Computer Model of a Mechanical Surrogate for Pedestrian Lower Extremity on Bumper Safety Evaluation

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Abstract

This study proposes engineering countermeasures for lowering the possibility of pedestrian lower-extremity injuries. Pedestrian safety tests are commonly performed using anthropometric test device (ATD) subsystems. The lower legform is used to assess lower-extremity injuries during impact. In this study, a computer model of a mechanical surrogate for a pedestrian lower extremity is developed based on the finite element (FE) method. Results from the FE model indicate that the surrogate passes all certification tests proposed by the European Enhanced Vehicle-Safety Committee (EEVC), meaning that the model is biofidelic and suitable for the assessment of lower-extremity injuries due to car-pedestrian impact. In addition, a bumper system is chosen as an example to demonstrate safety evaluation by this pedestrian lower extremity surrogate. Based on the demonstration results, the safety of the bumper system is investigated.

Keywords: Lower-extremity surrogate, Pedestrian safety, FE method, Biofidelity, Bumper system

1. Introduction

Pedestrian fatalities account for a large proportion of total traffic fatalities. Pedestrian accident statistics show that sedans are the most common type of car to cause pedestrian injuries (75%) and that most of pedestrian injuries are attributed to side impact with the front of a car (54%) [1]. Kalliske and Friesen collected 783 accident cases between pedestrians and automobiles and analyzed pedestrian injuries for ev-...
2. Methods

2.1 Lower-extremity surrogate

A finite element (FE) model of a mechanical surrogate for a pedestrian lower extremity is constructed. LS-DYNA software is utilized to develop the model. The model includes the femur, the tibia, a flesh layer, a skin layer, and a knee structure, which are defined by the European Enhanced Vehicle-Safety Committee/Working Group 17 (EEVC/WG 17) [8].

(1) Femur and tibia: The femur and tibia are solid elements. Because they are much stiffer than the outer soft layers, they are modeled as rigid bodies to reduce computation time. The flesh layer is a solid element whose material is Confor foam type CF-45. The mechanical properties of the foam were adopted from the studies of Yao et al. [10] and Cappetti et al. [11]. The skin layer is a solid element whose material is neoprene foam. The properties of the skin were those suggested by Konosu [12]. The dimensions, mass, center of gravity, and moment of inertia of every part of the model comply with the EEVC/WG17 definitions. The properties of the proposed structure and the corresponding EEVC requirements are listed in Table 1.

Table 1. Mechanical properties of lower legform model and EEVC requirements.

<table>
<thead>
<tr>
<th>Section</th>
<th>Length (mm)</th>
<th>Diameter (mm)</th>
<th>Mass (kg)</th>
<th>Inertia at C.G. (kgm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Femur section</td>
<td>432</td>
<td>70 ± 1</td>
<td>8.6 ± 0.1</td>
<td>0.217 ± 0.010</td>
</tr>
<tr>
<td>Tibia section</td>
<td>494</td>
<td>70 ± 1</td>
<td>8.6 ± 0.1</td>
<td>0.121 ± 0.010</td>
</tr>
<tr>
<td></td>
<td>494</td>
<td>494</td>
<td>4.8 ± 0.1</td>
<td>0.121 ± 0.010</td>
</tr>
</tbody>
</table>

* C.G. means Center of Gravity

(2) Knee structure: The FE model of the knee structure is composed of a pin-in-slot joint, two linear springs, and one torsional spring. The proposed knee structure is newly developed and different from those used by other researchers [9,13]. According to the EEVC/WG17 definitions, the knee structure should be able to move transversely and rotate, and include deformable elements to restrict knee movement. A pin-in-slot joint meets the requirement of transverse and rotational movement, and the linear and torsional springs are the deformable elements. In this joint, the ranges of motion are 22° for the bending angle and 8 mm for the shearing displacement (these are validated in a later section). The torsional spring measures the knee bending angle and bending moment and provides bending stiffness that represents the tensile properties of the medial collateral ligament and the lateral collateral ligament. The linear springs measure the knee shearing displacement and shear force, and provide shearing stiffness that represents the tensile properties of the anterior cruciate ligament and the posterior cruciate ligament. Fig. 1 shows the knee structure in the model. The spring is constructed from discrete elements, which have 3 translational and rotational degrees of freedom, respectively. The tensile property of the linear springs is assigned using a linear force-displacement curve, and the tensile property of the torsional spring is assigned using a nonlinear torque-angular displacement curve. The pin-in-slot joint and the springs in this knee structure can be easily utilized for designing a mechanical surrogate. In addition, the adjustable stiffness of the torsional and linear springs can resolve the existed anisotropic problem in the actual human knee.

