# Primary stability of uncemented acetabular liner of a total hip prosthesis using an axisymmetric finite element method based on computed tomography

**L Capitanu[1], L L Badita[2*] and V Florescu[3]**

[1]Institute of Solid Mechanics of the Romanian Academy, Bucharest, Romania
[2]National Institute of Research and Development for Mechatronics & Measurement Technique, Bucharest, Romania
[3]Institute of Civil Engineering, Bucharest, Romania

E-mail: badita_l@yahoo.com

**Abstract.** Primary stability has a great influence on behaviour during day-to-day activities and on the long-term success of uncemented acetabular components. Achieving this stability by pushing the acetabular bush (uncemented fixation) is a common practice today. In this paper, it was tried to establish a criterion for the success of fixing the acetabular cup by evaluating the short-term stability. The influence of the contact parameters established between the cup and the peripheral area of the acetabulum was investigated in this paper. These parameters depend on the intraoperative procedures: reaming and fixation. Reaming is considered without residual stress, but the acetabulum has been remodelled (a sphere has been adapted through an optimization algorithm). Uncemented fixation (press fit) involves a series of load impulses applied to the cup, simulating the impact strokes through the surgeon's hammer. Finite Element Method (FEM) was used for this simulation along with a pelvis model based on computed tomography (CT) data. Consequently, there is a postoperative state of residual effort. Assessing this state of effort could lead to a good estimation of short-term stability.

## 1. Introduction

Short-term stability of an implant is an important purpose for an orthopaedic surgeon performing a Total Hip Arthroplasty (THA) [1], [2]. Establishing some quantitative criteria, in addition to qualitative, empirical assessment, has become almost necessary. However, it is not only the primary stability (for a short term) that could be affected by the fixation level of the implant [3]. This aspect could have a big impact on what is called long-term stability [4], [5], [6], an indirect measure of the implant life.

Using cemented prostheses is probably the best way to achieve a good short-term stability. But in the last decades, more and more surgeons have abandoned this procedure on the basis that, over time, the presence of the cement layer has an adverse effect on bone development itself [7], [8], [9]. New surgical practice appeared and despite the fact that it was initially considered to be unsafe, in the recent years, almost all major orthopaedic clinics have already adopted it. Our study refers to the simulation of this surgical procedure for press-fitting a titanium cup that is part of a total hip prosthesis.

## 2. A simplified axisymmetric Finite Element Model

A simplified model with finite elements, with inhomogeneity of the material (a large volume of trabecular bone closed by a thin layer of cortical bone) was assumed. In the study of the mechanical

behaviour of the human body in contact with the prosthetic parts, there are some uncertainties that refer at the same time to the elastic behaviour, but also to the friction characteristics.

It is very important to set some general results or standards, which is almost impossible, given that the values of the parameters are very different among individuals. So, only few significant values for three parameters – Young's modulus of the spongy bone, the width of the interference (half of the difference between the diameter of the cup and the diameter of the reamed acetabulum) and the friction coefficient – are considered. It may be observed that the elastic parameters variation, the geometry variation, but also the variation of the dissipative term of the motion equation are analysed. In the same equation, by inserting the bush with constant speed, the inertial term will be cancelled.

Elastic support of the press-fitting mechanism is the spongy (trabecular) bone. It has been found that the elastic features of the bone in the acetabulum area will influence the general behaviour and, of course, the effectiveness of the press-fitting, which means – for the period immediately following the surgical procedure – the primary stability of the cup's bush. Analysing the literature data for the Young modulus of the spongy bone [10], [11], it was considered that its value varies between 0.1 and 2 GPa.

The thin layer of cortical bone (about 1 mm thick) will have no direct influence on the press-fitting mechanism. But, in an indirect way – by restricting the deformation of the spongy bone – it has a relevance that, however, has not been taken into account in this study. The value reported in [12] and used in this paper in all analyses for Young's modulus of the cortical bone is 15.6 GPa.

Fixing by means of press-fitting involves the use of a cup with a diameter larger than the reamed acetabulum. There is no doubt that a larger diameter will result in a higher interference width, which means an increased interference area. But, on the other hand, it will have to increase the size of the insertion force. Some studies show that there is a limit value for the insertion force, rather than the fact that the spongy bone undergoes permanent changes [13]. In general, the cups are chosen such that the difference between their diameter and the diameter of the acetabulum to be between 0.2 and 1 mm.

The friction mechanism that accompanies the contact has a great importance in the primary stability of the implant. By varying the values of this parameter, it was covered not only the large variety of prostheses used today, but also the possible values of the roughness of the acetabulum surface after the milling operation [14]. Prosthetic designers attempt to establish an optimal value for this friction parameter to have achievable values of insertion forces and, at the same time, a good implant stability. Friction coefficients values from 0.1 to 0.5 [15] are considered in these studies.

In the present evaluation an inhomogeneous model of simplified finite element, which includes both types of bones encountered in the real situation was used. It was an axial symmetric model with three components: the spongy bone for which flat, solid elements were used, the cortical bone (a very thin layer on the outer surface of the iliac bone), and the cup shaped with planar elements that are more rigid than the spongy bone (~ 200 ... 1000 times) or cortical one (~ 10 times).

