

Validation of an FE Lower Limb Model for a Child Pedestrian by Means of Accident Reconstruction

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ABSTRACT

Validation of a child FE model is a great challenge due to the lack of sufficient data for children. In their previous study, the authors have developed a methodology for validating a child FE model by reconstructing pedestrian accidents and comparing predicted and observed injuries. However, the study reconstructed only one accident case and more validation cases were needed for enhanced confidence of the estimated material property. The current study therefore reconstructed two additional child pedestrian accident cases. In addition, published 3-point bending test results of child long bones were also used for quantitative assessment of the material property.

The accident cases were taken from the PCDS and CIREN databases. Based on the information from the accident data, a multi-body simulation and optimization procedure were used to identify impact conditions. The estimated impact conditions were used to simulate the accident cases using the FE model with failure criteria. The predicted injuries using various sets of material parameters representing individual variation were compared with observed injuries. Published 3-point bending tests for child femora and tibiae were simulated using the models for these bones. The upper and lower bounds of the estimated individual variation of the material parameters were applied to the models to obtain predicted response corridors and to compare with the test data.

The results of the FE accident reconstruction showed that the observed lower limb fracture patterns were reproduced within the estimated individual variation of material property. It was also found that most of the 3-point bending test results fell within the predicted force-deflection response corridors. Combining these two results, the accuracy of the estimated average material parameters and their individual variations were quantitatively assessed.

INTRODUCTION

In recent years, pedestrian tests have been used as one of the evaluation items for automobile crash safety performance in regulations as well as third-party assessments in Japan, Europe, and other countries. Based on this circumstance, research on pedestrian protection is becoming increasingly important. Currently these tests focus on evaluating the adult head and lower limb regions, while the test used for child pedestrians only covers the head region. Nevertheless, the most frequent type of serious injuries to child pedestrians is injuries to the lower limb region [1]. In order to reduce injuries to child pedestrians, there is a need for an evaluation tool that covers a child lower limb in addition to that for the head region. The authors have developed an FE child pedestrian lower limb model as an evaluation tool for lower limb injuries to child pedestrians [2]. However, validation of the child FE model was a remaining issue because experiments using child specimens are difficult due to ethics issues, and applicable mechanical property data is limited.

Some of the past studies have developed a child FE model for crash safety. Mizuno et al. [3] applied a geometric scaling to an adult FE model to create geometry of a child model, and the impact response of the model was validated against that of a child dummy. However, the impact response specifications for the child dummy were determined by scaling the impact response data for adults [4]. Meyer et al. [5] developed an FE head-neck model using CT scans from a child subject, and validated the model against response data obtained by scaling adult response data. Since these past studies validated their child FE models against child response data estimated by assuming geometric similarity and scaling adult response data, the accuracy of child impact response estimation was an issue.

To address this issue, the authors have proposed a methodology for validating an estimated material property specified in a child FE model by reconstructing

child pedestrian accidents and reproducing injuries sustained by the child pedestrians, and the proposed methodology was shown to be valid with one particular child pedestrian accident case [6]. In the current study, two additional child pedestrian accident cases from the Pedestrian Crash Data Study (PCDS) and the Crash Injury Research and Engineering Network (CIREN) databases as well as published 3-point bending test results for child femora and tibiae were used to further validate the estimated material property and improve confidence level by increasing the number of validation cases.

METHODOLOGY

MODEL VALIDATION AGAINST ACCIDENT DATA

The model validation procedure proposed by the authors in their previous study [6] is illustrated in Figure 1. A multi-body simulation and an optimization technique were used to identify pedestrian-vehicle impact conditions from contact information described in the accident data. Next, a baseline child pedestrian FE model representing an average 6-year-old child anthropometry was scaled to a case pedestrian based on the information from the accident data to obtain a child FE model representing the child pedestrian for each accident case. In addition, the average material property for the cortical bones of the femur and tibia and its individual variation for the specific age corresponding to that of the case pedestrian were defined to be used for reconstructing each accident case. Using the impact conditions identified by a multi-body simulations and an optimization technique, crash simulations were performed with the FE model for the particular case pedestrian. The predicted lower limb injuries were then compared with the injuries sustained by the case pedestrian to validate the material parameters and their individual variations specified in the FE model.

Accident Data

In this study, one child pedestrian case with lower limb fractures was chosen from each of the PCDS (Case 1) and the CIREN (Case 2) databases for which contact locations on the pedestrian and vehicle were clearly defined and described in the accident data.