![Figure 1. Knee structure in the proposed model.](image)

2.2 Validation method

Certification tests of the lower-legform impactor are proposed by EEVC/WG17 [8]; they involve simulating pedestrian legs being side-impacted by cars. These certification validations include static and dynamic tests. The static tests consist of bending and shearing validations. A legform impactor without a foam covering and skin is used in these tests. The test setup for both bending and shearing tests involves mounting the tibia to a fixed surface. The difference between these two tests is the location of applied force. For the bending test, the force is applied 2 m from the knee centerline, and for the shearing test, it is applied 50 mm from the knee centerline. The validations were conducted under the conditions required by EEVC/WG17.

In the dynamic test, a legform impactor with a foam covering and skin is suspended horizontally by ropes and impacted by a linearly guided 9.0 ± 0.05 kg impactor at a velocity of 7.5 ± 0.1 m/s. The impactor is positioned on the tibia centerline, 50 mm from the center of the knee. The model for the dynamic simulation is shown in Fig. 2.

![Figure 2. Top view of dynamic certification test setup.](image)

The EEVC assigned corridors for both the static and dynamic certification tests. Simulation results that are within the EEVC/WG17 corridors indicate that the impactor is biofidelic.

3. Results and discussion

The results of the certification tests are described below.

(1) Static legform impactor bending and shearing certification tests: Two requirements must be satisfied in the static bending
certification test: (a) the applied force versus knee bending angle response must be within the limits and (b) the energy required to generate 15.0° of bending shall be 100 ± 7 J. Fig. 3 shows the test results, which fall within the assigned corridor. The energy required to generate 15.0° of bending is 103.29 J, which meets the requirement. One criterion must be satisfied in the static shearing certification test: the applied force versus knee shearing displacement response must be within the limits. The test results shown in Fig. 4 are within the specified range.

![Figure 3. Results of static knee bending certification test.](image)

![Figure 4. Results of static knee shearing certification test.](image)

(2) Dynamic legform impactor certification test: Three requirements must be met in the dynamic certification test: (a) the maximum upper tibia acceleration must be between 120g and 250g, (b) the maximum bending angle must be between 6.2° and 8.2°, and (c) the maximum shearing displacement must be between 3.5 mm and 6.0 mm. The test results show a maximum upper tibia acceleration of 174.41g, a maximum bending angle of 7.25°, and a maximum shearing displacement of 4.90 mm, all of which meet the requirements. A summary of results from the dynamic certification tests is shown in Table 2.

<table>
<thead>
<tr>
<th>Table 2. Dynamic certification results for lower leg surrogate.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper tibia acceleration (g)</td>
</tr>
<tr>
<td>EEVC requirements</td>
</tr>
<tr>
<td>Simulation results</td>
</tr>
</tbody>
</table>

Results from the FE model constructed in this study pass all certification tests proposed by EEVC/WG17, indicating that the model is biofidelic.

The automobile model in this study is adapted from the open database of the National Crash Analysis Center (NCAC) of the Federal Highway Administration/National Highway Traffic Safety Administration (FHWA/NHTSA) [14], and the vehicle model is a Dodge Neon. Since the computation time increases with increasing number of meshes, the automobile model was simplified to only the bumper system.

As shown in Fig. 5, the bumper system consists of three parts: (1) Fascia (outer surface); (2) Energy absorber (middle part); (3) Bumper beam (inner space). The energy absorber normally is made of a foam material and simulated by an 8-node solid element, and the fascia and bumper beam are simulated by Belytschko-Tsay shell elements. This type of shell element has high efficiency. There are 3 degrees of freedom in translation and rotation, respectively, at each node. The material parameters, namely density (ρ), Young’s modulus (E), yielding stress (σy), Poisson’s ratio (PR), damping coefficient (DAMP), and number of elements, of each part of the bumper system are shown in Table 3. The masses of the parts of the bumper system are 4.538 kg (fascia), 0.631 kg (energy absorber), and 6.787 kg (bumper beam), for a total of 11.956 kg.

![Figure 5. Exploded view of bumper system.](image)

In the bumper safety evaluation, the lower legform impactor was propelled at a velocity of 40 km/h parallel to the longitudinal axis of the car. The boundary conditions were set according to the regulation defined by EEVC/WG17 [8]. Then, the acceleration of the upper tibia, the bending angle, and the shearing displacement of the impactor in the lower-legform-to-bumper test.