The simplified axial symmetric model takes into consideration only a part of the iliac bone (acetabulum area) as shown in figure 1. Limits in which the movement restrictions are imposed can be seen. It is believed that the rigid cup has an imposed displacement and moves with a constant speed, as in researches made by Eko Saputra et al [16]. It was tried to evaluate the contribution of some important parameters for the variation of force that has to be applied to obtain the desired kinematic (insertion force) for the maximum effort in the contact area and ultimately for the extraction force required to extract the cup from the acetabulum, which also represents the resistance of the cup against unwanted movements.

The analytical expression of the insertion force is presented below:

$$F_i = \int_{A_c} \sigma_n (\sin \alpha + \mu \cos \alpha) \, dA \qquad (1)$$

where $\sigma_n$ is the normal stress on the contact area $A_c$, $\alpha$ is the angular coordinate of the contact surface

and $\mu$ the friction coefficient. It could be seen that the insertion force must counteract the elastic and friction forces.
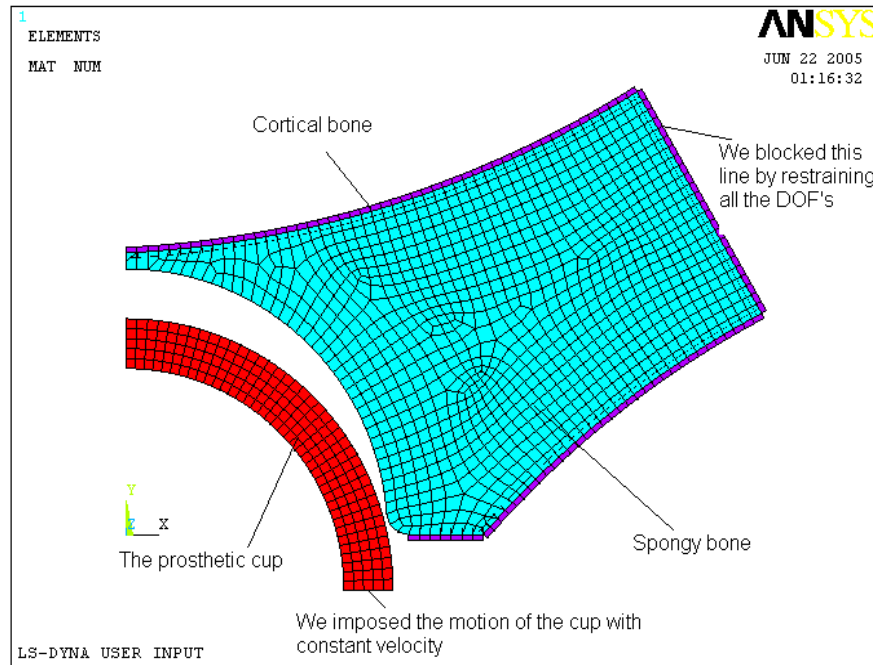


**Figure 1.** The simplified FE axial – symmetric model.

For the extraction force, it can be observed that, in this case, the elastic force due to the press fit contributes to the migration of the cup. In view of this, for the pulling force, there is the following expression:

$$F_p = \int_{A_f} \sigma_n(\mu \cos\alpha - \sin\alpha)\, dA \qquad (2)$$

where $A_f$ is the final area of interference.

The studied cases are listed in table 1. The standard values listed there for the parameters are used when other parameters are varied.

**Table 1.** Values used for analysis parameters.

| Parameter | Young's modulus of the spongy bone (GPa) | Width of the interference (mm) | Friction coefficient |
|---|---|---|---|
| | 0.1 | 0.2 | 0.1 |
| Friction coefficient | 0.2 | 0.4 | 0.2 |
| | 0.5 | 0.6 | 0.3 |
| Variation of the | 1 | 0.8 | 0.4 |
| parameter | 2 | 1.0 | 0.5 |
| | 0.1 | 0.2 | 0.1 |

## 3. Simplified FEM model results

In figure 2 the distribution of stresses and movements in the spongy bone can be observed for the following parameters: $E_{spongy}$ = 0.1 GPa; Interference = 1 mm; $\mu$ = 0.25. Of course, of all the cases studied, only one was chosen to show here the distributions of movements and stresses, for the rest of these having a similar aspect, but different values.



(a)



(b)

**Figure 2.** Distribution of the stresses (a) and displacements (b) in the spongy bone.

Variation of the insertion force when varies the elastic and friction parameters can be observed in figures 3, 4 and 5.



**Figure 3.** Variation of the insertion force dependent on the variation of the Young's modulus, *E.*
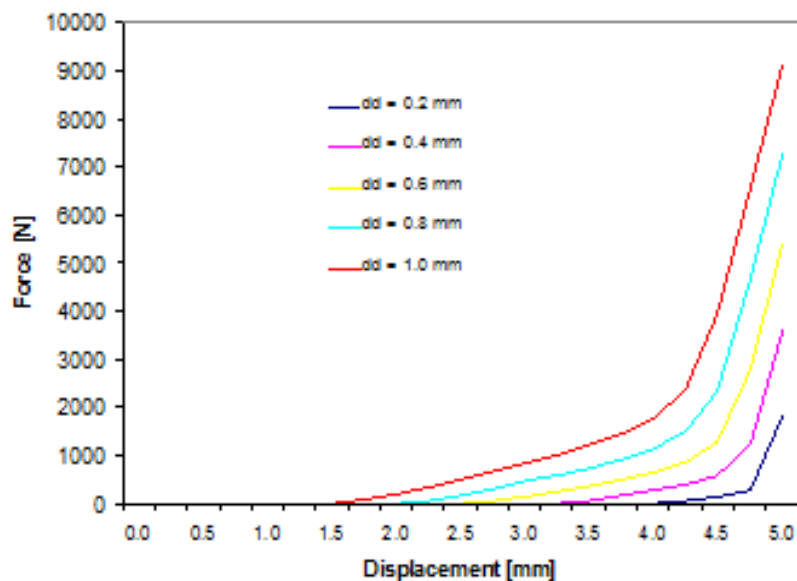


