

Development of a Finite Element Model for the Pedestrian Lower Limb

– Development and Validation of Bone Models –

Yuji KIKUCHI*

Yukou TAKAHASHI*

Atsuhiko KONOSU**

Hirotoishi ISHIKAWA**

ABSTRACT

A finite element model for the long bones of the human lower limbs that accurately reflects their anatomical structure has been developed. The average values of the material properties of the bones as well as their strain rate dependency were determined from an extensive survey of the related literature and were reflected in the model. The model has been validated against published quasi-static and dynamic 3-point bending test results, confirming that it accurately predicts the peak impact force. In addition, it was found that the strain rate dependency of the material properties has a significant effect on the accuracy of peak force prediction.

1. Introduction

According to accident statistics, pedestrians account for a high proportion of traffic accident fatalities in Japan, at approximately 30%.⁽¹⁾ This accounts for the high priority placed on pedestrian protection together with occupant protection in the improvement of automobile safety. The Pedestrian Crash Data Study (PCDS), a database of pedestrian accident surveys in the United States, was used to examine distribution of the site of injuries with the Abbreviated Injury Scale (AIS) rating 2 or more. As shown in Fig. 1, the lower extremity area was found to be the most frequently injured body region. While not typically life-threatening, lower limb injury is becoming an increasingly important concern in terms of long recovery time and social cost.

Pedestrian lower limb injuries, as shown in Fig. 2, can be classified into two types: bone fracture and damage to knee ligament. Working Group 17 of the European Enhanced Vehicle-safety Committee (EEVC) has proposed a component test procedure using a legform impactor for evaluating these injuries.⁽²⁾ However, recent studies have shown that the biofidelity of the impactor and validity of

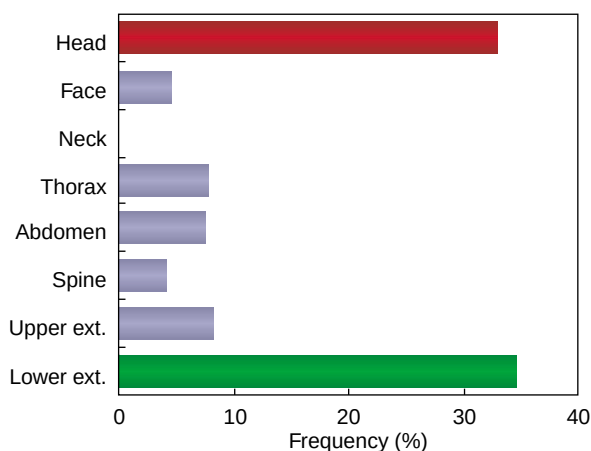


Fig. 1 Distribution of AIS2+ pedestrian injuries by body region (Source : PCDS database)

injury criteria should be further improved.^{(3) - (5)} The test procedure will require further study for application to lower limb injury evaluation in car-pedestrian accidents. This paper reports on the development of a finite element model

* Tochigi R&D Center

** Japan Automobile Research Institute

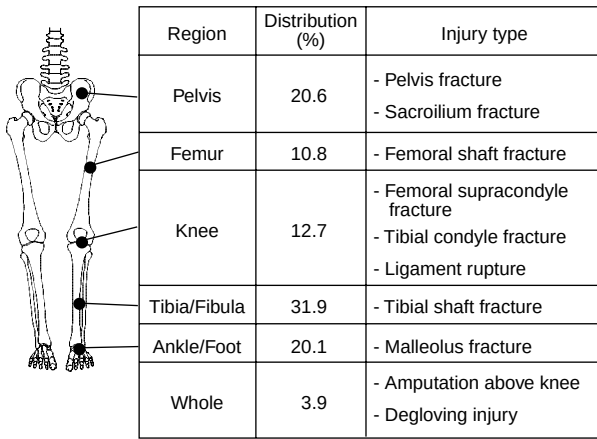


Fig. 2 Distribution of AIS2+ lower limb injuries by injury type (Source: PCDS database)

for the pedestrian lower limb with accurate anatomic structure. The purpose of this development was to identify the mechanism of injuries to the pedestrian lower limb and to establish an improved methodology for evaluating these injuries.

Large numbers of simulation models have been developed in the past in order to reproduce lower limb response in car-pedestrian accidents and identify the injury mechanism. Yang and Kajzer⁽⁶⁾ used the MADYMO program to develop a three-dimensional multibody model of the pedestrian lower limb with a focus on the anatomic structure of the knee. Since the thigh and leg were each modeled as single rigid segments, it was not possible to simulate bone fracture. Yang et al.^{(7) (8)} attempted to reproduce bone fracture by adding a joint element to this model in the upper part of the leg. The problem, however, was that this required the location of the bone fracture to be specified in advance. In these multibody models, the ability to simulate deflection of the bony structure and bone fracture is largely dependent on the modeling method, and it is difficult to construct a model that can be used under various different loading conditions.

Bermond et al.^{(9) (10)} developed a three-dimensional finite element model of the knee joint, but this has not been validated for accuracy of dynamic response. Schuster et al.⁽¹¹⁾ developed a three-dimensional non-linear finite element model for the pedestrian lower limb. The response of the bony structure, however, has only been validated under quasi-static conditions. Tamura et al.⁽¹²⁾ used a finite element model of the human body to study the tibial injury mechanism in passengers during a collision. However,

quasi-static data were used for the material properties of bones, and the repeatability of the dynamic response has not been validated. Although the finite element models developed in the past studies had broader applicability than that of multibody models, no model has succeeded in simulating the dynamic response of the bony structure accurately.

This report describes the development and validation of a bone model produced in the course of development of a finite element model for the pedestrian lower limb. As noted above, lower limb injuries to pedestrians are generally categorized as bone fracture or ligament damage. Past experimental studies showed that ligament damage rarely occurs when bone fracture takes place.⁽¹³⁾ A possible explanation for this is that a bone fracture interrupts the load path so that the force on the knee joint is reduced. In this study, the development of a bone model consequently incorporated a modeling method that was capable of simulating the interruption of the load path due to bone fracture. In order to accurately predict the dynamic response of the lower limb, the model also incorporated strain rate dependency of the material properties of bone that had not been taken into consideration in previous studies. Validation of this model was also carried out under dynamic conditions.

