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TRIBOLOGY IN BIOMATERIALS DESIGN AND SELECTION

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Abstract: *The paper presents review of selected biomaterials that were subject of tribological investigations at Tribology Center of Mechanical Engineering Faculty Kragujevac. Development, design and selection of biomaterials are primary determined by characteristics and nature of the tissue that is being replaced or supplemented. Modern material investigations at micro level loads enable insight into new aspects of material behaviour and offer possibilities for further improvement. Tribological investigations of biomaterials can contribute to their development by offering solutions for wear decrease and wear prediction. Short review is presented of tribological tests of four different biomaterials, namely Ti6Al4V alloy, stainless steel AISI 316LVM, ultrahigh-molecular-weight polyethylene (UHMWPE) and polymethyl methacrylate (PMMA).*

Keywords: *Biomaterials, Ti alloys, Stainless steel AISI 316LVM, UHMWPE, PMMA, Nanotribometer*

1. INTRODUCTION

Advances in biomedical materials research and development have made an enormous impact on the treatment of injury and disease of the human body. Biomaterials scientists and engineers pursue novel techniques and methodologies to study cells, their components, complex tissues and organs and their interactions with natural and synthetic materials and implanted prosthetic devices, as well as to develop and characterize the materials used to measure, restore, and improve physiologic function, and enhance survival and quality of life [1-4]. Biomaterials applications increased rapidly in the late 1800s. The first metal devices to fix bone fractures were used in late eighteenth to nineteenth century; the first total hip replacement prosthesis was implanted in 1938; and in the 1950s and 1960s, polymers were introduced for cornea replacements and as blood vessel replacements [1].

A biomaterial is any matter, surface, or construct that interacts with biological systems. Williams in 1987 defined a biomaterial as: "a nonviable material used in a medical device, intended to interact with biological systems". It is still true today, but the level of interaction of biomaterials with the biological system has changed dramatically. Biomaterial function has advanced from remaining relatively inert in the body to being "bioactive" and assisting with regeneration. Bioactive materials have the capability to initiate a biological response after implantation and research nowadays is aimed towards regeneration of a damaged tissue or whole organ through this interaction.

Multidisciplinary area of biomaterials development, design and selection is today one of the most promising field of research. There is no material suited for all biomaterial applications and new applications are continually being developed as medicine advances. There are number of questions that still need to be answered regarding the biological response to biomaterials and the optimal role of biomaterials in tissue regeneration. The usual approach to regain the function of the lost tissue was to replace it with a simple biomaterial. But as understanding of many aspects of tissues, disease, and trauma improved, the concept of attempting to repair damaged tissues emerged.

Development and application of biomaterials are directly determined by characteristics and nature of the tissue that is being replaced or supplemented. Desired or optimal properties of a biomaterial very much depend on its biomedical application. General strategies for guiding tissue repair by varying the

chemistry, structure and properties of biomaterials are subject of many ongoing research projects. Application-specific biomaterials solutions are investigated according to the major organ systems in the body. Important part of biomaterials development is also standardization and regulative to ensure safety and efficiency in this novel scientific area, including many social questions, as well.

Characterization of materials used in medical purposes represents a very complex field which is application dependent. Many interdependent parameters of the observed material are important to study and understand in relation to their material characteristics (thermomechanical processing, microstructural properties), mechanical properties (hardness, modulus of elasticity, tensile stress), corrosion and wear resistance, biocompatibility and osseointegration. It is important to determine their exploitation stability in a longer time, especially for biomaterials used for implants (knee and hip replacement, spinal implants, dental implants etc.) [2]. Investigations in relation to wear behavior, stress/strain development, characteristics such as design and selection of classical or 3D porous structure, grain orientation, porosity, functional stability (durability, biocompatibility, wear, corrosion) are realised in worldwide laboratories in order to provide quantitative parameters for further evaluation and improvement. Tribological behaviour of material is unavoidable, especially from aspect of wear. Tribological investigations by taking into account variation of contact materials and contact conditions can efficiently contribute to improvement of existing biomaterials and validation of newly developed.

Table 1. Materials and their medical use

Class of Material	
Metal	
Stainless steel	Joint replacements, bone fracture fixation, heart valves, electrodes
Titanium and titanium alloys	Joint replacements, dental bridges and dental implants, coronary stents
Cobalt-chrome alloys	Joint replacements, bone fracture fixation
Gold	Dental fillings and crowns, electrodes
Silver	Pacemaker wires, suture materials, dental amalgams
Platinum	Electrodes, neural stimulation devices
Ceramics	
Aluminum oxides	Hip implants, dental implants, cochlear replacement
Zirconia	Hip implants
Calcium phosphate	Bone graft substitutes, surface coatings on total joint replacements, cell scaffolds
Calcium sulfate	Bone graft substitutes
Carbon	Heart valve coatings, orthopedic implants
Glass	Bone graft substitutes, fillers for dental materials
Polymers	
Polymethylmethacrylate (PMMA)	Bone cement, intraocular lenses
Polyethylene (PE); Ultra-high-molecular-weight polyethylene (UHMWPE)	Hip and knee implants, artificial tendons and ligaments, synthetic vascular grafts, dentures, and facial implants
Polyvinylchloride (PVC)	Tubing, facial prostheses
Nylon	Surgical sutures, gastrointestinal segments, tracheal tubes
Silicone rubber	Finger joints, artificial skin, breast implants, intraocular lenses, catheters
Polyester	Resorbable sutures, fracture fixation, cell scaffolds, skin wound coverings, drug delivery devices
Natural Materials	
Collagen and gelatin	Cosmetic surgery, wound dressings, tissue engineering, cell scaffold
Cellulose	Drug delivery
Chitin	Wound dressings, cell scaffold, drug delivery
Ceramics or demineralized ceramics	Bone graft substitute
Alginate	Drug delivery, cell encapsulation
Hyaluronic acid	Postoperative adhesion prevention, ophthalmic and orthopedic lubricant, drug delivery, cell scaffold

Characterization of the surface layers on observed materials, to study local phenomena on the micro- and nano-scale, is realised by advanced microscopy techniques such as scanning electron microscopy (SEM), atomic force microscopy (AFM) or scanning force microscopy (SFM), transmission electron microscopy (TEM), auger electron spectroscopy (AES), etc. It is usually accompanied with chemical analysis such in case of Raman microscopy, or EDS analysis (Energy-dispersive X-ray spectroscopy (EDS or EDX), which is often integral part of SEM microscope. Wear mechanisms and microstructural characteristics can be determined by using these microscopy techniques.

