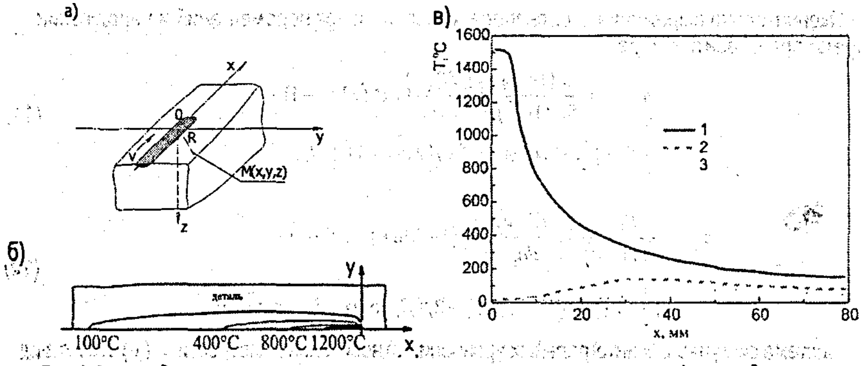

Рис.1. Распределение температуры в поверхностном слое металла; а) схема движения теплового источника, б) изотермы на поверхности образца, в) расчетные значения температуры по глубине образца. 1 - глубина 0 мм; 2 - глубина 1,0 мм; 3 - глубина 1,5 мм

Точность численного решения высока для внутренних точек и удовлетворительна для точек, примыкающих к границе области. Получены зависимости изменения теплового поля и напряжений как функции времени и координат.

Основными преимуществами применяемого метода по сравнению с другими существующими является необходимость дискретизации только границы области, при этом сохраняется высокая точность решения при уменьшении затрат машинного времени. Кроме того, учитываются условия закрепления закаливаемой детали и их влияние на процесс упрочнения.

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AN OVERVIEW OF THE SURFACE ENGINEERING METHODS USAGE FOR SURGICAL IMPLANTS PRODUCTION

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Introduction

Scientific description of different materials and technologies for medical purposes has been a subject of research and studies of different scientific fields, for instance material engineering, mechanics, technology or physics. Some groups of biomaterials were developed as result of many decades of research: polymer materials (UHMW PE, PEEK), ceramic materials (Al_2O_3, ZrO_2) and metallic alloys. They should be characterized by good mechanical properties, corrosion resistance, biocompatibility, and appropriate physical features. Major metals used in medical applications today include: commercially pure titanium and its alloys (largely the αβ alloys, Ti-6Al-4V and Ti-Al-Nb, and several β-Ti alloys), cobalt-based alloys (primarily Co-Cr-Mo line), stainless steel mainly type 316L) and Ni-Ti alloys [1-4]. Over the years, a variety of fabrication processes have been developed. At least two main methods of manufacture are used to make metallic implants: precise (lost wax) casting and hot forging. However, a number of researches focus on usage of powder metallurgy (P/M) techniques in this area [5-7]. Further development of new implant alloys is limited by the strong requirements put to biomaterials. Therefore, the surface engineering is a prospective solution of appearing demands [8-11].

Though many achievements have been made, many problems still need to be solved especially in the field of endoprostheses. Failure of total hip arthroplastics has appeared to be principally due to late aseptic loosening of acetabular components [1, 12]. Mismatch of Young' modulus of the biomaterials and the surrounding bone has been identified as one of reasons of implant loosening following stress shielding of bone [12]. However, the implanted material must be strong enough and durable to withstand the physiological loads placed upon it over the years.

The second cause has been a means of fixation. Widely used bone cement fixation allowed to restore the patient mobility in a relatively short period of time, whereas, it has been pointed as a cause of problems, because of failure due to postoperative infections, bone atrophy and implant loosening.

The third reason of failures has been shown the releasing of material from orthopedic implants into surrounding tissue by various processes, including corrosion, wear and mechanically accelerated processes such as stress corrosion, fatigue and fretting corrosion. Such metal ions and wear debris, concentrated at the implant-bone interface, might migrate through the tissue. Many publications have reported that the metal and polyethylene release had been associated with osteolysis, allergic reactions and clinical implant failure.

