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MATERIAL CHARACTERIZATION ISSUES IN FEA OF LONG BONES

Nikola D. Korunovic[1*], Miroslav D. Trajanovic[1], Dalibor M. Stevanovic[1], Nikola M. Vitkovic[1], Milos S. Stojkovic[1], Jelena R. Milovanovic[1] and Dragana S. Ilic[2]

[1] Faculty of Mechanical Engineering, University of Nis
Aleksandra Medvedeva 14, 18000 Nis, Serbia
nikola.korunovic@masfak.ni.ac.rs
miroslav.trajanovic@masfak.ni.ac.rs
dalibor.stevanovic85@gmail.com
vitko@masfak.ni.ac.rs
miloss@masfak.ni.ac.rs
jeki@masfak.ni.ac.rs

[2] Clinical Center Nis
Bulevar Zorana Djindjica 48, 18000 Nis
e-mail: gagus@ninet.rs
gagus@ninet.rs

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Abstract. *One of the main issues that arise during preparation of models for subject specific finite element analysis (FEA) of long bones is the accuracy of material characterization. This paper tends to identify the most common sources of material characterization errors, which are sometimes also interconnected with bone geometry reconstruction errors, in order to help in creation of more accurate finite element models of long bones. Reconstruction of patient's bone geometry is usually based on medical images obtained by means of computational tomography (CT). Material characterization is performed either by segmentation of the model to characteristic zones that are assigned typical averaged material properties, or by local material mapping, based on bone density values estimated from CT numbers. Some of the main factors that influence material characterization accuracy are the choice of material model, the approach to material properties averaging, x-ray tube parameters, scanner calibration, relations between CT image gray values and bone density and relations between bone density and elastic properties of the bone. The paper brings a comparison of numerical results obtained from a number of subject-specific analyses of human femur, in which the approaches to material modeling were varied. Material modeling was performed using either geometry segmentation with material properties averaging or local material mapping. The results of the analyses were examined and mutually compared, and the influence of material characterization errors to analyses results was identified and explained.*

1 INTRODUCTION

Finite element analysis (FEA) is widely accepted as a technique for subject-specific evaluation of stress state in human bones, bone fixators and implants, with the purpose of surgery planning or bone tissue strength evaluation [1, 2]. It is especially useful in non-standard cases of bone fractures or tissue degradation. Such are the cases when it is not easy to choose an appropriate fixator from a number of standard ones, or when it is necessary to design a custom implant and predict its mechanical properties. The use of FEA in surgery planning is most valuable for less experienced orthopedists, as it may help them choose an adequate fixator and place it correctly on the bone.

Surgery planning often requires that the analyses are conducted in a very short time, i.e. in a few hours, thus a compromise approach to preprocessing and processing phases of FEA must be taken. In such conditions, one of the main issues that arise during finite element (FE) model preparation is the accuracy of material characterization.

This paper tends to identify the common sources of material characterization errors in FEA of long bones. It also brings the comparison of numerical results obtained from a number of subject-specific analyses of human femur, in which the approaches to material modeling were varied. Material modeling was performed either by geometry segmentation with material properties averaging or by local material mapping. The results of the analyses indicate that material characterization strategies have a significant influence on results accuracy in subject specific FEA of long bones.

2 BONE MATERIAL CHARACTERIZATION

The differences in intensity and direction of the stresses in various bone segments cause the differences in bone remodeling process, so the structure and density of remodeled bone tissue vary inside the bone. As a consequence, elastic properties of the bone also vary from point to point. Bone material characterization for use in FEA is performed either by segmentation of the model to characteristic zones that are assigned typical (averaged) material properties, or by local material mapping, based on bone density values estimated from radiological density (i.e. gray values or HUs). The second approach gained popularity in subject-specific FEA of bones, as it can be used to quickly model the inhomogeneous material properties.

Some of the main factors that influence material characterization accuracy are the choice of material model, the approach to material properties averaging, x-ray tube parameters, scanner calibration, relations between CT image gray values and bone density and relations between bone density and elastic properties of the bone.