Some estimates on biomaterials use [1]:

- Total hip joint replacements: 448,000
- Knee joint replacements: 452,000
- Shoulder joint replacements: 24,000
- Dental implants: 854,000
- Coronary stents: 1,204,000
- Coronary catheters: 1,328,000

Materials selection for a medical device is complicated. The selection depends on a number of factors, including the mechanical loading requirements, chemical and structural properties of the material itself, and the biological requirements. Generally, materials used for interaction with biological systems can be: metals (Ti-based alloys, stainless steel), polymers (Polymethyl methacrylate**,** PMMA, Ultra-high-molecular-weight polyethylene, UHMWPE, Polylactic acid, PLA), ceramics (Al_2O_3, Hydroxyapatite), composites and many more. Classes of material current uses for biomedical purposes are shown in Table 1 [1].

2. TITANIUM ALLOYS

Titanium and titanium alloys has broad applications in a field of medicine [1-5]. Extensive research activities in relation to different ways of Ti alloys manufacturing and its further treatments and surface modifications are aimed at characteristics improvement. Ti alloys are applied in cases when high strength and low density are of primary importance [1], even though they posses low sliding wear resistance due to its low resistance to plastic shearing. They are particularly interesting for biomedical applications because of their excellent biocompatibility, relatively low Young's modulus and high corrosion resistance [5-14]. Elastic modulus of titanium based materials vary from 55 GPa (Ti-29Nb-13Ta-7.1Zr alloy) to 112 GPa (Ti6Al4V alloy) [2], what is lower compared to 316L stainless steel (200 GPa). Elastic modulus of the bone is approximately 30 GPa and development of new alloys is aimed at achieving elastic modulus close to that of the bone. Metal alloy of nickel and titanium, nickel titanium (NiTi), also known as nitinol, is one of the novel shape memory biomaterials having elastic modulus of approximately 48 Gpa. Titanium based materials are often used for joints (e.g. femoral stem in hip replacements).

Ti6Al4V alloy becomes osteointegrated in a very short period when in contact with a bone. Despite their good mechanical and chemical properties and low density (4.5g/cm^3) the use of Ti alloys for structural applications is prevented by their poor wear resistance what is attributed mainly to two reasons: low resistance to plastic shearing what as a consequence has weak counteracting of material to wear mechanisms (adhesion, abrasion, delamination); and high flash temperatures induced by friction during dry sliding implying forming of surface oxide which greatly influences frictional and wear behaviour of the observed system [2-10]. Due to this, many surface modification techniques are applied in order to improve tribological behaviour of Ti alloys [12-14]. However, the acting wear mechanisms during sliding wear of titanium alloys have not been sufficiently addressed and understood and needs further study [6, 7, 15].

Laboratory simulations regarding biomaterial behaviour can be conducted in different ways. If wear behaviour is considered, tribological investigations are usually performed with reciprocating sliding tester with high contact pressures (e.g. ball-on-flat sliding wear test), in dry conditions or with some solutions simulating body fluids (Ringer solution, distilled water etc.). Investigations on influence of the contact load, sliding speed and environment variation offer valuable information on Ti alloy behaviour.

Real time diagrams of friction coefficient Ti6Al4V alloy sliding against alumina (Al_2O_3) under dry and lubricated conditions for selected regimes of normal force and maximum linear speed are shown in Figs. 1 and 2. Optical micrographs of worn scars are also shown in Figs. 1 and 2. Maximum penetration depth of the ball into the observed sample is also given in Figs. 1 and 2, as recorded on nanotribometer. Tribological tests were realised on the CSM Nanotribometer and linear reciprocating sliding mode was used. Static body was 1.5 mm diameter alumina ball. Moving body was flat rectangular Ti6Al4V alloy sample. Testing was done with 0.5 mm stroke (0.25 mm half amplitude) in dry conditions and with Ringer's solution, in ambient air (temperature of 25 °C). Five values of normal force were selected (100 mN, 250 mN, 500 mN, 750 mN, 1000 mN) and three values of sliding speed (4 mm/s, 8 mm/s, 12 mm/s). The values of estimated maximum contact pressure were 0.68, 0.93, 1.17, 1.34 and 1.47 GPa, respectively. Duration of one test was 30000 cycles, whereat distance of two strokes represents one cycle. Selected sliding velocities corresponds to the range of speed characteristic for hip joints testing (0–50 mm/s) [16].