**Figure 4.** Variation of the insertion force dependent on the distance between the pole of the cup $d_{d.}$

It could be noticed that the variation of Young's modulus of the spongy bone and the interference width – figures 3 and 4, have a greater impact on the insertion force than the variation of the friction coefficient (figure 5). The contribution of the first two parameters is – as can be observed from figure 1 – in both terms of the sum below the integral. It could also be observed that there are two important stages in the press-fitting procedure. First of all, there is a linear increase of the insertion force that covers almost the entire movement of the cup (bush). At this stage, the spongy bone in the acetabulum area has an important radial deformation. In the last sequences, this radial deformation is accompanied by an elastic bending deformation of the entire spongy bone, which increases almost exponentially the required insertion force.
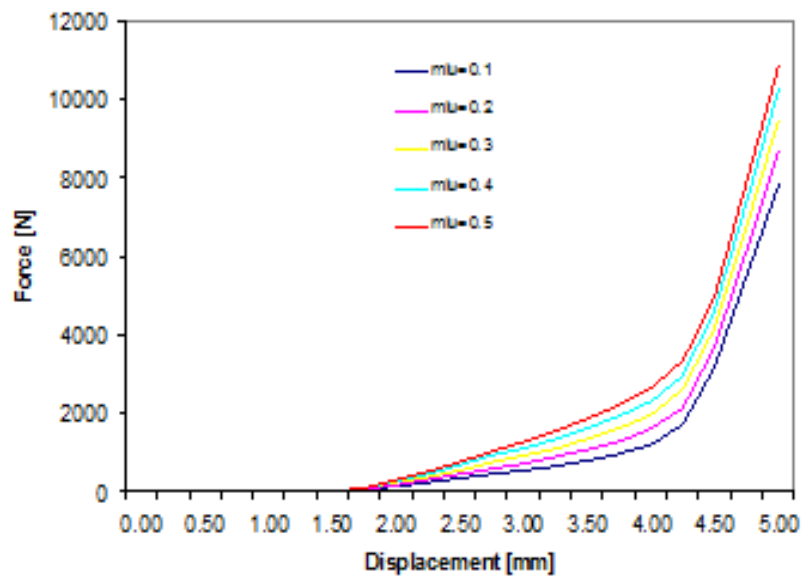
**Figure 5.** Variation of the insertion force dependent on the friction coefficient, *μ.*

In figures 6, 7 and 8, the circumferential distribution of the radial stress (in the fixing position) and the evolution of the maximum von Mises stresses are graphically represented. The remark made before still remains valid hereafter.
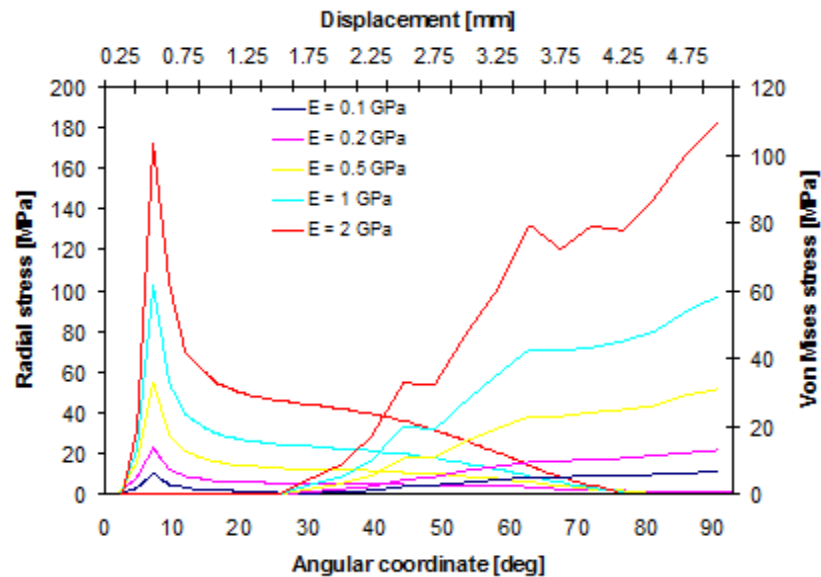


