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Development and Validation a Finite Element Model of the Thigh of a Chinese 50th Percentile Male

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Abstract:

Background: Currently, crash dummies and mathematical model of human were mainly developed based on the European human statistical data. The prediction accuracy for Chinese was remained to be studied.

Objective: The current study aims to explore the modeling methods of Chinese 50th percentile human body.

Method and material: Geometry model based on the CT data from a Chinese male volunteer close to the anthropometry of 50th percentile was scaled to the 50th percentile male model according to Chinese human body statistics. The rate dependent and different property in compression and tension of cortical bone was simulated with Ls-Dyna material model. Trabecular of femoral head was divided into three parts with different material parameters to imitate the regional variations of it. The femoral head was also validated by compression loading.

Results: The result shows that the proposed model has higher accuracy and biological fidelity.

Conclusions: It is believed that the proposed method, modeling accuracy, geometry scaling and fully verified could be used to develop accurate 50th Chinese human body Finite Element (FE) model.

Keywords: Finite Element Model; Chinese 50th Percentile; Modeling Method; Thigh;

1 Introduction

Recently, with the rapid increase in economic and computational power, crash dummies and Finite Element Human Model are increasingly used in vehicle safety research [1]. Most of the dummy models are designed based on the human body of the Europe and other developed countries. However, a large number of simulation and experimental studies have shown that different regions have a different response. Only in human dimension, China 50th percentile body's physical size is about U.S. 10th percentile human body size [2]. If we used the foreign dummies and Finite Element Human Model to evaluate Chinese human injuries in crashing is clearly inaccurate, due to physical difference between Chinese and foreign body. The results can not effectively evaluate the degree injury of Chinese human body .What's more, it is not conductive to improve the capability of Chinese designing automotive safety.

The objective of this study is to explore the modeling method of Chinese 50th percentile human body. In this paper we mainly establish the thigh finite element model as an example. The geometry directly reconstructed from the computed tomography (CT) scan data of a close to the anthropometry of 50th percentile male volunteer. The model includes accurate representations of the cortical and trabecular bone layers, flesh and skin. The material properties were rigorously defined based on the appropriate literature data [3][4].

2 Method

2.1 Geometry

The geometry reconstruction of the lower extremities was conducted by the Department of Radiology, The Third Xiangya Hospital of Central South University. A male volunteer with anthropometric characteristics (173.1 cm height and 69.7 kg weight) close to the 50th percentile male (170.8 cm/65 kg) was recruited. The geometries of the bony structures and soft tissues of the volunteer thigh region were reconstructed using the CT scanned images. Examples of these CT scans are shown in Figure 1 for thigh sections [4]. The scans obtained were manually processed using Mimics software so that the boundary of the structure could be easily recognized and three-dimensional surfaces were generated. Then the surface data was read into 3D structure design software for geometry repairing. The latest data was read into an ICEM CFD software (ICEM CFD 15.0, ANSYS Inc., USA) for mesh generation [5]. This software is efficient for meshing models with complex geometries by using the fine and high quality hex elements of the models.

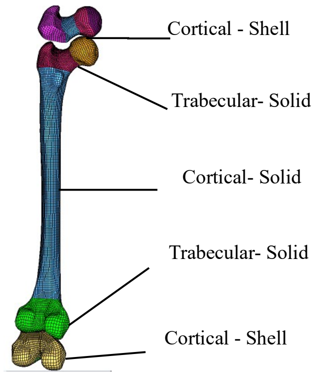


Figure 1. Example of thigh CT. Figure 2. Structure of bone models.

Two meshing approaches, structural and un-structural, were employed to achieve a good quality mesh. The structural mesh technique consists of filling the solid object with cubic blocks and projecting the outer and inner boundaries to the exterior and interior surfaces of the object ,while un-structural mesh technique were used to repair the poor quality elements by HyperMesh (Altair HyperWorks, Troy, MI). The bone models consist of the cortical and trabecular layers to represent the anatomical structure of the human bones as shown in Figure 2.

