Application of a Finite Element Lower Extremity Model for Simulation of Long Bone Fractures in Vehicle-Pedestrian Collision

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Abstract: ,A finite element model of lower extremity was used. to investigate biomechanical response and injury mechanisms of lower extremity in vehicle-pedestrian impact. The model consists of the pelvis, the femur, the tibia, the fibula, the patella, the foot bones, primary tendons, knee joint capsule, meniscus, and ligaments. This model is constructed using shell and solid elements also including linear spring-damper elements, it contains of 29777 nodes and 30873 elements. The whole model was improved and validated by simulation of shearing-bending cadaver tests described in literature. Then the FE model of lower extremity and an existing multi-body system (MBS) model of the pedestrian were used to reconstruct a real-world vehicle-pedestrian accident and lower extremity injuries. Results of injury reconstruction show that this FE model has a good biofidelity and biomechanical response, and can be applied to research on the injury mechanisms. It can be used also in the development of injury protective devices.

Key Words: FE model; lower extremity injury; injury reconstruction; vehicle-pedestrian accident.

1. Introduction

Road safety is a serious problem throughout the world, especially the pedestrian safety. To improve it we need adequate tools for validation of safety performance of vehicles. In the development of such tools the information about outcome from accidents is crucial. Also a good knowledge about human tolerances and injury mechanisms is necessary.

In 2007 in China, pedestrian accounts for about 26 percent of all the road crash related fatalities and for 18.6 percent of people injured [1]. To investigate in detail this problem, in Changsha at Hunan University, an ad-hoc working team was organized to develop an in-depth automobile traffic accident database. KONG et al. [2] collected and analyzed accident cases from Changsha occurred between 2000 and 2006. She showed that lower extremity injuries occupied 27.2 percent of all injuries in pedestrian accidents. Although, they were non-fatal, they caused long-term impairment and disability, and resulted in huge social and economical costs. These injuries occurred, duo to the initial impact by the frontal structures of a vehicle. Based on YANG [3], in such type of accident the most common injuries are to lower extremity as: long bone fractures, knee injuries (including femoral condyle and tibia condyle fractures, knee ligament tearing and rupture), patella fracture, ankle/foot dislocation and fracture. These injuries are generated due to various injury mechanisms. Thus understanding of these mechanisms is very important for protection of lower extremities.

Over the past decades, the injuries and tolerances of the human lower extremity in car-pedestrian impacts have been widely studied mainly experimentally using human cadaver specimens [4-5]. In these studies, fracture of the tibia and failure of the knee structure were frequently observed (Figure 1). KAJZER et al. [6-7] performed impact tests to the knee using cadavers at a low and a high velocity, and indicated that knee injury types differ due the shear and bending mechanism acting on the knee joint. Based on his experiments, the injury criteria and risk functions were presented by MATSUI et al. [8].

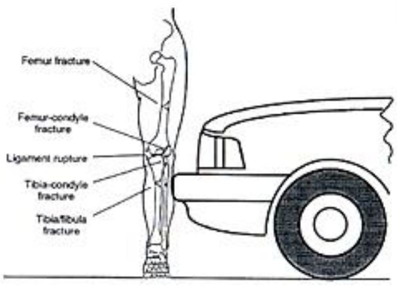


Figure 1 Major injuries of lower extremity in a car-to-pedestrian impact[9]

Currently the computer simulation techniques have been used for modeling of vehicle-pedestrian impacts. Multi-body system (MBS) and finite element method (FEM) are two approaches commonly used. YANG et al. [10] developed a MBS model of the lower extremity with a human-like knee joint to investigate the dynamic response and predict the injury risk of these structures exposed to a lateral impact loading. The MBS models are useful to predict kinematics, forces, accelerations etc. in a dynamic system. However, these models can't be used to calculate the stress and strain distribution within the human tissues. These disadvantages can be solved in finite element method. Several FE models were developed over the years. YANG et al. [11] developed a finite element model of skeleton of the human lower extremity to investigate dynamic responses to lateral impact loading. TAKAHASHI et al. [12] showed that the shear displacement and bending angle depend on the location of impact, and proposed injury criteria of knee ligaments based on the combination of knee displacement and bending angle. MIZUNO et al. [13] developed a finite element model based on "THUMS" and demonstrated that the deformation of a lower extremity is influenced by the level and direction of impact. Another finite element model of the lower limb model to simulate pedestrian impacts was developed in US by UNTAROIU et al. [14]. Similar models were developed within European project "HUMOS" and also in India.

