Vol. 36, No. 4 (2014) 354-360

Tribology in Industry

www.tribology.fink.rs

RESEARCH

Contact Stress Generation on the UHMWPE Tibial Insert

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Keywords:

Biomaterials
Stress distribution
Knee joint prosthesis
Wear

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A B S T R A C T

Total knee replacement (TKR) is considered, during last years, as a very successful surgical technique for removing knee joint deformities and eliminating pain caused by cartilage damage. In literature, as primary causes for knee joint endoprothesis damage are cited complex movements which cause occurrences of complex stress conditions, sagital radius conformity, sliding, types of materials etc. Aim of this study is analysis of contact stresses that occur on tibial implant for 15°, 45° and 60° knee flexion and 50 kg, 75 kg, 100 kg and 125 kg weight. Knee joint prosthesis model and finite elements method (FEM) analysis are done in software Catia V5. For this analysis we used ultra-high molecular weight polyethylene (UHMWPE) for tibial implant material and AISI 316, AISI 317, AISI 321, 17-4PH, CoCrMo, Ti6Al4V and SAE A-286 for femoral component materials. Results show that area of maximal contact stress is identified in medial and lateral part of tibial implant. Von Mises stress values vary regarding of flexion degree and weight, but values are approximate for types of chosen materials. Contact stress location corresponds to damage that occur on tibial implant during exploitation.

1. INTRODUCTION

Application of biocompatible materials in medicine is aimed to return form and function of replaced biological structures. In the orthopedic field, materials used in implantology should provide adequate ductility, corrosion resistance, wear resistance, biocompatibility and integration with bones. In modern orthopedic surgery usually are used stainless steel, superalloys based on cobalt, titanium and its alloys, ceramics, polymers and, rarely, composite materials. Bearing surfaces are more and more made from UHMWPE material because it has properties of joint cartilage and remarkable durability toward abrasion, friction durability, exceptional ductility, small density, biocompatibility and biostability. Besides application in complete knee replacement, UHMWPE material is used for manufacturing hip joint endoprosthesis components, shoulder joint etc. (Fig.1) [1-3].

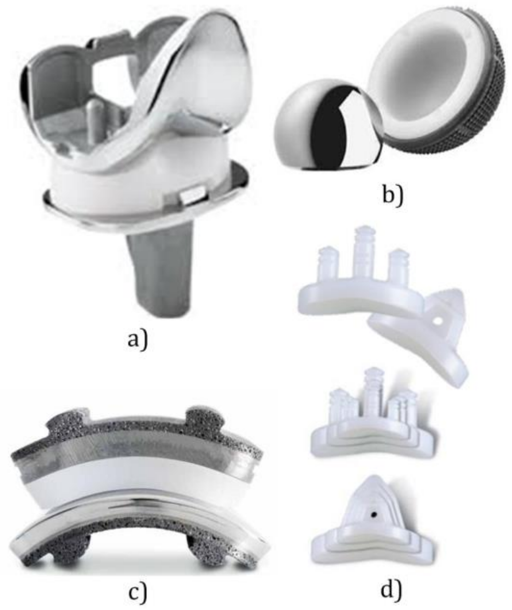

Fig. 1. UHMWPE endoprosthesis components for: a) Knee joint, b) Hip joint, c) Ankle joint, d) Shoulder joint [4].

TKR is seen as very successful surgical technique for removing knee joint deformities, eliminating pain caused by cartilage damage etc. By applying and adequately choosing modern materials, durability of knee joint endoprosthesis is significantly improved. Nowadays, material selection has great importance. Main factors responsible for knee joint damage as well as knee joint endoprosthesis damage are complex movements (such as internal – external rotation, flexion/extension of the knee, anterior – posterior and medial – lateral translation) which cause complex stress conditions, as well as sagital radius conformity. These factors lead to delimitation and adhesive/abrasive wear [5-9].
Knee joint endoprosthesis construction can be with fixed or mobile bearing. Considering improving patient life conditions, S. Affatato et al. consider that special attention should be paid on chemical features of UHMWPE material and ways they affect wear levels in different knee joint endoprosthesis constructions. Their results show that fixed bearing constructions have more significant area degradation affecting increased wear level and increased movement radius which leads to creating stronger contact stress [10].

Realistic knee joint movement simulation is hard to conduct in vitro with help of standard equipment. For that reason, researchers choose to construct special tribo-systems that can better mimic patient with knee endoprosthesis everyday activities. Determining wear mechanics of tibial implant made of UHMWPE entail keeping track of large number of factors such as metal materials application, contact area geometry, location of maximal stress values, maximum values of force and sliding friction [7].
Constructing special tribo-systems can be very expensive and time consuming. Computer simulations are very significant for predicting material behavior and behavior of final products during exploitation. Contact stresses location determined by cyclic loads can, conditionally, represent location where damage will occur. For that reason, researchers are conducting comparative analyses in vitro wear and finite elements method (FEM) analyses [11-13].