The trabecular layers in the epiphyses were modeled using solid elements surrounded by shell elements that represent the surface cortical layers (thickness of 1.0–3.0 mm) in order to keep the simulation time steps at reasonable values considering current computational power. Although the cavity inside the cortical tubular structure in diaphysis is filled with a yellow marrow, it is a fatty tissue and its contribution to high-energy impact response was deemed negligible. The cortical tubular structure was modeled using solid elements. The element size of the bony structures and most of the soft tissues (except flesh) generally ranges from 1.5 mm to 3 mm. The larger element size (~8 mm) was used for flesh, which is mostly responsible for impact energy absorption and transmission [6]. The quality of the FE model for computational reliability and stability was verified by checking the distribution of multiple element geometric parameters such as aspect ratio and element quality. The worst value of the mesh quality parameters was limited to values that would not cause model failure, taking aspect ratio ≤5 and the element quality≥ 0.3 for example.

Unfortunately, the CT data represents a close to the 50th percentile male so the resulting mesh needed to be scaled down to the Chinese 50th percentile based on China normal anthropometric measurements in Table 1 [7].Therefore, the scale factor of femur was 0.978.

Table 1.Femoral parameters scaling

Paramters(mm)	FE model	Chinese 50th	scale factor
Femur length	445.41	433.01	0.972
Femur head diameter	47.10	46.17	0.980
Femoral neck diameter	24.00	23.40	0.975
Femoral shaft diameter	26.90	26.29	0.977
Femoral condyle width	81.20	79.90	0.983
Femoral condyle length	62.5	61.40	0.982

2.2Material Property

Bone is a non-linear, viscoelastic, anisotropic and inhomogeneous material [8]. It consists of two types of material: cortical and trabecular bone. Cortical bone is the external part, organized in cylindrical shaped elements called osteons composed of concentric lamellae. Variations in the mechanical properties of cortical bone can be influenced by the number, orientation and size of the osteons. Similarly, the structure of individual lamellae changes orientation in different regions of the bone. This variation gives the bone its orthotropic material properties and allows the bone to respond to optimising the usual load paths through the structure [8].

Trabecular bone is quite porous and is organized in trabecules oriented according to the direction of the physiological load. Trabecular bone is a highly anisotropic structure composed of a large number of interconnected rods, plates and beams. Moreover, the configuration of the trabecular structures is highly variable and depends on the anatomical site. The role of trabecular bone is to absorb and dissipate energy much like structural foam but it contributes relatively little to the overall strength of the bone [9].

A great deal of information can be found in the literature about mechanical and ultimate properties for cortical bones. Goldstein et al. (1993) [10] summarised many different research projects that were conducted to investigate femoral tensile, compressive and shear cortical properties according to differences in age, sex , dryness and direction of load. The values chosen are shown in Table 2.

The trabecular tissue for the femur bone has not been studied as intensely as the cortical so that the values of Young's moduli and ultimate strengths in different directions are not available.

Table 2. LSDYNA bone material properties used in the model

Bone	Density	Modulus of elasticity	Yield stress	References
Femoral head cortical 1	1800	13.5	115	
Femoral head cortical 2	1800	6	115	Yamada
Femoral head cortical 3	1800	2	115	（1970）[15]
Femoral condyle cortical	1800	13	115	Ehler et al
Femur shaft 1	1900	17	115	（ 1970 ）
Femur head Trabecular 1	1500	0.7	9.3	[16][17]
Femur head Trabecular 2	1500	0.4	9.3	Untaroiu
Femur head Trabecular 3	1500	0.2	9.3	[18]
Femoral condyle trabecular	1500	0.7	9.3	

According to literature data, the femoral head subchondral bone in the weight-bearing region has an elastic modulus of approximately 1.5 GPa, which is lower than the corresponding value in the shaft region (6–21 GPa[12][13][14]). In addition, the elastic modulus of the cortical shell in the femoral neck region is 24% less than the shaft region. Based on these data reported in literature, the cortical bone layer of the proximal femur was divided into three small subcomponents: Femoral head cortical 1,2,3 with elastic moduli of 2, 6 and 13.5 GPa, respectively (Fig.3a). The yield stress of each respective component in the proximal femur region was scaled proportionally to its corresponding elastic modulus (13.5 GPa) and the assumed yield stress in the femoral shaft (115 MPa). Similar to the cortical bone, the stiffness and strength of femoral head trabecular bone decreases from the femoral head to the shaft region (Fig. 3b). Thus, the elastic modulus of femoral head trabecular bone was individually assigned to the Femur head Trabecular 1,2,3 with values of 0.7, 0.4, and 0.2GPa.

