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Effect of elliptical deformation of the acetabulum on the stress distribution in the components of hip resurfacing surgery

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Hip resurfacing surgery is a matter of controversy. Some authors present very good late results of 99% survival outcomes. However, national records of implants point to the series of complications connected with biomechanical flaws of the implant. These results implicate the experimental research on biomechanical properties of HRS.

The aim of the research was to define the nature of cooperation between the components of hip resurfacing surgery (HRS) and the influence of the deformation of acetabulum, the size of the implant and the nature of the bone surface on the stress distribution in the acetabulum and the femoral component.

The calculations were run with the use of the finite element method (FEM), using the ANSYS bundle for this purpose. Four decrete models of the studied system were made: a model with the elements of the system connected with glue, a perfect spherical model with cooperating surfaces, a model reflecting an elliptical deformation of the acetabulum, and a model with different sizes of the implant.

The results indicate that the stress values obtained for models with the ideally spherical acetabulum cannot cause significant deformation of cooperating implants. In the case of loads of the elliptically deformed acetabulum significant point stress concentrations can be observed in the spots of joint. The size of the acetabular and femoral components of HRS has influence on the stress concentration on the internal surface of the acetabulum as well as in the bone tissue surrounding the madrel of the femoral component. Moreover, physical properties of the base surface surrounding the HRS components have influence on the size of stress in the acetabulum and the femoral component.

Key words: *hip resurfacing, stress distribution, finite element analysis*

1. Introduction

Hip arthroplasty is considered to be one of the greatest achievements in orthopedic surgery. Despite constant technological development, especially the improvement of tribological conditions, the wear of the elements of the prosthesis still remains a key problem that limits the lifespan of the implant to 10–15 years [12]. Every year there is a rise in the number of performed arthroplasty surgeries. 332 000 hip arthroplasties were performed in the United States in 2010 [30], and 76 500 in Great Britain in 2012 [31]. Along with the rising number of surgeries, the number of complications increases. In 2013 in Great Britain 10 000 revision surgeries were performed, which constitutes 12% growth in the number of these surgeries.

In the last 20 years, resurfacing endoprosthesis has become one of the most popular types of implants. The concept of recreating the surface of hip joint using different materials dates back to 20th century [14]. Smith–Petersen applied a cobalt and molybdenum

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alloy [26]. Charnley continued his experimental study using polytetrafluoroethylene (PTFE) in the beginning of his experiments with the good result of 2–3 years. However, the complications led Charnley to developing the theory of low friction arthroplasty [7]–[9]. Tribological properties of the work of a metal head working in the polythene acetabulum and a big surface of contact resulted in 8-year lifespan of only 40% of implants [4], [5]. The new generation of metal-on-metal hip arthroplasty takes origin in the 1980s when Ring and McKee-Farrar applied big metal surfaces of a joint [2], [25]. In February 1991 McMinn implanted first metal-on-metal resurfacing prosthesis and interim results showed only 1 revision out of 446 HRS [18], [10], [24]. Along with the increasing number of HRS and prolonged observation period, the increasing number of complications requiring reoperation was noted, among which the most important included necrosis of the femoral head, femoral neck fractures and formation of pseudotumors. The National Records of Endoprostheses during 3 years of observations showed the necessity of revision in 4.5% of HRS (1.3% of THA) in Great Britain and 2.7% in Australia. The majority of revisions concerned young and physically active people [1], [21].

Clinical analysis and biomechanical research have proven that the revisions may be caused mainly by tribological features of MoM surfaces different from those theoretically assumed, and from the distribution of forces on the surfaces. A strict corelation ($p < 0.0001$) between the structure of HRS and the number of revision surgeries was proven. In the case of BHR (Smith & Nephew) 3.3% of revisions were observed and in the case of ASR (DePuy) – 7,5%. In the case of ARS, the femoral component was subjected to heat treatment and the acetabulum component had reduced value of the acetabulum inclination angle [16]. These parameters caused the stress concentration on a small surface and increased grindability. In the case of smaller sizes of HRS, 44 mm or smaller, significantly higher risk was observed in 7-year-long study than in the case of the implants of bigger sizes (48 mm and bigger) [1]. The method of implantation has also an influence on the work of the MoM surface. According to experimental studies on cadavers, the press-fit technique may lead to the deformation of the acelabulum and the deformation of the spherical surface of MoM joint [13]. Correct tribological factors, such as sphericity, the carbon content, reduced clearance, and the roughness of the surface, were disturbed, resulting in the increase of wear of cooperating MoM surfaces and the number of metal ions emitted to surrounding tissues and blood [11], [17]. The result of disturbed tribology is formation of a pseudotumor that destroys surrounding bone tissue.

