INJURY THRESHOLDS AND A MEASUREMENT TECHNIQUE FOR THE THIGH AND LEG OF A PEDESTRIAN DUMMY

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ABSTRACT

The new design of a modified lower limb of the Polar-II pedestrian dummy for enhanced biofidelity requires establishment of an alternative instrumentation to load cells. The objective of this study was to determine injury thresholds for the thigh and leg of the modified dummy, and to establish an alternative measurement technique. FE pedestrian impact simulations were run using full-body dummy and human models to determine injury thresholds for the dummy. Quasi-static mid-shaft 3-point bending tests of the modified femur and tibia were performed to compare bending moments calculated from strains with those from support forces to validate the measurement technique.

Keywords: PEDESTRIANS, DUMMIES, FINITE ELEMENT METHOD, INJURY CRITERIA, MEASUREMENTS

ACCIDENT STATISTICS from the OECD database (International Road Traffic and Accident Database, 2005) show that pedestrian accidents account for approximately 12-35% in European countries, 17% in the US, 47% in Korea, and 35% in Japan of all traffic fatalities. Despite the recent advancement in vehicle safety features for pedestrian protection, these numbers suggest that pedestrian protection is still one of the major issues in vehicle safety. In order to address this issue, a pedestrian dummy called Polar-II has been developed as a research tool to investigate mechanisms of pedestrian accidents at the full-scale level (Akiyama et al., 2001). Since head injuries dominate in pedestrian fatal injuries, the dummy primarily targeted at accurately representing whole-body kinematics to allow investigation of head impact conditions.

Pedestrian accident data from the Pedestrian Crash Data Study (PCDS) database show that leg fractures dominate in AIS 2+ pelvis and lower limb injuries followed by knee injuries for passenger car impacts. In contrast, for pedestrian impacts with Sport Utility Vehicles (SUVs), the contribution of pelvis and thigh fractures are significantly higher than those for passenger car impacts (Kikuchi et al., 2008). In order to enhance biofidelity of the Polar-II dummy so that it can be used for investigating injury mechanisms in impacts with various types of vehicles, a modified version of the femur, tibia and knee joint has been developed and the biofidelity of those components has been evaluated (Bose et al., 2007). For improving biofidelity of the thigh region, new deformable femur shafts made of polyoxymethylene (POM) were incorporated. The same material was used for the tibia shafts. Load cells used for the original version of the Polar-II thigh and leg were eliminated to accommodate longer femur and tibia shafts for enhanced biofidelity. Although the basic design of the modified knee joint remained the same, the vertical dimension was decreased as much as possible for the same purpose. Due to the elimination of the load cells, an alternative measurement technique capable of quantitatively assessing femur and tibia fractures needed to be established. In addition, although good biofidelity was experimentally confirmed at the component level by Bose et al., injury thresholds for the dummy thigh and leg needed to be determined by correlating dummy and human impact responses at the full-scale level.

The goal of this study was to determine injury thresholds for the thigh and leg of the modified dummy, and to establish an alternative measurement technique. Finite Element (FE) models for the femur, tibia and knee joint of the modified Polar-II dummy were developed and validated against dynamic 3-point and 4-point bending tests, respectively. The lower limb model was incorporated into the Polar-II upper body model developed and validated by Shin et al. (2006) to create a full-body model. FE impact simulations were run where the dummy model along with a full-body FE human model were hit by FE vehicle front
models with various geometries at 40 km/h. Moments in different cross-sections of the femur and tibia were compared between the dummy and human models to determine injury thresholds for the dummy by correlating dummy and human impact responses at the full-scale level.

The femur and tibia of the modified dummy were instrumented with strain gages. Quasi-static mid-shaft 3-point bending tests were performed using the strain-gaged modified femur and tibia. Bending moments calculated from strain measurements were compared with those determined from support forces to validate the measurement technique.

METHODOLOGY

FULL-SCALE FE DUMMY MODEL DEVELOPMENT: For determining dummy injury thresholds for the thigh and leg based on the impact response correlation at the full-scale level, a full-scale FE dummy model was developed using PAM-CRASH. The model development included component validation for the femur and tibia shafts and the knee joint, and incorporation of the component models into an existing FE upper body model.

