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Biomaterials for hip joint implants with particular emphasis on titanium and its alloys

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Abstract

Introduction: The primary objectives of replacing a natural hip joint with a prosthesis are to restore the function of a joint damaged by degenerative disease, advanced rheumatoid conditions, or mechanical injury, to eliminate pain caused by pathological joint changes, and to re-establish proper load transfer. Additionally, the procedure aims to restore motor functions, enabling the patient to move and perform essential daily activities such as walking, lifting, and carrying objects.

Material and methods: This article outlines the requirements for biomaterials used in the components of hip joint prostheses, presents various groups of titanium alloys, and describes their properties in detail. Special attention is paid to titanium and titanium alloys, including their mechanical and physical characteristics. The influence of Young's modulus on fretting and fatigue cracking in metallic alloys used in modular prosthetic components is also examined.

Results: Titanium alloys are generally not used for femoral heads due to their poor tribological performance. Without appropriate surface modification, titanium alloys are unsuitable for use in articulating components such as heads and acetabular cups, as they exhibit low abrasion resistance. They also demonstrate limited resistance to fretting damage, particularly at the junctions between the stem neck and head in modular prostheses.

Conclusions: To improve abrasion resistance, diffusion-based thermochemical surface treatments can be employed to harden the material's outer layer. However, the issue of fretting-induced damage at the neck region of hip endoprostheses remains unresolved. This problem is critical, as it often leads to corrosion and fatigue cracking of the stem neck, ultimately requiring prosthesis revision and incurring significant clinical and economic consequences. Plasma (ionic) nitriding has been shown to enhance surface properties such as hardness, wear resistance, resistance to fretting damage, fretting corrosion, and fretting fatigue, as well as overall fatigue cracking resistance.

Keywords: titanium alloys, nitriding, cobalt alloys, fretting, Young's modulus of elasticity

Introduction

Joint diseases are a serious social problem. The progress of civilization and the associated increase in car, motorcycle, and bicycle accidents, as well as joint diseases, an aging society, and the increasing average life expectancy of populations, combined with the sedentary lifestyle of seniors, result in unfavorable changes to the musculoskeletal system. The consequence is the growing demand for biomaterials that can replace a sick, damaged, or injured joint that requires a long-term stay in the human body. The main goals of replacing a natural hip joint with an endoprosthesis are reconstruction of a hip joint damaged as a result of degenerative changes, advanced rheumatoid disease, or mechanical injury; elimination of pain caused by pathological changes in the natural joint; enabling load transfer; restoration of motor functions; and the ability to move and perform basic human movements such as walking, lifting, and carrying objects. Endoprostheses implanted in the human body replace the function of the damaged element of the hip joint movement. If other treatments are ineffective, the only effective treatment method may be an invasive surgical method, consisting of replacing the naturally diseased joint with an endoprosthesis. The purpose of this article is to familiarize the reader with the requirements for materials used in hip endoprosthesis, the characteristics of the most commonly used materials, their properties, common problems, and ways to prevent them. Biomaterials must meet several criteria, demonstrating biotolerance and biocompatibility, being resistant to corrosion in a physiological environment, and characterized by high, long-term resistance to both constant and vari-

Authors' contribution: A – Research concept and design, B – Collection and/or assembly of data, C – Data analysis and interpretation, D – Writing the article, E – Critical revision of the article, F – Final approval of the article.

able stresses. Biomaterials in contact with tissues and body fluids of the human body should not cause allergic reactions, inflammations, diseases of the nervous system, blood clots, circulatory system disorders, etc.

The following factors influence biotolerance:

1. reactions of human implants with tissues and body fluids of the human body,
2. biomechanical, biophysical, and biochemical stability of biomaterials,
3. local mechanical damage, fretting,
4. fretting damage to the implant surface, ability to repassivate, and corrosion effects associated with this damage[1,2,4,23].

An endoprosthesis has a structure similar to the anatomy of a natural joint and is made of biomaterials with appropriate mechanical and biochemical properties, enabling the transfer of large loads and replacing natural bearing functions[1,2,4,5]. Titanium has very valuable properties, which led to its use in titanium-based alloys used in the production of hip endoprostheses. The main advantages of titanium that have influenced the use of alloys for orthopedic implant components are a high strength-to-density ratio, high resistance to general corrosion, biocompatibility, biotolerance, no susceptibility to pitting corrosion caused by chlorine ions (Cl^-) present in human body fluids, non-magnetic properties, and bioactivity (ability to form biochemical bonds with human bone). The high resistance of titanium alloys to general corrosion is related to their high affinity for oxygen and the formation of a stable, passive layer of titanium oxides on the surface, tightly adhered to the substrate. The passive layer of titanium oxides is resistant to chlorine ions and protects against pitting corrosion caused by chlorine ions (Cl^-). The passive layer is formed as a result of contact between metal and oxygen. In the case of mechanical damage to the passive layer, a necessary condition for repassivation is a free, unrestricted supply of oxygen to the damaged surface. Titanium alloys can repassivate in physiological solutions containing oxygen. In the absence of oxygen in physiological solutions, repassivation of the damaged surface of titanium alloys does not occur. Titanium alloys are particularly susceptible to pitting corrosion caused by fluoride ions (F^-), which may be present in the fluids of the human body. Fluoride ions can enter body fluids from toothpaste, mouthwash, and fluoridated water. Despite the many good properties of titanium alloys, there are still problems with material wear caused by poor tribological properties, fretting, fretting corrosion, and fatigue cracking caused by variable stresses. Fretting corrosion can cause allergic reactions and inflammation, leading to the rejection of the implanted prosthesis. Due to poor tribological properties, titanium alloys are particularly susceptible to fretting wear, which causes surface damage due to friction. Fatigue, fretting wear, fretting corrosion, and fretting fatigue are the most common causes of problems occurring in the neck and head joint connections of modular hip prostheses. Among the metal materials used for hip prostheses, cobalt alloys, titanium alloys, and chromium-nickel steels are most commonly used. One of the important parameters that determines the strength and durability of the implant-femoral connection is the value of Young's modulus of elasticity (E). Metal alloys used for hip endoprosthesis stems have a significantly higher Young's modulus of elasticity than the modulus of elasticity of compact bone. Young's modulus of elasticity of compact bone is 30 GPa; the titanium alloy Ti6Al4V is 115 GPa, and the cobalt alloy Co28Cr6Mo is 230 GPa. The large difference in Young's modulus of elasticity between a compact bone and a stem made of a metal alloy is, from the point of view of biomechanical compatibility, it would be desirable for the modulus of elasticity of the alloy from which the stem is made to be close to the modulus of elasticity of the femur so that there is no excessive concentration of stresses at the stem-bone connection boundary, which could lead to damage, separation, or loosening of the implant due to stress concentration and uneven bending of the stem and femur. From the point of view of fatigue crack resistance and resistance to fretting damage in the connection of the neck stem with the head of a modular prosthesis, an optimally high Young's modulus of elasticity is desirable. Progress in the treatment of hip joint diseases and injuries involves overcoming material barriers. The functions of biomaterials are constantly changing. One of the reasons for these changes is the increasing average human lifespan, which has major consequences for the development of biomaterials and a significant change in their role in the human body. The extension of human life is associated with increased requirements for longer durability and reliability of implants, especially since they must fulfill biomechanical functions in a long-term manner when subjected to variable loads. One of the most important parameters that determines the correct distribution of stresses between the stem and the femur is Young's modulus of elasticity E of the stem made of metal and bone. Young's modulus of elasticity also has a very large effect on the resistance to fretting damage of the neck surface with the head in modular endoprostheses. As Young's modulus decreases, the resistance to the fretting and fatigue cracking of the neck in the modular endoprosthesis connection decreases. The metal biomaterials used to make hip prosthesis stems have a modulus of elasticity much higher than bone. The much higher Young's modulus of elasticity of metal alloys used for prosthesis stems than the bone modulus creates difficulties in designing metal implants so that their biomechanical interaction with bone is compatible. The much higher modulus of elasticity of a metal alloy stem than that of bone can cause a decrease in bone density and bone resorption. The high modulus of elasticity can also contribute to the loosening of the implanted prosthesis.

