

Review of materials in medical applications

Pregled materialov v medicinskih aplikacijah

DAVID BOMBAČ¹, MIHA BROJAN², PETER FAJFAR¹, FRANC KOSEL², RADO TURK¹

¹University of Ljubljana, Faculty of Natural Sciences and Engineering, Aškerčeva cesta 12, SI-1000 Ljubljana, Slovenia; E-mail: david.bombac@ntf.uni-lj.si, peter.fajfar@ntf.uni-lj.si, rado.turk@ntf.uni-lj.si

²University of Ljubljana, Faculty of Mechanical Engineering, Aškerčeva cesta 6, SI-1000 Ljubljana, Slovenia; E-mail: miha.brojan@fs.uni-lj.si, franc.kosel@fs.uni-lj.si

Received: November 22, 2007

Accepted: December 14, 2007

Abstract: Medical implants are products that have to satisfy functionality demands defined by the human body as working environment. Ideally, they should have biomechanical properties comparable to those of autogenous tissues without any adverse effects and are regulated in order to ensure safety and effectiveness. The choice of material used for designing a medical implant is governed by biocompatibility, bioadhesion, biofunctionality, corrosion resistance, etc. In these review metallic biomaterials are divided into four subgroups: stainless steels, the cobalt alloys, titanium alloys, and other metals. Some attention is also given to shape memory, polymeric, and ceramic biomaterials for their innovative use in practical medical applications.

Izvleček: Medicinski vsadki so izdelki, ki morajo zadovoljiti stroge funkcijske zahteve, ki jih narekuje človeško telo kot delovno okolje. V idealnem primeru naj bi imeli biomehanske lastnosti primerljive avtogenim tkivom brez škodljivih efektov in so evidentirani, da zagotavljajo varnost in uspešnost. Izbira materiala pri načrtovanju medicinskih vsadkov je pogojena z biokompatibilnostjo, bioadhezijo, biofunkcionalnostjo, korozijsko odpornostjo, itd. V tem pregledu so kovinski biomateriali razdeljeni v štiri podskupine: nerjavna jekla, kobaltove zlitine, titanove zlitine in druge kovine. Nekaj pozornosti je posvečene tudi zlitinam s spominom, polimernim in keramičnim biomaterialom zaradi njihove inovativne uporabe v praktičnih medicinskih aplikacijah.

Key words: medical implants, biomaterials, AISI 316L, titanium alloys, shape memory alloys, polymeric biomaterials, ceramic biomaterials

Ključne besede: medicinski vsadki, biomateriali, AISI 316L, titanove litine, materiali z oblikovnim spominom, polimerni biomateriali, keramični biomateriali

INTRODUCTION

The development of medical implants utilizing new materials continues to attract considerable academic and commercial interest. The development of new biomaterials involves a complicated mix of materials science and cell biology. Collaboration of various experienced specialists such as material scientists, metallurgists, traumatologists, orthopedists, chemists, mechanical engineers, pharmacists and others in order to achieve better results in research, development and implementation of the extracted knowledge into the practice is of essential importance.

Biomaterials are nonviable materials used in a medical devices intended to interact with biological systems^[1] and cover several classes of materials, such as metallic, ceramic, and polymeric materials. Medical implants are products that have to satisfy functionality demands defined by the human body as working environment. Ideally, they should have biomechanical properties comparable to those of autogenous tissues without any adverse effects and are regulated in order to ensure safety and effectiveness. The choice of material used for designing a medical implant is governed by biocompatibility, bioadhesion, biofunc-

tionality, corrosion resistance, etc. To better understand implant material-biological organism interaction most of the studies are directed into the releases of particles from the material and offer screens for genotoxicity, carcinogenicity, cytotoxicity, irritation, sensitivity and sterilization agent residues^[2]. Focus of this contribution is on metallic, shape memory, polymeric and ceramic biomaterials used in modern medical applications.

METALLIC BIOMATERIALS

Metallic biomaterials are often used to support and/or replace components of the skeleton. They are used e.g. as artificial joints, bone plates, screws, intramedullary nails, spinal fixations, spinal spacers, external fixators, pace maker cases, artificial heart valves, wires, stents, and dental implants. They possess greater tensile strength, fatigue strength, and fracture toughness when compared to polymeric and ceramic materials. Most widely used metallic biomaterials for implants devices are 316L stainless steels, cobalt alloys, commercially pure titanium, and Ti-6Al-4V alloys^[3-6]. Originally, these materials were developed for industrial purposes. Their excellent mechanical properties and relatively high

Table 1. Mechanical characteristics of metal alloys used in medicine*

Tabela 1. Mehanske lastnosti kovinskih zlitin uporabljenih v medicini*

Characteristics	Stainless steel	Cobalt alloys	Titanium alloys
Stiffness	High	Medium	Low
Strength	Medium	Medium	High
Corrosion Resistance	Low	Medium	High
Biocompatibility	Low	Medium	High

* the scale is relative

corrosion resistance, which results in very small release of harmful toxins when exposed to bodily fluids, are the main reasons for these materials can be left inside the body for a longer period of time and are therefore appropriate for medical uses. In Table 1 some mechanical and biological characteristics of stainless steel, cobalt and titanium alloys are presented.

As additional information let us mention that production of metallic-based medical devices in general involves cutting operations (turning, milling, drilling, etc.); forming operations (pressing, hydroforming, forging, etc.) and other alternative machining operations (laser and waterjet cutting, different layer-by-layer sintering techniques such as direct metal laser sintering, selective laser melting, selective laser sintering, electron beam melting, and laser engineered net shaping).

In these review metallic biomaterials will be divided into four subgroups: stainless steels, the cobalt alloys, titanium alloys, and other metals.

Stainless steels

Stainless steel used for medical implants is mainly austenitic type 316L due to its resistance to corrosion, together with a wide range of other physical and mechanical properties coupled with inert, easily-to-clean surfaces. The chemical composition of type 316L stainless steel was developed to obtain stable austenitic structure which has numerous advantages, namely:

- Austenitic stainless steel has a face-centered cubic structure and is characterized by very low yield strength-to-tensile strength ratio and high form-

ability.

- To increase strength, cold working and successive strain aging treatment can be applied.
- Austenitic stainless steel is superior to ferritic stainless steel in corrosion resistance because the crystallographic atomic density of the former is higher than that of the latter.
- Austenitic stainless steel is essentially nonmagnetic.

The disadvantages of austenitic stainless steels generally are higher sensitivity toward pitting corrosion and stress corrosion cracking^[3]. Pitting corrosion causes deep pits on the metal surface. It is initiated when an oxidant such as dissolved oxygen reacts with chloride ions. Pitting is further accelerated by the existence of an oxygen concentration cell at the early growth stage. The chemical composition of type 316L (ASTM F138, F139) austenitic stainless steel where "L" denotes low carbon content is as follows: $\leq 0.030\%$ C, $\leq 1.0\%$ Si, $\leq 2.0\%$ Mn, $\leq 0.045\%$ P, $\leq 0.030\%$ S, 12.0-15.0 % Ni, 16.0-18.0 % Cr, and 2.0-3.0 % Mn^[1]. Its corrosion resistance is improved by adding molybdenum, increasing nickel and reducing carbon to less than 0.030 %. This steel has less than 0.03 wt.% carbon in order to reduce the possibility of in vivo corrosion. If the carbon content of the steel significantly exceeds 0.03 %, there is increased danger of formation of carbides such as Cr_{23}C_6 . These tend to precipitate at grain boundaries when the carbon concentration and thermal history have been favorable to the kinetics of carbide growth. In turn, this carbide precipitation depletes the adjacent grain boundary regions of chromium, which has the effect of dimin-

ishing formation of the protective chromium-based oxide Cr_2O_3 ^[7]. The presence of molybdenum as an alloying element in stainless steel reduces both the number and the size of nucleations and metastable pits. This is because bonds in the oxide film are strengthened and active sites caused by the formation of molybdates or of molybdenum oxyhydroxides are eliminated^[8].

Due to high content of chromium, 316L stainless steel forms a protective, adherent and coherent oxide film that envelops the entire outer surface. This oxide film, often called passive layer, is intentionally formed when device is manufactured as chromium in the surface layer reacts with oxygen creating Cr_2O_3 . The passive film serve as a barrier to corrosion processes in alloy systems that would otherwise experience very high corrosion rates^[9] and has ability of self-healing, when damaged, as chromium in the steel reacts with oxygen and moisture in the environment to reform the protective oxide layer^[10].

Cobalt alloys

Cobalt alloys may be generally described as nonmagnetic, corrosion and heat resistant where exhibit high strength even at elevated temperature and are also resistant to wear^[11]. Many of its properties originate from the crystallographic nature of cobalt, and formation of extremely hard carbides and the corrosion resistance imparted by chromium. Cobalt alloys are difficult to fabricate which is why their use has been limited, but continuous work led to the development of specialized casting methods^[12] and recently to selective laser sintering. Due to cobalt alloy excellent resistance to degradation in the oral envi-

ronment, the first medical use was in the cast of dental implants. Various in-vitro and in-vivo tests have shown that the alloys are biocompatible and suitable for use as surgical implants. Today the use of Co alloys for surgical applications is mainly related to orthopaedic prostheses for the knee, shoulder and hip as well as to fracture fixation devices. Joint endoprostheses are typical long-term implants, and the applied implant material must therefore meet extremely high requirements with regard to biocompatibility with the surrounding body tissue material and corrosion resistance to bodily fluids^[13].

Nowdays the CoCrMo cast and wrought versions of alloy are highly biocompatible materials^[14,15] and are widely used as orthopedic implant materials in clinical practice such as hip joint and knee replacement. The biocompatibility of CoCrMo alloy 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. Similar to AISI 316L stainless steel predominant oxide film is Cr_2O_3 with some minor contribution from Co and Mo oxides^[16,17].

In spite of the alloys excellent corrosion resistance, there is still some concern about metal ion release from orthopedic implants into the human body environment. Implant components fabricated from Co-Cr based alloys have been reported to produce elevated Co, Cr and Ni concentrations in the surrounding tissue^[18,19].

