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MODERN METHODS FOR INCREASING WEAR RESISTANCE OF HIP PROSTHESES SURFACES

LILIANA LAURA BADITA*, GHEORGHE ION GHEORGHE
National institute of Research and Development in Mechatronics and Measurement Technique, Pantelimon Road, no. 6-8, district 2, Bucharest, Romania

In this paper are presented the most common techniques used to increase wear resistance of hip prostheses. Total hip prosthesis (THP) is a bio-tribo-system with durability influenced by mechanical, thermal, chemical and biological factors. Depending on all these influences its durability is generally limited to almost 15 years. We decided to study and to improve tribological properties of hip prostheses, in order to increase the functionality and life span of these joint implants. The study of tribological properties and characterization of hip prostheses is realized by atomic force microscopy (AFM). In order to improve hip prostheses durability we used physical laser deposition (PLD) technique in order to deposit TiN layer on femoral head's surface. The main conclusion of this study is that TiN coatings offer the possibility to improve system properties, as demonstrated from scratch tests performed.

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1. Introduction

Hip joint is a ball joint restricted to three degrees of freedom (characterizing spins around three axes of a Cartesian system), which can theoretically be controlled by only six muscles. In reality, the joint mobility is controlled by 22 muscles, most of them having multiple functions. This joint forms the primary connection between the lower limb and the axial skeleton of the trunk and pelvis. The rounded femoral head attaches directly to the acetabulum and to the stem by a thin neck region.

The hip joint provides static and locomotion, playing an important role in human body positioning. It transmits the body weight data by the pelvis to the femur in mono- or bipodal support phase and swing phase of leg required during the movement.

Six different kinds of movements are possible in the hip joint: flexion and extension (on or from the spine and on or from the thigh), abduction and adduction of the femur, internal (medial) and external (lateral) rotation of the pelvis, thigh or spine.

These movements and the other functions of hip joint may be disturbed due to illness, accidents, and under the negative influence of many other lifestyle factors. The thin neck region is often prone to fracture in the elderly, mainly due to the degenerative effects of osteoporosis. The results of these disturbances and the final stages of different diseases cause joint disability and pronounced pain sensations. In such cases, pseudo joints implantation is the only way to alleviate pain, restore limb length and joint mobility.

These implantations are realized by a modern method of treatment - arthroplasty surgery – consisting in the regeneration of the joints fragments damaged by disease. Implantations of total hip prostheses (THP) are currently performed worldwide, with a number of approximately 1000/ day.

*Corresponding author: badita_l@yahoo.com

To restore the natural movement of the hip, THP consists of three parts (Fig. 1): an acetabular cup that replaces the hip socket; a ball that will replace the fractured head of the femur; a stem that is attached to the bone to add stability to the prostheses. In addition, there is a shell located outside the acetabular component.

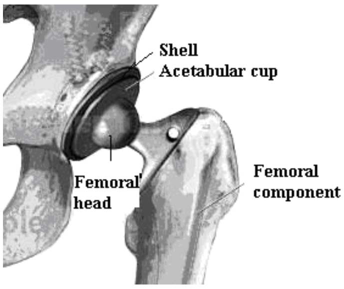


Fig. 1: Components of a total hip prosthesis [1].

Arthroplasty is properly performed if the moves produced after its realization are painless, and the surfaces of the friction torque work with a decreased friction coefficient. A hip prosthetic system is performing good only if it makes the evidence of 95% survival rates after 10 years of operation. This quality is considered polifactorial as a result of the influence of material properties, design of components, anchoring strategy and the accuracy of surgical techniques.

If the components were well positioned and secured, the limits of prostheses survival are related to the process of stability's loss. This means prosthetic components wear, accumulation of wear particles flows, loss of fixation. Wear is seen as the reason of osteolysis occurrence and the main cause for prostheses replacement.

2. Constructive changes

High surface pressure, produced by mechanical movements of the body, is the principal factor that deteriorates hip prostheses surfaces.

Wear may occur outside and inside the shell, head and stem of hip prosthesis. Inside the shell is a micro-movement which can lead to a destruction of the material, in time. Wear of a hip prosthesis depends on the number of friction cycles it is subjected to, and not to the time it stays in the patient's body. Wear rate of prostheses differs greatly from one patient to another because their activity is very different. For example, a person with average activity is about 1 million steps per year. Most active ones reach 3.2 million cycles per year. Older less active reach 0.2 to 0.5 million cycles per year. Men, younger than 60 years go with 40% more than those over 60 years. Men go 28% more than women.

