PAPER REF: 7203

INFLUENCE OF BONE QUALITY IN THE BEHAVIOUR OF GRAFT FIXATION IN ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION

C. Quintana[1(*)], **C. Rodriguez**[1,2], **I. Peñuelas**[1,2], **A. Maestro**[1,3]

[1]SIMUMECAMAT group, University of Oviedo, Gijón, Spain
[2]University Institute of Industrial Technology of Asturias, University of Oviedo, Gijón, Spain
[3]FREMAP Clinique, Gijón, Spain
[(*)]*Email:* covaquimbar89@gmail.com

ABSTRACT

This work compares, using FEA tools, the different behaviour of the anterior cruciate ligament reconstruction (ACL) depending on the bone quality of the patient. In the literature, some authors (Hernandez and Keaveny, 2006) analyse how the age of the patient and the risk of osteoporosis can affect the density of the bone, and thus, the elastic modulus of the material. With this basis, the aim of this paper is to give an approach of how the quality of the cancellous bone affects the later response of the fixation in ACL reconstructions, focusing on the tibia-graft-screw assembly. This will help with the preoperative, choosing the best relation between the diameter of the interferential screw, the diameter of the tibiae tunnel and the election of the graft in each case.

Keywords: biomechanics, ACL reconstruction, interference screw, postoperative response, bone quality.

INTRODUCTION

Anterior Cruciate Ligament (ACL) injury is the most common ligament injury in the knee for athletes. However, due to the adoption of healthier habits and the consequent increase in life expectancy, this injury is becoming increasingly common in people of wide ranges of ages and health conditions. In addition, the postoperative care requirements and the degree of satisfaction felt by patient will depend on the age and health of patient in terms of bone deterioration, and mainly the need of restoring a previous sport condition. To achieve the purpose of the paper, numerical models (based on finite element methods), as well as experimental results, will be used to reproduce the degradation of bone with the age of the patient. The variation of the cancellous bone quality will be modelled through the change of both the elastic modulus and the strength of the bone.

The behaviour of graft fixation will be studied in terms of compressive stresses and pull-out strength, since bone quality directly affects the pull out stress required to displace the screw. Thus, both the tibiae bone tunnel-interference screw joint and the tibiae bone tunnel-ACL joint have studied. The higher the compressive stress, the best the configuration. In addition, tensile tests have been simulated for checking the pull out strength of the reconstruction measuring the tensile load that causes the screw to slip. According to the age-related variations (mean +-SD) in the mechanical properties in the three age groups analysed (young, middle and old patients) found in bibliography (Ming Ding, 1997), three different Young's

Modulus have been used in this study (700 MPa for young, 900 MPA for middle and 600 MPa for old patients). It has also been found a direct relation between the stress in the bone caused by the interference screw and the elastic modulus of the bone. For these age groups, the ultimate stress (strength of the tibiae cancellous bone) considered for the numerical studies vary from 14 MPa for young patients to 7 MPa for old ones, been around 11 MPa for middle-age patients (Keaveny, 2001).

MATERIALS

In the present work, fresh porcine tibiae were used, storing them at the sacrifice moment. As plasty, porcine tendons of the *flexor digitorum* were chosen, storing at the sacrifice moment as well. Tendons were selected with the most similar and homogeneous diameter as possible, of approximately 4mm. In Figure 1, biological material used during the investigation is shown. All these material were frozen at -22ºC and, then, defrosted for the mechanical test. After the defrosting, the different samples were maintained wet until the execution of the mechanical test with saline solution, and were stored in hermetically sealed polyethylene bags.

Fig. 1 - Biological material

With the purpose of obtaining the geometric models and the mechanical behaviour models needed in the FEA analysis, the different materials were subjected to different test methodologies.

In the case of cortical bone, and basing our analysis in the results obtained with similar materials (Giddings, 2001), the Small Punch Test (**SPT**) was chosen to characterize the mechanical behaviour of this part of the bone. During the test, load-displacement values were registered. Some samples are shown in Figure 2.
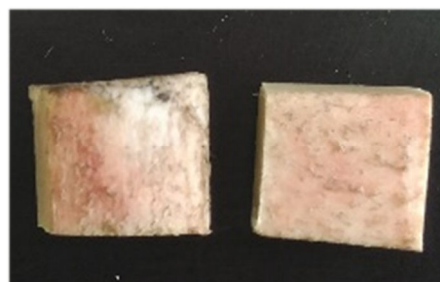

Fig. 2 - Cortical bone samples. SPT colocation

The characteristic curves obtained clearly define a linear elastic material. Thus, the mechanical parameter needed to define the cortical bone in FEA model is the elastic modulus. This parameter was obtained with the simulation (using a FEA model) of the test. The elastic modulus will be the value which best fits the numerical load-displacement curve to the experimental one. This value was E=25GPa, very closed to the bibliographically found for humans of 20GPa (Dorogoy, 2017). This value was chosen as reference because of the similarities, in terms of mechanical behaviour, between porcine and human bones. The Poisson's coefficient was selected basing on bibliography data (Peña, 2007), with a value of 0.35.

