METHODS OF COATING AND CONTROL OF THE SURFACES FOR HIP PROSTHESIS COMPONENTS

Valentin GORNOAVA [1,2], Gheorghe I. GHEORGHE[1,2], Dragos-Viorel BREZOI[1]

[1]Valahia University, Targoviste, România; [2] National Institute of Research and Development for Mechatronics and Measurement Technique, Bucureşti, România

Abstract*: The hip joint is a ball joint with three freedom degrees (characterizing rotations around the three axes of a Cartesian system), which can theoretically be controlled only by six muscles. In reality, the mobility of the joints is controlled by 22 muscles, most of whom wear material with multiple functions, which requires improved tribological performance through constructive changes and even the principle of operation. The strain relief can be achieved by increasing the durability of the materials constituting the prosthetic components, a different surface chemistry in order to reduce friction between adhesive components. This paper presents morphological studies of nanoparticle layers deposited on the surface of hip prostheses by AFM microscopy.*

Keywords: *nanotechnology, technological flowchart*

1. INTRODUCTION

Implantation of total hip prostheses is currently conducted around the world, with a total of approx. 1,000 prostheses per day is estimated that 80% of them are total hip prosthesis replacement of primary (THR). It is estimated that total hip prostheses have a lifespan between 15 and 20 years. An endoprosthetic hip is efficient only it give the proof of a survival rate by 95% after 10 years of operation. This quality is considered polifactorial as a result of the influence of material properties from who is manufactured prosthesis design components, anchoring strategy, accurate surgical techniques.

To restore natural movement of the hip, total hip prosthesis consists of three main elements: rod or tail, which fits into the femur and provides fixation of the prosthesis head or ball that replaces the femoral head and acetabular cup which replaces cavity (Figure 1). In addition there is also located outside the metal shell of the acetabular component. [1]

The destruction of the structure of a hip prosthesis depends on the size of the particle volume and the total volume thereof or the type of material. Basically, there is a material perfectly smooth, without any roughness. These rough pre-existing natural systems add wear, giving rise to surface roughness prosthetic bone or higher. Depending on the composition (ceramic, metal), such systems can be used for different periods of time. Articular surfaces are damaged mainly due to high surface pressure produced by mechanical movements of the body. The same happens in the case of hip prostheses.

Wear a hip prosthesis depends on the number of friction cycles that confronts and how much time is not in the body of the patient. Rate wear dentures varies greatly from one patient to another because their work is very different. An individual with average activity, do steps 1 million/year. Most Active reach 3.2 million cycles/year. The elderly, less active cycles are 0.2-0.5 million / year. Men younger than 60 years go by 40% more than those over 60 years. Men go with 28% more than women [2]

In this case, wear can occur in the following areas of a hip prosthesis:

- Outside and inside of the shell
- Femoral head
- Stem prosthesis.

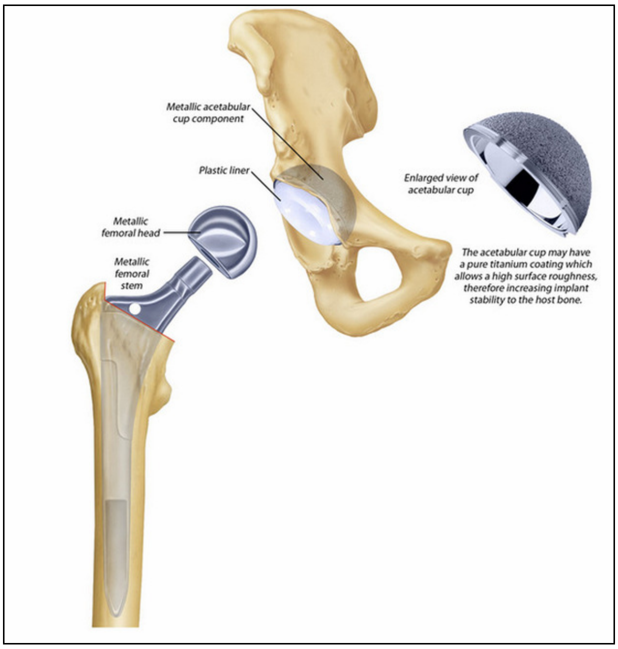


Figure 1. The componenets of the hip prostheses [3]

Inside of this the shell it is a micro movement that can lead to a destruction of the material time. In addition, indications for revision surgery leading to total hip prosthesis components include the wrong position, loss of acetabular cups or rod, significant wear of the polyethylene prosthesis instability, osteolysis and infection. A higher risk of wear or osteolysis occurs if need fixing screw that requires a particular design of the implant

2. MATERIALS USED FOR THE CONSTRUCTION OF HIP PROSTHESES

Materials commonly used to manufacture total hip prostheses must possess certain properties: chemical composition biocompatible to avoid adverse reactions of the human body, an excellent resistance to degradation (corrosion) in the human body acceptable strength to support the cyclic load borne by hinge, a low modulus to reduce the resorption of bone, as well as a high resistance to wear in order to minimize the generation of particles.

