Influence of Material Inhomogeneity and Transverse Isotropy on the Biomechanical Responses of Femur under Bending and Compression

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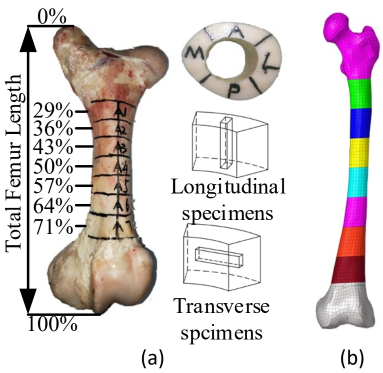
I. INTRODUCTION

Finite element (FE) models of long bones have been widely used, featuring homogeneous material parameters derived from the middle section of the bone [1-2]. However, long bones exhibit significant anisotropy due to their microstructural characteristics [3]. Therefore, the objective of this study was to improve understanding of the long bones' biomechanical properties by using FE models with inhomogeneous and transversely isotropic material constitutive laws.

Vehicle-pedestrian collisions can cause different loading conditions, including three-point bending and cantilever bending, based on the pedestrian's height and the vehicle's front structures. But it is the case that the occupant's skeleton experiences axial compressive loads in frontal and oblique frontal collisions involving vehicle occupants. As such, in order to better study the injury of femur in automobile collision, this study examined the impact of the material constitutive models and parameters on the femur's biomechanical responses in these conditions. The FE models used were homogeneous isotropic, homogeneous transversely isotropic, inhomogeneous isotropic, and inhomogeneous transversely isotropic elastic-plastic materials.

II. METHODS

THUMS V7 FE femur was adopted as the baseline homogeneous, isotropic model. To define inhomogeneity and transverse isotropy, we utilized three fresh bovine femurs obtained from different animals. Each femur was divided into seven sections equally based on normalized length (Fig. 1). Each bone section was further divided into four quadrants: anterior, posterior, medial, and lateral. For each quadrant, two transverse specimens and two longitudinal specimens were prepared. The variation of longitudinal and transverse mechanical properties of each section relative to the longitudinal mechanical properties of the middle shaft was described using the k ratio (Fig. 2). The k value is a proportional coefficient that represents the ratio dividing the material parameter of the specimens taken at a specific location and orientation by the material parameter of the longitudinal specimen of the mid-shaft section. The observed variation of elastic moduli of bovine femur from three-point bending testing is consistent with the findings of literature using ultrasound method to test human femurs, showing largest elastic moduli in the mid-shaft region [3,4].



Finite element models						
Normalized femur length	Young's Modulus(MPa)		Yield Stress(MPa)		Tangent Modulus(MPa)	
	Longitudinal specimens	Transverse spcimens	Longitudinal specimens	Transverse spcimens	Longitudinal specimens	Transverse spcimens
29%	14306	8255	24.9	11.88	12900	9148
36%	18106	8424	34.69	13.65	16326	9412
43%	17485	10605	31.44	14.86	15767	14054
50%	17300	9912	34.5	16.22	15600	12341
57%	15931	9745	29.01	13.28	14365	11984
64%	12516	8450	25.98	12.73	11286	10666
71%	10853	7884	17.99	11.99	9787	8975

(c)

Fig. 1. (a) Axial position gradient and bovine specimen preparation method. A total of seven tested regions were marked. (b) FE model of the inhomogeneous human femur with seven sections divided according to tested bovine femur sections. (c) Material parameter of FE model of the human femur.

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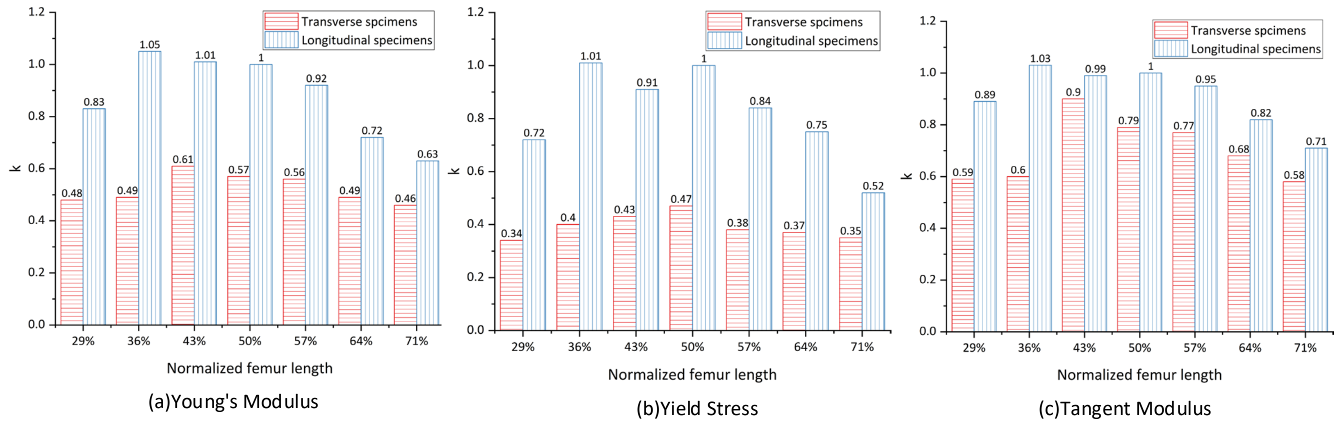

Fig. 2. Axial gradient variation of material parameters for the bovine femur. Mean values were referred to for developing computational simulations.

