

ORIGINAL ARTICLE

RADIAL HEAD ARTHROPLASTY WITH THE BIPOLAR IMPLANTS

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ABSTRACT
The aim: Of the research is to study of the stress distribution in the "bone-implant" system for various cases of elbow flexion and semipronation in the event of the radial head arthroplasty with the developed bipolar radial head (RH) implant and the bipolar KPS endoprosthesis.
Materials and methods: We developed a metal-metal bearing bipolar RH endoprosthesis providing an uncemented fixation due to a special porous coating and stem design. Using Solid Works software, we developed a geometrical model of the elbow joint with cartilaginous surfaces. Then, to refine the parameters of the finite-element model, calculation and visualization we transferred the model to the ANSYS complex.
Results: The developed bipolar RH endoprosthesis with metal-metal bearing is a stiffer construction compared to the KPS endoprosthesis. However, the displacement fields in the joint and the value of arising maximum strains in the "bone-implant" system with the RH endoprosthesis have a smaller deviation from the strains arising in the healthy elbow joint than those in the "bone-implant" system with the KPS endoprosthesis.
Conclusions: The developed bipolar RH endoprosthesis does not cause any critical impacts on the joint surfaces and ligamentous apparatus of the elbow joint. All the elements of the developed bipolar RH endoprosthesis satisfy the requirements of an operative structure strength and stiffness. Strain fields arising in the bipolar RH endoprosthesis have a smaller deviation from the strains in the healthy joint in comparison with those occurring in the "bone-implant" system with the KPS endoprosthesis.

KEY WORDS: radial head endoprosthesis, bone-implant system, radial head arthroplasty, elbow joint

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INTRODUCTION
Currently both monoblock and bipolar radial head (RH) endoprostheses are used in clinical practice [1,2]. Judet polyethylene-metal bearing bipolar cemented endoprosthesis was first introduced in 1988 and has been used successfully in various modifications since then [3-5]. Since 2004 the KPS polyethylene-metal bearing bipolar RH endoprosthesis (Poland) [6,7] has been widely used both in Poland and Ukraine. However, the KPS endoprosthesis provides the cement fixation only. We developed a metal-metal bearing bipolar RH endoprosthesis providing an uncemented fixation due to a special porous coating and stem design.

The stress fields distribution features in the radial head has already been investigated as well as the contact stress in the humeroradial and radioulnar joints [8-12]. There are some comparing data according the contact stress in the native elbow joint and in case of RH arthroplasty with various design implants [11-12]. Further biomechanical studies of the bone-implant system based on three-dimensional geometric models of the elbow joint in normal conditions and after implantation of a radial head endoprosthesis are actual and will be very useful for future elbow surgery.

THE AIM
The aim of this paper is a comparative analysis of the stress distribution in the "bone-implant" system for various cases of elbow flexion and semipronation in the event of the RH arthroplasty with the developed bipolar RH implant and the bipolar KPS endoprosthesis.

MATERIALS AND METHODS
Basing on the data provided by Zygote Media Group, Inc., U.S. (http://www.3dscience.com), we developed the necessary geometric model. It meets the requirements necessary for mathematical simulation. In order to obtain the most reliable data in the analysis of stresses, we added cartilaginous surfaces and ligaments to the geometric model. Using SolidWorks software, we developed a geometrical model of the elbow joint with cartilaginous surfaces. Each cartilage is 1 mm thick and is identical to the bone Spline-surfaces. We have transferred the model to the ANSYS complex (ANSYS Inc. Canonsburg, PA) to get the parameters of the finite-element model, calculation, and visualization. For further mathematical calculations we have considered the elbow joint in three positions: maximum supination, neutral and

