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Influence of Pre-Impact Pedestrian Posture on Lower Extremity Kinematics in Vehicle Collisions

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Abstract: lower extremities are the most frequently injured body regions in vehicle-to-pedestrian collisions and such injuries usually lead to long-term loss of health or permanent disability. However, influence of pre-impact posture on the resultant impact response has not been understood well. This study aims to investigate the effects of pre-impact pedestrian posture on the loading and the kinematics of the lower extremity when struck laterally by vehicle. THUMS pedestrian model was modified to consider both standing and mid-stance walking postures. Pedestrian in walking posture exhibited larger knee bending angle (40% for ipsilateral knee joint) and pelvic rotation angle (27.5% for z-direction pelvis rotation angle). The walking posture increased the injury risk of soft connection tissue about 20-30% and reduced the internal force in bone structure about 25% regardless of impact severity. Two-leg interaction, inertial effect, anatomical features of the knee and pelvis exhibited a coupled influence on lower extremity kinematics. Further research efforts are necessary to include more loading scenarios and to quantify the lower extremity injury risk in detail.

Keywords: Pedestrian injury, Pedestrian posture, Finite element human body model, Lower extremity kinematics

1 Introduction

Lower extremity is one of the most frequently injured body regions in vehicle-to-pedestrian collisions and lower extremity injuries comprise almost 32.6% of AIS2+ injuries[1]. Lower extremity injuries usually take long time to recover [2]. Pedestrian Crash Data Study (PCDS) had documented 521 pedestrian accidents at six sites in United States during the period from July 1994 to December 1998. Data in PCDS pointed out that 68% of the involved vehicles were passenger cars. At the instant prior to crash, 55% of pedestrians were in a walking state and 24% were stationary or not in an obvious motion. 63% of the pedestrians had two legs apart from each other in a lateral sight [1].

To investigate injury mechanisms of pedestrian lower extremities, many cadaveric tests were conducted, using full cadavers or cadaveric parts [3][4][5][6][7]. Full-scale pedestrian cadaveric tests with vehicle front-end revealed that bone fracture and ligament rupture were main injury patterns in lower extremity injuries [3][10]. In full cadaveric tests, pedestrians were always positioned in mid-stance walking posture with almost no overlap between the two legs. Thus the experiments were not capable to represent those nearly 40% crashes in which the two legs interacted. Pedestrian gait had been taken into consideration in accident reconstruction, while has not received enough attention in their influence on kinematics and injury [8][9]. Previous studies of reconstruction focused on relation between head impact location and pedestrian initial postures. Whether pedestrian posture could affect load path and distribution of local deformation in the lower extremities, causing differences in injury patterns and locations remains unclear. Recently the Crash Injury Research and Engineering Network (CIREN) recorded detailed whole body injury data in 67 pedestrian crashes. In-depth statistical analysis demonstrated different frequency of main injury patterns comparing to what cadaveric tests presented [10]. In real-world accidents the pedestrian would not always in walking posture and purely lateral impact can only be taken as a simplified setup for laboratory-based use. Whether it is the causation of the variation in results needs to be further investigated.

