

Effect of muscle contraction on the lower limb response in low speed car–pedestrian lateral impact: simulations for a walking pedestrian

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Q1: Au: Please mention article acceptance date.

Q2: Au: Please spell out abbreviations MCL, PCL and LCL.

Q3: Au: (1) The reference to Table 1 seems incorrect here (as there is no Table 1). Should it be Figure 1 instead? (2) The reference Mizuno et al. (2003) has been treated (here and elsewhere in the text) as Mizuno (2003). Is this correct?

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Effect of muscle contraction on the lower limb response in low speed car–pedestrian lateral impact: simulations for a walking pedestrian

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(Received 30 June 2009; final version received xx xx xxxx)

This paper investigates the effect of muscle contraction on lower extremity injuries for pedestrian walking posture in a car–pedestrian lateral impact at low speed. The full body model, pedestrian model with active lower extremities (PMALE), which was configured in a symmetric standing posture, has been repositioned in the walking posture. Finite-element simulations have then been performed using the PMALE in walking posture and front structures of a car. Two impact configurations, i.e. impact on the right and left legs, have been simulated. Two pre-impact conditions, that of a symmetrically standing pedestrian, representing a cadaver and an unaware pedestrian have been simulated for both the impact configurations. Stretch-based reflexive action was included in the simulations for an unaware pedestrian. It is concluded that (1) with muscle contraction, the risk of ligament failure decreases whereas the risk of bone fracture increases; (2) in lateral impacts, MCL could be considered as the most vulnerable and LCL as the safest ligament; and (3) for a walking pedestrian, PCL would be at a higher risk in the case of impact on the rear leg, whereas ACL would be at a higher risk if car strikes the front leg.

Keywords: PMALE; lower extremity model; finite-element model; dynamic simulation; muscle contraction; standing posture; walking posture; car–pedestrian impact; knee injury

Introduction

Pedestrians constitute 65% of the 1.17 million people killed annually in road traffic accidents worldwide (World Bank 2001). Epidemiological studies on pedestrian victims have indicated that together with the head, lower extremities are the most frequently injured body region ([7,14]). Pedestrian crash data study (PCDS; [7]) reports that passenger cars have the biggest share in vehicle–pedestrian accidents. Furthermore, the front bumper was the major source of injury to the lower extremity when injuries were caused by a vehicle structure [14]. This has posed a challenge for vehicle designers to design pedestrian-friendly car front structures. To devise effective pedestrian protection systems, it is essential to understand the injury mechanism.

So far, the lower limb injury mechanism in car–pedestrian crashes has been studied through tests on human cadaver specimens [2, 8–11] and simulations using validated passive lower limb computational models [1,3,4,13–17,20,22]. 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 remained unclear.

Of late, Chawla et al. [5,6] have investigated the probable outcome of muscle contraction using a lower limb

(single leg) finite-element (FE) model with active muscles (A-LEMS). More recently, Soni et al. [18,19] have extended the single leg model A-LEMS to a full body pedestrian model with active lower extremities (PMALE) and studied the effects of muscle contraction on the response of lower extremity for a symmetrically standing pedestrian (with legs in side-by-side stance) in full-scale car–pedestrian impact. However, pedestrian crash data study (PCDS; [7]) reported that prior to the crash, only 4% pedestrians were in the stationary standing posture whereas 55% were walking.

This study extends our earlier studies to investigate the effect of muscle contraction on the response of lower limb for pedestrian walking posture in full-scale car–pedestrian lateral impact at low speed. The PMALE, which was configured in the standing posture, has been repositioned in the walking posture. The real-world car–pedestrian lateral impact has then been simulated using the PMALE configured in the walking posture (PMALE-WP) and front structures of a validated car FE model. Two impact configurations, i.e. impact on the right and left legs of PMALE-WP, have been simulated. Two sets of simulations, i.e. with deactivated and activated muscles (including reflex action), mimicking an unaware walking pedestrian have been performed for both the impact configurations. Strains in knee ligaments and VonMises stresses in bones for two levels of muscle

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activation have then been compared to assess the effect of muscle contraction.

