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Numerical Analysis of Human Femoral Bone in Different Phases

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Abstract—the yearly occurrence of hip fracture due to osteoporosis & other reasons has been increased very rapidly. From a biomechanical point of view hip fractures are thought to be caused in real settings by different directions of loading. Thus, clarification of the loading directions under which the proximal femur & shaft is most susceptible to fracture would be helpful for elucidating fracture mechanics and establishing preventive interventions. Finite element analysis has been used extensively in the study of human bone loading checks and implant performances. In this paper we are presenting the femur bone analysis. The boundary conditions are applied here, generally producing excessive femoral deformation, and although it has been shown that the muscle/tendon forces influencing femoral deflections and dynamic loadings. It is hypothesize that careful application of physiologically based constraints can produce physiological deformation this causes straining, of the femur. Five boundary condition cases were applied to a finite element model of the Standardized Femur. This paper presents numerical analysis results of stresses and displacement in femur in a living dead phase& artificial implant. The aim of the work was to present the influence of different mechanical properties of these three bones & comparative study on the obtained results. The appropriate selection of the properties ensures correct results.

Index Terms: **Numerical techniques, FE Analysis, Mechanical properties.**

I. INTRODUCTION

Numerical analyses of biomaterial tissue are closely related to selection mechanical properties of implants. Results of these analyses are helpful in mechanical properties selection of biomaterials and implants. But it is to be remembered that these results are connected with preselected mechanical properties of bone tissue. Knowledge connected with mechanical and structural properties of bone tissues, especially hard tissues, is essential in order to carry out correct numerical and experimental analyses. From the biomechanical point of view, determination of hard tissues structure is crucial. For this Knowledge of the material properties is essential, both in diagnosis & implants. Stiffness of a bon implant system is particularly important. As Young modulus of bone changes with age. It is related with demineralization of bone. Increase of bone porosity is caused by different factors, for example osteoporosis which is characterized by decrease of bone mass, disordered micro architecture of bone and, as a result, decreased mechanical strength. These factors lead to increase of fracture risk. This paper literature data in adults (approximately 30-35years old) shows that maximum mass of bone tissue is reached in this age range. After the age of 40 loss of bone mass starts. In this way, about 0.5-1.25% of minerals of healthy body per year are lost. conversely, in osteoporotic bone the loss is in the range 2 to 5% per year. That is why in the osteoporotic bone the calcium intake is less exit is more which makes bone porous and brittle [16].

Table 1. Mechanical Properties of Bone in Different Phases (Live and Dead). [15]

Bone	Living Bone	Dead Bone
	Young modulus of bone, MPa	
Femur	17 260	20 202
Tibia	19 040	20 590
Fibula	18 540	21 080

Mechanical properties of bone tissue (tensile, bending and torsional strength) allows to evaluate stresses and strains in bones and select mechanical properties of implants. Comparative test results of mechanical properties of fresh and dead cortical bone are presented in Table 1 [15].

Need of Implants

Many surgeons suggest that there is not much difference overall between a standard implant and a custom one but some have seen slight differences. For the most part, modern implant components from every manufacturer are versatile enough to be successful in both men and women. With modular components, total customization is rarely necessary. In some instances though, surgeons find gender-specific and other custom implant designs useful. A large man, for example, with a bigger-than-typical offset might be best served with a custom-made hip implant using an extra-extended offset stem. You should speak with your hip surgeon about your specific needs. It makes sense to inquire about gender and other potential reasons for a customized or specialized implant. Understand that it generally has less to do with gender than it has to do with physique. Rest assured that there are many options available that will surely fit your specific circumstances [3]. After all surgeon will decide which implant to use based upon many factors including his or her experience with a particular implant, your situation, age, weight, gender and your lifestyle. For design & analysis point of view of custom implant the following flowchart will guide.

