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Computer and experimental analyses of the stress state in the cement hip joint endoprosthesis body

Računarska i eksperimentalna analiza naponskog stanja tela cementne endoproteze zgloba kuka

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Abstract

Background/Aim. One of the possible complications after implantation of a cement hip-joint endoprosthesis is fracture in the endoprosthesis body. Fractures arise from overload or material fatigue of which an implant is made. The purpose of this research was to define the intensity of maximum stress and the positions of a critical cross-section in the endoprosthesis body. **Methods.** Unilaterally changing forces which act on the hip joint during walking as well as the loads result in flexible deformations of the endoprosthesis body. Biomechanical analysis of the forces acting on the hip joint determine their direction and intensity, whereas on the basis of Gruen's classification of the endoprosthesis body loosening the level of fixation is established. The bodies of cement hip joint endoprosthesis are made of cobalt-chromium-molybdenum (CoCrMo) alloy, suitable for vacuum casting, are submitted to the analysis. Analysis of the critical stress in the endoprosthesis body was performed on the endoprosthesis body by means of the finite element method. The experimental verification of the obtained results was carried out on the physical prototype under laboratory conditions. **Results.** Computer analysis, by means of the finite element method, determined the stress state by calculation of the maximum Von Mises stress and critical cross-sections for different angles of the resultant force action. The results obtained by the computer and experimental method correlate and are comparable to the results of similar analyses conducted on various endoprosthesis types. **Conclusion.** The analyses described in the paper make the basis for improving the process designing of hip joint endoprostheses and their customization to each individual patient (custom made).

Key words:
arthroplasty, replacement, hip; fractures, stress; computer simulation.

Apstrakt

Uvod/Cilj. Jednu od mogućih komplikacija posle ugradnje cementne endoproteze zgloba kuka predstavlja prelom tela endoproteze. Prelomi nastaju kao posledica preopterećenja ili zamora materijala od koga je napravljan implantat. Cilj ispitivanja bio je da se odredi intenzitet maksmalnih napona i kritičnog preseka u telu endoproteze. **Metode.** Jednosmerno promenjive sile koje deluju u zglobu kuka prilikom hoda i opterećenja imaju za rezultat elastične deformacije tela endoproteze. Biomehaničkom analizom sila koje deluju u zglobu kuka određuju se njihov pravac i intenzitet, a na osnovu Gruenove raspodele razlabavljenja tela endoproteze i nivo uklještenja. Ispitivana su tela cementne endoproteze zgloba kuka izrađene od legure kobalt-hrom-molibden (CoCrMo) pogodne za livenje u vakumu. Analiza kritičnih napona u telu endproteze izvedena je na telu endoproteze metodom konačnih elemenata. Eksperimentalna verifikacija dobijenih rezultata sprovedena je na fizičkom prototipu u laboratorijskim uslovima **Rezultati.** Računarskom analizom pomoću metode konačnih elemenata određeno je naponsko stanje kroz izračunavanje maksimalnih Von Misses-ovih napona i kritični preseci za različite uglove delovanja rezultantne sile. Rezultati dobijeni računarskom i eksperimentalnom metodom u korelaciji su i uporedivi su sa rezultatima sličnih analiza na različitim tipovima endoproteza. **Zaključak.** Istraživanja opisana u radu predstavljaju osnovu za usavršavanje procesa projektovanja endoproteza zgloba kuka i njihovo prilagođavanje svakom bolesniku posebno (*custom made*).

Ključne reči:
kuk, artroplastika; prelomi usled zamora; simulacije; kompjuterske.

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Introduction

The primary aim of contemporary orthopedics and bone and joint traumatology is to establish the function and eliminate pain in injured or degeneratively changed joints. As far as the hip joint is concerned, establishing its function usually implies installation of implants, i.e. an artificial joint.

Implantation of a hip joint endoprosthesis is the most widespread and most successful surgical procedure in orthopedic surgery, which is every year performed on 800,000 to 1,000,000 patients [1]. The success of this procedure relies on a great number of factors among which the most significant include: surgical technique, material, endoprosthesis properties and general mental and physical state of the patient [2].