The aim of the study was to analize the character of cooperation between the components of hip resurfacing surgery (HRS): the influence of the elliptic deformation of the pelvis component of endoprosthesis – the acetabulum – on the value and stress distribution in the acetabulum and femoral component – the head, and the influence of the values of geometrical parameters of the implants (diameter) and the type of base on the value and distribution of stress on cooperating surfaces was evaluated.

2. Material and method

In this paper the cases of spherical acetabular (Fig. 1) and elliptically deformed acetabular were analysed. The elliptical deformation is caused by the change in the axial length of the cross-section of the acetabulum. An elliptic cross-section of the following dimensions was obtained: longer axis (A_L) bigger by 1 mm and the shorter axis (A_S) smaller by 1 mm than the diameter of the ideally spherical acetabulum (Table 1).

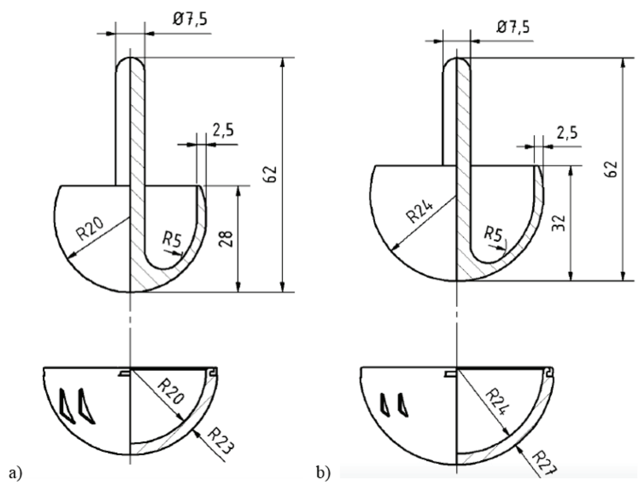

Fig. 1. The geometry and dimensions of the head and the acetabulum: a) the head – small diameter: ∅ 40 mm, b) big diameter: ∅ 48 mm

Table 1. Geometric parameters of the models analyzed implants

Model	Shape	Diameter
S1	Spherical	f = 40 mm
S2	Spherical	f = 48 mm
E1	Elliptically deformed	A_L = 39 mm. A_S = 41 mm
E2	Elliptically deformed	A_L = 47 mm. A_S = 49 mm

The calculations were run with the use of the finite element method (FEM), using the ANSYS software. In the proposed model, the pelvis has the shape of a hemisphere with a spherical socket inside, where the acetabulum is placed. The model of the femur has cylindrical shape with a spherical end that reconstructs the surface prepared for the femoral component (Fig. 2). A linear elastic model of material was adopted to model the acetabulum and the femoral component. The mechanical features of the ASTM F90 alloy were adopted. The linear elastic model of material was also adopted for the volumes reconstructing bony elements. Mechanical properties of the materials applied in the model are presented in Table 2.

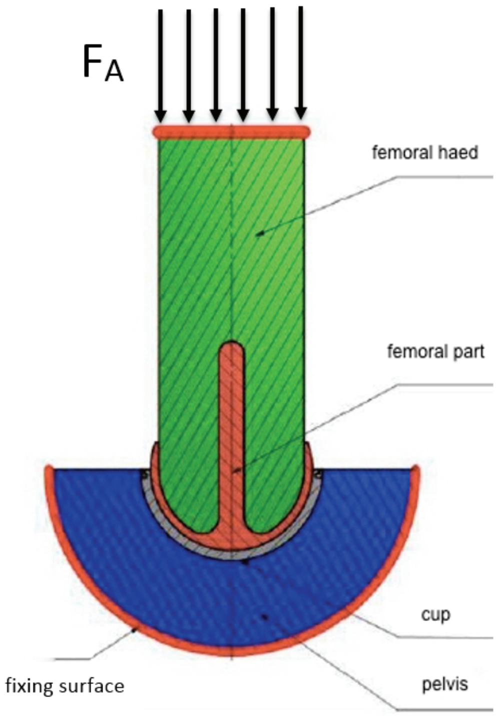

Fig. 2. Diagram describing support and load of analyzed model. Particular elements of the model were marked in the figure

