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FINITE ELEMENT MODELING AND ANALYSIS OF TRABECULAR BONE USING THE UNIT-CELL APPROACH

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ABSTRACT
We present alternative approach to modeling and analysis of trabecular bone using unit-cell approach. This approach is based on the Cellular Solids Theory as described by Gibson and Ashby (1997). In this mesostuructural model a macromechanical behavior of trabecular bone is explained by deformation of underlying bony structure. The trabecular structure is discretized using regular space-filling polyhedra. This can be seen as a special form of 3D Voronoi tessellation. With some modifications this method can be used to model structural anisotropy of trabecular bone. With the inclusion of material non-linearities, phenomena such as bone adaptation can be explained, in such a manner that is in good agreement with biological observations.
Keywords: trabecular bone, bone adaptation, cellular solids theory,

1. INTRODUCTION

Long bones (*ossa longa*) are subject of special interest in biomechanical research. These bones consist of two morphologically distinct areas, Figure 1. The outer part of these bones is built of mostly compact substance called the *substantia compacta*, or cortical bone. Inner area of the bone is a fine porous structure called the trabecular bone. While both areas are built of the same material, they exhibit different mechanical properties. These differences are due to bone's unique structure.
Bone's smallest building element is the osteon. The osteons are small rotationally symmetric structures typically 100 – 300 μm in diameter, and up to several millimeters in length (Cowin and Hegedus 1976). Osteons have lamellar structure, Figure 2. Each lamella is 1 – 5 μm thick (Lenz and Nackenhorst 2003), and can be seen as a composite shell consisting of ductile collagen matrix embedded with stiff hydroxylapatite crystals. The lamellae show anisotropic mechanical properties due to specific orientation of hydroxylapatite crystals. The interlamellar space is filled with layer of soft organic material such as bone cells and intramedullary fluid. Hence the bone material can be considered anisotropic and inhomogeneous at the level of single osteon. Mechanical properties of bone at subosteonal and level of single osteon are called micromechanical properties of bone.
Additionally, bone shows different mechanical properties on different levels of material localization. Therefore the multiscale analysis is a natural choice for modeling and analysis of mechanical properties of trabecular bone.

2. MECHANICAL MODELING OF TRABECULAR BONE STRUCTURE

In the course of time many different models and modeling techniques have been proposed to explain complex mechanical behavior of trabecular bone. Some of them are purely phenomenological and rely on many non-physical constants, which have to be obtained experimentally (Turner et al. 1990).

Other mechanistic investigations (Keaveny 2001, Bayraktar and Keaveny 2004) are based on finite element analysis (FEA) of 3D scans obtained by means of micro computed tomography (micro-CT), Figure 3. Since the resolution of modern micro-CT scanners is high, the directly meshed models contain large number of elements. Analysis of such models is obviously computationally very expensive. Besides, the presence of very thin and rough areas in such a model leads to unphysically high stress concentrations.
The third approach, which was used here, can briefly be described the following way. Trabecular bone shows similar morphological and structural properties as engineering cellular materials (solid foams). In this investigation the osteons are modeled as short axisymmetric bodies with varying cross-section across the length.This way, the mean macro-level density of bone tissue can be related to shape of trabeculae. The trabecular structure was discretized through regular tetrakaidecahedra (trucated octahedra), Figure 3. These polyhedra are regular isotropic cells with space-filling properties.
We postulated that due to predominantly axial loading of trabeculae, material anisotropy at subosteonal level can be neglected. Our primary interest was the investigation of bone properties during slow physiological loading. Hence, the actual rate-dependent viscoplastic bone material is approximated as elastic perfectly plastic material.
In case of unit-cells with uniform edges, as shown in Figure 4, the representative volume element (RVE) is a single unit-cell. RVEs can be arranged in either simple cubic lattice (Figure 4.a) or in space-filling arrangement (Figure 4.b). The discretization with uniform space-filling unit-cells can be seen as the special case of Voronoi tessellation (Roberts and Garboczi 2001). Our approach can be slightly modified to model non-uniform trabecular structure. For this purpose, cell-edges with different geometries, corresponding to discrete trabecular volumes must be generated. The number of edges corresponding to particular trabecular volume is function of chosen probability distribution function. These cell edges are then randomly distributed across the RVE. In case of such arrangement a RVE with 4x4x4 unit-cells was used. In similar manner, the damage of trabecular bone can be modeled by removing randomly chosen edges. Note that, the removal of more than 15% of cell-edges can lead to unconnected geometries. Hence highly damaged materials cannot be modeled this way. Damaged trabecular material was simulated using RVE consisting of 8x8x8 unit-cells.
Edges of unit-cells are discretized with solid three dimensional finite elements. This approach is especially appropriate for investigation of plastic yielding localization in trabeculae. However discretization of larger trabecular structures using this approach is computationally expensive. Therefore, cell edges can be discretized by simple three dimensional Timoshenko-beam elements. Their moments of inertia can be calculated from the mean diameter of osteonal models.

