Preliminary Validation of a Detailed Finite Element Model of a 50th Percentile Male Pedestrian

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Abstract

The pedestrian is one of the most vulnerable road users and comprised about 22% of the road crash-related fatalities in the world. While pedestrian protection regulations involving subsystem impact tests have been proposed, they cannot capture the whole vehicle-pedestrian interaction during car-to-pedestrian collisions (CPC). A few pedestrian finite element (FE) models representing 50th percentile male (M50) have been developed and validated previously. However, the existing FE models have several limitations, such as neglected/simplified body parts. To better predict crash-induced injuries observed in pedestrian accidents, a detailed pedestrian FE model was developed and preliminary validated in this study. The model geometry was reconstructed using a multi-modality protocol from medical images and exterior scanned data corresponding to a mid-sized male volunteer. The material properties of the pedestrian model were assigned based on the Global Human Body Models Consortium (GHBMC) M50 occupant model. The lower extremity and upper body of the FE model were validated against post mortem human surrogate (PMHS) test data recorded in valgus bending and lateral/anterior-lateral blunt impact tests, respectively. Then, the whole body pedestrian model was impacted by a vehicle FE model corresponding to a mid-size car used in PMHS tests. All validations were performed with LS-DYNA® software. In the component validations, the M50 model's responses were close to the mean corridor of the PMHS test data. The kinematic trajectories predicted by the M50 pedestrian FE model during CPC validation were close to the corresponding trajectories recorded on taller PMHS subjects. This is the first study to develop and validate the 50th percentile male pedestrian FE model including the internal organs, muscles, and vessels in order to evaluate various pedestrian injury mechanisms during traffic accidents. Compared to previous pedestrian models developed by various auto manufactures, which are proprietary, this pedestrian model will be publically available for academic researchers. Overall, good model predictions recommend using the M50 model in European New Car Assessment Program (Euro NCAP) protocol and in automotive safety research for the development of more pedestrian-friendly new vehicles.

Introduction

According to the World Health Organization (WHO), the total number of fatalities recorded in traffic accidents is about 1.25 million each year worldwide [1]. Among the road traffic deaths, in 2015, pedestrian fatalities represented 22% (World), 26% (Europe), and 22% (Americas) of total traffic fatalities. In the United States, approximately 5,376 pedestrians are killed and 70,000 injured annually in road crashes [2]. Historically, the major challenge of automotive safety research was to improve the design of both vehicles and restraint systems to reduce injuries for vehicle occupants. Consequently, the occupant fatalities have declined 46% since 1975 [3]. However, the pedestrian fatalities have been increased since 2006 [2] while the total traffic fatalities have been decreased. Therefore, the pedestrian protection during car-to-pedestrian collision (CPC) has generated increased attention with regulations.

Recently, subsystem tests with impactors corresponding to an adult/child head and adult upper/lower legs [4] were implemented in regulations. While these subsystem tests can help in reducing the stiffness of vehicle front-end components and consequently reduce the risk of injuries, neither the complex vehicle-pedestrian interaction nor the injury mechanisms can be characterized by these simple impact tests.

To characterize the whole-body response of vehicle-pedestrian interactions, several pedestrian FE models representing 50th percentile male have been developed and validated previously [5-11]. However, the existing models have several limitations related to limited geometrical data, neglected/simplified body parts, and simplified models of the material properties due to insufficient PMHS test data available at the time the models were developed and validated. The objective of this study was to develop and validate a detailed pedestrian FE model corresponding to a 50th percentile male anthropometry for characterize the whole-body response and assess the pedestrian injuries in vehicle-pedestrian interaction.

Methods

Development of the Detailed Finite Element Model of 50th Percentile Male Pedestrian

A detailed 50th percentile male pedestrian (M50-P) model was developed in LS-DYNA software (LSTC, Livermore, CA) based on anthropometry of a recruited male subject (26 years old, 175cm height, 78kg weight). The external anthropometry data was collected using a 3D scanner (Faro, Platinum Model arm, 8ft. (2.4m), Lake Mary, FL). The subject was able to preserve the head in the Frankfurt plane using an adjustable height photo target on the opposing wall of the lab. The external anthropometry data and surface scans were integrated together for the generation of a non-uniform rational basis spline (NURBS) patchwork of the average male outer surface in this neutral stand posture. A multi-modality protocol was used to acquire data in a pedestrian posture [12]. The final heel to heel distance (314.2mm) and H-point height (949.7mm) of M50-P model met the European New Car Assessment Program (Euro NCAP) standard (310±10mm heel to heel distance and 938mm±5% H-point height).