70 Methods

PMALE in walking posture

In this study, PMALE [19], which was configured in the symmetrically standing posture of a pedestrian with legs in side-by-side stance, has been repositioned in the walking posture. Relative angles between the body segments required to define the alignment of the walking posture (Table 1) have been taken from Mizuno [14]. A series of FE

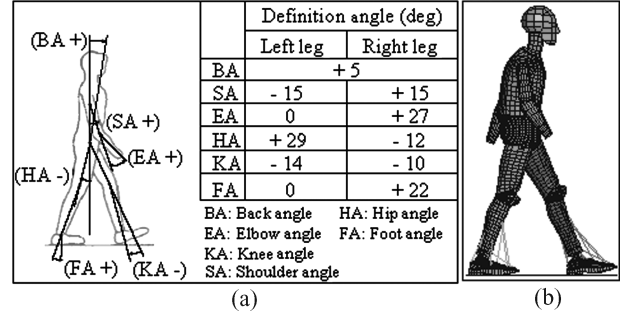


Figure 1. (a) Definition angles with sign conventions for pedestrian walking posture [14] and (b) PMALE in walking posture (i.e. PMALE-WP).

Table 1. Activation levels used in this study.

Lower extremity muscles	Activation levels	
	Left leg	Right leg
Vastus lateralis	0.5*	0.1*
Vastus intermedius	0.005	0.005
Vastus medialis	0.005	0.005
Rectus femoris	0.5*	0.1*
Soleus	0.2*	0.35*
Gastrocnemius medialis	0.2*	0.2*
Gastrocnemius lateralis	0.2*	0.3*
Flexor hallucis longus	0.005	0.005
Flexor digitorum longus	0.005	0.005
Tibialis posterior	0.005	0.005
Tibialis anterior	0.4*	0.1*
Extensor digitorum longus	0.4*	0.1*
Extensor hallucis longus	0.005	0.005
Peroneus brevis	0.005	0.005
Peroneus longus	0.4*	0.2*
Peroneus tertius	0.005	0.005
Biceps femoris (LH)	0.4*	0.1*
Biceps femoris (SH)	0.4*	0.1*
Semimembranosus	0.4*	0.1*
Semitendinosus	0.4*	0.1*
Piriformis	0.005	0.005
Pectineus	0.005	0.005
Obturator internus	0.005	0.005
Obturator externus	0.005	0.005
Gracilis	0.005	0.005
Adductor brevis 1	0.005	0.005
Adductor brevis 2	0.005	0.005
Adductor longus	0.5*	0.5*
Adductor magnus 1	0.25*	0.1*
Adductor magnus 2	0.25*	0.1*
Adductor magnus 3	0.25*	0.1*
Gluteus maximus 1	0.5*	0.15*
Gluteus maximus 2	0.5*	0.15*
Gluteus maximus 3	0.5*	0.15*
Gluteus medius 1	0.5*	0.05*
Gluteus medius 2	0.5*	0.05*
Gluteus medius 3	0.5*	0.05*
Gluteus minimus 1	0.005	0.005
Gluteus minimus 2	0.005	0.005
Gluteus minimus 3	0.005	0.005
Sartorius	0.4*	0.25*
Tensor fascia lata	0.005	0.005

Note: Activation levels labelled with asterisk are taken from Winter [21].

simulations have been performed with the PMALE (configured in the standing posture) to reposition its body segments in the walking posture. Figure 1 shows the PMALE configured in the walking posture (referred to as the PMALE-WP). In PMALE-WP, the left leg is placed in front of the right leg as shown in Figure 1. Here, PMALE-WP is configured such that its left leg (positioned in front) corresponds to the heel strike phase of the human gait cycle, whereas the right leg (positioned in rear) corresponds to the terminal stance phase. Upper body is leaned forward by five degrees with the vertical axis.

Simulation set-up

Figure 2 shows the simulation set-up used in this study. Here, the real-world car-pedestrian impact has been reproduced using the PMALE-WP and front structures of a validated car FE model. PMALE-WP represents a pedestrian walking on rigid ground in gravity field. The friction coefficient between the shoes and the ground has been set to 1.0. A car model with a total mass of 1158 kg (mass of the front structures is 355 kg and 803 kg is modelled as an added mass to account for the remaining car structures) is propelled with a speed of 25 km/h towards the PMALE-WP. Since in real-world car-pedestrian crashes, a moving car may strike a walking pedestrian from his or her right or left. Therefore, two impact configurations, i.e. impact on lateral side of the right leg (Figure 1a) and the left leg (Figure 1b), have been simulated. In both the impact configurations, the PMALE-WP is placed in front of the car model such that it interacts with mid-portion of the bumper, whereas the car model is positioned at a height above the ground such that it corresponds to the car rolling on its tyres.