Material properties are usually modeled as linear isotropic or anisotropic. While a number of studies have shown that anisotropic properties yield more accurate results [3, 4], they are much more tedious to determine and apply. It is also claimed that the difference in results is not worth the extra time and effort necessary for orthotropic material modeling [5]. This is even more relevant in the case of subject-specific bone analysis.

One of the earliest approaches used for bone material characterization in FEA is the assignment of averaged material properties to different bone segments. According to it, the bone is divided at least to cortical and spongious volume segments, but those may be further subdivided into zones that are expected to have notably different material properties [6, 7]. The idea behind the process is relatively simple, but it has a number of drawbacks. The major one is the use of average elastic properties that are assigned to different bone segments, as elastic properties of bone tissue may vary significantly throughout the bone volume. This is especially true for long bones like femur or tibia. Recognition and creation of zones inside the bone also requires significant time and effort, which does not speak in favor of this option, especially concerning subject-specific FEA. The mentioned drawbacks will be illustrated in an example presented in the next chapter.

On the other hand, local material mapping strategies imply the assignment of unique elastic properties to each finite element of the bone model, based on the density of bone tissue at the corresponding location, which is estimated from CT images. Elastic properties may be constant over the whole finite element, or they may vary spatially within elements volume. Empirical equations are used to establish relations between reported CT image gray values and tissue density, as well as between tissue density and elastic properties [8, 9]. As commercial FEA codes often accept only a limited number of material definitions (material cards), the whole range of possible values of an elastic property, like Young's modulus, is usually subdivided into a number of intervals, and each finite element is assigned a mid value of the interval to which it's calculated value belongs. While this approach enables a fast assignment of material properties to FE model, it must be performed carefully, as there are many issues that may influence the accuracy of results obtained by subsequent analyses. Some of those are:

- *Correlation between results accuracy and number of material intervals*. It is known that analysis results tend to converge to exact solution with a rising number of elements in a mesh [1, 10]. The same tendency is also present in the case of a rising number of material intervals [11]. Prior to an analysis, an optimal number of material definitions should be chosen through a convergence analysis, or an empirically estimated value for a certain bone type and mesh size should be used.

- *Calibration of CT dataset*. Depending on CT scanner type, CT tube parameters and even of oscillations in supply voltage, the reported CT numbers (gray values or HUs), in different scans of the same bone, may vary. Thus the calibration of CT dataset using a phantom is recommended, to establish a correct relation between reported CT numbers and radiological density of bone tissue ρ_{QCT} [8].

- *The relation between radiological density ρ_{QCT} and ash density ρ_{ash}*. It should ideally be equal to 1, but there are deviations present, which depend on bone tissue type and specific bone specimen. If possible, they should be taken into account [8].

- *The relation between ash density ρ_{ash} and apparent density ρ_{app}*. The ratio ρ_{ash}/ρ_{app} is close to 0.6 in human cortical bone, while for spongious bone it ranges from 0.34 to 0.62 [8]. There are also possible difficulties in experimental determination of the densities, as explained in [8]. For practical use, empirical relations that directly connect HU values and apparent density are often used, which either take into account or ignore all mentioned issues.

- *The relation between ash density ρ_{ash} or apparent density ρ_{app} and elasticity modulus $\boldsymbol{E}$*. The relations presented in literature are numerous [9, 12, 13], and they yield surprisingly different results, as shown in [9].

