

# Development of Injury Probability Functions for the Flexible Pedestrian Legform Impactor

Yukou Takahashi  
Honda R&D Co., Ltd.

Fumio Matsuoka  
Toyota Motor Corporation

Hiroyuki Okuyama  
Nissan Motor Company Ltd.

Iwao Imaizumi  
Honda R&D Co., Ltd.

## ABSTRACT

The goal of this study was to develop injury probability functions for the leg bending moment and MCL (Medial Collateral Ligament) elongation of the Flexible Pedestrian Legform Impactor (Flex-PLI) based on human response data available from the literature. Data for the leg bending moment at fracture in dynamic 3-point bending were geometrically scaled to an average male using the standard lengths obtained from the anthropometric study, based on which the dimensions of the Flex-PLI were determined. Both male and female data were included since there was no statistically significant difference in bone material property. Since the data included both right censored and uncensored data, the Weibull Survival Model was used to develop a human leg fracture probability function. As for the MCL failure, since one of the two data sources does not provide tabulated data, two MCL failure probability functions as a function of the knee bending angle developed using the Weibull Survival Model were averaged over the knee bending angle. The functions developed for the human leg fracture and the MCL failure were converted to those for the Flex-PLI using the results of the previous study that investigated the correlation between human and Flex-PLI injury measures using human and Flex-PLI finite element (FE) models and simplified vehicle models. 10% increase in failure tolerance due to muscle tone estimated from the literature was taken into account when converting the function for the MCL. Since the conversion of the MCL failure probability function was made using the correlation function between different human and Flex-PLI injury measures, the correlation between the different measures both for human was investigated for validating the conversion. The injury thresholds proposed for the Flex-PLI were evaluated in terms of equivalence to the current injury thresholds for the TRL legform specified in the global technical regulation (gtr) on pedestrian safety by developing an injury probability function for the tibia fracture measure used for the TRL legform. The results showed that the protection level provided by the currently proposed injury thresholds for the Flex-PLI is similar to that for the TRL legform in terms of injury probability.

**CITATION:** Takahashi, Y., Matsuoka, F., Okuyama, H. and Imaizumi, I., "Development of Injury Probability Functions for the Flexible Pedestrian Legform Impactor," *SAE Int. J. Passeng. Cars - Mech. Syst.* 5(1):2012, doi:10.4271/2012-01-0277.

## INTRODUCTION

According to the Japanese and US accident statistics [1,2], the lower limb is the most frequently injured body region in pedestrian serious injuries. In order to address pedestrian knee and leg injuries, the EEVC (European

Enhanced Vehicle safety Committee) subsystem test procedure using the legform impactor developed by TRL (Transport Research Laboratory) has been widely used. The test procedure was incorporated into the gtr on pedestrian safety [3] adopted by the World Forum for Harmonization of Vehicle Regulations (WP29) under the United Nations

Economic Commission for Europe (UNECE) in 2008. However, the improvement of the TRL legform in terms of the biofidelity and repeatability of its impact responses has been an issue.

In order to address this issue, Japan Automobile Manufacturers Association (JAMA) and Japan Automobile Research Institute (JARI) have developed the Flexible Pedestrian Legform Impactor (Flex-PLI) [4]. The biofidelity of the new legform has been significantly improved relative to the TRL legform by representing the flexibility of the bony structures (femur and tibia) as well as the anatomical structure of the human knee joint. The validity of the final version of the legform has been evaluated as a tool for regulatory testing by the international collaboration of the Flex-PLI Technical Evaluation Group (Flex-TEG) under UNECE/WP29. As the injury measures to be used for the Flex-PLI, the Flex-TEG has proposed the bending moment of the leg as well as the elongation of the knee ligaments inducing the Medial Collateral Ligament (MCL), Anterior Cruciate Ligament (ACL) and Posterior Cruciate Ligament (PCL). Although the injury thresholds associated with such injury measures have also been proposed by the Flex-TEG [5], no scientific background for determining the injury thresholds has been given.

The goal of this study was to develop injury probability functions for the human leg fracture and the knee ligament failure using the human impact response data available from the literature. The human injury probability functions were converted to those of the Flex-PLI by applying the results of the correlation analysis performed in a past study using human and Flex-PLI finite element (FE) models. In addition, the injury thresholds proposed for the Flex-PLI by the Flex-TEG were evaluated in terms of equivalence to the current injury thresholds for the TRL legform specified in the gtr on pedestrian safety by developing an injury probability function for the tibia fracture measure used for the TRL legform.

