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FEMUR MECHANICAL SIMULATION USING HIGH-ORDER FE ANALYSIS WITH CONTINUOUS MECHANICAL PROPERTIES

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Abstract. *Femur 3D finite-element (FE) analysis based on CT data requires both accurate geometry representation and mechanical properties assignment. A "structure based" method is suggested for the reconstruction of a FE model so that the geometry is represented by smooth surfaces extracted from the CT data including a separating surface between cortical and trabecular regions. Then, p-FE auto-mesh is generated. Within each region (cortical or trabecular) the heterogeneous density is described by continuous spatial functions that are generated using Least Mean Square (LMS) approximations based on the CT scan data. Finally the Young modulus is evaluated continuously according to the functional representation of the density. Numerical investigation shows good results of the suggested method compared to voxel-based method. Mechanical in-vitro experiments on fresh frozen femur measuring head deflection and strains at several points is currently used for FE model calibration and validation.*

The presented study is a first step towards a p-FEM simulation of human femurs based on CT data in order to optimize a protocol for cervical fracture surgery fixation.

1 INTRODUCTION

The use of three dimensional (3-D) finite element analysis (FEA) for orthopaedic application is well accepted for more than three decades [1]. These analyses may provide medical doctors with estimation on the risk for fracture or evaluate the consequence of medical treatment once fracture had occurred for bones in general and for the femur in particular. Although the bone is a complex biological tissue, the use of FEA is attractive because at the macro level it exhibits elastic linear behavior for loads in the normal range of regular daily activities [2]. The proximal femur consists of cortical (dense) and trabecular (cellular) tissues [3]. These constitutes were experimentally investigated to obtain homogenized mechanical properties. In several cases bone specimens Young's modulus were reported as a function of bone apparent density [2,4–9]. Although constitutive models of bones are better represented by an anisotropic elastic relationship, the available information from CT scans are insufficient at the present to determine the numerous material properties of an anisotropic domain and isotropic relationship is assumed. The use of Quantitative Computerized Tomography (QCT) enables the generation of accurate FE models for the human femur based on the standard DICOM format of CT scan output [10–15]. The accurate geometry can be obtained by an analysis of voxel coordinates while the Young modulus may be estimated according to the voxels intensity (given in Housfeld units - HU).

Two methods are commonly used for geometry construction and mesh generation: "voxel based" and "structure based". The "voxel based" method generates a mesh of 8-noded hexahedral elements, each enclosing a cubic predefined volume of the CT scan without any use of surfaces or solid body [1,16–18]. In the "structure based" method, the mesh is constructed on an accurate geometrical model of the bone (usually auto-meshing methods are utilized due to the geometric complexity) [12–15, 19, 20]. Generally speaking, "voxel based" models are more easily generated (via automated custom softwares) and accurate enough to estimate deflections or inside stresses, while "structure based" models are more accurate when surface strains and stresses are of interest [1,15]. In both methods, inhomogeneous mechanical properties are used. Determination of element's Young modulus is according to an averaged value (of evaluated density and empirical correlation to Young's modulus) of the voxels enclosed in the finite element volume [21].

Because the assignment of Young's modulus to a finite element is based on the averaged QCT value, a special attention should be given to the procedure. The averaged HU value is correlated to the bone density using linear relations [2, 5, 22]. This value is then used for Young modulus estimation usually through power-law relations as been widely reported [4–6,8,9,12,23]. It should be noted that a QCT pixel size is of an order of 1 mm, whereas typical specimen's dimension used for density calculation and Young modulus estimation (by experimentation) is of an order of 5 mm. Although studies on the influence of the finite elements size on the results were reported [1,21], these usually consider the computational accuracy versus meshing difficulties and computational time, yet no mechanical or biological justifications is provided for the characteristic size of elements in use. Keyak et al. reported on hexahedral 3 mm cubic elements as a good choice for "voxel based" mesh [16, 24], yet Viceconti et al. show that further refinement can result in an increasing error in FEA results [1]. Decreasing elements size does not lead necessary to convergence since both geometry and element material assignment change from one model to the other. Furthermore, although some studies show correlation between FE model results and fracture load in experiments, to the best of our knowledge only two studies investigate quantitatively the differences between strains computed by FEA and and these measured experimentally on a femur bone [12, 25]. In both studies only partial agreement is found, suggesting the need for further investigation by more reliable simulations.