Other Co alloys used in medicine are MP35N or CoNiCrMo (ASTM F 562) with a nickel content of 35 % used for car-

diovascular pacing leads, stylets, catheters and orthopaedic cables. Increased content of nickel exhibits an improved resistance to stress-corrosion cracking in aqueous solution, but also increase the possibility of nickel allergy reactions. Therefore these alloys are not ideal for orthopaedic applications. The biocompatibility of the wear particles produced can be troublesome because of the increased surface area of these small particles which are in direct contact with the surrounding medium or tissue material. In work-hardened or work-hardened and aged conditions, this alloy has very high tensile properties which are among the strongest available for implant applications. Other Co-based alloy is L-605 cobalt alloy or CoCrWNi (ASTM F 90) which is used for heart valves and in an annealed condition (ASTM F 1091) for surgical fixation wires. Its mechanical properties are approximately the same as those of Co-CrMo alloys, but after the material is cold worked the mechanical properties more than double.

Titanium and its alloys

Titanium and its alloy grades started gaining widespread usage as implant materials in the early 1970s. Material specifications and forms of them are detailed in a number of specifications, including ASTM and BS7252/ISO 5832 standards. The majority of Ti-based implants are made of commercially pure titanium (CP-Ti), mainly in Europe, and titanium alloy grade 5 Ti-6Al-4V (ASTM F 1472), mainly in North America. Although CP-Ti offer better corrosion resistance and tissue tolerance than stainless steel, its comparatively lower strength and unfavorable wear properties restricted its usage to certain applications such as pace-

maker cases, heart valve cages and reconstruction devices. Selection of titanium alloys for implementation is determined by a level and combination of the most desirable characteristics including immunity to corrosion, biocompatibility, shear strength, density and capability of osseointegration^[20]. The cold worked CP-Ti has been used for dental implant and maxillofacial applications. Interest in the Ti-6Al-4V alloy for total joint prostheses grew significantly in the United States toward the late 1970s because of its high strength, low elastic modulus, excellent corrosion resistance, and good tissue tolerance. Currently, its applications include hip and knee prostheses, trauma fixation devices (nails, plates, screws, and wires), instruments, and dental implants. Due to its relatively poor wear resistance, it was soon realized that the Ti-6Al-4V alloy was not suitable for bearing surface applications as are hip heads and femoral knees, without a coating or surface treatment^[21]. Application from titanium alloys for bone trauma fixation applications as spinal fusion devices, pins, bone-plates, screws and intramedullary nails are shown in Figure 1.

Number of medical devices being made from titanium alloys which are generally preferred comparing to stainless steel and Co-alloys and is increasing, because of their lower modulus of elasticity (it is closer to that of bone), superior biocompatibility and corrosion resistance^[22,23]. Recently, new titanium alloy compositions, specifically tailored for biomedical applications, have been developed. These first generation of orthopaedic alloys included Ti-6Al-7Nb^[24] and Ti-5Al-2.5Fe^[25,26], two alloys with properties similar to Ti-6Al-4V that were developed in response to concerns



Figure 1. Examples of implants made of metallic biomaterials
Slika 1. Primeri implantatov narejenih iz kovinskih biomaterialov

relating V to potential cytotoxicity^[27,28] and adverse reaction with body tissues^[29]. Further, biocompatibility enhancement and lower modulus has been achieved through the introduction of second generation titanium orthopaedic alloys including Ti-12-Mo-6Zr-2Fe^[30,31], Ti-15Mo-5Zr-3Al^[32], Ti-15Zr-4Nb-2Ta-0.2Pd and Ti-15Sn-4Nb-2Ta-0.2Pd alloys^[33] as well as the almost completely biocompatible Ti-13Nb-13Zr alloy^[34,35]. Titanium and its alloys have the inherent property to osseointegrate, enabling use in implants that can remain in place for longer period of time. Since titanium is non-magnetic, patients with titanium implants can be safely examined with magnetic resonance imaging which is convenient for long-term implants. Preparing titanium for implantation in the body involves subjecting it to a high-temperature plasma arc which removes the surface atoms, exposing fresh titanium that is instantly oxidized, whereas may be anodized to produce various colors^[36]. Titanium is also used for the surgical instruments used in image-guided surgery.

To stimulate osseointegration, limit resorption and thus increase the implant lifetime, some designs (cementless prostheses) use roughened bioactive coated surfaces. Hydroxyapatite (HA) coatings are used as bioactive surface of titanium implants, of which many surface treatment techniques have been developed^[37]. Those techniques are plasma spraying, electrophoretic deposition of HA and micro-arc oxidation. Another form of implant coating is diamond-like carbon (DLC) coatings. DLC coatings can address the main biomechanical problems with the implants currently used, e.g. friction, corrosion and biocompatibility^[38]. Metallic surfaces in contact with bodily fluids corrode as their surface dissolves and the dissolved metals enter the blood stream. The frequency of skin sensitivity to metals in patients with artificial joints is substantially higher than that in the general population^[39]. At present, the risk to patients to develop such skin reaction after implantation of artificial joints may be considered minimal, cf. Table 2.

Table 2. Sensitivity to metal
Tabela 2. Občutljivost na kovine

	Percent Metal Sensitive
General population	10 %
Patients with stable total joints	25 %
Patients with loose total joints	60 %

Other metals

Conventional metal materials with use of numerous surface coatings and porous designs have been developed to enhance biological fixation of medical implants to bone for use in orthopedic procedures^[40]. Other metals used for medical purposes include tantalum, gold, dental amalgams and other special metals. Although excellent clinical results of classical metal materials have been shown, they have several inherent limitations i.e. low volumetric porosity, relatively high modulus of elasticity and low frictional characteristics. To address the limitations of these solid metals, a new porous tantalum biomaterial has been developed. Porous tantalum is an open-cell tantalum structure of repeating dodecahedrons with an appearance similar to cancellous bone commercially known as trabecular metal. Tantalum is a transition metal that remains relatively inert in in-vivo. In the past multiple medical devices have been fabricated that utilize this material, including: pacemaker electrodes, foil and mesh for nerve repair, radiopaque markers, and cranioplasty plates^[41,42]. Tantalum-based implants have displayed an exceptional biocompatibility and safety record in orthopedic, cranio-facial, and dentistry literature^[43]. The basic structure of this po-

rous tantalum metal yields a high volumetric porosity, a low modulus of elasticity, and relatively high frictional characteristics. Porous tantalum implants are fabricated from the pyrolysis of a thermosetting polymer foam which in turn creates a low-density vitreous carbon skeleton with 98 % porosity and a repeating dodecahedron array of regular pores. Commercially pure tantalum is then deposited onto this interconnected vitreous carbon scaffold using chemical vapor deposition^[40,44,45].

Figure 2^[46], demonstrates the distinct microtexture and overall geometry of this nanostructured porous metal construct and some examples of medical implants.

Oxidized zirconium is a metal with a ceramic surface that was introduced commercially for knee arthroplasty components^[47,48]. Commercial medical alloy under trade name Oxinium consists of a zirconium and niobium alloy (Zr-2.5Nb) that has been oxidized to transform the surface into zirconia ceramic. The ceramic is not a coating but a transition of metal to ceramic approximately 4-5 μm thick that is produced by reaction to oxygen during a thermal treatment. The ceramic surface is a monoclinic zirconia ceramic^[49,50].

Because the bulk of the material is metal, it does not have the same risk of fracture as a monolithic ceramic head. Thus, the oxidized zirconium implant offers the potential to decrease wear and thereby increase the life span of implants. First oxidized zirconium implants were introduced in 1996 as simple implants. For total knee replacements it has not been used until 2001, and was later used also for hip replacements since 2003.



Figure 2. Microstructure of the porous nanostructured tantalum material and examples of medical devices

Slika 2. Mikrostruktura nanostrukturiranega poroznega tantalata in primeri medicinskih izdelkov

In Figure 3 ^[51], knee and hip replacement component from oxidized zirconium are shown.

SHAPE MEMORY MATERIALS

Shape memory alloys (SMA) have been given a lot of attention mainly for their innovative use in practical applications using shape memory effect (SME). The SME is unique property that some alloys possess according to which, after being deformed at one temperature, they re-

cover their original shape by increasing to the alloys second temperature. This effect arises from reversible and usually rate-independent martensitic transformation and resulting changes of crystal structure of the solid phases of the material. The low temperature phase is called martensite phase and the high temperature austenite phase. Large residual strains of even more than 10 % can be recovered in this way and the process is often referred to as free recovery. The return to the original shape starts at a temperature called austenite start transformation temperature A_s , and com-

plates at the austenite finish transformation temperature A_F . Pseudoelasticity or superelastic effect and shape memory effect are two separate mechanical effects which characterize the response of SMA. At constant high temperature, above temperature A_F , a mechanical loading-unloading cycle induces highly-nonlinear large deformations. At the end of the loading-unloading cycle no permanent deformations are present. The cycle usually presents a hysteresis loop, Figure 4a.

In the early 1960s, researchers at the U.S. Naval Ordnance Laboratory discovered the shape memory effect in an equiatomic alloy of nickel and titanium. Discovered alloy was patented and named Nitinol (Nickel-Titanium Naval Ordnance Laboratory). This discovery is considered a breakthrough in the field of shape memory materials and since that time, intensive investigations have been made to elucidate the mechanics of its basic behavior. The use of NiTi as a biomaterial is fascinating because of its superelasticity and shape

memory effect, which are completely new properties compared to the conventional metal alloys. Composition and metallurgical treatments have dramatic impacts on the described transition temperatures. From the applications point of view, NiTi can have three different forms: martensite, stress induced martensite (superelastic), and austenite. In martensite form alloy is soft and ductile and can be easily deformed. Superelastic NiTi is highly elastic, whereas austenitic NiTi is quite strong and hard. The NiTi alloy can exhibit all these properties depending on the temperature in which it is used.