Construction of a hip joint prosthesis

The hip joint prosthesis consists of the following elements:

1. stem,
2. head,
3. socket,
4. metallic shell.

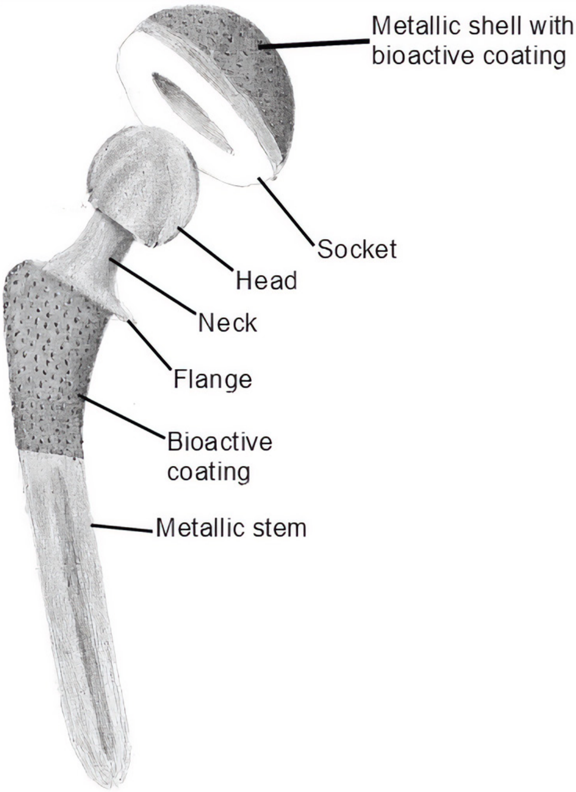

Figure 1. Schematic of typical cementless prosthesis components with total hip replacement (THR) with a combination; the head is made of cobalt alloy, and the socket (acetabulum) is made of ultra-high molecular weight crosslinked polyethylene (UHMWCLPE). Cementless hip prostheses have bioactive (hydroxyapatite composite) coatings to create a biochemical connection between the stem and the femur and the acetabulum bone from the pelvic bone.

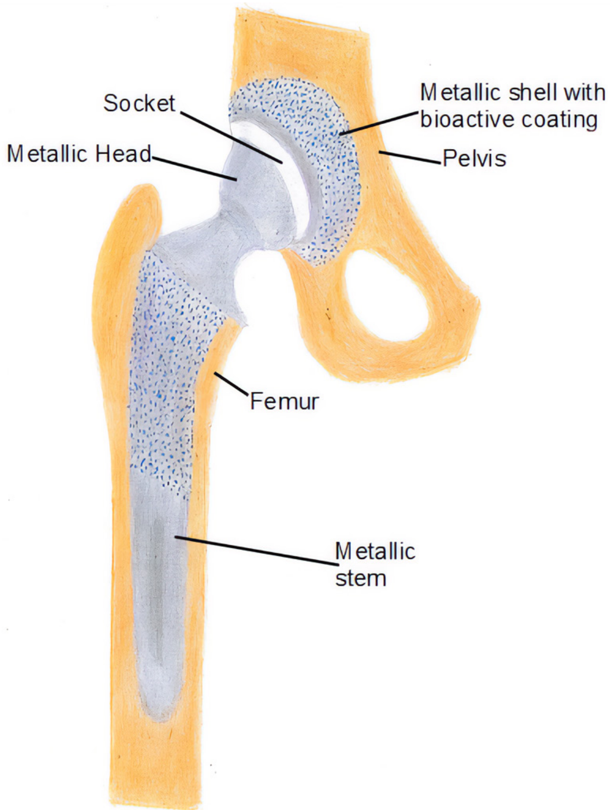

Figure 2. Schematic showing the implanted total hip cementless prosthesis.

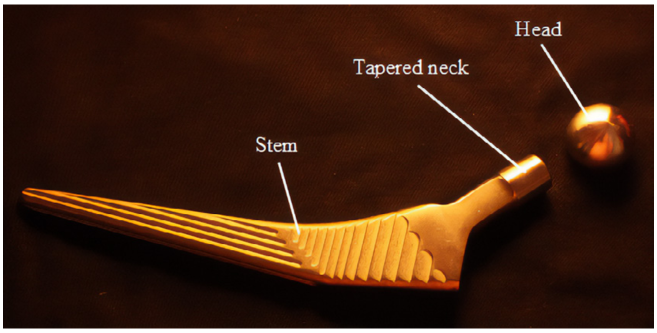

Figure 3. Cemented modular prosthesis with a tapered neck allowing connection with the head.

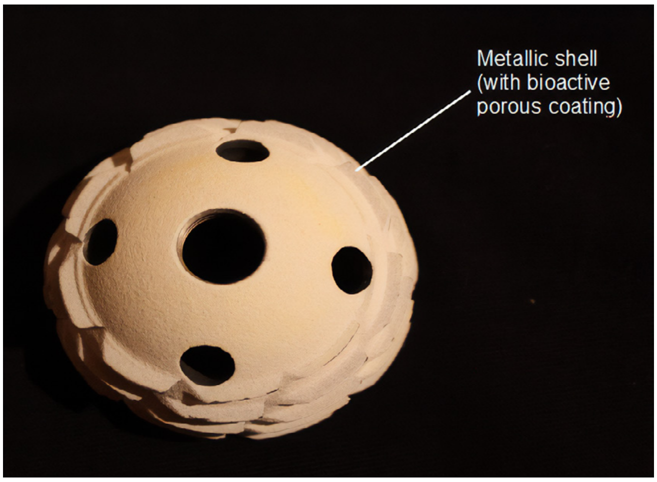

Figure 4. The shell is made of titanium alloy coated with a bioactive porous composite consisting of hydroxyapatite $Ca_{10}(PO_4)_6(OH)_2$ and tricalcium phosphate β-$Ca_3(PO_4)_2$ to create a biochemical connection with the pelvic bone. The shell has holes that are additionally used for attachment with pelvic bone screws for a more stable connection.

