

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 preimpact 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. Impact simulations were conducted under three severities, including 25, 33 and 40 kph impact for both postures. Global kinematics of pedestrian was studied. Rotation of the knee joint about the three axes was calculated and pelvic translational and rotational motions were analyzed. 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) with less constraint due to isolated single-leg interaction 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. Two-leg interaction, inertial effect, anatomical features of the knee and pelvis exhibited a coupled influence on lower extremity kinematics. Injury predictors such as tibia stress and collateral ligament stretching ratio was found to be associated with kinematics. The trade-off of injury risks induced by kinematics with different pre-impact postures is a challenge for vehicle front-end structure design. Further research efforts are necessary to include more loading scenarios and to quantify the lower extremity injury risk in detail.

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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.

METHODOLOGY

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].

In the present study, simulations were conducted using explicit LSDYNA software (LSTC, Livermore, CA, Version 971 R6.1.2). The whole simulation time for a single case was 100 ms and each took approximately 13 hours in physical time in a cluster of Intel Xeon E5-2650 processor with 64 cores.



(a). Standing posture (original posture)



(b). Mid-stance walking posture (adjusted posture)

Figure 1. Schematic of pedestrian pre-impact postures

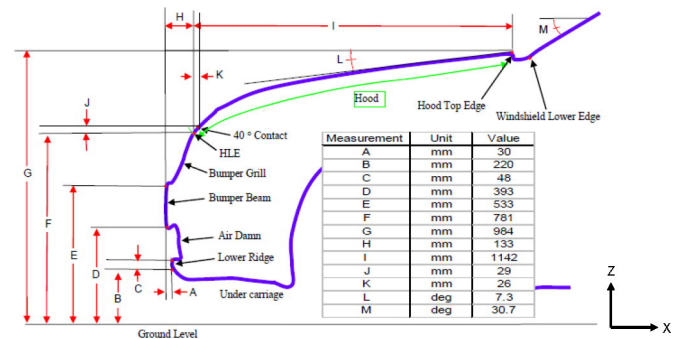


Figure 2. Geometry of the sedan model and global coordinate definition [12]

The pedestrian was positioned 5 mm in front of the bumper central outline to mimic the instant prior to a typical vehicle-to-pedestrian collision. The pedestrian was in standing still posture or mid-stance gait with vehicle striking the left side. The ipsilateral (struck side) leg was moving forward in walking case. The ground was modeled by rigid wall and the friction coefficient was set to 0.2 between the pedestrian and the ground. Global gravity field was assigned to simulate the gravity effect. Braking effect and subsequent pitching motion of vehicle front-end were not taken into account.

Characterization of Pedestrian Kinematics

Displacements of several body regions were measured relative to the vehicle coordinate system to characterize the global kinematics, including the head center of gravity, the acromia, the spine (First thoracic vertebra T1 and eighth thoracic vertebra T8), top of the sacrum, great trochanter, femoral and tibial shaft, and tibial plateau (both the medial and lateral sides) (Figure 3).

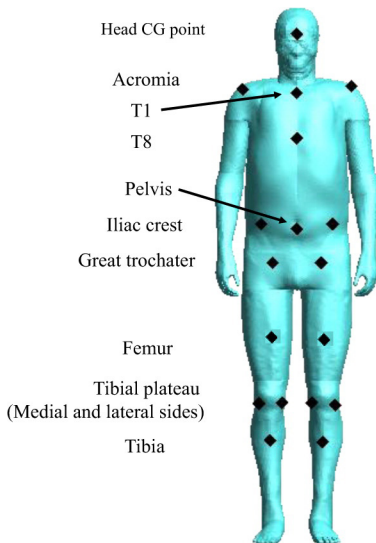


Figure 3. Marked points for recording pedestrian kinematics

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 4). 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-cosine 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.

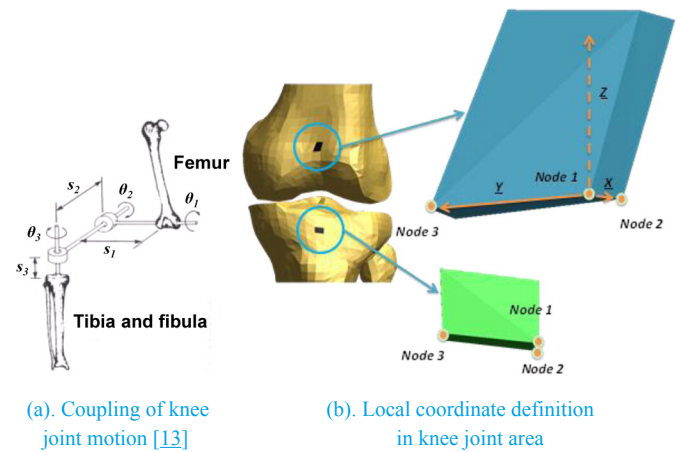


Figure 4. Schematic of knee joint motion coupling and local coordinate definition in femoral condyle and tibial plateau to calculate knee joint motion

Definition of Injury Predictors

In the present study, long bone fracture and ligament rupture were taken into consideration, since those were main injury patterns for lower extremities.