It is often assumed that wear rate reduction is related to increasing of the contact area between femoral component and tibial insert. In order to prove this hypothesis, S. Sathasivam et al. have perfomed laboratory test and FE analysis. They were prepared/designed pins, made of UHMWPE material, with 8 – 23 mm diameter which was moved along CoCr alloy plate under influence of load. After analysis, they concluded that decreasing of the contact areas leads to cracks formation which influence on increasing wear rate [14].

Considering the fact that the tibial insert is not only element made of UHMWPE material in implantology, great attention is paid to the acetabular cup wear analysis in total hip replacement. For instance, L. Căpitanu et al. have performed an experiment related to femoral head and acetabular cup wear after revision surgery of the total hip replacement. In order to get wider picture of the acetabular cup wear, they have performed FE analysis. They concluded that the main source of the acetabular cup damage was influenced by the abrasive wear particles. These particles were located at the active contact area of the hip prosthesis [15].

Theoretically and experimentally, A. Wang et al. showed that volumetric wear rate per motion cycle is proportional to $2/3^{rd}$ power of the applied load and$1/3^{rd}$ power of the contact area for the wear of UHMWPE under multi – directional motion. The experiments were conducted on MTS hip joint simulator where is performed three test: effect of the sliding radius or distance on the wear of UHMWPE, effect of the peak load on the wear rate of the UHMWPE, and effect of the contact area on the wear rate of the UHMWPE [16].

The topic of this paper is pointed to researching influence of different metal materials on creating and defining location of contact stresses on tibial insert made of UHMPWE materials with help of finite elements method and comparing results with results shown in literature.

2. MATERIALS AND METHODS

3D finite elements models are analyzed in order to define location of maximal contact stresses and location of possible damage of tibial implant depending on material type, body weight and knee flexion degrees. In these analyses the same model of knee prosthesis is used, but loads and angles are variable.

Case of fixed bearing tibial component is analyzed. Tibial implant is made of UHMWPE and femoral component is made of commonly used materials with properties listed in Table 1.

Table 1. Metalic materials properties applied in FE analysis [1].

Matrial type	R_m	R_{eh}	R_{ds}	E	OH	R
Stainless steel						
AISI 316	550	205	270	198	8	7,87
AISI 317	570	250	290	193	8,5	7,97
AISI 321	600	230	265	197	8	7,95
AISI630 (17-4PH)	1300	1500	450	202	9	7,82
Co Alloy						
Co-Cr-Mo	700	900	350	225	10	8,6
Ti Alloy						
Ti-6Al-4V	1020	1050	625	114	7,5	4,42
Fe Superalloy						
SAE A-286	110	600	370	201	9	7,92

Legend:
R_m, **MPa** - Tensile Strength; R_{eh}, **MPa** - Yield Strength; R_{ds},**MPa** - Dynamic Durability; **E,GPa** - Young's Module; **OH**, / - Wear Resistance; **R**, **g/cm³** - Density

Knee joint functional properties represent its primary usage – movement in sagital plane. Three different knee joint positions are simulated: 15°,45° and 60° of flexion [2,11]. Tibial component was fixed in every position and load was applied on femoral component. Loads correspond to human weight: 50 kg, 75 kg, 100 kg i 125 kg.

Knee prosthesis model and finite element (FE) analysis are made in Catia V5 (Fig. 2). Tibial implant is designed with fixed bearing and sagital radius, which resulted in lower movement comfort.

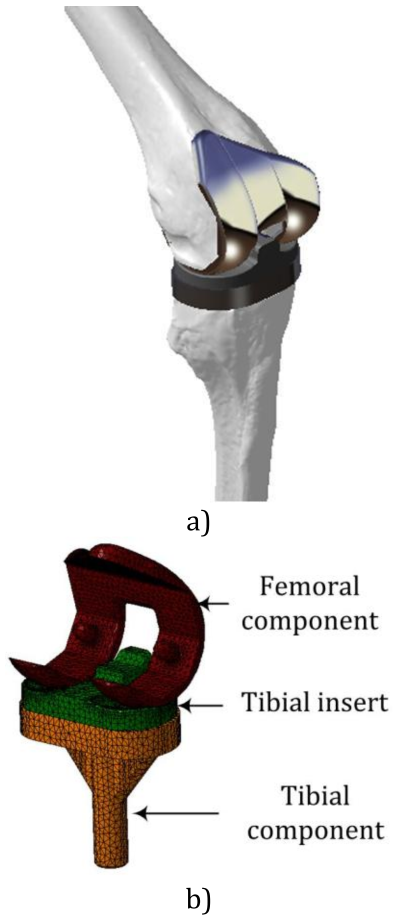

Fig. 2. a) Total knee endoprosthesis model [2], b) Total knee endoprosthesis FEA model.