There are three acceptance levels in the experiment according to the EEVC regulation [8]: (1) Maximum lateral tibia acceleration ≤ 150 g. (2) Maximum lateral knee bending angle ≤ 15°. (3) Maximum lateral knee shearing displacement ≤ 6 mm.
If the responses exceed these levels, injuries may occur to the lower leg and knee. At the maximum acceleration, the pedestrian might sustain a tibia fracture. The two most common injury mechanisms at the knee due to bending are ligament avulsion failure and diaphysis/metaphysis failure. The 6-mm shear displacement is based on a 4-kN shear and a 15° bending angle according to EEVC/WG17 [8]. Therefore, similar injury mechanisms due to the bending angle might occur when the shear displacement exceeds the acceptable level. In addition, a large shear displacement might also lead to epiphysis failure.

According to the regulations, a minimum of three impact points must be used for evaluation. The distance between impact points should be greater than 132 mm. In this study, the bumper was divided into three equal parts, from the middle of the bumper to bumper corners. The middle of the bumper, and 1/3 and 2/3 the distance from the centerline to the two corners, respectively, were used as impact points, as shown in Fig. 6. There are two contact settings in the model: (1) the distance between the outer foam of the legform impactor and the fascia, and (2) the distances among the parts of the bumper system.

Simulations of an impact are shown in Fig. 7, and the simulation results of each impact point are shown in Table 4. The values of shearing displacement of each impact point are all below the limits; however, the values of bending angle and acceleration exceed the limits by about 6% and 60%, respectively. This is due to the impact between bumper and impactor being at knee level, which resulted in small values of shearing displacement. The energy absorber was thin (about 50 mm), making the crush distance short and acceleration high.

Based on the EEVC criteria, the acceleration of upper tibia must be below 150g, so the crush distance should be 60-80 mm [15]. Since the space of bumper system is limited, increasing energy absorbing efficiency is very important [16]. In addition, the stiffness of the bumper beam also affects pedestrian injuries; that is, a lower stiffness leads to less severe injury.

**Table 3. Properties of each part of the bumper system.**

<table>
<thead>
<tr>
<th>Part</th>
<th>Thickness (mm)</th>
<th>LS-DYNA material type</th>
<th>Material property</th>
<th>No. of elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fascia</td>
<td>3.5</td>
<td>Piecewise linear plasticity</td>
<td>ρ=1.2×10³ kg/m³  E=2800 MPa  σy=45 MPa  PR=0.3</td>
<td>3670</td>
</tr>
<tr>
<td>Beam</td>
<td>1.956</td>
<td>Piecewise linear plasticity</td>
<td>ρ=7.89×10³ kg/m³  E=2.1×10⁶ MPa  σy=570 MPa  PR=0.3</td>
<td>2156</td>
</tr>
<tr>
<td>Energy absorber</td>
<td>49</td>
<td>Crushable foam</td>
<td>ρ=9.131×10³ kg/m³  E=30.6 MPa  DAMP=0.1  PR=0.3</td>
<td>11232</td>
</tr>
</tbody>
</table>

**Table 4. Results of simulation for bumper safety evaluation.**

<table>
<thead>
<tr>
<th></th>
<th>Acceleration (g)</th>
<th>Bending Angle (degrees)</th>
<th>Shearing Displacement (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>EEVC criteria</td>
<td>150</td>
<td>15.00</td>
<td>6.000</td>
</tr>
<tr>
<td>Center</td>
<td>233.12</td>
<td>15.90</td>
<td>1.704</td>
</tr>
<tr>
<td>Right</td>
<td>254.15</td>
<td>16.00</td>
<td>2.485</td>
</tr>
<tr>
<td>Left</td>
<td>245.00</td>
<td>16.09</td>
<td>2.736</td>
</tr>
</tbody>
</table>

4. Conclusion

FE models of a mechanical surrogate for a pedestrian lower extremity and an automotive bumper system were constructed. Static and dynamic certification test results indicate that the lower extremity model is biofidelic and thus suitable for simulating impact tests with the bumper system. The results from the impact simulation were used to evaluate the safety of the bumper system. Impact analysis shows that a thin energy absorber made of a stiff material can easily cause serious injuries to pedestrians, making it fail to meet EEVC requirements. In order to meet the requirements and prevent lower leg injuries, the developed computer model can aid the design of a safer bumper system. Since the surrogate is based on a computer model, tests can be conveniently repeated, reducing the cost of the design. Although the computer
simulation provides preliminary results for pedestrian lower leg injuries due to bumper impact, actual tests still are needed. Therefore, a physical limp surrogate based on the developed computer model will be implemented in a future study. In addition, new biomechanical data [17,18] has recently become available. It suggests that the biofidelity of the current legform might need to be improved. For example, a flexible pedestrian legform impactor (Flex-PLI) based on the new data was introduced by Konosu [12]. It also will be the future considerations for modifying this lower leg surrogate.

Acknowledgement

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References