STRESS ANALYSIS OF THE RADIAL HEAD REPLACEMENTS IN AN ELBOW ARTICULATION

Pavel Hlavoň*, Vladimír Fuis*, Zdeněk Florian**

The paper is focused on computational modeling of elbow articulation with radial head replacement. The main part of the project is aimed towards the creation of computational model of suitable partial endoprothesis of proximal part of the radial bone tissue which would keep the function of the elbow articulation while replacing the distance created by resection. The geometrical and computational models of radial head replacements and elbow joint were created. Computational models with these implants were compared with given physiological state of elbow articulation. The influence of friction and material characteristics of bone tissue and cartilage on changes in contact pressure (and therefore to the abrasion) were analyzed using those models.

Key words : *elbow joint, elbow articulation, FEM, radial head replacement, partial radio-humeral alloplasty, contact stress, computational modelling*

1. Introduction

In traumatology demolition of an elbow joint there are many cases, where the surfaces of the joint are damaged, mainly the surface of the radius proximal part. These parts of the joint have been removed during the operation (here we are dealing with spokebone). Function of the joint is not lost but it nevertheless causes a really strong pain in the patient's wrist. Upon the initiative of the medical team at Traumatological Hospital in Brno, efforts are invested to create possible partial joint replacements of the proximal end of the radius in the elbow articulation that would suitably replace the patient's missing bone tissue and, at the same time, to preserve the same functionality of the radio-humeral connection.

This contribution speaks about potential geometrical and computational models of a partial joint replacement and its stress-deformation analysis (different stress-strain analysis, contact pressures, displacements and deformation, sensitivity analyses), which can be used for comparing with physiological state and design the proper geometry of radial head replacements (hereinafter RHR only).

2. Material features

The bone tissue consists of bone cells of long cell bodies with numerous protrusions in the basic substance channels. The basic substance comprises an organic part (ossein) and an inorganic one in variable proportions. As one gets older, the inorganic part grows and

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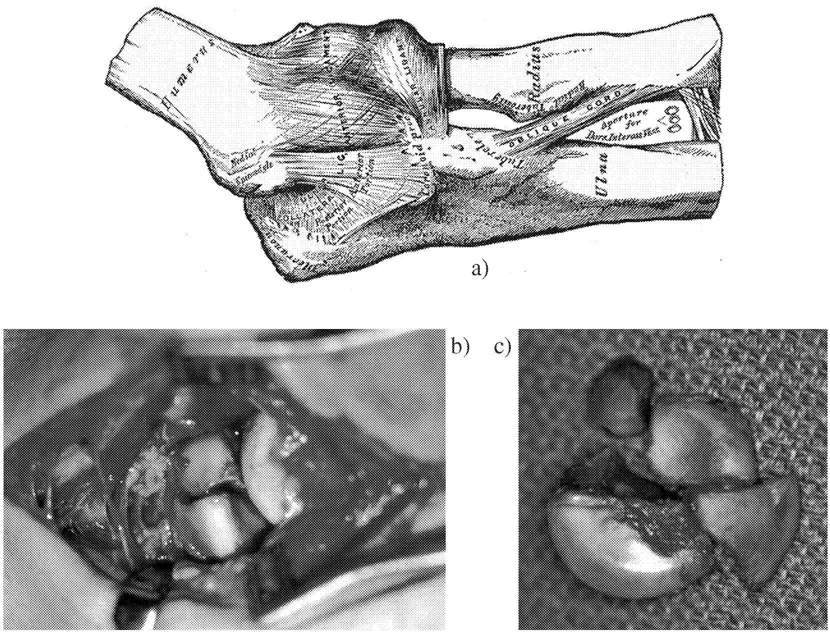

Fig.1: Elbow articulation : a) physiological state, b) proximal part of spokebone is damaged, c) removed damaged radius head

the bone becomes harder but more brittle and more connective. It is ossein that secures elasticity and flexibility in young bones. Bones involve two types of bone tissue; their percentage proportion differs in bones of various types. Compact bone tissue (substantia compacta) – is found especially in long bone diaphyses, see Figure 2a. Its growth activity and regeneration capacity are enormous. Trabecular bone tissue (substantia spongiosa) – is present in long bone epiphysis, see Figure 2b. It has a special structure, called bone architectonics, owing to their load in some bones. The bone can be described as solid, relatively hard, mineralized connective tissue of yellow and white color, having supportive and protective function.