The mesh of the M50-P model is mostly adapted from a 50th percentile male occupant (M50-O) model, but the other regions including thorax, abdomen, pelvis, and lower extremities were re-meshed to acquire the pedestrian posture. The material properties of the M50-P model were assigned mostly based on M50-O model as well. The body region models of this occupant model were developed and validated by different groups and assembled together [13-19]. Compared to the simplified 50th percentile male pedestrian (M50-PS) model [9], this model was developed as a detailed version that was modelled and included the muscles, vessels, internal viscera, and deformable brain components which was neglected/replaced to simplified parts previously (Figure 1). Overall, the M50-P model comprised of total number of 1,251,274 nodes and 2,324,738 elements (about 97% deformable elements).

Figure 1. FE model comparison between simplified (left) [9] and detailed (right) models

Validation of the knee joint under valgus bending

Combined valgus bending and shear loading of the knee joint have been recognized as primary injury mechanisms during a pedestrian accident [20]. To validate the injury biomechanics of the knee joint under lateral loading, a four-point bending PMHS test reported in the literature [20] was simulated (Figure 2). The knee joint FE model was extracted from the whole-body model in standing posture (straight knee) and then re-positioned manually to fit into the simulation setup [9]. The ends of the three bones (femur, fibula, and tibia) were positioned approximatively in the center of cups and then rigidly constrained to the bone cups in a depth of 76 mm (the interior length of the cup). Each part of the apparatus was defined as rigid, except the load cell part attached to the bone cup, which is deformable in order to calculate the bending moment using a cross-section card [21]. The bone cups were attached to extension beams that are linked to the rotational joint supports. The initial anterior-posterior knee axis was approximately parallel with the axis defined by the support centers. The support on the femur side was allowed to slide horizontally, while the other support was fixed. To load the knee joint under valgus bending, the extension beams were rotated at a knee angular rate of approximately 1deg/ms in correspondence with a 40km/h CPC impact [22].

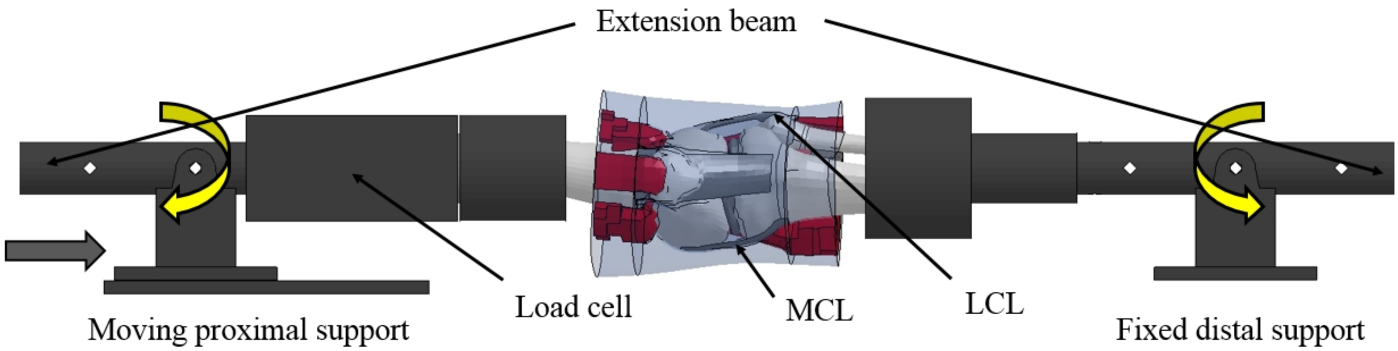

Figure 2. FE simulation setup of four-point knee joint bending

Validation of the thoracoabdominal region and pelvis under lateral impact loading