The present overview article aims at summarizing the most common methods for improving the implant osseointegration processes as well as tribological behaviors by usage of surface engineering methods. Some results of own research the tribological properties of biomaterials treated by surface method are also presented.

Corrosion resistance

The biocompatibility of titanium and CoCrMo alloys is closely related to its excellent corrosion resistance due to the presence of an extremely thin passive oxide film that spontaneously forms on the alloy surface. However, this protecting film may by damaged during surgical operation and hence cause corrosion. In the human body, implant devices are constantly exposed to living cells, tissues and biological fluids which produce a hostile environment for the survival of the implant. Additionally different corrosion processes appearing during implant utilization like stress, fatigue, and fretting corrosion increase metal release into surrounding tissue. Some of metal ions have been reported to induce allergic and inflammation reactions. One approach for preventing and reducing the potentially harmful metal ion release from orthopedic implant materials is to coat the surfaces of these materials. Various methods are used to form a barrier to corrosion on metallic surfaces [13,14]. The most popular are anodic oxidation, ion implantation, and thin coatings deposited by PVD or CVD methods . Many authors reported that TiN or CrN coatings on implant alloys are an effective barrier to metal release. Recently the DLC coating are developed as very promising layer due to its excellent biocompatibility and corrosion resistance [15,16]. A number of works pointed at the glow discharge-assisted technique, such as nitriding, oxynitriding or carbonitriding, as the most prospective method due to possibility of precise control the layer formation process with regard to the structure and the phase composition [17,18].

2. Osseointegration

The fixation of the artificial implant with acrylic bone cement, which is still one of the most popular means of fixation, unfortunately suffers from numerous problems [19,20]. Even though affixation through the use of bone cement proved somewhat successful in older patients, it was often a failure in more active and young patients. This is probably due to the brittle nature of the cement. It has been found that during polymerization, the toxic monomer can be released into the system. Additionally, the temperature during this process grows significantly, which is sufficient to cause heat necrosis of the surrounding tissues.

This has led to the rapid development of another solution of implant fixation. There have been used different techniques in creating mechanical fixation or chemical bonding [21,22]. The researches are conducted in two directions of morphological fixation of the implants to bone: to produce porous surface on the metallic implant stem or by using of bioactive ceramic.

The morphological fixation of the implants to bone via bony in-growth into the structure determines the dimensions of pores [23-25]. The commonly accepted range of pore size is reported to be between 150 and 400μm. This macroporous structure can be produced by different methods like precise casting (lost wax), plasma spraying metallic powder on an implant, and powder metallurgy method with traditionally sintering or electro-discharge compaction (EDC). However, there are still some serious problems concerning the possibility of microcracking and delamination of porous layers due to high thermal gradient, which caused reduction in fatigue strength.

Biochemical methods of surface modification utilize current understanding of the biology and biochemistry of cellular function and differentiation. Much has been learned about the mechanism of cells adhering to substrates and the role of growth factors in promoting differentiation of cells. The goal of biochemical surface is to mobilize biomolecules on biomaterials for the purpose of inducing specific cell and tissue responses.

Over the years, a variety of bio-active ceramics have provided an alternate means of implant fixation. Bio-active ceramics tend to form direct strong bonds with natural bones. However, it should be realized that the brittle nature of the ceramic materials and their poor bending strengths prevent the extensive use of these materials. Presently efforts are underway to coat the metallic parts of the implant that will be in contact with the natural bone with bio-active ceramics [26]. One of the most promising bioactive materials for clinical use is considered the hydroxyapatite (HA). Many studies have indicated that HA is biocompatible with hard tissues of human beings (its chemical composition is very similar to those of a component of bone) and exhibits osseoconductive properties. The mechanisms of hydroxyapatite behaviour during bone restoration processes are good known and widely reported in the literature [27,28]. HA coatings have been applied on various substrates by a wide range of surface deposition techniques such as plasma spraying, high velocity oxy-fuel (HVOF) spraying, ion beam sputerring, pulsed laser ablation, electrophoretic deposition, rf magnetron sputerring, sol-gel and conventional ceramic processes that involves pressing and sintering [29,30]. Among these surfacing processes, thermal spray techniques offer the attractive prospect of economy end efficient deposition of HA. Implant alloys coated with plasma sprayed HA have been widely used in clinical (Fig.1).

