EFFECT OF MUSCLE CONTRACTION IN LOW SPEED CAR-PEDESTRIAN IMPACT – SIMULATIONS FOR WALKING POSTURE

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ABSTRACT

This paper investigates the effect of muscle contraction on lower extremity injuries in low-speed car-pedestrian lateral impacts for a walking pedestrian. The full body model, PMALE, which was configured in symmetric standing posture, has been repositioned in the walking posture. FE simulations have then been performed for its impact with the front structures of a car. Two impact configurations, i.e. impact on the right and on the left leg 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 reflex action was modeled for the unaware pedestrian. It is concluded that (1) with muscle contraction, risk of ligament failure decreases whereas 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 case of impact on rear leg whereas, in case of impact on front leg, ACL would fail.

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 after the head, the lower extremities are the 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 was 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. To devise effective pedestrian protection systems, it is essential to understand the injury mechanism.

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; Soni et al. 2007). However, the major shortcoming in these experimental and computational studies was that they did not account for reflex muscle action. Therefore, effects of pre-crash muscle contraction on the response of lower limbs in car-pedestrian crashes remained unclear.

Of late, Soni et al. (2007) have investigated the probable outcome of muscle contraction using a lower limb (single leg) FE model with active muscles (A-LEMS). More recently, Soni et al. (2008) 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. They concluded 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. However, Pedestrian Crash Data Study (PCDS) (Chidester et al. 2001) reported that prior to the crash, only 4% pedestrians were found standing stationary whereas, a majority, i.e. 55%, was walking.

The present study extends our earlier studies to investigate the effect of muscle contraction on the response of lower limb for the walking pedestrian in low speed car-pedestrian lateral impact using FE simulations. The PMALE, which was configured in standing posture, has been repositioned in walking posture in the current study. The real world car-pedestrian lateral impact has been simulated using the PMALE configured in the walking posture and front structures of a validated car FE model. Two impact configurations, i.e. impact on right and on left leg have been simulated. This is to account for the equal chances of impact on either leg of a walking pedestrian in real world crashes. Two sets of simulations, i.e. with deactivated muscles (cadaveric) and with 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 activation have been compared to assess the effect of muscle contraction.

METHODS

PMALE in Walking Posture

In the present study, PMALE (Soni et al. 2008), which was configured in symmetrically standing posture of a pedestrian with legs in side by side stance, has been adopted as the base model. Body segments of the PMALE configured in the standing posture have then been repositioned in the walking posture in the current study. Relative angles between the body segments required to define the alignment of the walking posture (Table 1) are taken from Mizuno et al. (2003).

Table 1 Definition angles for pedestrian walking posture (Mizuno et al. (2003)

	Definition Angle	
	Left Leg	Right Leg
BA (deg)	+5	
SA (deg)	-15	+15
EA (deg)	0	+27
HA (deg)	+29	-12
KA (deg)	-14	-10
FA (deg)	0	+22

A series of FE simulations have then been performed with the PMALE in standing posture to reposition its body segments in the walking posture. Figure 1 shows the PMALE in walking posture (referred as PMALE-WP) obtained after the repositioning process. In PMALE-WP, right leg (positioned in rear) corresponds to the terminal stance phase of the human gait cycle whereas; left leg (positioned in front) corresponds to heel strike phase. Upper body is leaned forward by 5 degrees with the vertical axis.

Simulation Setup

Figure 2 shows the simulation setup used in the present 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 coefficient of friction between shoe and ground is set to 1.0 as suggested for grooved rubber on road (Li K.W. et al. 2006). 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 with a speed of 25 kmph towards the PMALE-WP. Since in real world car-pedestrian crashes, a car may hit any one of the two legs of a pedestrian; therefore, to account this variability, two impact configurations, i.e. impacting the right leg (Figure 1 (a)) and the left leg (Figure 1 (b)) on the lateral side, 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.