## DEVELOPMENT OF HUMAN INJURY PROBABILITY FUNCTIONS

### LEG FRACTURE

In car-pedestrian accidents, the most frequent scenario is that a pedestrian being hit laterally by a car while crossing a road [6]. In such a case, the lower limb of a pedestrian on the struck-side is primarily subjected to a loading in a Lateral-Medial (L-M) direction. For this reason, the results of the dynamic mid-shaft 3-point bending tests of the leg in a L-M direction obtained from Kerrigan et al. [7] were used to develop a human leg fracture probability function. Kerrigan et al. combined the dataset from their own tests with those from the past studies by Nyquist et al. [8] and Kerrigan et al. [9,10] to develop an injury probability function.

Table 1 summarizes the specimen information and the test results from the combined datasets presented in Kerrigan et al. [7]. As for the dimensions of the specimens, Nyquist et al.

[8] measured the distance from the bottom of the foot to the top of the tibial plateau (Tibia Height), whereas Kerrigan et al. measured the length of the tibia (Tibia Length). Kerrigan et al. [7] geometrically scaled the bending moment data at fracture using the standard lengths of 460.7 mm and 378.7 mm for Tibia Height and Tibia Length, respectively. The dimensions of the Flex-PLI are almost identical to those of the TRL legform, which were determined from the anthropometric data for the average adult male in a standing position from the study done by the University of Michigan Transportation Research Institute (UMTRI) [11, 12, 13]. In the current study, it was decided to use the standard lengths from the UMTRI study (483 mm and 402 mm for Tibia Height and Tibia Length, respectively), to apply geometric scaling consistent with the dimensions of the Flex-PLI. By assuming the constant mass density and the Young's modulus, the leg bending moment data at fracture were geometrically scaled using the following equation:

$$M_{scaled} = \lambda_L^3 M = (L_{ref} / L)^3 M \quad (1)$$

where

M : measured bending moment at fracture

$M_{scaled}$  : geometrically scaled bending moment at fracture

$\lambda_L$  : length scale factor

L : measured length (Tibia Height or Tibia Length)

$L_{ref}$  : standard length (Tibia Height or Tibia Length)

Table 2 presents a list of the geometrically scaled bending moment data along with the standard lengths and the length scale factors used for the data scaling.

**Table 1. Specimen information and leg fracture moment data from the literature**

| Test No.  | Data Source | Age | Gender | Stature (mm) | Weight (kg) | Anatomical Measurement |      | Fracture Moment (Nm) |
|-----------|-------------|-----|--------|--------------|-------------|------------------------|------|----------------------|
|           |             |     |        |              |             | Type                   | (mm) |                      |
| N-126     | A           | 58  | M      | 1740         | 73          | TH                     | 480  | 224                  |
| N-129     | A           | 57  | M      | 1780         | 99          | TH                     | 500  | 349                  |
| N-147     | A           | 57  | M      | 1780         | 84          | TH                     | 405  | 431                  |
| N-127     | A           | 56  | M      | 1760         | 79          | TH                     | 465  | 237                  |
| N-124     | A           | 64  | M      | 1770         | 82          | TH                     | 490  | 287                  |
| N-118     | A           | 54  | M      | 1820         | 68          | TH                     | 520  | 395                  |
| N-132     | A           | 57  | M      | 1870         | 45          | TH                     | 445  | 264                  |
| N-148     | A           | 57  | F      | 1630         | 75          | TH                     | 420  | 254                  |
| N-152     | A           | 51  | F      | 1630         | 68          | TH                     | 430  | 274                  |
| K(a)-134L | B           | 44  | M      | 1702         | 73          | TL                     | 420  | 416                  |
| K(b)-D1   | C           | 54  | M      | 1905         | 88          | TL                     | 445  | 463                  |
| K(b)-D2   | C           | 54  | M      | 1905         | 88          | TL                     | 450  | 485                  |
| K(b)-D3   | C           | 68  | M      | 1651         | 51          | TL                     | 385  | 290                  |
| K(b)-D4   | C           | 68  | M      | 1651         | 51          | TL                     | 385  | 309                  |
| K(b)-D5   | C           | 65  | F      | 1727         | 60          | TL                     | 378  | 416                  |
| K(b)-D6   | C           | 75  | M      | 1778         | 65          | TL                     | 395  | 306                  |
| 9.1       | D           | 66  | M      | 1829         | 79.8        | TL                     | 397  | 277                  |
| 9.2       | D           | 69  | M      | 1702         | 81.6        | TL                     | 418  | 433                  |
| 9.3       | D           | 62  | M      | 1829         | 60.8        | TL                     | 416  | 259                  |
| 9.4       | D           | 54  | M      | 1880         | 117.9       | TL                     | 479  | 482                  |