The deterioration process has a complex mechanism, combining abrasive wear, adhesive wear, third body wear and fatigue wear. Wear by fretting and corrosion wear are added to these ones. In THP these wear process mechanisms are combined in different times, depending on the strains acting on the prosthesis [2].

Destroying the structure of a THP depends on the size and volume of particles and their total or type of material. Basically, there is no perfectly smooth material, without any roughness. In these rough natural systems a plus depreciation will appear, causing bone or prosthetic surfaces with high roughness. Particles resulting from friction are comprised of: bone, polymethylmethacrylate - generally used for cementing, metal alloys of prosthesis, or particles resulting from metal corrosion. In turn, these particles will reach the primary friction surfaces (the components of the prosthesis), increasing wear at this level.

3. Methods to reduce wear

Completely removing the problems associated with the use of hip prostheses, i.e. loosening and fractures, rejection physiological reactions of the body and materials wear, has not succeeded. Therefore, it is necessary to realize a resistant prosthesis, with high mechanical properties and anticorrosive composition. It was tried to improve tribological performance through structural changes and even the operating principle. Surface engineering offers an alternative to reduce wear, wear particles production and release of ions in metal-metal couplings. Factors that could contribute to this reduction are: increasing the components durability, a different surface chemistry to reduce adhesive friction and coated components that will be undestroyed.

Increased durability of components. First, the basic materials used for construction of hip prostheses were polyethylene and ceramics. Joints made of ceramics have better survival than polyethylene. Ceramic heads were placed in polyethylene joints to reduce the degree of wear. Zirconium oxide has a higher resistance to fracture, and bending lower wear rates against polyethylene in vitro [3]. Prosthetic joints of aluminium oxide surface provide very good protection against adhesive wear and poor protection against abrasive wear. It is a fragile surface which resists breaking. However, ceramics, with extremely low wear rates, is associated with a risk of failure because of the inherently low cracking toughness of the materials. So fixing ceramic acetabular components may be a problem [4, 5]. To improve the wear resistance of ceramic and polyethylene prostheses, have appeared hip prostheses made of metal without cement.

Metal hip prostheses have emerged, with increased interest in the late eighties, with the introduction of so-called “second generation metal components”. Improvements that were introduced with this generation were suitable space and a higher and more uniform hardness of metal. Metallic materials are resistant to breakage, even if they are relatively soft. Metallic biomaterials have different influences on the human body, according to: concentration of metal, exposure time and route of administration. Titanium and titanium alloys are commonly used for these kinds of prostheses due to their high mechanical strength, good corrosion resistance and biocompatibility. However, the compounds have poor tribological characteristics during dry foil, including a high friction coefficient, low wear resistance and high susceptibility to seizures. In particular, Ti-6Al-4V compound has high resistance to stress and fatigue. Zirconium alloy is relatively soft when compared with cobalt-chrome alloy femoral heads. It may deform in contact with acetabular shell materials in the case of dislocation.

Loss of acetabular cup occurs as a result of insufficient implant’s fixation and is manifested as movement between prosthesis and bone. Another factor restricting the development and incorporation of bone graft implants is low porosity of the implants often used. To resolve this limitation, new structural porous biomaterials have been developed. Trabecular metal (TM) (Zimmer Inc., Warsaw, IND) is a material with high porosity (80%) made of tantalum and was approved for use in acetabular cups [6]. Three dimensional structure of the TM (Fig. 2) is composed of a series of interconnected dodecahedron pores that have an average diameter of 550 :m. The pores size is within limits considered optimal for bone and soft tissue development and is similar to trabecular bone.

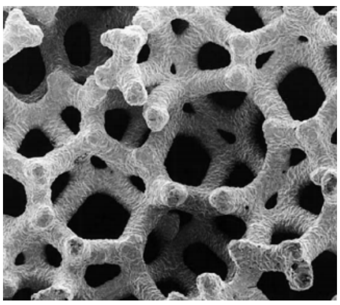


Fig. 2. Trabecular Metal (TM). An image made by scanning electron microscopy demonstrates highly porous microstructure.