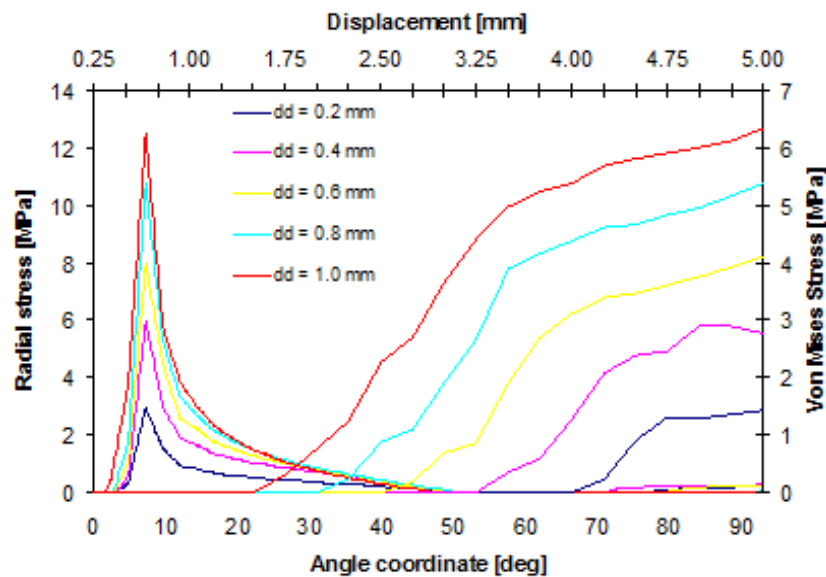**Figure 6.** The influence of the Young's modulus on the radial stress (left) and von Mises stress (right).

**Figure 7.** The influence of the interference width on the radial stress (left) and the von Mises stress (right).
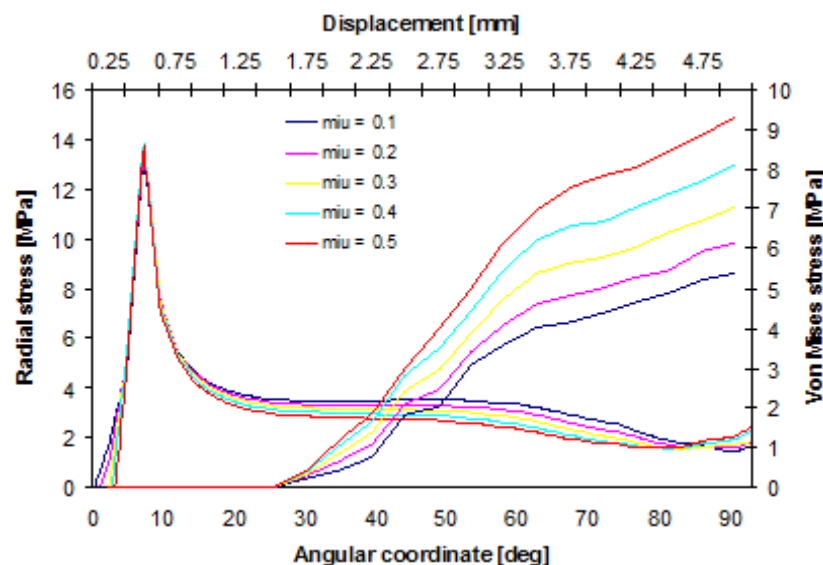


**Figure 8.** The influence of the friction coefficient on the radial stress (left) and the von Mises stress (right).

The pulling force can be calculated by Eq. (2), integrated on the entire contact surface. For example, the variation of the Young's modulus of the elastic support for contact (spongy bone) between 0.1 and 2 GPa could obtain pulling force values between 130 and 1270 N. A negative value of this pulling force means that the cup will not remain in acetabulum after the loading will be removed (the cup will be rejected by the elastic forces born by pressing into the iliac bone).

**4. FE model based on computed tomography**
Due to the anisotropic mechanical behaviour and of its complex geometry, the iliac bone is not easy to model. An important part of the recent studies refers to the possibility of obtaining some viable finite element models based on some detailed investigation methods, such as computed tomography (CT-

scanning). In this part of the work, a finite element model of the iliac bone created within an EU project, namely "Virtual Animation of the Kinematics of the Human (VAKHUM)" [17] is used.

This model is generated by the CT scans, which characterize the elastic properties (evaluated by specific tomodensitometry relationships) on a 3D mesh, obtained by approximating the real geometry of the bone. To avoid the excessive distortion of the elements, special methods are used, such as relaxation and Laplacian smoothing. Through these procedures, the entire anisotropic bone elasticity is replaced by a heterogeneous set of isotropic elements. A good image of the complexity of the model could be provided by the fact that it contains almost 15.000 knots and 13.000 elements. The material characteristics (density and Young modulus) range between the following limits [17]: $\rho = 0.741 \dots 0.780$ (g/cm$^3$) and $E = 1531 \dots 22849$ (N/mm$^3$).

The inner surface of the acetabulum was considered to be reamed, and the subchondral bone was totally removed, given that the contact was established between the outer surfaces of the cup and the spongy bone (see figure 9). A reduced model, so as in figure 10, was used hereafter.
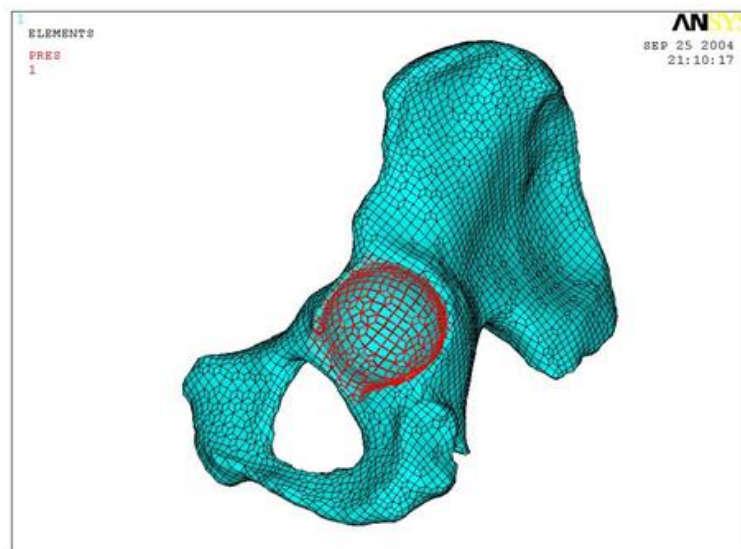