maximum pronation. Each position is considered in a different degree flexion position at the elbow joint: 0°, 30°, 60°, 90°and 120°. Thus, we have had 45 geometric models: 15 for each calculation case – the finite-element model of the native elbow joint, the finite-element model of the developed bipolar RH implant and the finite-element model of the bipolar KPS endoprosthesis. The stress distribution in the elbow joint was set on grounds of the experimental studies [13]. It is assumed that when resting on a wrist joint, 70-75% of the stress is transmitted through the radial bone, and 25-30% – through the ulnar one respectively. The strain was set by the force applied along the forearm axis and was divided as 173 *H* on the radial bone and 74 *H* on the ulnar bone. The load conditions were the same for all the computational models. The interaction contacts of the cartilage contact surfaces are represented as friction contact. The coefficient friction of the contact pair is equal to 0.01. To create a geometric models of elbow joint with the developed bipolar RH implant and with bipolar KSP endoprosthesis we cut off the RH of the native elbow joint model and replaced in it the appropriate endoprosthesis. Figure 1 shows of the geometric models of elbow joint with bipolar RH endoprosthesis in a neutral semipronation and 90° elbow flexion.

Each geometric model had its own finite -element mesh. The correct simulation of the contact interactions was the principal feature of this construction. The number of elements varied depending on the model.

The first step was to construct the model of the native elbow joint with annual ligament, lateral ligaments and medial ones. The mutual influence of the native elbow joint ligaments is very important on the possible stress changes. The ligaments were created as springs with the stiffness 28500 N/m of annual ligament, and the stiffness 15500 N/m of the lateral and medial ligament. These data were obtained from the experimental studies presented in the paper [14]. Figure 2 shows the geometrical model of native elbow joint with the contact pairs and the native elbow joint ligaments fixation scheme.

For all of the models we consider the connection between the radius and ulna like a movable hinged fixation, enabling the movement along the radius. Since the orientation of the elbow joint was set relatively to the position of the humerus, we neglected the shoulder joint impact on the elbow joint. There was the truncated part of the humerus fixed in its upper part that we considered in all models. The Figure 3 shows the calculation scheme of the native elbow joint in the position of maximum supination with 90° elbow flexion.

In the mathematical models with RH implants were considered all contact pairs. For the developed bipolar RH implant, the contact zone of the metal head and humerus cartilage was set as a contact with the friction coefficient equal to 0.05. The stem – head contact zone (metal-metal bearing) was set with the friction coefficient equal to 0.2. For the bipolar KSP RH implant the friction contacts were set with the following coefficients: polyethylene head – humerus cartilage- 0.05; metal stem and polyethylene head – 0.1 (Figure 4).

When constructing the physical model of the elbow joint, we considered the physical and mechanical characteristics of the material properties presented in Table I.

Taking into account the considerable amount of visual information and number of the calculations done, it doesn't seem to be possible to bring them all within the limited scope of the publication. We give only one option for each mathematical models as an example. For instance, the results of calculation for a native elbow joint are presented as stress and strain fields for the supination position with the elbow angle of 90° (see Figure 5-6).

The results presented below in the tables have errors of the finite element model not exceeding 1%. The estimation of the calculation error was carried out according to the technique presented in the works [15,16]. For this, in each model, a comparison of the averaged nodal results of the displacement vector projections with interpolation ones, which are calculated by the points of integration in the finite element, is carried out.

RESULTS

The result's calculation of the model has shown that the obtained stress distribution and stress fields are close to the experimental data [9,10,12,14]. This means that our assumptions in the construction of the model can be used for the assessment of the radius head arthroplasty. Besides, the calculation shows that the model commits a plane-parallel displacement in the direction of the power plane. Considering the model of the elbow joint we further simplified the mathematical model replacing the elastic springs with the boundary conditions of plane-parallel movement of the ulna and radius in the direction of the force action.

Figures 7-10 present the calculation results of the stress distribution and the stress fields for the model with the developed bipolar RH endoprosthesis in the neutral semipronation and 0° flexion in the elbow.

Figures 11-14 present the calculation results of the stress distribution and the stress fields for the model with the KPS bipolar endoprosthesis in the positions of maximum supination and 90° elbow flexion.

Tables II-V show a comparison of the calculation results for the three models.