The torso of the whole-body FE model was validated against PMHS test data subjected to blunt lateral/anterior-lateral impact loading [23]. Based on the test scenario, eight FE simulations performed using the pedestrian FE model under various combinations of impacted regions (pelvis, abdomen, and thorax) and impactor velocities. To avoid interference between arm and impactor during validation, the upper extremities were removed but a concentrated mass corresponding to each arm was applied near the scapular region to maintain the pedestrian total mass (Figure 3. a-c). The pelvis was loaded laterally with the impactor aligned adjacent to the greater trochanter (Figure 3. a). The center of the impactor was aligned to the xiphoid process for the thorax impact and 7.5 cm down from the xiphoid process for the abdomen impact. Then, the impactor was rotated 60° around the vertical (z) axis before impact to the model (Figure 3. b, c).

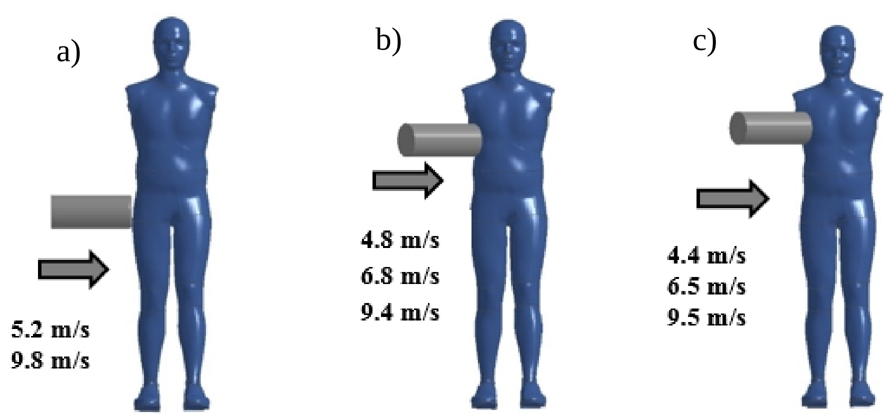

Figure 3. FE simulation setup of lateral and anterior-lateral impact loadings at upper body
a) pelvis, b) abdomen, c) thorax

Validation of the whole body

The whole-body FE model was validated in a CPC scenario based on PMHS test data [24]. The M50-P model was positioned laterally at the centerline of a mid-sized sedan FE model used in testing to represent the most frequent accident scenario [25]. The posture of the FE model was set to mid-stance with the legs apart walking towards the vehicle centerline and the rearward leg being impacted first by the vehicle (Figure 4). This configuration was based on the pedestrian testing protocol of Euro NCAP [4]. In FE simulation, prior to the impact at approximately 5ms, the gravity acceleration was assigned to the pedestrian model and a force corresponding to its body weight was applied upward by the ground part to initiate the foot-ground contact. Then, the pedestrian model was impacted by the vehicle FE model which was validated by its manufacturer against pedestrian subsystem test with 40km/h initial velocity [26]. During testing, markers were attached to the PMHS head, first thoracic vertebra (T1), and sacrum and their kinematics were recorded relative to the car using the high-speed video [24]. Nodes corresponding to these markers were defined in the pedestrian FE model at the head's center of gravity (CG), and about 60mm behind of the T1 and sacrum due to the consideration of the screw length (Figure 4) [9]. In addition, a Telemetry Data Acquisition System (TDAS) bag was attached to the PMHS during testing. While the bag dimensions and the attachment to the PMHS was not documented [24], the TDAS mass (about 4.3 kg) was added as four concentrated mass nodes rigidly to the lumbar spine (L1-L4). Static/dynamic friction coefficients were defined as the average values reported in literature [27]: 0.26/0.25 (fabric-to-steel) for vehicle-to-pedestrian contact and 0.61/0.45 (fabric-to-fabric) for pedestrian-to-pedestrian contact.