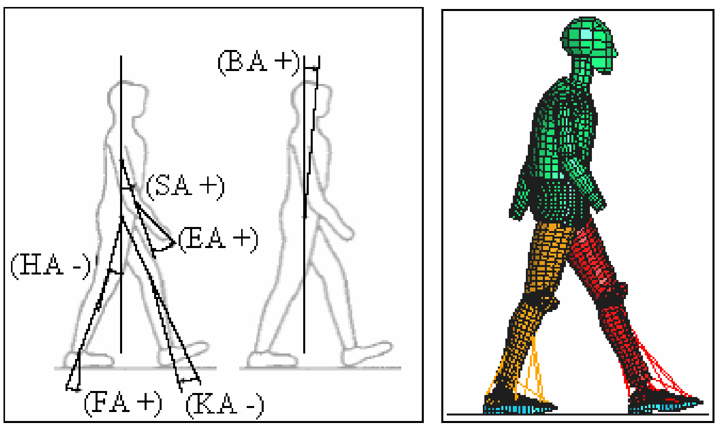

Figure 1 (a) Definition angles with sign conventions for pedestrian walking posture (Mizuno et al. 2003) and (b) PMALE in walking posture (i.e. PMALE-WP)

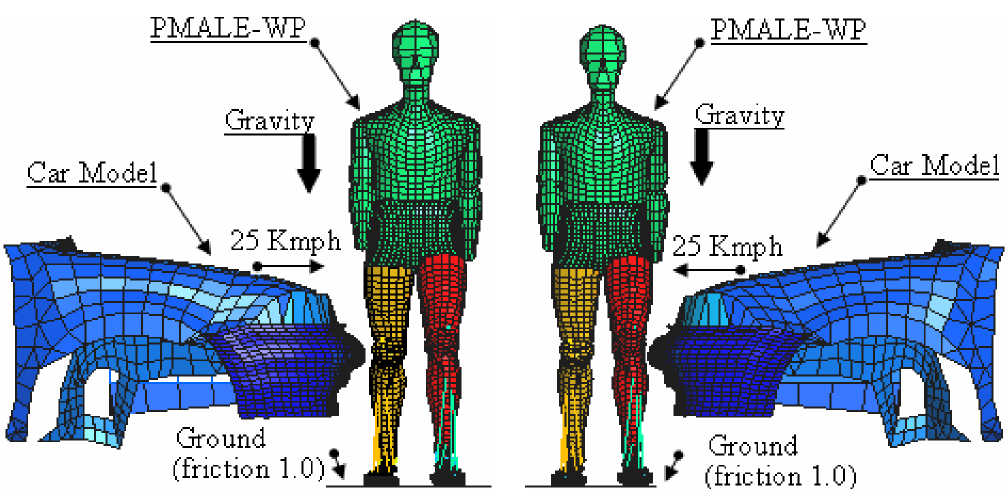

Figure 2 Simulation set up used in the present study for (a) impact on right leg and (b) impact on left leg

Pedestrian Pre-Impact Conditions

Two pre-impact pedestrian conditions, i.e., one with deactivated muscles (cadaveric) and the other with activated muscles (including reflex action) for an unaware pedestrian have been simulated for both the impact configurations 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; thereby increasing the force produced by that muscle.

Cadaveric Condition - In this condition, a cadaver aligned in walking posture has been simulated. To model a cadaver in FE simulation, all the muscles in 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. Here, we have considered that prior to the impact, pedestrian's right leg (in rear) is in terminal stance phase (i.e. right heel is about to leave the ground) of the human gait cycle and left leg (in front) is in heel strike phase (i.e. left heel is just landed on the ground). To model an unaware pedestrian in such walking posture, right leg muscles have been assigned the activation levels corresponding to the terminal stance i.e. 60% gait whereas, muscles in the left leg have been assigned the activation levels corresponding to heel strike i.e. 0% gait. Values of these muscle activation levels (Table A1 in Appendix A) have been taken from the electromyography (EMG) levels recorded in human subjects during the gait cycle by Winter (1987).

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 "stretch reflex" (Vander et al. 1981).