- *Material mapping algorithm*. As already mentioned, two approaches are used to assign material properties to finite elements of subject-specific models, and those imply the assignment of either spatially variable or constant material properties. Material properties that vary spatially inside each finite element cannot be simply defined in FEA codes. Thus they have not been used very much, although they are reported to yield more accurate strain results [14]. The constant values assignment approach has many variations, depending on how the average value of a material property is calculated from CT numbers. The main difference between those variations is reflected in the averaging technique used. According to the simplest one, density values are assigned to every finite element node, based on the nearest value on CT sampling grid, and then a weighted average of nodal values is assigned to the element [14]. According to another technique, average element density is calculated from eight points surrounding element centroid [14]. The aforementioned techniques are found to produce inaccurate results when elements are significantly larger than the spacing of CT sampling grid. One of the more advanced techniques starts with averaging of densities from CT grid points that belong to a finite element [11]. A consecutive improvement of this technique implies that an average HU value for a finite element is calculated by numerical integration of HU field [15]. According to a further improved technique, Young's modulus field is first calculated from HU field and then averaged by numerical integration [16], to compensate for the fact that the averaging procedure is not commutative, as modulus-density relations are usually nonlinear. Other approaches are also present in literature, like the one based on the intermediate step of material blocks creation [17]. This approach is said to be more accurate than the others, when finite element size is larger than the volume occupied by voxels chosen for property averaging, or when element size is comparable to or smaller then voxel size. Material mapping algorithms may also assign inaccurate (lower than real) values of material properties to boundary layers of finite elements, as some of the voxels used for averaging do not or only partially belong to bone tissue. Material mapping algorithm is certainly one of the most important factors affecting bone material characterization accuracy, and an universal one is still not recognized.

- *The dependence of material mapping results on mesh size*. This issue is partially connected to the choice of material mapping algorithm, as already explained. If finite elements are too large, a number of factors that influence FEA results accuracy may be present. At some locations within FE model, element size may be large comparing to the thickness of cortical bone, so it may also cover a large portion of spongious bone. As a result of material properties averaging, a too small value of Young's modulus or other elastic property may be assigned to it. Averaging of material properties may be the cause of inaccuracies even if a finite element covers only a single tissue type, in cases when the tissue is highly porous, as spongious bone for example. The mentioned problem is also interconnected with CT image voxel size.
- *Connection between material mapping results and accuracy of geometry reconstruction*. If cortical bone is much thinner then CT image voxel size, gray values of voxels that contain a thin layer of cortical bone may be much lower than the usual equivalent of cortical bone, as averaging algorithms also take into account large portions of adjacent soft tissue or spongious bone. Thus, threshold based methods used for bone surface reconstruction may omit some voxels, and the surface of the cortical bone in corresponding areas may not be detected at all. The simplest way to avoid this situation is to reduce the lower threshold value, so a greater portion of the surface is recognized. Unfortunately, in this case some of the tissue surrounding cortical bone (periosteum) is also recognized as a cortical bone, and so are the denser segments of spongious bone. Thus, the reconstructed bone surfaces may be inaccurate, e.g. the reconstructed outside surface may be larger than the real one. This approach represents a trade-off that enables bone surface reconstruction to be performed more quickly. However, in this case the averaging of material properties assigned to finite elements situated on the bone surface may produce too low values of elastic properties, as density of connective tissue that surrounds the bone gets included in the calculation.

3 MATERIAL CHARACTERIZATION STUDY

In order to explore the influence of material characterization approaches to the results of FEA of long bones, a study was performed on a number of subject-specific models of human femur that were assigned material properties using a number of different approaches.

3.1 Finite element models and material characterization

A CT scan of lower extremities, characterized by pixel size of 0.782mm, was used as a basis for creation of 5 different FE femur models, with the following features:

- **Model 1** (Figure 1). Only the external femur surface was reconstructed from CT images. Femur volume was meshed with quadratic tetrahedron elements, using average edge size of 3mm at the bone surface and fast element growth towards the inside of the bone. Material mapping was performed in order to assign material properties to all elements, using the empirically obtained relation between HU values and bone density (Eq. 1) and relation between bone density and Young's modulus taken from [12] (Eq. 2). Three variations of the model were created, (**model 1a**, **model 1b** and **model 1c**), using 20, 100 and 300 material definitions, covering the range of Young's modulus values from 0 to 19GPa.

