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Heterogeneous Femur Model Reconstruction for Improved FEA Analysis

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Abstract – *The objective of this research is to reconstruct the femur bone as a heterogeneous model and to examine the mechanical response of the proximal femur bone under mechanical loading using finite element analysis. Contrary to setting a uniform Young's modulus for the femur bone during analysis, the femur bone was reconstructed as a layered heterogeneous model with each layer having its own assigned Young's modulus. Computerized Axial Tomography (CT) slices of a fresh frozen femur of a 30 year old male donor were downloaded from (www.bgu.ac.il/~zohary/CT FF.html) which was uploaded by Yosibash et al. [1]. The images were imported into MIMICS 11.02 and used to convert 2-D CT scans into a layered 3-D Stereolithography (STL) image. The proximal femur was divided into ten layers in the longitudinal direction of the proximal femur based on their Hounsfield unit (HU) values. The Hounsfield unit obtained from MIMICS was used to calculate the density of the bone as well as the corresponding Young's modulus. Constant Poisson's ratio of 0.3 was used in the simulation. It is hypothesized that by including the variation in density within the finite element analysis model (FEA), it would be possible to replicate the experimental results obtained by Yosibash et al. [1].* ***Copyright © 2012 Praise Worthy Prize S.r.l. - All rights reserved.***

Keywords*: Femur Bone, Heterogeneous Modeling, Computerized Axial Tomography*

I. Introduction

Combination of FEA and CT scans is of vital importance for the mechanical response of patients in orthopedics and biomechanics. According to Yosibash et al. [1],[2], the simulation of a patient's femur bone response to mechanical load is limited due to complex geometry and anisotropic properties of the bone. Determining the mechanical response of a patient when subjected to different loads on the femur bone using FEA will help the orthopedists in treating their patients.

Shim et al. [3] investigated the use of sparse CT datasets for auto-generating accurate FEA models of the femur and pelvis. The authors used high order cubic Hermite basic functions for mesh generation and least-square fits to the data set, and this procedure reduced the data for automesh generation significantly. In a related study, Viceconti et al. [4] analyzed four different methods, namely mapped mesh, tetra mesh, voxel mesh, and hexa mesh, for automatic mesh generation of human femur. They concluded that tetra mesh is the best method when a solid model is available while hexa mesh produced the most accurate results, but required significant amount of computational resources. Furthermore, Monetti et al. [5] analyzed the structure of human femur for prediction of biomechanical strength in vitro. The authors concluded that 3-D nonlinear structural anisotropic texture which was measured from High Resolution Magnetic Resonance Images (HRMRI) can successfully predict the mechanical properties of human femur. In a related study, Wirtz et al. [6] evaluated the bone material properties in order to determine the anisotropic FE-simulation of the proximal femur. They concluded that constructing an FEA based simulation model of bony structures, the apparent density-dependent characterization was the best way to analyze individual and local bone variation on age.

Also, Bianchi et al. [7] investigated the quantitative and comparative stress analysis of human femur under two different static situations using 3-D photoelasticity. It was found that both the maximum tensile and compressive stress were in the region below the subtrochanter. On the other hand, the tensile stress did not vary along the borders of diaphysis. Again, Perilli et al. [8] determined the compressive strength on local variations in microarchitecture in cancellous bone of the proximal femur. The authors concluded that because of the heterogeneity of cancellous bone, the region where the bone is weaker has a higher probability of failure.

Furthermore, Birnbaum et al. [9] investigated the material properties of trabecular bone structures. They found that horizontally orientated trabecular was only 30% stable when compared to vertically oriented trabecular structure. This implies that the vertical trabecular structure has an advantage over the horizontal trabecular bone for reconstruction of the acetabulum. Also, Yang et al. [10] examined the anisotropic application of Hooke's law to cancellous bone and wood.

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According to their study, the cancellous bone was dependent on solid volume fraction of orthotropic Hooke's law while hardwoods and softwoods are dependent on bulk density. Brown and Ferguson [11] studied the mechanical property distribution in the cancellous bone of human proximal fermur. The authors found that the stiffness and yield strength values at any given point in the proximal fermur were proportional to one another. They tested the cancellous in the primary trabeculation system in the longitudinal compression and obtained a stiffness value that was 160 to 400 percent above the overall proximal fermur average. Contrarily, the apparent stiffness of this same primary system when tested in the transverse section resulted in only about 50 to 70 percent of the value observed for longitudinal test.

