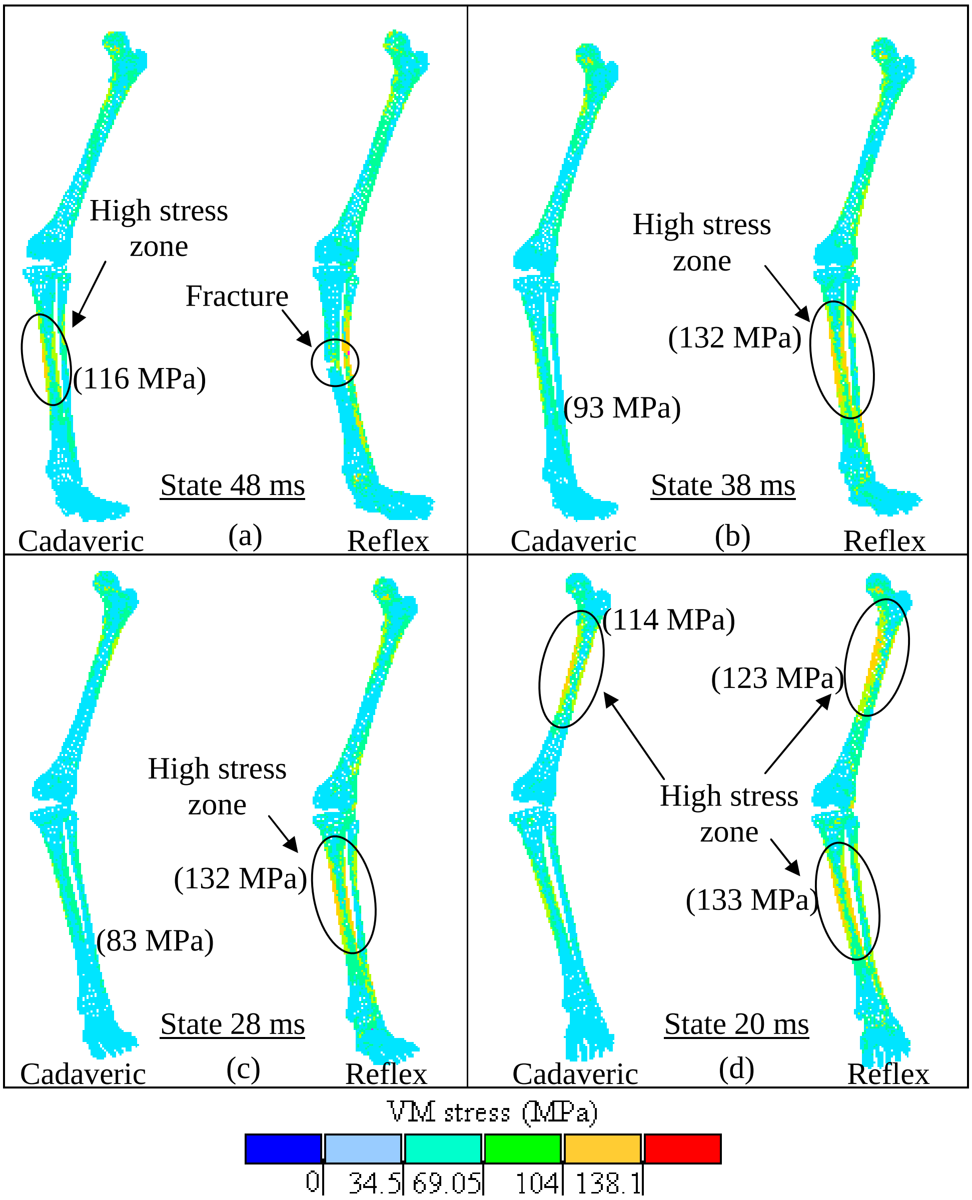

Figure 5 VonMises stress distribution in bones (at time of maximum stress) for (a) central (b) 150 mm (c) 300 mm and (d) 450 mm offset configurations in both cadaveric and reflex conditions