The implants are made of a wide variety of alloys, containing from two to eight metal. The implants used in orthopedic surgery are made from three main different classes of alloys, each showing characteristic components, along with those made of pure metals:

- pure metals are titanium-Ti-Ta tantalum and gold-Au;
- alloys based on iron-iron with a high content of chromium Cr, called stainless steels;
- alloys based on cobalt-Co, with a chromium content of 25-30% Cr, 5-7% Mo molybdenum and small amounts of other metals such as nickel Ni, manganese Mn, zirconium Zr, tin Sn , namely about 20% chromium Cr, 10% Ni nickel and up to 15% tungsten W;
- alloys based on titanium Ti, with 70-90% or more of titanium, Ti, containing small amounts of other metals: aluminum, Al, vanadium, V, niobium, Nb, tantalum Ta, manganese Mn, zirconium -Zr and tin Sn.

Currently, the most used materials are high molecular weight polyethylene (UHMWPE) - for acetabular cup and titanium alloy (Ti-6Al-4V) or Co-Cr for the metal parts of the prosthesis.

3. METHODS FOR REDUCING THE WEAR

Engineered surfaces gives the alternative opportunity to reduce wear, wear particle production and it release of ions in metal-metal joints. [5]

The factors that could contribute to this reduction are:
- Increasing the sustainability components;
- A different surface chemistry to reduce friction tape;
- Coated components remain undamaged;
- Methods used for coating.

The thin films are deposited on the prostheses by using different techniques. The most common technique for depositing a coating are physical and chemical techniques: physical vapor deposition, chemical vapor deposition, thermal sprayed coating used for plasma, plasma vacuum, plasma electrolytic oxidation, deposition by laser pulses, the method for polymer replication.

4. THE THICKNESS OF THE COATING

According to other researchers, the thickness of the coating layer plays an important role in connection for the hardness obtained. A layer of spheres with three folds (folds) for a CoCr coatings produce higher values than a simple layer, while a fully porous layer leads to a decrease in shear force. Bone is unable to grow arbitrary porous coatings. It was achieved a higher interfacial force for coverage of 0.5 mm compared with coatings 1 and 1.5 mm thick. Thus, it was concluded that coating thicknesses is greater than 0.5 mm and doesn't offer no mechanical advantage. Turner and his colleagues have made a comparison of the efficiency of the different porous titanium coating, bead, fiber and powder. The bone contact measurements, it was observed that the differences between different types of coatings are not significant even after one month or after 6 months (Table 1).

Table 1. Percentage of different bone contact porous coatings for prostheses after 1 month and 6 months in animal experiments (Turner) [4]

Coating with titanium	1 month	6 months
Metalic fiber	21.8 ± 2.5%	43.2 ± 4.5%
Siterizate sphere	34.0 ± 4.2%	33.3 ± 2.9%
Spraying with plasma	32.5 ± 3.8%	41.9 ± 5.2%

Bone resorption were observed in the proximal if the rod was covered on its whole length. The most important conclusion of these experiments was is why it is better if the femoral stem is not fully covered (remote) to counteract osteoporosis force protection. There is also the possibility that the excessive rigidity of the prosthesis to be responsible for this phenomenon.

The experimental observations in the case of porous coatings of prosthetic components for noncementless linkage were the following:
- An increase of material will occur if the material is inert and if there is no movement at the location of the implant;
- Pore size should possess a minimum of 50 µm and 400 µm to ensure maximum fixed and rapid stabilization of the implant;
- Based implant should be slightly smaller than the implanted adjusted to ensure primary stability and produce direct contact between bone and porous surface;