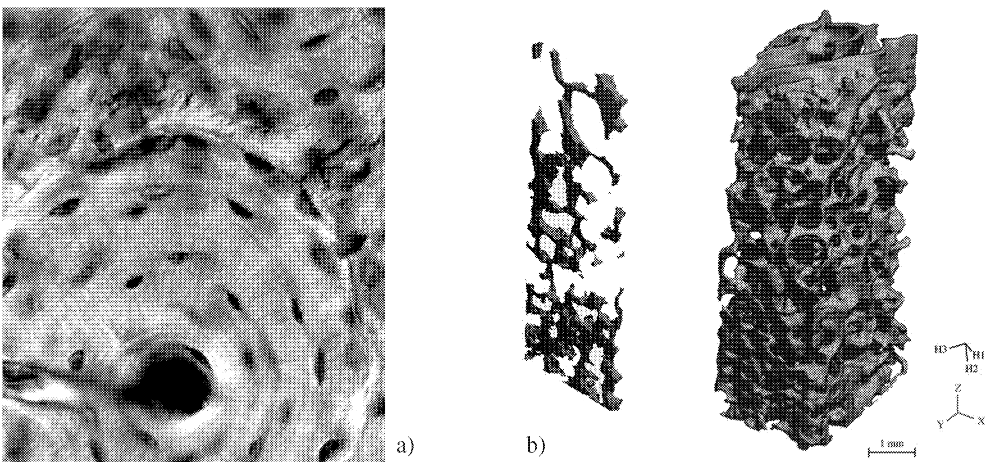

Fig.2: a) compact bone tissue, b) trabecular bone tissue

Experimental measuring of mechanical properties and features of the bone tissue results in values within a wide interval. This interval width is caused by many factors affecting the measurement. These components involve factors that depend on the donor's physiological condition, such as age, sex or internal structure of the sample, among others. Other factors arise directly out of experimental measurement – sample size, current physical state of the sample, the fact whether the sample is fresh or embalmed. Measurement results are also considerably affected by the method of sample storage and conservation; tests of frozen samples show results that are different from samples that have been taken out of a biological solution. These differences are caused by the fact that the bone tissue dramatically changes its properties within a short period after the sample is taken out of the donor's body. Considered features in this problem are mentioned in references. The Young's elasticity module and Poisson numbers that were taken into account are shown in Table 1.

Compact bone tissue can be considered as an inhomogeneous anisotropic high-elastic biomaterial with nonlinear behavior. The mechanical properties are influenced by many factors, as it was already mentioned. Computational models of this tissue, which are nowadays used, behave like an isotropic or orthotropic linear continuum that considerably simplifies the reality, see References [2], [3].

The trabecular bone tissue is made of trabeculars and also ground substance. From a review study can be seen that the substantia spongiosa is a heterogeneous material. The density and the form of lay-outs vary in different part of bone tissue. Therefore the material features can be found in a wider interval than substantia compacta. Accordingly here in spongiosa, the linear isotropic material is commonly used in computational modeling.

Material, tissue	Elasticity module [MPa]	Poisson's number [–]	Bibliography
Compacta	18000	0.3	[1]
Thicker spongiosa	600	0.3	[2]
Average spongiosa	500	0.3	[1], [2]
Thinner spongiosa	200	0.3	[2]
Cartilage	50	0.45	[2], [3]
RHR – corrosion resistant steel	210000	0.3	[1], [2], [3]
RHR – titanium	90000	0.3	[1], [2], [3]
RHR – Polyethylene	500	0.3	[1], [2], [3]
Cement	3000	0.3	[1], [2]

Tab.1: Material features used for computational modeling

3. Geometrical model

The elbow articulation geometrical model was created using computer tomography (CT) at the representation method clinic, St. Ann's University Hospital in Brno. The computer tomography provided profiles of the whole embalmed elbow joint, i.e. the point of contact of three bones – humerus, cuboids bone and radius. The tomography output data were supplied in standard DICOM and JPG formats, each profile scanned with an axial distance of 0.6 mm.