For the inhomogeneous model, the FE femur shaft model was divided into seven sections, with the mid-shaft having the same material parameters as the THUMS V7 femur model and the parameters of the other six sections (Fig. 1(b)) defined by mapping k values to human femur. For transversely isotropic models, the axial material parameters were the same as the isotropic model for the respective sections, and those in the other two directions were the material parameters in the axial direction of the mid-shaft multiplied by the k (Fig. 1(c)). LS-DYNA Material 24 was used for isotropic materials, while Material 157 was used for transversely isotropic materials. This study conducted 12 simulations, including three loading conditions and four material conditions for each loading. Three loading conditions included three-point bending test, cantilever bending test with the proximal epiphysis fixed, and compression test with a fixed proximal epiphysis and a rigid wall applied at the distal end. The loading rates were 10 mm/s and 5 mm/s for the bending and compression tests, respectively.

III. INITIAL FINDINGS

The maximum loading force at 20 mm displacement using isotropic, seven-section inhomogeneous materials in both three-point bending and cantilever bending decreased by 6.5% and 11.1%, respectively, compared to models using isotropic, homogeneous materials. In the axial compression test, the maximum load at 3 mm displacement decreased by 20.8%. The maximum loading force of the models using transversely isotropic, inhomogeneous materials exhibited the same variation compared to models using transversely isotropic, homogeneous materials. Furthermore, when comparing isotropic and transversely isotropic models that use homogeneous materials, the maximum loading force decreased by 23.2% (three-point bending), 23.0% (cantilever bending) and 28.0% (axial compression), respectively. Using inhomogeneous materials, the maximum load of the anisotropic model is reduced by 22.3% (three-point bending), 22.6% (cantilever bending) and 23.2% (compression) compared to that of the isotropic model.

In the three-point bending condition, the maximum stress of the isotropic material model appeared in the rear of the femur, while the maximum stress of the transversely isotropic material model appears in the front. In the cantilever bending condition, the maximum stress of the isotropic model appeared at the anterior proximal end of the femur, while the maximum stress of the transversely isotropic homogeneous model was generated at the posterior proximal end, and the maximum stress location of the transversely isotropic inhomogeneous model was the middle of the anterior femur (Fig. 3(d)).

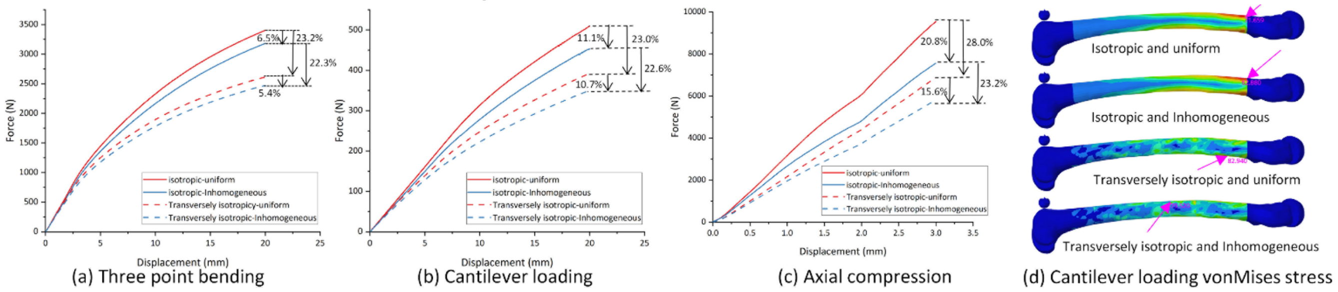

Fig. 3. Comparison of force-displacement curves and vonMises stress.

IV. DISCUSSION

This study highlights the importance of considering the inhomogeneous and transversely isotropic properties of long bones in FE modelling. The effect of transversely isotropic properties was largest, producing differences in force up to 28%. The effect of inhomogeneity was also considerable, producing differences in force up to 20.8%. More importantly, inhomogeneous and transversely isotropic materials lead to changes in the location of maximum stress. The limitations of this study include: (1) deriving material constant relations based on bovine specimens while bovine and human gait models are different; and (2) defining material directions using the global coordinate, which can approximate femur initial axial direction but does not update during bending; (3) defining heterogeneous models using discrete material change modes. Future studies incorporating detailed human femur testing and smooth section interface modeling are highly recommended.

V. REFERENCES

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