

A 50th Percentile Female Human Body Model for Research on Equitable Occupant Protection

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ABSTRACT – The application of virtual human body models in vehicle safety research has expanded significantly, providing advantages over traditional anthropomorphic test devices. Key advantages include the flexibility to adjust body shape and posture and the ability to accurately assess injury risk at the tissue level. In this study, a virtual human body model representing a 50th percentile female occupant was developed to enhance research on gender and sex in injury assessment. The model was generated by scaling the existing 5th percentile occupant model taking into account the anatomical features of the female skeleton. The mechanical properties of the developed model were validated by comparing with those measured in post-mortem human subject tests described in the literature. Vehicle frontal and lateral collision simulations were conducted using male and female occupant models of different body shapes. The injury risks were analyzed and compared between the models.

INTRODUCTION

In recent years, the use of virtual human body models (HBMs) has been expanding in vehicle safety research. In vehicle collision tests, the injury risk of occupants is assessed using anthropomorphic test devices (ATDs). The structure of ATDs is standardized, and the degree of freedom for posture changes is limited. Common ATDs represent the body types of a small female, an average male, and a large male. Compared to ATDs, HBMs can flexibly adjust posture and body type. ATDs evaluate injury risk based on geometric measurements from sensors such as accelerometers and deflection gauges. On the other hand, HBMs can precisely evaluate injury risk at the tissue level of the human body based on strain values occurring in parts representing bones and internal organs.

Recently, sex-equitable injury risk assessment has been discussed. Forman et al. (2019) pointed out that female occupants have a higher risk of injury to chest and lower limbs, including the ankles, in frontal collisions compared to male occupants. As a general trend, females have smaller skeletons and muscles than males. Some researchers have highlighted the necessity of evaluating injury risk for occupants assuming an average female body type. Since developing and standardizing new ATDs takes many years, HBMs are considered promising for assessing injury risks for various occupant body types. This paper aims to develop an HBM representing a 50th percentile female occupant. Furthermore, using the developed model, frontal and lateral collision simulations were conducted to examine the impact of differences in body type on occupant injury risk.

METHODS

Generation of 50th Percentile Female Model

Total Human Model for Safety (THUMS) is a HBM developed to analyze human body injuries in vehicle collisions. The latest version of THUMS, Version 7 series, more precisely represents the human body structure including the thoracolumbar skeleton and abdominal organs. In this study, THUMS AF50 Version 7.1, a 50th percentile female occupant model with the features of Version 7, was developed (Figure 1). This model was based on the female 5th percentile model and scaled to a female 50th percentile. The biofidelity of the thorax and ankles, which are considered to be frequently injured, was improved.

Scaling and Adjustment. THUMS AF05 Version 7 is a model generated from CT scan images of a real woman with a 5th percentile body size, reflecting her anatomical features. The AF05 model was scaled to the 50th percentile body size based on the statistical data (Park et al. 2021). Table 1 shows the mean dimensions and scaling factors (ratio of 50th percentile dimensions to 5th percentile dimensions) for each body part. Furthermore, the head, thorax, and pelvis which form three-dimensional shapes were scaled so that their dimensions along the three axes approximate the statistical values of 50th percentile females (Fujita et al. 2025). Notably, the female thorax is characterized by smaller dimensions in the anterior-posterior direction compared to males. The thorax of the AF50 model was adjusted within the range of measurement data variability to represent the characteristics (Weaver et al. 2014, Figure 2).

Improvement of Foot and Ankle Ligament Model. The ligament models around the foot and ankle were represented by 2mm-thick shell elements. The shapes

were revised based on anatomical illustrations to more accurately reproduce the mechanical properties of the ankle joint. The ends of the tibiofibular ligament (TFL), calcaneofibular ligament (CFL), and deltoid ligaments (DL: ATTL, PTTL, TCL, and TNL) were faithfully represented at their attachment points to the bone. The talocalcaneal ligament (TCL) and long pedal ligament (LPL), which had been omitted, were added. The material model of the ligaments was assumed to be isotropic elastic. The properties of the major ligaments were defined based on measurement data from the literature (Siegler et al. 1988) to approximate the average stiffness (Figure 3).