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 of the PMALE-WP have been recorded from the simulations. 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 (stabilization duration), PMALE-WP has been stabilized under gravity load in each simulation. At the end of first 50 ms, car front impacts the right leg or the left leg of the stabilized PMALE-WP. Ligament strains and VonMises stresses in bones have been recorded from the 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.

Impact on Right Leg

In this section we present the results obtained from the simulations of impact on right leg in both cadaveric and reflex conditions.

<u>Strain in Knee Ligaments</u> - Figure 3 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 3 (a) compares the strain time history in ACL for both the conditions. It is observed that upto 30 ms, ACL remained nearly unstrained in both the conditions. At about 30 ms, strain in ACL has kicked-in and then increased for the remaining portion of the simulations in both the conditions. However, active muscle forces in the reflex condition (peak strain 2.96%) have significantly reduced the strain in ACL as compared to the cadaveric condition (peak strain 4.37%).

PCL: Strain time history in PCL is compared for both the conditions in Figure 3 (b). It is observed that upto 28 ms, PCL is strained equally (approx. 3.5%) in both the conditions. However, after 30 ms, strain in PCL has suddenly increased in the cadaveric condition and reached to the peak value of 13.1% at around 45 ms. Whereas, in the reflex condition, active muscle forces have shared the load and hence reduced the strain in PCL (peak strain reached only up to 8% at 50 ms).

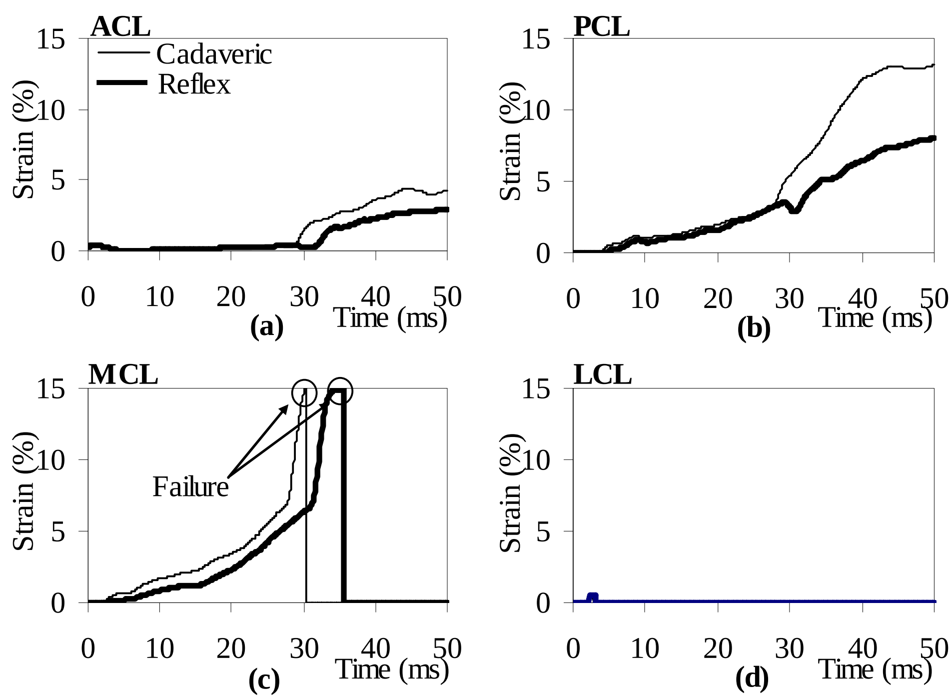

Figure 3 Comparison of strain time history in knee ligaments (a) ACL (b) PCL (c) MCL and (d) LCL of the right leg

MCL: MCL strain for both the conditions is shown in Figure 3 (c). 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). Effect of rupture of MCL is reflected as a sudden increase in strain in ACL (Figure 3(a)) and PCL (Figure 3(b)) around 30 ms in both cadaveric and the reflex conditions.