Data Source A: Nyquist et al., 1985

B: Kerrigan et al., 2003(a)

C : Kerrigan et al., 2003(b)

D: Kerrigan et al., 2004

Anatomical Measurement Type TH: Tibia Height

TL: Tibia Length

**Table 2. Standard length, scale factor and geometrically scaled leg fracture moment**

| Test No.  | STD Length (mm) | $\lambda_L$ | Scaled Moment (Nm) |
|-----------|-----------------|-------------|--------------------|
| N-126     | 483             | 1.006       | 228.2              |
| N-129     | 483             | 0.966       | 314.6              |
| N-147     | 483             | 1.193       | 731.1              |
| N-127     | 483             | 1.039       | 265.6              |
| N-124     | 483             | 0.986       | 274.9              |
| N-118     | 483             | 0.929       | 316.5              |
| N-132     | 483             | 1.085       | 337.6              |
| N-148     | 483             | 1.150       | 386.3              |
| N-152     | 483             | 1.123       | 388.3              |
| K(a)-134L | 402             | 0.9571      | 364.8              |
| K(b)-D1   | 402             | 0.9034      | 341.3              |
| K(b)-D2   | 402             | 0.8933      | 345.8              |
| K(b)-D3   | 402             | 1.0442      | 330.1              |
| K(b)-D4   | 402             | 1.0442      | 351.8              |
| K(b)-D5   | 402             | 1.0635      | 500.4              |
| K(b)-D6   | 402             | 1.0177      | 322.6              |
| 9.1       | 402             | 1.0126      | 287.6              |
| 9.2       | 402             | 0.9617      | 385.2              |
| 9.3       | 402             | 0.9663      | 233.7              |
| 9.4       | 402             | 0.8392      | 284.9              |

As shown in Table 1, three out of twenty data in the combined datasets were from female specimens. Since geometric differences were compensated for by applying the geometric scaling, the statistical significance of the difference in the bone material property between female and male was

investigated. Lindahl et al. [14] performed some quasi-static tensile tests using the test pieces taken from the femur and humerus specimens of various age and gender, and found that the gender differences in the failure strength, failure displacement and elastic modulus were statistically insignificant. Takahashi et al. [15] investigated the elastic modulus and the ultimate stress of the femoral cortical bone for male and female as a function of age. The results of F-test showed that the difference in the variance was statistically insignificant for both the elastic modulus and the ultimate stress at the 5% significance level ( $p=0.167$  for the elastic modulus,  $p=0.309$  for the ultimate stress), t-test was then applied to both datasets, showing that the difference in the mean was statistically insignificant at the 5% significance level ( $p=0.681$  for the elastic modulus,  $p=0.753$  for the ultimate stress, two-sided test). Based on these results, it was decided to include the geometrically scaled data for both female and male when developing an injury probability function.

The geometrically scaled bending moment data listed in Table 2 were subjected to Grubbs' test, showing that the data for the test number N-147 was a significant outlier at the 5% significance level (two-sided test). For this reason, the data for N-147 was omitted as an outlier. The data for the unsealed fracture moment listed in Table 1 show that the data for N-147 differs from the mean value by only 1.03 SD, while the difference is 3.596 SD when calculated using the scaled bending moment data presented in Table 2. This means that the geometrically scaled data for N-147 introduced much larger variability, suggesting that the Tibia Height of 405 mm



Documents

Document Sharing Hub

specified for N-147 in Nyquist et al. [8] is likely to be erroneous.