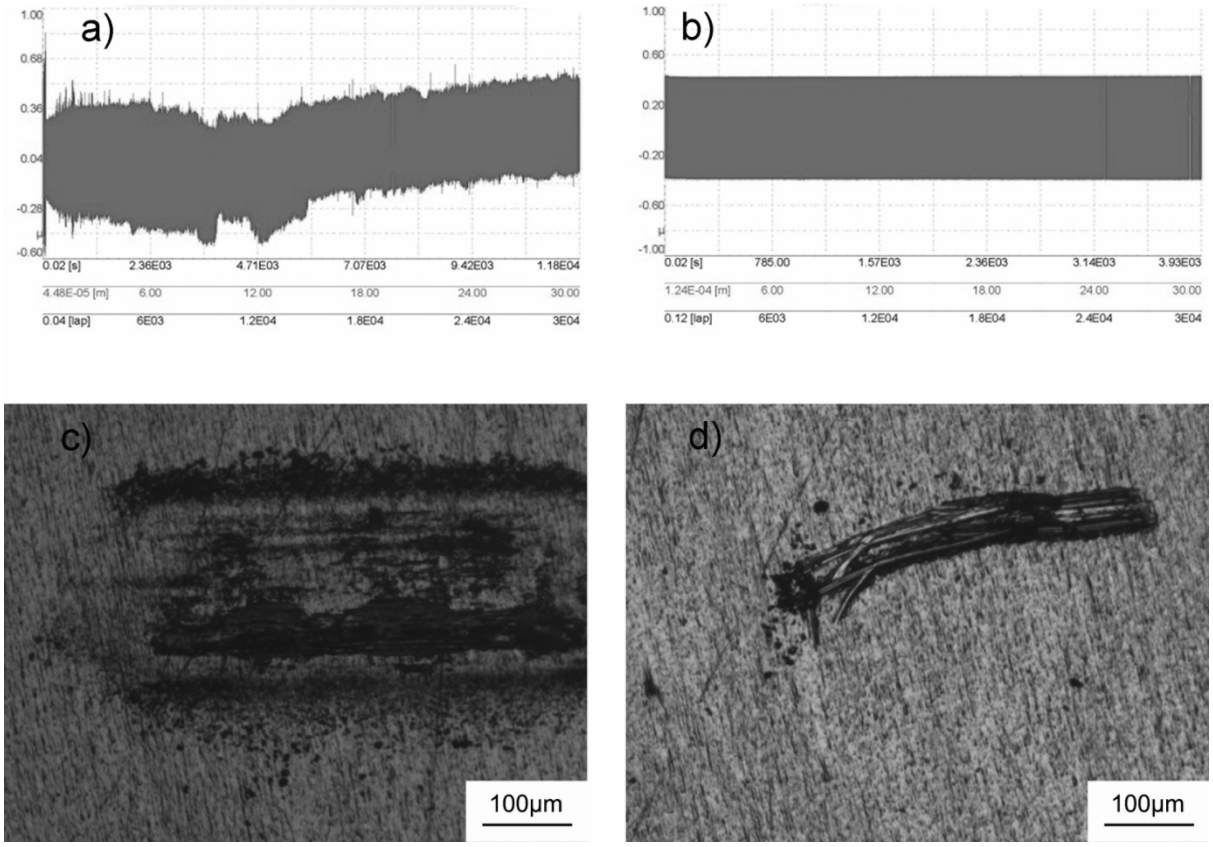

Figure 1. Diagrams of friction coefficient and optical micrograph showing wear track on Ti6Al4V sample:
a), c) Dry, v=4 mm/s; F_N=100mN; Penetration depth: 2.7µm;
b), d) Dry, v=12 mm/s; b F_N=1000mN; Penetration depth: 18.0µm;

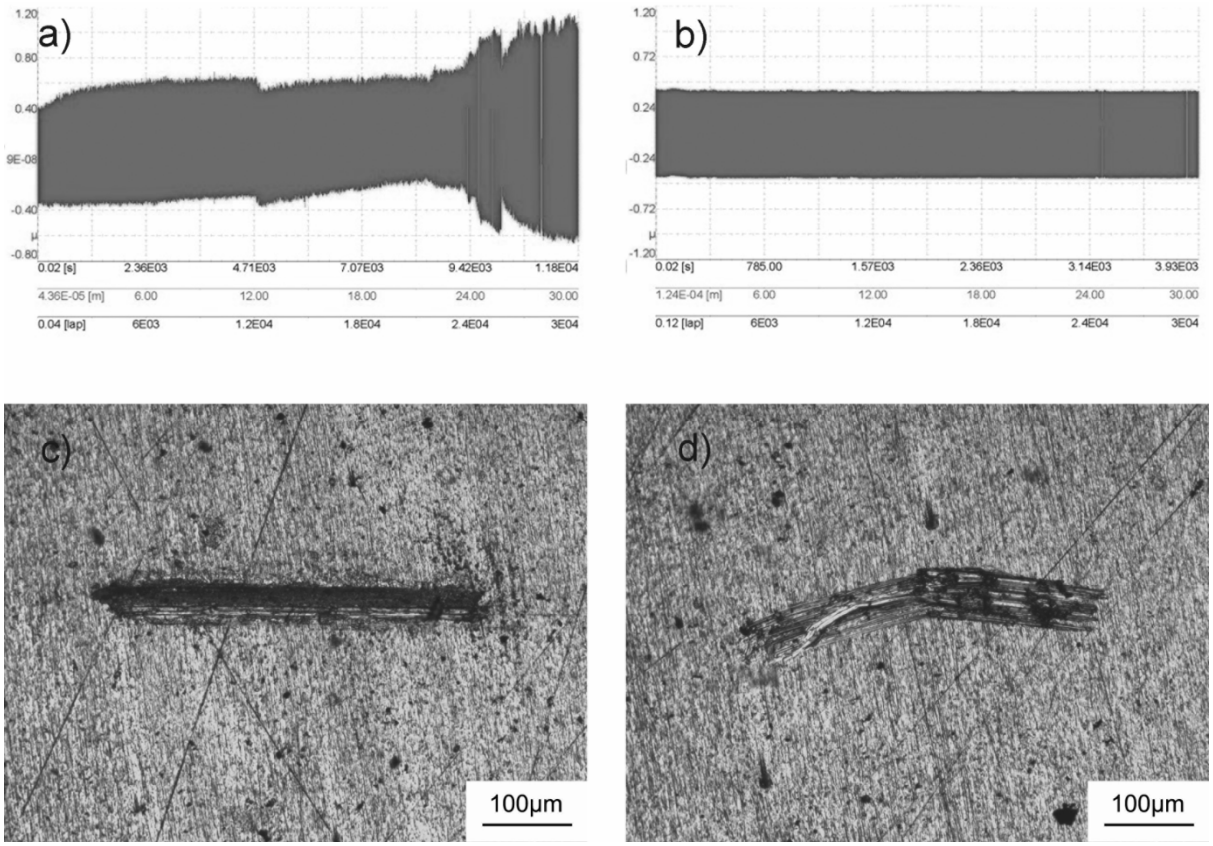

Figure 2. Sample 3: Diagrams of friction coefficient and optical micrograph showing wear track on Ti6Al4V sample:
a), c) Ringer, v=4 mm/s; F_N=100mN; Penetration depth: 25.2µm;
b), d) Ringer, v=12 mm/s; b F_N=1000mN; Penetration depth: 0.27µm;