In China however, we lack such basic finite element models to evaluate the injury risks and mechanisms. Therefore more domestic research is necessary to understand this and to improve pedestrian protection. Though FANG et al. [15] developed a FE lower extremity model; it didn't show a good biofidelity and couldn't predict the knee ligaments injuries. Therefore the purpose of the current study is to improve this preliminary model, and to use it in accident reconstruction and to study injury mechanisms.

2. FE model of lower extremity

The Human Body Model (HBM) of lower extremity has been previously developed using LS-DYNA 3D code. In the current study we checked it re-meshed some parts of the model, made a new selection of material properties and validated the model against results from cadaver tests.

2.1. Geometry and meshing

The geometry and mass distribution of the model are based on measurements of anatomical data available from Viewpoint Company. In the model all main anatomical structures of the 50th percentile male from the hip joint to the foot are represented. It consists of: hip, pelvis, femur, knee joint, tibia, fibula, patella, foot bones and soft tissues (Figure 2). These bones are composed of elements representing spongy bone and compact bone. The soft tissues are composed of elements representing, ligaments, tendons and meniscus. In the current model the skin, muscles and cartilage are not considered.

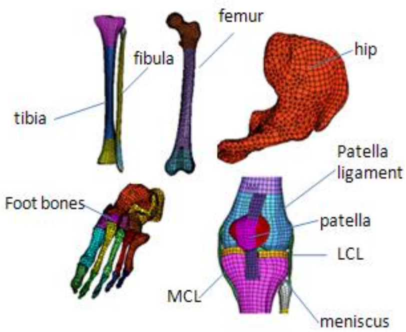


Figure 2 The HBM of lower extremity FE model

The whole model is meshed by Hypermesh (Altai, ver.8.0) and contains 53 components, 29124 nodes, 18630 solid elements and 12427 shell elements. The effective mass is 13.4 kg.

Shell, solid and linear spring-damper elements are used in construction of different regions of the model. The diaphysis regions of the cortical parts of the femur and tibia are meshed using shell elements, while the epiphysis regions the cortical bones were meshed using solid elements. The all cancellous bones are meshed with solid elements. The elements representing cancellous bone are connected with the elements of cortical bone according to tied contact interface. The menisci of the knee are meshed using solid elements on the tibia platform using a tied-node contact interface. This solution will constrain the movement of the menisci in relation to tibia due to dynamic deformation of the knee. The major ligaments: the medial collateral ligament (MCL), the anterior cruciate ligament (ACL), the posterior cruciate ligament (PCL), the lateral collateral ligament (LCL) and the patella ligament (PL) are modeled with shell elements. Tied contact interface were defined between the elements of ligaments and the cortical bone of the femur, tibia and fibula. Some spring elements with a low stiffness are used to represent the knee capsule.

The quality of elements was controlled when mesh was generated. The targets jacobian was larger than 0.7 and aspect ratio less than 0.5. In the current model only 2% of the shell elements' jacobian is less than 0.7. The average of the solid elements' jacobian is 0.7, with the minimum of 0.49. In this manner, the results of the calculations wouldn't be influenced by the mesh quality.

2.2 Material properties

The majority of material properties are derived directly from the literatures [16-17]. The behavior of tissues such as bones, ligament, cartilage and tendons are non-isotropic visco-elastic [18]. To simplify the model, in current study, cancellous and cortical bones are modeled using isotopic elastic-plastic material (Table 1). The material properties of the ligaments are defined using an elastic-plastic model with rate-dependent characteristics. The ultimate strain is used to define the threshold to initiate the element elimination process. Finally the menisci are modeled using an elastic material (table 2).

Table 1 Material properties of the elements representing bones

Bones	ρ (kg/m^3)	E(MPa)	δ (MPa)	ν	δ $_{max}$	ε $_{max}$ (%)
Femur cortical bone	2080	13500	115	0.3	124	1.7
Femur head bone	1900	900	9.3	0.315	9.3	13.4
Femur proximal	2000	616	6.6	0.3	6.6	13.4
Tibia cortical bone	1900	20033	129	0.315	134	1.6
Fibula cortical bone	1000	15000	125	0.3	134	1.6
Tibia cancellous	1000	445	5.3	0.3	5.3	13.4