The next step upon model generating made use of the RHINO software for curve, area and body generation, gradually importing tomography profile. A border circumferential bone curve was gradually generated at these profiles through splines; and these profiles were put together with a concrete shift in the axial direction in a perspective view. The

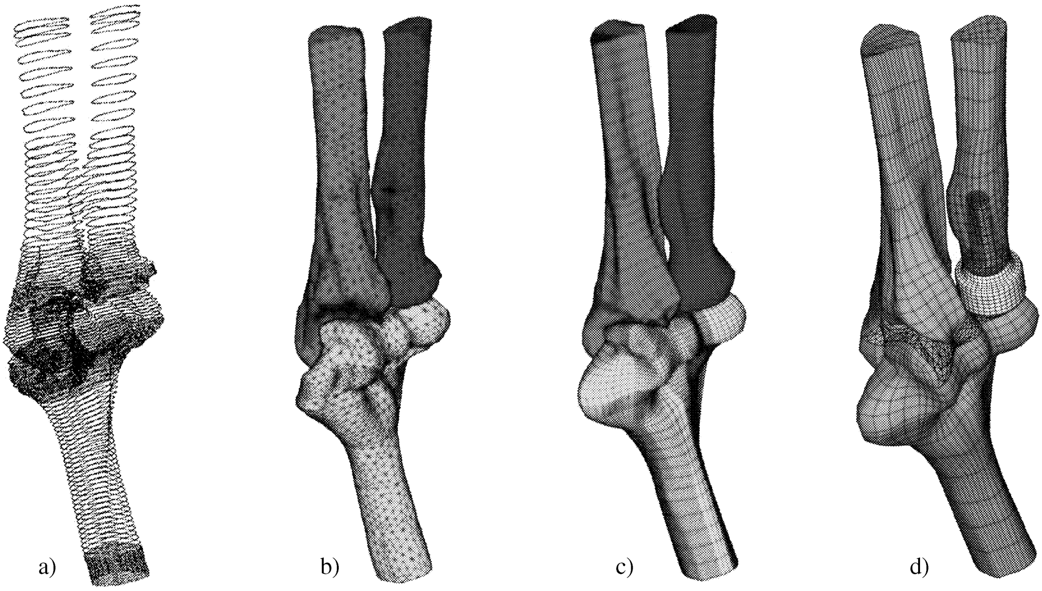

Fig.3: a) 'wire' model (RHINO), b) free mesh model (ANSYS), c) mapped mesh model (ANSYS), d) model with RHR (ANSYS)

entire model, containing curves and surfaces in 2D, was exported to the IGES format. This permits to continue with the model's geometry in software forms of the final element method (hereinafter FEM only), see Figure 3.

In order to make geometric and computational models, the FEM software was used (ANSYS system). The 3D model was imported and further adjusted in the selected ANSYS finite-element calculation program. The generated curves and surfaces were used to create volumes, where the finite-free-elements and mapped mesh were generated on the applied volume entities.

4. Computational model

Computerized modeling and simulations were separated into the three main parts.

The first part was stress analysis for *physiological state* of articulation. Two types of models were created in this part. Cartilage was also imported between the two areas in the articulation. Contact areas were created with two exact radii – R16 and R35. Within the first phase of computational modeling the bones were considered as a compact bone tissue with the average Young modulus of compacta and spongiosa. The whole material was considered and used as isotropic with linear behavior. In the second phase, the geometric model was further corrected. The bone tissue was elaborated as the compact tissue with trabeculars and also cartilage was used on the contact surfaces of spoke and cuboids' bones.

The generated mesh system quality is dependent on the type of FEM elements which are used. In general, there are two possibilities in using 3D elements. SOLID 92 – these elements have linear or quadratic base function. Mesh production is very simple and fast. One can use the automatic mesh generation as well. Their disadvantage is bigger number with opposite of hexahedrons. Solving time is directly influenced through this unfavorable fact. In this application the number of used volumes was 27 and 139337 elements. SOLID 95

– on the other hand, these elements adapt the real geometry of the model and also stress gradient. Its disadvantage is longer time for mesh generating, because we have to create a particular net by hand. There is no simple possibility to modify this mesh. With this application, 306 volumes and 67298 elements were achieved.

In the further sequence, created volumes for the surface of bones were covered by SHELL 181 elements with different mechanical features corresponding with compact bone tissue that would were to separate the trabeculars. Cartilages on the contact surfaces were filled with SOLID 95 elements.