In vast majority of medical applications is utilized superelastic effect. Result of slightly nickel richer alloys is superelasticity which could be exhibited in a narrow temperature range of the human body. The Simon Inferior Vena Cava filter was the first SMA cardiovascular device. It is used for blood vessel interruption for preventing pulmonary embolism^[52]. The Simon filter is filtering clots that travel inside blood-

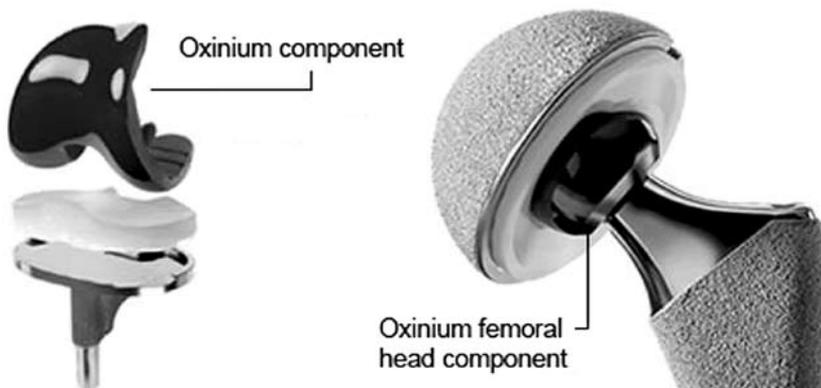


Figure 3. Microstructure of the porous nanostructured tantalum material and examples of medical devices

Slika 3. Mikrostruktura nanostrukturiranega poroznega tantala in primeri medicinskih izdelkov



Figure 4. The stress-strain diagram, SE stents, Nitinol clamps, orthodontic wire

Slika 4. Diagram napetost-raztezek, SE stent, Nitinol spona, ortodonska žica

stream. The device is made of SMA wire curved similarly to an umbrella which traps the clots which are better dissolved in time by the bloodstream. For insertion, the device is exploiting the shape memory effect, i.e. the original form in martensitic state is deformed and mounted into a catheter. When the device is released, body heat causes the filter to return to its original shape.

Stents are most rapidly growing cardiovascular products which are used to maintain the inner diameter of a blood vessel. The product has been developed in response to limitations of balloon angioplasty, which resulted in repeated blockages of the vessel in the same area. NiTi alloys have become the material of choice for superelastic self-expanding (SE) stents^[53], Figure 4b.

The self-expanding nitinol stents are produced in open state and later compressed and inserted into the catheter. The basic open form is obtained mainly by SMA tubing, the final shape is then obtained by alternative machining operations such as laser cutting. Stents can also be produced from wire and laser welded or coiled

striped etched sheet. During the operation procedure, when the catheter is in correct position in the vessel, the self-expanding stent is pushed out and then it expands against the vessel wall due to a rise in temperature.

Superelastic SMA wires have found wide use as orthodontic wires as well, Figure 4c. Nitinol wires have large elastic deformation combined with a low plateau stress. This results a smaller number of visits to the orthodontist due to the larger elastic stroke and more comfort due to lower stress levels. In dental medicine, special plates for fixing a loose tooth have become available, Figure 4c. They are produced with laser cutting from sheet metal.

Orthopedic implants far exceed any other by weight or volume. They are used as fracture fixation devices, which may or may not be removed and as joint replacement devices. As shown in Figure 4a, bone and nitinol have similar stress-strain characteristics, which make nitinol a perfect material for production of bone fixation plates, nails and other trauma implants^[54]. In the past, acceptability of nitinol as material for

permanent bone implants was conditioned by releasing Ni^{3+} ions from NiTi and integrity of contact between the bone and the implant. To solve these two vital problems, coating the bioactive layer on the device surface has been introduced^[55].

Medical equipment is also a branch where nitinol has found its place. Because modern surgery is aiming less invasive operations, smaller diameters of tubing are very important. Reducing the diameters of medical devices was possible compared to polymer materials due to superelasticity when tubing of NiTi alloys became available.

POLYMERIC BIOMATERIALS

Polymeric biomaterials are used as a substitute for metal alloys in trauma and orthopedic implant devices or as an aid at surgical procedures. First polymeric material, used in medicine since the 1960s, was ultra-high molecular weight polyethylene (UHMWPE). It is a thermoplastic with extremely long chains and molecular weight numbering between 2 and 6 million. The longer chain serves to transfer load more effectively to the polymer backbone by strengthening intermolecular interactions. Results of this are a very tough material, with the highest impact strength of any thermoplastic presently made. UHMWPE is highly resistant to corrosive chemicals, with exception of oxidizing acids and has extremely low moisture absorption, very low coefficient of friction, characteristic of self lubrication and high resistance to abrasion. The mechanical and tribological properties of UHMWPE favor its use as a bearing material in many joint replace-

ment devices. UHMWPE is used in buttons to resurface the patella in total knee arthroplasty, in sleeves to permit semi-constrained rotation in elbow and wrist arthroplasty designs, and in counterfaces inserted into the glenoid in shoulder arthroplasty. However, for the purposes of this review we will focus on the most common uses of UHMWPE in medical devices, those of tibial bearings in knee arthroplasties and of acetabular bearings in hip arthroplasties^[56]. UHMWPE was first used clinically in 1962 and emerged as the dominant bearing material for total hip and knee replacements in the 1970s. Since the 1980s UHMWPE is successfully used for spine implants^[57]. Thus, even though UHMWPE components are typically in no imminent danger of wearing through during a patient's lifetime, the generation of particulate debris from the articulating surface has been associated with osteolysis and loosening of implants^[58-61]. To address these problems a highly crosslinked UHMWPE materials were clinically introduced in 1998 and has rapidly become the standard of care for total hip replacements^[62-68]. Another important medical advancement for UHMWPE in the past decade has been the increase in use of fibers for sutures, where maximum strength and minimum weight are required. It is ideal for orthopedic implants, for example, as it allows smaller implants to be used, and is flexible and resistant to abrasion. Similarly, its strength can be used for surgical instruments for minimally invasive procedures^[69].

Polymethyl methacrylate (PMMA) is the synthetic polymer of methyl methacrylate and is in field of medical technologies and implants used because of its good de-

gree of compatibility with human tissue. In orthopaedics, PMMA bone cement is used to affix implants and to remodel lost bone. It is supplied as a powder with liquid methyl methacrylate (MMA). When mixed these yield dough like cement that gradually hardens. Surgeons can judge the curing of the PMMA bone cement by pressing their thumb on it. Although PMMA is biologically compatible, MMA is considered to be an irritant and a possible carcinogen and therefore PMMA has also been linked to cardiopulmonary events in the operating room due to hypotension^[70-75]. Bone cement acts like a grout and not so much like a glue in arthroplasty. Although sticky, it does not bond to either the bone or the implant, it primarily fills the spaces between the prosthesis and the bone preventing motion. A big disadvantage to this bone cement is that it heats to quite a high temperature while setting and because of this it kills the bone in the surrounding area. It has a Young modulus between the one of cancellous bone and the one of cortical bone, thus it is a load sharing entity in the body not causing bone resorption^[76]. Dentures are often made of PMMA, and can be color matched to the patient's teeth and gum tissue. In cosmetic surgery, tiny PMMA microspheres suspended in some biological fluid are injected under the skin to reduce wrinkles or scars permanently. PMMA also is used for replacement intraocular lenses in the eye when the original lens is removed in the treatment of cataracts. Hard contact lenses are frequently made of this material. Soft contact lenses are often made of a related polymer, where acrylate monomers containing one or more hydroxyl groups make them hydrophilic^[77-80].

Another polymeric material used as biomaterials for trauma, orthopedic, and spinal implants is polyetheretherketones (PEEK). It is thermoplastic polymer obtained from aromatic dihalides and bisphenolate salts by nucleophilic substitution. The bisphenolate salt is formed in situ from bisphenol and either added sodium or added alkali metal carbonate or hydroxide and have extraordinary mechanical properties. PEEK is partially crystalline and has unusual property in exhibiting two glass transition temperatures, at approximately 140 °C and 275 °C, depending on cure cycle and precise formulation. It is also highly resistant to thermal degradation^[81]. By the late 1990s, PEEK had emerged as the leading high-performance thermoplastic candidate for replacing metal implant components, especially in orthopedics^[82,83] and trauma^[84,85]. Numerous studies documenting the successful clinical performance of PEEK polymers in orthopedic and spine patients continue to emerge in the literature^[86-91]. Recent research has also investigated the biotribology of PEEK composites as bearing materials and flexible implants used for joint arthroplasty^[92-95]. Due to interest in further improving implant fixation, PEEK biomaterials research has also focused on compatibility of the polymer with bioactive materials, including hydroxyapatite (HA), either as a composite filler, or as a surface coating^[96-100]. As a result of ongoing biomaterials research, PEEK and related composites can be engineered today with a wide range of physical, mechanical, and surface properties, depending upon their implant application. Mechanical properties of PEEK implants can be tailored by preparing carbon-fiber-reinforced (CFR)

Table 3. Typical average physical properties of UHMWPE, PMMA and PEEK
Tabela 3. Značilne povprečne mehanske lastnosti UHMWPE, PMMA in PEEK

	UHMWPE	PMMA	PEEK	CRF-PEEK
Polymer type	Semi-crystalline	Amorphous	Semi-crystalline	Semi-crystalline
Degree of crystallinity [%]	39 – 75	Noncrystalline	30 – 35	30 – 35
Molecular weight [10 ⁶ g/mol]	2 – 6	0.1 -0.8	0.08 – 0.12	0.08 – 0.12
Poisson ratio	0.46	0.35	0.36	0.38 – 0.40
Specific gravity	0.932 – 0.945	1.180 – 1.246	1.3	1.4 – 1.6
Flexural modulus [GPa]	0.8 – 1.6	1.5 – 4.1	4	20 – 135
Tensile strength [MPa]	39 – 48	24 – 49	93	170 – >2000
Tensile elongation [%]	350 – 525	1 – 2	30 – 40	1 – 2

composites with varying fiber length and orientation. Comparison of physical properties of UHMWPE, PMMA, PEEK and CFR-PEEK is presented in Table 3 [81].