The stem is usually made of a metal alloy (cobalt alloy, titanium alloy, or austenitic stainless steel). The heads of hip joint endoprostheses are often made of cobalt Co28Cr6Mo alloys or dense corundum ceramics. The stem and shell are coated with a bioactive porous hydroxyapatite composite. The hip joint socket can be made of ultra-high molecular weight crosslinked polyethylene (UHMWCLPE), cobalt alloy, or corundum ceramic. The design of a hip joint prosthesis should provide an appropriate range of joint movements, load transfer, resistance to overload, resistance to vibration, damping, stimulation of bone tissue (remodeling bone tissue), abrasion resistance, and the possibility of performing surgery. The hip joint prosthesis consists of a ball-ended stem and a socket (hollow hemisphere). Cobalt or titanium alloys and stainless austenitic steels are used to construct the stems. The socket is made of ultra-high-molecular-weight-crosslinked polyethylene (UHMWCLPE, or corundum ceramics. The head is most often made of a cobalt alloy and dense corundum ceramics, Al_2O_3. The acetabulum is most often made of ultrahigh molecular weight cross-linked polyethylene (UHMWCLPE), because an important advantage of the acetabulum made of UHMWCLPE polyethylene is the elastic property of the polymer contributing to the cushioning of the prosthesis during overloads and the elastic properties of the prosthesis.

Types of hip joint prostheses

Due to the construction of the prostheses, hip joint prostheses are divided into:
1. One-piece – where the stem and head are one component,
2. Modular – where the stem and head are two components connected (**Figure 3**).

Due to the design of the prostheses stem, hip joint prostheses are divided into:
1. With stems not having a collar,
2. With stems having a collar.

To ensure proper uniform load, stability of the femur-implant connection system, and bone load, the appropriate biomaterial must be selected, and the structure, shape, and geometric dimensions must be properly designed to ensure proper uniform load on the bone. It is necessary not only to select the appropriate biomaterials but also to correctly design the shape and geometric dimensions of individual elements of the prostheses. Bone underload is one of the factors causing bone tissue loss. A very important advantage of prostheses with stems with a collar is the beneficial effect of the collar on the remodeling of bone tissue and the stability of the bone-stem connection. The stresses stimulate and intensify bone-forming cells (osteoblasts) and contribute to the formation of new bone tissue. Bone tissue is constantly undergoing reconstruction. For the proper reconstruction of bone tissue, load, stress, and proper stress distribution are needed. The stresses activate bone-forming cells (osteoblasts) responsible for the formation of new bone tissue. Stems with a collar result in a more favorable stress distribution than stems without a collar (flangeless), which is very important and effective in preventing the phenomenon of underload. A very important advantage of stems with flanges is that the transfer of stresses to the bone is similar to that in the case of natural biofouling. In the case of a stem with a collar, the load is transmitted through the collar to the bone mainly compressively, in a direction approximately parallel to the femur. In the case of the stem with a flange, the load acts mainly on the femur. A very important advantage of prostheses with stems with a collar is the beneficial effect of the collar on the proper distribution of stresses, on the stimulation and intensification of osteoblasts of bone-forming cells, on the occurrence of bone tissue remodeling, on the formation of new bone tissue, on the reduction of stress concentration at the border of the bone-stem connection, and on connection stability.

Types of prostheses depending on the method of fixing the metal stem in the femur and the socket covered with a metal coating in the pelvic bone.

Depending on the method of attaching a metal stem to the femur and covering the socket made of metal with the pelvic bone, two types of prostheses are distinguished:
1. **Cemented** - the stem in the femur and the acetabulum of the hip joint in the pelvic bone are fixed with bone cement (**Figure 3**),
2. **Uncemented** (Cementless) (**Figure 1**) - the stem and metallic shell are most often made of a titanium alloy. The acetabulum consists of an insert made most often of ultra-high molecular weight polyethylene $Ca_{10}(PO_4)_6(OH)_2$ and tricalcium phosphate β-$Ca_3(PO_4)_2$ to create a biochemical connection with the femur and the pelvic bone.

Composite hydroxyapatite coatings are used on components of prostheses made of titanium alloys because the titanium contained in them is a bioactive metal and creates a strong chemical bond with the titanium alloy[19-20]. The hydroxyapatite composite coating dissolves at a certain rate in the environment of the human body, and the components present in it (Ca and P) participate in the formation of new bone tissue (**Figures 1, 2, and 4**).

Classification of biomaterials for hip joint prostheses components

Biomaterials for hip joint prostheses are divided into the following groups:
1. metallic,
2. ceramic,
3. polymeric,
4. composite.

Materials with biocompatible and tolerated properties should be used for the elements of endoprostheses, i.e., they do not cause allergic, inflammatory, irritating, or carcinogenic effects when in contact with tissues and body fluids.

Requirements for hip prosthesis components

Individual elements of prostheses (stem, head, and socket) of the hip joint perform an appropriate function; therefore, specific requirements are set that must be met by individual elements of the hip joint prostheses. Individual elements of the prosthesis are also subjected to different types of forces, loads, and stresses. For example, the stem of a prosthesis works mainly in compression, bending, and torsion, while the head and socket work mainly in compression and abrasion. The stems of implants used for prostheses are commonly made of metal alloys.

Requirements for biomaterials for hip joint prosthesis stems

– high biocompatibility and biotolerance (does not cause allergic reactions, inflammation, formation of blood clots, toxicity, or carcinogenicity),
– non-magnetic,

- possibility of performing imaging after endoprosthesis insertion, e.g., using computed tomography and magnetic resonance imaging,
- appropriate mechanical properties,
- tensile strength,
- yield point,
- favorable strength-to-yield strength ratio,
- appropriate surface hardness in the core (core hardness),
- plasticity (elongation, narrowing),
- appropriate surface compressive stresses,
- fatigue strength (high resistance to the initiation of fatigue cracks under the influence of variable stresses),
- have an optimally high Young's modulus of elasticity E,
- appropriate impact strength (resistance to brittle fracture),
- biomechanical properties stimulating the remodeling of bone tissue (destruction of old bone tissue and creation of new bone tissue),
- have an appropriate design, and the bioactive biomaterials used for individual elements should ensure the long-term stability of the implant, its proper functioning, and its reliability.
- when using cemented endoprosthesis, the part of the surface of the stem and the acetabular casing that is in contact with the bone should have an appropriate structure and surface topography that allows filling the space between the bone and the implant with cement and creating a mechanical attachment.

An important influence on the proper remodeling of bone tissue is the biomechanical interaction between the implant inserted into the thigh bone and the process of creating and resorbing bone tissue, which depends on the size, distribution, and changes of stresses and strains in the bone.

Classification of metallic biomaterials for hip prostheses components

Metallic biomaterials for hip prostheses are divided into three groups of metal alloys

1. titanium-based alloys (titanium alloys),
2. cobalt-based alloys (cobalt alloys),
3. austenitic stainless steel.

Three groups of metal alloys are most commonly used for hip joint prostheses stems:[34,35,36].

1. two phase α+β titanium-based alloys (titanium alloys), Ti6Al4V, Ti6Al7Nb,
2. cobalt-based alloys (cobalt alloys), Co28Cr6Mo,
3. austenitic stainless steel (316L).