Component model validation: Bose et al. (2007) performed dynamic 3-point bending tests using the modified version of the femur, tibia and knee joint of the Polar-II dummy. FE models for the modified femur and tibia were created to validate the models against Bose et al. The models were developed using solid elements. Non-linear elastic-plastic material model (Type 36) was used to represent material property of the femur and tibia shafts (POM). Since the study by Bose et al. showed that the material had minimal strain rate dependency, no strain rate effects were incorporated in the models. The material parameters were identified such that the predicted moment-deflection response matches that from the experiment. The attachment rigs used in the experiment to provide simply supported boundary conditions were modeled using solid elements and treated as rigid bodies. Fig. 1 shows the femur and tibia models set up in 3-point bending for mid-shaft loading. The moment-deflection response was compared for mid-shaft and proximal third loading of the tibia, and mid-shaft, proximal third, distal third loading of the femur. The same loading speed of 1.5 m/s as that used in the experiment was applied to the impactor for all loading conditions. For the tibia loading only, the impactor was surrounded by a model representing Confor foam used in the experiment. The span length was set at 334 mm and 404 mm for the tibia and femur, respectively, to replicate the experiment.

The knee joint model consisted of rigid femoral condyles and tibial plateau with external shapes represented by shell elements, deformable meniscus modeled by solid elements, and four knee ligaments modeled by bar elements. The adaptors at both ends of the knee joint and the load cell connected to the adaptor on the tibial side were modeled as rigid bodies. The aluminum pipes at both ends of the knee joint to provide 4-point bending configuration were also modeled as rigid bodies, and rotational and translational degrees of freedom were given to the joint elements that connect the pipes to the space. A fork that loads the pipes vertically was modeled using a rigid body definition. The loading speed of the fork corresponding to 1°/ms of pipe rotation was applied to the fork model. Fig. 2 shows the FE model of the knee joint in 4-point bending configuration. A fixed joint element was set at the center of the load cell model to allow output of load cell force and moment. The output was used to calculate inertially compensated moment at the knee.

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**Fig. 1 – Tibia and Femur Component Model in 3-Point Bending**
joint (Fig. 3) using the following equation:

\[ M_K = M_F + V_F (x_F + x_K) + M \ddot{u}x_K - I \ddot{\theta} \]  

(1)

where

- \( M, I \): Mass and moment of inertia of the proximal knee segment
- \( M_K \): Knee moment
- \( M_F \): Moment from the load cell
- \( V_F \): Shear force from the load cell
- \( u \): Circumferential linear displacement
- \( \theta \): Angular displacement
- \( x_F \): COG to knee center distance
- \( x_K \): COG to load cell center distance

For model validation, the inertially compensated knee moment was compared with the experimental data.

**Full-scale model development**: The femur, tibia and knee joint models were incorporated into an existing FE upper body model developed and validated by Shin et al. (2006) to develop a full-scale dummy model. The upper body model consisted of the following deformable parts: ribs, rib damping materials, neck joint, thoracic joint, lumbar joint, shoulder joint stopper rubber, shoulder pads, thoracic jacket, abdominal flesh, and pelvis flesh. Other components were modeled as rigid bodies and connected to each other by joint elements. The femur, knee and tibia models were assembled and connected to the upper body model at the hip joints using joint elements. Rigid body foot models were also connected to the tibia models at the ankle joints using joint elements. The shape of the thigh and leg skin was taken from the original Polar-II, and solid elements were used to represent the thigh and leg flesh. Fig. 4 shows the full-scale FE model for the Polar-II dummy with modified thigh, knee and leg.

**FULL-SCALE FE IMPACT SIMULATION**: For determining injury thresholds for the thigh and leg of the modified dummy based on the impact response correlation at the full-scale level, full-scale FE impact simulations were run using the whole-body dummy model and a whole-body FE human model developed by the authors in their previous studies (Kikuchi et al., 2008). The human FE model contains deformable pelvis and lower limbs extensively validated at the component level (Takahashi et al., 2003, Kikuchi et al., 2006). The upper body was represented by rigid bodies connected by joint elements. The neck and the

![Fig. 2 – Knee Joint Component Model in 4-Point Bending](image)

![Fig. 3 – Knee Joint Moment Calculation](image)
lumbar were divided into seven and five segments to simulate the same degrees of freedom as those in a human body. The full-scale human FE model was validated at the full-scale level in terms of upper body trajectories and pelvis and lower limb injury prediction for small sedan and SUV impacts against the full-scale pedestrian sled tests using Post Mortem Human Subjects (PMHS) published by Kerrigan et al. (2005a, 2005b, 2008).