Property of polymeric materials reviewed earlier (UHMWPE, PMMA and PEEK) is that they are not bioabsorbable and biodegradable. Due to high cost of secondary operations, where implants are extracted scientists started research and development of medical devices which are made from bioabsorbable and biodegradable materials. These types of materials have ability to gradually decline in the strength of implant during the healing process where the area of fracture is filled with bone and connective tissue. The rate of dissolving can be engineered in order to be consistent with the rate of new bone growth and under ideal conditions, a bioabsorbable polymer could encourage bone healing while the body slowly metabolizes it [101]. The main advantage of biodegradable implants is elimination of need for a second surgery whereas the main disadvantage is a ten-

dency to release harmful acids and other toxins during the dissolving process. The physical properties required for reliable bioabsorbable and biodegradable implants such as appropriate initial strength, initial modulus of elasticity, controlled strength, etc. could be engineered carefully [102]. In general bioabsorbable polymers used in medicine are thermoplastics, linear-chain, partially crystalline or totally amorphous with a definitive melting temperature and/or a glass transition region. The most used medical polymers are polyglycolide acid (PGA), polylactide acid (PLA) and poly(ϵ -caprolactone) (PCL). Polymers prepared from glycolic acid and lactic acid have found a multitude of uses in the medical industry, beginning with biodegradable sutures first approved in the 1960s [103]. Since that time other medical devices, based on lactic and glycolic acid, as well as other materials, including poly(dioxanone), poly(trimethylene carbonate) copolymers, and poly(ϵ -caprolactone) homopolymers and copolymers, have been accepted for use as medical devices [104].

In addition to these approved devices, a great deal of research continues on polyanhydrides^[105] polyorthoesters^[106] and other materials^[107,108].

Fibers from PGA exhibit high strength and modulus and are too stiff to be used as sutures except as braided material. Sutures of PGA lose about 50 % of their strength after two weeks and 100 % at four weeks and are completely absorbed in 4-6 months^[101]. PGA is immunological inert, but it leads to slight non-specific lymphocyte activation, as it induces inflammatory mononuclear cell migration^[109].

Lactide is the cyclic dimer of lactic acid, which exists as two optical isomers, D and L. L-lactide, is the naturally occurring isomer, and DL-lactide is the synthetic blend of D-lactide and L-lactide. The homopolymer of -lactide (LPLA) is a semicrystalline polymer. PGA and LPLA exhibit high tensile strength and low elongation and consequently have a high modulus that makes them more applicable than the amorphous polymers for load-bearing applications such as in orthopedic fixation and sutures. Poly(DL-lactide) (DL-PLA) is an amorphous polymer having

lower tensile strength and higher elongation and much more rapid degradation time making it more attractive as a drug delivery system. The degradation time of LPLA is much slower than that of DL-PLA requiring greater than 2 years to be completely absorbed^[110]. Copolymers of ϵ -caprolactone with DL-lactide have been synthesized to yield materials with more rapid degradation rates^[111]. A copolymer of ϵ -caprolactone with glycolide that has reduced stiffness compared to pure PGA is being sold as a monofilament suture.

Non-specific inflammatory reactions in living tissue associated with the degradation of the implant sometimes lead to a clinical complication, either to small fluid accumulation under the skin needing no treatment, or to painful fluid accumulation under the skin treated by aspiration with a needle^[112]. Polymeric bio-implants are made by traditional melt-molding techniques such as blow molding and injection molding, extrusion, vacuum forming, fiber spinning and sintering technique. They could be produced in the final or semifinal form. The main requirements for polymer bioimplants are sterile production and sterile techniques. To gain better mechanical



Figure 5. Bone-fracture fixation polymer screws, rods and plates
Slika 5. Polimerni vijaki, žablji in plošče za fiksacijo fraktur kosti

properties the self-reinforcing technique has been introduced^[101]. Examples of polymeric material medical devices are sutures, small staple devices, drug delivery devices, small pins, bone-plates and screws, depicted in Figure 5.

CERAMIC BIOMATERIALS

In order to avoid the problems associated with random dissolution which include uncontrolled physical degradation, particulate release and long-term durability, the materials need to remain essentially insoluble only to be removed by specific cell activity. Ceramic biomaterials have been developed that not only act as suitable substrates for bone mineralization by osteoblasts but are essentially insoluble in biological media and are resorbed when acted upon by osteoclasts^[113]. Zirconium dioxide or zirconia ceramics (ZrO_2) is a bioinert non-resorbable metal oxide which has a good chemical and dimensional stability, and a high strength and toughness^[114]. Currently, zirconia ceramic is being recognized for its high strength and surface finish, making this material potentially suitable for the highly loaded environments found in joint replacement. Biomedical grade zirconia was introduced approximately 20 years ago to solve the problem of alumina brittleness and the consequent potential failure of implants^[115]. Zirconia ceramic has been used to manufacture femoral heads for total hip replacements since the late 1980s^[115]. Its color and excellent biocompatibility and mechanical properties have made it attractive for dental applications^[116-120]. A prerequisite for successful bone implant integration is direct bone apposition which

was observed at bone-zirconia interfaces in histological^[121-123] and ultrastructural^[124] studies suggesting that ZrO_2 may also be a suitable implant material. On the one hand, biomedical grade zirconia exhibits the best mechanical properties of oxide ceramics: this is the consequence of phase transformation toughening, which increases its crack propagation resistance. The stress-induced phase transformation involves the transformation of metastable tetragonal grains to the monoclinic phase at the crack tip, which, accompanied by volume expansion, induces compressive stresses^[125]. On the other hand, due to this meta-stability, zirconia is prone to ageing in the presence of water^[126]. Up to date clinical reports appear to be again somewhat opposite where some results show excellent behavior of some femoral heads after several years in vivo^[127] while others show poor follow up results^[128] with severe wear and osteolysis around the implant. Few case studies report surface degradation of zirconia implants, which could be related to ageing^[129,130].

Alumina ceramics (Al_2O_3) have been used for implants and prostheses for several decades now^[131]. The material is characterised by its excellent biocompatibility^[132] and high strength, hardness and fracture resistance^[133,134]. The resultant high wear resistance is of particular interest for implant components with articulating surfaces like artificial joints. The outstanding wear resistance is the major reason for the predominant use of this material for the femoral joint head^[135]. A common material pairing used for hip arthroplasty is a femoral joint head made of alumina and an acetabular liner made of UHMWPE. Inauspiciously alumina is not suitable for

implant components with bone contact, because the material is bioinert and thereby no bony ongrowth, and subsequently loosening of the implant occurs^[136].

No difference between the biocompatibility of zirconia and that of alumina ceramics has been found in the biological reaction *in vivo*. Furthermore, the wear factor of UHMWPE against zirconia ceramic is 40-60 % less than that against alumina ceramic counterfaces and 10-20 % less than that against SUS 316L metal counterfaces^[137]. Hydroxyapatite ceramic (HAC) granules are used successfully world-wide as a bone substitute material because of their high biocompatibility. In an orthotopic site, such as a bony defect, bone formation occurs on ceramic surfaces. This newly formed bone bonds tightly to the ceramic surface without any mechanical interlock^[138]. Bioactive glasses have been used in many medical applications. However, due to their poor mechanical properties, these glasses cannot be used in load-bearing applications, whereas metallic alloys are still the materials of choice. It was recognized early on that one of the main applications of bioactive glasses could be coatings for prosthetic metallic implants. These coatings serve two purposes: improving the osseointegration of the implants, and protecting the metal against corrosion from the body fluids and the tissue from the corrosion products of the alloys. Unfortunately, most of the attempts to coat metallic implants with bioactive glasses have had limited success due to poor adhesion of the coating and/or degradation of the glass properties during the coating procedure, typically enameling, or flame or plasma spray coating^[139].

In recent years, transition metal nitrides like TiN, ZrN, TiAlN, NbN, TaN and VN were successfully used as protective coatings against wear and corrosion in order to increase the life expectancy of surgical implants and prosthesis^[140-142].

CONCLUSIONS

As presented in the article, a variety of different materials and processing technologies are available for medical applications. Which material should be used depends on the type of injury. Medical implants used for temporary healing should be made of conventional metallic biomaterials. The question of the long-term effects of bio-metal, on patients is very important. Further studies relating to long-term effects of materials on biological tissues are necessary, and are likely to lead to an increased understanding of the biocompatibility of materials in the future. Medical implants used for permanent healing are made of titanium alloy due to its inertness and good material characteristics, or polymeric degradable materials due to their properties and possibility of using them together with medicaments and bone healing stimulants. SMA medical devices and implants have been successful because they offered a possibility of performing less invasive surgeries. Nitinol wires in medical instruments are more kink resistant and have smaller diameter compared to 316L or polymer devices. Research to develop a porous SMA which enables the transport of body fluids from outside to inside the bone is currently underway. It was also an intention to provide an overview of uses of polymers in medicine, the field, where is expected an increased number of

applications. To address the limitations of current surgical techniques a diverse and large number of roles exist for bone biomaterials that are capable of incorporation into the natural process of healing. We expect that in the future, surgeons will have available even more products that will accelerate patient recovery and eliminate secondary surgeries.

In the future research and development of a new perspective that conflates in solid-state physics, biological science as well as materials engineering. The perspective is one that recognizes that future new advances in all these areas will be based on a fundamental understanding of the atomic and molecular infrastructure of materials that has resulted from the past achievements of physics and chemistry. Major advances will be achieved when the novel behavior, in particular the quantum mechanical behavior, that nanoscale structures possess, can be controlled and harnessed. The next decade should see the emergence of new technologies based on nano-systems with not only improved but hopefully also fundamentally new physico-chemical properties produced at reasonable costs. Experimental and theoretical research should lead to industrial applications yielding important breakthroughs.

POVZETEK

Pregled materialov v medicinskih aplikacijah

Medicinski vsadki so izdelki, ki morajo zadovoljiti stroge funkcijske zahteve, ki jih narekuje človeško telo kot delovno okolje.

V idealnem primeru naj bi imeli biomehanske lastnosti primerljive z avtogenim tkivom brez škodljivih učinkov in so evidentirani tako, da zagotavljajo varnost in uspešnost. Izbira materiala pri načrtovanju medicinskih vsadkov je pogojena z biokompatibilnostjo, bioadhezijo, biofunkcionalnostjo, korozijsko odpornostjo, itd. V tem pregledu so kovinski biomateriali razdeljeni v štiri podskupine: nerjavna jekla, kobaltove zlitine, titanove zlitine in druge kovine. Nekaj pozornosti je posvečeno tudi zlitinam s spominom, polimernim in keramičnim biomaterialom zaradi njihove inovativne uporabe v praktičnih medicinskih aplikacijah.