We obtain an accurate smooth representation of the bone geometry obtained by an analysis of the QCT data, where an internal smooth surface is computed that separates the cortical and trabecular regions. The geometrical description in conjunction with an auto-mesher provide a p-FE mesh containing large p-elements [26]. Within each region (cortical or trabecular) the bone density is evaluated according to a moving average method on HU-values. Thereafter, two different spatial functions are generated based on LMS approximation which describe the density of the bone according to the x, y, z coordinates (using continuous functions). Finally the Young modulus is estimated by a continuous function based on the density, without the need to assign a discrete value for an entire element. The motivation for this approach is that the geometry should be represented as accurately as possible, and the Young modulus should be evaluated according to a similar volume as the test specimens used for its estimation in actual experimentation. Therefore, a moving average method was used so within each voxel the QCT value is computed as an average of the QCT values in the surrounding volume (see [21]). Another point of interest is that mechanical properties are independent on the FE mesh. This requires an accurate numerical solution on bone's surface, so validation against in-vitro experiments (strains measured by strain-gages on bone's surface) will be possible. Although the bone is known to be anisotropic and inhomogeneous, most studies simplify these difficulties by considering an isotropic inhomogeneous material, as we shall follow in our study and concentrate our attention in the current work on a new method for the geometry, mesh representation, and mechanical properties assignment for bones.

The presented work is a preliminary study in the context of a larger project in which a p-FEM simulation of a human femur based on QCT data is investigated and compared to experimental observations, and will be reported in a future publication.

2 GENERATING THE p-FE MODEL BASED ON QCT DATA

2.1 Mesh generation

The suggested method for the construction of the mesh is a "structure based" method, i.e. the geometrical representation of the surfaces are as accurate as possible. The mesh is generated on after a solid body is created having smooth surface representation. The solid body is generated by a commercial CAD software (SolidWorks 2004) according to

IGES surfaces approximated by reverse engineering software (RapidForm) from the CT data (semi-automatic procedure defines exterior surface points for that purpose). This model has two distinct regions for the trabecular and cortical regions, separated by an internal surface. The cortical region is defined using thresholds of the CT HU values and can be defined along the diaphysis and up to a small distance above the lesser trochanter. Due to the geometry complexity of the proximal femur an auto-mesher is used obtaining high-order tetrahedral elements mapped by a sub-parametric function to the standard element.

2.2 Determination of Young modulus

At the current step of our research bone mechanical properties are assumed to be isotropic with inhomogeneous Young modulus and a constant Poisson's ratio [10, 12]. Although transversely isotropic or orthotropic properties were reported [5, 6, 23], the isotropic approach is widely used especially in the trabecular bone where material principal directions are currently difficult to predict using clinical QCT protocols.

The suggested method uses a spatial field, defined by a set of 3D continuous functions, in order to assign bone heterogeneous Young's modulus. The spatial field is evaluated according to the following steps (special procedures created using MatLab 6.5 for steps 1-3):

1. *Data extraction*: A 3D array of HU values is extracted from the CT, DICOM format. To each member of this array we assign the coordinates of the voxel midpoint. Then the apparent density is calculated for each voxel according to [9], or [2] with proper phantoms used.

2. *Moving average*: A new averaged value for apparent density is computed for each voxel according to the voxels located inside a 5mm cube around the voxel of interest (volume size can be defined according to specimen's size used for mechanical properties determination). An array that covers a volume larger than the bone size prevents the side effects, i.e. loss of data close to the surface. False average value is prevented because averaged values are only obtained for voxels that are inside the solid body.