$$\rho_{app}[g/cm^3] = 0.1957 + 0.001053\, HU \tag{1}$$

$$E[N/mm^2] = 6950 \cdot \rho_{app}^{\ 1.49} \tag{2}$$

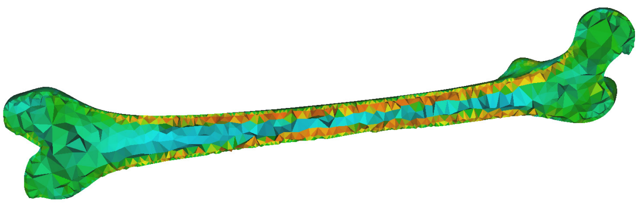


Figure 1: Cross section of femur FE model, with mapped material properties (model 1)

- **Model 2** (Figure 2). The same external surface that was used to create model 1, was used as external surface of model 2. In this case the inner surface of the cortical bone was also constructed. It was done by merging the surface of medullary cavity, reconstructed from CT images, and two surfaces that were offset from external femur surface in epiphyses areas, using offset value of 1 mm. The offset surfaces were created in order to approximate the thin part of cortical bone that could not be properly reconstructed from CT images. Two zones inside femur model were created, one inside the inner cortical bone surface, and one between the inner and outer cortical surfaces. The zoned model was meshed using the same mesh parameters as in the case of model 1. Material mapping was performed using the same relations that were used for model 1.

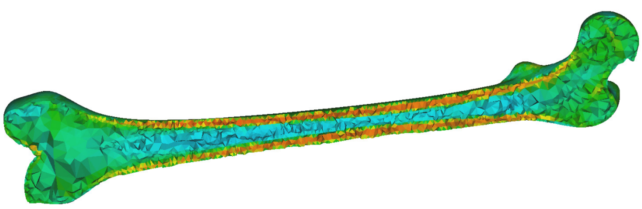


Figure 2: Cross section of zoned femur FE model, with mapped material properties (model 2)

- **Model 3** (Figure 3). Separate zones inside the model were created for cortical bone and medullary cavity. The volume of spongious bone was divided into a number of zones, according to usual distribution of trabecular density, as described in [6]. Material characterization was performed by assignment of constant Young's modules to the elements belonging to cortical bone zone and to the elements belonging to spongious bone zones (Table 1). A very small value of Young's modulus (1MPa) was also assigned to elements inside medullary cavity, where bone marrow is situated. Material properties were defined using trial and error approach, in order for model 1 to have the same value of maximum displacement as model 2.

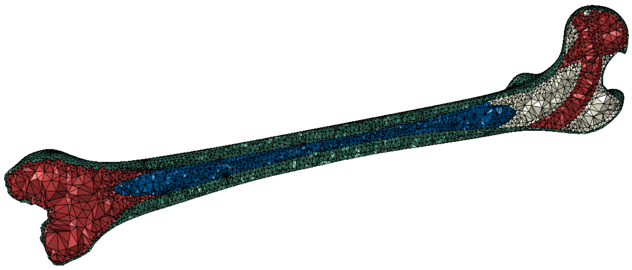


Figure 3: Cross section of zoned femur FE model, with constant averaged material properties (model 3)

Compared to model 2, model 1 contains larger elements near the surface, each of them partially covering the cortical bone and partially the spongious bone. Model 2 contains considerable smaller elements near the external surface, most of them covering only the cortical bone and some of them also covering only the small portions of spongious bone.

Boundary conditions and loads were chosen such as to simulate one legged stance, according to [18] (Figure 4).

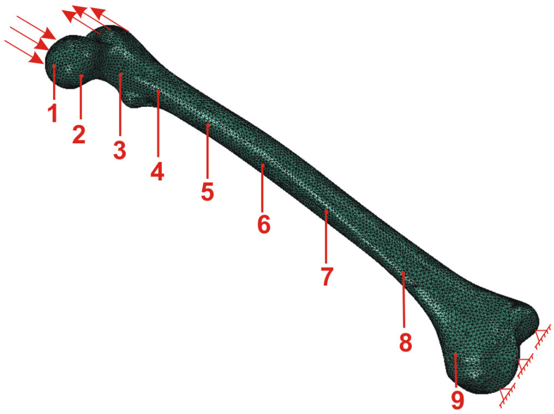


Figure 4: One of FE femur models with symbolically presented boundary conditions and loads. The points that were used to monitor analysis results are also shown.