Two groups of biomaterials are currently most commonly used for the heads of hip prostheses:

1. cobalt alloys, Co28Cr6Mo,
2. ceramic materials, most often corundum (Al_2O_3)

The ASTM Co28Cr6Mo alloy, subjected to thermomechanical (forging) processing, is most commonly used for hip joint prosthesis stems and heads from the cobalt group.

Three groups of biomaterials are currently most often used for hip sockets:

1. ultra-high-molecular-weight crosslinked polyethylene (UHMWCLPE),
2. ceramic materials, most often corundum (Al_2O_3),
3. cobalt alloys, most often Co28Cr6Mo.

Biochemistry and biophysics of titanium and titanium alloys

Titanium is an attractive metal for alloy implants due to its good biocompatibility and mechanical properties. Titanium alloys are a group of metallic materials widely used for long-term implants in orthopedics. Titanium has many valuable properties, but it is not without its flaws. The advantages and disadvantages of titanium are presented below:

Advantages of titanium

- good biocompatibility features,
- not cause allergic reactions,
- has good resistance to general corrosion,
- resistant to pitting corrosion caused by chlorine ions (Cl^-)
- no toxicity,
- it is not carcinogenic,
- does not cause inflammation,
- low density,
- high strength-to-density ratio,
- favorable strength-to-yield strength ratio,
- has non-magnetic properties and good imaging properties in medical diagnostics,
- ability to spontaneously passivate damage to the passive oxide layer in a humid and oxygen-containing environment,
- is a bioactive metal (it can create biochemical connections with bone).

Disadvantages of titanium

- a long, energy-absorbing, expensive process,
- to the high chemical reactivity of titanium to the interstitial elements, high solubility in the liquid, and high tendency to absorb gases in the solid state at high temperatures during refining, heat treatment, and plastic processing, it is necessary to use expensive vacuum furnaces during melting, heat treatment, and expensive protective gas (high-purity argon) during plastic processing.
- titanium has a high coefficient of friction, low wear resistance, poor tribological properties, a tendency to create friction joints during moving contact, and low resistance to fretting, fretting corrosion, and fretting fatigue.
- not resistant to fluorine ions (F^-).

The structure of titanium alloys depends on chemical composition, parameters of plastic processing (thermo-mechanical treatment), and heat treatment[1,2]. Titanium alloys have a much lower Young's modulus than cobalt alloys and corrosion-resistant austenitic steels. Despite many beneficial features of the Ti6Al4V alloy, the key issue is still its low resistance to fretting abrasion and fretting corrosion in the neck and head joints of modular endoprostheses. The use of these alloys raises numerous concerns regarding their mechanical properties, which cause a biomechanical mismatch between the stem and the bone at the implantation site. The Ti6Al4V alloy, which is commonly used for prosthesis stems and acetabular casings, does not meet the requirements for chemical composition because this alloy contains the alloying components of vanadium (V) and aluminum (Al), which are considered toxic elements. Vanadium has a particularly adverse effect on the liver and kidneys. Aluminum causes damage to nerve cells, the central nervous system, brain diseases, memory problems, and Alzheimer's disease. The negative impact of these elements on the human body will result in the search for new vanadium-free alloys and the replacement of vanadium with the following elements: niobium (Nb), zirconium (Zr), and tantalum (Ta), which increase the biocompatibility of the titanium alloy. Most research on new alloys is focused on alloys containing the following alloying elements: Ti, Zr, Nb, and Ta. These elements are classified as elements that do not cause harmful interactions with the tissues and fluids of the human body. Vanadium-free titanium alloys contain these alloying elements in their chemical composition and have greater resistance to general corrosion, biotolerance, and a low Young's modulus of elasticity.

The first titanium-based alloy was a two-phase alloy ($\alpha + \beta$) Ti6Al4V, whose chemical composition was originally designed for the aviation, rocket, and armaments industries and was first melted in 1954 in the USA, in Chicago, and then, in the late 1950s, it was used on implants in orthopedics[2].

This alloy is criticized for having the following unfavorable properties:

1. mismatch of Young's modulus of elasticity and Young's modulus of elasticity of bone,
2. having in the chemical composition alloying elements (Al, V) that have a negative effect on the human body,
3. having low abrasion resistance.

The release of aluminium (Al) and vanadium (V) alloying elements into the human body, whose stems are made from the titanium alloy Ti6Al4V, occurs mainly in modular prosthesis connections as a result of fretting and fretting corrosion. The aluminum (Al) present in this alloy contributes to damage to nerve cells, neurotransmissions, and brain diseases. Vanadium (V) causes cytotoxic reactions, neurogenic disorders (Alzheimer's disease, memory loss), and has an adverse effect on the kidneys and liver. Despite drawbacks and criticism, the Ti6Al4V alloy is still widely used for implants in orthopedics. Alloys with higher purity, containing fewer interstitial elements, are marked as ELI (Extra Low Interstitial). By reducing the content of interstitial elements, the plasticity of the alloy increases. The Ti6Al4V ELI alloy is characterized by a lower content of iron (Fe) and interstitial elements, mainly oxygen (O) and hydrogen (H), which significantly improves fracture resistance[2,5]. The passive layer of titanium oxides is resistant to pitting corrosion caused by chlorine ions (Cl^-) but is not resistant to pitting corrosion caused by fluorine ions (F^-). Fluoride ions may enter bodily fluids from toothpaste, mouthwashes containing fluorine (F^-) ions, and fluoridated water. Titanium is a light metal. The mechanical properties of titanium strongly depend on its purity[21-33]. This metal exists in two allotropic varieties: low-temperature Tiα and high-temperature Tiβ. The Tiα variety crystallizes in a hexagonal system with a hexagonally compact A3 crystallographic lattice. At 883°C, the Tiα variety transforms into the high-temperature Tiβ variety with a regularly spatially centered A2 crystallographic lattice. The production of titanium on an industrial scale began relatively late, only in the 50s of the twentieth century, even though it was discovered as early as the 18th century. The reason for such late production of titanium on an industrial scale was the difficulty in extracting titanium from titanium ores due to its high affinity for oxygen, nitrogen, carbon, and hydrogen.

At temperatures above 470°C, titanium absorbs oxygen (O), carbon (C), nitrogen (N), and hydrogen gases, and the plastic properties are reduced; therefore, heat treatment and forming processes of titanium must be carried out in a vacuum or an atmosphere of protective gases (argon or helium). The mechanical properties of titanium depend strongly on the content of interstitial elements (O, N, C, H). Interstitial elements, such as oxygen, nitrogen, carbon, and hydrogen, reduce the plasticity of titanium. The hydrogen in titanium contributes to its embrittlement by forming titanium hydrides (TiH_2), which are needle-shaped with titanium. Titanium is not resistant to hydrofluoric acid (HF). The mechanical and physicochemical properties of titanium can be changed within a wide range by adding appropriate alloying elements, which have a significant impact on the transformations, phase, and chemical composition, as well as the temperature range of individual phases. The main alloy additions introduced in specific amounts into titanium to increase strength are aluminum (Al), tin (Sn), molybdenum (Mo), manganese (Mn), chromium (Cr), and iron (Fe). Alloying elements dissolving in titanium increase its strength. Alloying elements introduced into titanium increase or decrease the allotropic transformation temperature $\alpha \leftrightarrow \beta$.