2. Structure and Material Properties of Bone

The structure of long bones is shown in Fig. 3. The surface of the epiphysis is covered by cortical bone, which has high stiffness, while the interior is filled with trabecular bone, which has a honeycomb-like structure and relatively low stiffness. The diaphysis forms a tubular structure of continuous cortical bone that grows progressively thicker toward the mid-section of the bone. There is no trabecular bone in the interior of the diaphysis, which is instead filled with yellow bone marrow. Consequently, prediction of the mechanical response of the bony structure requires the model to separately characterize and incorporate the

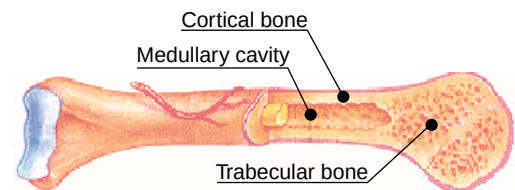


Fig. 3 Structure of long bones

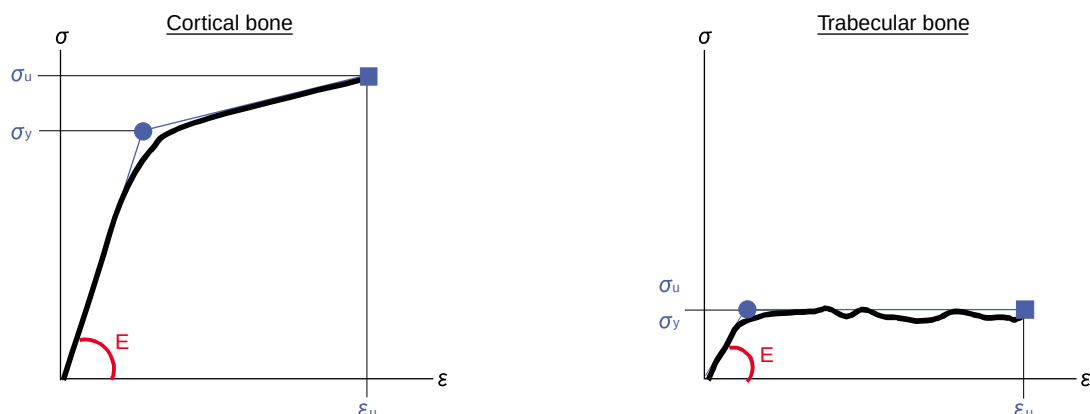


Fig. 4 Typical stress-strain curves for bone and parameters specified in the model

properties of the cortical bone and trabecular bone in order to take into account the difference in stiffness between these bone components.

The typical stress-strain curves for cortical bone and trabecular bone are shown in Fig. 4. To simplify modeling, the stress-strain relationships were approximated with the two phases shown by the blue lines, and the four parameters of the Young's modulus (E), the yield stress (σ_y), the ultimate stress (σ_u), and the ultimate strain (ϵ_u) were determined through extensive literature survey.^{(14)–(51)} For the trabecular bone, the properties of the plastic region were approximated by a constant stress ($\sigma_y = \sigma_u$). It was

anticipated that the material properties of the bones would correlate with age when the survey results were compiled. Therefore the values for each property were plotted against age as the horizontal axis. The data set was not separated by gender, since past studies have shown no gender difference at the material property level.⁽⁵²⁾

Fig. 5 shows the survey results for the Young's modulus and ultimate stress of femoral cortical bone. The data are presented under the categories of tension and compression. Clinical findings indicate that bone fracture is generally initiated in the portion subjected to tension. Consequently, these figures show that bone has less strength

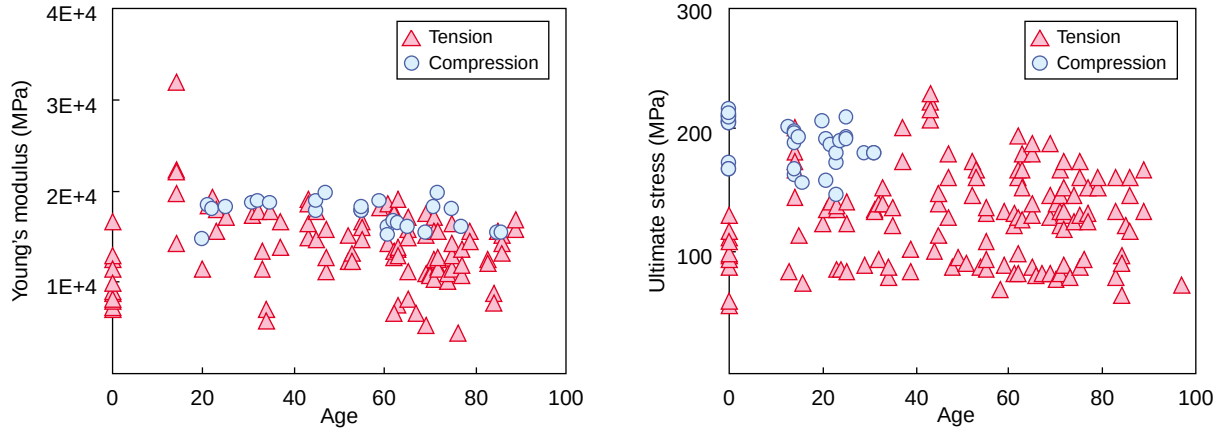


Fig. 5 Tensile and compressive properties of femoral cortical bone as a function of age