Table 2 Material properties of elements simulating soft tissues

Soft tissue	ρ (kg/m^3)	E(MPa)	δ (MPa)	ν	δ $_{max}$	ε $_{ma(}$(%)
MCL,PCL,LCL,MCL	1100	345	29.8	0.22	36.4	15-20
Meniscus	1500	250		0.3		

2.3 Methodology of model validation

The tibia segment of the model had already been validated against the published three-point bending test of leg specimens [19]. In the current study whole FE lower extremity model is validated against the cadaver experiments in shearing and bending conditions conducted by KAJZER et al. [6]. The shearing and bending test don't represent conditions of a typical pedestrian accident but two simplified conditions when we can expect different patterns of deformation of the knee region (Figure 3). The simulation of these conditions was commonly used in validation of other FE models of lower extremity [11, 12, 13]. In his study cadavers were laid on a stable table, and the lower extremity was preloaded with a force of 400N to simulate the upper body weight The femur bone was fixed at the upper and lower locations with rigid support, a foam-covered impactor of 6.25 kg hit the ankle region (bending configuration) and below the knee joint (shear configuration) at a velocity of 40 km/h. He also published dynamic pressure-strain characteristics of the foam from the impactor. The relative movement of femur and tibia has also been published [6, 22] and the pattern of tibia movement: targets P1 and P2.

In the current study, the simulation test set-up is performed according test procedure described above but in vertical position (Figure 4). Therefore mass distribution of the pelvis is adjusted to 400N to simulate the upper body mass. Two points (P1 and P2) are selected on the lower extremity to indicate the displacement of the tibia in relation to the femur. This displacement is compared with the results from the experiments. The cadaver tests are very hard to reproduce due to its anatomical differences. Therefore four tests were used to validate the dynamic response and prediction capability of injuries by the HBM-lower extremity in two loading conditions: shearing (S) and bending (B). The data from tests 16S and 14B published by KAJZER et al. [6], and tests 8S and 7B published by MIZUNO et al.[13] was used in model validation because the size and the weight of the cadaver were similar to that of 50 percentile model.

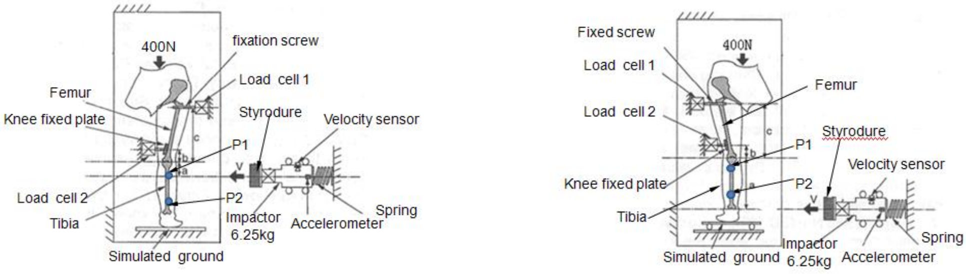


Figure 3. Knee loading test in shearing (left) and bending test (right) [6]

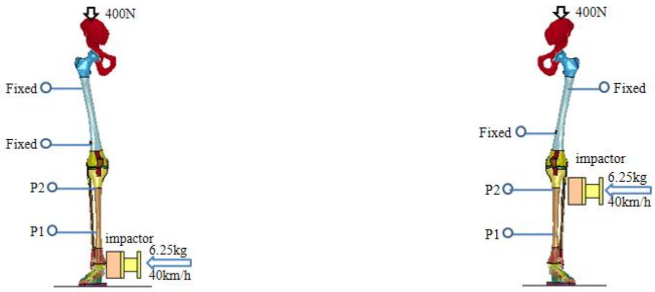


Figure 4. FE simulation of shearing (left) and bending (right) test condition

2.4 Validation: Comparison of displacement

The comparison of displacements of selected points on tibia from simulation and experiments at 40 km/h is shown in Figures 5 and 6. One can see that modified model shows very good agreement between calculated displacements of points P1 and P2 of tibia, but there is also some difference between the simulation and cadaver tests. The first difference can be observed during the initial 3ms, there is no displacement of the point P1 and P2, due to the free movement of impactor before contact to the bones. Another reason can be the fact that the current model is lacking muscles and that would influence the contact force between impactor and the model, and then also influence the movement of the P1 and P2. The third reason may be the way to make a model of the knee joint. For simplification in this model spring elements are used.