A passive oxide film forms spontaneously on titanium alloys in an oxygen-containing environment. Titanium's high resistance to general corrosion is related to its high affinity for oxygen and the formation of a stable, passive layer of titanium oxides on the surface, tightly adhering to the substrate. Titanium has a strong chemical affinity for oxygen, which causes its repassivation, i.e., the formation, in contact with an environment containing oxygen, of a tight, non-conductive, and well-adhering layer of TiO, Ti_2O_3, and TiO_2 oxides, called the passive layer, which protects titanium against further oxidation. Titanium can self-passivate in the air and in physiological solutions containing oxygen. Oxygen has a strong influence on the increase in the strength of titanium, which is why this

Table 1. Chemical composition of unalloyed commercial (technical) pure (CP) titanium grades in wt% according to ASTM F 67

ASTM F67	Ti	C (max)	Fe (max)	N (max)	O (max)	H (max)	Other
Grade 1	99.5	0.08	0.20	0.03	0.18	0.15	-
Grade 2	99.2	0.08	0.25	0.03	0.20	0.15	-
Grade 3	99.1	0.08	0.25	0.05	0.30	0.15	-
Grade 4	99.0	0.08	0.50	0.05	0.40	0.15	-
Grade 7	99.2	0.08	0.30	0.05	0.25	0.15	0.2Pd

element is sometimes treated in appropriate amounts as an alloying element. Titanium has lower strength in comparison to titanium alloys but is characterized by greater resistance to general corrosion. Titanium is resistant to the effects of seawater and chloride solutions. Titanium is deformed by slipping and twinning. The mechanism and share of deformation depend on the content of interstitial elements in titanium: hydrogen (H), nitrogen (N), carbon (C), and oxygen (O). As the amount of interstitial elements increases, the share of deformation by twinning increases. Commercial Technical Titanium or Commercial Pure Titanium is marked as CT or CP-Ti. It is technically pure titanium, containing limited, permissible amounts of impurities, mainly interstitial elements (C, H, O, N) and Fe. This titanium is mainly used for melting titanium-based alloys. Technically pure titanium crystallizes in the structure of the α phase with a hexagonally compact crystallographic lattice of the A3 type. The Ti-α variant occurs up to the allotropic transformation temperature, approximately 883°C. After exceeding the allotropic transformation temperature, a β phase with a cubically centered crystallographic lattice of the A2 type is formed. The higher the number next to the symbol of a given type of titanium, the more impurities it has, and the greater its strength and lower plasticity. Depending on the grade of technically pure titanium, the permissible content of interstitial elements is specified in the ASTM F67 standard. Commercially pure titanium which is unalloyed ranges in purity of 99.5 to 99% Ti[2].The chemical composition of the main grades of commercially pure titanium (CP) is given in Table 1. According to the ASTM (American Society for Testing and Materials) standard, technically pure titanium is divided into four main grades: Grade 1, Grade 2, Grade 3, and Grade 4, with different levels of purity. These grades contain from 0.2 to 0.5 iron (Fe) and trace amounts of carbon (C), hydrogen (H), oxygen (O), and nitrogen (N). Impurities affect the properties of titanium and have a decisive impact on its application. A slight change in the chemical composition, mainly impurities with interstitial atoms (C, H, O, N), has a significant impact on the change in mechanical properties and potential application. Also, a slight change in the amount of interstitial elements significantly affects the temperature of allotropic transformations of titanium. In addition to the four basic grades of commonly used CP-Ti (technically pure titanium) commercial titanium, grades with improved corrosion resistance in reducing environments with low noble elements are used, designed grades 7,11,16,17,26, and 27, containing small amounts of palladium (Pd) or ruthenium (Ru). The most corrosion-resistant titanium grade in reducing environments is grade 7, containing Pd. Alternatives to grade 7 are other cheaper grades containing smaller amounts of Pd or containing Ru instead of Pd. The noble elements Pd or Ru are added to pure titanium or titanium alloy to increase corrosion resistance, especially to crevice corrosion where there is not enough oxygen to form a passive oxide layer. The introduction of a higher content of interstitial elements into titanium results in an increase in strength properties. In titanium alloys, the temperature of the beginning and end of the allotropic transformation α↔β depends on the chemical composition of the alloy. In titanium alloys, the elements stabilizing the α phase are aluminum, carbon, oxygen, and nitrogen. Carbon (C), oxygen (O), and nitrogen (N), because they have a small atom size, form interstitial solid solutions with titanium. Aluminum (Al) with titanium forms a multi-node solid solution. The elements stabilizing the β phase include the following elements: vanadium, niobium, tantalum, molybdenum, chromium, cobalt, and manganese. The introduction of phase-stabilizing elements β will favor the occurrence of the β phase at room temperature and below. When a given element exceeds the solubility limit in solid solutions of the α and β phases, intermetallic compounds are formed[1,2,4,6].

Selected physical and mechanical properties of titanium are presented in **Tables 2 and 3**.

Titanium is characterized by a high strength-to-density ratio and higher corrosion resistance than titanium alloys. Since technical titanium does not have sufficient mechanical properties for orthopedic implants, various alloying elements are introduced into technical titanium to improve the desired properties. To increase the strength of titanium, various alloying elements are introduced during its melting. The main alloying elements present in titanium alloys are Al, V, Mo, Nb, Zr, Ta, Mn, Fe, and Cr. Depending on the type of element introduced, there is a different impact on the mechanical, physical, and corrosion properties of a given titanium alloy. Alloying elements that dissolve in titanium increase its strength. Fe, Cr, and Al have the greatest impact on the increase in strength. During rapid cooling of the α+β two-phase alloy, a diffusion-free martensitic transformation β→α' occurs, resulting in the formation of titanium martensite marked as α' with an acicular structure. Titanium martensite (α') doesn't have as high a hardness as martensite in carbon and alloy steels.

Table 2. Selected physical and mechanical properties of titanium[1,2,4,6].

Properties	Values
Atomic number	22
Atomic mass	47.90
Allotropic transformation temperature	882°C
Crystallographic structure (lattice type)	Below the allotropic transformation temperature of Tiα↔Tiβ, 882°C, the α phase with a compact hexagonal lattice (HZ), A3, occurs. Above the allotropic transformation temperature α↔882°C, the Tiβ phase with a space-centered crystallographic lattice (RPC), A2, occurs.
Lattice constants	Ti- α, A3 (a = 0.295 nm, c = 0.4683 nm) c/a = 1.59
Color	Dark gray
Melting temperature	1668°C
Density	4.50 g/cm^3
Hardness in the annealed condition	70 HRB
Tensile strength in the annealed condition	240 MPa
Poisson's ratio	0.40
Young's modulus of elasticity	100 GPa
Thermal conductivity coefficient	22.0 W/mK
Thermal expansion coefficient	$8.6 \cdot 10^{-8}$ K^{-1}
Types of oxides formed on the surface during spontaneous oxidation in an oxygen-containing environment	TiO_2, Ti_2O_3, TiO TiO_2 oxide is formed outside the passive layer, then Ti_2O_3, and even deeper, TiO. TiO_2 oxide is the most stable.
Abrasion resistance	Titanium has very poor abrasion resistance. It has a high coefficient of friction. During contact movement of elements, it creates adhesive joints. Titanium is not suitable for use on elements subject to friction without appropriate thermal and chemical treatment. In order to increase abrasion resistance, it is necessary to modify the surface properties. Ion nitriding can be used to improve abrasion resistance.
Resistivity	0.420 Ωm
Electronic structure	$s1^2 2s^2 p6d^2 4s^2$
Crystallographic relationship of the transformation α↔β	(0001) α‖ (110) β <1120>α‖<111>β

Table 3. Mechanical properties of commercially pure titanium grades in wt% according to ASTM F 67 in the annealed condition.