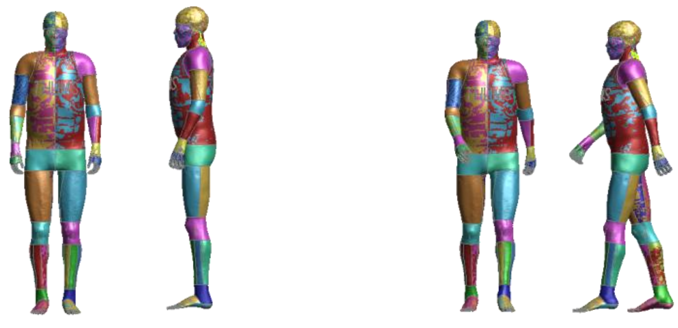
Human body model proved to be an efficient tool in identifying the contribution of pre-impact posture in kinematics and injury risk [11]. The computational approach was taken in this study for its capability of eliminating individual diversity of the samples in cadaveric tests. There have been many studies related to pedestrian safety protection with THUMS (Total Human Body Model for safety) [14][20][21]. In the present study, THUMS pedestrian model was modified to simulate full-scale vehicle-to-pedestrian impact with two pre-impact postures (standing and walking) under 3 velocities, including 25, 33 and 40 kph, to investigate the influence of velocity on pedestrian kinematics and the coupling interaction of velocity and pre-impact postures. Different velocities of vehicle were considered to investigate the potential influence in a wide range of the initial energy of the whole dynamic event. Since vehicle front-end structure is deformable and so its resistance and local deformation to the leg contact change with different velocities. Higher contact velocity results in flatter contour of the deformed front-end structure, and consequently the different contact surface for lower extremity affects the knee joint bending angle and other kinematical characteristic. Lower extremity kinematics of pedestrian was investigated, including knee joint spatial bending motion and pelvic spatial translation and rotation. The goal of this study is to analyze influence of pre-impact pedestrian posture on lower extremity kinematics and mechanisms, and possible effects to injury risks of lower extremity.

2 Methodology

2.1 Model setup

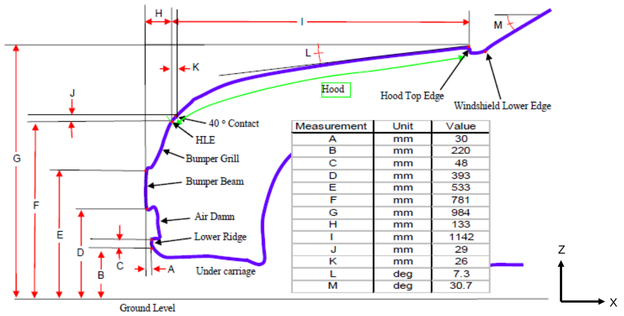
Two gait states were adopted in the current study, the original THUMS pedestrian model (Version 4.0.1) in a standing posture (Figure 1a) and the mid-stance walking posture modified from the standing model [9]. The posture adjustment was made by applying quasi-static loading at the extremities with the upper body fixed as rigid. All the long bones of the upper and lower extremities were rigidified, with the knee, elbow and other essential connecting joints left deformable. In the pre-simulations for generating different gait postures, the torso was also converted to rigid body to improve the computational efficiency. 1-D beam element was attached to the distal end of the target extremity. Free end of the beam element was given a prescribed slow motion to pull the extremity to the predetermined position. Unlike a prescribed motion is assigned to bone structure directly, such an indirect loading applying method can avoid interfere in joint motion. Final postures of the model are illustrated in Figure 1.

A detailed front-end model of a mid-size sedan, used in the experiments by Subit et al. (2008), was used in this study [12]. The front-end structures included engine hood, upper/lower grill, bumper assembly, lower stiffener and other structures relevant to pedestrian lower extremity impact. The A-pillar was rigidly fixed. The geometry of the sedan model and the global coordinate system definition were illustrated in Figure 2. The adopted model had already been validated by Watanabe et al. against the cadaveric test results including data of accelerometers and markers in their study [11].



(a) Standing posture (original posture) (b) Mid-stance walking posture (adjusted posture)