Table 2. The mechanical properties of materials applied in the model

Material	*E* [GPa]	*v* [–]
Alloy CoCr ASTM F90	210.0	0.3
Trabecular bone	1.0	0.3
Cortical bone	17.3	0.3

The geometric model was divided into the finite elements mesh. The solid element type heksa 20 was used. The element has 20 nodes and 3 degrees of freedom in every node (u_x, u_y, u_z). The nodes on the external spherical surfaces of the solid modeling the pelvis were devoid of all degrees of freedom, immobilizing the solid, which is a necessary condition required by the applied calculation method. Additionally, the conditions of symmetry were imposed on the nodes on the surface passing through the axis of the model. The load with the use of force F_A was applied to the upper base of the solid simulating the femur (Fig. 2). The value of the load was adjusted based on literature and equaled 4000 N, which is the maximum value of the resultant force on a hip joint during walking, on the stage of the load of one lower limb [3]. A displacement along the axis was applied to the same nodes that were loaded with the use of force in the previous step. The value of the displacement was adjusted based on the results of the simulation of a load with the use of force. This proceeding was necessary to define the parameters of the load in the target model that took into account the contact mechanics between the head and the acetabulum. When taking into account contact problems, the only way of implementing the load is the enforcement by the displacement of one of the cooperating elements. In the presented case, the first calculation step was performed taking into account the load with the use of force in order to define the value of the displacement of the head with regard to the acetabulum, caused by a defined force. The obtained value was applied in the second calculation step that took into account the contact mechanics as the value of the enforcement of the load on the studied system. On the surfaces of the joint a symmetric and deformable surface contact was applied. The stiffness coefficient of the contact equaled 1. The adopted coefficient of friction equaled 0.05. The backlash between the surfaces was eliminated (the contact pairs were closed) and the preliminary blending of the volumes was excluded.

In addition, the influence of mechanical properties of the bone in which the acetabulum is embedded on the values and distribution of stresses in the acetabulum and the femoral component was analyzed. To that end, the material properties of coritical bony tissue or cancellous tissue were ascribed to the elements belonging to the support volumes.

3. Results

Based on the calculation the stresses distributions in femoral components and acetabulum were determined. The impact of implant size (diameter) and elliptical deformation of the acetabulum on the value of reduced stress on the working surfaces of the femoral components and the acetabulum were analyzed. In the case of perfectly spherical acetabulum the maximum stress values did not exceed 30 MPa. It was noticed that in the smaller diameter implant (∅ = 40 mm), the stress on the inner surface of the acetabulum was by about 25% higher than that of the implant with diameter ∅ = 48 mm (Fig. 3).

Character and value of the stress on the working surface of femoral components and acetabulum also depend on the type and the mechanical properties of the bone in which they are embedded. When the substrate represents properties corresponding to cancellous bone tissue, the resulting stresses are lower than in the case of a substrate having the mechanical properties of cortical bone (Fig. 4).

The main problem, which is addressed in the article was the question of the effect of deformation of the elliptical acetabulum on the state of stress in the working elements of femoral components and acetabulum. The calculations show that the elliptical deformation of the acetabulum results in a quintuple increase in the stress on the surface of the cap and the acetabulum. A very uneven distribution of stress is characteristic, with a maximum value at the outer edge of the femoral components and the acetabulum,