In [Table 1](#), the data for the test numbers starting with 'N' (nine out of twenty data) are from Nyquist et al. [8]. In the Nyquist paper, it is stated that the peak moment was attenuated by approximately 10% due to filtering. However, no scientific evidence is given to the validity of the 10% attenuation. For this reason, the data from Nyquist et al. were interpreted such that the fracture moment is always larger than the 'attenuated' peak bending moment presented in the Nyquist paper, hi contrast, other data presented in [Table 1](#) are from Kerrigan et al. [7,9,10], where unfiltered peak bending moment data are given. Since Nyquist et al. gives the bending moment lower than the fracture moment and Kerrigan et al. gives the exact fracture moment, the current study utilized the Survival Analysis to develop an injury probability function because this analysis is capable of dealing with both types of data. When only the bending moment data above which fracture occurred are available (as with the data from Nyquist et al.), the data are treated as right censored data. When the exact fracture moment data are known (as with the data from Kerrigan et al.), the data are treated as uncensored data. The Weibull distribution was chosen because the distribution is capable of representing asymmetric distribution and ensuring zero injury probability at zero loading represented by the injury measure. One of the most commonly used statistical procedures to develop an injury probability function is the logistic regression analysis. However, the current study chose not to employ this procedure because the study done by Kent et al. [16] showed that when the procedure is applied to the dataset that contains data at the exact occurrence of injury (uncensored data), then unrealistic injury probability functions may be given, such as the injury probability functions in the form of decreasing functions. This is because the procedure in theory treats the data as doubly censored (no injury data and injury data are treated as right censored and left censored data, respectively).

The leg fracture probability function was determined using the Survival Analysis and Weibull distribution based on the geometrically scaled leg fracture moment data presented in [Table 2](#), with the data for N-147 eliminated as an outlier. The leg fracture probability function was formulated as shown in [Equation \(2\)](#):

$$P = 1 - \exp[-\exp\{5.77458 \cdot \ln(M) - 34.51175\}] \quad (2)$$

where

P : leg fracture probability

M : leg bending moment (Nm)

The intercept and scale values of the probability function were statistically significant ( $p < 0.01$ ). [Figure 1](#) plots the leg fracture probability function determined in this study. The horizontal axis represents the human leg bending moment, and the vertical axis represents the leg fracture probability.

The leg bending moment corresponding to 50% probability of fracture is approximately 370 Nm.

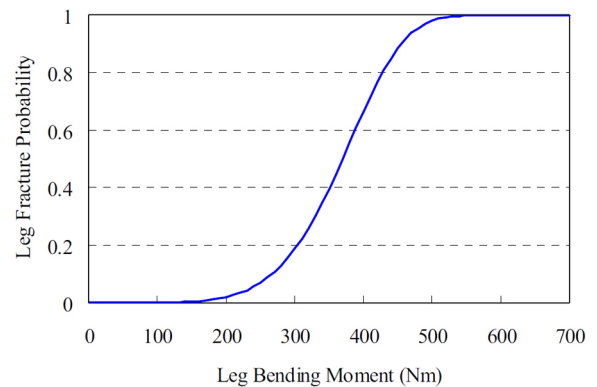


**Figure 1. Injury probability function for human leg fracture**

## KNEE LIGAMENT (MCL) FAILURE

The knee joint on the struck side of a pedestrian in a car-pedestrian collision is generally subjected to a valgus bending. For this reason, the results of the dynamic valgus bending tests using human knee specimens were taken from the literature to develop a human MCL failure probability function.

Ivarsson et al. [17] developed a human MCL failure probability function as a function of the knee valgus bending angle based on the results of the dynamic 4-point valgus bending tests of isolated human knee joints performed by Bose et al. [18]. Similar to the procedure employed in this study for developing the leg fracture probability function, Ivarsson et al. used the Weibull Survival Analysis to develop the probability function. In addition, they used two different definitions of the MCL failure timing and developed a probability function for each of the definitions. Definition A defined the initial peak timing of the knee bending moment as the timing of MCL failure, and Definition B defined the overall peak timing of the knee bending moment as the timing of MCL failure. Bose et al. performed eight dynamic 4-point valgus bending tests of the knee joint. In the post-test necropsy, no knee ligament failure was observed in one out of eight tests, a complete MCL failure was observed in one out of eight tests, and a partial failure of the MCL was observed in six out of eight tests. Since only a partial failure of the MCL took place in most of the tests, it is most likely that the failure probability be overly estimated when Definition A is applied. For this reason, the current study used the MCL failure probability function from Ivarsson et al. based on their Definition B. [Equation \(3\)](#) formulates the relationship between the knee valgus bending angle and the MCL failure probability developed by Ivarsson et al.

$$P = 1 - \exp[-\exp\{7.43354 \cdot \ln(\beta) - 21.937127\}] \quad (3)$$