Pedestrian pre-impact conditions

Two pre-impact pedestrian conditions, i.e. with deactivated and activated muscles (including reflex action) for an unaware pedestrian, have been simulated for both the impact configurations in the present study. These conditions

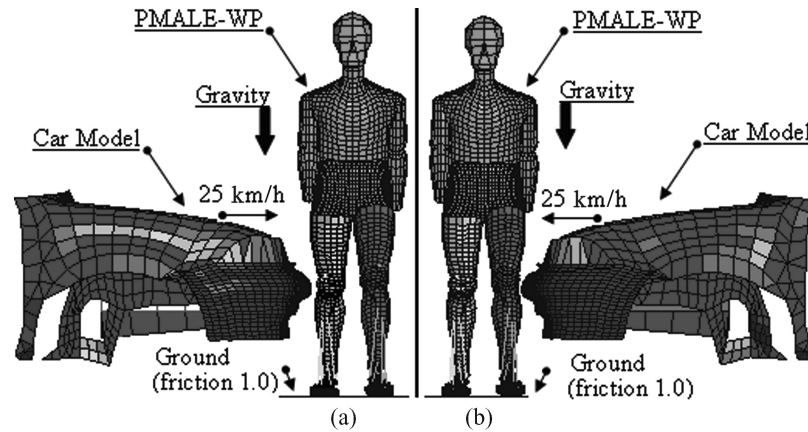


Figure 2. Simulation set-up used in this study for impact on (a) the right and (b) left legs.

are called the 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.

Cadaveric condition. In this condition, a cadaver which is aligned in a walking posture has been simulated. To model a cadaver in the FE simulation, all the muscles in the PMALE-WP have been assigned the minimum value of 0.005 as the initial activation level. The reflex action is disabled. As a result, in this condition, activation levels in each muscle remain at the minimum value (i.e. 0.005) for the entire duration of the simulation. Therefore, all the muscles function at their minimum capacity.

Reflex condition. In this condition, a pedestrian who is walking on road and is unaware of an impending crash has been simulated. Prior to the impact, the pedestrian's right leg (in rear) is in the terminal stance phase (i.e. right heel is about to leave the ground) of the human gait cycle and the left leg (in front) is in the heel strike phase (i.e. left heel has just landed on the ground). To model an unaware pedestrian in such a walking stance, right leg muscles have been assigned the activation levels corresponding to the terminal stance i.e. at 60% gait, whereas muscles in the left leg have been assigned the activation levels corresponding to the heel strike, i.e. at 0% gait. Values of these muscle activation levels (Table 1) have been taken from the electromyography (EMG) activities recorded in volunteers during the gait cycle by Winter [21].

A stretch-based involuntary reflex action has also been enabled in this condition. For enabling the reflex, a threshold value of elongation is to be defined in Hill material card of a muscle. When the elongation in muscle crosses the threshold value, stretch reflex in a muscle gets activated. However, the increase in muscle force starts only after a certain time known as reflex time. This delay between the activation

of stretch reflex and the onset of increase in muscle force represents the time taken by the signal to travel through the central nervous system (CNS) circuitry (muscle-spinal cord-muscle). A delay of 20 ms has been assigned to all the muscles in PMALE-WP (Ackerman 2002). This mimics the ability of live muscle to respond to a small stretch produced by an outside agency. In medical terms, this kind of reflex action is known as the 'stretch reflex' (Vander et al. 1981).

Data analysis

Two nodes at both femur and tibia have been selected to obtain the nodal time history in simulations. Relative movements of selected nodes are then used to calculate relative tibia displacements and knee joint angles. Sign conventions used here are as per the SAE standards. An element elimination approach has been enabled to simulate the failure in the ligaments and bones. Knee kinematics, strain time history of each knee ligament and VonMises stress contours in bones of the impacted leg of PMALE-WP have been recorded from the simulations. The response in cadaveric and reflex conditions has then been compared to determine the role of muscle contraction.

Results and discussion

In all, four simulations, each of 100 ms duration, have been performed in the present study. For the first 50 ms (stabilisation duration), PMALE-WP has been stabilised under gravity load in each simulation. At the end of first 50 ms, car front impacts the lower extremity of the stabilised PMALE-WP. Knee kinematics, ligament strains and VonMises stresses in bones have been recorded from these simulations to assess the effect of muscle contraction. Results presented here are for the impact duration, and the initial time (i.e. 0 ms) corresponds to the time of contact.