**Figure 9.** The axial symmetric model with finite element of the contact established between the outer surface of the cup and the spongy bone.
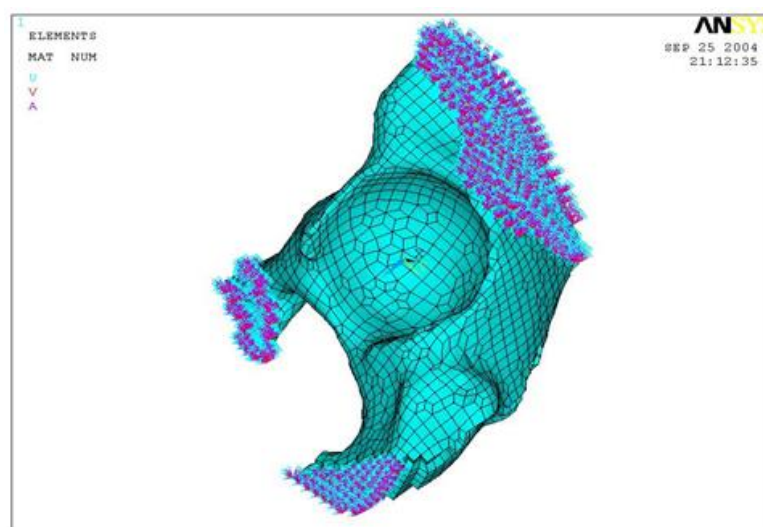


**Figure 10.** The reduced model with the kinematic conditions on nodes.

The initial position of the cup is taken into account, as in figure 11, where the following notations were introduced:

$R$ – the outer radius of the cup;

$r$ – radius of the reamed acetabulum.

In these conditions, the distance between the pole of the cup and the pole of the acetabulum (gap $d$) can be calculated by the relation:

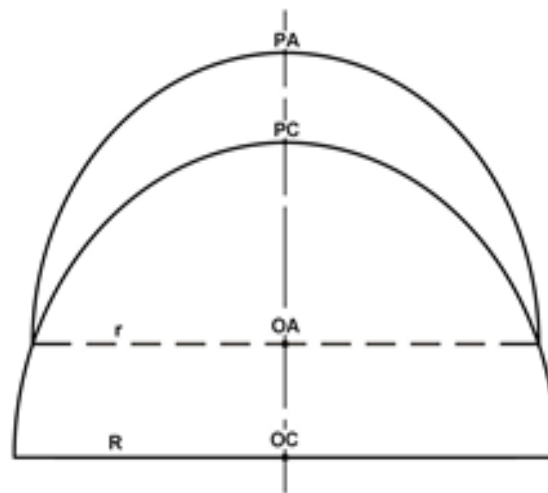$$d = \sqrt{R^2 - r^2} + r - R \tag{3}$$



**Figure 11.** Initial position of the cup on the reamed acetabulum.

It is considered that the insertion force is applied to the cup, by impact, as in the normal surgical procedure (see figure 12).
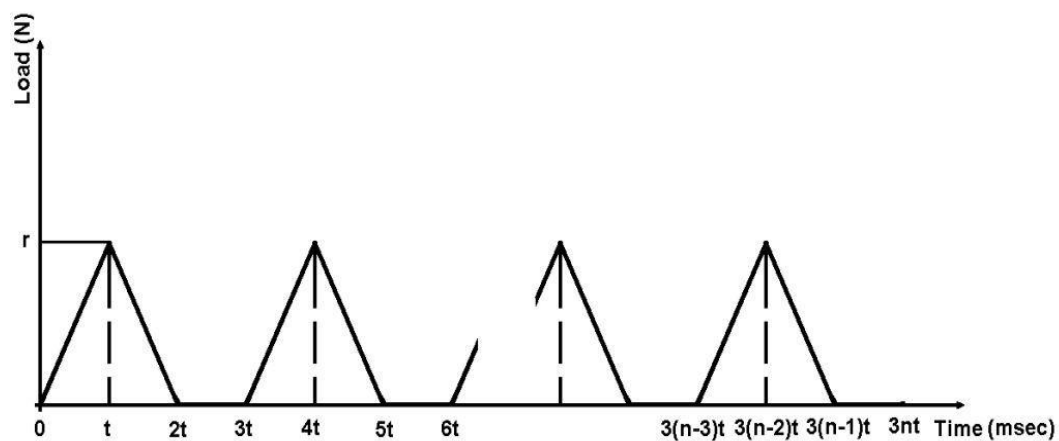