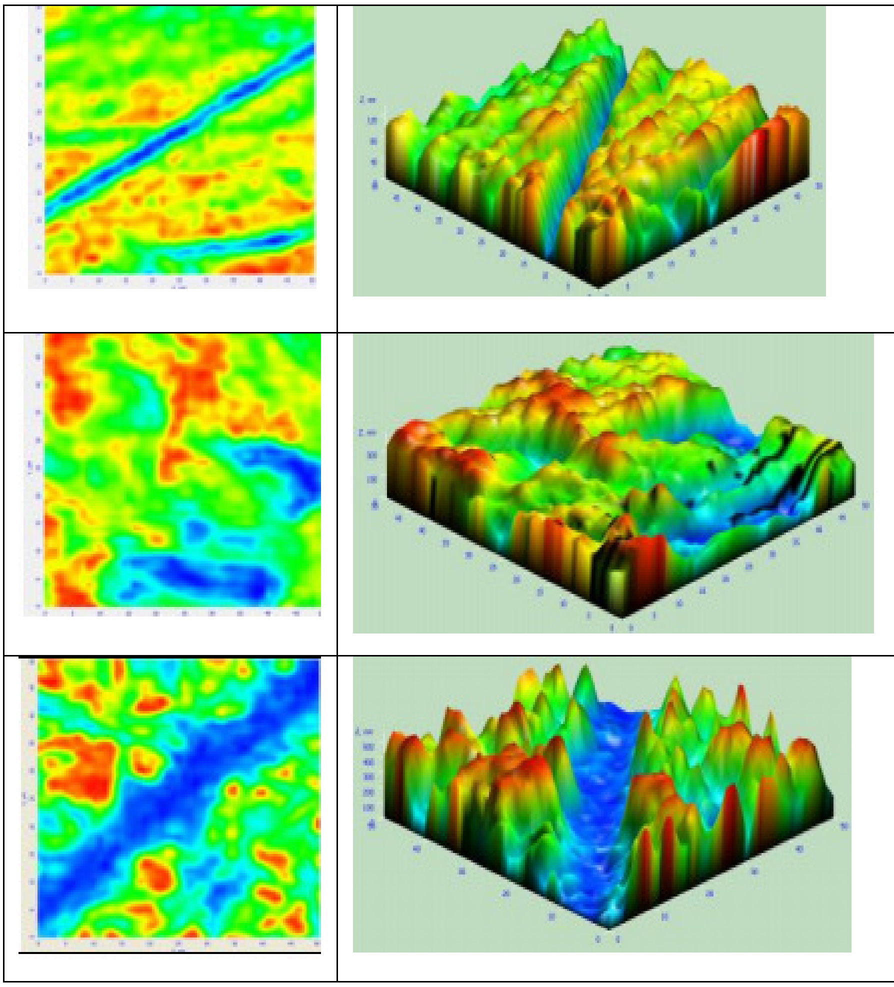
- A reduction in the load should be observed for at least 3 weeks to ensure bone growth into the pores, and thus promote rapid fixing side;
- the prosthesis stem should be covered only about to accommodate the transfer of force to the physiological conditions and to prevent osteoporosis force protection in the area.

5. APPLICATION

The surface layer of TiN deposited at 10,000 pulses is higher than the uniformity of the deposited layer at 5000 pulses., After scanning the 2D and 3D images in this transformation. It can be seen these images and small surface defects, but their size is reduced. The average surface roughness of the TiN layer is deposited on 10000 26 334 nm pulses

The TiN layer deposited at 5000 pulses has a lower surface uniformity, which could be the result of more pronounced unevenness of the substrate. These results can be seen in Table 1, after scanning the 2D and 3D images in this transformation. Relatively little surface roughness values differ from one area to another, so that the TiN layer deposited at 5000 pulses roughness average value of the results of these measurements is 40 013 nm.

The surface of TiN layer is deposited to 20000 pulses uniformity largest of the three tests performed (5000 pulses, pulses 10000 and 20000 pulses).



Amount of sampling	2601
Max	223.407 nm
Min	0 nm
Peak-to-peak, Sy	223.407 nm
Ten point height, Sz	108.38 nm
Average	70.178 nm
Average Roughness, Sa	22.4144 nm
Second moment	76.2224
Root Mean Square, Sq	29.7473 nm
Surface skewness, Ssk	1.13235
Coefficient of kurtosis, Ska	2.17863
Entropy	5.88607
Redundance	-19.2337

Amount of sampling	2601
Max	129.152 nm
Min	0 nm
Peak-to-peak, Sy	129.152 nm
Ten point height, Sz	65.8396 nm
Average	107.623 nm
Average Roughness, Sa	16.0471 nm
Second moment	110.625
Root Mean Square, Sq	25.5987 nm
Surface skewness, Ssk	-2.47863
Coefficient of kurtosis, Ska	5.71456
Entropy	4.95425
Redundance	-27.2712

Amount of sampling	2601
Max	22.034 nm
Min	0 nm
Peak-to-peak, Sy	22.034 nm
Ten point height, Sz	11.3069 nm
Average	14.0003 nm
Average Roughness, Sa	2.14117 nm
Second moment	14.2813
Root Mean Square, Sq	2.81941 nm
Surface skewness, Ssk	-0.71623
Coefficient of kurtosis, Ska	1.97222
Entropy	4.25952
Redundance	-134.467

Figure 2. For a complete characterization of TiN coatings deposited by a number of pulses was used the atomic force microscopy, the research took place in MEMS and NEMS Laboratory of Mechatronics INCDMTM

This can be seen from 2D images of the scanning AFM and transformation of this image in 3D. The average roughness of the deposited layer is 2527 nm to 20000 pulses. It is the smallest value of the average roughness of the layer of these samples demonstrated uniformity and surface coating, visible from AFM scans.

6. CONCLUSIONS

The titanium and its blends are materials commonly used due to their low density, high mechanical strength, good corrosion resistance in body fluids, low density, biocompatibility and lack of toxicity. However, mixtures have poor tribological characteristics during dry sliding, including a high friction coefficient, low wear resistance and high susceptibility to seizures. In particular, Ti-6Al-4V mixture has high resistance to tension and fatigue. Ti-6Al-4V is a mixture of titanium microstructure is thermomechanically processed to create the desired amount of two-phase fine particles optimal mechanical properties.

7. REFERENCES

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