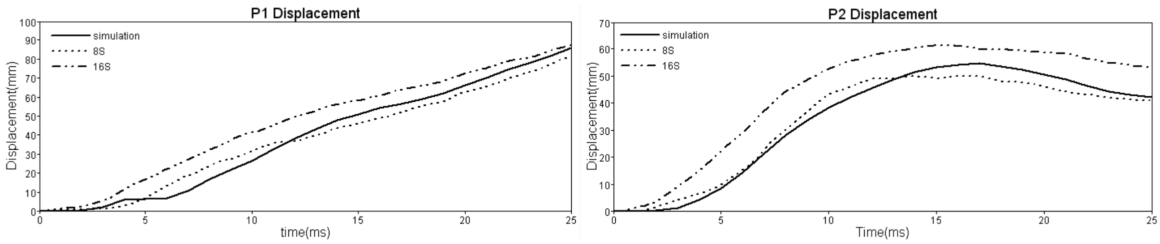


Figure 5. Comparison of displacement between FE simulation and experiment results (shearing test)

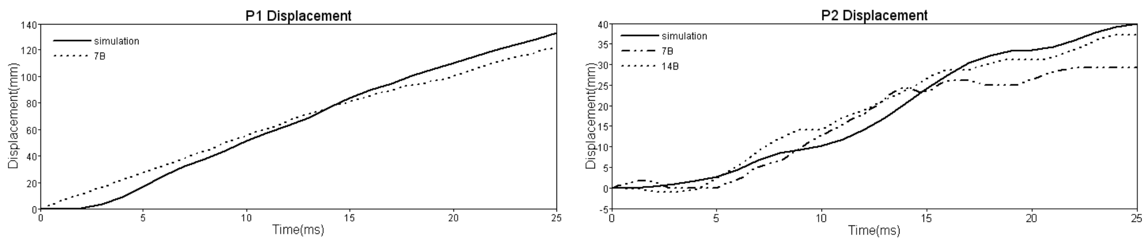


Figure 6. Comparison of displacement between FE simulation and experiment results (bending test) result from Mizuno et al. [13]

2.5 Validation: Comparison of injures

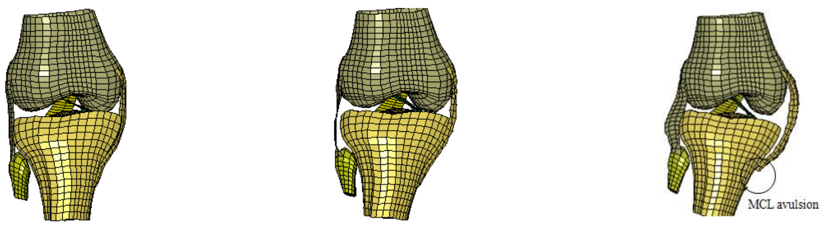
Table3 lists the injuries predicted by the lower extremity model in the impact simulations.

Table 3 Lower extremity injuries from the simulation of Test 8S/16S and 7B/14B

No. of simulation	Lower extremity injury
8S/16S	Fracture of femoral diaphysis
	Fracture of fibular diaphysis
	ACL avulsion
7B/14B	MCL avulsion

In the simulation 8S and 16S, the femoral diaphysis was fractured at the location of the fixation plate, and the fibula also fractured, these injuries were not observed in simulated experiments. However, this type of injuries was frequently observed (70% of the all cadaver tests) in other cadaver tests performed in the same experimental set-up. The ACL avulsion was observed both in simulations and comparable experiments. The time of ACL avulsion was also correctly predicted in simulations, at 5.9 ms after the impact, only 1 ms later than in the 16S experiment. The prediction capability of injuries in shearing configuration of the model is also acceptable.

In the simulation 7B, the injury to medial collateral ligament (MCL) was observed (Figure 7), but in the tests 7B and 14B, only femoral diaphysis and the proximal femoral fracture was found, respectively. A reason of these differences between prediction of injuries by the model and real experiments can be simplification of the knee joint. It isn't strong enough to transfer the force to the femoral condoyle in the process of deformation. However the type of injury predicted by the model is commonly observed in another study of the knee exposed to bending performed by KERRIGAN [20]. Also the time of MCL avulsion in the simulation can be compared with the average time (9.7 ms) based on his study. Thus the whole validation results can be treated as acceptable.