The developed bipolar RH endoprosthesis with metal-metal pair of friction is a stiffer construction (Table II) in comparison with the KPS endoprosthesis. However, the displacement stress fields in the joint and the value of arising maximum stress in the "bone-implant" system with the RH endoprosthesis have a smaller deviation from the stress arising in the healthy elbow joint than those in the "bone-implant" system with the KPS endoprosthesis. This means that the developed RH endoprosthesis causes less strain effect on the elbow joint than the KPS endoprosthesis. The value of the stress depends on the material physical and mechanical characteristics. A significant difference in the endoprosthesis material properties also defines the discrepancy in the values of the arising stress. In the developed bipolar RH endoprosthesis with metal-metal pair

Fig. 1. Two geometric models of elbow joint with bipolar RH endoprosthesis in a neutral semipronation and 90° elbow flexion.

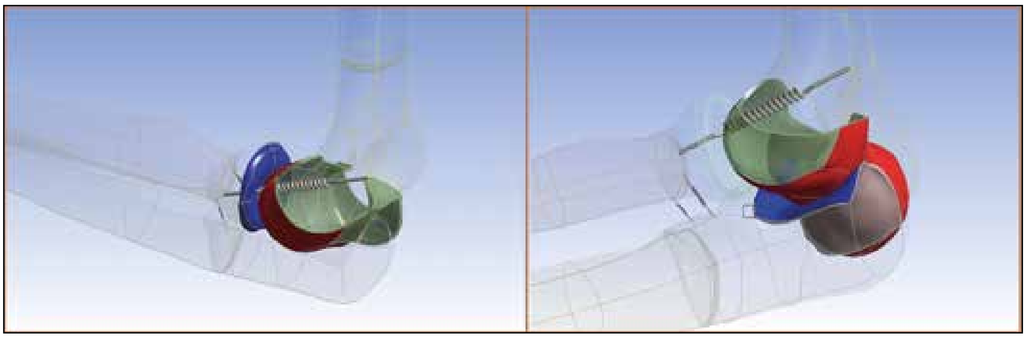

Fig. 2. Contact pair of the radial and humerus articular cartilage surfaces and contact pair of the ulnar and humerus articular cartilage surfaces as well as ligaments-fixation scheme in the native elbow joint model.

Fig. 3. The geometric native elbow joint model in maximum supination and 90° elbow flexion.

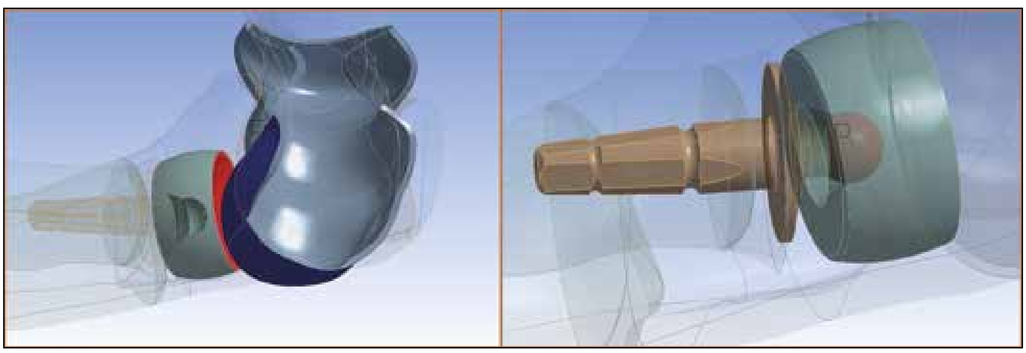

Fig. 4. Contact pairs in the developed RH implant and in the KPS RH implant.

of friction the greatest contact stress fields occurs on the metal head and the capitellum in the position of pronation and the elbow angle of 0° in the neutral position and the elbow angle of 90°. However, in all other calculation cases under consideration the values stress distribution and the stress fields are similar to the native elbow joint. The highest stress occurs in the elbow joint with 90° angle of flexion for all cases.