The dummy and human FE models were subjected to lateral impact from three different FE vehicle models representing a small sedan, mini-van and SUV. The pedestrian models (dummy and human) were set up in a standing position with the lower limbs rotated 10° about the lateral-medial axis (left limb forward) to simulate the orientation used in the experiment by Kerrigan et al. (2005a) against which both models were validated (Shin et al., 2006, Kikuchi et al., 2008). The models were hit laterally from their right by the vehicle models at 40 km/h. Section force output was set up at the proximal third, mid-shaft and distal third sections of the right (struck side) femur and tibia to compare section moments between the dummy and human models and determine injury thresholds for the dummy. Fig. 5 shows the model set-up for the dummy and human models with three different vehicle models.

QUASI-STATIC 3-POINT BENDING TESTS: The modified version of the Polar-II tibia and femur was instrumented with strain gages and subjected to quasi-static 3-point bending tests. Four strain gages (KYOWA KFEL-5-120) were affixed to the surface of one cross-section to measure strains along the longitudinal axis of the tibia and femur. Three different cross-sections were chosen for the measurement – proximal third, mid-shaft and distal third, resulting in 12 strain gages for each specimen. Two test specimens were prepared for each of the tibia and femur, and two tests were run for each specimen. Test rigs providing a simply supported boundary condition similar to those used by Bose et al. (2007) for dynamic 3-point bending tests of the modified tibia and femur were attached to both ends of the test specimen. The specimen with the test rigs were placed on two load cells that measure vertical support forces. The span lengths used for the component model validation of the tibia and femur were also used in these tests (334 mm for the tibia, 404 mm for the femur). Each specimen was loaded at mid-shaft at a loading speed of 10 mm/min. The radius of the tip of the indenter was 10 mm. Fig. 6 shows the 3-point bending test set-up for the tibia and femur.

Bending moment at each cross-section was calculated from the strain data using the elastic beam theory. Fig. 7 shows a schematic of the cross-section of a specimen to which strain gages were affixed. The axis definition for bending moment calculation is also illustrated in the figure. Under the assumptions of the beam theory, the longitudinal strain at arbitrary locations in the cross-section can be expressed as:

![Fig. 4 – Polar-II FE Model](image)

![Fig. 5 – Full-scale Dummy and Human FE Model Set-up in Vehicle Impacts](image)
where

\[ \varepsilon(x, y) = \varepsilon_{\text{axial}} + A_x x + A_y y \quad (2) \]

where

- \( \varepsilon_{\text{axial}} \) : Longitudinal strain produced by axial loading
- \( A_x, A_y \) : Constants defining the strain field
- \( \varepsilon, \varepsilon_{\text{axial}}, A_x, A_y \) are functions of time. In order to determine strain field at the cross-section, three unknown parameters need to be determined. Thus, the fourth strain gage in the cross-section is redundant and was used to check validity of the data processing. The three unknown parameters can be determined by solving the following equation at each time step:

\[
\begin{pmatrix}
\varepsilon_1 \\
\varepsilon_2 \\
\varepsilon_3
\end{pmatrix} =
\begin{pmatrix}
1 & x_1 & y_1 \\
1 & x_2 & y_2 \\
1 & x_3 & y_3
\end{pmatrix}
\begin{pmatrix}
\varepsilon_{\text{axial}} \\
A_x \\
A_y
\end{pmatrix}
\quad (3)
\]

where the suffix denotes strain gage number, and \( \varepsilon_1 \) through \( \varepsilon_3 \) denote longitudinal strains measured by the corresponding strain gage. Once the three unknown parameters are determined by solving equation (3), the bending moment \( M \) can be given as:

\[ M = -EA_y I_x \quad (4) \]

where \( E \) is the Young’s modulus and \( I_x \) is the geometric moment of inertia of the cross-section. Exactly the same Young’s modulus value for the POM material as that identified from the component FE model validation was used for moment calculation. The calculated bending moment was compared with the bending moment obtained from the support loads using the following equations:

\[ M_{\text{mid}} = \frac{1}{4} FL \quad (5) \]

\[ M_{\text{third}} = \frac{1}{6} FL \quad (6) \]

where \( M_{\text{mid}} \) and \( M_{\text{third}} \) are mid-shaft and proximal or distal third moment, respectively, \( F \) is the sum of the support forces from the two support load cells, and \( L \) is the span length.
RESULTS