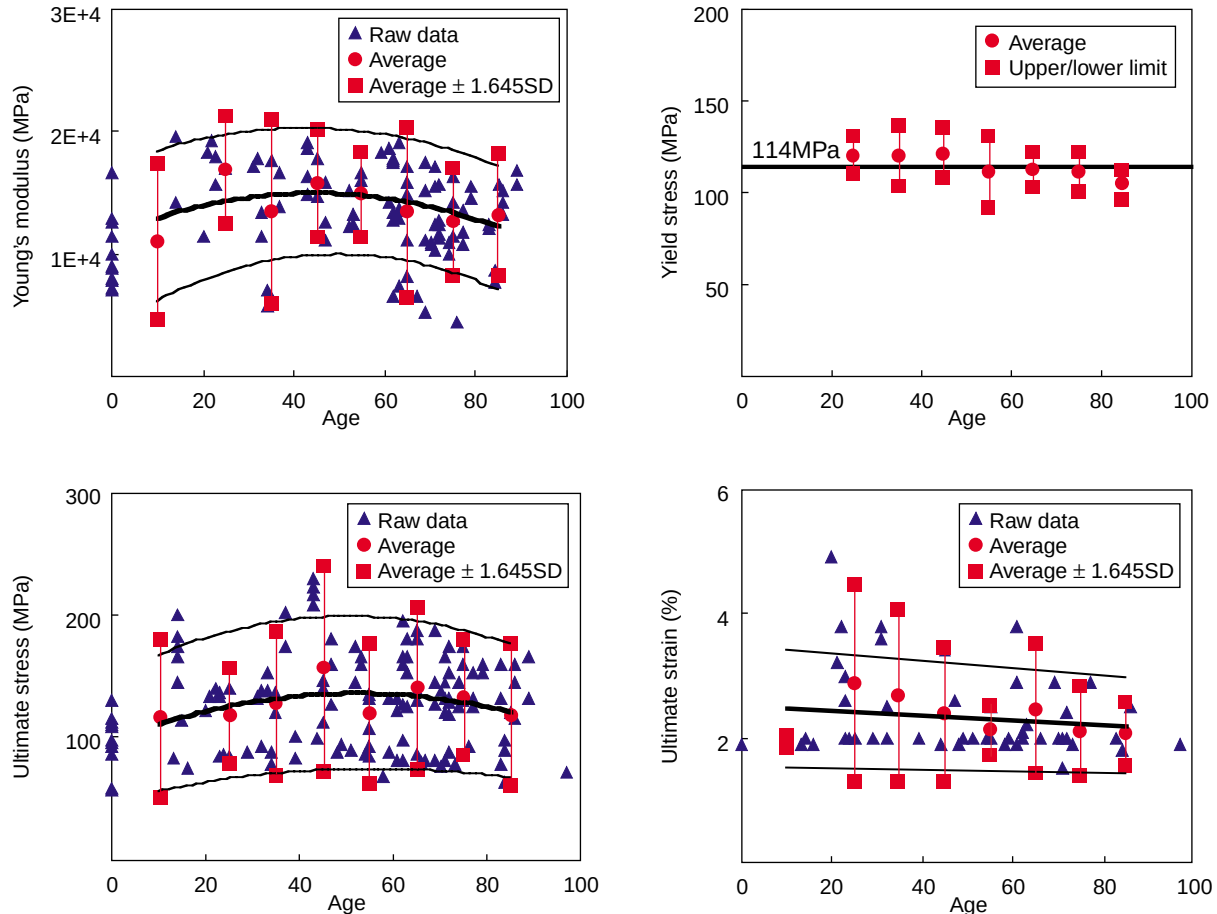


Fig. 6 Material property corridor of femoral cortical bone as a function of age

in tension than in compression, and the results are consistent with the clinical observations. In this research, therefore, the properties of bone in tension were used for the purposes of bone fracture prediction. Fig. 6 shows the survey results for the above four material properties with respect to femoral cortical bone. The values for properties other than the yield stress were divided into age groups at 10-year intervals and the averages were taken (red circles), and upper and lower limits were calculated (red squares) so as to include 90% of the data on the assumption of normal distribution. These were interpolated using a quadratic function to establish a corridor for the properties as shown in the graphs. For the ultimate strain, linear interpolation was chosen due to the relatively small amount of data and its wide variety. The only data obtained for the yield stress were the average values for the 10-year age groups (red circles) and the upper and lower limits (red squares). Therefore the correlation with age was not taken into account, and it was decided to use uniform averages instead.

Table 1 Differences in the material properties of the cortical bone in the femur and tibia (age 20 – 89, average)

Material property	Femur	Tibia	Ratio
Young's modulus (MPa)	16843	23843	1.42
Yield stress (MPa)	114	129	1.13
Ultimate stress (MPa)	132	157	1.19
Ultimate strain (%)	2.8	3.1	1.09

It is apparent from these graphs that the values for bone properties are affected even more by individual difference than by age. Use of these collected figures makes it possible to construct a model that incorporates the age and bone strength level of the pedestrian to be analyzed. The only data obtained for the tibial cortex were averages. Therefore, as shown in Table 1, the ratio of the average value for each property between the tibia and femur was used in order to obtain the corridor for the tibia by scaling that of the femur. No data at all were found for the fibula. The values for the properties of the femur were therefore used as initial estimates, and these were later adjusted to agree with test results during the process of validation, described below. Since there was also very little data on the properties of trabecular bone, it was not possible to determine material properties for each bone separately. The trabecular bone in the femoral head was found to have high stiffness compared to other regions, however, and this region was given different properties, as shown in Fig. 7. The stiffness of trabecular bone is about 1/10 that of cortical bone, and it was anticipated that the contribution of trabecular bone to the overall material properties of the bone is relatively small. Therefore, considering the small amount of data available, it was decided to use average values without defining the correlation with age. The ultimate strain for trabecular bone was set at 13.4%, using the only source that refers to this property.⁽⁵³⁾ A value of 0.315 was used as Poisson's ratio for cortical bone and trabecular bone in all the bones, taken from an average of the small amount of data available.^{(23) (27) (47)}