In the case of the trabecular bone, Figure 3, and with the aim of characterize this material, fresh trabecular bone samples, conveniently measured, were subjected to compression tests.
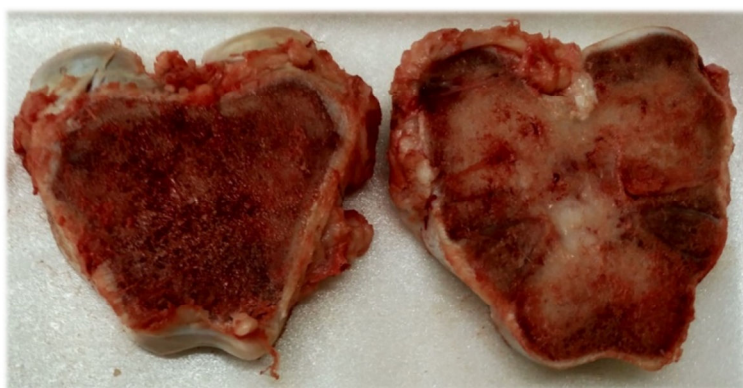

Fig. 3 - Trabecular porcine bone samples

With the characteristic curves, mechanical parameters will be obtained for defining this material. From our results, and coinciding with other authors approaches (Burstein, 1975), the trabecular bone behavior could be considered as elastic-perfectly plastic. Thus, the mechanical parameters needed will be the elastic modulus (E) and the yield stress (Sy).

Three cases of study have been considered in this paper: young patient, middle-age patient and old patient. These cases reveal the difference, in terms of bone density and mechanical strength, between the three population cases. In our FEA model, the values for elastic modulus and yield stress considered were shown in Table 1.

Table 1 - Trabecular bone parameters

	E (MPa)	Yield Stress(MPa)	σ_{true} (MPa)	Ultimate Strain(%)	ε_{true}
Young person	700	14	14.42	3	0.012837
Medium-age	900	11	11.275	2.5	0.01072
Old	600	7	7.14	2	0.0086

The trabecular bone exhibits anisotropy, that is, its elastic modulus and its yield stress are different in both directions, transversely and longitudinally. This orthotropic characteristic is defined with the equations which establish the relation between the elastic modulus and the Poisson's coefficient obtaining the shear modulus:

$$D_{1122} = \frac{E_1}{2(1+\mu)} \tag{1}$$

$$D_{2233} = \frac{E_2}{2(1+\mu)} \tag{2}$$

where the Poisson's coefficient for trabecular bone has a value of 0.35.

In relation to tendons (Figure 4), they were characterized using uniaxial tensile tests, but with different fastener devices, which allowed pressure to be increased with load.

Fig. 4 - Tendon during tensile test

Afterwards, the mathematical model of the material behavior was done for specific soft biological tissues. Basing the model in bibliographic results and taking into account the constitution of the tissue, tendons were modelled as transversely isotropic and longitudinally anisotropic and hyper-elastic behavior. That means, that the collagen fibers that compound these kind of tissues, are in one and unique direction of the tendon, the longitudinal one. This is why the only stress different to zero, and positive (tensile stress) will be the longitudinal stress.

Experimental procedure reveals that tendons are materials which exhibit large deformations at low loads. These tissues present a strong non-linearity (Figure 5). Therefore, the proper formulation needs to be found in order to achieve the constitutive material model that better

adjust to these requirements, assuming that the tissue is incompressible or quasi-incompressible (Odgen, 2001).