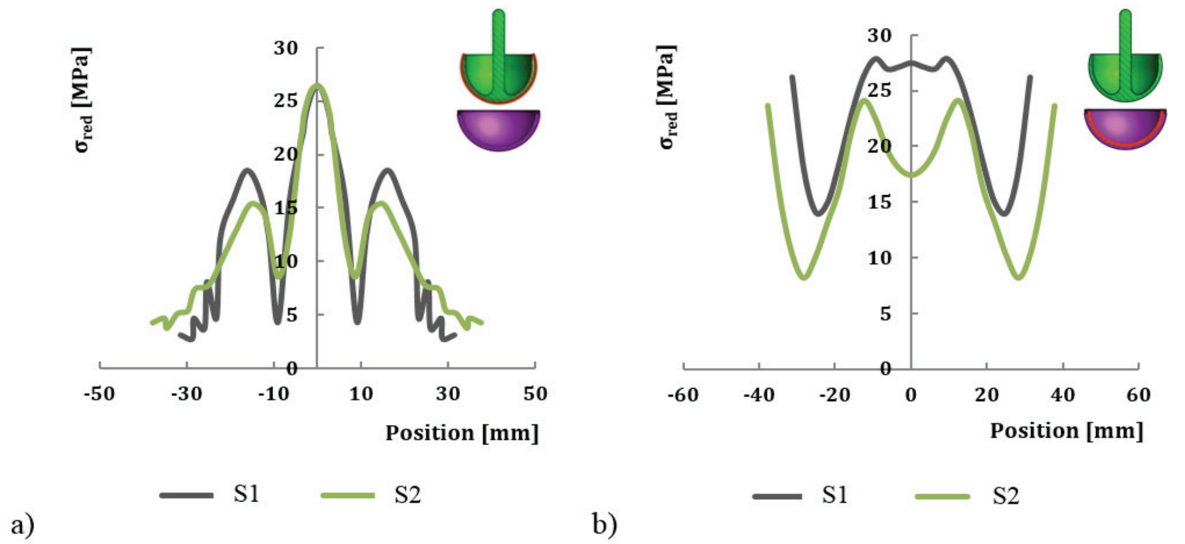

Fig. 3. The comparison of the stress values on the external contour of the head (a) and internal contour of the acetabulum (b) for different diameters of the head (∅ = 40 mm, ∅ = 48 mm)

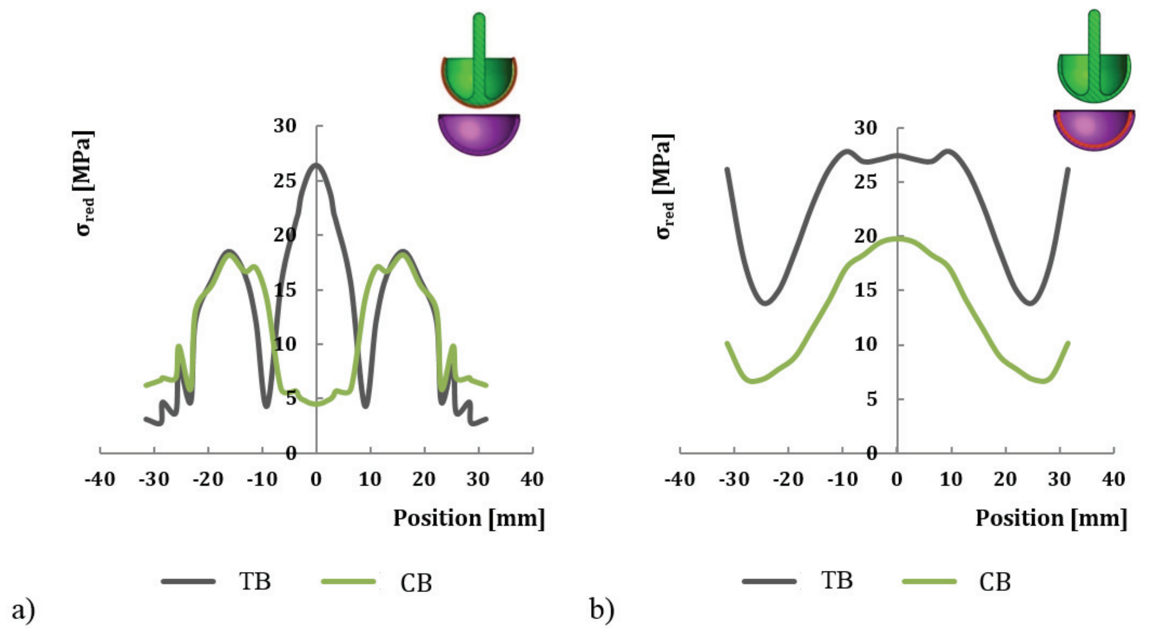

Fig. 4. The comparison of the stress values on the external contour of the head (a) and internal contour of the acetabulum (b) for different types of supporting material (TB – trabecular bone, CB – cortical bone).
The case of prosthesis with the head diameter ∅ 40 mm