Figure 1. Schematic of pedestrian pre-impact postures

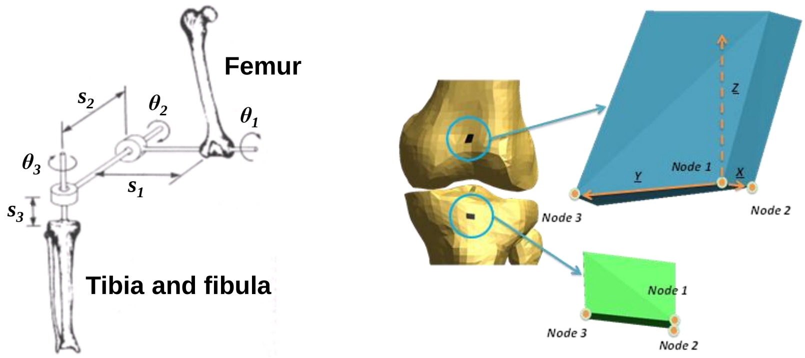


Measurement	Unit	Value
A	mm	30
B	mm	220
C	mm	48
D	mm	393
E	mm	533
F	mm	781
G	mm	984
H	mm	133
I	mm	1142
J	mm	29
K	mm	26
L	deg	7.3
M	deg	30.7

Figure 2. Geometry of the sedan model and global coordinate definition [12]

2.2 Characterization of pedestrian kinematics

In the post process, both ipsilateral and contralateral knee joint bending angles in the three directions were calculated by the relative rotation of the local coordinate systems defined on femur and tibia using the Grood-Suntay method [13]. The two local coordinate systems were defined based on nodal information in femoral condyle and tibial plateau areas, respectively, to represent the structure motion(Figure 3). This decision was made considering the much higher stiffness and less deformation of the femoral condyle and tibial plateau areas comparing to adjacent tissues. Relative bending angle between the two subjects in the three directions, flexion, abduction/adduction, tibial external/internal rotation, can be obtained by inverse-consine calculation of projection of femoral axis on the corresponding tibial axis, and the formula is referred to [13]. Similarly, pelvic spatial motion can be calculated with respect to the global coordinate system. Ligament stretching ratio was used as an indicator to monitor ligament injury risk. For each of the concerned ligament, the stretching ratio was calculated by summing up of piecewise length between adjacent nodes in linear consequence along its length.



(a) Coupling of knee joint motion [13] (b)Local coordinate definition in knee joint area

Figure 3. Schematic of knee joint motion coupling and local coordinate definition in femoral condyle and tibial plateau to calculate knee joint motion

3 Results

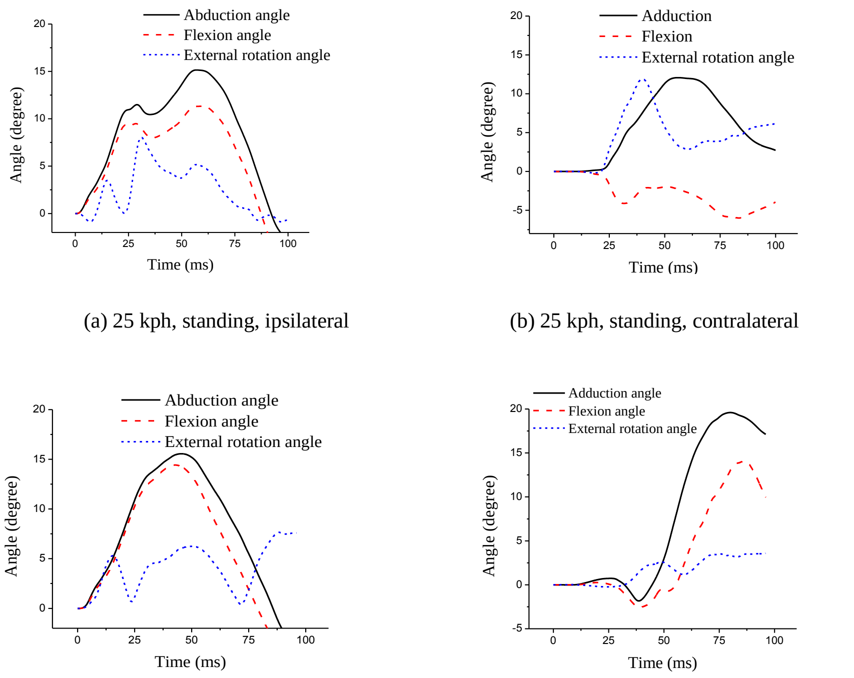
3.1 Knee joint kinematics

The knee joint motion for standing and walking postures under vehicle velocity of 25 kph is illustrated in Figure 4 as a representative of the three impact severities. At 25 kph vehicle impact speed the curves exhibit more details as the lower speed corresponding with long impact process. On the contrary larger velocity case corresponds with shorter time in impact process, periods induced by different mechanisms cannot be obviously distinguished in the curves.