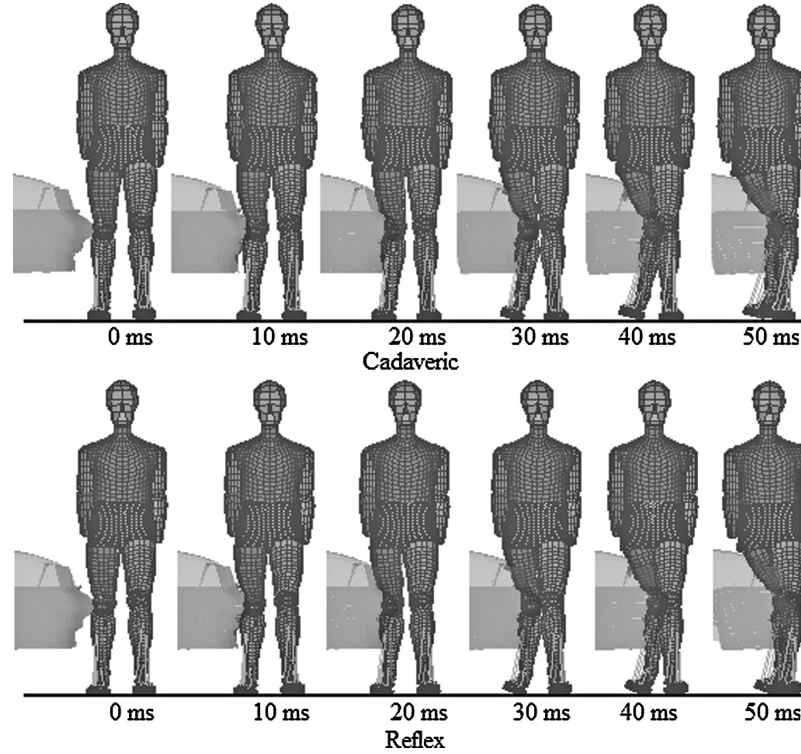


Figure 3. Snapshots during the right leg impact for the cadaveric and reflex conditions.

Right leg impact configuration

185 This section presents the results of the right leg impact configuration. Figure 3 illustrates few snapshots of the simulations for both cadaveric and reflex conditions.

Knee joint kinematics

The linear and angular tibia displacement time histories relative to femur for both cadaveric and reflex conditions

are compared in Figures 4a and 4b, respectively. It is evident 190 that active muscle forces have significantly affected the knee joint kinematics in the reflex condition compared to the cadaveric condition.

In simulations for both cadaveric and reflex conditions, it has been observed that the car front impacts the knee joint 195 of the right leg in the lateral side. As a result, both tibia and femur at the knee level are forced to move in the medial direction. Due to this, even though the overall kinematics

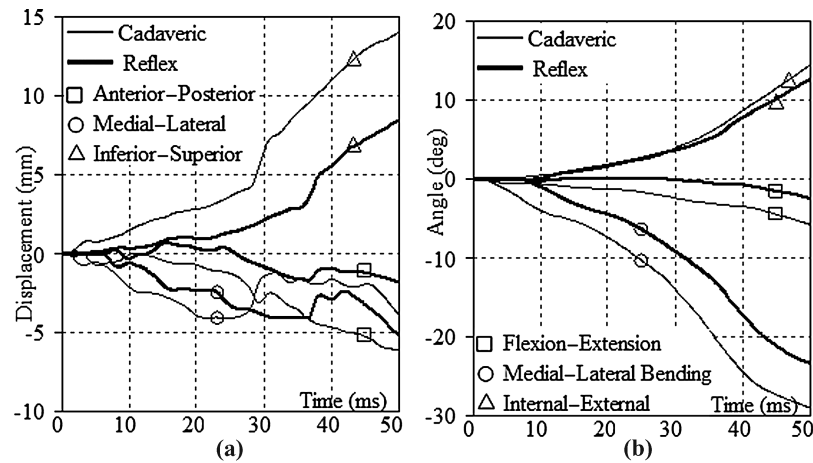


Figure 4. Comparison of relative (a) tibia displacements and (b) knee angles of the right leg.

(Figure 3) appears similar, relative medial tibia displacement (Figure 4a) and medial knee bending angle (Figure 4b) increases in the initial 20–25 ms. At about 28–30 ms, the lower leg moves in the posterior direction and eventually increases the posterior tibia displacement (Figure 4a) and the knee flexion (Figure 4b). High momentum in the car front further pushes the left knee joint medially and hence the medial bending angle increases further. Later, around 32–36 ms, the right foot leaves the ground and frees the lower leg to move away from the femur in the inferior direction which eventually sets the motion in the right leg. Effect of this event is noticeable as further increase in the inferior tibial displacement (Figure 4a) and the medial bending (Figure 4b).