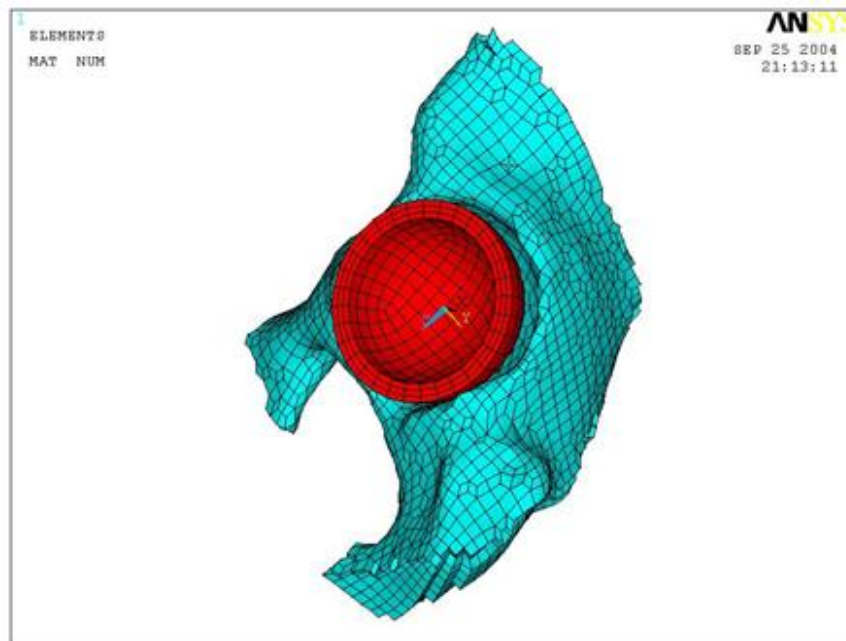**Figure 12.** Insertion force applied by impact.

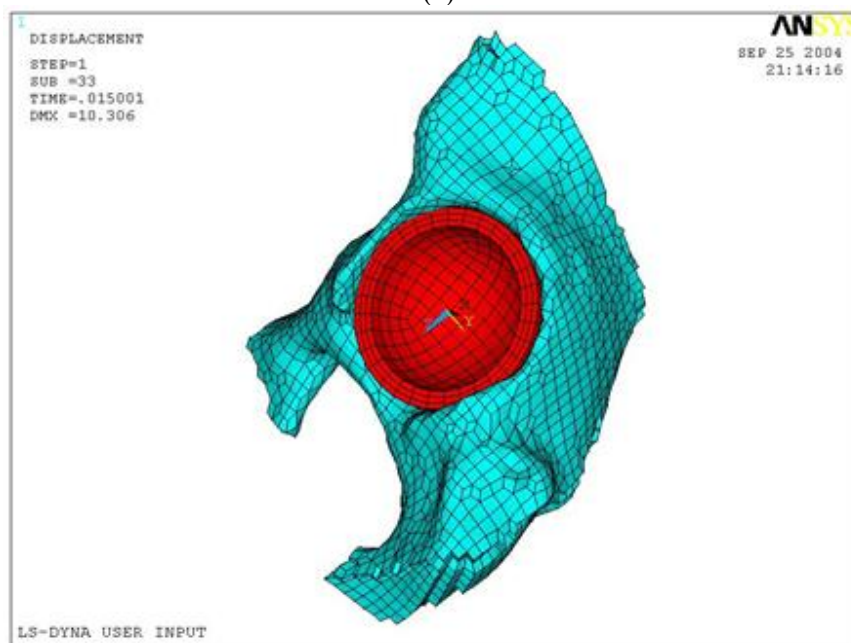In this figure the following notations are entered:

$F$ – pulse force value;

$3t$ – duration of a force pulse;

$n$ – number of the force pulses.

By varying the numerical value of the parameters ($F$, $t$, $n$) a proper load is obtained to acquire a desired displacement of the cup. The initial and final positions of the cup are shown in figures 13a and 13b, respectively.



(a)



(b)

**Figure 13.** The cup in the initial position (a) and in the final position (b)

Next, a numerical application was made.

    Analyses with finite elements were performed for the following input parameters values:
    - Loading parameters: $F = 170$ N; $t = 0.5$ msec; $n = 10$;
    - Geometric parameters: $R = 24.5$ mm; $r = 24$ mm.