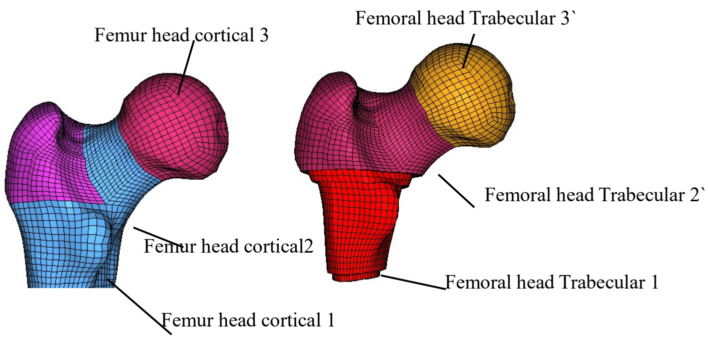


Figure 3. Proximal femur material properties of (a) cortical bone and (b) trabecular bone.

2.3 Model Validation

The thigh model was validated by comparing the FE results with quasi-static and dynamic tests in Table 3.

2.3.1 three-point bending Validation

The femur models were validated against dynamic 3-point bending tests and quasi-static for the femur in the lateral-medial direction.

Table 3. Model validations

FE model	Load type	velocity	Location/direction	References
Femur	Three-point bending (QS)	0.1m/s	Mid-shaft/AP	Yamada.（1970）[15]
Femoral shaft	Three-point bending (D)	1.2m/s	Mid-shaft/LM	Kerrigan et al.（2003）[18]
Femoral head	Compression (QS)	0.5mm/s	Head/ stance configurations	Keyaket al.[19][20]
Thigh	Three-point bending (D)	1.0m/s	Mid /LM	Kerrigan et al.（2003）[18]

AP: anterior-posterior; LM: lateral- medial; D: dynamic; QS: quasi-static.

Yamada (1970) [15]and Ehler et al（1970）[16][17]conducted three-point bending tests on wet femurs obtained from human cadavers of 20 to 39 years of age, as shown in Figure 4. To simulate this series of experiments, the femur was along its posterior surface and the center of the anterior surface of the femoral shaft was gradually compressed at a constant velocity of 0.1 m/s to failure.

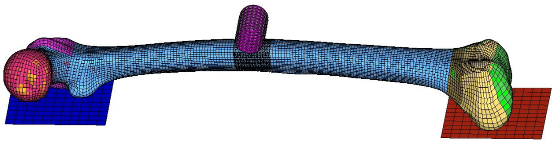


Figure 4: Simulation set-up for three-point bending test of the femur.

The femur was potted in potting cups with rollers at their distal and proximal ends. The roller part of the potting cup was placed on the support plate, and the support load was measured by a load cell placed underneath the plate. The ram had a circular tip and was rigidly attached to a servo-hydraulic test machine. The rigid ram was used for femur bending. The ram was displacement controlled at approximately 1.2 m/s, and displacement transducers were used to monitor the force-deflection response.

It is known that bones are strongest in compression, weaker in tension, and weakest in shear[21]. Therefore, long bones under bending loading usually break in tension and then, the fracture propagates in shear. While in several loading conditions femoral shaft fractures were observed in compression (due to the material model of cortical bone with similar tension and compression material properties) To avoid this situation, the element of the femur closeing to the impactor entity does not define failure.

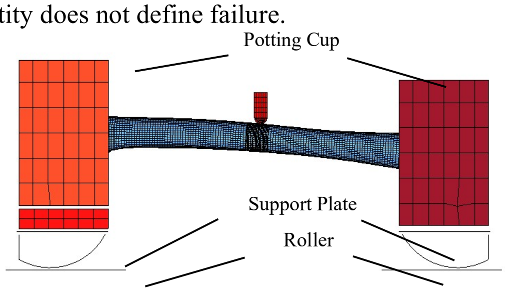
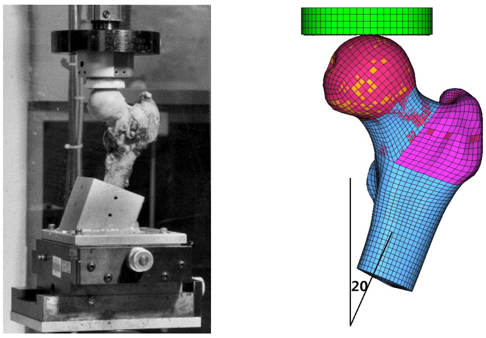


Figure 5.Schematic diagram of femur mid-shaft 3-point bending model.