Mesh of the FE models is defined to represent materials characteristics and structural properties that define the way structure will react on load. Optimal model elements are parabolic tetrahedrals with 2.5mm side.

Analyzed models correspond only to numerical, because key factors such as muscles and ligaments are not taken into consideration. Simulation represents axial load, while effects of torsion forces and flexion forces are not taken into consideration [2,17].

Several body assumptions are introduced for FEA simulation:

- Tibial implant is deformed body with Poason coefficient of 0.4;
- Tibial implant elasticity module is 1048MPa;
- Lower area of tibial implant is attached on fixed area of tibial component;
- Femoral component is rigid body
- Tibial component is rigid body.

Contact between femoral component and tibial implant is simulated like rigid body (femoral component) deforms soft body (tibial implant) [2,17].

3. RESULTS AND DISCUSSION

With rising application of UHMWPE as biomaterial, many researchers try to define mechanics and cause of tibial insert (made of this material) damage in exploitation. Considering that in vitro experiments cannot fully describe everyday activities, FE method is unavoidable in results analysis [11-14].

FE analysis defines stress distribution on tibial insert for different knee flexion values show in Fig. 3. Stresses are approximately equal for body weights and knee flexion degree.

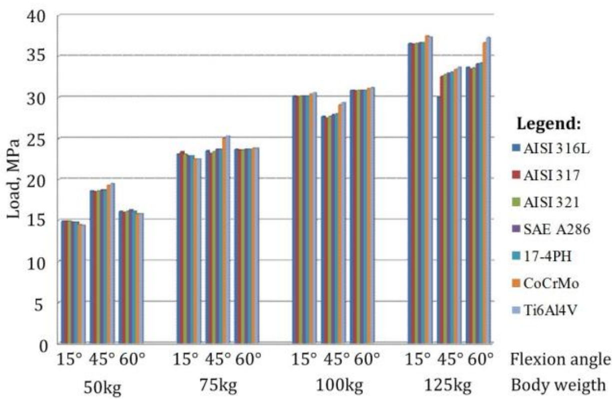

Fig.3. Histogram of the tibial insert contact stress depending on flexion angle degree, body weight, and femoral component material type [2].

In the first scenario, values of tibial insert contact stresses and metal femoral component for body weight of 50 kg are considered. In the case of 15^0 of the knee flexion maximum values of contact pressures are around 15 MPa and are approximately equal. In the case of 45^0 knee flexion, maximum values of contact pressures are around 20 MPa and Co and Ti alloys create somewhat bigger pressure on UHMWPE in comparison with other alloys. In the case of 60^0 knee flexion maximum values of contact pressures are around 16.3 MPa and are approximately equal.

In the second scenario values of tibial implant contact pressures and metal femoral components for body weight of 75 kg and different knee flexion degrees are considered. In the case of 15^0 knee flexion maximum values of contact pressures are around 23.4 MPa. In the case of 45^0 knee flexion, maximum values of contact pressures are around 25.3 MPa. In the case of 60^0 knee flexion maximum values of contact pressures are around 23.8 MPa.

In the third scenario values of tibial implant contact pressures and metal femoral components for body weight of 100 kg and different knee flexion degrees are considered. In the case of 15^0 knee flexion maximum values of contact pressures are around 30.6 MPa. In the case of 45^0 knee flexion, maximum values of contact pressures are around 29.4 MPa. In the case of 60^0 knee flexion maximum values of contact pressures are around 31.2 MPa.

In the fourth scenario values of tibial implant contact pressures and metal femoral components for body weight of 125 kg and different knee flexion degrees are considered. In the case of 15^0 knee flexion maximum values of contact pressures are around 37.4 MPa. In the case of 45^0 knee flexion, maximum values of contact pressures are around 33.7 MPa. In the case of 60^0 knee flexion maximum values of contact pressures are around 37.3 MPa.

Results obtained in FE analyisis are in agreement with D.J. Van den Heever et al. which were got similar results while analyzing three different knee joint endhoprothesis constructions [6, 18]. Also, F. Živić et al. were examined dynamic friction coefficient behavior during dry sliding contact between UHMWPE and Al_2O_3 materials. Examination was performend on the nanotribometer. They used five load values in range 100 – 1000 mN and three sliding speeds in range 4 do 12 mm/s. They showed that this coefficient has relatively low value and maximal contact stress values concide with our results [3].