0 ms 6 ms 9 ms

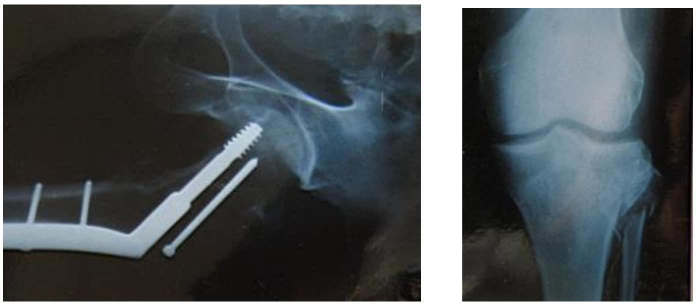
Figure.7 MCL avulsion in the simulation of bending test

3 Pedestrian accident analysis

3.1 Case description

One case was selected from the database existing at Hunan University. This database is developed by accident research team that carried out in-depth investigations of vehicle traffic accident from Changsha city, the capital of Hunan province. The team is in co-operation with traffic authority and hospital in Changsha. Each accident case contains detailed information about configuration of accident, vehicle deformation, contact interactions and also description of injuries. The injuries are coded according to the Abbreviated Injury Scale (AIS).

Case information: A male pedestrian of 50 years old was impacted by a small family car (SFC) when he tried to cross the road but stopped near the middle yellow line to wait the approaching car to pass firstly. The driver discovered the pedestrian and emergency braked the car, however, still hit the pedestrian. The pedestrian suffered several injuries. The injuries to the lower extremities were: intertrochanteric fracture (AIS 3) (Figure 8a), tibia & fibula proximal fracture (AIS 2) (Figure 8b). These injuries were due to impacted by the bumper and hood edge initially.



(a) (b)

Figure. 8 The injuries' X-ray pictures

3.2 Pedestrian accident reconstruction

Multi-Body System (MBS) pedestrian models and car models were used to reconstruct the accident considering the final position of the car and pedestrian. This way the impact velocity and relative impact position of the car and the pedestrian could be determined. This information was used as the input in the reconstruction of injuries to lower extremity by FE model (Figure 10a).

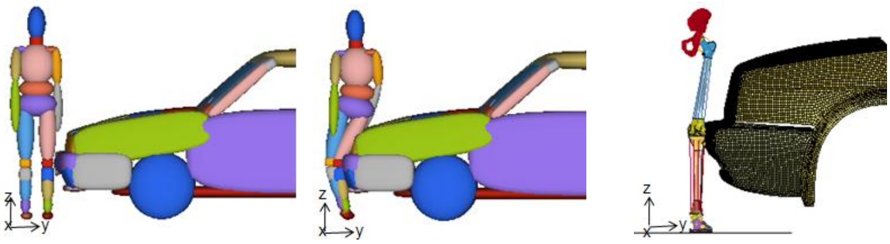
The MBS model used in reconstruction need to show a good biofidelity to correctly predict the whole accident event. The MBS pedestrian models used in the current study were developed and validated by YANG, et al. [21-22]. The model consists of 15 ellipsoids representing head, neck, chest, abdomen, hip, upper and lower extremities and connected by 16 spherical joints. The model can predict the lower extremity fracture by two frangible joints between tibia segments. The tolerance reference of joints is for the force 4 kN and for the moment 200 Nm.

The MBS car model was developed based on the geometry data of the car that was involved in accident. The car front structure and shape were represented by several ellipsoids. The stiffness characteristics of the bumper system and hood system of the model were defined according to the stiffness corridors based on results from Euro-NCAP reported by MARTINEZ et al. [23].

The accident reconstruction was performed in the MADYMO code. The car model and the braking distance were taken from accident records. The position and orientation of the pedestrian relative to the car model was determined based on the case record, as well as the contact points. The pedestrian initial velocity was defined as 0 m/s because he didn't move at the impact moment. The goal of the accident reconstruction process was to estimate the impact speed when the final position of the car and pedestrian, the final distance between the pedestrian and the car in simulation was similar to these from the accident.

3.3 Modeling of the lower extremity injuries

In the real-world accidents, different impact conditions lead to different injures of lower extremity. In order to correlate physical parameters with injuries to the lower extremity in lateral impact, the validated HBM-lower extremity model was used to reproduce them. Von Mises stress of bones was used as indicator of injuries. The FE reconstruction of lower extremity injuries was carried out based on the conditions that come from the output of the MBS accident reconstruction (Figure.9b). These conditions contain the impact speed of the car and the position of the pedestrian relative to the car and the initial posture of the lower extremity when initial impact to the pedestrian occurred. A dynamics process of 70 ms was computed using LS-DYNA code. The phenomenon of fracture was simulated by element elimination when they reached ultimate stress.