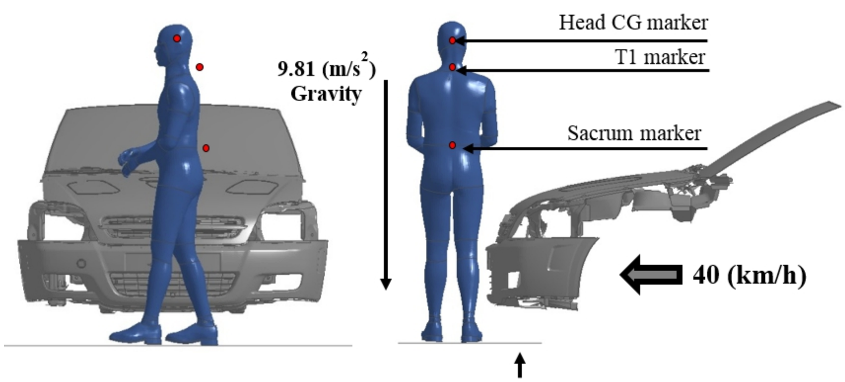

Figure 4. FE simulation setup of the car-to-pedestrian impact

Results

Overall, the M50-P model responses were within the PMHS test corridor and observed similar trend as the M50-PS model at component level validation [9]. In the whole-body validation, the M50-P model predicted responses close to the PMHS test data corresponding to the taller specimen.

Validation of the pedestrian model at component level

The knee bending stiffness of the M50-P model showed a relatively similar trend as the curves corresponding to the M50-PS model and scaled PMHS test data (Figure 5). In terms of injury, the M50-P model predicted the ruptures of Anterior Cruciate Ligament (ACL) and Medial Collateral Ligament (MCL) in a knee angle range from 10.7° to 15.4°.

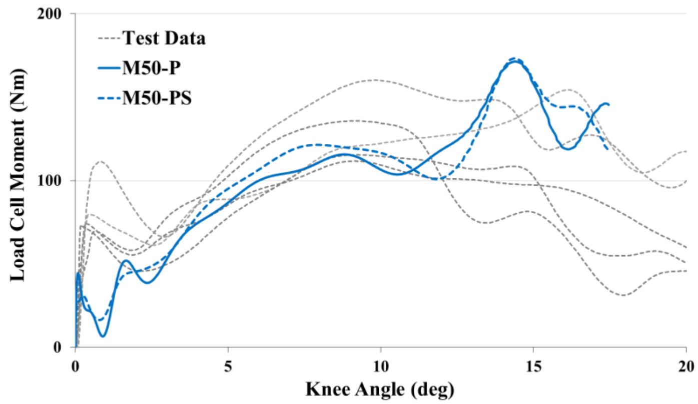

Figure 5. Responses of knee joint under bending loading: PMHS tests vs M50 models

The upper body validation responses of M50-P model under blunt lateral/anterior-lateral loadings at the regions of pelvis, thorax, and abdomen are mostly within the PMHS test corridor developed by 14 PMHS test specimens (53.8 ± 13.9 age, 67.2 ± 16.2kg weight). Compared to the M50-PS model response, the force time histories predicted by the M50-P model under high velocity pelvic lateral loading showed lower peak forces (Figure 6). In the abdomen and thorax validation, the responses of M50-P model under anterior-lateral impact loading showed closer to the mean values of test corridor than that of the M50-PS (Figure 7,8).

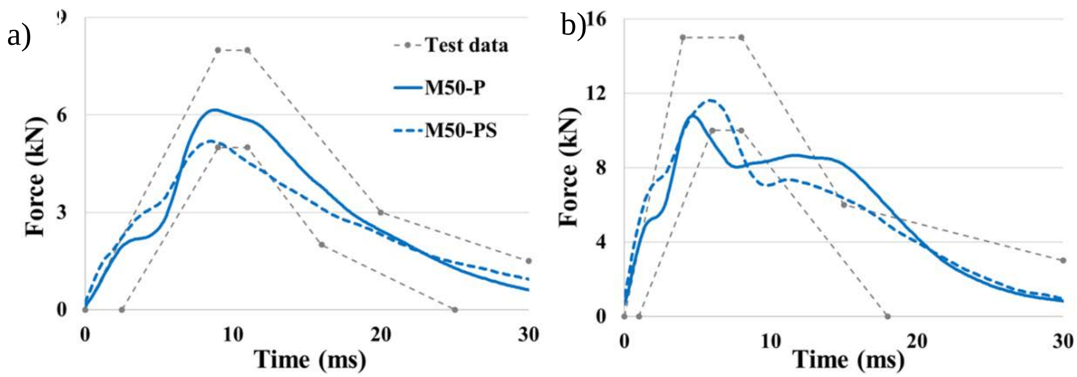

Figure 6. Time histories of impact force at pelvis: PMHS test vs. M50 models
a) 5.2m/s, b) 9.8m/s initial impactor velocity