COMPONENT MODEL VALIDATION: The results of the tibia and femur dynamic 3-point bending validation are shown in Fig. 8 and Fig. 9, respectively. The predicted moment-deflection responses matched well with the experimental results from Bose et al. (2007). The comparison of the inertially compensated knee moment along with the comparison of knee ligament forces for the MCL, ACL and PCL are presented in Fig. 10. Except for the initial spike seen in the moment-angle response probably due to hard contact between the fork and the pipes, the model prediction matched well with the experimental results from Bose et al. The results of the model validation were quantitatively assessed using the method proposed by Rhule et al. (2002) for dummy biofidelity evaluation. Fig. 11 shows a schematic for illustrating the evaluation procedure applied to the model validation results. The method uses the ratio R of Dummy Cumulative Variance (DCV) to Cadaver Cumulative Variance (CCV) as the index for quantitatively evaluating the dummy biofidelity. CCV, DCV and R are given by the following equations:

\[
CCV = \sum_{n=0}^{N} |CV(n)|
\]

\[
DCV = \sum_{n=0}^{N} |DV(n)|
\]

\[
R = DCV / CCV
\]

In the current case, deviations of the dummy component FE simulation results from the dummy component test results were used for DV. For CV, one standard deviation from multiple cadaver tests was used in the original procedure. In this case, 10% of actual dummy response variations were assumed and used in the calculation of CCV. The results of R number calculation for all the validation cases are summarized in Table 1. R<2 is considered ‘Good’ biofidelity for dummy evaluation, and all of the FE predictions fell within this criterion.

FULL-SCALE FE IMPACT SIMULATION: Fig. 12 through Fig. 14 show time sequence of the pedestrian kinematics at 10 ms interval for both human and dummy FE models for impacts with the small sedan, mini-van and SUV, respectively. The kinematics was shown in posterior view for easier understanding of the motion of the struck-side limb shown in red. Due to relatively flat geometry of the front end of the mini-van, the impact load from the bumper was widely distributed to the lower limb, resulting in smaller knee bending. For the SUV, due to the lack of structure below the bumper, the knee joint

<table>
<thead>
<tr>
<th>Femur 3-point bending</th>
<th>Tibia 3-point bending</th>
<th>Knee 4-point bending</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prox loading</td>
<td>Mid loading</td>
<td>Dist loading</td>
</tr>
<tr>
<td>R</td>
<td>1.030</td>
<td>0.370</td>
</tr>
</tbody>
</table>
Fig. 8 – Moment-Deflection Comparison for Tibia

Fig. 9 – Moment-Deflection Comparison for Femur

Fig. 10 – Comparison of Knee Moment and Ligament Forces
was subjected to significant bending. The low hood with the passenger car resulted in large thigh rotation, causing large knee joint bending. These characteristics of the pedestrian lower limb kinematics were essentially the same between the human and dummy models. The difference in the lower limb kinematics due to difference in vehicle front end geometry resulted in different bending moment time histories. Fig. 15 compares moments at distal third, mid-shaft and proximal third cross-sections of the struck-side femur between the human and dummy models. The same comparison was made for the tibia in Fig. 16. The comparison of the first peak bending moment at each cross-section along with the ratio of the peak bending

Fig. 12 – Kinematics for Small Sedan Impact

Fig. 13 – Kinematics for Mini-van Impact

Fig. 14 – Kinematics for SUV Impact
moment between the human and dummy were summarized in Table 2. The average ratio of the peak moment over three different vehicles was also calculated for each cross-section of each bone. The bending moment thresholds for 50% probability of fracture of the tibia and femur were taken from Ivarsson et al. (2004). Since thresholds were available only for the mid-shaft of each bone, the MRI scans of a human volunteer used to develop the lower limb FE model (Takahashi et al., 2003) were reviewed to determine the ratio of the geometric moment of inertia for the proximal and distal third cross-sections relative to the mid-section. The ratio was used to scale the threshold moments for the mid-shaft to other cross-sections. The results of the fracture threshold calculations were summarized in Table 3.