3.2 Knee joint presented large variation for different pedestrian pre-impact postures

For the ipsilateral knee joint, peak value of abduction from the two pre-impact postures were similar, i.e., 15.2 degree for standing (Figure 4a) and 15.6 degree for the walking posture (Figure 4b).

Load transfer pattern variation was revealed by difference in curve shapes for the two postures. The ipsilateral side in the standing posture exhibited a dual peak phenomenon in the abduction and flexion curve (Figure 4a). Mechanism for this is the contact of the two legs in knee joint area. Descent of the ipsilateral abduction curve was at the same instant when contralateral knee joint initiated to move around 25 ms. Flexion angle varied along with abduction in the curve shape regardless of pre-impact posture. This can be explained as flexion was a natural motion of knee joint and was easy to trigger during a complicated spatial motion. Therefore, it was initiated by the lower extremity to adjust itself into a more flexible and relaxed configuration. Excessive abduction was an abnormal motion from anatomical view due to external loading. When knee joint deviated from the initially perfect lateral loading condition to some kind of oblique bending condition, flexion would emerge to mitigate internal tensity in knee joint caused by lateral bending. Tibial external rotation exhibited a significant oscillation (Figure 4) and small peak value compared to flexion and abduction/adduction curves. Overall, the pre-impact posture did not present a significant influence on the ipsilateral knee joint motion. In the initial period the motion curves for the ipsilateral legs were similar before two legs started interaction. Difference in the external rotation curve was induced by the swing of ipsilateral leg in walking posture.



(a) 25 kph, standing, ipsilateral

(b) 25 kph, standing, contralateral

(c) 25 kph, walking, ipsilateral

(d) 25 kph, walking, contralateral

Figure 4. Time histories of knee joint bending angle in three directions

Two-leg interaction induced larger difference in kinematics of contralateral knee joint. For the ipsilateral knee joint, peak value of abduction from the two pre-impact postures were similar, i.e., 12.5 degree for standing (Figure 4a) and 20.1 degree for the walking posture (Figure 4b). There was a 40% difference. For pedestrian in standing posture, contact at the lateral side of the femoral condyle resulted in femoral internal and in-coronal-plane rotation. This led to an extension motion and the contralateral knee joint was loaded in a "locked" state (Figure 5). The extension was the consequence of local geometry around the contact location and such motion brought more constrained adduction motion of the contralateral knee joint, that is, the knee joint was loaded towards a more deflected direction from the normal state. For walking posture the contralateral leg impacts directly with the front-end structures without obstruction from the other leg. Prior to the contact, the femur head had already been loaded by acetabulum due to pelvic motion. Thus there was an original descent in the flexion and adduction curve. Peak value was also higher comparing to the ipsilateral side. As the thigh did not cling to vehicle surface, the far-side leg possessed larger space to deform. Thus the lateral bending angle in knee joint was larger for the walking case (Figure 6).

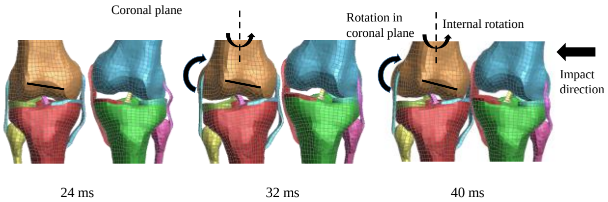


Figure 5. Two-leg interaction in the standing case. Femoral condyles impact lead the contralateral knee joint to a "locked" state to some degree as an extension emerged