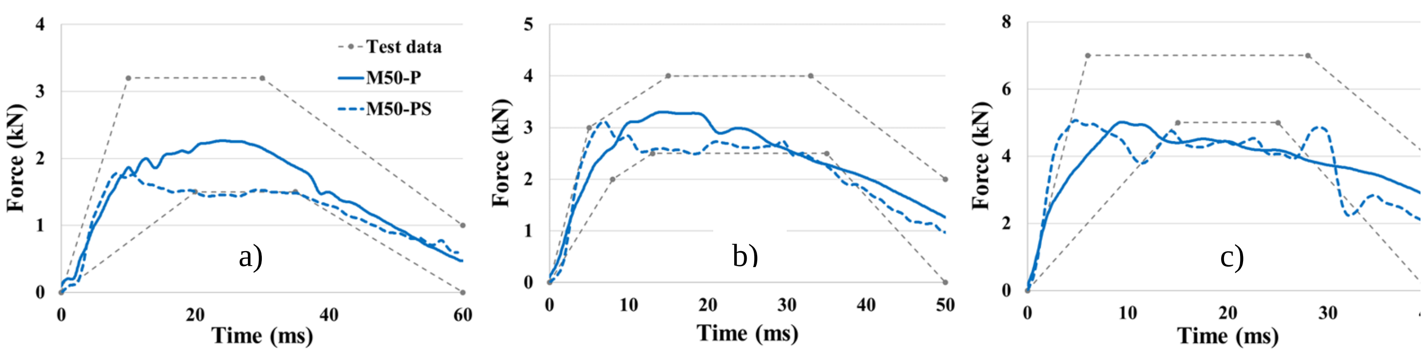

Figure 7. Time histories of impact force at thorax: PMHS test vs. M50 models
a) 4.4m/s, b) 6.5m/s, c) 9.5m/s initial impactor velocity

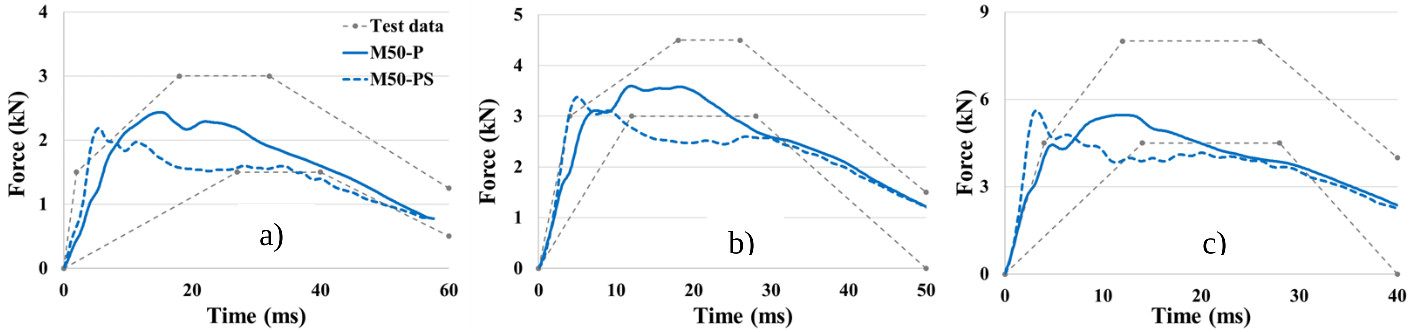

Figure 8. Time histories of impact force at abdomen: PMHS test vs. M50 models
a) 4.8m/s, b) 6.8m/s, c) 9.4m/s initial impactor velocity

Validation of the pedestrian model in car-to-pedestrian impact

The kinematic trajectories of FE model was recorded during CPC simulation and compared against the PMHS test data [24]. The beginning and the end of the outputs were defined at the time of initial right leg-bumper contact and the initial time of the head-vehicle contact, respectively. Overall, the kinematic trajectories predicted by M50-P model in frontal plane were close to the corresponding trajectories recorded on taller PMHS specimens (Figure 9).