DISCUSSION

The radial head is an important secondary stabilizer of the elbow joint against valgus stress and posterior dislocation [17-22]. With the rupture of the medial ligamentous complex, which usually occurs in these injuries, the head of the radius becomes even more important as an elbow stabilizer against valgus stress [23-25]. Resection of the radial head in patients with a fracture-dislocation of the elbow can lead to instability and contracture. At the technical opportunity, it is necessary to perform osteosynthesis of the radial head. However, in most patients with fractures and fractures of the III-IV type by Mason-Hotchkiss endoprosthetic of the radial head should be performed [26-27]. Medium-term results from data of different authors show that the results of endoprosthetics with monoblock bipolar endoprostheses of the radial head are similar [28-31]. Our biomechanical computer-aided compar-

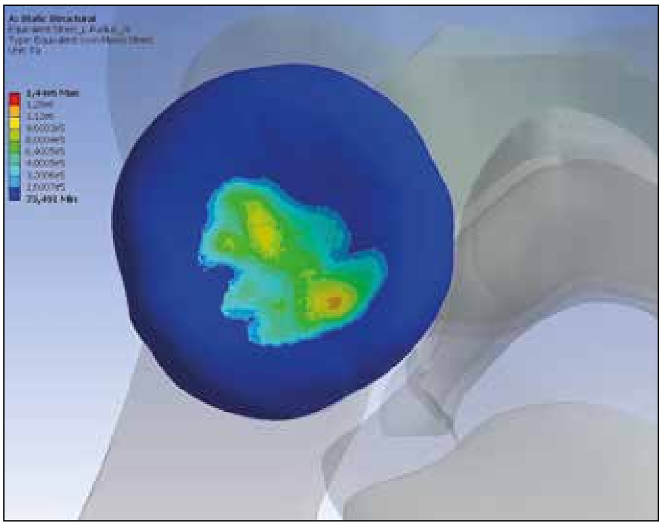

Fig. 5. Equivalent stress distribution in the radial cartilage contact zone.

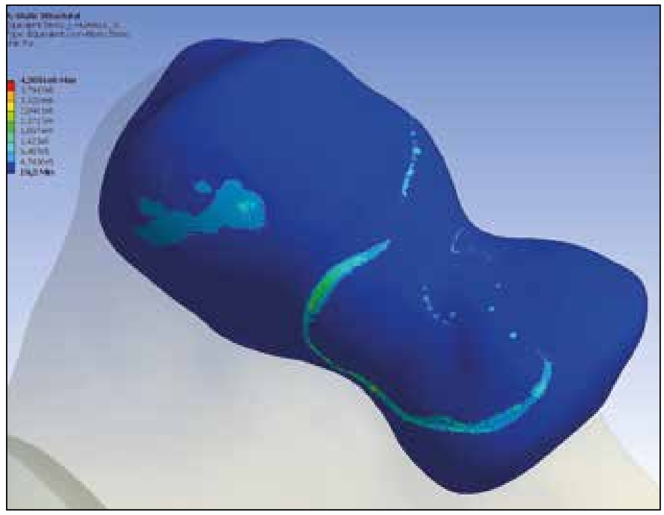

Fig. 6. Equivalent stress distribution in the humerus cartilage contact zone.

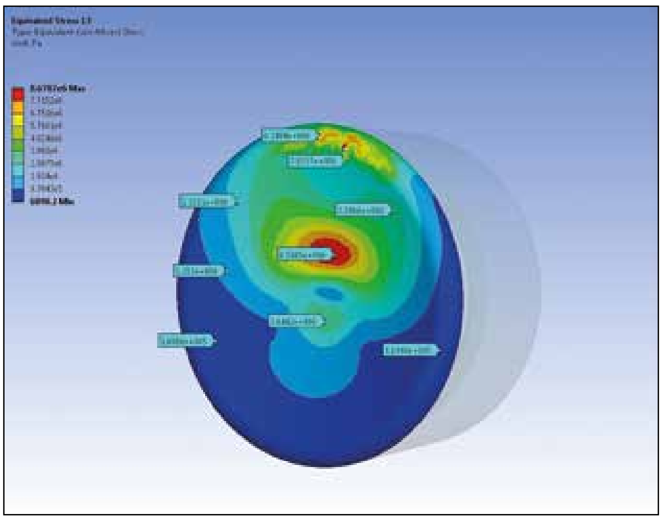

Fig. 7. Equivalent stress distribution in the metal head and humerus cartilage contact.