Designation Unalloyed grades	Tensile strength (UTS) (min) MPa	0.2 Yield Strength (YS) (min) MPa
ASTM Grade 1	240	170
ASTM Grade 2	345	275
ASTM Grade 3	448	380
ASTM Grade 4	550	480
ASTM Grade 7	340	280

Production of titanium

Titanium is obtained by two methods:

1. On a laboratory scale, titanium of the highest purity(containing more than 99.5% Ti, and very small amounts of oxygen, nitrogen, carbon, hydrogen, and iron) is produced in small quantities in the laboratory using the so-called iodide method, mainly for special applications requiring very high purity and research purposes.

The iodide method involves thermal dissociation of titanium tetraiodide at high temperatures (1400 °C) according to the reaction:

$TiI_4 <=> Ti + 2I_2$

2. On an industrial scale, the so-called technical titanium or commercial pure titanium (CP) is obtained by the Kroll method, developed in 1932, by reducing $TiCl_4$ with magnesium according to the reaction:

The Kroll method involves the reduction of titanium tetrachloride with magnesium in a vacuum furnace, in a noble gas atmosphere, argon (Ar) at a temperature of 850°C.

$TiO_2 + 2C + 2Cl_2 \rightarrow TiCl_4 + 2CO$ $TiCl_4 + 2Mg \rightarrow Ti + 2MgCl_2$

Due to low production costs, the Kroll method is widely used on an industrial scale. Obtained as a result of the so-called reduction, "titanium sponge" is melted in vacuum and arc furnaces, in which the electrode is pressed and welded to the titanium sponge. The titanium sponge obtained as a result of the reduction is melted twice in vacuum and arc furnaces to reduce the content of interstitial elements (H, N, C, O) to the permissible content for a given grade of technical titanium. The interstitial elements oxygen (O), nitrogen (N), and carbon significantly increase the hardness and strength and reduce the ductility of titanium.

Classification of titanium alloys according to the type of microstructure

Titanium alloys are classified according to the microstructure present at room temperature. According to the type of microstructure, titanium alloys are divided into:

1. α alloys,
2. near α alloys,
3. α+β alloys,
4. stable β alloys,
5. near β,
6. metastable β alloys.

The most popular Ti-6Al-4V titanium alloy, containing 6 wt.% aluminum and 4 wt.% vanadium, was introduced in 1956 in the USA (Chicago) as an alloy for aerospace and military applications and later for medical applications. A new generation of titanium alloys with niobium and zirconium was introduced in 1979.

Table 4. Selected experimental and applied titanium alloys for implant applications[2].

Alloy design and type	Developed in
α+β Alloys	
Ti6Al4V	USA
Ti6Al4V ELI	USA
Ti6Al7Nb	Switzerland
Ti5Al2.5Fe	Germany
Ti3Al2.5Fe	USA
Ti15Zr4Nb2Ta0.25Pd	Japan
Ti5Al3Mo4Zr	Japan
β Alloys	
Ti15Mo5Zr3Al	Japan
Ti29Nb13Ta4.5Zr	Japan
Ti13Nb13Zr	USA
Ti12Mo6Zr2Fe	USA
Ti16Nb10Hf	USA
Ti35Nb7Zr5Ta	USA
Ti15Mo28Nb0.2Si0.26O	USA

Characteristics of the Ti6Al4V alloy

The introduction of an appropriate amount of β and α phase stabilizing elements into a titanium alloy results in an alloy with a two-phase microstructure (α+β). The percentage of a given phase depends on the amount of the alloying element present in the alloy. The two-phase microstructure is characterized by higher fatigue resistance than the single-phase (α+β) or single-phase β alloys, which is a very important advantage. The proportion (percentage division) of the α and β phases can be regulated by the temperature used during the plastic forming process (forging) and heat treatment. In alloys with a higher content of the aluminum alloying element (Al), stabilizing the α phase, there will be more of the α phase in the two-phase microstructure. Increasing the purity of the alloy by introducing the Ti-6Al-4V ELI alloy with a lower content of interstitial element impurities (O, N, C, H) (Extra Low Interstitial) and iron than the Ti-6Al-4V alloy significantly improves the plasticity of the alloy. Plastic processing and heat treatment have a significant effect on the type of the resulting microstructure and mechanical properties of the implants[9-22].The Ti6Al4V alloy is commonly used for the femoral stem and metallic shell housing of the hip socket. This alloy is characterized by good general corrosion resistance in the environment of the human body. Since titanium is a bioactive metal, i.e., it can create chemical bonds with bone tissue, this alloy is often used to produce cementless endoprostheses with bioactive coatings made of hydroxyapatite composites. This alloy has an α+β two-phase microstructure. The β phase is obtained in this alloy by adding stabilizing elements, which include vanadium. Aluminum reduces the specific weight of the alloy and increases its mechanical properties. By applying different types of heat treatment to this alloy, different types of microstructures can be obtained: equiaxed, lamellar, and characterized by different mechanical properties (Table 5). This alloy is characterized by low resistance to fretting corrosion in the neck and head joints of modular prostheses because it is characterized by a low ability to repassivate in an environment of low oxygen content and a lower modulus of elasticity. Young's longitudinal modulus of elasticity E of the Ti6Al4V alloy is more similar to that of bones compared to the Co28Cr6Mo alloy commonly used for stems and heads of hip joint endoprostheses. The modulus of elasticity has a significant impact on the fatigue strength of the occurrence of fretting, fretting corrosion, and the resistance to fatigue cracking of the neck in the neck-head connection in modular prostheses. The values of Young's modulus of elasticity of selected alloys compared with the value of the modulus of elasticity of the compact bone of the femur are presented in Table 6. The influence of Young's modulus of elasticity on the fatigue strength of selected alloys for prosthetic implants is presented in Table 7. As Young's modulus of elasticity increases, the fatigue crack resistance of metal alloys increases. Titanium alloys are partially susceptible to fretting and galling, which cause surface damage and wear due to friction, creating adhesive and friction joints during hot or cold working. Also, titanium alloys are susceptible to pitting corrosion caused by fluorine (F^-) ions that may be present in the body fluids of the human body. Fluoride ions may enter body fluids from toothpaste, mouthwash, and fluoridated water. Titanium oxides are sensitive to fluoride ions, which is a particular problem because fluoride toothpaste, fluoride mouthwashes, and fluoridated water contribute to the supply of the most aggressive ions, causing pitting corrosion in titanium alloys used for orthopedic and dental implants[1,3]. Titanium and its alloys tend to absorb gases (oxygen, nitrogen, and hydrogen) at elevated temperatures. The alpha case, also called the "white layer" in titanium and titanium alloys, is a surface layer enriched in oxygen, formed at temperatures above 480°C due to oxygen absorption on the surface and its diffusion into the surface layer. The presence of alpha case on the surface of finished denture components is unacceptable because it significantly reduces fatigue resistance. The alpha case is caused by an improperly applied atmosphere during thermoplastic treatment, shaping of implants in the dies of forging presses, or during heat treatment. To prevent the occurrence of alpha cases, thermoplastic treatment and heat treatment should be carried out in a protective atmosphere free of oxygen. The occurrence of the alpha layer can be avoided by carrying out heat treatment in vacuum furnaces containing argon. If it is already present on the surface, it can be removed by machining, e.g., milling, or by electrochemical treatment in solutions containing hydrofluoric acid (HF). It should be borne in mind that hydrofluoric acid is highly corrosive. During the microstructure analysis inspection, finished components must be checked for the presence of the alpha phase using an optical microscope. The type of microstructure has a significant impact on the mechanical properties of alloys. Table 5 presents the influence of equiaxial and lamellar microstructures on the mechanical properties of Ti6Al4V and Ti6Al4V ELI alloys determined using standard test methods according to ASTM E8 and ASTM E399 specifications.