V delu so predstavljeni različni materiali in procesne tehnike, ki so mogoče za medicinske aplikacije. Izbira materiala je pogojena s tipom poškodbe. Medicinski vsadki, ki so začasno v telesu naj bi bili narejeni iz klasičnih kovinskih biomaterialov. Vprašanje kako na pacienta vpliva dolgoročni efekt stika tkivo-kovina je izjemnega pomena. Potrebne so nadaljne študije v zvezi z dolgoročnim vplivom materialov na biološka tkiva, ki bodo po vsej verjetnosti vodile povečanemu razumevanju biokompatibilnosti materialov v prihodnosti. Medicinski vsadki, namenjeni permanentni uporabi so običajno narejeni iz titanovih zlitin, zaradi njihove inertnosti in dobrih lastnosti, ali iz razgradljivih polimernih materialov zaradi posebnih lastnosti in možnosti hkratne uporabe z zdravili ali stimulantni rasti kosti. Medicinski pripomočki in vsadki narejeni iz materialov s spominom (SMA) so razširjeni in uspešni, ker ponujajo možnost izvršiti manj invazivnih kirurških posegov. Žice in medicinski instrumenti iz nitinola so bolj odporne proti vozljem in omogo-

čajjo manjše premere v primerjavi z nerjavečim jeklom AISI 316L ali polimernimi pripomočki. Veliko raziskav in razvoja je narejenega tudi na poroznem SMA, ki omogoča prenos telesnih sokov proti kostem. Namen študije je bil tudi zagotovitev pregleda uporabe polimerov v medicini, ki je področje kjer pričakujemo povešanje števila aplikacij.

Za odpravo omejitev kururških tehnik obstaja veliko število različnih vlog biomaterialov, ki so se sposobni vključiti v naravni proces zdravljenja. V prihodnosti pričakujemo, da bodo imeli kirurgi še precej več produktov, ki bodo pospešili pacientovo okrevanje in izključili potrebe po sekundarnih operacijah.

REFERENCES

- [1] RATNER, B.D., SCHOEN, F., HOFFMAN, A., LEMONS, J. (2004): *Biomaterials Science: An Introduction to Materials in Medicine*. Elsevier Science & Technology Books, ISBN: 978-0-125-82463-7.
- [2] BALAZIC, M., BROJAN, M., BOMBAC, D., CARAM, R. JR., KOSEL, F., KOPAC, J. (2007): Titanium and titanium alloy applications in medicine. *Surface Engineered Surgical Tools and Medical Devices*. Editors: Jackson, J., Ahmed, W., Springer, ISBN: 978-0-387-27026-5.
- [3] SUMITA, M., HANAWAB, T., TEOH, S.H. (2004): Development of nitrogen-containing nickel-free austenitic stainless steels for metallic biomaterials - review. *Materials Science and Engineering: C*; Vol. 24, No. 6-8, pp. 753-760.
- [4] *Annual Book of ASTM Standards*. Section 13 - Medical Devices and Services: Medical Devices, 1999.
- [5] WILLIAMS, D.F. (1993): Volume 14: Medical and Dental Materials. *Materials Science and Technology: A Comprehensive Treatment*. Editors: Cahn, R.W., Haasen, P., Kramer, E.J., Wiley-VCH, ISBN: 978-3-527-26825-2.
- [6] BREME, J., BIEHL, V. (1998): Metallic biomaterials. *Handbook of Biomaterials Properties*. Editors: Black, J., Hastings, G., Springer, ISBN: 978-0-412-60330-3.
- [7] BOMBAC, D., BROJAN, M., KRKOVIC, M., TURK, R., ZALAR, A. (2007): Characterization of titanium and stainless steel medical implants

- surfaces. *RMZ - Materials and geoenvironment.*; Vol. 54, No. 2, pp. 151-164.
- [8] ILEVBAR, G.O., BURSTEIN, G.T. (2001): The role of alloyed molybdenum in the inhibition of pitting corrosion in stainless steels. *Corrosion Science.*; Vol. 43, No. 3, pp. 485-513.
- [9] LIU, C., BI, Q., MATTHEWS, A. (2003): Tribological and electrochemical performance of PVD TiN coatings on the femoral head of Ti-6Al-4V artificial hip joints. *Surface and Coatings Technology.*; Vol. 163-164, pp. 597-604.
- [10] NEWSON, T. (2002): Stainless Steel – A Family of Medical Device Materials. *Business briefing: medical device manufacturing & technology.*
- [11] DAVIDS, J.R. (1998): *Metals Handbook.* ASM International.
- [12] SHETTY, R.H., OTTERSBERG, WI-I. (1995): Metals in Orthopedic Surgery. *Encyclopedic Handbook of Biomaterials and Bioengineering.* Editors: Wise, D., Trantolo, J.D., Altobelli, E.D., Yaszemski, J.M., Gresser, J.D., Schwartz, E.R., Marcel Dekker, ISBN: 978-0-824-79649-5.
- [13] SURY, P., SEMLITSCH, M. (1978): Corrosion behavior of cast and forged cobalt-based alloys for double-alloy joint endoprostheses. *Journal of Biomedical Materials Research.*; Vol. 12, No. 5, pp. 723-741.
- [14] KATTI, K.S. (2004): Biomaterials in total joint replacement. *Colloids and Surfaces B: Biointerfaces.*; Vol. 39, No. 3, pp. 132-142.
- [15] LONG, M., RACK, H.J. (1998): Titanium alloys in total joint replacement - a materials science perspective. *Biomaterials.*; Vol. 19, No. 18, pp. 1621-1639.
- [16] MILOSEV, I., STREHBLow, H.-H. (2003): The composition of the surface passive film formed on CoCrMo alloy in simulated physiological solution. *Electrochimica Acta.*; Vol. 48, No. 19, pp. 2767-2774.
- [17] KOCIJAN, A., MILOSEV, I., PIHLAR, B. (2004): Cobalt-based alloys for orthopaedic applications studied by electrochemical and XPS analysis. *Journal of Material Science: Materials in Medicine.*; Vol. 15, pp. 643-650.
- [18] OKAZAKI, Y., GOTOH, E. (2005): Comparison of metal release from various metallic biomaterials in vitro. *Biomaterials.*; Vol. 26, No. 1, pp. 11-21.
- [19] JACOBS, J.J., SKIPOR, A.K., PATTERSON, L.M., HALLAB, N.J., PAPROSKY, W.G., BLACK, J., GALANTE, J.O. (1998): A prospective, controlled, longitudinal study of metal release in patients undergoing primary total hip arthroplasty. *Journal of Bone and Joint Surgery.*; Vol. 80 A, No. 10, pp. 1444-1458.
- [20] RACK, H.J., QAZI, J.I. (2006): Titanium alloys for biomedical applications. *Materials Science and Engineering: C.*; Vol. 26, No. 8, pp. 1269-1277.
- [21] WANG, K. (1996): The use of titanium for medical applications in the USA. *Materials Science and Engineering: A.*; Vol. 213, No. 1-2, pp. 134-137.

- [22] DOWSON, D. (1992): Friction and wear of medical implants and prosthetic devices. *ASM Handbook.*; Vol. 18, ASM International, Gereland, Materials Park, OH, pp. 656-664.
- [23] PARK, J.B., LAKES, R.S. (1992): *Bio-materials - An introduction*. 2nd ed, Springer, ISBN: 978-0-306-43992-6.
- [24] SEMLITSCH, M.F., WEBER, H., STREICHER, R.M., SCHOÖN, R. (1992): Joint replacement components made of hot-forged and surface-treated Ti-6Al-7Nb alloy. *Biomaterials.*; Vol. 13, No. 11, pp. 781-788.
- [25] BOROWY, K-H., KRAMER, K-H. (1985): On the properties of a new titanium alloy (TiAl5Fe2.5) as implant material. *Titanium '84 Science and Technology.*; Vol. 2, Munich, Deutsche Gesellschaft Für Metallkunde EV, pp. 1381-1386.
- [26] ZWICKER, R., BÜHLER, K., MÜLLER, R., BECK, H., SCHMID, H-J., FERSTL, J. (1980): *Titanium '80 Science and Technology*. Warrendale, PA, TMS-AIME, pp. 505-514.
- [27] STEINEMANN, S.G. (1980): Corrosion of surgical implants-in vivo and in vitro tests. *Evaluation of Biomaterials*. Editors: Winter, G.D., Leray, J.L., de Groot, K., John Wiley, ISBN 978-0-471-27658-6.
- [28] STEINEMANN, S.G. (1985): Corrosion of titanium and titanium alloys for surgical implants. *Titanium '84 Science and Technology.*; Vol. 2, Munich, Deutsche Gesellschaft Für Metallkunde EV, 1373-1379.
- [29] LAING, P.G., FERGOSUN, A.B. JR., HODGE, E.S. (1967): Tissue reaction in rabbit muscle exposed to metallic implants. *Journal of Biomedical Materials Research.*; Vol. 1, No. 1, pp. 135-149.
- [30] WANG, K., GUSTAVSON, L., DUMBLETON, J. (1993): The characterization of Ti-12Mo-6Zr-2Fe. A new biocompatible titanium alloy developed for surgical implants. *Beta Titanium in the 1990's*. The Minerals, Metals&Materials Society, pp. 49-60.
- [31] WANG, K., GUSTAVSON, L., DUMBLETON, J. (1993): Low modulus, high strength, biocompatible titanium alloy for medical implants. *Titanium '92 Science and Technology*. The Minerals, Metals & Materials Society, 2697-2704.
- [32] STEINEMANN, S.G., MAÜSLI, P-A., SZMUKLER-MONCLER, S., SEMLITSCH, M., POHLER, O., HINTERMANN, H-E., PERREN, S.M. (1993): Beta-titanium alloy for surgical implants. *Titanium '92 Science and Technology*. The Minerals, Metals & Materials Society, pp. 2689-2696.
- [33] OKAZAKI, Y., ITO, Y., ITO, A., TATEISHI, T. (1993): Effect of alloying elements on mechanical properties of titanium alloys for medical implants. *Materials Transactions JIM.*; Vol. 34, No. 12, pp. 1217-1222.
- [34] KOVACS, P., DAVIDSON, J.A. (1993): The electrochemical behavior of a new titanium alloy with superior biocompatibility. *Titanium '92 Science and Technology*. The Minerals, Metals& Materials Society, pp. 2705-2712.
- [35] MISHRA, A.K., DAVIDSON, J.A., KOVACS, P., POGGIE, R.A. (1993): Ti-