It can be clearly seen (Figs. 1 and 2) that change of contact conditions (load, sliding speed, dry sliding or with Ringer's solution) produced significant change in tribological behaviour of Ti6Al4V samples. It is obvious that some regimes should be avoided when using Ti6Al4V alloy. Low sliding speed produced variation of the friction coefficient (Figs. 1a, 2a). In case when Ringer's solution was present, friction coefficient curve started to rise after approximately 24000 cycles, indicating further change of the behaviour. High sliding speed produced steady friction coefficient during the whole test (Figs. 1b, 2b). Worn tracks are of rather smaller dimensions at 4 mm/s sliding speed than at 12 mm/s sliding speed, especially in case of dry sliding (Figs. 1c, 1d and Figs. 2c, 2d). If optical micrographs were compared in cases of dry sliding and sliding with Ringer's solution (Figs. 1c, 2c and Figs. 1d, 2d), it can be noticed that worn tracks are significantly smaller at sliding with Ringer's solution. Presence of the Ringer's solution produced lowering of the wear level, in comparison with dry sliding.

3. STAINLESS STEEL

Application of metals as biomaterials is based on their high strength and resistance to fracture and design to resist corrosion. Many orthopedic devices are made of metal, such as stents, hip and knee joint replacements. Out of the eight coronary stents approved by the US Food and Drug Administration (FDA), seven are made from 316L stainless steel [17]. Plates and screws that hold fractured bone together during healing also are made of metal. They are sometimes retrieved after successful healing, but in other cases they are left in place. Metallic devices are also used to fuse segments of the spine together when the disk has degenerated and as dental root prosthetic implants.

Metals as biomaterials have advantages over ceramics or polymers because they are strong, tough, and ductile (or deformable, particularly as compared to ceramics). However, they are susceptible to corrosion [18] causing adverse effects on the healing process, which led to application of alloys of titanium or cobalt-chrome. Biocompatibility is an issue with metal implants and allergic reactions can occur. Metals also exhibit high density and much greater stiffness than most natural materials they replace, leading to undesirable stress shielding. For instance, after implantation of metal joint replacements loss of adjacent bone has been observed because the bone is not exposed to normal levels of mechanical loading. Solution to these problems has been investigated by application of shape memory alloys (e.g., nitinol) that can be bent or deformed and still return to their original shape when the stress is released.

Stainless steel AISI 316LVM is a Molybdenum alloyed vacuum remelted stainless steel for the production of both temporary and permanent implants. Chemical composition of AISI 316LVM steel is given in Table 2. ASTM F138-03 defines standard specification for wrought 18 chromium-14 nickel-2.5 molybdenum stainless steel bar and wire for surgical implants. DIN standard designation of the steel is DIN: X 2 CrNiMo 18 15 3. It is a vacuum melted to achieve the extremely high levels of purity and cleanliness required for surgical implants. Elastic modulus of AISI 316LVM stainless steel is approximately 200 GPa.

Table 2. Chemical composition of AISI 316LVM steel

C max	Si	Mn	P max	S max	Cr	Ni	Mo	Cu max	N max
0.025	0.6	1.7	0.025	0.003	17.5	14	2.8	0.10	0.10

Stainless steel AISI 316LVM has a very good resistance in physiological environments to general and integranular corrosion due to high purity and low ferrite content; pitting and crevice corrosion due to the high molybdenum content. However, it is highly susceptible to localized forms of corrosion [19]. Many authors investigated possibilities to predict, prevent or lower corrosion of biomedical grade 316LVM stainless steel surface [18, 19, 20].

Real time diagrams of friction coefficient of AISI 316LVM stainless steel sliding against alumina (Al_2O_3) under dry and lubricated conditions for selected regimes of normal force and linear speed are shown in Figs. 3 and 4. Optical micrographs of worn scars are also shown in Figs. 3 and 4. Penetration depth of the ball into the observed sample (Figs. 3 and 4), as recorded on nanotribometer (minus sign) indicated that wear debris was accumulated within the contact zone, during the sliding.