For this purpose, a modified Weiss's model (Weiss, 1996) is done. The most proper analytical expressions for tendon tissue were obtained. These expressions have been used by Calvo *et al.* (Calvo, 2009) in previous works.

$$\bar{\varphi} = c_{10}(\bar{I_1} - 3) + \overline{\varphi f} \tag{3}$$

$$\overline{\varphi f} = 0, \ \overline{I_4} < \overline{I_{40}} \tag{4}$$

$$\overline{\varphi f} = \frac{c3}{c4} \left(e^{c4(\overline{I4} - \overline{I40})} - c4(\overline{I4} - \overline{I40}) - 1 \right),$$
$$\overline{I_4} > \overline{I_{40}} \ y \ \overline{I_4} < \overline{I_{4ref}} \tag{5}$$

$$\overline{\varphi f} = c5 \sqrt{\overline{I4}} + \frac{1}{2} c6 Ln(\overline{I4}) + c7, \qquad \overline{I_4} > \overline{I_{4ref}}$$
$$(6)$$

The constants c10, c3, c4, c5, c6 y c7 are obtained by fitting experimental data of both tested tendons (Figure 5) applying an error function minimization between experimental stress and analytical stress. $\overline{I_{40}}$ e $\overline{I_{4ref}}$ are the minimum and maximum values that define the tendon exponential curve.
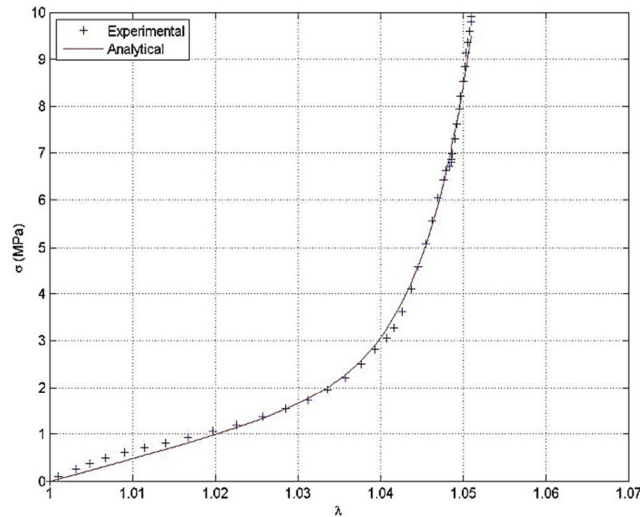

Fig. 5 - Characteristic curve tendon 1. Tensile test

As far as the FEA software does not have a proper model to characterize this material, a new subroutine (uanisohyper_inv) was programed in Fortran for Abaqus, taking into account that this tendon only has a unique fibers family.

Finally, the interference screw mechanical model, Figure 6, was deduced from the results of compression tests done over interference screws, from which load-displacement curves were obtained. The elastic modulus can be determined fitting the numerical curve to the experimental one. In this case, the obtained elastic modulus was 8GPa, very similar to the value proposed by other authors for this type of mixture (PLLA+HA) in the same proportions

(Jianping, 2006). This material (75% PLLA+25%HA) is an elastic-plastic material. Thus, it is important to know the yield stress, too. In this case, the value of yield stress for this specific material is 100MPa, with a Poisson's coefficient of 0.45.

Fig. 6 - Interference screw.

NUMERICAL MODEL

FE Analyses in this study were performed using commercial software ABAQUS. Two dimensional (2D) model of the simulation is shown in Figure 7.
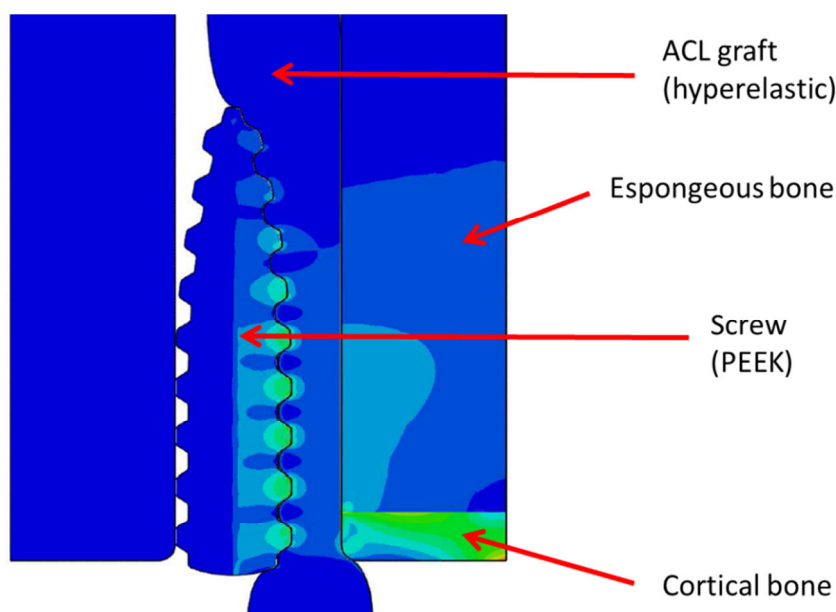

Fig. 7 - FEA model of ACL reconstruction

Asymmetric conditions were utilized and eight-node brick elements type plane stress, two of them node linear and the other one bilinear with reduced integration and hourglass control (CPS3, CPS4 and CPS4R within ABAQUS) were used.