For Case 1, the case vehicle was traveling in a two-lane roadway, and the case pedestrian was running across the roadway from east to west. The pedestrian was hit by the vehicle on her right side, wrapped onto the hood and slid off the right side after being carried for approximately ten meters. Two dents were identified on the hood presumably due to the contact with the pelvis and head of the pedestrian. The pedestrian rotated about the vertical axis during the collision, resulting in face contact to the hood. The pedestrian was an 8-year-old female with 138 cm of height and 63.8 kg of weight.

She sustained right tibia and fibula fractures due to the impact from the vehicle. The estimated vehicle speed upon impact was 40 km/h. The vehicle collided with the pedestrian without braking. The summary of the accident is given in Figure 2.

For Case 2, the pedestrian was crossing the five-lane roadway in the cross walk in front of a parked bus. As he noticed the case vehicle, the pedestrian jumped backwards attempting to get back in front of the parked bus. He had the left foot on the ground with the upper body moving backward when the right front bumper struck his left lower limb. His right foot was in the air. He went up a little on the fender, then was thrown straight forward onto the ground where he rolled. The pedestrian was a 11-year-old male with 148 cm of height and 56.7 kg of weight. He sustained left tibia fracture due to the contact with the vehicle. The estimated vehicle speed upon impact was 35 km/h. The driver's attempt to brake did not succeed in decelerating the vehicle prior to the impact. The summary of the accident is presented in Figure 3.