Verhulp et al. [12] investigated the load distribution in the healthy and osteoporotic human proximal fermur during a fall to the side using micro-finite element models. The objective of their study was to determine the contribution of trabecular bone to the strength of the proximal fermur for nonhabitual load. They calculated the load distribution and estimated the yield and ultimate loads from linear analyses. According to their findings, the distributions of the maximal principal strain and effective strain in the entire model suggested that the contributions to bone strength of the trabecular and cortical structures were similar.

In a related study conducted by Lotz et al. [13], it was found that the percentage of total load supported by cortical and trabecular was approximately constant for all loads, but differed in loading location. The cortical bone carried 30% of the load at the subcapital region, 50% at the mid-neck, 96% at the base of the neck, and 80% at the intertrochanteric region. This was in contrast to the assumption that that cortical bone carries 75% of the load in the femoral neck and 50% of the load at the intertrochanteric region. The authors also found that during gait, the principal stresses concentrated within the primary compressive system of trabeculae and in the cortical bone of the intertrochanteric region. During a fall, the trabecular stresses concentrated within the primary tensile system of trabeculae. The distribution of stress for the osteoporotic femur was similar to the normal. Keyak et al. [14] examined a new automated method of generating patient-specific three-dimensional finite element modelling of the bone. This method used digital computed tomographic (CT) scan data to derive the geometry of the bone and estimated its inhomogeneous properties. Stress, strain, and strain energy in a human proximal fermur in vivo was predicted by this method. Three-dimensional loading conditions from literature that corresponded to stance phase of gait were used and maximum principal compressive stresses of 8-23 MPa were computed for the medial femoral neck. Wirtz et al. [15] developrd an orthotropic FE "model femur". They used CT scans of a pair of cadaveric femurs at equal distances of 2mm to construct three-dimensional geometry of the proximal femur. Young's modulus, Poisson's ratio, ultimate strength in pressure, tension and torsion, shear modulus were assigned to each FE of the same "CT-density-characterized" volumetric group. Using that method they developed realistic simulations of the dynamical processes of bone remodeling under different loading conditions.

Yosibash et al. [16] investigated the yield prediction capabilities of finite element (FE) models using four different yield criteria, and the spread in the predicted load between isotropic and orthotropic material models. They used subject-specific high order FE models of two human femurs that were generated from CT scans with inhomogeneous orthotropic or isotropic properties, and loaded at the head by compression force. The authors computed strains and stresses by both the orthotropic and isotropic FE models and these were used to determine the load that predicteds yielding using four different yield criteria: Von Mises, Drucker–Prager, maximum principal stress, and maximum principal strain. They loaded one of the femurs by a simple load until fracture, and the force resulting in yielding was compared with the FE predicted force. The surface average of the maximum principal strain criterion along with the orthotropic FE model best predicted both the yield force and fracture. They concluded that there was a non-negligible influence on the predictions if orthotropic or isotropic material properties are applied to the FE model, and all stress-based investigated yield criteria had a small spread in the predicted failure. Recently, Trabelsi and Yosibash [17] performed quantitative computer tomography (QCT) scans on four fresh-frozen human femurs. They developed high-order FE models depending on empirically determined isotropic properties and on MM-based inhomogeneous orthotropic properties. They found that the strains and displacements computed by FE models having MM-based inhomogeneous orthotropic properties and the FE-results having empirically based isotropic properties and the experimental results are in close agreement.

In this work, a new method in the analysis of femur bone is used. The femur bone is divided into ten layers longitudinally based the difference in the bone density in the longitudinal direction. Then a finite element analysis is performed using ANSYS. This analysis is then compared with Yosibash et al. [1].

II. Methodology

The images were imported into MIMICS 11.02 and used to convert 2-D CT scans into a layered 3-D Stereolithography (STL) image. The proximal femur was divided into ten layers in the longitudinal direction of the proximal femur based on their Hounsfield unit (HU) values. The Hounsfield unit obtained from MIMICS was used to calculate the density of the bone, as well as the Young's modulus. Equation (1) represents the objective function used for characterizing the variation in the HU values across the different sliced images of the proximal femur:

$$Min\left[\frac{|CT_{iz}-CT_{nz}|}{CT_{nz}}\right]\leq\varepsilon \qquad (1)$$

where:
ε : accepted error.

Equation (2) was used to calculate the density in the bone:

$$\rho = 1.067\ HU + 131 \qquad (2)$$

ρ : density of the bone

HU: Hounsfield unit values obtained from MIMICS

Equation (3) was used in calculating the Young's modulus of elasticity from the determined apparent density:

$$E = 0.004\cdot\rho^{2.01} \qquad (3)$$

where:
E: Young's modulus of elasticity.