LCL: It is observed that LCL (Figure 3 (d)) 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 4 compares the VonMises stress distribution in the bones (i.e. femur, tibia and fibula) of the right leg at 34 ms in both cadaveric and reflex condition. 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 lateral femoral condylar region and 118 MPa at the medial side of mid 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.

Impact on Left Leg

Now, we present the results obtained from the simulations of impact on left leg in both cadaveric and reflex conditions.

Strain in Knee Ligaments - Figure 5 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 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 peak ACL strain has reached the ligament failure limit of 15% in both the conditions. However, active muscle forces in the reflex condition (47 ms) have delayed the failure by 7 ms as compared to the cadaveric condition (40 ms).

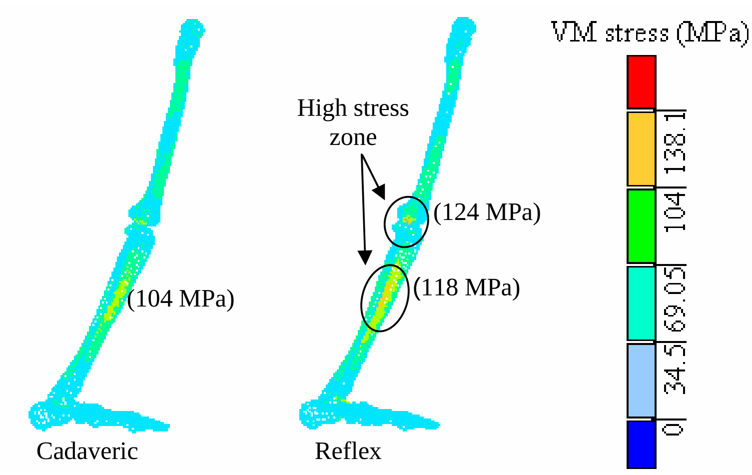

Figure 4 Comparison of VonMises stress distribution in bones (peak stress values are also given) of the right leg in both cadaveric and reflex conditions at 34 ms state

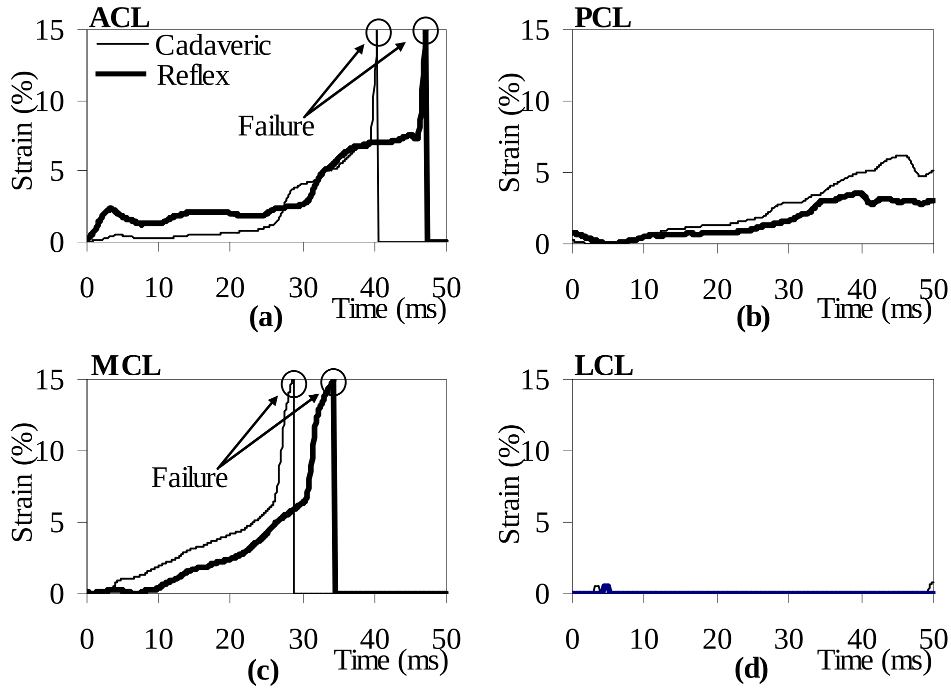

Figure 5 Comparison of strain time history in knee ligaments (a) ACL (b) PCL (c) MCL and (d) LCL of the left leg

PCL: Strain time history in PCL is compared for both the conditions in Figure 5 (b). 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%).