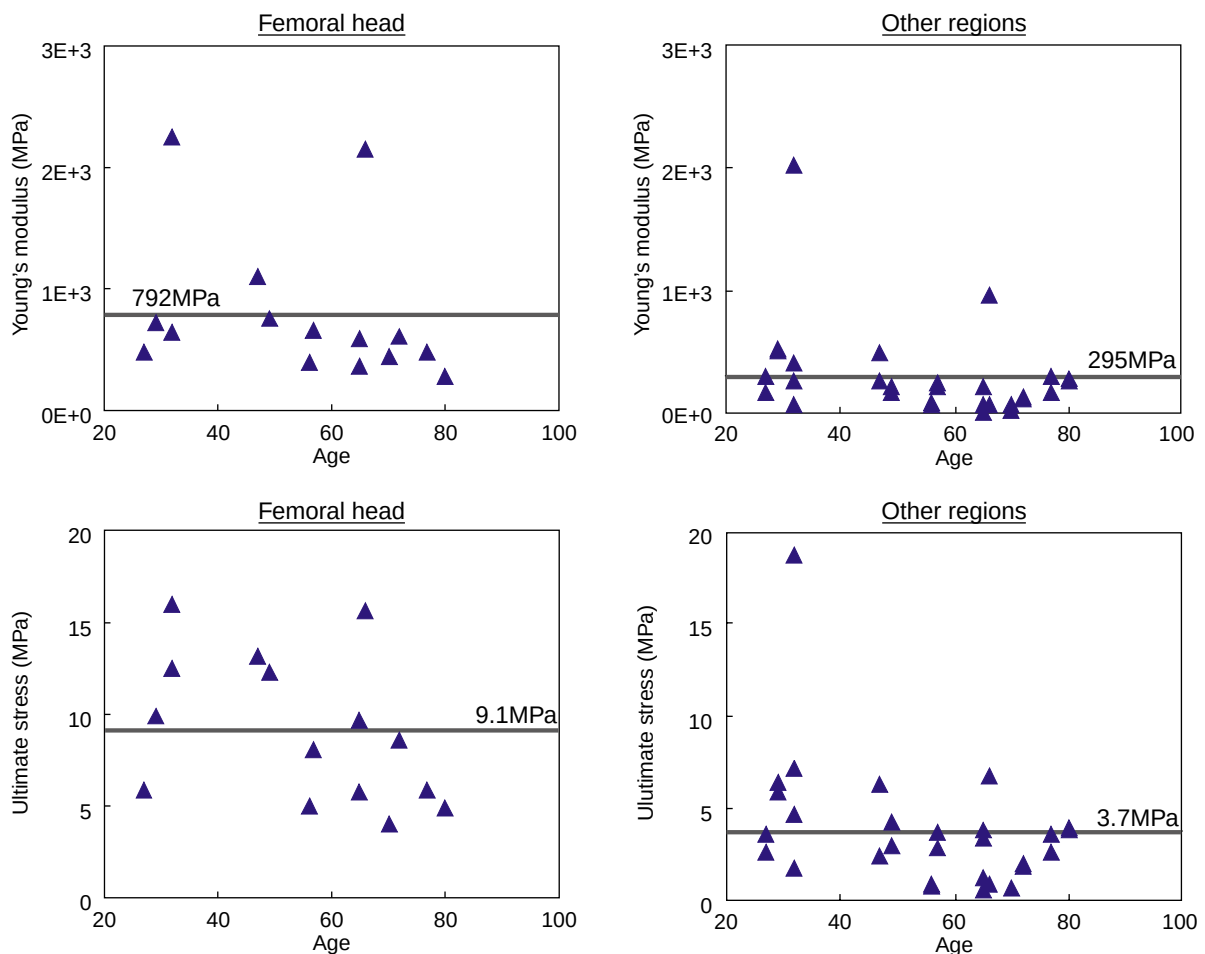


Fig. 7 Material properties of the trabecular bone for the femoral head and other regions

Carter and Hayes⁽⁵⁴⁾ ⁽⁵⁵⁾ experimentally examined the effect of strain rate on the stiffness and strength in cortical bone and trabecular bone, and found that the stiffness and strength of bone are proportional to the strain rate raised to the power of 0.06 (Fig. 8). Based on these findings, the relationship between the strain rate and dynamic factor was defined as shown in Fig. 9. It is also apparent in Fig. 8 that a strain rate dependency exists with regard to the Young's modulus. Therefore, this relationship was applied to every region of the stress-strain curve, including the elastic region.

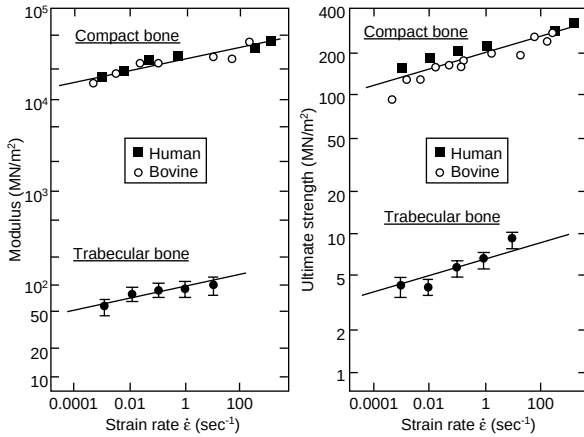


Fig. 8 Strain rate sensitivity of the material properties of bone

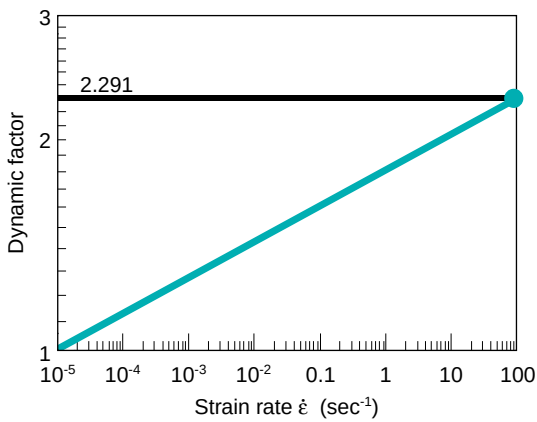


Fig. 9 Dynamic amplification factor versus strain rate specified in the model

3. Model Description

The external shape of the bone model was obtained from the H-DummyTM ⁽⁵⁶⁾ finite element model of the human lower limb developed by Nihon ESI. The shape of this model is anatomically accurate since it is based on the human bone model from the Viewpoint DatalabTM that was generated from scanned images of the human body. ⁽⁵⁷⁾ The configuration of the bone model is consistent with that of human long bone shown in Fig. 3. As shown in Fig. 10, the epiphysis is modeled with a two-layered structure that uses shell elements for the surface cortical bone and solid elements for the internal trabecular bone. The diaphysis is modeled using solid elements for the cortical bone in order to simulate the internal cavity. A closer look inside cortical bone, as shown in Fig. 11, shows the existence of