Surface contact option within ABAQUS was applied between surfaces of the interference screw, the trabecular bone and the tendon, being the screw the master surface for bone-screw and screw-tendon contacts, meanwhile the master surface for bone-tendon contact is the bone.

The interaction contact between the screw-bone and bone-tendon was type friction defined by a tangential behavior with a friction coefficient of 0.1 and a normal behavior type hard-contact. In all cases geometric properties were taking into account.

Contact controls were created to have a properly stabilization during the simulation. These controls are chosen with automatic stabilization (with a factor of 0.001) in order to let the program decide the value that best fits in each step.

In the case of the screw-tendon contact, the interaction was defined by a frictionless behavior, with a null friction coefficient as tangential behavior and a hard-contact normal behavior. Both interaction properties were created with active geometric properties active and using automatic stabilization.

As boundary conditions, it is important to note that the tendon is fitted in the superior edge simulating its position in femur hole. The cortical bone is also fitted in its edges in order to have a stable position in reconstruction.

It is also important to define properly the geometric relations. Our model started with the surgeon (A. Maestro) guidelines, based on his own experience. These geometric relations are highly dependent on screw diameter. As far as this value is 7mm, the tibia hole is considered as 8mm and the tendon diameter as 8mm (doubled-tendon of 4mm).

All material properties were defined using the information described in section "Materials" of this paper, using in the case of the tendon, the subroutine programmed. That is, Fortran command window was used in order to call the subroutine at the same time the model was running.

The simulation was divided in four steps. The first step was used to apply the minimum longitudinal tensile load (Y-axis) to the tendon required to maintain a tensile stress over it in order to avoid the tendon could be coiled around the screw when this entity is being introduced into the tibia hole.

Second step was created to apply minimum load in transverse direction (X- axis) in order to let the screw to advance into the tibia hole without interference.

Next step is used to properly positioning, in transverse (X-axis) direction, the interference screw. It is known that there are two possibilities of ACL reconstruction. One of them is based on the use of symmetry between the tibia hole and the screw axis, making both symmetry axis coincident. With this geometry position, the tendon was forced to be double but in both extremes of the screw (Figure 8).
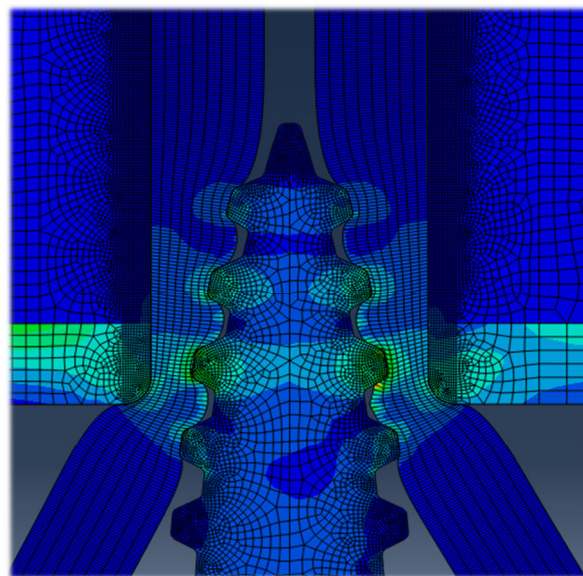