It is apparent from Figure 4 that, in the right leg impact configuration, the inferior tibial displacement and the medial knee bending remain consistently higher than other knee kinematics' parameters in both cadaveric and the reflex conditions. However, in the reflex condition, active muscle forces have pulled tibia close to the femur and therefore significantly reduced the relative linear and angular tibial displacements in all the directions. It is observed that the peak inferior tibial displacements have reduced by a factor of 1.67 and the peak medial bending angle has reduced by a factor of 1.25.

Strain in knee ligaments

Figure 5 illustrates the calculated strain time history in knee ligaments of the right leg of PMALE-WP for both cadaveric and reflex conditions. It is apparent that strains in knee ligaments have reduced significantly in the reflex conditions as compared to the cadaveric condition.

ACL. Figure 5(a) compares the strain time history in ACL for both the conditions. It is observed that till 30 ms, ACL remained nearly unstrained in both the conditions. At about 30 ms, strain in ACL has started and then increased for the remaining portion of the simulations in both the conditions. This sudden rise in the ACL strain in both the conditions can be related with MCL failure (Figure 5c) in both the cadaveric and reflex conditions. It is observed that in the reflex condition, peak strain in ACL has dropped by a factor of 1.47 compared to the cadaveric condition. This can be attributed to the combined effects of the reduced inferior tibial displacement (Figure 4a) and medial knee bending (Figure 4b) in the reflex condition than in the cadaveric condition.

PCL. Strain time history in PCL is compared for both the conditions in Figure 5b. It is apparent that reduction in strain due to muscle action is more prominent in PCL than in ACL. In the reflex condition, the peak strain value in PCL has dropped by a factor of 1.78 compared to the cadaveric condition. Relative displacement of tibia in the inferior direction is the major source of strain in the PCL. Figure 4a illustrates that in both cadaveric and reflex conditions, the tibia is moving away from the femur. However, active muscle forces in the reflex condition have pulled the tibia towards the femur and therefore tibial displacement in the inferior direction (away from the femur) has reduced (Figure 4a). This has slackened the PCL in the reflex condition.

MCL. MCL strain for both the conditions is shown in Figure 5c. It is observed that peak MCL strain has reached the ligament failure limit of 15% in both the conditions. However, in comparison to the cadaveric condition (30 ms), failure is delayed by 5 ms in the reflex condition (35 ms).

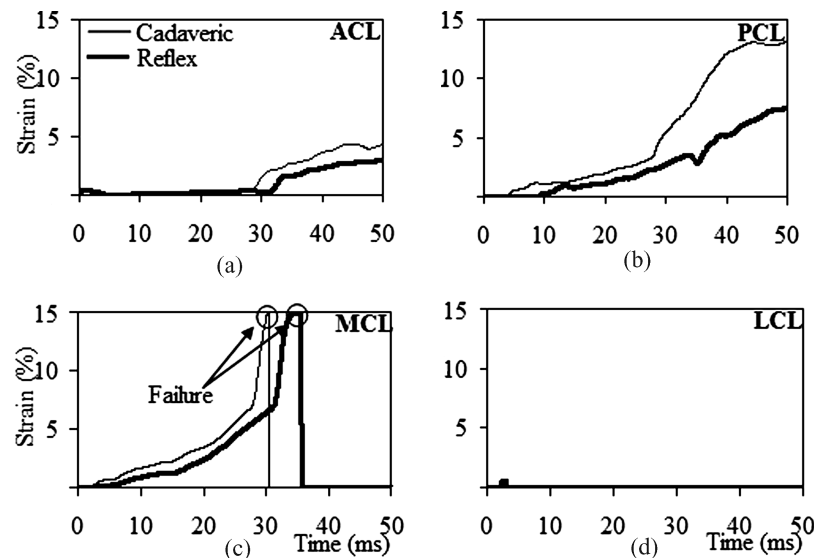


Figure 5. Comparison of strain time history in knee ligaments (a) ACL, (b) PCL, (c) MCL and (d) LCL of the right leg.

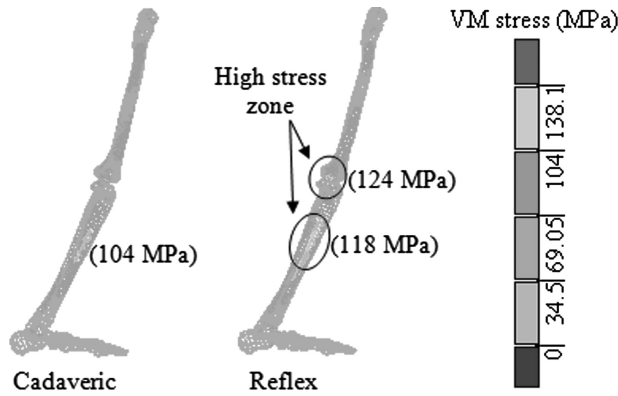


Figure 6. Comparison of the VonMises stress distribution in bones (peak stress values are also given) of the right leg in both the cadaveric and reflex conditions at 34 ms state.