Nevertheless, the HA coatings often flake off as a result of poor ceramic/metal interface bonding, which may cause the surgery to fail [31].

New kinds of coatings are also developed, often including advanced technologies. Authors [32] researched coatings with layers of fibrillar type-I collagen prepared on titanium, cobalt and their alloys to improve initial osteoblast adhesion and implant-tissue integration. An increase in the surface density of reactive groups due to chemical and electrochemical pretreatment of the metal surfaces enabled the formation of a higher number of covalent linkages between the metal oxides and the collagen layers and thus significantly enhanced adhesion and proliferation of osteoblast-like cells.

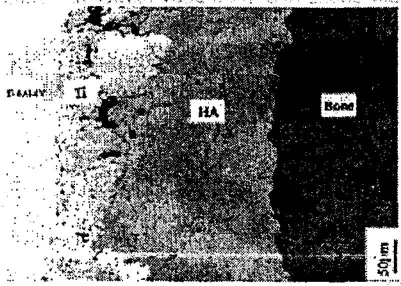
a 

Fig. 1. Crossection of HA coating on the titanium alloy implant

Another way is to make a microcomposite using metal fibers or particles to reinforce the hydroxyapatite. Authors [33-35] have pointed the excellent bioactivity of biocomposite fabricated from HA and titanium powders by powder metallurgy method.

Very prospective are also active bioglasses used as the coating as well as the composite layer on the metallic implant surface. A number of studies focused the beneficial influence of used bioglass on the corrosion resistance and the bone restoration process due to the ability of apatite formation in body fluid environment [36].

Tribological properties

Artificial implants are designed to restore function and relief pain but many have failed, especially in the long term, for a variety of reasons [38]. Failure of total hip arthroplastics has been shown to be principally due to late aseptic loosening of acetabular components [39]. Currently, the majority of hip prostheses are metal articulating with ultra-high-molecular-weight polyethylene (UHMWPE) and it is the general consensus that failure is due to UHMWPE particulate debris, which induces bone resorption [12,40,41]. In recent years, there has been resurgence in metal-on-metal articulating designs; motivated mainly by the fact that wear debris from PE acetabular components can lead to foreign body reactions, aseptic loosening and implant failure. Among metals that are commonly used as biomaterials, cobalt-chrom-molybdenium (Co-Cr-Mo) alloys are the preferred materials for self-bearing applications because of their excellent wear resistant [3,42]. The use of titanium alloys as bearing surfaces in total human replacements was widespread during the 1970s. However, following revision surgery, after several years use, concerns were raised regarding the creation of unwanted, titanium based debris.

Different solutions aimed at improving the tribological properties of the femoral and knee head-acetabular system are being developed. The most common used are the light ion implantation (C^+, O^+, N^+), various nitriding processes, thin wear resistance coatings like TiN, ZrO_2, DLC. Many authors emphasized the excellent influence of surface treatments on changes in the dominant wear mechanism and hence on reduction in the wear loss of investigated materials [9,43,44]. However, most of them concentrated research in laboratory conditions, very often a far cry from the nature in real joints. Compared to the volume of literature on improvement of implant materials wear resistance due to various surface engineering techniques, there are relatively few works on the long-term results. Some of them reported the deterioration of friction conditions and the danger of rapid abrasive wear. In vivo observations confirmed the resisting of third body wear mechanism, especially in a case of TiN coated titanium alloy femoral head [45].

The author own research showed ambiguous results. The ion implantation of Co-Cr-Mo alloy with oxygen and carbon showed an increase in the surface hardness, especially in a case of high-dose C^+ implantation – by near 100% [46]. The tribological tests carried out in the ball-on-disc system with alumina ball confirmed good protecting properties of hardened surfaces. However, pin-on-disc wear tests, when the contact area between sliding materials was larger (comparable to the one in artificial hip joint) did not caused the improvement of friction conditions resulting from ion implantation. The friction process was more unsteady in compare to unimplanted alloy. The resistances to moving lower at the beginning of process, rapidly changed as a consequence of forming hard wear debris and three-body abrasion. Hence the increase of wear loss in the case of implanted materials was observed.