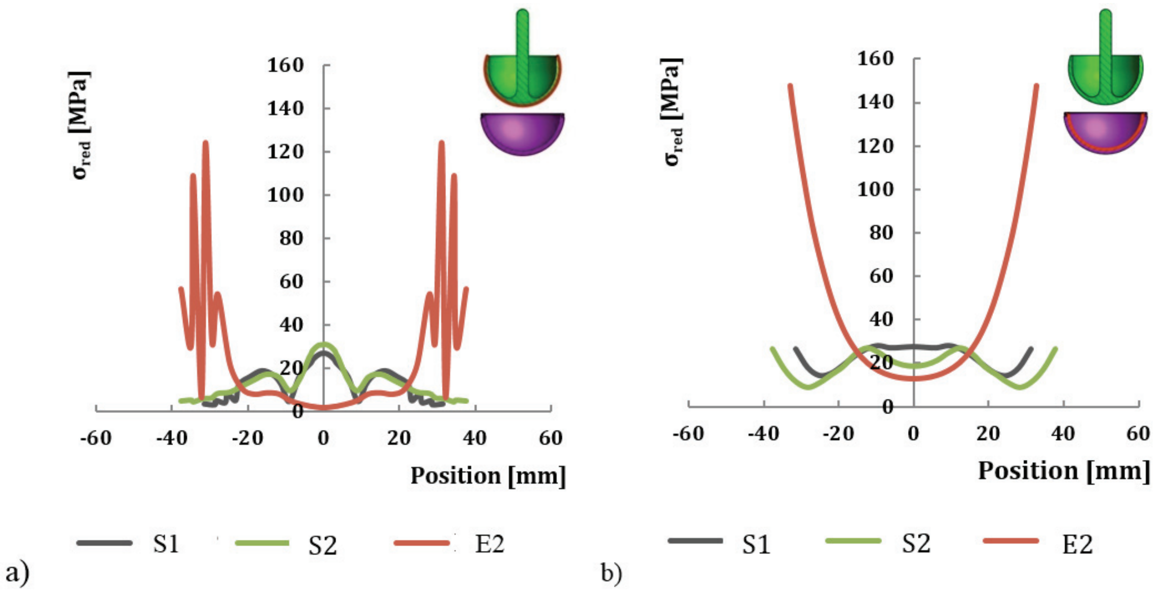

Fig. 5. The comparison of the stress values on the external contour of the head (a) and internal contour of the acetabulum (b) for the spherical S1 and S2 (∅ = 40 mm, ∅ = 48 mm), and elliptically deformited acetabular E2 (∅ = 48 mm) cases

and the lowest at the middle part (Fig. 5). The stress values observed in the current system were significantly lower than the border of plasticity of the CoCr alloys, and caused only insignificant distortions of the metal structures. However, a very uneven distribution of stress is reported to create disadvantageous conditions for the work of these elements.

4. Discussion

The aim of the study was to define the nature of cooperation between the components of hip resurfacing surgery (HRS). From the biomechanical perspective, the mechanical parameters of the base surface which cooperates with the components of the prosthesis are crucial. In the conducted simulations two types of base surfaces were analyzed, namely the surfaces having the parameters correlating to cortical bone and the cancellous bone. The results obtained showed that the base with the lower physical properties (cancellous bone) contributes to the higher stress values in the acetabulum and the head. Similar results were obtained by Murat and Mao, who applied the three-dimensional model of FEM and said that simulated low values of modulus of bone (principal stresses) were much more diverse than the stress values for elastic modulus greater than 5 GP [20].

The obtained stress values for the ideally spherical acetabulum cannot cause significant deformations of cooperating implants. Additionally, it is crucial to notice the surface nature of the cooperation between implants and the lack of point concentrations of stress in the spot of the joint between the implants, which is the ideal situation from the tribological perspective.