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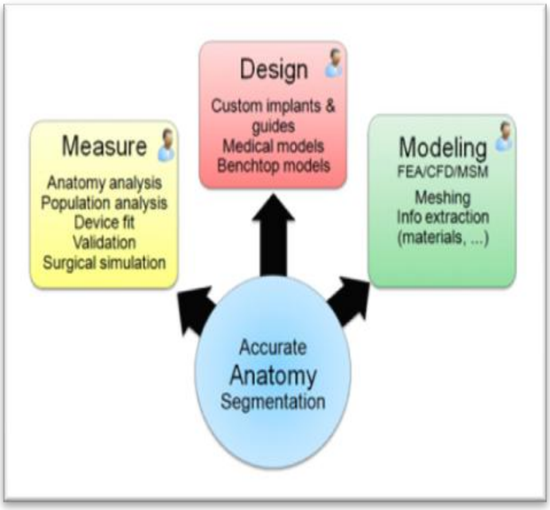

II. MATERIAL AND METHODS

The femur can undergo only a 1.25 % decrease in length before fracturing. For this case study we have considered Femur of adult male age 33yrs of length l=457 mm, shaft diameter D=30 mm and diameter of tube d=11 mm was used in the research .The femur model is developed in order to carry out numerical analyses, the model modified with the use of Ansysversion12.0. In the preprocessing stage meshing of the geometrical model is done. The obtained numerical model consisted of 45300 elements. Calculations were carried out for the femur of different mechanical properties corresponding with the living stage (fresh bone) and the dead (desiccated bone). In order to obtain more satisfactory results, a intermediate values were also preferred. The following mechanical properties were selected [12]:

- for the living phase (model 1) – EI=17260 MPa, υ = 0.3
- for the intermediate phase (model 2) – EII=18600 MPa, υ=0.3
- for the dead phase (model 3) – EIII=20202 MPa, υ=0.3.
- Calculation of the maximum compressive force (Fc) that can be applied to the femur without it fracturing in live phase is

$\delta = 0.0125L = FL/AE$,

$F = 0.0125\pi[(14^2-5.5^2)x10^{-6}](1.5x10^{10})$

$= 97600N$

Young's Modulus for bone is 1.50×10^{10}Pa. The following boundary conditions were applied to carry out the numerical analysis:

- lower part of the femur was steady fixed (all degrees of freedom),
- the models were loaded with the following forces [8,9]:
- resultant of reduced body mass (0,80G) and abductor muscle force R (head of femur),
- Muscle reaction forces M (gluteus),
- Muscle reaction forces T (iliotibial).

Loading scheme of the femur was presented in Fig. 2.

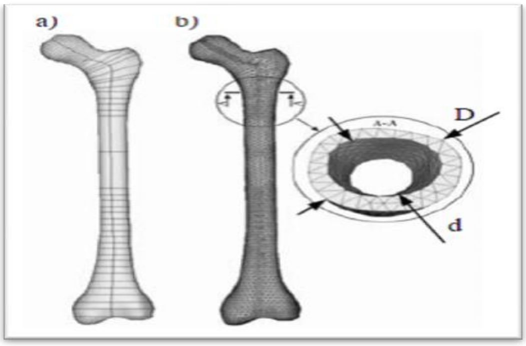

Fig. 1. Model of femur

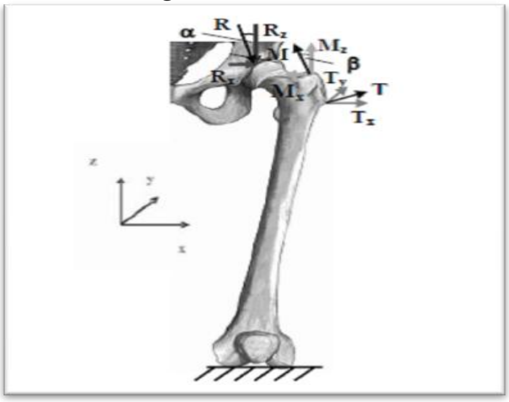

Fig. 2. Loading scheme of model

a) Geometrical b) isolated$\alpha=16^o$, $\beta=21^o$

Table 2. Maximum displacements distribution in the plane Ux, Uy and Uz and stress distribution [8]

Model	Displacement. Ux, m	Displacement. Uy, mm	Displacement. Uz, mm	Stress, MPa1
1	-13.14	1.20	-3.99	49.4
2	-12.19	1.11	-3.70	49.4
3	-11.23	1.02	-3.41	49.4