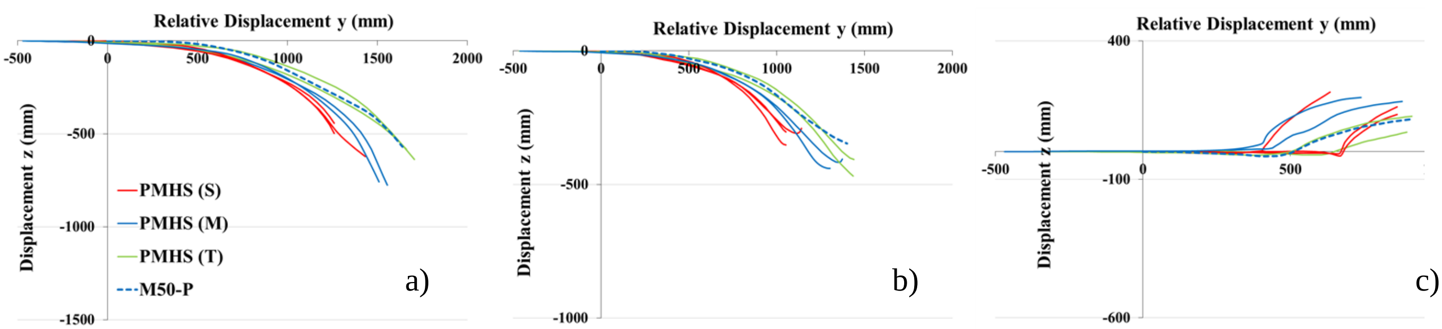

Figure 9. Kinematic trajectories relative to the vehicle: PMHS test vs. M50-P model
a) Head CG, b) T1, c) Sacrum, S: shorter, M: mid-size, T: taller specimen

Discussion and Conclusions

This study presents the development and preliminary validation of a detailed pedestrian FE model corresponding to a 50th percentile male anthropometry. This FE model is distinct from previous models in two aspects. First, the model geometric data were reconstructed using a multimodality protocol from medical images and exterior scanned data corresponding to an average male volunteer. The previous pedestrian FE models were developed using individuals with anthropometry different from 50th percentile male [10, 22] or commercial databases [8, 10] which possessed limitations (e.g. only exterior surfaces of bones). Second, unlike the simplified FE model comprised neglected/simplified body parts, this model was developed as a detailed version including muscles, vessels, and internal viscera. While the detailed FE model is as computationally efficient compared to the simplified pedestrian model M50-PS [9], it can better predict the injury mechanisms and biofidelic responses in vehicle-pedestrian interaction.

The results predicted by the M50-P model in component validations showed similar trends and mostly within the PMHS test corridor. The load cell bending moment predicted by the M50-P model was lower than in testing at the beginning of loading, probably caused by possible differences in the inertial properties of the test setup (model setup vs. physical setup). The knee joint stiffness of M50-P model showed mainly similar but higher peak value observed at about 15°. While this model was developed as detailed version, a complex of human knee joint was not modelled explicitly. Improvement of the knee model may include improving the geometry of knee capsule (which was approximated, not reconstructed from medical images), updating material model of ligaments, and/or modeling the synovial fluid. In terms of injury, the M50-P model predicted MCL and ACL ruptures which is the most frequent injuries observed in all 40 PMHS knee bending test (52.5% MCL ruptures and 10 % ACL & MCL ruptures) [20].

From the upper body validation under lateral/anterior-lateral impact, the M50-P model predicted biofidelic responses compared to the corresponding PMHS test data. However, a different pattern and lower peak value of impact force predicted by M50-P model were observed in higher impact velocity at pelvis. Since the pelvic flesh plays a significant role in lateral impact pelvic test, it is believed that a more biofidelic material model of flesh may improve the response of the model in this loading condition. Therefore, it is recommended to update the flesh material model when dynamic compression test data on pelvic flesh specimens becomes available. For the anterior-lateral impact responses on the thorax and abdomen, the M50-P model predicted high biofidelic responses compared to the M50-PS model. This better performance could be the effect of adding the internal organs into the rib cage of M50-P model that improved the damping properties of the upper body.