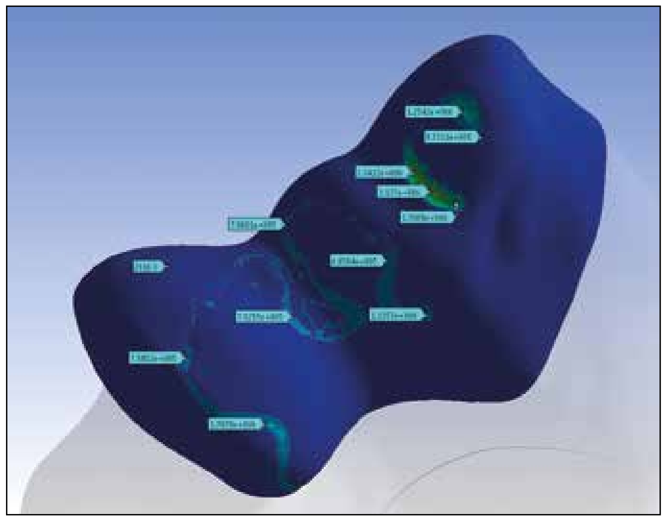

Fig. 8. Equivalent stress distribution in the humerus cartilage contact zone.

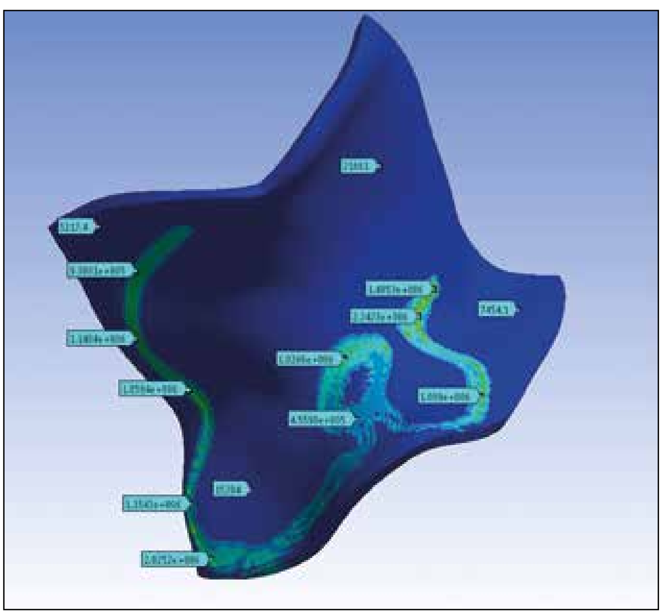

Fig. 9. Equivalent stress distribution in ulnar cartilage contact zone.

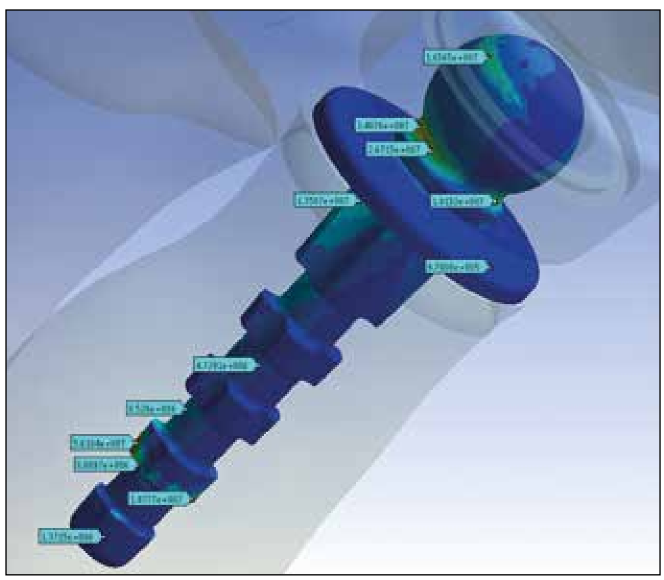

Fig. 10. Equivalent stress distribution in the stem of the developed bipolar RH endoprosthesis.