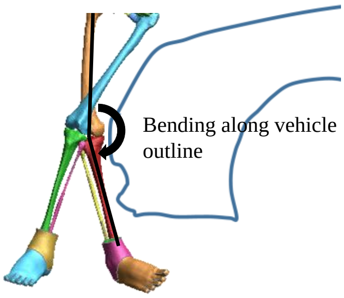


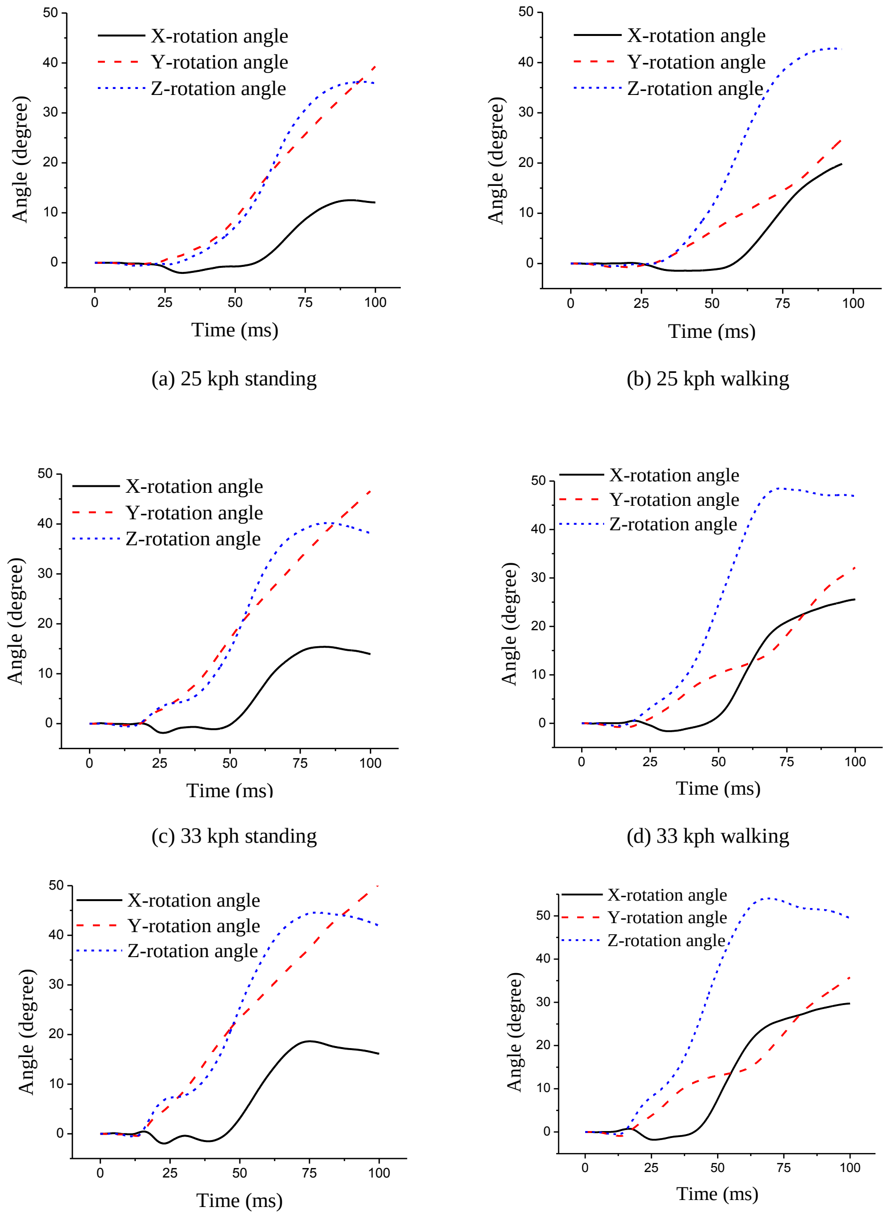
Figure 6. Schematic of lower extremity motion with walking posture (76 ms, 25 kph).