Mechanical Properties of bone

The bone density can be defined as the bone mass per total volume of bone including any holes. The bone density calculated in this way represents the mean density of the apparent material specimen and is also known as the 'apparent density'. Note that the bone apparent density is not equal to the bone tissue density. Let introduce the bone volume fraction *V V*representing ratio of bone volume over total volume,[17]

$$V\,V = V\,B / T\,V$$

Where*B V* and *T V* are the bone volume and total volume, respectively. Assuming that the bone tissue is uniform with density *tissue* ρ, the relationship between apparent density and bone tissue density can be written as

$$Apparent\ V\ tissue\ \rho = V\ \rho$$

Experimental investigations showed approximately linear dependence of the apparent density of bone tissue on bone porosity. General principles of the Finite Element Modeling of structures are used here. The bone structure is usually modeled by 3D finite elements in order to capture the bone geometry.

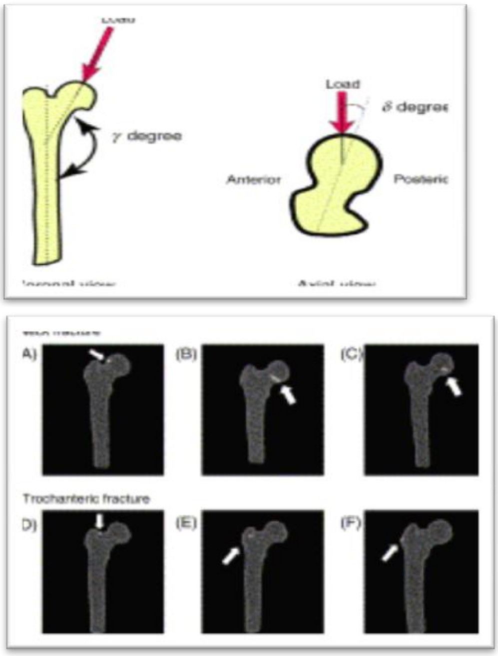

Fig 2a -Femur Geometry & Loading Conditions.

Fig-2a shows Femur Geometry & different Fracture types loading conditions corresponding to each of the predicted fracture sites. Arrow: Predicted fracture site. (A) (B) (C) With predicted fracture sites located at the sub capital region, the type was classified as neck fracture. (D) With predicted fracture sites at the base of the femoral neck, the type was classified as trochanteric & (E) (F) with predicted fracture sites on the trochanteric region.

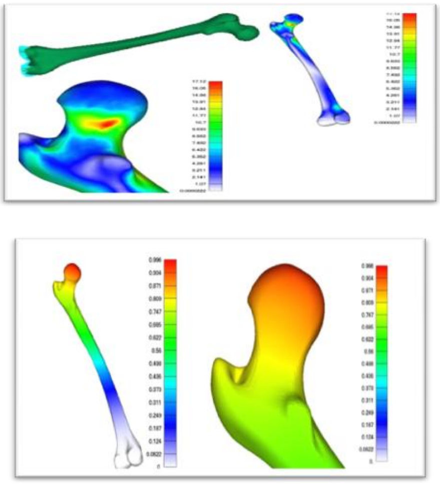

Fig 2 b- Stress Distribution in Proximal & stem region of femur

III. RESULTS

The presented results clearly indicate the numerical analysis for different mechanical properties was presented in Table 2 and in Figs3 to 7. Obtained displacements and stresses are reduced in values.

The results indicate differences in displacement and stresses depending on mechanical properties of the analyzed femur bone model. The results indicate that the displacements appear mostly in the frontal plane (ux). The highest values are observed in femur head region. It was also observed that increase of the Young modulus caused the decrease of displacements in the transversal plane. It is caused by higher limpness of the bone. The fig 2b shows the stress distribution where red & green colourindicates higher & lower stresses respectively. The highest displacements was observed in region of the femur head, which for the model 1 was equal to ux=-13.14 whereas the lowest value was observed in the model 3 (dead phase) - ux=-11.23. The lowest values of displacements were observed in the sagittal plane (uy) and along the bone axis (uz). For the model 1 the values were equal to uy=-0.01 mm and uz=1.54 mm (lower region) and uy=1.20 mm and uz=-3.99 mm (femur head region). For the model 2 the displacements were in the range 0.01-1.11 mm and uz=-3.70-1.43 mm. For the model 3 the displacement values were the lowest and were in the range uy=-0.01-1.02 mm iuz=-3.40-1.32 mm.