3. *Spatial field approximation*: A 3-D polynomial functions (up to 4th degree) is evaluated based on least mean squares (LMS) so to describe the density as function of x,y,z coordinates.

$$\rho_{app}[g/cm^3] = \sum_{i=0}^{4} \sum_{j=0}^{4} \sum_{k=0}^{4} a_{ijk} x^i y^j z^k \tag{1}$$

   for the femur head other spatial field is used based on spherical coordinate system.

4. *Elastic modulus evaluation*: The Young modulus, represented as a 3-D function, is used for each point within an element when stiffness matrices are computed (2).

2.3 Loads and constrains

Loads and constrains are defined according to the specific experiment being performed. It is quite common to use full constrain on the distal part of a proximal femur model [10, 12, 17]. This constrain is according to the bone mounting in ex-vivo experiments.

2.4 FE solver

A p-FE code StressCheck[1] Version 7.0.2 is used on a PC (Pentium 4 CPU, 2.60 GHz, 1 GB RAM). The p-version solver allows high-order polynomial function spaces which provides exponential convergence rates in energy norm over coarse meshes (see details in [26]). One of the major advantages of p-FEMs over traditional h-FEMs is the ability to monitor the numerical error by inspecting the convergence of the results as the polynomial degree is increased over the mesh. In our studies the p-level for geometry based models (having large elements and accurate boundary representation) is increased from 1 to 5 and the five results at progressively larger degrees of freedom (DOFs) are monitored.

3 THE TWO FE MODELS INVESTIGATED

3.1 A slice of a bone

As a first step the suggested method for the generation of the FE model was investigated numerically on a part of the bone. Three models were created based on a clinical CT scan of a 59 years old female (140 kVp, spiral scan, 2mm slice thick with resolution of 0.863 mm pixel size). A 36 mm slice of the upper diaphysis including the lesser trochanter was chosen (this section contains both trabecular and cortical bone tissues). Figure 1 shows a schematic representation of the femur location at which the slice of interest is analyzed. The first two (3A, 3B) were created according to the "voxel-based" method with different element sizes (3.2 mm and 2.4 mm respectively) and an averaged value of evaluated voxel's Young modulus was assigned [10]. The third model (3C) was created according the new suggested method.

In the elements representing the trabecular bone a heterogeneous Young modulus was assigned according to [6]:

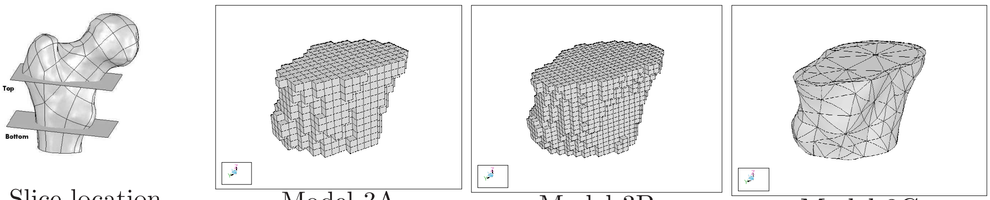
$$E[MPa] = 1904\rho_{app}^{1.64} \tag{2}$$

with a constant value of 0.4 for Poisson's ratio [10]. For the cortical region homogeneous Young modulus ($E = 16GPa$) was used [12].

The bone slice is subjected to two different load cases: one is 700 N axial load distributed uniformly on the upper surface and is clamped at the bottom surface, whereas

[1] StressCheck is trademark of Engineering Software Research and Development, Inc, St. Louis, MO, USA

the other is an axial uniform displacement of the top surface generating a 0.1% strain in the z direction over the whole model.



Slice location Model 3A Model 3B Model 3C

Figure 1: Different models for the bone slice model, A-B using voxel based method, C using spatial field method.