Complications which may arise after a hip joint endoprosthesis implantation, and which require revision, may be classified into early complications including luxation and infection and the late ones comprising aseptic loosening, dislocation, bone tissue fractures in the endoprosthesis zone and endoprosthesis fractures.

The occurrence of fractures in the hip joint endoprosthesis body, as a complication, accounted for 1.3% of all complications due to which the revision had to be performed after the endoprosthesis implantation in the period which was analyzed from 1979 to 2010 [3]. It occurs due to material fatigue or overload and is the result of a number of factors. These include as follows: the occurrence of high stress exceeding the limits of the material strength, mechanical properties of the endoprosthesis' material, aseptic loosening and endoprosthesis geometry [4]. In order to decrease the occurrence of endoprosthesis fractures, frequent analyses are undertaken aimed at defining the load values leading to fractures as well as the zones in which this phenomenon occurs. Theoretical and experimental analyses of the intensity and distribution of load in the endoprosthesis body are demonstrated in the papers by Katouzian and Davy [4] and Bennet and Goswami [5], in the femur in the papers by Weinans et al. [6] and Schileo et al. [7], whereas in the assembly, which is the result of endoprosthesis implementation in the femur, is shown in the paper by Peters et al. [8].

In this paper the results of computer and experimental analyses of the BB2 type cement endoprosthesis are given. The principal aim of the analysis was to improve the processes of designing hip joint endoprosthesis and customize them to a specific patient.

Methods

Analysis of the stress state was conducted and the critical cross-section of the endoprosthesis body defined in the conditions of maximum load during monopedal leg support. Theoretical analyses were conducted by simulating the endoprosthesis' behaviour through the application of the finite element method in the product development software CATIA v5R21. Experimental analyses were implemented in the laboratory for material testing.

Hip joint endoprosthesis biomechanical loads

The task of a hip joint endoprosthesis is to shift the load occurring in the hip joint due to the mass of upper body part (torso) and additional loads to the lower extremity. The maximum loads in the state of quiescence occur when the body is supported by one leg (monopedal support) [9]. In that case, the external force, G_1 , (Figure 1) resulting from the weight of the upper body and unsupported lower extremity, as well as from the internal muscular force, ($\vec{M}$), acts on the hip joint. The value of $\vec{G}_1$ force is around 80% of the body weight - G_1 (equation 1).

$$G_1 = 0.8G \tag{1}$$

The direction and intensity of the resultant force ($\dot{R}$) acting in the centre of the hip joint rotation was determined by adding up the vectors of external ($\dot{G}_1$) and internal ($\dot{M}$) forces, (equation 2).

$$\vec{R} = \vec{M} + \vec{G}_1 \tag{2}$$

On the basis of the steady state (3) and geometrical requirement (4), the intensity of the muscle force $\dot{M}$ (6) as well as of the resultant force $\dot{R}$ (7) was defined.

$$\vec{M} \cdot \vec{r} = \vec{G}_1 \cdot \vec{l} \tag{3}$$

$$l = 2 \cdot r \tag{4}$$

$$\vec{M} \cdot \vec{r} = \vec{G}_1 \cdot 2\vec{r} \tag{5}$$

$$\vec{M} \cdot \vec{r} = 2 \cdot \vec{G}_1 \tag{6}$$

$$\vec{R} = 3 \cdot \vec{G}_1 \tag{7}$$

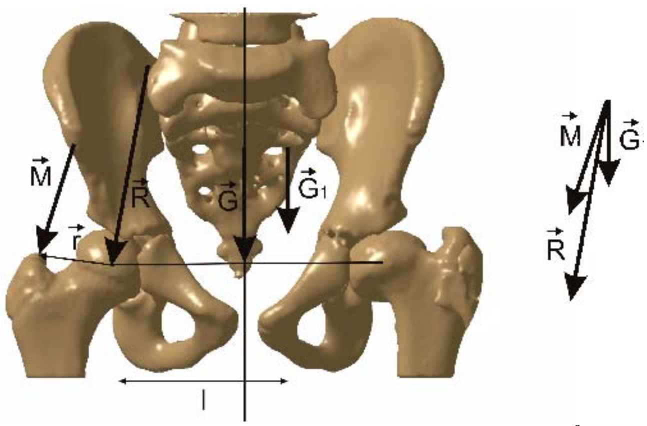

Fig. 1 – Distribution of load in monopedal support [9].