Along the years, combinations of these materials came from in vitro studies conducted by researchers and clinical information published in literature. More sustainable joints were developed including highly cross-linked polyethylene – metal, metal – metal and ceramic – ceramic combinations. Composite materials with high mechanical strength, and elasticity modulus closer to that of the bone were developed in this way.

Metal – metal combination has been recommended to prevent wear prostheses in young and active patients. Cobalt – chromium – molybdenum (CoCrMo) compound is a favourite material for manufacturers and researchers to support metal – metal surfaces. Ceramic – ceramic prosthesis has the lowest rate of attrition between different prosthesis [7]. Ceramic particles induced less macrophage reaction and decreased cytokine secretion compared with high-density polyethylene particles. After tests, it was shown that survival rates for combination ceramics – polyethylene is only 97% ± 2% compared with 100% metal – metal [8].

Graphite transfer process. One way to control the process of friction is to use ceramic materials. It was made from metallurgy powders a titanium-graphite composite material with titanium particles of 150 :m and a mean particle size of graphite of 105 :m. Wear was not observed in the presence of lubrication conditions with 8% graphite composite. Factors affecting the friction properties are reduction of tangential efforts to the contact surface and wear particles transport by lubricating saline solution.

Ions implementation. Using ionic implants to change metal surfaces components of orthopaedic implants can be an effective way to reduce wear particles from the interface. In 1991 were introduced femoral heads treated with friction ions. One of the studies that show the effects of femoral head treated with ions friction on polyethylene friction is that of Maruyama and colleagues in 2000 [9]. It is believed that the ions implementation on Co-Cr femoral heads can be used to reduce clinical wear rates of polyethylene cups with 20%. Nitrogen ions implementation is a method of reducing wear on metal – metal hip prostheses components.

Kr ions irradiation surface. Budzynsky et al. [10] studied the effect of radiation damage on the friction coefficient, wear and microstiffness of Ti-6Al-4V compound after irradiation with 250 MeV krypton ions. Wear was significantly reduced after irradiation with fluence of 10^{14} cm^{-2}. Microstiffness of Ti-6Al-4V compound increased over 25% after irradiation with high fluence. Surface's microhardness after irradiation increased with fluence $D \geq 1 \times 10^{13}$ cm^{-2}.

Coating methods. The need for resistant prostheses, with anti-corrosive composition and improved mechanical properties led to application of thin films materials with superior properties onto the prostheses surfaces. Bioactive materials coatings can protect against reactivity of the metal/ ceramics with the body fluids and, at the same time, can cause fixation and growth of bone on the hip prosthesis. Metal–matrix composite (MMC) coatings reinforced with ceramic particles are promising materials for improvement of various mechanical properties. A thin hydroxyapatite (HA) coating applied on metal substrates [11], combines the mechanical and osteoconductive properties of metals with the biocompatibility of ceramics [12]. Nanoparticles and nanocrystalline materials that are readily manufactured and can substitute less performant bulk materials can also be used as coatings in bone replacements, prostheses, and implants [13].

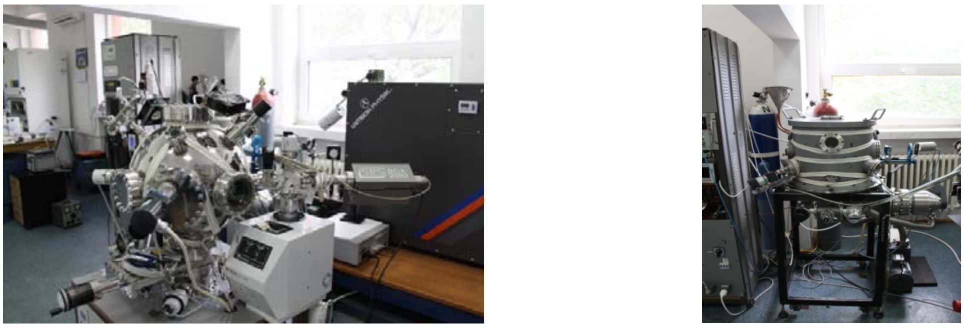
Thin films are deposited using different techniques. The most common techniques to deposit a thin film are physical vapour deposition (PVD), chemical vapour deposition (CVD), thermal sputtering. Plasma electrolytic oxidation can also be used to create hard coatings on metal substrates. The first difference between CVD and PVD is that in a CVD process the reaction occurs on the surface and in the PVD reaction occurs in the atmosphere and the compound is then deposited onto the surface.