Generally can be said, that contact task is a nonlinear one. Heftiness and the length of computing are directly proportional to the generated mesh, numbers of the elements and the quantity of sections, where the contacts take place. There were two contact places in this task, between radius – humerus and humerus – cuboids bone. For each of them the TARGE 170 and CONTA 174 elements were used. Boundary conditions were chosen on the bottom part of brachial bone – zero displacements in every axis. Numerical results for this part of computational modeling are stated in the left part of Table 2.

The second part was stress analysis for *elbow* articulation with possible radial head replacements. After the contact problem mentioned above, that ran over for :

– option without cartilages – contact with two cortical bones,
– option with cartilage at the radial and humeral contact surfaces,
– version with cartilage at the humeral and cuboids bone contact surfaces.

... the geometrical model was further adjusted and modified. During a traumatological demolition of contact areas in the proximal part of the radial bone remains are operatively removed. Missing bone tissue was not replaced by any tissue and this distance expressed in shifting of spoke bone and therefore the patients feel the strong pain in the wrist and carpal parts. The instigation of medical team was to create such a radial replacement that would suitably compensate missing bone tissue and the function of articulation would be kept.

To the geometrical model of the physiological articulation the possible geometrical model of radial head replacement was created. Also a cement layer model, which is able to fix joint replacement in the porous bone tissue, was implied in the following tasks. The FEM mesh of the geometrical models of radial head replacement and cement layer was realized through the hexahedrons SOLID 95 with material features according to Table 1.

In the following analysis, the project was focused on solving the nonlinear contact tasks for physiological state and also between the partial joint replacements at the radius & cartilage at the humerus. The geometry and the design of the radial replacement had to be suitably optimized, so as to best correspond to the contact stresses of the radio-humeral connection in physiological state. Models were developed of several alternative types of replacement of the proximal end of radius.

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For preliminary computational modeling the alternative of replacements with a conical direct stem and with jointed conical direct stem were selected. The analysis was developed also for two different curvatures of the bearing-on contact surface – for radius R16 and R35. Also the superficial part of the humeral cartilage exhibited a different curvature. In this section, several contact tasks were solved in various parts of the computational model, also for various geometry and friction, various materials of proximal replacements.

Border conditions of the humerus inferior part were selected as zero movement at all three coordinate axes and zero turns. Since the contact was closed at the beginning in this contact task, power conditions were predetermined in the respective z-direction of the fixed coordinate system at the superior surfaces of the radius's section and zero movements in x and y directions. The dynamic load at all nodes of the selected surfaces in the upper part of the radius's section roughly corresponded to the static load of an individual weighing approximately 80 kg, falling on one hand.

5. Conclusions

There are shown main outputs from the several parts of numeric analyses. The first one was stress analysis for *physiological state* of articulation. Contact compositions were solved with nonsymmetrical approach and also with symmetrical state of the contact areas. Force load in all nodes of the selected areas of the upper radial part replied to a static burden approximately 80 kilograms body-weight of any individual after fall to the one hand – in static meaning.

Maximal values of contact pressures for both contact parts at the humeral bone are near to 3 MPa. The dispersion of the values in different phases is caused by the chosen types and density of the particular mesh that were generated in the contact faces. It turned out that the contact pressures are not essentially influenced by the type of the contact tasks. Also the friction has no bigger effect to these values, as it will be shown later in further analyses with radial head replacements. By this computational modeling was proved that a cartilage plays significant role at the level of contact pressures. The values of contact pressures were about 30 % lower at incriminated places against the contact with only compact bones. Attrition is then lower as well.

In the second part of the computational modeling, the best correspondence to the physiological state, exhibits the proximal replacement made of high-pressure polyethylene, the difference of which makes 0.8 %. The replacements made of titanium or corrosion resistant