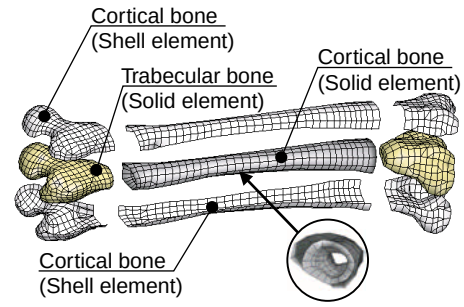


Fig. 10 Structure of femur model

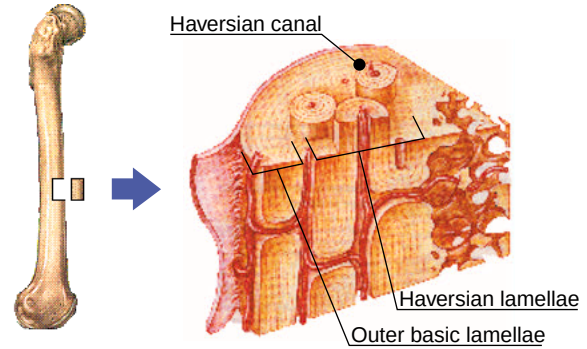


Fig. 11 Structure of cortical bone

Haversian lamellae enclosing large numbers of Haversian canals through which blood vessels pass. On the surface of the Haversian lamellae are outer basic lamellae that have even greater stiffness. ⁽⁵⁸⁾ The material property of cortical bone found in the literature is related to these Haversian lamellae. Therefore, further shell elements were added to the outside of the diaphysis, where the cortex is thick and the outer circumferential lamellae are considered to have a significant effect. In this way the geometric moment of inertia of the diaphysis was corrected. As the literature did not contain quantitative data on the material properties of the outer circumferential lamellae, however, the data for the Haversian lamellae were used for this layer instead. The thickness of the lamellae was determined so that the geometric moment of inertia was identical to that calculated by applying beam theory to the published load-deflection curves from quasi-static 3-point bending tests conducted for each bone.

For the purpose of bone fracture modeling, material model Type 16 (solid element) and Type 105 (shell element) in PAM-CRASHTM were chosen for bones since they have fracture modeling functions. The Element Elimination option in these material models was applied in order to simulate bone fracture. This option steps the stiffness of certain elements down to zero when the elements reach the value set for the ultimate strain. For the strain rate dependency of material properties, the relationships in Fig. 9 were incorporated into the model using the PAM-CRASHTM logarithmic interpolation method. ⁽⁵⁹⁾ ⁽⁶⁰⁾

4. Model Validation

The bone models were validated under quasi-static conditions against the results from quasi-static 3-point bending tests with human long bones conducted by Yamada

et al.⁽⁶¹⁾ Though the test conditions are not described in detail in Yamada et al., they are assumed to have been as shown in Fig. 12. The literature contains load-deflection properties for human long bones, including such bones of the lower limb as the femur and tibia, in the form of average values. A model as shown in Fig. 14 was created and used for comparison of load-deflection curves. The computer simulations were run utilizing PAM-CRASH™. Since this computer program is for dynamic analysis, settings for the strain rate dependency of material property were excluded in order to simulate quasi-static conditions. The tests conducted by Yamada et al. used bones from persons aged 20 – 39 years. Therefore, average values calculated for the corresponding age group from the age corridor determined above were used. The material property values used in the computer simulation are shown in Table 2. A comparison of the simulation and test results for load-deflection properties of the femur, tibia, and fibula is shown in Fig. 15. The test results show slightly higher loads as compared to the simulation results when deflection is in the vicinity of approximately 5 – 10 mm, while the load and deflection at the point of bone fracture show excellent correlation for every bone.

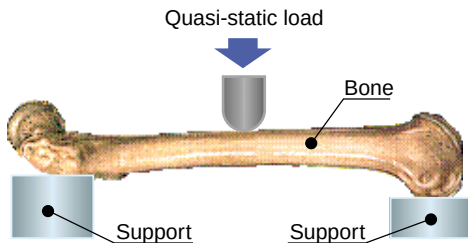


Fig. 12 Assumed quasi-static 3-point bending test setup for long bone

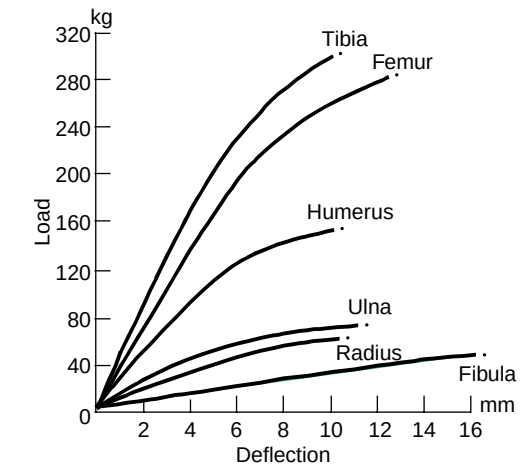


Fig. 13 Load-deflection curves for human long bones

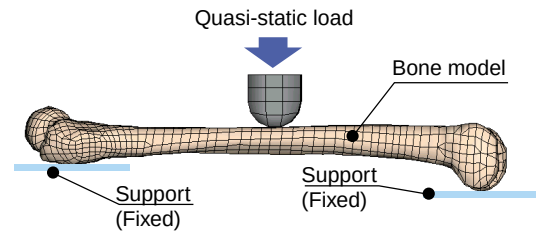


Fig. 14 Schematic diagram of femur model

Table 2 Material property of bones used in the model for quasi-static validation (for age 20 – 39, average)

Material Name	Density (kg/m ³)	Young's modulus (MPa)	Yield stress (MPa)	Yield strain (%)	Poisson's ratio	Ultimate stress (MPa)	Ultimate strain (%)
Femur cortical	1850	14 317	114.1	0.80	0.315	123.3	2.0
Tibia cortical	1850	20 033	129.0	0.63	0.315	138.1	1.5
Fibula cortical	1850	17 406	103.2	0.59	0.315	107.0	0.9
Trabecular	600	295	3.7	1.25	0.315	3.7	13.4
Femoral head trabecular	600	792	9.1	1.15	0.315	9.1	13.4

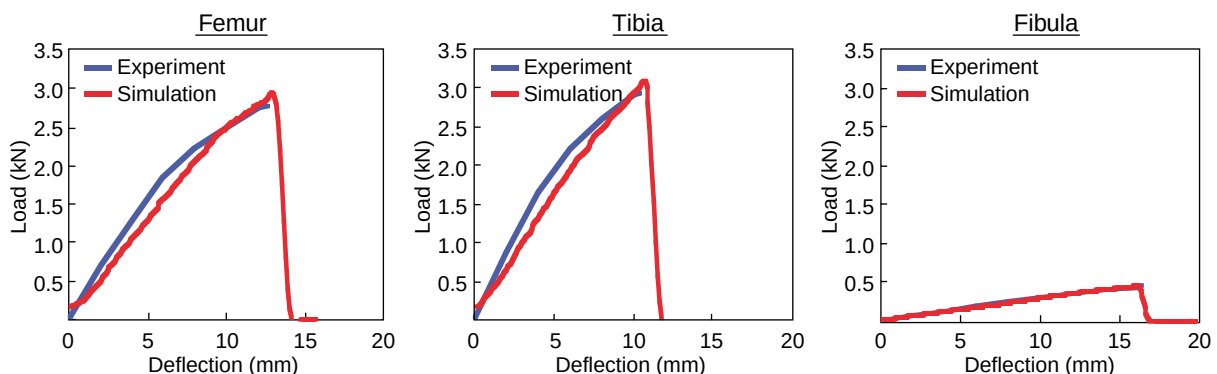


Fig. 15 Comparison of load-deflection response of bone in quasi-static 3-point bending