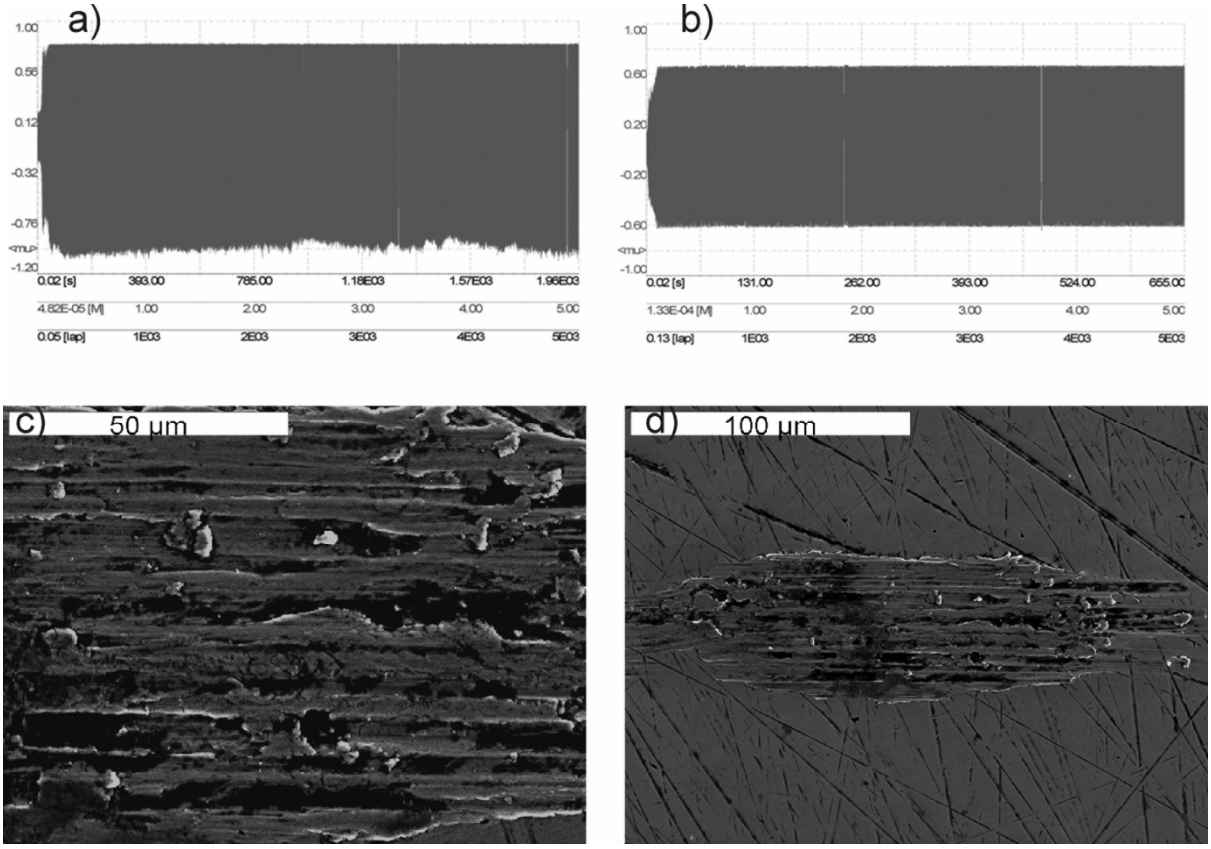

Figure 3. Diagrams of friction coefficient and optical micrograph showing wear track on AISI 316LVM sample in dry conditions: a), c) v=4 mm/s; F_N=100mN; Penetration depth: -1.9µm; b), d) v=12 mm/s; b F_N=750mN; Penetration depth: -0.7µm;

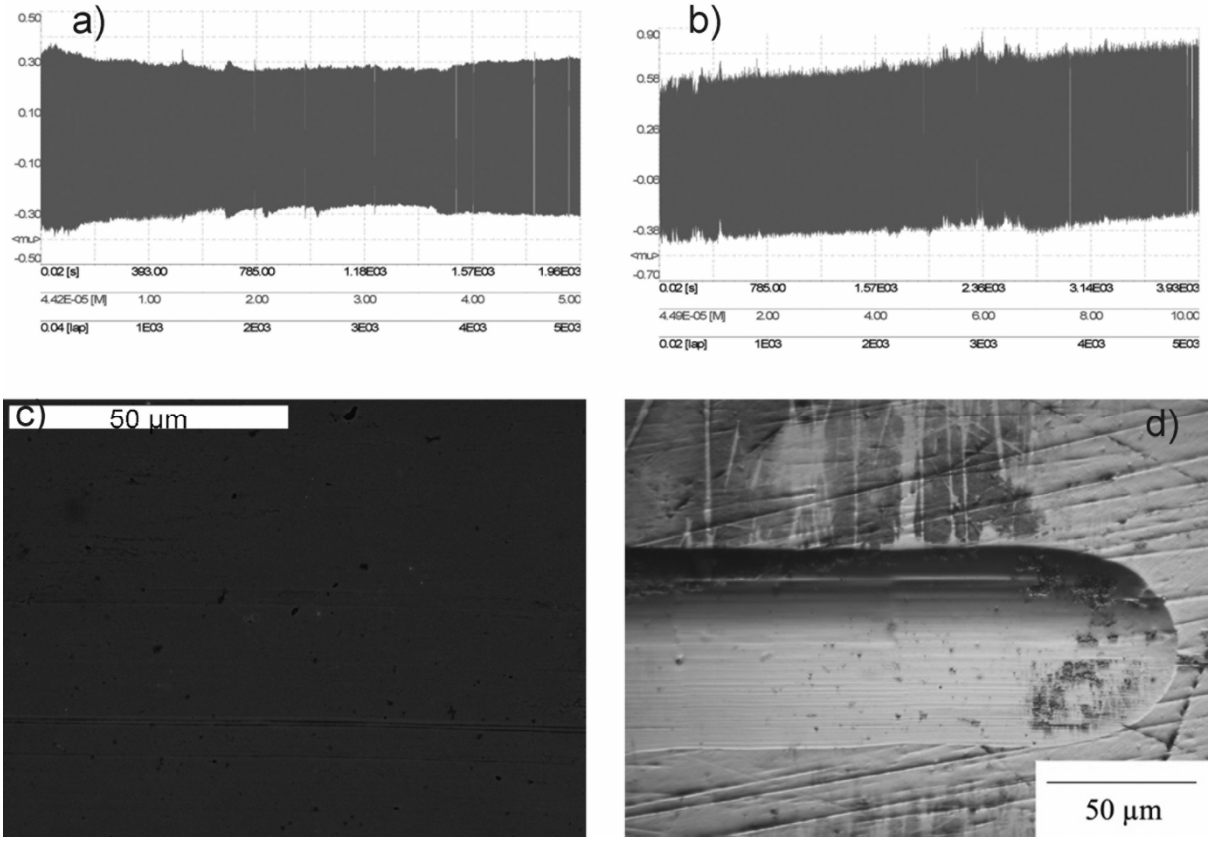

Figure 4. Diagrams of friction coefficient and optical micrograph showing wear track on AISI 316LVM sample: a), c) Ringers' solution; v=4 mm/s; F_N=1000mN; Penetration depth: 0.2µm; b), d) Ringers' solution with PMMA particles; v=12 mm/s; b F_N=750mN; Penetration depth: 3.6µm;