The main goals of the study were defined as follows:

- To compare the results of the analyses obtained using FE models to which material properties were assigned using either material mapping approach or combined zoning-mapping approach (model 1 vs. model 2).
- To compare the results of the analyses performed on FE models which were assigned material properties in two different ways, by combined zoning - material mapping procedure or by zoning and constant average material properties assignment (model 2 vs. model 3).

- To indentify some of the ways in which material characterization strategy affects the results of subject specific FEA of long bones.

3.2 Results

Nodal displacements and equivalent Von Misses stress field were monitored over whole models and especially at selected nodes, shown in Figure 4. Before the study was performed, the influence of the number of different material definitions to analysis results was investigated using model 1 (Figure 5). Because the results obtained using three different numbers of material definitions were very similar, the study was continued using only model 1b (containing 100 material definitions), which will in further text be referred to as model 1. In order to properly compare the results, model 2 was also mapped using 100 material definitions.

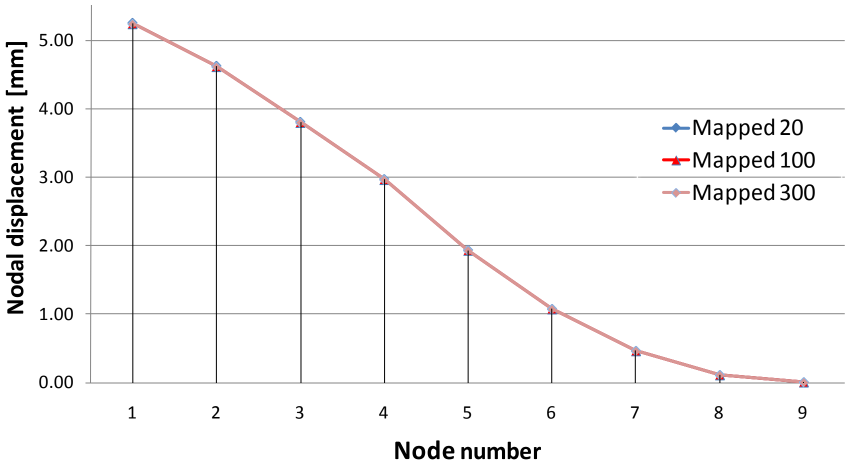


Figure 5: Magnitude of displacement at selected nodes, obtained by analysis of femur models in which material properties were defined using 20, 100 or 300 material cards (1a, 1b and 1c).

Although the same material mapping equations were used for models 1 and 2, the difference between nodal displacements obtained using the two models is notable, and it ranges from 4.7 to some 11.3%, for nodes 1 - 8 (Figure 6). As quadratic elements were used, it was not expected that mesh size variation alone should produce such a difference. Thus, it is speculated that the observed difference is the consequence of material mapping approach. It is probable that the larger portion of model 2 was assigned higher values of Young's coefficient than in the case of model 1. As it contains smaller elements near the surface, material properties averaging process may not have taken into account as large portions of spongious bone as it did in the case of model 1, where the elements near the surface are larger.

As already mentioned, material properties of model 3 were adjusted, in order to achieve the same maximal displacement as in the case of model 2, under the same loading conditions (Figure 7). As it may be seen from the graphic, displacement value at node 1 obtained by FEA of the two models is the same, but at nodes 2-5 it is noticeably different. That means that the deformed shape of model 3 is different from deformed shape of model 2. As it may be seen from Table 1, equivalent Young's modulus of cortical bone ($E_{cortical}$=12.7 GPa), that is defined in order to achieve the same maximal displacement over the model, is quite low considering the range of values reported for human cortical bone (10.9 - 20.6 GPa) [19]. On the other hand, the range of Young's modulus values for model 2, which results from material mapping process, is approximately 0 - 19 GPa. It is hard to determine this range for cortical bone only, as some elements at epiphyses have gotten smaller modulus values, because tissue density is averaged based on pixels that partially cover the spongious bone.