Fig. 8 - FEA model of ACL reconstruction with symmetry conditions

In this work, the other option was chosen. That is, the symmetry axis of the screw is not coincident with the symmetry axis of the tibia hole. In this case, the doubled-tendon was positioned at one side of the screw, between the trabecular bone and the cited screw (Figure 9).
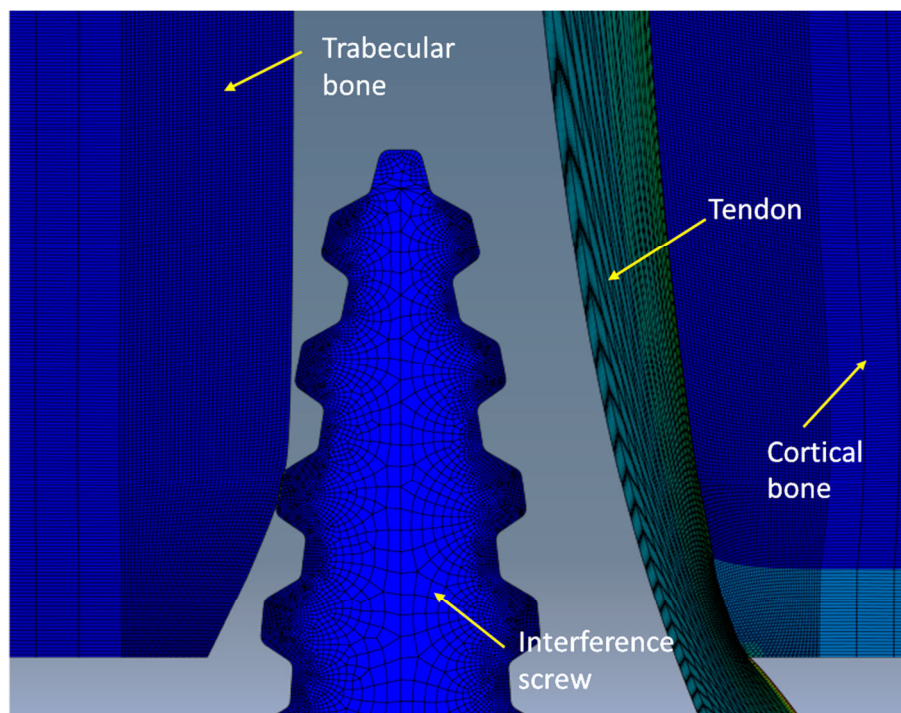

Fig. 9 - FEA model of ACL reconstruction with asymmetry conditions

In the last step, the screw is totally introduced in the tibia hole. This provokes an interference with the trabecular bone which is partly damage by the screw. However, as we said in the previous section, since the tendon is a material that only works in the fiber family direction (longitudinal), it does not suffer important damage with the screw interaction.

RESULTS

Simulation was used to determine the compressive stress in trabecular bone after screw introduction in tibiae hole during the surgery. The three cases of study were simulated with the different trabecular bone properties. The main objective is to determine how material properties of the bone affect the results of the ACL reconstruction. In Figure 10, the transverse stress in the trabecular bone during the interference screw introduction are shown.

The results of the study determine that compressive stress in trabecular bone after surgery for a middle-age patient is approximately a 30% higher than in the case of a young patient (less than 30 years). While, for old patients (older than 70 years), the stress value decreases a 70% with reference to the middle-age one. However, the displacement produced by the slip of the screw into the trabecular bone is similar for the three cases under study (Figure 11).

The final position of the reconstruction in numerical simulation model is shown below, in Figure 12.
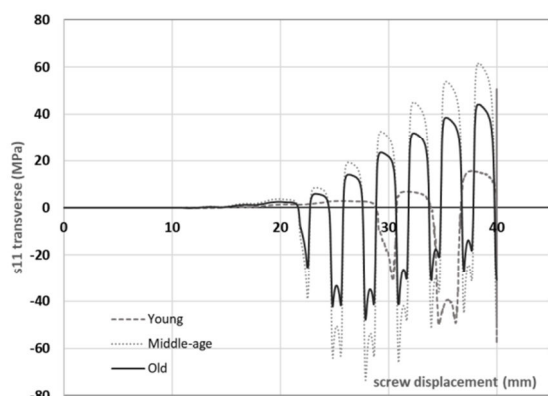
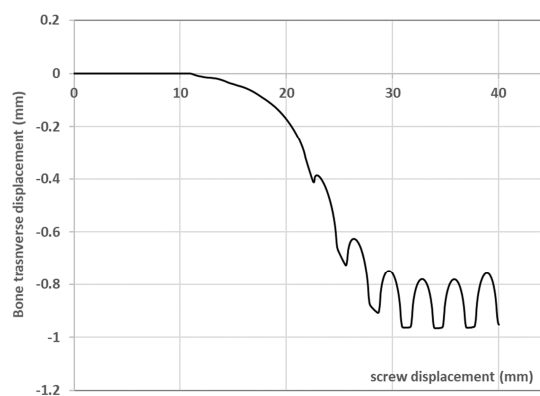Fig. 10 - Transverse Stress screw-bone interaction