In the case of the elliptically deformed acetabulum, significant point concentrations of stress were observed. In the conducted simulations, the stress value reached 150 MPa and was significantly lower than the elastic limit of the ASTM F90 alloy. Nevertheless, it should be noted that the conducted simulations reflect only the quasi-static state of the load. It can be assumed that in the real conditions of the exploitation of the prosthesis, when the system is subjected to dynamic variable loads, the stress values can be similar to the elastic limit of the alloy. In that case, the damage of the cooperating surfaces and permanent deformation of the implant can occur. Fend et al. drew attention to the possibility of the deformation of the acetabulum and its consequences. In his research, conducted on cadavers, concerning the impact of the implantation of the acetabular component using the uncemented press-fit method, he observed the deformation of the acetabulum with the articulation of MoM and the increase of the content of metal ions in the surrounding tissues. By virtue of the application of frozen preparations, in order to eliminate a calculation error, he took the measurements directly after implantation, after 24 hours, and after 7 days. It confirmed the results of our study, concluding that the deformations of the acetabulum in the realistic conditions can be bigger [1]. Yew et al. also declared the risk of the deformation of the acetabulum embedded suing the press-fit technique. According to their study, a factor reducing the risk of deformation is the thickness of the walls of the acetabulum [29]. Markel et al. obtained different results in the course of his research conducted on cadavers. They found that during the implantation of acetabulum, metal bearing shell reduces the size from 0.32 to 0.22 mm (p = 0.019), and that distortion during implantation and mechanical properties of the substrate bone determines the total output of the distortion [19].

Beside the implantation technique and the structure of the implant, the size of the implant has a significant influence on the formation of the places of stress concentration, bigger grindability and incorrect remodeling. The nominal diameter of the surface of joint influences the stress values in the components forming the HRS system. Smaller influence can be observed on the surface of the femoral component where the maximum differences in stress do not exceed 5%. On the internal surface of the acetabulum, which directly cooperates with the head, the differences are bigger and reach 50%.

Comparing the stress in the mandrels of the femoral components of HRS, depending on the nominal diameter of the femoral head with the external nominal diameter ∅ = 40 mm, the stress values were by approximately 40% higher than the stress values in the mandrel of the head with the nominal diameter ∅ = 48 mm. Therefore, it may be stated that in the case of the femoral component with a smaller nominal diameter, there is a higher probability of the implant damage during its exploitation. These observations were confirmed during clinical observations, where the higher risk of revision surgeries was determined proportionally to the increase of the diameter of the head of the resurfacing endoprosthesis – from 9% in the case of the diameter 44 mm and less, to 17% in the case of the diameter 55 mm and more [6].

Bigger stress values in the mandrel may also result in the higher level of stress in the bone tissue adjacent to the mandrel. This may be the reason of exceeding values of the mechanical incentives tolerated by the bone tissue and the decrease of its physical properties [22], [27] (it is connected with the essence of the remodeling process of the bone tissue). Similar results were obtained by Ong et al., who, using the method of finite elements, analyzed the biomechanical conditions and the reconstruction of the bone tissue around the femoral component of the BHR type of HRS, based on the stress distribution. They observed the stress concentration in upper lateral and lower and medial femoral component, around the mandrel and on its end, which may result in the adverse reconstruction of the bone tissue and the loosening of the endoprosthesis [23]. Watanabe et al. observed smaller stress concentration within the head but significantly bigger stress along the mandrel and on its end, which can be justified by tight fit and the flexion moment occurring in the mandrel [28]. Huiskes, who studied the stress distribution in the closer part of the femur after the uncemented implantation of the Wagner's resurfacing endoprosthesis using the finite element method, stated that bone resorption can only occur in the peripheral part of the femur neck and in the central part of the femoral head [15].

Foregoing results of clinical trials and experimental studies arouse controversy and have not explained all the problems connected with hip resurfacing surgery. Numerous reports about promising clinical results on one hand, and adverse data from national records of endoprostheses and the results of biomechanical study showing the imperfections of metal-on-metal HRS indicate that the research on the optimisation of the construction of the endoprostheses should be continued, and the indications of the implantation must be strictly applied.

5. Conclusions

Elliptical deformation during static load of the spherical acetabulum in HRS causes stress concentration in the area of joint and indicates the risk of the damage of the cooperating surfaces and permanent deformation of the implant during the dynamic load.

The diameter of acetabular and femoral components of HRS influences the stress concentration on the internal surface of the acetabulum and in the bone tissue surrounding the mandrel of the femoral component.

Physical properties of the base surface surrounding the components of HRS influence the stress value in the acetabulum and the femoral part. The base surface having weaker physical properties corresponding to cancellous bone generates bigger stress in the acetabulum and the femoral component of the implant during the load.

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