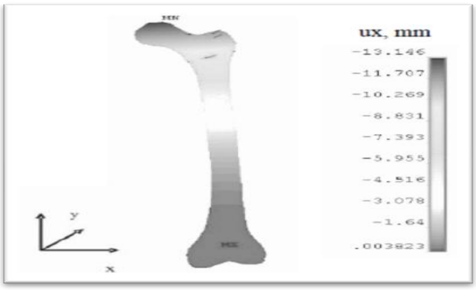

Fig.3. Displacements Distribution in the Bone in the Frontal Plane (Ux) - Model 1

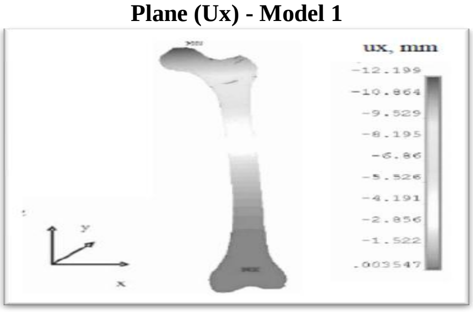

Fig. 4. Displacements Distribution in the Bone in the Frontal Plane (Ux) - Model 2

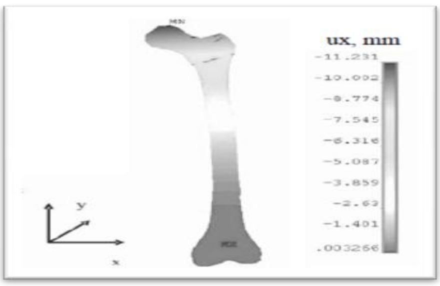

Fig. 5. Displacements Distribution in the Bone in the Frontal Plane (Ux) - Model 3

The Stress investigation of the femur showed that stresses, for the all analyzed bone models, are analogous and did not exceed the value of 50 MPa. The applied loading generates tensile stresses in the exterior region of the bone and compression stresses in the stem region.

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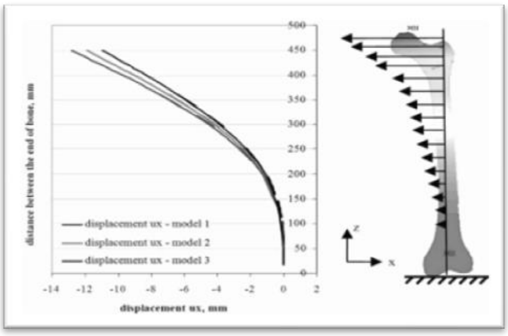

Fig. 6. Diagram of Displacement Distribution in the Frontal Plane (Ux)

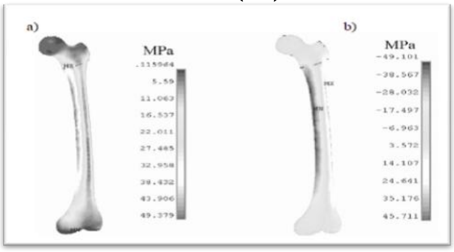

Fig. 7. The Stresses in the Bone – A, Compression and Tensile Stresses – B

IV. CONCLUSION

The numerical analysis was carried out in order to evaluate displacements and stresses in femur of 33 year old man. Knowledge of the problems is essential, both in diagnosis of bone system illnesses as well as in appropriate selection of fixation system. It is known that these conditions must be correlated with geometry and mechanical properties of bone (dependent on a bone structure). On the basis of the established, appropriate boundary conditions and obtained results, it can be stated that:

- displacements of femur depend on its mechanical properties,
- value of Young modulus significantly influences the displacements,
- higher value of displacements for all axes (x, y, z) was observed for the bone in the living phase, because of the higher elasticity in comparison with the bone in the dead phase,
- mechanical properties of bone (Young modulus) do not exert an significant influence on stresses,
- the obtained results can be used in further biomechanical analysis of femur – implant fixation.

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