2. SIMULATION AND RESULTS

The initial investigation has been performed on single-osteon models. Osteons were loaded over the yield point, both in axial and transversal direction. Quasi-static loading continued until the osteonal cross-section has completely yielded. Both initial yield load and load corresponding to fully yielded cross-section were recorded, and are represented in Figure 5., as a function of osteonal geometric ratio R/r. The plastic straining in osteon is localized in so called plastic hinges. The larger the curvature of single osteon, the smaller the plastic hinge is, and the larger is the plastic strain on the surface.
We postulate that osteocytes (bone cells with mechanosensory function) can perceive inelastic strains. Upon their perception, bone tissue starts complex biochemical process of bone adaptation. Damaged material is removed and replaced with a fresh bone material. This process is localized, and takes place only in certain areas of osteon (Cowin 2004), probably corresponding to plastic hinges. The process of bone adaptation continues until current loads produce elastic strains only. As a result of this process the increase in density of trabecular tissue can be observed. On mesomechanical level, can this be related to increase in volume of single trabeculae. This process is graphically represented in Figure 5b. With small changes in relative density, the bone is able to increase the elastic modulus for almost 100%.
Subsequent investigations were concentrated on the dependence of macromechanical elastic properties of trabecular tissue on its relative density. For this purpose RVEs were loaded uniaxialy, and mean macroscopic elastic moduli and Poisson ratios were recorded. Macromechanical elastic moduli were found to be linearly dependent on the relative density of trabecular tissue, Figure 6. The results are in contrast to those predicted by the theory of Gibson and Ashby. This discrepancy can be explained by

non-uniform geometry of cell-edges. Two mechanisms of deformation are dominant in case, Figure 7. Very porous structures are represented with cell-edges with low geometric ratio (R/r). In this case the exclusive deformation mechanism is the pure bending of cellular edges with narrowly localized yielding. At higher tissue densities, corresponding to thicker cellular edges, non-pure bending is the dominant deformation mechanism, Figure 7.. The tissue density, corresponding to change of underlying deformation mechanism, corresponds roughly to start of quasi-linear segment in Figure 6.

Non-uniform trabecular structure was discretized with Timoshenko beam elements. Moments of inertia of beam elements were calculated from the mean diameter of osteons. The results of FE-simulations were "smeared out" over the volume of RVE. Figure 8. shows the astonishingly good agreement between results obtained with standard solid-element based simulation and those based on Timoshenko beam elements.

Figure 9. shows that macromechanical material properties not only depend on relative density but also on number of each discrete beam cross-section governed by the probability distribution. Although structures generated using Gaussian and uniform distribution of cellular edges' cross-sections are of similar relative density, they show different macroscopic stiffness. This can be related to localized failure occurring at very thin cross sections, which are more present at structures generated using the uniform distribution, and which are virtually absent from those generated by the Gaussian distribution. This can significantly influence the overall macromechanical properties of trabecular tissue.

By randomly removing cell edges from the isotropic virgin cellular structure we were able to simulate the structural damage. For example, the pathological processes such as the osteoporosis can be modeled this way. It was found out that the decrease of mechanical properties associated with the damage is linear in nature, Figure 10.

Figure 1. Typical cross section of a long bone, showing trabecular and cortical tissues (Ströhla 2005)

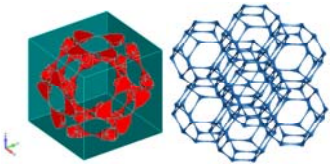


Figure 2. Morphology of trabecular bone (http://en.wikipedia.org/wiki/Image:Illu_compact_spongy_bone.jpg)

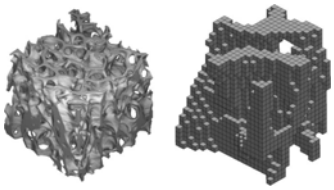
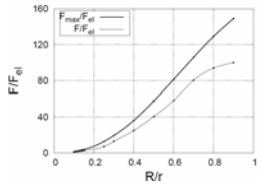


Figure 3. Micro-CT scan and the finite element model of trabecular structure (Keaveny 2004)

Figure 4. FE-model of unit-cell a) arrangement of unit-cells in cubic lattice, b) space-filling arrangement



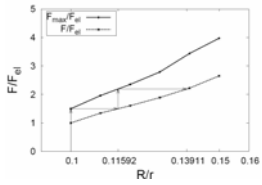


Figure 5. Proposed bone adaptation model. a) Dependence of yielding on trabecular geometric ratio. b) Bone adaptation at small trabecular geometric ratios.

Figure 6. Dependence of relative elastic modulus a) and Poisson ratio b) on relative density of trabecular tissue

Figure 7. Deformation mechanisms of unit-cell edges a) pure bending of cell-edges with highly localized yielding, b) non pure bending of cellular edges.

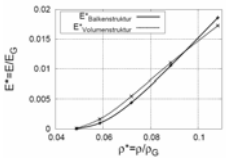


Figure 8. Comparison of results obtained with different discretization techniques. Solid line shows results obtained from models discretized with Timoshenko beam elements. Dashed line represents results obtained from those discretized with 3D solid finite elements.

Figure 9. Influence of structural non-uniformity on mechanical properties of modeled trabecular material.

Figure 10. Influence of structural damage on mechanical properties of modeled trabecular material.

3. CONCLUSION

In this paper a novel approach to modeling and simulation of trabecular tissue is presented. The model is shown to be able to capture the dependence of tissue's macromechanical properties on its porosity. The model can be adapted to account for structural non-uniformities, such as the non-uniform distribution of porosity and structural damage. Processes of bone adaptation occurring at the osteonal level can also be simulated using this approach.

4. REFERENCES

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