3.3 Pelvic motion and relation to pre-impact postures

Pelvis played a predominant role in load transfer from lower extremities to upper torso [6]. In the present study, pre-impact posture of pedestrian induced differences in pelvic motion. As mentioned above, contralateral knee joint in standing case was "locked". Lower extremity motion was more constrained in the coronal plane. Walking would bring complicated nonplanar configuration to initial condition which can be viewed as a "disturbance" to the equilibrium of the lower extremity. Specifically, although the initial states of the pelvis were practically the same for the two postures, it exhibited significantly higher rotation along the body z-axis during the impact in walking posture. (Figure 7).

For pedestrian in standing posture, pelvis rotation about Y-direction and Z-direction were significant and the magnitudes were close in a large time range (Figure 7a, c, e). Walking posture presented a significant higher rotation about the Z-direction rather than X- direction and Y-direction. The reverse Z-direction rotation of pelvis would turn the human body to a position with larger contact area with the vehicle front-end. In 40 kph cases, the largest Z-direction rotation was 21.6 degree for standing and 29.8 degree for walking. There was a 27.5% difference. Hip joint got closer to vehicle. Pelvis was consequently more backward in the impact direction, indicating that it acquired less momentum from the vehicle (Figure 8). Less amount of momentum transfer corresponded with smaller contact force which was beneficial to pedestrian safety. The pedestrian pelvis displacement in the global X-direction is shown in Figure 9. The marked point was at the conjunction point between sacrum and L5 vertebra. Thus the X displacement of the specific point partially reflected the inertial effect of the upper torso, that is, how much the upper torso got involved in the X-direction motion. Take 40 kph cases for example, the X-direction displacement was -368 mm for standing case and

-334 mm for walking case. There is a 10.2% difference of X-direction displacement, which means that locally the lumbar components acquired more momentum in the standing case. The results were consistent with what was revealed by the rotation curves. The relatively larger Z direction rotation and smaller X displacement in the walking posture implies less constraint on the pelvis motion and a potentially lower injury risk.