3.2 Proximal femur analysis accompanied by an experiment

A fresh/frozen cadaver bone (30 years old male) passed CT scan (140 kVp, 250 mAs, axial scan, 0.75 mm slice thick and resolution of 0.781 mm pixel size). Strain-gages were positioned at four points of interest (see Fig. 2): two on femur neck (superior and inferior) and two on the femur shaft (lateral and medial). Rosette was in use for the shaft lateral point while uni-axial strain-gages were used otherwise. The bone was mounted within a cylindrical steel base using PMMA, and was inclined in 7° (typical bone position in human body while standing) before subjected to deflection rate of 0.5 mm/min, up to load of 1500N (measurements were taken also for 15°, 20° and vertical posture). A finite element model was generated according to the suggested method based on CT scan data. Trabecular bone Young modulus was assigned according to (2) and cortical bone Young modulus was assigend according to [6]:

$$E[MPa] = 2065\rho_{app}^{3.09} \tag{3}$$

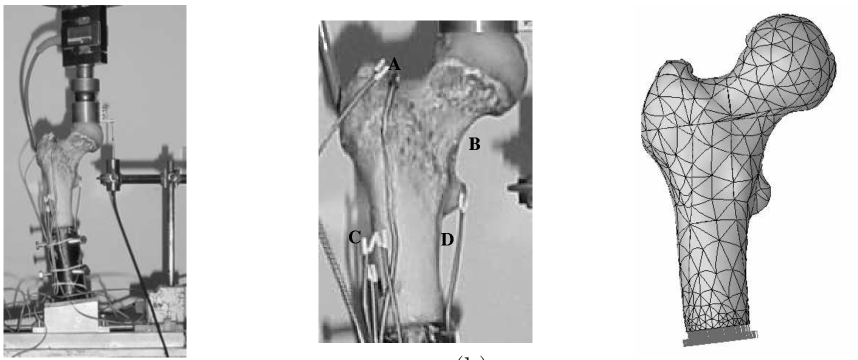
Clamped boundary conditions were applied to the distal part of the model and a displacement in the z direction was applied to the top face of femur head as in the experiment (Fig. 2).

4 RESULTS

4.1 A slice of a bone

Figure 3 presents the Young modulus distribution in the model at the mid-hight plane. This distribution causes differences in the results along an ellipse inside the trabecular region and along a medial-lateral line across the center of the bone as shown in Figure 4.

Results of the three models show qualitative agreement, i.e. same trends are seen in the displacement plots and in the graphs describing the displacement and strains along the



(a) (b) (c)

Figure 2: (a) Fresh/frozen femur under mechanical experiment; (b) Strain-gages positioning on the proximal femur: A - neck superior, B - Neck inferior, C - Shaft lateral, D - Shaft medial; (c) equivalent FE model base on CT scan data

selected curves. Yet, the spatial-field method used in conjunction with the geometrical-realistic model show almost constantly lower displacements and strains compared to the "voxel-based" methods. For example, the maximum displacement (u_z) of model 3C was 0.019 mm comparing to 0.022 mm according to models 3A/B (the overall displacement pattern in the three different models is presented in Figure 5). This trend is also observed in the displacements along the medial-lateral line and in the strain ε_{zz} along the ellipse (Figure 4).

4.2 Proximal femur results

Linear response was observed in all experiments performed on the proximal femur. The strain gages readings demonstrate a linear response with a linear regression fit of $R^2 > 0.999$. Due to large diversity in bone HU data 4 different spatial fields were used in the FE model: one for the cortical and three for the trabecular region (lesser trochanter, greater trochanter and femoral head). That way LMS approximation regression coefficient of $R^2 \geq 0.965$ was achieved when approximating the Young modulus by a continuous function. The FE analysis showed some difference (up to 30%) between the predicted femur head displacement values and those measure experimentally. Strain prediction is much less accurate and can get to more than 100% error.