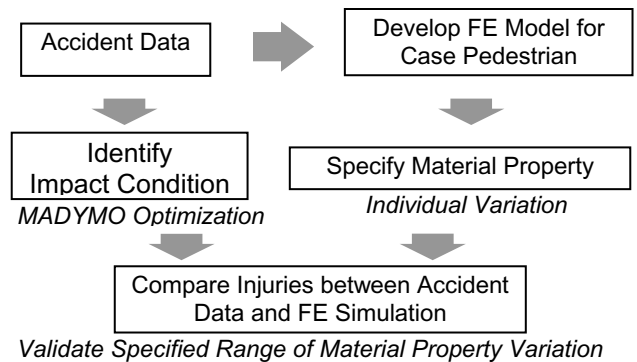


Figure 1 Schematic diagram of validation methodology

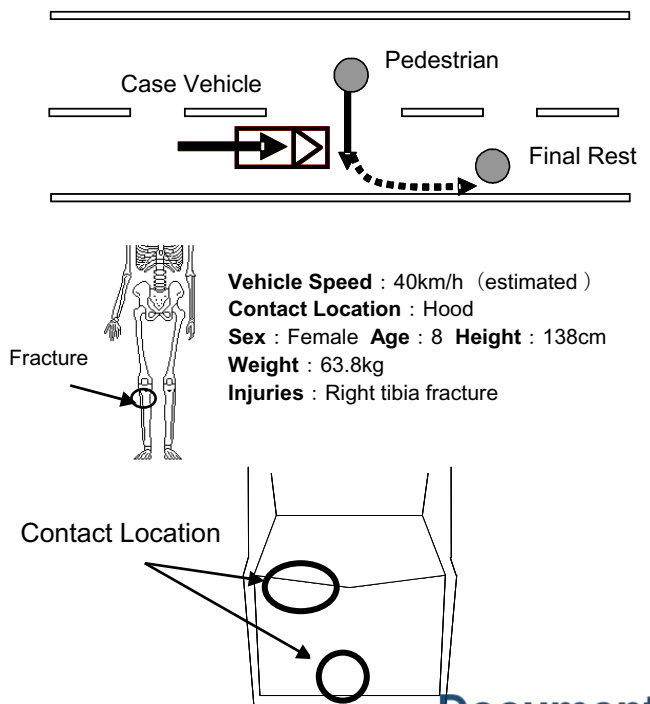


Figure 2 Summary of accident Case 1

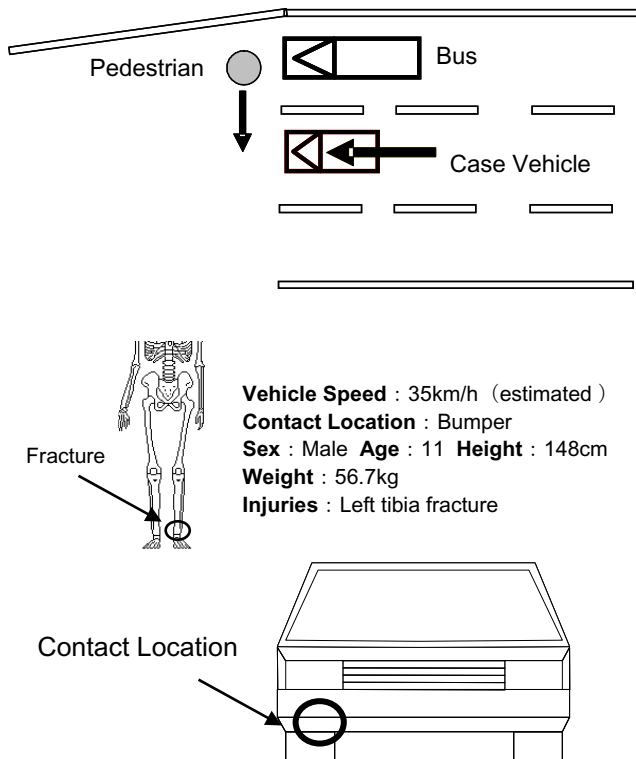


Figure 3 Summary of accident Case 2

Impact Condition Identification

Given that iterative simulation runs are necessary to identify impact conditions using an optimization calculation, MADYMO was used for simulating pedestrian-vehicle impact in order for pedestrian kinematics simulations to be run in relatively short time. The pedestrian model is based on the six-year-old child model from the MADYMO pedestrian model database, and was scaled to the anthropometry of the case pedestrian by referring to an anthropometry database (GEBOD). The geometry of the vehicle model was created by converting a surface model from the ViewPoint database into a MADYMO facet model. The stiffness characteristics of the vehicle front were estimated from an available FE vehicle model with front geometry similar to that of the case vehicle. Figure 4 shows the MADYMO vehicle-pedestrian impact model used in this research for both Case 1 and Case 2. A commercially available optimization software package was used in combination with MADYMO to identify the impact conditions using an optimization technique. The number of iterative simulation runs for optimization was set at 200. Small ellipsoids (targets) were placed in positions on the pedestrian and vehicle models corresponding to the pedestrian and vehicle contact locations, and the relative distances between them were calculated. As shown in Equation (1), the sum of the relative distances between the corresponding targets on the pedestrian and vehicle models was defined as an objective function, and optimization calculations were performed to minimize the objective function.

$$\text{Objective function} = \sum (\text{relative distance between contact location}) \dots\dots(1)$$

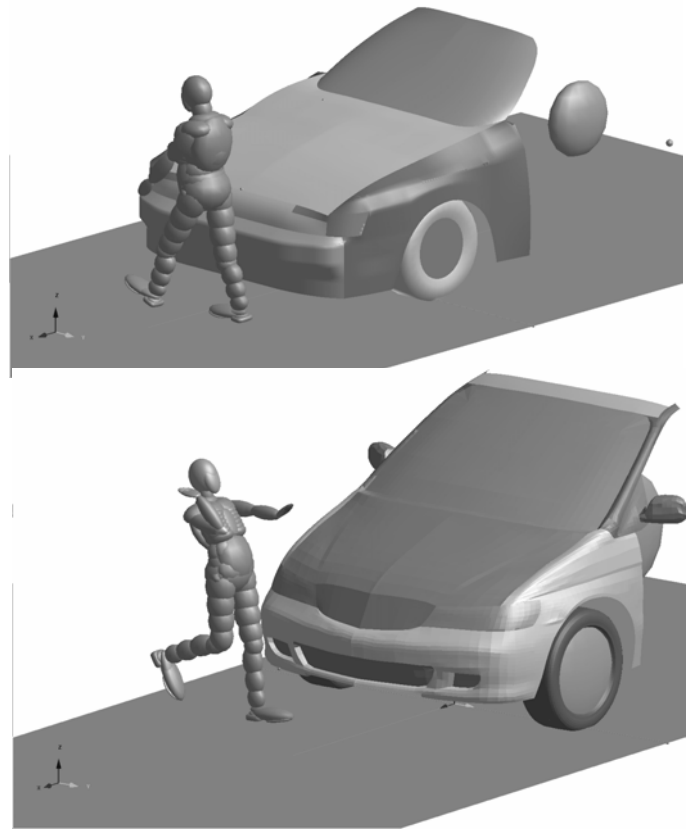


Figure 4 Car-pedestrian impact simulation models using MADYMO (Upper: Case1, Lower: Case2)

The relative distance between contact locations used in the objective function was determined at the onset of the contact force between the corresponding contact locations on the pedestrian and vehicle.