2.3.2. Femoral head validation

The femoral head test was used to validate the material properties of the bone material model of the femur particularly in the area of the femoral head as well as the failure mechanism in the proximal femur. Since the physical experiments are tests to failure, the model should be able to correctly replicate the fracture locations and ultimate load experienced in the physical experiment at the initiation of fracture.

Keyak et al. (1998) conducted stance-like loading test on the femoral head and neck obtained from 10 female and eight male cadavers with an average age of 70.3 years and an age range of 52 to 92 years. One femur from each cadaver was randomly selected for investigation in a stance-like loading configuration. Force was applied to the femoral head and directed within the coronal plane at 20 degrees to the femoral shaft axis while the femoral shaft was fixed as shown in Figure 6(a). The femoral head and neck were compressed at a loading rate of 0.5 mm/s until failure. The simulation setup for the stance-like loading to the femoral head is shown in Figure 6(b). In the simulation, the femur posture was set to be the same as that used in the experiment.



(a) Experimental setup (Keyak et al. 1998) (b) Simulation setup Figure 6: Stance-like loading to the femoral head

2.3.3 Thigh validation

Skin and muscle are viscoelastic material. In the collision, the skin and muscle mainly in compression. There was little information about muscle material, it was modelled using the viscoelastic material with solid element. The skin was used elastic-plastic material to simulate with thickness1mm, setting the modulus of elasticity of the skin 1 MPa , using the nodes overlap to connect the muscle and bone. The thigh model validation was same with femur shaft validation in Figure 7.

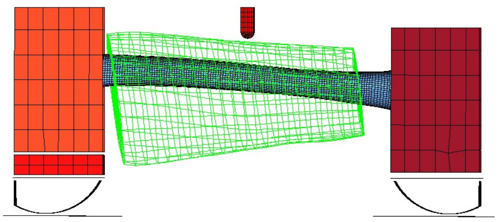


Figure 7. The FE model setup of thigh combined loading validation

3 Results

In quasi-static three-point bending test of the femur, the force–deflection response predicted by the femur model was compared with Yamada (1970) and Ehler et al（1970）. The response of the model was nearly linear and close to the corresponding test data (Figure 8) In the simulation, a bone fracture was predicted in the femoral mid-shaft at about a 3.82 kN, which was within the range of the experimental data (2.4–4.2 kN).

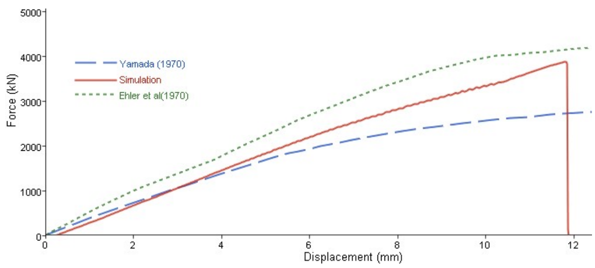


Figure 8: Force-displacement curves of quasi-static femur mid-shaft 3-point bending.

In dynamic three-point bending test of the femur, the force-deflection and moment-deflection response predicted by the femur model was compared with Kerrigan et al.（2003）The global response of the model close to the corresponding test data (Figure 9,10). The simulation results exhibited a good match with the experimental results, suggesting that the bone models can accurately reproduce the dynamic stiffness and failure properties of the actual bones.

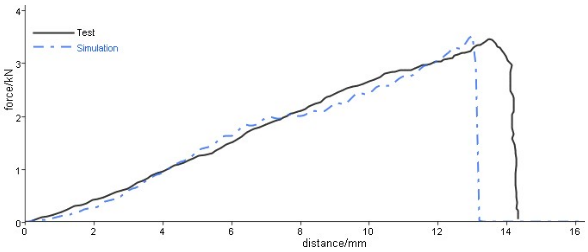


Figure 9: Force-displacement curves of dynamic femur mid-shaft 3-point bending.

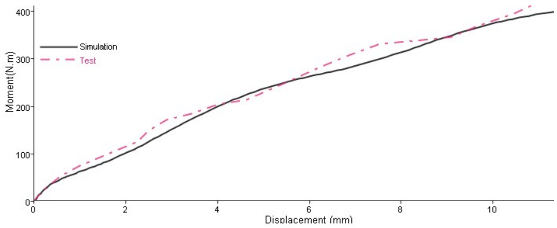


Figure 10: Moment-displacement curves of dynamic femur mid-shaft 3-point bending.