In motion, dynamic forces which occur in specific phases of the step are variable and significantly greater than the static ones. For the body mass of 100 kg the reduced maximum load of the hip joint in motion is around 4,000 N [9]. Figure 2 shows the load variations on the hip joint in specific phases of the step.

Hip joint endoprosthesis body modeling

Computer models obtained by discretisation of geometric models generated in the computer aided designing software's (CAD) [10] constitute the basis for analyses by means

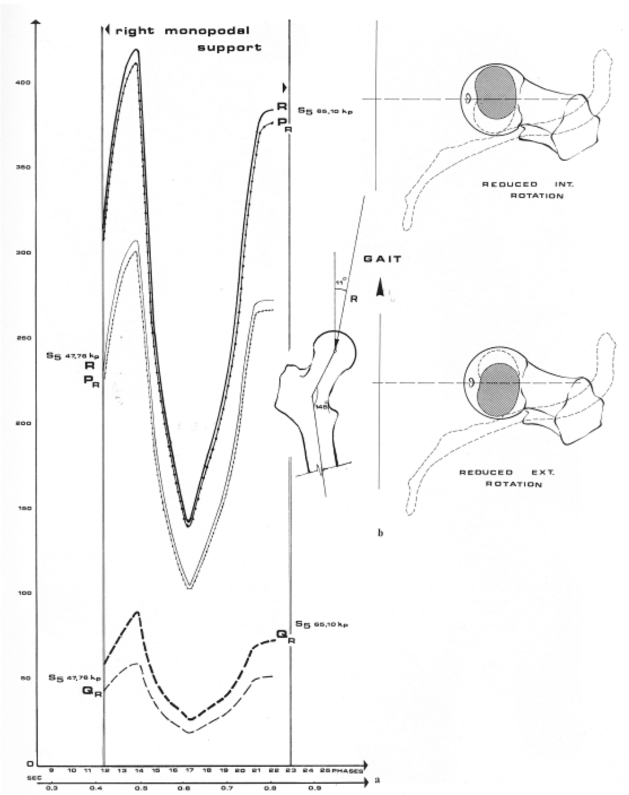

Fig. 2 – The hip load variations in different phases of the step [9].

of the finite element method, which results in creation of the mesh of three-dimensional finite elements. Figure 3 shows the geometric (3a) and discretized (3b) model of the hip joint endoprosthesis.

from the biomechanical analysis, whereas the experimental part of the analysis was conducted on the physical endoprosthesis model.

In computer simulation the properties of the material of which the physical model of the endoprosthesis body BB2 was made was adopted Cobalt-Chromium-Molybdenum (Co-Cr-Mo) alloy (ISO 5832/IV 1978) which is treated as a linear isotropic material in FEM analyses whose mechanical properties are shown in the Table 1.

Table 1

Mechanical properties of the Co-Cr-Mo material

Parameters	Values
Flexibility module (MPa)	211
Paoson coefficient	0.33
Yield limit, $R_{0.2}$ (MPa)	680
Tensile strength, R_m (MPa)	720

Co-Cr-Mo – Cobalt-Chromium-Molybdenum

In the maximum stress analysis the most unfavourable case was adopted as a high fixation when, according to Gruen's zones of loosening, the endoprosthesis body was fixed in the zones 3, 4, and 5 (Figure 4a), that is, 1/3 of the height distally. Figure 4b shows constraints in the discretised model of the endoprosthesis body.