Tests were conducted in dry conditions and with presence of two different solutions: Ringer's solution and Ringer's solution with particles of Polymethylmethacrylate (PMMA), in order to simulate human body environment in which implanted component function. It can be clearly seen (Figs. 3 and 4) that change of contact conditions (load, sliding speed, dry sliding, sliding with Ringer's solution or with presence of Ringer's solution with PMMA particles) produced significant change in tribological behaviour of AISI 316LVM samples. Abrasive wear mechanism can be clearly seen, accompanied with adhesive wear and pitting. Samples sliding in the presence of pure Ringers' solution exhibited smooth worn tracks (Fig. 4c) in comparison to all other conditions. It can be clearly seen that presence of PMMA particles produced severe forms of pitting corrosion at leading zones of the worn track (Fig. 4d). This is in consistence with findings of other authors who reported development of localized attack types of pitting corrosion [19]. Severe effects of wear debris on bone regarding biological response have been studied, but it is clear that component also would exhibit worsening of its surface properties leading to further unwanted forms of wear. Pitting corrosion can adversely affect both biocompatibility and mechanical strength of the implant and even lead to complete mechanical failure of the component [19, 20].

4. POLYMERS: ULTRA-HIGH-MOLECULAR-WEIGHT POLYETHYLENE (UHMWPE)

Polymers are large molecules synthesized from smaller molecules, called monomers. Plastics are polymers that are rigid solids at room temperature and generally contain additional additives. Some common plastics used in biomedical applications are polymethyl methacrylate (PMMA) for intraocular lenses, and ultrahigh-molecular-weight polyethylene (UHMWPE) for the articulating surfaces of orthopedic implants. Over the years, ultra-high-molecular-weight polyethylene (UHMWPE) has emerged as the material of choice for fabricating one of the bearing components in various arthroplasties, such as acetabular cups, acetabular cup liners, tibial inserts, etc. [21]. Components made of UHMWPE have performed admirably in vivo. The only major concern is wear and the effect of the wear particles on the in vivo longevity of the prosthesis.

UHMWPE is a linear, low-pressure, polyethylene resin. It has both the highest abrasion resistance and highest impact strength of any plastic. Combined with abrasion resistance and toughness, the low coefficient of friction of UHMWPE yields a self-lubricating, non-stick surface. Static and dynamic coefficients are significantly lower than steel and most plastic materials. Elastic modulus of UHMWPE is approximately 0.69 GPa. ASTM F648-00 defines standard specification for Ultra-High-Molecular-Weight Polyethylene powder and fabricated form for surgical implants.

Despite its superior mechanical performance, the standard UHMWPE is subject to fatigue failure and it produces too many wear particles, leading to shortening of the component life and revision surgeries. It can also absorb small amounts of fluids or retain small amounts of air in the microscopically small pores (about 0,1 volume percent) that can cause great deterioration of the material during the long years in the patient's body. The main problem is production of wear particles from the UHMWPE surfaces. The wear particles enhance inflammatory reaction and eventually lead to development of osteolysis (bone dissolving disease). Along with the extensive application of UHMWPE, the understanding of polymer tribology is becoming increasingly important. Many authors have investigated tribological performance of UHMPWE [21-25].

Many strategies for reducing wear of this material have been investigated [25]. Widely accepted approach is crosslinking of the polymer. The cross-linked UHMWPE has been introduced in clinical practice in 1998, although it was known since 1960's that irradiation of UHMWPE increases its wear resistance. In the process called cross-linking, the irradiation "glues" the long polyethylene molecules to keep together. On the other hand this process deteriorates mechanical characteristic of the material and above a certain limit (maximal irradiation dose) it becomes brittle and unusable for mechanical applications. Understanding wear mechanisms occurring during the contact of UHMWPE with other biomaterials is important to be able to predict in-vivo life of the component. Small number of reports has been published regarding a correlation between in vitro and in vivo wear rate results [21].

The effect of different types of lubricant environment and varied regimes (normal load and sliding speed) on the tribological performance of UHMWPE against Al_2O_3 was examined with a reciprocating sliding setup of the nanotribometer (Figs. 5 and 6).