The delay in MCL failure can be attributed to the reduced medial knee bending (Figure 4b) in the reflex condition compared to the cadaveric condition.

265 *LCL*. It is observed that LCL (Figure 5d) has remained unstrained in both the conditions. This can be ascribed to the lateral impact which forces tibia to bend medially and consequently keeps the LCL slackened.

VonMises stresses in bones

Figure 6 compares the VonMises stress distribution on the bones (i.e. femur, tibia and fibula) of the right leg at 34 ms in both the cadaveric and reflex conditions. It is apparent that stresses in bones have increased significantly in the reflex condition as compared to the cadaveric condition.

It is observed that in the reflex condition, stresses in the bones have reached up to 124 MPa at the femoral lateral condylar region and 118 MPa at the medial side of middle one-third of tibia, whereas it has reached only up to 104 MPa in the cadaveric condition. This can be attributed to the higher compressive forces caused by the muscle pull in the reflex condition.

Left leg impact configuration

The results of the left leg impact configuration are presented in this section. Figure 7 illustrates few snapshots of the simulations for both cadaveric and reflex conditions.

Knee joint kinematics

Linear and angular tibial displacement time histories relative to femur for both cadaveric and reflex conditions are compared in Figures 8a and 8b, respectively. It is evident

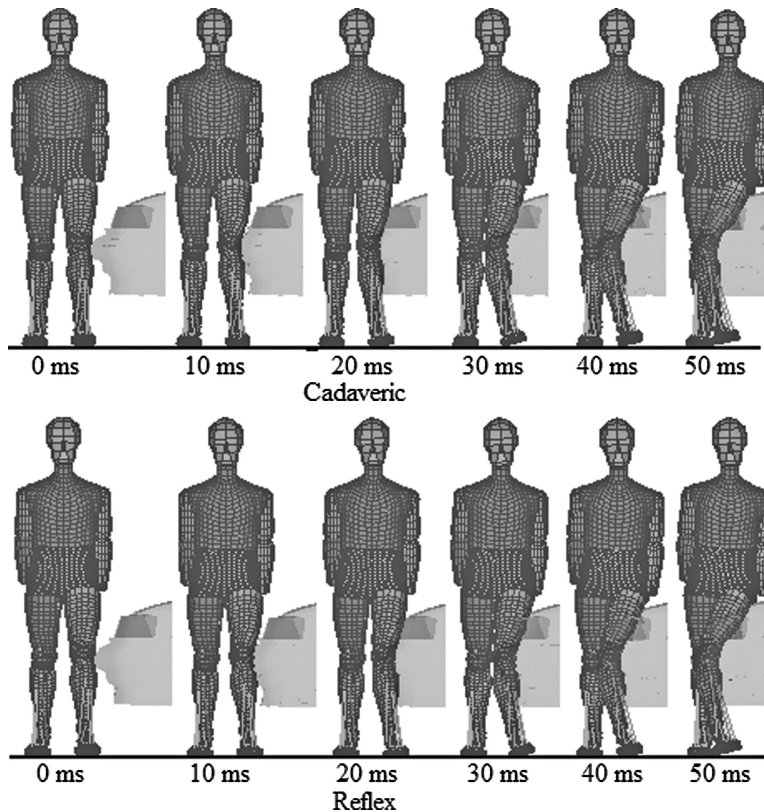


Figure 7. Snapshots during the left leg impact for the cadaveric and reflex conditions.

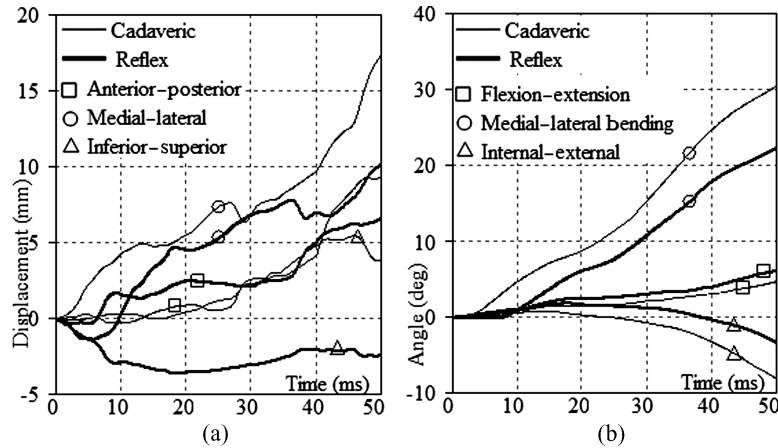


Figure 8. Comparison of relative (a) tibia displacements and (b) knee angles of the left leg.

that active muscle forces have significantly affected the knee joint kinematics in the reflex condition compared to the cadaveric condition.