Computer analysis of the stress state (static behaviour) of the endoprosthesis body

In the phase of simulation of exploitation conditions the forces and constraints obtained by biomechanical analysis of the human locomotor system was defined. These included as follows: the force intensity of 4,000 N acting on the surface of the conical element of the endoprosthesis' proximal part

a) b)

Fig. 3 - Models of the hip joint endoprosthesis body: a) geometric, b) discretised

To define the geometric model of the endoprosthesis body the appropriate modules of the CAD software Dessault Systems Catia V5r21 were used for surface (Generative Shape Editor) and solid modeling (Part Modeling), whereas for discretisation and analysis of the static behaviour the Generative Structural Analysis module was employed. The discretised model consists of solid isoparametric finite elements of hexahedron of the 2 mm side length, thus having the endoprosthesis body circumscribed by the mesh of 26971 of finite elements.

Maximum stress analysis

Maximum stress analysis of endoprostheses was performed by means of computer simulation of the static behavior of endoprosthesis and experimental verification of the obtained results. Simulation has been carried out by employing the finite element method for the conditions resulting

under the angle of α[7] (Figure 5). The femoral inclination angle (as well as of the anatomical endoprosthesis axis) is 6° in relation to the central axis. The degree of fixation equals 1/3 of the endoprosthesis body height distally. The variable value in the analysis is the resultant force action angle ranging from 15–20°. The analysis examined the position and intensity of maximum Von-Mises stress on the endoprosthesis body.

Analysis of the stress in the endoprosthesis body

The experimental definition of the stress in the endoprosthesis body was performed under laboratory conditions on the physical model cast in vacuum using the Cr-Co-Mo alloy. Stress measurements were carried out by means of strain gauges LY 1 × 3/120, manufactured by HBM, the one of which was placed on the lateral surface at the point maximum stress according to the computer analysis, and the two

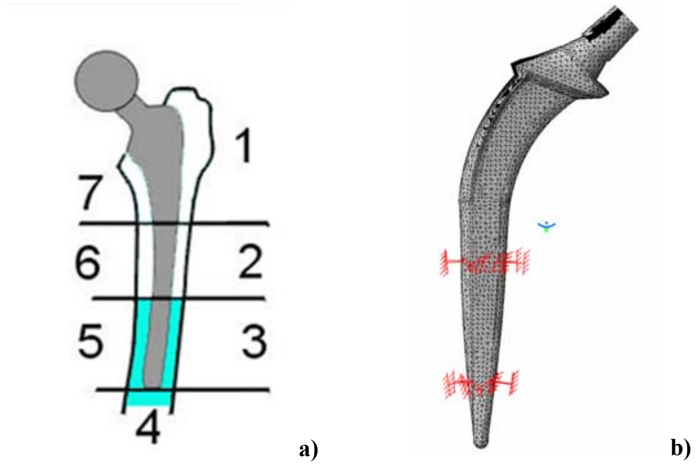

Fig. 4 – Fixation of the endoprosthesis body according to: a) Gruen zones of loosening; b) the computer model constraints.

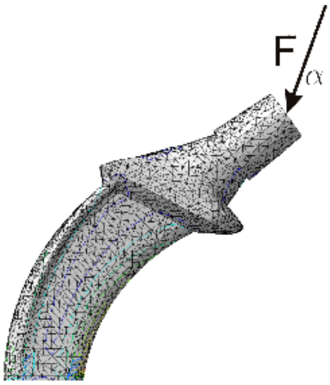

Fig. 5 – Direction of the load action in the analysis of the static behaviour.

of them on the lateral sides at the same level. The physical model was fixed by epoxy resin in the metal holder at 1/3 of the height distally. At the tensile strength testing machine (Figure 6) the resulting load was 4,000 N with the action direction corresponding to the direction angle of the resultant force action of 20°.

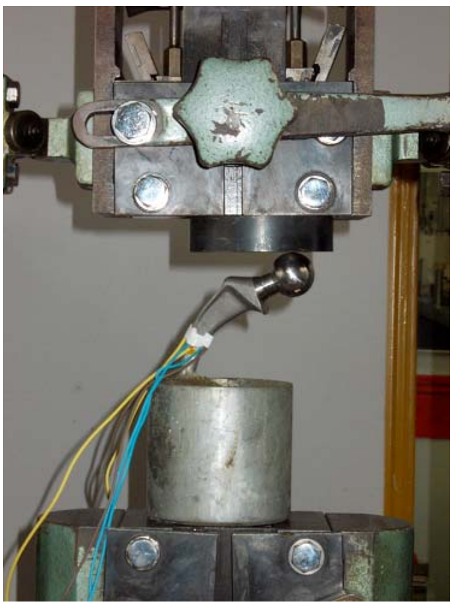

Fig. 6 – Experimental stress analysis.