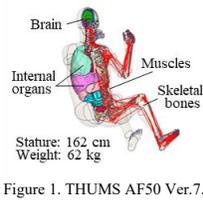


Figure 1. THUMS AF50 Ver. 7.1

Table 1. Interarticular Length and Scaling Factors

		50F [mm]	5F [mm]	Scaling factor
Body	Neck-C7	110.6	103.6	1.07
	T1-T12	257.4	240.0	1.07
	L1-S1	160.6	147.5	1.09
Upper arm		209.1	192.5	1.09
Lower arm		215.0	195.2	1.10
Upper leg		391.4	354.4	1.10
Lower leg		386.7	358.5	1.08
Stature [mm]		162	153	-
Weight [kg]		62	49	-

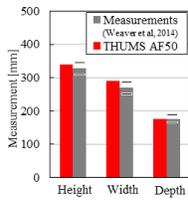


Figure 2. Ribcage Dimensions

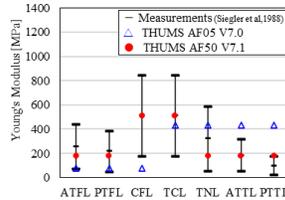


Figure 3. Material Properties of Ligaments

Validation of Biomechanical Fidelity

The mechanical impact responses of the THUMS AF50 model were validated to results of Post Mortem Human Subjects (PMHS) tests reported in the literature. A total of 38 cases of loading configurations were validated for each body part and for the whole-body model. These validation cases are available at <https://www.toyota.co.jp/thums/about/>. This section describes some of the validations for the thorax and ankles, where the risk of injury for female occupants is higher than that for male.

Thoracic Validation. The load responses of the anterior thorax under belt loading were evaluated (Figure 4). In the test conducted by Kent et al. (2004), a belt was wrapped around the chest of a PMHS positioned supine on a platform, and both ends of the belt were pulled posteriorly at an average speed of 0.9 m/s. Based on the test results, a corridor (mean ± standard deviation) of load-chest deflection curves for a 50th percentile male was generated. This corridor was adjusted for a 50th percentile female using the correction method proposed by Eppinger et al. (1984). The PMHS test was simulated using THUMS AF50,

and the resulting load-deflection curve was compared with the PMHS corridor.

Ankle Validation. The mechanical properties of the ankle joint under eversion loading conditions were evaluated (Figure 5). In the test conducted by Roberts et al. (2018), a 2 kN pre-load was applied to the proximal end of the PMHS lower leg, and a fixture secured to the foot was rotated outward. Based on the test results, a moment-rotation angle corridor for a 5th percentile female ankle was generated. This corridor was adjusted for a 50th percentile female using the method described above. PMHS test was simulated using THUMS AF50, and the resulting moment-angle curve was compared with the PMHS corridor.

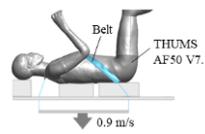


Figure 4. Anterior Thorax Belt Loading

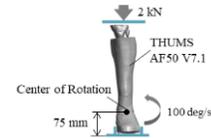


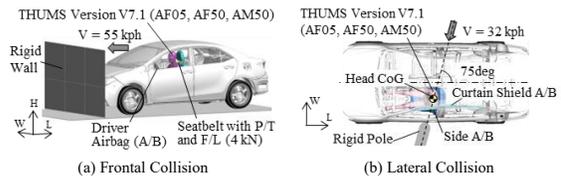
Figure 5. Ankle Joint Eversion

Application to Collision Simulation

The effect of occupant body size differences on crash injury risk was investigated as a first use case. Frontal and lateral collision simulations were conducted with three occupant models (AF05, AF50, and AM50: 50th percentile male) seated in a vehicle model (Figure 6).

Simulation Condition. A mid-size sedan was assumed as the vehicle model. Two collision scenarios were considered: a frontal collision against a flat rigid wall at 55 kph, and a side collision against a rigid pole at 32 kph. The fore-aft position and height of the seat were adjusted according to each occupant’s body size, and the occupant models were positioned accordingly. The right foot of the occupant was placed on the accelerator pedal, while the left foot was positioned on the footrest. A seatbelt was fitted to each occupant model, incorporating a force limiter set to 4 kN. The airbag was also deployed in the simulations.

Injury Risk Output. The 95th percentile value of maximum principal strain (MPS95) in the major skeleton was calculated as an index of injury risk. In the THUMS, the fracture threshold is defined as a principal strain of 3% (Shigeta et al. 2009).