- 13Nb-13Zr: a new low modulus, high strength, corrosion resistant near-beta alloy for orthopaedic implants. *Beta Titanium in the 1990's*. The Minerals, Metals & Materials Society, pp. 61-72.
- [36] EDWARDS, J. (1997): *Coating and Surface Treatment Systems for Metals*. Finishing Publications and ASM International, pp. 39-40, ISBN 0-904477-16-9.
- [37] CHEN, J.Z., SHI, Y.L., WANG, L., YAN, F.Y., ZHANG, F.Q. (2006): Preparation and properties of hydroxyapatite-containing titania coating by micro-arc oxidation. *Materials Letters.*; Vol. 60, No. 20, pp. 2538-2543.
- [38] DEARNALEY, G., ARPS, J. (2005): Biomedical applications of diamond-like carbon (DLC) coatings: a review. *Surface and Coatings Technology.*; pp. 2518-2524.
- [39] HALLAB, N., MERRITT, K., JACOBS, J.J. (2001): Metal sensitivity in patients with orthopaedic implants. *Journal of Bone and Joint Surgery.*; Vol. 83 A, No. 3, pp. 428-433.
- [40] BOBYN, J.D., STACKPOOL, G.J., HACKING, S.A., TANZER, M., KRYGIER, J.J. (1999): Characteristics of bone ingrowth and interface mechanics of a new porous tantalum biomaterial. *Journal of Bone and Joint Surgery.*; Vol. 81 B, No. 5, pp. 907-914.
- [41] BLACK, J. (1994): Biological performance of tantalum. *Clinical Materials.*; Vol. 16, No. 3, pp. 167-173.
- [42] LEVINE, B.R., SPORER, S., POGGIE, R.A., DELLA VALLE, C.J., JACOBS, J.J. (2006): Experimental and clinical performance of porous tantalum in orthopedic surgery. *Biomaterials.*; Vol. 27, No. 27, pp. 4671-4681.
- [43] KATO, H., NAKAMURA, T., NISHIGUCHI, S., MATSUSUE, Y., KOBAYASHI, M., MIYAZAKI, T., KIM, H.-M., KOKUBO, T. (2000): Bonding of alkali- and heat-treated tantalum implants to bone. *Journal of Biomedical Materials Research.*; Vol. 53, No. 1, pp. 28-35.
- [44] ZARDIACKAS, L.D., PARSELL, D.E., DILLON, L.D., MITCHELL, D.W., NUNNERY, L.A., POGGIE, R.A. (2001): Structure, metallurgy, and mechanical properties of a porous tantalum foam. *Journal of Biomedical Materials Research.*; Vol. 58, No. 2, pp. 180-187.
- [45] HACKING, S.A., BOBYN, J.D., TOH, K., TANZER, M. and KRYGIER, J.J. (2000): Fibrous tissue ingrowth and attachment to porous tantalum. *Journal of Biomedical Materials Research.*; Vol. 52, No. 4, pp. 631-638.
- [46] <http://www.zimmer.com/z/ctl/op/global/action/1/id/33/template/MP/navid/312>.
- [47] RIES, M., SALEHI, A., WIDDING, K., HUNTER, G. (2002): Polyethylene wear performance of oxidized zirconium and cobalt-chromium knee components under abrasive conditions. *Journal of Bone and Joint Surgery.*; Vol. 84 A, No. 2, pp. 129-135.
- [48] HUNTER, G., LONG, M. (2000): Abrasive wear of oxidized Zr-2.5Nb, CoCrMo, and Ti-6Al-4V against bone cement. *6th World Biomate-*

- rials Congress Transactions. Society For Biomaterials, Minneapolis.*
- [49] BENEZRA, V., MANGIN, S., TRESKA, M., SPECTOR, M., HUNTER, G., HOBBS, L.W. (1999): Microstructural investigation of the oxide scale on Zr-2.5Nb and its interface with the alloy substrate. *Biomedical Materials: Drug Delivery, Implants and Tissue Engineering*. Editors: Neenan, T., Marcolongo, M., Valentini, R.F., Materials Research Society Symposium Proceedings, Vol. 550, Materials Research Society, pp. 337-342, ISBN: 978-1-558-99456-0.
- [50] HUNTER, G. (2001): Adhesion testing of oxidized zirconium. *Transactions of the Society for Biomaterials.*; Vol. 24, pp. 540.
- [51] <http://www.oxinium.co.uk/patients/material.php>
- [52] YANLI, C., CHUNYONG, L., SHENGLI, Z., ZHENDUO, C., XIANJIN, Y. (2006): Formation of bonelike apatite - collagen composite coating on the surface of NiTi shape memory alloy. *Scripta Materialia.*; Vol. 54, No. 1, pp. 89-92.
- [53] DUERIG, T.W., TOLOMEI, D.E., WHOLEY, M. (2000): An overview of superelastic stent design. *Minimally Invasive Therapy & Allied Technologies.*; Vol. 9, No. 3-4, pp. 235-246.
- [54] MORGAN, N.B. (2004): Medical shape memory alloy applications the market and its products. *Materials Science and Engineering A.*; Vol. 378, No. 1-2, pp. 16-23.
- [55] DUERIG, T.W., PELTON, A., STOECKEL, D. (1999): An overview of nitinol medical applications. *Materials Science and Engineering A.*; Vol. 273-275, pp. 149-160.
- [56] KURTZ, S.M., MURATOGLU, O. K., EVANS, M., EDIDIN, A. A. (1999): Advances in the processing, sterilization, and crosslinking of ultra-high molecular weight polyethylene for total joint arthroplasty. *Biomaterials.*; Vol. 20, No. 18, pp. 1659-1688.
- [57] KURTZ, S.M. (2004): *The UHMWPE Handbook: Ultra-High Molecular Weight Polyethylene in Total Joint Replacement*. Academic Press, New York, ISBN: 978-0-124-29851-4.
- [58] LEWIS, G. (1997): Polyethylene wear in total hip and knee arthroplasties. *Journal of Biomedical Materials Research.*, Vol. 38, No. 1, pp. 55-75.
- [59] SCHMALZRIED, T.P., KWONG, L.M., JASTY, M., SEDLACEK, R.C., HAIRE, T.C., O'CONNOR, D.O., BRAGDON, C.R., KABO, J.M., MALCOLM, A.J., HARRIS, W.H. (1992): The mechanism of loosening of cemented acetabular components in total hip arthroplasty. Analysis of specimens retrieved at autopsy. *Clinical Orthopaedics and related research.*; Issue 274, pp. 60-78.
- [60] XENOS, J.S., HOPKINSON, W.J., CALLAGHAN, J.J., HEEKIN, R.D., SAVORY, C.G. (1995): Osteolysis around an uncemented cobalt chrome total hip arthroplasty: Polyethylene wear. *Clinical Orthopaedics and related research.*; Issue 317, pp. 29-36.

- [61] LIVINGSTON, B.J., CHMELL, M.J., SPEC-
TOR, M., POSS, R. (1997): Complica-
tions of total hip arthroplasty as-
sociated with the use of an acetab-
ular component with a Hylamer
liner. *Journal of Bone and Joint
Surgery.*; Vol. 79 A, Issue 10, pp.
1529-1538.
- [62] LI, S., BURSTEIN, A.H. (1994): Ultra-
high molecular weight polyethyl-
ene. The material and its use in to-
tal joint implants. *Journal of Bone
and Joint Surgery.*; Vol. 76 A, No.
7, pp. 1080-1090.
- [63] PREMNATH, V., HARRIS, W.H., JASTY,
M., MERRILL, E.W. (1996): Gam-
ma sterilization of UHMWPE ar-
ticular implants: an analysis of the
oxidation problem. *Biomaterials.*;
Vol. 17, No. 18, pp. 1741-1753.
- [64] SAUER, W.L., ANTHONY, M.E. (1998):
Predicting the clinical wear per-
formance of orthopaedic bearing
surfaces. *Alternative bearing sur-
faces in total joint replacement.*
Editors: Jacobs, J.J., Craig, T.L.,
American Society for Testing and
Materials, West Conshohoken,
ISBN: 978-0-803-12490-5.
- [65] MCKELLOP, H.A. (1998): Wear as-
sessment. *The adult hip.* Editors:
Callaghan, J.J., Rosenberg, A.G.,
Rubash, H.E., Lippincott-Raven
Publishers, Philadelphia, ISBN:
978-0-397-51704-6.
- [66] EYERER, P., ELLWANGER, R., FEDEROLF,
H.-A., KURTH, M., MADLER, H.
(1990): Polyethylene. *Concise en-
cyclopaedia of medical and den-
tal materials.* Editors: Williams,
D., Cahn, R., Pergamon, Oxford,
ISBN: 978-0-080-36194-9.
- [67] FISHER, J., DOWSON, D. (1991): Tribol-
ogy of total artificial joints. *Pro-
ceedings of the Institution of Me-
chanical Engineers, Part H.*; Vol.
205, No. 2, pp. 73-79.
- [68] KLEIN, P.G., GONZALEZ-OROZCO, J.A.,
WARD, I.M. (1991): Structure and
morphology of highly oriented ra-
diation crosslinked polyethylene
fibres. *Polymer.*; Vol. 32, No. 10,
pp. 1732-1736.
- [69] [http://www.dsm.com/en_US/html/
hpf/dyneema_purity.htm](http://www.dsm.com/en_US/html/hpf/dyneema_purity.htm).
- [70] ELLIS, R.H., MULLVEIN, J. (1974): The
cardiovascular effects of methyl-
methacrylate. *Journal of Bone and
Joint Surgery.*; Vol. 56 B, No. 1,
pp. 59-61.
- [71] GRESHAM, G.A., KUCZYŃSKI, A., ROS-
BOROUGH, D. (1971): Fatal fat
embolism following replacement
arthroplasty for transcervical frac-
tures of femur. *British Medical
Journal.*; Vol. 2 (5762), pp. 617-
619.
- [72] HOMSY, C.A., TULLOS, H.S., ANDERSON,
M.S., DIFERRANTE, N.M., KING,
J.W. (1972): Some physiological
aspects of prosthesis stabilization
with acrylic polymer. *Clinical Or-
thopaedics and related research.*;
Issue 83, pp. 317-328.
- [73] PHILLIPS, H., COLE, P.V., LETTIN, A.W.F.
(1971): Cardiovascular effects of
implanted acrylic bone cement.
British Medical Journal.; Vol.
3(5772), pp. 460-461.
- [74] ORSINI, E.C., BYRICK, R.J., MULLEN,
J.B.M., KAY, J.C., WADDELL, J.P.
(1987): Cardiopulmonary function
and pulmonary microemboli dur-
ing arthroplasty using cemented or