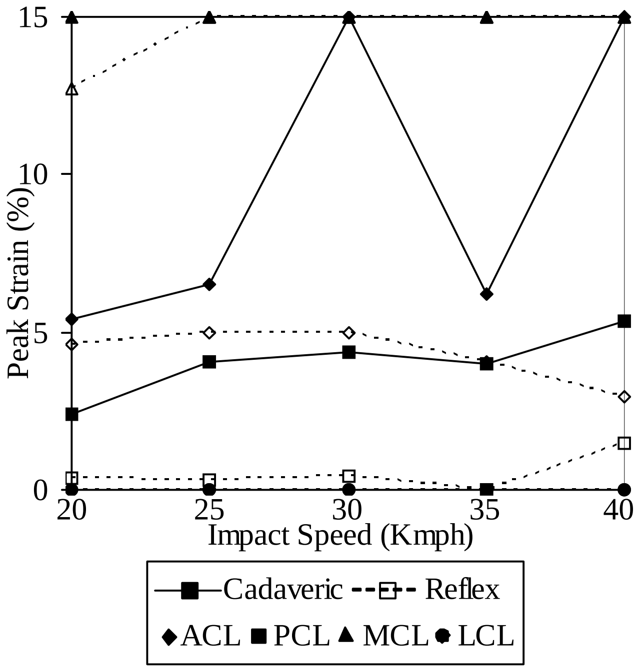

Figure 6 Comparison in peak ligament strains for impact at different speeds in both cadaveric and reflex conditions

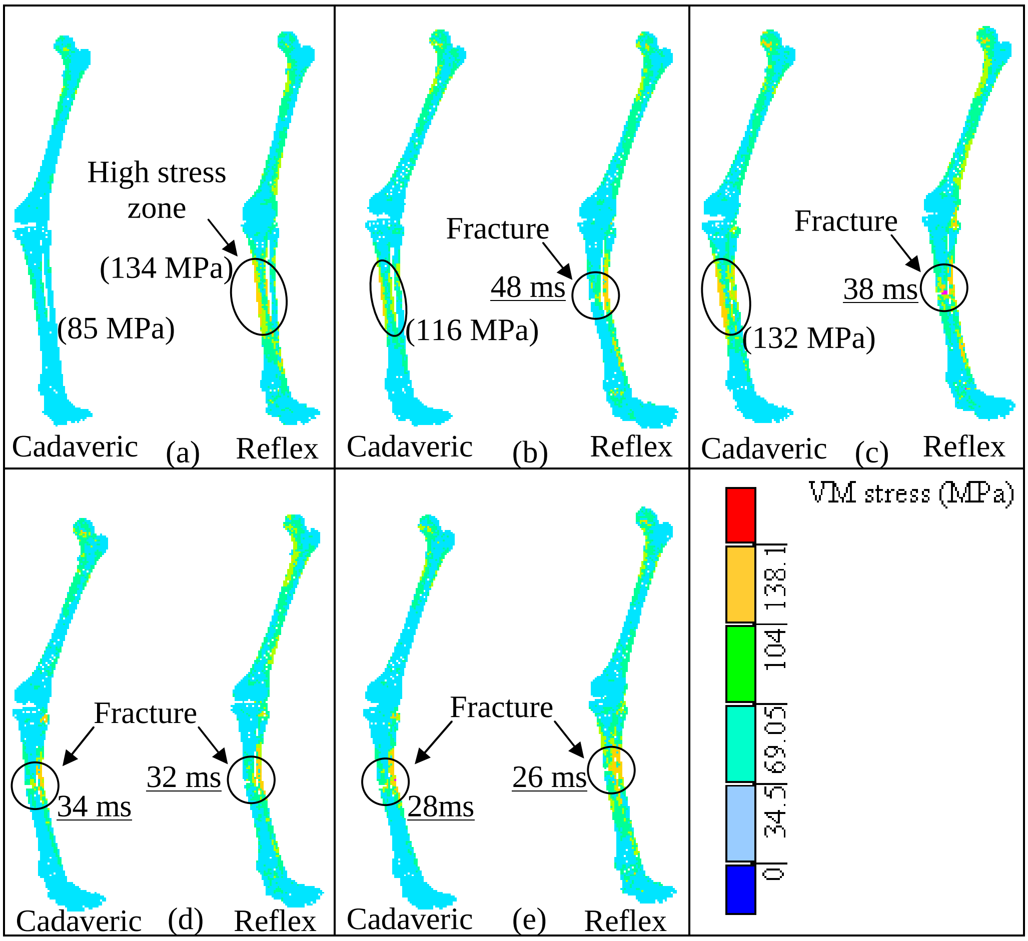

Figure 7 VonMises stress distribution in bones (peak stress values are also given) for (a) 20 kmph (b) 25 kmph (c) 30 kmph (d) 35 kmph and (e) 40 kmph impact configurations in both cadaveric and reflex conditions