Pulsed laser deposition (PLD) is a PVD technique, a good way to make amorphous and crystalline, dense and porous films by controlling conditions of the laser system. Bell et al. showed that a film comprising several layers of TiN, TiC, Ag and PAH on a Ti alloy, made using PLD, has toughness and an order of magnitude higher than Ti alloy. These coatings can also have a nanocrystalline structure.

4. Experiments

In order to improve resistance of femoral heads we made studies about deposition of TiN thin films on stainless steel disks by PLD. We used this material due to the fact that it is a hard biocompatible material with excellent resistance to abrasion.

PLD experiments were realized at National Institute for Laser, Plasma and Radiation Physics using a KrF* laser (λ = 248 nm, $\tau_{FWHM} \approx$ 25 ns, ν = 10 Hz), into a deposition chamber (Fig. 3a), with stainless steel reaction chamber (Fig. 3b) to 5000, 10000 and 20000 pulses. Cylindrical samples (22.5 mm in diameter and 10 mm in height), made of 316L stainless steel, with hardness of 150 HV_{30} were used as substrates. The experimental conditions during deposition process were: substrate temperature 500 ^{0}C; laser fluence ~ 5 J/cm^2, at energy of 500 mJ; the target-substrate separation distance 5 cm; the pressure inside the deposition chamber ~ 2x10^{-3} Torr.



(a) (b)

Fig. 3. Deposition (a) and (b) stainless steel reaction chambers [14].

The obtained TiN thin layers have been characterized using Atomic Force Microscopy (AFM), working in the noncontact mode. In our institute, it is used an atomic force microscope NTEGRA Probe NanoLaboratory. It is composed of a base unit where we have the sample, a special measuring head made of a cantilever with a pointed end, and an optical viewing system. AFM images are processed using Nova SPM software. In this way tribological parameters that offer information about the uniformity of these surfaces were obtained, like:

- maximal and average height of surface;
- roughness of the studied surface describes its uniformity;
- ten point height (S_z) expresses surface roughness by the selected five maximal heights and hollows, nm;
- surface skewness (S_{sk}) characterizes the non-symmetry of distribution;
- coefficient of kurtosis (S_{ka}) characterizes the distribution spread.

Wear and adhesion resistance of the deposited layers was evaluated with the tester presented in Fig. 4 (from Institute of Solid Mechanics of Romanian Academy). This system has a diamond tip allowing considerable simplification of the test device. The movement is, in this case, pure sliding, common in many tribometers with continuous or alternate motion. The tested body is the 316L stainless steel disk coated with TiN. The applied loads were in the range of (2.5 – 125) N. The test was performed at 22°C to better simulate conditions that the coatings are subjected to. Each scratch was analysed at the end by optical microscopy and atomic force microscopy (AFM).

Fig. 4. General view of scratch tester with a diamond spherical segment.

5. Results

Obtained TiN layers have been characterized by atomic force microscopy (Fig. 5).