- non-cemented components. *Journal of Bone and Joint Surgery.*; Vol. 69, No. 6, pp. 822-832.
- [75] KAUFMANN, T.J., JENSEN, M.E., FORD, G., GILL, L.L., MARX, W.F., KALLMES, D.F. (2002): Cardiovascular Effects of Polymethylmethacrylate Use in Percutaneous Vertebroplasty. *American Journal of Neuroradiology.*; Vol. 23, No. 4, pp. 601-604.
- [76] MILLER, M.D. (2004): *Review of Orthopaedics 4th Edition*. W.B. Saunders Company, ISBN: 978-0-721-60364-3, 129.
- [77] OLSON, R.J., CRANDALL, A.S. (1998): Silicone vs polymethylmethacrylate intraocular lenses with regard to capsular opacification. *Ophthalmic Surgery and Lasers.*; Vol. 29, No. 1, pp. 55-58.
- [78] OSHIKA, T., NAGAHARA, K., YAGUCHI, S. (1998): Three-year prospective, randomized evaluation of intraocular lens implantation through 3.2 mm and 5.5 mm incisions. *Journal of Cataract & Refractive Surgery.*; Vol. 24, No. 4, pp. 509-514.
- [79] KRUGER, A.J., SCHAUERSBERGER, J., ABELA, C., SCHILD, G., AMON, M. (2000): Two-year results: sharp vs rounded optic edges on silicone lenses. *Journal of Cataract & Refractive Surgery.*; Vol. 26, No. 4, pp. 566-570.
- [80] FINDL, O., BUEHL, W., MENAPACE, R., SACU, S., GEORGOPOULOS, M., RAINER, G. (2005): Long-term Effect of Sharp Optic Edges of a Polymethyl Methacrylate Intraocular Lens on Posterior Capsule Opacification: A Randomized Trial. *Ophthalmology.*; Vol. 112, No. 11, pp. 2004-2008.
- [81] KURTZ, S.M., DEVINE, J. N. (2007): PEEK biomaterials in trauma, orthopedic, and spinal implants. *Biomaterials.*; Vol. 28, No. 32, pp. 4845-4869.
- [82] LIAO, K. (1994): Performance characterization and modeling of a composite hip prosthesis. *Experimental Techniques.*; Vol. 18, No. 5, pp. 33-38.
- [83] MAHARAJ, G.R., JAMISON, R.D. (1993): Intraoperative impact: characterization and laboratory simulation on composite hip prostheses. *Composite Materials for Implant Applications in the Human Body: Characterization and Testing*. Editors: Jamison, R.D., Gilbertson, L.N., ASTM, Philadelphia, ISBN: 978-0-803-11852-2, 98-108.
- [84] KELSEY, D.J., SPRINGER, G.S., GOODMAN, S.B. (1997): Composite implant for bone replacement. *Journal of Composite Materials.*; Vol. 31, No. 16, pp. 1593-1632.
- [85] CORVELLI, A.A., BIERMANN, P.J., ROBERTS, J.C. (1997): Design, analysis, and fabrication of a composite segmental bone replacement implant. *Journal of Advanced Materials.*; Vol. 2, No. 1, pp. 2-8.
- [86] TOTH, J.M., WANG, M., ESTES, B.T., SCIFERT, J.L., SEIM, III H.B., TURNER, A.S. (2006): Polyetheretherketone as a biomaterial for spinal applications. *Biomaterials.*; Vol. 27, No. 3, pp. 324-334.
- [87] BRANTIGAN, J.W., NEIDRE, A., TOOHEY, J.S. (2004): The Lumbar I/F Cage for posterior lumbar interbody fu-

- sion with the variable screw placement system: 10-year results of a Food and Drug Administration clinical trial. *The Spine Journal.*; Vol. 4, No. 6, pp. 681-688.
- [88] BRANTIGAN, J.W., STEFFEE, A.D., LEWIS, M.L., QUINN, L.M., PERSENAIRE, J.M. (2000): Lumbar interbody fusion using the Brantigan I/F cage for posterior lumbar interbody fusion and the variable pedicle screw placement system: two-year results from a Food and Drug Administration investigational device exemption clinical trial. *Spine.*; Vol. 25, No. 11, pp. 1437-1446.
- [89] AKHAVAN, S., MATTHIENSEN, M.M., SCHULTE, L., PENOYAR, T., KRAAY, M.J., RIMNAC, C.M., GOLDBERG, V.M. (2006): Clinical and histologic results related to a low-modulus composite total hip replacement stem. *Journal of Bone and Joint Surgery.*; Vol. 88 A, No. 6, pp. 1308-1314.
- [90] GLASSMAN, A.H., CROWNINSHIELD, R.D., SCHENCK, R., HERBERTS, P. (2001): *A low stiffness composite biologically fixed prosthesis.* Clin Orthop Relat Res (393), pp. 128-136.
- [91] KÄRRHOLM, J., ANDERBERG, C., SNORRASON, F., THANNER, J., LANGE-LAND, N., MALCHAU, H., HERBERTS, P. (2002): Evaluation of a femoral stem with reduced stiffness. A randomized study with use of radiostereometry and bone densitometry. *Journal of Bone and Joint Surgery.*; Vol. 84 A, No. 9, pp. 1651-1658.
- [92] WANG, A., LIN, R., STARK, C., DUMBLETON, J.H. (1999): Suitability and limitations of carbon fiber reinforced PEEK composites as bearing surfaces for total joint replacements. *Wear.*; Vol. 225-229, Part 2, pp. 724-727.
- [93] JONES, E., WANG, A., STREICHER, R. (2001): Validating the limits for a PEEK composite as an acetabular wear surface. *27th Annual Meeting of the Society for Biomaterials in conjunction with the 33rd International Biomaterials Symposium.*
- [94] JOYCE, T.J., RIEKER, C., UNSWORTH, A. (2006): Comparative in vitro wear testing of PEEK and UHMWPE capped metacarpophalangeal prostheses. *Bio-Medical Materials and Engineering.*; Vol. 16, No. 1, pp. 1-10.
- [95] MANLEY, M., ONG, K., KURTZ, S.M., RUSHTON, N., FIELD, R.E. (2007): Biomechanics of a PEEK horse-shoe-shaped cup: comparisons with a predicate deformable cup. *Transactions of the 53rd Orthopedic Research Society.*; Vol. 32, 1717.
- [96] YU, S., HARIRAM, K.P., KUMAR, R., CHEANG, P., AIK, K.K. (2005): In vitro apatite formation and its growth kinetics on hydroxyapatite/polyetheretherketone biocomposites. *Biomaterials.*; Vol. 26, No. 15, pp. 2343-2352.
- [97] FAN, J.P., TSUI, C.P., TANG, C.Y., CHOW, C.L. (2004): Influence of interphase layer on the overall elastoplastic behaviors of HA/PEEK biocomposite. *Biomaterials.*; Vol. 25, No. 23, pp. 5363-5373.
- [98] TAN, K.H., CHUA, C.K., LEONG, K.F., CHEAH, C.M., CHEANG, P., ABU

- BAKAR, M.S., CHA, S.W. (2003): Scaffold development using selective laser sintering of polyetheretherketone - hydroxyapatite biocomposite blends. *Biomaterials.*; Vol. 24, No. 18, pp. 3115-3123.
- [99] ABU BAKAR, M.S., CHENG, M.H.W., TANG, S.M., YU, S.C., LIAO, K., TAN, C.T., KHOR, K.A., CHEANG, P. (2003): Tensile properties, tension - tension fatigue and biological response of polyetheretherketone - hydroxyapatite composites for load-bearing orthopedic implants. *Biomaterials.*; Vol. 24, No. 13, pp. 2245-2250.
- [100] HA, S.W., KIRCH, M., BIRCHLER, F., ECKERT, K.-L., MAYER, J., WINTERMANTEL, E., SITTIG, C., PFUND-KLINGENFUSS, I., TEXTOR, M., SPENCER, N.D., GUECHEVA, M., VONMONT, H. (1997): Surface activation of polyetheretherketone (PEEK) and formation of calcium phosphate coatings by precipitation. *Journal of Materials Science: Materials in Medicine.*; Vol. 8, No. 11, pp. 683-690.
- [101] MIDDLETON, J.C., TIPTON, A.J. (2000): Synthetic biodegradable polymers as orthopaedic devices. *Biomaterials.*; Vol. 21, No. 23, pp. 2335-2346.
- [102] TÖRMÄLÄ, P., POHJONEN, T., ROKKANEN, P. (1998): Bioabsorbable polymers: materials technology and surgical applications. *Journal of Engineering in Medicine.*; Vol. 212, No. 2, pp. 101-111.
- [103] GILDING, D.K., REED, A.M. (1979): Biodegradable polymers for use in surgery - polyglycolic/poly(lactic acid) homo-and copolymers: 1. *Polymer.*; Vol. 20, No. 12, pp. 1459-1464.
- [104] BARROWS, T.H. (1986): Degradable Implant Materials: A Review of Synthetic Absorbable Polymers and Their Applications. *Clinical Materials.*; Vol. 1, pp. 233-257.
- [105] DOMB, A.J., AMSELEM, S., LANGER, R., MANIAR, M. (1994): Polyanhydrides as carriers of drugs. *Biomedical Polymers: Designed-To-Degrade Systems*. Editor: Shalaby, S.W., Hanser Gardner Pubns, pp. 69-96, ISBN: 978-1-569-90159-5.
- [106] HELLER, J., DANIELS, A.U. (1994): Poly(orthoesters). *Polymers: Designed-To-Degrade Systems*. Editor: Shalaby, S.W., Hanser Gardner Pubns, pp. 35-68, ISBN: 978-1-569-90159-5.
- [107] KOHN, J., ABRAMSON, S., LANGER, R. (2004): Bioresorbable and Bioerodible Materials. *Biomaterials science*. Editors: Ratner, B.D., Hoffman, A.S., Schoen, F.J., Lemons, J.E., Academic Press, pp. 115-126, ISBN: 978-0-125-82463-7.
- [108] SHALABY, S.W., JOHNSON, R.A. (1994): Synthetic absorbable polyesters. *Biomedical Polymers: Designed-To-Degrade Systems*. Editor: Shalaby, S.W., Hanser Gardner Pubns, pp. 1-34, ISBN: 978-1-569-90159-5.
- [109] SANTAVIRTA, S., KONTTINEN, Y.T., SAITO, T., GRONBLAD, M., PARTIO, E., KEMPPINEN, P., ROKKANEN, P. (1990): Immune response to poly-