The bone model should also be validated under dynamic conditions since the impact between the pedestrian lower limb and the vehicle during an accident is a dynamic phenomenon. Dynamic validation of the bone model was carried out using the results from dynamic 3-point bending tests on many of the lower limb bones conducted by Kress et al.⁽⁶²⁾ Kress et al. used the test setup shown in Fig. 16 to obtain average peak forces occurring when an impactor weighing 50 kg is made to impact with the bone at a velocity of 7.5 m/s. Table 3 shows the test conditions and the average peak forces. The simulations employed the material properties corresponding to the average age for each set of conditions as calculated from the corridor. The tests applied impact to the femur and tibia from two directions, Lateral-Medial (L-M) and Anterior-Posterior (A-P), while the fibula only had L-M impact applied.

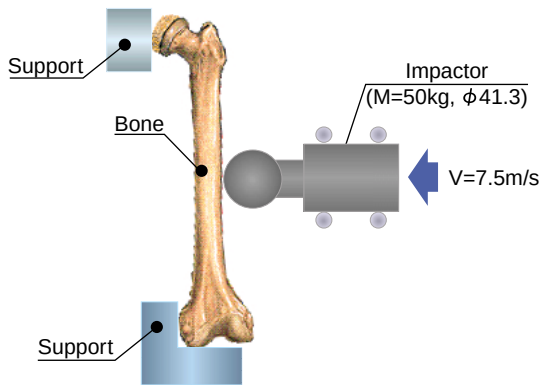


Fig. 16 Test setup for dynamic 3-point bending

Table 3 Test results of dynamic 3-point bending

Specimen	Impact plane	Age (Average)	n	Peak force (kN)
Femur	L-M	71.0yr	21	5.48 (SD 1.17)
	A-P	69.2yr	50	5.70 (SD 0.46)
Tibia	L-M	74.3yr	33	4.07 (SD 1.22)
	A-P	69.9yr	95	4.85 (SD 2.08)
Fibula	L-M	72.5yr	21	0.57 (SD 0.28)

* A-P : Anterior to Posterior / L-M : Lateral to Medial

Simulation models of the kind shown in Fig. 17 were created for all 5 of these cases and the peak forces were compared. The figure also shows deformation diagrams for the time of bone fracture, and the elements that exceed the ultimate strain threshold specified in the PAM-CRASH™ Element Elimination option are set not to be displayed. The mass of these elements is distributed in the surrounding nodes, and consequently they are conserved even after exceeding the threshold. The stiffness, however, diminishes to zero, so that load transmission does not take place in these elements to either side. In reality, butterfly-type fracture patterns are often observed in bone fractures caused by bending. However, in the simulation results in Fig. 17, the elements in the row immediately under the impactor are subject to Element Elimination. Therefore, this model is not capable of simulating the microscopic mechanism of actual bone fracture. This kind of model can, however, reproduce the interruption of the load path due to bone fracture, which is important for the analysis of pedestrian injury. A comparison of the peak forces between the simulation results and the test results is shown in Fig. 18. In the simulations, two models were employed: one with and one without strain rate dependency, in order to clarify the effect of the strain rate dependency of the material properties in dynamic loading conditions. It was found that the simulation results from the model without strain rate dependency showed a peak force of 1/2 of the test results

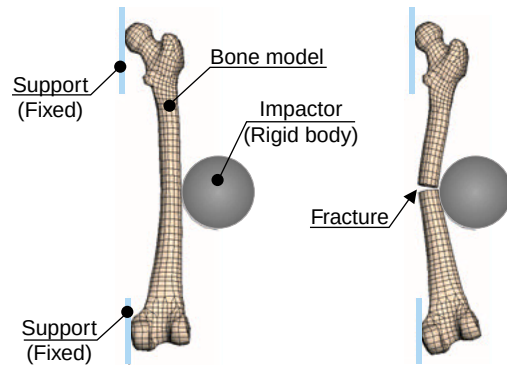


Fig. 17 Femur model setup for 3-point dynamic bending

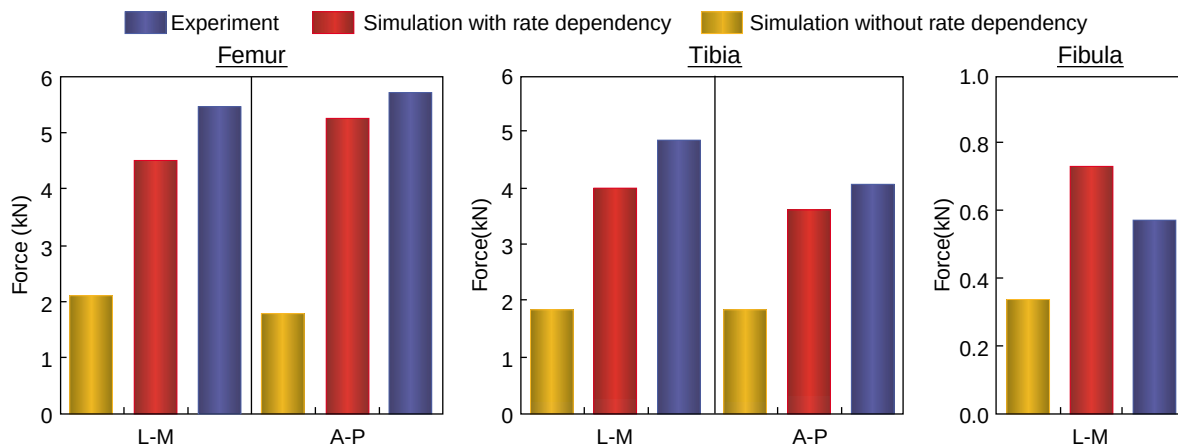


Fig. 18 Comparison of peak force between experiment and simulation

or less, while incorporation of strain rate dependency yielded results that were closer to the test results. This suggests that it is essential for the model to incorporate strain rate dependency of the material properties of bone in order to reproduce the injury mechanism of the pedestrian lower limb under dynamic conditions.