(a) Frontal Collision

(b) Lateral Collision

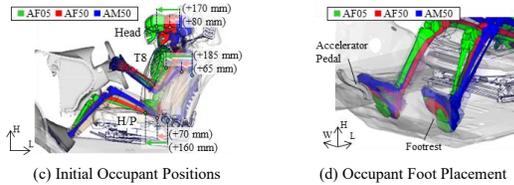


Figure 6. Simulation Conditions of Vehicle Collisions

RESULTS

Validation of Biomechanical Fidelity

Thoracic Validation. Figure 7 shows the simulation result of the load-chest deflection curve obtained using THUMS AF50. The curve of THUMS was within the PMHS corridor from the start of compression to 40 mm of chest deflection.

Ankle Validation. Figure 8 shows the simulation result of moment-rotation angle curve for the ankle of THUMS AF50. The curve of THUMS was generally within the corridor. Although the THUMS moments were lower than those of the PMHS corridor at the rotation angle less than 15 degrees, it was judged to be sufficiently accurate because the injury risk occurred at angles greater than 15 degrees.

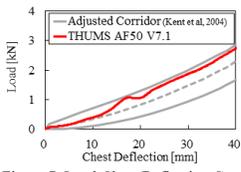


Figure 7. Load-Chest Deflection Curve in Anterior Thorax Belt Loading

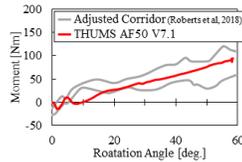


Figure 8. Moment-Rotation Angle Curve in Ankle Joint Eversion

Collision Simulation

The results of the vehicle collision simulations were analyzed to compare injury risks among the three occupant models representing different body sizes and genders. The cortical bone strains (MPS95) of the major skeleton are shown in Figure 9. The ankle (talus) strain during frontal collision was higher in both AF05 and AF50 compared to the AM50. Conversely, thoracic (rib) and lumbar (pelvis) strains during both frontal and lateral collisions, and lower limb (femur, tibia) strains during frontal collision were higher in AM50 compared to AF05 and AF50.

DISCUSSION

The simulation results indicated that female occupants (AF05 and AF50) exhibited higher talus strains than male occupant (AM50) in frontal collisions. The finding is consistent with the previous report on ankle injury risks in the field data. To explore the possible causes for this trend, the kinematics of the lower extremities during frontal collisions were examined. In

all occupant models, forward displacement of the lower limbs caused the right foot to move away from the center of the pedal, resulting in eversion of the ankle joint. The ankle eversion stretched the medial ligaments (DL) and induced bending compression in the lateral bones such as the distal fibula and the talus (Figure 10). The region of concentrated strain in the distal fibula corresponded to the typical fracture site observed during ankle eversion, which is generally classified as Weber Type B. While the moment applied to the ankle generally increases with occupant mass, the ankle joint stiffness may decrease in small-sized occupants. Consequently, the female occupants exhibited greater ankle rotation than the male occupant, leading to higher strain in both the ligaments and the bones (Figure 11). In lateral collision, the male occupant showed higher risks to the thoracic and lumbar regions, possibly due to the smaller distance to the door trim.

CONCLUSION

A female occupant HBM with a 50th percentile body size was developed. The frontal collision simulation results indicated that the lower stiffness of the ankle joints of female occupants was a possible reason for the higher risk of injury compared to the male occupants. The study has limitations that simulations were conducted assuming specific conditions such as vehicle type and occupant body size, while actual vehicle collisions occur under various conditions. Additionally, since the model was created by simple scaling the existing HBM, it may not fully capture the anatomical features of the bones in certain regions as precisely as recent morphing methods that utilize numerous landmark points (Lindgren et al. 2024). Further studies are needed to consider the effects of various factors that may influence injury risk.

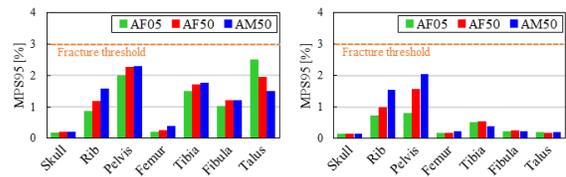


Figure 9. MPS95 in Major Bones



Figure 10. Distribution of Strain in Ankle Bones and Ligaments

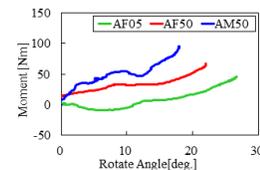


Figure 11. Moment-Rotate Angle Curves of Ankle Joint

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