Table 5. Mechanical properties of annealed *Ti6Al4V and Ti6Al4V ELI alloys forgings tested according to ASTM E 8 and ASTM E399 standards

Microstructure	Tensile Strength UTS)	Yield strength YS (MPa)	Elongation, El (%)	Reduction area, (RE) (%)	Critical Stress Intensity Factor K_{1C} ($MPa \cdot m^{1/2}$)
Ti-6Al4V Equiaxial	1010	950	15	35	60
Ti6Al4V lamellar (plate)	950	885	13	25	75
Ti6Al4V ELI Equiaxial	900	835	17	40	90

* Annealed 2 hours at 700 °C

Table 6 presents a comparison of the longitudinal Young's modulus of selected alloys with the longitudinal modulus of the cortical bone of the femur tested according to the ASTM E8 specification.

Table 6. Values of Young's modulus of elasticity (E) of selected metal alloys compared to the compact bone of the femur tested according to ASTM E 8 standard

Material	The value of Young's modulus E GPa
Cortical bone of the femur	30
Ti13Nb13Zr	80
Ti6Al4V	115
Ti6A4V- ELI	110
Ti6Al7Nb	105
Austenitic stainless steel (316L)	210
Co28Cr6Mo	230

Among the mechanical properties of the alloys in the annealed state presented in Table 7, the Ti13Nb13Zr alloy has the lowest fatigue strength because it has the lowest longitudinal Young's modulus of elasticity, whereas the highest fatigue strength is observed in the CoCr28Mo6 alloy because it has a higher Young's modulus of elasticity. As the Young's modulus of elasticity increases, the fatigue strength increases.

Conclusions

1. Titanium is used for titanium alloys used in implants because it has desirable properties. The most important advantages of titanium that have a beneficial effect on the properties of titanium alloys used in orthopedic implants include an excellent combination of mechanical properties, a high strength-to-density ratio, low density, good resistance to general corrosion, and non-magnetic properties. Titanium and titanium alloys are characterized by poor tribological properties: a high coefficient of friction and low abrasion resistance.

2. Titanium alloys are characterized by low resistance to fretting damage in the connections of the stem neck with the endoprosthesis head. Fretting damage in the connections of modular endoprostheses is a serious problem because it can lead to fretting corrosion and fatigue cracking. Fretting damage to the neck of the stem occurring in modular endoprostheses made of titanium alloys is caused by elastic micro-deformations and micro-movements resulting from the relatively low Young's modulus of longitudinal elasticity (E). Another unfavorable property that influences fretting damage to the neck of the endoprosthesis stem is the relatively low hardness of titanium alloys in the annealed state and low abrasion resistance. The resistance of titanium alloys to fretting damage can be increased by strengthening the surface with thermochemical treatment using the ion nitriding technique. Additionally, resistance to fretting damage can be obtained by designing a connection in which the neck is small, larger in diameter, and, therefore, more rigid. The release of aluminium (Al) from the Ti6Al4V alloy due to fretting and fretting corrosion from modular hip joint protheses into the human body primarily contributes to nerve cell damage, neurotransmission disorders, and brain diseases (Alzheimer's, memory loss). The release of vanadium (V) from this alloy into the human body mainly causes cytotoxic reactions, neurological disorders, brain diseases(Alzheimer's disease, memory loss), and damage to the kidneys and liver.

3. Titanium nitrides (TiN, T_2N) are "gold" in color, very hard, and brittle. Titanium nitride zones in the nitrided layer reduce fatigue crack resistance. A titanium nitride layer in the top layer of the hip joint prosthesis stem is not allowed. The titanium nitride layer cracks under the influence of bending stresses. Cracking the nitride layer is a place of stress concen-

Table 7. Mechanical properties of selected annealed alloys for hip joint prosthesis implants compared to the Ti13Nb13Zr alloy[2,4].

Alloy designation	State	Modulus of elasticity [GPa}	Yield strength YS 0.2 [MPa]	Tensile strength UTS [MPa]	Elongation [%]	Fatigue strength for 10^7 cycles FST [MPa]
CoCr28Mo6	Forged	230	1,600	1,800	8	800
Ti6Al4V	Forged	115	900	980	12	730
Ti6Al4V ELI	Forged	110	880	950	13	680
Ti6Al7Nb	Forged	105	860	900	15	600
Ti13Nb13Zr	Forged	80	840	880	18	500

tration where a fatigue crack starts. Only the solid solution zone increases fatigue crack resistance.

4. During the nitriding of titanium alloys, hydrogen (H_2) should not be present in the furnace atmosphere because hydrogen has a small diameter and, at elevated temperatures, easily diffuses into titanium alloys and forms titanium hydrides (TiH_2), which contributes to hydrogen embrittlement. Titanium hydrides have the shape of long, thin, brittle needles and contribute to hydrogen embrittlement. To prevent hydrogen embrittlement, the melting process, heat and thermomechanical treatment, and heat treatment processes should be carried out in protective atmospheres free of hydrogen.

5. The alpha case is caused by an improper protective atmosphere during the technological process. The alpha case is hard and brittle, and reduces resistance to fatigue cracking. The cause of the alpha case of a titanium alloy element is thermomechanical treatment or heat treatment in atmospheres containing oxygen, nitrogen, and hydrogen. To prevent alpha cases, it is necessary to carry out thermomechanical and heat treatment of components made of titanium alloys in a protective atmosphere.

6. Due to significant differences in properties between a stem made of a metal alloy and compact bone, there is no biomechanical compatibility.

7. It is possible to produce a solid metal alloy that has Young's elastic modulus (E) equal to that of bone and has good resistance to fatigue cracking.

8. Bone is a completely different type of material from metal alloys; it has a different microstructure and mechanical properties.