In simulations for both cadaveric and reflex conditions, it has been observed that the car front impacts the knee joint of the left leg in the lateral side. As a result, both tibia and femur at the knee level are forced to move in the medial direction. Due to this, even though the overall kinematics (Figure 7) remains similar, relative medial tibial displacement (Figure 8a) and medial knee bending angle (Figure 8b) increase consistently during the entire duration of both the simulations. At about 26–28 ms, the left foot leaves the ground and eventually sets the motion in the left leg. Since the left leg is in forward stance, inertia of the moving left leg further increases the anterior tibia displacement (Figure 8a) and the knee extension (Figure 8b).

It is apparent from Figure 8 that in the left leg impact configuration, medial tibial displacement and medial knee bending have remained consistently higher than other knee kinematics' parameters in both the cadaveric and reflex conditions. However, in the reflex condition, active muscle forces have pulled tibia close to the femur and therefore significantly reduced the relative linear and angular tibial displacements in all the directions, except the knee extension. It is observed that the peak medial tibial displacement (Figure 8a) has reduced by a factor of 1.71 and the peak medial bending angle (Figure 8b) has reduced by a factor of 1.36. However, the peak value of knee extension (Figure 8b) has increased by 1.35 times in the reflex condition as compared to the cadaveric condition. This could be attributed to the higher active force in the knee extensor muscles i.e. vastus lateralis and rectus femoris in the reflex condition (peak values of 984 and 305 N in the vastus lateralis and the rectus femoris, respectively) as compared to the cadaveric condition (peak value below 20 N in both the muscles).

It is interesting to note that in the reflex condition tibia has moved in the superior direction (i.e. towards femur)

(Figure 8a) whereas it has moved in the inferior direction (i.e. away from the femur) in the cadaveric condition. This is due to the higher active muscle forces in the reflex condition which have pulled tibia towards the femur. More knee extension in the reflex condition may also have contributed in moving the tibia towards the femur.

Strain in knee ligaments

Figure 9 illustrates the calculated strain time history in knee ligaments of the left leg of PMALE-WP for both cadaveric and reflex conditions. It is evident that strains in knee ligaments have reduced significantly in the reflex condition as compared to the cadaveric condition.

ACL. Figure 9a compares the strain time history in ACL for both the conditions. It is observed that the peak ACL strain has reached the ligament failure limit of 15% in both the conditions. However, active muscle forces in the reflex

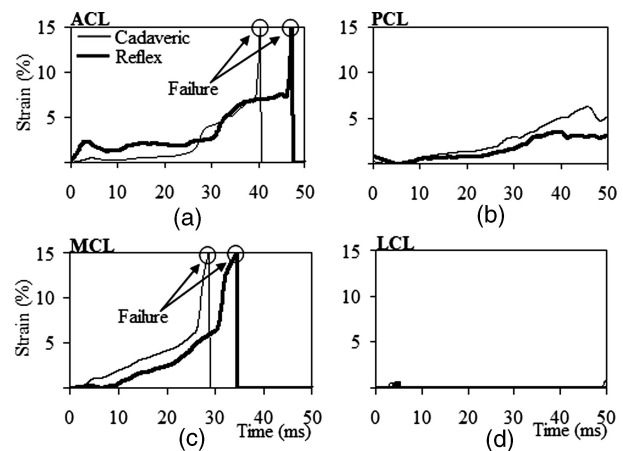


Figure 9. Comparison of strain time history in knee ligaments (a) ACL, (b) PCL, (c) MCL and (d) LCL of the left leg.

condition (47 ms) have delayed the failure by 7 ms as compared to the cadaveric condition (40 ms). The delay in ACL failure can be attributed to the combined effects of reduced anterior tibial displacement (Figure 8a) and medial knee bending (Figure 8b) in the reflex condition as compared to the cadaveric condition.