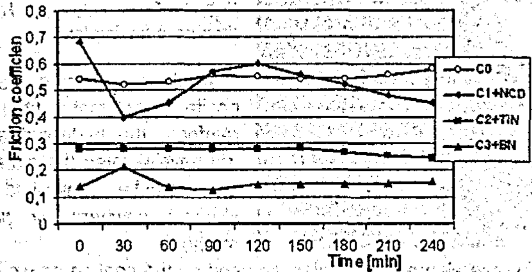


Fig.2. Results of comparative tribological tests of CoCrMo alloy (Vitalium) with various hard coatings

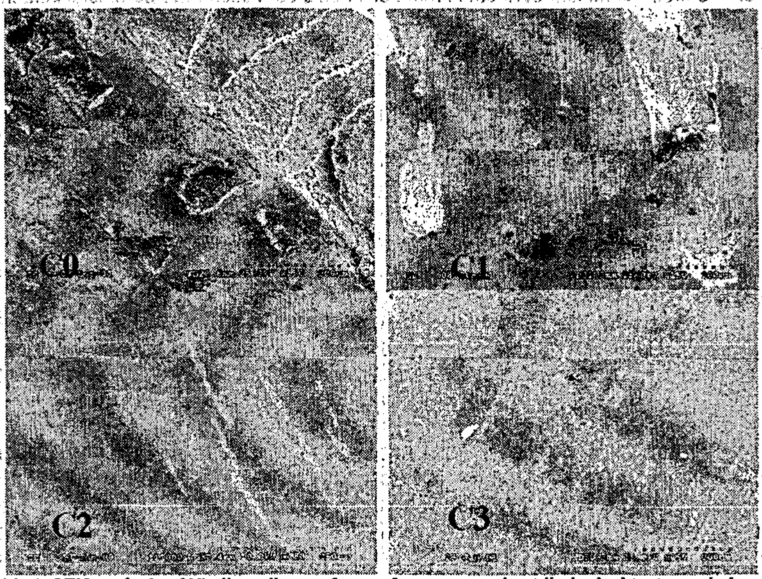


Fig.3. SEM analysis of Vitalium disc surfaces after comparative tribological tests carried out on symulator of hip joint friction: (C0) not treated alloy; (C1) with NCD coating; (C2) with TiN coating; (C3) with BN coating

In a case of thin ceramic layers on CoCrMo alloy the improvement of tribological properties was generally observed (fig.2 and fig.3) [47,48]. Nonetheless, there might appear the danger of local coating delamination. Debris, staying in a friction zone, might cause the abrasive wear (what is visible on the worn surface with NCD coating – fig.3C1). The thin layers deposited on titanium implant alloys showed the protecting effect during tribological tests only in a case of low range of load (till contact pressure of p_{max}=2MPa). Increasing of load entailed the rapid damage of hard coating (due to strong disparities of ceramic layer and metallic ground mechanical properties) and intensified abrasive wear [49,50].

Conclusion

Surface engineering is considered a very prospective method of improving functional properties of implant alloys. There are some problems regarding materials in biomedical applications, concerning mainly with elongation of joint prostheses life time.

A large number of research showed excellent affect of ion implantation, oxidation and nitriding processes as well as thin ceramic coatings, like TiN, ZrO_2, DLC, on corrosion resistance and biocompatibility of biomaterials. For intensification of osseointegration the hydroxyapatite and bioglass coatings are the most promising solution.

Many works are focused on improwing tribological properties of implant alloys by ion implantation, nitriding or by using thin wear resistant coatings. In fact, hardening of sliding materials surface layers changed wear mechanism and decreased wear loss, especially in conditions of low friction velocity and high load. However, some long-term in-vivo observations as well as the author's own investigations showed the danger of local coating delaminating and hence intensification of abrasive wear, entailing necessity of reimplantation.

Acknowledgements

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