- glycolic acid implants. *Journal of Bone and Joint Surgery.*; Vol. 72, No. 4, pp. 597-600.
- [110] BERGSMAN, J.E., DE BRUIJN, W.C., ROZEMA, F.R., BOS, R.R.M., BOERING, G. (1995): Late degradation tissue response to poly(L-lactide) bone plates and screws. *Biomaterials.*; Vol. 16, No.1, pp. 25-31.
- [111] SCHINDLER, A., JEFFCOAT, R., KIMMEL, G.L., PITT, C.G., WALL, M.E., ZWIEDINGER, R. (1977): Biodegradable polymers for sustained drug delivery. *Contemporary Topics in Polymer Science.*; Vol. 2, pp. 251-289.
- [112] BOSTMAN, O., HIRVENSALO, E., MAKINEN, J., ROKKANEN, P. (1990): Foreign-body reactions to fracture fixation implants of biodegradable synthetic polymers. *Journal of Bone and Joint Surgery.*; Vol. 72, No. 4, pp. 592-596.
- [113] LANGSTAFF, S., SAYER, M., SMITH, T.J.N., PUGH, S.M. (2001): Resorbable bioceramics based on stabilized calcium phosphates. Part II: evaluation of biological response. *Biomaterials.*; Vol. 22, No. 2, pp. 135-150.
- [114] PICONI, C., MACCAURO, G. (1999): Zirconia as a ceramic biomaterial. *Biomaterials.*; Vol. 20, No. 1, pp. 1-25.
- [115] CHRISTEL, P., MEUNIER, A., DORLOT, J.M., CROLET, J.M., WITVOET, J., SEDEL, L., BOUTIN, P. (1988): Biomechanical compatibility and design of ceramic implants for orthopedic surgery, Bioceramics: material characteristics versus in vivo behaviour. *Annals of the New York Academy of Sciences.*; Vol. 523, pp. 234-256.
- [116] AHMAD, I. (1998): Yttrium-partially stabilized zirconium dioxide posts: an approach to restoring coronally compromised nonvital teeth. *International Journal of Periodontics & Restorative Dentistry.*; Vol. 18, No. 5, pp. 454-465.
- [117] MEYENBERG, K.H., LUTHY, H., SCHÄRER, P. (1995): Zirconia Posts: New All-Ceramic Concept for Nonvital Abutment Teeth. *Journal of Esthetic Dentistry.*; Vol. 7, No. 2, pp. 73-80.
- [118] FRITZSCHE, J. (2003): Zirconium oxide restorations with the DCS pre-cident system. *International Journal of Computerized Dentistry.*; Vol. 6, No. 2, pp. 193-201.
- [119] TINSCHERT, J., NATT, G., MAUTSCH, W., SPIEKERMANN, H., ANUSAVICE, K.J. (2001): Marginal fit of alumina- and zirconia-based fixed partial dentures produced by a CAD/CAM system. *Operative Dentistry.*; Vol. 26, No. 4, pp. 367-374.
- [120] GLAUSER, R., SAILER, I., WOHLWEND, A., STUDER, S., SCHIBLI, M., SCHÄRER, P. (2004): Experimental zirconia abutments for implant-supported single-tooth restorations in esthetically demanding regions: 4-year results of a prospective clinical study. *The International Journal of Prosthodontics.*; Vol. 17, No. 3, pp. 285-290.
- [121] AKAGAWA, Y., HOSOKAWA, R., SATO, Y., KAMAYAMA, K. (1998): Comparison between freestanding and

- tooth-connected partially stabilized zirconia implants after two years' function in monkeys: a clinical and histologic study. *Journal of Prosthetic Dentistry.*; Vol. 80, No. 5, pp. 551-558.
- [122] AKAGAWA, Y., ICHIKAWA, Y., NIKAI, H., TSURU, H. (1993): Interface histology of unloaded and early loaded partially stabilized zirconia endosseous implant in initial bone healing. *Journal of Prosthetic Dentistry.*; Vol. 69, No. 6, pp. 599-604.
- [123] SCARANO, A., DI CARLO, F., QUARANTA, M., PIATTELLI, A. (2003): Bone response to zirconia ceramic implants: an experimental study in rabbits. *Journal of Oral Implantology.*; Vol. 29, No. 1, pp. 8-12.
- [124] ALBREKTSSON, T., HANSSON, H.A., IVARSSON, B. (1985): Interface analysis of titanium and zirconium bone implants. *Biomaterials.*; Vol. 6, No. 2, pp. 97-101.
- [125] GARVIE, R.C., HANNINK, R.H.J., PASCOE, R.T. (1975): Ceramic Steel?. *Nature.*; Vol. 258, pp. 703-704.
- [126] LAWSON, S. (1995): Environmental degradation of zirconia ceramics. *Journal of the European Ceramic Society.*; Vol. 15, No. 6, pp. 485-502.
- [127] CATON, J., BOURALY, J.P., REYNAUD, P., MERABET, Z. (2004): Phase Transformation in Zirconia Heads after THA Myth or Reality. *Proceedings of the Ninth BIOLOX Symposium.*; Vol. 26-27, pp. 73-74.
- [128] ALLAIN, J., LE MOUËL, S., GOUTALLIER, D., VOISIN, M.C. (1999): Poor eighth year survival of cemented zirconia-polyethylene total hip re-
placements. *Journal of Bone and Joint Surgery.*; Vol. 81 B, No. 5, pp. 835-842.
- [129] HARAGUCHI, K., SUGANO, N., NISHII, T., SAKAI, T., YOSHIKAWA, H., OHZONO, K. (2001): Phase transformation of a zirconia ceramic head after total hip arthroplasty. *Journal of Bone and Joint Surgery.*; Vol. 83 B, No. 7, pp. 996-1000.
- [130] CATLEDGE, S.A., COOK, M., VOHRA, Y., SANTOS, E.M., MCCLENNY, M.D., MOORE, K.D. (2003): Surface crystalline phases and nanoindentation hardness of explanted zirconia femoral heads. *Journal of Materials Science: Materials in Medicine.*; Vol. 14, No. 10, pp. 863-867.
- [131] HULBERT, S.F., YOUNG, F.A., MATHIEWS, R.S., KLAWITTER, J.J., TALBERT, C.D., STELLING, F.H. (1970): Potential of ceramic materials as permanently implantable skeletal prostheses. *Journal of Biomedical Materials Research.*; Vol. 4, No. 3, pp. 433-456.
- [132] GRISS, P., WERNER, E. (1980): Alumina ceramic, bioglass and silicon nitride. A comparative biocompatibility study. *Mechanical Properties of Biomaterials*. Editors: Hastings, G.W., Williams, D.F., John Wiley, pp. 217-225, ISBN: 978-0-471-27761-3.
- [133] GREEN, D.J. (1998): *An Introduction to the Mechanical Properties of Ceramics*. Cambridge University Press, ISBN: 978-0-521-59913-9.
- [134] MUNZ, D., FETT, T. (1999): *Ceramics: Mechanical Properties, Failure Behaviour, Materials Selection 1st*

- Ed.*, Springer Verlag, ISBN: 978-3-540-65376-9.
- [135] WILLMANN, G. (1998): Überlebensrate und Sicherheit von keramischen Kugelköpfen für Hüftendoprothesen (Survival rate and reliability of ceramic femoral heads for total hip arthroplasty). *Materialwissenschaft und Werkstofftechnik.*; Vol. 29, No. 10, pp. 595-604.
- [136] FISCHER, H., NIEDHART, C., KALTENBORN, N., PRANGE, A., MARX, R., NIETHARD, F.U., TELLE, R. (2005): Bioactivation of inert alumina ceramics by hydroxylation. *Biomaterials.*; Vol. 26, No. 31, pp. 6151-6157.
- [137] YEN, S.K., GUO, M.J., ZAN, H.Z. (2001): Characterization of electrolytic ZrO₂ coating on Co-Cr-Mo implant alloys of hip prosthesis. *Biomaterials.*; Vol. 22, No. 2, pp. 125-133.
- [138] HOLTGRAVE, E.A., DONATH, H. (1995): Response of odontoblast-like cells to hydroxyapatite ceramic granules. *Biomaterials.*; Vol. 16, No. 2, pp. 155-159.
- [139] LOPEZ-ESTEBAN, S., SAIZ, E., FUJINO, S., OKU, T., SUGANUMA, K., TOMSIA, A. P. (2003): Bioactive glass coatings for orthopedic metallic implants. *Journal of the European Ceramic Society.*; Vol. 23, No. 15, pp. 2921-2930.
- [140] BRAIC, M., BALACEANU, M., BRAIC, V., VLADESCU, A., PAVELESCU, G., ALBULESCU, M. (2005): Synthesis and characterization of TiN, TiAlN and TiN/TiAlN biocompatible coatings. *Surface and Coatings Technology.*; Vol. 200, No. 1-4, pp. 1014-1017.
- [141] LENG, Y.X., SUN, H., YANG, P., CHEN, J.Y., WANG, J., WAN, G.J., HUANG, N., TIAN, X.B., WANG, L.P., CHU, P.K. (2001): Biomedical properties of tantalum nitride films synthesized by reactive magnetron sputtering. *Thin Solid Films.*; Vol. 398-399, pp. 471-475.
- [142] HÜBLER, R., COZZA, A., MARCONDES, T.L., SOUZA, R.B., FIORI, F.F. (2001): Wear and corrosion protection of 316-L femoral implants by deposition of thin films. *Surface and Coatings Technology.*; Vol. 142-144, pp. 1078-1083.