<table>
<tr><th rowspan="2">Contact area vs. state type</th><th colspan="2">Physiological state</th><th colspan="2">Model with RHR conical direct stem</th><th colspan="2">Model with RHR jointed conical stem</th></tr>
<tr><th>R16</th><th>R35</th><th>R16</th><th>R35</th><th>R16 with turning</th><th>R16 w/o turning</th></tr>
<tr><td>Proximal radius & RHR</td><td colspan="2"></td><td>5.295 MPa</td><td>5.440 MPa</td><td>6.021 MPa</td><td>5.356 MPa</td></tr>
<tr><td>Number of all contact elements</td><td>600</td><td>720</td><td>1730</td><td>1930</td><td>3300</td><td>3300</td></tr>
<tr><td>RHR & humerus polyethylene</td><td rowspan="3">1.998 MPa</td><td rowspan="3">2.684 MPa</td><td>2.014 MPa</td><td>2.770 MPa</td><td rowspan="3">1.862 MPa</td><td rowspan="3">3.042 MPa</td></tr>
<tr><td>RHR & humerus titanium</td><td>2.074 MPa</td><td>3.023 MPa</td></tr>
<tr><td>RHR & humerus steel</td><td>2.085 MPa</td><td>3.049 MPa</td></tr>
<tr><td>Ulna & humerus</td><td>2.105 MPa</td><td>2.124 MPa</td><td>2.105 MPa</td><td>2.124 MPa</td><td>2.105 MPa</td><td>2.124 MPa</td></tr>
<tr><td>RHR with joint</td><td colspan="4"></td><td>23.954 MPa</td><td>17.578 MPa</td></tr>
<tr><td>Substeps & iterations</td><td>7/19</td><td>7/12</td><td>25/57</td><td>25/62</td><td>8/54</td><td>8/20</td></tr>
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Tab.2: Results of the stress analysis in elbow articulation with RHR

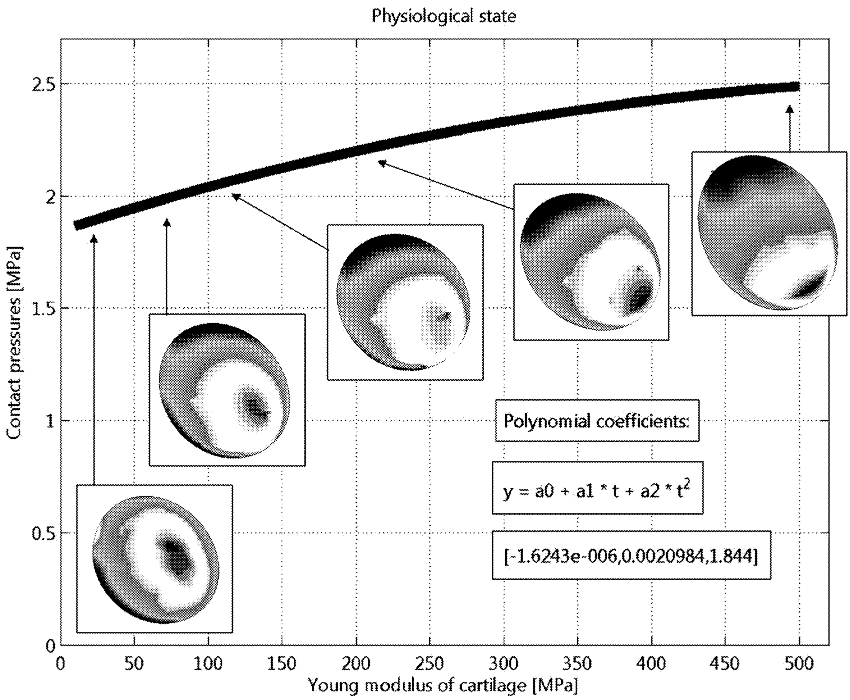

Fig.4: Contact pressure dependence on the Young modulus of cartilage – physiological state