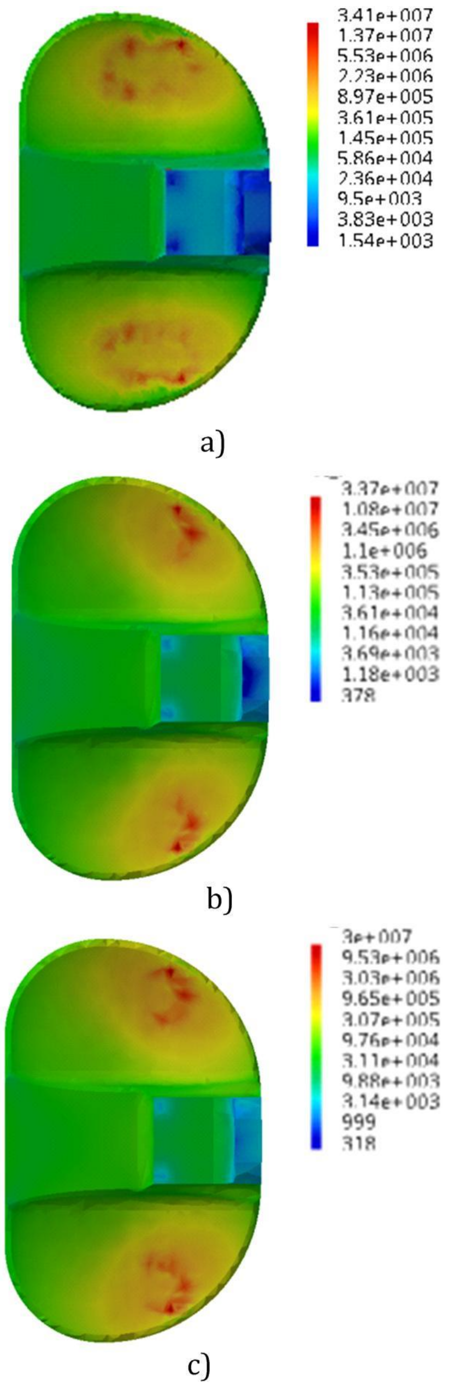

Fig. 4. Von Misses Stress for flexion of: a) 15°, b) 45°, and c) 60° [2].

FE analyses showed that maximal contact stress occur on lateral and medial part of tibia in the middle of contact between tibial implant and femoral component. According to J.J Rawlinson et al. contact pressure values can be linked with piting occurrence [19].

In the case of 15^0 knee flexion maximal contact pressures are identified in the dent of tibial implant. S. O'Brien et al. developed computer model for predicting tibial implant (made of UHMWPE materials) wear. They concluded that calculated factor of wearing articular area is 1.0370.22_10_7 mm3/Nm. Their results showed contact stress values of 26.2 MPa which coincides with our analysis for weight of 70 kg [20].

Medial side of endoprothesis have bigger areal conformity than lateral side, because of shape of femoral condil component. That is the reason why both sides have approximately same contact area in the case of small flexion degree. With increase of knee flexion degree, contact area and maximal stresses are moving anterior. In that case maximal pressures on lateral side are higher. Similar results are found by C.H.Cho et al. in their study [21,22].

4. CONCLUSION

Knee joint represents key link for ensuring erect posture, walking and amortizing concussion during contact with the surface. In every phase of movement there are static and dynamic pressures that can lead to degenerative changes on knee joint and knee joint endoprothesis. Deeper understanding of forming and development of degenerative changes mechanics asks for precise geometrical, kinematical, mechanical and tribological knee joint models. In our study we used 3D knee joint endoprothesis models with enlarged sagital radius and corresponding kinematics. Considering that exponential dependency between stresses and wear is identified in literature, we were changing prosthesis materials and gait cycle parameters and analyze contact pressure location. Results show uniform pressure distribution in lateral and medial part of tibial implant which corresponds to results shown in literature regarding wear occurrences. Further research is aimed at developing integral system for gait analysis for clinical needs. This system will aid surgeons for selecting optimal prosthesis design for every single patient. Thus, prosthesis will be lasting, and its revisions will be delayed.

Acknowledgement

This work presents a part of the research work under the projects "Application of Biomedical Engineering in Preclinical and Clinical Practice", supported by Ministry of Education and Science of Serbia (Grant No. III-41007), "Gait Analysis Lab" supported by Faculty of Medical Sciences, University of Kragujevac, Serbia (Grant No. JP 20/10), and BioEMIS, "Studies in Bioengineering and Medical Informatics" (530423 - TEMPUS - 1 - 2012 - 1 - UK - TEMPUS – JPCR), funded by EC-EACEA.

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