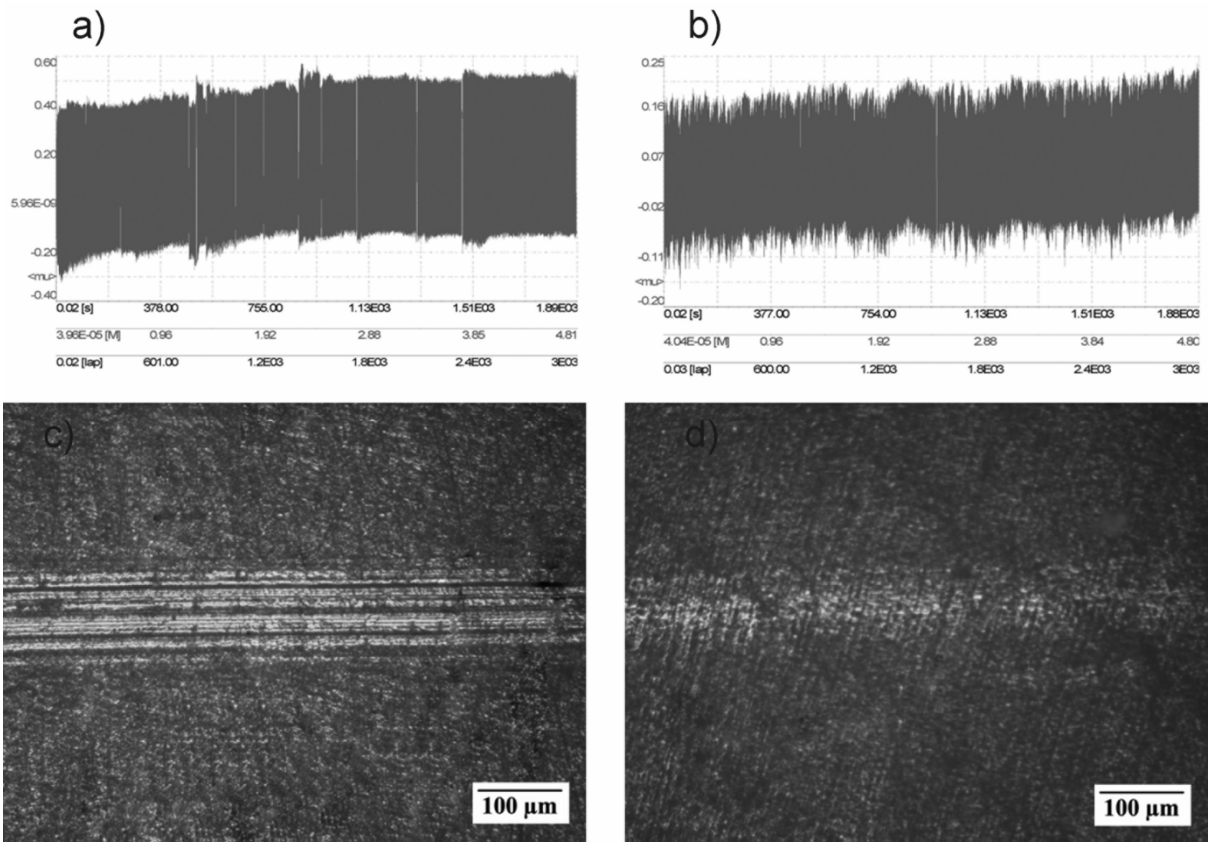

Figure 6. Diagrams of friction coefficient and optical micrograph showing wear track on UHMWPE sample: a), c) Ringers' solution with PMMA particles; v=4 mm/s; F_N=100mN; Penetration depth: 39.0µm; b), d) Distilled water; v=4 mm/s; b F_N=100mN; Penetration depth: 0.2µm;

Tests were conducted in dry conditions and with presence of distilled water and two different solutions: Ringer's solution and Ringer's solution containing particles of PMMA, in order to simulate human body environment in which implanted component function. It can be clearly seen (Figs. 5 and 6) that change of contact conditions (load, sliding speed, environment) produced significant change in tribological behaviour of UHMWPE samples. It can be clearly seen that the highest wear was produced for sliding with Ringer's solution containing particles of PMMA. Final penetration depth recorded by tribometer in that case was 39.0µm in comparison with other testing environments: 1.5µm, 2.5µm and 0.2µm respectively in cases of dry sliding and sliding with Ringer's solution and distilled water.

5. POLYMERS: POLYMETHYL METHACRYLATE (PMMA)

Bone cements have been used very successfully to anchor artificial joints (hip joints, knee joints, shoulder and elbow joints) for more than half a century. Artificial joints are anchored with bone cement. The bone cement fills the free space between the prosthesis and the bone and plays the important role of an elastic zone. This is necessary because the human hip is acted on by approximately 10-12 times the body weight and therefore the bone cement must absorb the forces acting on the hips to ensure that the artificial implant remains in place over the long term. Bone cement chemically is polymethyl methacrylate (PMMA). PMMA was used clinically for the first time in the 1940s in plastic surgery to close gaps in the skull. Comprehensive clinical tests of the compatibility bone cements with the body were conducted before their use in surgery. Also, until recently, the only polymer used to replace or augment bone itself (as opposed to the articulating surfaces) was polymethylmethacrylate (PMMA), or bone cement.

The ready bone cement is a compound consisting of 90 % of polymethylmetacrylate**,** (PMMA), the rest are mainly crystals of barium sulfate or Zirconium oxide that make the resulting product radio-opaque. The microscopic structure of bone cement is made by two substances glued together. One substance are the small particles of pre-polymerized PMMA so called "pearls". These pearls are supplied as a white powder. The other substance is a liquid monomer of MMA (MethylMetacrylate). Both substances are mixed together at the operation table with added catalyst that starts the polymerization of the monomer fluid. Polymer's properties can be predicted and explained by understanding the polymer structure

on the atomic, microscopic, and macroscopic scale. Amorphous polymers such as PMMA are brittle, hard plastics at room temperature [26]. It is isotropic and very bioinert material. Elastic modulus is up to 2.65GPa.

The primary functions of bone cement, when used to anchor artificial joints, are to secure the orthopedic implants to bone and transfer mechanical loads from the implant to the bone. The femoral stem and acetabular cups are cemented, screwed or press fit into place. Approximately 50% of all orthopedic implants utilize bone cement to achieve implant fixation [27]. PMMA is the most commonly used bone cement. Local tissue damage due to chemical reactions during polymerization, the high shrinkage of the cement after polymerization, the stiffness mismatch between bone and the cement are some drawbacks associated with PMMA-based bone cements [27]. Loose cement particles also mediate osteolysis of the bone and are highly unwanted to occur.

On the other hand, when PMMA is used to replace or augment bone itself, sliding contacts with different biomaterials become even more important to study, especially under micro-level forces. Many studies have been conducted focused on solving problems of PMMA application [28-33]. The effect of different environment on the tribological performance of PMMA against AISI 316LVM stainless steel at reciprocating sliding setup of the nanotribometer is shown in Fig. 7. These results are part of the investigation where five values of normal force were selected (100 mN, 250 mN, 500 mN, 750 mN, 1000 mN) and three values of sliding speed (4 mm/s, 8 mm/s, 12 mm/s).