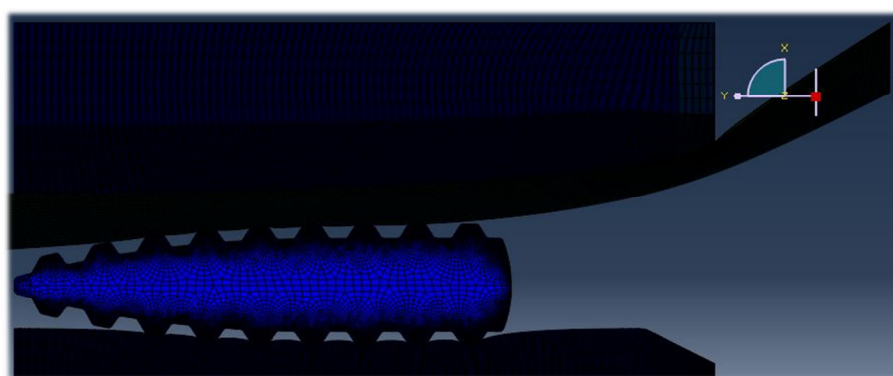Fig. 11 - Displacement produced by the slip of the screw into trabecular bone

Fig. 12 - Final position of the introduction of the screw into tibia hole

CONCLUSION

The main purpose of this work was to make a FEA model that reproduced the Anterior Cruciate Ligament reconstruction in order to know the effect of bone quality (with three age cases under study) in the reconstruction final results.

As conclusion, the worst quality of the trabecular bone is obtained at old age, when the compressive stress is lower than in the other two cases of study. It is important to remark that the middle-age patient has the best bone quality taking into account the results of this study.

Finally, the transverse displacement of trabecular bone produced by the interference screw action during its introduction on the tibia hole, is less than 1mm in all cases of study.

ACKNOWLEDGMENTS

REFERENCES

[1] C.J. Hernandez, and T.M Keaveny. A biomechanical perspective on bone quality. National Institute of Health, 2006, 39(6); pp. 1173-1181.

[2] Ming ding, Michel Dalstra, Carl C. Danielsen, Jesper Kabel, Ivan Hvid, Frank Linde. Age variations in the properties of human tibial trabecular bone. Bone Joint Surg [Br] 1997; 79-B; pp. 995-1002.

[3] Tony M. Keaveny, Elise F. Morgan, Glen L. Niebur, and Oscar C. Yeh. Biomechanics of trabecular bone. Annu. Rev. Biomed. Eng. 2001. 3; pp. 307-33.

[4] V. L. Giddings, S. M. Kurtz, C. W. Jewett, J. R. Foulds, and A. A. Edidin, "A small punch test technique for characterizing the elastic modulus and fracture behavior of PMMA bone cement used in total joint replacement" Biomaterials, vol. 22, no. 13, pp. 1875-1881, 2001.

[5] A. Dorogoy, D. Rittel, K. Shemtov-Yona, and R. Korabi. Modeling dental implant insertion, J. Mech. Behav. Biomed. Mater., vol. 68, no. January, pp. 42-50, 2017.

[6] Peña, E., delPalomar, A.P., Calvo, B., Martinez, M.A., Doblaré, M."Computational modelling of diarthrodial joints. Physiological, pathological and pos-surgery simulations". Archives of Computational Methods in Engineering 14 (1), pp. 47-91. 2007.

[7] Burstein, A.H., Zika, J.M, Heiple, K.G, Klein, L. Contribution of collagen and mineral to the elastic-plastic properties of bone. Journal Bone Jt Surg.57-A, pp. 956-961. 1975.

[8] Weiss, J.A., Maker, B.N., Govindjee, S. "Finite element implementation of incompressible, transversely isotropic hyperelasticity". Computer Methods in Applied Mechanics and Engineering 135, pp. 107-128. 1996.

[9] Ogden, R.W." Nonlinear Elasticity, Anisotropy, Material Stability and Residual Stresses in Soft Tissue." Lecture Notes. CISM Course on Biomechanics of Soft Tissue Udine. 2001

[10] Calvo, B., Peña, E., Martins, P., Mascarenhas, T., Doblaré, M., Jorge, R.N., Ferreira, A. "On modelling damage process in vaginal tissue." Journal of Biomechanics 42 (5), pp. 642-651. 2009.

[11] F. Jianping, T. Chak-Yin, and T. C.P., "Computation of flexural properties of HA/PLLA composite using a cell model approach," Acta Mech. Solida Sin., vol. 19, no. 1, pp. 18-25, 2006.