MCL: MCL strain for both the conditions is shown in Figure 5 (c). 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 (29 ms), failure is delayed by 6 ms in the reflex condition (35 ms). Effect of rupture of MCL in both the conditions is reflected as a sudden increase in strain in ACL (Figure 5 (a)) between 29-32 ms in both the conditions.

LCL: It is observed that LCL (Figure 5 (d)) 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.

<u>VonMises Stresses in Bones</u> - Figure 6 compares the VonMises stress distribution on the bones (i.e. femur, tibia and fibula) of the left leg at 36 ms in both cadaveric and reflex condition.

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 120 MPa at medial side of mid 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.

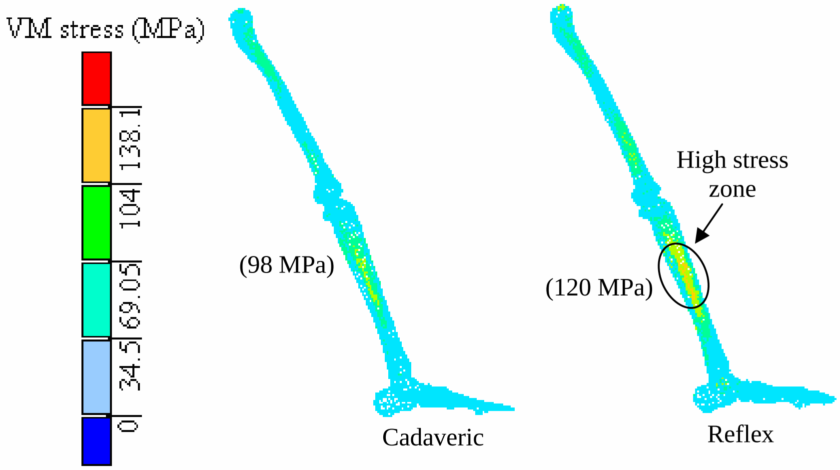

Figure 6 Comparison of VonMises stress distribution in bones (peak stress values are also given) of the left leg in both cadaveric and reflex conditions at 36 ms state

CONCLUSIONS

In the present study, effect of muscle contraction on the response of lower limb in low speed lateral impact has been studied for the pedestrian walking posture. The full body model with active lower extremities i.e. PMALE, which was configured in standing posture, has been repositioned in the walking posture. 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 leg and on the left leg have been simulated. For each impact configuration, two sets of simulations, i.e. one with deactivated muscles (cadaveric condition) and the other with activated muscles (including reflex action) mimicking an unaware walking pedestrian have been performed. Differences in responses of a cadaver and an unaware pedestrian have been then studied. To assess the effect of muscle activation, strains in knee ligaments and VonMises stresses in bones have been compared. It has been concluded that:

1. For both impact configurations, peak strains in knee ligaments were lower in the reflex condition (with active muscles) as compared to the cadaveric condition. This supports 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 the 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 the 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 found to be significantly higher than that in ACL. This suggests that in case of impact on 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. This indicates that in case of impact on front leg of a walking pedestrian, ACL would be at a higher risk.

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APPENDIX-A

Values of activation levels used in the present study to model the 42 active muscles in each leg are listed in the Table A.1. These values are taken from Winter (1987). Here, right leg muscles are modeled for 60 % gait (i.e. terminal stance) and left leg muscles are modeled for 0 % gait (i.e. heel strike).

Table A-1 Activation levels in muscles of left and right leg (Note: Activation levels labeled with (*) are taken from Winter (1987))

Lower extremity muscles	Activation levels	
	Left	Right
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 Digitorium Longus	0.005	0.005
Tibialis Posterior	0.005	0.005
Tibialis Anterior	0.4*	0.1*
Extensor Digitorium	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