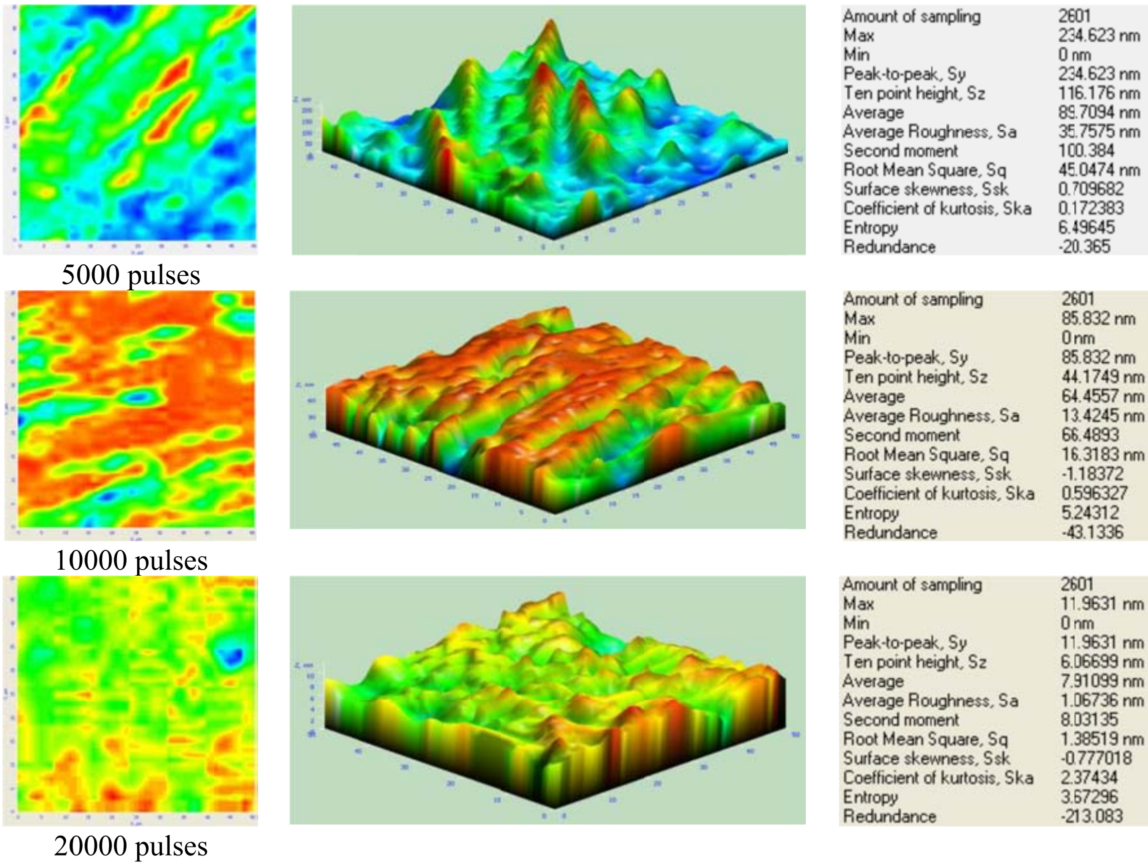


Fig. 5. Roughness of 316L stainless steel substrate coated with TiN layer measured using NTEGRA Atomic force Microscope.

TiN layer deposited at 5000 pulses has a surface with lower uniformity, which could be the result of more pronounced irregularity of the substrate. The surface of TiN layer deposited at 10000 pulses has a higher uniformity than that of the layer deposited at 5000 pulses. It can be seen from these images small surface defects, but their size is reduced. The surface of the TiN layer deposited at 20000 pulses has the highest uniformity of the all 3 realized samples (5000 pulses, 10000 pulses and 20000 pulses). It is the lowest mean roughness value of these samples, as demonstrated by the uniformity of surface coating, visible from AFM scanning.

It was observed that layers roughness values vary depending on the number of pulses the coating has been deposited. A decrease of roughness value with increasing number of pulses was observed.

The thickness of the deposited layers obtained by PLD has been determined using OM, AFM and SEM investigations, performed on polished cross sections of specimens tested for scratch. Average thickness determined by AFM was 1,67 μm for 5000 pulses, 2,11 μm for 10000 pulses and 2,72 μm for 20000 pulses. Generally, it was observed an increase of the layer's thickness with the increase of laser pulses number.

Surfaces of TiN layers of approximately 2 μm thick have been subjected to scratch tests. The applied loadings, between 2.5 N and 125 N created scratches with different depth and width. Surfaces obtained after the scratch tests have been characterized using AFM in order to determine different tribological parameters (Fig. 6).