For Case 1, vehicle velocity, pedestrian velocity and pedestrian position (the position of the pedestrian relative to the vehicle in the vehicle lateral direction) were used as the design variables for optimization. The vehicle velocity range was defined as the estimated vehicle velocity $\pm 50\%$, and the pedestrian velocity range was set between 1.0 and 3.0 m/s, with reference to indications in the accident data that the pedestrian was running. The range of the pedestrian position was set between -0.1 and 0.2 m (relative to the center of the vehicle; positive direction pointing left side of the vehicle) based on the location of the dent due to the contact with the pelvis of the pedestrian identified on the hood of the vehicle. In terms of the pedestrian posture, Mizuno [7] discretized the pedestrian stances in a gait cycle as shown in Figure 5. This could have been used as one of the design variables in the optimization calculation.

However, since posture information was available from the accident data for Case 1, an appropriate gait stance was chosen from the discretized postures rather than defining the postures as one of the design variables. S-2 stance in Figure 5 was chosen for Case 1 based on the statement in the accident data that the pedestrian left lower limb was in forward position and the pedestrian rotated about the vertical axis during the collision with the vehicle in such a way that the pedestrian face hit the hood of the vehicle. The sum of the pelvis relative distance and head relative distance was defined as the objective function because the pelvis and head contact locations were identified in the accident data.

In terms of Case 2, the only identified contact locations were on the pedestrian lower limb and the lower bumper (airdam) of the vehicle. Thus, the lower limb relative distance alone was defined as the objective function. The accident data states that the pedestrian head did not contact the vehicle. Based on this statement, an additional constraint was introduced that the maximum contact force between the head and vehicle equals zero. The pedestrian posture shown in Figure 6 was created for Case 2 based on the statement in the accident data that the upper body of the pedestrian was moving backward when the bumper struck his left lower limb, and his right foot was in the air. The vehicle velocity range was defined as the estimated vehicle velocity $\pm 50\%$, and the pedestrian velocity range was set between -2.0 and 0.0 m/s, with reference to indications in the accident data that the pedestrian was attempting to avoid collision with the vehicle with his upper body moving backward. The range of the pedestrian position was set between -0.9 and -0.7 m to cover between the contact location on the lower bumper and the lateral end of the vehicle.



Figure 5 Discretized gait stances



Figure 6 Posture used for Case 2 crash reconstruction

Injury prediction using FE model

FE model set-up

The identified impact conditions from the optimization calculation (vehicle velocity, pedestrian velocity and pedestrian position) were used to set up the FE model, and car-pedestrian impact simulations were run using PAM-CRASH. An upper body model consisting of rigid segments linked by joint elements was integrated into the lower limb FE model developed by the authors based on MRI scans from a six-year-old child [2] to create a baseline pedestrian model. The joint characteristics of the upper body model were determined by scaling those from an adult pedestrian model developed by Takahashi et al. [8][9] using the scaling method described by Langhaar [10]. The mass and moment of inertia of each upper body segment were scaled from those specified in a commercially available rigid-body human model for an average adult (H-Dummy [11]) based on the distance between the joints that connect the segment to the subsequent segments. The baseline model was then scaled to the anthropometry of the case pedestrian based on the anthropometry information of the case pedestrian from the accident data. The stance of the model was modified to represent the stance used in the impact condition optimization. Since the femur and tibia were connected using ligament models rather than using a joint element for the knee joint, the shape of the knee ligament models in these stances was obtained by running dynamic knee joint bending simulations. This procedure is shown in Figure 7. The vehicle model employed in the MADYMO optimization was simply converted to a PAM-CRASH model.

To model bone fractures, material Type 36 and Type 143 (Elastic/Stiffening-Plastic with Failure for Solid and Shell Elements) with Element elimination option in PAM-CRASH were adopted. This option gradually decreases the stiffness of the element down to zero over 100 time steps of calculation once it is activated when the equivalent plastic strain reaches the criterion specified for each material.