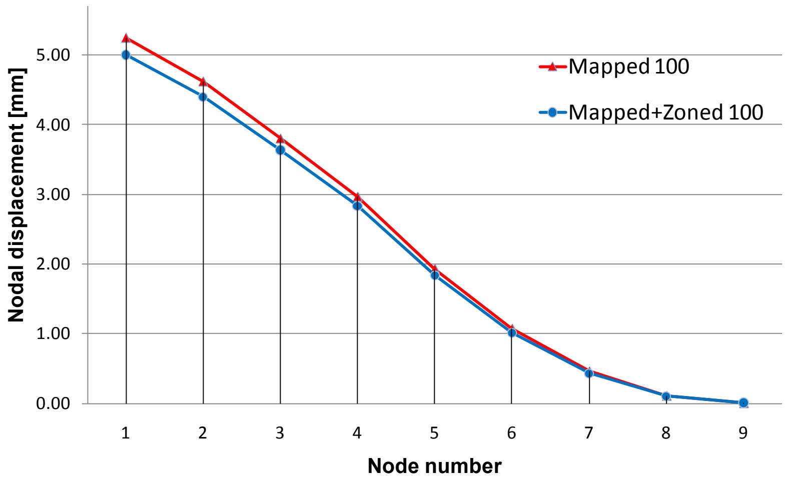


Figure 6: Displacement magnitude at selected nodes, obtained by analysis of models 1 and 2 (mapped material properties and zoned + mapped material properties)

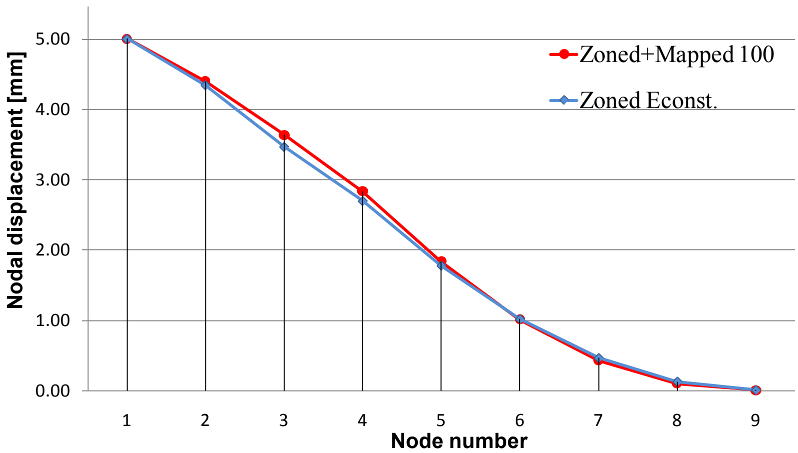
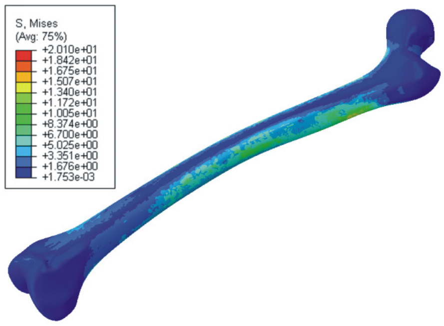


Figure 7: Total displacement at selected nodes, for models 2 and 3 (zoned + mapped material properties and zoned with constant material properties)