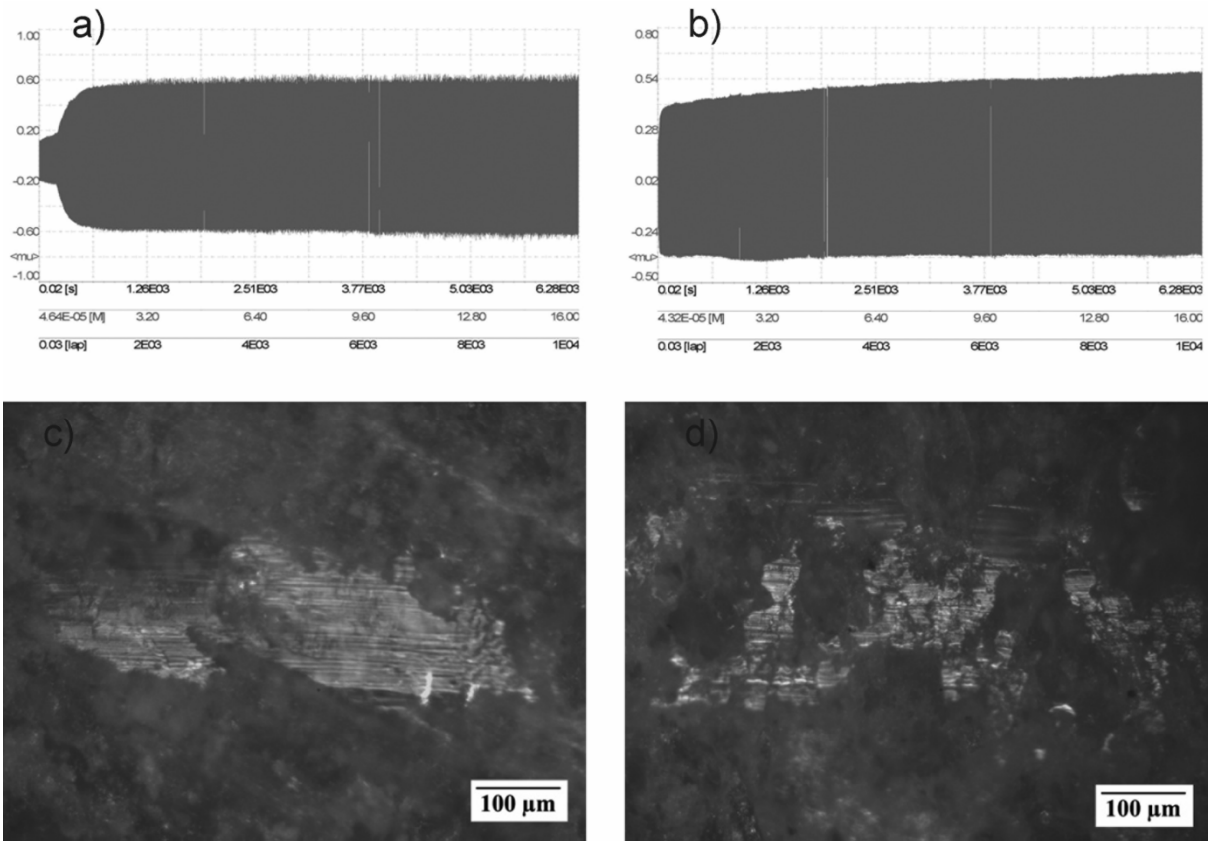

Figure 7. Diagrams of friction coefficient and optical micrograph showing wear track on PMMA sample: a), c) Dry; v=4 mm/s; F_N=500mN; Penetration depth: -0.5µm; b), d) Ringers' solution; v=4 mm/s; b F_N=500mN; Penetration depth: 0.8µm;

It can be clearly seen that the wear behaviour of a PMMA substrate sliding against AISI 316 steel ball is different for dry conditions if compared to sliding with Ringers' solution (Fig. 7). Many studies have been focused on understanding wear mechanisms in relation to polymer sliding contacts, also in case of microwear process. According to [29, 30] microwear process of PMMA involves three distinct stages. First, there is a plateaulike upheaval during which no wear takes place. This is followed by distortion of the surface to produce projections, again involving no wear. Finally, the surface is subjected to wear and wear particles are produced. A special feature of PMMA is that the deformation rapidly proceeds to destruction. This is in consistence with findings of our study. It can be clearly seen that plateaus can be defined throughout the worn track (Fig. 7d), what is more distinctively exhibited for sliding with Ringers' solution than for dry sliding. This phenomenon was also more

pronounced for higher normal load applied than for low loads. It is also proved by other authors [30] that wear and fatigue crack growth of PMMA sliding against steel, are very sensitive to organic fluids, related to the absorption of the liquid into the polymer, which, in turn, could result in surface plasticization or reduction of the minimum stress for the onset of cracking. Results of our study also showed that more cracks appeared in case of sliding with Ringers' solution. In case of dry sliding, larger regions with clear abrasive wear were produced. Ringers' solution produced decrease of wear in all cases of observed sliding speeds and normal loads.

6. CONCLUSION

Nowadays, technological innovation has progressed at such an accelerated pace that it is has permeated in almost any aspect of our lives. It is especially pronounced in a field of medicine, by integrating engineering that offer tools for advancement in health care (biomaterials, biosensors, image processing, etc.). Multidisciplinary area of biomaterials research comprises and blends number of research fields, aiming to achieve ultimate goal from prosthetics to regeneration. There are still many challenges to address, from manufacturing of biomaterials and their characterization, to design and selection of material for specific use in medicine and biomaterial interactions with human tissues. Important demand from all biomaterials in use today, is to prevent process of wear or at least to minimize it to the highest possible extent. Tribological investigations can offer valuable insight in material behaviour thus contributing to advances in this complex field of research.

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