The diagram obtained for external mechanical work is graphically represented in figure 14 and von Mises stresses corresponding to those shown in three intermediate loading steps are shown in figures 15a-15c.
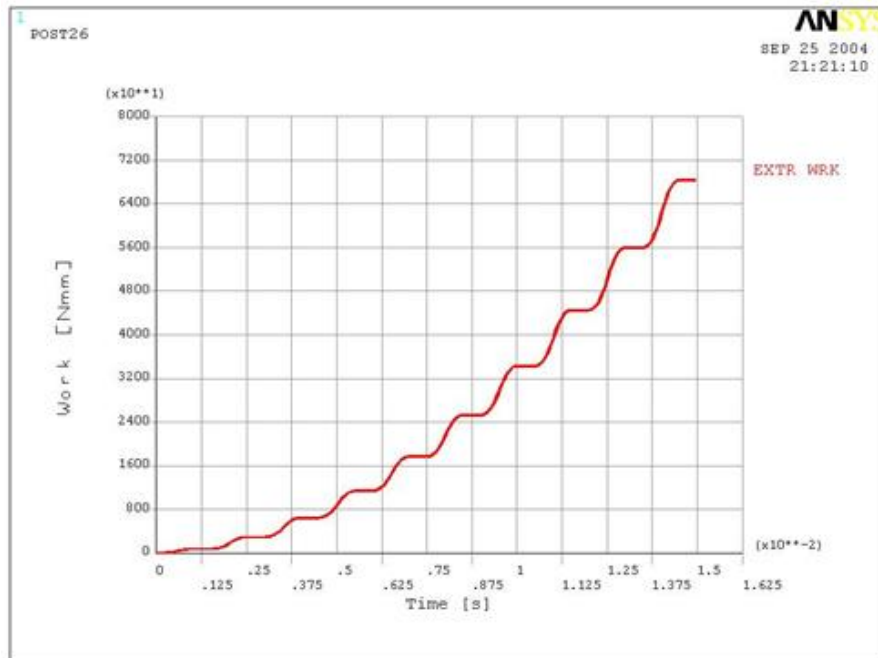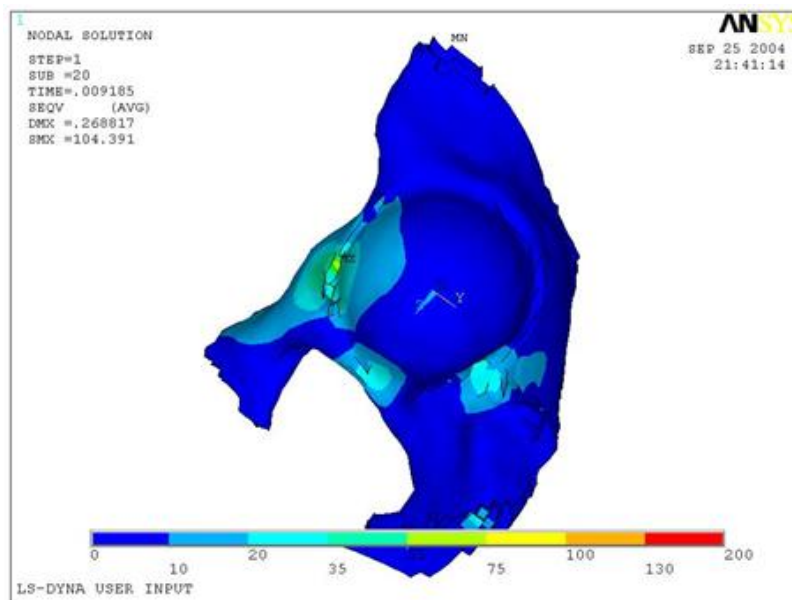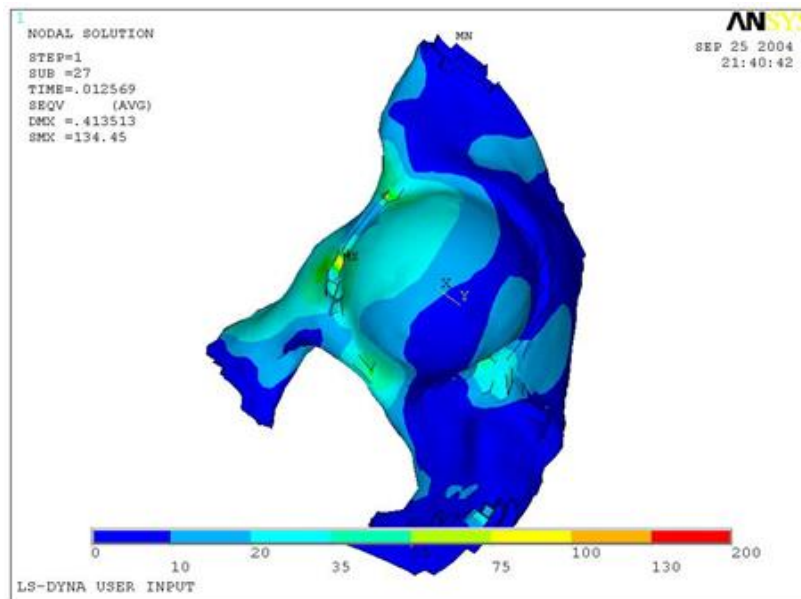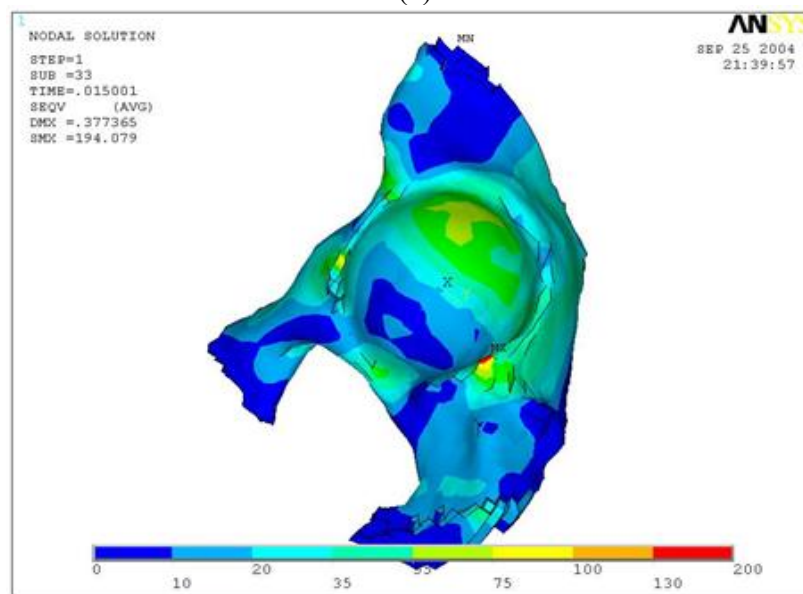


**Figure 14.** External mechanical work.



(a)

(b)



(c)

**Figure 15.** Maximum von Mises stresses corresponding to (a) 6th pulse, (b) 8th pulse and (c) 10th pulse.

Results of the analyses carried out (presented in Section 2) show the influence of the elastic parameter (Young's modulus of the spongy bone), of the geometry (the interference width) and of the friction coefficient on the required values of the insertion force, on the distribution of the contact pressure that will remain after fixation and the stress level in the elastic support of the contact – spongy bone.