This research showed that the bone model developed was able to accurately predict the peak impact force in quasi-static and dynamic 3-point bending. In actual pedestrian accidents, however, the bumper frequently impacts the pedestrian lower limb immediately below the knee joint. Therefore, it is possible that the distribution of shear force and bending moment differs from 3-point bending. In addition, due to the microscopic internal structure of the long bone as illustrated in Fig. 11, the material properties are different in the axial and transverse directions.⁽⁶³⁾ A material model that simultaneously incorporates the anisotropy of a material, the strain rate dependency, and a fracture model is currently not available in PAM-CRASH™. Therefore, the model developed in this study focused on reproduction of bone fractures and assumed isotropic material properties. Future study should focus on the development of a material model that takes the above three elements into account simultaneously and the model validation should occur under conditions that more closely approximate actual pedestrian accidents.

5. Conclusion

A finite element model of the long bone of the human lower limb was developed that accurately represents the anatomic structure of the bone and incorporates material properties determined through extensive survey of the literature. The model was validated against test results from quasi-static and dynamic 3-point bending. The following conclusions were reached as a result:

- (1) It was found through survey of the literature on material properties of bone that stiffness and strength are lower in tension than in compression. In relation to age, stiffness and strength tend to be at their peak around the age of 40, while individual difference has a greater effect even than age in terms of the amount of variation.
- (2) The bone model developed in this study was found to correlate very well with test results in terms of the load and displacement magnitude at the time of bone fracture in quasi-static 3-point bending, and in terms of the load at the time of bone fracture in dynamic 3-point bending.
- (3) It was found to be essential for the model to incorporate strain rate dependency of the material properties, which was not taken into consideration in previous research, in order for it to accurately simulate the dynamic response of bone.
- (4) The reflection of anisotropy in the material properties of the bone model and the model validation under conditions more closely approximating actual accidents were identified as issues for the future.

References

- (1) International Road Traffic and Accident Database (1995)
- (2) European Enhanced Vehicle-safety Committee: "EEVC Working Group 17 Report – Improved test methods to evaluate pedestrian protection afforded by passenger cars" (1998)
- (3) Matsui, Y., Ishikawa, H., Sasaki, A., Kajzer, J., Schroeder, G.: "Impact response and biofidelity of pedestrian legform impactors", IRCOBI (1999)
- (4) Konosu, A., Ishikawa, H., Tanahashi, M.: "Reconsideration of Injury Criteria for Pedestrian Subsystem Legform Test – Problems of RIGID Legform Impactor –", 17th ESV Conference, Paper Number 263 (2001)
- (5) Takahashi, Y., Kikuchi, Y.: "Biofidelity of Test Devices and Validity of Injury Criteria for Evaluating Knee Injuries to Pedestrians", 17th ESV Conference, Paper Number 373 (2001)
- (6) Yang, J.K., Kajzer, J.: "Computer simulation of impact response of the human knee joint in car-pedestrian accidents", SAE paper 922525 (1992)
- (7) Yang, J.K., Rzymkowski, C., Kajzer, J.: "Development and validation of a mathematical breakable leg model", IRCOBI (1993)
- (8) Yang, J.K., Kajzer, J.: "Mathematical model of pedestrian lower extremity", 14th ESV Conference (1994)
- (9) Bermond, F., Ramet, M., Bouquet, R., Cesari, D.: "A finite element model of the pedestrian knee-joint in lateral impact", IRCOBI (1993)
- (10) Bermond, F., Ramet, M., Bouquet, R., Cesari, D.: "A finite element model of the pedestrian leg in lateral impact", 14th ESV Conference (1994)
- (11) Schuster, P. J., Chou, C. C., Prasad, P.: "Development and Validation of a Pedestrian Lower Limb Non-Linear 3-D Finite Element Model", STAPP CAR CRASH JOURNAL, Vol.44 (2000)
- (12) Tamura, A., Miki, K., Yang, K.: "A Tibial Mid-Shaft Injury Mechanism in Frontal Automotive Crashes", 17th ESV Conference, Paper Number 355 (2001)
- (13) Cesari, D., Cavallero, C., Roche, H.: "Mechanisms producing lower extremity injuries in pedestrian accident situations", 33rd AAAM (1989)
- (14) Goldstein, S., Frankenburg, E., Kuhn, J.: "Biomechanics of bone", Accidental Injury (1993)
- (15) Hirsch, C., Evans, F.G.: "Studies on some physical properties of infant compact bone", Acta Orthop. Scand., 35 (1970)
- (16) Sedlin, E.D., Hirsch, C.: "Factors affecting the determination of the physical properties of femoral cortical bone", Acta Orthop. Scand., 37 (1966)
- (17) Lindahl, O., Lindgren, A.G.H.: "Cortical bone in man. II. Variation in tensile strength with age and sex", Acta Orthop. Scand., 38 (1967)
- (18) Hirsch, C., Dasilva, O.: "The effect of orientation on some mechanical properties of femoral cortical specimens", Acta Orthop. Scand., 38 (1967)
- (19) Carter, D.R., Caler, W.E., Spengler, D.M., Frankel, V.H.: "Fatigue behaviour of adult cortical bone: the influence of mean strain and strain range", Acta Orthop. Scand., 52 (1981)