The kinematic responses of whole body FE model in CPC simulation predicted closer to the corresponding taller specimen as were classified by Kerrigan et al., not to a mid-size specimen [24]. While the height of M50-P model is close to the mid-size specimens (176cm FE model vs. 172.9/174.3cm specimens), the gender difference and discrepancy of body weight (77.3kg FE model vs. 90.6/92.9kg female specimens) are possible factors for disagreement. Since an interaction between pelvis and vehicle hood influences the trajectory of head CG and T1, a discrepancy of the height of greater trochanter (907mm FE model vs. 802/826mm specimens) also affected to the response. Recently, the light truck and vans (LTV) sales have been increased in US, thereby a new concern regarding pedestrian protection has emerged [28]. Nearly 88.7% of pedestrian fatalities were recorded with front impact by the passenger cars and light trucks including Sport Utility Vehicle (SUV), pickup, and van in 2015 [2]. In addition, a traffic accident statistics analysis that the pedestrian morbidity and mortality are higher when struck by the LTV than that of passenger cars [28, 29] and the injury risks of the head and chest are greater when struck by the LTV than in sedan car. Hence, the additional validation using M50-P model with different type of vehicles (e.g. SUV, truck, and vans) are suggested.

The pedestrian model presented in this study is the first step in developing detailed human FE models that can provide the specific injury biomechanics and biofidelic responses in pedestrian accident scenario. To cover whole population with respect to the anthropometry in pedestrian protection, the 5th percentile female and 95th percentile male detailed FE models are currently under development as well. Overall, good model predictions recommend using the M50-P model in new car assessment program protocol and automotive safety research to improve the pedestrian protection.