QUASI-STATIC 3-POINT BENDING TESTS: The bending moment calculation from strain measurement requires three strain gages per cross-section. Since four strain gages were affixed to one

![Graphs showing moment comparison for different sections of bones](image)

Table 2. Peak Moment Comparison (moment in Nm)

<table>
<thead>
<tr>
<th></th>
<th>Small sedan</th>
<th>Mini-van</th>
<th>SUV</th>
<th>Ave.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prox</td>
<td>Dummy</td>
<td>83.6</td>
<td>30.2</td>
<td>61.5</td>
</tr>
<tr>
<td></td>
<td>Human</td>
<td>48.3</td>
<td>34.7</td>
<td>71.4</td>
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<td></td>
<td>Correlation factor</td>
<td>1.73</td>
<td>0.87</td>
<td>0.86</td>
</tr>
<tr>
<td>Mid</td>
<td>Dummy</td>
<td>123.0</td>
<td>30.2</td>
<td>61.5</td>
</tr>
<tr>
<td></td>
<td>Human</td>
<td>68.6</td>
<td>34.7</td>
<td>71.4</td>
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<tr>
<td></td>
<td>Correlation factor</td>
<td>1.79</td>
<td>0.87</td>
<td>0.86</td>
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<td>Dist</td>
<td>Dummy</td>
<td>137.0</td>
<td>26.4</td>
<td>36.2</td>
</tr>
<tr>
<td></td>
<td>Human</td>
<td>63.4</td>
<td>21.0</td>
<td>37.8</td>
</tr>
<tr>
<td></td>
<td>Correlation factor</td>
<td>2.16</td>
<td>1.26</td>
<td>0.96</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
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<th>Small sedan</th>
<th>Mini-van</th>
<th>SUV</th>
<th>Ave.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prox</td>
<td>Dummy</td>
<td>63.1</td>
<td>53.7</td>
<td>89.0</td>
</tr>
<tr>
<td></td>
<td>Human</td>
<td>61.4</td>
<td>22.1</td>
<td>79.0</td>
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<td>Correlation factor</td>
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<td>2.43</td>
<td>1.13</td>
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<td>70.4</td>
<td>37.8</td>
<td>88.8</td>
</tr>
<tr>
<td></td>
<td>Human</td>
<td>32.1</td>
<td>22.5</td>
<td>83.0</td>
</tr>
<tr>
<td></td>
<td>Correlation factor</td>
<td>2.19</td>
<td>1.68</td>
<td>1.07</td>
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<tr>
<td>Dist</td>
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<td>76.6</td>
<td>29.2</td>
<td>75.7</td>
</tr>
<tr>
<td></td>
<td>Human</td>
<td>30.2</td>
<td>36.2</td>
<td>85.1</td>
</tr>
<tr>
<td></td>
<td>Correlation factor</td>
<td>2.54</td>
<td>0.81</td>
<td>0.89</td>
</tr>
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</table>
cross-section in the bending tests, bending moment in the cross-section can be obtained from four different combinations of three out of four strain gages. Fig. 17 shows an example of the comparison of calculated bending moment of the mid-shaft tibia from these four different combinations of the strain gages. The numbers in the legend of the graph represent combinations of three out of four strain gages. The four curves are almost exactly coincident, suggesting validity of the procedure used in the current study. Fig. 18 and Fig. 19 compare bending moments calculated from strains with those obtained from support forces for the tibia and femur, respectively. Although two tests were performed for each specimen, all the results were very similar and the data from only one representative test are presented for each bone. Only one combination of the strain gages was chosen as well. The bending moment calculated from strains matched the bending moment obtained from the load cells quite well for roughly initial half of the moment time histories in all

| Table 3. Tentative Moment Threshold (moment in Nm) |
|---------------------------------|-----------------|-----------------|-----------------|
| Femur.                          | 1 ratio         | Human threshold (Nm) | Correlation factor | Dummy threshold (Nm) |
| Proximal                        | 0.975           | 436               | 1.15             | 501                |
| Mid-shaft                       | 1.000           | 447               | 1.17             | 523                |
| Distal                          | 0.946           | 423               | 1.46             | 617                |
| Tibia                           |                 |                   |                  |                    |
| Proximal                        | 1.211           | 378               | 1.53             | 578                |
| Mid-shaft                       | 1.000           | 312               | 1.65             | 515                |
| Distal                          | 0.820           | 256               | 1.41             | 361                |

![Fig. 17 – Section Moment from Different Combinations of Strain Gages (see text)](image1)

![Fig. 18 – Moment Calculated from Strain and Support Force for Tibia 3-point Bending](image2)

![Fig. 19 – Moment Calculated from Strain and Support Force for Femur 3-point Bending](image3)
cases. The difference between the moment from strains and load cells was larger for larger deflection.