The fracture force of the femoral head and neck due to stance-like loading predicted by the model was 7.1 kN. On the other hand, the fracture force obtained experimentally from 18 specimens ranged from 3.1 to 15.0 kN with an average of 8.4±3.0 kN (Keyak et al. 1998) Figure 11.The fracture force predicted by the model was slightly lower than the average fracture force obtained experimentally but still within one standard deviation.

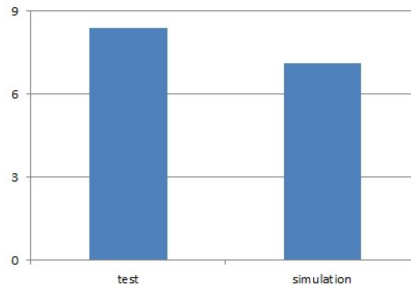


Figure 11: Fracture forces distribution in stance-like loading of the femoral head

In dynamic three-point bending test of the thigh, the force-deflection and moment-deflection response predicted by the thigh model was compared with Kerrigan et al.（2003） .The global response of the model close to the corresponding test data (Figure 12,13). The simulation results exhibited the force-deflection was slightly higher than the experiment at the beginning but still within one standard deviation. The moment-deflection had a good match with the experimental results.

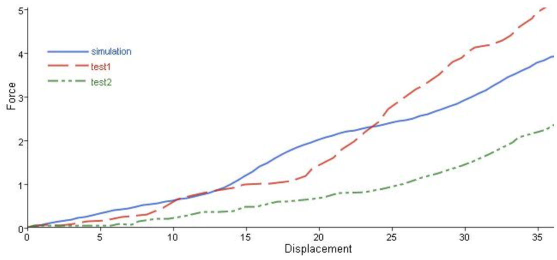


Figure 12: Force-displacement curves of dynamic thigh mid-shaft 3-point bending.

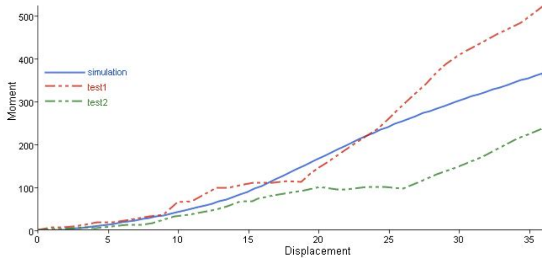


Figure 13: Moment-displacement curves of dynamic thigh mid-shaft 3-point bending.

4 Discussions

As shown in the previous sections, the FE model of the 50th percentile male replicates the forces and moment observed in the component-level tests. The FE simulation result was within the range of the physical test results although generally toward the high side of the test corridors.

As shown earlier in Figure 9,10,12 and 13, the simulation results showed that force-deflection and moment-deflection response of the femur and thigh in dynamic 3-point bending tests had a similar trending with the physical tests.

In Figure 11. The fracture force of the femur head predicted by the model was slightly lower than the average fracture force obtained experimentally but still within one standard deviation. This indicates that the FE model provides a reasonably good estimate of the location of initial fractures that will occur in crash loadings of the thigh.

As with any numerical model of a physical event there are improvements and refinements that could be made. For example, physical testing would help to provide more accurate orthotropic and rate sensitive properties of the bones. Also, the boundary conditions of the FE model should be accurated.

The most common numerical problem with the FE model involved element negative volume distortions of muscle. This often results in negative volume run terminations since elements made of the muscle material would experience very high distortions as the cortical bone failed.

In general, the FE model provides a good tool for exploring the forces and fracture mechanisms in the thigh since it accurately replicates the forces and failure patterns

5 Conclusions

An FE model of the 50th percentile male thigh was developed for use with the LS-DYNA non-linear dynamic FE software. The model features precise geometrical models of the bones including the actual cross-sectional geometry, non-linear failing material properties for the bones. The model results were compared to physical tests of femoral dynamic 3-point bending and quasi-static tests, femur head and thigh test. The comparisons showed that the FE model correctly replicates the peak forces and reactions for these component-level tests within the range of physical tests. The results of these validation experiments indicate to the authors that the FE model of the thigh will provide accurate and reliable results in exploring other types of thigh loadings that may be difficult to replicate with physical tests. The availability of a FE model provides a valuable analysis tool for understanding injuries to the thigh and designing vehicle interiors that minimise the number and severity of thigh injuries in the field. It is believed that the proposed method, modeling accuracy, geometry scaling, experiments scaling and fully verified, could be used to develop accurate 50th Chinese human body finite element (FE) model.

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