(a) 25 kph standing

(b) 25 kph walking

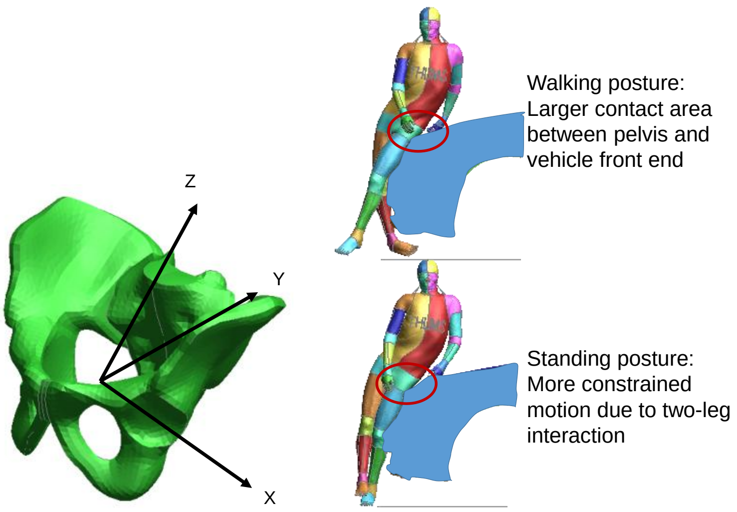
(c) 33 kph standing

(d) 33 kph walking

(e) 40 kph standing

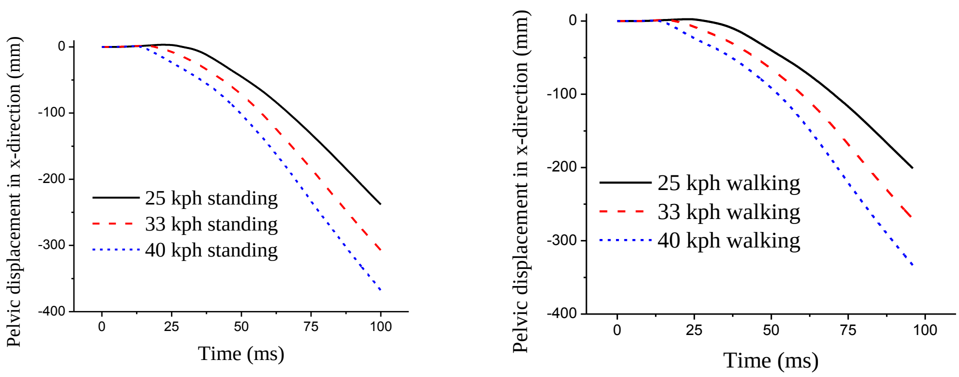
(f) 40 kph walking

Figure 7. Pelvis kinematics in global coordinate system



(a) Pelvis rotation (b) Pedestrian kinematics

Figure 8. Schematic of pelvis rotation and pedestrian kinematics



(a) Standing (b) Walking

Figure 9. Pelvis displacement history in X-direction

3.4 Injury of lower extremities

Abduction/adduction of knee joint (conventionally referred to as the lateral bending angle) was chosen to model the relative movement between femur and tibia besides the knee joint kinematics itself in the present study. They were also considered as a main injury predictor, especially for the collateral ligaments [15][16][17]. This study was intended to focus on the lower extremity kinematics, therefore specific knee joint kinematical and injury parameters such as lateral shearing displacement, tibial anterior/posterior displacement, ligament maximum principal strain and local stress of the other connection soft tissues were not included. Excessive abduction/adduction tended to cause large collateral ligament stretching in the tensile side or and may result in ligament rupture. Table 1 listed the maximum ligament stretching ratio and occurrence time for ipsilateral MCL and contralateral LCL. Generally, walking posture presented higher stretching ratio corresponding to higher risk of rupture reflected by maximum stretching ratio of ligaments in Table 1. For example, in 40 kph cases, ipsilateral MCL maximum stretching ratio was 18.8% for walking case and 16.6% for standing case. Overall collateral ligament stretching in the walking cases was 20-30% larger than that in the standing cases. The stretching ratio was approximately linearly proportional to the lateral bending angle due to the anatomical features of knee structure. Shearing displacement and axial tension of knee joint also contributed to ligament stretching. As the ligaments served as main connection tissue in knee joint structure, their mechanical behavior was dependent on knee joint local kinematics, which was highly relevant to pre-impact postures. Especially for the contralateral LCL the stretching ratio exceeded 30% under 40 kph velocity. Generally regardless of pre-impact posture, LCL stretching ratio increases with velocity.

The phenomenon that contralateral LCL presented larger stretching ratio and corresponding higher injury risk relative to ipsilateral MCL was already addressed by Kerrigan in his cadaveric tests [10]. LCL had a smaller cross section area and was weaker in structure. Thus the stretching ratio was larger under similar axial force along the fiber direction. And the contralateral leg undertook the axial tension induced by pelvis rotation. The pulling motion in vertical direction added on the axial force in knee joint ligaments. Together with lateral bending motion, the axial tension force in contralateral LCL was magnified. Such phenomena were not observed in previous cadaveric tests and field data, probably because the boundary condition was not perfect lateral impact for the realistic accident and there might be rotation during the impact process. The mechanism behind the difference of response in ipsilateral and contralateral legs is still under investigation.