Measurement of the stress variations resulting from the endoprosthesis body deformations was made by means of the corresponding DMS acquisition device, while the extent of deformations was measured by a displacement sensor of HBM W103 type.

To determine the real values of boundary stress at the critical cross-section of the hip joint endoprosthesis body, the samples were cut out of the physical model and a test specimen made (Figure 7) and subject to tension up to the point of tearing using the tensile strength testing machine.

Fig. 7 – A test specimen resulting from cutting the physical model out.

Results

Analysis of the stress state of the endoprosthesis body, under the set conditions, by applying the finite element method proved that the maximum Von Mises tension stress appears in the dorsal surface of the endoprosthesis body.

Figure 8 shows the position of the maximum Von Mises stress within the critical cross-section zone. The spectral diagram in the same figure demonstrates numerical stress values depending on the color.

On the basis of analyses the resulting maximum equivalent stress is 449 MPa (Table 2), which was significantly lower than the yield stress for Co-Cr-Mo alloy (720 MPa) according to the Table 1). Therefore, it was concluded that the reductions in the load action angle would not lead to exceeding the maximum acceptable stress values for the given material of the hip joint endoprosthesis body, but to significantly influence the stress increase at the critical cross-section. In all the conducted analyses the critical cross-sections were nearly in the same position (Figure 8, marked by a circle) coinciding with the common place for endoprosthesis' fractures (Figure 9). Decreasing the angle of the resultant force action direction results in increasing the stress at the critical cross-section (along with the linear dependency), which is evident in Figure 10. This means that if the endoprosthesis is placed in "Varus" position, implemented femoral head with longer neck or increased "offset", the angle of the resultant force action direction decreases while the stress at the critical cross-section increases.

Table 2

Maximum equivalent Von Mises stress

Load angle α (°)	Maximum equivalent Von-Mises stress, σ_{ekv} (MPa)
15	449.5
16	431.7
17	413.8
18	395.8
19	377.7
20	359.4

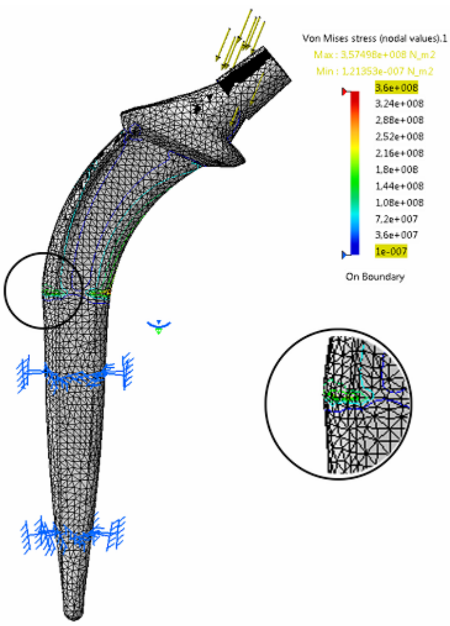

Fig. 8 – Equivalent Von Mises stress.

Fig. 9 – Endoprosthesis body fractures due to material fatigue.

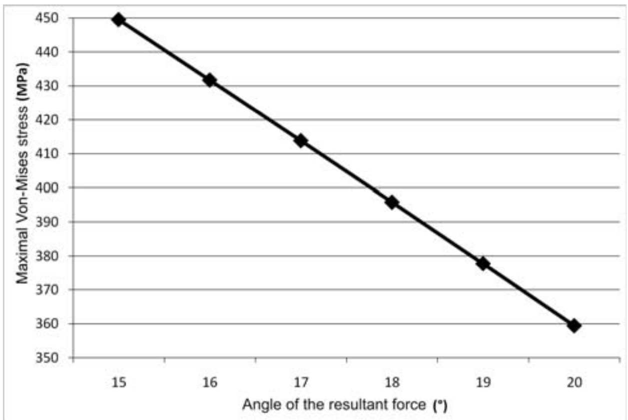

Fig. 10 – Dependency of the stress at the critical cross-section on the force action angle.