The proximal femur was divided using MATLAB 2007R into ten layers in the longitudinal direction based on the differences in the density between regions through the femur bone, as shown in Figure 1.

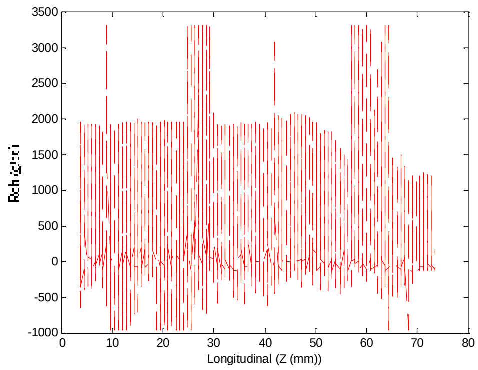

Fig. 1. Density distribution in the longitudinal direction

As illustrated in Fig. 1, the densities are changing along the longitudinal direction. Hence, ten layers are obtained based on the density distribution in the longitudinal direction, as shown in Fig. 2.

Trabecular and cortical regions in the bone have different densities.

Trabecular is a spongy region, and so its density is lower than that of cortical region which is the hard and dense part of the bone. Hence, the first three layers of the bone are divided into two regions: core which has the small density, and the outer region which has the high density.

Figure 3 explains the core region and the outer region of the proximal femur created using PRO/Engineer Wildfire 3.0.

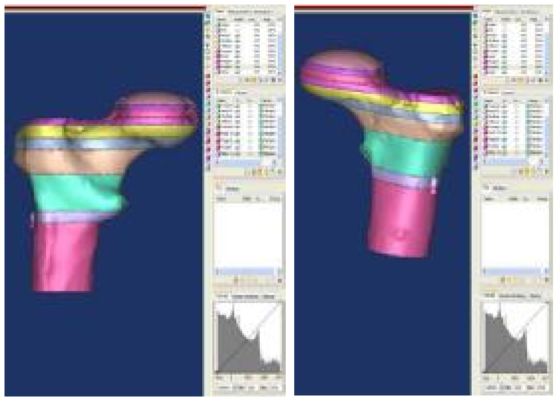

Fig. 2. Layered Heterogeneous Femur Model Reconstructed images from Mimics

The densities of those two regions were obtained using the following procedure:

1. For each layer the average density was obtained. The points in the same layer that have a density more than the average was determined, and their average was calculated. That average was used to calculate the Young's modulus of elasticity for the outer region in the layer.
2. For each layer the average density was obtained. The points in the same layer that have a density less than the average was determined, and their average was calculated. That average was used to calculate the Young's modulus of elasticity for the core region in the layer.

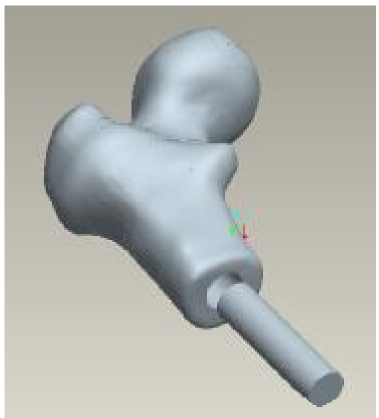

Fig. 3. The outer region and the core region of the proximal femur

To perform FEA on the proximal femur, the 3-D STL model was converted to a CAD model and imported into Pro/Engineering Wildfire as an IGES file. The model was subsequently sliced to 10 sections with each being assigned its respective Young's modulus and a Poisson's ratio of 0.3. The models were meshed in Pro/Mechanica and the loads corresponding to 500, 1000, and 1500 N were applied separately on the proximal femur at 0°, 7°, 15°, and 20° with the axial direction of the proximal femur.

III. Results and Discussion

A surface plot of the CT variation in the bone shows that the femur bone needs to be reconstructed to a minimum of 10 layers to capture the entire CT variation in the vertical z-axis present in the structure as illustrated in Fig. 2.

Stresses, displacements, and strains were obtained using FEA. Fig. 4 is an example of the results that were obtained.