(a) (b)

Figure 9. The kinematics to the pedestrian and FE model for injury reconstruction

3.4 Results: Pedestrian accident reconstruction with MBS model

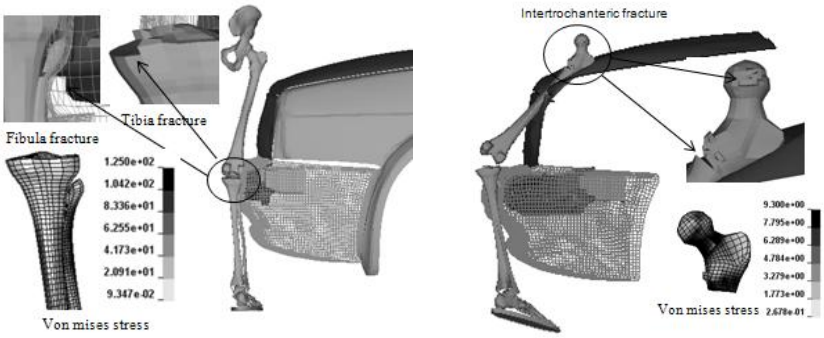
Several computations were performed before the goal could be fullfield. Finally the model could correctly predict the final position of the car and pedestrian, the final distance between the pedestrian and the car, the contact interaction between them. Also the force level of frangible joints between tibia segments was above the tolerance value used in the study. It indicated the fracture risk to this segment. Therefore the results of the final reconstruction were acceptable and the simulation could be used to calculate input parameters (velocity) for FE reconstruction. The velocity components of the car at the impact moment are shown in Table 4.

Table 4. Initial velocity of the car for FE reconstruction

Parameters	Case
Velocity $v_x/(m \cdot s^{-1})$	0.032
Velocity $v_y/(m \cdot s^{-1})$	-7.432
Angular velocity $\omega_z/(rad \cdot s^{-1})$	-0.046

3.5 Results : Injury reconstruction with FE model

The injury reconstruction results are shown on figures presented below. The Figure 10a shows that the tibial plateau and proximal fibula fractured duo to impact by the bumper. The Von Mises stress of the proximal fibula was up to 125 MPa. The Figure 10b shows that the femur intertrochanteric cancellous bone fractured duo to impact by the hood edge and the Von Mises stress was up to 9.3 MPa.



(a) Tibia and fibula (b) Femur intertrochanteric

Figure 10. The stress distribution of the components & fracture phenomenon

4. Discussion

The first part of the paper describes the development and the validation process of the advanced FEM of lower extremity. The tibia segment was validated by three-point-bending test and the whole FE model by result from cadaver tests. Though the P1 and P2 displacement show a good agreement with the experiments, some possible errors of geometry of this model can exist. The lack of muscles also can influence on the response of the model.

In current study, a real-world accident case was also reconstructed with MBS and FE methods. The MBS models (pedestrian and car) were used to simulate the kinematics of human body components. The output of the MBS was defined as the initial boundary conditions which were used for reconstruction of injuries with HMB-lower extremity FE model.

The results of the MBS reconstruction were compared with the records from the police and hospital data. They showed good agreement with the real-world accident.

The injury reconstruction with FE model shows that the result could reflect the injury mechanism of lower extremity in the vehicle-pedestrian impact. The Von Mises stress (125 MPa) of the proximal fibula and tibia predicted by the model is in the interval (100-125 MPa) of the ultimate stress reported by BEILLAS [17]. The cancellous bones of proximal tibia and femur intertrochanteric region were also failure. It showed good agreement with fractures observed on CT scan of the lower extremity (Figure 8). These results demonstrate that bone fractures can be generally predicted by this model.

The results from current study indicated that the detailed real-world accident case information obtained by in-depth investigation is very useful to evaluate the validity of FE model, by comparing the injuries response with the real injuries information.

5 Conclusions

A finite element HBM-lower extremity was improved and further validated. It shows a good biofidelity and has capability to predict long bone injuries in lateral impact to pedestrian.

The calculated Von Mises stress 125 MPa of the proximal fibula and tibia from injury reconstruction can be a relevant predictor for long bone fracture.

This study showed that the combination of MBS accident reconstruction and injury reconstruction with Finite element model is a good tool in investigation of injury mechanisms in real-world accidents.

6 Acknowledgements

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