Based on experimental measurement of the strain gauges dilatation in critical zones, it may be established that the stress on the dorsal endoprosthesis surface at the critical cross-section was linearly increasing with the increased load, while it remained constant on the lateral sides, corresponding to the stress state of the computer model obtained by the FEA method application. By conversion of the obtained results on the basis of common relations for Hooke's law, the maximal stress on the endoprosthesis body was calculated (420 MPa).

Experimental analysis on the tensile strength testing machine determined the yield strength and tensile strength of the material. The resulting values were $R_{p0,.2}$ = 652 MPa for the yield strength and R_m = 695 MPa for the tensile strength.

Discussion

Evaluation of the achieved results may be performed through analysis and critical reflection on the results obtained by simulation of the behaviour of the endoprosthesis computer model by means of the finite element method and experimental verification.

In accordance with vector analysis of the composition of forces acting in the hip joint under the static conditions for the monopedal support, the resultant force intensity ($\dot{R}$) acting in the hip joint is $4 \cdot \dot{G}_1$ while $\dot{G}_1$ is the load resulting from the torso mass and the mass of the leg which is in the air. The total load in this case is $3 \cdot 0.8 \cdot G$, i.e. approximately 2,400 N. Under the dynamic conditions, the maximum value of the resultant force acting in specific phases of the step is where G is the body mass weight. The higher value was used as a referential parameter in the computer model analysis and during the laboratory verification with the assumption that the patient body mass is 100 kg (as well as the resultant force intensity of 4,000 N).

Simulation of the endoprosthesis body's behaviour under exploitation conditions points to the fact that the maximal Von-Mises stress on the endoprosthesis body is significantly lower than the acceptable stress of Cr-Co-Mo alloy which is used for manufacturing of BB2 endoprostheses. The resulting values (360–450 MPa) match the values obtained by means of the finite element method, for the six endoprostheses types (150–600 MPa) [5]. The indicated differences in numerical values also result from the properties of the material used for the endoprosthesis manufacturing. In the computer model the material properties' values were used from the relevant literature and they differ according to the experimental measurements and in the specific example are lower by around 3.6%. In addition, the critical cross-section zones in BB2 endoprosthesis also coincide with the ones in the mentioned paper and corresponded to the point where the force acting on the endoprosthesis established the maximum bending moment [11].

A significant correlation between these factors was on the basis of impact analysis of the change of the resultant force action direction angle on the stress state at the critical cross-section.

The results of experimental measurements of the maximum stress intensity (420 MPa) are within the boundaries of the results obtained by the finite element method (deviations are within the range of 1.5–14.4 % depending on the load action angle). The measured values are significantly lower than the values of the material properties of which the physical model is made, which is also supported by experimental analysis of the material. The noted deviations in the results obtained by the theoretical and experimental analysis arise, on one hand, from the difference between the real data and those found in the literature on material properties and on the other, from the objective effects in experimental analyses including as follows: the dimensions of the used strain gauges (strain gauges are significantly larger than the maximum stress zone which leads to errors in defining the stress), bone cement [12] and the problem of precise positioning of the endoprosthesis during measurements (load angle error).

Future analyses should also encompass the effects of bone cement on the connection between the endoprosthesis and the femur, as well as the mechanical properties of the bone tissue itself [4, 6].

Conclusion

The results obtained in this study support the hypothesis that the application of theoretical analyses of endoprostheses by means of the finite element method is a reliable alternative to experimental analyses. The results of the finite element method application have a significant role in defining the shape and selection of the implant material, both from the medical and engineering point of view. From the orthopedic point of view, it is possible to analyse the effect of the surgical technique used for the endoprosthesis implantation. The computer model offers an opportunity to analyse the impact of various load actions on the endoprosthesis. In contrast, the application of these methods in the development of endoprostheses enables engineers to optimize the shape and dimensions of endoprostheses according to the requirements of surgical method, the type of disease and the patient himself/herself.

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