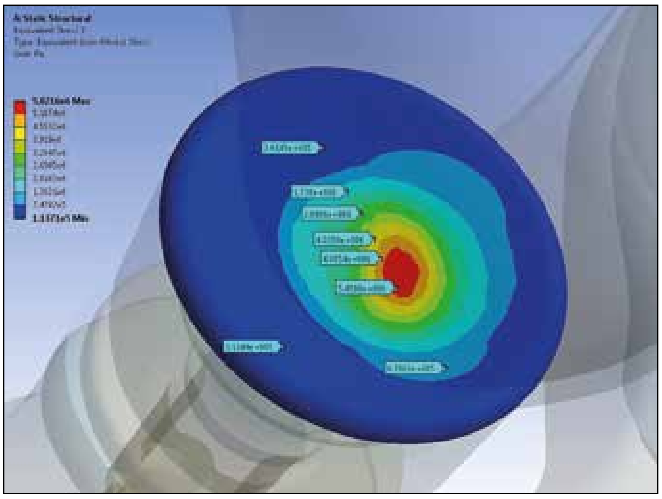

Fig. 11. Equivalent stress distribution in the polyethylene head and humerus cartilage contact zone.

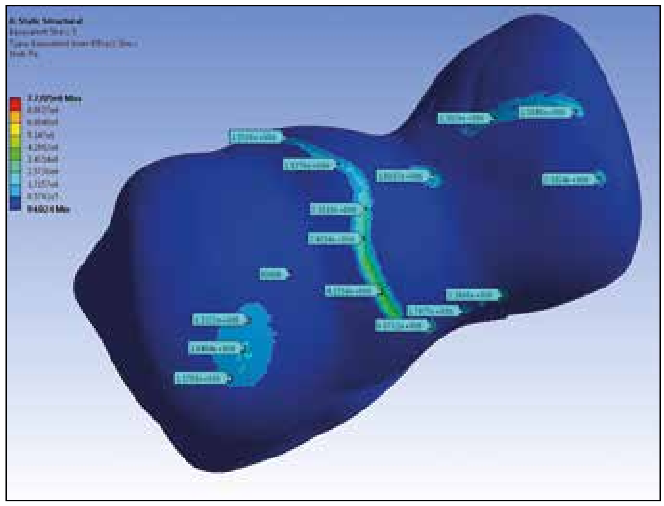

Fig. 12. Equivalent stress distribution in the humerus cartilage contact zone.

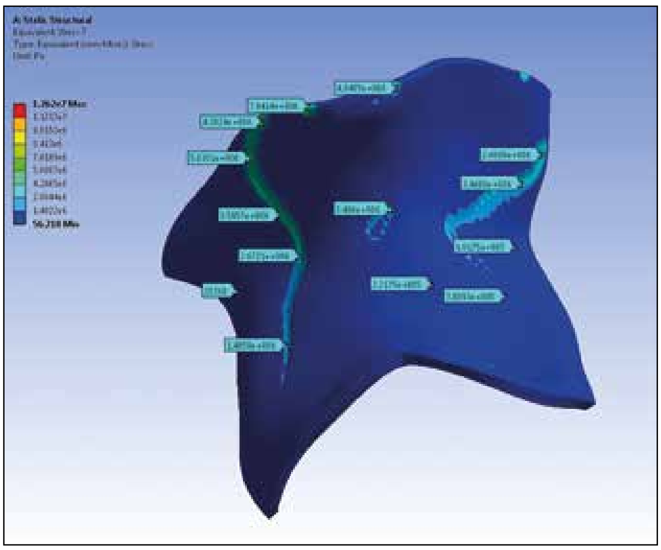

Fig. 13. Equivalent stress distribution in the ulnar cartilage contact zone.

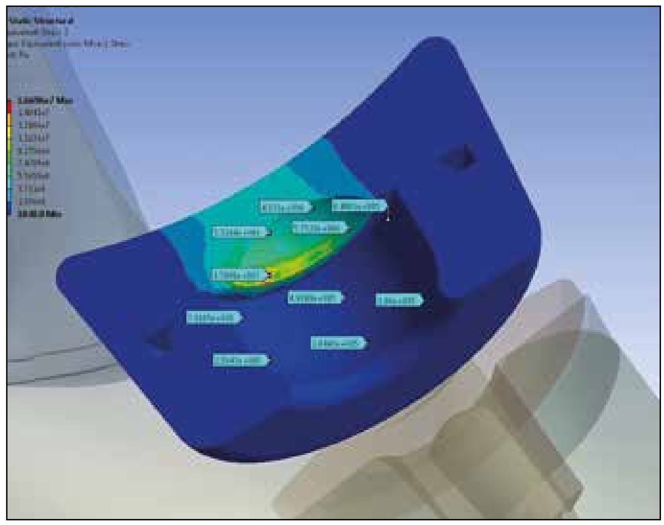

Fig. 14. Equivalent stress distribution in the polyethylene head of the KPS endoprosthesis.