- (20) Evans, F.G., Bang, S.: "Differences and relationships between the physical properties and the microscopic structure of human femoral, tibial and fibular cortical bone", *Am. J. Anat.*, 120 (1967)
- (21) Wall, J.C., Chatterji, S.K., Jeffery J.W.: "Age-related changes in the density and tensile strength of human femoral cortical bone", *Calcif. Tiss. Intl.*, 27 (1979)
- (22) Carter, D.R., Spengler, D.M.: "Basic science and pathology - Mechanical properties and composition of cortical bone", *Clin. Orthop. Related Research*, 163 (1982)
- (23) Lappi, V.G., King, M.S., LeMay, I., "Determination of elastic constants for human femurs", *Journal of Biomechanical Engineering*, 101 (1979)
- (24) Hight, T.K., Brandeau, J.F.: "Mathematical modeling of the stress strain - strain rate behavior of bone using the Ramberg-Osgood Equation", *Journal of Biomechanics*, 16 (1983)
- (25) Cezayirlioglu, H., Bahniuk, E., Davy D.T., et al.: "Anisotropic yield behaviour of bone under combined axial force and torque", *Journal of Biomechanics*, 18 (1985)
- (26) Bonfield, W.: "Advances in the fracture mechanics of cortical bone", *Journal of Mechanics*, 20 (1987)
- (27) Katsamanis, F., Raftopoulos, D.D.: "Determination of mechanical properties of human femoral cortical bone by the Hopkinson bar stress technique", *Journal of Biomechanics*, 23 (1990)
- (28) Lotz, J.C., Gerhart, T.N., Hayes, W.C.: "Mechanical properties of metaphyseal bone in the proximal femur", *Journal of Biomechanics*, 24 (1991)
- (29) Saha, S., Hayes, W.C.: "Tensile impact properties of human compact bone", *Journal of Biomechanics*, 9 (1976)
- (30) Yoon, H.S., Katz, J.L.: "Ultrasonic wave propagation in human cortical bone-II, Measurements of elastic properties and microhardness", *Journal of Biomechanics*, 9 (1976)
- (31) Reilly, D.T., Burstein, A.H.: "The Mechanical properties of cortical bone", *Journal of Bone and Joint Surgery*, 56-A (1974)
- (32) Currey, J.D., Butler, G.: "The Mechanical properties of bone tissue in children", *Journal of Bone and Joint Surgery*, 57-A (1975)
- (33) Burstein, A.H., Reilly, D.T., Martens, M.: "Aging of bone tissue: Mechanical properties", *Journal of Bone and Surgery*, 58A (1976)
- (34) McCalden, R.W. et al.: "Age-related changes in the tensile properties of cortical bone", *Journal of Bone and Joint Surgery*, 75-A (1993)
- (35) Mather, B.S.: "The symmetry of the mechanical properties of the human femur", *J. Surg. Res.*, 7 (1967)
- (36) Ruff, C.B., Hayes, W.C.: "Sex differences in age-related remodeling of the femur and tibia", *Journal of Orthopaedic Research*, 6 (1988)
- (37) Goldstein, S.A. et al.: "The mechanical properties of human tibial trabecular bone as a function of metaphyseal location", *Journal of Biomechanics*, 16 (1983)
- (38) Currey, J.D.: "The mechanical properties of bone", *Clin. Orthop. Related Research*, 135 (1978)
- (39) Leichter, I. et al.: "The relationship between bone density, mineral content and mechanical strength in the femoral neck", *Clin. Orthop. Related Research*, 163 (1982)
- (40) Parfitt, A.M. et al.: "Relationships between surface, volume, and thickness of iliac trabecular bone in aging and in osteoporosis", *J. Clin. Invest.*, 72 (1983)
- (41) Williams, J.L., Lewis, J.L.: "Properties and an anisotropic model of cancellous bone from the proximal tibia epiphysis", *Journal of Biomechanical Engineering*, 104 (1982)
- (42) Stone, J.L., Beaupre, G.S., Hayes, W.C.: "Multiaxial strength characteristics of trabecular bone", *Journal of Biomechanics*, 16 (1983)
- (43) Gibson, L.J.: "The Mechanical behavior of cancellous bone", *Journal of Biomechanics*, 18 (1985)
- (44) Linde, F., Hvid, I.: "Stiffness behaviour of trabecular bone specimens", *Journal of Biomechanics*, 20 (1987)
- (45) Rice, J.C., Cowin, S.C., Bowman, J.A.: "On the dependence of the elasticity and strength of cancellous bone on apparent density", *Journal of Biomechanics*, 21 (1988)
- (46) Pugh, J.W., Rose, R.M., Radin, E.L.: "Elastic and viscoelastic properties of trabecular bone, dependence on structure", *Journal of Biomechanics*, 6 (1973)
- (47) Behrens, J.C., Walker, P.S., Shoji, H.: "Variations in strength and structure of cancellous bone at the knee", *Journal of Biomechanics*, 7 (1974)
- (48) Ducheyne, P., Heymans, L., Martens, M., et al.: "The mechanical behavior of intercondylar cancellous bone of the femur at different loading rates", *Journal of Biomechanics*, 10 (1977)
- (49) Linde, F., Hvid, I.: "The effect of constraint on the mechanical behavior of trabecular bone specimens", *Journal of Biomechanics*, 22 (1989)
- (50) Courtney, A.C. et al.: "Age-related reductions in the strength of the femur tested in a fall-loading configuration", *Journal of Bone and Joint Surgery*, A77 (1995)
- (51) McElhaney, J.H.: "Dynamic response of bone and muscle tissue", *Journal of Applied Physiology*, 21 (1966)
- (52) An, Y.H., Draughn, R.A.: *Mechanical Testing of Bone and the Bone-Implant Interface*, CRC Press, 2000, p.68
- (53) Lindahl, O.: "Mechanical properties of dried defatted spongy bone", *Acta Orthop. Scand.*, 47 (1976)
- (54) Carter, D.R., Hayes, W.C.: "The compressive behaviour of bone as a two-phase porous structure", *Journal of Bone and Joint Surgery*, 59-A (1977)
- (55) Carter, D.R., Hayes, W.C.: "Bone compressive strength: The influence of density and strain rate", *Science*, 194 (1976)
- (56) H-Dummy™ Version 1.6 User's Manual, Hankook ESI, Engineering Systems International, 1998
- (57) Viewpoint Premier CATALOG, 10th Anniversary Edition, p. 81
- (58) Mori, O., et al.: *Anatomy 1*, Tokyo, Kanehara & Co., LTD., 1998, p. 24
- (59) Pam System International, PAM-CRASH™ PAM-SAFE™ Version 1998 Notes Manual, 1998
- (60) Pam System International, PAM-CRASH™ PAM-SAFE™ Version 1998 Reference Manual, 1998
- (61) Yamada, H.: *Strength of biological materials*, The Williams & Wilkins Company, 1970

- (62) Kress, A. et al.: "Fracture patterns of human cadaver long bone", IRCOBI (1990)
- (63) Evans, F.G.: Mechanical properties of bone, Charles C Thomas Publisher

■ Authors ■



Yuji KIKUCHI



Yukou TAKAHASHI



Atsuhiro KONOSU



Hirotoshi ISHIKAWA