5 DISCUSSION

The maximum difference of 15% between displacement results in the numerical investigation is large but in the same order of magnitude compared to the modelling and

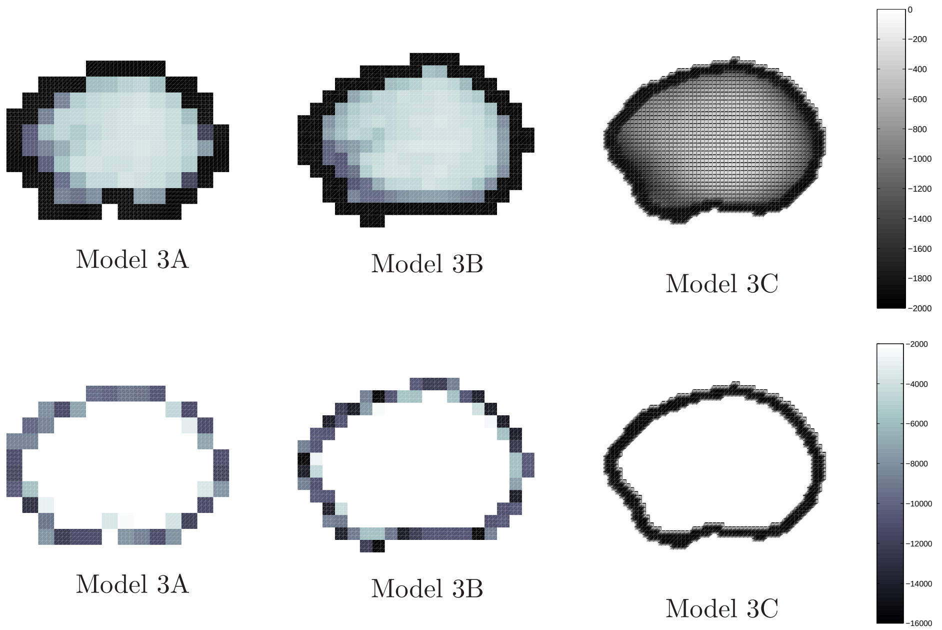


Figure 3: Young modulus distribution across the bone slice at the curves surface, for trabecular and cortical regions (first and second row respectively).

experimental errors - for example, the correlations between QCT data and Young modulus can be less accurate (with $R^2 = 0.60 \div 0.92$) [2, 4, 5, 8, 22]. That comparison allows one to assess the error associated with the ”voxel-method” for which Young modulus is computed by averaging. Elements in two regions have questionable values: these at the trabecular-cortical interface, and these at the outer surface of the bone where part of the element encloses a cortical bone and part is bone-free. In the spatial field method a full separation between cortical and trabecular regions exists, therefore no underestimation of the cortical contribution to bone stiffness is evident. That concept was validated in the results showing smaller displacements and strains in the spatial field model. Underestimation of bone Young's modulus in voxel based methods was described by [27] and considered to be important.

The ability of predicting the mechanical response of the femur depends on the ability to approximate the density values using a continuous functions. The fields that were evaluated herein showed good approximation to the original values with $R^2 \geq 0.965$ $(R^2 > 0.99 inslicemodel(3C))$. In the current research higher polynomial degrees of the LMS approximation lead to ill-conditioned matrices. Therefore, separation between cortical and trabecular regions is essential due to different functional representation as well

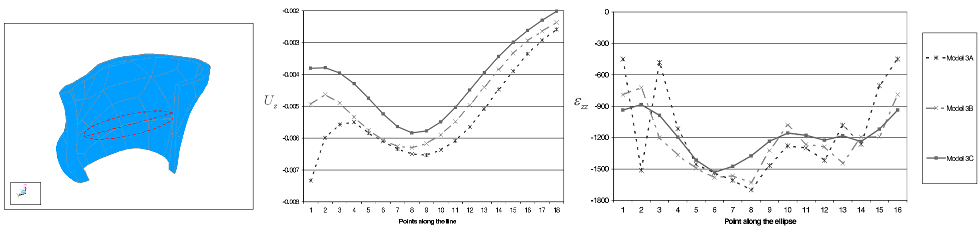
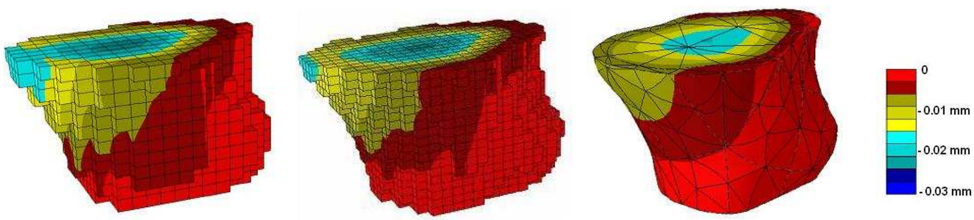


