

Trabecular Tissue Heterogeneities Determined by Large Scale Finite Element Models

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INTRODUCTION:

The trabecular bone remodeling process takes place on bone surfaces rather than in the middle of trabeculae, which results in higher mineral content in the interstitial bone compared to the surface layers. Thus, trabecular tissue is better defined as a heterogeneous material than a homogeneous one. When performing finite element analysis on trabecular bone, the homogeneous material property assumption has been employed to simplify the modeling procedure and good results were reported from this simplification [1, 2]. Others have concluded that the intra-specimen variations in tissue modulus, if large, may have appreciable effects on trabecular apparent properties [3, 4].

Though previous studies have demonstrated the importance of heterogeneity in modeling trabecular bone, only empirical or plausible correlations are available in the literature to capture the trabecular tissue heterogeneity. Our purpose of this study is to demonstrate how to accurately determine the trabecular bone tissue heterogeneity by relating the tissue modulus to tissue Hounsfield units with either power or linear equation. Our long term objective is to model the failure analysis of trabecular bone where heterogeneity should play a more significant role.

METHODS:

Twelve trabecular bone cubes with 4.5mm size average (min 3.5mm) were excised from medial and lateral femoral condyles of 6-month old New Zealand white rabbits. The largest dimension of each specimen was oriented along the trabeculae direction (anterior-posterior). The specimens were scanned in a micro-CT system (ACTIS 150/225 FFi-HR CT, BIR Inc., Chicago, IL) with 14µm nominal resolution at 82kV and 97mA. After scanning, all specimens were compressed between two parallel platens along the anterior-posterior direction. The platens were polished and lubricated with mineral oil. The specimens were preconditioned with 0.4% strain for 20 cycles. After preconditioning, the load was increased to 5% to guarantee that the yield point was exceeded. The apparent stiffness was calculated from the linear region of the stress vs. strain curves.

Finite element models were constructed from image stacks which were degraded to 56µm voxel size. As the element size was much smaller than half the thickness of a trabecular strut (~170µm), the apparent moduli could be determined with sufficient accuracy [5]. The Otsu method [6] was used to segment the original scans. The degraded scans were forced to have the same volume fraction as the original ones. The tissue modulus for each element was correlated to the Hounsfield units. All elements were assumed to be isotropic and were given a Poisson's ratio of 0.3. Along the anterior-posterior direction as the mechanical test, the boundary condition was set to be frictionless on one surface and a 0.4% strain applied on the opposite surface.

The tissue modulus was related to Hounsfield units with either power or linear equation. The coefficients of the equations were program adjusted following the Powell multi-dimensions minimization algorithm to minimize the following weight function

$$I = \frac{1}{n} \sum_{i=1}^n \left(\left| \frac{(App_{i}^{FEA} \cdot stif_{i}^{FEA} - App_{i}^{MTS} \cdot stif_{i}^{MTS})}{App_{i} \cdot stif_{i}^{MTS}} \right| \right) \quad (1)$$

where n is the sample numbers. A pre-calibration was carried out using 84µm models. From this pre-calibration, two samples having more than 40% FEA-MTS apparent stiffness errors were suspected to have micro damages (average error of others was 7%). These two samples together with another one which was too large for the whole calibration program to run efficiently were eliminated from the calibration step but saved for later evaluation purpose.

RESULTS:

The best-fit power correlation was achieved as (Figure 5)

$$E (GPa) = 16.54 (NHU)^{0.23} \quad (2)$$

where the NHU is the normalized Hounsfield units (HU/10000). The corresponding weight function value I, i.e. the averaged absolute percentage error is 2.86%. The variation of weight function value I vs. power is shown in Figure 3. The R² value for apparent stiffness regression with refined specimens was 0.9894 and 0.836 for all twelve

specimens (Figure 4). The tissue modulus was ranged from 9.84GPa (0.204) to 13.89GPa (0.456). The coefficient of variation (COV standard deviation/mean) is 6.4% (0.57%).

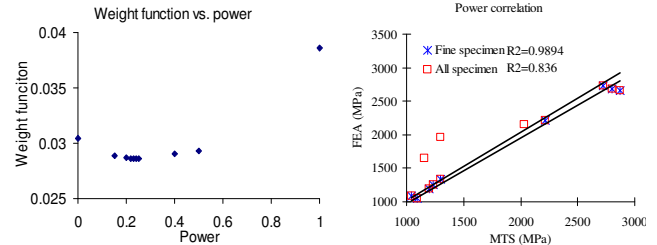


Figure 3. The weight function value I vs. power. Figure 4. The regression R² for refined specimens and all specimens. (Intercept=0)

The calibrated linear correlation equation was (Figure. 5)

$$E (GPa) = 12.1 (NHU) + 8.932 \quad (3)$$

the minimum weight function value I was 2.89% which is a little larger than that of the power function 2.86%. The R² is 0.9891 for refined specimens and 0.8348 for all specimens (Figure 6). The tissue modulus was ranged from 10.2GPa (0.116) to 14.49GPa (0.675). The COV is 6.2% (0.48%).

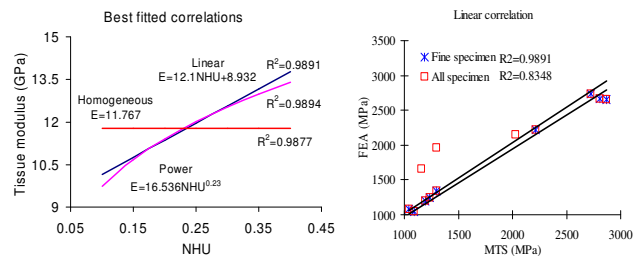


Figure 5. Curves for best fitted linear power and homogeneous correlation. Figure 6. Best fitted power and linear correlations.

DISCUSSION:

We have demonstrated a tissue modulus calibration method which can determine the tissue heterogeneity with a considerable accuracy. It could also be easily applied on other independent variables such as tissue density. In our work, we have shown that the tissue modulus could be expressed in either power or linear correlation with Hounsfield units without significant difference. A number of previous finite element analyses on examining the trabecular tissue modulus have been performed on varieties of specimens regarding different species and anatomic locations. The trabecular tissue modulus from 5.7GPa (1.6) to 17.3GPa (2.62) [7] was achieved using finite element methods. The result from our work was well within the ranges of these previous studies. Although there was limited advantage shown for heterogeneous models over homogeneous models (Figure 5. R²= 98.77%) with respect to linear stiffness analysis, the heterogeneous model should be more accurate for nonlinear failure analysis. Also, this limited advantage was caused by the low material COV. The advantages are supposed to become significant when the material COV is larger than 20% [3].

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