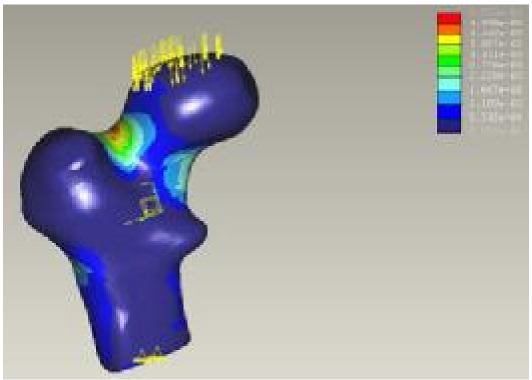

Fig. 4. Strain on the proximal femur loaded by a force of 1500 N at an angle of 15°

Fig. 5 shows that the displacements at the head part of the proximal femur obtained from MIMICS at 15° and 20° were close to those obtained by Yosibash et al. (2007a) while the results at the other angles showed significant differences. For angles 0 and 7, may be more layers are needed to give precise results that are close to those of Yosibash et al. [1]. The error in the head displacement compared to the experimental results is 22%, 58%, 8%, and 4% at the angles, 0, 7, 15, and 20, respectively.

It can be noticed from Figures 6, 7, and 8 that the error in strain between the proposed FEA and the experimental work of Yosibash et al. [1] is maximum in the case of strain at shaft medial. The error reached to around 122%. This error can be minimized by taking more layers at that critical region. On the other hand, the error in strain at neck inferior is minimum. It reached to only 11% which is an acceptable value in this kind of finite element analysis.

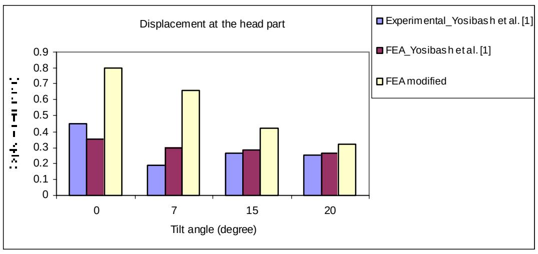

Fig. 5. Comparison of displacement at the head part between proposed FEA at 1500 N and FEA and experimental work of Yosibash et al. [1]

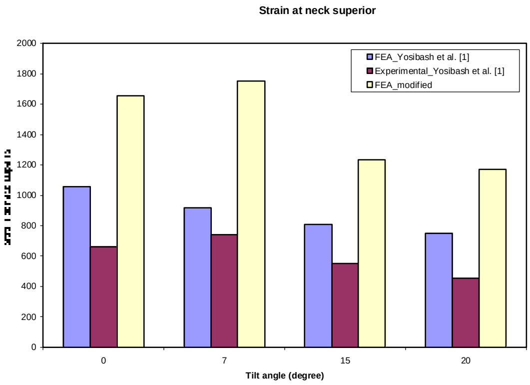

Fig. 6. Comparison of strain at neck superior between proposed FEA at 1500 N and FEA and experimental work of Yosibash et al. [1]

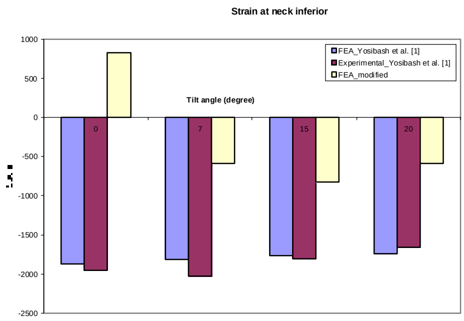

Fig. 7. Comparison of strain at neck inferior between proposed FEA at 1500 N and FEA and experimental work of Yosibash et al. [1]

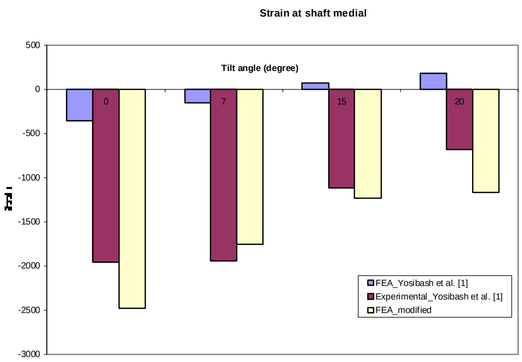

Fig. 8. Comparison of strain at shaft medial between proposed FEA at 1500 N and FEA and experimental work of Yosibash et al. [1]

IV. Conclusion

Ten layers of the proximal femur bone were obtained using the HU values obtained from MIMICS. The first three layers were divided into a core region and an outer region depending on the variation of the values of densities. Promechanica/structure was used for FEA. Displacement and strain of the proximal femur bone at different sections have been obtained. The proposed finite element analysis helped in precise simulation of the loading on the femur bone. The results obtained showed that for angles 0 and 7 more layers are needed to give precise results that are close to those of Yosibash et al. [1]. Precise results have been obtained for strain at neck inferior, where the error between the proposed FEA and the experimental work by Yosibash et al. [1] reached to only 11%.

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