PCL. Strain time history in PCL is compared for both the conditions in Figure 9b. It is observed that in the reflex condition strain in PCL has remained lower than the cadaveric condition for the entire duration of the simulation. It is found that peak strain in PCL has dropped by a factor of 1.78 in the reflex condition (3.5%) as compared to the cadaveric condition (6.2%). The relative displacement of tibia in the inferior direction is the major source of strain in PCL. Figure 8a illustrates that in the cadaveric condition, tibia is moving away from the femur (i.e. in the inferior direction), whereas in the reflex condition, active muscle forces have pulled the tibia towards the femur (i.e. in the superior direction). This has slackened the PCL in the reflex condition.

MCL. MCL strain for both the conditions is shown in Figure 9c. It is observed that the peak MCL strain has reached the ligament failure limit of 15% in both the conditions. However, in comparison to the cadaveric condition (29 ms), failure is delayed by 6 ms in the reflex condition (35 ms). The delay in MCL failure can be attributed to the reduced medial knee bending (Figure 8b) in the reflex condition as compared to the cadaveric condition.

LCL. It is observed that LCL (Figure 9d) has remained unstrained in both the conditions. This can be ascribed to the lateral impact which forces tibia to bend medially and consequently keeps the LCL slackened.

VonMises stresses in bones

Figure 10 compares the VonMises stress distribution in the bones (i.e. femur, tibia and fibula) of the left leg at 36 ms in both the cadaveric and reflex conditions. It is apparent that

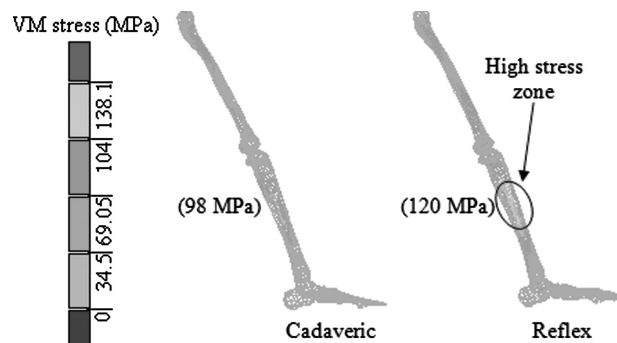


Figure 10. Comparison of the VonMises stress distribution in bones (peak stress values are also given) of the left leg in both the cadaveric and reflex conditions at 36 ms state.

stresses in bones have increased significantly in the reflex condition as compared to the cadaveric condition.

It is observed that in the reflex condition, stress in the bones has reached up to 120 MPa at the medial side of middle one-third of tibia, whereas it has reached only up to 98 MPa in the cadaveric condition. This can be attributed to the higher compressive forces caused by the muscle pull in the reflex condition.

Conclusion

In this study, effects of muscle contraction on the response of lower limb for pedestrian walking posture in low speed lateral impact have been studied. The full body model with active lower extremities i.e. PMALE, which was configured in a standing posture, has been repositioned in the walking posture (PMALE-WP). The real-world car-pedestrian lateral impact has been simulated using the PMALE-WP and front structures of a validated car FE model. Two impact configurations, i.e. impact on the right and left legs have been simulated. For each impact configuration, two sets of simulations, i.e. with deactivated and activated muscles (including reflex action), mimicking an unaware walking pedestrian have been performed. Differences in response between a cadaver and an unaware pedestrian have been studied. To assess the effect of muscle activation, knee kinematics, strains in knee ligaments and VonMises stresses in bones have been compared. Finally, the following conclusions can be drawn from the present study.

1. For both 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 both 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. In all four simulations, MCL has failed, whereas LCL remained nearly unstrained. This implies that in lateral impacts, MCL could be considered as the most vulnerable and LCL as the safest ligament.
4. In the right leg impact configuration, strain in PCL is seen to be significantly higher than strain in ACL. This suggests that in the case of impact on the rear leg of a walking pedestrian, PCL would be at a higher risk than ACL.
5. In the left leg impact configuration, ACL has failed in both the conditions, whereas strain in PCL remained well below the failure limit. This indicates that in the case of impact on the front leg of a walking pedestrian, ACL would be at a higher risk than PCL.

Acknowledgements

The authors would like to acknowledge the support from the Transportation Research and Injury Prevention Program (TRIPP) at Indian Institute of Technology Delhi and the Volvo Research Education Foundation.

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APPENDIX

Values of activation levels used in this study to model active muscles in each leg are listed in Table 1. These values are taken from Winter [21]. Here, right leg muscles are modelled for 60% gait (i.e. terminal stance) and left leg muscles are modelled for 0% gait (i.e. heel strike).