Figure 4: Results along the curves inside the trabecular region at the mid-hight of the slice for the different FE models. Left: axial displacement (u_z) along the medial-lateral line. Right: strain in axial direction (ε_{zz}) along the ellipse.



Model 3A Model 3B Model 3C

Figure 5: Axial displacement (U_z) for the three different bone slice models. Model 3C shows a stiffer behavior.

as for obtaining well-conditioned matrices so to evaluate reliably the unknown coefficients representing the function. Therefore the spatial-field representation cannot be straightforward implemented of other structure based meshing methods such as [15, 20]. Other complications could rise due to high density-gradients in osteoporotic region, yet again several fields can be used, e.g. one for the healthy trabecular region and other for the osteoporotic one.

The differences found between proximal femur analysis and the experimental measurements demands model improvement. That improvement is in progress and a better functions for LMS approximation is sought, using other $\rho - E$ relations including orthotropic material properties. The prediction error reported here is better compared to errors reported in similar studies [12, 25].

Difficulties may be encountered also when trying to generate the mesh in cortical regions that are very thin, as in the grater trochanter, neck and the femoral head. Meshing the thin region with 3D tetra elements is a difficult task, so the effect of neglecting the cortical shell (in the thinner regions) will be examined in the future.

The bone's complicated geometry introduces distorted elements at several points when an auto-mesher was used. This problem had local influence only on the strains (insignif-

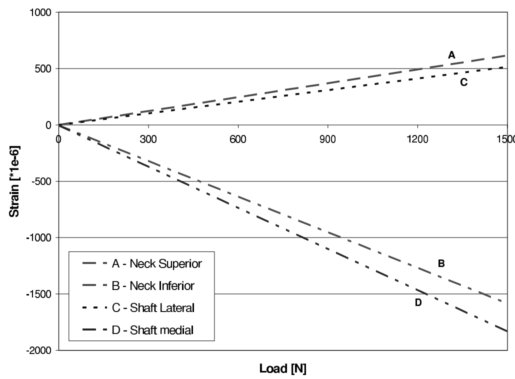


Figure 6: Linear regression for strain vs. force measurements (all linear regression fits show $R^2 > 0.999$).

icant influence on the displacement) at several isolated points only. Although the voxel-based method is widely used, it produces results of lower accuracy, and weak singularities at elements edges and vertices deteriorating the convergence rate and numerical accuracy due to the abrupt change in the Young modulus.

6 SUMMARY

A new method was suggested for a more reliable simulation of the mechanical response of bones based on an accurate representation of bone's geometry, accurate estimation of the trabecular-cortical interface, a spatial description of the inhomogeneous Young modulus and high order FE analysis. This method is developed based on conceptual ideas that the geometry has to be represented accurately and evaluation of mechanical properties should not ignore the experimental methods that were used for the reported correlations between QCT and mechanical properties. The suggested method (using "structure based" mesh) shows better performance compared to the common "voxel-based" mesh especially when realistic femur QCT data in the trabecular region is used for Young modulus determination, and when strains on the surface are needed to be compared to experimental observation (strain-gauges).

Currently the new method is applied for the FE analysis of a proximal femur loaded axially. The strains at the surface are being compared to experimental results performed on the fresh/frozen femur based on which the FE model was created (using QCT scans). This comparison will be reported in a future publication. When orthotropic material properties will be computable from QCT scans the methodology detailed herein could be improved with a minimal effort to provide a more realistic simulation of the mechanical response of the femur.

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