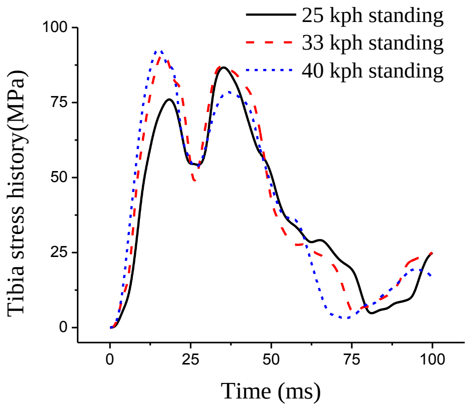
Table 1. Maximum stretching ratio of collateral ligaments for all cases

	Ipsilateral MCL		Contralateral LCL	
Velocity (kph) and posture	Max stretching ratio (%)	Occurrence time (ms)	Max stretching ratio	Occurrence time (ms)
25 standing	10.9	56	14.9	76
25 walking	13.3	42	22.3	78
33 standing	13.4	46	19.8	56
33 walking	16.6	38	26.3	68
40 standing	16.6	42	24.7	50
40 walking	18.8	34	31.2	60

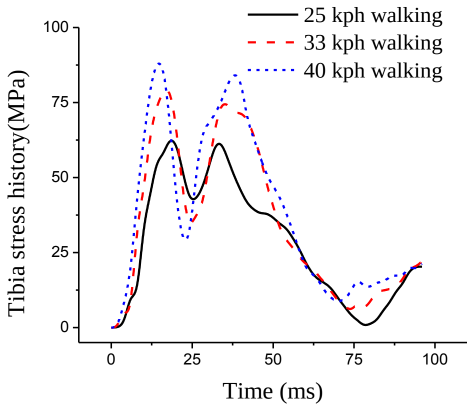
Component test in current regulations towards pedestrian lower extremity protection was established to represent the impact of ipsilateral leg. For the contralateral leg impact, axial tension and interaction of the two legs should be taken into consideration in future.

Another injury pattern always emerged in accidents was tibia fracture, which is commonly caused by instantaneous impact load from bumper and the subsequent bending load during continuous vehicle intrusion. Injury risk of tibia could be assessed by local stress in the shaft area. The Von-Mises stress history of the element with the global maximum stress value in the ipsilateral tibia was shown in Figure 10.

On the contrary, standing posture always presented larger tibial stress, especially in low velocity case. In 25 kph cases, the peak stress in standing posture (75 MPa) was 25% larger than that in walking case (60 MPa). The phenomenon could be attributed to the two-leg interaction. The curves showed a dual-peak shape in which the first one was due to bumper impact and the second one was induced by tibial lateral bending. With overlap of two legs in standing posture, the contralateral leg inertia prevented the ipsilateral tibia from freely doing the X direction motion and thus more mass was involved in the system, resulting in higher internal force in lower extremities The larger deformation brought by more unconstrained motion in walking posture generally led to higher injury risk in knee joint ligaments but it can decrease the stress in hard tissues, like bones. For pedestrian safety protection design, the conflict of different injury risk distribution induced by pre-impact postures should receive more attention.



(a) Standing



(b) Walking

Figure 10. Stress history of ipsilateral tibia

4 Conclusions

This study investigated the influence of pre-impact postures on the kinematics of pedestrian lower extremities in vehicle-to-pedestrian impact. A 50th adult pedestrian model in both standing and walking postures were taken into account under different impact velocities. Global kinematics of pedestrian was studied. Rotation of the knee joint around three axis was calculated and pelvic translational and rotational motion were analyzed. For the walking case, the contralateral knee joint lateral bending angle was 40% larger than the standing case. Stretching ratio for all ligaments was 20-30% larger, and pelvic Z-direction rotation was 27.5% larger in the walking cases. On the contrary, pelvic X-direction translation was 10.2% smaller and tibial stress was 25% smaller in the walking case. Pedestrian in walking posture exhibited larger knee bending angle and pelvic rotation due to isolated single-leg contact with vehicle and nonplanar characteristic from the leg swing. The walking posture increased the injury risk of soft connection tissue about 20-30% and reduced the internal force in bone structure about 25% regardless of impact severity. The trade-off of injury risks induced by kinematics with different pre-impact postures was a challenge for vehicle front-end structure design. Further research efforts are necessary to include more loading scenarios and find a balance between protection of long bones and soft tissues.

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