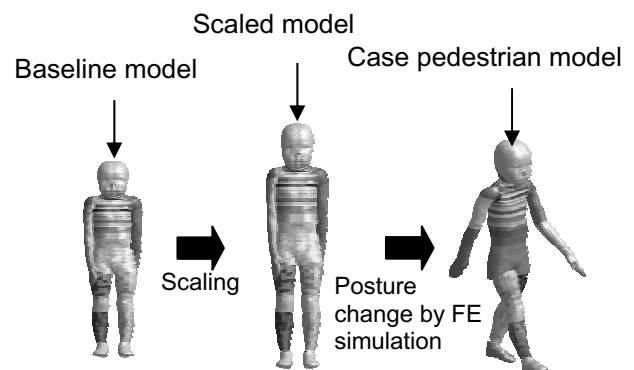


Figure 7 Creation of child FE model representing case pedestrian

Bone material property

Due to limited data available for children, it was necessary to estimate bone material parameters for a child to simulate child lower limb fractures using a FE model. For femur and tibia, the authors estimated the average bone material parameters for human children as a function of age from published dog test results [12] and correlation between human and dog ages [13], while the relationship between material property and age for adult was determined by Takahashi et al. [8]. In general, there are significant individual differences in terms of the material property of human bodies. When attempting to predict injuries by reconstructing an accident, it is difficult to determine in advance where the material parameters of the case pedestrian should be positioned within these individual variations. It was therefore necessary to estimate a range of individual variations in material properties for a child to see if observed injuries could be predicted within the upper and lower limits of these variations. The ratio of average material properties and their standard deviations determined for adults by Takahashi et al. [8] was applied to the material properties of children. In addition, the estimated average material properties for children were scaled in such a way that material parameters are consistent with those determined by Takahashi et al. [8] at age 25 to compensate for the difference between dog and human material properties. Figure 8 shows the Young's modulus for femur as a function of age for example. Similarly, such functions were also determined for the yield stress, ultimate stress and ultimate strain. These functions yield the average and upper and lower limits of the bone stress-strain curve for each age as schematically shown in Figure 9. The boxes around the yield and failure points represent the range of individual variations in stress and strain at these points. In the current research, the upper and lower limits of the individual variation were determined by calculating 1.645 SD of each material parameter variation that covers 90% of population when the normal distribution of the material parameter was assumed. If the injuries sustained in real accidents could be reproduced by any of the stress-strain curves within this range, the hypothesized range of bone material properties could be judged as valid. The confidence level would be higher as the number of cases reproduced increases. Since there are countless stress-strain curves within the established range, model validation would require an enormous number of simulations. When validating an FE child model against real accidents, only the occurrence or non-occurrence of injury (in this study lower limb bone fracture) is compared because loading time histories to the pedestrian are not available from the accident data. In the present research, the probability of bone fracture was assumed to be determined by the amount of energy that could be absorbed by the bone until failure. Based on this assumption, the stress-strain curves at the maximum and minimum absorbed energy were defined as the upper and lower limits of the range of bone material property variations as shown in Figure 10. The properties between these limits were proportionally interpolated along the diagonal lines that connect points representing maximum

and minimum absorbed energy for yield and failure points. For notation purposes, the stress-strain curves determined in this way were represented by the parameter K defined in Equation (2) that corresponds to the deviation of yield and failure points from the average curve along the diagonal lines. The parameter K is -1 and +1 for the minimum and maximum bone energy absorption, respectively (Fig.10). Since the change in the parameter K would vary both the stress and strain simultaneously, a parametric study for bone fracture prediction by changing K between -1 and +1 would evaluate the entire region of the individual variations at the yield and failure points represented by the boxes shown in Figure 9 given the assumption that the probability of bone fractures is determined by the bone energy absorption.

$$\begin{aligned}
 K &= \frac{(\sigma_u - \sigma_{u_ave})}{(\sigma_{u_max} - \sigma_{u_ave})} \\
 &= \frac{(\epsilon_u - \epsilon_{u_ave})}{(\epsilon_{u_max} - \epsilon_{u_ave})} \\
 &= \frac{(\sigma_y - \sigma_{y_ave})}{(\sigma_{y_max} - \sigma_{y_ave})} \\
 &= \frac{(\epsilon_y - \epsilon_{y_ave})}{(\epsilon_{y_max} - \epsilon_{y_ave})} \dots\dots\dots (2)
 \end{aligned}$$

where,

σ_u, ϵ_u : Ultimate stress, strain
 $\sigma_{u_ave}, \epsilon_{u_ave}$: Average ultimate stress, strain
 $\sigma_{u_max}, \epsilon_{u_max}$: Maximum ultimate stress, strain
 σ_y, ϵ_y : Yielding stress, strain
 $\sigma_{y_ave}, \epsilon_{y_ave}$: Average yielding stress, strain
 $\sigma_{y_max}, \epsilon_{y_max}$: Maximum yielding stress, strain