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References

[1] WHO, 2016, "Road traffic injuries."
[2] NHTSA, 2015, "Traffic Safety Facts," 2015file:///C:/Users/Student/Downloads/2015%20Pedestrians%20Traffic%20Safety%20Fact%20Sheet%20(1).pdf.
[3] IIHS, 2016, "General statistics,"http://www.iihs.org/iihs/topics/t/general-statistics/fatalityfacts/passenger-vehicles.
[4] EURONCAP, 2017, "Pedestrian Testing Protocol,"file:///C:/Users/Student/Downloads/euro-ncap-pedestrian-testing-protocol-v84%20(2).pdf.
[5] Arnoux, P. J., Cesari, D., Behr, M., Thollon, L., and Brunet, C., 2005, "Pedestrian lower limb injury criteria evaluation: a finite element approach," Traffic Inj Prev, 6(3), pp. 288-297.
[6] Iwamoto, M., Omori, K., Kimpara, H., Nakahira, Y., Tamura, A., Watanabe, I., Miki, K., Hasegawa, J., and Oshita, F., "Recent Advances in THUMS: Development of Individual Internal Organs, Brain, Small Female, and Pedestrian Model," Proc. 4th European LS-DYNA Conference.
[7] Mo, F. H., Arnoux, P. J., Cesari, D., and Masson, C., 2014, "Investigation of the injury threshold of knee ligaments by the parametric study of car-pedestrian impact conditions," Safety Sci, 62, pp. 58-67.
[8] Takahashi, Y., Kikuchi, Y., Konosu, A., and Ishikawa, H., 2000, "Development and validation of the finite element model for the human lower limb of pedestrians," Stapp car crash journal, 44, pp. 335-355.
[9] Untaroiu, C. D., Pak, W., Meng, Y., Schap, J., Koya, B., and Gayzik, S., 2018, "A Finite Element Model of a Midsize Male for Simulating Pedestrian Accidents," J Biomech Eng, 140(1).
[10] Watanabe, R., Katsuhara, T., Miyazaki, H., Kitagawa, Y., and Yasuki, T., 2012, "Research of the relationship of pedestrian injury to collision speed, car-type, impact location and pedestrian sizes using human FE model (THUMS Version 4)," Stapp car crash journal, 56, p. 269.
[11] Takahashi, Y., Kikuchi, Y., Konosu, A., and Ishikawa, H., 2000, "Development and validation of the finite element model for the human lower limb of pedestrians," Stapp Car Crash J, 44, pp. 335-355.
[12] Gayzik, F., Moreno, D., Geer, C., Wuertzer, S., Martin, R., and Stitzel, J., 2011, "Development of a full body CAD dataset for computational modeling: a multi-modality approach," Annals of biomedical engineering, 39(10), p. 2568.
[13] Beillas, P., and Bertet, F., 2012, "Performance of a 50th Percentile Abdominal Model for Impact: Effect of Size and Mass," J Biomechanics, 45(Suppli. 1), p. S83.
[14] DeWit, J. A., and Cronin, D. S., 2012, "Cervical spine segment finite element model for traumatic injury prediction," J Mech Behav Biomed Mater, 10, pp. 138-150.
[15] Mao, H., Zhang, L., Jiang, B., Genthikatti, V. V., Jin, X., Zhu, F., Makwana, R., Gill, A., Jandir, G., Singh, A., and Yang, K. H., 2013, "Development of a finite element human head model partially validated with thirty five experimental cases," J Biomech Eng, 135(11), p. 111002.
[16] Yue, N., and Untaroiu, C. D., 2014, "A numerical investigation on the variation in hip injury tolerance with occupant posture during frontal collisions," Traffic Inj Prev, 15(5), pp. 513-522.
[17] Shin, J., Yue, N., and Untaroiu, C. D., 2012, "A finite element model of the foot and ankle for automotive impact applications," Ann Biomed Eng, 40(12), pp. 2519-2531.
[18] Shin, J., and Untaroiu, C. D., 2013, "Biomechanical and injury response of human foot and ankle under complex loading," J Biomech Eng, 135(10), p. 101008.
[19] Untaroiu, C. D., Yue, N., and Shin, J., 2013, "A finite element model of the lower limb for simulating automotive impacts," Ann Biomed Eng, 41(3), pp. 513-526.
[20] Bose, D., Bhalla, K. S., Untaroiu, C. D., Ivarsson, B. J., Crandall, J. R., and Hurwitz, S., 2008, "Injury tolerance and moment response of the knee joint to combined valgus bending and shear loading," J Biomech Eng, 130(3), p. 031008.
[21] LSTC, 2013, "LS-DYNA KEYWORD USER'S MANUAL."
[22] Untaroiu, C., Darvish, K., Crandall, J., Deng, B., and Jenne-Tai, W., 2005, "A finite element model of the lower limb for simulating pedestrian impacts," Stapp car crash journal, 49, p. 157.
[23] Viano, D. C., 1989, "Biomechanical responses and injuries in blunt lateral impact," No. 0148-7191, SAE Technical Paper.
[24] Kerrigan, J. R., Crandall, J. R., and Deng, B., 2007, "Pedestrian kinematic response to mid-sized vehicle impact," International journal of vehicle safety, 2(3), pp. 221-240.
[25] Fredriksson, R., Rosén, E., and Kullgren, A., 2010, "Priorities of pedestrian protection—a real-life study of severe injuries and car sources," Accident analysis & prevention, 42(6), pp. 1672-1681.
[26] Kerrigan, J. R., Parent, D. P., Untaroiu, C., Crandall, J. R., and Deng, B., 2009, "A new approach to multibody model development: pedestrian lower extremity," Traffic Inj Prev, 10(4), pp. 386-397.
[27] Kothari, V., and Gangal, M., 1994, "Assessment of frictional properties of some woven fabrics."
[28] Lefler, D. E., and Gabler, H. C., 2001, "The emerging threat of light truck impacts with pedestrians," Rowan University.
[29] Henary, B. Y., Crandall, J., Bhalla, K., Mock, C. N., and Roudsari, B. S., "Child and adult pedestrian impact: the influence of vehicle type on injury severity," Proc. Annual Proceedings/Association for the Advancement of Automotive Medicine, Association for the Advancement of Automotive Medicine, p. 105.