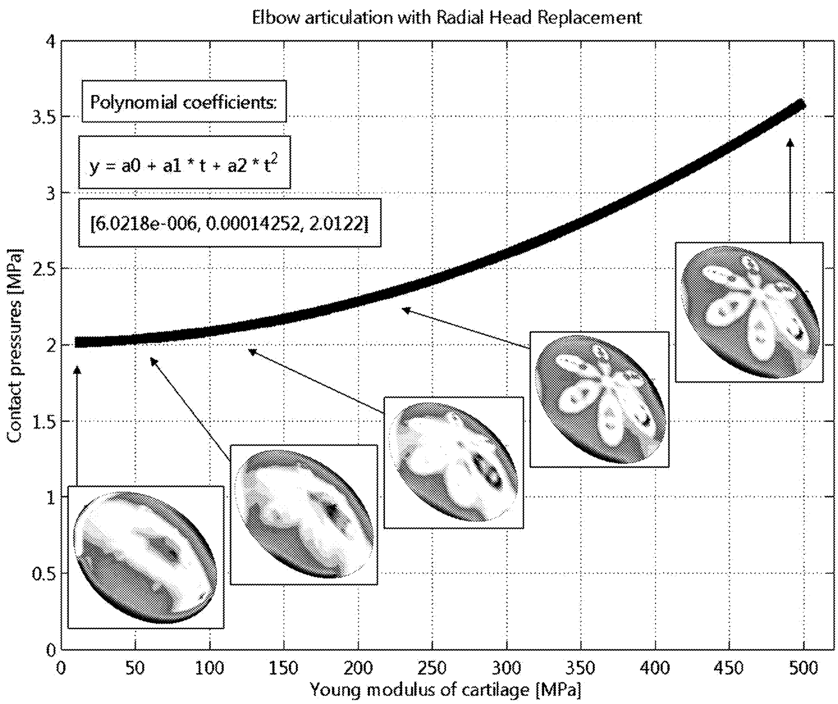

Fig.5: Contact pressure dependence on the Young modulus of cartilage – elbow with RHR

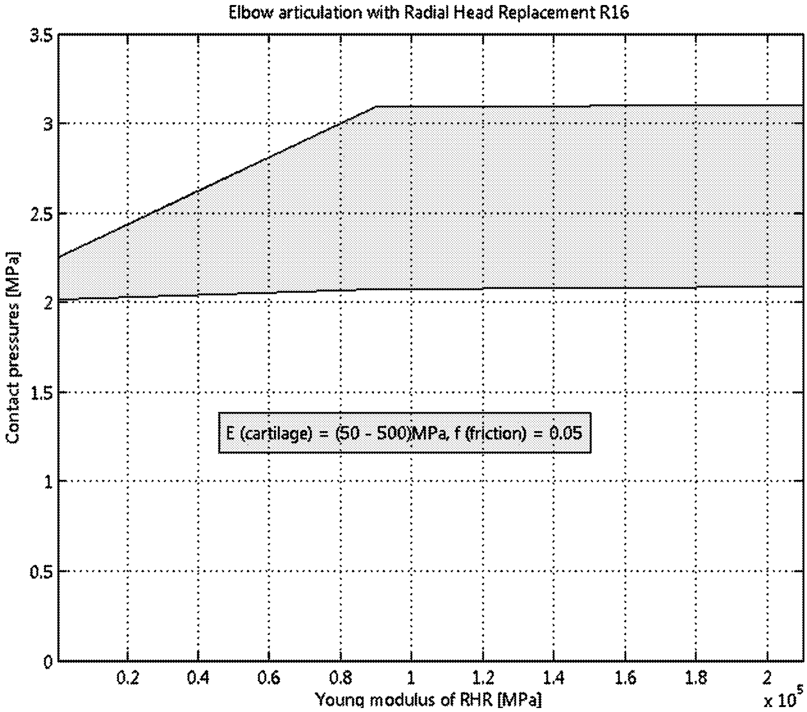

Fig.6: Contact pressure dependence on the Young modulus of RHR

steel differ from the physiological state by titanium (4–12) % and steel (4–13.5) %. The analysis also shows that if we increase the number of elements by about 50 %, the contact pressure will decrease by about 2.7 %, see Table 2. Thus, it is clear that the number of elements needs not to be increased in this area. For the other RHR type was also achieved fair correspondence in contact pressures. Jointed RHR with conical stem was made only for curvature R16 of contact areas between proximal part of replacement and cartilages on the humerus. There are shown two numbers for different geometrical position of radius against humerus. The value (without turning) naturally corresponds with geometrical posture of physical state and it is around 1 MPa bigger than physiological state.

In further analyses, it was determinated how contact pressures for physiological state and for RHR depend on modification of Young's modulus of cartilage. Young's modulus of cartilage was chosen after background research in the wide interval from 10 to 500 MPa. Apparently contact pressures in physiological state change with slight degressive function from 1.9 MPa to 2.5 MPa, see Figure 4. On the contrary, contact pressures in the elbow articulation with RHR shift with progressive function from 2 MPa to 3.5 MPa, see Figure 5. For the trendiest value of Young's modulus, about 50 MPa, the CP values are in good accordance. And also at Figure 6, it can be seen how dependent contact pressures are if the material of radial head replacement and material features of used cartilage are change.

Acknowledgments

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