DISCUSSION

In the past study by Bose et al. (2007), the femur, tibia and knee joint of the modified dummy were validated against human responses, suggesting good biofidelity at a component level. However, some differences were identified in a comparison of the responses of the femur and tibia between the full-scale FE dummy and human models subjected to the same impact conditions. For this reason, only the first peak of the femur and tibia moment time histories was compared to determine tentative fracture tolerance values for the dummy. The difference was most significant with the femur moment time histories for the SUV impact. Since the pelvis was directly loaded by the hood leading edge of the SUV, and the pelvis gives a boundary condition to the proximal end of the femur, a possible explanation on this difference is the lack of biofidelity of the current rigid pelvis of the dummy. In addition, pedestrian accident statistics show that the most predominant AIS 2+ pelvis and lower limb injuries in SUV impacts are pelvis fractures (Kikuchi et al., 2008). In order to cover important injuries for a variety of vehicles with the pedestrian dummy by improving biofidelity of femur responses of the dummy at the full-scale level particularly with SUV impacts, the current rigid pelvis of the dummy needs to be modified to a more compliant structure. A future study on further modification of the pedestrian dummy should focus on the biofidelity enhancement of the pelvic region. In addition, full-scale kinematic and kinetic responses with the modified pelvis and lower limb components need to be validated.

Determination of the dummy injury thresholds based on the response correlations from the full-scale FE simulations using the dummy and human models assumes that the FE human model predictions of the forces and moments in the femur and tibia are valid not only in pure lateral direction but in other directions. Since in past studies the lower limb component responses of the FE human model have been validated primarily in pure lateral direction and the upper body kinematics and pelvis and lower limb injury prediction capabilities have also been validated in pure lateral impacts (Takahashi et al., 2001 and 2003; Kikuchi et al., 2006 and 2008), they need to be further validated in different loading directions.

Some of the recent studies indicate potential effect of the musculature on the overall bending stiffness of the knee joint. For example, Lloyd et al. (2001) showed that the average contribution of muscles to knee valgus stiffness is $10 \pm 6.3\%$ of externally applied moment. Although the FE human and dummy models used in this study both lack muscle forces and thus the response comparison was made under the same condition, the effect of the musculature need to be further investigated when simulating the actual car-pedestrian accidents.

Moment measurements of the femur and tibia using strain gages were validated in quasi-static condition. Quasi-static and dynamic 3-point bending tests performed by Bose et al. (2007) using the tibia and femur of the modified dummy showed minimal rate sensitivity of the femur and tibia bending responses. This suggests that the measurement techniques validated in quasi-static condition should work in dynamic conditions as well. Although the deviation of the cross-section moments calculated from strains from those from the load cell support forces become larger for larger deflection probably due to the assumptions introduced by the application of the elastic beam theory, the trend is still qualitatively the same and dummy injury criteria could be determined by introducing a scaling factor for higher deflection level.

CONCLUSION

FE models for the femur and tibia along with the knee joint of the modified Polar-II pedestrian dummy were developed and validated against dynamic 3-point and 4-point bending tests, respectively. FE simulations were run using full-scale FE dummy and human models and FE vehicle front models with different geometries to compare femur and tibia responses during pedestrian impacts. The comparison showed that the difference between the dummy and human responses was significant particularly for SUV impacts.
impact, suggesting the need for the modification to the rigid dummy pelvis. Tentative dummy injury thresholds for the femur and tibia were determined by scaling human tolerances from literature based on the response correlations obtained from the full-scale FE simulations. The results of the quasi-static 3-point bending tests using the femur and tibia of the modified dummy with strain gages showed that the moments in the femur and tibia can be reasonably estimated from the strain measurements.

REFERENCES