9. Young's modulus of elasticity characterizes the stiffness of the metal alloy. The resistance to fatigue cracking and fretting fatigue of metal alloys depends strongly on Young's modulus of elasticity, E.

10. Young's high elastic modulus (E=230 GPa) of the Co28Cr6Mo alloy has a beneficial effect on the following properties of hip joint components made of this alloy:

a) increases resistance to fretting damage to the cervical surface in the connection between the stem and the head in modular endoprostheses. In the case of a stem made of an alloy with a lower elastic modulus, micro-movements and elastic micro-deformations of the neck occur in a modular endoprosthesis, which leads to damage to the passive oxide layer and, consequently, to fatigue cracking of the neck. Micromovements and elastic micro-deformations with low oxygen supply to the gap in the junction of the neck and head of the modular endoprosthesis do not allow passivation.
b) increases the resistance to fatigue cracking of hip joint endoprostheses stems,
c) increases the resistance to contact fatigue of a head made of this alloy,
d) increases resistance to fretting damage to the cervical surface in the connection between the stem and the head in modular endoprostheses,
e) increases resistance to fretting corrosion in the neck and head joints in modular prostheses,
f) increases resistance to fretting-fatigue cracking.

11. The Co28Cr6Mo cobalt-based alloy is characterized by good tolerance in the environment of tissues and body fluids and better resistance to crevice and fretting corrosion compared to titanium-based alloys and austenitic stainless steels. The Co28Cr6Mo alloy is characterized by the best mechanical properties among metal alloys used for endoprosthesis components. High resistance to fatigue cracking is associated with the high Young's modulus of elasticity E (high stiffness) of this alloy.

12. The most common cause of fractures of hip endoprosthesis stems made of metal alloys is fatigue. It is estimated that the frequency of fractures due to the fatigue mechanism is over 80%. Therefore, improving the resistance to fatigue fracture of materials used for endoprosthesis stems is very important. The consequences of the fracture of the hip endoprosthesis stem are the necessity of re-exchanging the stem, which is associated with new experiences, usually greater problems associated with the removal of the fractured stem that was fixed in the femur, and the occurrence of damage to the femur during the removal of the stem connected to the bone (fixed in the femur), creating a strong connection of the stem with the bone. In addition, there are additional costs associated with the purchase of a new, expensive endoprosthesis and costs associated with the replacement with re-implantation of the endoprosthesis. Recovery may be much longer, especially due to the possibility of complications associated with the re-exchange of the endoprosthesis. Therefore, the selection of appropriate materials has a very important impact on the success rate (success of the performed implantation (performed procedure) of the prosthesis, reliability, and recovery time (convalescence)).

13. Stress shielding is a phenomenon of bone tissue loss (reduction in bone density) as a result of which there is a decrease in mechanical properties caused by insufficient bone loading, improper distribution of stresses and strains associated with the introduction of the implant, and the creation of a zone in which the bone tissue of the bone is drastically relieved at the implant compared to other parts of the bone. "Stress shielding" is related to the difference in the modulus of elasticity, the lack of compliance with the size of the bone module, and the metal implant with a significantly higher Young's modulus of elasticity. The load is mainly transferred by the implant and not by the bone. The high modulus of the implant made of a metal alloy with a high Young's modulus of elasticity increases the mismatch between the bone modulus and the implant, which increases stress shielding, which is responsible for the decrease in radiological bone density. Bone tissue is constantly remodeling. Bone loading has a significant impact on the formation of new bone tissue (remodeling—the reconstruction of bone tissue).

14. The collar endoprosthesis stem contributes to the favorable loading of the femur and reduces the effect caused by the difference in the modulus of elasticity of the implant and the bone. Bone loading stimulates (intensifies the formation of new bone tissue) the remodeling of bone tissue.

15. Young's modulus of elasticity has a significant effect on the resistance of metal alloys to fatigue cracking resistance. With a decrease in Young's modulus of elasticity, fatigue cracking resistance decreases.

16. The Ti13Nb13Zr alloy is not used for hip joint replacement components because it has low resistance to fatigue cracking as a consequence of the low Young's modulus of elasticity, E.

17. Titanium alloys in the annealed condition have a Young's modulus of elasticity that is closest to the modulus of elasticity of compact bone.

18. Titanium and its alloys are capable of forming a passive oxide layer in an oxygen-containing environment.

19. Titanium alloys are not resistant to corrosion in a reducing environment.

20. Titanium alloys for modular hip joint endoprostheses have low resistance to fretting damage and fretting corrosion of the neck of the stem in modular connections.

21. Low resistance to fatigue damage is caused by the relatively low Young's modulus of elasticity and poor tribological properties of titanium alloys. Titanium alloys have a low resistance to crevice corrosion in modular endoprosthesis connections.

22. Currently used cobalt alloys have greater resistance to fretting damage, fretting corrosion, and fatigue cracking than currently used titanium-based alloys.

23. Cobalt-chromium-molybdenum Co28Cr6Mo is commonly used on endoprosthesis heads in the material combination (system) metal on polyethylene or metal on ceramic.

24. Ultra-high molecular weight polyethylene (UHMWPE) with cross-linking is widely used as a socket in hip endoprostheses because it has good abrasion resistance and load absorption properties.

25. Compact bone has a low longitudinal modulus of elasticity (Young's modulus (E= 30 GPa)).

26. The fatigue strength of metal alloys used for hip joint prosthesis stems depends strongly on the alloy microstructure and Young's modulus of elasticity. As Young's modulus of elasticity increases, the fatigue strength increases.

27. The Co28Cr6Mo alloy has the highest Young's elastic modulus (E=230 GPa) of all metal alloys used for hip replacement components.

28. The higher resistance to fretting damage, fretting corrosion, and fatigue cracking of cobalt alloys than of titanium alloys used for modular endoprostheses stems from a significantly higher Young's modulus of elasticity and higher abrasion resistance.

29. The Co28Cr6Mo alloy has the best mechanical properties of all metallic materials used for components of hip joint endoprostheses. High Young's modulus of elasticity has a significant impact on the mechanical properties.

30. Titanium alloys are resistant to pitting corrosion caused by chlorine ions (Cl^-) but are not resistant to pitting corrosion caused by fluorine ions (F^-).

31. Austenitic stainless steels are not resistant to pitting corrosion caused by chlorine ions (Cl^-).

32. The most common cause of pitting corrosion of austenitic stainless steels is non-metallic impurities, mainly sulfur from iron ore.

33. Due to the limited lifetime of endoprostheses, it is advisable to continue work aimed at improving the materials used and searching for new ones that will enable a more durable connection of the endoprosthesis elements with compact bone and reduce the number of generated wear products.

Note

The information in this article is for informational and educational purposes and is not meant as medical advice or recommendation. Only a qualified orthopedic surgeon can determine which material implant system is best for a person. There are many factors that the surgeon uses when recommending hip implants (state of health, weight, age, life activity, anatomy, etc.). Any questions and concerns regarding specific types of materials and implants should be discussed with an orthopedic surgeon. Every patient's case is unique, and each patient should follow specific instructions from his or her doctor. The information in this article does not replace the orthopedic surgeon's specific instructions.

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