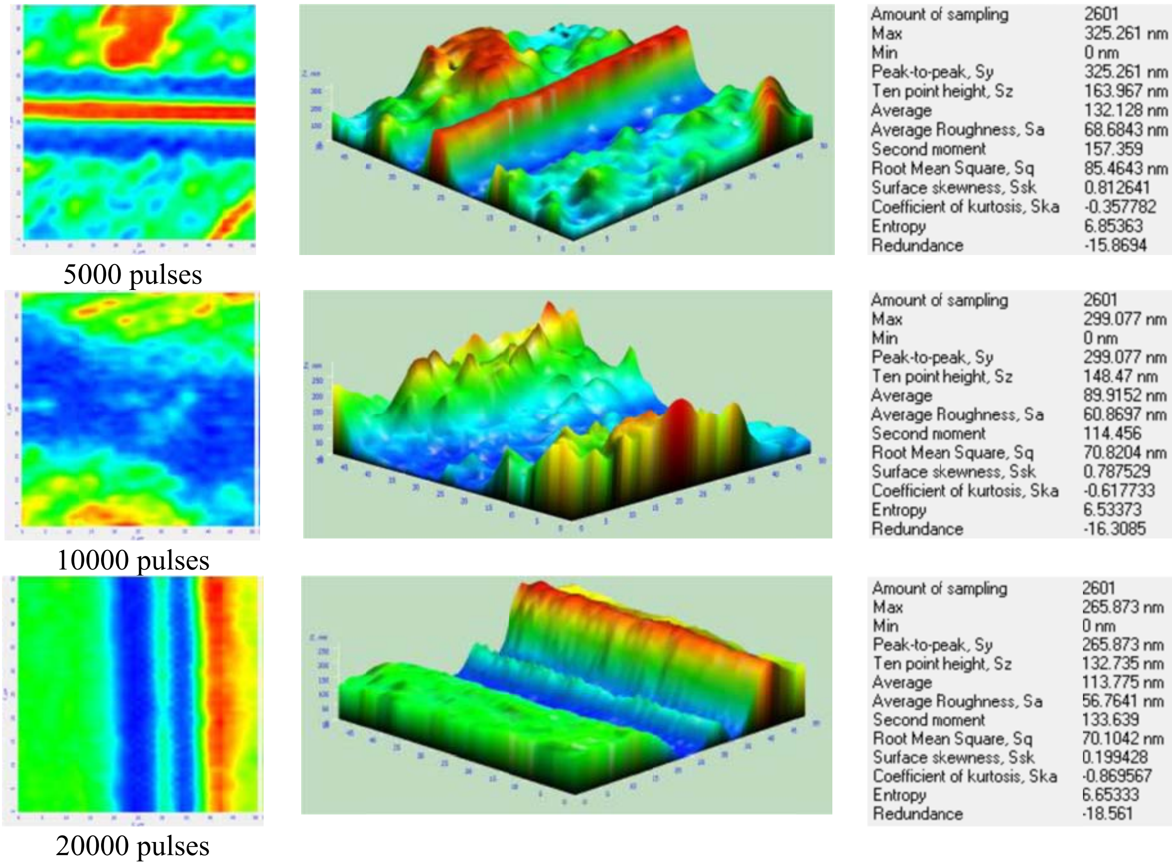


Fig. 6. AFM characterization of 316L stainless steel surfaces coated with TiN after the scratch tests at 10 N.

The roughness values of the TiN surfaces varied depending on the applied loads. These values increased with the increase of the applied load. Lower values of the roughness compared with the values of the uncoated surfaces were determined. In this way was demonstrated that TiN layers have a more uniform surface even after the scratch tests. This can be a proof for a higher resistance of TiN layers.

The main conclusion of these experiments and observations is that TiN nanostructured coatings offer the possibility to improve the properties of hip prostheses. Given the fact that the thickness of the layer increases with the number of pulses and considering that the surface of the substrate has shown some minor defects, it was concluded that the layer becomes more uniform along with a more complete coating of initial defects.

5. Conclusions

All clinical and theoretical studies conducted demonstrated the need for resistant prostheses, with high mechanical properties and anti-corrosion composition. To improve the mechanical properties of prosthesis, they were made of different materials, were processed or coated with various materials, which have superior properties.

An alternative perspective was the use of new materials for friction torque (head of alumina, zirconia head) in order to reduce the number of wear particles or completely change their nature, metal – metal or ceramics – ceramics.

Progress has been achieved through the use of very hard alloys and biocompatible materials. Certainly, an important step for future is to coat areas of hip prostheses with nanostructures. For example, femoral head coating offers the opportunity to improve hip prostheses properties.

Experiments presented in this paper demonstrate that TiN coating offers the possibility to improve the properties of hip prostheses. Atomic force microscopy was a useful technique to characterize the topography of these thin layers. Analyzing the experimental results obtained the main conclusion is that TiN protective coatings deposited by PLD technique with 20000 pulses can be an alternative technology to ensure scratch resistance of femoral heads.

An important conclusion is that nanostructured coatings offer the opportunity to improve hip prostheses properties.

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