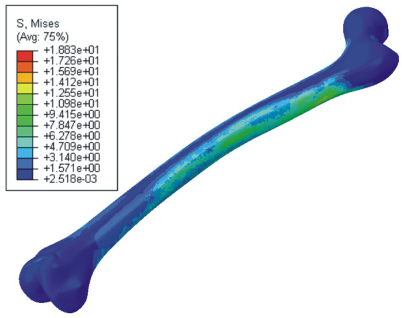
The conclusions that may be drawn from comparison of analysis results obtained using models 2 and 3, are:

- It is hard to determine the average material properties that will be assigned to different bone segments when zoning approach is used.
- Even if results obtained using local material mapping and zoning / average material properties assignment approaches are similar in one part of FE model, they may be different in the other part.
- There will always be some error present when zoning / average material properties assignment approach is used, as in reality elastic properties of cortical bone vary from point to point.

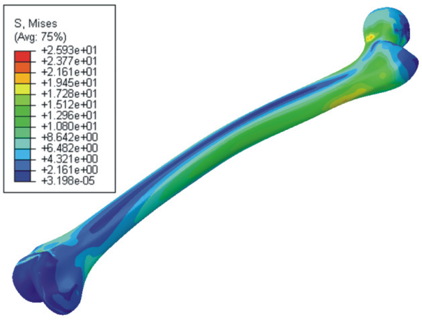
Equivalent stress field obtained by FEA of models 1 - 3 is shown in Figure 8. It is obvious that numerically obtained stresses at the surface of models 1 and 2 are very similar, while the stresses at the surface of model 3 are quite different. This difference is the consequence of difference in Young's modulus distribution, which is variable in first two cases, and constant in the third. On one side, it is more realistic to have variable Young's modulus over the model, but as the consequence of the nature of material mapping, and especially of averaging techniques used, it happens that for models 1 and 2 the maximal value of stress is located rather unrealistically, deep under the bone surface.



Model 1



Model 2



Model 3

Figure 8: Equivalent stress field obtained by FEA of models 1-3

The maximal values of displacement, stress and strain obtained by FEA of all three femur models are given in Table 1. It may be noticed that the analysis of the zoned model, where constant averaged material properties were used, yields considerably smaller value of maximal strain.

Femur model	Material characterization	Number of elements	Number of nodes	Max. displacement u[mm]	Max. Eq. Stress [MPa]	Max. strain
1a	Mapped, 20 material definitions	59 926	98 298	5.50	20.26	0.00192
1b	Mapped, 100 material definitions	59 926	98 298	5.49	20.10	0.00202
1c	Mapped, 300 material definitions	59 926	98 298	5.49	20.26	0.00201
2	Zoned + Mapped, 100 material definitions	149 712	217 762	5.24	18.33	0.00200
3	Zoned, $E_{cortical}$=12.7GPa, $E_{trabecular}$=0.07-0.3GPa	149 712	217 726	5.25	25.93	0.00839

Table 1: Material characterization, model sizes and maximal values of displacement, stress and strain field variables for various femur models used in the study.

4 CONCLUSIONS

Common sources of geometry material characterization errors in subject specific FEA of long bones, according to literature and experience of the authors, were identified and discussed in the paper. The study was performed in order to investigate this topic in more depth, in which various approaches were used to create a number of subject specific models of human femur. The same boundary conditions were applied to all finite element models used in the study. Comparison of the obtained results led to the conclusions that follow.

In the presented study, local material mapping approach was compared with zoning / constant material properties approach, as well as with combination of those two. The difference of up to 11.3% in nodal displacements between mapped and zoned / mapped models was noticed. It is considered to be the consequence of combined influence of material properties averaging technique used during material mapping and size of finite elements located near the bone surface. The difference in deformed shapes between mapped and zoned models was also significant, which is the consequence of material characterization approach, as the same mesh was created on the two compared models. It is concluded that zoning / constant material properties assignment method will always produce a significant error in stress / strain prediction and that it is hard to estimate an averaged value of Young's modulus which will represent the elasticity of bone tissue over a whole long bone. A tendency of mapped FEA models to report the maximal stress values deep beneath the bone surface is also identified as a consequence of material characterization approach.

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