Table I. Physical and mechanical properties of materials

Material	Density ρ,kg/m^3	Young's modulus E	Poisson ratio μ	Ultimate Tensile Strength σ+	Ultimate Compressive Strength σ-
Cortical bone	1800	18.0 GPa	0.3	130 MPa	200 MPa
Spongy bone	500	400 MPa	0.3	18.1	28.6
Hyaline cartilage	1100	10 MPa	0.3	25 MPa	25 MPa
Polyethylene UHMWPE	930	830 MPa	0.3	42 MPa	55.2 MPa
Steel S31673	8000	200 MPa	0.3	500	200

Table II. Comparison of deformation in three computational models

Upper arm position	Elbow angle	Computational models		
		Native elbow joint	Developed RH endoprosthesis	KPSendoprosthesis
		Values of maximum deformation in the joint (mm)		
Supination	Angle 0°	1.667	0.874	3.17
	Angle 30°	1.707	0.641	3.213
	Angle 60°	2.66	1.445	3.863
	Angle 90°	3.079	2.66	4.232
	Angle 120°	2.814	2.612	4.013
Neutral	Angle 0°	0.709	1.437	2.682
	Angle 30°	0.798	0.631	2.704
	Angle 60°	1.611	1.762	3.524
	Angle 90°	4.045	2.404	4.356
	Angle 120°	2.432	2.384	4.085
Pronation	Angle 0°	0.755	1.142	2.369
	Angle 30°	0.758	1.18	2.458
	Angle 60°	1.448	1.673	3.365
	Angle 90°	3.829	2.908	4.569
	Angle 120°	2.635	2.406	4.135

Table III. Comparison of the maximum equivalent stress in the radial head contact zone for three computational models

Joint position	Elbow angle	Computational models		
		Native elbow joint	Developed RH endoprosthesis	KPS endoprosthesis
		Values of the maximum equivalent stress in the radius cartilage contact zone (MPa)		
Supination	Angle 0°	3.026	4.952	6.548
	Angle 30°	1.616	7.104	8.468
	Angle 60°	1.655	7.548	8.763
	Angle 90°	1.44	5.821	6.93
	Angle 120°	1.216	6.322	7.357
Neutral	Angle 0°	4.229	8.678	8.975
	Angle 30°	2.995	15.357	12.73
	Angle 60°	4.924	15.465	13.56
	Angle 90°	7.798	17.003	10.684
	Angle 120°	2.403	16.241	11.358
Pronation	Angle 0°	4.24	10.076	9.007
	Angle 30°	5.051	12.006	11.567
	Angle 60°	2.837	14.704	10.805
	Angle 90°	7.294	12.9	5.939
	Angle 120°	3.984	12.33	7.864

ative study of two bipolar prostheses shows some advantages.

The dynamic loading, to which the implants are subjected, along with the corrosive activity of physiological fluids, can increase the wear of the friction pair. Tribocorrosion is defined as "irreversible transformation of a material in tribological contact, caused by the simultaneous physical, chemical and mechanical interaction of the surface" [32]. In recent decades, patients with a metal-metal friction pair after total hip arthroplasty (THA) have experienced inflammatory reactions, often with signs of tribocorrosion at the junction of the head and neck. Tribocorrosion occurs not only on the bearing surfaces, but also at the metal /

Table IV. Comparison of the maximum equivalent stress in the humerus capitellum contact zone for three computational models