There are two types of forces that characterize the cup fixation:

- insertion force;
- pulling force.

Comparing the values in figure 3 with the values calculated by Eq. (2), it can be seen that the pulling force is less than the insertion force. Indeed, for insertion, the friction at the contact surface and elastic

response of the bone act against the applied force. For pulling, only the friction has an opposite direction, the elastic force generated by the spongy bone helping to extract the cup.

## 5. Conclusions

Using even simplified models to calculate the force required for insertion, and the level of fixation given by the pull force value, FEM could be a suitable method for such sensitivity analyses and could provide a good way to quantitatively describe the phenomena that were evaluated before.

Results of the analyses carried out show the influence of the elastic parameter, of the geometry and of the friction coefficient on the required values of the insertion force, on the distribution of the contact pressure that will remain after fixation and the stress level in the elastic support of the contact - spongy bone.

Insertion force and pulling force are the two types of forces that characterize the cup fixation.

The obtained results demonstrate that the pulling force is less than the insertion force. Indeed, for insertion, the friction at the contact surface and the elastic response of the bone act against the applied force. For pulling, only the friction has an opposite direction, the elastic force generated by the spongy bone helping to extract the cup.

The application mode by impact of the insertion force, as it is really happening in the actual surgical procedure was presented. A 3D FE model was used, closer to the actual iliac bone. These numerical analyses allow the assessment of the final gap and of the maximum efforts level, depending on the impact load applied by surgeon.

## References

[1]     Phillips AT, Pankaj, Usmani AS and Howie CR 2004 The effect of acetabular cup size on the short-term stability of revision hip arthroplasty: a finite element investigation *Proceedings of the Institution of Mechanical Engineers, Part H* **218** pp 239-249

[2]     Schmitz P, Gueorguiev B, Zderic I, Pfeifer C, Nerlich M and Grechenig S 2017 Primary stability in total hip replacement: A biomechanical investigation *Medicine* **96** pp e8278

[3]     Macdonald W, Carlsson L V, Charnley G J and Jacobsson C. M 1999 Press-fit acetabular cup fixation: principles and testing *Proceedings of the Institution of Mechanical Engineers, Part H* **213** pp 33-39

[4]     Thanner J 1999 The acetabular component in total hip arthroplasty: Examination of different fixation principles *Acta Orthopaedica Scandinavia* 286 (supplement)

[5]     Iorio R, Puskas B, Healy W L, Tilzey J F, Specht L M and Thompson M S 2010 Cementless acetabular fixation with and without screws: analysis of stability and migration, *The Journal of Arthroplasty* **25** pp 309

[6]     Ni S H, Guo L, Jiang T L, Zhao J, and Zhao Y G 2014 Press-fit cementless acetabular fixation with and without screws *International Orthopaedics* **38** pp 7-12

[7]      Kinov P 2013 *Arthroplasty* (IntechOpen)

[8]     Vaishya R, Chauhan M and Vaish A 2013 Bone cement *Journal of Clinical Orthopaedics and Trauma* **4** pp 157–163

[9]     Su W, Zeng M, Hu Y, Zhu J, Wang L and Xie J 2017 Cup revision involving retention of a fixed but malpositioned acetabular component in patients with poor general conditions *Medicine* **96** pp e8622

[10]    Dalstra M, Huiskes R, Odgaard A and van Earing L 1993 Mechanical and textural properties of pelvic trabecular bone *Journal of Biomechanics* **26** pp 522-535

[11]    Madi K, Aufort G, Gasser A and Forest S 2010 Prediction of the elastic modulus of the trabecular bone based on X-ray computed tomography $6^{th}$ *World Congress on Biomechanics Singapour, Singapore IFMBE Proceedings* **31** pp. 800-803

[12]    Spears I R, Morlock M M, Pfleiderer M, Schneider E and Hille E 1999 The influence of friction and interference on the seating of a hemispherical press-fit cup: a finite element investigation *Journal of Biomechanics* **32** pp 1183-1189

[13]   Adler E, Stuchin S A and Kummer F J 1992 Stability of press-fit acetabular cups *Journal of Arthroplasty* **7** pp 295-301

[14]   Shirazi-Adl A, Dammak M and Paiement G 1993 Experimental determination of friction characteristics at the trabecular bone/porous coated metal interface in cementless implants *Journal of Biomedical Materials Research* **27** pp 167-175

[15]   Valenta J et al. 1993 *Biomechanics*, Prague: Academia Ed. pp 78-93

[16]   Saputra E, Anwar I B, Jamari J and van der Heide E 2013 Finite Element Analysis of Artificial Hip Joint Movement during Human Activities *Procedia Engineering* **68** pp 102-108

[17]   van Sint Jan S 2005 The VAKHUM project: virtual animation of the kinematics of the human *Theoretical Issues in Ergonomics Science* **6** pp 277–279