For knee ligament injury, ligament stretching ratio was selected as the injury predictor. Comparing to other stress/strain-based predictors, stretching ratio was of higher robustness, and the response would not be highly dependent on mesh quality, especially for those cases with large local deformation in knee ligaments. The ligament length was calculated by sum of piecewise element length in ligament fiber direction (Figure 5). Stretching ratio was the value of the elongation part divided by the original length.

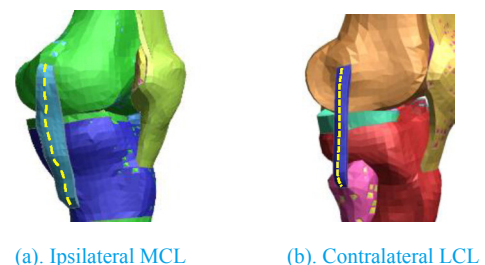


Figure 5. Calculation method of ligament length

For tibial fracture, plastic strain exceeding 1.5% may be regarded as the injury predictor and corresponding threshold in THUMS [19]. Von-Mises stress was not regarded as an injury predictor. However, it could reflect the local force load to the bone structure. Thus it was chosen to evaluate local contact force in tibia.

Element elimination of knee joint ligaments and bone structure was deactivated in the simulations.

RESULTS

Global Kinematics

The traces of the markers on the pedestrian body are shown in [Appendix A](#). [Table 1](#) lists the peak excursion values referring to vehicle coordinates in the X-Z plane of different body parts.

Table 1. Peak excursions of different body parts

Body parts	Peak excursion (mm)					
	25 kph		33 kph		40 kph	
	Standing	Walking	Standing	Walking	Standing	Walking
Head X	570	572	719	729	850	845
Head Z	-92.1	-65.1	-127	-120	-169	-151
T8 X	527	540	661	646	736	741
T8 Z	-84.3	-60.1	-118	-104	-143	-140
Pelvis X	330	371	407	421	470	473
Pelvis Z	-23.1	-22.5	-21.5	-23.5	17.6	19.2
Knee R X	82.5	210	109	187	131	233
Knee R Z	151	77.5	220	146	289	201
Heel R X	-15.4	265	-43.5	204	-94.7	86.3
Heel R Z	152	96.5	241	140	323	201

In high velocity case, generally the peak excursion was larger. The current cases lasted 100 ms, which did not cover the rebounding phase of the whole impact process for pedestrian. Thus, the peak value occurred at 100 ms.

Considering the influence of pre-impact postures on excursion, walking cases presented drastically larger excursion in right knee joint and heel in X direct. That was because there was no two leg interaction and the right leg could move towards vehicle. For pelvis and thorax, excursion in walking cases was slightly larger, about 5-10%.

Knee Joint Kinematics

The knee joint motion for standing and walking postures under vehicle velocity of 25 kph is illustrated in [Figure 6](#) 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. All schematics of the knee joint kinematics curves are shown in [Appendix B](#).

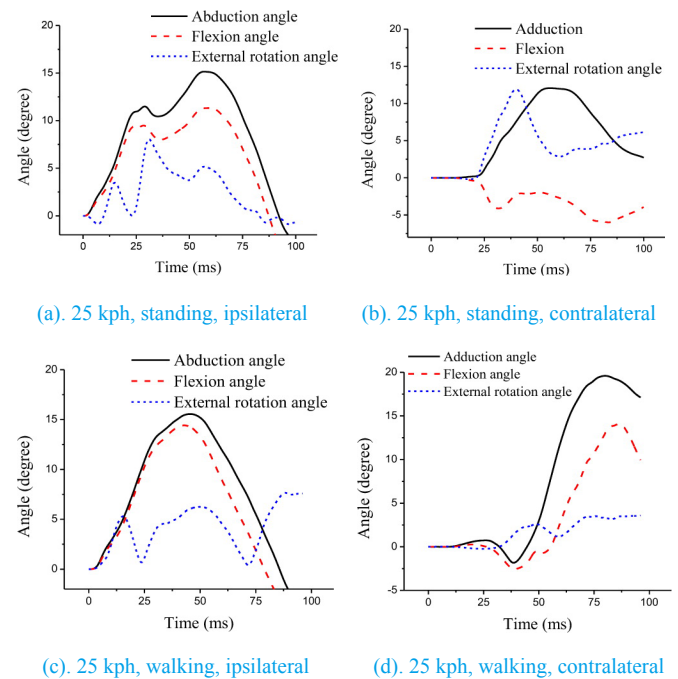


Figure 6. Time histories of knee joint bending angle in three directions

Knee joint presented large variation for different pedestrian preimpact 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 6a](#)) and 15.6 degree for the walking posture ([Figure 6b](#)).

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 6a](#)). 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 tension in knee joint caused by lateral bending. Tibial external rotation exhibited a significant oscillation ([Figure 6](#)) 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.