Upper arm position	Elbow angle	Computational models		
		Native elbow angle	Developed RH endoprosthesis	KPS endoprosthesis
		Values of the maximum equivalent stress in the humerus cartilage contact area (MPa)		
Supination	Angle 0°	3.953	3.467	5.588
	Angle 30°	3.16	2.961	4.878
	Angle 60°	4.887	5.021	7.381
	Angle 90°	4.269	7.72	8.24
	Angle 120°	3.863	5.496	6.276
Neutral	Angle 0°	2.831	4.277	4.732
	Angle 30°	2.637	4.818	4.956
	Angle 60°	4.499	11.958	8.248
	Angle 90°	12.667	7.809	7.679
	Angle 120°	6.517	6.164	6.38
Pronation	Angle 0°	3.258	8.641	3.643
	Angle 30°	3.022	11.143	4.369
	Angle 60°	4.201	7.586	5.651
	Angle 90°	10.67	10.789	5.88
	Angle 120°	4.507	12.082	4.78

Table V. Comparison of the maximum equivalent stress in theulnar articular surface contact zone for three computational models

Upper arm position	Elbow angle	Computational models		
		Native elbow joint	Developed RH endoprosthesis	KPS endoprosthesis
		Values of the maximum equivalent stress in the ulnar cartilage contact zone (MPa)		
Supination	Angle 0°	2.052	4.34	1.507
	Angle 30°	4.763	4.96	2.024
	Angle 60°	6.438	5.863	4.878
	Angle 90°	6.456	7.941	5.438
	Angle 120°	6.231	6.767	5.024
Neutral	Angle 0°	2.97	2.561	2.864
	Angle 30°	3.088	4.034	3.587
	Angle 60°	6.989	6.42	6.881
	Angle 90°	7.151	16.091	8.238
	Angle 120°	6.725	6.774	6.256
Pronation	Angle 0°	3.651	3.896	3.338
	Angle 30°	3.11	4.052	4.866
	Angle 60°	6.655	6.042	8.224
	Angle 90°	6.91	6.933	10.061
	Angle 120⁰	6.698	6.35	6.548

metal joints, where micromotions are possible between them. However, these facts do not apply to friction pairs with heads of small diameter and made of high-carbon steel alloys [33]. Wear particles present in metal-polyethylene implants are of the size required for phagocytosis by macrophages, which is considered to be the cause of aseptic instability [34]. On the other hand, the particles formed by the implants in the metal-to-metal friction pair are on the nanometer scale, which reduces the number of macrophages. However, the distribution of these particles in the body can have different biological effects and may be responsible for cytotoxicity, hypersensitivity, and ultimately carcinogenesis [32-34]. All this fully applies to large contact areas and heavy loads, which is absent in our case. The developed bipolar RH endoprosthesis with metal-metal pair of friction was applied in the period from 2014–2017 in 14 patients (6 men and 8 women) aged 43 to 72 years. The average follow-up period was 18 months (from 6 up to 24). There were no revisions till now. We didn't notice any allergic or cytotoxic reaction as well as pceudotumor. The use of developed bipolar RH endoprosthesis allows us have been obtaining in all patients positive results within 2 years after surgery, the average score according to Mayo Elbow Performance Score was 88.5 points [35-37]. Further clinical observation of patients provides additional information about of the possible aseptic inflammation and subsequent instability of the developed bipolar RH endoprosthesis.

CONCLUSIONS

1. The developed bipolar RH endoprosthesis with metal-metal pair of friction does not cause any critical impacts on the joint surfaces and ligamentous apparatus of the elbow joint. The character of the stress distribution and the stress fields are similar to the native elbow joint.
2. All the elements of the developed bipolar RH endoprosthesis match to the requirements of a strength and stiffness of imlants.
3. The carried out comparative biomechanical analysis of the stress in the "bone-implant" system numerical models with the developed bipolar RH endoprosthesis and KPS endoprosthesis has shown the similar meaning in the different structures of the elbow joint as well as parts of implants.
4. The developed bipolar RH endoprosthesis with a metal-metal pair of friction is a stiffer construction.
5. The stress distribution and the stress fields in case of the bipolar RH endoprosthesis implantation have a smaller deviation comparing with the same parameters in the KPS endoprosthesis case.

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